Hearing Aids

Harvey Dillon

Second Edition

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This book is designed to educate, and be a reference tool for, students and professionals engaged in hearing rehabilitation through the use of hearing aids. While the book aims to have a wide scope, it is not the intention of author or publisher to present all the information that is available on this topic. The reader should obtain and read all the material that the reader considers necessary to competently perform hearing rehabilitation.

Audiology is an ever-changing science. Research and clinical experience are continually broadening and changing our knowledge, in particular our knowledge of proper treatment. Every effort has been made to make this book as accurate as possible. However, there may be mistakes both typographical and in content. This text should therefore be used only as a general guide and not as the ultimate source of information on hearing aids.

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PREFACE

Rationale for the book

This book collects together knowledge about hearing aids that should be useful to student clinicians, practicing clinicians, and engineers involved in developing improved hearing aids. I have aimed to make the book both practically useful and theoretically sound. Issues are explained rather then described. Wherever possible, practical recommendations are based on empirical research, and where there is no research to draw upon, the tentative nature of the recommendation or conclusion is indicated.

The book is designed to be read on a number of levels. For readers who need only an overview of a topic, the synopsis at the beginning of each chapter should suffice. Most readers will hopefully be enticed to read further. To this end, what I consider to be the essential information on each topic is marked with a blue bar in the margin. These marked paragraphs are designed to be read, and to be understandable, without the intervening paragraphs. I think of the marked material as a thin book buried inside this thick book. Some academic courses may wish to restrict themselves to just this material, which includes approximately half the book.The remaining material provides a greater level of detail. Finally, the most detailed comments are tucked away as footnotes and as sections in small print. The material is further segmented by presenting the most theoretical material in green boxes, the most practical material in blue boxes, and summary information in pink boxes.

The issue raised in the first edition of what to call a person with a hearing impairment who is seeking help to overcome the difficulties caused by his or her hearing loss remains. The possible terms of *patient*, *client*, *consumer*, and *customer* all seem to offend someone. At different parts of the encounter, different terms seem most appropriate. The person could be a patient when his or her hearing is being assessed, a consumer when he or she is deciding whether to buy an advanced (and expensive) hearing aid or a more basic one, a client as he or she works through communication problems with the advice and guidance of the clinician, and a customer when evaluating whether the total package has been good value. From the perspective of hearing impaired people, what the clinician says and does to help people is important, as is the attitude of the clinician towards people. The term used to describe these people is much less important than the attitude of the clinician and probably only becomes important if it affects that attitude. This book takes an extremely client-centered (or patient-centered!) approach to rehabilitation. In keeping with the most common usage, and in line with a survey of what most people attending a hospital outpatient hearing clinic expect to be called (admittedly a biased sample),[1302a] I have mostly adopted the term "patient" throughout this book. Feel free to mentally replace it with your preferred term if you wish.

Changes to the second edition

All chapters have been completely reviewed. Some have had only minor changes; some have had major changes, and a new chapter has been added reflecting the huge importance of directional microphones and the growing range of digital signal processing strategies. The degree of change necessary in each chapter was determined by the answer to two questions:

- What research has been published in the last decade that has caused a change of (my) understanding of the issues, or that should cause some change in clinical practice?
- What technological advances have been made in the products (hearing aids, ALDs and implanted devices) that clinicians need to know about?

If the answer to *both* questions was "not much", then I didn't change much either. In many places, evidence that emerged during the last decade either strengthened or challenged assertions made in the first edition. Where this was the case I have changed the text accordingly. I have tried hard to convey the level of certainty on any assertion that seems justified on the basis of the evidence available. I have changed nuances about this certainty in many, many places. Any assertions with the words *may* or *possibly* have a low level of evidence behind them, and should be taken as possibilities yet to be confirmed.

A major change is the prominence given to open fittings. The use of open earmolds is actually far from new and was covered in the first edition. However, the wide availability of effective feedback cancellation and thin-tube canal fittings have made these fittings used for, and valuable to, a large proportion of clients. The implications of open fittings are described in many places throughout the text.

Thanks to many

I am incredibly grateful to an army of friends, experts all, from around the world who graciously read and commented on sections, chapters, or several chapters of drafts of this book. I like to think this process has given the book the advantages (consistency and inter-relatedness) of a single-author book plus the breadth and width of knowledge that can only be achieved in a multi-author book. (You, the reader, are in serious trouble if it has actually worked the other way around.) To all of the following I give my great thanks for their help and generosity in reviewing and/or providing information: Harvey Abrams, Iris Arweiler, Martina Bellanova, Virginia Best, Arjan Bosman, Eric Burwood, Peter Busby, Sharon Cameron, Simon Carlile, Teresa Ching, Terry Chisolm, Laurel Christensen, Cynthia Comptom, Robyn Cox, Huanping Dai, Ole Dyrlund, Kris English, Carol Flexer, Mark Flynn, Kirsty Gardner-Berry, Megan Gilliver, Helen Glyde, David Hartley, Heike Heuermann, Louise Hickson, Larry Humes, Earl Johnson, Dirk Junius, Gitte Keidser, Alison King, Linda Kozma-Spytek, Sophia Kramer, Frances Kuk, Ariane Laplante-Lévesque, Stefan Launer, Dawna Lewis, Brian Moore, Hans Mulder, Kevin Munro, Graham Naylor, Anna O'Brien, Unn Siri Olsen, Wendy Pearce, Rainer Platz, David Preves, Henning Puder, Gary Rance, Jason Ridgway, Gabi Saunders, Richard Seewald, Karolina Smeds, Pauline Smith, Michael Stone, Robert Sweetow, Janette Thorburn, Peter Van Gerwen, Andi Vonlanthen, Wayne Wilson and Justin Zakis. Many of the useful comments received from another army of expert colleagues, as listed in Edition 1, are retained in this edition also. Thanks also to Steven Banning for careful editing. Any faults in the final product are, of course, mine alone, and I would appreciate learning of them as soon as possible. Comments can be sent to: publisher@boomerangpress.com.au.

For over thirty years I have been in an organization dedicated to effective habilitation and rehabilitation of people with hearing impairment, using clinical methods founded on research-based evidence. Much of my knowledge and beliefs on the topics in this book have been shaped by the talented people who have educated and inspired me concerning both research and practical clinical issues. Foremost among these people are the late Denis Byrne, and my close collaborators Teresa Ching and Gitte Keidser. Internationally, I would particularly like to thank Arthur Boothroyd, Don Dirks and Harry Levitt for their teaching, encouragement and friendship. With each year since his passing, I am more and more missing Stuart Gatehouse's insights and contributions.

Finally, and most importantly, I would like to thank my wife, Fiona Macaskill, without whom neither the first edition of the book nor this updated edition would have come into existence. Fiona has kept our family intact during the two 3-year periods that each edition intruded most unreasonably into family life. Fiona's great clinical expertise has also provided me with many audiological insights that I would never have otherwise gained.

To Fiona, Louisa and Nicholas, for their continued patience and understanding.

CHAPTER 1

INTRODUCTORY CONCEPTS

Synopsis

Hearing aids partially overcome the deficits associated with a hearing loss. For a sensorineural hearing loss, there are several deficits to be overcome. Some sounds are inaudible. Other sounds can be detected because part of their spectra is audible, but may not be correctly identified because other parts of their spectra (typically the high-frequency parts) remain inaudible. The range of levels between the weakest sound that can be heard and the most intense sound that can be tolerated is less for a person with sensorineural hearing loss than for a normal-hearing person. To compensate for this, hearing aids have to amplify weak sounds more than they amplify intense sounds. In addition, sensorineural impairment diminishes the ability of a person to detect and analyze energy at one frequency in the presence of energy at other frequencies. Similarly, a hearing-impaired person has decreased ability to hear a signal that rapidly follows, or is rapidly followed by, a different signal. Hearing-impaired people are also less able to separate sounds on the basis of the direction from which they arrive. This decreased resolution (frequency, temporal, and spatial) means that noise, or even other parts of the speech spectrum, will mask speech more than would be the case for a normal-hearing person.

The physiological origins of sensorineural hearing loss include loss of inner hair cell function, outer hair cell function, reduced electrical potential within the cochlea, and changes to the mechanical properties of the cochlea. The resulting auditory deficits mean that a person with a sensorineural hearing impairment needs a signal-to-noise ratio greater than normal in order to communicate effectively, even when sounds have been amplified by a hearing aid. In contrast, a conductive impairment simply attenuates sound as it passes through the middle ear, so the amplification provided by hearing aids comes close to restoring hearing to normal.

To understand how hearing aids work, the physical characteristics of signals must be understood. These characteristics include the rate at which sound fluctuates (frequency), the time taken for a repetitive fluctuation to repeat (period), the distance over which its waveform repeats (wavelength), the way sound bends around obstacles (diffraction), the strength of a sound wave (pressure and sound pressure level), the break-up of a complex sound into pure tone components at different frequencies (spectrum), or into several frequency bands (octave, one-third octave or critical bands), and the degree to which a body of air vibrates when it is exposed to vibrating sound pressure (velocity and impedance).

The amplifiers inside hearing aids can be classified as linear or nonlinear. For sounds of a given frequency, linear amplifiers amplify by the same amount regardless of the level of the signal, or what other sounds are simultaneously present. By contrast, the amplification provided by a nonlinear amplifier varies with the amplitude of the signal input to the amplifier. The degree of amplification can be represented as a graph of gain versus frequency (gain-frequency response), or as a graph of output level versus input level (I-O curve). The highest level produced by a hearing aid is known as the saturation sound pressure level (SSPL). SSPL is usually estimated by measuring the output sound pressure level for a 90 dB SPL input (OSPL90). The sound output by a hearing aid can be measured in the ear canal of an individual patient, or in a small coupler or ear simulator that has a volume similar to that of a real ear.

Hearing aids are described according to where they are worn. In order of decreasing size these categories are: body, spectacle, behind-the-ear, in-the-ear, in-the-canal and completely-in-the-canal. For behind-the-ear hearing aids, further categorization is needed to distinguish between styles where the hearing aid receiver (the output transducer) is within the hearing aid case or within the ear canal.

Decreasing size has been a constant trend during the history of the hearing aid. This history can be divided into six eras: acoustic, carbon, vacuum, transistor, digital, and wireless. The last of these eras, which we are just entering, promises to hold advances at least as significant as in the eras that preceded it.

Hearing aids are designed and fitted to lessen the specific problems faced by hearing-impaired people and so improve their life quality. To better appreciate what hearing aids can and cannot do, we will briefly review the ways in which hearing abilities deteriorate when a hearing loss occurs.

1.1 Problems Faced by People with Hearing Impairment

Hearing loss is a multifaceted loss of hearing ability. Except where noted, the following descriptions apply to the most common form of hearing loss, a ***sensorineural hearing loss***.

1.1.1 Decreased audibility

Hearing-impaired people do not hear some sounds at all. People with a severe or profound hearing loss may not hear any speech sounds, unless they are shouted to at close range. People with a mild or moderate loss are more likely to hear some sounds and not others. In particular, the softer phonemes,[a] which are usually consonants, may simply not be heard. For example, the sequence of sounds *i e a ar*, might have originated as *pick the black harp*, but could be heard as *kick the cat hard*. To make sounds audible, hearing aids have to provide amplification, and this they do very well.

Hearing-impaired people also have trouble understanding speech because essential parts of some phonemes are not audible. To recognize speech sounds, the auditory system must determine which frequencies contain the most energy. The vowel *oo* for example, is differentiated from the vowel *ee* by the location of the second intense region (the second ***formant***), as shown in Figure 1.1. If, for example, a hearing loss caused all frequencies (and therefore all formants) above 700 Hz to be inaudible, as indicated by the shaded region, the two sounds could not be differentiated. Although both sounds could be detected, the similarity of their first formants would make them sound almost identical.

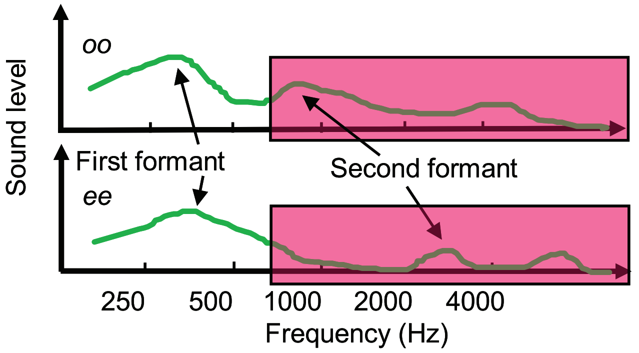

Figure 1.1 Similarity of the two vowels, oo and ee, when the second formant is inaudible because of hearing loss (pink area).

The high-frequency components of speech are weaker than the low-frequency components.[227] Furthermore, for approximately 90% of hearing-impaired adults and for 75% of hearing-impaired children, the degree of impairment worsens from 500 Hz to 4 kHz.[1113] Most commonly, therefore, hearing-impaired people miss high-frequency information. Because the loudness of speech mostly originates from the low-frequency components, hearing-impaired people may not realize that they are hearing *less* of the speech signal, even when they cannot understand speech in many circumstances. Statements such as *speech is loud enough, but not clear enough* and *if only people would not mumble* are common.

To help overcome this difficulty, a hearing aid has to provide more amplification for frequencies where speech has the weakest components and where hearing loss is the greatest (i.e. usually the high frequencies). Hearing aids are very good at providing different amounts of gain in different frequency regions, and for many years, hearing aid selection consisted primarily of prescribing and adjusting the amount of gain provided at each frequency. This was achieved by selection of an appropriate model of hearing aid and by variation of the tone controls.

1.1.2 Decreased dynamic range

As implied above, soft sounds can be made audible merely by amplifying them. Unfortunately, it is not appropriate to amplify all sounds by the amount needed to make soft sounds audible. A sensorineural hearing loss increases the threshold of hearing much more than it increases the threshold of loudness discomfort.[1695] In fact, for mild and some moderate hearing losses, there is likely to be very little increase in loudness discomfort level, even though the threshold of hearing has increased by up to 50 dB.[846, 1393, 1579] Consequently, the ***dynamic range*** of an ear (i.e. the amount by which the discomfort threshold exceeds the threshold of audibility) with a sensorineural impairment will be less than that of a normal-hear-

[a] Phonemes are the basic sounds of speech, such as individual consonants or vowels.

ing ear. Another consequence is that each increase of sound level will produce a bigger loudness increase for a hearing-impaired person than for a normal-hearing person,[716] which is referred to as recruitment.

This problem of decreased dynamic range is shown pictorially in Figure 1.2. For normal-hearing Norm (a), a wide range of sounds in the environment can fit between Norm's threshold of hearing and the loudest level he can comfortably tolerate (the white region). For Sam, the range of sound levels in the environment exceeds his dynamic range from threshold to discomfort. Part (b) shows what happens without amplification: weak to moderate sounds are not heard. Part (c) shows what happens when there is enough amplification to make the weak sounds audible: the medium to intense sounds now become excessively loud. If the sounds in the environment are to fit within Sam's dynamic range, a hearing aid must give more amplification to weak sounds than it does to intense sounds. This squashing of a large dynamic range of levels in the environment into a smaller range of levels at the output of the hearing aid is called ***compression***. In essence, a compressor is nothing more than an amplifier that automatically turns itself down as the sound gets stronger.

Hearing aids are very good at reducing the dynamic range of the signal and compression can be applied to this task in several ways. As we shall see in Chapters 6 and 10, we are not certain of the best way to decrease dynamic range, but on the bright side, we have several good alternatives from which to choose.

1.1.3 Decreased frequency resolution

Another difficulty faced by people with sensorineural hearing loss is separating sounds of different frequencies. Different frequencies are represented most strongly at different places within the cochlea. In an unimpaired cochlea, a narrowband sound (i.e. one containing power within a restricted range of frequencies) produces a clearly defined region of relatively strong vibration centered on one position on the basilar membrane. In turn, this produces a clearly

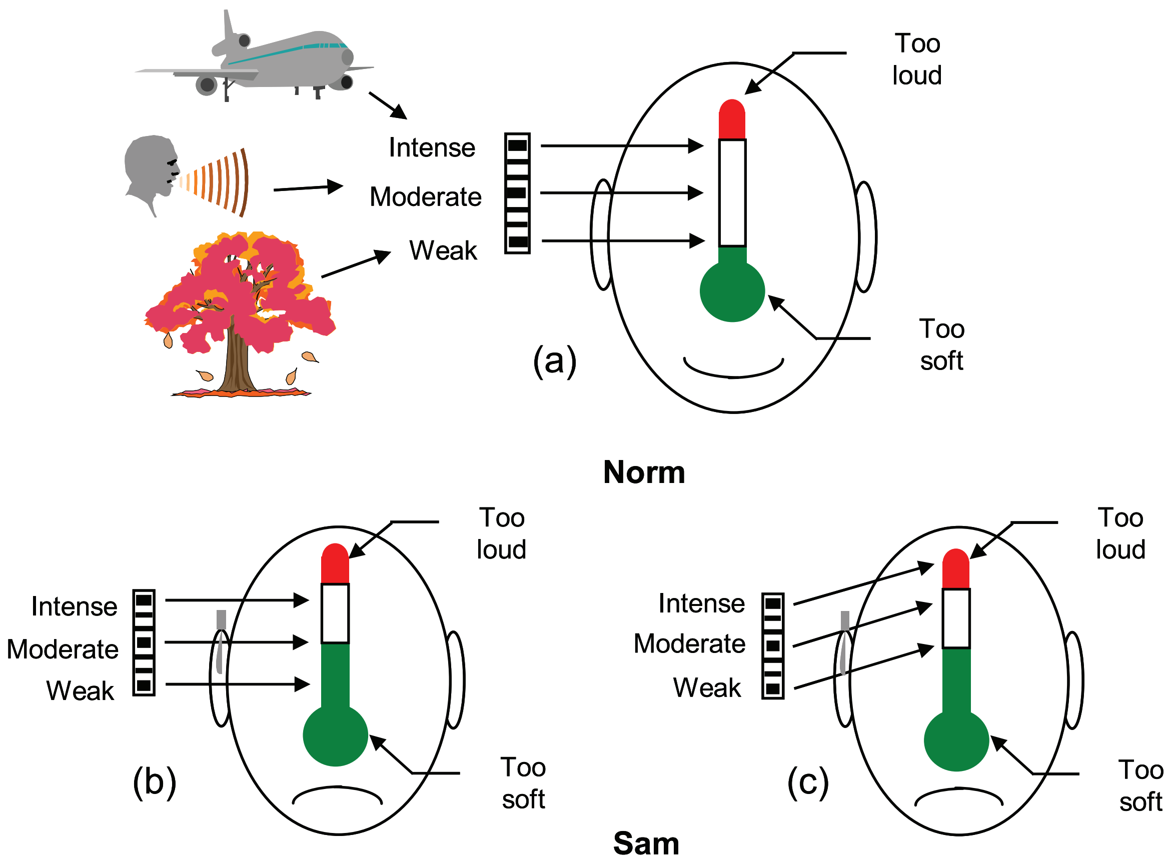

Figure 1.2 The relationship between the dynamic range of sounds in the environment and the dynamic range of hearing for: (a) normal hearing, (b) sensorineural hearing loss without amplification, and (c) sensorineural hearing loss with a constant amount of amplification for all input levels.

defined region of activity within the auditory cortex. For a complex speech sound, each small frequency region containing intense components of that sound also produces a narrow, clearly defined region of activity within the cochlea.

If a background noise contains some energy at a frequency close to one of the components of the speech sound, the normal-hearing ear can do a good job of sending separate signals to the brain, one for each region of intense activity in the cochlea. The brain can then consider all the spectral information it is getting, as well as visual information (e.g. from lip-reading), information about the direction of arrival of the sounds (by comparing the sounds received by each ear), and information about the context of the message (especially if it is speech). Armed with all this information, the brain can then partly ignore the activity originating from the noise, and decode the activity represented by the target speech. That is, the ear has ***frequency resolution*** or ***frequency selectivity*** sufficiently precise to enable the brain to separate speech from noise, provided the speech component and the noise are sufficiently separated in frequency, given their relative levels.[b]

A person with sensorineural hearing loss has decreased frequency resolution. The outer hair cells normally increase the sensitivity of the cochlea for frequencies to which the corresponding part of the cochlea is tuned. When the outer cells lose their amplifying ability, the cochlea therefore loses some of its frequency selectivity. Psychoacoustically, this shows up as flatter ***masking curves*** and ***tuning curves***.[1969] The significance of this deficit is that even when a speech component and a noise component have different frequencies, if these frequencies are too close the cochlea will have a single broad region of activity rather than two separate regions. Consequently, the brain is unable to untangle the signal from the noise.

The situation is represented pictorially in Figure 1.3. For the sound spectrum shown in (a), a normal-hearing cochlea would send a message to the brain that two separate bundles of energy existed in the region around 1000 Hz. One of these bundles may have originated from a talker that the listener was trying to understand, while the other may have originated from some interfering sound. The impaired cochlea, by contrast, may send a message to the brain that there is

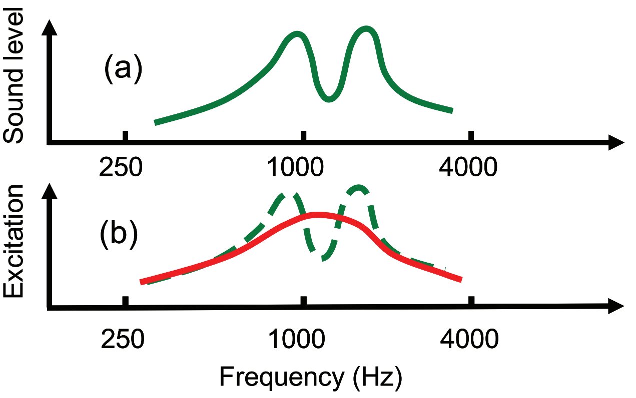

Figure 1.3 (a) Sound spectrum, and (b) possible representation in the auditory system for normal hearing (dotted green line) and sensorineural hearing impairment (solid red line).

just a broad concentration of energy around 1000 Hz. Consequently, the brain has no chance of being able to separate the signal from the noise.

Decreased frequency resolution can adversely affect speech understanding even without the presence of noise. If frequency resolution is sufficiently decreased, relatively intense low-frequency parts of speech (e.g. the first formant in voiced speech sounds) may mask the weaker higher frequency components (e.g. the second and higher formants, and high-frequency frication noise from the vocal tract). This is referred to as ***upward spread of masking***[387, 1146] and is evident in the neural responses of cats with noise-induced hearing losses.[1563] Neural fibers that would normally respond in synchrony with the second formant are instead captured by the waveform of other harmonics, especially by the more intense first formant.[1194] The degree of reduced frequency selectivity, and its impact on speech understanding, increases with the degree of hearing loss. Artificially degraded frequency selectivity appears to have a greater effect when applied to high frequencies than to low frequencies.[1102] An appropriately prescribed hearing aid will minimize the amount of upward (and downward) spread of masking by making sure that there is no frequency region in which speech is much louder than for the remaining regions.

There is a second reason why decreased frequency resolution is a problem. Even normal-hearing people

[b] As a minimum requirement, to be separately processed by someone with normal hearing, the two frequencies have to be further apart than one critical band (which is explained in Section 1.2.1).

have poorer resolution at high intensity levels than at lower levels. Hearing-impaired people, especially those with severe and profound loss, have to listen at high levels if they are to achieve sufficient audibility. Consequently, their difficulty in separating sounds is partly caused by their damaged cochlea and partly caused by their need to listen at elevated levels.[484]

The extent to which inadequate frequency resolution affects speech understanding is still being debated. It is clear that frequency resolution gradually decreases as the amount of hearing loss increases. It seems highly probable that, for mild and some moderate hearing losses, decreased speech intelligibility is mostly caused by decreased audibility (i.e. some parts of speech lying below threshold). For people with severe and profound loss, and for some people with moderate loss, decreased frequency resolution is very likely to also play a significant role.[624] Certainly it is true that speech intelligibility for such people is poorer than can be explained on the basis of decreased audibility alone.[293]

Once speech and noise in the same frequency region, and arriving from the same direction, get mixed together inside the electronics of a hearing aid, there is as yet no way the hearing aid can separate these to significantly enhance intelligibility. All hearing aids can currently do to minimize the problems caused by decreased frequency resolution is to:

- keep noise out of the hearing aid by picking up a signal remotely and transmitting it to the hearing aid (Sections 3.4 to 3.11);
- use a directional microphone to emphasize wanted sounds coming from one direction and/or partially suppress unwanted sounds coming from other directions (Section 2.2.4 and Chapter 7); and
- provide an appropriate variation of gain with frequency so that the low-frequency parts of speech or noise do not mask the high-frequency parts of speech (Sections 10.2 and 10.4) and so that frequency regions dominated by noise are not louder than frequency regions dominated by speech (Section 8.2).

1.1.4 Decreased temporal resolution

Temporal resolution is a very general term that is used to mean different things. First, intense sounds can mask weaker sounds that immediately precede them or immediately follow them. This ***temporal masking*** happens to a greater extent for people with sensorineural hearing impairment than for people with normal hearing,[388, 1970] and adversely affects speech intelligibility.[624] The increased temporal masking is likely caused by the impaired cochlea not being able to increase its sensitivity after the masking sound ceases, as happens in a normal-hearing cochlea.[1372]

Many real-life background noises fluctuate rapidly, and normal-hearing people extract useful snippets of information during the weaker moments of the background noise, referred to as ***listening in the gaps***. Hearing-impaired people partially lose this ability to hear during the gaps in a masking noise, particularly if they are elderly.[177, 492, 540, 791, 1403, 1435] The ability to hear weak sounds during brief gaps in a more intense masker gradually decreases as hearing loss gets worse.[293] Part of the reason for decreased gap-listening ability is that even normal hearers loses some gap-listening ability as ***signal-to-noise ratio*** (***SNR***) increases, and a higher SNR is invariably needed for hearing-impaired people to just understand speech.[119]

Another aspect of temporal resolution is the ability to use the information contained within the cycle-by-cycle timing of the waveform at any point on the basilar membrane. This is referred to as the ***temporal fine structure*** of the waveform, and those who are least able to use it are also least able to understand speech during the gaps in a masking noise.[1077] Decreased ability to use temporal fine structure may be caused by reduced precision in the timing of neural firing. Whatever the physiological cause, degrading the fine structure of the waveform, either by jittering the waveform[1102] or by replacing it with sinusoids in a way that preserves the overall spectral shape,[761] causes speech intelligibility in noise to decrease.

Hearing aids can help a little in compensating for decreased temporal resolution ability. Fast-acting compression, where the gain is rapidly increased during weak sounds and rapidly decreased during intense sounds, will make the weaker sounds more audible in the presence of preceding stronger sounds, and so will make them *slightly* more intelligible.[1240] Unfortunately, it will also make unwanted weak background noises more audible.

1.1.5 Physiological origins of hearing loss

Many abnormalities within the outer ear or middle ear can cause a ***conductive hearing loss***. These abnormalities include: no external ear or ear canal (***atresia***), a perforated or absent ear drum, fixated ossicles in the middle ear, fluid in the middle ear arising from infection, and a disconnection of one ossicle from the next (or a complete absence of ossicles) in the middle ear. All of these mechanisms result in an attenuation of vibration prior to sound entering the cochlea.

Within the cochlea, either the ***inner hair cells*** (***IHC***), ***outer hair cells*** (***OHC***), or both, can cease to function normally over at least some portion of the cochlea and hence over some range of frequencies. If only the OHC cease to function normally, then thresholds are elevated, dynamic range is reduced, and frequency and temporal resolution are both degraded. If only the IHC cease to function normally, then thresholds are again elevated, but frequency resolution remains at or close to normal. The timing of signals within the brainstem may become less precise, due to either a reduced number of functioning IHCs or a reduced number of synapses connecting to each IHC. When IHCs cease to function, then it is common for the spiral ganglion cells to which they connect to progressively die over the next year or two. Either abnormalities within the hair cells, abnormalities in the synapses connected to the hair cells, or complete destruction of individual hair cells can cause both IHC and OHC to have reduced (or absent) functioning.

Alternatively, the hair cells may function poorly because the cochlear "battery" (the stria vascularis) generates insufficient voltage.[1573] Normally, ions are pushed towards the hair cells by the voltage created by the stria vascularis. They are able to flow through the cells when the stereocilia (hairs) are bent by movement of the basilar membrane. The hearing loss resulting from inadequate stria operation is called ***strial sensorineural loss***. Another cause of hearing loss within the cochlea is a change to the physical properties (e.g. stiffness) of structures within the cochlear duct (***cochlear conductive loss***). Any defect that interferes with the conversion of vibrations in the cochlea to nerve signals is called sensory hearing loss.

When everything up to and within the cochlea works normally, but there is a defect in the connection to the auditory nerve or defective transmission along the auditory nerve, the hearing loss is referred to as ***neural***. Partially overlapping with this, when the OHC function normally, but either the IHC, their connection to the auditory nerve, or the auditory nerve itself are defective, the loss is referred to as ***auditory neuropathy spectrum disorder***. This disorder is common in children who are born with a condition or in a manner that requires them to spend time in neo-natal intensive care.[40]

When the IHC within some region of the cochlea completely cease transmitting information to the auditory nerve, that part of the cochlea has come to be referred to as a ***dead region*** (Section 10.3.3), and convenient tests for detecting dead regions are available. [1233, 1235, 1602 936] Unfortunately, there is as yet considerable overlap between the defining characteristics of dead regions, auditory neuropathy spectrum disorder, and neural hearing loss, although the terms are not synonymous.

Sensorineural hearing loss is thought mostly to be caused by deficiencies in IHC and/or OHC function, and so should really be called sensory hearing loss. It is likely that many people have more than one of the underlying causes of cochlear hearing loss.[1573] Consequently, this book will use the simple division of hearing loss into conductive, sensorineural, and auditory neuropathy spectrum disorder when discussing hearing aid requirements.

1.1.6 Deficits in combination

Each of the above aspects of a hearing loss (decreased audibility, dynamic range, frequency and temporal resolution, and the occurrence of dead regions) can cause a reduction in intelligibility. Any combination of these can cause a hearing-impaired person to understand much less than a normal-hearing person in the same situation, even when the hearing-impaired person is wearing a hearing aid. Looked at another way, the hearing-impaired person needs a better SNR than does a normal-hearing person, if they are to understand the same proportion of words in a speech signal.[1434]

A further contribution to SNR deficit comes from ***auditory processing disorders***. These disorders of the brainstem, mid-brain, or auditory cortex can exist independently of any peripheral hearing loss or can be the direct consequence of an impaired cochlea sending deficient signals to the brainstem.[299]

One auditory processing disorder that has been well studied is the ability to separate a target from competing speech on the basis of direction of arrival. The deficit in SNR experienced by hearing-impaired people is much greater when target speech and the competing sounds are spatially separated (i.e. usual real-life conditions) than when they all come from the same direction, or when they are presented through headphones (i.e. usual test conditions in the clinic).[131, 177, 178, 492, 604, 1335, 1400] This deficit, which occurs even when sounds are made comfortably audible by individually prescribed amplification, is referred to as ***spatial processing disorder***, or alternatively as a deficit in ***spatial release from masking***.

The magnitude of this SNR deficit for spatialized sounds, which can be measured clinically with the Listening in Spatialized Noise Sentences (LiSN-S)[241] test, is, on average, proportional to the magnitude of the sensorineural loss.[628, 1142] This spatial deficit is slightly exacerbated by aging, even after the effects of hearing loss have been allowed for.[628, 1288]

The binaural auditory processing system that enables normal-hearing people to focus attention in one direction and suppress sounds from other directions is adversely affected by the cochlear distortions occurring in sensorineural hearing loss.

On average, the SNR required for a given level of speech intelligibility increases as the amount of sensorineural hearing loss increases. Several factors affect the degree of additional SNR needed because of hearing loss. The SNR deficit is greater:

- if the signal competing with a talker fluctuates greatly in amplitude, such as occurs for a single competing speaker;[1403]
- if the hearing-impaired people are significantly older than the normal-hearing people to whom they are compared;[299, 470, 1403]
- if the speech and competing signal(s) come from different directions;[628, 1142, 1400]
- if the competing signal has spectral gaps that cause it to be a more effective masker for people with sensorineural loss (and hence reduced frequency selectivity) than for people with normal hearing.[444]

If a speech-in-noise test administered to a patient uses a non-fluctuating noise with a smooth spectrum, and there is no spatial differentiation of signal or noise, the degree of SNR deficit measured will underestimate the deficit that the patient will experience in real life relative to the normal-hearing people around them.

Across experiments, it appears that every 10-dB increase in four-frequency average hearing loss requires a 1 dB to a 3 dB increase in SNR to keep speech intelligibility in noise constant, when averaged across research participants.[109, 169, 444, 628, 914, 1657, 1924] The largest of these values occurs when the target and competing sounds are spatially separated.[628] It may be that reduced ability to separate speech from noise using spatial cues is one of the major reasons hearing-impaired people still have trouble understanding speech despite wearing well-fitted hearing aids.

Part of the reason for the wide variation in SNR deficit across experiments, and for the even larger variation between individuals within experiments, is the extent to which the hearing-impaired people being assessed received audible speech across the full frequency range of speech. If elevated hearing thresholds prevent them from hearing speech in one part of the frequency range, even after amplification, then the overall SNR must be increased if the remaining part of the frequency range is to provide them with the same total amount of information that a normal-hearing person listening in the same noise would receive. When normal-hearing people are deprived of audibility by adding noise to simulate a hearing loss, their speech intelligibility scores deteriorate to levels similar to those of people with equivalent mild to moderate hearing loss.[483, 1968] Nonetheless, even with appropriate amplification, on average a significant SNR deficit remains for people with sensorineural hearing loss and on average, the greater the loss, the greater the deficit.[169, 293, 628, 1395, 1435]

The situation regarding speech intelligibility is far simpler with conductive losses. These appear to cause a simple attenuation of sound, so that provided the hearing aid can adequately amplify sound, the normal cochlea can resolve sounds entering it just as well as the cochlea of someone with normal hearing. Hearing aids are thus very beneficial for people with conductive loss. As conductive loss increases, the proportion of sound reaching the cochlea by bone conduction also increases (Section 15.1.2). Consequently, the similarity of input to the two cochleae also increases, and the ability of the brain to combine these two signals to selectively attend to sounds coming from one direction decreases. Hearing aids increase the proportion of sound received by air conduction, but with large conductive losses, some mixing within each cochlea of the signals reaching the left and right hearing aids is likely to remain, thus reducing spatial listening abilities compared to normal hearing.

When the cochlea ceases sending signals in one frequency region ("dead regions"), the nerve cells within the auditory cortex that normally receive these signals are likely to break their existing connections and instead respond to adjacent frequencies that are being more effectively sent by the cochlea,[935, 1166] or even to a different modality, such as vision. This is an extreme example of ***neural plasticity*** which in general has profound implications for auditory rehabilitation. The person affected may require many months to fully learn to make use of amplified sound.[63, 1382] In the most extreme case of a person with long-term profound hearing loss receiving a cochlear implant, the ability to use the signals now coming from the cochlea may remain minimal for ever. Neural plasticity is therefore limited.

1.2 Acoustic Measurements

1.2.1 Basic physical measures

The acoustic quantities of frequency, period, wavelength, diffraction, pressure, sound pressure level (SPL), waveform and spectrum must be understood before some parts of this book will make much sense.

Frequency describes how many times per second a sound wave alternates from positive pressure to negative pressure and back to the starting value. Frequency is measured in cycles per second, or, more usually, Hertz (Hz) or kiloHertz (kHz).

Period is the time taken for a repetitive sound wave to complete one cycle. Period is measured in seconds (s) or milliseconds (ms) and is equal to one divided by the frequency.

Phase describes the timing of a sound, or one component of a sound, relative to some other aspect of the sound or relative to another sound. One complete period corresponds to a ***phase shift*** of 360° or 2π radians. Two sounds are ***out of phase*** when their waveforms are proportional to each other but have opposite polarity. This corresponds to a phase shift of 180° at all frequencies.

Wavelength describes the distance a sound wave travels during one period of the wave. It is measured in meters (m) and is equal to the speed of sound (which in air is 345 m/s) divided by the frequency of the sound. Low-frequency sounds therefore have large wavelengths (several meters) and high-frequency sounds have small wavelengths (a few centimeters).

Diffraction describes the way in which a sound wave is altered by an obstacle. When a sound meets an obstacle, like a head, the size of the wavelength compared to the size of the obstacle determines what happens. Sounds with wavelengths much smaller than the obstacle cannot bend around the obstacle and so cause a sound shadow to occur on the side of the obstacle away from the sound source (i.e. the sound is attenuated). Such obstacles will also cause the sound pressure to increase on the side closest to the source. Sounds with wavelengths much larger than an obstacle will flow smoothly, without attenuation, around the obstacle, giving much the same sound pressure at all points around the obstacle.

Pressure describes how much force per unit area a sound wave exerts on anything that gets in its way, such as an eardrum. It is measured in Pascals (Pa), mPa or μPa.

Sound pressure level (SPL) is the number of decibels (dB) by which any sound pressure exceeds the arbitrary, but universally agreed reference sound pressure of 2×10^{-5} Pa (i.e. 20 μPa). It is equal to 20 times the logarithm of the ratio of the actual sound pressure to the reference sound pressure. When pressure doubles, the SPL increases by 6 dB; when pressure increases ten times, the SPL increases by 20 dB.

RMS stands for the root-mean-square value of a signal. It is a way of representing, with a single number, the strength of a fluctuating signal over a certain time. This method reflects the average power of the signal.

Waveform describes how the pressure of a sound wave varies from moment to moment in time. The waveform of a pure tone, for example, is a sinusoid.

Spectrum describes the mixture of pure tones that, when added together, produce a particular complex sound over a specific portion of time. A complete spectrum specifies the amplitude and phase of every pure tone component in the complex sound, but often we are interested in only the amplitude spectrum. When the complex tone is ***periodic*** (i.e. each cycle looks like the preceding cycle), the pure tone components are called ***harmonics***. The frequencies of these harmonics occur at integer multiples of the ***fundamental frequency***, which is the frequency at which the complex wave itself repeats. A ***Fourier transform*** is a mathematical operation that enables the spectrum to be calculated if the waveform is known. Conversely, an inverse Fourier Transform enables the waveform to be calculated if the spectrum is known. A spectrum and a waveform are thus two different ways of describing the same sound.

Octave bands and ***one-third octave bands*** are frequency regions one octave and one-third octave wide respectively. The spectrum of acoustic signals is often analyzed by filtering the signals into adjacent octave or one-third octave bands and measuring the RMS level of the components that fall into each of the bands. An ***octave*** corresponds to a doubling of frequency. The origin of *octave* is that in Western music the eighth note in the scale has a frequency twice that of the first note in the scale.

Critical bands are frequency regions within which it is difficult for the ear to separate sounds of different frequencies.[1560] Sounds spaced apart by more than a critical band are more likely to be separately recognized by the brain, at least by normal-hearing people. A more accurate conceptualization of this grouping of sounds is that as far as the signals in each auditory nerve are concerned, the cochlea processes sounds through ***auditory filters*** centered at each and every place (and hence frequency) in the cochlea. Although these band-pass filters (Section 2.5.1) progressively attenuate sounds as their frequency departs from each filter's centre frequency, the width of each filter can be characterized by its ***equivalent rectangular bandwidth (ERB)***.[c, 1217] For centre frequencies above 1000 Hz, an ERB is approximately 1/6 of an octave wide.[1217] The bandwidth in Hz therefore increases proportional to the centre frequency of the band. As centre frequency decreases below 1000 Hz, the bandwidth (in Hz) decreases, but the relative bandwidth (in octaves) increases. By 100 Hz, the ERB is about 30 Hz or half an octave wide.[843]

Impedance describes how easily a medium (e.g. air) vibrates when a sound pressure is applied to it. In free air, impedance is equal to the ratio of sound pressure to ***particle velocity*** (the velocity at which particles in the medium vibrate back and forth as the sound wave propagates). The impedance of a medium has a constant value that depends only on the physical characteristics (density and elasticity) of that medium. In tubes, impedance is defined differently, and is equal to the ratio of sound pressure to ***volume velocity***. Volume velocity is defined as particle velocity multiplied by the cross sectional area of the tube. Volume velocity can be thought of as the total quantity of sound flowing back and forth through any plane perpendicular to the length of the tube.

1.2.2 Linear amplifiers and gain

The gain of any device relates the amplitude of the signal coming out of the device to the amplitude of the signal going into the device. Gain is calculated as the output amplitude divided by the input amplitude. This applies whether the input and output signals are electrical signals, with their amplitudes measured in Volts, or whether they are acoustical signals measured in Pascals. If an input signal of 20 mPa were amplified to become an output signal of 200 mPa, the gain of the hearing aid would be ten times. Expressing gain in this way best reflects what a ***linear amplifier*** does: it makes everything bigger by multiplying the input signal by a fixed amount. This same amplifier system would multiply an input signal of 1 mPa up to an output signal of 10 mPa.

More commonly, and more conveniently, the input and output amplitudes are expressed as a level in decibels (e.g. dB SPL). Gain is then calculated as the output level minus the input level and is expressed in decibels. In the first example above, the input signal would have a level of 60 dB SPL, the output signal would have a level of 80 dB SPL so the gain would be 20 dB. Over a wide range of input signal levels, this same linear amplifier will always cause the output signal level to be 20 dB greater than the input signal level.

The relationship between input and output SPL for a particular frequency is often shown in an ***input-output (I-O)*** diagram. Figure 1.4 shows the I-O diagram for a hearing aid that is linear for all input levels up to 85 dB SPL. The linear portions of I-O diagrams are straight lines at an angle of 45°, because any increase in input level results in the same increase in output level.

The behavior of a linear amplifier is not affected by how many signals it is amplifying at the same time. If signal A is amplified by, say, 30 dB when it is the only signal present at the input, then it will still be amplified by 30 dB even when several other signals are simultaneously being amplified by the device.

[c] The terms *critical band* and *equivalent rectangular bandwidth of the auditory filter* (or just *ERB*) will be used interchangeably as they represent the same concept. Older and less precise measurements of critical bandwidth indicated wider bandwidths than are now known to apply.

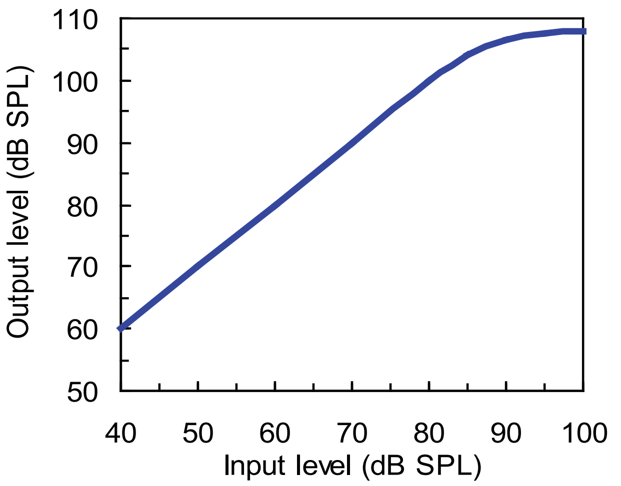

Figure 1.4 Input-output diagram for a hearing aid with 20 dB gain, showing how the output SPL depends on the input SPL, for some particular signal.

The gain of electrical amplifiers often depends on frequency, and the gain of a hearing aid always does. To fully describe the gain of a linear amplifier it is thus necessary to state its gain at every frequency within the frequency range of interest. This is referred to as the ***gain-frequency response*** of the device, and is usually shown graphically. The solid line in Figure 1.5 shows an example of the gain-frequency response of an in-the-ear hearing aid. This is sometimes just called the ***gain curve***.

Sometimes people will say things like *the gain of the hearing aid is 30 dB*. Such a statement is ambiguous and therefore of little value. It may refer to the gain at the frequency at which the gain is the greatest, the gain at some (unspecified) reference frequency, or the gain averaged across some (unspecified) particular frequencies. People will also shorten the term *gain-frequency response* to *frequency response*. This also is ambiguous, but usually what is meant is the manner in which gain varies with frequency, irrespective of what the actual gain is at any frequency. For example, the broken line in Figure 1.5 could be said to have the same frequency response as the solid line, because the two curves have the same shape, even though they have different gains at every frequency. As a practical issue, the two curves shown in Figure 1.5 could be produced from the same hearing aid just by varying its volume control. For a gain-frequency response to convey useful information, it is important that the measurement conditions, especially the position of the volume control, be stated.

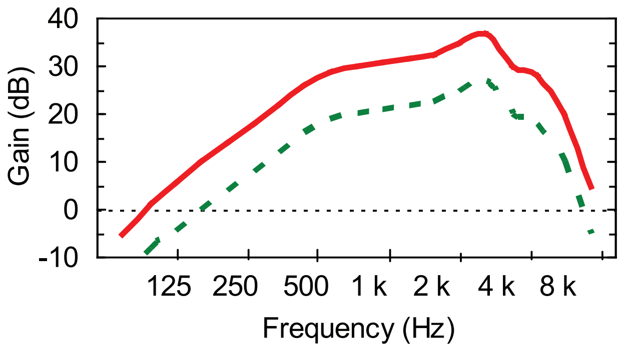

Figure 1.5 Gain-frequency response of an in-the-ear hearing aid at maximum volume control position (solid red line) and reduced volume control position (broken green line).

For non-linear devices, such as those employing compression, the amount of amplification depends on the level, and possibly other characteristics of the input signal. Measurements of gain-frequency response are thus only meaningful if the signal used is specified.

1.2.3 Saturation sound pressure level

All amplifiers become nonlinear when the input or output signals exceed a certain level. This happens because amplifiers are unable to handle signals larger than the voltage of the battery that powers the amplifier. For many reasons it is often desirable to limit the maximum output of the hearing aid to be even less than the limit imposed by the battery voltage and below the limit imposed by the receiver. The highest value of SPL that a hearing aid can produce is called the ***saturation sound pressure level*** (***SSPL***). As with gain, the SSPL varies with frequency, and a useful measure is the SSPL response curve. Figure 1.6 shows the SSPL response curve of one in-the-ear hearing aid.

Terms closely related to SSPL are ***output sound pressure level for a 90 dB SPL input level*** (***OSPL90***) and ***maximum power output*** (***MPO***). The term

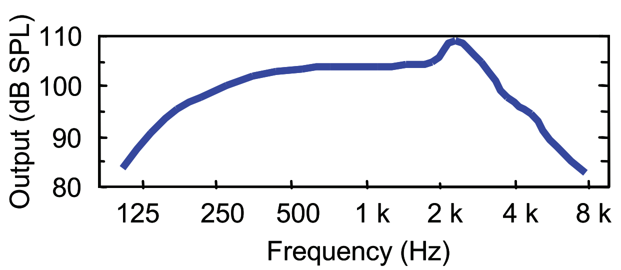

Figure 1.6 Saturated sound pressure level frequency response of an in-the-ear hearing aid.

MPO, although often used, is not an appropriate term, because the quantity being measured is SPL, not power. The term OSPL90 is the most precise term as it states how the maximum output level of the hearing aid is measured: the hearing aid is exposed to a signal of 90 dB SPL. This level is chosen as the standard input level because it is nearly always large enough to saturate the hearing aid (unless the volume control is set to a very low level). Devices that amplify in a nonlinear manner are considered in detail in several later chapters.

1.2.4 Couplers and real ears

The discussion of gain and SSPL has referred to the SPL at the output of the hearing aid without saying where or how the SPL is measured. There are two choices. Hearing aids are meant for ears, so the first important place to measure the output of a hearing aid is in the ear canal of a hearing aid wearer. The only practical way to measure this is with a soft, thin ***probe-tube*** attached to a microphone. Two different types of ***real-ear measurement,*** performed with such a probe-tube, are discussed in some detail in Sections 4.3 and 4.4.

It is also desirable to be able to measure a hearing aid in a standard way that does not require it to be mounted in a person's ear. It is inconvenient to enlist a human assistant every time a hearing aid has to be checked, and the response measured will vary from person to person. Hearing aids can be measured in a standard way by coupling them to a ***coupler***. Couplers are small cavities. The hearing aid connects to one end of the cavity, and the other end of the cavity contains a microphone, which in turn is connected to a sound level meter. The cavity in the most commonly used coupler has a volume of 2 cm^3, so the coupler is called a 2-cc coupler. Couplers, and their more complex, but realistic cousins, ***ear simulators***, are described more fully in Section 4.1.1.

Couplers and ear simulators are indispensable for confirming that hearing aids are operating correctly. Due to individual differences in ear canal geometry and the way the hearing aid is coupled to the ear, the hearing aid's performance should normally be measured in the ear of each individual hearing-impaired person.

1.3 Types of Hearing Aids

A hearing aid is essentially a miniature public-address system. Its essential components are:

- one or more microphones to convert sound into an electrical signal;
- an amplifier to increase the strength of the electrical signal; in the process it will also alter the balance of the sound, usually giving more emphasis to high-frequency sounds and weak sounds than it does to low-frequency sounds and intense sounds;
- a miniature loudspeaker, called a receiver,[d] to turn electricity back into sound;
- a means of coupling the amplified sound into the ear canal; and
- a battery to provide the power needed by the amplifier.

Microphones and receivers are jointly referred to as ***transducers*** because they convert one form of energy into another. The amplifiers in nearly all hearing aids now use ***digital signal processing***, which means that the amplifiers also contain circuits to turn the continuous (i.e. ***analog***) electrical signals into numbers, mathematically manipulate the numbers, and turn the new numbers back into analog acoustic signals at the output of the hearing aid.

Hearing aids can be categorized in many ways. The simplest way to categorize them is by the place where they are worn, which also implies what the maximum size of the hearing aid must be. The largest type of hearing aid is the ***body aid***. These aids are typically about 60 x 40 x 15 mm (very approximately 2 x 2 x 0.5 inches). As implied by their name, they are worn somewhere on the body: in a pocket, in a pouch around the neck or on the belt. They are connected, via a cable containing two or three wires, to a receiver, from which the amplified sound emerges. The receiver usually plugs into an earmold custom-made for the individual's ear canal and concha.

A ***behind-the-ear (BTE)*** hearing aid is considerably smaller. BTEs are also two-piece hearing aids. The microphone and electronics are mounted in the characteristic banana-shaped case, or in some artistic variations of it. In the long-established form of the BTE

[d] Microphones used to be called transmitters, which explains why something that emits sound should be called a receiver.

the receiver is also mounted in the case. The sound from it is conveyed acoustically via a tube to a custom earmold or to a soft dome that retains the open end of the tube within the ear canal.

A more recent variation of the BTE is the ***receiver-in-the-ear canal*** (***RITE***), in which the receiver is located within the ear canal rather than in the BTE case, and an electrical cable rather than an acoustic tube runs from the electronics to the ear canal. RITE BTEs are also known as ***RIC, RITC*** and ***CRT*** (canal receiver technology). The two types of BTE are sometimes referred to by their manner of connection between the case and the piece in the ear canal: traditional BTEs are known as ***standard-tube*** or ***thin-tube*** BTEs, and RITE BTEs are known as ***thin-wire*** BTEs. The traditional BTE style (with any tube diameter) will be referred to in this book as a ***receiver-in-the-aid*** (***RITA***) BTE to contrast with the RITE terminology.

Figure 1.7 shows standard-tube, thin-tube, and thin-wire BTEs attached to ear canal fittings. As is customary, the standard-tube BTEs are attached to custom molds, whereas the thin-tube and thin-wire BTEs are attached to flexible, modular, standard-sized canal fittings. The tubes between the BTE case and the canal fitting are available in a small range of standard lengths, and the canal fittings are available in a small range of standard diameters.

The next type is the ***in-the-ear*** (***ITE***) hearing aid. These vary in size from full concha styles that, as their name implies, fill the entire concha as well as about half the length of the ear canal. A smaller variation of the ITE hearing aid is the ***half-concha*** or ***half-shell*** ITE, which fills only the lower portion of the concha (the cavum) up to the crus-helias. Another variation fills only the upper portion of the concha (the ***cymba*** ITE) and connects to the ear canal with RITE technology. A further variation is the ***low profile*** ITE, which does not extend outwards from the ear canal sufficiently to fill the concha. The various features of the ear are defined in Figure 5.2.

When an ITE hearing aid occupies a sufficiently small portion of the cavum concha and its outer face is parallel to the ear canal opening, it is referred to as an ***in-the-canal*** (***ITC***) hearing aid. (One would expect from the name that an ITC hearing aid would fit entirely within the ear canal, but the name is more a reflection of marketing-inspired optimism than an accurate description of where the hearing aid is located.)

Hearing aids that *do* fit entirely within the ear canal are known as ***completely-in-the-canal*** (***CIC***) hearing aids. These hearing aids use components small enough that none of the hearing aid need protrude into the concha. Removing these hearing aids from the ear can be difficult, so often a small handle, similar to

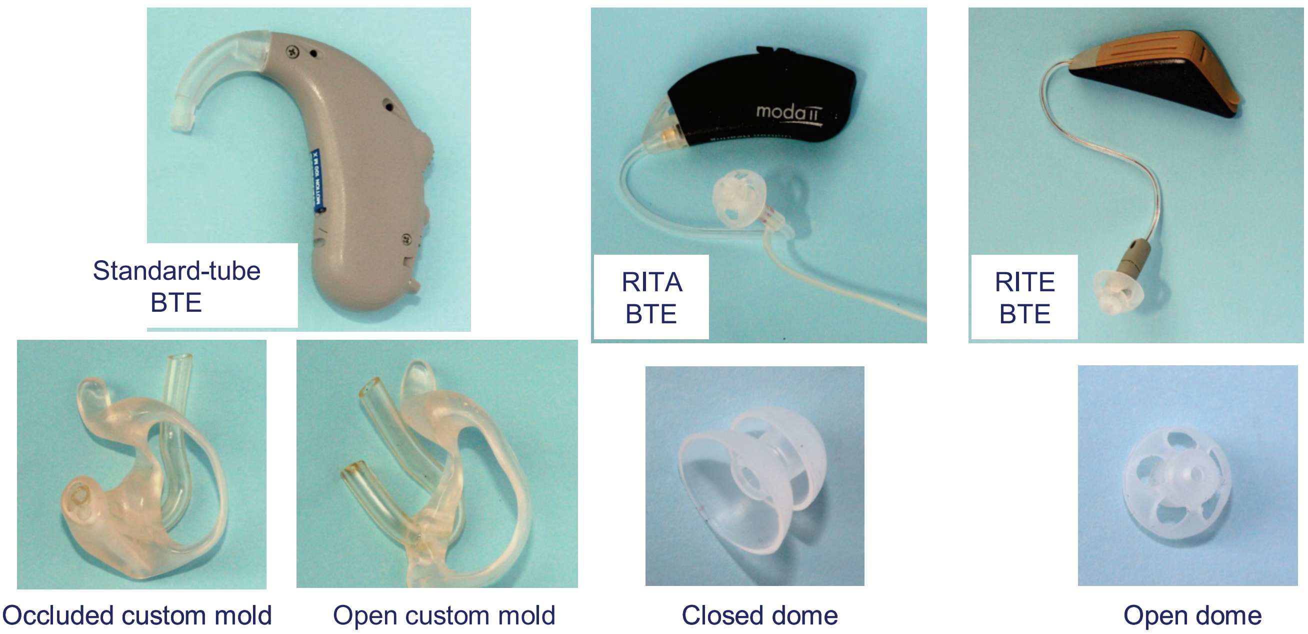

Figure 1.7 BTE hearing aids and ear fittings. The standard-tube BTE in the top left can be attached to an open or occluded custom mold. The RITA BTE and the RITE BTE can be attached to custom molds or to pre-formed canal fittings that can be either open or occluded.

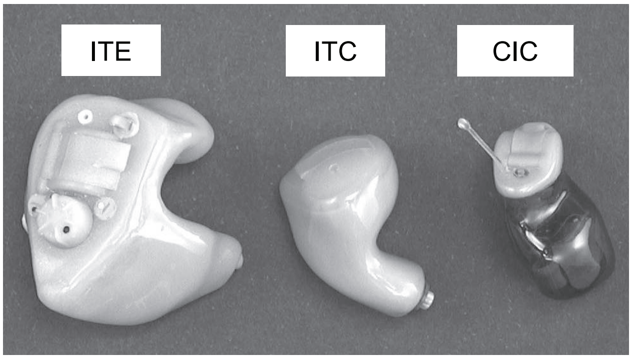

Figure 1.8 An ITE, an ITC and a CIC hearing aid.

nylon fishing line with a small knob on the end, is attached to the hearing aid and this does extend into the concha. When the medial end of a CIC hearing aid is within a few millimeters of the eardrum, the CIC is referred to as a ***peri-tympanic CIC***.

Figure 1.8 shows ITE, ITC, and CIC hearing aids. The typical locations of the major components within a standard-tube BTE and ITC hearing aid are shown in Figure 1.9. Component positioning is similar for thin-tube BTEs except that they do not have an earhook. Thin-wire BTEs also do not have an earhook, and the receiver is not in the case. Many thin-tube and thin-wire BTEs are made as small as possible and do not have volume controls or switches, other than the battery compartment door.

Another type of hearing aid is the ***spectacle*** or ***eyeglass*** aid. As the name suggests, these are a combination of spectacles and one or two hearing aids. There are actually two types of spectacle aids. In the first type, the side frame of the spectacles (the ***bow***) contains all the hearing aid components. These were the first type produced and were bulky in appearance. In current models, the part of the bow that fits behind the ear on a conventional pair of spectacles is sawn off, and a short adapter is glued on in its place, as shown in Figure 1.10. The spectacle hearing aid (essentially a BTE) attaches to this adapter and a tube leads from the hearing aid receiver to the ear. These are less conspicuous, and the frontal appearance is little different from the appearance of the spectacles alone. This is particularly true when the hearing aid couples into the ear without an earmold, using only tubing.[1865]

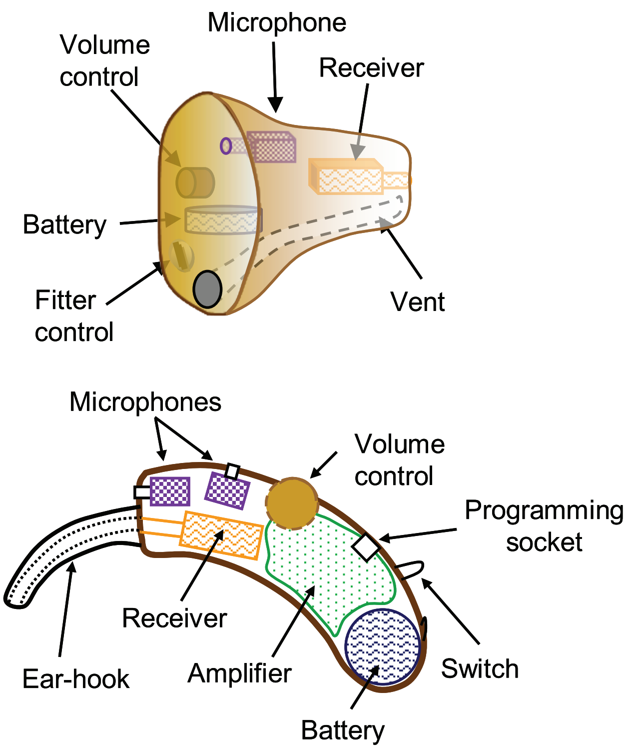

Figure 1.9 The typical location of components in an ITC and a BTE hearing aid.

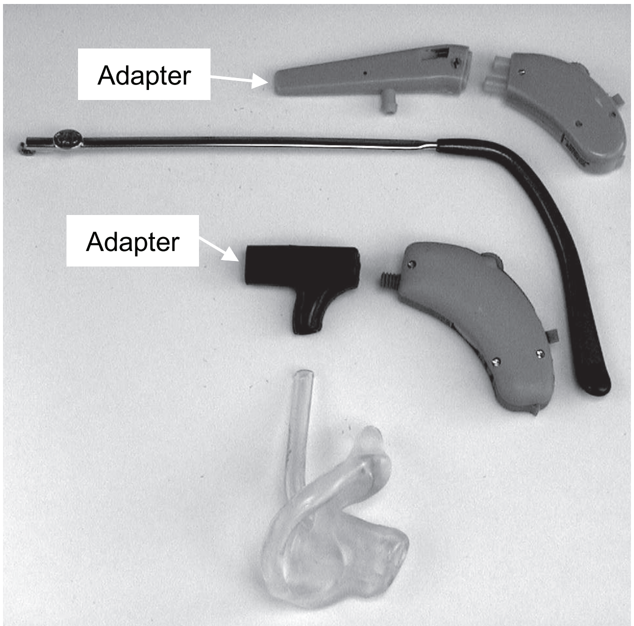

Figure 1.10 Spectacle adapter system showing two different adapters and BTE hearing aids, an earmold, and a spectacle bow. The bow would be cut at the white line and the left half inserted in the adapter.

Decreasing component size (and hence case size), the use of thin (near invisible) tubing, the avoidance of custom manufacture, and higher reliability, have led to BTEs being by far the most common style in most, if not all, markets. Most of the rest fitted are ITE, ITC or CIC hearing aids.

1.4 Historical Perspective

The biggest change to hearing aids over the last century is that they have become smaller. The quest to make them smaller and less conspicuous has been a constant driving force behind technological progress. Sometimes performance has had to be sacrificed to reduce size, but sometimes performance has increased because of the size reductions. Current hearing aids have better fidelity (wide bandwidth and low distortion), greater adjustment flexibility, and greater adaptability to changing listening situations than has been possible at any time in the past.

The following brief historical review of hearing aid technology is heavily based on an excellent and authoritative chapter by Sam Lybarger (1988), who created many of the innovations in hearing aids over many decades. For those who would like to read a more detailed history, please see Lybarger (1988) and Berger (1984). Fitting procedures for hearing aids are not described here because they are covered in Section 10.1.

The history of hearing aids may be divided into six eras: acoustic, carbon, vacuum tube, transistor, digital and wireless. The new technology added within

Figure 1.11 Three acoustic-era instruments:
(a) the Auricle,
(b) the horn, and
(c) the speaking tube.

each era has enabled significant improvements in performance, cosmetic appearance or both. Most of the technological features mentioned in this section are described in later chapters, and may have limited meaning to some readers at this stage.

1.4.1 The acoustic era

The acoustic era began the first time someone cupped a hand (or a possibly a paw) behind an ear. This produces 5 to 10 dB of gain at mid and high frequencies by collecting sound from an area larger than the ear can by itself.[409, 1865] It also shields the ear from sounds coming from the rear, thus working as a very effective noise reduction system, at least for mid- and high-frequency sounds.

A more effective acoustic aid is formed by anything with a shape like a ***trumpet***, ***horn*****,** or ***funnel***. Illustrations of horns appeared in 1673 and in 1650.[114, 789] The principle is to have a large open end to collect as much sound as possible. This energy is transferred to the ear by having a gradual reduction in area along the length of the trumpet or funnel. If the area decreases too quickly, most of the sound just reflects back out again instead of travelling into the ear. Ear trumpets therefore have to be both wide and long to be effective.

The quest to make hearing aids smaller has been around a long time. Lybarger (1988) reports that the idea of coiling the trumpet to make it smaller dates at least as far back as 1692. The desire to conceal the hearing aid also has a long history. Ear trumpets have been "hidden" inside top hats, armchairs, fans and beards.[636]

If the open end of an acoustic hearing aid is moved closer to a talker, it picks up more intense sound as well as a greater area of sound. The ***speaking tube***, as shown in Figure 1.11, is designed to do this and consists of a horn-shaped open end attached to a long tube that terminates in a narrow earpiece. If the talker speaks directly into the horn end, the signal-to-noise ratio at the input to the device is much better than that which the listener would receive naturally. As well as amplifying, the device thus improves signal-to-noise ratio, and by an amount not possible with even the most sophisticated one-piece hearing aids now available.

Given that people, even those with normal hearing, still cup their hands behind their ears in adverse listening situations, one cannot say that the acoustic era ever ended.

1.4.2 The carbon era

A carbon hearing aid, in its simplest form, consisted of a ***carbon microphone***, a ***battery*** of 3 to 6 Volts and a ***magnetic receiver***, connected in a simple series electrical circuit. The carbon microphone contained carbon dust, granules, or spherical globules.[e] When sound hit the microphone diaphragm, movement of the diaphragm pushed the bits of carbon closer together, or pulled them further apart, thus changing the electrical resistance of the microphone. This fluctuating resistance caused the electrical current to fluctuate in a similar way, and when this current passed through a coil inside the receiver, it created a fluctuating magnetic field inside the receiver. This fluctuating magnetic field pushed and pulled against a permanent magnet, thus making the receiver diaphragm move in and out, in synchrony with the original sound hitting the microphone. The sound level out of the receiver (when coupled into a small cavity) was 20 to 30 dB greater than the input to the microphone.[1096]

To achieve greater gain, a ***carbon amplifier*** was invented. If one microphone and receiver pair could increase the sound level, it was reasoned that a second pair (but with the two pairs sharing a single diaphragm) could increase it more. The carbon amplifier thus consisted of a coil, which vibrated a diaphragm, which moved some carbon granules or globules to produce a bigger fluctuating electrical current.

The first carbon hearing aid, a large table model called the Akoulallion, appeared in 1899,[114, 610] and the first wearable model (variously called the Akouphone and the Acousticon[836]) appeared shortly after in 1902. The microphone and receiver, which is the entire hearing aid other than the (huge) battery, are shown in Figure 1.12. Carbon hearing aids continued to be used through to the 1940s, but were satisfactory only for people with mild or moderate losses.

During the carbon era, the idea of amplifying different frequencies by different amounts (to suit the hearing loss) emerged. This was achieved by selecting

[e] The replacement of dust by granules and then by globules in 1901 is a good example of the technological refinements that inevitably follow each major change of technology.

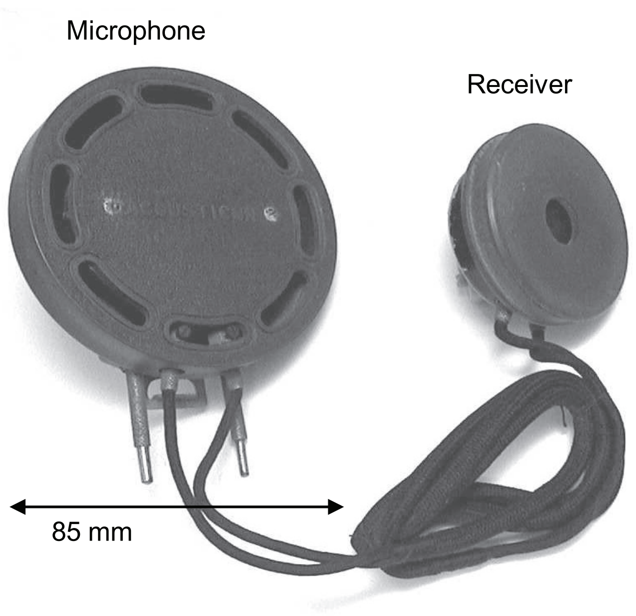

Figure 1.12 A carbon aid (The Acousticon) without its battery.

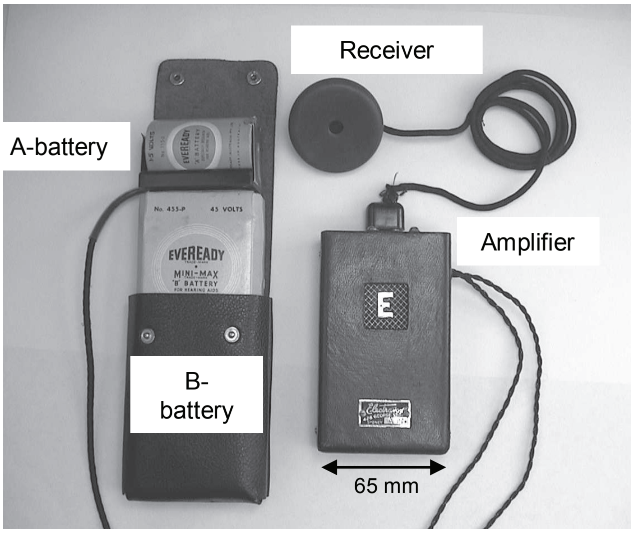

Figure 1.13 A relatively late vacuum tube hearing aid, with its two separate batteries.

different combinations of microphones, receivers, and amplifiers. Couplers, initially with a volume of 0.5 cc, first emerged during the carbon era, as did high-quality condenser measurement microphones.

Assistive listening devices, which are hearing aids that are not worn entirely on the hearing-impaired person, also emerged during the carbon era. Johnston (1997) recalls seeing a microphone in a church pulpit wired to several hand-held receivers in selected pews around 1916, but they were probably in use ten years before this.[1321]

1.4.3 The vacuum tube era

The vacuum tube electronic amplifier was invented in 1907 and applied to hearing aids in 1920.[1096] The vacuum tube allowed a small voltage, which came from the microphone, to control the fluctuations in a large current. By combining several vacuum tubes in succession, very powerful amplifiers (with 70 dB gain, able to output sounds up to 130 dB SPL) could be made, thus increasing the range of hearing losses that could be helped. The increasing sophistication of electronics also allowed the gain-frequency response shape to be better controlled than for carbon aids.

The biggest problem with vacuum tube hearing aids was their total size. The size of the vacuum tubes themselves reduced enormously over time, driven by military requirements, but two batteries were needed to make them work. A low voltage *A* battery was needed to heat the filament of the tubes, and a high voltage *B* battery was needed to power the amplifier circuits. Vacuum tube hearing aids became practical during the 1930s, but until 1944 their batteries were so large that the batteries had to be housed separately from the microphone and amplifier (see Figure 1.13). In 1944, vacuum tube and battery technology had advanced sufficiently to make possible a one-piece hearing aid. Batteries, microphone, and amplifier were combined into a single body-worn package, which connected to an ear-level receiver via a cable. There were further creative attempts to conceal the hearing aid during the vacuum tube era. This included enclosing the electronics, except for the transducers, inside a large pen-shaped case (the Penphone).[611] Microphones were concealed inside broaches and wristwatches, and receiver cords were enclosed within strings of pearls.[611]

Earmold venting, magnetic microphones, piezoelectric microphones and compression amplification were also devised during the vacuum tube era.[1096] Piezoelectric substances have a crystal structure that generates a voltage when twisted or bent. In a microphone, the bending happens because a diaphragm is connected to one corner or end of the piezoelectric crystal. The early origin of compression amplification is surprising; compression seems to have been largely forgotten until the 1980s, but then became the dominant type of advanced amplification in the late 1990s.

1.4.4 The transistor and integrated circuit era

The transistor became commercially available in 1952.[1639] So dramatic was the reduction in battery power required, that all new hearing aids used transistors rather than valves by 1953.[610] The reduction in battery size, and the small size of transistors relative to valves, meant that from 1954 all the component parts of the hearing aid could be moved up to the head. Head mounting had several advantages: clothing did not create noise as it rubbed the microphone, the body did not have such adverse effects on the tonal balance of sound coming from different directions, cables were no longer required, and true binaural hearing aids were possible.

First amongst the head-mounted aids were barrette hearing aids and spectacle or eyeglass aids. The barrettes, which had an external receiver like a body aid, came in a variety of shapes and were worn on or under the hair (or on the body on ties, lapels, or collars).[f] Several were made to resemble jewelry.[1639] Spectacle aids had all the hearing aid components built into the temple pieces (the bow) of spectacles. With a rapid and continual reduction in the size of all the components, they could soon all be moved behind the ear, either as part of the spectacle bow, as a self-contained curved package that attached to a sawn-off standard spectacle bow, or finally as a stand-alone BTE hearing aid. During the following ten years, the BTE took over from the eyeglass aid as the dominant style, and remained so until the mid-1980s in the USA, and until the 1990s in much of Europe.

With further decreases in the size of components, ITE aids started to appear in the mid and late 1950s.[1639] The first ITEs were so large by today's standards that Lybarger (1988) referred to them as *out of the ear* hearing aids.

Two big leaps in the performance and size of components occurred during the 1960s. First, in 1964, the integrated circuit (IC) was applied to hearing aids. This meant that multiple transistors and resistors could be combined into a single component that was similar in size to any one of the individual transistors that it replaced. Second, in 1968, a piezoelectric microphone was combined with a relatively new type of transistor (the ***field effect transistor***, or ***FET***) inside a small metal can. For the first time, a small rugged microphone with a reasonably smooth, reasonably wide frequency response could be used in hearing aids.[916] A few years later, directional microphones emerged, using the same technology.

Microphone technology further improved in 1971 when the electret/FET microphone (described in Section 2.2) was developed.[917] These brought about even better responses and even smaller sizes. During the transistor era, receiver volume decreased from 1800 mm^3 to 39 mm^3 (Knowles model FS), whereas microphone volume decreased from 5000 mm^3 to 23 mm^3 (Knowles model TM). Most of the shrinkage in receiver volume occurred prior to 1970, so perhaps there may not be huge reductions in receiver size in the future, although further reduction in microphone volume is likely.[504]

By the early 1980s, ITE aids had become small enough for most of the components to fit within the ear canal portion of the aid, thus creating the ITC hearing aid.[653] With further improvements in battery chemistry, amplifier efficiency and transducer size, the entire hearing aid could finally be located inside the ear canal by the early 1990s. The CIC had arrived, and at last the hearing aid was invisible! This placement of the hearing aid also carried some acoustic advantages, as the useful sound-collecting and shielding properties of the pinna could be used when wearing a hearing aid. Also, wind noise was greatly decreased. In the United States, ITE/ITC/CIC hearing aids took over from BTEs as the dominant hearing aid style in the mid 1980s.

Some of the advances during the transistor era include:

- zinc-air batteries, that allowed a halving of battery volume for the same electrical capacity (Section 2.12);
- improved filtering, that led to more flexible response-shaping and multi-channel processing of sound (Section 2.5);
- miniature potentiometers, that allowed the clinician to adjust the amplification characteristics of even very small hearing aids;
- wireless transmission hearing aids, in which the hearing aid connects to or contains a wireless receiver that is tuned to a transmitter worn by a talker some distance away (Section 3.4);

f Barettes were larger versions of the very modern RITE BTEs "invented" early this century.

- class D amplifiers, that decreased the battery drain required to achieve a given output level with minimal distortion;
- improved understanding of the acoustics of earmolds and ear shells, that allowed more appropriate gain-frequency responses to be achieved,[909] and occlusion and feedback problems to be decreased, but not solved (see Chapter 5); and
- use of two microphones within a hearing aid, so that the user can select directional or omni-directional performance as needed (Section 2.2.4).

A further very significant advance, which could arguably be placed in the next era, was the application in 1986 of digital control circuits and digital memories to hearing aids. These circuits replaced potentiometers, and because they occupied little space inside the hearing aid, many "controls" could be included in a hearing aid. These circuits thus enabled the amplification characteristics of hearing aids to be adjusted by the clinician with greatly increased flexibility and precision. A by-product of digital control circuits was that the user could also conveniently change the hearing aid's characteristics (usually with a remote control), which made multi-memory hearing aids practical, even in ITC or CIC hearing aids.

1.4.5 The digital era

As was just mentioned, digital electronics first met hearing aids when the digital circuits acted as the controls for an otherwise conventional hearing aid. The real revolution came when the sound waveform itself was converted to a series of numbers and manipulated using digital circuits.

Research into digital processing began in the 1960s within Bell Laboratories.[1045] However, because of the slow speed of computers, the necessary calculations could not be performed quickly enough for the signal to come out of the laboratory hearing aid as rapidly as a signal was put into it! It was not until the late 1970s that computers were fast enough for the output to keep up with (but be delayed slightly behind) the input, and it was not until the 1980s that power consumption and size were decreased sufficiently to make a wearable hearing aid.

Because the first aid was a body aid and did essentially the same things to sound as ear-level analog hearing aids, it was not a commercial success and quickly ceased to be available. Finally, in 1996, fully digital BTE, ITE and ITC hearing aids became commercially available, although several years previous to that, an analog hearing aid with a digital feedback reduction system was available.[502] Some excellent reviews of the development (and future!) of digital hearing aids have been written by Levitt (1987, 1997).

Advantages already seen for digital technology include:

- further increases in the flexibility and precision with which response shaping and compression characteristics can be controlled;
- intelligent automatic manipulation of the gain and frequency response of the hearing aid, depending on how much signal and noise the hearing aid estimates is arriving at the aid in each frequency region;
- intelligent manipulation of the way gain varies for sounds coming from different directions (i.e. directivity), so that noise is minimized;
- increased gain without feedback oscillations occurring;
- reduction in size and power required from the battery, relative to an analog aid that manipulates sound in the same way;
- lowering in frequency of high-frequency sounds carrying crucial speech information, so that they can be perceived in a frequency range where the aid user has better hearing;
- automatic connection to a telephone held to the ear; and
- hearing aids that learn, and then automatically apply, the amplification preferences of their wearers in different environments.

It is extremely likely that digital signal processing will confer many more advantages in the near future. These will probably include further improvement to feedback control, and better methods for reducing the effects of background noise, at least in certain circumstances. The impact of digital processing will be discussed in many places throughout this book. Although hearing aid usage rates and satisfaction did not improve significantly from the late 1980s to the late 1990s,[732, 946, 1390] surveys performed since 2000 are showing improvement in usage,[834, 1869] satisfaction[953, 1869] and listening comfort.[834]

1.4.6 The wireless era

Wireless electromagnetic transmission makes possible the transmission of signals without the degradation caused by noise and reverberation that inevitably accompanies propagation of sound waves. Although hearing aids have been able to connect via cable to wireless receivers for many years, increasingly the wireless receivers are being built into, or snapped onto, head-worn hearing aids. There are four broad applications:

- remote reception: to receive signals sent across a room from a microphone and transmitter worn by the talker;
- coordinated control of bilateral hearing aids: to manually or automatically adjust amplification in the left and right hearing aids in a synchronized manner;
- connectivity to communication devices: to receive audio signals from devices such as mobile phones, computers, personal stereo players, or satellite navigation systems;
- binaural array hearing aids: to provide a full audio connection of the left and right hearing aids, thus enabling super-directional hearing that improves speech intelligibility in noisy places.

The first and last of these options enable an increase in intelligibility in noise that far outstrips anything achievable without wireless transmission. They offer the possibility that in noisy places, people with hearing loss will hear even better than those with normal hearing. If so, hearing aids will no longer signal that the wearer has a disability, but rather signal that the wearer has super hearing. This should dramatically increase both the take-up of hearing aids by people with hearing loss, and the benefit and satisfaction experienced by those who wear hearing aids.

CHAPTER 2

HEARING AID COMPONENTS

Synopsis

Hearing aids are best understood as a collection of functional building blocks. The manner in which a signal passes through these blocks in any particular hearing aid is indicated in a block diagram. The first block encountered by an acoustic signal is a microphone, which converts sound to electricity. Modern miniature electret microphones provide a very high sound quality, with only very minor imperfections associated with internal noise and sensitivity to vibration. Directional microphones, which have two entry ports, are more sensitive to frontal sound than to sound arriving from other directions. These enable hearing aids to improve the signal-to-noise ratio by several decibels (depending on acoustic conditions) relative to omni-directional microphones, and hence can improve the intelligibility of speech in noise. Dual-microphone hearing aids can be switched automatically or by the user to be either directional, or omni-directional, as required in different listening situations.

The small signals produced by microphones are made more powerful by the hearing aid amplifier. All amplifiers will distort the signal, by peak clipping it, if they attempt to amplify the signal to too high a level. Excessive distortion decreases the quality and intelligibility of sounds. To avoid distortion, and to decrease the dynamic range of sound, compression amplifiers are used in most hearing aids. These amplifiers decrease their gain as the level of the signal put into them increases, in much the same way that a person will turn down a volume control when the level becomes too high.

Amplifiers can represent sound in an analog or a digital manner. The signals within analog amplifiers have waveforms that mimic the acoustic waveforms they represent. Digital systems represent signals as a string of numbers. Performing arithmetic on the string of numbers alters the size and nature of the signals these numbers represent. Fully digital circuits may be constructed so that they process sounds in ways specific to each device, or may be able to perform any arithmetic operation, in which case the type of processing they do depends on the software that is loaded into them.

Filtering a signal is a common way in which hearing aids alter sound. Filters can be used to change the relative amplitude of the low-, mid- and high-frequency components in a signal. When the filters are made with variable, controllable characteristics, they function as tone controls operated by the user or the clinician. Filters can also be used to break the signal into different frequency ranges, so that different types of amplification can be used in each range, as required by the hearing loss of the hearing-impaired person.

Receivers are miniature headphones that use electromagnetism to convert the amplified, modified electrical signals back into sound. Their frequency response is characterized by multiple peaks and troughs, which are partly caused by resonances within the receivers, and partly caused by acoustic resonances within the tubing that connects a receiver to the ear canal. Inserting an acoustic resistor, called a damper, inside the receiver or tubing will smooth these peaks and troughs. A damper absorbs energy at the frequencies corresponding to the peaks, and this improves sound quality and listening comfort.

There are several other ways to put signals into hearing aids. A telecoil senses magnetic signals and converts them to a voltage. A radio receiver senses electromagnetic waves and converts them to a voltage. A direct audio input connector enables an electrical audio signal to be plugged straight into the hearing aid.

Users operate hearing aids via electromechanical switches on the case of the hearing aid, or via a remote control. The hearing aid performs all its functions by taking electrical power from a battery. These batteries come in a range of physical sizes and capacities, depending on the power needed by each hearing aid, and the space available.

This chapter will describe the bits and pieces that make up a modern hearing aid. These pieces comprise the transducers that convert sound to and from electricity, and the things that alter sound while it is represented in electrical form. These electronic parts will be considered as functional boxes, rather than as electrical circuits or digital mathematical algorithms. Combinations of these functional boxes are represented by block diagrams. As we will see, the function of these boxes can be discussed irrespective of whether they are implemented as analog or digital circuits. Modern hearing aids have digital amplifiers, which may be supplemented by analog circuits at the input and output of the hearing aid.

2.1 Block Diagrams

Two decades ago, the operation of most hearing aids, for any setting of an aid's controls, could be understood just by looking at a few graphs of gain versus frequency and maximum output level versus frequency. Hearing aids are now more complicated, and the only way to understand how a more complex hearing aid changes a signal is to understand its ***block diagram***.

A block diagram shows what operations a device carries out on signals within the hearing aid, and in what order each of these operations is carried out. Block diagrams also usually show the location of fitter and user controls within the processing chain. This helps the clinician understand what effect varying a control will have, and, just as importantly, what effects it will not have.

Figure 2.1 shows the symbols used in this book for each of a number of blocks. Most of these blocks will be more fully described later in this chapter. Some of the blocks have synonyms. The ***AC/DC converter***, used in conjunction with compression amplifiers, is sometimes called a ***level detector***, an ***averaging circuit***, or an ***envelope detector***. The ***adder*** is sometimes called a ***summer***.

An arrow drawn diagonally through a block indicates that some characteristic of the block can be altered, usually by a fitter or user control, but sometimes by the output of another block. Figure 2.1 shows an example for a variable gain amplifier, but any of the blocks except for the microphone or the receiver can be made variable in this manner. Most of the blocks have an input (or several inputs) and an output.

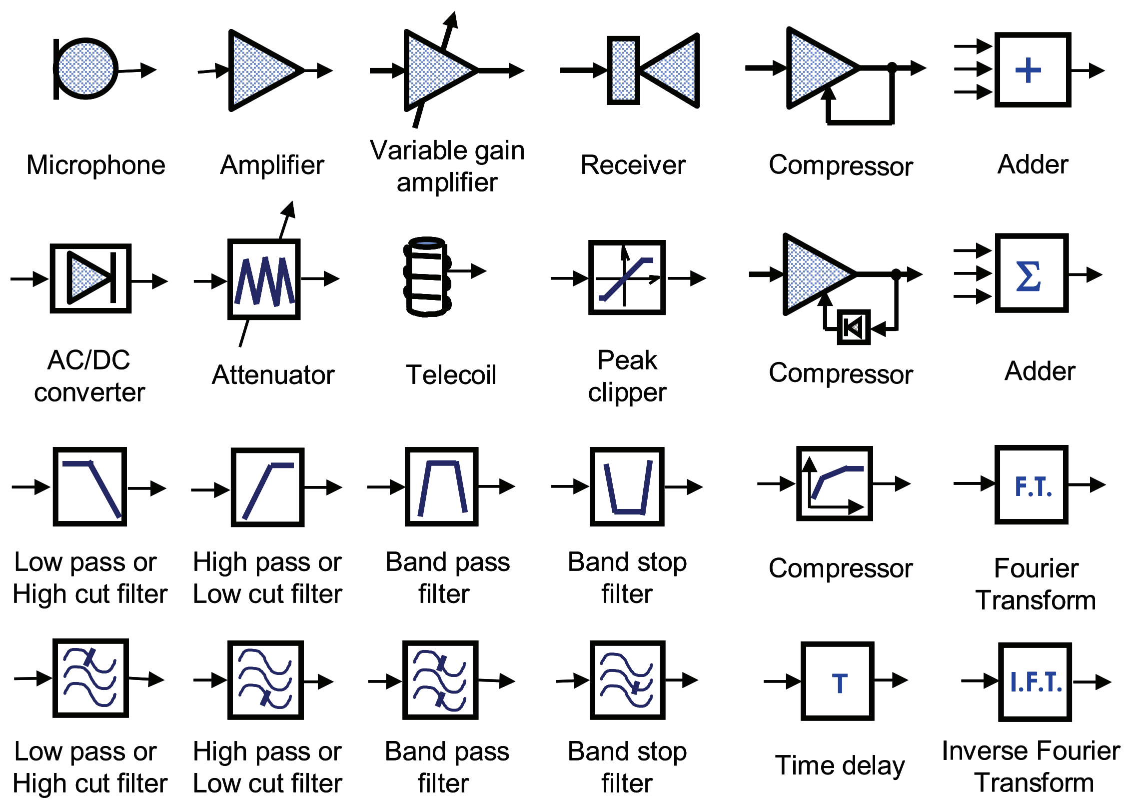

Figure 2.1 Symbols used in block diagrams.

Unfortunately, there is no absolute standardization for block diagram symbols. The symbols and alternative symbols shown in Figure 2.1 are commonly used, but further alternatives exist. The origins of some symbols are easy to see. The filters and the compression amplifier shown in the third row simply show the graph that results when the electroacoustic performance of the block is measured in the most meaningful way. This is a gain-frequency response in the case of a filter and an input-output diagram in the case of a compression amplifier.

In more advanced hearing aids, the signal may be altered in ways that would be too complex to represent as a combination of simple blocks. In this case, the appropriate symbol is just a box with a short description of the process written inside the box.

Some conventions and rules govern how block diagrams are drawn. Arrows on the lines connecting blocks show that a signal is being passed *from* one block *to* another. A convention is that signals mostly flow from left to right, although it is often necessary to make exceptions in complex diagrams, and the arrows on the connecting lines make this clear. One rule is that when the output of a device is fed simultaneously to the input of several other blocks, the entire output signal goes to each of the blocks to which it is connected. Another rule is that an input cannot be driven by more than one output. Connecting two outputs to one input would produce a logical conflict as both devices try to tell another device what its input signal should be. Where possible, block diagrams are drawn so that there are no crossing lines connecting the various blocks, although sometimes this cannot be avoided. Lines that cross and connect to each other usually have the connection indicated by a dot at the point of intersection. Figure 2.2 shows a block diagram of a three-channel hearing aid with a peak clipper in the high-frequency channel and a compressor in the low- and mid-frequency channels.

Let us consider the following examples of what can be learned from a block diagram like the one shown in Figure 2.2. You may like to re-read this paragraph after learning more about the individual blocks, and particularly about the effects of compressors, later in this chapter and in Chapter 6. The microphone converts input signals to electricity and these electrical signals are amplified by a compression amplifier of some type. The resulting signal is split into its low-frequency, mid-frequency and high-frequency components. The low-frequency band of signal is attenuated by a selectable amount and amplified by a compression amplifier. The mid-frequency band is amplified by a selectable amount and further amplified by a compression amplifier. The high-frequency band is amplified by a selectable amount, and then peak clipped. The three parts of the modified signal are recombined and amplified by a user- or fitter-controlled amount, before being delivered to the receiver. The effects of the controls on the operation of the aid can also be deduced, but the preceding description illustrates the information that can be easily read from a block diagram.

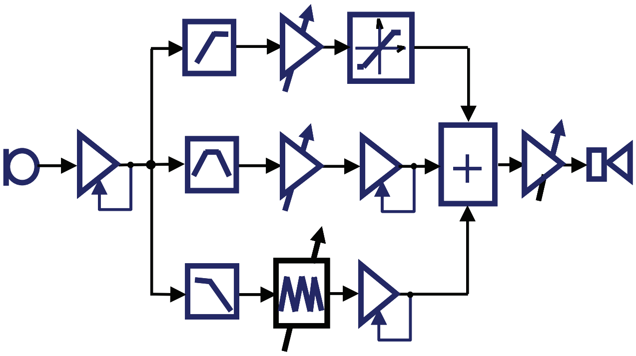

Figure 2.2 A three channel compression hearing aid.

2.2 Microphones

The function of the microphone is to convert sound into electricity. Because it changes energy from one form to another it is known as a ***transducer***. For a perfect microphone (and microphones *are* close to perfect), the waveform of the electrical signal coming out of the microphone is identical to the waveform of the acoustical signal going into the microphone. Microphones act in a linear fashion, so every time the pressure of the input signal doubles, for instance, the output voltage also doubles, until the output reaches the highest voltage that the microphone can deliver. The ratio of the size of the output voltage to the size of the input sound pressure is known as the ***sensitivity*** of the microphone. Typical hearing aid microphones have a sensitivity of about 16 mV per Pascal, which means that sounds of 70 dB SPL produce a voltage of around 1mV.

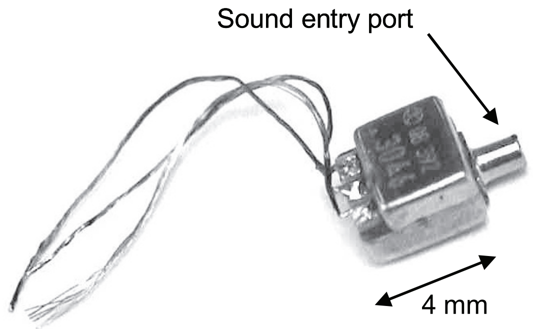

Figure 2.3 An electret microphone.

2.2.1 Principle of operation

Microphones can be made using several fundamentally different types of technology, but since the 1980s hearing aids have used only one type of microphone: the electret microphone.[917, 1606] Figure 2.3 shows the external appearance of an electret microphone, although some electret microphones are cylindrical in shape. Figure 2.4 shows a cross-section to illustrate the operating principle. Sound waves enter through the inlet port and reach one side of a very thin, very flexible plate with a metallized surface, called the ***diaphragm***. Pressure fluctuations within the sound wave cause the diaphragm to move up and down (by an extremely small amount, invisible to the naked eye). A small air space separates the diaphragm from a rigid metal plate, called the ***back-plate***. Coated onto the back-plate is some thin teflon material called an ***electret***. The diaphragm is held away from the back-plate by some bumps in the back-plate. The back-plate has holes in it to allow movement of air through it.

The electret material gets its name from the fact that it has a permanent electric charge comprising an excess of electrons on one side of it, and a shortage of electrons on the other. These electrical charges attract opposite electrical charges onto the diaphragm and the back-plate. When sound pressure forces the diaphragm towards or away from the electret, the changing distance between the diaphragm and the electret changes the electrical force between the opposing charges, which is another way of saying that the voltage between the back-plate and diaphragm varies. This, of course, is the point of the device, because the vibrating diaphragm has turned a sound wave into an electrical voltage. An amplifier is built into the same container as the rest of the microphone. Its job is to allow this voltage to deliver a larger current to the main hearing aid amplifier than would otherwise be possible. The microphone amplifier is sometimes referred to as a ***FET***, because it is made using a type of transistor known as a Field Effect Transistor. Alternatively, it is referred to as a ***buffer amplifier*** (because it stops the main amplifier from loading down the microphone), or as a ***follower*** (because the voltage out of the microphone amplifier follows or equals the voltage between the diaphragm and the back-plate).

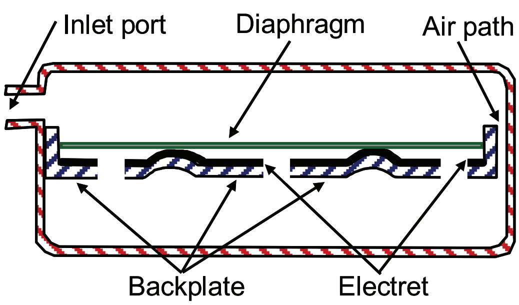

Figure 2.4 Cross section of an electret microphone.

Completely electronic microphones are also available. The ***silicon microphone*** (also referred to as the ***solid state***, ***integrated*** or ***micro-electro-mechanical systems (MEMS)*** microphone) is made by etching away parts of a block of silicon, and depositing layers of other materials onto it, using techniques similar to those used to make an integrated circuit. Microphone manufacturers expect that when problems of low sensitivity and high internal noise are solved (which is very close), it will eventually replace the electret microphone and should be smaller as well as being more reliable and reproducible (which will assist in making directional microphones with highly predictable characteristics). The silicon microphone should also eventually be cheaper as it can be made in a more automated fashion, and can potentially be made from the same block of silicon as the integrated circuit used for the main hearing aid amplifier.

2.2.2 Frequency response of microphones

Electret microphones have frequency responses that are essentially flat, although variations from a flat response occur both by design and by accident. A low cut is often intentionally introduced into electret microphones used in hearing aids. The low cut makes the hearing aid less sensitive to the intense low-frequency sounds that often surround us. These may not be perceived, even by normal-hearing people, but they can cause the microphone or the complete hearing aid to overload unless the microphone attenuates them.

Achieving the low cut is simple: a small passage-way between the front and back of the diaphragm allows low-frequency sounds to impact almost simultaneously on both sides of the diaphragm, thus reducing their effectiveness in moving the diaphragm. The larger the opening, the greater the attenuation, and the greater the frequency range over which attenuation occurs. The opening also equalizes the static air pressure between the front and back of the diaphragm, just as the Eustachian tube does for the ear. Microphones with different degrees of low cut are often used in custom hearing aids to help achieve a desired gain-frequency response for the hearing aid as a whole.

The second variation from a flat response is the result of an acoustic resonance within the microphone case. A resonance occurs between the air in the inlet port (an ***acoustic mass***), and the volume of air next to the front of the diaphragm (an ***acoustic compliance***, or spring). The mechanical compliance of the diaphragm itself, and of the air behind the diaphragm, also contribute to the resonance, which is called a ***Helmholtz resonance***.[a] This resonance causes a peak in the gain-frequency response, typically about 5 dB high and centered somewhere in the range from 4 to 10 kHz. The shorter and wider the inlet port, the higher is the resonant frequency, and consequently the greater is the high frequency bandwidth. Figure 2.5 shows a microphone gain-frequency response with a low cut below 500 Hz and a resonance at 5 kHz. Above the resonant frequency, and because of the resonance, the sensitivity of the microphone decreases as frequency increases.

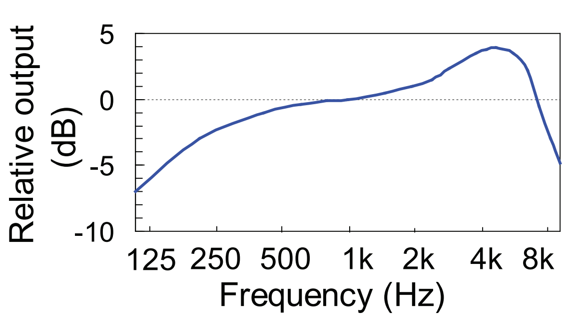

Figure 2.5 Frequency response of a typical electret microphone with tubing on its input port.

2.2.3 Microphone imperfections

The major imperfection with microphones is that they eventually break down if they are exposed to adverse chemical agents, like perspiration.

Less dramatically, all electronic components generate small amounts of ***random electrical noise***, and microphones are no exception. The noise is partly the result of random motion of air molecules against the diaphragm, and partly the result of random electrical activity within the internal microphone amplifier.[1780] This noise, when sufficiently amplified by the main hearing aid amplifier, is sometimes audible to the hearing aid user in quiet environments, particularly if the user has near normal hearing at any frequency. Microphone noise is greatest in those microphones that use an internal acoustic path to steeply roll-off the low-frequency response of the microphone.

Another imperfection of microphones is that as well as being sensitive to sound, they are sensitive to ***vibrations***. This occurs because if the microphone is shaken, the inertia of the diaphragm causes it to move less than the outer case of the microphone. Consequently, the diaphragm and the case move relative to each other, just as they do for a sound wave, so the microphone generates a voltage reflecting the magnitude and frequency of the vibration. Why does this matter?

The first consequence of a microphone's sensitivity to vibration is that any vibrations will be amplified into an annoying sound. For example, rubbing of the hearing aid case (e.g. by clothing next to a body aid) will be audible. Direct vibration of the body, such as occurs when running on a hard surface, may also be audible as an unwanted thumping noise.

The second consequence is that when the hearing aid receiver operates it creates vibrations as well as sound. The microphone picks up some of these vibrations, converts them to an electrical signal, and they are then amplified by the hearing aid and passed to the receiver, which creates further vibrations. If the

[a] At the Helmholtz resonant frequency, the air in a tube and volume to which it is connected vibrates responsively, just as a mass on a spring vibrates easily at its resonant frequency.

mechanical transmission of the vibrations from the receiver to the microphone is strong enough, and/or if the gain of the hearing aid is high enough, then this feedback loop may cause an audible oscillation, usually at a low frequency. Hearing aid designers avoid this by careful mounting and placement of the microphone and receiver, but if either of these become displaced, the hearing aid can become unstable due to this internal feedback loop.

Displacement of the transducers from their proper position is more likely to occur for in-the-ear (ITE), in-the-canal (ITC) or completely-in-the-canal (CIC) hearing aids because of their small size and custom manner of construction. A repair consists of repositioning either or both transducers. ***Internal feedback*** that is not sufficiently strong to cause an audible oscillation can be detected from the coupler response of the hearing aid. It is indicated by bumps in the frequency response that are present at high volume control or gain settings, but which disappear at lower volume control or gain settings.

A further imperfection is that microphones will overload and distort the sound if the input sound pressure is too great.

Another possible imperfection can occur with bad design or construction of a hearing aid. If the microphone is mounted with a long thin tube on its inlet port, the Helmholtz resonance referred to in the preceding section is moved downward in frequency. This causes a larger peak in the gain-frequency response, and a rapid decrease in gain for frequencies above this peak frequency.

The last imperfection is that microphones are subject to ***wind noise***. When wind hits an obstacle like a head, a pinna, or a hearing aid, turbulence is created.[458] Turbulence inherently contains marked pressure fluctuations, so the microphone indiscriminately converts these to sound: in this case an audible noise dominated by low- and mid-frequency components. Even moderate wind speeds can produce extremely high SPLs at the microphone input, sometimes sufficient to overload the microphone.[1954]

How directional microphones work

Microphones can be made to have a sensitivity that depends on direction of arrival by feeding sounds to both sides of the diaphragm from two separate inlet ports (the open ends of the microphone tubing), as shown in Figure 2.6. The directional properties of the microphone depend on two delays:

- The maximum external time delay is the time taken for sounds arriving from the front or rear to get from one inlet port to the other, and is approximately equal to the distance between the ports divided by the speed of sound in the vicinity of the head.*
- The internal time delay arises because the rear port contains an acoustic damper or resistor (Section 2.7). This combines with the cavity at the back of the diaphragm to create a low-pass filter that passes most of the amplified frequencies without attenuation, but with some delay that is inherent to all filters.[255]

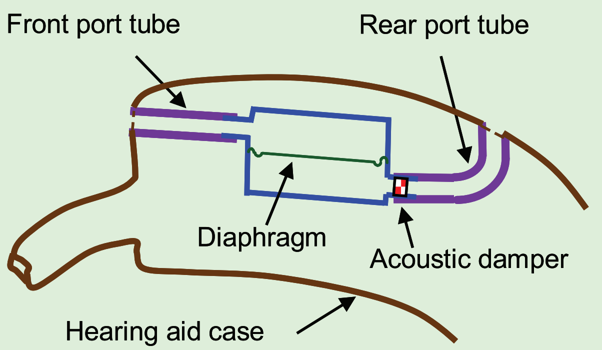

Figure 2.6 Diagram showing the sound paths in a directional microphone.

Sound coming from the rear direction hits the front port later than the rear port. However, the sound entering the rear port is delayed as it passes through the internal low-pass filter. If the internal and external delays are equal, then sound from the rear will reach both sides of the diaphragm at the same time, and there will be no net force on the diaphragm. Such a microphone is insensitive to sounds from the rear. If the internal delay is less than the external delay, the microphone will be insensitive to sounds coming from other directions.

* The effective speed of sound is lower than usual near the surface of the head because of waves that diffract around the head in both directions.[1121]

Keeping the microphone inlet away from the wind flow can minimize the amount of wind noise. A cosmetically unacceptable, but very effective way to do this is to place some plastic foam over the microphone port.[651] A cosmetically better way to achieve it is to place the microphone port inside the ear canal, as occurs for CIC hearing aids.[1954] The deeper the faceplate within the canal, the greater will be the avoidance of wind noise. A further option is to have a large microphone port opening covered by a mesh screen or dome.[459] This reduces noise because the pressure fluctuations across the large opening partially cancel each other, which cannot occur across a small opening. A more effective option that is suitable for some people is to wear a light scarf, with sufficiently open weave that it does not obstruct sound or create feedback oscillation.[191] This will prevent wind from hitting the hearing aid and the pinna, and much more importantly will prevent turbulence created by the head from reaching the microphone port.

2.2.4 Directional microphones

Directional microphones suppress noise coming from some directions, while retaining good sensitivity to sounds arriving from one direction.

The directional sensitivity of microphones is usually indicated on a ***polar sensitivity pattern***. Figure 2.7 shows the polar sensitivity pattern for a microphone like the one described in the accompanying panel. This particular response shape is called a ***cardioid***, because of its heart shape. By changing the ratio of the internal delay to the external delay (see panel), a whole family of patterns can be generated as discussed further in Chapter 7. The opposite of a directional microphone is an ***omni-directional*** (i.e., non-directional) microphone, which has a single port and a polar pattern in the shape of a circle.

What directional pattern is most desirable? In many real-life situations, unwanted noise arrives more or less equally from all directions. Even if the noise originates from only one or two sources, room reflections cause the energy to arrive at the aid wearer from all directions. By contrast, if the aid wearer is standing close to the person he or she wants to hear, the wanted signal will arrive mostly from directly ahead. A good directional microphone should therefore have maximum sensitivity for sounds arriving from directly ahead, but the sensitivity averaged across all other possible directions should be as low as possible if intelligibility in noisy environments is to be maximized.

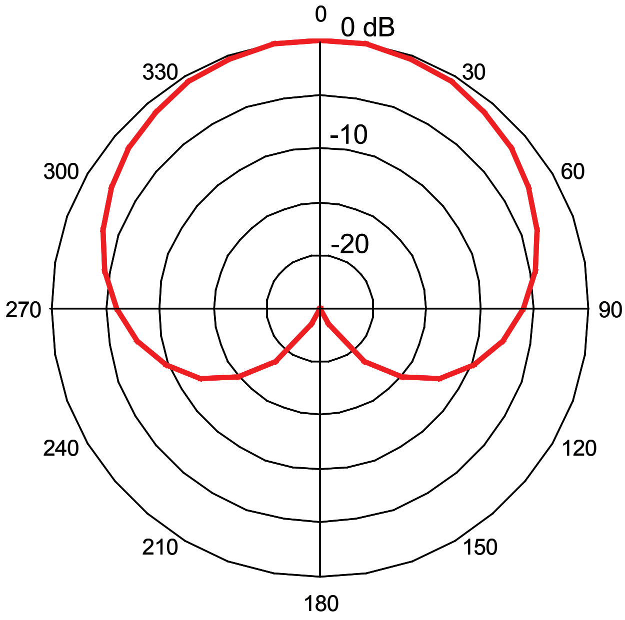

Figure 2.7 Directional sensitivity (in dB) of a microphone with a cardioid sensitivity pattern.

This ratio of sensitivity for frontal sounds relative to sensitivity averaged across all other directions is referred to as the ***directivity index (DI)***.[b] The "other directions" can be restricted to just the horizontal plane (giving a two-dimensional DI) or to all directions in space (giving a three-dimensional DI).

Sometimes, the ratio of frontal sensitivity to rearward sensitivity is quoted in hearing aid specifications. This ***front-to-back ratio*** is a misleading measure because it says nothing about the effectiveness of the hearing aid in suppressing noise arriving from directions other than precisely behind the aid wearer.

Neat polar patterns like the one shown in Figure 2.7 occur only for hearing aids suspended in free space, because when hearing aids are worn, the head introduces a polar pattern of its own. In fact, even omni-directional microphones act as though they are

[b] More precisely, the ***directivity factor*** is the ratio of power out of the microphone for a frontal source, to the power out of the microphone when sound comes equally from all directions. The ***directivity index*** is the decibel equivalent of this directivity factor.

somewhat directional when they are placed on the head, although the direction of maximum sensitivity is more to the side than the front, particularly for BTE hearing aids. Directionality occurs because the head and pinna attenuate the sound when they come between the source and the microphone, and boost the sound when the microphone is positioned between them and the source. These boosting and attenuating effects of head diffraction increase in magnitude as frequency rises (Figure 15.6).

Directional microphones have been available for decades, in both BTE and ITE hearing aids, but have most commonly been used in BTE hearing aids.[1458] Many early hearing aids appear to have had poorly designed directional microphones, such that they were barely more directional than an omni-directional microphone when mounted on the head.[91]

When hearing aid wearers need to hear people behind them, it is counterproductive to use a directional microphone. There are two solutions to this, both of which are in commercial use. The first is for a hearing aid to include both a directional microphone and an omni-directional microphone, and for the user (or the hearing aid itself) to select the microphone most appropriate to each situation.

The second, much more common, solution is for the hearing aid to incorporate two separate omni-directional microphones, each with one inlet port. When an omni-directional sensitivity pattern is needed, the output of one microphone is selected or the two outputs are added together. When a directional sensitivity pattern is needed, the two microphones are used in a different combination. In this ***dual-microphone*** technique, the output from the second microphone is electronically delayed and subtracted from the first microphone, making an exact electronic equivalent of the processing that happens acoustically in a single directional microphone.

Whichever solution is used, the user can switch between directional and omni-directional modes, which also enables the user to appreciate the advantages of the directional mode.[1460]

Figure 2.8 shows the directivity index versus frequency for various representative hearing aids. Note, however, that depending on the design (primarily the port spacing and internal low-pass filter or delay used), particular hearing aids can have directivity indices larger or smaller than those shown. Directivity indices even larger than those shown in Figure 2.8

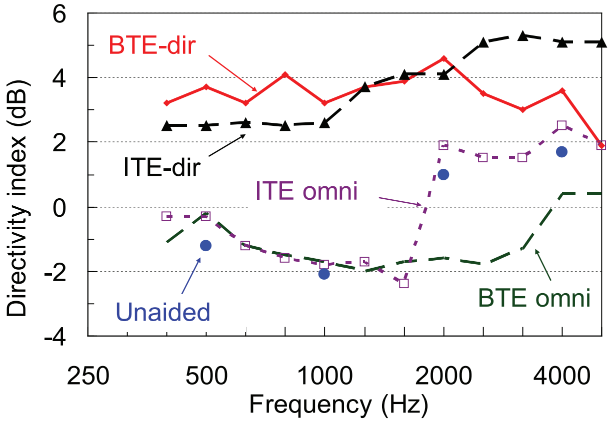

Figure 2.8 Directivity index, measured in the horizontal plane, for a directional and omnidirectional BTE and a directional and omnidirectional ITE.[1504] Also shown (as blue dots) is the directivity index for an unaided ear, measured on KEMAR.[454]

can be obtained with hand-held, chest-worn, spectacle-mounted, or bilateral microphone arrays. An omni-directional microphone has a DI of 0 dB when measured in free space, but when mounted on the head its DI varies from about -1 dB in the low frequencies to 0 to 2 dB in the high frequencies, depending on the microphone location.

Directional microphones inherently have a low-cut gain-frequency response shape, progressively reducing gain below about 2 kHz. An electronic filter can be used to boost the low-frequency gain and so partially compensate for this, but such a filter also boosts the internal microphone noise, which may then become excessive.

The interfering effect of background noise is the single greatest problem reported by hearing aid wearers. Directional microphones are particularly important to hearing aids because they are the only form of signal processing that can improve the ***signal-to-noise ratio (SNR)*** in a way that significantly improves intelligibility. The benefits of directional microphones are well established.[705, 923, 1034, 1824] The extent to which a conventional directional microphone improves the ability to understand non-reverberant speech in noise can be estimated by averaging the directivity across frequency. The benefit decreases as the environment becomes more reverberant, unless the source of the wanted signal is very close to the listener.

Chapter 7 gives a much more detailed description of conventional and advanced directional microphones and microphone arrays – the technology, their measurement, and their advantages and limitations.

2.2.5 Microphone location

Hearing aid microphones are usually located within the hearing aid, but can be located in an accessory such as a hand-held microphone, a wireless transmitter, or a satellite microphone located on the opposite side of the head. These devices will be covered in Chapters 3 and 17, and the acoustic effects of different microphone locations will be covered in Section 4.3.2.

2.3 Amplifiers

The basic function of an amplifier is simple. Its job is to make a small electrical signal into a larger electrical signal. Because the microphone has already converted the sound to electrical voltages and currents, the amplifiers can do three things. First, they can make the voltage larger, but not affect the current. Second they can make the current larger, but not affect the voltage. (We have already met one of these inside the microphone case.) Third and most commonly, they can make both the voltage and the current larger.[c] All three options result in the signal having more power when it comes out of the amplifier than when it entered. Of course, this additional power must come from somewhere. The job of the amplifier is to take power from the battery and transfer it to the amplifier output in a manner controlled by the input signal. Thus, the output waveform (either voltage or current or both) is simply a larger version of the input waveform.

2.3.1 Amplifier technology

The key element in an analog amplifier that allows a current to be controlled by a smaller current (or by a small voltage) is the ***transistor***. Although a single transistor will provide amplification, amplifiers nearly always are made up of several transistors and resistors connected together to provide better performance than is achievable with a single transistor. These multiple transistors and resistors are made, using photographic and chemical techniques, into an ***integrated circuit (IC)***. Transistors can be made using one of two broadly different types of technology: bipolar and CMOS (Complementary Metal Oxide Semiconductor). Each type has advantages: bipolar transistors tend to have lower internal noise; and CMOS transistors tend to use less battery power. Both types are used in hearing aids, and both can have acceptably low noise and power consumption. For hearing aid applications, an IC amplifier can contain from a few dozen to a few thousand transistors, depending on the complexity of the hearing aid. As nearly all hearing aids now rely on digital signal processing (Section 2.4), the analog amplifiers described in this section perform minor roles, typically at the input and output of the hearing aid, rather than providing the main amplification as they did in completely analog hearing aids.

Complete amplifiers also need other electrical components. ***Diodes***, which allow current to flow one way but not the other, are used to sense the size of signals, and are built into the IC. ***Capacitors*** are needed for various purposes, including the making of filters. If they are small enough, these are also built into the IC. If not, separate, discrete capacitors have to be used.

In most hearing aids, the ICs are mounted onto ***circuit boards*** with electrical connections already printed on them, such as the one shown in Figure 2.9. These cir-

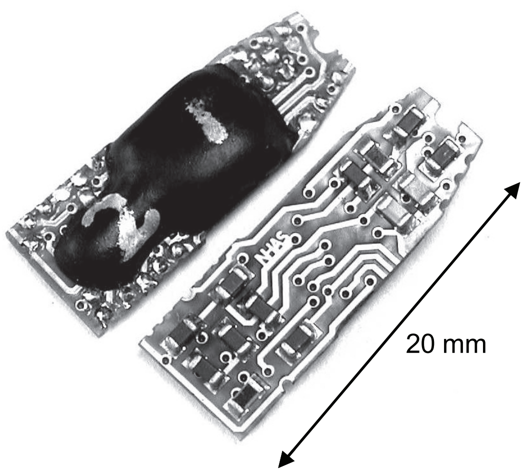

Figure 2.9 An amplifier board from a high-power BTE hearing aid. Integrated circuits are mounted on one side (under the protective coating) and individual components are mounted on the other, The protective coating protects the ICs against physical damage and the ingress of moisture and contaminants.

[c] For readers not familiar with voltage and current, a water analogy might be useful: voltage is the equivalent of water ***pressure***, whereas current is the equivalent of water ***flow***.

cuit boards can be made of fiberglass (which is rigid), or plastic (which is flexible). Alternatively, they can be made of rigid ceramic, in which case they are referred to as ***substrates***. The boards fill two functions. First, they provide the electrical connections between discrete components (like capacitors) and the IC. Second, they make it easier for the person (or machine) assembling the hearing aid to connect other devices (like battery terminals and volume controls) to the amplifier than if the connection had to be made directly to the IC. The circuit boards with their IC(s), interconnections, and sometimes other components are often referred to as ***hybrids*** (because they contain different types of electronic devices), although sometimes this term is reserved for boards made of ceramic material.

2.3.2 Peak clipping and distortion

An ideal amplifier would have the gain-frequency response required, would generate no noise internally, and would not distort the signal, no matter how large the input signal was. Real amplifiers live up to this ideal to varying degrees. The most noticeable deviation from ideal occurs when signals get too large for an amplifier to handle properly.

Amplifiers cannot produce signals larger in voltage than some specified maximum. This maximum is usually equal to, or related to, the battery voltage. If the biggest signal in the amplifier (usually the output signal) is near this maximum, and either the input signal level or the gain of the amplifier is increased, then the amplifier will clip (remove) the peaks of the signal. An exception to this occurs for amplifiers containing compression limiting, as discussed in Section 2.3.3.

Figure 2.10 shows the output waveform that results from ***peak clipping*** when the input signal is a sine wave. The dotted line shows what the output signal would be if no peak clipping occurred. Because the output is no longer a sine wave, it contains components at frequencies not in the input signal. These additional components are called ***distortion products***. When the input is a sine wave, the distortion products occur at frequencies that are harmonics (i.e. integer multiples) of the input frequency. Consequently, the process is called ***harmonic distortion***. All amplifiers create some distortion, and all amplifiers create large amounts of distortion if the signal is sufficiently peak clipped. If the peak clipping is symmetrical, the distortion products occur only at odd harmonics of the input frequency. If it is asymmetrical, then both even

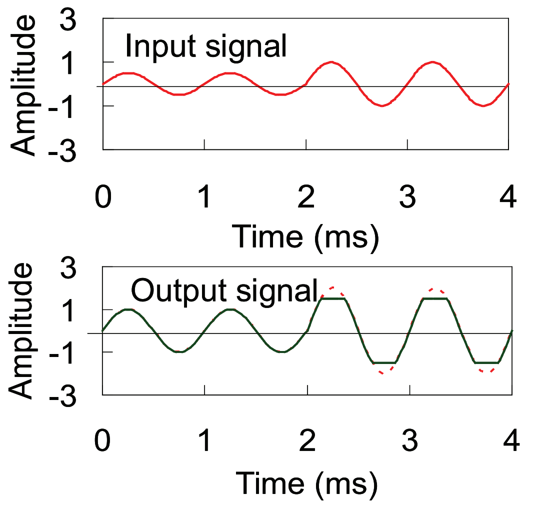

Figure 2.10 A signal that has been linearly amplified (from 0 to 2 ms), but amplified and peak clipped (from 2 to 4 ms) whenever the output signal reached the maximum amplitude that the amplifier was capable of delivering.

and odd harmonics are likely to be produced. Usually, the low order harmonics (the second and the third) are the most powerful. Consequently, distortion is sometimes quantified by expressing the power of these two components relative to the power of the wanted signal. More commonly, the power of all the distortion products is summed and expressed relative to the power of the wanted output signal component. This is referred to as ***total harmonic distortion (THD)***.

Distortion degrades the quality of speech and other signals when present in excessive amounts.[16, 370, 701, 858, 969, 1734, 1769] When present in larger amounts it also degrades intelligibility.[369, 370, 620, 826] Even when the distortion represents only 10% of the total signal power, speech quality is adversely affected.[1031] Section 10.7.2 will discuss under what circumstances peak clipping is acceptable, and sometimes even recommended, for use in hearing aids.

When a more complex signal is peak clipped, the distortion products occur at frequencies that are harmonics of all the frequencies in the input signal, and at frequencies that are combinations of all the harmonics. If two tones, with frequencies f_1 and f_2 are input, for example, distortion components will occur at $2f_1$, $3f_1$, $4f_1$, $2f_2$, $3f_2$, $4f_2$, f_2-f_1, $2f_2$-f_1, $2f_1$-f_2, $3f_1$-f_2, to name but a few frequencies. Although the mechanism causing the distortion is exactly the same as for harmonic distortion (peak clipping is the most common cause), the result is called ***intermodulation distortion***, because the distortion products arise from the modulation (mutual alteration) of every component in the input signal by every other component in the input signal.

Although peak clipping and distortion has been discussed here in the context of amplifier performance, hearing aid microphones (uncommonly) and receivers (more commonly) can also peak clip a signal. Peak clipping and its resulting distortion, or even worse forms of distortion, also occur when digital signal processors overload.

Analog hearing aids typically had several amplifiers in a chain, culminating in the output amplifier that used most of the battery current. Several types of output amplifiers (class A, class B, class D and class H) were used, depending on the power of the hearing aid and other design considerations.[436] Digital hearing aids typically use a pre-amplifier prior to the analog-to-digital converter and may have an output amplifier, to achieve the highest possible output power, following the digital to analog converter (Section 2.4.6).

2.3.3 Compression amplifiers

Section 1.1.2 discussed how people with sensorineural hearing impairment have dynamic ranges smaller than normal, so that less amplification is required for intense input sounds than for weak input sounds. Chapter 6 and Section 10.4 will further elaborate on how the amount of amplification could and should decrease as input level increases. It is the job of the compression amplifier to achieve this change of amplification when input level changes. The concept is simple and dates back to 1937.[1695] *A **compressor** is nothing more than an amplifier that turns down its own gain as the input to (or the output from) the amplifier increases.*

Figure 2.11 shows the block diagram of a ***feedback compressor***. The signal at the feedback point, *F*, is fed to the level-detecting device, whose job is to convert the rapidly varying audio signal into a more slowly varying control signal. The size of the control signal represents the level of the signal at *F* averaged in some desired way over some appropriate duration. The control signal is fed back into the control input, *C*, of the compression amplifier and tells the compression amplifier how much gain should be applied to the input signal. We can immediately see why the instantaneous waveform at *F* cannot be applied directly to the control input of the compression amplifier: if the compressor gain decreased every time the instantaneous waveform increased in size, the compressor would distort the detailed shape of the waveform. The compressor is meant to leave the fine detail in the waveform unchanged, while it more gradually varies the gain applied to the waveform. A compression amplifier is also called an ***automatic gain control (AGC)*** or an ***automatic volume control ()***. The last term is used only when the compressor vares the gain very slowly.

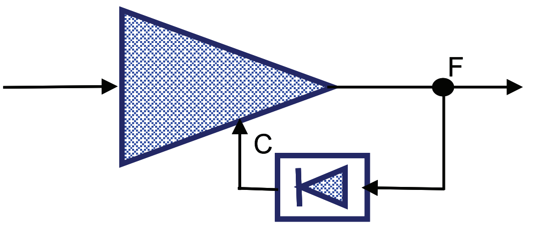

Figure 2.11 Basic feedback type compression amplifier block diagram.

Figure 2.12 shows the effect that a compressor might have on an input signal that varies in level. Notice that the difference in level between the low and high intensity parts of the signal has been decreased, but the detail of the waveform has not been significantly

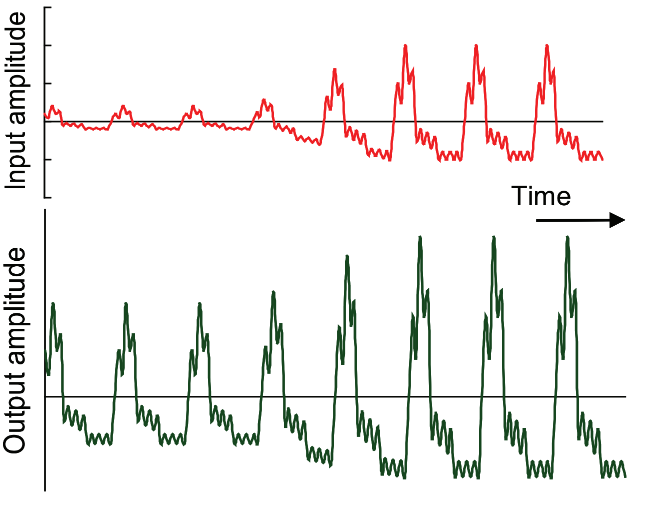

Figure 2.12 Effect of a compressor on a waveform varying in level.

affected. The compressor functions just as if a human finger inside the hearing aid had rapidly, but smoothly, turned down the volume control as soon as the output signal increased in level.

2.4 Digital Circuits

The amplifiers and signals discussed in the preceding sections all strictly relate to ***analog*** technology. In analog technology, which has been around since the invention of the telephone, an electrical voltage (or current) is *analogous* to the acoustic sound pressure, hence the name. When the sound pressure increases from one moment to the next, so too does the electrical signal produced by the microphone. A newer technology, which had already been used in research for twenty years, became available in commercial head-worn hearing aids in the mid-1990s. This, of course, is ***digital*** technology, which almost all hearing aids now sold use. The advantages of digital technology include greater predictability of operation, less internal noise, and particularly the ability to do complex operations in small ICs that consume little power.

As in analog hearing aids, digital hearing aids use a microphone to convert sound to an analog voltage. As shown in Figure 2.13, an analog-to-digital converter changes this voltage to a series of numbers, using the principles outlined in the next section. The hearing aid's digital signal processor then performs arithmetic on these numbers to manipulate the sound (Sections 2.4.2 to 2.4.5), before the numbers are turned back into an analog signal by the digital-to-analog converter (Section 2.4.6).

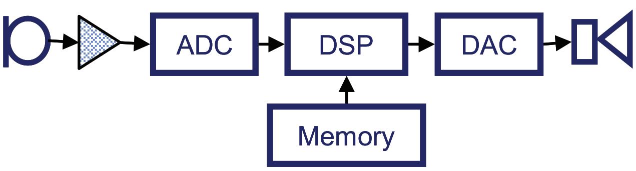

Figure 2.13 Basic components of a digital hearing aid incorporating a pre-amplifier, analog-to-digital converter (ADC), digital signal processor (DSP), digital-to-analog converter (DAC), and memory to hold the desired amplification characteristics.

2.4.1 Analog-to-digital converters

In digital technology, sound is represented as an ever-changing string of numbers. It is the job of the ***analog-to-digital converter (ADC)*** to change the analog electrical voltage coming from the microphone into these numbers. ***Sampling*** is the first step in this process. A signal is sampled by first noting the size of the signal at regular intervals in time, and totally ignoring the value of the signal at other times between these sampling points. If we want the sampled signal to be a good representation of the original signal, these samples must be obtained very often. They must follow each other more quickly than the signal waveform can make marked changes in direction. It can be shown mathematically that no information about the original signal is lost provided the ***sampling frequency*** (also called ***sampling rate***) is greater than twice the highest frequency component present in a complex signal. Thus, if a hearing aid is to faithfully amplify signals up to, say, 10 kHz, the sampling frequency has to be at least 20 kHz. This means the waveform is sampled every 1/20,000 of a second, or every 50 µs.

A hearing aid has to contain a low-pass filter to make sure that signals going into the analog-to-digital converter are indeed lower in frequency than half the sampling frequency.[d] This filter is called the ***anti-aliasing filter***. It gets this name because if a signal component with a frequency *greater* than half the sampling frequency gets into the analog-to-digital converter, the hearing aid will amplify this signal as though it has a frequency *lower* than half the sampling frequency. That is, signals with excessive frequency ***alias*** themselves down to lower frequencies. The anti-aliasing filter, which may be an intrinsic part of the analog-to-digital conversion process, prevents this undesirable distortion from occurring. Because no filter is ideal and some sound signal at frequencies just above the anti-aliasing cut-off frequency passes through the filter, the sampling frequency is usually selected to be 10 to 20% higher than the theoretical minimum.

Having sampled the waveform, which in Figure 2.14 occurs at each of the arrows, each of the resulting samples then has to be represented as a number. The designer of a digital system decides how many different numbers are going to be allowed. Suppose, for simplicity, only the eight integers from -3 to +3

[d] The frequency equal to half the sampling frequency is referred to as the ***Nyquist*** frequency.

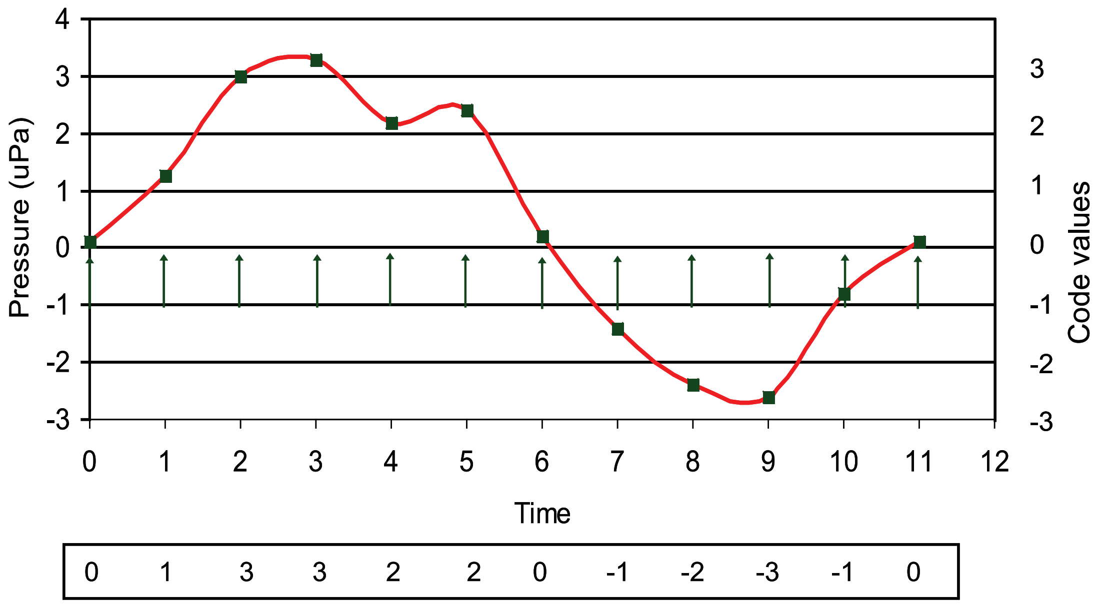

Figure 2.14 An analog pressure waveform (plotted using the left hand scale), the sampled values (shown by the dots on the waveform), the sampling signal (represented by the regular series of arrows), and the digitized codes (read from right hand scale) that approximate and represent the sampled values (enclosed in the rectangle beneath the time axis).

were to be allowed. The waveform shown at the top of Figure 2.14 would be represented by the numbers (called the ***code***) shown at the bottom of the figure, because these numbers are the allowable code values closest to the actual sample values. The sampled waveform has now been ***digitized***.

One more step remains. The digitized code values are broken up into ***bits***.[e] The word *bit* is a contraction of the words ***bi****nary digi****t***. The numerals we use in everyday life are allowed to have only one of ten values (from 0 to 9), and we make up bigger numbers by combining them using multipliers of 10, 100, 1000 and so on. Binary digits are allowed to have only two values (0 or 1), and we make up bigger numbers by combining them using multipliers of 2, 4, 8, 16, and so on. The numbers shown at the bottom of Figure 2.14 are repeated in Table 2.1, and the corresponding bits are shown there. Because we need to convey negative and positive numbers, the numbers can be arranged so that the first bit represents the sign of the number.

Table 2.1 Break-up of the digitized code values of Figure 2.14 into three-bit words.

Digitized code		Fours	Twos	Ones
0	=	1	0	0
1	=	1	0	1
3	=	1	1	1
3	=	1	1	1
2	=	1	1	0
2	=	1	1	0
0	=	0	0	0
-1	=	0	0	1
-2	=	0	1	0
-3	=	0	1	1
-1	=	0	0	1
0	=	1	0	0

[e] In reality, choosing the closest allowable number and expressing this number in the form of bits occur in a single operation inside the ADC.

A little thought will show that just as we can represent eight numbers with a three-bit word, we can represent 16 numbers with a four-bit word, 32 numbers with a five-bit word, and so on. Home compact disk (CD) players use a 16-bit word to represent sounds, and this word length means that 65,536 different numbers can be represented. Hearing aids use a similar number of bits. Eight bits comprise the ***byte*** that computer enthusiasts talk about when bragging about how much memory their computer or hard disk has. The memory capacity of these devices is usually measured in Megabytes (a million bytes), Gigabytes (a billion bytes) and Terabytes (a trillion bytes). Digital hearing aids require a small amount of memory, to enable their electroacoustic performance to be programmed to suit particular aid wearers.

Why is it necessary to break the sampled values up into bits? First, it is convenient for computers, because they can most efficiently represent signals as either *on* or *off* and these can easily be thought of as the two values of a binary digit.[f] More importantly, having only two allowable values makes a signal almost incorruptible when it is stored, transmitted, or used in any way. Suppose that inside a hearing aid, a "0" corresponds to 0 Volts, and a "1" corresponds to 1 Volt. What will happen if electrical noise inside a hearing aid causes the 0 V signal to be turned into, say, 0.1 V as the signal is passed from one part of the hearing aid to the next? Nothing! The next stage of the hearing aid knows that signals are allowed only to be 0 V or 1 V, so it treats the corrupted signal as if it was the closest allowable value, which is 0 V. The internal hearing aid noise has caused no error whatsoever, whereas in an analog signal the noise would be inextricably mixed up with the signal and would eventually get passed to the hearing aid wearer. Note that this advantage applies only to noise generated internally within the digital part of the hearing aid, not noise picked up by the hearing aid or created within the microphone, pre-amplifier, or ADC.

2.4.2 Digital signal processors

Apart from protecting against noise, conversion to digital form carries a second advantage. Once sound has been represented as a series of numbers, we can modify the sound just by doing arithmetic with the numbers. For example, if we wished to amplify a sound by 6 dB, then we must double the amplitude of the sound. Simply multiplying the number representing each sample of the sound by 2 will do this. For greater amplification, we multiply each sample by a larger number. Suitable combinations of arithmetic operations accomplish other changes to the sound. For example, to make a low-pass filter, we can take each sample of the sound and add to it some fraction of the preceding sample. We can think of this as averaging or smoothing a series of numbers, which decreases the size of any rapid fluctuations that are present (i.e. the high-frequency components). Using arithmetic, we can modify the sound in just about any way that we can with analog electronics. Fortunately, digital electronics can do more than simply mimic analog electronics as we will see in Chapters 7 and 8. There are two contrasting types of digital signal processors used in hearing aids. These could be referred to as ***hard wired*** and ***general arithmetic processor*** hearing aids, and are described in the next two sections.

2.4.3 Hard-wired digital processing

In hard-wired digital hearing aids, different parts of the processor each perform some specific function (e.g. a compressor, or a filter). These blocks are connected together in a particular, fixed order. That is, the samples of the sound wave are passed through the various blocks of the processing in a particular order, and each block can do only the function (e.g. filtering, compression) that is has been designed to do.

Another way of thinking about this is that if the digital hearing aid is represented as a block diagram (just as with an analog aid), it can process sounds in *only* the way represented by that particular block diagram. The amounts by which digital hearing aids amplify and filter can be programmed in a very flexible manner. Thus there is no disadvantage in a digital aid being hard-wired, provided the block diagram is appropriate for the aid wearer's hearing loss, and provided the parameters of each block (e.g. compression ratio, filter corner frequency) can also be adjusted to values appropriate to the aid wearer. Digital hard-wired aids currently on the market have amplification characteristics that can be adjusted very flexibly.

[f] At the deepest level inside a computer or digital signal processor, a "0" is represented by a transistor with voltage across it and a "1" by a transistor with no voltage across it, or vice versa.

2.4.4 General arithmetic digital processing

An alternative to the hard-wired digital aid is a digital aid that simply has an arithmetic processor at its heart. What would such an aid do? As with a computer, it would do whatever its software told it to do! If its software told it to filter the signal into three parallel bands, compress the signals in each band, and add these signals together, then the general arithmetic processor would function just as if it was a three-channel compression hearing aid. If some different software was loaded into the hearing aid, then the aid could function as a single-channel peak-clipping hearing aid. There is no fundamental limit to what hearing aids of this type *could* do. Unfortunately, to date we have a fairly restricted list of things that we would like them to do, as discussed in Chapters 6 to 8. (Of course, we would *like* them to amplify sound so that the output is always comfortable, always intelligible, and never has any noise in it. Unfortunately this is not a very productive wish unless we are able to say exactly what operations the digital signal processor should perform on the sound to achieve this delightful state of affairs.)

If a hearing aid incorporating a general arithmetic processor has to be configured to a particular block diagram before it can function, is it useful to have a general processor of this type? Yes, and the extreme flexibility will probably be useful in four ways. First, such an aid can truthfully, and usefully, be marketed as several different types of hearing aid, depending on the software loaded into the aid by the manufacturer, after the aid has been assembled.[1394] The resulting cost saving to the manufacturer may eventually be passed on to the aid purchaser. Second, the manufacturer may market the aid as a super-flexible aid, in which the aid wearer can switch between different signal processing schemes (each with its own block diagram) using a remote control. Third, as new and improved processing schemes are developed (and hopefully proven), aid wearers could purchase new software that enabled their existing hearing aid to perform the new type of processing, provided the hearing aid has adequate processing power. Fourth, the time between development of a new algorithm and it becoming available to hearing aid wearers should be considerably less than the time needed to develop an application-specific IC.

Thus, we will have to view hearing aids as we now view computers: there is a hardware component and a software component (in addition to the software in the fitting system), and either can be upgraded without necessarily changing the other. There are now companies who produce signal processing software for hearing aids, and sell this to other companies who produce the hearing aids in which this software is installed. In the future, it is possible that some of these software companies will also sell directly to clinicians.

These flexibility advantages come at a price, however. Any particular set of calculations requires more processing power (and hence battery current) to perform than if the calculations were to be done in a purpose-designed, hard-wired digital circuit.[1008] Another way of looking at this is that for the same battery current, hard-wired processors can do more complex operations than general arithmetic processors.

General arithmetic processor hearing aids are sometimes described as ***open platform***,[1394] because it is open to people other than the IC manufacturer to write software that will run on the IC. Some hearing aids now on the market are open platform devices.

Although hard-wired and general arithmetic processors have been presented as the only two alternatives, they are actually just the two extremes of a continuum. General arithmetic processors can contain dedicated hard-wired circuits to handle frequently repeated calculations in an efficient manner. Conversely, hard-wired processors can contain a small general arithmetic processor that can control how parts of a hard-wired circuit are configured. For some time, most digital hearing aids are likely to be a hybrid of hard-wired circuits and general arithmetic processors. Over time, the proportion of processing handled by the general arithmetic processor is increasing.

The implication of all this is that the clinician should pay little attention to what manufacturers say about *how* the signal processing is performed. The clinician should simply ask what processing features are included that benefit patients and how flexibly that processing can be tailored to the needs of the patient. Whether the hearing aid is made out of digital hard-wired circuits, digital general-purpose circuits, analog circuits, or for that matter play-dough, is irrelevant. Hearing aids do not provide greater benefit just because they are constructed with some particular technology.[1386]

2.4.5 Sequential processing, block processing, and hearing aid delay

The way that hearing aids amplify sounds invariably depends on frequency. To achieve frequency-dependent amplification, digital hearing aids process sounds in two broadly different ways. The first way, which is analogous to how analog hearing aids operate, is to process the incoming signal ***sequentially***. At any given time, the computer is processing the current sample of the input signal, although the processing it does to that sample often will depend on the values of the preceding samples. For a slow-acting compressor, for example, the gain given to the current sample may depend on the value of many thousands of previous samples.

An alternative to this is ***block processing*** (also known as ***frame processing***, or ***windowing*** the signal). In this approach a number of input samples (typically 64, 128, 256, or 512) are taken in by the hearing aid before *any* computations on them are performed. Processing a complete block of input data at one time enables a ***Fourier transform*** to be calculated, with the result that the complete block is now represented by an amplitude and phase at every frequency (i.e. a frequency spectrum), rather than as the instantaneous value of the waveform at each point in time. The greater the number of input samples processed at once, the more finely the individual frequencies can be specified within the hearing aid. We say that the signal is now represented in the ***frequency domain*****.** Fourier transforms require a lot of arithmetic to be calculated, so efficient calculation methods must be used. The most common method is known as the ***fast Fourier transform (FFT)***. After the complete block of transformed data has been processed (i.e. altered in the desired way), an ***inverse FFT*** is used to convert the block back into the time domain. Each sample in the block is then output by the hearing aid, one sample at a time.

What the hearing aid does with this information depends on what we are trying to accomplish. It may monitor the spectrum from block-to-block to deduce when the hearing aid is whistling because of feedback, and then automatically change the amplification conditions until the whistle disappears. Alternatively, it may provide one frequency response characteristic when sound has dominant high-frequency components and a different characteristic when sound has dominant low-frequency components. As a further example, the hearing aid may measure the way the spectrum changes from block-to-block, deduce whether the signal is predominantly noise or predominantly speech, and alter the amplification characteristics in an appropriate manner.

Although block processing enables complex operations to be performed, it has a disadvantage if the blocks are too long: the output samples are delayed with respect to the input samples by at least the length of the block. Even with sequential processing, the output signal is delayed with respect to the input. Delays occur in the ADC and during filtering, and potentially in other signal processing algorithms. Filters delay the signal in a complex manner that is best characterized by the ***group delay*** at each frequency. The group delay at each frequency describes how much the envelope of signal components in this frequency region will be delayed.[330]

Unfortunately, not all the sound received by the aid wearer is delayed, as low-frequency sounds reach the eardrum via the vent, leakage paths around the earmold, or through the bones of the head and into the ear canal (Section 5.3.1). All of these paths bypass the hearing aid. For open-canal hearing aids, the acoustic (unamplified) sound path may dominate the delayed (amplified) sound path up to 1000 or 1500 Hz.

Any delay in the amplified sound path, even including the very short delay found in analog hearing aids, can disrupt the resulting gain-frequency response of the complete system for people with mild or moderate losses. The acoustic (non-delayed) and amplified (delayed) sound paths will partially cancel at particular frequencies but add constructively at frequencies intermediate to these. The resulting series of peaks and troughs in the frequency response is referred to as ***comb filtering***. The problem is greatest within the frequency range where the sounds arriving via the acoustic and amplified sound paths have similar magnitudes. The greater the delay, the greater is the likelihood that cancellation will occur within this frequency range.

Even within the amplified sound path, some filtering methods delay low frequency sounds to a greater extent than high frequency sounds. This is particularly likely to occur in hearing aids that use filter bandwidths that mimic those of the cochlea.

For various reasons, therefore, the output of a hearing aid will be delayed with respect to the input, and it is very likely that some frequencies will be delayed more than others. How much delay matters depends on which perceptual consequence of delay we assess.

Delays of around 5 ms in the amplified sound relative to the low-frequency, non-delayed sound can just be detected by the hearing aid wearer in ideal circumstances where the delayed version can easily be compared to a version with no or minimal delay.[17, 658] Delays of 5 ms can be differentiated from delays of 10 ms, but lead to equally acceptable sound quality.[166] As the delay increases beyond about 10 ms, the sound quality decreases, particularly for the user's perception of his or her own voice quality, though delays of up to 20 ms may be tolerable.[17, 1722, 1723, 1727] Similarly, when the low frequencies are delayed with respect to the high frequencies, differential delays as small as 5 ms can be detected.[1203] Delays around 10 ms are disturbing when the aid wearer speaks, and delays of 15 ms affect the intelligibility of incoming speech.[1725]

Delays of the complete signal longer than 30 ms disturb the production of speech by the aid wearer.[1723] Longer delays of 40 ms or more put the auditory information out of synchronization with visual information and so may disturb lip-reading, particularly for good lip-readers.[1167, 1744] Other research, however, has indicated that sound can be delayed with respect to vision by several times this amount before auditory-visual asynchrony is reported.[706]

Overall, it appears that the maximum acceptable delay in digital hearing aids is more determined by the effect of delays between different frequency regions, rather than by audio-visual dis-synchrony.

Because the amplified and acoustically transmitted sounds, with their disparate timing, interact over an octave or two, the resulting disturbance of the phase response is likely to affect the ability of hearing aid wearers to localize sounds. More research is needed on this issue.

The maximum acceptable delay, and its variation with frequency, is of great interest to hearing aid designers, as signal processing features such as compression and adaptive noise suppression can be made to work more effectively the more the signal is delayed. With a long delay, the hearing aid effectively gets to peek ahead at the way the signal is changing so that the hearing aid can change its amplification characteristics smoothly and more appropriately, but still in time to react to changes in the sound. Most hearing aids are designed to have delays of less than about 5 ms which is possibly conservative given the research into acceptable delays and the probable benefits of peeking ahead.[450]

2.4.6 Digital-to-analog converters

After the digital signal processor has altered the sound in some desired manner, the hearing aid must present the modified and amplified sound to the aid wearer. As there is no use presenting the aid wearer with a string of numbers, the modified numbers must be converted into an acoustic signal. This conversion is the job of the ***digital-to-analog converter (DAC)*** combined with the hearing aid receiver. Digital devices have traditionally done this by having a digital-to-analog converter that outputs an analog voltage, which in turn is fed to a receiver of some type to make the final conversion to sound.

To minimize power consumption, digital hearing aids use a different solution. The multiple bits that comprise each sample are converted into a single bit that changes at a rate many times higher than the sample rate. The converter is referred to as a ***digital-to-digital converter***. The high-speed serial output from this converter is fed to the receiver, which averages out the high-speed variations in the digital signal (i.e. acts as a low-pass filter) to produce a smooth analog signal. The digital-to-digital converter and the receiver thus combine to make up the digital-to-analog converter. The electronic parts of the digital-to-analog converter can be located either with all the other amplifier parts or inside the metal can that houses the receiver.

2.4.7 Specifications for digital hearing aids

Digital hearing aids have the same types of specifications, such as gain, maximum output, range of frequency response adjustment, compression characteristics, internal noise, and current consumption as did their analog predecessors. With digital hearing aids, however, some additional specifications indicate the likely audio quality and processing capabilities of the hearing aid. The following six specifications significantly affect the sound quality and sophistication of processing that hearing aids provide.

Instructions per second: Digital processors are characterized by the number of instructions or operations (such as multiplication or addition) that they can do in a second. A particular processor, for example, may be able to do 40 ***MIPS***, which stands for 40 million instructions per second.[g] Complex signal processing schemes generally require a greater number of instructions per second than less complex schemes. As examples, compression is more complex than peak clipping, multi-band processing is more complex than single band processing, and the more effective varieties of automatic feedback suppression are more complex than tone controls. For a given integrated circuit, increasing the number of instructions per second, to perform more complex processing, will increase current consumption and thus decrease battery life. Unfortunately, one cannot assume that a hearing aid that is calculating 40 MIPS is performing more complex processing than one that is calculating 10 MIPS, because each "instruction" in the lower-speed hearing aid may be more complex than in the higher-speed hearing aid.

Sampling rate: The sampling rate, or sampling frequency (Section 2.4.1), describes how many times per second the hearing aid samples the input signal. The major impact of the specification is that the hearing aid can amplify sounds only up to about 40 to 45% of the sampling frequency, with the absolute theoretical maximum being 50%. A second impact is that if the sampling rate is unnecessarily high, the complexity of the processing that the hearing aid can perform will be unnecessarily limited. This occurs simply because the hearing aid has to do each of the operations on more speech samples every second than may be justified by, say the upper frequency limit of the hearing aid receiver. Consequently, fewer operations can be performed on each sample. The bandwidth of any hearing aid is limited by the component that has the most restricted bandwidth, so there is no advantage in other components having an excessively high bandwidth.

Number of bits: Section 2.4.1 showed that we can represent each sample of the audio waveform by a number, which in turn is represented as a string of bits. The greater the number of bits, the greater the number of analog voltage levels that we can represent. If there are too few levels, the digital approximation of the original signal is too coarse. The errors made by selecting the nearest allowable level are equivalent to adding noise to the signal, and this is referred to as ***quantization noise***. Thus, the greater the number of bits, the better the digital approximation of the signal, and the less the quantization noise. The amount of quantization noise, compared to the biggest signal that can be represented without overload, can easily be estimated. The noise is approximately $6b$ dB below the biggest signal, where b is the number of bits. A 12-bit system will therefore have quantization noise 72 dB below the highest signal. When the largest possible signal was input to the hearing aid, the SNR would thus be 72 dB. While this sounds like a very high SNR, if the input level were to be decreased by 70 dB (which would occur for a 30 dB SPL signal input to a hearing aid that accepted an input signal of 100 dB SPL before it overloaded), the SNR would be only 2 dB, which does not seem so acceptable! Hearing aids may use different numbers of bits in different parts of the aid, depending on the dynamic range needed in each part. Also, clever coding schemes can be used to make a smaller number of bits produce sound as good as that produced by simple coding schemes with a greater number of bits. When comparing the performance of different digital hearing aids, specifications for the number of bits should thus be interpreted warily. In general though, the more bits the better, as the hearing aid will be able to handle a greater dynamic range of signals without adding excessive noise of its own.

Current consumption**:** The current consumption, and hence battery life and feasible battery size, depends on the instruction rate, the voltage at which the integrated circuit operates, and the technology used to make the integrated circuit. None of this is under the control of the clinician, or need be understood by the clinician, but the consumption directly affects the size and hence appearance of the finished hearing aid. Current consumption to power the digital processing, for a given number of instructions per second, is spiraling steadily downward and should continue doing so for as long as general computer technology continues to improve.

Processing delay: As mentioned in the previous section an excessive time delay from input to output of the hearing aid will degrade signal quality, particularly for the aid wearer's own voice, but longer delays facilitate more sophisticated signal processing that

[g] An alternative label for the MIP is the ***MOP***, which stands for millions of operations per second.

enables the hearing aid to react promptly but smoothly to changes in the signal dynamics. Hearing aids with long delays do not necessarily have more sophisticated processing than hearing aids with short delays.

***Physical size*:** Complex circuits, especially when they contain a lot of computer memory of the type needed by programmable hearing aids, can require an integrated circuit several mm by several mm (or approaching a quarter inch by a quarter inch). Because transducers have been slowly shrinking in size for the last fifty years, the size of the integrated circuit can have a big effect on the finished size of the hearing aid.

2.4.8 Digital versus analog hearing aids

Because of the many advantages of digital hearing aids, they have fully replaced analog hearing aids in that no new analog hearing aids are being designed. The biggest advantage is that they can perform more complex processing than is possible in analog hearing aids. Some operations just cannot realistically be done in analog aids (e.g. block processing to finely represent signals in the frequency domain), and many operations can be done with less power and circuit size if done digitally. Digital hearing aids are also able to make decisions about how to process the sound, depending on what they sense the overall acoustic environment to be. Further, provided they have enough processing capacity, digital circuits containing a general arithmetic processor can potentially be updated with new processing schemes as knowledge advances or hearing loss changes.

By the end of the last century, digital technology had advanced sufficiently rapidly that digital hearing aids required less power and volume than analog hearing aids performing operations of similar complexity. The power and size advantage of digital aids have further increased since then. The hearing aid can be made smaller if the integrated circuit is smaller. If the integrated circuit consumes less power, a smaller battery can be used for the same battery life. Consequently, the hearing aid can again be made smaller. Alternatively, more complex operations can be performed for the same battery current. The only significant disadvantage of digital hearing aids is the longer delay between the input and output as mentioned in Section 2.4.5. This is a small price to pay for the many advantages that digital signal processing confers.

It is, however, worth stating what digital aids cannot do. Digital hearing aids are sometimes referred to as *providing CD sound quality*. This analogy is only partly true: they use the same type of technology as compact disk players. Once a sound is converted into digital form, it is possible for the sound to be manipulated without the hearing aid adding any significant noise of its own. Unfortunately, by the time digital hearing aids get to manipulate the sound, it already has noise mixed in with it, because background noise enters the hearing aid along with the wanted sound. Also, the hearing aid microphone will add noise to the signal before the sound is converted to digital form. By contrast, compact discs are usually recorded in very quiet studios under ideal conditions. Consequently, little noise gets in along the way, and the result, for CDs, is a virtual lack of background noise. *This is never likely to be true for hearing aids*.

2.5 Filters, Tone Controls and Filter Structures

Tone controls and the filters on which they are based are extremely important as they enable hearing aids to have different amplification characteristics in different frequency regions.

2.5.1 Filters

The basic electronic structure that causes gain to vary with frequency is the ***filter***. Filters are known by their effect on signals:

- ***High-pass filters*** provide more gain to high-frequency sounds than to low-frequency sounds, which gives the sound a treble, or shrill quality.
- ***Low-pass filters*** provide more gain to low-frequency sounds than to high-frequency sounds, which gives the sound a muffled or boomy quality.
- ***Band-pass filters*** provide more gain to frequencies in a certain band than to either higher or lower frequencies.
- ***Band-stop filters*** provide less gain within a restricted range of frequencies than for all other frequencies.

Hearing aids make extensive use of high-pass, low-pass, and especially band-pass filters. Filters can also be designed to have an arbitrary response shape, such as that shown in Figure 2.15.

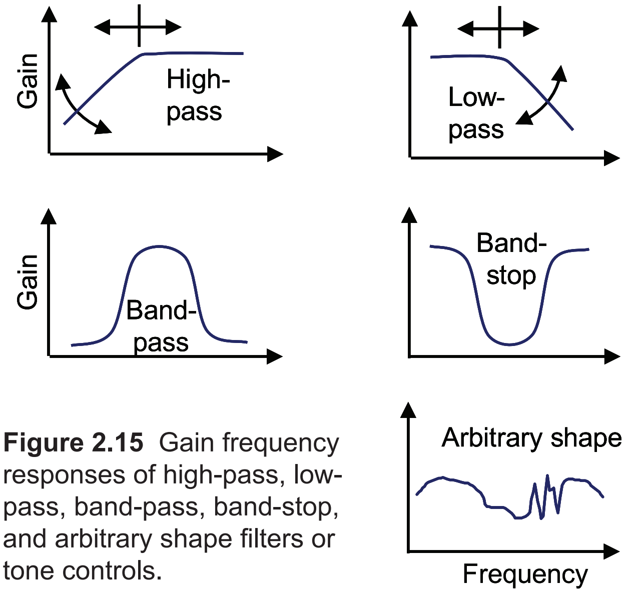

Figure 2.15 Gain frequency responses of high-pass, low-pass, band-pass, band-stop, and arbitrary shape filters or tone controls.

Almost all the filtering performed in hearing aids is achieved by mathematical manipulations while the signal is in digital form, and therefore represented by numbers, as discussed in Section 2.4.1. In one commonly used filtering method, an output sample is calculated by adding a fraction of the current input sample to suitable fractions of each of the previous *n* samples, where *n* is the *length* of the filter. (To gain access to the previous input samples, they are temporarily stored, and the oldest one is discarded every time a new sample is added.) These filters are said to have a ***finite impulse response (FIR)*** because once the input signal ceases, the output completely dies away a short, but finite time later. The time is equal to that taken for the *n* samples to pass by. Even complex, arbitrary response shapes are easy to implement as FIR filters in digital hearing aids, so they can be used to provide the gain-frequency response desired for the hearing aid.

In a second filtering method, the output sample at a given time is also made to depend on the *output* samples at previous times. Every input signal will therefore have an effect on the output that theoretically lasts forever (though its effect continuously and rapidly gets smaller with time). These filters are said to have an ***infinite impulse response (IIR)***. Their advantages compared to FIR filters are that complex filter shapes can be generated with fewer computations, and they generally cause less delay to the signal. Their disadvantage is that they can become unstable and oscillate, or alter the signal in unwanted ways, if the precision of the computations is not sufficiently high.

2.5.2 Tone Controls

The function of tone controls is the same in hearing aids as in home stereos: they cause the gain of the amplifier to vary with frequency. Tone controls get their name because they affect the ***tonal quality***, or ***timbre***, of sounds passing through them. A tone control can be constructed by making some aspect of the filter adjustable. A high-pass filter, for example, can have its response varied by changing the ***corner frequency*** (also called the ***cut-off frequency***) of the filter, or by changing the ***slope*** of the filter, as indicated by the arrows in Figure 2.15. Slopes of filters are commonly integer multiples of 6 dB per octave (e.g. 6, 12, 18, 24 dB per octave).

2.5.3 Filter structures

Filters can be combined in serial, parallel or serial-parallel arrangements.

Serial structures

Figure 2.16 shows a block diagram of a hearing aid comprising one low-pass, and one high-pass filter. The arrangement is referred to as a serial structure, because all sounds pass through all the blocks, one

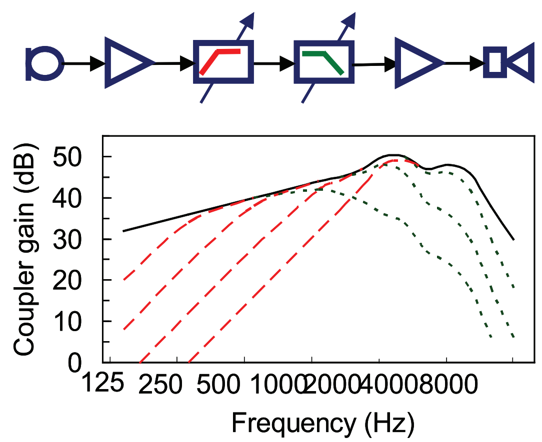

Figure 2.16 Block diagram of a serial structure, single-band hearing aid, and a range of low cut (dashed red curves) and high cut (dotted green curves) variations that might be made to the basic response (solid line).

Terminology: Multi-band or multi-channel?

The terms ***multi-band*** and ***multi-channel*** are usually used interchangeably, although some authors and hearing aid companies differentiate between them. Many hearing aids selectively filter those parts of a signal that lie within a certain frequency range, and process these parts differently from those parts of the signal at other frequencies. It is extremely important what this processing is (e.g. amplification, compression) but there is no form of processing that would dictate whether the group of signal components be called a *channel* rather than a *band* or vice versa. Some people call all the frequencies that pass through any individual compressor a *channel*, and all the frequencies whose amplitudes can be controlled with a single gain control a *band*. With this terminology, some hearing aids have more bands than they do channels.

It could perhaps be helpful to use *band* to mean the components or frequency range in question, and to use *channel* to mean the electronic chain of devices or mathematical operations through which this band of signal components pass.

after the other. This arrangement was common in analog hearing aids, but is little used now. The figure also shows the range of frequency responses that such a hearing aid can typically provide. Although combining high- and low-pass filters in a serial structure allows some flexibility of response shape, the flexibility offered by serial structures is usually inadequate unless one of the filters can be made to have an arbitrary response shape.

Parallel structures

Parallel structures, such as shown earlier in the block diagram in Figure 2.2, generally allow more flexibility, even with simple filters. The filters divide the sound into adjacent frequency regions. These are variously called ***bands*** or ***channels***. Parallel structures are simple conceptually, as sound in each frequency region can be amplified (or changed in various other ways) more or less independently of sound in other regions.[h] After the parts of the signal falling within each band have been amplified to the required degree, the parts are recombined in the adder. A possible disadvantage is that because each of the band-pass filters has a gradual transition from frequencies that are passed to frequencies that are rejected, individual frequency components can simultaneously create activity within two or more nearby channels. When the outputs from all the channels are recombined, the multiple versions of a single frequency component can recombine in a destructive manner at some frequencies, and in a constructive manner at other frequencies, thus imparting undesired ripples in the gain-frequency response.[i] Another form of distortion that can occur is for different filters to delay the signals passing through them by different amounts, which slightly smears out in time the overall signal after the channels have been recombined (Section 2.4.5).

Serial-parallel structures

While the distortions referred to in the previous paragraph can be avoided with careful design, a structure that guarantees these distortions do not occur is the serial-parallel structure, like that shown in Figure 2.17. The parallel bank of band-pass filters (or equivalently

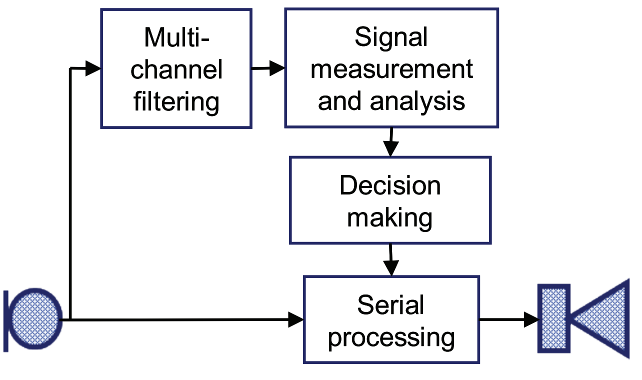

Figure 2.17 Series signal flow and parallel analysis structure used in some hearing aids. Modification of the signal occurs only in the serial path.

h Control of gain in each frequency region will be independent of gain in the other frequency regions provided each of the filters has a slope sufficiently steep to prevent sound from leaking to adjacent frequency bands.

i Adverse combination of the output of adjacent filters can be avoided if the filters all impart the same delay, and this delay is the same at all frequencies.

a Fourier transform) is used to determine the level and other characteristics of the signal present within each channel. These characteristics are then used to determine the filter characteristics that are to be imparted to the signal itself by the serial filter. If the serial filter is an FIR filter, it is relatively easy to ensure that the time and frequency distortions referred to do not occur.

2.6 Receivers

The receiver, which externally looks just like the microphone shown earlier in Figure 2.3, converts the amplified and modified electrical signal into an acoustic output signal.

2.6.1 Principle of operation

The receiver operates by magnetic forces. Figure 2.18 shows the receiver's operating principle and the basis of construction.[504] Current passes through a coil that encloses a piece of metal, temporarily turning it into a magnet. As the current alternates in direction, this piece of metal, called an ***armature***, is alternately attracted and repelled by two permanent magnets. The armature is very thin and can bend, so the end of the middle arm of the armature is free to move up and down between the magnets. The free end of the armature is linked by a drive-pin to the diaphragm, so that the diaphragm also vibrates backwards and forwards, and this produces the sound. Only a portion of the diaphragm is shown in Figure 2.18. This transducer seems simple, but making all this in such a way that it has a wide frequency response, consumes little power, leaks little magnetic field outside the case, and occupies almost no volume is a major technological feat.

An important implication of the construction is that the receiver will peak-clip, and hence not operate linearly, once the armature travels sufficiently far that it touches either of the magnets. Greater output can be obtained only by using a receiver with a bigger diaphragm, which increases the size of the receiver, or with the magnets further apart, which then requires greater electrical power for the receiver to operate.

2.6.2 Frequency response of receivers

Figure 2.19 shows the frequency response of a receiver connected to the tubing used in a BTE hearing aid and earmold. What causes all these bumps and dips? Mostly, it is the tubing. This tubing comprises a short length of tubing inside the hearing aid, and for a standard tube BTE, the earhook, and finally the flexible tubing terminating at the tip of the earmold or the dome inside the ear canal. For a thin-tube BTE, the earhook and earmold tubing are replaced by a single length of small-diameter tubing. The combined length of these tubes typically has a length of 3 inches, or 75 mm. The ear canal end of the tubing opens out into the ear canal itself, which, being wider than the tube, has an acoustic impedance less than that of the tube. The hearing aid end of the tubing system connects, eventually, to the receiver. Because the receiver is so small, it has a high acoustic output impedance (compared to the impedance of air in the tube). This means that, acoustically, the tube has one end almost open and one end almost closed.

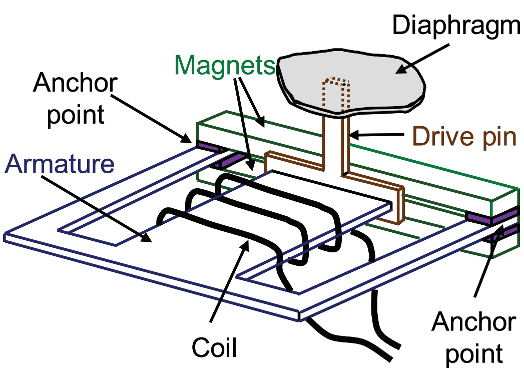

Figure 2.18 Principle of operation of the moving coil receiver.

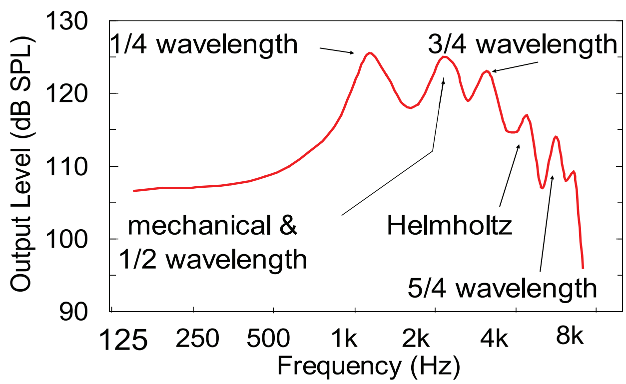

Figure 2.19 Frequency response of a receiver in a BTE hearing aid.

Such tubes have ***wavelength resonances*** at frequencies approximately equal to odd multiples of the speed of sound divided by four times the length of the tube (just like an organ pipe, an oboe or a didgeridoo).[j] This produces resonances at around 1 kHz, 3 kHz, and 5 kHz. The resonance at 4 kHz appears to be a ***Helmholtz resonance*** between the mass of air in the tube and the volume (and hence compliance) of air inside the receiver.[336] The bump at 2 kHz is primarily caused by the ***mechanical resonance*** of the receiver: the mass of the diaphragm and springiness, or compliance, of the diaphragm combined with the compliance of the air inside the receiver. Even this resonance, however, is affected by the tubing. At its resonant frequency, the receiver actually has a low acoustic impedance, so at this frequency the tube acts as though it is acoustically open at both ends. It then has a resonance at multiples of the speed of sound divided by twice the length of the tube. This occurs at about 2 kHz, so the second bump in Figure 2.19 is actually a resonance of both the receiver and the tubing.[336]

Figure 2.20 shows the frequency response typical in a receiver for an ITE, ITC or RITE hearing aid. There are only two peaks, one somewhere in the range 2.2 to 3 kHz, and one around 5 kHz. The first of these is the mechanical resonance in the receiver. It is often at a frequency higher than occurs in BTE hearing aids because ITE/ITC/RITE aids usually have a smaller receiver with a lighter, stiffer diaphragm. Different resonant frequencies in the hearing aid can be achieved by the designer selecting different model receivers and by changing the amplifier's output impedance. It is desirable for receivers to have a peak in the 2.5 to 3 kHz range because the unaided adult ear has a natural resonance in this frequency range (again the result of a quarter wave resonance, this time of the ear canal). Consequently, a receiver resonance at this frequency helps the hearing aid restore the natural resonance and gain that is lost when the hearing aid is inserted in the ear via any earmold that mostly closes the ear canal. The ear canal resonance is retained in hearing aids where the ear canal remains mostly open, which affects the frequency response of the receiver when measured in the ear canal (see Sections 4.2.1 and 5.3.6). The higher frequency peak around 5 kHz in ITE, ITC and CIC hearing aids is predominantly caused by a quarter-wave resonance of the receiver tubing.

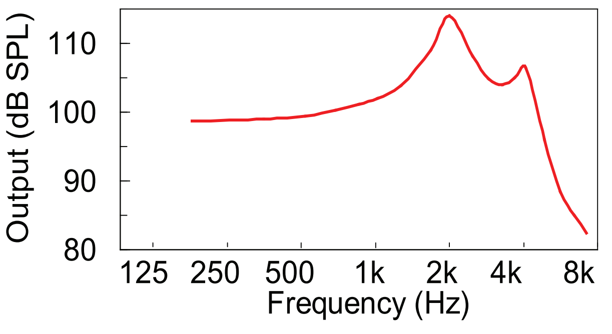

Figure 2.20 Frequency response of a receiver in an ITE, ITC or RITC hearing aid, connected to a 2 cc coupler via a tube 10 mm long and 1 mm inner diameter.

When coupled with suitable tubing (see Chapter 5) and dampers (next section), receivers can have a smooth, wide frequency response to 8 kHz or more that allows a very good sound quality to be achieved.[925] It is not yet possible, however, to achieve a flat response out to 8 kHz in a very high-power receiver.

2.7 Acoustic Dampers

Does it matter whether the receiver response has bumps and dips, and does it matter what causes them? The answer to both questions is yes! Peaks and troughs (especially peaks) in the gain-frequency response adversely affect both speech intelligibility and quality of the amplified sound.[404, 822, 1032, 1243, 1830] One way they decrease intelligibility is that the hearing aid inappropriately gives each amplified sound a peak at the frequency of the peak in the hearing aid response. In some cases, this inappropriate peak will change the identity of a sound to a different sound that truly has a peak at that frequency.[1681] Peaks become objectionable if they rise by more than about 6 dB above the smooth curve joining the dips.[454, 1830] Multiple peaks

[j] Wavelength resonances are created by the sound wave reflecting backwards whenever the tube it is travelling in changes its diameter, and hence its impedance. An open end and a closed end are extreme examples of a change in impedance. The tube with one open and one closed end is said to have a quarter wave resonance, because at the resonant frequency, the length of the tube equals one quarter of a wavelength. The resonant frequency therefore mostly depends on tubing length, but very thin tubing causes the resonant frequencies to decrease slightly.

more adversely affect quality when they are spaced apart by about an octave (which is about their spacing in hearing aids!) than when they are much more closely or widely spaced.[1243] Peaks caused by the receiver and tubing affect the shape of the maximum output curve of the hearing aid just as much as they affect the shape of the gain-frequency response. Such peaks make it unnecessarily difficult to get all sounds loud enough without some sounds becoming excessively loud (Section 10.7).

Understanding the cause of the peaks and troughs is important because the cause of a peak determines how its size can be decreased. Placing an acoustic resistor, also called a damper, in the tubing at an appropriate place decreases the peaks. One type of damper consists of a fine mesh (like a fly screen designed to stop *extremely* small insects) inserted across a small metal cylinder or ferrule, as shown in Figure 2.21. As the particles of air move backwards and forwards, in response to the sound wave in the tube, they lose energy when they have to change course slightly to avoid the wires in the mesh so that they can flow through the holes in the mesh. The more quickly the particles are flowing, the more energy they will lose when the mesh is added. In a tube, the particles flow most quickly:

- at the resonant frequencies;
- at the open end of the tube; and
- at any location a half-wavelength away from an open end.

Thus, a damper will decrease the receiver output most at the resonant frequencies, but only if the damper is placed in an appropriate place (Section 5.5).

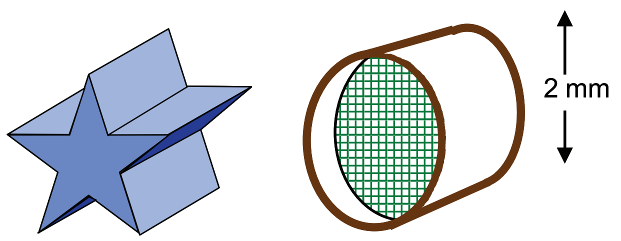

Figure 2.21 A star damper and a fused-mesh damper that can be inserted inside #13 tubing of internal diameter 1.93 mm.

Apart from the fused-mesh dampers just described, dampers are also made from sintered stainless steel (fine particles of metal). Another variety, made of plastic, looks like a star-shaped prism, and is known as a star damper. Dampers are also made from lamb's wool and from plastic foam. The degree to which a damper decreases resonant peaks depends on the impedance of the damper, which is determined by the fineness, length, and number of air paths through the damper. Fused-mesh dampers and sintered-steel dampers are available in a range of standard impedances. The impedance of star dampers, lamb's wool dampers, and foam dampers is varied by using different lengths of the material.

Dampers can be placed in the tubing connected to a receiver or, in a few hearing aids, in the inlet port of the microphone. Some receivers have dampers built-in when they are manufactured. Damping in different places to achieve specific effects will be covered in Section 5.5.

2.8 Telecoils

A ***telecoil*** is a small coil of wire that produces a voltage when an alternating magnetic field flows through it. The magnetic field to be picked up by the telecoil is generated by an electrical current that has the same waveform as the original audio signal. This magnetic field may occur as a by-product of some device, such as from a loudspeaker or a receiver in a telephone, or may be generated intentionally by a loop of wire around a room or other small area. The process of an electrical current inducing a voltage in a coil some distance away is called ***induction***. Induction loop systems are discussed in more detail in Section 3.5.

To increase the effectiveness of a telecoil, the wire is coiled around a rod made of ***ferrite*** material. Like iron, but even more so, ferrite provides an easy path for magnetic fields to flow through. It thus "attracts" and concentrates the ***magnetic flux***. If more flux flows through the coil, then more voltage is generated by the coil, which is desirable so that the audio signal is large compared to the internal noise generated by the hearing aid. Coils can also be made more sensitive by increasing the cross-sectional area or by increasing the number of turns, but both of these increase the physical size of the coil, and hence of the complete hearing aid.

Not all hearing aids include a telecoil, although nearly all high-powered BTE hearing aids and many other

BTE and ITE hearing aids do. The hearing aid user can select the coil, instead of the microphone, for amplification by switching the hearing aid to the ***T*** (for Telecoil) position. Most hearing aids now have a program selector switch, rather than a separate *M-T* switch. The telecoil is selected by switching to the ***telephone program***. In other programs, (or on older hearing aids, in the ***M*** position) only the microphone is connected to the hearing aid amplifier. Some hearing aids can be set up to have a program in which the microphone and coil provide signals at the same time, which is also achieved in the *MT* position of an older hearing aid, as shown in Figure 2.22. The *MT* combination is useful if the aid wearer wants to receive both the acoustic and magnetic signals simultaneously or in quick succession, but has the disadvantage that any acoustic noise present will be amplified even if the aid wearer is trying to listen only to the magnetic signal.

When the hearing aid is switched to the *T* program and the input comes from a room loop, the only sounds amplified will be magnetic signals reaching the hearing aid. Although this is also true when the input comes from a telephone, the microphone of the telephone will pick up any sounds reaching it, and the telephone side-tone will cause all these sounds to emerge as magnetic signals, which the hearing aid will sense and amplify. To reduce local noise while listening to the telephone in the *T* program, the hearing aid wearer should therefore cover the *telephone* microphone (when not speaking!).

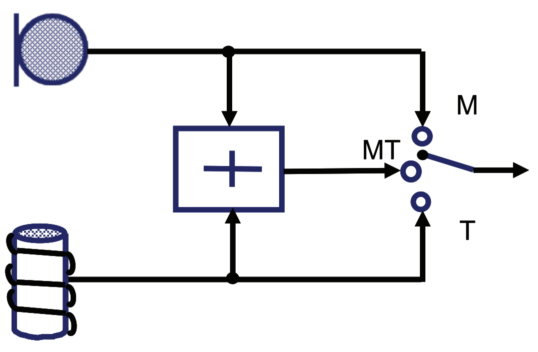

Figure 2.22 Block diagram of the input stage of a hearing aid with M, T, and MT selector switch.

2.9 Audio (Electrical) Input

An alternative way to get an audio signal into a hearing aid is to connect it via an electrical cable. This is referred to as ***direct audio input (DAI)***. The electrical audio signal may have originated from equipment such as an MP3 player, a hand-held microphone, or an FM wireless receiver (Sections 3.6.3 and 3.11.1). If the device producing the signal is itself receiving, or has previously recorded, a clear signal (i.e. with little added noise or reverberation) the device should also be able to output a clear signal to the hearing aid. Provided the signal put into the hearing aid is not so large that it overloads the hearing aid, and not so small that it is obscured by noise generated within the hearing aid, then the hearing aid too will be able to output a clear signal. Furthermore, the hearing aid will be able to shape the signal in the right way for the individual aid wearer: the frequency response, maximum output, and other amplification characteristics applied to the signal will be just as if the hearing aid microphone picked up the sound directly.

In fact, the direct audio input connector is normally connected into the same part of the hearing aid (the input amplifier) to which the hearing aid microphone is connected. This means that the size of the signal should be about the same as for signals sent by the microphone, which is about 1 mV for typical input levels. In some hearing aids, the input connector and the microphone are simply connected together. In older hearing aids, they are connected via a physical switch, so that the user can select either the microphone input, the audio input, or a mixture of both. More commonly, a single pushbutton on the hearing aid or remote control cycles through programs, one of which connects the DAI signal to the amplifier. Alternatively, the hearing aid automatically selects the DAI signal when a DAI connector or "boot" is attached to the hearing aid.

2.10 Remote Controls

Remote controls serve the same function for hearing aids that they do for televisions or video players: they allow the user to vary the way a device works without having to actually touch it. The advantage of a remote control for hearing aids is primarily one of size. Because hearing aids are so small, it is difficult to fit many, or sometimes any, user controls on them. Also, because a hearing aid is located in or behind the ear, the user cannot see the controls, and so may have trouble locating a control, particularly if the hearing aid does indeed have more than one.

Buttons on the remote control are easier to operate than those on the hearing aid, partly because they are larger, and partly because the user can look directly at the controls while they are being operated. Alternatively, some users like to operate the remote control while it is in their pocket because it does not draw attention to the hearing aid, as can occur when the aid itself is manipulated.

A remote control usually has a volume control. It may also enable the user to select an alternative program or programs (Section 3.3.2). Other features that are commonly provided on remote controls include selection of telecoil, electrical audio input, directional versus omni-directional microphone response, tone control, and on-off switch.

Remote controls work by transmitting signals to the hearing aid. These signals can have any effect on the aid that a switch actually located on the hearing aid could have. Various methods of transmission are used in hearing aids currently available. Some of these are explained in more detail (in the context of transmitting audio signals) in Chapter 3.

Infra-red. This uses the same technology as used for television remote controls, and transmits an infra-red light wave. The remote control must be within "sight" of the hearing aid and pointed towards it. The hearing aid contains an infra-red detector on its exterior. Bright sunlight (rarely a problem in Europe, it seems) may interfere with transmission.

Ultrasonic. The remote control transmits an acoustic wave too high in frequency to be heard (by humans), but which can be received by the hearing aid microphone. It also requires line-of-sight operation.

Radio wave. An electromagnetic radio wave is transmitted by the remote and received by a small aerial within the hearing aid.

Magnetic induction. The control signals are transmitted from the remote to the hearing aid by creating a magnetic field at a frequency above the audible range. The hearing aid receives this using either a special purpose coil or the same telecoil that receives audio magnetic signals.

Each of these methods has its advantages,[1865] as summarized in Table 2.2. Magnetic induction and radio wave remote controls have largely or completely replaced ultrasonic and infrared remote controls.

A concern that is sometimes raised about the use of remote controls is their potential to interfere with ***pacemakers***. Because pacemakers are designed to sense small voltages, it is very sensible to be concerned about interference from sources of electrical or magnetic energy. However, because remote controls put out only a small amount of power (compared, for example, to a mobile phone), it is likely that they can be operated safely in close proximity to a pacemaker. Radio waves are attenuated by the body by an amount that increases with frequency, but magnetic induction fields are not attenuated at all by the body. Because there are multiple brands of pacemakers and multiple brands of remote controls, it is difficult for manufacturers of either type of device to give any general guarantees about freedom from interference. Hearing aid manufacturers may be able to provide minimum safe distances for their specific remote controls for

Table 2.2 Advantages of different remote control technologies. *Interference* refers to interference of remote control operation by other devices.

	Ultrasonic	Infrared	Radio Waves	Magnetic Induction
Freedom from interference		✓		
Operated from any position			✓	✓
Simultaneous bilateral operation			✓	✓
Simple technology	✓	✓		✓

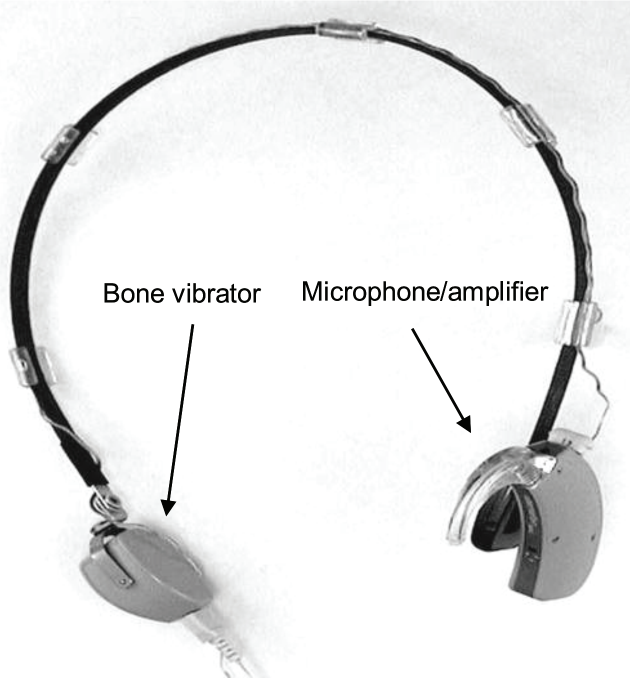

Figure 2.23 A bone conductor hearing aid

pacemakers that have immunity to interference conforming to IEC601-1-2.

Experimental evaluations have failed to show any interference of pacemaker operation by a remote control, but have shown that a remote control can interfere with active pacemaker programming and with telemetry read-out of a pacemaker.[1489] If a remote control has to be used for someone with a pacemaker, a cautious approach would be to preferentially select one based on infrared or ultrasonic transmission (if available), or failing that, very-high frequency radio transmission, unless the manufacturer of the specific pacemaker or the manufacturer of the specific remote control can provide an assurance that interference to the pacemaker is not possible.

2.11 Bone Conductors

Bone conductors are an alternative output transducer intended for people who, for various reasons, cannot wear a receiver coupled to the ear canal. Bone conductor transducers directly vibrate the skull, which via several transmission paths, transmits these vibrations to the cochlea. The bone conductor works on a similar principle to the receiver. One difference is that instead of a vibrating light diaphragm, it has a heavy mass that is shaken by the audio current passing through a coil. The inertia of this mass causes it to resist being shaken, so the case of the vibrator shakes in the direction opposite to that of the mass. It is this vibration of the case that is transferred to the skull. For efficient transfer of power, the transducer has to be held firmly against the skull by means of a tight headband or spectacle frame. This force, combined with the small contact area of some bone conductors, creates sufficient pressure on the scalp that the devices are uncomfortable to wear.[k] Continued wearing can create a permanent indentation in the skull of an aid wearer. Fortunately, high-output, low-distortion vibrators with larger contact areas that spread the force out sufficiently to produce safe contact pressure, are becoming available. Bone conductor transducers require considerable power, so they are usually driven by high-powered hearing aids. The hearing aid amplifier output is connected to the bone conductor transducer, instead of its usual receiver, by wires emerging from the hearing aid or by a plug and socket arrangement, as shown in Figure 2.23.

2.12 Batteries

The battery is the source of the increased signal power that the hearing aid delivers to the aid wearer. The important characteristics of the battery are its ***voltage***, its ***capacity***, the ***maximum current*** it can supply, its ***electrical impedance*** and its ***physical size***.

2.12.1 Principle of operation

Batteries (which are really called ***cells***)[l] generate electricity by putting two different materials (called the ***electrodes***) in close proximity in a medium (called the ***electrolyte***) that conducts electricity in the form of ions. Charged particles are attracted from one of the materials to the other via the electrolyte, and this can continue only if electrons can get from one electrode to the other via an external electrical circuit. This external current of electrons, is of course, the current

[k] The pressure significantly exceeds the closure pressure of blood capillaries, so the skin, and possibly bone, underneath the vibrator loses its blood supply, which causes the tissue loss after sustained use.[1474]

[l] Formally, a battery is a number of cells connected together to give a higher voltage or current, although the terms are used interchangeably in everyday use, and this book will refer to cells as batteries.

that the hearing aid amplifier makes use of. The process continues until one of the electrodes is used up, in that it can no longer supply charged particles and electrons: the battery is "flat".

2.12.2 Operating voltage

The voltage generated by a battery depends solely on the type of materials used for the electrodes. The batteries most commonly used for hearing aids use Zinc and Oxygen as their negative and positive electrodes, respectively, so the batteries are known as ***Zinc-air*** batteries. These batteries, whatever their physical size, generate approximately 1.4 Volts when not connected to anything and approximately 1.25 V when in use. When the zinc is close to being depleted, the battery voltage drops suddenly, and the hearing aid gets weaker, more distorted, and eventually ceases to operate once the voltage becomes too low. Few hearing aids will operate well once the battery voltage drops below about 1.0 V. Some hearing aids become unstable when the battery is near the end of its life, and the hearing aid generates and emits a low-frequency tone or noise. This can sound like a motor-boat, and the phenomenon is called motor-boating. Such sounds can be thought of either as a fault, or as a useful indicator that the battery is nearly dead! Most hearing aids sense the voltage and intentionally generate a sound to warn of the impending death of the battery. Some hearing aids automatically switch off once the voltage drops below the minimum voltage needed for the digital circuits to operate properly. They may also automatically decrease the hearing aid OSPL90 as the voltage approaches this limit so that the hearing aid gets gradually weaker, rather than stopping abruptly.

Other combinations of materials that are sometimes used in hearing aids are Mercuric Oxide and Zinc, which generate 1.35 Volts. Still available, but rarely used, are batteries comprising Silver Oxide and Zinc which generate 1.5 Volts. Body-level hearing aids use larger batteries, such as AA or AAA size. These have Manganese Dioxide and Zinc as their electrode materials and also generate 1.5 V. Other batteries sometimes used in body aids use Lithium instead of Zinc as their negative electrode and one of several materials as the positive electrode. These are more expensive, have a higher capacity, and generate 3 V.

2.12.3 Capacity and physical size

Batteries last longer the more electrode material they contain. Bigger batteries therefore last longer than smaller batteries with the same chemistry. The electrical ***capacity*** of a battery is measured in ***milliamp hours (mAh)***. A battery with a capacity of 100 mAh, for example, can supply 0.5 mA for 200 hours, 1 mA for 100 hours, or 2 mA for 50 hours. There is an upper limit to how much current a battery can supply at any instant. If the current gets too high, even for a frac-

Batteries: practical tips

- The sticky tabs on zinc air batteries restrict air from getting in to the zinc electrodes. The battery will not operate until the tab is removed, but once it is removed, the battery has a shelf life of only a few weeks.
- If the sticky tab was too well sealed to the battery, the battery will not be useable until the air has had time to percolate into the battery. This can be speeded up by leaving the battery a few minutes before putting it into the hearing aid.
- If a new battery appears to be dead, leave it for a few minutes after removing the tab - it may make a miraculous recovery!
- If a hearing aid is left unused for a period (weeks or months), the battery should be removed to protect the hearing aid (especially the battery contacts) from potential battery leakage and corrosion.
- For a high-powered hearing aid, it is worth investigating the battery life and sound quality obtainable with an HP battery

tion of a second, the battery voltage will drop excessively because of the internal resistance of the battery. Momentary intense noises will therefore cause the voltage to momentarily decrease, perhaps so much that the hearing aid temporarily ceases to operate, giving a very distorted sound. Bigger batteries can generally supply bigger maximum currents, as well as having a larger mAh capacity. AA and AAA batteries have typical capacities of 2000 and 800 mAh, respectively.

High-powered hearing aids need the greatest current, and some batteries are advertised as being more able to supply the high currents these hearing aids need without losing too much voltage. These are referred to as ***HP*** (***High Performance*** or ***High Power***) batteries, and may have the prefix *H* in their name. These are also zinc-air cells, but have bigger holes to allow oxygen in at a faster rate and use an electrolyte that causes less voltage drop during high current demand. HP batteries should give a longer life than a standard battery if the hearing aid has a high peak-current demand, but will give a shorter life if the hearing aid has a low peak-current demand. HP batteries should be used if a hearing aid draws more than about 8 mA (for a size 13 or 312 battery) or more than about 18 mA (for a size 675 battery) when the hearing aid is saturated.[1152] Peak-current demand can be assessed in a test box equipped with a battery pill, by applying a 500 Hz signal at 90 dB SPL with a high volume control setting. Read the current as soon as you switch on the sound. For hearing aids intended for open-fit applications, also test with a higher frequency, say 2000 Hz, as the hearing aid may be configured to give no output for low-frequency sounds.

Table 2.3 shows the capacity of good zinc-air batteries of various sizes. Some brands claim greater capacity than those shown, and some have less capacity. Mercuric oxide batteries have capacities about half that of zinc-air batteries of the same size. Two labeling systems are used: the labels in the first column are most common, while those in the second column are specified in international standards.[795] Figure 2.24 shows each of the batteries drawn full size.

Zinc-air batteries are the preferred type of non-rechargeable cell because they are the cheapest (per mAh), they do not have to be changed as often as mercury or silver batteries, and they have less adverse environmental consequences than mercury batteries when they are discarded. Mercuric oxide batteries are now rarely used because the hazardous nature of mercury when discarded, and especially if ingested. In the past, all zinc-air batteries contained small amounts of mercury, but mercury-free varieties are increasingly available and may colloquially be called green batteries.

Table 2.3 Names and typical capacities of zinc air batteries of various sizes and the hearing aid styles in which they are most commonly used.

Type	Standard Label	Capacity (mAh)	Hearing aid types
675	PR44	600	BTE
13	PR48	300	BTE, ITE
312	PR41	175	BTE, ITE, ITC
A10 (or 10A, or 230)	PR70	90	BTE, CIC
A5	PR63	35	CIC

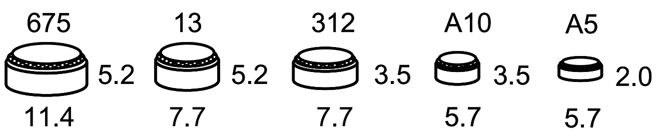

Figure 2.24 Hearing aid batteries of various types drawn full size, with typical dimensions shown in mm. Minimum and maximum allowable dimensions are 0.1 to 0.2 mm smaller and larger than these dimensions.

2.12.4 Rechargeable batteries

Some manufacturers offer hearing aids with ***rechargeable batteries***. Their major advantage is increased convenience from not having to change batteries. Rechargeable cells can be discharged and recharged several hundred times, so the battery only has to be replaced every one to three years. Many elderly hearing aid wearers find that changing batteries is the most difficult part of wearing hearing aids, because of the fine manipulation skills required. Rechargeable cells avoid this inconvenience, and also avoid damage to battery drawers resulting from attempted insertion of cells in reverse polarity. The major disadvantages of rechargeable cells is that their capacity is only around 10% of that of a non-rechargeable cell of the same size, so that recharging must be performed regularly, as often as every night, which requires some basic discipline from the user. Another issue is the increased power consumption of wireless technology. In many cases rechargeable batteries will not be able to provide sufficient current when the wireless feature is enabled, and/or the discharging/re-charging cycle will be too short.

Rechargeable cells used in hearing aids (and many other electronics devices) are most commonly Nickel Metal Hydride (NiMH) construction (Nickel hydroxide for the positive electrode and a metal alloy hydride for the negative electrode). They generate 1.2 V, which remains relatively constant as the cell is discharged. Less commonly now, body aids running on rechargeable AA or AAA batteries may use Nickel and Cadmium as their electrodes (NiCad), which generate 1.3 V. These have lower capacity than NiMH cells of the same size, more rapidly lose capacity if they are not fully discharged each cycle, and the Cadmium creates toxicity when disposed of. NiCad cells should be fully discharged regularly to prevent the loss of capacity that can occur if the battery is repeatedly charged and partially discharged (referred to as the memory effect).

A potential advantage of rechargeable batteries is that they can be recharged from a solar cell, making them especially suitable for places in the world where there is neither a reliable supply of disposable batteries nor reliable mains electrical power. Both body aids (with the solar cell on the outside of the case) and BTE aids (with the solar cell on a separate charging unit that recharges the hearing aid overnight) are available.[1388]

In the future, fuel cells, running on hydrocarbons like methanol, may replace current batteries. They are rechargeable, just by replacing the expended methanol.

2.13 Concluding Comments

Although the major components in hearing aids (transducers, amplifiers, and batteries) have existed in some form for over a century, there has been a dramatic improvement in their quality and a dramatic reduction in their size over this time. These technological advances have enabled hearing aids to provide amplification in increasingly sophisticated and effective ways.

The advent of digital technology has revolutionized the *methods* hearing aids use to change sounds. In some respects, however, the *result* (the sound coming out of the hearing aid) is no different from the sound emerging from analog hearing aids. Fortunately, there are several ways, as we will see in Chapters 7 and 8, of how digital hearing aids can modify sounds in ways that were never possible with analog hearing aids.

In the following chapter, we will see how the individual components are combined to provide complete hearing aids and amplification systems.

CHAPTER 3

HEARING AID SYSTEMS

Synopsis

Components can be combined into hearing aids in an extremely customized manner, such that individual components are selected for each patient and are located in the position that best suits each ear. At the other extreme are modular aids, including some ITC hearing aids, and all BTE hearing aids, that are prefabricated in a totally standard manner. Many hearing aids fall somewhere between these extremes.

Increasingly, the hearing aids on each side of the head communicate with each other by wireless transmission so that their amplification characteristics (directionality, noise reduction processing, compression characteristics and input source) can remain coordinated as the environment changes or as the user varies a control. In some cases, the complete audio signal is transferred from one side of the head to the other, which enables a telephone signal to be heard in both ears, and will enable super-directional microphones to be developed.

Hearing aid amplification characteristics are programmed from a computer, via a suitable wired or wireless interface, to suit the hearing capabilities of each user. Often, more than one program is put into the hearing aid so that different amplification characteristics can be selected, automatically by the hearing aid or manually by the user, in different listening conditions.

The most effective way to make speech more intelligible is to put the microphone near the lips of the person talking. This markedly decreases noise and reverberation, but requires a means of transmitting the signal from the microphone to the hearing aid wearer some distance away. Methods to do this currently include (1) magnetic induction from a loop of wire to a small telecoil inside the hearing aid, (2) radio transmission of a frequency-modulated, or digitally modulated electromagnetic wave, (3) infrared transmission of an amplitude-modulated electromagnetic wave, and (4) acoustic transmission of an amplified sound wave. Each of these systems has strengths and weaknesses compared to the others. The first three offer a very large potential increase in signal-to-noise ratio, and hence intelligibility. It can, however, be a challenge to adjust the hearing aid and wireless system together so that both the wireless system and the hearing aid individually provide maximum benefit to the wearer without the wireless input and microphone input signals interfering with each other. Increasingly, wireless receivers are being built into hearing aids, considerably improving cosmetic appearance and convenience. A major application of these systems is to make teachers more easily understood in classrooms, but they can be used by children and adults in other situations as well.

Wireless reception is also enabling hearing aids to conveniently accept electrical signals from a range of audio devices, including televisions, MP3 players, computers and mobile phones. In many cases, this connection requires an intermediary wireless relay device, as the current consumption of the ubiquitous Bluetooth receivers and transmitters precludes them from being directly built into hearing aids. While there have been problems with mobile phones causing interference in hearing aids, this problem is decreasing due to improvements in hearing aid design and changes in the mobile phone transmission system. Use of a mobile phone via hearing aids is now often trouble-free. There is the potential for hearing aids to become the audio portal to the world, and possibly not just for hearing-impaired people.

Assistive listening devices enable hearing aid wearers to receive sounds other than just by the amplification provided in a self-contained hearing aid. Assistive listening devices include the transmitter/receiver pairs already described for remotely sensing and sending speech or music, and devices that alter sound at its source (such as a telephone amplifier). Other types of assistive listening devices enable the aid wearer to detect alerting sounds (e.g. the doorbell, a telephone ring, a smoke alarm). Some do this by transmitting the sounds wirelessly to the hearing aids; others convert sound to other sensory modalities (such as flashing lights or vibrating shakers).

The long-established distribution and fitting system for hearing aids is being somewhat challenged by over-the-counter hearing aids, and their more modern cousin, hearing aids sold over the internet, and even by disposable hearing aids.

Chapter 2 described all the bits and pieces that go to make up a hearing aid. This chapter will describe how these bits and pieces are combined to make complete hearing aids, including hearing aid systems that transmit and receive signals across a distance, including in the classroom.

3.1 Custom and Modular Construction

The basic styles of hearing aids (body, BTE, ITE, ITC, RITC and CIC) have already been introduced in Section 1.3. The ITE, ITC and CIC styles can be completely ***custom-made*** for the individual hearing aid wearer. Alternatively, any of the styles can be manufactured in totally standardized shapes and sizes, which is referred to as ***modular*** construction. Hearing aids can also be constructed in an intermediate way, which is referred to as ***semi-custom*** or ***semi-modular*** construction. The following sections will describe the differences between each of these construction techniques.

3.1.1 Custom hearing aids

Custom hearing aids (ITEs, ITCs, and CICs) take full advantage of the size and shape of an individual aid wearer's ear. Construction begins when the clinician makes an ***ear impression*** and sends it, or a laser-scanned image of it, to the hearing aid manufacturer. The manufacturer, through either a casting or a laser-scanning technique, uses the impression to make a hollow ear shell that fits snugly within the ear canal and (if appropriate) the concha of the aid wearer.

Customization of the hearing aid components happens to different degrees. Construction is based around the ***faceplate***. This is a flat or contoured sheet of plastic that is trimmed to size and becomes the outer surface of the hearing aid. Most commonly, the faceplate will come from the factory with the amplifier board, microphone, volume control, battery compartment, telecoil and switch (if appropriate) already assembled. The receiver is likely to be loosely attached to the remaining components, because its position relative to the other components has to be adjusted for each custom aid to make the best use of the space available within each ear. Figure 3.1 shows a faceplate with all the other components attached, next to the ear shell to which it is about to be fitted.

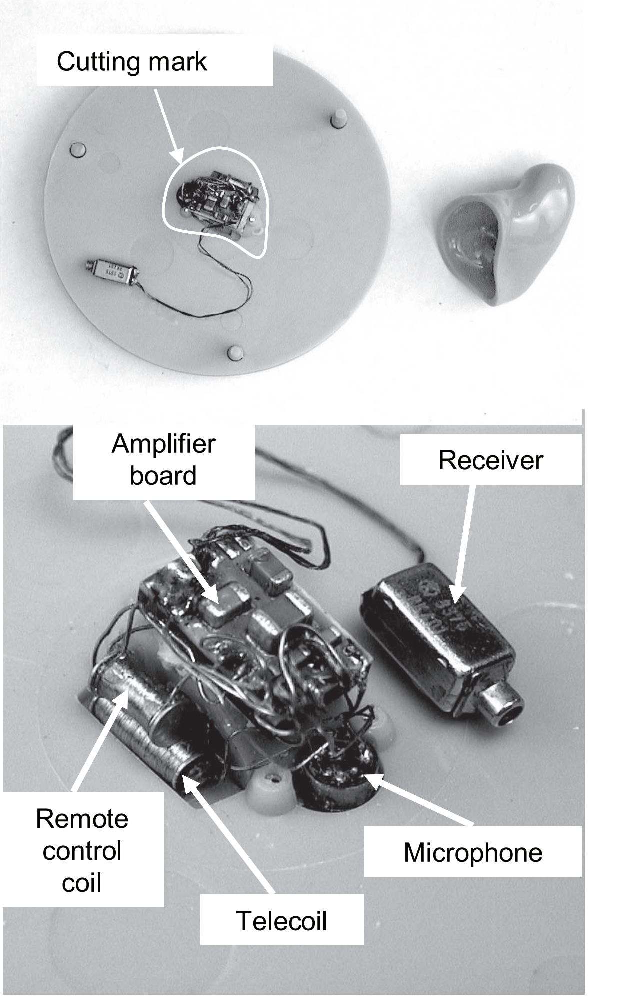

Figure 3.1
(Top) A faceplate with components mounted, next to a shell for an ITC hearing aid. After gluing the two parts together, all material outside the cutting mark is removed.
(Bottom) A close-up of the components mounted on the faceplate.

Sometimes, one or more of the other components are also not positioned or attached until the individual hearing aid is being constructed so that they too can be positioned to take advantage of the shape of the individual ear. This enables as much material as possible to be cut from the outer part of the shell before the faceplate is attached, which makes the aid as small as possible for the components chosen. Finally, the faceplate is glued to the shell and any excess trimmed off. This raises the obvious problem of how repairs are carried out on these aids. Most repairs require the faceplate to be cut away from the shell, which can usually (but not always) be done without damage to either part.

3.1.2 Modular hearing aids

Modular ITE/ITC/CIC hearing aids are those in which the hearing aid components are manufactured as a standard package. ITC hearing aids, in particular, have been made in a variety of cases having standard shapes. These can be thought of as *ready-to-wear* hearing aids, and physically fitting these aids to the ear merely comprises selecting the case with the shape that best matches the person's ear canal and concha.

Modular hearing aids have several advantages and disadvantages. First, the module can be manufactured and tested in a more automated manner, which lowers cost and increases reliability. Second, it can be attractive for the client and the clinician to be able to fit a hearing aid as soon as a hearing assessment has been carried out, rather than having to take ear impressions and complete the fitting at a later appointment. The disadvantages can be severe. For many ears, there may be no standard case that fits in a cosmetically or functionally acceptable manner. The aid may fall out too easily, or there may be so much leakage around the case that the hearing aid whistles at gain settings lower than those needed by the aid wearer for adequate audibility.

Low-cost mail-order hearing aids are, by necessity, modular devices. Some modular hearing aids have a foam sleeve around the canal section of the aid, which reduces the problem of a loose fit and feedback oscillation. The disadvantage is that the foam material deteriorates and regularly has to be replaced.

Another type of modular hearing aid that appeared during the early part of this century was the ***disposable ITC*** hearing aid, which was succeeded by a ***disposable BTE*** hearing aid. These hearing aids contain an embedded cell (i.e. battery), and the complete hearing aid must be replaced when the cell fully discharges. These devices have an extremely low cost (per device, but not per year) combined with good quality electronics and transducers, and the ITC version was available in a range of gain-frequency response shapes. Their frequency response, internal noise, and distortion compared favorably with much more expensive hearing aids.[1242, 1882] These performance characteristics make it clear that the high cost of hearing aids is more related to the high development costs, low manufacturing volumes, and high marketing, distribution, fitting and support costs, than to the cost of device production. Like other modular hearing aids, comfort and retention for the disposable ITC version were not as good, on average, as for custom hearing aids.[1882]

BTE and body-level hearing aids could also be called modular hearing aids, as the electrical and mechanical components have a fixed size and shape, which are then connected to either an individual earmold or a standard (i.e. pre-manufactured) dome located in the ear canal. When used with a standard dome, either as a tube fitting or with a receiver in the ear canal (RITE), BTEs are completely modular and can therefore be fit to the client on the same day as the client's hearing is assessed. BTEs are not, however, usually referred to as modular aids, because no one has yet invented any way to make a custom BTE aid (apart from the custom earmold).

3.1.3 Semi-modular, semi-custom hearing aids

ITE or ITC hearing aids that combine a standard module with a custom-made ear shell can be referred to as semi-custom or semi-modular hearing aids. The modules are usually clipped, rather than glued, to the individual ear shell, which makes repairs faster, cheaper and unlikely to damage the earshell or faceplate, as can occur in a custom aid. The disadvantage is that because the components cannot be rearranged to take advantage of the individual ear's geometry, a semi-modular hearing aid will generally be larger than a custom hearing aid with the same components.

There is a continuum from fully custom to fully modular aids. At the fully custom extreme, the position of any major component relative to any other major component can be varied, and the manufacturer individually selects some of the components for each hearing aid wearer. At the fully modular extreme, the entire hearing aid is manufactured in a totally standardized manner (such as the ready-to-wear aids mentioned in the preceding section).

Most CIC/ITC/ITE hearing aids sold lie closer to the fully custom extreme. They typically combine a glued-on faceplate with the battery compartment, volume control, and programming socket (if not in the battery compartment) in a fixed position relative to each other. The microphone and integrated circuit are often also fixed in position relative to the faceplate. The receiver is individually positioned within a custom shell.

3.1.4 Hearing aid reliability

Hearing aids have to function in an adverse environment – rain, sweat, cerumen, hair-sprays, hair gels and humidity are all conducive to failure of the electronics or transducers, primarily through causing corrosion. The components that fail most are those exposed to the air and those that have moving parts. These comprise the battery contacts, transducers, volume controls and switches.

Several innovations in many hearing aids are enabling hearing aid reliability to be significantly increased. These features include:

- ***Automatic volume controls,*** including wide dynamic range compression, make it unnecessary for many hearing aids to incorporate a manual volume control.
- ***Electrical programming,*** instead of fitter-adjusted screw controls, reduces the number of components with moving parts.
- ***Water-repellant fabric,*** e.g. Gore®, covering the microphone inlet port, impedes the entry of water into the microphone.
- ***Water-proof membranes*** block the entry of moisture (even as vapor) and cerumen into the receiver, but allow sound waves to pass through unimpeded.[167]
- ***Gaskets and water-proof fabric*** enable the zinc-air battery to obtain the oxygen supply it needs to operate without allowing water into the battery compartment.
- ***Nanocoating,*** a lacquer containing nano-particles, makes the surface so smooth that water beads on the surface instead of spreading over it, making water entry through joins less likely.
- ***Swipe controls,*** that sense movement of the finger, don't require any moving mechanical parts or apertures that enable the ingress of moisture.

3.2 Linked Bilateral Hearing Aids

It has long been common to fit a hearing aid to both ears, but only this century have these two hearing aids been able to exchange information with each other so that they can coordinate the manner in which they operate. They achieve this exchange by wireless transmission from each device to the other.

Why might this be desirable? The reasons fall into two categories: convenience and performance. When a hearing aid wearer wishes to adjust the volume, most commonly he or she will want the volume of both hearing aids to vary. It is convenient if the aid wearer can adjust the volume control on just one hearing aid, and have the gain of the other hearing aid automatically vary in the same manner. As the second aid no longer needs its own volume control, the limited space available in it can be used for a program selector switch (see Section 3.3.2), which of course simultaneously changes the program in both hearing aids. Manipulation of the two controls is then much easier for the aid wearer.

The performance advantages of linked bilateral hearing aids are less marked but still worth having. Ensuring that the two hearing aids make the same choice of microphone directivity at any instant minimizes the likelihood that the hearing aids will distort the timing and level differences between the ears that people use to localize sounds (Sections 7.1.4 and 15.1.1). Co-ordinating compression and adaptive noise suppression in the two hearing aids minimizes the distortion of inter-aural level differences that will otherwise occur. Linked processing appears to slightly improve localization and naturalness,[1671] with the greatest benefit likely arising from the linking of directional microphone settings.[887]

Linked bilateral hearing aids can also make excellent decisions about when to switch the input from microphone to telecoil. A single hearing aid making its own decisions may mistakenly switch to telecoil simply because the person is walking past some piece of equipment that happens to be emitting a strong magnetic signal. Linked hearing aids, however, can compare notes on the strength of the signal they are receiving. A telephone held to one ear will produce a much stronger magnetic field in the closer hearing aid than in the more distant one. Such a disparity in magnetic field strengths is unlikely to occur for more distant sources, so the presence of a handset next to one hearing aid is much more reliably detected. The hearing aid near the telephone can then automatically switch to telecoil, and the microphone on the other hearing aid can be switched off, or left as normal, whichever the clinician has decided is more appropriate for that client.

Similarly, linked binaural hearing aids can better determine whether a strong tonal component in the output is being caused by amplification of a musical sound (in which case it appears in the output of both hearing aids) or by feedback oscillation (in which case it appears at a high level in the output of only one hearing aid). This comparison helps avoid the distortion that digital feedback cancellation systems (Section 8.2.3) can otherwise cause to musical sounds.

At the time of writing, many linked bilateral hearing aids exchange control information, such as volume settings, program settings, directional microphone settings, and information about how much gain a compressor is currently providing. These control signals have in common a very slow rate of change. Consequently, the information can be transmitted with a very low radio-frequency bandwidth, which requires very little power to be transmitted, and which in turn requires very little battery current.

Some hearing aids, however, transmit a full audio bandwidth signal from ear to ear. In some hearing aids, this is achieved by ***near-field magnetic inductive coupling***, like the audio magnetic induction that will be described in Section 3.5, but operating at much higher frequencies, such as in the 1 to 10 MHz range. In other hearing aids, electromagnetic radio-frequency transmission in a much higher frequency range (0.9 to 2.4 GHz) is used.[a]

Whatever technique is used, if a full audio bandwidth signal can be transmitted from one ear to the other, signals sensed on one side of the head can be transferred to the hearing aid on the other side of the head, enabling several useful modes of operation:

- Transferring the microphone signal on one side to the ear on the other side (after amplification) makes a wireless CROS hearing aid (Section 17.1).
- Unlike a conventional CROS hearing aid, the transfer can be in either direction, and the direction can change from time to time depending on where the dominant talker is situated. This is good for conversations in cars where sitting positions are fixed and conversation partners are limited (at least in number!) so the SNR is regularly much larger on one side of head than the other.[1496].
- Transferring a telecoil signal on one side to the other ear enables a telephone call to be heard in both ears simultaneously.
- Most excitingly, the signals from microphones on both sides of the head can be combined in each hearing aid to produce a greater degree of directivity to the front, or indeed any other direction. (Section 7.1.4).

At the time of writing, these full-bandwidth, ear-to-ear transmission systems consume sufficient battery current that it is necessary they be used only for a limited time, such as in the noisiest of situations, but this will likely change in the near future. The feature is exciting as it will facilitate the development of super-directional hearing aids that in many noisy places will enable people with mild or even moderate loss to understand speech more clearly than people with normal hearing.

3.3 Programming the Hearing Aid

3.3.1 Programmers, interfaces, and software

The clinician changes the contents of the digital control circuits using a programming device. Although there remain some small special-purpose programming devices that can be used with particular brands of hearing aids, hearing aids are almost always programmed via a computer. Virtually all manufacturers have adopted a common standard for storing data and sending information from computers to hearing aids. That standard is called ***NOAH*** (as in: "we are all in the same boat").

The NOAH standard specifies how common data (like the audiogram and age of the client) should be stored, and how information should be sent to and received from the hearing aid.[1515] To program hearing aids from different manufacturers, specific software provided by that manufacturer is needed. However, once the client's data has been entered, those data can be accessed from any manufacturer's program, so that potential fittings from different manufacturers can be compared.

Because the hearing aid has to be sent electrical signals different from those that computers can provide, an interface between the computer and the hearing aid is required. One such interface, a small box with suitable sockets, or the same circuitry incorporated into other test equipment, is called the ***HiPro*** (**h**earing

[a] The higher the frequency, the more efficiently it can be transmitted with a small aerial, but the more the signal is attenuated by the head as the signal propagates from one ear to the other.

instrument **pro**grammer) interface. The HiPro device is still in common use but there are now more convenient alternatives:

- The NOAH-link wireless interface connects via a cable to the hearing aid in exactly the same way as for the HiPro, but has a Bluetooth wireless connection to the computer. As the NOAHlink is worn around the neck of the patient, the patient is free to move around in an unrestricted manner while the hearing aid is being programmed (as long as he or she stays within wireless range of the computer - about 10 m). Fine-tuning of the hearing aid could therefore be done in any environment if the clinician uses a lap-top computer.
- The NOAH-link wireless interface can plug into the nEARCom, a hook-shaped device worn around the patient's neck, that contains a 10.6 MHz inductive transmitter/receiver module to send signals to and from the hearing aid. The transmitter-receiver modules are manufacturer-specific, but up to five of them can be inserted in the nEARCom at the same time.
- Several manufacturers have manufacturer-specific transmitter/receiver devices that plug into a USB socket in the clinician's computer and communicate wirelessly with a matching transmitter/receiver in the hearing aid. At least one uses a digitally modulated 2.4 GHz signal, which provides very fast communication with the hearing aid.

These three alternatives all enable the patient to not be tethered to the programming computer. The last two alternatives also have the advantage of not needing to plug programming cables into the hearing aids.

3.3.2 Multi-memory or multi-program hearing aids

The data sent to a hearing aid by the computer and interface is stored in a memory inside the hearing aid. If one set of data can give the hearing aid one set of performance characteristics (e.g. gain, frequency response, microphone directionality) then several sets of data can give the hearing aid several sets of performance characteristics. Each set of performance characteristics is called a hearing aid program. Either the wearer or the hearing aid itself can then switch between programs whenever appropriate. Why might a hearing aid wearer need to change programs? It is not as if a drama can be heard on one program and a comedy on another.

The first reason is that sounds entering the hearing aid can have acoustic properties that differ vastly from one environment to another. For optimal listening, the hearing aid should have different amplification characteristics in each environment. Of course, the hearing aid could sense the acoustic environment and automatically change the amplification characteristics (and many hearing aids do). It is possible, however, that the user can do a better job of selecting the optimal characteristics than an automatic circuit.

There is a second reason for needing more than one program, and an automatic circuit cannot satisfy this need. Depending on the circumstances (such as the level of interest in a particular talker, or the wish to listen to non-speech sounds), listeners sometimes wish to optimize intelligibility, and sometimes wish to optimize comfort. These goals can require different amplification characteristics.[864, 1886] An automatic circuit, no matter how smart, cannot know which of these (or other) listening criteria is most important at any given time.

In most multi-memory hearing aids, all of the parameters that can be adjusted in one program can be independently adjusted in the other program or programs. The user, at the press of a button, can thus access the sound qualities of two or more entirely different amplification characteristics if that is how the clinician programs the aid. Most commonly the listening programs are adjusted to be identical except for one or two key parameters, or maybe just the means of selecting different inputs, such as a telecoil, or FM or directional microphone. Methods for prescribing multiple memory hearing aids are covered in Section 10.6.

3.3.3 Paired comparisons

If hearing aids are given two or more programs, then they can rapidly be switched between two programs during the fitting process. This enables the hearing aid wearer to compare two responses in quick succession and state which is preferable. The clinician can use these preferences to fine-tune the response when the hearing aid is initially programmed and at any follow-up appointments. Procedures for doing ***paired comparisons*** are covered in Section 12.2.

3.4 Remote Sensing and Transmitting Hearing Aid Systems

When a sound wave travels away from its source its power spreads out over an ever-increasing area and so it gets weaker. This causes two types of sound quality degradation. First, the decreased level is more easily masked by background noise. Second, reflected sounds, in the form of ***reverberation***, add delayed versions of the original sound to the direct sound. Reverberant sound is smeared out in time and not surprisingly is much less intelligible than direct sound, particularly when the room has a long reverberation time. Noise and reverberation thus both cause intelligibility to diminish as the listener gets further from the source.

Figure 3.2 shows a typical variation of SPL with distance from the talker in a room. The ***critical distance***, also referred to as the ***reverberation radius***, is defined as the distance from the source at which the level of the reverberant sound equals the level of the direct sound. Beyond the critical distance, reverberant sound dominates. The larger the room, and the less reverberant it is, the greater is the critical distance. In classrooms, the critical distance is often in the range 1 to 2 m, and in a living room is typically a little less than 1 m.

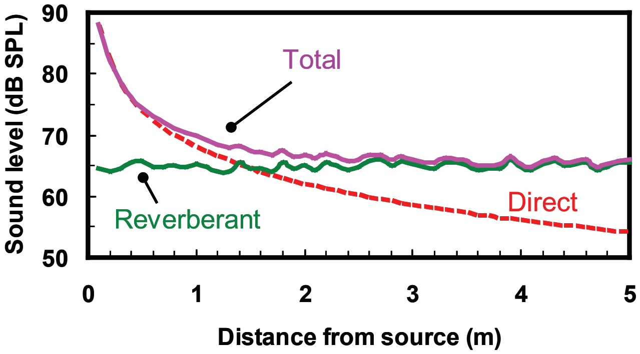

Figure 3.2 Variation of SPL with distance from a source within the room, showing the direct field, the reverberant field, and the total SPL. The critical distance, at which the direct and reverberant fields are equal in level, occurs at about 1.5 m in this room.

> **A critical concept in sound**
>
> The concept of ***critical distance*** is, well, *critical* to our understanding of why remote sensing and transmitting systems are needed, and to our understanding of how well directional microphones can work in different circumstances (Section 7.3.1).

Critical distance, in metres, can be calculated from the formula:[175, 1971]

$$d_c = 0.1\sqrt{\frac{QV}{\pi T_{60}}} \quad \ldots 3.1,$$

Where V is the volume of the room in m^3, T_{60} is the reverberation time of the room in seconds, and Q is the directivity factor of the source. A source with high directivity projects more sound forward than it does in other directions. The directivity factor, Q, for a human talker varies from around 1.3 in the low frequencies up to around 4.0 in the high frequencies.[113]

A solution to the problem of reverberation masking the direct signal is to pick up the signal where it is strongest and clearest (next to the talker's mouth, much closer than the critical distance), and transmit this strong, clear signal to a hearing aid wearer either as an electromagnetic wave or as a magnetic field, rather than as a sound wave. Provided the hearing aid wearer has the equipment necessary to turn the electromagnetic wave or magnetic field back into a sound wave, the wearer can hear the signal as clearly as though his or her ear was right next to the talker's mouth. There are three types of wireless transmission systems used to get the signal from the talker to the listener, and these are covered in the following three sections.

3.5 Induction Loops

There is an intimate connection between electricity and magnetism. ***Induction loops*** take advantage of this by converting an audio signal into an electrical current that flows through a loop of wire, and hence into a magnetic field that travels across space at the speed of light. This field is sensed by a coil of wire, and ***induces*** an electrical voltage in the coil (see Section 2.8). This voltage is then amplified and converted by a receiver back into sound. Figure 3.3 shows the complete path from talker to listener. The

Practical tips: Room loops

- Immediately above (or below) the wire forming a loop, the magnetic field flows almost horizontally. A vertical coil, as in a hearing aid, will not pick up much signal. Room loops therefore have to be a little larger than the area over which the loop has to work.
- Building steel near the loop can greatly weaken the strength of the magnetic field and change its direction.
- Purpose-designed loop amplifiers should include a compressor so that the magnetic field is always close to the optimum strength, even for soft talkers.
- Magnetic fields spill over outside the loop, so two loops with different audio signals in the one building should be well separated.
- Many home audio appliances have sufficient power to drive a small loop directly, but an additional, high-wattage volume control (and electronics expertise) will be needed if they are to drive the loop and a loudspeaker with an appropriate balance.
- Loops at floor level can run over doorways without adversely affecting performance.

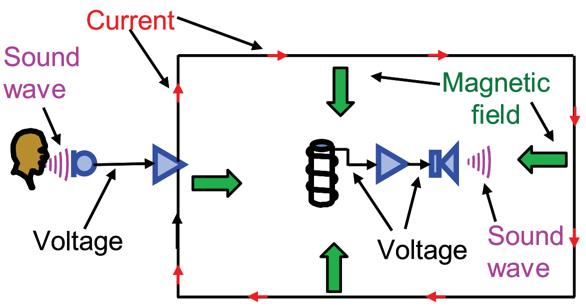

Figure 3.3 The complete chain, from sound wave in to sound wave out, for a magnetic loop induction system.

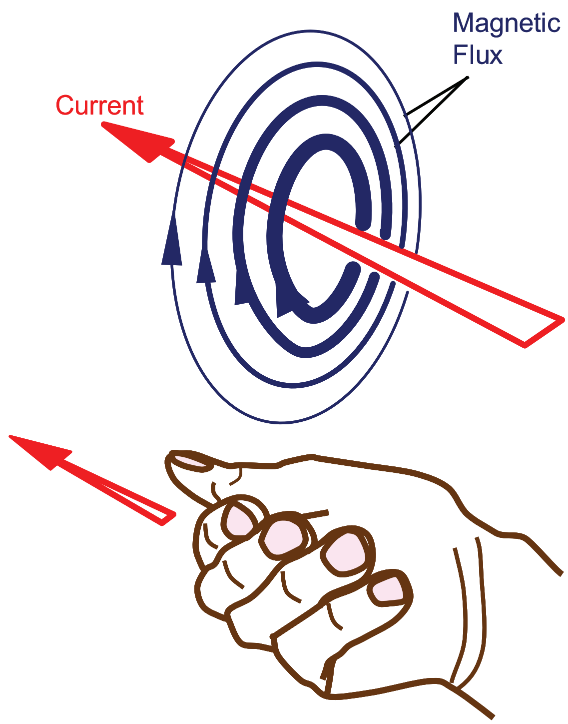

Figure 3.4 Lines of magnetic flux flowing around a conductor carrying a current and the right hand rule showing how the curled fingers help to visualise the direction of the magnetic field caused by current flowing in the direction in which the thumb points.

loop that emits the magnetic field can be as large as a length of wire around the perimeter of an auditorium or as small as a device that can fit behind the ear, alongside the usual hearing aid. In-between are loops that surround an individual listener (on the floor or attached to a favorite chair) and loops that are worn around the neck of the listener. The coil that picks up the magnetic signal is invariably mounted inside the listener's hearing aid, although many very small hearing aids do not contain such a coil.

3.5.1 Field uniformity and direction

Although magnetic fields emerge outwards from the wire and current that cause them, the ***magnetic lines of force***, and the resulting ***magnetic flux***, which can be thought of as the flow of magnetism, actually flow in circles *around* the current that causes the field, as shown in Figure 3.4. As the circles become more distant from the current, the magnetic force and the magnetic flux become weaker. To visualize the flow of magnetism, angle your right thumb at right angles to the fingers on your right hand (with your hand flat), and then curl your fingers. If your thumb points in the direction of the electrical current, the curled fingers will show the circular path taken by the magnetic field around the line of the thumb. (In fact, engineers call this the *right-hand rule* and use it to deduce which way around the circle the magnetism is flowing.)

Let us apply this to an imaginary loop on the floor in the room in which you are now sitting. Suppose the loop is hidden away in the corner where the floor meets the walls (as it often is in practice). Suppose the current is flowing clockwise around the room when you look down on the floor, as shown in Figure 3.5. Now put your thumb next to the imaginary wire next to one of the walls in the room. Notice where your curled fingers are pointing, and imagine the complete circles of magnetism (of various diameters) around each wire. Immediately above each wire, your fingers, and hence the magnetic field, should be pointing horizontally into the room. Within the room, at floor level, the magnetic field should be pointing straight down, no matter which section of the wire you think of as the source of the magnetism.

This last fact is very fortunate. Recall that as the circles get further from the wire, the magnetism gets weaker. As you move away from one section of wire,

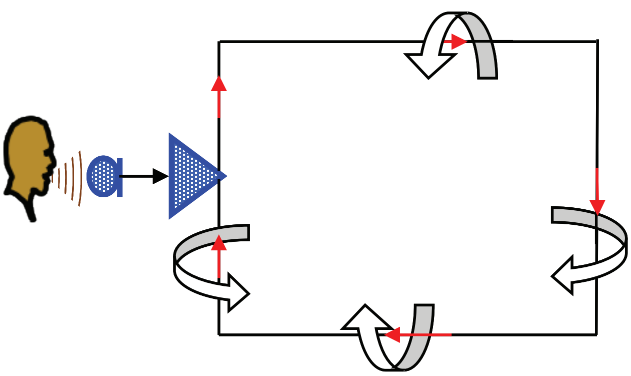

Figure 3.5 A complete induction loop system, showing how lines of magnetism from all parts of the loop add constructively within the region enclosed by the loop.

however, you are always moving closer to another section of wire. Also, if the wire is at floor level and the aid wearer is seated or standing, the magnetic field just inside the loop is more horizontal than vertical. (Use your right hand to see this for yourself.) The result of all this is that at head height, the vertical part of the magnetic field has a nearly constant strength over most of the room, except for just inside the loop, where the total field is strong, but the vertical part is weak. This is important because if the receiving coil in the hearing aid is mounted vertically, it will pick up only the vertical part of the magnetic field.[b] To work optimally with room loops, the telecoil in a hearing aid should therefore be oriented vertically when the hearing aid is worn. Unfortunately, the magnetic field generated by a telephone held near the ear may well have its maximum strength in some direction other than vertical, and this direction varies from phone to phone. As the footnote indicates, significant (but not extreme) deviations from vertical are acceptable.

Although we have talked about the current going around the loop in one direction, if the source is an audio signal, the direction will reverse many times per second, corresponding to the positive and negative pressure variations in the original acoustic wave. Consequently, the circular magnetic fields will also reverse their direction many times per second. It is actually this constantly changing magnetic flux that enables a telecoil to sense the magnetism and produce an audio voltage. (The earth's magnetic field does not induce a voltage in a coil, because the earth's field has a constant strength and direction).

3.5.2 Magnetic field strength

The strength of the magnetic field near the center of the room is directly proportional to the magnitude of the current in the loop and to the number of turns in the loop, and is inversely proportional to the diameter of the loop. International standards (IEC 60118-4, BS7594) specify that the long-term rms value of the magnetic field should be 100 mA/m (that is, milliamps per meter).

[b] A magnetic field at some intermediate angle, such as 30° from the vertical, can be considered to have a vertical component added to a horizontal component, in the right mix of strengths to produce the actual angle of the field. The vertical component has a field strength equal to the actual field strength times the cosine of the angle the field is from the vertical. Angles up to 45 degrees therefore have relatively little effect on the size of the vertical component, but when the field is horizontal, a vertically oriented coil will detect nothing. Similarly, a horizontally oriented telecoil will pick up nothing if immersed in a vertical magnetic field.

The actual strength of the field at the center of a circular loop of diameter *a* meters, with *n* turns around the loop, can be calculated using Equation 3.2:

$$H = \frac{nI}{a} \qquad .... 3.2,$$

where *H* is the magnetic field strength, in Amps per meter, and *I* is the rms value of the current, in Amps. For a square loop, of size *a* by *a* meters, the magnetic field strength is 10% less than the value calculated by Equation 3.2. If the long-term rms strength of the loop has to be 100 mA/m, the loop must be able to output an rms level of at least 400 mA/m (and preferably 560 mA/m), so that excessive peak clipping can be avoided during the more intense sounds in speech. Field strengths are often expressed as dB re 1 A/m, so 100 mA/m is equivalent to -20 dB re 1 A/m.

It is important that the magnetic output of loops be not much weaker than this. The loop will not be the only electrical wire in a building producing a magnetic field. All the building wiring will also be producing magnetic fields with a frequency of the electricity power supply (50 or 60 Hz, depending on the country) and at harmonics of that frequency. This constitutes magnetic interference or background noise (which has a characteristic low-pitched hum or buzz). If the audio magnetic field is too weak, the SNR will not be adequate.[c]

In small hearing aids, with small coils, this telecoil sensitivity is achieved by using a separate pre-amplifier for the coil signal. While the user could compensate for a weaker field by turning up the volume control, this is inconvenient, particularly if the user needs to switch frequently between the telecoil and microphone positions. Also, compensation is possible only if the hearing aid has a volume control, if the control has adequate reserve, and if the gain can be sufficiently increased without causing feedback. Even on telecoil position, feedback oscillations can occur if the gain is too great. Just as acoustic waves cause feedback by leaking back from the receiver to the microphone, so too magnetic fields can cause feedback in the *T* position by leaking back from the receiver to the telecoil.

Ideally, all telephones would also emit a magnetic field strength of 100 mA/m, but unfortunately, some don't. Very old telephones produced a satisfactory magnetic field strength, because they accidentally leaked a lot of magnetism. Newer telephones and public telephones have been designed to emit a magnetic field specifically for use by hearing-impaired people (e.g. ANSI C63.19 requires cell phones to emit a field strength of at least 125 mA/m).[970] The problem lies in the telephones in-between, some of which are still in use, which were designed to be efficient for their acoustic output, a consequence of which was that only a very weak magnetic signal leaked out.

Taking into account the standards for room loops and telephones, a hearing aid wearer should be able to conveniently switch from microphone to telecoil mode without changing the volume control if the hearing aid produces the same output for a magnetic field strength of 60 to 100 mA/m as it does for an acoustic input of around 65 dB SPL.[840a]

3.5.3 Loop frequency response

The frequency response of a loop and telecoil system can sometimes be unsatisfactory, although this need not be so. Because the hearing aid acoustic response will have been carefully adjusted to suit the aid wearer, it is important that the combined response of the loop and hearing aid telecoil not be too different from the acoustic response. One exception to this is that some additional cut for frequencies below about 500 Hz *may* be beneficial (for some people in some situations), as this is the frequency region where magnetic interference is most likely to occur. Unfortunately, this may also be the most important frequency region for people with profound hearing losses. Fortunately, multi-memory hearing aids (Section 3.3.2) often make it possible to adjust the response separately for the telecoil and microphone operation, so that the best telecoil response for an individual aid user can be selected. Some remote controls even allow the user to select a low-tone cut when needed, such as in rooms with a lot of magnetic interference (fluorescent lights, and lights with dimmers operating are particularly troublesome). Adaptive noise reduction algorithms (Section 8.1) should also be effective at decreasing the buzz from magnetic interference (along with the

c Background interference levels can be measured, or can be listened to with the help of a hearing aid set to telecoil position and a stethoclip or temporary earmold, before a loop is installed, to ensure that the background interference will not be too great.

Practical tips: Installing or improving a loop

- Make the loop as small as you can get away with. Consider using two or four loops for large areas.
- The resistance of the loop (which can be measured with a multimeter, or calculated with the equation below) should not be less than the amplifier is able to safely drive. Four ohms or more is usually safe, but wherever possible, read the amplifier specifications!
- A room of size 5 m by 5 m can be looped with two turns of 0.4 mm diameter wire, powered by an amplifier of 10 Watts (or more). Alternatively, thicker wire could be used for convenience with a 10 Watt, 3 to 5 ohm resistor added in series to provide the necessary minimum total resistance. Figure 3.6 shows how a twin-core cable can be connected to provide the two turns required.

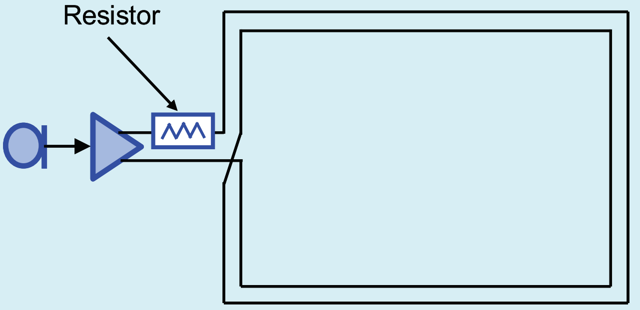

Figure 3.6 The connections needed to make a loop of two turns using a single run of cable that has two separate wires. The location of an optional series resistor is also shown.

If you do not like equations, read no further!

- For a room covered by b loops connected in series, each of n turns, with each turn in each loop having a perimeter of p meters, made of wire with a diameter d mm (excluding the insulation), and producing a maximum rms field strength of 0.4 A/m, the following can be calculated to design a loop or to check an existing loop. Where the area is covered by a single loop of n turns (the most common configuration), b equals 1.

$$\text{Minimum amplifier power} = \frac{bp^3}{2800n\,d^2}\,Watts$$

$$\text{Minimum wire diameter to avoid overheating} = \sqrt{\frac{p}{62n}}\,mm$$

$$\text{Corner frequency} = \frac{7610}{nd^2\log_{10}(446\,p/d)}\,Hz$$

$$\text{Loop resistance} = 0.022\frac{bnp}{d^2}\,Ohms$$

These equations, which are in part derived from Philbrick (1982) and from British Standard 7594, assume that no external resistor is used to increase the total resistance. If one is used, the minimum amplifier power and the corner frequency are both increased by the ratio of total resistance to resistance of the loop itself. An amplifier power of twice the minimum power calculated above is desirable to minimize peak clipping.

components of speech at those same frequencies) because of the constant or slowly changing nature of the magnetic interference.

There are two reasons why a user might experience a different frequency response in the telecoil position than in the microphone position. First, the loop may emit a weaker magnetic signal for high-frequency sounds than it does for low-frequency sounds. This can happen because the electrical impedance of the loop comprises an ***inductance*** as well as a ***resistance***. An inductance has an impedance that increases with frequency, so the total impedance of the loop starts to rise once the frequency exceeds a certain frequency known as the ***corner frequency***. (At the corner frequency, the impedance of the inductance equals the impedance of the resistance). If the loop is powered by a conventional audio power amplifier, the current, and hence the magnetic signal, will both decrease as frequency rises above the corner frequency. The solution is to make sure the corner frequency is 5 kHz or higher. This can be achieved by using either:

- wire with a small diameter (provided it does not overheat);
- a special current-drive power amplifier (with a high output impedance);
- installing several loops to cover the total area;
- very few turns, or even just one turn, in the loop;
- a graphic equalizer; or
- an external series resistor.

Each of the last three options may require a more powerful amplifier. Simultaneously achieving the right resistance, field strength, and frequency response, without overheating the wire in the loop, is easy for small loops, and difficult (i.e. expensive and impractical) for large loops.

An alternative solution is to use a grid of small loops, which must be placed under the carpet or mat, rather than around the perimeter of the room.[35] This also minimizes spillover from the loop to adjacent rooms. Such systems, embedded in mats, can be purchased commercially, or can be connected such as shown in Figure 3.7.

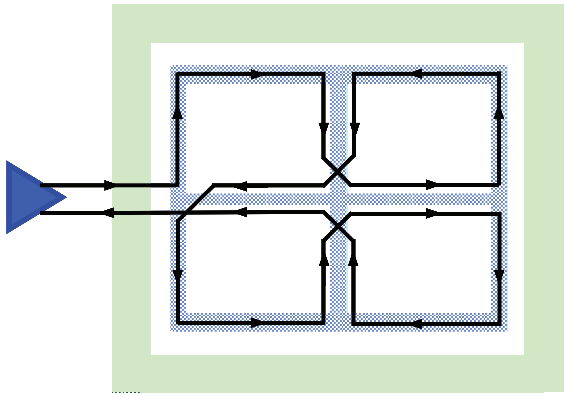

Figure 3.7 Multi-loop connection to cover a large area, but still achieve an adequately high corner frequency. The field strength vertical component is weak within the shaded blue areas, which could correspond to aisles in an auditorium, and also within and outside the outer green shaded area.

An even more effective solution is to use two separate loop systems in complex patterns covering the same area. The second loop is driven by a second amplifier, which produces a signal 90° out of phase with the signal from the first amplifier. Each complex loop provides a field in the dead spots of the other loop, and the 90° phase shift prevents the two magnetic signals from cancelling each other in places where they are both strong. This combination, which is called a ***phased-array loop***, results in a very uniform magnetic field, even if the area is large, and minimal spillover outside the area of the loops.

The second reason why the telecoil frequency response might be different from the acoustic response lies in the hearing aid itself. Coils inherently produce a voltage that rises with frequency. The hearing aid designer can compensate for this, either partially or completely, by the way in which the telecoil connects to the hearing aid amplifier, but the end result will likely have a low-frequency roll-off.[d, 840a] The shape of the telecoil response, relative to the microphone response, is evident from the specification sheet for the hearing aid, or by measuring each response in a test box (see Section 4.1.8).

[d] It would therefore be desirable for room loops to include a bass boost to compensate for the low cuts that usually occur in hearing aid telecoil circuits. Such a boost would also improve the ratio of signal levels to interference, which is usually very low-frequency dominated.

Practical tips: Wearing a transmitter and microphone

- Wearing the microphone on a head-mounted boom just below the mouth will result in a SNR about 10 dB better than clipping it to the lapel or dangling around the neck, and the signal transmitted will not be affected by extreme head turns away from the microphone.
- Clipping the transmitter microphone to the lapel will result in a SNR about 10 dB better than clipping it at waist level. (Many people clip it far too low, especially if the waist is the most obvious feature when looking down!)
- A directional microphone on the transmitter can compensate for it being worn further from the mouth.[1058]
- Wearing a transmitter with self-contained microphone under clothing, although convenient, will likely produce clothing-rubbing noises for the recipient as well as an attenuated signal, and should be avoided.

3.6 Radio-frequency Transmission

Radio-frequency transmission provides a more portable way to get a signal from a talker to a listener without corrupting the signal by noise or reverberation. The talker wears a small ***transmitter***. The transmitter may contain a microphone, in which case the transmitter is worn around the neck. Alternatively, a microphone is attached to the transmitter by a cable, in which case the transmitter is clipped to a belt or worn in a pocket and the microphone is clipped to the lapel or worn on the head. The connecting cable may also serve as an aerial for the transmitter.

The ***receiver*** is worn by the hearing-impaired person:

- it can connect electrically to the hearing aid by a cable (i.e. direct audio input);
- it can connect by magnetic induction (via a loop around the neck or by a silhouette coil) to the hearing aid;
- it can be clipped onto the hearing aid; or
- the whole receiver can be incorporated within the hearing aid as discussed in section 3.6.3.

In radio-frequency transmission, the audio electrical signal is not directly converted to another form of energy (as occurs in magnetic induction from a loop to a telecoil), but instead modifies or ***modulates*** the characteristics of an electromagnetic wave. This electromagnetic wave is called the ***carrier***. In the absence of an audio signal, the carrier is a sinusoidal wave. It can convey information only when the audio signal alters some aspect of the carrier. A variety of analog or digital modulation techniques can, in principle, be used, but the two most commonly associated with short-range transmission are frequency modulation, and frequency-hopping spread-spectrum modulation. These are outlined in the following two sections.

3.6.1 Frequency modulation

In the hearing aid field, it is the carrier frequency that is most commonly altered, so we refer to this as ***frequency modulation*** or ***FM***. Figure 3.8 shows an audio wave, an unmodulated carrier, and the resulting modulated carrier. The job of the receiver is to detect the carrier and then produce a voltage that is proportional to (i.e. a replica of) the original audio signal. This extraction of the modulating waveform is called ***demodulation***.

There are other forms of modulation that could be used, the most common of which is ***amplitude modulation***, in which the audio signal modulates the amplitude, rather than the frequency of the carrier. The advantage of using modulation (of either sort) is

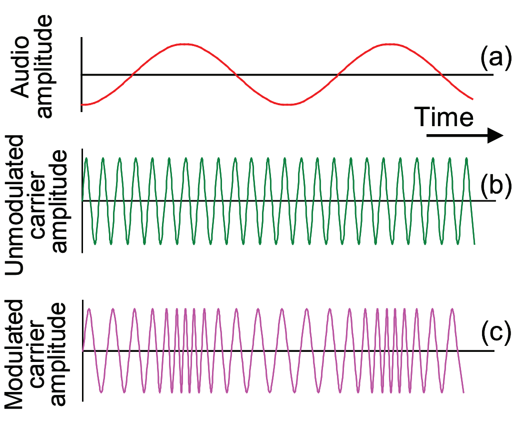

Figure 3.8 The waveform of a carrier before (b), and after (c), it has been frequency modulated by a sinusoidal audio signal (a).

that the strength of the audio signal coming out of the receiver does not depend on the strength of the carrier wave, and hence does not depend on the distance between the transmitter and the receiver. As the carrier wave becomes weaker, however, the receiver will progressively add noise to the audio signal. When the carrier becomes extremely weak, reception will cease entirely.

There are probably many hundreds of electromagnetic waves, coming from many hundreds of transmitters, passing through the room in which you are now sitting. How does the receiver select just one of these before it demodulates the audio signal it contains? The receiver is tuned to be most sensitive to a particular carrier frequency. Only when the receiver frequency matches the frequency sent out by the transmitter will the receiver pick up the transmitted signal. What happens if two transmitters are sending out signals at the same frequency? There is certainly the potential for much confusion in the receiver. Licensing authorities minimize the problem by designating different parts of the electromagnetic spectrum (i.e., different carrier frequencies) for different types of transmitters. In various countries, the frequency bands available for hearing aid devices are 37, 43, 72-76, 173, 183, 216, 900 and 2400 MHz. Within each of these bands, a number of different transmission frequencies are allowed, and the narrow frequency region around each is referred to as a ***transmission channel***.

With FM transmission and reception, an additional phenomenon helps when a receiver is exposed to two different transmissions at the same carrier frequency, or to two carrier frequencies that are only slightly different. Because the demodulator works by "locking on" to the carrier, and then measuring how much its frequency varies with time, it can lock on to a strong carrier even if a weaker carrier is simultaneously present. This phenomenon of demodulating only the stronger signal is known as the ***FM capture effect*** because the receiver is captured by the stronger signal. If two transmitters of the same output power are generating the two signals, then the stronger of the two signals at the receiver will be the one originating from the closer of the two transmitters. The field intensity coming from a transmitter decreases in inverse proportion to the square of the distance from the transmitter (known as the ***inverse square law***), just as for acoustic waves. Radio-frequency waves pass through non-conductive obstacles (such as brick walls) extremely well. They are attenuated by large conductors such as sheet metal walls, and, to a lesser extent, by the human body.

The FM capture effect can be used to advantage in schools. Two classrooms can use the same transmission channel provided the classrooms are sufficiently far apart for all children in each class to receive their own teacher's signal much more strongly than they receive a signal from teachers in any other classroom operating at the same frequency. Unfortunately, it is not always clear just how far this should be. The inverse square law does not apply exactly if there are large metal objects nearby, which is common in buildings. Long metal objects can cause signals from distant transmitters to be received with stronger than

Practical Tips: Fixing unreliable reception

- Check that the transmitter and receiver batteries are fresh, or fully charged in the case of rechargeable batteries.
- Make sure that the problem is not a faulty connecting cord, plug or audio shoe (try a new one, or for intermittent operation, wiggle the cords).
- Check that the transmitter and receiver aerial wires are not cut off or curled up.
- Rearrange the room so that the teacher and children are closer together and and/or further away from large metal objects.
- Choose a different transmitter (and receiver) channel.
- On hand-held transmitters, make sure the user is not accidentally covering the microphone ports with his or her hands or fingers.
- If possible, substitute each component (transmitter, receiver, hearing aid) to identify the component in which the problem lies.

expected intensity. Metal objects can also cause reflections of the electromagnetic wave. This reflection can cancel the signal coming directly from the transmitter, thus causing the signal strength to be very low at certain places in the room. A receiver at these positions will not be able to adequately detect the carrier and a ***dropout*** occurs. The listener will then hear only noise. More sophisticated receivers will detect that a drop-out has occurred and will ***mute*** or ***squelch*** the output signal, so that silence occurs when the receiver detects that it is not receiving a carrier wave.

Because of the potential for signal strength to be much lower or higher than expected, a better solution where FMs are to be used in multiple nearby classrooms is for the transmitters in each classroom to operate at different carrier frequencies, and for the receivers to automatically synchronize to the transmitter in that classroom when the students enter the room.

The discussion in the preceding paragraphs about potential interference between multiple transmitters applies to systems that are designed to operate with a single transmitter. There are now several systems available that enable the output of several microphones and transmitters to be fed to a single receiver. This means that several people, each with their own transmitter, can talk to the hearing aid wearer and the feature is useful for team teaching or classroom activities in which other students are important talkers.

3.6.2 Digital modulation techniques

An alternative modulation technique particularly well suited to the transmission of digital data (including digital audio) is the combination of ***differential binary phase-shift keying***[e] and ***frequency-hopping spread spectrum***.

In binary phase-shift keying, the 1's and 0's of the digital data are represented by the phase of the radio-frequency carrier. For example, every time a digital 1 occurs, the phase of the carrier is changed by 180 degrees, which simply means inverting it, as shown in Figure 3.9. Every time a digital 0 occurs, the phase is left unchanged. A receiver can therefore recover the digital data by detecting whenever the phase of the carrier changes. The transmission method still has the same disadvantage as FM however, in that another transmitter sending the same frequency can interfere with reception.

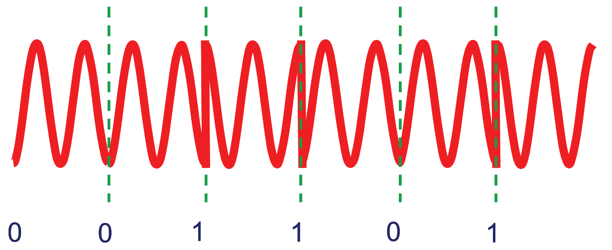

Figure 3.9 Transmission of digital data by differential binary phase-shift keying. A phase reversal occurs every time a digital 1 occurs, but not when a digital 0 occurs.

An alternative that is less prone to interference is frequency-hopping spread spectrum. Many times per second, the transmitter hops to a new carrier frequency, in a pre-arranged but seemingly random sequence. If the receiver knows this same sequence, it hops at the same time, so the transmission of information carries on in an uninterrupted manner. At each new carrier frequency, the receiver continues to detect changes in the phase of the carrier, and so continues to recover the digital data. The advantage is that because the transmitter sends such a small amount of power at each frequency, the signal is less likely to interfere with other transmitters, or be interfered by them, than when both transmitters are sending all their information in the same narrow frequency range. To keep the amount of information as low as possible, the bit rate of the digitized audio signal is reduced by one of several available data compression algorithms. In the receiver, the encoded signal is decoded to reconstruct a close approximation of the original signal.

Bluetooth is the most well-known example of a frequency-hopping, spread-spectrum system. Transmission occurs by hopping between 79 channels, each 1 MHz wide, in the range from 2,402 MHz to 2,480 MHz. Hops occur 1600 times per second, and if interference occurs on any carrier frequency, the transmitter and receiver agree to skip that frequency in the future, thus further decreasing interference to and from conventional narrow-bandwidth transmitters. (This is called ***adaptive frequency hopping***, as the sequence adapts to avoid interference.) Bluetooth

[e] *Differential* means it is the phase changes that are significant, not the actual phase; *binary* means that only two different phases are used; and *phase-shift keying* means that it is the phase that is key to the value of the binary data.

systems are intended for short-range transmission, typically 30 ft, or 10 m, and have several applications within audiology. Section 3.3.1 referred to the Bluetooth-enabled Noah-link device used to program the hearing aid from a computer. Section 3.11.1 will show how Bluetooth can help connect a mobile phone to the hearing aid via a Bluetooth link to an intermediate accessory.

Unfortunately, Bluetooth transmitters so far consume too much battery current for them to be built into the hearing aid itself. The high frequency of operation (2,400 MHz or 2.4 GHz), does however enable operation with very small antennae. Purpose-designed frequency-hopping systems operating in this frequency range are now starting to be incorporated within hearing aids.

Another disadvantage of Bluetooth, for some applications, is that the handshake[f] transmission protocol it employs imparts substantial delay to audio signals. The sound is then desynchronized from visual input, perhaps by enough to disturb lip-reading. A more serious interference problem occurs if the user can hear the original sound source, plus the delayed version arriving via Bluetooth transmission, just as discussed in Section 2.4.5. The Bluetooth delay creates no problem if there is no visual information and no audio information reaching the user except that which arrives via the Bluetooth path.

Recently, a new low-power Bluetooth standard that uses less power and has less audio delay has been published, so hearing aids in the future may have Bluetooth built into them.

3.6.3 Coupling to the hearing aid

The audio signal coming from the wireless receiver is useful only if it can be delivered to the ears of the hearing-impaired user. The simplest form of output is for the wireless receiver to directly drive an earphone of some sort. The major disadvantage is that wireless receivers do not usually contain sophisticated (or sometimes any) tone controls or adjustable forms of compression. It is thus not possible to adjust the amplification characteristics to suit the requirements of the individual aid wearer. Individual amplification needs can be met more accurately if the wireless receiver output is coupled to the person's own hearing aid. This coupling can be achieved in four ways:

- Electrically, from a body-worn receiver via a cable to the hearing aid's direct audio input connector, assuming it has one.
- Inductively, from a body-worn receiver via a loop worn around the user's neck, which then sends a magnetic signal to the hearing aid telecoil, or via a small coil mounted in a thin plastic case that is positioned behind the wearer's ear, right beside the wearer's own BTE hearing aid. This coil is known as a ***silhouette*** coil, because its case has a profile similar to that of a BTE hearing aid. It is also known as an ***inductive earhook***.
- Electrically, from a receiver mounted into a small boot that plugs into the bottom of the BTE hearing aid.
- Electrically, from a receiver completely integrated inside the hearing aid.

Each of these methods of coupling to the hearing aid has its advantages and disadvantages, and may affect the gain-frequency response of the combined amplification system in different ways.[703] Direct electrical connections provide a well-defined signal. Those using a cable are less convenient, cosmetically undesirable, and less reliable after continued use.[831] An important advantage is that speech-operated switching and adaptive systems (see next section) are most easily possible with direct connection. Also stereo (i.e. dichotic) listening is possible if the source produces a stereo signal.

If a body-level receiver is used, the neck loop is cosmetically superior (particularly important to teenagers) and there are no cables outside the clothing to interfere with an active lifestyle or to be grabbed by small hands. Disadvantages are that the low frequencies can be attenuated, and the strength of the magnetic coupling (and hence of the audio signal) can be decreased when the head is inclined to either side. (Boring school lessons can produce a 90-degree bend of the neck, by which angle none of the magnetic

[f] In a handshake protocol, transmission only continues once the receiver sends back a signal acknowledging that the signal has been received without interference, which takes time. The actual delay in Bluetooth systems depends on exactly which protocol the designers have selected.

signal is picked up by the telecoil, possibly making the lesson even less interesting!) Also, magnetic signals are prone to interference from nearby electrical apparatus. The silhouette coil has all the disadvantages associated with the presence of a cable, and the potential for interference. Its two advantages are that a stereo connection is possible and unlike with the neck loop, there is no change of signal strength with changes in head position.

The coupling methods that have the best combination of reliability and cosmetic acceptability is when the wireless receiver is mounted *inside* the hearing aid, or is mounted in a small boot that plugs onto the bottom of the hearing aid (Section 3.11.1). These solutions are also the most used.

If the patient has near-normal hearing in any frequency region, the wireless receiver should be coupled to the ear with an open earmold or ear shell so that the wireless system does not reduce unaided signal reception – i.e. the level of sounds arriving acoustically from nearby talkers who are not wearing a transmitter.[966] The open coupling will, of course, affect the gain-frequency response of the complete system (Section 5.3.1). This applies whether the wireless receiver is coupled directly to the ear or is coupled via a hearing aid.

The digital signal processing in hearing aids can cause interference in a nearby wireless receiver, which in turn can pass the interference on as an audio signal to the hearing aid, which then produces the interfering sounds in the user's ear.[79, g] This is an entirely different mechanism from the interference that mobile phones commonly cause in hearing aids, as explained in Section 3.11.3. It is therefore essential that a listening check be performed whenever a hearing aid and a wireless receiver are first coupled together, in addition to any other electroacoustic tests that are carried out.

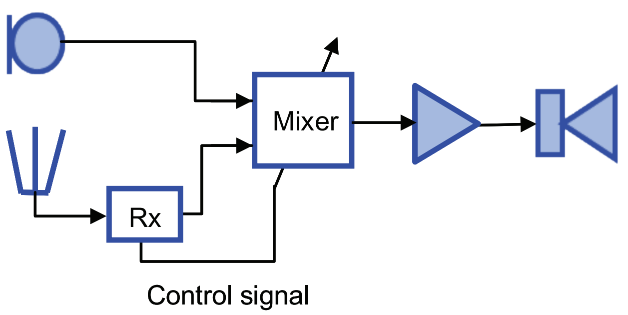

Figure 3.10 A system in which the hearing aid automatically selects either just the local microphone signal or the FM signal from the receiver (Rx) combined with an attenuated version of the local microphone signal.

3.6.4 Combining wireless and local microphones

When children (or adults) are working in small groups, or otherwise need to hear more than one other person talking, or need their hearing aids for own-voice monitoring, it is not satisfactory for the hearing aid to receive only the signal coming from the wireless transmitter, because the transmitter may be far away from the person talking. Wireless systems overcome this problem in a number of ways. The most common solution is for the aid wearer to hear a mixture of sound coming from the transmitter, and sound being picked up by the hearing aid microphone, also called the ***local microphone*** (see also Section 2.9). While this allows a nearby talker to be heard, the hearing aid microphone continues to pick up noise and reverberation even when the teacher is talking into the transmitter, potentially removing most of the advantage provided by the wireless system.[155, 329, 699] The problem is minimized by mixing the two signals, as shown in Figure 3.10, so that the signal from the FM system is more intense than the signal from the hearing aid microphone. This ratio is referred to as ***FM advantage, FM priority*** or ***FM precedence***. In some hearing aids, the hearing aid microphone is attenuated to achieve this FM advantage only when the receiver detects that it is receiving a valid radio-frequency signal from a transmitter.

[g] In many cases, the interference is not audible if the transmitter is close to the receiver, because the signal received from the transmitter is then much stronger that the interfering signal produced by the hearing aid. The interference therefore reduces the operating distance of the wireless system. A cure to this problem is to use a channel frequency that is not disturbed by the hearing aid. With cochlear implants, the problem is more likely to occur because the coupling coil uses much more power for transmitting energy and data on a 5 MHz carrier frequency, harmonics of which cause interference. The solution is to use the radio only on "clean" channels, which the cochlear implant manufacturer or FM manufacturer can advise.

FM systems and advanced nonlinear hearing aids

There should be no more problems coupling an FM system to an advanced, nonlinear hearing aid than there are to a linear, peak-clipping hearing aid. The hearing aid must be properly adjusted to operate by itself, and the output level from the FM receiver must be properly adjusted (see next panel). Once these adjustments are achieved, the signal from the wireless receiver is processed in exactly the same way as a signal picked up by the hearing aid microphone.

As with any non-linear hearing aid, the aid should be measured with a broadband signal with a speech-like spectrum rather than a pure tone (Section 4.1.3). If the hearing aid contains adaptive noise suppression, this function should be disabled during measurement unless the test signal contains speech-like modulations.

Amplification, compression, noise suppression, output limiting, transposition, and possibly even feedback suppression, will all operate normally. They will operate just as they do for acoustic input signals with overall levels around 65 dB SPL (or as they do for acoustic signals of varying input levels if the FM transmitter does not normally operate in compression). Microphone directionality will not be relevant if the FM system is operating in a mode that attenuates the hearing aid microphone.

There is a great dilemma in setting up the mixture in a combined signal: the clearest signal from the teacher is received in the wireless-alone condition, the worst signal from the teacher is obtained using the local microphone alone, and a signal of intermediate clarity is obtained in the combined position.[329, 699] The dilemma is that when children are asked which operating mode they prefer, the order is exactly reversed,[329] presumably because the children feel increasingly detached from their immediate environment as their local microphone becomes less dominant.

The best solution to this dilemma lies in an automatic switch within the wireless system that attenuates the hearing aid microphone when someone speaks into the transmitter microphone, but that restores full sensitivity to the local microphone when the transmitter has nothing to send. These systems, referred to as ***speech-operated switching (SOX)***, or ***voice-operated switching (VOX)***. Unaccountably (to this author) such systems are less commonly available at the time of writing than in previous decades.

A commonly-used solution aimed at solving the same problem is a processing scheme known as ***dynamic FM***. In this system, the degree of FM advantage automatically increases as the background noise level (sensed by the transmitter) increases. As would be expected, increasing the priority given to the FM signal increases speech intelligibility when there is background noise.[1779, 1927]

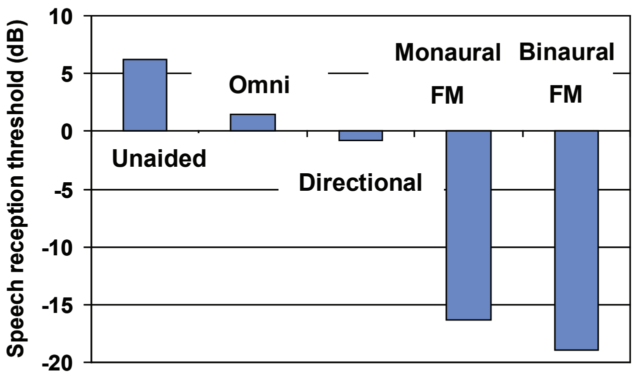

Figure 3.11 Speech reception threshold for different amplification conditions, averaged across the two experimental sites reported in Lewis et al, 2004.

The improvement in speech intelligibility in noise offered by wireless transmission is potentially huge, because the signal picked up by the transmitter microphone is likely to have a SNR 20 dB greater than the SNR picked up by the hearing aid microphone. Figure 3.11 shows the speech reception threshold for adults wearing hearing aids or FM systems.[1059, h] The advan-

[h] Note that the discussion in this section uses FM and wireless interchangeably. While most wireless systems are now FM, this may not stay the case forever, but the issues of how to combine wireless/FM signals with acoustic inputs will nonetheless remain.

Adjusting the wireless receiver output controls

Adjustment, or at least verification, of the operation of an FM system in conjunction with the hearing aid is essential. If, as is usually the case, the FM system will be used in a combined FM+HA condition, verification should also be performed in this condition. The aim of the adjustment is that at the output of the hearing aid the signal from the FM will be 10 dB greater than the signal picked up by the local microphone, on the assumption that the signal at the local microphone will be around 65 dB SPL.[1, 511] While it might seem that this could be achieved by ensuring the hearing aid has a 10 dB greater output level for an 80 to 85 dB SPL speech signal (typical of voice level at the transmitter microphone) as it does for a 65 dB SPL speech signal input directly to the hearing aid in normal operation, compression limiting in the transmitter and wide-dynamic range in the hearing make this an inappropriate adjustment method.

An alternative method, specified in AAA(2008) guidelines,[1] instead introduces the concept of transparency. The system is said to be in its transparent condition when 65 dB SPL input to the hearing aid with the FM connected but muted results in the same output level as 65 dB SPL input to the FM transmitter, when averaged across the mid frequencies (but preferably at all frequencies). The required 10 dB FM advantage will then be achieved if, as is commonly the case, the transmitter limits the speech level sensed by its microphone to the electrical equivalent of 75 dB SPL. Because the signals are mixed at the input of the hearing aid, the 10 dB FM advantage is not affected (at least when both signals are present simultaneously) by any subsequent compression in the hearing aid.

The whole adjustment procedure can be carried out with a test box and 2-cc coupler. Ideally the test signal will be speech. If actual speech, or a signal with speech-like dynamics is not available, speech-weighted noise can be used, but adaptive noise reduction in the hearing aid should be disabled.

Some implications of this procedure are:

- The actual level received by the aid wearer compared to when the hearing aid is used in isolation depends on the extent to which the hearing aid microphone is attenuated when the FM system is connected and turned on.
- If background levels around the aid wearer are less than or greater than 65 dB SPL, the FM advantage will be greater than or less than 10 dB, respectively.
- The variation in FM advantage with changing background noise level is reduced if the signal level sent by the transmitter varies automatically with background noise level (i.e. dynamic FM).
- Variations to the procedure may be necessary for different FM systems, such as when the transmitter limiting level is different from 75 dB SPL, as can occur for different types of transmitter microphones.

tage offered by the wireless system far exceeds that offered by the directional microphones. Applying the FM system to both ears improves the speech reception threshold by a few dB over application to just one ear. A small improvement is understandable because the FM system in the second ear lowers the exposure of that ear to noise around the aid wearer, thus reducing central masking of the clean signal provided by the FM system to the first ear. Unfortunately, the second FM also attenuates signals of interest originating from people other than the transmitter wearer.

The relatively small improvement offered by having the wireless signal connected to the second ear suggests another solution to the problem of combining the two inputs. The FM can be connected to one ear, and the other ear can be left with just the hearing aid microphone input, or with a combined input. Provided the listener can always move his or her head to compensate for different directions of acoustic sources, this arrangement produces very similar performance, for both the wireless and the acoustic inputs, as having automatically switched inputs to both ears.[1422] None-the-less, for optimal results in all situations, the best system seems to be one that automatically gives the wireless signal a large (e.g. 20 dB) level advantage when the transmitter receives input from a nearby source, delivers it to both ears, and switches the wireless signal off otherwise. The only exception seems to be where the wearer of the transmitter and another person are simultaneously talking, and the hearing aid wearer needs to monitor both talkers, which is a challenging situation even for people with normal hearing.

3.7 Infra-red Transmission

Infra-red radiation is the same type of electromagnetic energy as radio waves, except that it occurs at a much, much higher frequency (approximately 10^{14} Hz). For frequencies slightly higher than this, electromagnetic radiation is perceived by humans as a red light, hence the term infra- (meaning *below*) red. Transmission of audio signals via infra-red electromagnetic waves also requires that the carrier (the infra-red wave) be modulated by the audio wave.

In this case, it is more convenient to use amplitude modulation. In particular, the infra-red wave is pulsed on and off, with the audio signal directly or indirectly controlling the timing of the pulses. The infra-red receiver first detects the pulsing infra-red carrier, and then demodulates the pulses to recover the audio signal. More complex modulation systems combine amplitude modulation with either frequency modulation or spread-spectrum modulation to improve rejection of background noise which arises from other sources of infra-red energy.

Just as for FM radio-frequency signals, the output of an infra-red system can directly drive a headphone, or can be coupled electrically or inductively to a hearing aid. It is more common for infra-red systems to be used directly with an earphone or earphones, although there is no fundamental reason why this should be so.

Because infra-red operates at almost the same frequency as light waves, infra-red radiation behaves in the same way as light waves. It travels in straight lines, is easily blocked by opaque obstacles, and reflects (with some attenuation) off flat, light-colored surfaces, like ceilings and white-boards. Direct sunlight may interfere with transmission. You can experiment (but not now!) with how well infra-red systems work in the presence of obstacles and reflectors by noting the conditions in which your TV or video remote control works or does not work.

A variation on transmitting information by infra-red light is to transmit it with visible light as the carrier.[740] The intensity of light from a fluorescent light can be modulated at a rate far higher than is perceptible to the human eye and brain. Audio waveforms, or digital information, can be coded by varying the frequency of the intensity fluctuations. Just as for infra-red systems, a receiver detects the light and demodulates the intensity fluctuations to recover the information. If the recovered information is an audio waveform, it can be presented over headphones or coupled to a hearing aid as described in Section 3.6.3. If the information is data, such as captioned text, it can be displayed on a screen. The major advantage is that rooms already have lights that can be converted to act as light modulators. The major disadvantage is that on bright days, outside light may swamp the modulated light, making detection impossible.

3.8 Classroom Sound-field Amplification

Unlike the three systems just described, sound-field amplification systems get sound to the listener using acoustic waves propagated across the room. They work on the premise that if SPL, and hence SNR, is adversely affected by distance from the talker and background noises (see Figure 3.2), both of these problems can be minimized by amplifying the wanted sound and positioning a loudspeaker near the listener. The most common application is a classroom, and the most basic system thus consists of a microphone, an amplifier, and one or more loudspeakers.

An obvious limitation of this basic system is that the teacher either has to remain next to a fixed microphone, or carry round a microphone with a long cable. An essential addition is therefore a radio-frequency (e.g. FM) or infra-red link[i] between the teacher and the amplifier, which enables the teacher to move freely around the room. Figure 3.12 shows the block diagram of such a system with four loudspeakers.

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[i] One system has three infra-red receivers located at different positions around the room so that there is a clear line of sight from the teacher's transmitter to at least one of the receivers at all times.

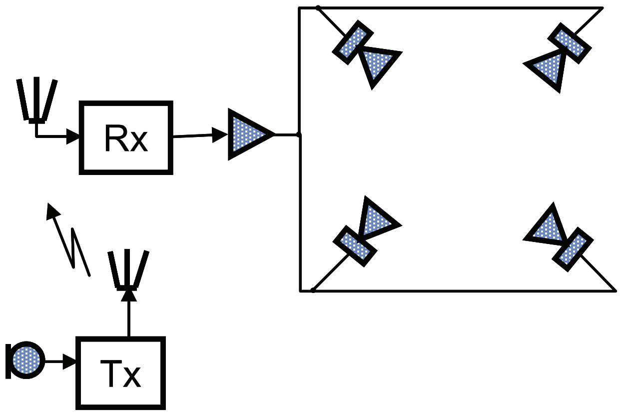


Figure 3.12 Block diagram of a sound-field amplification system comprising a transmitter worn by the teacher, a receiver and amplifier mounted somewhere convenient in the room, and four loudspeakers distributed around the room.

Sound-field amplification systems have several advantages over the three other types of transmission systems already discussed. First, they do not require the listeners to wear any special equipment. This simplifies supply logistics (except for screwing loudspeakers onto a wall as required for some systems), and increases system reliability because there is nothing worn on the body for the children to break! Second, the improved sound clarity is available to all children in the classroom, not just those with hearing aids or cochlear implants. Children with normal hearing thresholds will also benefit educationally from receiving a clearer signal.[371, 1156, 1953] Third, the advantages will therefore be available to those who have a temporary conductive hearing loss. This is particularly advantageous for some indigenous populations where the incidence of conductive loss can be high. The fluctuating nature of the loss makes it difficult for children to be fitted with individual hearing aids at all times when they need it. In some cases, cultural factors can lead to rejection of individual devices. An initially unexpected advantage of sound-field systems is that teachers report considerably less voice fatigue, and decreased incidence of voice nodules when they use these systems.[553]

Of course, there are also disadvantages. Sound-field amplification systems can increase the sound level in typical classrooms only by about 10 to 15 dB before feedback oscillation occurs.[j] This means that the ratio of signal to noise is also likely to be increased by this same amount, which is very worthwhile if noise, rather than reverberation, is limiting speech intelligibility. The increase in SNR may be less if background noise levels also rise when the system is used, but they are more likely to fall.[1025a] The most sophisticated sound field systems monitor noise levels in the classroom and automatically increase their gain and apply some degree of high-frequency emphasis when the background noise levels rise.

Although sound-field systems can increase the ratio of direct to reverberant sound received by each child, the magnitude of the increase depends on the distance between the child and the closest loudspeaker, the directivity of the loudspeaker, the distance between the child and the teacher and the critical distance in the room. Unfortunately, each loudspeaker adds to the reverberant sound as well as to the direct sound, so if reverberation rather than noise is limiting speech intelligibility, classroom amplification is not a good solution. There are several ways to minimize, but not completely avoid, this problem:

- Loudspeakers can be positioned in the ceiling or high on the wall near each corner of the classroom, so that as many children as possible are as close as possible to one of the loudspeakers.
- A directional vertical column loudspeaker at head height when seated can be used. These loudspeaker arrays contain several individual loudspeakers which cause the array to radiate much more sound horizontally than they do vertically. This increases the critical distance (by increasing Q in equation 3.1), thus increasing the ratio of direct to reverberant sound received by each child.
- Where there are only one or two children in the classroom who need assistance, each child needing assistance can have a small loudspeaker positioned on their desk directly in front of them.[47, 796] This system is referred to as ***a desk-top FM***

[j] The maximum gain achievable without feedback whistling depends on the amount of sound absorbing material in the room. Adding soft furnishings will allow higher gains to be achieved (and will also improve sound quality in the absence of the system). Higher gain is also achievable in systems that have a feedback management system, such as a small degree of frequency compression (see Section 8.2.4).

Sound field amplification works in classrooms because:

- It increases the level of sound (good for people with hearing loss).
- It increases the ratio of direct to reverberant sound for anyone closer to the loudspeaker than they are to the teacher (good for anyone close enough to a loudspeaker).
- It increases the signal to noise ratio (good for everyone).
- Its benefits are immediate.[1187]
- It is immediately apparent to the teacher when there is a technical fault.

system. Figure 3.2 is again relevant, but this time with the loudspeaker considered as the source. The arrangement is relatively clumsy if the child has to move around, singles out the child assisted as different, and provides minimal benefit to most of the rest of the children in the class. It is a good solution, however, in any situation where the listener is static, such as for an infirm adult who frequently watches TV from bed or their favorite chair.

In principle, the use of column arrays in all four corners would give the highest ratio of direct to reverberant sound for all children in the class, but such a system would be expensive, and no such system is commercially available. Whichever of the three available options is used, there is great benefit in making the room surfaces as acoustically absorbent as possible to minimize reverberant sound.

The increase in sound level and SNR achieved by sound-field systems is greatest if the teacher wears a microphone on the head, positioned on a boom so that the microphone is close to the lips (but not directly in front of them to avoid popping sounds and to avoid blocking lip-reading).[554] At this close distance to the mouth, the microphone picks up very little reverberant sound, and very little noise, as can be seen from Figure 3.2.

As with other remote transmission systems, the basic system is suitable for only a single talker. Team teaching, or interactive work between the teacher and the children, requires that two transmitters be used. The outputs can be added, or in more sophisticated systems, connected within an automated network that determines on the basis of pre-determined priorities, or on the basis of voice activity at each transmitter, which transmitter the receiver responds to at any time. Automatically selecting one microphone, rather than adding together the output of multiple microphones, avoids the addition of multiple noise sources and prevents the hearing aid wearer from receiving multiple voices simultaneously. Several commercial systems that facilitate multiple microphones are available. The second transmitter must be passed around if the advantages are to be achieved whenever a child is the talker. Children generally react well to having control of the microphone when reading or making presentations to the rest of the class.

It will be apparent that classroom amplification systems and wireless systems for individual children share a need for the teacher to wear a microphone and transmitter. Where both types of systems are in use in the same classroom, it is convenient for the teacher if the same microphone and transmitter can send signals to both the classroom amplifier/loudspeaker and the individual wireless receivers. One commercial system has achieved this level of compatibility in a seamless manner, albeit by building into the teacher's "transmitter" two separate transmitters: an FM transmitter to send to the children wearing individual wireless receivers, and a digital modulation transmitter to send to the classroom amplifier/loudspeaker.

3.9 Comparative Strengths and Weaknesses of Magnetic Loops, Radio-frequency Wireless Systems, Infra-red, and Sound-field Amplification Systems

The four remote transmission hearing aid systems can be compared in a number of ways: effectiveness in improving SNR, effectiveness in increasing the ratio of direct signal to reverberant signal, convenience, reliability, and cost. We will assume that the sound-field system has been implemented with a wireless link between the teacher and the amplifier. We will also assume that the application is a classroom containing several hearing-impaired children, each of whom already has an individually fitted hearing aid with direct electrical input connection and/or telecoil, as appropriate. Table 3.1 shows which of the systems have advantages over the others in each of these areas.

Each of the first three systems, because they transmit energy in a non-acoustic form, are able to deliver large increases in SNR, and large decreases in reverberant energy, provided the microphone is placed sufficiently close to the teacher's mouth. These improvements can be 20 dB or more, and are therefore much more substantial than can be achieved with a classroom sound-field amplification system.[47] The increase in intelligibility in noise is so great that, if use of a remote transmission system is possible, its use may enable some patients with severe loss to continue wearing hearing aids rather than receive a cochlear implant.[1160] Of course, these options are not mutually exclusive: remote transmission systems improve the performance of cochlear implants in noise just as much as they do the performance of hearing aids.[549] Individual sound-field systems, with the speaker very close to the recipient, provide benefit intermediate to that of a radio-frequency or infra red system and a classroom sound-field system.

The sound-field amplification system, however, is probably the most convenient to use provided hearing assistance is needed only in one place. The receiver, amplifier and loudspeaker components do not have to be touched, and there is therefore only one transmitter to be handled. Convenience rapidly decreases, however, once there are two or more talkers to be amplified, unless the system has been designed to work with two transmitters simultaneously.[1373]

The magnetic loop system is less convenient to install. Apart from installation of the loop itself and power amplifier, unless the talker is always in the same position in the room, a radio-frequency system (e.g. FM) is needed to convey the signal from the microphone to the loop amplifier. Once set-up, however, it is extremely convenient in that the children need no additional device provided their hearing aids have a T position.

A consideration with both magnetic loops and radio-frequency systems is that all children have to switch to the T position or direct audio input position of their hearing aids to receive the magnetic/radiated signal, and they will then be detached from their local acoustic environments. This problem is avoided if their hearing aids can be adjusted to receive both the radiated/magnetic signals and acoustic input, but the SNR advantage is somewhat degraded, as discussed in Section 3.6.4.

The sound-field amplification system provides the most consistent performance, partly because there is only one mobile component, and partly because it is immediately apparent to the teacher if the system

Table 3.1 Relative advantages of each of the remote transmission systems, for application in a classroom containing several hearing-impaired children. For each criterion, a greater number of check marks indicates a greater relative advantage.

	Magnetic Loops	Radio-frequency wireless transmission	Infra-red transmission	Sound-field amplification
SNR improvement	✓✓	✓✓	✓✓	✓
Reverberation decrease	✓✓	✓✓	✓✓	✓
Convenience		✓	✓	✓
Consistency and reliability				✓
Privacy			✓✓	✓
Low cost				✓

fails, so corrective action can be taken immediately. Magnetic loops are the next most consistent, again because there are few mobile components, provided the system has been installed so that it produces a sufficiently strong magnetic signal. While magnetic interference can occur, the sources of interference do not usually come and go, so systems can remain interference-free if they are initially so.

Radio-frequency wireless systems can sometimes be less consistent in operation, partly because of ***drop-outs***, partly because of ***interference*** (which can come from distant places at unpredictable times), and partly because there may be several receivers and even transmitters, each with their own cables, connectors and batteries to be maintained. In the classroom, infra-red systems are less prone to interference, but may have numerous dropouts whenever the transmitter and the receiver are not directly facing each other. This likely to happen as the teacher or children move around the classroom. Infra-red systems are particularly well suited to applications where the listeners are all facing in the same direction and where the transmitter always directly faces the listeners. In indoor situations where this is true, infra-red systems are free from interference and dropouts. The design of the room will affect the likelihood of dropouts. If the teacher faces a whiteboard, the IR will reflect off it far more readily (and hence reach the receiver(s) with sufficient strength) than if the teacher faces a blackboard.

Infra-red systems are the clear winner when it comes to privacy: there is virtually no spillover outside the room in which the transmitter is located. Apart from any confidentiality issues, the lack of spill-over means that many rooms in a building can be fitted with identical systems, without any interference between adjacent rooms. The other three systems radiate signals outside the room in which they are being used, with the extent of radiation being greatest for radio-frequency wireless systems, although eavesdropping requires access to a receiver with the correct carrier frequency. If confidentiality is not an issue, electromagnetic spillover is not a problem provided that the equipment used in different classrooms operates at different frequencies. It is advantageous to have a choice of frequencies so that interference can be avoided. Spillover of array loops and phased array loops is considerably less than for simple perimeter loops.

Finally, simple magnetic loops and sound-field amplification systems are the cheapest to install and maintain, as only one device per classroom is needed, rather than one device per child. Only sound-field systems provide an improved signal to children who are not wearing any individual hearing device.

3.10 Assistive Listening Devices

Any devices that help hearing-impaired people detect sounds or understand speech, but which are not worn totally on the head or body are referred to as ***assistive listening devices*** or ***ALDs***. ALDs can be used in conjunction with hearing aids, or instead of hearing aids. The various wireless systems described in Sections 3.6 to 3.9 are all ALDs, and the inter-connection devices described in Section 3.11.1 are also sometimes referred to as ALDs. They can be categorized into those that improve intelligibility and those that detect environmental events.

ALDs that improve intelligibility. Intelligibility is improved by locating a microphone closer to the source than is usually possible for the user to be or by directly connecting a source device (e.g. TV) to a transmitter. ALDs that improve speech intelligibility can be further categorized according to their purpose:[1678]

- *One-to-one communication,* such as for listening to a conversation partner in a car or in a noisy or reverberant place. An example is a personal wireless system (e.g. infra-red or radio-frequency FM) with the receiver coupled to a hearing aid or coupled to headphones and used without a hearing aid.
- *Group listening systems,* such as sound-field systems, infra-red systems, and magnetic loop systems. While in principle it should be possible for a user to purchase a receiver and use it with transmitters available in public places like theatres, the variety of system types (infra-red and radio-frequency), carrier frequencies, and modulation types, makes it unlikely that any single receiver will be compatible with multiple venues that the user frequents. Magnetic induction systems do not have this disadvantage, provided the hearing aid has a telecoil.
- *Television devices,* which either pick up the television audio via a plug-and-socket, or use a microphone placed very close to the television

loudspeaker. The signal can be delivered to the user via hard-wired connection or a wireless link, which in turn can be either radio-frequency transmission or magnetic loop induction. The loop can be room-sized, chair-sized, or ear-sized. Of course, radios, home stereos, or any other electronic media source can, in principle, be connected in the same way. Clinicians should also make the patient aware of the very significant intelligibility advantage that closed captioning provides for TV watching.[642]

- *Telephone devices*, which include:
  - an amplified telephone;
  - an amplifier that is inserted between a regular telephone and its wall socket;or
  - a coupler that picks up the acoustic signal coming from the receiver and amplifies it, to create either a stronger acoustic output, a stronger magnetic output, or an electrical signal that can drive a neck loop or a silhouette coil.

Other connection options are considered in the following sections.

The increase in SNR offered by ALDs with the microphone near the sound source is so huge that clients would benefit greatly if they were to be more frequently used. Their use is commonplace by children in school, but not so by adults. Investigations have shown that the substantial improvement in SNR does not necessarily lead to perceived benefit great enough to overcome the perceived logistical disadvantages.[153, 818, 1060] Some limiting factors are the user's willingness to involve communication partners in their use, cosmetic appearance (decreasingly so), the logistics and discipline needed for battery charging, the sensitivity of the FM system relative to the hearing aid microphone, and expense. When the conversation is over, the microphone must be retrieved, which may be an awkward way to terminate a conversation.

Successful use of wireless systems by adults is, however, possible with appropriate instruction and demonstration, although cost will be an obstacle for some.[298, 1340] The increased clarity that an ALD can provide, relative to acoustic reception across a noisy and reverberant room, will be staggering to many patients if they are only given the opportunity to hear it.

Some devices referred to as ALDs are really just very large hearing aids, comprising a microphone, an amplifier, and headphones, worn on the body of the patient or held against the ear of the patient. They have the advantage of large controls, self-evident operation, robustness, and are less likely to be lost, making them particularly suitable for some clients in nursing homes.[146, 1042]

The greater the degree of hearing loss the user of an ALD has, the more important it is that the electroacoustic characteristics of the device match the hearing characteristics of the listener. This can be achieved with real-ear measurement, just as for hearing aids, as further explained in Chapter 4.

ALDs that alert the user to environmental events. Alerting ALDs comprise a sensor of some type linked to an output that can be easily detected by the hearing-impaired person (flashing light, vibrator, or intense low-frequency sound). The most common sensors/detectors or triggers are:

- a telephone ring sensor;
- a baby cry sensor;
- a smoke detector;
- an alarm clock; and
- a doorbell.

The ALD comprises one or more of these detectors or sensors, plus one or more output transducers, the most common of which are:

- a loud sound with low, or adjustable pitch;
- a bright flashing light; and
- a vibrator that can be placed under the mattress or pillow, or carried in the pocket in the form of a pager.

Depending on the particular ALD, the detector and output transducer may be within the same package, or may be physically separated, in which latter case they can be connected by wire or, increasingly, by wireless transmission.

The most effective type of output transducer depends on the situation, and in principle should depend on the degree of loss of the patient. Averaged across a representative range of commonly occurring hearing losses, however, a 520 Hz square wave is much more effective at waking people with a hearing loss, as well as waking people from a range of other sub-populations, than either the standard 3000 Hz high-pitched tone used in almost all smoke alarms, a bed vibrator, or a flashing light.[195, 196] The flashing strobe light was shown to be least effective, waking only 27% of the participants.[196] It seems likely that the higher the sensation level of a sound while awake, the greater the chance that the sound will wake the person while asleep. Selection of devices must take into account whether the patient is certain to be wearing hearing aids at all times when the alert provided by a device is likely to occur. Selection should also take into account the impact that very loud alarm signals can have on normal-hearing people within the same household.

There is the potential for expert counseling to greatly increase the use that patients make of ALDs. One limiting factor, which you can overcome, is to ensure that they are actually *told* about ALDs, are assisted in *selecting* one or more that suit their individual needs, and have the opportunity to *trial* it or them. Although clients may report to you the needs they are aware of, there are some they may not think of, such as the need to hear a smoke detector at home. They are unlikely to be aware of how much more easily they could be hearing people at a distance.

A more extensive description of ALDs can be found in Comptom (2002), who advocates positioning ALDs and hearing aids as parts of a total communication system for the patient. Each patient's needs at home, at work/school, and for leisure should be reviewed. For each of these situations, there may be needs in the categories of face-to-face communication, reception of media, telecommunications, and alerting to environmental sounds that could be better met with ALDs than with hearing aids alone. These needs can be portrayed within a matrix showing situation by signal category. The level of need within each of the elements of this matrix increases with the degree of hearing loss.

3.11 Connectivity and Convergence

It is now very common for people with normal hearing to spend a significant part of each day with things in their ears. Most commonly these are earphones (i.e. receivers) to enable people to listen to mobile phones and MP3 players. Other electronic devices that produce audio signals include radio receivers, home entertainment systems, portable video players, satellite navigation systems, personal digital assistants, and computers. People with hearing loss have the same needs to listen and communicate, and it is inconvenient to remove a hearing aid before inserting an earphone. Removal of the hearing aid also takes away the individually shaped gain-frequency response, to the detriment of intelligibility and tonal quality, especially for people with a moderate or greater loss and/or a sloping hearing loss. Consequently, there is a growing need for hearing aids to connect easily to these other devices, or to actually fulfill the function of these other devices, as described in the next two sections.

3.11.1 Connecting electronic devices to hearing aids

Methods for connecting radio-frequency wireless systems to hearing aids have already been discussed in Section 3.6.3. These same methods (direct electrical audio input, neck loop/telecoil, and silhouette coil) can be used to connect any audio device. Because of the difficulty of plugging anything into a small hearing aid, and the inconvenience of having a cable attached to the hearing aid, most manufacturers are offering better alternatives.

Figure 3.13 shows two alternatives for connecting a distant audio source, such as a TV, to the hearing aids. In the upper diagram, a wireless transmitter picks up the TV audio with a microphone (located as close as possible to loudspeaker) or connects directly to the TV with an audio cable. The transmitter sends the TV audio as a radio-frequency signal which is picked up directly by the hearing aids.

In the lower diagram, the transmitter at the TV sends the audio to an interface device[k] via Bluetooth or a proprietary wireless transmission. The interface device then re-transmits the audio to the hearing aids via a low-power, low-frequency (several MHz), very short-range transmission method, usually magnetic induction.[l]

[k] At the time of writing, the interface devices referred in this section have trade names including Dex, iCom, SmartLink, Streamer, Surflink streamer, Tek Connect, and uDirect.

[l] These proprietary short-range systems are designed to use much less battery current than is required for Bluetooth, and thus have less impact on the hearing aid battery life than if the Bluetooth transmitter/receiver were to be built directly into the hearing aid.

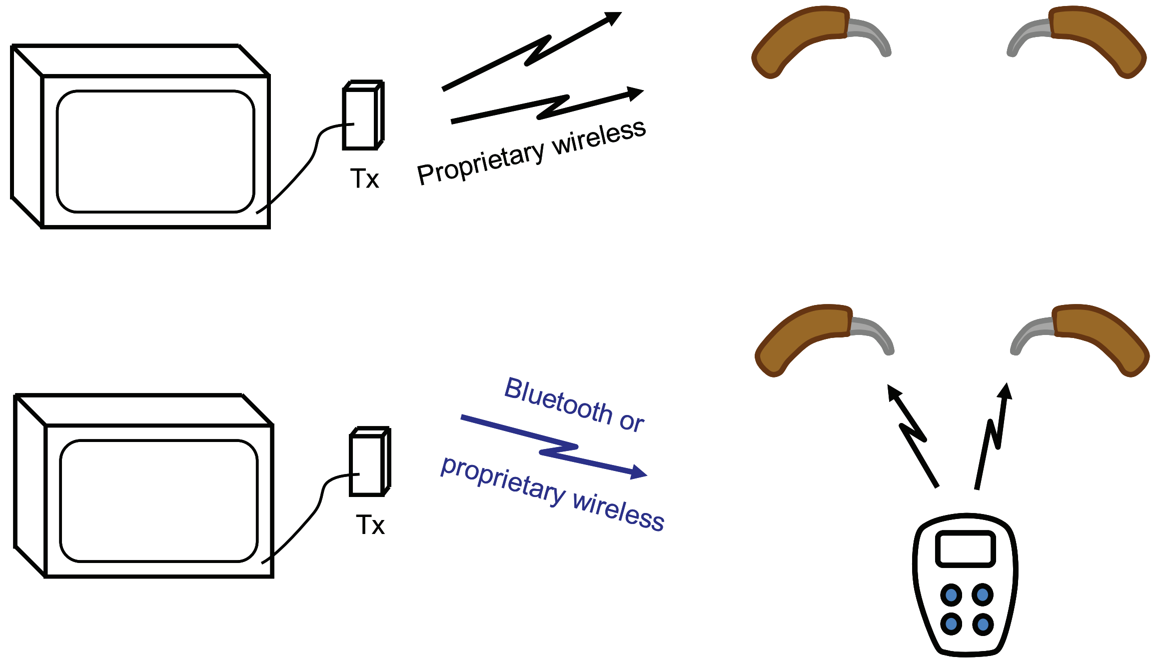


Figure 3.13 Wireless connection of a TV to hearing aids. In the upper figure, the transmitter (Tx) transmits directly to the hearing aids. Transmission methods available include FM in the 150 to 220 MHz range and proprietary digital modulation in the 850 - 900 MHz and 2.4 GHz ranges. In the lower figure, the transmitter at the TV sends a proprietary wireless or Bluetooth signal to an intermediate device worn on or near the person. The intermediate device retransmits the signals to the hearing aids by a low-power, low-frequency transmission method, such as inductive magnetic coupling.

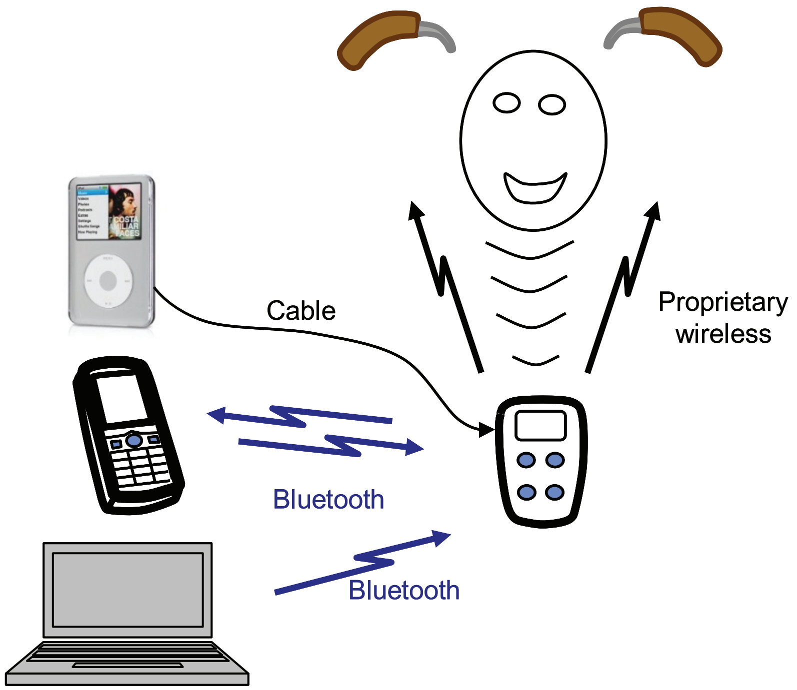


Figure 3.14 Connection of an MP3 player, mobile phone, or laptop to hearing aids via an interface device. The interface device picks up the aid wearer's voice with its microphone, and transmits it to a mobile phone via Bluetooth.

The signal is sent by inductive coupling rather than radiated energy, which has the advantage that the signal strength decays very rapidly with distance, and thus decreases the likelihood of one person inadvertently sending signals that interfere with the hearing aids of others. The interface must be hand-held or worn on the body. Some are designed to be worn around the neck, with the neck loop providing the induction sending coil. Alternatively, transmission from the interface device to the hearing aid is via a proprietary, very-low power, ultra-high frequency (UHF), digitally modulated, electromagnetic wave, such as at 2.4 GHz. In either case, the hearing aid can be programmed so that it attenuates the internal hearing aid microphone whenever the hearing aid recognizes that the interface device is sending an audio signal via wireless.

Although the use of an intermediate interface device seems more complex, the same device also provides a way to connect other devices such as mobile phones and MP3 players to the hearing aids, as shown in Figure 3.14. As the interface device has to be carried by the hearing aid wearer, it may be designed to fulfill additional purposes, such as a remote control or a highly-directional hand-held microphone. An additional advantage is that because the interface transmits to both hearing aids, the audio source can be heard in both ears, which improves intelligibility in noisy places, particularly for occluding fittings.[1414] This can be a stereo signal when the source is in stereo, like MP3 players, or the same signal in both ears when the source is mono, like a mobile phone. As the interface may be paired with several different devices, it likely has an internal priority system that specifies which of the signals it receives should be sent to the hearing aid. Signals from the mobile phone are likely to be prioritized over other signals.

Alerting ALDs can be fully integrated with hearing aids in a similar manner, as shown in Figure 3.15. The sensor (from telephone, doorbell, or smoke alarm) transmits a signal to a remote control carried by the hearing aid wearer. The remote control vibrates, illuminates a display, and also wirelessly transmits an audible alert to the hearing aids, or a signal to a bed shaker if the remote control is mounted in its overnight charging station.

3.11.2 Convergence

An alternative to connecting the hearing aid to an external audio device is for the functions of the hearing aid and an audio device to be met within a single, ear-worn device. This is a trend that will increase in the near future as it becomes increasingly possible to implement wireless transmitters, receivers, and other functions within hearing aids without placing an unacceptably high drain on the hearing aid battery.

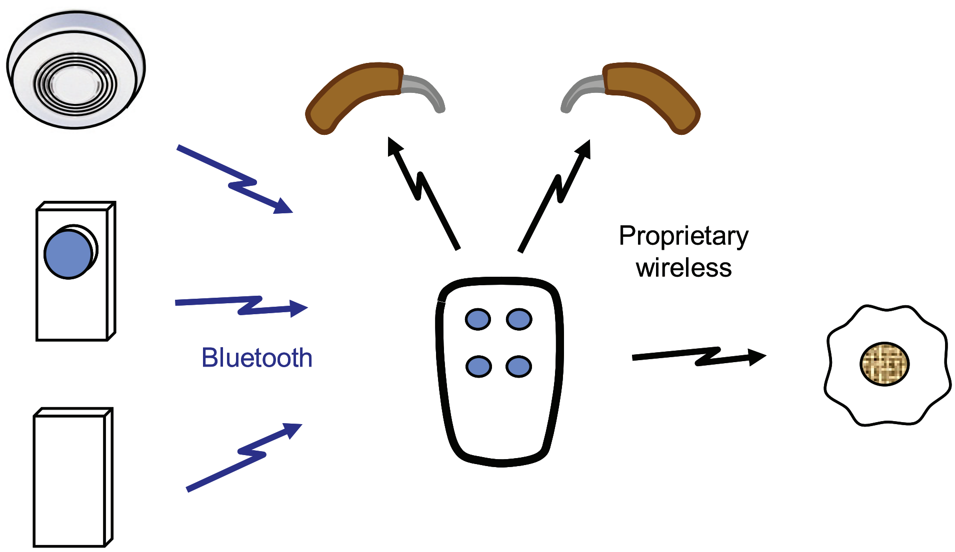


Figure 3.15 Connection of environmental sensing devices (smoke detector, telephone interface, doorbell) to a remote control, and from the remote control to hearing aids or a bed shaker.

An obvious first step would be for the hearing aid(s) to become the hands-free microphone/receiver for a mobile phone. The hearing aid(s) would then also deliver the audio output to the ear for any other electronic devices with compatible wireless transmission.

As mentioned in Section 3.2, fully linking hearing aids on opposite sides of the head will create super-directivity that will enable better-than-normal hearing in some noisy situations. If so, even people with normal hearing would benefit from such devices in noisy places. As well as facilitating conversation in difficult situations, these devices will simultaneously protect against noise-induced hearing loss. Future devices may therefore be better thought of as combined speech enhancers, hearing protectors, and audio output systems for a variety of electronic devices. For those wearers who happen to have hearing loss, the devices would also provide amplification and shaping of sound! In the longer-term future, complete additional functions, such as a mobile phone, satellite navigation system, or personal digital assistant (all voice activated and with audio rather than visual output) could perhaps be incorporated within something that looks like a current hearing aid package.

An alternative direction of convergence would be for the amplification and other signal processing needed for a hearing-impaired person to be incorporated within other devices (e.g. a mobile phone). Such devices are already available, at least experimentally.

3.11.3 Interference between mobile phones and hearing aids

Interference of the hearing aid by mobile phones has been a major, but decreasing, problem over the past 15 years. Interference should not be surprising as mobile phones are designed to transmit strong electromagnetic fields, and hearing aids contain many conductors that act as aerials when immersed in an electromagnetic field. As well as the radio-frequency signals that phones are designed to emit, they also emit audio-frequency electrical and magnetic signals of various magnitudes depending on their physical layout. Although interference from mobile phones was originally sufficiently great that hearing aid wearers could experience interference from nearby users of mobile phones,[1048] the resistance of hearing aids to interference has now progressively improved so that at most, interference is caused when the user holds his or her phone close to the head.[137, 219, 1565]

The level of interference experienced depends on:

- the design of the hearing aid;[962]
- the carrier frequency and modulation method used in the telephone transmission system;
- the distance of the mobile phone from the nearest phone tower and the presence of intervening obstacles (and hence the power transmitted by the phone);
- the design of the phone (both location of the aerial and layout of conductors carrying current);
- the distance between the telephone and the hearing aid; and
- the orientation of each device relative to the other.

Over the last decade, hearing aids have been made progressively less sensitive to interference by using shorter conductors, adding capacitors to bypass radio-frequency signals that are picked up by wires and components within the hearing aid, and by coating the inside of the hearing aid case with a conductor to act as a radio-frequency screen.[219, 1043, 1045] Hearing aids with a high level of immunity to radio interference are said to be ***hardened***.

The modulation method used in the Global System Mobile (GSM) phone system is particularly efficient at *producing* interference.[968] The phone transmits short bursts of radio-frequency energy approximately every 5 ms, resulting in the carrier being amplitude modulated at a rate of 217 Hz. If the radio-frequency signal picked up by the hearing aid is strong enough to overload any transistor in the integrated circuit, the

Ensuring compatibility: Check the specs and/or try before you buy!

Patients should be advised to try a mobile phone with their hearing aid before finalizing their purchase of the mobile phone or hearing aid. Ideally, the trial should occur in a region of low wireless signal strength (visible on the screen of the mobile) so that the mobile is transmitting at maximum power during the trial.

The lower the IRIL value for the hearing aid, which can be found in most data sheets (see text), the more likely it is that the mobile phone and hearing aid can by used with the same ear.

distortion that results acts as a demodulator, producing an audio signal with a fundamental frequency of 217 Hz and at every integer multiple of 217 Hz. That is, the interference, which sounds like a buzz, spreads throughout the entire bandwidth of the hearing aid. Third-generation phone systems preferentially use a spread-spectrum modulation method (called Wideband Code Division Multiple Access; W-CDMA), and when this is demodulated, the audio interference produced is a less noticeable white noise.

In some countries, both phone systems operate in parallel and smart phones automatically switch between systems depending on which system has the highest signal strength at any moment. Because of this switching, and because the mobile phone increases its transmission power when it is in a low signal strength area, hearing aid users can find that the interference they experience can come and go, seemingly without any reason.

There is very little that can be done to make a hearing aid more resistant to interference if interference suppression has not been designed into the hearing aid. One option that is available is to keep the hearing aid and mobile phone further apart. This separation can be achieved by connecting the hands-free output socket of the mobile phone to either a neck loop, one or two silhouette coils, or a direct audio input socket of a hearing aid. Any of these coupling methods are likely to significantly increase speech clarity relative to just holding the mobile phone against the hearing aid with the hearing aid microphone input selected.[1677] Speech clarity improves because of some combination of reduced interference from the mobile phone, reduced pick-up of acoustic background noise near the hearing aid wearer, and better preservation of the signal spectrum via inductive loop coupling than by acoustic transmission from the mobile phone speaker to the hearing aid microphone.

If a neck loop is not available, a regular mobile phone hands-free kit can be used, but with the earpiece positioned beside the hearing aid. The hearing aid should again be switched to telecoil if available, as the hands-free receiver is likely to emit a satisfactory magnetic signal. If not, the hearing aid can be switched to microphone, but the loose and open coupling from the earpiece to the hearing aid microphone is likely to cause poor signal quality.

The best results will be obtained by combining a phone that emits the lowest possible level of electrical and magnetic signals in the vicinity of the hearing aid, with a hearing aid that is as insensitive as possible to electrical and magnetic signals (other than those intentionally picked up by the telecoil).

ANSI/IEEE standard C63.19 makes this choice easier for clinicians and patients by providing a rating system for both mobile phones and hearing aids. Mobile phones intended to work with hearing aids without a telecoil are rated as M1, M2, M3, or M4, with M1 being worst (i.e. largest emissions). Mobile phones intended to work with hearing aids in telecoil mode are similarly rated as T1 to T4.[m] The best ratings are likely to be obtained with the decreasingly common clamshell (i.e. flip-phone) mobiles,[968, 1589] as most of the electronics are maximally distant from the ear, and hence the hearing aid. Hearing aids are also rated, with M1 hearing aids being the most sensitive to interference and M4 being the least sensitive.

To check compatibility of a particular phone with a particular hearing aid, their rating numbers are added. If the total is 4, the combination is just usable; if it is 5 the combination is good; and if it is 6 or more, the combination has excellent freedom from interference.

An alternative method for quantifying susceptibility to interference is specified in IEC 60118-13. The ***input-related interference level*** (***IRIL***) is the level of an acoustic signal input to the hearing aid that produces the same output level as is produced by an interfering radio-frequency wave with specified characteristics. An IRIL of 55 dB SPL is considered likely to result in an acceptable level of interference when other people nearby use a mobile phone. The lower the IRIL, the less the interference picked up by the hearing aid. IRIL values considerably lower than 55 dB SPL are needed for the hearing aid wearer to directly use a mobile phone.

The issues in this section relating to interference by mobile cell phones also apply to interference by cordless handsets designed to operate in a local area with a range of several hundred meters. These phones, such as those conforming to the Digital European Cordless Telephone (DECT) system are becoming increasingly common, but fortunately have a lower power output than mobile phones, and so produce less interference.

m Phones cannot achieve a T3/T4 rating unless they first achieve an M3/M4 rating.

3.12 Concluding Comments

Hearing aid amplification systems involving remote transmission can provide a much higher level of performance than individual devices worn solely on the head. Remote transmission devices such as radio-frequency wireless, infra-red, induction loop, and sound-field systems should be used whenever good intelligibility is critical, such as in schools or business meetings. High performance does not always require high expense. The telecoil, a low-cost device that has been available in hearing aids for decades, is capable of being used much more often than is now the case.

A high rate of innovation in hearing aid systems is occurring and will continue. The intermediate devices, usually worn around the neck, that interface external devices to hearing aids provide a workable, but slightly clumsy solution. It is likely that with advances in integrated circuit technology, external devices (TVs, MP3 players, mobile telephones etc) will routinely transmit directly to hearing aids within the next few years – the challenge will be to find easy ways for the patients to control, at any time, whether they wish to hear one of the wireless transmissions, the acoustic input, or a combination of the two.

Hearing aid, mobile telephone, and computer technologies have much in common, so hearing aids will benefit from advances made in the other higher volume fields. For example:

- Hearing aids on each side of the head will be able to communicate with each other to achieve more dramatic noise reduction than has so far been possible, probably enabling people with mild hearing loss to hear better than people with normal hearing in many noisy environments.
- A hearing aid wearer may be able to position several miniature remote microphone-transmitters in his or her immediate vicinity, and a hearing aid with in-built wireless receivers may be able to combine the outputs to produce even more dramatic noise reduction.
- A hearing aid wearer may place a variety of sensors within the home to alert him or her through the hearing aid to important environmental signals, without a need for any other intermediary device.

A more immediate problem for audiology is to find how (or whether) low-cost, non-professionally fit hearing aids can best help hearing-impaired people. These devices, known as direct-to-client hearing aids, can be obtained by over-the-counter sale and internet/mail-order. They have the potential to give an increased proportion of the hearing-impaired population a first-hand experience of amplification with minimal financial commitment.

Cheap devices ***can*** have excellent electroacoustic performance,[1242] ***can*** give considerable benefit,[1173, 1882] and ***can*** be selected such that they match prescription targets adequately well.[1882] They are less likely to fit in the ear well, in which case increased irritation, feedback oscillation, poor retention in the ear, and hence decreased satisfaction are the result.[n, 1763, 1882] In all those cases where the physical fit or electroacoustic performance is not well suited to the hearing-impaired person, they also have the potential to reinforce negative and outdated beliefs about the limited effectiveness of hearing aids.

Unfortunately, many of the low-cost devices on the market have insufficient high-frequency gain relative to their low-frequency gain, and hence have gain-frequency responses unsuitable for typical mild and moderate hearing loss configurations.[239, 271] Also, many of the devices have high equivalent input noise, making them unsuitable for people with hearing loss and people with normal hearing alike, despite their claims to enhance even normal hearing for quiet sounds.

Difficulties with non-professionally fit hearing aids include the unknown degree of loss of the candidate, the unknown possibility of surgical correction, the inability to select hearing aid features, the inability to fine-tune the devices, the possibility of the hearing aid being dangerously loud, and the likelihood of a sub-optimal physical fit that causes feedback oscillation or is physically uncomfortable.

There is so far no substitute for a skilled clinician ensuring that a hearing aid is electroacoustically and physically well matched to the intended user. Current research into sophisticated self-fitting hearing aids may, however, help create more effective commoditized devices in the near future (Section 8.5).

[n] These experimental results were obtained with modular ITC devices. The comfort and fit that would be obtainable with over-the-counter thin-tube dome canal fittings is not yet known.

CHAPTER 4

ELECTROACOUSTIC PERFORMANCE AND MEASUREMENT

Synopsis

The performance of hearing aids is most conveniently measured when the hearing aid is connected to a coupler. A coupler is a small cavity that connects the hearing aid sound outlet to a measurement microphone. Unfortunately, the standard 2-cc coupler is larger than the average adult ear canal with a hearing aid in place, so the hearing aid generates lower SPL in this coupler than in the average ear. This difference is called the real-ear-to-coupler difference (RECD) a quantity that is worth measuring in infants because they have ear canals considerably different from the average adult. A more complex measurement device, which better simulates the acoustic properties of the average adult ear canal, is called an ear simulator.

Test boxes provide a convenient way to get sound into the hearing aid in a controlled manner. These sounds can be pure tones that sweep in frequency, or can be complex, broadband sounds that, like speech, contain many frequencies simultaneously. Broadband sounds are necessary to perform meaningful measurements on many nonlinear hearing aids. Increasingly, it is necessary for the test sound to approximate the spectral and temporal properties of speech so that the various signal processing algorithms in the hearing aid alter the gain in a manner representative of actual use. The measurements most commonly performed using test sounds are curves of gain or output versus frequency at different input levels, and curves of output versus input at different frequencies. The curve of output versus frequency when measured with a 90 dB SPL pure tone input level is usually taken to represent the highest levels that a hearing aid can create. Some other test box measurements that are less commonly performed are measures of distortion, internal noise, and response to magnetic fields. These measurements are used to check that the hearing aid is operating in accordance with its specifications.

Test box measurements are but a means to an end. That end is the performance of the hearing aid in an individual patient's ear. This performance can be directly measured using a soft, thin probe-tube inserted in the ear canal. Real-ear performance can be expressed as real-ear aided response (REAR; the level of sound in the patient's ear canal), real-ear aided gain (REAG; the level of sound in the ear canal minus the input level of sound near the patient) or as real-ear insertion gain (REIG; the level of sound in the ear canal when aided minus the level in the same place when no hearing aid is worn). Each of these measures requires the probe to be carefully located, but the requirements for probe placement are a little less critical for REIG than for REAG or REAR.

Both types of real-ear gain are different from coupler gain, partly because of the real-ear-to-coupler difference already mentioned, and partly because the input to the hearing aid microphone is affected by sound diffraction patterns around the head and ear. The changes in SPL caused by diffraction are referred to as microphone location effects. Insertion gain is further different from coupler gain because resonance effects in the unaided ear form a baseline for the insertion gain measurement. This baseline, referred to as the real-ear unaided gain, provides the link between the REAG and the REIG.

Many factors can lead to incorrect measurement of real-ear gain. These factors include incorrect positioning of the probe, squashing of the probe, blockage of the probe by cerumen, background noise, and hearing aid saturation. Fortunately, there are some simple checks one can do to verify measurement accuracy.

Feedback oscillation is a major problem in hearing aids. It happens when the amplification from the microphone to the receiver is greater than the attenuation of sound leaking from the output back to the input. Clinicians must be able to diagnose the source of excess leakage. Other problems that often have simple solutions include no sound output, weak output, distorted output, and excessive noise.

We cannot know what a hearing aid does unless its performance is measured. A block diagram shows us what types of things a hearing aid does, but it requires a measurement to determine the extent to which it does these things to the sound.

4.1 Measuring Hearing Aids in Couplers and Ear Simulators

Hearing aids are most conveniently measured in ***couplers*** and ***ear simulators***. The availability of standard couplers and simulators allows measurements to be made in different places and at different times under identical conditions.

4.1.1 Couplers and ear simulators

A coupler is simply a cavity. It has a hearing aid connected to one end and a microphone connected to the other. The coupler provides a repeatable way to connect the hearing aid to a microphone, and hence to a sound level meter, without sounds leaking out to other places. The standard coupler most commonly used for hearing aids has been around for over 60 years and has a volume of 2 cubic centimeters.[1524] This volume was chosen because it approximated the volume of the adult ear canal past the earmold when a hearing aid is worn (i.e. the ***residual ear canal volume***) and the equivalent volume of the eardrum and middle ear. Unfortunately, it is not a good approximation and is a very poor approximation of the acoustic impedance of the ear at high frequencies.

The SPL generated in any cavity by a hearing aid depends directly on the impedance of the cavity, which in turns depends on the volume of the cavity, and on the nature of anything connected to the cavity. In the average adult ear, the residual ear canal has a physical volume of about 0.5 cc.[828] This volume acts as an acoustic spring, or more formally, an ***acoustic compliance***. The ear canal, of course, terminates in the eardrum, on the other side of which is the middle ear cavity. The compliance of the middle ear cavity and eardrum together act as if they have a volume of about 0.8 cc.[1972] The combined 1.3 cc volume determines the impedance for low-frequency sounds.[1027] As frequency rises, the mass of the eardrum and ossicles causes their impedance to rise, while the impedance of the residual ear canal volume falls. Consequently, for increasing frequency, the total impedance does not decrease as much as would be expected for a simple cavity.

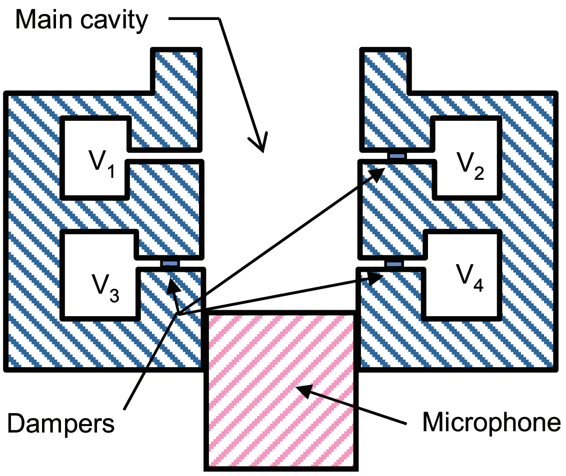


Figure 4.1 Simplified internal structure of a four-branch ear simulator.

An ear simulator mimics the ear's variation of impedance with frequency. Figure 4.1 shows the concept behind one ear simulator. As well as the main cavity, with a volume of 0.6 cc, the simulator shown has four side cavities, each with volumes from 0.10 to 0.22 cc, connected to the main cavity by small tubes, three of which contain dampers. As frequency rises, the impedance of these tubes rise and they effectively close off, thus causing the effective total volume to gradually fall from 1.3 cc to 0.6 cc.

One ear simulator with four cavities is widely known as the Zwislocki coupler and was sold commercially as the Knowles DB100 ear simulator. Other ear simulators still commercially available are the Bruel & Kjaer 4157 and the GRAS RA0045 ear simulators. They operate on the same principles, except they have two side cavities instead of four. The three simulators have a very similar variation of impedance with frequency. They match average ear transfer characteristics (from acoustic input to SPL at the microphone, corresponding to the eardrum) from 100 Hz to 10 kHz, but can be used as a coupler from 20 Hz to 16 kHz.

Several standards published by the American National Standards Institute (ANSI) and International Electrotechnical Commission (IEC) specify how hearing aids should be tested (Section 4.1.9) but allow measurement in either 2-cc couplers or ear simulators. To correctly interpret a hearing aid specification sheet, it is essential to determine whether the data refer to coupler or ear simulator performance because the gain and OSPL90 are higher in an ear simulator,[a] and whether the hearing aid has been measured in a test box or on an ***acoustic manikin***. An acoustic manikin comprises a head and torso, with an ear simulator incorporated inside each ear. As we shall see, the choices of coupler versus ear simulator, and test box versus manikin make a big difference to the numbers quoted.

Couplers and ear simulators have to connect to any type of hearing aid, and to achieve this, a range of adapters is used. Figure 4.2 shows several couplers, simulators and adapters, and Figure 4.3 shows some details and dimensions of 2-cc couplers. An essential concept is that of the ***reference plane***. This is a plane, at right angles to the longitudinal axis of the ear canal, located at the point in the ear canal where the earmold or ear shell usually terminates (defined in the standards to be approximately 13 mm from the eardrum). An ear simulator (and very approximately,

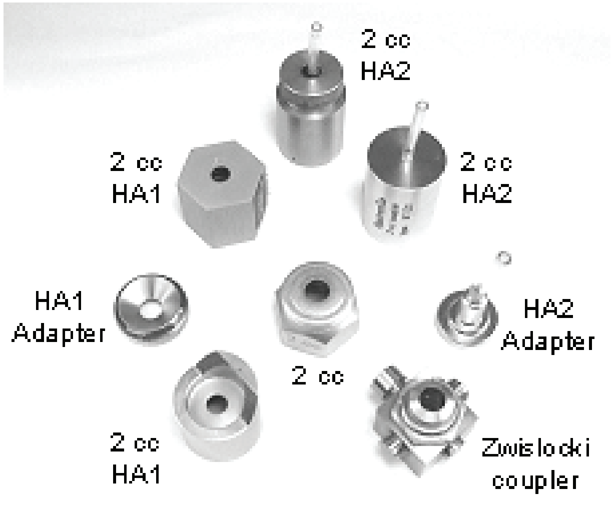
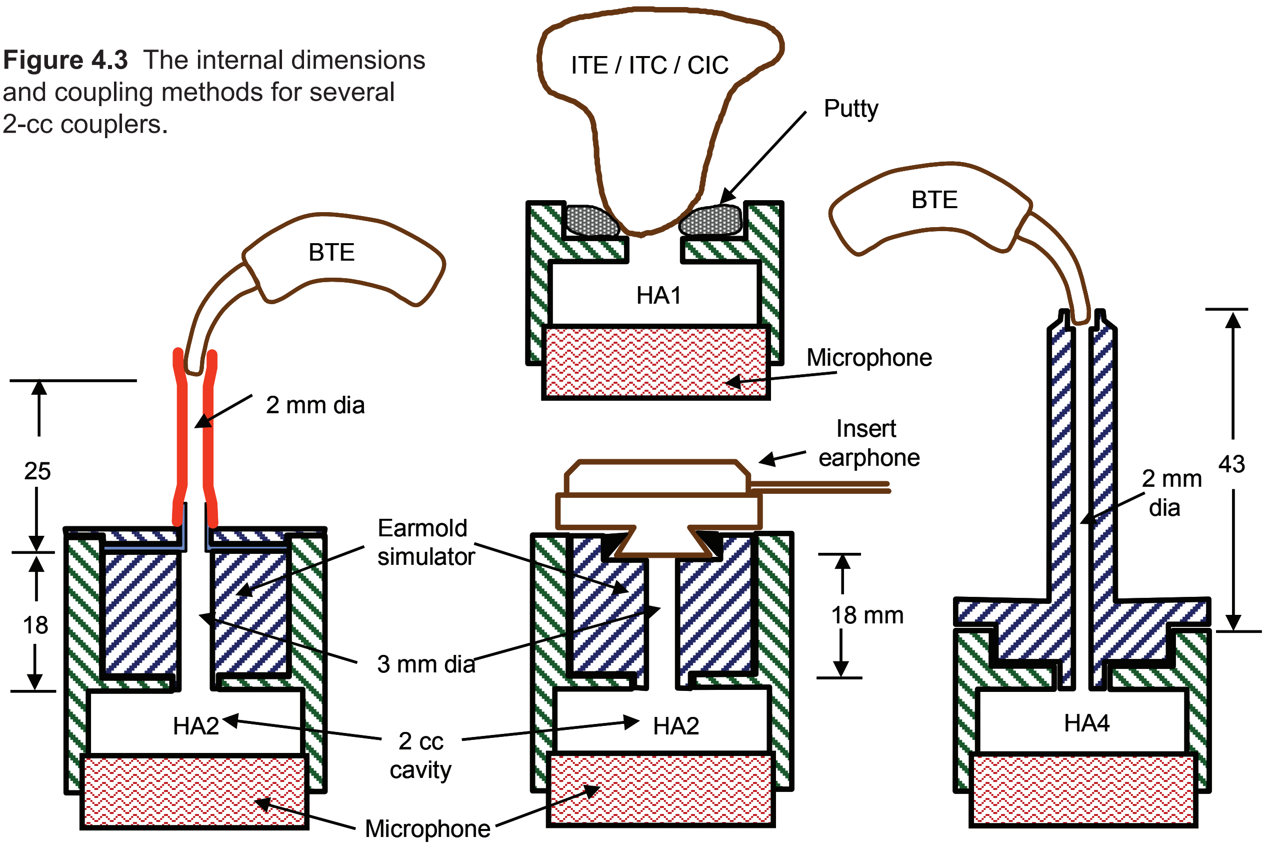


Figure 4.2 Several couplers and their adapters, and an ear simulator.

Figure 4.3 The internal dimensions and coupling methods for several 2-cc couplers.



[a] On a hearing aid specification sheet, the use of a 2-cc coupler may be signified just by mention of ANSI S3.7, ANSI S3.22 or IEC 60318-5 (previously known as IEC 60126 and IEC 126). Use of an ear simulator may be signified by ANSI S3.25, or IEC 60318-4 (previously known as IEC 711 and IEC 60711).

a coupler) represents the acoustic impedance of the residual ear canal volume and middle ear from this point inward. ITE, ITC and RITE BTE hearing aids usually terminate at this point, so these hearing aids are directly connected to a coupler or ear simulator. BTE (standard tube) and body aids, however, connect to the real ear via an earmold, so an ***earmold simulator*** is added between the coupler or ear simulator and the hearing aid. In addition, BTE hearing aids use tubing when connecting to a real ear, so they also require tubing when connecting to the coupler or simulator. CIC hearing aids are measured in the same way as ITE and ITC hearing aids, although in actual use they may be inserted beyond the reference plane.

ANSI S3.7 describes a 2-cc coupler as being used in several different applications, the most important of which are:

- The ***HA1 coupler*** has no earmold simulator and is used for ITE and ITC aids, which are connected to the coupler via putty. It is also used for BTE hearing aids not intended to be used with earmolds – typically thin tube BTEs that terminate in a dome, or RITE BTEs.
- The ***HA2 coupler*** includes an earmold simulator, which is connected to the BTE hearing aid via tubing, or into which a receiver for a body aid snaps. The HA4 coupler is a variation of the HA2 coupler intended for BTE or spectacle aids in which the tubing diameter from the hearing aid to the medial tip of the earmold is a constant 2 mm diameter. Although this tubing configuration is commonly used in BTE hearing aids, use of the HA4 coupler is less common.

The 2-cc coupler variations were defined in standards prior to the invention of the thin-tube RITC and RITA BTEs. Because these hearing aids have a single tube stretching from the BTE case to the reference point in the ear canal, rather than a detachable earhook, it is not appropriate to use the HA2 coupler (or else the length, as well as the diameter of the sound path would be inappropriate). It has become common practice to measure these hearing aids by connecting the output of the thin tube (for a RITA BTE) or the sound outlet (for a RITC BTE) directly to the input port of a HA1 2-cc coupler. An adaptor, only a few mm in length, is often used to achieve a more reliable, reproducible, faster method of attachment to the coupler than is possible with putty.

The advantage of the ear simulator over the 2-cc coupler is that because the ear simulator has the same variation of impedance with frequency as the average ear, a hearing aid generates the same SPL in an ear simulator as it does when inserted to the reference plane in an average adult ear.[b] This equivalency assumes that the hearing aid is coupled to the ear simulator in the same way that it is coupled to the ear canal. (The effect of different coupling methods on hearing aid response is covered in Chapter 5.) Even the ear simulator, however, cannot show the SPL that would be present in an individual ear, which is our ultimate interest.

Connection methods for ear simulators are similar to those used for 2-cc couplers. A connection option for the ear simulator is an ear canal extension attached to the opening of the ear simulator at the reference plane. The dome of a thin-tube, instant-fit BTE hearing aid is then inserted into the canal simulator, just as it would be inserted into the ear. A major difference from the other measurement configurations so far described is that there may be significant leakage of sound from the ear simulator, and significant transmission from the sound field directly into the ear simulator. Some simulators have ear canal extensions that enable the tubing from the hearing aid to be sealed to the simulators with putty.

The disadvantages of the ear simulator compared to the 2-cc coupler are its higher cost and the potential for the small openings inside the simulator to become blocked. Both 2-cc couplers and ear simulators will produce inaccurate results if:

- the sound bore of an ITE/ITC/CIC hearing aid is poorly sealed to the coupler or simulator;
- the tubing connecting to a BTE hearing aid becomes stiff and does not properly seal at either end;
- the o-ring connecting a button receiver wears out; or

[b] The simulator response matches the average real-ear response only over the frequency range for which leakage around the earmold or ear shell is insignificant in the real ear.

- the pressure equalization hole becomes blocked or excessively open.[c]

Except for the blocked pressure equalization hole, the remaining faults will all decrease the apparent low-frequency gain and power of the hearing aid being measured, and may also create a spurious mid-frequency resonance (see *Vent-associated resonance* within Section 5.3.1).

The equivalent of a coupler for bone-conduction hearing aids is the ***artificial mastoid***. It provides a standard way to measure the force output of a bone-conductor hearing aid over the frequency range 125 to 8,000 Hz, although this is not necessarily the same force that would be exerted on a human mastoid (ANSI S3.13, IEC 60318-6).

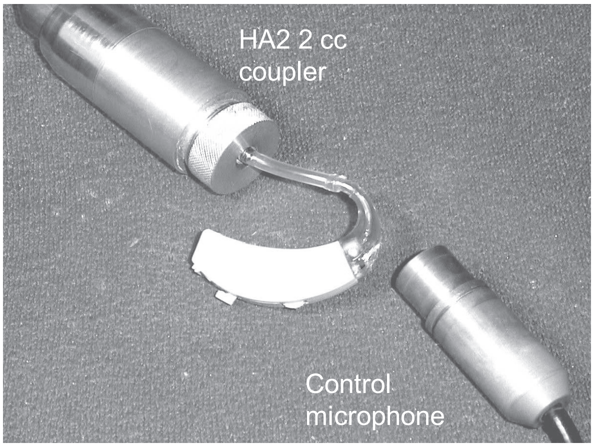


Figure 4.4 A hearing aid connected to a coupler, with a control microphone positioned next to the hearing aid microphone.

4.1.2 Test boxes

The coupler and ear simulator provide a way for the *output* from the hearing aid to be measured. Just as important is the means to put a controlled sound *into* the hearing aid. A ***test box*** generates sounds of a required SPL at the hearing aid microphone. A test box includes a tone and/or noise generator, an amplifier, a loudspeaker, and a control microphone. The ***control microphone*** (also called a ***reference microphone***) is placed next to the hearing aid microphone, as shown in Figure 4.4. The control microphone monitors the SPL reaching the hearing aid from the loudspeaker. If the input level is higher or lower than the desired level, the control microphone circuit automatically turns the volume of the sound coming from the test box speaker down or up, respectively, until the required level is obtained.

The control microphone works in one of two ways. With the ***pressure method***, the control microphone is placed as close as possible to the hearing aid microphone while the measurement is taking place. The control microphone does its job of correcting the field during every measurement. With the ***substitution method***, the control microphone is placed in the test position *prior* to the actual measurement. During a calibration measurement, the control microphone measures the SPL present at each frequency, and stores any discrepancy between the actual and desired SPLs. During all subsequent measurements, the test box adjusts its outputs to compensate for these discrepancies.[d]

As well as providing a connection and a home for all the bits and pieces needed to measure a coupler or simulator response, the test box performs two other important functions. First, it attenuates ambient noise by having a lid that seals well to the box, by being constructed with solid, dense walls, and by containing absorbent material inside. Second, it minimises

[c] Both 2-cc couplers and ear simulators include a very fine pressure equalization hole, created by drilling a hole and then partially filling it with a fine wire. With some designs it is possible for the fine wire to be accidentally removed by someone fiddling with the device.

[d] Because the pressure method removes any diffraction effects caused by the hearing aid, but the substitution method does not, the two methods can give different results for high-frequency sounds. If the substitution method has to be used, then the results can be made to agree by performing both calibration and measurement with the control microphone and the hearing aid next to each other, as with the pressure method. A complication is that with some equipment, there is only one microphone, and it is used as both the control microphone and as the coupler microphone. In such cases, when the microphone is moved from the control position to the coupler to measure the hearing aid's output, a dummy microphone that matches the size of the real microphone has to be positioned at the place just vacated by the real microphone. Annex A of ANSI S3.22 refers to this as the Equivalent Substitution Measurement and gives a full description.

Practical tip: quick calibration check

A full calibration of a test box will check that the correct input levels are generated and that the SPL at the coupler or simulator microphone is correctly displayed. The following is *not* a substitution for a full calibration, but is a valuable quick and simple check:

- If possible, take the coupler or simulator off the microphone. Otherwise, take any earmold simulator off the coupler or ear simulator.
- With test boxes that use a separate control microphone, place the control microphone next to the coupler microphone and measure the frequency response with an input level of 90 dB SPL. With test boxes that use the substitution method, *Calibrate* or *Level* first, and then measure the frequency response for a 90 dB SPL input level with the measuring microphone at the calibration or levelling position.
- If the coupler/simulator could be removed, the output measured by the coupler/simulator microphone should be 90 ± 2 dB SPL at all frequencies.
- If the coupler/simulator could not be removed, then the output should be 90 ± 2 dB up to 500 Hz.

This quick procedure will *not* reveal an improperly calibrated test box if the control microphone and the coupler/simulator microphone are both out of calibration by the same amount, and in the same direction. With two-microphone systems, this is less likely to happen than just one of the microphones becoming out of calibration.

the amount of reflected sound that reaches the hearing aid. The resulting decrease in reflections, and hence in standing waves, makes it easier for the control microphone to achieve the desired SPL at the hearing aid input.

For omni-directional microphones, when using the pressure method of calibration, it is important only that the control microphone and the hearing aid microphone be close to each other and the same distance away from the loudspeaker. (This arrangement makes sure that neither of them will provide an acoustical barrier that could cast an acoustic shadow over the other one.)

For directional microphones, it is important that the sound from the loudspeaker hits the hearing aid at the same angle that it would when the hearing aid is being worn and the source is directly in front of the person wearing the hearing aid. Often, this will mean that the line joining the two inlet ports of the directional microphone will pass through the center of the loudspeaker, with the front port closest to the loudspeaker. Make sure you know whether the test box has the loudspeaker in the lid, base or front of the test box, and whether it is centered or off to the side! Unless the test box has specifically been designed to measure directional hearing aids with the lid closed, directional hearing aids should be measured with the lid of the test box open, in a quiet room. (An open lid decreases the strength of any reflections arriving at the hearing aid from the wrong direction.) Directional hearing aids will usually have to be supported in the desired orientation by a piece of putty or absorbing foam, whereas omni-directional hearing aids can just lie flat on the surface of the test box.

If the gain obtained with the directional hearing aid "facing" the loudspeaker is not substantially greater across the low and mid frequencies than the response obtained with the hearing aid facing away from the loudspeaker, the test box is not suitable for measuring the hearing aid in the directional mode. As reflections will usually interfere with the measurement, it is usually not possible to accurately measure the directivity pattern of a directional hearing aid in a test box.

The measurements are useless unless the microphone(s) in the test box have been calibrated, and the calibration checked at regular intervals. Opinions vary about what those regular intervals should be, but a full calibration once every two years, plus a quick calibration check (see above panel) once a week may be reasonable.

4.1.3 Measurement signals

Test boxes use one (or both) of two different types of measurement signals. The traditional measurement signal is a ***pure tone*** that automatically sweeps in frequency over the desired frequency range (typically from 125 Hz to 8 kHz). For several reasons, it is more appropriate to measure modern hearing aids with ***broadband*** test signals. These signals have a wide range of frequencies present simultaneously. The test box uses a Fourier Transform (see Sections 1.2 and 2.4.5) or a swept filter to determine the level in each frequency region of the signal coming out of the hearing aid. Because the analyzer stores the level of each frequency component at the input to the hearing aid, it can calculate the gain at each frequency. If the hearing aid is operating linearly, measurement with pure tones will give exactly the same gain-frequency response as measurement with a broadband signal, no matter what the spectral shape of the broadband signal. Why then is it worth using the more complex measurement stimulus?

Most hearing aids intentionally do *not* amplify linearly over a wide range of input levels. In hearing aids currently available, the most common cause of nonlinearity is compression which, as we have seen in Section 2.3.3, involves an amplifier whose gain depends on the input signal. Suppose a hearing aid amplifier includes a high-pass filter (i.e. a low tone cut) followed by a compressor. If such a hearing aid were to be measured with a swept pure tone signal, then as frequency increases, the signal level passed by the filter to the compressor would increase. Consequently, the compressor would increasingly turn down the gain, thus partially (or even wholly) undoing the effect of the filter. However, if a broadband signal of any fixed spectral shape[e] were to be input to the aid, the compressor would settle down to a particular gain. Analysis of the output spectrum would reveal that the filter had its full effect on the spectrum of the input signal. Swept pure tones and broadband noises would thus reveal very different response shapes.

Which is the *real* response of the hearing aid? Neither! Real input signals, such as speech sounds, are not narrowband signals like swept pure tones, nor are they signals whose spectra remain fixed with time. Rather, they are signals with a complex spectrum that varies in shape from moment to moment. If the compressor changes gain rapidly compared to the duration of speech syllables, the response measured with the broadband input signal will not show how the levels of two succeeding sounds with different spectral shapes are affected. Thus neither measurement tells the full story, but overall the measurement made with the broadband stimulus is more realistic.

If we imagine a more complex hearing aid, such as the three-channel compression aid shown in Figure 2.2, the gain-frequency response measured will also depend on the shape of the input spectrum. Imagine that two different signals are used: Signal A with intense components in the low frequencies and Signal B with intense components in the high frequencies. For Signal A, the compressor in the low-frequency channel will turn its gain down greatly, but for Signal B, the peak clipper in the high-frequency channel will clip heavily.

The most realistic assessment of the effect of a hearing aid on speech will occur when the input spectrum has a spectrum similar to that of speech. Broadband signals used in test boxes thus usually have such a spectrum. Test signals, all with a long-term spectrum matching that of speech, include:

- spectrally shaped random noise;
- a repetitive waveform with a ***crest factor*** (the ratio of a waveform's peak value to its rms value) similar to that of speech, one example of which is ***pseudo-random noise*** (ANSI S3.42);
- a series of very short tone bursts that vary rapidly in frequency and amplitude to match both the spectrum and dynamic range of speech;
- speech sounds that have been processed to remove the fine detail that provides most of the intelligibility while retaining the temporal fluctuations in amplitude of real speech, such as one of the ICRA noises;[475]
- speech syllables extracted from multiple languages and pasted together to sound like speech, referred to as the International Speech Test Signal (ISTS; IEC 60118-15);[757]
- actual continuous speech.

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[e] Signals that have a spectral shape that does not vary with time (other than the random fluctuations that occur in noise signals) are referred to as *stationary* signals.

Unfortunately, there are several representations of the long term speech spectrum. One well-researched spectrum is based on measurements of 21 different languages/accents around the world,[227] and one commonly used spectrum is a simplification of older, less comprehensive measurements of speech and has been incorporated in standards.[53] With typical compression ratios of around 2:1, measurement signals based on the ANSI S3.2 standard lead to high-frequency gains around 5 dB lower than those based on the internationally derived long-term speech spectrum.[877]

Matching the long-term spectral shape to speech is a good start, but the dynamics of the test signal are also important.[1586] Wide dynamic range compression will cause slightly different gains to be measured for speech than for test signals with less modulation. The gain measured with unmodulated test signals exceeds the gain measured with speech by increasing amounts as the compression ratio, the ratio of release time to attack time, and the number of channels increases. [720]

For hearing aids with multiple bands of compression (which nearly all hearing aids have) gain measured with a pure tone sweep will be less than the gain measured with a speech-shaped signal, because all the signal power present at any instant resides at one frequency, and hence falls in a single hearing aid channel. The signal power in this channel will be much greater than would occur for a broadband sound at the same SPL, so the compressor reduces the gain more for the pure-tone signal. Differences increase with the number of channels of compression, and are greatest for the high frequencies (where speech sounds have the weakest power per channel).

Even more marked differences in gain occur for hearing aids with adaptive noise reduction. These hearing aids are designed to decrease their amplification in each channel as the SNR in each channel decreases (see Section 8.1). Such hearing aids may treat swept pure tones and stationary noise test signals as though they were background noise and decrease their amplification accordingly. The amplitude fluctuations in the test signal are therefore particularly important. Indeed, the magnitude of the adaptive noise suppression may be estimated by the difference in gain measured with a modulated versus an unmodulated test signal.[475] Hearing aids with feedback cancellation (most now have this) may treat swept pure tones as feedback oscillation and decrease the gain at the measurement frequency.

The more advanced the hearing aid being tested, the more important it is for the test signal to simulate those characteristics of real speech used by the hearing aid to control its amplification characteristics. Real speech is obviously a valid signal, the only problem being that its highly variable nature may require a measurement to last for 30 seconds or more before the gain-frequency response averaged over the measurement time is stable.

4.1.4 Gain-frequency response and OSPL90-frequency response

The measurements most commonly performed on hearing aids are the gain-frequency response and OSPL90-frequency response. Figure 4.5 shows an example of each, obtained with a BTE hearing aid in an HA2 style 2-cc coupler and measured with a swept

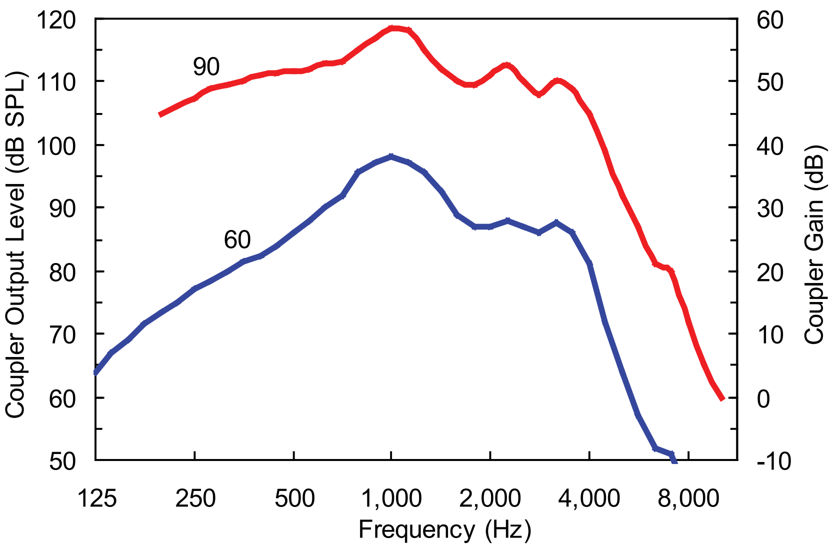

Figure 4.5 Gain-frequency response (measured with a 60 dB SPL input level) and OSPL90-frequency response of a BTE measured in a 2-cc coupler with a swept pure tone. The 60 dB curve can be read against either vertical axis; the OSPL90 curve must be read against the left hand axis.

Terms used to summarize the gain-frequency response

High-Frequency Average (HFA) Gain: Average of the gains at 1000, 1600 and 2500 Hz (ANSI S3.22).

Special Purpose Average Gain: Average of the gains at three frequencies, each separated by 2/3 octaves. This is used for hearing aids with unusual frequency responses (ANSI S3.22).

Frequency Range: This is the range of frequencies between the lowest and highest frequencies whose gains are 20 dB below the HFA gain (ANSI S3.22).

pure tone signal. The gain-frequency response was obtained with an input signal level of 60 dB SPL. The results can be shown with either of two different, but related, vertical axes. The left-hand axis shows the output in dB SPL. The gain at any frequency can be calculated as the output SPL at that frequency minus the input SPL. This gain is shown directly on the right-hand axis in Figure 4.5.

Both IEC 60118-0 and ANSI S 3.22 standards specify that hearing aid maximum output should be measured using a 90 dB SPL input signal, and both standards use the term OSPL90 to describe the measurement. This level is high enough to cause most hearing aids to reach their highest possible output level at each frequency. When the hearing aid output has reached its limit for any input signal, it is said to be ***saturated***. Hearing aids with a steeply rising response will often not be saturated at low frequencies, so in such instances, the measurement will underestimate the true maximum output of the hearing aid at low frequencies. At mid and high frequencies, for many hearing aids, there will be an increase in output level as the input rises from 90 to 100 or 110 dB SPL, but this is nearly always small enough to be of no consequence. The vertical axis of the OSPL90-frequency response graph is always shown in dB SPL.

When a broadband measurement signal is used, the results are always meaningful if the vertical axis shows gain. The problem with showing output level is that broadband signals (whether the input or output signal) have their power spread continuously across frequency, so SPL can be measured only by combining all the energy within some finite analysis bandwidth.[f] The bigger the analysis bandwidth, the greater will be the SPL measured. The actual SPL thus depends on whatever analysis bandwidth the designer of the test equipment arbitrarily chooses. One solution is to show the level as the SPL that exists in each band of frequencies 1 Hz wide.[g] The most commonly used solution is to show the level that exists in each one-third octave band. These complications do not arise if the vertical axis displays gain, because the same analysis bandwidth is used to measure the input signal and the output signal. Consequently, the gain is largely independent of the analysis bandwidth chosen, and is less dependent on the input spectrum than is the case with the output signal.

The differing characteristics of narrow and broadband signals cause a further complication when measuring the maximum output of a hearing aid. The OSPL90 measured with a swept pure tone indicates how large a signal the hearing aid can produce at each frequency when all the power of the hearing aid is concentrated into that same narrow frequency region. The output measured with a broadband signal, by contrast, indicates how large a signal the hearing aid can produce in *each* frequency region when it is simultaneously producing signals in *all* frequency regions. The rela-

[f] In fact, for random noise, sound pressure exists at every frequency, so there is an infinitesimal amount of sound pressure at any particular frequency.

[g] This is sometimes expressed as *SPL per √Hz*, but this is a misleading expression as it is the underlying pressure density (*Pa/√Hz*) that must be multiplied by the square root of bandwidth to obtain SPL. To convert an *SPL per √Hz* value to SPL in a wider band (such as might correspond to one channel in a multichannel hearing aid), add *10 log(B)*, where *B* is the bandwidth in Hz of the channel concerned. The situation is even more complex if the level in each 1 Hz band changes significantly within the channel.

tionship between these quantities depends on whether the maximum output of the hearing aid is determined separately within each hearing aid channel, or determined after (or before) the different frequency components of the signal are filtered into the hearing aid's channels. Compared to the maximum output measured with pure tones, broadband sounds underestimate the maximum output at each frequency for single-channel limiting devices, and overestimate the broadband output (i.e. total SPL) for multi-channel limiting devices.

The amount of gain measured with a hearing aid depends on where the volume control and all the fitter controls are set. The volume control should either be in the full-on position, in which case the ***full-on gain*** is obtained, or else should be at the ***reference-test setting***, in which case the resulting gain curve is referred to as the ***basic frequency response*** (IEC 60118-0) or the ***frequency response curve*** (ANSI S3.22). The purpose of reducing the volume control to a reference position is to set the hearing aid so that it is not saturated for mid-level input signals. The reference position is achieved when the high-frequency average (HFA; average of 1.0, 1.6 and 2.5 kHz) output, measured with a 60 dB SPL input signal, is 17 dB below the HFA OSPL90. In all cases, OSPL90 is measured at the full-on gain setting. For measurement of both gain and OSPL90, all other controls are usually set to the position that gives the widest frequency response with the greatest average gain. These settings must be recorded on the measurement; otherwise, the measurement is meaningless.

The two standards emphasize measurement of the gain-frequency response for an input level of 60 dB SPL. For non-linear hearing aids, it is more meaningful to display the gain for each of a range of input levels. Two commonly chosen sets of levels are: 50, 60, 70, 80 and 90 dB SPL (ANSI S3.42, IEC 60118-15); and 50, 65, and 80 dB SPL.

4.1.5 Input-output functions

Whereas a gain-frequency response shows the gain (or output level) versus frequency for one input level, an ***input-output function*** shows the output level versus input level for one frequency or for one broadband test signal. It is thus the same type of data, but is displayed in a different manner. Because all hearing aids become nonlinear at high input levels, and because many are nonlinear over a wide range of input levels, the input-output (I-O) function is an invaluable tool for understanding how a hearing aid modifies sound. Let us examine what we can learn from an I-O function.

Figure 4.6a shows the I-O diagram for a hearing aid with two compressors. Also shown are some lines that would correspond to the I-O function for a linear hearing aid with different amounts of gain. Note that all of these dotted lines are at an angle of 45°. Look in particular at the dotted line labeled *30*. For

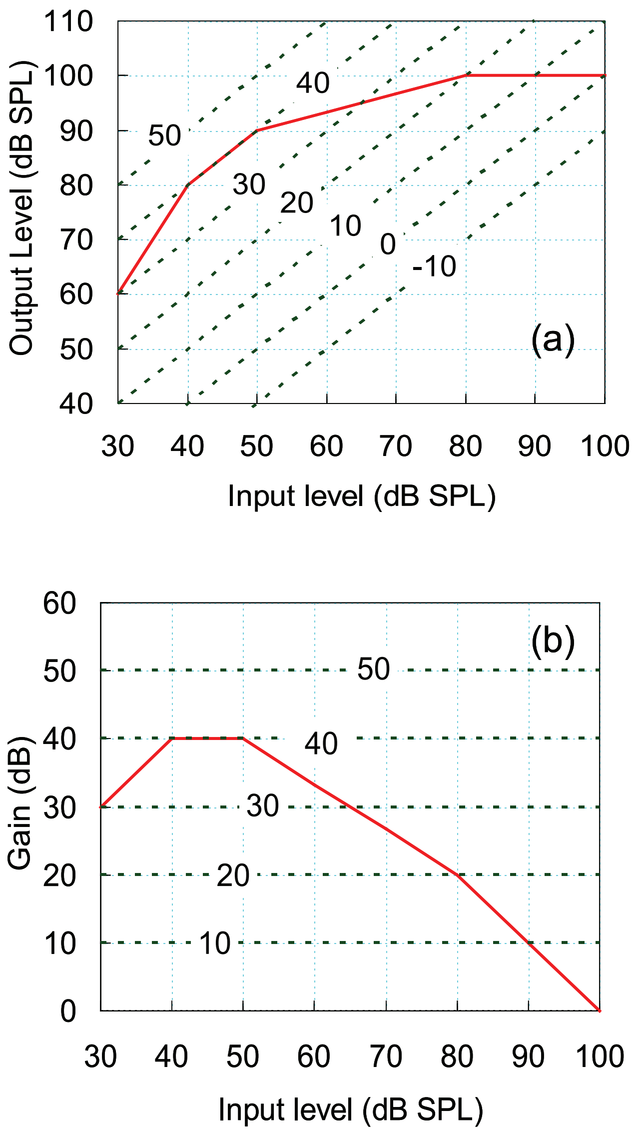

Figure 4.6 (a) Input-output diagram of a non-linear hearing aid at 2 kHz (solid red line) and lines of constant gain (dotted lines). (b) Gain-input diagram of the same amplification characteristics.

Understanding gain, attenuation, compression, and expansion on the I-O diagram

Make sure that you really understand the four terms *gain*, *attenuation*, *compression*, and *expansion*.

- Gain and attenuation each describe how large the output signal is, compared to the input signal. They correspond to different regions on the I-O diagram: above and below the 0-dB gain diagonal, respectively. They also correspond to different regions on the gain-input diagram, above and below the horizontal axis respectively.
- Compression and expansion each describe the effect of the amplifier on the dynamic range of a signal that varies in amplitude over time. They correspond to different slopes on the I-O diagram: less than and greater than 45° respectively. Linear operation corresponds to a slope of exactly 45°. On the gain-input diagram compression and expansion correspond to a negative slope and positive slope, respectively.

A compressor makes the signal's dynamic range smaller, regardless of whether the output level is smaller or larger than the input level. By contrast, an expander increases dynamic range.

For a small range of levels around a given input level a hearing aid can, in principle, simultaneously:

- amplify and compress,
- amplify and expand,
- attenuate and compress, or,
- attenuate and expand.

In practice, amplification combined with expansion at low levels, linear amplification at low to mid levels, and compression at mid and high levels is most common.

every point on the line, the output is 30 dB more than the corresponding input. The line thus represents the I-O function for a hearing aid with a fixed gain of 30 dB. Notice that the upper (or left-most) lines have the greater gain. Lines below the 0-dB gain diagonal have negative gains, and thus represent attenuation by the hearing aid. As a general principle, gain increases for movements vertically upwards or horizontally to the left, or both simultaneously, on an I-O diagram. Gain remains constant for movement simultaneously upwards and to the right at 45°.

Now turn your attention to the hearing aid's I-O curve in Figure 4.6a. It comprises four sections. For input levels between 40 and 50 dB SPL, the hearing aid behaves in a linear fashion, with a constant gain of 40 dB. Above the linear region, from 50 to 80 dB input level, the line still slopes upwards, but with a slope of less than 45°. Any increase in input level thus results in a smaller increase in output level. This effect, of course, is compression. For an input level of 50 dB SPL, the output is 90 dB SPL and the gain is therefore 40 dB. As the input level increases, the I-O function crosses several lines of constant gain. The hearing aid gain is thus decreasing (as one would expect for a compressor) and when the input level reaches 80 dB SPL, the gain has decreased to 20 dB.

The highest level section of the curve is horizontal, which is referred to as ***limiting*** because the output cannot rise above a certain limit, in this case 100 dB SPL. From the I-O diagram alone, we cannot tell whether this limit is set by peak clipping or by compression limiting. As the input level increases beyond 80 dB SPL, the gain decreases further. In fact, the gain decreases by 1 dB for every 1 dB increase in input level. The hearing aid begins to act as an attenuator (i.e. an earplug) at this frequency for input levels greater than 100 dB SPL, where the output level becomes smaller than the input level.

For input levels less than 40 dB SPL, ***expansion*** (the opposite of compression) is occurring. Whenever the input level decreases, so too does the gain. Expansion, which is also called ***squelch*** and ***noise-gating***, is used in some hearing aids, and is useful for decreasing the audibility of very low-level sounds, including hearing aid internal noise. Such a reduction in audibility is good as long as all the noises made inaudible really are unwanted noises. If the expansion threshold is too high (e.g. 55 dB SPL) expansion will reduce the gain of soft speech sounds and adversely affect speech intelligibility.[1926]

The gain-input curve shown in Figure 4.6b shows exactly the same amplification characteristic as Figure 4.6a. To understand this curve, follow the shape of the curve as you re-read the description in the previous paragraphs of how gain changes with input level.

Measuring the I-O curves or gain-input curves for two different settings of the volume control will reveal how the volume control affects the operation of the compressor, as we will return to in Section 6.2.3.

4.1.6 Distortion

The concepts of harmonic distortion and intermodulation distortion have already been introduced in Section 2.3.2 in the context of peak clipping. Mechanisms other than peak clipping can give rise to distortion within hearing aids, but peak clipping is the most common cause and produces the largest amounts of distortion. ***Harmonic distortion*** is measured by putting a pure tone into the hearing aid, and then analyzing the output waveform to measure the distortion components relative to the total power of the signal.

The relative size of the distortion components can be expressed in a few ways. First, it can be expressed in dB or it can be expressed as a percentage.[h] Second, the distortion can be expressed separately for each harmonic (usually just the second and third), or it can be expressed for all the harmonics summed together. When the power (which is proportional to the square of the pressure) in all the harmonics is summed, the final number is referred to as ***total harmonic distortion (THD)***:

$$THD\ \% = 100\sqrt{\frac{p_2^2 + p_3^2 + p_4^2 + ...}{p_1^2 + p_2^2 + p_3^2 + p_4^2 + ...}}\ \% \qquad .... 4.1,$$

$$THD\ dB = 10\log_{10}\left(\frac{p_2^2 + p_3^2 + p_4^2 + ...}{p_1^2 + p_2^2 + p_3^2 + p_4^2 + ...}\right)\text{dB} \qquad ..... 4.2,$$

where p_n is the pressure of the n'th harmonic. The first harmonic (of amplitude p_1) is the fundamental (the frequency of the input signal), which represents the undistorted part of the signal. The clinician will never have to use Equations 4.1 and 4.2; test boxes do these calculations automatically and display the results.[i] A distortion of 1% is equivalent to -40 dB, 3% is equivalent to -30 dB, 10% to -20 dB, and 30% to -10 dB.

The standards specify that distortion should be measured with medium level signals (60-70 dB SPL, depending on frequency) with the volume control set to the reference test position. It is just as relevant however, to know what distortion occurs at lower and, particularly, higher input levels. The distortion results may be displayed as distortion versus frequency at a particular input level, or distortion versus level at a particular frequency.

Harmonic distortion measurement can be a misleading indicator of hearing aid distortion for low and high input frequencies. For high input frequencies, the harmonics that the distortion adds to the pure-tone input signal will all fall above the response range of the receiver, and so will not be discernable in the acoustic output, although the hearing aid amplifier may be clipping heavily. The distortion will, however, be audible and objectionable when a more complex (broadband) input signal is used and the distortion products occur at many frequencies throughout the audio range. (See intermodulation distortion in Section 2.3.2.) For low frequencies, a hearing aid with a steeply rising response will emphasize the harmonics of a low-frequency signal if the peak clipping in the hearing aid precedes the filter that causes the steeply rising response (which might be the receiver and tubing system). In this case, the distortion for broadband signals will not be as bad as one would expect based on the harmonic distortion measures.

[h] A square root is taken before calculating the percentage so that the final ratio refers to sound pressures, or voltages, rather than intensities or powers.

[i] ANSI S3.22 allows an alternative and preferred distortion formula, in which only the power of the fundamental component appears in the denominator. The two versions give almost identical results for THD less than 20%.

A method of measuring distortion applicable to broadband signals is the ***coherence*** between the input signal and the output signal. Coherence quantifies the proportion of the output signal at each frequency that is linearly related to the input signal at the same frequency. It ranges from 1, when there is no noise or distortion, down to 0, when the output is not at all linearly related to the input. Coherence measurements can be adversely affected by hearing aid delay and fast acting compression. A measure similar to THD can be deduced from coherence as follows:[1461]

$$THD\ coherence = 100\sqrt{\frac{1 - coherence}{coherence}}\ \% \qquad \text{..... 4.3.}$$

Distortion can be measured for the following purposes:

- ensuring that a hearing aid continues to meet its published specifications – measurements may be made following a repair on the hearing aid, or in response to adverse comments by the aid wearer about the sound quality;
- comparing the fidelity of two different hearing aids;
- establishing whether a hearing aid employs compression limiting or peak clipping (because it cannot be deduced from the I-O function) – the THD of compression limiting hearing aids should always be less than 10%, whereas the THD of peak clipping hearing aids will rise rapidly above this once peak clipping commences; or
- determining the highest input level that can be passed through the hearing aid without significant distortion (particularly important for music; see Section 10.6.1).

4.1.7 Internal noise

As mentioned in Chapter 2, microphones and amplifiers generate noise. The internal noise of a hearing aid is quantified by expressing it as the ***equivalent input noise (EIN)***. The EIN is the amount of noise that would have to be applied to the input of a noiseless hearing aid with the same gain-frequency response, if the noise coming out of this noiseless hearing aid were to be the same as that coming from the hearing aid under test. It is sensible to express noise relative to the input of the hearing aid for three reasons. First, most of the noise in a well-designed hearing aid originates from the microphone, and most of the remaining noise originates from the input amplifier. Second, and because of the origin of the noise, the output-referred noise will vary markedly with the position of the volume control, whereas input-referred noise will be less affected by the position of the volume control and other fitter controls. Third, if the noise were expressed as the output noise, high-gain hearing aids would always be noisier than low-gain aids, even though the wearers of these aids (people with severe or profound hearing loss) may be less aware of the internal noise than the wearers of low-gain aids.

The EIN is calculated by measuring the magnitude of the noise at the output of the hearing aid and subtracting from this the gain of the hearing aid for soft sounds. Two such types of measurement can be performed. In the simpler type of measurement, the total output noise SPL is measured, and the HFA gain is subtracted. This measurement does not reveal how much noise is present in each frequency range so it cannot be used to reliably compare the noisiness of two aids with different gain-frequency responses. It is, however, suitable for ensuring that a hearing aid is operating within its specifications.

Practical tip: Measuring internal noise

- To ensure that ambient noise does not affect the measurement of output noise, close the test box lid, and if necessary, place putty over the microphone inlet port. (It ***is*** necessary if adding the putty causes the output noise to decrease.)
- When the gain is measured for the purposes of the noise measurement, the input level must be low enough for the hearing aid to be in its linear region. (This may make internal noise measurement difficult for hearing aids with a fixed very low compression threshold or with low-level expansion that cannot be disabled).
- Ideally the test box should be in a test booth, or other very quiet place, for this measurement.

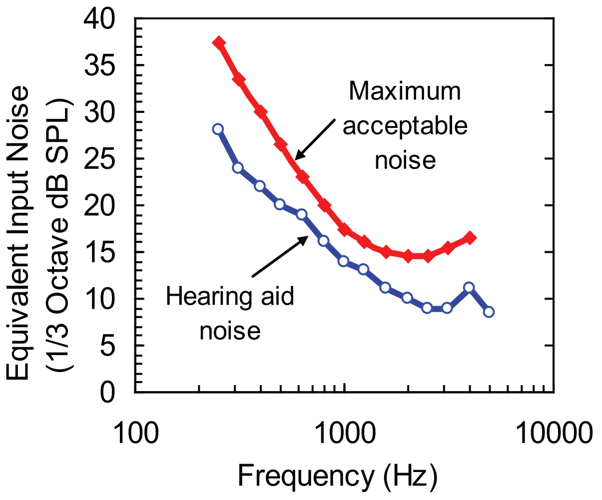

Figure 4.7 Equivalent 1/3-octave input noise of a typical hearing aid as a function of frequency, and maximum acceptable 1/3-octave noise.

A more thorough method of measuring internal noise is to filter the output signal into bands (usually 1/3 octave, or one octave) and so measure the level of the output noise that falls within each band. The equivalent input noise at each frequency is then calculated by subtracting from these output levels the gain at the center frequency of each band. The result is a graph of equivalent input noise (for the measurement bandwidth used) as a function of frequency. The input referred noise for a typical hearing aid is shown in Figure 4.7, along with the maximum equivalent input noise considered acceptable by listeners.[1112]

4.1.8 Magnetic response

The principles of magnetic induction have been explained in Sections 2.8 and 3.5. Measurement of the magnetic response is straightforward if the test box contains a loop to generate a magnetic field and impossible if it does not. The only precautions are:

- make sure that the volume control is at its reference position when measuring magnetic frequency response; and
- orient the hearing aid as it would normally be oriented in regular use.

The standards specify that the magnetic response of a hearing aid be measured with a field strength of 31.6 mA/m. The hearing aid output is referred to as the SPL for a vertical inductive field (SPLIV; ANSI S3.22) or as the SPL in a magnetic field (SPLI; IEC 60118-0).[j] The shape of the magnetic frequency response should be similar to the shape of the acoustic frequency response. There will, however, be some differences, because the coil probably will not have a resonance to match the microphone Helmholtz resonance, and because the coil response may have been given an additional low cut by the aid designer (see Section 3.5.3). The result of the magnetic response is displayed as a graph of output SPL versus frequency for the specified input magnetic field strength. The concept of *gain* does not strictly apply because the input and output quantities are different.

Because telephones are such an important source of magnetic signals, ANSI S3.22 specifies a Telephone Magnetic Field Simulator that generates magnetic signals similar in level and field shape pattern to those generated by a telephone. The output of the hearing aid is referred to as the SPL for an inductive telephone simulator (SPLITS).

ANSI S3.22 and IEC 60118-0 specify another method to compare acoustic and telecoil sensitivity. The terms ***equivalent test loop sensitivity (ETLS)*** and ***relative simulated equivalent telephone sensitivity (RSETS)*** are calculated as the output signal (SPLIV and SPLITS, respectively) for telecoil input minus the output signal for an acoustic input level of 60 dB SPL. They are intended to describe how much the user would have to change the volume control so that the acoustic output will be the same when listening via the telecoil as when listening via the microphone. The first term applies to room-loop use; the second term applies to telephone use. Values of ETLS and RSETS close to 0 dB (i.e. the volume control does not have to be adjusted) are most desirable.[1459] Unfortunately, it is common to find that the volume control has to be increased to achieve a comfortable output level with magnetic input signals.[840a] This creates a major problem when the hearing aid does not have a volume control.

[j] IEC 60118-0, in revision at the time of writing, also specifies a measurement made with an input of 10 mA/m and the hearing aid oriented for maximum sensitivity.

4.1.9 ANSI, ISO and IEC standards

Frequent references have already been made to IEC and ANSI standards, and some of their similarities and differences have been outlined. Most countries in the world adopt international standards, which comprises IEC standards and International Standards Organization (ISO) standards. IEC standards specify electrical equipment, including hearing aids and methods for measuring them, whereas ISO standards specify human-related standards such as the normal threshold of hearing, and procedures for measuring hearing. The USA has the ANSI system to accomplish both purposes. Tables 4.1 and 4.2 list several standards that are directly relevant to hearing aids. A standard published by ISO that is particularly relevant to hearing aids is ISO12124 (2007) *Procedures for the measurement of real-ear acoustical characteristics of hearing aids*, which specifies procedures for testing hearing aids on the ears of patients. This will soon be incorporated into IEC 61669.

Table 4.1 Some ANSI standards relevant to hearing aids. Year is the date of the latest revision. Date of subsequent reaffirmations are not shown.

Number	Year	Title	Comment
C63.19	2011	Methods of measurement of compatibility between wireless communications devices and hearing aids	Specifies how to rate emission levels for mobile phones and immunity levels for hearing aids
S3.7	1995	Method for coupler calibration of earphones	Defines the 2-cc coupler (HA1, HA2, HA3 and HA4) (and also the 6-cc coupler for supra-aural earphones)
S3.13	1987	An artificial headbone for the calibration of audiometer bone vibrators	Specifies the impedance and shape of an artificial mastoid used for measuring bone-conduction hearing aids
S3.22	2009	Specification of hearing aid characteristics	Specifies test conditions, procedures and tolerances for coupler measurements, including that a 2-cc coupler be used
S3.25	2009	An occluded ear simulator	Specifies the acoustic characteristics of occluded ear simulators, and shows the mechanical design of a Zwislocki ear simulator and an IEC 2-branch ear simulator
S3.35	2010	Methods of measurement of performance characteristics of hearing aids under simulated real-ear working conditions	Specifies how to use a manikin and ear-simulator to measure aided gain, insertion gain, and directivity index
S3.36	1985	Specification for manikin for simulated *in-situ* airborne acoustic measurements	Specifies both physical shape and free-field response of a manikin
S3.37	1987	Preferred earhook nozzle thread for postauricular hearing aids	Applies only to BTEs with threaded nozzles
S3.42	1992	Testing hearing aids with a broad-band noise signal	Specifies spectrum of noise approximating the speech spectrum and analysis methods using that noise
S3.46	1997	Methods of measurement of real-ear performance characteristics of hearing aids	Defines terms and specifies how to measure hearing aids on patients

Table 4.2 IEC standards relevant to hearing aids. Additional years, where shown, refers to the date of an amendment. Note that parts 1, 2 and 6 of IEC 60118 will soon be withdrawn and their updated contents incorporated into a new 60118-0.

Number	Year	Title	Comment
60 118-0	1983 1994	Hearing aids - Part 0: Measurement of electroacoustical characteristics	Specifies conditions for testing a hearing aid in a sound field, such as in a test box, including that an ear simulator be used
60 118-1	1999	Hearing aids - Part 1: Hearing aids with induction pick-up coil input	How to test telecoil response
60 118-2	1983 1993 1997	Hearing aids - Part 2: Hearing aids with automatic gain control circuits	How to measure I-O curves and attack and release times
60 118-4	2006	Hearing aids - Part 4: Induction loop systems for hearing aid purposes - magnetic field strength	Specifies 100 mA/m long-term level
60 118-5	1983	Hearing aids – Part 5: Nipples for insert earphones	Defines the dimensions of nipples for insert earphones used with body aids
60 118-6	1999	Hearing aids - Characteristics of electrical input circuits for hearing aids	Specifies impedance and sensitivity to ensure compatibility with external devices
60 118-7	2005	Hearing aids - Part 7: Measurement of the performance characteristics of hearing aids for production, supply and delivery quality assurance purposes	Specifies test conditions, procedures and tolerances
60 118-8	2005	Hearing aids - Part 8: Methods of measurement of performance characteristics of hearing aids under simulated *in situ* working conditions	How to measure a hearing aid mounted on a manikin
60 118-9	1985	Hearing aids - Part 9: Methods of measurement of characteristics of hearing aids with bone vibrator output	How to measure bone conductor hearing aids
60 118-12	1996	Hearing aids – Part 12: Dimensions of electrical connector systems	Specifies the plugs and sockets that connect to hearing aids
60 118-13	2004	Hearing aids – Part 13: Electromagnetic compatibility (EMC) product standard for hearing aids	Specifies immunity required from mobile phones for bystander compatibility and user compatibility
60 118-14	1998	Hearing aids – Part 14: Specification of a digital interface device	Specifies the interface that allows a computer to program the hearing aid
60 118-15	2009	Methods for characterizing signal processing in hearing aids	Specifies a speech-like signal, set of standard audiograms for pre-setting hearing aids, and signal analysis methods
60 318-4	2010	Electroacoustics – Simulators of human head and ear – Occluded-ear simulator for the measurement of earphones coupled to the ear by ear inserts	Specifies an occluded ear simulator. Standard replaces IEC 711 (later 60 711)
60 318-5	2006	Electroacoustics – Simulators of human head and ear – Part 5: 2 cm3 coupler for the measurement of hearing aids and earphones coupled to the ear by means of ear inserts	Defines the 2-cc coupler, and methods of coupling to it for different hearing aid styles Standard replaces IEC 126 (later 60 126)
60 318-6	2007	Electroacoustics – Simulators of human head and ear - Mechanical coupler for measurements on bone vibrators	Specifies the impedance and shape of an artificial mastoid used for measuring bone-conduction hearing aids Standard replaces IEC 60 373
61 669	2001	Electroacoustics – Equipment for the measurement of real-ear acoustical characteristics of hearing aids	Equipment for real-ear gain measurement on patients
60 959	1990	Provisional head and torso simulator for acoustic measurements of air conduction hearing aids	Specifies both physical shape and free-field response of a manikin

4.2 Real-Ear-to-Coupler-Difference (RECD)

The difference between the SPL a hearing aid delivers to the ear canal and the SPL it delivers to a coupler (for the same input to the hearing aid) is called the real-ear-to-coupler difference (RECD):

RECD = Canal SPL – Coupler SPL … 4.4.

The RECD concept is enormously useful in hearing aid fitting and audiometry, especially for fitting babies whose very small ear canals must be allowed for. Knowledge of a patient's RECD allows us to more accurately interpret hearing thresholds measured with insert earphones, and allows us to more accurately adjust a hearing aid in a coupler so that it achieves the desired performance in the patient's ear.

Although the RECD concept seems very simple (i.e. how much greater is the SPL in a real ear than in a coupler) the measurement and application of RECD is complicated by a number of issues. Chief among these reasons is that measurement of RECD has not yet been defined in international standards, so a variety of different things are all called "RECD". Many are affected by factors other than the acoustics of the ear canal and the coupler.

4.2.1 Factors affecting RECD

Ear canal volume. Recall from Section 4.1.1 the prime reason why SPL in the ear canal is larger than in the coupler: the residual ear canal has a volume smaller than that of the standard 2-cc coupler routinely used to measure hearing aids and to calibrate insert earphones for audiometry. Consequently, if a receiver attempts to push air cyclically in and out of the end of a tube, the receiver and tube will generate more pressure when the tube terminates in the ear canal than when it terminates in the larger 2-cc coupler. As female ears are, on average, smaller than male ears, female RECDs are, on average 1 to 2 dB higher than male RECDs. The difference between real ears and a 2-cc coupler occurs even for a standard insertion depth in the real ear. If the earmold is inserted deeper into the ear canal, however, the residual volume will be even smaller, the SPL generated will be even greater, and hence the RECD will be larger, at least for occluding earmolds.

Leakage, vents and open fittings. Earmolds rarely, if ever, fit in the ear without leakage. The leakage may be intentional (a vent, see Section 5.3) or may be accidental (via unintentional slit leaks). By contrast, it is easy to connect a tube to a coupler with zero leakage. As Section 5.3.1 will show, leaks and vents allow low-frequency sound out of the ear canal, which causes a smaller SPL within the ear canal, and hence reduces RECD in the low frequencies relative to a well-sealed ear canal. Leakage can easily be so great that RECD becomes negative at 250 Hz and even at 500 Hz. There is likely to be more leakage, and hence smaller (or more negative) RECD values at 250 Hz for custom molds than for foam ear tips that expand within the ear canal. The solid line in Figure 4.8 shows the RECD, derived from three experiments on adult ears, for unvented custom earmolds.[1277, 1283, 1286] The gradual decrease towards negative values as frequency decreases indicates that there was some leakage present around the earmolds for the people on whom these data were measured.

Table 4.3 shows RECD values for occluded, but typically leaking, earmolds relative to HA1 and HA2 2-cc couplers. The final row shows a typical ear simulator to coupler difference. It is different from the RECD shown in the first row primarily because there is no leakage for both the ear simulator and the coupler measurement.

Table 4.3 RECD: SPL generated in the average real ear minus the SPL generated in a 2-cc coupler (dB) for occluded, but typically leaky, earmolds.[1276, 1277, 1283, 1285, 1286] Values in row three are for deeply inserted earmolds or hearing aids.[98, 660] The final row shows the ear simulator to HA1 coupler difference.[204]

	250	500	1 k	1.5 k	2 k	3 k	4 k	6 k
Standard insertion depth, re HA1 coupler	-2.5	4.0	6.5	8.5	10.0	9.0	10.0	10.5
Standard insertion depth, re HA2 coupler	-2.0	4.5	7.0	8.0	7.5	2.5	2.5	5.5
Deep insertion depth, re HA1 coupler	6.0	8.0	10.0	12.5	15.0	19.0	20.0	23.0
Ear simulator to HA1 coupler difference	3.5	3.5	4.5	7.5	8.0	10.0	12.5	14.5

We can think of an open-canal fitting as an extremely large vent combined with retention of the ear canal resonance. Because of the large vent, the RECD for an open-canal fitting will therefore display very negative values in the low frequencies. Because of the retained resonance, the RECD for an open-canal fitting will display a value larger than for occluded earmolds in the high frequencies, particularly around the canal resonant frequency, which on average is centered at 2.7 kHz for an unobstructed canal, and slightly lower in frequency for a canal with an "open fitting" inserted. This increase in SPL, and hence gain relative to an occluded fitting, occurs because for frequencies near the ear canal resonance, the resonance causes SPL at the eardrum to be greater than at any other location in the ear canal.[k] Both the low-frequency and the high-frequency effect of the open canal on the RECD can be seen in the dashed line in Figure 4.8. Data for the effect on gain-frequency responses of open fittings relative to occluded fittings are given in Section 5.3.1 covering venting effects.

Tubing. The diameter of tubing affects the amount of vibrating air that flows back and forth between the tubing and the ear canal or coupler. (This flow is technically called the ***volume velocity.***) If the tubing profile used for the real-ear part of the RECD measurement is different than for the coupler part, then the high-frequency RECD will apparently change, although this really reflects the change in the sound delivery system rather than something inherent to ear canals versus couplers. Because HA2 2-cc couplers are very convenient to use, they often are used for the coupler part of an RECD measurement, even though the wide-bore earmold simulator they contain rarely realistically simulates the actual hearing aid earmold or tubing used for audiometric measurements. Because of the widening of the sound path in a HA2 coupler, a greater high-frequency SPL is generated in the HA2 coupler than the HA1 coupler, so RECD values will be smaller in the high-frequencies (by about 7 dB at 4 kHz) for a HA2 coupler than when a HA1 coupler has been used, as shown in Table 4.3.

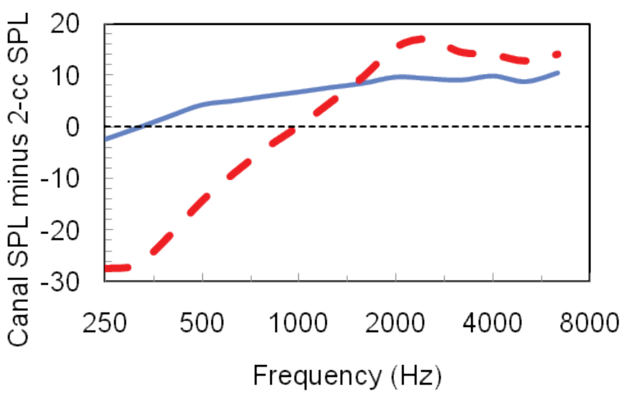

Figure 4.8 RECD: SPL generated in the average adult real ear canal minus SPL generated in an HA1 2-cc coupler. Data for the occluded ear (solid line) are averaged across three experiments.[1277, 1283, 1286] Data for the open canal fitting (dashed line) were measured in KEMAR relative to the SPL in an HA1 2-cc coupler.

Transducer type. The RECD value will reflect the differing volume (and hence impedance) of the ear canal relative to the coupler only if the transducer and tubing system used delivers the same volume velocity in each case, irrespective of the differing "loads" that the canal and coupler present to the tubing system. The volume velocity coming from the source (the receiver and tubing) will in fact be unaffected by the load (the ear canal or coupler) only if the acoustic impedance of the source is much greater than that of the load. When this is true, the RECD will depend only on the acoustic properties of the ear and the coupler, and not at all on the source used to measure it. At any frequency for which the apparent source impedance does not greatly exceed the ear canal impedance, RECD will be smaller than that expected from the difference in effective volume of the ear canal relative to the coupler.

Unfortunately, a transducer such as the ER3A insert earphone that is often used to measure RECD does not have an impedance many times higher than that of ear canals and couplers at all frequencies.[1286, 1544] BTE hearing aids can also be used as the transducer, but if they have no damping in the earhook, they may even have an impedance *lower* than that of the ear around their resonant frequencies.[1286, 1544] Consequently, the RECD *does* vary with the transducer used to measure

[k] The unoccluded ear canal resonance corresponds to that of a tube almost closed at one end and open at the other. When the ear canal is occluded, the nature of the resonance is changed to that of a tube almost closed at both ends, and its frequency is moved to a much, much higher frequency, outside the frequency range of hearing aids.

it.[1283, 1285, 1286] RECD can vary by up to 10 dB across transducers. The effect of the transducer is greatest when a hearing aid with an undamped earhook is used, when the tubing connected to the earmold is longest, and when an HA2 coupler is used. The problem is therefore greater for adults' earmolds than for infants' earmolds.

This measurement error can be reduced by performing the RECD measurement in as similar a manner as possible to the application to which the RECD measurement will be applied. For correction of thresholds measured with an ER3A insert phone through foam tips, for example, there will be ***no*** systematic measurement error if the RECD is also measured using an ER3A insert phone, connected to foam tips for the real-ear part of the measurement, and connected to the same type of coupler used to calibrate the audiometer. For application to hearing aid adjustment, RECD should preferably be measured using the patient's own mold and own hearing aid. The latter is often not feasible, but error can be minimized by fitting a hearing aid with a damped sound delivery system. Despite the potential for systematic errors with particular combinations of transducers and sound delivery systems to the ear, it is commonly the case that application of the RECD process can give highly accurate predictions of real-ear SPL.[1211, 1277, 1278, 1280, 1587, 1598]

The combination of different canal volumes, different insertion depths, different degrees of leakage, different middle ear impedances, and the interaction of these with different measurement systems (transducer, tubing, earmold/eartip, and coupler type) creates an enormous range of RECD values across individuals. Some of this range is real, and is the reason for measuring RECDs (at least for infants), some of it is just measurement error and the mixing up of RECDs measured, and hence defined, in different ways. One experiment showed up to 40 dB variation in RECD across adult subjects, a good deal of which is undoubtedly measurement error.[1555] In the mid frequencies, variations in RECD are determined by the equivalent volume of the ear, so a 40 dB variation in RECD would correspond to the largest ears being 100 times the volume of the smallest ears. In the absence of measurement error, this could be true only if the research subjects accidentally included a different species, perhaps about the size of an elephant or a mouse.

By contrast to the situation with insert earphones, supra-aural earphones such as the TDH39/49 have a low acoustic source impedance, so the volume of the ear canal has relatively little effect on the SPL generated within it. Of course, inter-person variations in leakage around the earphone still cause variations in low-frequency SPL, and inter-person variations in ear canal length still cause variations in high-frequency SPL, but the mid-frequency SPL is very predictable. The RECD measured with high-impedance insert phones thus has no relevance to measurements made with low-impedance supra-aural phones. The *concept* of RECD is still relevant for supra-aural phones, but the coupler is different, and so too are the mean values of RECD. Measurement of RECD is more difficult because of the size of the earphones.

4.2.2 Measurement of RECD

There is value in measuring RECD if it is likely that the ear canal is significantly different from that of an average adult. The two categories of clients for whom differences are most likely are children younger than about five years of age, and people who have had surgery on their ear canals. RECD is most easily measured with a real-ear gain analyzer that has been designed to perform RECD measurements. The probe microphone that records the SPL for the real-ear part of the measurement should be inserted just as described in Section 4.3.1 for measurement of real-ear aided gain. Some older real-ear gain analyzers will also require you to use the probe tube for the coupler part of the measurements, but more modern equipment will use the coupler microphone for this part of the measurement.

If RECD is measured to allow for the effects of individual ear canal acoustics on hearing thresholds, then individual earmolds will usually not be available and the measurement will be made using foam or other flexible ear inserts. Figure 4.9a shows the real-ear part of the measurement. The coupler part is shown in Figure 4.9b if a HA1 coupler is used, and in Figure 4.9c if a HA2 coupler is used.[1]

[1] The 25 mm length of tubing used when measuring BTE hearing aids should be removed, because unlike when an earmold and tubing is involved, the HA2 coupler already allows for the length of the tube inside the foam ear tip.

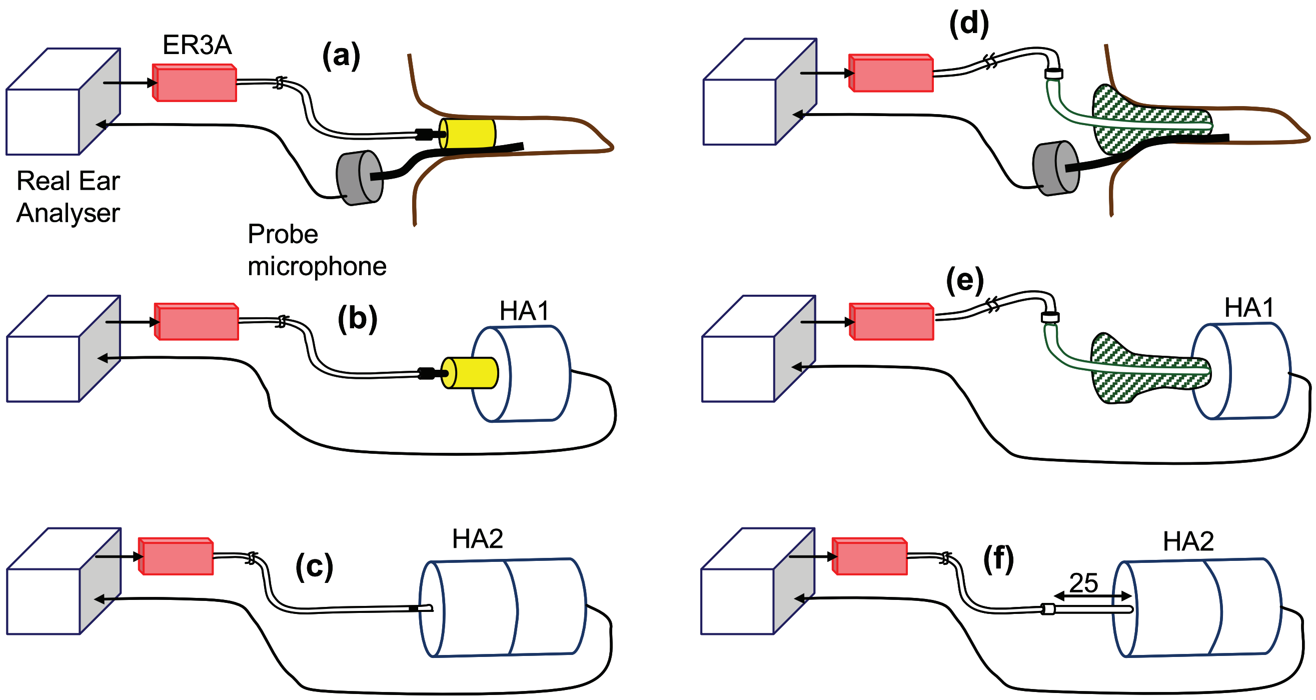

Figure 4.9 Measurement of RECD using a real ear analyser, insert phones (e.g. ER3A), and a probe microphone. For ITEs, or for correction of hearing threshold measurements, the insert phone is connected (a) the patient's ear via a foam tip, and either (b) a HA1 coupler via a foam tip, or (c) directly to a HA2 coupler. For BTEs that will be fit with a custom earmold, the insert phone is connected to (d) the patient's ear by the individual earmold and tubing, and either (e) a HA1 coupler via the individual earmold and tubing, or (f) a HA2 coupler by 25 mm of tubing. Note that the RECD values obtained with a HA1 coupler will be different from those obtained with a HA2 coupler.

If RECD is measured to allow for the effects of individual ear canal acoustics when adjusting the hearing aid, and if the hearing aid is a BTE that will be fit with a custom earmold, then it is better to include the patient's actual earmold in the measurement, as shown in Figure 4.9(d). Parts (e) and (f) show the corresponding 2-cc parts of the measurement for HA1 and HA2 couplers respectively.

An alternative method of measuring RECD, available with just some hearing aids, is to use the hearing aid itself, in conjunction with the software used to program the hearing aid. The measurement is enabled by positioning a probe tube in the ear canal, with the other end of the probe tube connected to either:

- a hearing aid microphone inside the faceplate of an ITE hearing aid or case of a BTE hearing aid;[1938] or
- a special boot plugged into the direct audio input of the (BTE) hearing aid.

In either case, the hearing aid can then generate and deliver the stimuli, measure the response in the ear canal, and use the RECD measurement to adjust the hearing aid to allow for the individual ear canal characteristics. The only limitation in accurately predicting the final hearing aid response is that if the stimuli are generated electronically within the hearing aid, transmission of sounds through the vent into the residual ear canal is not measured in the individual ear (Section 5.3.1).

Because patients often have similar earmolds in each ear, and reasonably symmetrical ear canals, RECD values for the two ears usually match to within 3 dB, for both children and adults.[1276, 1279] If time or the behavior of an infant does not permit RECD to be measured in both ears, it therefore seems reasonable to use the RECD measured on one ear to calculate the gain that will be achieved in both ears.

There is great value in measuring RECD for infants, for whom RECD values are significantly higher than for adults (Section 16.2.2), but great care must be taken in the measurement and in the way the measurement is used. Otherwise, the measurement, or its application, has the potential to create errors larger than the inter-person variations it seeks to allow for in the obtained:

- ***high frequency gain*** – if the probe is not inserted sufficiently deeply or if the wrong type of 2-cc coupler is used;
- ***mid-frequency gain*** – if an inappropriate transducer is used for the measurement; or
- ***low-frequency gain*** – if sound leaking out through a vent, and sound directly entering the ear canal via a vent, are not properly allowed for.

The need for increasing depth and precision of probe-tube placement as frequency increases are exactly the same for RECD measurement as they are for measurement of real-ear aided gain (Section 4.3.1).

4.2.3 RECD and REDD

A quantity closely related to RECD is the ***Real Ear to Dial Difference*** (***REDD***). This quantity is equal to the SPL in the ear minus the setting (in dB HL) on the dial of an audiometer:

REDD = Canal SPL – Dial HL … 4.5.

For a properly calibrated audiometer, the SPL in the coupler used to calibrate the earphones will be equal to the dial setting plus the ***Reference Equivalent Threshold SPL*** (***RETSPL***), the values for which are known for commonly used headphones, insert earphones, and couplers:

Coupler SPL = Dial HL + RETSPL … 4.6.

Equations 4.4 to 4.6 can easily be combined to show:

REDD = RECD + RETSPL … 4.7,

and this equation can also be seen from Figure 4.10 which summarizes the relationships between these quantities.

As RETSPL values are fixed for any particular earphone, REDD varies across patients to just the same degree that RECD varies.

4.3 Real-Ear Aided Gain (REAG)

It is the gain of a hearing aid in the individual's ear that matters. While the equations given in this and the next section enable real-ear gain to be estimated from coupler gain, there is always inaccuracy in the estimate caused by variations in ear sizes, fit of the hearing aid or mold to the ear canal, size of the sound tube and vent path, and the location of the microphone relative to the pinna or concha. Furthermore, there is some small variation between nominally identical hearing aids. Consequently, measurement of individual real-ear gain is important unless the fitting software can predict it within about 5 dB.

Real-ear gain is measured by placing a probe-tube, connected to a microphone, into the ear canal. There are two fundamentally different types of real-ear gain. The first of these is the real-ear aided gain, described in this section. The second is the real-ear insertion gain, described in the next section. Some prescription procedures prescribe in terms of real-ear aided gain, and some prescribe in terms of insertion gain. Most

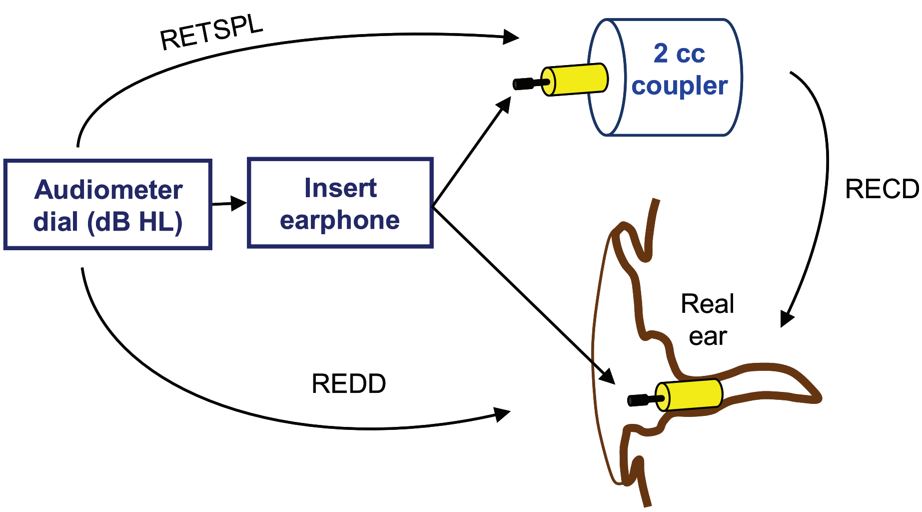

Figure 4.10 Relationship between RETSPL (coupler minus dial), RECD (real ear minus coupler), and REDD (real ear minus dial). After Munro and Lazenby, 2001.

enable either type of prescription. Measurement of real-ear aided gain is a necessary step in the calculation of insertion gain, so much of the information in this section is also relevant to insertion gain. The term ***real-ear gain*** will be used in a general sense: it will apply to both real-ear aided gain and insertion gain.

The ***real-ear aided gain (REAG)***, expressed in dB, is defined as the SPL near the eardrum, *A*, minus the SPL at some reference point outside the head. This reference point is variously defined as the level in the undisturbed field, *F*, or the level at a ***control microphone*** mounted on the surface of the head, *C*, as shown in Figure 4.11. The control microphone, also called a ***reference microphone***, is mounted either just above or below the ear. ANSI S3.46 specifies the level at the control microphone as the ***field reference point***. The real-ear measurement equipment uses the signal from the control microphone to regulate the sound level near the ear to the required level. This concept is the same as applied to control the level in test boxes (Section 4.1.2). The control microphone thus removes diffraction effects from the free field to the surface of the head. These diffraction effects are minimal when the sound source is located directly in front of the aid

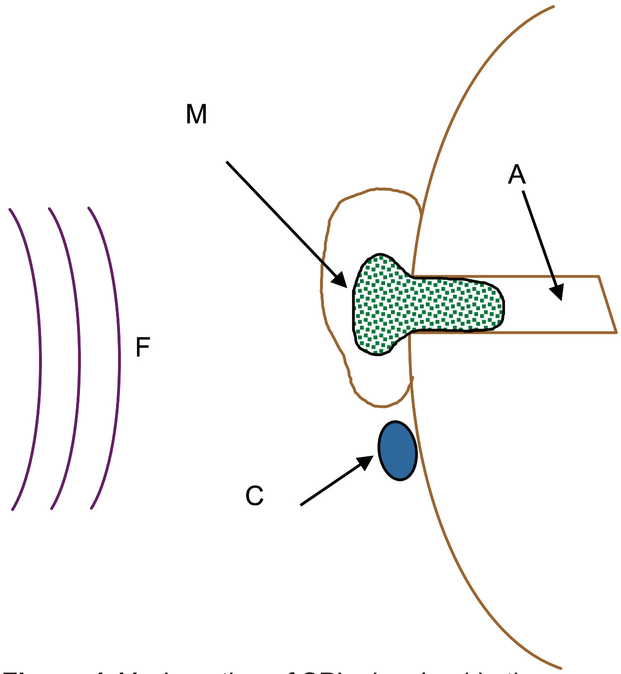

Figure 4.11 Location of SPLs involved in the measurement of real-ear aided gain. F is located in the undisturbed sound field (e.g. with the head absent), C is at the control microphone location on the surface of the head, M is at the hearing aid microphone port, and A is within the residual ear canal close to the eardrum.

Theoretical explanation: Standing waves in the ear canal

Because some of the power transmitted down the ear canal is reflected back by the eardrum, SPL can vary markedly along the length of the ear canal. A probe tube in the residual cavity actually senses the addition of the inwards-going and outwards-going waves. At the eardrum, these waves add almost in phase, so the pressure is a maximum at this point. As the reflected wave travels back from the eardrum, a phase shift develops between the incident and reflected waves (because the reflected wave has traveled further). Consequently, the two waves add less constructively. At a distance back from the eardrum approximately equal to one quarter of the sound's wavelength, the two waves are half a cycle out of phase, and partially cancel. Because this distance depends on wavelength, it therefore depends on frequency, as shown in Figure 4.12. A probe microphone placed at this position (called a *node*) would misleadingly indicate that there was very little sound travelling along the ear canal at this frequency.

The position of the node will not precisely be one quarter of a wavelength back from the eardrum because some phase shift occurs as the wave is reflected from the eardrum.[1865a]

Because the pattern of SPL versus distance looks like a wave that is always in the same place, the pattern is called a standing wave. For positions between the eardrum and the node of the standing wave, the waves partially cancel, or partially add, but the total sound pressure is always less than that at the eardrum.

Note that the discussion about how the incident and reflected waves combine near the eardrum did not require any assumptions about where the incident sound wave started. *Consequently, the variation of SPL with distance near the eardrum is exactly the same when a hearing aid is inserted in the ear as when the person is listening unaided.* Of course the actual SPL will be affected by the input level and by the hearing aid gain. In the case of unaided listening, SPL will also be affected by the frequency of the sound relative to the resonant frequency of the ear canal.

wearer. At 45°, the maximum effect is only about 4 dB, and this occurs in the 500 to 1000 Hz range.[1733] Measurement equipment thus actually displays the SPL at location *A* minus the SPL at location *C,* but for sounds from the front, this is approximately equal to *A-F.*

Synonyms for real-ear aided gain (REAG) are ***real-ear aided response (REAR)***, ***in-situ gain***, and ***real-ear transmission gain***. Some authors have used the word *response* instead of *gain* when they refer to a complete graph of gain versus frequency, rather than the gain at a specific frequency. More commonly, *response* implies that the measurement result is expressed as the absolute level of sound (i.e. in dB SPL), whereas *gain* expresses the result as the difference between two SPLs (i.e. in dB).

Probe microphones directly measure SPL, not gain, so the REAG curve is actually based on a measurement of REAR. The relationship between the two is simply:

REAR = REAG + input SPL … 4.8.

If the input is a pure tone, equation 4.8 is unambiguous. If the input is a broadband stimulus, then both the input SPL and the REAR SPL refer to the stimulus power falling within defined frequency regions (usually each 1/3 octave in bandwidth – see Section 4.5.7). Note that REAR is not just a property of the hearing aid – it is equally affected by the gain of the hearing aid and the level and spectral shape of the stimulus used to measure it.

4.3.1 Positioning the probe for REAG measurement

Measurement of REAG is straightforward. The precise details depend on the particular equipment used, but in all cases a flexible probe tube is inserted into the ear canal so that the SPL in the residual canal is sensed while the hearing aid is in place and operating. The probe is usually inserted first, and then the hearing aid or earmold. The only tricky part of the measurement is obtaining the correct depth of insertion. Provided the probe tube is past the tip of the aid or mold, its position does not matter for frequencies up to 2 kHz for closed earmolds and up to 1 kHz for open earmolds. Up to these frequencies, the wavelength is much bigger than ear canal dimensions, so the same SPL exists at all locations within the residual ear canal. As frequency increases above these values,

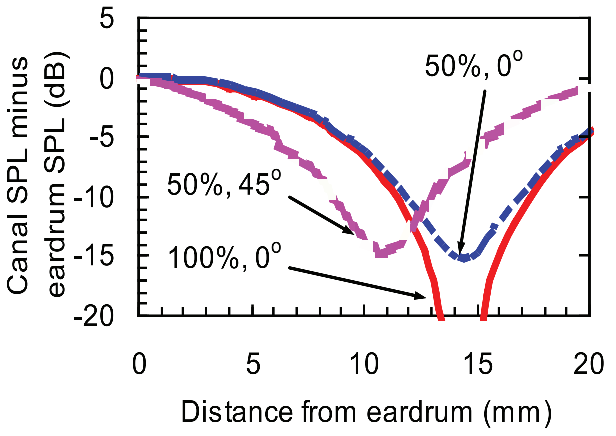

Figure 4.12 Calculated pattern of SPL in the ear canal versus distance from the eardrum at a frequency of 6 kHz. The red solid curve is for total reflection from the eardrum with no phase shift at the drum. As examples of the change in pattern caused by less than perfect reflection of sound, the blue dashed line is for 50% power reflected from the drum with no phase shift, and the pink speckled line is for 50% reflected with a 45 degree phase shift at the drum. There is significant inter-person variation in reflectance from the eardrum.[1865a]

however, placement becomes more critical because of standing waves in the residual ear canal.

Figure 4.12 shows the variation of SPL in the ear canal for a frequency of 6 kHz, for example. We are interested in the sound pressure at the eardrum, and if we wish to limit the error caused by the standing wave phenomenon to, say, 2 dB, we cannot afford to have the microphone more than 6 mm, which corresponds to 0.1 wavelengths from the eardrum. As frequency increases (and wavelength decreases) it becomes necessary to have the probe-tube tip closer and closer to the eardrum. (The maximum distance in wavelengths stays the same). Table 4.4 shows how close the probe tube must be to the eardrum if the error due to standing waves is to be kept to within 1, 2, 3, 4 or 5 dB. As a further example, if we wished to measure the REAG up to 8 kHz, and were willing to tolerate an error of only 1 dB, the probe tube would have to be within 3 mm of the eardrum. At the other extreme, if we were concerned about REAG only up to 3 kHz, and were willing to tolerate 5 dB error, the probe tube could be up to 18 mm away from the eardrum, which means that almost any location within the residual ear canal would be acceptable.

Table 4.4 Maximum distance from the eardrum (in mm) if the error induced by standing waves is not to exceed the values shown in the first column. These values, which can be used as a guide to practice, are based on calculations that assume complete reflection from the eardrum. (The distances increase insignificantly for partial reflections, but decrease by 25 to 50% for a phase shift of 45° at the eardrum.)

Standing wave error (dB)	Frequency (Hz)						
	2k	3 k	4k	5 k	6k	8k	10 k
1	13	9	6	5	4	3	3
2	18	12	9	7	6	4	4
3	22	14	11	9	7	5	4
4	24	16	12	10	8	6	5
5	27	18	13	11	9	7	5

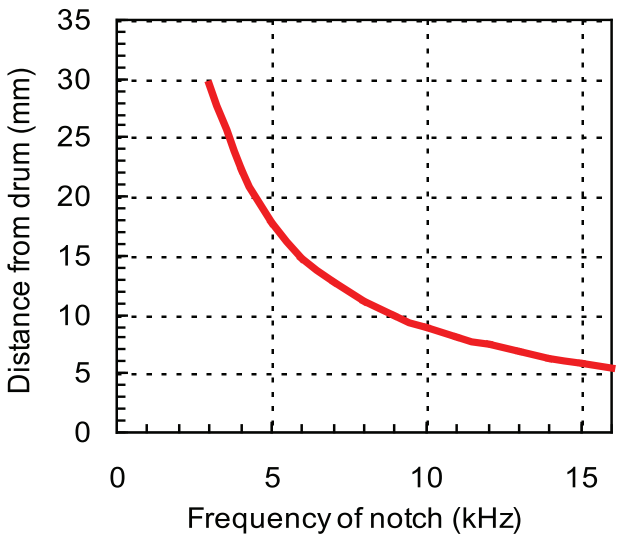

Figure 4.13 Distance from the eardrum at which SPL in the ear canal will be a minimum.

Practical tip: Positioning the probe tube for REAG measurements

- Position a marker on the probe tube approximately 30 mm from the open end.
- Generate a continuous tone at 6 kHz, and move the probe inwards, smoothly and continuously, starting at the entrance to the ear canal, while monitoring the SPL sensed by the probe microphone. Find the position at which the SPL is a minimum by moving the probe past the minimum a few times. When the probe is positioned at the SPL minimum, the probe tip should be 15 mm from the acoustic center of the eardrum (compare to Figures 4.12 and 4.13).[1731] Be aware that movements of your hand or the client's head can affect the amount of sound entering the ear canal, and hence give a misleading impression of which position corresponds to the node.[1741]
- Move the probe tube in by the amount necessary to position it the desired distance from the eardrum. For example, to position the tip 6 mm from the eardrum, insert it a further 9 mm. The extent of extra insertion is monitored by noting the movement of the marker.
- Some real-ear gain analyzers have a probe insertion section of the menu that facilitates probe placement using the notch in the frequency response created by the standing wave. The 6 kHz method described here can, however, be easily used with any real-ear gain analyzer. Also, the 6 kHz method avoids the problem of spurious notches in the frequency spectrum being created by loudspeaker or room acoustics, because all measurements are done at a single frequency.
- If the tip of a flexible probe tube gently touches the eardrum, physical damage and pain from mechanical force are unlikely, but loudness discomfort or pain from the acoustic sensation is possible.
- The skin near the eardrum in the final section of the canal can be very sensitive. The probe can cause pain if it is pushed into the canal wall in this area.
- In the ***average adult*** ear, 6 mm from the eardrum umbo corresponds to 18 mm past the ear canal entrance (1.5 mm more for males and 1.5 mm less for females), or 29 mm from the inter-tragal notch.[203, 1542] If you rely on these dimensions instead of the acoustic technique described above, view the location of the probe tip relative to the eardrum otoscopically during insertion and use a smaller insertion depth if appropriate.

Where the final distance from the eardrum can be estimated (see panel), and one is particularly interested in accurate high-frequency measurements, but is unwilling to further insert the probe tube, then Table 4.4 can be used in reverse. The left-hand column shows the correction that should be added for each frequency at each of the distances included in the body of the table. The correction allows us to estimate the SPL present at the eardrum based on the SPL measured some distance from it.

The theory in this section assumes that the incident sound wave is a plane wave[m] progressing smoothly down the canal. This may not be true in the first few millimeters immediately past the tip of the mold or aid, because the sound wave has to make a transition from the narrow sound bore to the wider canal.[204] There is some uncertainty about whether this creates a significant error in real ears, as experiments to investigate the effect of extending the probe past the tip of the mold have been confounded by the probe then being at different distances from the ear drum.[238, 1588] To be cautious, the probe tube preferably should not be positioned in this region, unless the hearing aid itself terminates within about 6 mm of the eardrum (i.e. at least a few mm past the second bend of the ear canal). For such deeply seated hearing aids, accurate measurements can certainly be made up to 6 kHz without the probe tube extending past the medial tip of the hearing aid.[1588] Accurate measurements to 8 kHz still requires extension of the probe tube closer to the eardrum.

4.3.2 Relationship between REAG, coupler gain and ear simulator gain

There are several reasons why the REAG will differ from the 2-cc coupler response of the hearing aid, both on average, and for individual aid wearers.

First, for a given test stimulus level, the actual input to the hearing aid will be greater for the REAG measurement than for the coupler measurement, at least for CIC, ITC and low profile ITE hearing aids. This occurs because for these aids, the diffraction effects from the free field to the microphone inlet port (*M-F* in Figure 4.11) are greater than the diffraction effects to the head surface (*C-F* in Figure 4.11) that are removed by the control microphone.

Table 4.5 shows the ***microphone location effects (MLE)*** from the undisturbed sound field to the microphone inlet port for each type of hearing aid, for two directions of sound waves. With the exception of body aids, microphone location effects are limited to the high frequencies, where the wavelength of sound is comparable in size to the obstacles creating the diffraction effects: the head and pinna. Microphone location effects are greater the more the concha remains unfilled by the hearing aid.

Table 4.5 Microphone location effects (MLE) due to body, head, pinna, concha, and canal diffraction and resonance: SPL at the hearing aid microphone port minus SPL in the undisturbed sound field, for two directions of the incoming signal. Body aid data are based on Kuhn & Guernsey (1983), CIC data from Cornelisse & Seewald (1997), and the remainder are based on Storey & Dillon (unpublished).

Aid type	Source	Frequency								
		125	250	500	1 k	2 k	3 k	4 k	6 k	8 k
Body	0°	2	3	5	3	2	1	0	0	0
BTE	0°	-1	0	0	0	3	2	1	1	2
ITE	0°	-1	0	1	1	3	5	7	3	2
ITC	0°	0	1	1	1	5	8	10	2	-2
CIC	0°	0	0	0	1	3	6	8	2	-5
BTE	45°	0	1	1	2	5	5	4	4	3
ITE	45°	0	2	3	3	5	7	9	7	5
ITC	45°	0	2	3	3	6	10	13	8	1
CIC	45°	2	3	3	4	6	10	13	10	0

m A plane wave has uniform pressure across the wave front, which, over the area of interest, forms a flat plane.

The second reason why REAG will differ from coupler gain is that the hearing aid terminates in a smaller volume in the real ear than when it connects to a 2-cc coupler, as discussed in Section 4.2. Thus, the real-ear to coupler difference (RECD) directly affects the relationship.

The third reason is that the hearing aid may use different coupling in the individual ear than when it is measured in the coupler. In particular, the sound bore may be different (for BTE and body aids), and the vent will not be included in the coupler measurement (and should not be, as discussed in the next chapter).

Finally, if the hearing aid leaves the ear canal largely open, the canal resonance affects the acoustic impedance of the ear canal within the bandwidth of hearing aids, and hence the sound pressure level generated in the ear. (This effect is allowed for in this book as a vent effect.)

These differences are summarized in Equation 4.9, which assumes that the coupler gain and the REAG are obtained with the same volume control setting. The equation also assumes that the hearing aid is linear.[n] If the coupler gain is measured in an HA1 coupler, then the HA1 RECD (Table 4.4) must be used, and if the coupler gain is measured in an HA2 coupler, then the HA2 RECD (Table 4.3) must be used.

$$\text{REAG} = \text{coupler gain} + \text{RECD} + \text{MLE} + \text{sound bore effects} + \text{vent effects} \quad \ldots 4.9.$$

Sound bore effects and vent effects will be explained and quantified in Chapter 5. If an average value of RECD is assumed (Table 4.3), then Equation 4.9 can be used to predict REAG on the basis of coupler gain. Alternatively, if the RECD is measured for an individual aid user, then a more accurate prediction of REAG can be made. RECD is most worth measuring when it is most different from average. This occurs for infants[1914] and for ears with middle-ear pathology (Sections 4.2 and 16.4.3).[544]

We can write a similar equation to relate REAG to ear simulator gain:

$$\text{REAG} = \text{ear simulator gain} + \text{MLE} + \text{sound bore effects} + \text{vent effects} \quad \ldots 4.10.$$

Note that the RECD term is missing, although REAG can still be predicted most accurately from ear simulator gain if the individual variation of RECD from average is taken into account (Section 11.4).

4.3.3 Detecting incorrect aided measurements

No physical measurement should be believed just because some buttons are pressed and a number or graph is obtained! Measurement of REAG is no exception, because several factors (discussed in Section 4.5) can affect the validity of the measurement. How can one know whether a measurement is correct? Fortunately, we can have some strong expectations of what the REAG should be, and for vented aids (including open-canal fittings) there are some additional quick checks that can be done.

Practical tip: Checking the aided measurement

If the hearing aid has a vent larger in diameter than the probe tube, then a REAG measurement can be checked as follows:

- Withdraw the probe tube from the ear, leaving the hearing aid in place.
- Inspect the tip of the probe for blockage by wax.
- Re-insert the probe tube, but via the vent hole this time, to the same depth as previously.
- Repeat the measurement.
- Withdraw the aid or mold and the probe tube together and make sure that the probe is extending beyond the aid or mold.

[n] For nonlinear hearing aids measured with pure tones, MLE must be divided by the compression ratio (see Section 6.2.2) applicable to that frequency and level, because sound diffraction affects the level of sound at the input to the hearing aid. For broadband sounds and few channels of compression, the situation is more complex. These considerations apply wherever MLE appears in this chapter.

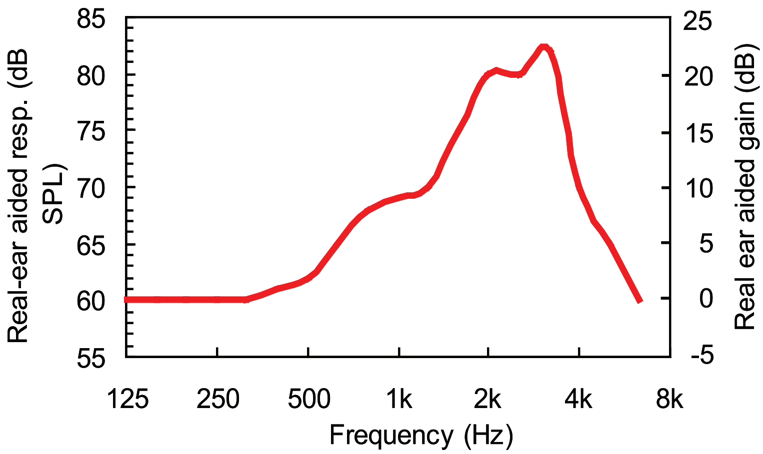

Figure 4.14 Typical REAG display for a vented, low to medium gain hearing aid, measured with a 60 dB SPL swept pure tone, displaying the expected low frequency plateau.

First, if the hearing aid is vented or not especially tight, low-frequency sounds will enter the residual ear canal directly via these air paths. As discussed in detail in Section 5.3.1, the SPL of these sounds in the canal will equal the SPL of the test stimulus outside the head. Consequently, if the gain is not too high, the aided response should show a low-frequency plateau (a horizontal line), as shown in Figure 4.14. If the measurement is expressed in ear canal dB SPL and a swept pure tone is used, the amplitude of the line should equal the test stimulus level. If the display is expressed as dB of gain, then the plateau should be at 0 dB gain, no matter what type of stimulus is used. (Figure 4.14 shows both of these vertical axes, although measurement equipment usually shows only one of these axes at a time.)

Second, the measurement can be repeated through the vent as described in the accompanying panel. If the probe fills most of the vent (i.e. their diameters are similar), the low-frequency response may change in a manner consistent with a reduction of vent size (see Section 5.3.1). If the mold or aid has a lot of leakage around it, or if the vent diameter is more than 50% larger than the probe diameter, then the two responses should be extremely similar. If not, something is wrong with one of the two measurements.

Finally, but less practically, Equation 4.9 or 4.10 can be applied at one or two frequencies and the result compared to the measurements. The discrepancy should never be more than 15 dB, should rarely be more than 10 dB, and will often be less than 5 dB.[431]

The most common causes of inaccurate measurements are wax blockage, probe tips pushed into the ear canal wall, probes excessively squashed by a tight earmold,[1782] and analyzer buttons pushed in the wrong sequence. If the hearing aid is accidentally left turned off, the result is instantly recognizable: one sees the REAG of the vent and leakage paths alone - typically 0 dB gain at low frequencies and attenuation at higher frequencies (Section 5.3.1). This measurement, even if it was not intended, is called the ***real-ear occluded gain (REOG)***. It shows how well the hearing aid functions as an earplug when it is turned off.

4.4 Insertion Gain

The second type of real-ear gain is called ***real-ear insertion gain (REIG)***. This gain tells us how much extra sound is presented to the eardrum as a result of inserting the hearing aid in the ear. Figure 4.15 shows the ear in its unaided and aided states. Insertion gain is defined as the SPL at the eardrum when aided, *A*,

Theoretical summary: real-ear gains

With reference to Figure 4.15, the following four equations summarize the relationships between the two types of real-ear gains.

REUG = U - C … 4.11

REAG = A - C ... 4.12

Insertion gain = REAG – REUG … 4.13

Consequently, for the same test level at position C in the aided and unaided measurements:

Insertion gain = A-U ... 4.14

Note that equations 4.13 and 4.14 are unchanged if the reference points for REUG and REAG are chosen to be F rather than C.

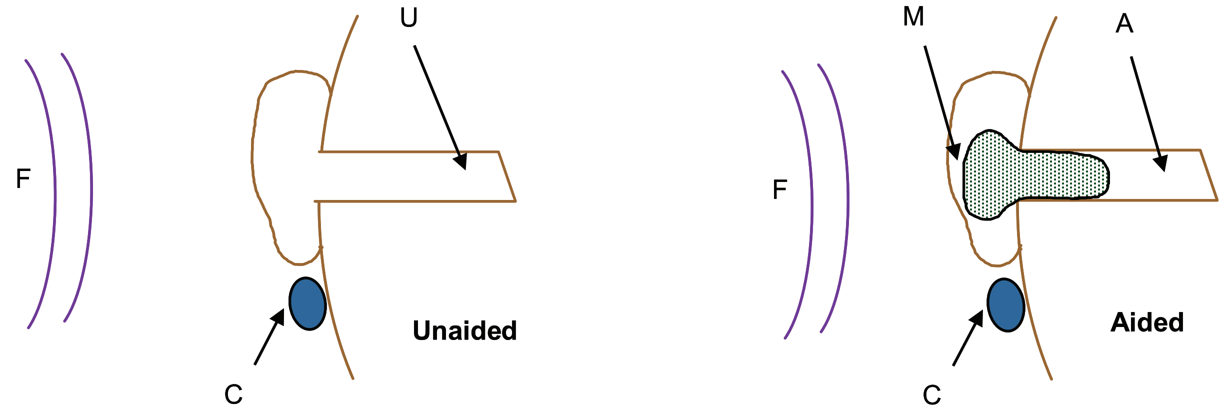

Figure 4.15 Location of SPLs involved in the measurement of insertion gain. F is located in the undisturbed sound field (with the head absent), C is at the control microphone location on the surface of the head, M is at the hearing aid microphone port, A is at the eardrum when aided, and U is at the eardrum when unaided.

minus the SPL at the eardrum when unaided, *U*. The key distinction between insertion gain and REAG is that insertion gain takes into account the amount of "amplification" the person was getting from the resonances in his or her concha and ear canal, prior to inserting the hearing aid. This natural amplification, called the ***real-ear unaided gain (REUG)***, is lost (partially or totally depending on how open the earmold is) when a hearing aid is inserted. Before a hearing aid can provide additional signal, it must first provide at least this much gain. Insertion gain can be thought of as the net result: the REAG provided by the hearing aid minus the REUG provided by the unobstructed ear.

The top half of Figure 4.16 shows a typical REUG for an adult person for sounds incident from 0° and with no head-mounted control microphone in place. It also shows the REAG for a hypothetical hearing aid. How much "gain" does the hearing aid provide at, say 3 kHz? The question is, of course, ambiguous. The SPL at the eardrum is 30 dB more than the SPL outside the head, so the REAG is 30 dB. However, the person's unaided concha and ear canal provide 16 dB of gain (the REUG) at this frequency, so the net effect of the hearing aid, that is, the insertion gain, is only 14 dB.

Consistent with its definition, insertion gain measurement is a two-step process. In the first step, the unaided response is obtained. This is the baseline for the second step: measurement of the aided response. The insertion gain finally displayed is then the difference between these two measures, although real-ear gain analyzers often show either or both of the two intermediate results as well.

4.4.1 Positioning the probe for insertion gain measurement

The position of the probe for insertion gain measurement is much less critical than it is for REAG measurement. Although we are interested in the increase in SPL at the eardrum caused by inserting the hearing aid, the same increase will occur at other points within the ear canal medial to the tip of the mold or aid. The increase in SPL from a mid-canal position to the eardrum does not depend on the source of the sound. *Consequently, the increase in SPL at the ear-*

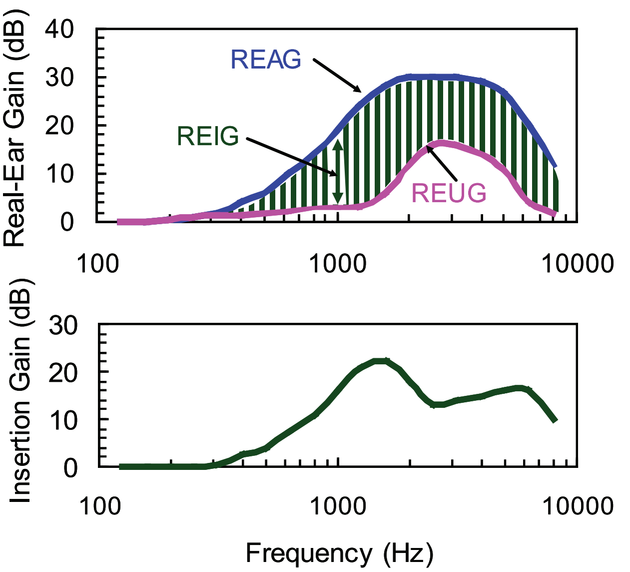

Figure 4.16 Real ear unaided and aided gains (top half). The difference between these curves is the insertion gain, shown as the shaded region in the top half and as the curve in the lower half.

Practical tip: Positioning the probe for insertion gain measurement

- Inspect the ear canal for excessive wax or abnormalities.
- Insert the aid or mold in the ear and note where its lateral surface lies with respect to some landmark on the ear: the ear canal entrance for a CIC, or the inter-tragal notch or tragus for larger hearing aids – see arrow in Figure 4.17(a).
- Remove the aid or mold and lay the probe alongside the inferior surface (the part that touches the floor of the ear canal and concha). The tip of the probe should extend approximately 5 mm past the tip of the aid or mold to avoid the transition sound field.
- Mark the probe, or position a sliding marker on the probe, at the position where the selected ear landmark would be when the mold or aid is inserted – see Figure 4.17(b).
- Insert the probe tube until the marker lines up with the selected landmark on the ear, and measure the REUG – see arrow in Figure 4.17(c).
- Insert the hearing aid, leaving the probe tip in the same position, and measure the REAG – see Figure 4.17(d). To leave the probe tip in the same position, the marker should be inserted 1 to 3 mm more for the aided measurement than for the unaided measurement, because of the more winding path followed by the probe in the aided condition as it conforms to the ear canal wall.[1490]

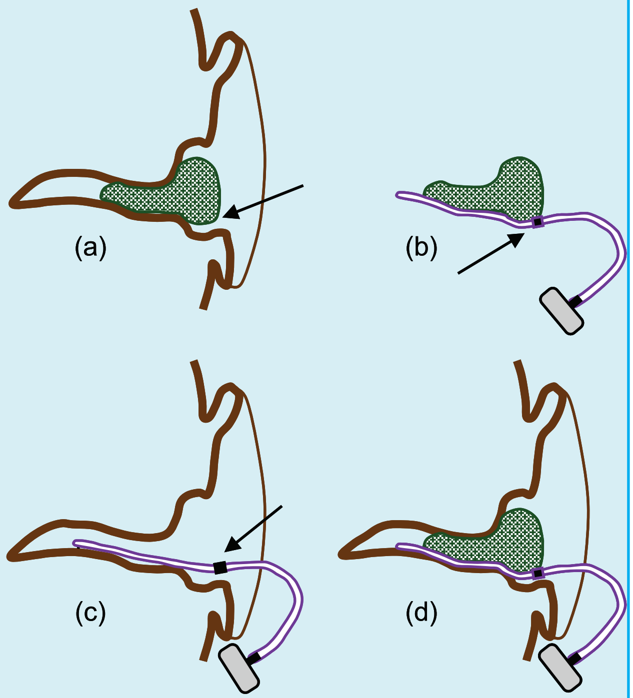

Figure 4.17 Probe positioning for measuring insertion gain: (a) noting a landmark on the ear; (b) marking the probe; (c) measuring the unaided response; (d) measuring the aided response.

drum can be measured at a mid-canal point provided the probe tube is in the same place for the unaided and aided measurements.

4.4.2 Relationship between insertion gain, coupler gain and ear simulator gain

Insertion gain will differ from coupler gain for several reasons. As with REAG, insertion gain should exceed coupler gain because it has the benefit of head, pinna, and concha diffraction (i.e. the microphone location effects), and because the volume of the residual ear canal is smaller than a 2-cc coupler (i.e. the RECD effect). However, insertion gain should be less than coupler gain because the measurement of insertion gain involves subtracting REUG. These opposing adjustments to coupler gain approximately, but coincidentally, cancel each other for ITE and ITC aids up to 3 kHz. Consequently, for ITE and ITC aids, up to 3 kHz, insertion gain approximately equals coupler gain (within a few dB) for the average adult.[o] For other types of aids, there is a net difference between insertion gain and coupler gain, even on average. These factors are summarized in the following equations:

Insertion gain = coupler gain + RECD + MLE
- REUG + sound bore effects
+ vent effects .....4.15.

[o] For frequencies higher than 3 kHz, insertion gain exceeds coupler gain for CIC, ITC and ITE hearing aids, because the sum of RECD and MLE exceeds REUG.

Table 4.6 Average real-ear unaided gain (REUG) for adults.[1733] Measurements are given for two sound field directions, and with and without the use of a head-mounted control microphone. Use of this microphone removes head diffraction effects but leaves ear diffraction and canal resonance effects in place. 0° corresponds to frontal incidence and 45° is towards the ear being tested.

Source angle	Control microphone present	Frequency (Hz)								
		125	250	500	1 k	2 k	3 k	4 k	6 k	8 k
0°	No	0	1	2	3	12	16	14	4	2
45°	No	0	1	3	5	13	20	18	9	3
0°	Yes	0	0	0	1	12	14	12	3	1
45°	Yes	0	0	0	1	12	17	15	7	2

Vent effects and sound bore effects are explained in Sections 5.3.1 and 5.4.1 respectively. The difference between coupler gain and insertion gain, measured with no venting and with the same sound bore in each measurement is often referred to as ***CORFIG***, which stands for **co**upler **r**esponse for a **f**lat **i**nsertion **g**ain.[921] That is:

CORFIG = coupler gain - insertion gain ... 4.16.

Comparison of equations 4.15 and 4.16, shows that:

CORFIG = REUG - RECD - MLE ... 4.17.

This equation clearly shows the three factors that cause insertion gain to be different from coupler gain: the gain of the unaided ear, the difference in effective volume between the ear canal and the coupler, and head diffraction effects to the microphone. Average values for RECD and microphone location effects have already been given in Tables 4.3 and 4.5 respectively. Table 4.6 gives average values for REUG for two different directions of the incident sound.

Average values for CORFIG for each type of hearing aid, consistent with these separate values of RECD, diffraction, and REUG are given in Table 4.7. We can summarize the two uses of CORFIG by writing:

Insertion gain = coupler gain - CORFIG + sound bore effects + vent effects … 4.18,

Coupler gain = insertion gain + CORFIG - sound bore effects - vent effects ... 4.19.

CORFIG values relate insertion gain to coupler gain at the same position of the volume control. CORFIG values are most often used to find the coupler gain that is equivalent to a certain insertion gain. In turn, these coupler gains are used to select an appropriate hearing aid and/or to adjust the hearing aid in a test

Table 4.7 CORFIG factors for each type of hearing aid when measured in a 2-cc coupler of the type shown, derived from Tables 4.3, 4.5 and 4.6. All values assume a typically leaky fitting and an earmold of typical length (stopping just short of the second bend). If the mold is deeper than standard, as often occurs for a CIC, CORFIG values will be smaller by an amount that depends on the depth of the mold or aid.

Hearing aid type		250	500	1 k	2 k	3 k	4 k	6 k
Body aid	HA2	0	-7	-7	2	12	11	-1
BTE	HA2	3	-2	-4	1	11	10	-2
BTE	HA1	3	-2	-3	-1	5	3	-7
ITE	HA1	3	-3	-4	-1	2	-3	-9
ITC	HA1	3	-3	-4	-3	-1	-6	-8
CIC	HA1	3	-2	-4	-1	1	-4	-8

box. If the hearing aid is specified at its maximum volume control setting (full-on gain), but used at a mid volume control setting, it is appropriate to add reserve gain to the CORFIG values. This reserve gain is the amount by which the user can turn up the volume control from the position at which the target real-ear gain is achieved. To summarize this:

$$\text{Coupler gain}_{\text{max v/c}} = \text{Insertion gain}_{\text{used v/c}} + \text{CORFIG} - \text{sound bore effects} - \text{vent effects} + \text{reserve gain} \qquad \text{... 4.20.}$$

This section has not dealt specifically with ear simulator gain. Ear simulator gain can always be calculated from 2-cc coupler gain simply by adding ear-simulator-to-coupler difference (Table 4.3) to the coupler gain.

4.4.3 Detecting incorrect insertion gain measurements

Despite the in-built resistance to errors of insertion gain measurement, one still must be able to detect incorrect measurements. Because the measurement is done in two stages (unaided and aided) this involves being able to spot errors in either stage. We can legitimately have very strong expectations about what the first unaided measurement should look like. For an adult, it should look like the unaided curve in Figure 4.16. Of course, it would ordinarily not look exactly like this, or else there would be little point in measuring it for each person. All valid REUR curves for people with ear canals with normal anatomy must, however, have certain features in common:

- There ***must*** be a low-frequency plateau at the level of the test stimulus (if expressed in dB SPL and measured with a swept pure tone) or at 0 dB (if expressed as a gain with any stimulus).
- There should be a peak somewhere between 2.2 kHz and 3.2 kHz with an elevation above the low-frequency plateau of between 12 and 22 dB.

The ranges in point (b) comprise plus or minus three standard deviations around the mean values of a sample of 20 adults.[1732] There will be an occasional person (probably with observably very long or very short ear canals) where REUG goes outside this range. There will also be people who have had surgery to their ear canals (e.g. a mastoidectomy) that alters the shape of the ear canal. Fitting procedures for such people are covered in Sections 10.2.4, 11.4, and 16.4.3.

The second half of the insertion gain measurement is obtaining the aided response. Methods for checking the validity of this measurement have already been covered in Section 4.3.3.

4.4.4 Accuracy of insertion gain measurements

If an insertion gain measurement is not simply wrong, then how accurate can one expect it to be? This can be deduced from repeated measurements made with a variety of measurement methods.[455, 922] Over most of the frequency range, the difference between a single measurement and the average of many measurements (i.e. the true value) has a standard deviation of 3 dB. This means that 95% of measurements would be within 6 dB (two standard deviations) of the true value. In the high frequencies, because of the effects of standing waves and the impossibility of ensuring that the probe is in *exactly* the same place for the aided and unaided measurements, the standard deviation rises to 5 dB. There is also a slight tendency for some probe-tube systems to underestimate high-frequency gain. This underestimation may occur as a result of some constriction of the probe tube by the mold or aid or because of inaccurate placement of the probe in either the aided or unaided condition.[455, 1782]

4.5 Practical Issues in Real-Ear Testing

The following practical issues affect the measurement of both REAG and insertion gain, although not necessarily to the same degree.

4.5.1 Probe calibration

Probe microphones have an inherently non-flat frequency response, because of the effect of the long thin probe tube. Real-ear gain equipment corrects for the response by including a calibration step in the measurement or by applying a correction stored in memory. Often, the probe microphone is calibrated against the control microphone, which does have a flat response. In this calibration step, the clinician should hold the tip of the probe tube closely against the control microphone inlet port, but without blocking the inlet port of either microphone. If the measurement system does not include a special clip to hold the two microphones together, it can be done with putty or with the fingers. The fingers and hand should be kept out of the way: that is, they should not be in a direct line between the speaker and the two microphones. Figure 4.18 shows the probe being held against the control microphone for one commercial analyzer.

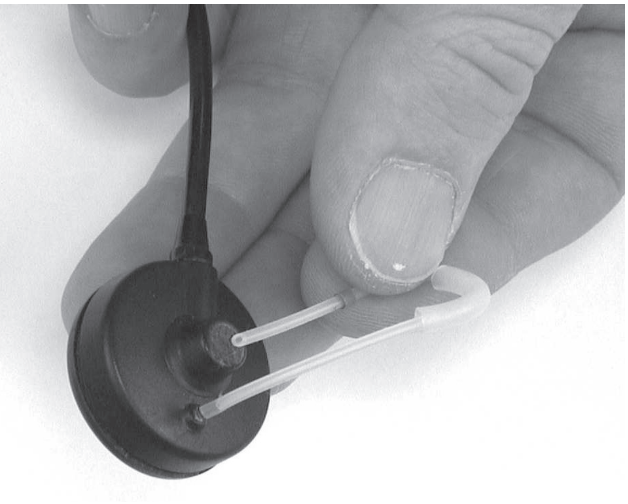

Figure 4.18 Positioning of the probe microphone against the control microphone during calibration.

4.5.2 Control microphones

Just as for measurements in a test box, a ***control microphone***, also called a ***reference microphone,*** is used to regulate the input SPL to the desired value. Most commonly, the pressure method of calibration is used, in which the control microphone operates while the actual measurement is taking place.[p] If the hearing aid wearer moves between the aided and unaided parts of an REIG measurement, the control microphone compensates for the movement thus avoiding the measurement error that would otherwise occur. For a perfectly stationary patient, the control microphone does not affect the REIG values obtained.

One of the advantages of insertion gain is that it is a difference measurement. This means that, at any frequency, provided everything is the same for the aided as for the unaided measurement (e.g. head position, probe tube position, stimulus level, and probe calibration), none of those factors will affect the final result, even if the probe microphone is totally uncalibrated. This makes insertion gain measurement resistant to errors. Despite this advantage, it is best to use the control microphone, in the pressure calibration mode, because of the protection it gives against errors created by the movements of the person being tested.

With open fittings, however, it is advisable to turn the control microphone off while the hearing aid is present, irrespective of whether REAG or REIG is being measured. Open fittings allow significant amounts of sound to leak back from the ear canal to both the control microphone and the hearing aid microphone.[1022, 1264] Unfortunately, the propagation to the two points is different, so the way the leaked sound combines with the incoming sound field is also different. Furthermore, inside the hearing aid, feedback cancelling algorithms effectively remove the leakage signal prior to amplification, but no such cancelling happens at the control microphone.

The signal sensed by the control microphone will therefore (usually) be greater than that amplified by the hearing aid, causing an artificially low gain measurement. This error increases with the gain of the hearing aid, increases with the openness of the fitting and decreases as the control microphone is moved further away from the ear. It is not only open-canal hearing aids for which this error occurs. An error is likely for any hearing aid where the gain achieved is possible only because of the feedback cancelling algorithm.[1022]

The solution is to turn the control microphone off for the aided measurement. Prior to this, perform a calibration and/or unaided measurement with the control microphone on. During the measurement the analyzer will set and then “remember” how intense the electrical signal delivered to the loudspeaker has to be to obtain the desired sound level in each frequency region. Then, with the patient and clinician remaining in the same position, perform the aided measurement with the control microphone turned off, during which the analyzer will ensure that the loudspeaker emits the same sound level as previously.[q]

An alternative solution is to keep the control microphone active, but to place it on the opposite side of the head, as is done for measurement of CROS hearing aids (panel in Section 17.1.1). Not all test equipment enables the control microphone and probe microphone to be physically separated sufficiently to use this solution.

p ANSI S3.46 refers to this as the “modified pressure method with concurrent equalization” (MPMCE).

q ANSI S3.46 refers to this as the “modified pressure method with stored equalization” (MPMSE).

Practical tip: Checking for corruption by background noise

Either of the following tests is probably enough to ensure that noise is not unduly affecting measurement accuracy. The first test alone may, however, misleadingly inspire confidence in the results, when the hearing aid is measured using equipment that employs signal averaging.

- Repeat the measurement and ensure that all the fine bumps and dips also appear in the second response.
- Using an aid operating in its linear region, decrease the input signal level by 5 dB. If the two curves are 5 dB apart (or equal if expressed in dB gain) and equally smooth in appearance, then background noise is not a problem at either test level. For nonlinear hearing aids the output should decrease by 5 dB divided by the compression ratio (CR). This is equivalent to the gain increasing by 5(CR-1)/CR dB.

4.5.3 Effects of wax

The most dramatic effect that wax can have on real-ear gain measurement is when it fills the tip of the probe tube. The equipment then incorrectly indicates that the signal level in the ear canal is very low. Apart from ingress of wax into the probe tube, real-ear gain measurement should not greatly be affected by cerumen in the canal. Cerumen should not have much effect on low-frequency real-ear gain until there is enough of it to fill a significant proportion of the residual ear canal volume. Cerumen should not have much effect on high-frequency real-ear gain until there is enough of it to fill a significant proportion (e.g. one-third) of the cross sectional area of the canal at any point. There are, however, no empirical data on this issue.

4.5.4 Contamination by background noise

Real-ear gain measurement equipment employs a filter in the measurement chain to help discriminate against background noise. For swept pure tone measurement, the filter tracks the stimulus frequency. For broadband stimuli, the analysis process (usually a Fourier Transform) is in essence a large set of very narrow band-pass filters. Some equipment also employs signal averaging to improve measurement accuracy. Because of these techniques, real-ear gain measurement systems are resistant to ambient noise, but are by no means immune to it. Swept pure tones or warble tones are much more resistant to background noise than broadband test signals. This is because all the signal energy is concentrated in one narrow region rather than being spread over the entire frequency range and it is easier to filter out the noise in the other frequency regions.

Some equipment monitors the consistency of repeated measurements at each frequency and rejects measurements that it considers are corrupted by background noise. Such equipment may give a warning message when background noise levels are excessive, or it may just lengthen the measurement time indefinitely. Other equipment gives no specific indication of noise corruption (see accompanying panel). Place the real-ear gain equipment in the quietest place available (the test booth is ideal if it is large enough for the loudspeaker to be at least 0.5 m from the patient). It is not, however, essential to have noise levels as low as those needed for audiological assessments, so locations other than the sound booth are likely to be suitable.

Wherever the equipment is located, you should identify the lowest signal level at which measurement is possible and then avoid testing at lower levels. The quieter the place, the lower the levels at which you will be able to test, which is especially useful for nonlinear aids. Being able to test at 65 dB SPL is the bare minimum that is acceptable, and being able to test down to 50 dB SPL is very desirable. If you cannot test at 65 dB SPL with broadband test signals, you either have to get a quieter location, measure RECD at a high level and adjust the hearing aid in the test box, test only with warble tones (see panel), or test only at higher levels and infer the response at lower levels from coupler measurements. The last two options greatly complicate testing nonlinear hearing aids. Apart from moving to a quieter location, the easiest option is to measure RECD and adjust the hearing aid in the test box, just as is done for pediatric fittings.

What to do when you cannot measure real-ear gain with a broadband signal

As most hearing aids are intentionally non-linear, it is best to use a broadband test signal. If one is not available, or if background noise limits its use to higher input levels, the following alternatives avoid the measurement of real-ear gain at low input levels with a broadband signal:

- Verify the gain-frequency response at high input levels only, and rely on the fitting software or a test box to confirm that all compression ratios and compression thresholds are correct.
- Verify the real-ear I-O curves, rather than gain-frequency response, using narrow band signals like warble tones.
- Measure the patient's RECD curve, calculate a corrected coupler-gain target, and do all measurements and adjustments in a test box (Section 11.4).

If on any occasion you do not have any test equipment, all you can do is accept the adjustment provided by the manufacturer's software and then evaluate the response subjectively using the methods described in Chapter 12. The ideal, however, is to have a real-ear gain analyzer with broadband measurement capability, located in a quiet place such as a test booth.

4.5.5 Hearing aid saturation

Nothing is simple. Contamination by background noise could always be avoided by using a sufficiently high level for testing. Unfortunately, if too high a test level is used, a hearing aid will saturate (i.e. limit in some way), and the result obtained will not be indicative of the performance of the hearing aid at lower input levels. In addition to high-level saturation, most hearing aids are intentionally non-linear over a wide range of input levels. To find out how a nonlinear hearing aid performs at several input levels it is necessary to actually measure the aid at these levels.[r]

Input-output functions are particularly useful for sorting out in what ways a hearing aid is nonlinear, and can usually be measured in the ear just as easily as in the coupler (Section 4.1.5). Preferably though, the clinician will have become thoroughly familiar with the characteristics of the hearing aid (from test box measurement, the specification sheet, or the fitting software) long before its performance is measured on a hearing aid wearer.

4.5.6 Loudspeaker orientation

People listen to sound coming from all directions, but are probably interested in sound coming from approximately frontal directions more often than from any other direction. Why then would we want to measure the real-ear response of a hearing aid from any azimuth other than from directly in front? (Angle, in the horizontal plane measured from the front, is called ***azimuth***.) The answer is that measurement from another azimuth may be more reliable. There is conflicting evidence as to whether an azimuth of 0° or 45° provides the best resistance to errors caused by movement of the patient,[922, 1726] but 90° should not be used.[794] Whichever direction in the range 0 to 45° is chosen (and the choice is not critical) small errors arise because the control microphone controls the test level only at the position of the control microphone itself, not at the position of the hearing aid microphone.

For CIC hearing aids, the unaided ear and the hearing aid microphone have about the same directionality, because in both cases sound is picked up after it has entered the ear canal. The insertion gain will therefore be about the same no matter what azimuth is chosen, assuming no probe tube or head movements occur during testing. For other hearing aids, where the microphone is not located within the ear canal, insertion gain will depend to some extent on azimuth, but below 5 kHz, and within the range from straight ahead around to 45° on the hearing aid side, azimuth has little effect on insertion gain.

[r] For hearing aids that are linear over a wide range of input levels, the gain that is measured at any input level within the linear range will also apply at any other input level within the range.

Practical tip: Positioning the aid wearer and the loudspeaker

- Whatever angle of the source relative to the head is chosen, position an interesting object in the direction in which the patient should be facing and have the patient look directly at it during the measurement. If 45° is chosen (see text), two such objects will be needed, one on either side of the source, for testing left and right ears.
- Choose a test position about 0.5 to 0.75 m away from the source. This is a compromise. If the spacing is too close, a small head movement can result in a large change of angle between the source and the patient. If the spacing is too large, room reflections are more likely to cause significant standing waves in the vicinity of the head.
- Avoid large, flat reflecting surfaces near the client, including walls which should be at least 0.4 m (16 inches) away, and preferably further away (ANSI S3.35; 2010).
- The tester should stand well back from the client, and should remain in the same position for aided and unaided testing, so as not to alter the reflections from the room.

For measurement of REAG, rather than insertion gain, the gain-frequency response obtained will depend on azimuth. For omni-directional hearing aids, head baffle effects will cause high-frequency gain to increase as azimuth increases from 0° to 60° or more (depending on frequency and aid type). For directional hearing aids, the combined effects of head baffle and microphone directionality will also make the gain and response shape depend a little on azimuth, with maximum gain at most frequencies occurring for azimuths between 20° and 50°. It may be most reasonable to test directional aids with an azimuth of about 30°, but any choice from 0° to 45° could not be criticized because the directionality of hearing aids is currently not large enough to have substantial effects on the measured response.

4.5.7 Measurement signal characteristics

There are two aspects to consider in selecting a signal for measuring real-ear gain. The first of these is choosing a signal that will make nonlinear hearing aids operate in a realistic manner. This issue is no different from measuring hearing aids in a test box and has been covered in Section 4.1.3. Some analyzers use speech as the test signal, and this has the advantage that non-linear processing schemes in hearing aids will operate in a manner more realistic than they may operate for other test sounds.

One possible speech stimulus is live speech, produced by the clinician, a person accompanying the client, or even by the client. This is a compelling counseling tool when combined with a display of hearing thresholds and unamplified speech levels.[378] Its use as a fitting verification tool is very problematic, even if the measured hearing aid output is compared to an REAR target, because the long term-level and the spectral shape of the input signal are both poorly controlled. Different clinicians, with different voice spectra or levels, would then consistently prescribe different gain-frequency responses, which is absurd. When the long-term rms spectrum and dynamic range of amplified speech (live or recorded) in the ear canal is shown superimposed on the patient's thresholds and optionally discomfort levels, all expressed as dB SPL in the ear canal, the result is called a ***speech-o-gram***, or a ***speech map*** (see Figure 9.3 for an example). The measurement process is called ***live speech mapping***.

Live speech mapping

Although live (i.e. not recorded) speech is a poor choice for verification (see main text), it can be used to assist in making special programs for frequent communication partners, as well as for counseling generally. For example, live speech mapping, while a very softly spoken spouse is talking, could be used to guide the adjustment of a special program that the aid wearer selects for extended communication with that person.

Whenever a speech map is used for verification, it is essential that the display show the prescribed speech target levels, not just the amplified speech and hearing thresholds, or the temptation to apply excessive gain to maximize audibility will be irresistible to the clinician. The result will be perfect audibility (though probably not perfect intelligibility) and a hearing aid so loud the client will not wear it. Even if the upper edge of the speech map is below the patient's discomfort levels, the aid can still cause loudness discomfort, because discomfort levels are measured one frequency at a time, whereas the upper limits of the speech signal will sometimes be present in several frequency bands simultaneously.

Output levels greater than prescribed will likely produce so much loudness the patient will ask for the hearing aid to be turned down, which will likely decrease intelligibility below the original value. For moderate hearing loss or greater, gains that produce the desired loudness actually result in the lower levels of the speech map being below threshold, counter-intuitive as this may be. A direct comparison of the amplified speech to the prescription target is thus essential if the appropriateness of the audibility of the amplified speech is to be judged.

The SPL levels shown in any display of level versus frequency depend greatly on the signal bandwidth used by the analyzer. A common choice is to display bands 1/3 of an octave wide. This is a good choice, as broadband sounds will, very approximately, start to become audible in a frequency region when their 1/3 octave levels rise above hearing threshold expressed as the SPL of the weakest pure tone that can be heard in that frequency region. This approximation becomes less and less true as hearing thresholds increase, because the wider auditory filters that accompany sensorineural hearing loss cause signal power to be summed across wider bandwidths. Audibility, as expressed in a speech-o-gram, should therefore be taken as a guide only. The greater the loss, the more the speech-o-gram underestimates audibility, but the poorer the ear is at analyzing the separate frequency components of the audible sound, and the less information the person can extract (see Section 1.1.3).

The second aspect is choosing a signal type that assists in control of the signal level. Although the measurement equipment uses a control microphone, this controls the level precisely only at the location of the control microphone. How well the level is controlled at other positions around the pinna depends on the signal bandwidth and the test environment. If reflections (e.g. from room boundaries, nearby objects, the subject's shoulders) cause standing waves to develop in the vicinity of the head, then large variations in SPL will occur within small regions, especially for the higher frequencies. Standing waves have their most pronounced effects for pure tones, because very pronounced minima (nodes) can occur when the reflected wave cancels the direct wave.

As stimulus bandwidth is broadened, the acoustic field becomes smoother, because it is impossible for a range of frequencies to all have a node at exactly the same point in space.[462] Consequently, the control microphone does a better job of keeping SPL constant at points a small distance from it if the signal has the widest possible bandwidth. Bandwidths of about 1/6 to 1/3 octave provide a reasonable compromise between getting a smooth acoustic field while retaining a frequency-specific stimulus. These bandwidths are commonly achieved by using warble tones, narrow bands of noise, or a broadband noise. In the latter case, the analyzer creates the necessary bandwidth as it analyzes the output.

4.6 Aided Threshold Testing and Functional Gain

Prior to the introduction of probe-tube equipment, hearing aid real-ear gain was tested by finding the hearing thresholds in a sound field while the person was aided and while he or she was unaided. The difference between these thresholds is known as ***functional gain***. Except in certain circumstances discussed below, and in the absence of measurement error, functional gain is identical to insertion gain.[455, 1155] If the hearing aid is operating in a nonlinear region for either measurement, then insertion gain and functional gain are equal only if the insertion gain is measured with the input level equal to the aided threshold.

The similarity and difference between the two gains can be summarized as follows:

- For insertion gain, the field level is the same for the unaided and aided measurements, and the acoustic effect of inserting the hearing aid on eardrum SPL is measured.
- For functional gain, the eardrum level is the same for the unaided and aided measurements, and the acoustic effect of inserting the hearing aid on field SPL is measured.

In both cases, the difference is the effect of inserting the hearing aid on the transfer function from sound field to eardrum. Although insertion gain and functional gain are similar in concept, they have different measurement errors associated with them. These random measurement errors, particularly the not-so-small errors inherent in measurement of functional gain, prevent the measured insertion gain from being *precisely* equal to the measured functional gain.

Insertion gain has a number of advantages over functional gain because insertion gain:

- is more accurate;
- can be measured in less time;
- gives results at many finely spaced frequencies instead of just the audiometric frequencies;
- can be measured at a range of input levels (see below);
- is not affected by the problem of masked aided thresholds (see below); and
- requires the hearing aid wearer only to sit still.

A severe disadvantage of functional gain testing is that it can conveniently be performed at only one input level for each patient – the level at which threshold is obtained. This is not a problem for linear hearing aids, but for nonlinear hearing aids, gain varies with input level, and we are explicitly interested in the gain at different input levels.

Another disadvantage of aided threshold testing (and therefore of functional gain testing) is that for people with near-normal hearing at any frequency, aided thresholds will often be invalid. A problem occurs when noise in the environment, or noise internal to the hearing aid, masks the test signal.[1114] The result is a functional gain that is lower than the insertion gain of the hearing aid. In this case, it is the insertion gain rather than the functional gain that properly portrays the increase in audibility provided by the hearing aid to most signals in the environment. A further complication with aided threshold testing occurs when hearing aids with very slow compression characteristics are tested: the gain of the hearing aid when a stimulus is presented may be affected by the level of the preceding stimulus, and by the time interval between stimuli presentations.[1006]

Because of these advantages, insertion gain (or alternatively, REAG or REAR) has replaced functional gain in the clinic. In some circumstances, however, it may be useful to measure aided thresholds in a sound field. Aided thresholds may be the only alternative for a child that is too active to allow a probe tube to be inserted (but see Chapter 16 for some suggestions with this).

Aided thresholds also have the advantage of being able to check the entire hearing aid and hearing mechanism, including the middle ear, cochlea and some aspects of the central auditory system. Obtaining an aided threshold provides a check that signal at each frequency will at least be audible, and this check may be especially valuable for clients with a profound hearing loss. Insertion gain measurement may correctly indicate that the hearing aid has a gain of 50 dB, but if the hearing aid OSPL90 is less than the person's thresholds nothing will be heard at that frequency.[1705] Aided threshold measurement at least alerts us to this extreme problem, but certainly does not assure us that the OSPL90 is optimal. (Although it is not recommended, insertion gain can even be measured on a cadaver, so high levels of gain are no guarantee of audibility!)

An additional advantage of aided threshold measurement is that, unlike inserting a probe tube, the measurement process itself cannot induce feedback oscillations (Section 4.7.3). Probe tube-induced feedback can be a problem for wearers of high-gain hearing aids and deeply seated hearing aids. Measuring the real-ear gain of deeply seated CICs (or other deeply seated hearing aids) can be difficult because of the incompressibility of the bony part of the canal. The probe tube reportedly can also become squashed because of the greater incompressibility of the bony canal. Two solutions to this problem are:

- Order the hearing aid with a purpose-drilled hole through which a probe tube can be inserted. The hole can be filled after testing. The only limitation is that the hearing aid has to have enough space available. This is less likely to be available if it also contains an internal vent. Remember that external vents are always an option.
- Another novel suggestion is to measure the functional gain of these aids, but using circumaural earphones both as the sound source and to attenuate room noise.[98] Functional gain measured in this way is equivalent to that measured in the sound field because the gain of a CIC is independent of signal azimuth.[1597] The major limitations are those

discussed above – masking by ambient noise if aided thresholds are too good, and the measured gain being applicable to only low input levels if the hearing aid is nonlinear. If circumaural earphones (ones that seal against the head) were to be used for this testing, this approach would also have the advantage of substantially attenuating background noise. This makes functional gain testing possible in less than perfect situations.

In summary, it does not seem time-effective to measure functional gain or aided thresholds *if* real-ear gain can be measured, and if a systematic procedure has been used to select and fine-tune hearing aid OSPL90. Aided threshold and functional gain measurement may at best be justified for some profoundly impaired patients, for some very young children, and for some wearers of CIC hearing aids.

4.7 Feedback in Hearing Aids

4.7.1 The feedback mechanism

Feedback oscillation (when a hearing aid whistles) is a major problem with hearing aids. The term ***feedback*** literally means that some of the output of the hearing aid manages to get back to the input of the aid (i.e., it is fed back to the input). Of course, when it does get back to the input, it is amplified along with every other signal arriving at the input. Unfortunately, it is not just any other signal. It has already traveled a complete loop from the microphone, through the amplifier, through the receiver, into the residual ear canal volume, and then back to the microphone via some path, as shown in Figure 4.19. If it has grown stronger while traversing around that loop once, then it will grow stronger still the next time, and the next and the next, and so on.

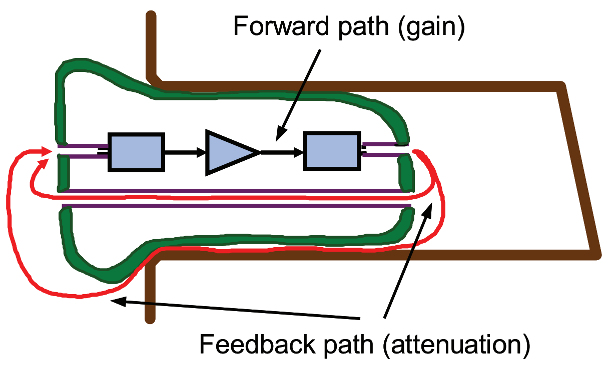

Figure 4.19 The feedback mechanism in hearing aids.

The process will stop only when the signal is so strong that the hearing aid changes its operating characteristics sufficiently because the signal has grown so large. For a linear hearing aid, this will be when the output limits by peak clipping or compression limiting; for a nonlinear hearing aid, it may be when the gain of the hearing aid decreases because of compression. Until this limiting occurs, the signal grows every time it passes around the loop no matter how small the original signal was. In fact, there does not need to be an original signal. An infinitesimally small random sound can start the process, and such sounds are always present.

Why doesn't this feedback process happen all the time? Sound is, in fact, always feeding back from the output to the input. It is just that the audible oscillations (whistling) can develop only when enough of it feeds back. Unfortunately, we (but not in *this* book!) loosely use the term *feedback* to mean the audible oscillation that results from the combination of feeding back a signal and then amplifying it sufficiently to cause an oscillation.

How much signal has to be fed back to create this unwanted oscillation? A moment's thought will reveal that if oscillations occur only when the signal gets larger every time it goes around the loop, then *oscillations can happen only if the amount of amplification through the hearing aid is greater than the amount of attenuation from the ear canal back to the microphone*. Thus, only if the real-ear aided gain of the hearing aid, (from input to the residual ear canal volume) is greater than the attenuation (from the residual ear canal volume back to the microphone) can continuous feedback oscillations occur. We can express this in another way by saying that the ***open-loop gain*** of the hearing aid (the total gain travelling forward through the hearing aid amplifier and transducers, and then backward through the leakage path) has to be greater than 0 dB.[714]

Suppose, for example, that a test signal emerging from a hearing aid had an SPL of 90 dB SPL in the residual ear canal, but had an SPL of 60 dB SPL by the time it leaked back to the microphone inlet via a vent. This hearing aid could not whistle if the REAG was less than 30 dB, but the hearing aid may whistle if the REAG was greater than 30 dB. One might expect that if the REAG were 31 dB, then the signal would get 1 dB stronger every time it went around the loop. However, every new sound out of the hearing aid adds to the sound that was already there. The combined signal leaks back to the microphone, and so the signal can grow stronger only if the sound adds in phase with the other oscillation already present at the hearing aid output. This requirement involving the phase response of the hearing aid is the second condition needed for feedback oscillation to occur: the total delay around the entire loop must be an integer number of periods of the feedback signal. Stated differently*, for oscillation to occur, the phase shift around the entire loop must be an integer multiple of 360°*, because 360° is the phase shift associated with a complete cycle.[s]

Because of the large delay in digital hearing aids, phase shift increases rapidly with frequency, so if the gain criterion for feedback is met, there are very likely one or more frequencies for which the phase criterion also will be met. Paradoxically, while digital processing has allowed the development of feedback-path cancellation (Section 8.2.3), digital processing has also made it more likely that feedback oscillations occur. The net effect of digital technology on the gain that can be achieved without oscillation is, however, beneficial.

We can turn these two requirements around the other way to determine at what frequency (or occasionally, frequencies) a hearing aid will oscillate: a hearing aid will oscillate at any frequency at which the forward gain is greater than the leakage attenuation, and at which the phase shift around the entire loop is an integer multiple of 360°.

When the sound combines with the sound already in the ear canal in this constructive way, it is called ***positive feedback***, irrespective of whether there is enough gain relative to the attenuation of the return path to actually cause oscillations. Positive feedback acts to increase the gain of the hearing aid. Indeed, a whistling hearing aid can be considered to have infinite gain at the frequency of oscillation: it has an output for no input at all. When the complete loop has a phase shift of 180°, 540°, or 900°, and so on in 360° steps, the sound fed back partially cancels any incoming sound. The effective gain of the hearing aid is decreased and we refer to this process as ***negative feedback***. Negative feedback cannot cause oscillations. As with everyday use of these technical terms, positive feedback causes something to increase, whereas negative feedback causes something to decrease. For hearing aids, the "something" is their gain.

Notice that hearing aid OSPL90 has not been mentioned. For a given amount of attenuation and phase shift, only the gain determines whether feedback oscillations will occur. A high-power hearing aid and a low-power hearing aid, adjusted to have the same gain, are equally likely to whistle. It is tempting to think that the high-power aid needs a tighter earmold to "hold back the sound." This is simply not correct; it is a high-gain hearing aid that needs a tight mold. The only effect of OSPL90 is that a high-power hearing aid will whistle more intensely than a low-power aid, should they both happen to whistle because of their gain, phase shift, and attenuation of the leakage path.

Anything that increases the amount of sound fed back to the microphone increases the likelihood of feedback oscillation. Some common causes are:

- positioning a sound reflector near the hearing aid, such as a telephone, or the brim of a hat; even movements of a telephone away from the ear by as little as 10 mm might avoid the problem;[1721]
- talking or chewing, such that the ear canal changes shapes and creates a sound path past the mold;
- growth of the ear canal, particularly in children; and
- shrinkage of the mold when it becomes old.

4.7.2 Effects of feedback on sound quality

Excessive feedback has two adverse effects. The first, audible whistling, is obvious, although sometimes it is obvious to everyone in the room *except* the person wearing the hearing aid. This happens if the aid

[s] To be precise, this is the phase condition needed for feedback when the open loop gain equals 0 dB exactly. The further the open loop gain increases above 0 dB, the greater is the range of open loop phase responses for which feedback can occur. For an open loop gain of 6 dB, for example, oscillation will occur if the phase response of the loop is anywhere within ± 60° of an integer multiple of 360°. The general condition is that $g \cos\theta > 1$, where g is the multiplicative open loop gain, and θ is the phase shift around the loop. The term $g \cos\theta$ can be thought of as the in-phase part of the open loop gain.

wearer has so much hearing loss at a frequency that even maximum output from the hearing aid at this frequency is inaudible. This highly embarrassing situation is much less common than in the past, thanks to the widespread use of feedback cancellation (Section 8.2.3). A hearing aid can oscillate at a frequency only if there is enough gain at that frequency, and there is no point in providing gain if the aid wearer cannot hear a signal at maximum output SPL at that frequency. As hearing aids have become more flexible, it is relatively easy to decrease gain in specific frequency regions to avoid this problem.

The second problem is subtler. When the hearing aid gain is set a few dB below the point at which the aid continually oscillates, the signal feeding back will still cause the gain to increase at frequencies where the feedback is positive, and to decrease the gain where the feedback is negative. Feedback thus induces extra peakiness in the hearing aid response and these peaks occur at the potential feedback frequencies.[357] Every time a sound with components at these frequencies is put into the hearing aid, the hearing aid ***rings*** for a little while after the signal has ceased. (The ringing mechanism is in fact very similar to the reason why a bell continues to vibrate and sound after it has been struck: the hearing aid or bell stores energy and gradually releases it at this frequency over the next few hundredths of a second.) Most people have experienced this ringing effect, also known as ***sub-oscillatory feedback***, when a public address system is turned up to the point where it is almost continuously oscillating. The sound quality is annoying.

Both the increased peakiness, and the ringing effect rapidly decrease as the gain of the hearing aid is decreased below the point at which feedback oscillation becomes continuous, as shown in Figure 4.20. By 10 dB below the onset of whistling, positive and negative feedback can at most cause the gain to increase by 3 dB and decrease by 2 dB, respectively. It is difficult to say how far below onset the gain must be decreased for the ringing sound to disappear, as it depends on how peaky the hearing aid response is without any feedback being present. However, 5 or 6 dB of gain reduction is likely to be sufficient.

4.7.3 Probe-tube measurements and feedback

A probe tube can *cause* feedback. Inserting the probe between the mold and the canal wall creates small additional leakage paths on either side of the probe, as shown in Figure 4.21. This leakage decreases the attenuation for the return part of the loop. A hearing aid may thus whistle when it is being measured but be totally satisfactory otherwise.

Even if there is no leakage around the probe tube, there can be leakage through the wall of the probe. The tip of the probe is in the residual ear canal, so the full output of the hearing aid exists at all points within the probe tube. This high-level acoustical signal vibrates the walls of the probe and hence the air outside the probe near the hearing aid microphone. Both of these leakage paths are significant only for high-gain hearing aids and some CICs. All other hearing aids will have been made with molds or shells sufficiently loose that the extra leakage created by

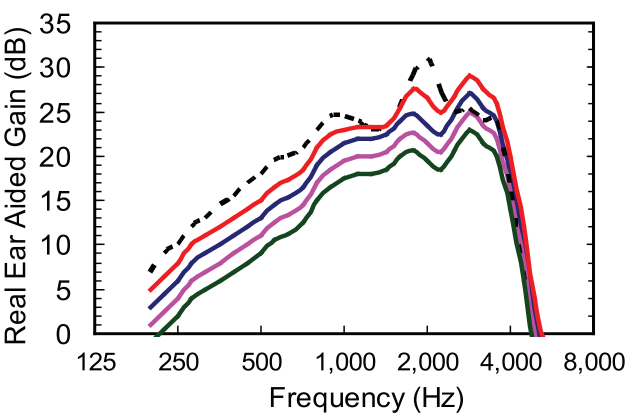

Figure 4.20 Coupler gain of a hearing aid with the volume control adjusted in 2 dB steps. One further increase resulted in oscillation.

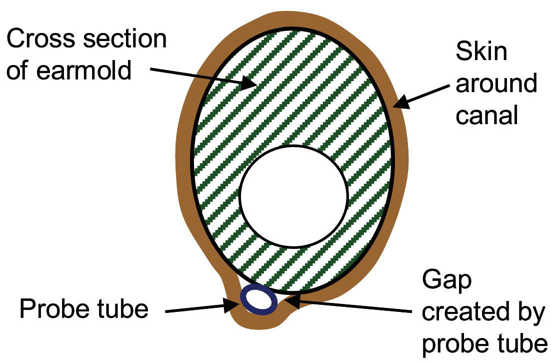

Figure 4.21 Leakage paths created by the insertion of a probe tube between an earmold or shell and the ear canal.

Practical tip: Avoiding probe-induced feedback

- Decrease the gain of the hearing aid by 10 dB (or so) below that which is required and then measure the shape of the gain-frequency response. Mentally add 10 dB to the gain at each frequency when comparing it to the target gain.
- Put some thick lubricating jelly on the mold or shell on the surface where it contacts the probe tube.

the probe tube is insignificant. Do not attempt to measure the gain of a hearing aid that is oscillating (i.e. whistling). The oscillation can adversely affect the operation of the hearing aid at all frequencies. Agnew (1996) discusses many aspects of feedback in detail in an excellent review.

4.8 Troubleshooting Faulty Hearing Aids

Clinicians do not usually make major repairs to hearing aids. Many repairs, however, are minor and can be done by anyone who can diagnose the problem. Also, when a patient returns a faulty aid, it is usually the clinician who has to decide whether to return the aid to the manufacturer for repair or take some other action. It is inconvenient and unnecessarily expensive for aids to be returned to a manufacturer when a repair could have been done on the spot in a few minutes.

It is useful for the clinician to be able to hear the output of the hearing aid, and there are several ways to achieve this.

- A simple accessory is a ***stethoclip***, as shown in Figure 4.22. A stethoclip (also known as stetoclip, stethoset, and stetoset) allows the clinician to hear the output of the hearing aid without having to wear it. For high-powered hearing aids, a damper, or several dampers, can be placed in the stethoclip tubing to decrease the output to comfortable levels for a normal-hearing person.
- A custom earmold (made to fit the clinician!) can be attached to a long tube that has an enlarged and flexible cupped end.
- There are several electronic devices available in which the hearing aid is connected to a coupler, and the output of the coupler is amplified and presented through headphones. These have the advantage that a comfortable listening level can easily be obtained, even for high-powered hearing aids.
- For BTE hearing aids that terminate in a dome, the hearing aid (with a fresh dome) can be worn directly by the clinician.
- Most real-ear gain analyzers come with a set of headphones that allow the clinician to hear the sounds present in the client's ear canal. Whenever the probe microphone is inserted, the clinician can listen to the sound while the client identifies precisely what aspect of the sound quality is unacceptable. *This method is invaluable if the clinician is in any doubt about the nature of the noise or distortion that the client is describing.*

Some hearing aid companies offer courses showing how to cut open custom hearing aids and effect straightforward repairs. On the other hand, opening a hearing aid usually (if not always) voids any warranty, and a repair attempted, but badly done, may make it impossible for the manufacturer to then repair the aid.

Figure 4.22 A stethoclip attached to a CIC hearing aid.

When diagnosing faults in hearing aids, it is important to be clear about the distinction between noise, distortion, and interference:

- ***Noise*** in a hearing aid output is an unwanted part of the output that is present whether or not a signal is being input to the aid. It may originate totally from within the hearing aid in which case it is referred to as internal noise, it may be an amplified version of some external noise (e.g. air conditioning noise), or it may originate from some external non-acoustic source, in which case we will refer to it as interference.
- ***Interference*** is the creation of a noise in the output of a hearing aid by a magnetic, electrostatic, or electromagnetic field near the hearing aid.
- ***Distortion*** is an unwanted part of the output that is present only when a signal is being amplified. It will usually be audible as a signal of poor quality rather than as something that is present in addition to, or in the absence of, the signal.

Interference in hearing aids by other electronic devices has received a lot of attention during their design. This effort has occurred because the signal transmitted by some digital mobile telephones is particularly effective at interfering with hearing aids, as described in Section 3.11.3.

The following tables list some possible causes of hearing aid faults, and the remedial action required for each, grouped according to the symptom. The comments apply to all types of hearing aids, except where otherwise indicated. If the hearing aid operates intermittently from one second to the next, look particularly for problems with the battery contacts (see Table 4.8 and 4.9) or the connections to the transducers. If the hearing aid output diminishes in strength or quality each day, returning to good performance each morning, look for cerumen build-up in the wax guard, dampers, sound bore or receiver screen. This is referred to as the rainforest effect because each day the high humidity in the ear canal reactivates and expands the dried-out cerumen lodged in the hearing aid (see Table 4.8).

Table 4.8 The audio output from the hearing aid is weak.

Possible cause	Diagnosis	Remedy
Weak battery	Test battery or try a new one	Replace battery
Dirty battery contacts	Visual inspection	Clean with spirits soaked into a cotton bud
Corroded battery contacts	Visual inspection	Clean with abrasive paper, or return to manufacturer
Clogged sound bore or receiver	Visual inspection	Clean with loop
Clogged wax-guard (ITE/ITC/CIC)	Visual inspection, plus output restored when wax-guard removed	Replace wax guard
Clogged damper (BTE)	Output restored (and hearing aid feeds back) when earhook is removed	Replace damper
Clogged microphone inlet port	Visual inspection, or thump audible when the aid is tapped	Clean inlet port with a fine pick. Replace tubing if it is perished.
Inadvertent re-programming or de-programming	Check program settings (only applicable to programmable aids)	Re-program. Return to manufacturer if fault re-occurs.
Faulty microphone	Aid works on telecoil or audio input (if present), and internal noise audible at high volume control setting	Send to manufacturer
Faulty amplifier or transducer	No other discernable fault	Send to manufacturer

Table 4.9 There is no audible sound from the hearing aid. Consider all of the items in Table 4.8, plus the following.

Possible cause	Diagnosis	Remedy
Dead battery	Test battery or try a new one	Replace battery
Bent battery contacts	Visual inspection, plus jiggling battery compartment causes intermittent operation	Bend contacts carefully (this may provide a temporary cure only), or send to manufacturer for replacement of contacts
Faulty wiring	No other discernable fault	Send to manufacturer

Table 4.10 The output from the hearing aid is distorted.

Possible cause	Diagnosis	Remedy
Weak battery	Test battery or try a new one	Replace battery
OSPL90 excessively decreased (if a peak clipper)	Problem disappears for low input levels or higher OSPL90 settings	Increase OSPL90, or fit a hearing aid with compression limiting and/or wide dynamic range compression
Dirty battery contacts	Noise occurs when battery or battery compartment is moved slightly	Clean contacts with eraser
Faulty transducer or amplifier	No other discernable fault	Send to manufacturer

Table 4.11 The output of the hearing aid is noisy

Possible cause	Diagnosis	Remedy
Faulty volume control or tone control	Noise increases or decreases markedly when the control is moved slightly	Send to manufacturer for replacement of component
Interference from computer, electric motor, transmitter, mobile phone, car ignition, or other electromagnetic source	Interference noise is present at particular times, or in particular places	Avoid the source of interference, or upgrade hearing aid to one with greater immunity to interference
Hearing aid is switched to the T position!	Hum disappears and signal reappears when switched to M position	Re-instruct user about the function and use of the M-T switch, or disable the T position
Dirty battery contacts	Noise changes when battery or battery compartment is moved slightly	Clean contacts with eraser
Faulty transducer, wiring, or amplifier	No other discernable fault	Send to manufacturer
Faulty microphone	Noise like radio static which increases with changes to gain	Send to manufacturer

As discussed previously, feedback is always caused by a signal leaking from somewhere back to an earlier point in the chain. The source of the feedback can sometimes be detected with a stethoclip by positioning the open end of the tubing at each of the points where sound could be escaping.[1151] Figures 4.23 and 4.24 show the major points in ITE/ITC/CIC hearing aids and BTE hearing aids, respectively, at which sound leaks.

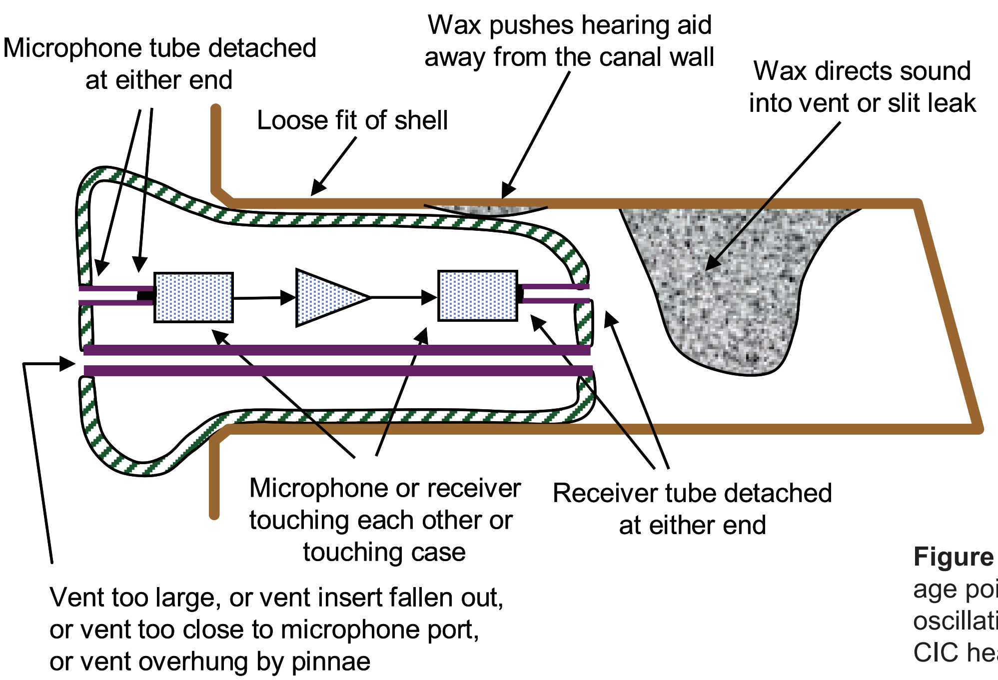

Figure 4.23 Common leakage points, leading to feedback oscillation, in ITE, ITC, and CIC hearing aids.

Table 4.12 Feedback causes an ITE/ITC/CIC hearing aid to whistle

Possible cause	Diagnosis	Remedy
Shell improperly inserted	Visual inspection	Re-instruct client on insertion technique
Shell no longer fits ear snugly (especially for unvented aids)	Whistling stops when thick lubricating jelly is smeared over the canal stalk, or when a Comply™ Soft Wrap or E-A-R Ring™ Seal encircles the canal stalk	Add build-up material to shell, or re-make shell
Venting insert or plug has fallen out	Visual inspection, compared to record of fitting on file	Insert (and glue in!) a new venting insert or plug
Microphone or earphone has moved and is touching the case or the other transducer	Whistling continues when the microphone inlet port is blocked with a finger	Return to manufacturer for re-positioning
Microphone tubing detached from microphone or case	Whistling continues when the microphone inlet port is blocked with a finger	Return to manufacturer for re-attachment
Receiver tubing detached from receiver	Whistling continues when the outlet hole is blocked with a finger	Return to manufacturer for re-attachment
Receiver tubing detached from tip of earshell	Visual inspection; whistling continues when the outlet hole is blocked with a finger	Grip carefully with fine tweezers, reposition, and glue (or return to manufacturer)

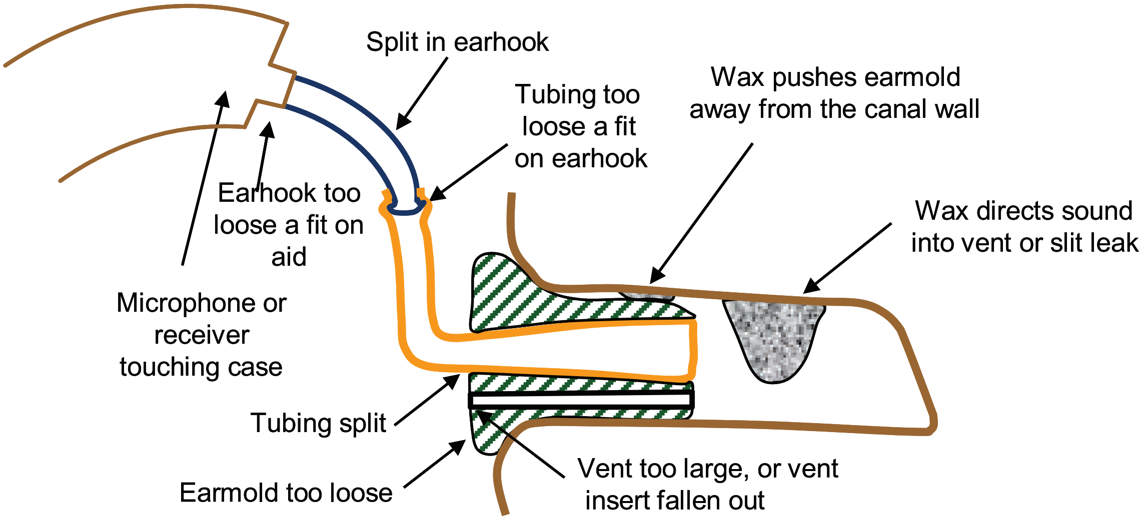

Figure 4.24 Common leakage points, leading to feedback oscillation, in BTE hearing aids.

Table 4.13 Feedback causes a BTE hearing aid to whistle

Possible cause	Diagnosis	Remedy
Mold improperly inserted	Visual inspection	Re-instruct client on insertion technique and/or modify mold shape
Mold no longer fits ear snugly (especially for unvented aids)	Whistling stops when thick lubricating jelly is smeared over the canal stalk, or when an E-A-R Ring™ Seal or Comply™ Soft Wrap encircles the canal stalk	Re-make mold
Microphone tubing detached from microphone or case	Whistling continues when the microphone inlet port is blocked with a finger	Open case and re-attach tubing, or return to manufacturer for re-attachment
Receiver tubing detached from receiver	Whistling continues when the outlet hole of the aid case is blocked with a finger	Open case and re-attach tubing, or return to manufacturer for re-attachment
Receiver tubing detached from case of hearing aid	Visual inspection; whistling continues when the outlet hole of the aid case is blocked with a finger	Grip carefully with fine tweezers, reposition, and glue, or open and re-glue, or return to manufacturer
Split in earhook or leak at junction of earhook and aid (i.e. hook is too loose)	Visual inspection; whistling continues when finger is placed over tip of earhook	Replace earhook
Split in tubing, or tubing a loose fit on the earhook	Visual inspection; whistling continues when finger is placed over tip of earmold	Replace tubing

4.9 Concluding Comments

Every clinician has to be competent at measuring hearing aids. The clinician must know the different types of hearing aid gains and be familiar with the various methods of displaying performance. If a clinician cannot confidently measure a hearing aid in a test box, the clinician has no way to determine whether a hearing aid is operating to the manufacturer's specifications. If a clinician cannot confidently measure a hearing aid in a patient's ear, the clinician has no way to determine whether the hearing aid is adjusted as closely as possible to the prescription target for that patient.

With the increasing sophistication of hearing aid fitting software, and the advent of trainable hearing aids (Section 8.5), the time may come when real-ear measurements are not needed as a routine part of hearing aid fitting, but rather are kept as an invaluable troubleshooting tool. However, until one can be confident that the predicted gain shown on a manufacturer's screen is a close approximation of the gain in an individual client's ear and of the prescription target, measurement of real-ear gain is an important part of the hearing aid fitting process.[2, 4, 449]

Although strong opinions are sometimes expressed about whether REIG, REAG, or REAR are the most useful responses to measure, and although each has small advantages and disadvantages (Section 10.2.4), none of them have a significant overall advantage over the others, and using any one of them has a significant advantage over using none of them.[449]

Soon to come will be hearing aids that have an internal microphone that senses the SPL within the ear canal. These microphones will fulfill several purposes (including active occlusion reduction, detection of hearing aid wearer's own voice, and detection of receiver malfunction), but will also allow measurement of RECD without needing any probe tube. They will also allow a complete measurement of REAG without any other equipment, as the two microphones will allow measurement of both the input SPL and the output SPL of the hearing aid. Alternatively, the hearing aid will generate the test signal electronically, with no need for a sound field.

Already available are hearing aids that are measured by attaching one end of a probe tube to the usual hearing aid microphone inlet, and positioning the other end, as usual, in the ear canal. RECD is measured while the hearing aid generates a sound electronically, and from this, REAG is calculated. Also already available are hearing aids in which the fitting software initiates measurement of real-ear gain, notes the discrepancy from target, and automatically adjusts the hearing aid to minimize the discrepancy.

As with all measurements, hearing aids can be measured reliably and accurately only if the clinician has an appreciation of what can go wrong with each measurement, techniques for minimizing the chance of an error occurring, and an understanding of what a correct measurement would look like.

CHAPTER 5

HEARING AID EARMOLDS, EARSHELLS AND COUPLING SYSTEMS

Synopsis

An earmold or earshell is molded to fit an individual's ear and retains the hearing aid in the ear. Pre-molded canal fittings, available in a range of standard sizes and shapes, are an alternative way to connect the hearing aid to the ear canal. Whether custom molded or pre-formed the ear fitting retains the sound bore, which is a sound path from the receiver to the ear canal. In many cases the fitting provides a second sound path, referred to as a vent, between the air outside the head and inside the ear canal. Where no vent exists, as in high-gain hearing aids, the fitting is said to be occluding. Where the cross-section of the ear canal remains largely unfilled for its entire length, the hearing aid is said to be an open fitting, or an open-canal fitting. The three functions of an ear fitting are thus physical retention, transmission of amplified sound to the ear canal, and control of the direct sound path between the ear canal and the air outside the head.

There is a wide variety of physical styles of both earmolds and earshells. These styles vary in the extent of the concha and canal that they fill. These variations affect the appearance, acoustic performance, comfort, and security of retention of the hearing aid.

One unwanted consequence of a hearing aid can be an occlusion effect, in which the aid wearer's own voice is excessively amplified by bone-conducted sound. For many hearing aid fittings, vent selection is a careful juggle between choosing a vent that is big enough to avoid an unacceptable occlusion effect, but not so big that it causes feedback oscillations, or limits the ability to achieve sufficient low-frequency gain and maximum output. For patients with mild or moderate hearing loss, the choice will often be an extremely open fitting comprising a BTE connected to thin tubing, or a wire connection for a RITE style, terminating in a pre-formed, flexible, perforated, dome-shaped canal fitting.

For any ear fitting with a vent or other direct path to the outside air, the speech range of frequencies can be subdivided into the vent-transmitted frequency range, the amplified frequency range, and the mixed frequency range that is intermediate to these. Hearing aids perform very differently in each of these ranges.

The shape of the sound bore that connects the receiver to the ear canal affects the high-frequency gain and output of hearing aids. Sound bores that widen as they progress inwards (horns) increase the high-frequency output. Conversely, those that narrow (constrictions), whether by design or as a consequence of poor construction technique, decrease the high-frequency output. Horns have to exceed a certain length if they are to be effective within the frequency range of the hearing aid.

Dampers are used within the sound bore to smooth peaks in the gain-frequency response. Careful choice of the placement and resistance of the damper can also control the mid-frequency slope of the response.

The key to a well-fitting earmold is an accurate ear impression. This requires an appropriate material (medium viscosity silicone is good for most purposes), a canal block positioned sufficiently deeply in the canal, and smooth injection of the impression material. Tighter earmolds or shells that reduce leakage of sound from the ear canal can be achieved by a variety of techniques. These techniques include taking an impression with the patient's jaw open, patting down the impression material before it sets, using viscous impression material, and building up the impression in the patient's ear.

Earmolds are made from a variety of materials. The most important difference between materials is hardness. Soft materials provide a better seal to the ear, but they deteriorate more rapidly, can be more difficult to insert, and are more difficult to modify and repair. Earmolds and earshells are routinely constructed by computer-aided manufacture in which lasers guide the "printing" of plastic based on a scanned image of the ear impression.

The earmold (for a BTE), the earshell (for an ITE, ITC, or CIC), and the pre-molded canal fitting (for a thin-tube or RITE BTE) will collectively be referred to as ear fittings. An ear fitting performs three essential functions:

- it couples sound from the hearing aid receiver to the aid wearer's ear canal via the ***sound bore*** (a tube), and consequently affects the gain-frequency response of the hearing aid;
- it controls the extent to which the inner part of the ear canal is open to the air outside the head (the ***venting***), and consequently affects the gain-frequency response, and electroacoustic comfort of the hearing aid; and
- it retains the hearing aid in the ear in a comfortable way.

There is an initially bewildering array of ear fitting styles and materials, some of them proprietary to particular manufacturers of earmold or hearing aids, and some with multiple names. This chapter will help the clinician select an ear fitting that achieves a required combination of sound bore acoustic characteristics, venting characteristics, and retention characteristics.

Consider, for instance, the two earmolds shown in cross-section in Figure 5.1. These two earmolds look completely different. Earmold *(a)* is a very bulky earmold that completely fills the concha, but it has a vent drilled through the mold. Earmold *(b)* contains very little material in the concha and in the canal, and may be referred to as a ***CROS mold*** or ***Janssen mold***. Provided, however, the cross-sectional area of the drilled vent in earmold *(a)* equals the cross-sectional area of the open space between the sound bore and the canal walls in earmold *(b)*, and provided the two sound bores have the same length and internal diameter, the two molds will have extremely similar acoustic effects on the gain-frequency response and OSPL90 of the hearing aid. Earmold *(a)* will, of course, be retained much more tightly in the ear than earmold *(b)*.

In this book, *any opening between the inner part of the ear canal and the free air outside the ear will be called a* ***vent***, irrespective of whether it has been formed by drilling a hole (Figure 5.1*a*) or by forming the canal portion of the ear fitting so that it does not completely fill the cross-sectional area of the ear canal (Figure 5.1*b*). These methods can be combined to provide a vent path comprised of a hole drilled through the concha part of the mold or shell leading to an open area within the canal portion. Conversely a canal dome fitting can have several openings in the dome within the canal, and no obstruction in the rest of the canal or in the concha. Yet another way to make a vent is to grind a groove along the outer surface of the mold or shell, all the way from the canal tip to the faceplate. This is called a ***trench vent*** or an ***external vent***.

Ear fittings can be ***occluding***, ***open***, or anywhere in-between. Occluded fittings are those that have no intentional air path between the inner part of the ear canal (the ***residual canal volume***) and the outside air. Occluded fittings therefore have no vent: the ear fit-

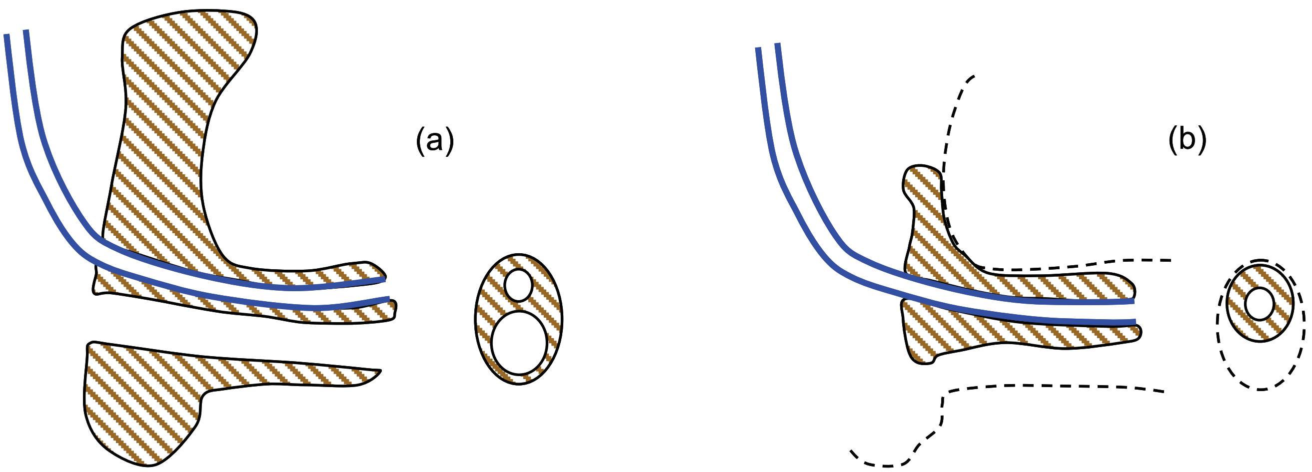

Figure 5.1 Cross sections of (a) a full concha earmold with a wide vent and (b) a Janssen mold that would have extremely similar acoustical properties, but different retention properties. See also Figure 5.4 for perspective views of these molds.

ting completely fills the cross-section of the canal, for at least part of its length. Occluded ear fittings may, and usually do, have a ***leakage path*** around them as a consequence of: imprecision in the impression of the ear; imprecision of the mold or shell made from the impression; incorrect size or shape of a pre-molded fitting; or flexibility of the ear canal. This leakage path has properties similar to those of a vent, and is sometimes referred to as a ***slit-leak vent***.

Open canal (***OC***) fittings are those that leave the canal almost completely open for its entire length, most commonly achieved with an open dome fitting as shown in Figures 1.7 and 5.4. The term ***non-occluding*** is more general; in fact, so general that it is practically meaningless. By non-occluding, some people mean that there is some vent path, no matter how small. Other people would describe an ear fitting as non-occluding only if most of the canal cross-sec-

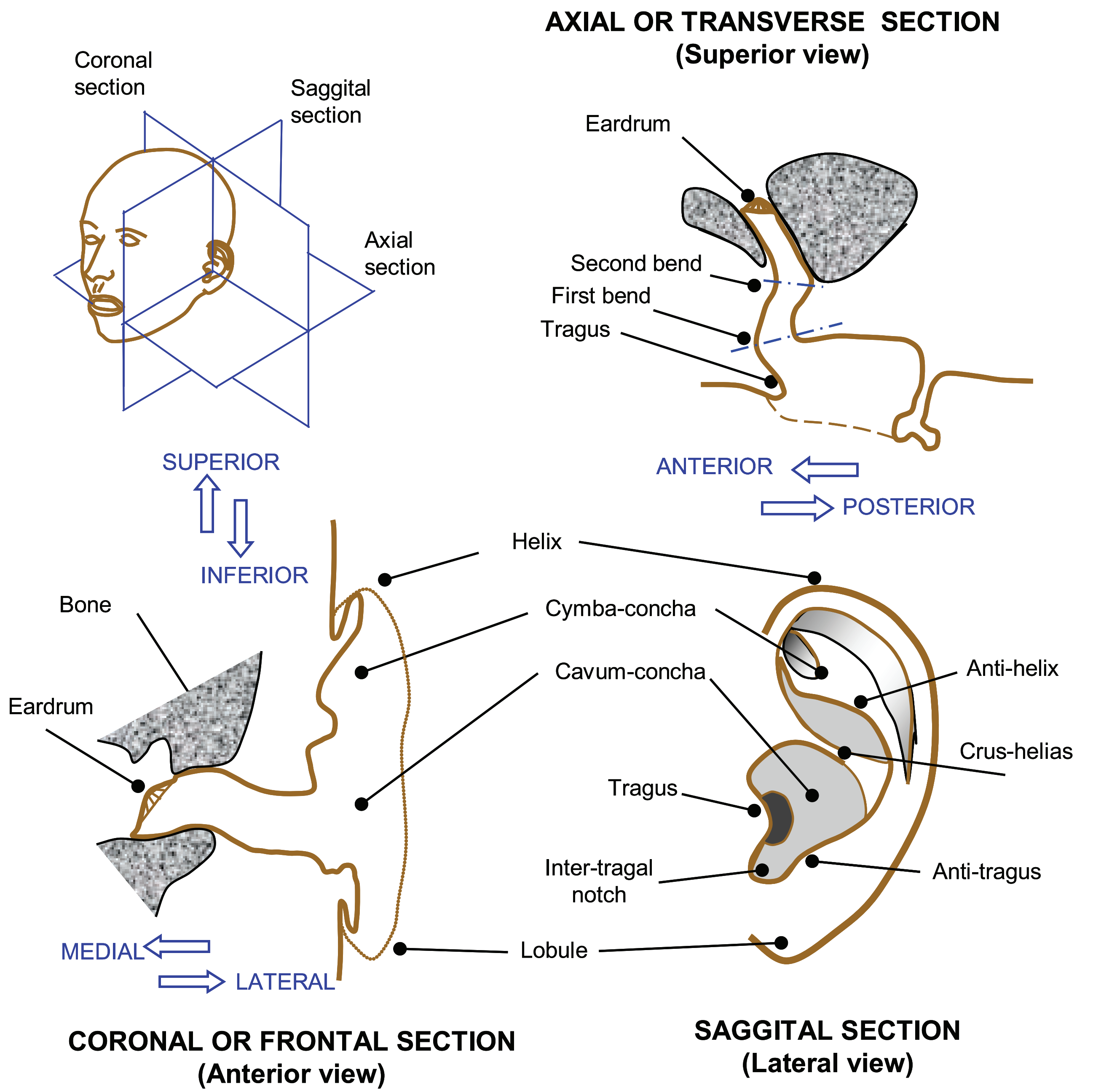

Figure 5.2 Side view and cross section of the external ear, drawn to average full-size dimensions and typical shape,[1542, 1684] and the names given to various parts of the ear.[1617]

tion were left open for its entire length. Some people may define *non-occluding* in terms of the hearing aid wearer's subjective impression of the earmold, as discussed in Section 5.3.2. There is, of course, a continuum of openness from being completely occluded to being completely open, and the term non-occluding, if used at all, should be used in a way that makes its intent clear. All "non-occluding" ear fittings are, in fact, partly occluding, even if the occlusion is minimal.

5.1 Earmold, Earshell and Canal Fitting Physical Styles

Earmolds and shells of different styles fill different portions of the concha and the canal. The parts of the molds and shells can be described by the corresponding parts of the ear in which they fit. Let us therefore review some names for parts of the ear, and for an earmold or earshell[37] as shown in Figures 5.2 and 5.3. Some features of the ear have particular significance for hearing aid fitting. The inner half of the ear canal, the ***bony canal***, is bounded by smooth skin only 0.2 mm thick overlaying bone[36] and is very sensitive to applied force. In the outer half of the canal, the ***cartilaginous canal***, the skin is much thicker, overlays cartilage, and is less sensitive. Cerumen is produced by glands and these are located only in the cartilaginous part of the canal. Earmold manufacturers refer to the section of the canal just inside the ear canal opening as the ***aperture***, and the corresponding part on the earmold can be called the ***aperturic seal*** (because the earmold most readily seals to the ear canal in this region).

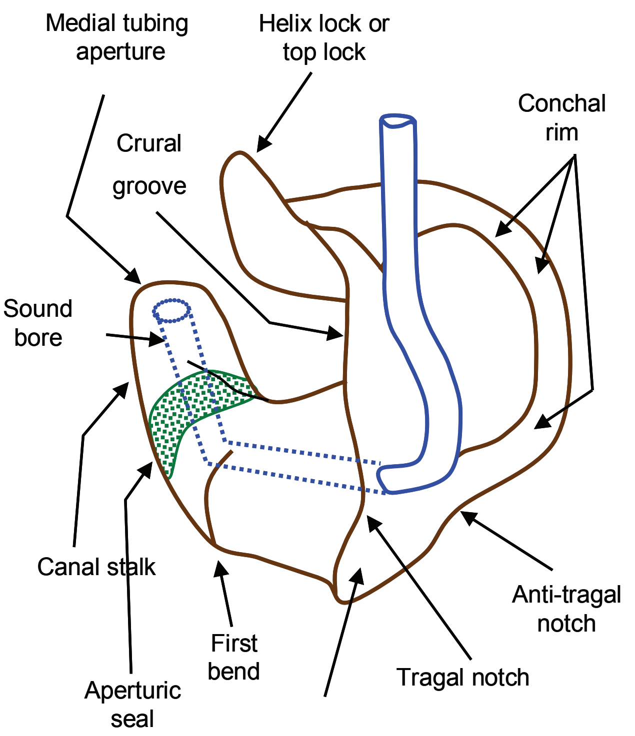

Figure 5.3 Names given to various parts of an earmold or ear shell, based in part on Alvord, Morgan & Cartright (1997).

The earmold has two easily recognizable bends. The ***first bend*** (the most lateral bend) although a pronounced feature on a mold or impression, is less evidently a bend when looking at the ear. The posterior surface of the tragus is continuous with the posterior wall of the canal. The first bend is in fact coincident with the ear canal entrance or a few mm inside the canal, depending on where one considers the entrance to be. The ***second bend*** marks the start of the transition from the cartilaginous canal to the bony canal, first on the posterior wall, and further in on the anterior wall. The first and second bends are much more acute for some people than for others. When people have a sharp first bend, however, they also tend to have a sharp second bend, so that the most inner and most outer segments of the canal tend to be parallel to each other when viewed as a transverse section.[1420]

5.1.1 BTE earmold styles

One of the difficulties in describing different styles of earmolds is the lack of standardization of names. Although the American National Association of Earmold Laboratories (NAEL) agreed on some standard names in 1976,[319] many new styles have been invented and re-invented since then. Some earmolds are usefully given a descriptive name (e.g. ***skeleton***), some are named after their inventor (e.g. ***Janssen***) and some are confusingly named after the application in which they were originally used (e.g. ***CROS***), even though they are used more frequently in other applications.

Figure 5.4 shows a number of earmold styles that are available from different earmold manufacturers. The names may vary from manufacturer to manufacturer. The diagram does *not* include styles that differ only in the diameter of the sound bore. Every one of the styles shown could include a sound bore that widens or constricts along its length, so it is unnecessarily confusing to give a new name to an earmold on the basis of its sound bore internal diameter(s). The effects of sound bore variation, and names for several commonly used sound bore shapes, will be covered in Section 5.4. Those in the bottom row are pre-molded canal fittings supplied by hearing aid manufacturers, and are typically used with thin tubing. These have become the most commonly used method of holding the tube of a BTE in the ear canal.

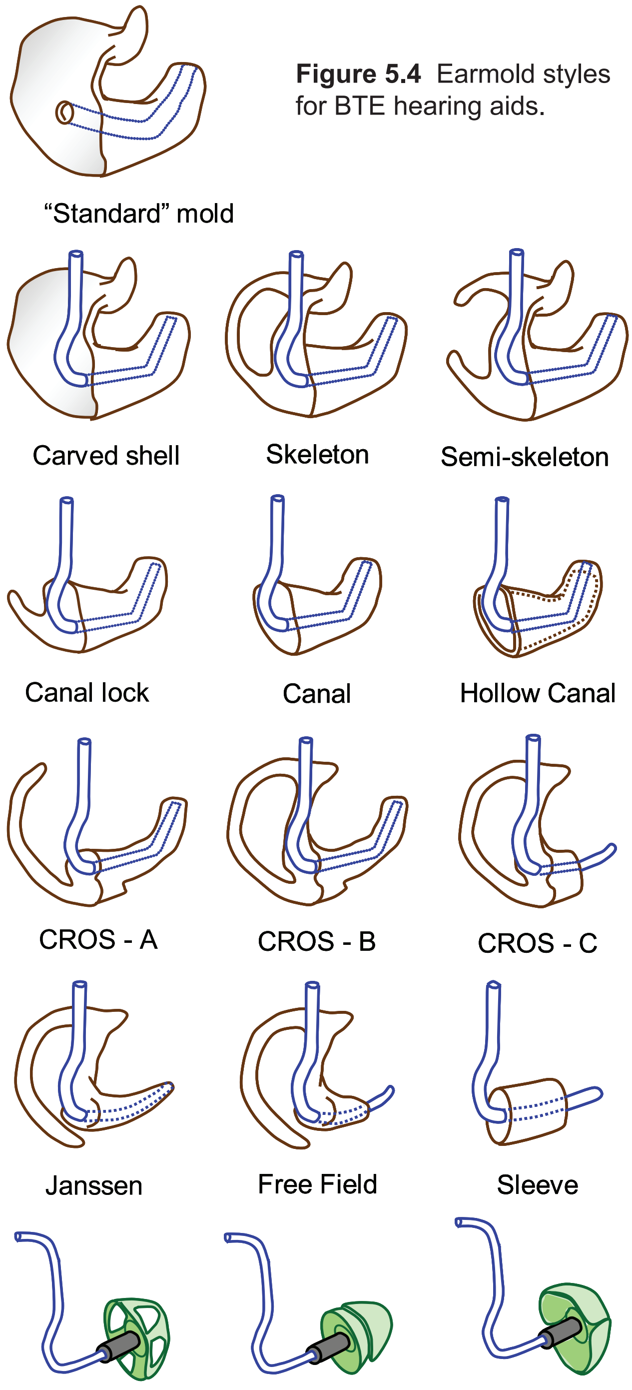

Figure 5.4 Earmold styles for BTE hearing aids.

The ***receiver mold*** (confusingly called a ***standard*** or ***regular*** mold, despite being rarely used these days) is the only one that can be used for a body aid: a button receiver clips firmly into the ring on the surface of the mold. It can also be used for a BTE aid by clipping a plastic angle piece into the ring. A length of tubing connects the angle piece to the hearing aid earhook. For BTE use, however, its disadvantages (leakage of sound, appearance, potentially decreased high-frequency response) outweigh its advantage (easy replacement of tubing).

A better way to enable tubing to be easily replaced is to have an elbow mounted in the earmold, to which the tubing is connected, as shown in Figure 5.5a. The sound bore inside the mold consists of a drilled hole rather than a tube. To avoid decreasing the high-frequency response of the hearing aid, the internal diameter of the elbow should be the same as that of the tubing. One particular brand of elbow that achieves this is known as a ***Continuous Flow Adapter (CFA)***™, as shown in Figure 5.5b.

The top seven earmolds shown in Figure 5.4 can be ordered as occluding earmolds, or they can be ordered with vents drilled through them. The remaining six molds and the open dome can never be completely occluding, because the canal portion of the mold does not fill the entire cross-section of the ear canal at any point along its length.

The most commonly used fittings are the pre-molded, ***dome***-shaped canal fittings and thin sound-bore tubing shown in the bottom row of Figure 5.4. These

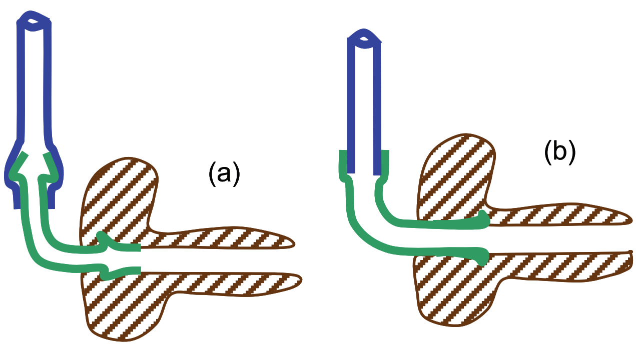

Figure 5.5 Two types of elbows used in BTE earmolds. In (a) the tubing fits around the elbow, which creates some constriction. In (b) the tubing fits inside the elbow.

have a soft flexible flange or flanges and come in a range of diameters, and tubing lengths. There are essentially two styles, the ***open dome*** with holes in the flange which is designed to leave the canal as open as possible, and the ***closed dome*** with no holes which is designed to seal the canal as completely as possible. Because the fitting is completely enclosed within the ear canal, and because it is mostly combined with thin tubing and very small BTE hearing aids, the open and closed dome styles, and the very similar tulip style, enable a very inconspicuous overall appearance. The same designs are used for RITE hearing aids, except that the central portion of the dome contains the receiver, encased in plastic, so the sound bore is only 1 or 2 mm long. For RITE hearing aids, the thin tube from the hearing aid is replaced by an equally thin electrical connection to the receiver, so the result is just as inconspicuous.

The dome and tulip fittings also have the advantage of not requiring an ear-impression to be made, and so facilitate same-day fitting. Selection of the correct dome size and tubing length is important: if the dome is too large it will be uncomfortable; if too small it will fall out, or itch as it moves around. If the tubing is too long or short then either the hearing aid will not sit comfortably behind the ear or the dome will not sit comfortably within the ear canal. The open dome fittings fulfill the same function acoustically as the ***sleeve mold***[235] and ***vented hollow canal mold***,[998] and have largely superseded them. The latter two designs may, however, be better retained in the ear.

The top four molds in Figure 5.4 are shown with the ***helix lock*** segment intact. Each of these molds can be ordered with the helix lock removed, or the helix lock can be cut or ground away by the clinician. Retaining the helix lock helps the mold stay in place, and thus maximizes security of the aid, provided the user can fully insert the mold with the helix lock properly tucked in under the helix and anti-helix. By helping retain the earmold in its correct position, the helix lock can also slightly decrease the likelihood of feedback.[1012, 1189] Unfortunately, many people cannot tuck in the helix lock properly, in which case its presence pushes the mold *out* of position, thus *increasing* feedback. The helix lock area of the mold can also create pressure discomfort. Some patients find it easier to insert the mold if the helix lock is removed.[1188] Consequently, some clinicians order molds without a helix lock for all patients, whereas others start with it in, and remove it if it creates problems.

There are systematic procedures that can be followed for determining how open (i.e. non-occluding) an earmold should be for a particular aid wearer, as outlined in Sections 5.3 and 5.7. It is not so clear how to systematically choose between molds that differ only in appearance, fragility, and degree of retention properties (e.g. shell versus skeleton versus semi-skeleton, or CROS-A versus CROS-B). There is no difference in the retention properties or occlusion properties of a shell versus a skeleton, because the material removed to turn a shell into a skeleton comes from the center of the concha region. As a general rule, the mold becomes less firmly anchored in the ear as more and more segments are removed from around the rim of the concha, and as the diameter of the canal stalk is decreased below the diameter of the ear canal itself. For people who have pinnae that move excessively during talking, chewing and head turning, however, the mold or shell may be best retained if it makes minimal contact with the concha, in which case a canal-sized ear fitting may be optimal.[841]

No matter what style of earmold or earshell is selected, there must be a ***retention region*** somewhere on it. A retention region is an area where the earmold or shell pushes against the skin if it were to start moving out of the ear canal. The part of the ear against which the retention region pushes might be the canal wall, the tragus, anti-tragus, or the helix. If the retention region is too small or not sufficiently angled against the exit motion, the earmold will work its way out of the ear. If the retention region is too large or excessively angled against the exit motion, it will be hard to insert the earmold.[1417]

5.1.2 ITE, ITC, and CIC earshell styles

Because the electronics of the hearing aid are inside the shell for an ITE, ITC or CIC hearing aid, there are fewer possibilities for alternative shell styles within each of these classes of hearing aids. ITE hearing aids that extend above the crus-helias are classified as ***full-concha ITEs***, those that are fully contained below the crus-helias are referred to as ***half-concha ITEs***, and those that fit entirely above the crus-helias could be called ***cymba-concha ITEs*****,** although they are too new to have developed a terminology. These differences are best seen from a lateral view, as shown in Figure 5.6.

If the full-concha or half-concha ITEs do not extend laterally sufficiently far to fill the concha, they are referred to as ***low-profile ITEs***. ITC hearing aids

Figure 5.6 Lateral view of different ITE, ITC, and CIC hearing aid styles, with the visible part of faceplate shown hatched.

that extend only part of the way along the posterior-medial wall of the tragus are sometimes referred to as ***mini-canal*** hearing aids. Mini-canal hearing aids can be thought of as low-profile ITCs.

The distinction between ITEs, low-profile ITEs, ITCs, CICs, and deeply seated (long-wear) CICs can best be seen in an axial section through the ear, as shown in Figure 5.7. The faceplate of an ITE (whether low- or high-profile) is approximately parallel to the plane containing the lateral surfaces of the tragus and helix. The faceplate of an ITC, by contrast, is approximately at right angles to the posterior-medial surface of the tragus. The faceplate of CIC hearing aids may be at the ear canal entrance or medial to the entrance. Any hearing aid that extends to within a few mm of the eardrum is referred to as ***peri-tympanic***, or as ***deeply-seated***, but it is rare for any hearing aid other than a CIC to extend this far. One such device on the market is inserted by a clinician and is then worn continuously for several months until the battery is depleted, when the whole hearing aid is disposed of and a new one inserted by the clinician. As the hearing aid itself is out of reach of the patient's fingers, a remote control of some type is needed to turn the hearing aid on or off or change the volume control.

Many of the comments made about BTE molds are also true of earshells. In particular, earshells can be occluding or partly occluding and, in general, the hearing aid becomes less securely anchored in the ear as more of the concha material is removed. Despite this generality, ITC and CIC hearing aids, with little or no material in the concha, can usually be retained in the ear provided that an appropriate impression technique is used (see Section 5.8).

When the vent path in an ITE or ITC hearing aid is opened out by removing some of the shell at the medial and lateral ends of the vent, leaving only a short, wide vent path, as shown in Figure 5.29b, the style is called an IROS vent.[a]

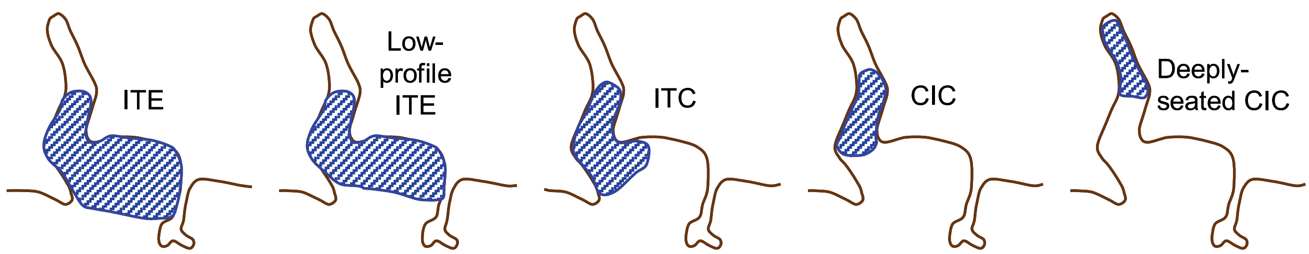

Figure 5.7 Axial view of typical placements for ITE, low-profile ITE, ITC, CIC, and low-profile CIC hearing aids.

[a] The term IROS stands for ***Ipsilateral Routing of Signals***, and was named to contrast with ***Contralateral Routing of Signals*** (see Section 17.1) which was the context in which such open earmold styles were first used.

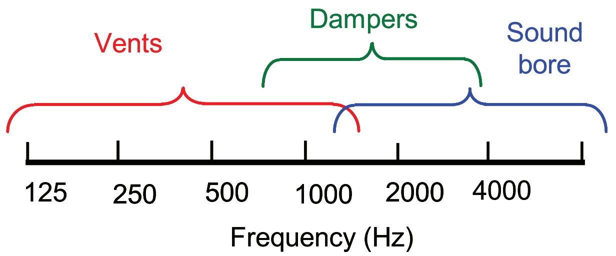

Figure 5.8 Frequency regions affected by each of the components of the hearing aid coupling system.

5.2 Overview of Earmold, Earshell and Canal Fitting Acoustics

The ear fitting affects three broad acoustic characteristics of the hearing aid: the shape of the gain-frequency response of the aid when it is mounted in the ear, the self-perceived quality of the patient's voice, and the likelihood of feedback oscillation.

There are also three acoustic aspects of the coupling system: the ***sound bore,*** the ***damping***, and the ***venting***. These primarily affect the frequency response in different frequency regions, as shown in Figure 5.8. Sound bore dimensions affect only the mid and high-frequency response (above 1 kHz for BTE aids and above 5 kHz for ITE/ITC/CIC aids). Damping mainly affects the response shape in the mid-frequency region (from 800 Hz to 2500 Hz for BTE aids, and from 1500 Hz to 3500 Hz for ITE/ITC/CIC aids) although it has some effects outside this range. Venting mainly affects the low-frequency response, from 0 Hz up to approximately 1 kHz, although if the vent is large enough, such as with an open-canal fitting, it affects the entire frequency range because it leaves the open-ear resonance largely intact.

5.3 Venting

Although this section on venting may seem to be excessively comprehensive, it is the author's experience that the effects of vents and leakage paths lie behind much of the seemingly inexplicable behavior of hearing aids that is encountered when their response is being measured in the ear. A good understanding of vents, including the venting effects of open-canal hearing aids, is essential to hearing aid fitting.

The vent size is selected with the aim of achieving the target gain, but without the ear canal being excessively occluded, and without the hearing aid oscillating. It is often not possible to completely achieve all three aims. These three issues are covered in Sections 5.3.1 to 5.3.3 respectively. Vent size also has implications for the effectiveness of directional microphones, adaptive noise suppression and compression, as discussed in Section 5.3.4.

Vents enable an exchange between air in the ear canal and air outside the air. This air exchange helps avoid excessive moisture build-up. The venting action can enable people with perforated eardrums to wear hearing aids, provided the perforation is not too large.[38]

Our understanding of the effects of vents will be greatly increased if we first understand the concept of ***acoustic mass***. A vent is a column of air surrounded by the walls of a tube. Air, like any other substance, has mass, and therefore has inertia. For a vent to transmit sound, this inertia has to be overcome (or else the air does not move). Overcoming inertia is much easier at low frequencies than at high frequencies and is much easier for small masses than for large masses. Pick up a small weight, like a pen and shake it sideways in front of you at a rate of once per second (i.e. 1 Hz). Now increase the rate to 3 or 4 Hz. You will notice the increase in force that you have to provide. Now pick up a heavier weight, like a 1-kg (2-pound) bag of sugar or flour and repeat the exercise. The higher frequency will require considerable force, and if you provide only a very small force at the 3 Hz rate, then only a very small motion will result.

The analogy is that the column of air in a vent will not move much and so not transmit much sound if the stimulating frequency is high and if the vent has a large acoustic mass. As shown in the panel, vents have a high acoustic mass if they are long and narrow.

Real vents are not always tubes with the same diameter at all points. The concept of acoustic mass helps us understand how the performance of a vent with a varying diameter differs from that of a constant diameter tube. For a vent like the one shown in Figure 5.9

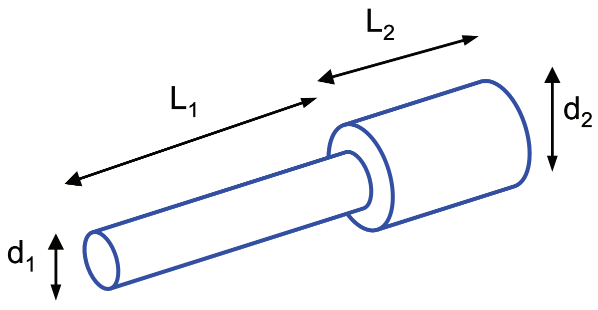

Figure 5.9 A vent made up of two tubes of different lengths and diameters.

Underlying theory: Calculating the acoustic mass of vents from their diameter and length

Although it is never necessary in clinical practice to calculate acoustic mass, the calculation formula is simple and helps our understanding of how changing the vent dimensions will vary the effects of a vent. Acoustic mass is not the same thing as the physical mass of the air in the vent. The acoustic mass of a column of air (i.e. a tube) of length* *L* (in meters) and cross sectional area *A* (in square meters) is equal to:[113]

$$M_a = 1.18\ (L/A) \qquad ..... 5.1.$$

The units are kg/m^4, but by analogy with electrical inertia, the units can be referred to as Henrys. The quantity 1.18 is the density of air in kg/m^3. Because vents are usually circular in shape we can make the calculation more convenient. If the internal diameter of the vent is *d* (in mm) and the length is *l* (in mm), the acoustic mass can be calculated as:

$$M_a = 1500l/d^2 \qquad ..... 5.2.$$

The acoustic mass of a vent increases as the vent gets longer or narrower. *Thus long vents transmit less sound than short vents, and narrow vents transmit less sound than wide vents.*

As an example, a vent 20 mm long with diameter 2 mm would have an acoustic mass of 8100 Henrys when the end correction is allowed for. As a second example, a 1 mm diameter vent in a hollow canal earmold with a shell thickness of 0.7 mm would have an acoustic mass of only 2250 Henrys.

* To precisely calculate acoustic mass, it is necessary to add a length correction to each end of the vent that opens out into a larger space (such as free air at the lateral end or the residual ear canal at the medial end). Each end correction is equal to 0.4 times the diameter of the tube. Thus, the tube in the example above actually acts like a tube that is 21.6 mm long and the tube in the second example acts like a tube that is 1.5 mm long. The end correction can be neglected for vents that are much longer than they are wide, but otherwise should be included, and the correction is especially important for short or wide vents.

the total acoustic mass equals the sum of the acoustic masses of each segment. In this particular case, the acoustic mass of the narrow segment will be much greater than the acoustic mass of the wider segment, so the total acoustic mass will be approximately equal to the acoustic mass of the narrow segment. The acoustic mass of vents with more than one diameter has a practical application to vents with adjustable apertures and vents that have been widened at one end.[999]

Because it can be difficult to predict exactly what size a vent should be, the clinician may need to adjust the vent after a preliminary fitting has been made on the aid wearer. One way to do this is to enlarge the vent diameter by drilling or grinding, or decrease the vent diameter by filling it with wax or plastic materials that cure, and then re-drilling it if necessary. Vents can be modified more quickly and easily if they are ordered with an exchangeable ***vent insert plug***. One such system is shown in Figure 5.10. It comprises a vent tube connected to a widened out cylindrical receptacle at the lateral end of the vent. A "tree" of inserts, any one of which seats firmly in the receptacle, completes the system. The inserts all have the same length (2.5 mm) but differ in the diameter of their internal hole. The different inserts thus change the acoustic mass of the vent but only provided:

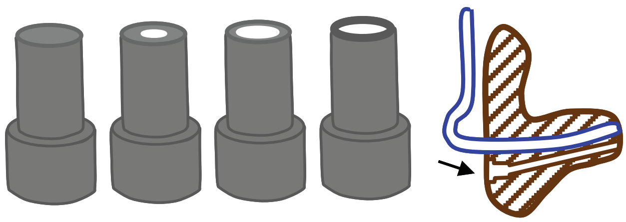

Figure 5.10 The inserts (larger than life-size) from a vent insert system, and the earmold and vent receptacle (approximately life-size) into which they fit. Positive Venting Valve (PVV) and Select-A-Vent (SAV) are two such systems commercially available.

- the rest of the vent (the vent tube) is not so long or so thin that its acoustic mass dominates the total mass,[336] and,
- the leakage around the mold or shell is not so big that the size of the vent is inconsequential.

It will commonly be the case that the inserts with the largest and second largest holes will have almost identical effects (because the total vent mass is dominated by the vent tube), and the inserts with the smallest and second smallest holes will have similar effects to each other (because the natural leakage dominates the venting effect). The insert system is nevertheless worthwhile in that it offers an easy way to obtain two or maybe three effectively different vents. If one does need the maximum flexibility in venting, then:

- for the narrowest inserts to be useful, leakage must be minimized by making the mold a tight fit (which may be uncomfortable and is not very sensible if one ends up using a wide insert); and
- for the widest inserts to be useful, the vent tube must be short and wide, which may not be possible if the ear canal is narrow, and the canal has to contain other large objects, like a horn for BTEs or a large receiver for ITEs.

It is thus useful to be able to predict approximately how much venting is necessary, and this is taken up in the next three sections.

5.3.1 Effects of vents on hearing aid gain and OSPL90

Vents (including leaks and open fittings) affect the low-frequency gain and OSPL90 of hearing aids by allowing low-frequency sounds *out* of the ear canal and by allowing low-frequency sounds *in* to reach the residual ear canal volume without passing through the hearing aid amplifier. These are two separate effects of vents, so let us consider them in turn, and then consider their combined effects.

Effects of vents on the amplified sound path

When amplified air vibrations emerge from the sound bore into the ear canal, they generate sound pressure in the canal. It is this sound pressure that is sensed by the eardrum. The smaller the residual canal volume (the space between the sound bore exit and the eardrum), the greater will be the SPL generated. If there is an escape route, such as a vent (including an open canal), some of the injected vibrations will leave by that route rather than contribute to the sound pressure within the canal. How much sound leaves and how much stays? The proportion leaving depends on the impedance of the escape route relative to the impedance of the residual canal and middle ear. The vent pathway, being an acoustic mass, has an impedance that rises with frequency. Conversely, the residual ear canal volume, being primarily an acoustic compliance, has an impedance that falls as frequency increases.

Table 5.1 Effect of different sized vents, in dB, relative to a tightly sealed earmold or shell, on the gain of the amplified sound path. Note that the vent acoustic masses shown do not allow for the effects of leakage around the mold.[430, 1355, 1816]

Vent size	Vent acoustic mass (Henrys)	Frequency (Hz)								
		250	500	750	1000	1500	2000	3000	4000	6000
Unvented, average fit		-4	-2	-1	-1	1	0	0	0	0
1 mm	26,700	-5	-2	-1	-1	1	0	0	1	1
2 mm	7,000	-11	-3	-1	-1	1	1	1	1	2
Closed dome		-10	-8	-3	-2	-2	-1	1	-2	0
IROS (ITE/ITC)	4,700	-16	-11	-4	-3	2	4	2	-1	0
3.5 mm	2,400	-21	-12	-6	-4	1	2	2	1	1
Janssen (ITE)	2,100	-23	-13	-3	-3	1	6	4	-1	1
Open dome	830	-30	-24	-16	-12	-8	-3	5	0	0

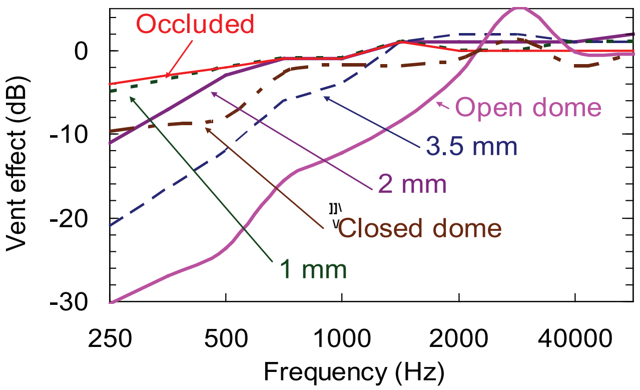

Figure 5.11 Effect of different sized vents on the frequency response of amplified sound, relative to the response with a tightly fitting earmold or earshell.[430, 431, 1355]

For both these reasons, the vent becomes more attractive as an escape route as frequency decreases. Consequently, for sounds injected into the ear canal by the amplifier and receiver, the vent provides a low cut to the frequency response.

The extent of the low-frequency cut depends on the size of the vent (because the vent size determines its acoustic mass). Figure 5.11 shows the degree to which vents of different sizes cut the low-frequency response of the amplified sound path relative to a tightly fitted earmold or earshell. Because these data are so useful in selecting a vent, they are presented in tabular form at the audiometric frequencies in Table 5.1. The results are consistent with those of Kuk, Keenan and Lau (2009). Vents of other sizes, but with the same acoustic mass, would have the same effect on the amplified sound path.[b, 998] Calculation of acoustic mass for the closed dome is not appropriate as transmission is affected by the compliance and mass of the dome, as well as leakage around its edge.

Effects of vents on the vent-transmitted (acoustic) sound path

Vents will transmit low-frequency sound waves no matter which end of the vent they enter. Sound waves reaching the head will thus be transmitted directly *into* the ear canal by a vent. This sound path is totally non-electronic. The range of frequencies over which the vent transmits sounds into the ear canal without attenuation is the same as the range over which it attenuates sound that has been electronically amplified. In particular, sounds are transmitted into the ear canal without significant attenuation up to the vent Helmholtz resonant frequency (Section 2.2.2). Above that frequency, the vent increasingly attenuates sound directly entering the ear canal from outside the head, so the hearing aid, when turned off, begins to act like an earplug.

Section 4.3.3 introduced the ***real-ear occluded gain (REOG)*** as the SPL in the canal with the hearing aid turned off, relative to the SPL in the incoming field. The sound wave causing this SPL reaches the canal primarily via the vent (and leakage) path. The REOG is the equivalent of the REAG, but for the ***vent-transmitted sound path***, rather than for the amplified sound path. Figure 5.12 shows REOG for several earmold styles. Also shown for reference is an average REUG curve. As vent diameter increases, the REOG curve becomes increasingly similar to the REUG curve. The largest vents (i.e., open-canal fittings) leave the REUG almost intact.[1105, 1264, 1937] The small acoustic mass caused by sound moving through holes in the dome or other device used to hold the tip of the tube in place moves the open-canal resonance to a slightly lower frequency.

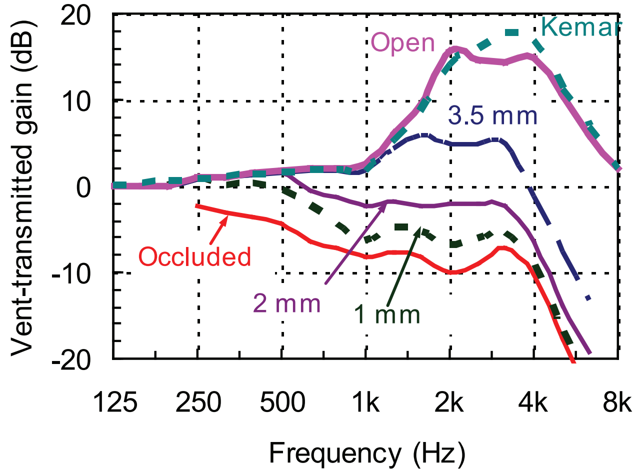

Figure 5.12 REOG of the vent-transmitted sound path for vents of different sizes in an earmold or shell with a mean canal stalk length of 7 mm[430, 431] and for an open fitting.[557, 1264, 1937] The dotted blue line shows a typical REUG curve.

[b] The data in Figure 5.11 relating to vented earmolds were obtained with vents averaging 17 mm in length. Equation 5.2 can be used to generalize the data to vents of different lengths. For example, if the vent length were to be halved, the acoustic mass (and hence the size of the low cut) would remain constant if the vent diameter were to be decreased by $\sqrt{2}$.

It is also useful to consider the equivalent of insertion gain when the hearing aid is turned off: the ***real-ear occluded insertion gain (REOIG)*** shows the SPL in the canal with the hearing aid turned off relative to the SPL in the ear canal with no device in the ear. The REIG is simply the insertion gain of the vent-transmitted sound path:

REOIG = REOG – REUG ……5.3.

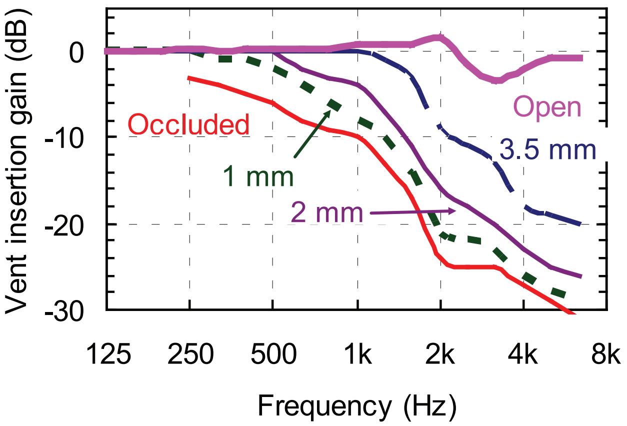

Figure 5.13 Insertion gain of the vent-transmitted sound path (REOIG) for vents of different sizes in an earmold or shell with a mean canal stalk length of 7 mm,[430, 431] and for an open fitting.[557, 1264, 1937]

Achieving high-quality, low-frequency sound

If a patient needs a gain of 0 dB below some frequency, no electronics can compete with the low distortion, flat frequency response that a vent can provide.

For such people, an earmold or shell that attenuates sound below this frequency should be used only if:

- The required high-frequency gain cannot be achieved if a vent is used, or
- It is important for the patient to have the benefits of directivity in the low frequencies.

REOIG is shown in Figure 5.13 for various vent sizes, including open fittings. Similar data have been published by Kuk, Keenan and Lau (2009).

Open fittings have close to 0 dB REOIG gain over the entire frequency range. This has a major advantage for people whose hearing is close to normal at any frequency. For people with reverse sloping losses, for example, any mold other than an open fitting will likely cause the complete hearing aid fitting to act as an earplug for the high frequencies because of the restricted bandwidth of the hearing aid amplifier and receiver.

REOIG is also called ***insertion loss***, and it is sometimes mistakenly thought that hearing aids have to produce gain equal to insertion loss before they provide any net benefit. This is simply incorrect: as equation 5.3 shows, insertion loss is affected both by the amount that the ear fitting blocks incoming sound (i.e. REOG) and by the loss of the open-canal resonance (i.e. REUG), and only the second of these must be compensated for by the hearing aid gain.[c, 1264]

Effects of vents on the combined amplified and vent-transmitted sound paths

The hearing aid user does not hear either the amplified sound path, or the vent-transmitted sound path, in isolation.[911] Rather, as Figure 5.14 shows, sounds arrive at the eardrum via both routes. The sounds arriving via each path combine in the residual ear canal volume.

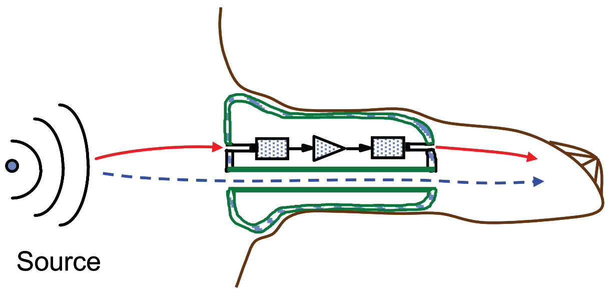

Figure 5.14 Sound travels from a source to the eardrum via the amplified path (solid red line) and the vent or leakage path (dashed blue line). An ITE is shown but the same principle holds for BTE or body aids.

[c] As a thought-experiment, imagine improving the fit of an earmold to the ear, and thus reducing leakage around the earmold. Insertion loss will now be larger than before, yet the modification will actually increase the low-frequency gain of the hearing aid

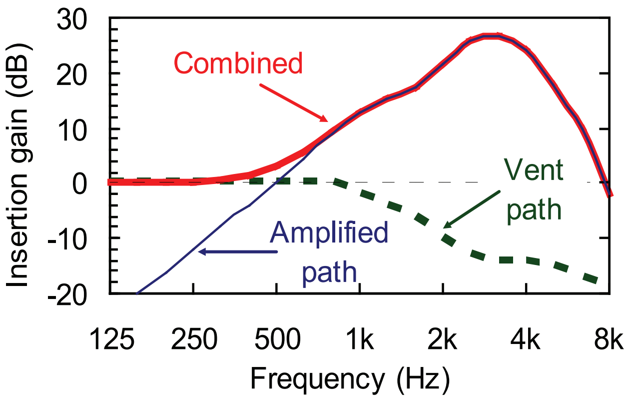

Figure 5.15 Insertion gain of the vent-transmitted path and the amplified path and the way these might combine to form the insertion gain of the complete hearing aid.

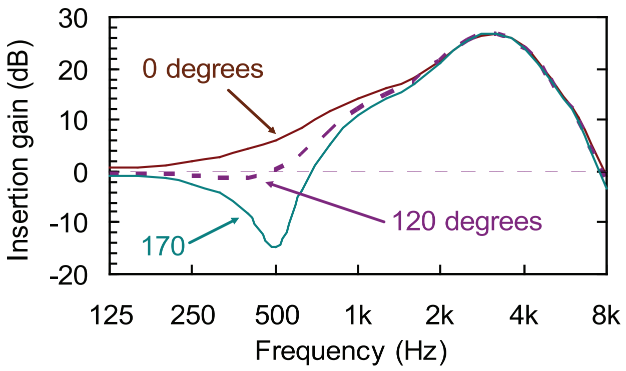

Figure 5.16 Insertion gain of the combined response for phase differences of 0, 120, and 170 degrees between the vent-transmitted and amplified sound paths shown in Figure 5.15. The combined path in Figure 5.15 assumed a phase difference of 90 degrees.

Figure 5.15 shows an example of how the two paths combine.[431] Notice that whenever the insertion gain of one path exceeds the insertion gain of the other path by 10 dB or more, the insertion gain of the combined paths is almost the same as the insertion gain of the path with the higher gain. This is because the amount of sound arriving via the path with the lower gain is inconsequential compared to the sound arriving via the dominant path. When the insertion gain or real-ear aided gain of the hearing aid, or the 1/3-octave spectrum of a speech signal in the ear canal, is measured, the only curve that is apparent is the combined response. It is evident from Figure 5.15, however, that this combined curve arises from two entirely different paths, and it is useful to divide the response into three separate regions: the ***vent-transmitted region***, the ***amplified region***, and between these two, the ***mixed region***.

In the vent-transmitted region, which can extend up to 1500 Hz in open-canal hearing aids, the microphone, amplifier, and receiver play no part in the sound received. In the amplified region, however, the vent can have an effect if it attenuates part of this region by allowing sound out of the ear canal, as detailed earlier.

In the mixed region, the final result depends on how the vent and amplified paths combine, which in turn depends on the phase difference between the two paths. Figure 5.16 shows how the phase difference directly affects how the two paths can combine in the mixed region. A dip or notch occurs in the combined response when the phase difference between the two paths is close to 180 degrees. In practice, the notch is rarely more than 10 dB deep, because deep notches require a phase difference of almost exactly 180 degrees at the frequency at which the two paths have identical gains. Minor dips probably do not have any adverse perceptual consequences, although peaks in the frequency response can.[454] In fact, when a pronounced dip does occur, it provides a convenient marker of the middle of the mixed region.

Because digital hearing aids have significant delay, in the range of 3 to 10 ms,[450] the relative phase of the amplified and vent-transmitted paths changes con-

Practical tip: Matching a real-ear gain target

- *Adding or widening a vent moves the insertion gain towards 0 dB* (causing a low-frequency gain reduction in the amplified region, but causing a low-frequency gain increase if the aid was previously acting like an earplug).
- *Varying an electronic tone control has no effect in the vent-transmitted region*, and therefore may have no effect at all if the tone control only affects the same frequency range as covered by the vent-transmitted region.
- *Varying an electronic tone control can have unpredictable effects in the mixed region*, with the result depending on the phase relationship between the two sound paths.

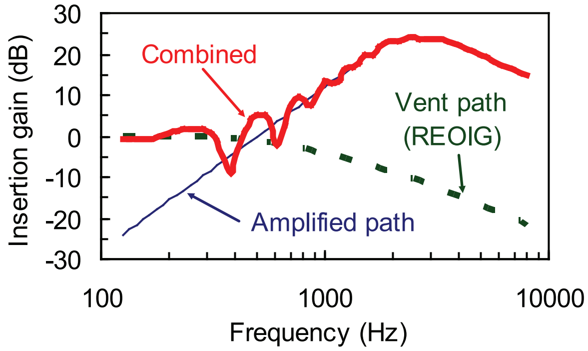

Figure 5.17 Insertion gain of the combined response when sound in the amplified path is delayed by 4 ms with respect to sound in the vent path.

tinuously and rapidly throughout the mixed region. Consequently, the combined response is marked by an alternating series of peaks and troughs, corresponding to in-phase addition and out-of-phase cancellation respectively, over the frequency region for which the two paths have gains within about 10 dB of each other, as shown in Figure 5.17. This modification of sound is called ***comb filtering***; the "teeth" of the comb are separated by the inverse of the relative delay of the two paths, in this case 250 Hz arising from a 4 ms delay.

It must be remembered that because vents affect the sound coming out of a hearing aid, they affect the maximum output in the same way they affect the gain. An OSPL90 control, for example, will have no effect on maximum output in the vent-transmitted region.

There has been some research producing conflicting results, for understandable reasons, on whether people prefer the low-frequency response of hearing aids to be achieved by the use of vents or by the use of electronic tone cuts. Conventional electronic low cuts, when combined with a well fitting earmold will produce negative low-frequency gains (i.e. an attenuation of sound), whereas vents decrease the low-frequency gain only to 0 dB and then provide 0 dB gain for all lower frequencies, as discussed above. Consistent with this, Cox & Alexander (1983) and Kuk (1991) found that vented hearing aids produce superior sound quality. Lundberg et al. (1992), however, used a more complex filter that better simulated the real effect of the vent and consequently found no difference in perceived sound quality. In general, the quality of an amplified sound will also depend on distortion, and this depends on the signal level relative to the level at which the hearing aid saturates. Distortion is never a problem with vent-transmitted sound!

5.3.2 Venting and the occlusion effect

When an ear canal is occluded by a mold or a shell, people with low-frequency hearing thresholds less than about 50 dB HL often complain that their own voice sounds hollow, boomy, like they are speaking in a drum or a tunnel, or that it echoes. These are all descriptions of the ***occlusion effect***.

Figure 5.18 shows the increase in SPL, relative to the SPL in an unoccluded canal, measured in one person's ear canal, as the person talked. The subject wore an occluding earmold with no sound bore. The length of the canal stalk was progressively shortened. For these measurements, a reference microphone in front of the subject was used to remove the effects of any variation in vocal effort. Data for the octave centered on 315 Hz are shown, because this is the frequency range in which the occlusion effect was largest. As the canal is progressively blocked by earmolds of increasing insertion depth, the SPL rapidly increases, then decreases slightly, and then rapidly decreases. A similar variation of SPL with canal stalk length has been reported by Mueller (1994) and by Pirzanski (1998). These variations are caused by changes in three things as the canal stalk is lengthened:

Measuring REOG or REOIG

If a probe tube has already been inserted into the ear to measure real-ear gain, then measurement of REOG or REOIG takes only as long as it takes to turn the hearing aid off and hit the "measure" button on the analyzer. The result is instructive as it is immediately apparent how much of the real-ear gain curve is actually the result of sound entering via the vent and hence unaffected by the hearing aid electronics and transducers. Also, the frequency range over which REOIG is close to 0 dB (or REOG is at or above 0 dB) is a good single-figure indicator of "openness" with strong implications for the amount of low-cut filtering, freedom from occlusion, and amount of feedback-inducing leakage for that fitting.

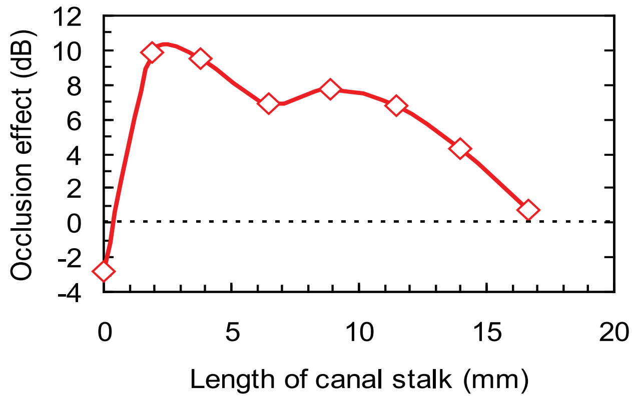

Figure 5.18 Increase in ear canal SPL (relative to no earmold) for the octave centered on 315 Hz when an aid wearer talks. Ear canal length was measured from the ear canal entrance along the center axis of the ear canal. For this person, the transition from cartilaginous to bony canal, as evidenced by the texture of the impression surface, commenced 9 mm into the canal (on the posterior wall, at the second bend) and completed 16 mm into the canal (on the anterior wall).

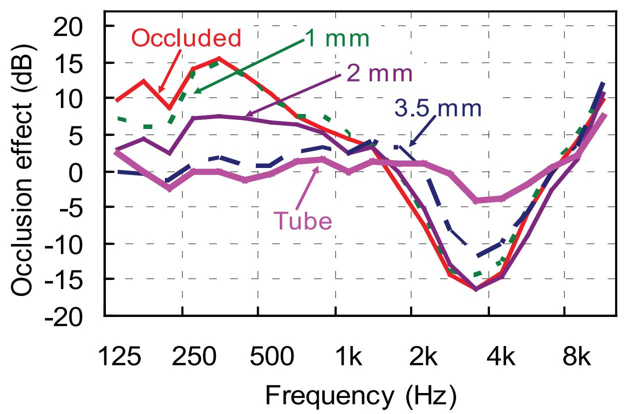

Figure 5.19 The mean increase in SPL (relative to no earmold) in the ear canal for 10 subjects, as they talked while wearing earmolds with vents of different sizes.[1157]

Experience the occlusion effect yourself

1. First say the vowels *ah*, *ee*, and *oo*. Notice that they sound approximately equally loud.
2. Now block both ear canals by squashing the tragus firmly across the ear canal with your fingers.
3. Repeat the same sounds and notice that the *ee* and *oo* sounds have become much louder than the *ah* sound. The *ee* and *oo* sounds are also much boomier than before.

Go on, do it now!

- The seal to the ear increases, thus trapping more of the bone conducted sound within the residual canal.
- The residual volume decreases, which by itself would lead to a higher SPL.
- The area of the vibrating cartilaginous canal wall that causes the occlusion sound decreases. By itself this would lead to a lower SPL. The decrease in SPL from this cause occurs at a rate faster than the decrease in volume, particularly as the end of the canal stalk approaches the end of the cartilaginous canal. Once the canal stalk fills the cartilaginous portion, only the bony portion of the ear canal remains, and this is not an effective generator of occlusion sound. Because the same (temporal) bone[d] surrounds all sides of the bony canal, phase differences between the top, bottom, front, and back walls are presumably minimal.

There are at least two ways to decrease the SPL induced by the occlusion effect. The first is to open up the residual ear canal volume with a vent, with the extreme vent being an open-canal fitting.

Figure 5.19 shows the increase in SPL generated in the ears of 10 subjects as they talked while they were wearing earmolds with vents of different sizes.[e] Subjects' ratings of the acceptability of their own

[d] The ***tympanic plate*** forms the floor and anterior walls of the bony canal; the ***squamous part*** forms the roof and posterior walls; but these bones are both parts of one rigid ***temporal bone***.

[e] The vents and subjects are the same as those for which the data in Tables 5.1 and 5.2 were obtained. Data for an occluded skeleton earmold was very similar to that shown for the occluded carved shell mold. A reference microphone in front of the subject was used to control for any variation in vocal effort.

Why the occlusion effect occurs

As can be inferred from Figure 5.2, the residual ear canal is bounded by the eardrum, the medial end of the mold or shell, and the walls of the canal comprising the cartilaginous section and the bony section. If any one of these boundaries vibrates with respect to the others, the volume of the residual canal changes, and an intense sound pressure is generated within the residual ear canal. What can cause such a vibration? When a person speaks, vibrations in the vocal tract are coupled to all the bones of the skull (including the jaw), and to any tissues connected to these bones.[1861] Because the jaw has a mass only one fifth of the rest of the skull, and is more loosely connected to the rest of the body, it vibrates to a much greater degree than the rest of the skull. Consequently, in the cartilaginous portion of the canal, the inferior and anterior canal walls (which are in close contact with the jaw) will vibrate with respect to the other two walls (which are in close contact with the temporal bone) thus generating a sound within the residual canal volume. When a person is not wearing a hearing aid this does not create a problem, as there is no enclosed cavity within which significant sound pressure can be generated. The air vibrations created by the vibrating canal wall just leak out into the outside air.

Why is the occlusion effect most noticeable for the *ee* and *oo* vowels? Looked at one way, their first formant is around the frequency of maximum occlusion effect (300 Hz) and so is most reinforced. Looked at differently, these vowels are formed as closed vowels, so there is a higher SPL present in the vocal tract than for open vowels like "ah".[927]

voices were significantly correlated to the degree of low-frequency SPL increase (r=0.63) and of course to the size of the vent. Such a correlation accounts for only 40% of the variance in reactions to occlusion, and presumably individual differences in the extent of canal wall vibration, ear canal length and volume, canal stalk length, and psychological tolerance to variation in voice quality all contribute to the end resulting perception. Similar results have been shown for CICs and for BTEs with a variety of earmold styles and vent shapes. Whatever the shape of the vent, the occlusion-induced SPL decreases, and perceived own-voice quality increases, as the acoustic mass of the vent decreases.[902, 997, 998]

The occlusion-induced SPL increases with the acoustic compliance of the ear canal, possibly because those with a large compliance have a larger area of vibrating canal wall, or possibly because the wall vibrates more easily and so transmits more sound into the canal.[251, 997] Not surprisingly, the perception of occlusion is more strongly connected to objectively measured SPL increase in the ear in which occlusion is greatest than to the SPL increase in the other ear.[902]

It is clear that a 1 mm diameter vent is not large enough to decrease the occlusion effect because a vent this small does not increase the venting significantly beyond that which occurs by leakage around the mold. Conversely, with the 3.5 mm diameter vent, the opening is sufficiently large to hold the SPL increase to a few dB. With an increase this small, own-voice quality is rated as normal.[997, 1157] Open-canal fittings, being even more open (see Figures 5.11 to 5.13), produce virtually no occlusion-induced sound.[902, 998, 1105]

A 2 mm vent is only partially effective in solving the occlusion problem. It decreases the size of the SPL increase but does not eliminate it. For each patient, own-voice quality will become more acceptable as the mold or shell is made more open. A 2 mm vent can be regarded as a good starting point for fixing the occlusion problem, but in many cases, the vent will have to be widened to 3 mm before the patient is satisfied with the sound of his or her own voice. All the diameters mentioned in this and the preceding paragraphs relate to vent lengths similar to those used in the study (17 mm). For very short vents, such as those occurring in hollow canal earmolds, much smaller diameter vents are needed to have the same acoustic mass, and hence the same acoustic effects.[998]

The second way to solve the problem of the occlusion effect is not to create one in the first place! As explained earlier, if the mold or shell completely fills the cartilaginous portion of the canal, there will be less occlusion-generated sound compared to molds or shells that terminate within the cartilaginous portion.[197, 927] While this sounds like an easy solution, there are practical difficulties for some hearing aid

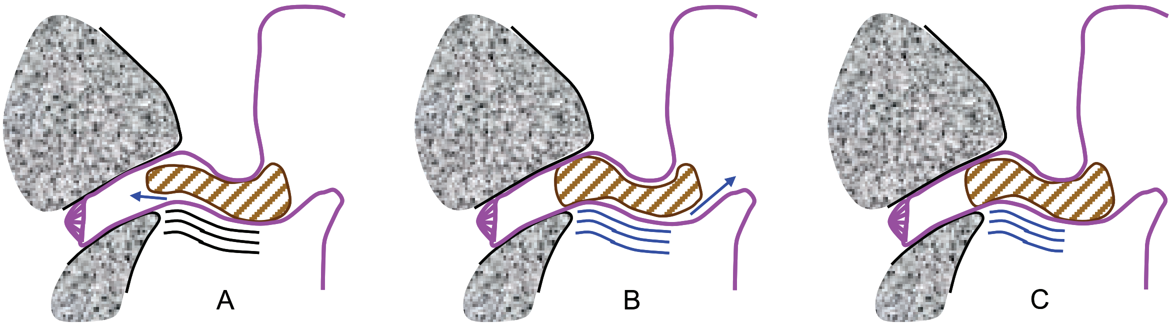

Figure 5.20 Axial view of earmolds or shells that produce a very strong occlusion effect (A), and a very weak occlusion effect (B). The mold or shell shown in (C) will produce a weak occlusion effect and will also have minimal leakage of sound from the hearing aid. In each case, the wavy lines show the vibrating anterior wall and the arrow shows the primary direction in which bone conducted sound will travel once it enters the ear canal. The looseness of fit in each diagram has been exaggerated for clarity.

wearers. First, extra care must be exercised when taking the impression, as detailed in Section 5.8.2. Second, the resulting earshell (or less commonly, earmold) may be difficult for the aid wearer to insert or remove. Third, the earshell may be uncomfortable when worn for long periods if it extends into the bony portion of the canal.[1882] Comfort is improved if the shell is made with a soft tip.[1361] If the soft material is made of compressible foam, the disadvantage is that the foam has to be replaced regularly, and if it is made of a soft plastic, the life of the earmold or hearing aid may be decreased.

Whatever the length of the canal stalk, it is important that the most medial parts of the earmold or shell be in close contact with the canal walls.[927, 1491] Earshells that receive a heavy build-up during manufacture cause less occlusion SPL than those made from open-jaw impressions (which achieve a tight fit in the lateral, flexible parts of the canal).[1421] Figure 5.20 show three earmolds of the same length. For earmold A, the tight seal near the ear canal entrance and the loose fit more medially traps the vibration-induced sound within the ear canal and so ensures that a high level reaches the eardrum. For earmold B, by contrast, the path of least resistance is outwards, so little vibration-induced sound will reach the eardrum. Earmold C should produce intermediate effects. Addition of a vent to any of the three will decrease whatever occlusion effect does occur.[f]

In the future, ***active occlusion cancellation*** (see Section 8.5) in an occluded mold or shell may provide a third alternative to ensure that hearing aid wearers are not bothered by the sound of their own voice.

For people with more than about 60 dB loss at 250 Hz and 500 Hz, the occlusion effect should not be a problem.[g] These people need significant low-frequency amplification, even for the high input levels typically encountered when the aid wearer talks, so it does not matter if there is an increased sound level when they speak. The only complication is that the hearing aid amplified sound will add (constructively or destructively, depending on the phase relationship) to the occlusion-generated sound, and this can affect the shape of the frequency response in the low-frequency region for the person's own voice. When the electronically amplified sound is out of phase with the bone conducted sound, increasing the degree of low-frequency amplification can cause a *decrease* in the SPL in the residual canal. This may be the reason behind the anecdotal reports of solving the occlusion effect by *increasing* low-frequency gain.[1767]

[f] The level of vibration-induced sound reaching the eardrum may be affected in another way by earmold tightness. It may be that inserting a tight earshell or mold into the canal partly couples the canal walls and so decreases the vibration of the tissues on one side of the canal relative to the tissues on the other side. If so, even the unfilled part of the cartilaginous canal would become a less effective generator of sound. This suggestion is merely an untested supposition.

[g] One might expect the occlusion effect to be a much greater problem for patients with low frequency thresholds close to normal than for patients with low-frequency thresholds of 40 or 50 dB HL, as the occlusion-induced sound represents a lower sensation level for the latter group, but this does not appear to be the case.[251]

Alternatively, hearing aid wearers may not have the perception or terminology to adequately differentiate deficient versus excessive low-frequency amplification of their own voice.[991]

The clinician can use real-ear gain analyzers to monitor the magnitude of sound when the patient speaks, whether this arises from bone conduction alone or from the combination of bone conduction and amplification.[1259, 1491] It is unlikely that this measurement is worthwhile doing routinely as, in the end, the amount of occlusion that is acceptable depends on a subjective judgment by the aid wearer rather than on a physical measurement, which in any case is subject to significant measurement error.[902] Measurement of occlusion would, however, be worthwhile when the aid wearer continues to complain about the sound of his or her voice, even after the clinician believes venting or deep seating should have largely eliminated the occlusion sound build-up. For deeply seated molds or shells, measurement will be possible only if the mold or shell contains a hole through which the probe tube can be placed. The skin of the bony canal does not have sufficient flexibility to enable the probe to be placed between the hearing aid and the canal wall without affecting either comfort or leakage.

One "solution" to the occlusion effect that does *not* work is telling the patient that he or she will get used to the altered sound of their own voice.[683, 902]

So far, the "occlusion effect" has been defined as the increase in SPL that occurs in the ear canal when the aid wearer talks. Occluded earmolds also create additional low-frequency sound level during chewing, and even walking; the generation mechanism and solutions are the same. Another consequence of a fully occluded earmold or shell is that the lack of ventilation and increase in moisture may increase the likelihood of external ear disease. Information on the effectiveness of ventilation is scarce. One study showed that even a 2 mm vent did not decrease reports of itchiness and moisture in the ear canal.[1107]

5.3.3 Effects of vents and leaks on feedback oscillations

As explained in Section 4.7.1, feedback oscillation occurs when the attenuation of signal leaking from the ear canal back to the microphone is less than the forward gain given to the signal by the hearing aid. Measurements of the amount of signal leaking back at each frequency can therefore be used to deduce the maximum possible insertion gain before feedback oscillation occurs, at least for hearing aids without feedback canceling algorithms (see Section 8.2.3).[h]

The maximum possible insertion gain without feedback oscillation is shown in Tables 5.2, 5.3, and 5.4 for BTE, ITE and ITC hearing aids, respectively. (Unfortunately, comparable data are not available for CIC hearing aids.) Feedback cancelling algorithms typically allow about an additional 10 to 15 dB of gain to be obtained. Not surprisingly, the maximum achievable gain decreases as the mold or shell becomes more open. In Table 5.2, data for shell and skeleton styles have been combined because the same maximum insertion gain is possible for both styles. Kuk (1994) and Pirzanski et al. (2000) also reached the conclusion that concha bulk does not affect leakage.

The information contained in Tables 5.2 to 5.4 can be used to select the maximum possible vent size without feedback. It is simplest to first compare the value in the 3 kHz column to the target insertion gain at 3 kHz, as this frequency usually provides the strictest constraint. This occurs because target REAG curves often have a maximum at 3 kHz, reflecting the maximum at this frequency in the REUG curve. Consequently, hearing aids must provide a REAG at least equal to the REUG before they start to give the aid wearer a greater signal level than is received without a hearing aid. The maximum gain is needed at low input levels, so it is this low-level target gain that should be compared to the maximum achievable gain.

h As explained in Chapter 4, IG equals REAG minus REUG. In turn, REAG equals the microphone location effect from the free field to the microphone plus the hearing aid gain from the microphone to the ear canal. The maximum value of the gain from microphone to ear canal (without oscillation) is approximately equal to the attenuation of the sound leaking back from the ear canal to the microphone. Consequently, the maximum insertion gain equals the leakage attenuation, plus the microphone location effect, minus the REUG. The maximum achievable gain at each frequency can thus be determined. When maximum gain is determined by turning up the gain of a particular hearing aid until it oscillates (e.g. Gatehouse, 1989; Kuk, 1994), the results are applicable only to the frequency at which oscillation first occurs, which depends on the particular hearing aid used.

Table 5.2 Maximum possible insertion gain (in dB) before feedback oscillation, for BTE hearing aids connected to hard acrylic earmolds with vents of different sizes. The data are average results for ten subjects.[430, 994] Higher insertion gains are possible with tight earmolds, soft earmold materials, and feedback cancellation algorithms.

Vent size	Vent acoustic mass (Henrys)	Frequency (Hz)							
		500	750	1000	1500	2000	3000	4000	6000
Occluded average fit		65	66	64	60	56	41	45	50
1 mm	26,700	65	64	61	58	52	39	45	47
2 mm	7,000	60	60	57	54	49	36	41	48
3.5 mm	2,400	51	53	52	48	43	31	35	41
Tube	800	41	43	42	40	34	23	26	37
Open dome	830	55	49	42	39	31	19	27	30

Table 5.3 Maximum possible insertion gain (in dB) before feedback oscillation, for ITE hearing aids containing vents of different sizes.[1816] The occluded tight shells had a special build-up during shell construction. Higher insertion gains are possible with feedback cancellation algorithms.

Vent size	Vent acoustic mass (Henrys)	Frequency (Hz)								
		250	500	750	1000	1500	2000	3000	4000	6000
Occluded tight fit		62	56	56	56	47	41	23	24	12
Occluded average fit		62	54	52	49	44	33	24	22	13
1.5 mm	14,200	61	57	54	53	48	37	26	25	15
2 mm	8,000	54	50	46	46	42	33	24	23	13
IROS	4,700	44	42	40	38	38	32	19	16	12
Janssen	2,100	42	41	40	39	36	31	17	16	13

Table 5.4 Maximum possible insertion gain (in dB) before feedback oscillation, for ITC hearing aids containing vents of different sizes.[1816] The occluded tight shells had a special build-up during shell construction. Higher insertion gains are possible with feedback cancellation algorithms.

Vent size	Vent acoustic mass (Henrys)	Frequency (Hz)								
		250	500	750	1000	1500	2000	3000	4000	6000
Occluded tight fit		58	52	49	52	45	39	31	33	13
Occluded average fit		52	48	44	45	42	37	23	28	11
1.5 mm	14,700	47	47	44	45	39	34	28	31	12
2 mm	7,800	44	41	38	38	38	32	21	27	17
IROS	4,500	39	34	31	31	29	26	15	23	7

The maximum vent size should be selected with considerable caution. For a given vent size, there is some variation between people in the maximum gain achievable without feedback. This variation is greatest for unvented styles because leakage around the mold or shell is more variable than leakage from a vent of known dimensions. Also, when the hearing aid gain is set a few dB below the point at which the aid continually oscillates, sound quality is adversely affected, as explained in Section 4.7.2. The maximum useable gain may therefore be only about 7 dB greater than the values shown in the tables, leaving a 3 to 8 dB safety margin on average when the feedback cancellation algorithm is operating.

The bottom row in Table 5.2 contains data for an open-dome canal fitting from a different data source.[994] The results should be similar to those for the tube fitting which extended 7 mm past the ear canal entrance. It is unclear why there are large differences in the low frequencies but differences in insertion depth may partly account for the discrepancy. The amount of low-frequency gain that can be achieved before feedback increases with insertion depth.[928]

It might be expected that the shape of the vent, rather than just its acoustic mass, would have an effect on the likelihood of feedback oscillation. It is sometimes recommended that vents be widened at the medial end rather than the lateral end, so that the vent has a reverse horn shape for sounds exiting from the ear canal. While a reverse horn does decrease the amount of high-frequency sound leaking back to the microphone (relative to a vent that has been widened at the lateral end), the differences are confined to frequencies above 6 kHz.[431] Consequently, the variation in shape due to the horn effect has little effect on feedback oscillation, because oscillation usually occurs at frequencies below 6 kHz. Reverse horns nonetheless sometimes have a practical advantage in that there may be more space available at the medial end of an ITC hearing aid than is available on the faceplate. The vent can therefore be made more open (i.e. a lower acoustic mass) than would be possible with a constant diameter vent, with a consequent decrease in occlusion but also in maximum achievable gain before feedback oscillation.[999]

Short vents are slightly less likely than long vents to induce feedback oscillation, when both vents have the same acoustic mass and hence produce the same own-voice occlusion.[715] The reason is that the half-wavelength resonance in long vents occurs within the frequency range in which hearing aids have considerable gain, and this wavelength resonance increases the efficiency with which sounds near the resonant frequency enter, traverse and exit the vent.

A more effective way to decrease high-frequency leakage while maintaining a low acoustic mass to decrease the occlusion effect is to use a cavity vent.[1111] This vent comprises a cavity within the earmold, accessed via two small openings at the medial and lateral ends, which combine to form an acoustic low pass filter. It requires a relatively large earmold to be effective, involves a more complex construction technique, and is therefore little used.

5.3.4 Interaction of vents with digital signal processing algorithms

Vents and directivity

Well-designed directional microphones have a directional pattern across the entire hearing aid bandwidth. This directivity, however, will be apparent to the user only at those frequencies where the amplified sound path dominates over the vent-transmitted sound path. To maximize the benefits of a directional microphone, the amplified sound path should therefore extend to as low a frequency as possible, which in turn means the vent should be as small as possible.

Because directionality depends on cancellation of the signal picked up at one microphone port by signal picked up at the other port (Section 7.1.1), vent-transmitted sounds can materially decrease directivity. Even when the aid-transmitted sound is 10 dB more intense than the vent-transmitted sound, the latter can change the direction at which maximum sensitivity occurs, and can significantly increase the response to rearward sounds.

Directivity rapidly disappears as the level of vent-transmitted sound comes within 5 dB of the aid-transmitted sound. Because wide dynamic range compression causes gain to decrease as input increases, aid-transmitted sound dominates vent-transmitted sound over a wider frequency range at lower input levels than at higher input levels. Consequently, the frequency range over which directivity is available is least at high levels, which unfortunately is precisely where it is most needed.[101] In short, vents in general and open fittings in particular decrease the effectiveness of directional microphones.

Vents and adaptive noise reduction

Just as with directivity, adaptive noise reduction relies on electronic attenuation of sound (Section 8.1) at specific frequencies, and so applies only to the aid-transmitted sound path. Vent-transmitted sound therefore creates a floor below which sound cannot be attenuated, no matter how poor the SNR is. Open fittings therefore decrease the effectiveness of adaptive noise reduction.

Vents and internal noise

The level of internal noise, like other amplified sound, will be decreased in the low frequencies by vents. For people with near normal low-frequency hearing, perception of internal hearing aid noise will therefore be minimized by making the vent as large as possible.

Vents, compressor action, and battery current

Although a large vent may cause a hearing aid to provide no amplification over an extended low-frequency region, the vent does not affect the sound reaching the microphone, and therefore does not affect the sound reaching any compressors within the amplifier. Consequently, low-frequency sounds may still be activating or even dominating the activity of a compressor, even though all the low-frequency sound heard by the patient arrives via the vent. This is an undesirable situation if the hearing aid has only two or three channels of compression, and hence has compression channels so wide that they encompass both the vent-transmitted region and also the aid-transmitted region. A further disadvantage is that the processing of the low-frequency sounds will consume some battery current, which is wasted if these output signals are inaudible.

Both problems are eliminated by the use of a greater number of compression channels. The gain of all channels in the vent-transmitted region should be set to a low value to minimize battery current used to drive the receiver. In hearing aids intended only for open fittings, the manufacturer may switch off the low-frequency channels completely. Irrespective of how you or the manufacturer sets the compression characteristics in the low-frequency channels, do not expect to see any evidence of compression in the output if these channels are in the vent-dominated frequency region. If they are in the mixed region (the extent of which will vary with gain, and hence with input level), the effective compression ratio will be less than the actual compression ratio operating inside the hearing aid.

5.3.5 Parallel versus Y (or diagonal) vents

Other than their propensity to cause feedback oscillations, one of the biggest difficulties with vents is fitting them in! This is especially a problem at the medial end. Figure 5.21 shows an alternative way to fit in a vent when space is tight. This is called a ***Y-vent***, ***diagonal vent***, or ***angle vent***, as opposed to the ***parallel vent*** shown in Figure 5.1, and which has been assumed until now. *The Y-vent should be avoided unless there is absolutely no alternative, as it creates two serious problems.*[336] High-frequency sounds propagating down the sound bore will be partially reflected at the Y junction where the sound bore meets the vent tube. This reflection decreases high-frequency gain and also makes high-frequency feedback oscillation more likely.

If a Y-vent absolutely has to be used, the sound bore and the vent tube should intersect as close to the medial end of the mold as possible. Furthermore, the diameter of the sound bore medial to the Y-junction should be widened as much as possible. This decreases the impedance of this section of the sound bore and minimizes the loss of high-frequency energy back up the vent. Of course, if there is room for extensive widening, there is probably room to avoid the Y-vent altogether!

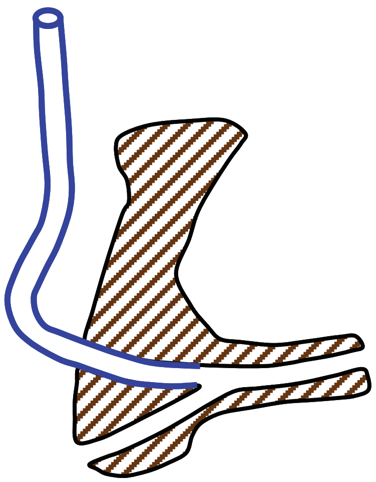

Figure 5.21 Cross section of a Y-vent (or diagonal vent) in a BTE earmold.

5.3.6 Open-canal fittings in summary

Open-canal fittings, when combined with thin tubing and a small BTE, give a very inconspicuous hearing aid, arguably more discreet for patients with hair down to the top of their ears than even a CIC.[i] Most of the acoustic properties of open-canal fittings have been covered in appropriate places within this chapter and in Sections 4.2.1 and 4.5.2. The comfort and inconspicuousness of open-canal fittings have made them very successful, although they none-the-less have disadvantages. This section draws together the acoustic characteristics of these devices. Many of these issues are covered in more detail in an excellent review by Mueller and Ricketts (2006).

Gain-frequency response. Open-canal fittings leave the canal sufficiently open that even with the hearing aid turned off, sound at every frequency reaches the eardrum at almost the same level as when no hearing aid is worn. Put more technically, the patient's REOG almost equals REUG, which is equivalent to having REOIG of close to 0 dB at all frequencies. Because the open canal acts as a very large vent, the hearing aid is ineffective at increasing SPL in the ear canal at low frequencies, and is thus limited to providing high-frequency amplification. Compared to using the same amplifier and receiver in a closed fitting, an open fitting will produce about 3 to 5 dB more gain around the resonant frequency of the REOG curve (typically 2000 to 3000 Hz) because the resonance is still present, but much less gain at lower frequencies and about the same gain at higher frequencies (see Figure 5.26). In some mid-frequency region, the sound entering the canal directly has a similar magnitude to the "amplified" sound, so the two paths combine constructively and destructively at adjacent frequencies to produce alternating peaks and troughs in the response. Insertion depth has relatively little effect on the overall gain-frequency response achieved, because its major effect is on the low-frequency gain of the aid-transmitted sound, which is insignificant compared to the low-frequency vent-transmitted sound.

Own-voice occlusion. The open canal allows the sound created by canal wall vibrations to escape, so the open fitting successfully avoids the occlusion effect that makes the aid wearer's own voice unacceptably boomy for more closed hearing aids.

Feedback oscillation. Unfortunately, high-frequency sounds also escape from the open canal, severely limiting the amount of high-frequency gain that can be achieved without feedback oscillation. Fortunately, these hearing aids have become feasible because of the development of feedback cancelling, which digital hearing aids have made possible. Even with feedback cancellation, it is frequently not possible to obtain the prescribed high-frequency gain. However, the ability to avoid occlusion will often outweigh the disadvantage of not achieving the high-frequency gain that is optimal for speech intelligibility.

Microphone directivity and adaptive noise reduction. The second disadvantage of open fittings is that, because the low-frequency sound is dominated by sound directly entering the ear canal, it is not possible to achieve directivity or adaptive noise reduction for low frequencies, where noise is often most intense. In the future, when microphones become much more directional (Section 7.1.4) the loss of this high directivity will be a major disadvantage. Fortunately, new methods for reducing occlusion with a closed earmold may become available (Section 8.5) and open-fittings will possibly become less dominant than they now are, though still much used.

Prescription targets and real-ear gain measurement. A hearing loss of a certain magnitude and configuration requires a specified increase in SPL at the eardrum to compensate, irrespective of how the sound is delivered to the eardrum. The real-ear prescription target for an open fitting is therefore the same no matter whether an open fitting, a closed fitting, or a vented earmold is used (or whether a thick tube, a thin tube, a BTE or a CIC is used). Of course, in all these cases, different coupler gains will be needed to achieve the same real-ear gain. This is not to say that an identical real-ear gain will actually be achieved for an open fitting versus a hearing aid with a minimal vent. When it is not possible to simultaneously avoid own-voice occlusion and achieve the prescribed high-frequency gain, the compromise struck with an open-canal fitting will be in the direction of having no occlusion but not achieving the prescribed gain. Real-ear gain can be verified by measuring REAG or REIG, just as with any other hearing aid. The only change needed is that it is necessary to turn off the control microphone unless the hearing aid is set to a particularly

[i] Replacing the open dome with a closed dome or canal-only mold gives just as inconspicuous fitting, of course, as the only change is within the ear canal.

Table 5.5 Typical dimensions (length x internal diameter) in mm of sound bore systems in BTE hearing aids. There is also a tube several mm long and about 1 mm in diameter inside the hearing aid case for BTEs other than the RITE style.

	Earhook BTE	Thin-tube BTE	RITE BTE
Earhook	20 x 1.3	-	-
Tubing	40 x 1.9	55 x 0.8 + 6 x 1.0	2 x 1.2

low gain (Section 4.5.2). Any hearing aid, no matter how open or closed, that relies on feedback cancelling to achieve the gain adopted will likely need to be measured with the control microphone turned off.

There has been much ill-advised information written about open fittings, prescription targets and real-ear measurement. There is nothing about preservation of the open ear resonance, and the entry and exit of sound acoustically, that in any way invalidates real-ear prescription targets, or the measurement of REIG, REAG, or REAR speech maps to verify real-ear performance.

In most of the issues discussed above, an open fitting is not fundamentally different from the vented hearing aids that have been used for many decades. It's just that the more open the canal is, the more pronounced each issue becomes. Vented hearing aids also minimize occlusion problems, provide no gain for low frequencies, preclude low-frequency directivity and noise reduction, limit the high-frequency gain that is achievable before feedback oscillation occurs, and undergo constructive and destructive multi-path addition of signals.

5.4 The sound bore: tubing, horns and constrictions

The sound bore provides the path between the receiver and the residual ear canal volume. It has a much greater length in BTE hearing aids than in other types, and hence has a much greater effect on the gain-frequency response for BTEs. There are three types of sound bore systems used in BTE hearing aids, as shown in Table 5.5.

The total length of a BTE sound bore (other than for an RITE) ranges from 60 to 85 mm for adults.[336] For earhook-style BTEs, the final 10 to 20 mm of the tubing is contained within the earmold itself. The sound bore in RITE, ITE, ITC and CIC hearing aids is much shorter, and typically contains a tube from 2 to 10 mm long of diameter 1.0 to 1.5 mm.[j] As explained in Section 2.6.2, the sound bore creates resonances, the frequencies of which are determined primarily by the length of the sound bore, but are also affected by its diameter.

So far, all the diameters mentioned in this chapter have been internal diameters, because it is this that affects the passage of sound *along* a tube. The thickness of the tubing wall, and hence the outer diameter, affects the leakage of sound out *through* the walls of tubing. Such leakage can be a problem in high-gain hearing aids, and tubing with extra thick walls is available. Inner and outer diameters of commonly used tubing, along with their NAEL classifications, are shown in Table 5.6. For new tubing, the #13 super thick wall (double wall) tubing provides 2 dB more attenuation of sound leaking through the walls than is obtained

Table 5.6 Diameters, in mm, of commonly used tubing.[1913]

Tubing type	Inner diameter (mm)	Outer diameter (mm)
#12 Standard	2.16	3.18
#13 Medium	1.93	3.10
#13 Thick wall	1.93	3.31
#13 Super thick	1.93	3.61
#15 Standard	1.50	2.95
#16 Standard	1.35	2.95
Thin-tube	0.80	1.40

[j] At least one RITE hearing aid contains a long and convoluted sound bore within the earmold, to achieve a lower resonant frequency than would otherwise occur, and hence create additional OSPL90 and gain in the mid-frequencies.

Theory: How acoustic horns work

Horns help overcome the ***impedance mismatch*** between the acoustic impedance of a receiver and the much lower acoustic impedance of the ear canal. If the receiver and the ear canal are directly connected together, or are connected via a constant diameter tube, much of the power is reflected back from the medial end of the tube rather than being transferred to the ear canal. By gradually changing the diameter of a connecting tube, and hence its impedance, there is a more gradual transition from the high impedance receiver to the low impedance canal, and hence less power is reflected. This gradual transition is effective only for those high frequencies for which the wavelength is less than or comparable to the dimensions of the tube. Because reflections are less marked, so too are standing waves in the tube. Consequently, the response is less peaky, resulting in improved sound quality.

The effects of horns can be quantified. The approximate boost provided to high frequencies can be calculated from:

$$Boost = 20 \cdot \log_{10}\left(\frac{d_o}{d_i}\right) \text{ dB} \qquad \text{... 5.3.}$$

That is, horns with the biggest outlet diameters will give the biggest high-frequency boost. However, this boost only occurs for frequencies well above the ***horn cut-off frequency***, f_h. Below the cut-off frequency, no boost occurs. For a continuous horn with an exponentially growing diameter, the cut-off frequency can be shown to be:[113]

$$f_h = \frac{c \cdot \log_e\left(d_o / d_i\right)}{2\pi l} \text{ Hz} \qquad \text{… 5.4,}$$

where c is the speed of sound, and $\log_e$ is the natural logarithm (shown as *ln* on most calculators). Thus, the shorter the horn, the higher the cut-off frequency of the horn. As an example, for a horn with an inlet diameter of 2 mm, an outlet diameter of 4 mm, and a length of 25 mm, the horn cut-off frequency is 1520 Hz. The boost *commences* at this frequency and does not reach its full extent until an octave higher than this.

If a horn is made in a stepped manner, as in Figure 5.22, the stepped portion has an additional effect: standing waves will occur within the widened section of tube because reflections occur at each change of diameter. The ***quarter-wavelength resonances*** caused by these reflections can be used to shape and extend the frequency range of the hearing aid.[427, 909, 910]

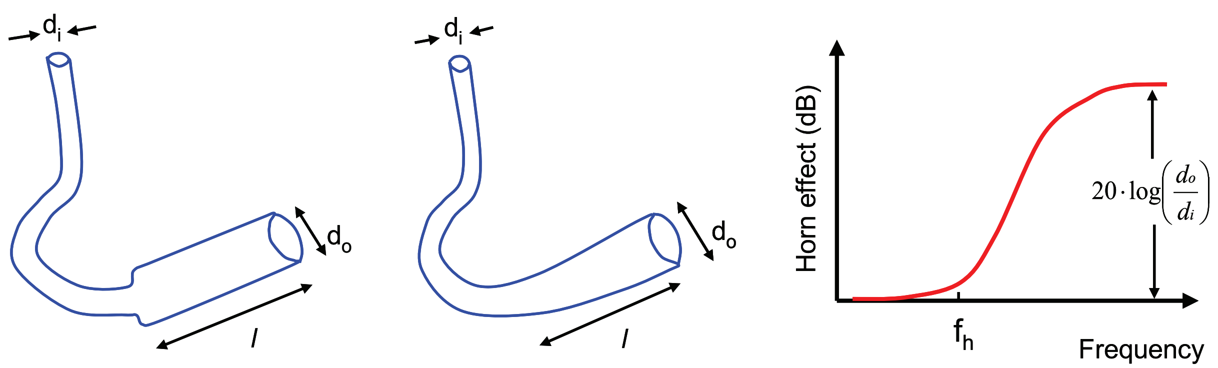

Figure 5.22 Two acoustic horns, one stepped and one continuous, each with inlet diameter di, and outlet diameter do, and the boost (an increase in gain and maximum output) given to the frequency response by the continuous horn.

with #13 standard tubing.[550] This difference may seem small but is consistent with the difference in wall thickness. Because a common source of leakage is the junction between the tubing and the earhook, the thicker wall tubing may provide greater advantages over time if it is better able to retain the integrity of this joint.

In areas with high humidity, moisture-resistant tubing should be used. This tubing is made of a different plastic that decreases the likelihood of moisture droplets forming inside. The tubing is stiffer than conventional tubing.

5.4.1 Acoustic horns and constrictions

Varying the internal diameter of the sound bore along its length will modify the high-frequency response of the hearing aid. If the diameter increases (either smoothly or in steps) it is referred to as an ***acoustic horn***, and if it decreases it is referred to as an ***inverse horn***, ***reverse horn***, or ***constriction***.

Horns

Acoustic horns increase the efficiency with which high-frequency power is transferred from the receiver to the ear canal, and hence increase both the gain and the maximum output in the high-frequency region. This boost is achieved only above a certain frequency, which depends on the ratio of the inlet and outlet diameters of the horn, and on its length (see panel). The shorter the horn, the higher the range of frequencies affected. Horns attached to a BTE earmold can be much longer than horns contained within ITE hearing aids, and are thus able to boost amplification over more of the frequency range where a boost is often needed. Horns in BTE fittings can typically provide significant boost at 3 kHz and above, while those within ITEs cannot provide any significant boost below 6 kHz.

Horns can be built into BTE earmolds in a number of ways. A simple method is to insert tubing only a few millimeters into the earmold. The outlet diameter of the horn will then be determined by the size of the hole drilled into the medial end of the earmold. Although a horn with one or more steps can be made in this way, the method has two major disadvantages:

- The length of the horn will always be less than the sound bore length of the earmold (typically 15 to 22 mm), so the boost may not extend sufficiently far down in frequency.
- The tubing is poorly retained in the earmold. Furthermore, glue has to be applied at the lateral end of the mold, and over time this will cause the tubing to stiffen, and crack, right at the point where the tubing is most stressed in daily life.

One alternative is to use an elbow securely mounted in the lateral end of the mold, to which the tubing is attached. This has the advantage that the tubing can be replaced without having to replace the mold, or without doing any gluing.

Another alternative is to use a molded plastic horn, such as a ***Libby horn***.[1062] It is very common for BTE hearing aids to have insufficient high-frequency gain relative to their mid-frequency gain and relative to the prescribed frequency response. Very commonly then, it is desirable to include the widest horn possible. One limiting factor is simply fitting it in, especially if the mold also has to fit a vent.

Figure 5.23 shows two ways in which a Libby 4 mm horn can fit into a mold. In the method on the left, the horn is fully inserted through the mold, for which a hole of approximately 5 mm diameter is required. In the method on the right, however, the final 15 mm or so of the horn is cut off and then the remainder of the horn is glued into the lateral end of the mold. Because

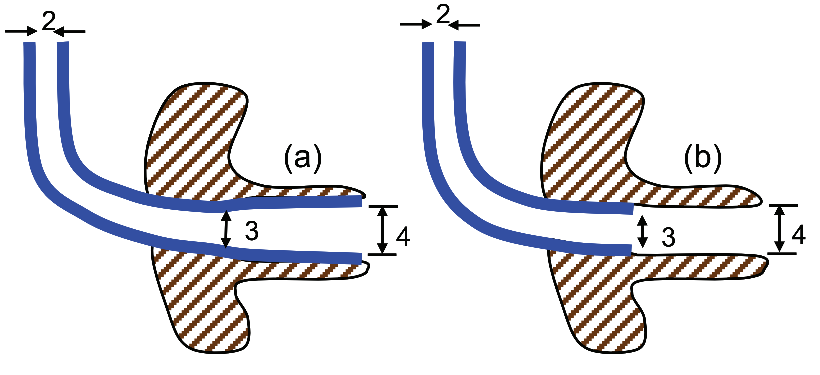

Figure 5.23 A Libby 4 mm horn (a) fully inserted into the earmold, and (b) partially inserted, with the mold forming the final section of the horn. Diameters are in mm.

the mold itself forms the final section of the horn, only a 4 mm hole has to be drilled into the canal portion of the mold. Furthermore, it is the area of the sound bore that matters, not the shape, so an oval-shaped outlet hole can be made if needed.

Using the same method of construction, a 3 mm Libby horn requires only the same space that is needed to fully insert a 2 mm constant diameter tube. The potential disadvantage of this ***half-tubing*** construction method is that if the horn is glued at the lateral extremity of the mold, the life of the tubing is shortened.

Both the horn effect and the quarter wavelength resonance rely on there being a difference between the inlet and outlet diameters of the horn. Instead of increasing the outlet diameter, the inlet diameter can be decreased relative to a #13 tube. This is the basis of the ***Lybarger high-pass tubing*** configuration: a tubing of internal diameter 0.8 mm (the same diameter as a modern thin tube) connects the earhook to the final 15 mm section of 1.93 mm diameter tubing. The narrow inlet tubing decreases the mid-frequency gain, however. The small size of the outlet diameter makes it particularly suitable for creating a horn in an earmold for an infant.

Note that amplification at 4 kHz cannot be increased just by ***belling*** (i.e. gradually widening), or drilling, the last 5 mm of sound bore at the medial tip of an earmold. Such a practice does make a horn, but because it is very short, its major effects will be above 6 kHz. Belling is likely to increase the frequency of the various tubing resonances, which, in a particular fitting, may be advantageous or disadvantageous. Figure 5.24 shows the effects of drilling a 4 mm diameter hole of different lengths at the medial end of an earmold. Notice that the bore has to be at least 10 mm long before a worthwhile effect is achieved at 4 kHz.

An exception to this is when the original, *nominally* constant diameter, tube had been inadvertently constricted at the medial end. The magnitude and extent of high-frequency reduction depends strongly on the degree to which the tube is accidentally constricted when it is glued into a more or less tightly fitting hole in the earmold. A short horn can sometimes provide a significant high-frequency boost, not because of a

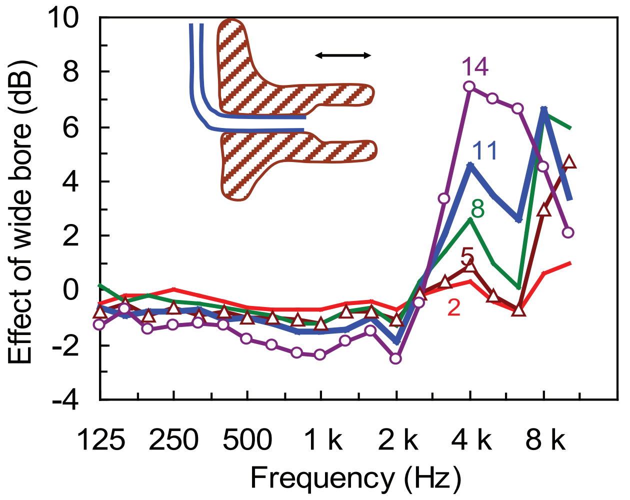

Figure 5.24 The effect of drilling a 4 mm diameter hole at the medial end of an earmold, relative to a constant 2 mm diameter sound bore. The number next to each curve shows the length, in mm, of the widened bore.

horn effect, but because adding the horn removes the constriction.[910] However, horns can also suffer the same fate. If they are squeezed into a hole that is too small for them, the high-frequency boost obtained will be less than that expected.[910]

Table 5.7 shows the effect on gain and OSPL90 provided by various acoustic horns relative to a constant diameter #13 tube. The overall high-frequency boost provided by a horn of specified dimensions is reliable – it is an acoustic inevitability - and so does not depend on the characteristics of an individual patient's ears or the model of hearing aid.[429] The boost in dB at specific spot frequencies will, however, vary from person to person and from aid to aid. This is because the horn shifts resonant frequencies slightly, as well as providing the overall high-frequency boost. The reason for the small decrease in gain at low frequencies is that horns slightly increase the total volume connected to the hearing aid. The data for the Lybarger high-pass tubing were measured with the one-piece molded plastic version (ER-12 HP).

Thin-tube sound bores do come at a price. As Table 5.7 shows, relative to a wider #13 tube, a constant diameter thin-tube sound bore attenuates high frequencies.[k] Thin-tube hearing aids could well use a horn, but so

[k] Precise values for the thin tube relative to 2 mm tubing will undoubtedly vary somewhat with the diameter and length of the tubing used, but the general trend of more gain around 750 Hz (caused by a lower resonant frequency) and less gain above 1500 Hz will be generally true.

Table 5.7 Effect on gain and OSPL90 (in dB) of various sound bore profiles relative to a #13 (2 mm) tube of truly constant diameter, for the same receiver.[428, 429, 1902]

Sound bore	Frequency (Hz)								
	250	500	750	1000	1500	2000	3000	4000	6000
Libby 4 mm	-1	-2	-3	-3	-1	-2	6	10	6
Libby 3 mm	-1	-1	-2	-2	1	1	5	5	2
CFA #2 horn	0	0	-1	-1	0	-1	4	6	4
CFA #3 stepped bore	0	0	-1	-1	0	-1	4	6	2
Lybarger high-pass tube	2	4	0	-11	-13	-12	-10	-1	-1
6C5	0	1	0	0	0	0	-4	-6	-11
6C10	0	2	0	-2	-1	-5	-10	-12	-17
1.5 LP tube	1	3	0	-9	-10	-9	-10	-10	-12
Thin-tube (0.9 mm)	0	1	4	3	-7	-8	-4	-5	-8
RITC	-1	-4	-5	-7	-10	-6	-3	-6	-6

far none have been made available other than the very short widening at the medial end to which the dome connects. Possibly this is because thin-tube delivery systems were originally developed for use with open fittings, for which the achievable high-frequency gain is severely limited by feedback oscillation. Because of their comfort and appearance, however, thin-tube hearing aids are now also used with closed-canal fittings, so a horn version of the tubing would increase the range of hearing losses for which these hearing aids are effective, and in the process improve sound quality by decreasing the peak-to-trough ratio of the gain-frequency response. Acoustically, the effect would be similar to the Lybarger high-pass tube available decades ago. Manufacturers, please take note!

Constrictions

Constrictions have the opposite effects of horns: they decrease the efficiency with which high-frequency power is delivered to the ear canal. They are rarely needed, partly because hearing loss usually is greatest at high frequencies, partly because hearing aid receivers become less effective above their primary resonance of 2 to 3 kHz, and partly because multichannel hearing aids make it easy to decrease gain in specific frequency regions. By combining a constriction with a small cavity achieved by widening the sound bore, an even greater degree of high-frequency cut can be achieved.

Figure 5.25 shows the dimensions of three constricting sound bores. The 6C5 and 6C10 configurations are part of a family of sound bores, and are so named because at ***6*** kHz, they nominally ***C***ut the response by ***5*** and ***10*** dB, respectively.[909] The 6C5 configuration can be made by inserting 14 mm of #16 tubing inside an earmold with a 3 mm horn.[911] To make the 6C10 configuration, 13 mm of #19 tubing is inserted inside 13 mm of #13 tubing which is inserted inside a 3 mm horn.[910] The 1.5 LP (low-pass) configuration is available as a one-piece molded tube (ER-12LP). The acoustic effects of constrictions are shown in Table 5.7.

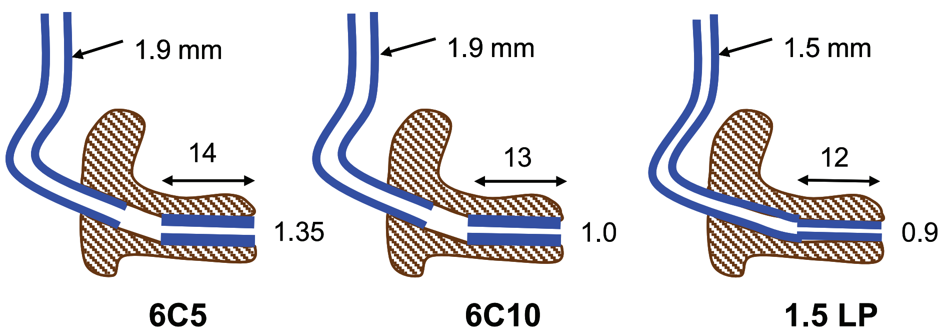

Figure 5.25 The dimensions of the constriction configurations known as 6C5, 6C10, and 1.5 LP.[909]

5.4.2 Tubing insertion depth

For closed and minimally vented hearing aids, increasing the length of the canal stalk decreases the volume of the residual ear canal and hence increases the gain at all frequencies, but a little more for high frequencies than for low frequencies. For open-canal fittings, increasing insertion depth again increases gain across frequency, but this time the effect is greatest for low frequencies, and the mechanism is totally different (see panel). Figure 5.26 shows the SPL change at the eardrum position of a KEMAR manikin as insertion depth increases. As the same receiver and electrical drive level was used for both measurements, the measurements illustrate several other features of open- and closed-dome thin-tube responses:

- Relative to the closed response, the venting effect of the open dome produces a low cut that reaches a slope of exactly 12 dB per octave.
- The high acoustic mass of the thin-tube sound bore moves the first resonance down to a frequency (in this case 500 Hz) much lower than expected from wavelength resonance alone and lower than observed with #13 tubing.
- The open system has a higher output around the canal resonance frequency range of 2 to 4 kHz.

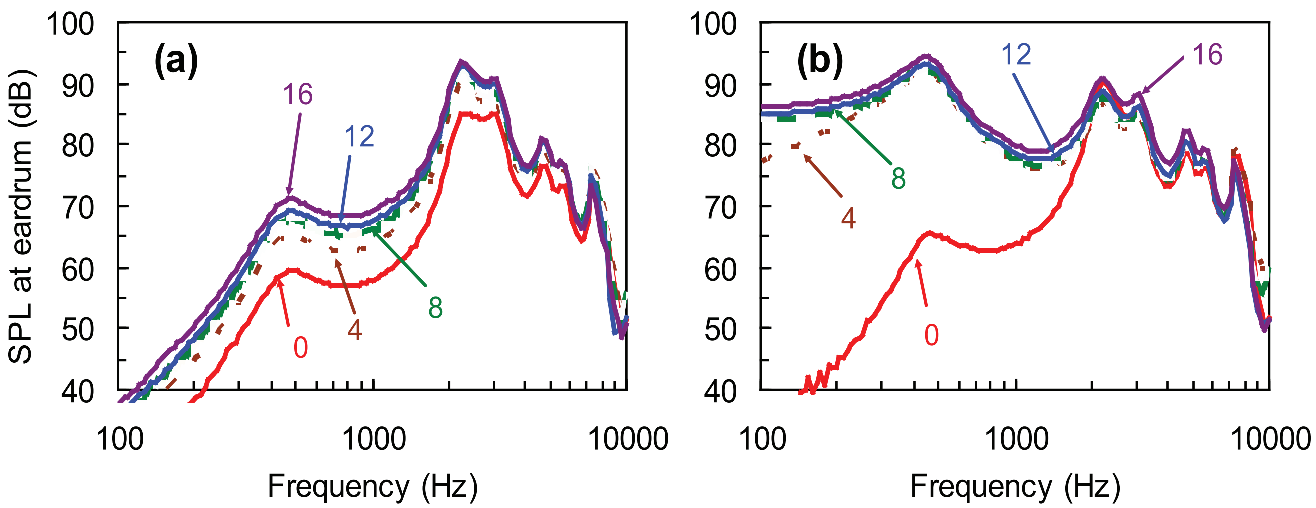

Figure 5.26 Response in an ear simulator of a receiver connected to a thin-tube sound delivery system terminating in (a) an open-dome canal fitting and (b) a closed-dome canal fitting. The numbers in each figure show the insertion depth of the tubing, in mm, past the ear canal entrance. The closed dome did not fully seal the canal for the two shallowest insertions.

Why insertion depth changes ear canal SPL in open fittings

Below about 1500 Hz, the impedance of the vent (i.e. the path from the tip of the tubing to the outside air) is much lower than the residual canal impedance, and so the vent impedance dominates the total impedance and hence determines the SPL in the canal. As insertion depth increases, so too does the length of the vent path, the vent impedance, and hence the canal SPL. For high frequency sounds, deeper insertion still causes the residual canal to have a smaller volume and greater impedance, but this advantage is offset by the tube opening being closer to the eardrum, and hence losing some of the sound pressure increase along the canal that the open-canal resonance provides.

5.5 Dampers

Dampers are used to decrease gain and maximum output at frequencies corresponding to resonances in the sound bore (Sections 2.6.2 and 2.7). Dampers are most effective if they are placed at locations where the resonance causes the fastest flow of air particles. For wavelength resonances, particle velocity is least at the end of the tube that joins to the receiver. In a BTE, the 1, 3, and 5 kHz resonances are therefore damped increasingly effectively as the damper is moved down the sound bore from the receiver towards the lateral tip of the earmold. Conversely, the receiver resonance near 2 kHz is damped slightly more effectively if the damper is placed near the receiver than at the tip of the earhook. Figure 5.27 shows the effect of a 1500-ohm damper placed at the tip (earmold end) and the nub (hearing aid end) of an earhook.

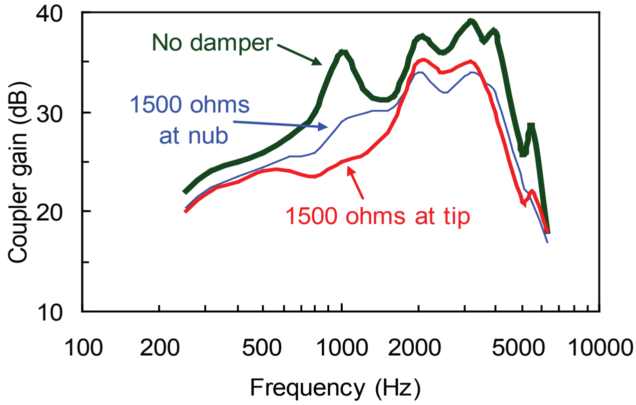

Figure 5.27 Frequency response of a hearing aid with no damper, and with a 1500 ohm damper placed at each end of the earhook.

Table 5.8 shows the effect on gain and OSPL90, in dB, of dampers with different impedances when they are placed at each end of the earhook.

In some cases it might be necessary to use dampers at both ends of the earhook to achieve a sufficiently smooth gain and OSPL90 response that is sufficiently close to the target gain.

Dampers can also be placed in the tubing of RITE/ITE/ITC/CIC hearing aids, but they are likely to need replacing more frequently because of blockage by wax and moisture. Some receivers have dampers built into the receiver outlet. In such cases, the entire receiver may need to be replaced when the damper becomes blocked.

Dampers can also be inserted in microphone tubing to decrease the high-frequency response of a hearing aid (typically to prevent feedback). Dampers are available with diameters of 2.08, 1.78, 1.37 and 1.12 mm to accommodate these various applications.[1847] For the three smaller sizes, the damping screens are not encased within a metal ferrule.

Dampers are currently not used in thin-tube sound bores, possibly out of concern that they would more easily block with moisture, and possibly because thin tubes inherently have greater damping. (No location

Table 5.8 The typical effect of the dampers placed at the nub or tip end of the earhook of a BTE hearing aid. Values shown are averages across several hearing aids; the attenuation varies slightly from hearing aid to hearing aid. The colors are the coding system used by Knowles Inc. Other values available are 1000 ohms (brown), 3300 ohms (orange), and 4700 ohms (yellow).

Damper impedance (ohms)	Damper position	Frequency (Hz)								
		250	500	750	1000	1500	2000	3000	4000	6000
330 (gray)	Nub	0	0	0	-1	-1	-1	-1	0	0
680 (white)	Nub	0	-1	-1	-3	0	-1	-1	0	0
1500 (green)	Nub	-1	-2	-3	-7	-1	-2	-4	-1	-1
2200 (red)	Nub	-1	-1	-2	-6	0	-3	-3	-4	-1
330 (gray)	Tip	0	0	-1	-3	-1	-1	0	-1	0
680 (white)	Tip	0	0	-2	-6	-1	-1	-1	-1	-1
1500 (green)	Tip	-1	-2	-6	-11	-3	-1	-2	-4	-1
2200 (red)	Tip	-3	-4	-9	-16	-4	-1	-1	-5	-1

Practical tip: Shaping a BTE mid-frequency response with dampers

- To maximally damp 1 kHz and minimally damp 2 kHz, place the damper as close to the earmold as possible. The tip of the earhook is the most practical position. Dampers placed at the medial end of the earmold would be more effective, but quickly become clogged with wax and moisture, and should therefore be avoided.
- To maximally damp 2 kHz, and minimally damp 1 kHz, place the damper as close to the receiver as possible. The end of the earhook where it attaches to the hearing aid case is the most practical position, but only some earhooks have been designed to accept fused mesh and sintered steel dampers in this position.

across the cross-section of a thin tube is far from the wall where damping inevitably occurs when the moving air rubs past the stationary wall). As Figure 5.26 shows, thin-tube sound bores nonetheless have pronounced resonances.

5.6 Specific Tubing, Damping and Venting Configurations

Earmold laboratories and hearing aid manufacturers offer a range of earmolds or sound delivery systems that have particular combinations of tubing, venting and damping. These earmolds or sound delivery systems are given particular names or numbers. Instead of thinking of each of these as a unique style with unique acoustic properties, identify the shape and dimensions of the sound bore, the size of the vent path, and the value and location of any dampers. The acoustic performance of the complete earmold will readily be understood on the basis of these three elements.

5.7 Procedure for Selecting Earmold and Earshell Acoustics

The following steps can be used to select suitable earmold acoustics prior to fitting the hearing aid. The procedure assumes that a tentative choice for the rest of the hearing aid has already been made (see Chapter 11). Sometimes, performing these steps will show that the first tentative choice was not a good one, and that a new choice should be made.

Step 1: Find the maximum vent size possible. The target insertion gain is calculated, and the appropriate one of Tables 5.2, 5.3 or 5.4 is used to find the *maximum vent size* that can be used without feedback. For non-linear hearing aids, the target gain for low-level inputs should be used, as gain is greatest for low-level inputs. Remember to add an allowance (perhaps 15 dB) if feedback cancellation is available, and subtract a safety factor (perhaps 7 dB).

Step 2: Estimate the minimum vent size needed. Based on the patient's hearing thresholds at 250 and 500 Hz, estimate the *minimum vent size* needed to overcome the occlusion effect. Good research data on this are not available, but as a guide, low-frequency losses greater than 50 dB do not need a vent, and low-frequency losses less than 30 dB must have at least a 2 mm vent, and preferably will have a larger one. Although a 1 mm vent is usually too small to have any effect on occlusion, the inclusion of such a vent will make it easier for the clinician to drill or grind a wider vent if it proves necessary.

Step 3: Decide on the vent size. Given the constraints determined in steps 1 and 2, this will be an easy choice for many patients, as the maximum and minimum vent sizes will be the same, or the maximum vent size will be slightly larger than the minimum. More difficulty will be found for patients with near normal low-frequency hearing thresholds and 60 to 90 dB loss in the high frequencies. The maximum vent size will turn out to be less than the minimum vent size. These are indeed difficult fittings, and will certainly require feedback-canceling algorithms. It is worthwhile ordering adjustable vents for patients with difficult audiograms, and it is probably worth while ordering adjustable vents for anyone where there is not a big range between the minimum vent size and the maximum vent size. Equivalently, for hearing aids with canal fittings rather than earmolds, there are some choices in the degree of openness of the fitting and, if necessary, the fitting can be changed without changing the hearing aid itself. If the result of this step 3 is selection of an open-canal fitting, then step 4 is not necessary, as cosmetic considerations will sug-

Practical tip: Will it fit?

Before finalizing your choice of sound bore and vent, ensure that the larger diameter of the canal stalk on the impression is at least 2 mm bigger than the sum of the sound bore diameter and vent diameter.

gest that a thin-tube bore be used, for which there is no choice of horns or constrictions.

Step 4: Select a sound bore profile. The details outlined in this step are aimed at earmolds for BTE hearing aids, partly because of the limited effectiveness of horns for RITE/ITE/ITC/CIC hearing aids, and partly because selection of the sound bore for these hearing aids is carried out by the manufacturer. For BTEs, the sound bore has the greatest effect on the response shape in the 2 kHz to 4 kHz octave, so selection can be based primarily on these frequencies. First calculate the slope (in dB/octave) of the coupler gain target (i.e. the 4 kHz target gain minus the 2 kHz target gain). Then calculate the maximum slope of the coupler gain of the chosen hearing aid. If the target slope rises more steeply than the hearing aid slope (as will often be the case) then a horn of some type is required. If, in rare instances, the target slope of the response from 2 kHz to 4 kHz is *less* than the hearing aid gain slope, then a constriction of some type is required. If the target slope and the hearing aid slope are approximately the same, a #13 tube is appropriate.[1]

The data in Table 5.9 will help you decide which horn or constriction to choose. Simply find the sound bore configuration whose slope in the 2 to 4 kHz octave matches the discrepancy you calculated between the target slope and the slope of the hearing aid you are going to fit. Note that the Lybarger high-pass tube achieves its slope, relative to a #13 tube, by suppressing the mid-frequencies rather than boosting the high frequencies, as shown in Table 5.7. Also note that the 1.5 LP tube achieves its cut from 500 Hz to 2 kHz, and so has little effect on the slope above 2 kHz. If you adopt a simple rule (without doing any calculations) of always choosing the biggest horn that will fit in the ear canal, you will be right most of the time!

Table 5.9 Sound bore profile needed to resolve a discrepancy between a target response slope (calculated as 4 kHz gain minus 2 kHz gain) and the response slope available from the hearing aid in a coupler.

Sound bore profile	2 to 4 kHz slope (dB/octave)
Libby 4 mm horn	12
Libby 3 mm horn	4
CFA #2 horn	7
CFA #3 stepped bore	7
Lybarger high-pass tube	11
#13 glued-tube	-1
6C5	-6
6C10	-7
1.5 LP tube	-1

If the selection of a hearing aid and earmold is based on insertion gain instead of coupler gain, Table 5.9 can still be used to select the sound bore. In this case, base the selection on the discrepancy between the target insertion gain slope and the expected insertion gain slope for the chosen hearing aid.

Step 5: Select a damper. Selection of the damper is most efficiently made after the hearing aid has been fitted. The first reason for this is that Table 5.8 does not show what effect the damper has on the frequencies between the audiometric frequencies. Second, the precise effects of dampers at a given frequency will vary depending on whether that frequency coincides with a peak or with a trough. Third, peaks and troughs in the insertion gain response of a hearing aid will depend on the shape of the individual's real-ear unaided response. Fourth, unlike a vent or a sound bore profile, the size of a damper can be changed at a fitting appointment within a few seconds. It therefore seems easiest to make the first measurement of real-ear gain with whatever damper comes as standard with the hearing aid, and then to vary the damper as needed, guided by the data in Table 5.8 and the information in the panel *Shaping a BTE mid-frequency response with damping*.

[1] The CORFIG figures used in this book for BTE hearing aids are based on the 2 cc coupler response being obtained with a HA2 earmold simulator, but the insertion gain being obtained with a constant diameter (#13) sound bore in an occluded, but typically leaky, earmold. Consequently, when there is no discrepancy between the target coupler response and the actual hearing aid response, the appropriate sound bore is a #13 tube.

Practical tip: Changing dampers quickly

Dampers can be changed quickly and without damaging the damper to be removed if a small stock of spare earhooks for the hearing aids most commonly used is maintained. These earhooks can be pre-loaded with dampers of different sizes.

5.8 Ear Impressions

Although this chapter has commenced with the acoustic effects of earmolds and earshells, there can be no earmold or earshell without first taking an impression. This section describes techniques and materials for making an impression

5.8.1 Standard ear impression techniques

Examine the ear canal. Taking an ear impression begins with an otoscopic examination. It is essential to pull the pinna back and to visualize the eardrum and the canal walls so that any abnormalities are known prior to taking the impression.

- Do not proceed if cerumen is present in amounts large enough to disrupt the accuracy of the ear impression. Opinions vary as to how much this is, but it certainly depends on the accuracy required. More cerumen can be tolerated for a low-gain, vented, BTE earmold or full concha ITE hearing aid than for a high-gain occluded aid or a CIC.
- Do not proceed if there is any visible sign of outer or middle ear infection or inflammation, or a distended or perforated eardrum. Medical clearance should first be obtained in each of these cases.
- Do not make a deep impression if the ear canal widens sufficiently (relative to the outer parts of the canal) that removal of the impression will be difficult. It is common for ear canals to widen slightly in the anterior-posterior dimension just medial to the second bend.[36] Ear impression material is able to compress *slightly* during removal (Section 5.8.3), but there is a limit. An extreme case of widening is when a mastoidectomy has been performed to remove diseased portions of the mastoid bone. ENT clearance should be sought before taking an impression of such an ear. An additional block(s) is used to pack the surgically created cavity prior to inserting the block that protects the eardrum.
- Trim any hair in the concha that is long enough to be cut with scissors, as it will be less likely to distort the impression or to get caught in the impression and make removal difficult.

Insert a canal block. A ***canal block*** is a small amount of cotton wool or foam that fills the cross-section of the ear canal to prevent impression material flowing further into the canal than is required. Canal blocks are also called ***oto-blocks***, ***impression pads*** and ***ear-dams***. The resistance to flow provided by the block enables the impression material to completely fill the canal cross-section right down to the desired length, rather than gradually tapering off in width. A piece of strong thread is knotted around the block to aid in removal, although the thread is usually needed only if the block has to be removed without an impression being taken. Alternatively, the cotton or foam is attached to a tube (see panel and Figure 5.28).

Blocks can be custom-made or, more conveniently, can be purchased in a range of sizes, with pre-tied thread. The correct size must be used: blocks that are too small may get pushed down the canal by the impression material, or allow the material to flow around the block. Blocks that are too large will not go in far enough, and may be uncomfortable. Blocks with decreased thickness in the medial-lateral dimension are available for taking deep canal impressions. The block is most conveniently inserted by pushing it with an illuminated plastic stick, referred to as an ***ear-light***, ***oto-light***, or ***light-stick***. Your little finger should be braced against the side of the head whenever using

Caution: unintended consequences

- A canal block of adequate size *must* be used. Impression material has been known to enter the middle ear cavity, either by entering through an existing undetected perforation, or by the pressure of syringing rupturing the eardrum.[958, 1564]
- Widened ear canals close to the eardrum, as a result of either an atypically narrow entrance, or of mastoid surgery, can result in ear impression material being so wedged in that surgical removal is needed.[800, 958]

the ear-light. If pushing the cotton block deeper suddenly becomes easier, beware – the canal may widen suddenly and removal of the impression will be difficult. The depth of insertion of the canal block is very important. Earmold or aid manufacturers can make the finished device with a canal stalk shorter than the impression, but they cannot make it longer. Err on the long side of what you need. The block should be at or past the second bend, unless you are sure you need a very short canal stalk on the finished device. If you wish the earmold or shell to be shorter than the impression you can mark the desired length on the impression.

Mix the impression material. Use only the recommended proportions of the ingredients. Although changing the mixture may let you decrease the viscosity (i.e. make it runnier for easy syringing) or vary the setting time, the change will probably also adversely affect the finished impression. Excessive liquid in a liquid/powder acrylic, for example, will make the impression more readily melt or change shape in heat[1249] and will increase the amount it shrinks.[14] Mixing must be thorough but fast. It should be done with a spatula on a disposable pad or cleanable surface. The reasons for this are to avoid:

- possible adverse health consequences for the clinician arising from repeated absorption of impression chemicals through the skin;
- contaminating the impression material with sulfur-based substances that can leak from hand lotions and from latex gloves;[1913] and
- raising the temperature of the impression material, because setting time decreases as temperature increases.

Similarly, the impression material can be scooped into the syringe with the spatula, or by pushing the inverted syringe at an angle around the mixing pad.

Fill the ear. Partially depress the syringe (or gun) until the material is starting to flow out of the tip. Pull the pinna up and back so that the syringe can be inserted as far as possible. Syringe extension tips can be used for long narrow canals. Depress the syringe until the material has covered the syringe tip to a depth of about 6 mm (0.25 inches). Continue to depress the syringe plunger but simultaneously withdraw the syringe tip at the rate required to keep it buried by the same amount. After the canal is filled, and the concha is nearly filled, lower the plunger end of the syringe and push the syringe tip upwards along the back of the concha (close to the anti-tragus and anti-helix) towards the helix. After the cymba-concha is filled, raise the plunger end of the syringe and push the syringe tip down the front part of the cavum-concha (close to the tragus). Finish syringing when the concha is completely filled and is slightly overflowing on all sides. The earmold/earshell laboratory must be able to recognize all the landmarks on the ear, and this is not possible unless the concha is over-filled. The whole operation should be one complete motion with a constant pressure being applied to the syringe. Finishing in such a way as to give an approximately flat external surface will make it easy to glue the impression to a container for shipping should that be necessary.

Mark the impression. If the hearing aid is an ITE or ITC, and will receive a directional microphone, then scratch a horizontal line across the face of the impression. This line will assist the manufacturer to position the two ports in the same horizontal plane to maximize frontal directivity.

Wait. After 7 to 10 minutes (depending on the impression material and the temperature), test the impression for hardness by momentarily indenting it with a fingernail or other sharp object. If the indent fully disappears, the impression is sufficiently cured. If the canal is particularly twisty, and/or the canal is particularly long, wait a few minutes longer than usual, so that the impression is less likely to tear as it is extracted.

Remove the impression. Have the patient open and close his or her jaw a few times. Pull down, then back, then up on the pinna. These movements help break the bond between the impression and the ear. Extract the helix part of the impression (the ***helix lock***). Grasp the impression and pull it out with whatever twisting motions seem to best suit the individual ear.

Inspect the ear. Make sure that nothing has been left behind!

Inspect the impression. Make sure there are no fold marks, gaps, or bubbles. These blemishes can be tolerated in parts of the impression that will be cut away before the finished product is made, but nowhere else. Good quality in the canal stalk is particularly important. If in doubt, make a second impression. Leave the canal block attached to the impression. Its angle relative to the impression will give the manufacturer some clues about the direction taken by the canal medial to the end of the impression material. *Never*

Has the impression reached the bony canal?

For hearing aids that are intended to extend into the bony canal, assess the following:

- If the second bend is not clearly visible in the impression, try again – the impression is too short.
- Inspect the impression under a magnifying glass – skin in the bony canal is smoother and less porous than in the cartilaginous canal, and this difference in texture is observable in a good quality impression.

lengthen the impression by adding impression material after the impression has been removed from the ear. It cannot be done accurately and is likely to result in discomfort and increased feedback.[1420]

Annotate and send the impression. Pack the finished impression in a shipping container in a manner suitable for the impression material (Section 5.8.3). Any distortion of the impression during shipping will be reproduced in the final product. Alternatively, scan the impression with a laser scanner and electronically transmit the scanned image to the manufacturer. However the impression is sent, annotate the impression, accompanying documentation, or scanned electronic record with notes on any abnormalities observed in the ear canal or requests for the finished otoplastic to differ from the impression. Mark any hollows in the impression that are caused by bumps in the ear (or else the technician may assume the hollows are the result of a poor impression technique and fill them). If the ear canal is particularly mobile when the jaw moves, mark the mobile region unless the impression has been taken with the jaw open (see later).

Clean up. Appropriate infection control measures are important, as they are for real-ear gain measurement, but it is beyond the scope of this book to cover the degree of risk and type of control measures required. While cerumen itself may not be an infectious substance, it is extremely difficult to ascertain visually whether there is any blood or other fluids mixed in with it.[889] All cerumen should therefore be treated as potentially infectious.[890] Extremely high care and fastidious disinfection (or disposal) of equipment should be taken if any infection is evident in the ear or if any procedure inadvertently causes bleeding in the ear canal.

5.8.2 Ear impression techniques for CICs and high-gain hearing aids

CIC hearing aids and high-gain hearing aids may require the earshell or earmold to fit the ear canal more tightly than is necessary for other hearing aids. A tight fit may be needed to avoid feedback oscillation or, in the case of a CIC, to retain the hearing aid in the ear.

CIC hearing aids are retained in the ear by the bends in the ear canal and by the variations in cross-sectional area and shape that occur along the axis of the canal. The widening that usually occurs at the second bend is particularly important.[1420] It is essential, therefore, that impressions for CIC hearing aids be sufficiently deep that they clearly contain the second bend, and preferably extend at least 5 mm past the second bend. With the canal block inserted this deeply, the impression material is able to expand the cartilaginous canal along its entire length, and so provide a more secure fit.[1419]

CIC hearing aids, and to a lesser extent, ITC hearing aids may also require a good fit to ensure that the hearing aid does not work its way out of the ear. Movement of the hearing aid can occur because the ear canal changes shape when the patient moves his or her jaw. In particular, the anterior-posterior dimension of the canal between the first and second bend gradually increases as the jaw opens. As the jaw opens, the condyle of the mandible moves forward, and this pulls the anterior wall of the canal forward.[1360] Averaged across ears, the canal width increases by 10% for a jaw opening of 25 mm measured between the upper and lower incisors.[1359] If the patient's back teeth are missing, the patient has poorly fitting dentures, or the patient has a temporo-mandibular joint disorder with some other cause, the jaw can over-close, and the

CIC impression technique in brief

- Take an impression to 5 mm past the second bend.
- Use a medium-to-high viscosity silicone impression material (e.g. Otoform A/K, Westone, Silicast, Steramold, Otosil, Dahlberg).[1419]
- Use an open-jaw technique.

variation in canal size with jaw motion will be even greater than normal.[652]

The solution to an overly mobile canal is to take an ear impression with the jaw held open, by having the patient bite on a 25 mm spacer until the material has cured.[1418] Although it might be thought that the thicker canal stalk in the mold or earshell would be uncomfortable when the jaw is shut, CICs made from open-jaw impressions are reported to be equally comfortable as, or more comfortable than, those made from closed-jaw impressions.[546] Pirzanski (1997a) comments that discomfort may result from a *loosely* fitting hearing aid because the patient repeatedly pushes it in deeper than it was intended to go, in an attempt to achieve a more secure fit or to prevent feedback oscillation.

Should open-jaw impressions be made for all CIC hearing aids? Possibly, although it seems probable that the need is greatest for those patients whose ear canals change shape the most. The clinician can estimate the degree of movement by observing the canal wall movement with an otoscope or, more intimately, by feeling the degree of motion with an inserted finger.[1153]

Greatly excessive movement of the canal wall may preclude the use of a CIC hearing aid.[652] A potential, though untried, solution may be to make the lateral part of the hearing aid a *loose* fit in the canal, but to make the medial part a tight fit within the bony canal (which does not move). A very soft material for at least the medial part of the hearing aid would be essential for comfort.

An alternative way to make a tightly fitting earmold or shell is to progressively build-up the size of the impression in the ear of the patient. The three-stage impression technique (see panel) uses ear impression material with different viscosities to make a tight but very accurate impression of the ear.[543] Each stage expands the walls of canal, with the greatest expansion occurring in those places where the ear has the greatest flexibility.

Both the open-jaw technique and the three-stage technique increase the width of the earmold in the anterior-posterior dimension just inside the ear canal entrance.[1110, 1419] The relative effectiveness of the two techniques is not known, but the open-jaw technique is considerably faster. A third option is to take a single-stage, closed-jaw impression and have the earmold or hearing aid manufacturer apply special build-up around the aperturic seal region. This is less effective than the three-stage technique because the manufacturer cannot know the flexibility of the patient's ear canal, but even manufacturer's build-up provides enough attenuation for most high gain hearing aids.[1110]

Deep-insertion CICs obviously require very deep impressions. The cotton block must be located at

The three-stage technique in brief

1. Take an impression using a viscous (i.e. non-runny) impression material, and with an embedded piece of tubing extending to the cotton block. The embedded tubing provides pressure relief as the impression is being removed from the ear, which is particularly valuable for steps 2 and 3.
2. Apply medium viscosity impression material over the surface of the canal stalk of the impression. Insert the impression, and apply a gentle pressure to it for a few seconds to assist re-distribution of the uncured material. Remove the impression when dry, ensure that the tube is unobstructed, and then re-insert the impression. Use the air pump of an impedance meter to ensure that a static pressure seal can be achieved. There should be no leakage for 5 seconds after applying a pressure of 200 daPa, while the patient opens and closes his or her jaw. If a leak occurs, repeat this step 2.
3. For this step, the patient's head should be turned to the side and resting on a pillow. Insert a new cotton block, at least as deeply as the first one, fill the ear canal with a low-viscosity material, and quickly re-insert the impression, thus forcing out much of the low-viscosity material. The coating left behind on the impression makes a closer match to the fine structure of the ear.

The earmold manufacturer should be instructed not to add any build-up and not to buff the earmold. For further details, see Fifield, Earnshaw & Smither (1980). For a faster, but less accurate procedure, omit step 3.

In summary: Five ways to produce a tightly fitting, but comfortable, earmold or earshell

- Take the impression with the jaw open.
- Take a two- or three-stage impression.
- Request special build-up during earmold construction.
- Use a viscous (non-runny) silicone impression material.
- Pat down the impression material before it hardens.

When used in isolation, items near the top of the list are probably more effective than items near the bottom of the list.[1110] The use of *all* the techniques simultaneously has not been investigated and is definitely *not* recommended.

or within a few mm of the eardrum, and must itself occupy only a few mm. A cotton block designed exclusively for this purpose, as shown in Figure 5.28, uses cotton only about 2 mm deep, securely attached to the end of a hollow plastic tube. During extraction of the impression, the tube allows the pressure medial to the impression to equalize to the ambient air pressure, minimizing pain during removal. Because the cotton block is so close to, or even touching, the eardrum, the impression material should have a very low viscosity to minimize the pressure it can exert against the eardrum during injection of the impression. The medial surface of the cotton block can be coated with lubricant prior to insertion to ease removal when the impression material has hardened.

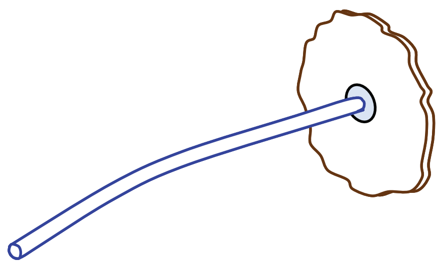

Figure 5.28 A thin cotton block with attached pressure-relief tube, used for taking very deep impressions adjacent to the eardrum.

5.8.3 Ear impression materials

At least three types of materials are used for taking ear impressions. In each case the impression sets when two materials are mixed and undergo a chemical reaction:

- ***Acrylic*** material (e.g. ethyl-methacrylate) is mixed by combining a liquid and a powder. An example is Audalin™.
- ***Condensation-cured silicone*** material (e.g. dimethyl-siloxane) is mixed by combining two pastes. Examples are Otoform-K™, Siliclone™, Blue Silicast™, and Micro-sil™.
- ***Addition-cured silicone*** material (e.g. polyvinyl-siloxane, vinyl-polysiloxane) is mixed by combining two pastes. Examples are Otoform A/K™, Reprosil™, Pink Silicast™, Silasoft™, Mega-Sil™, Dur-a-sil Equal™, Matrics™, Silhouette Plus™.

Ear impression materials must have a certain combination of properties if they are to lead to a tightly fitting, but comfortable, earmold or earshell.

Viscosity. Low viscosity (i.e. runny) materials are easy to syringe and are least likely to expand or distort the ear canal.[m] They were previously recommended for making impressions for CIC hearing aids, so as to make the most faithful reproduction of the ear canal. Unfortunately, this recommendation may have overlooked the changes in ear canal size that are caused by jaw motion (Section 5.8.2). Pirzanski (1997a) recommends that low viscosity material *not* be used when a

[m] Viscosity is measured in units of mPa s. Unfortunately, values are rarely quoted, so we have to describe viscosity by the ambiguous terms of low, medium and high.

tight fit is required, as it does not sufficiently inflate the cartilaginous portion of the canal. High viscosity material is more likely to push hairs against the canal wall rather than encapsulate them, and so will be less painful to remove from a hairy canal.[1417] One technique that has been recommended for deeply seated CIC hearing aids is to use a low viscosity material for the bony canal, followed one minute later by a higher viscosity material for the remainder of the impression.[1685] The intent is that this combination will expand the cartilaginous canal but not the bony canal. The comfort and feedback advantages of the technique have not been quantified.

Dimensional stability. If an ear impression shrinks in the hours or days following its making, the earmold or earshell made from the impression will also be smaller to the same degree unless some compensatory build-up is applied during manufacture of the earmold or shell. Such build-up typically is applied, but the overall precision of the process from impression to finished earmold or shell is obviously greatest if shrinkage is minimized. In the first 48 hours after the impression is made, the linear dimensions of addition-cured silicone material shrink by 0.1% or less.[1342, 1343] By contrast, condensation-cured silicones shrink by 0.5%,[1342] and acrylic materials shrink by 2 to 5%.[14, 1343]

Stress relaxation. When a force is applied to an impression to remove it, it stretches, compresses, and twists as it is pulled through the bends and tight parts of the ear canal. After removal, it is desirable that the impression spring completely back to the size and shape of the ear canal. The extent to which this happens is called its stress relaxation. Silicone materials have excellent stress relaxation properties;[n] acrylic materials do not. Forces can also be applied to an impression during shipping, so all impressions should be suitably protected. Crumpled tissues or other lightweight packing material are adequate to protect silicone impressions. Acrylic materials must be more effectively protected against distorting forces, so the concha portion of the impression should be glued to the liner of the shipping container in such a way that nothing can press against the canal stalk.

Tensile strength. Some force has to be applied to the ear impression to remove it from the ear. It would be a disaster if the impression tore at this stage, so impression materials have to have an adequate tensile strength. Tearing is very rare and is likely to occur only when more medial parts of the canal are considerably wider than more lateral parts or when the impression materials have been mixed in the wrong proportions.

Hardness. Softer impression material is flexible when set and hence easier to extract from the ear than harder impression material. Do not confuse viscosity (how easily the material flows before it is set) with hardness (how easily it deforms after it has set), as these have nothing to do with each other.

Release force. Ear impression materials are designed to conform closely to minute variations in the surface of the ear and ear canal. This makes the impression adhere to the skin so, to make removal possible, a release agent is built into the impression material. This produces the oily feel of a completed impression.

5.9 Earmolds and Earshells

5.9.1 Earmold and earshell construction

Earmolds and earshells (jointly referred to as ***otoplastics***) are made, usually by a specialist earmold or hearing aid manufacturer, from the impression obtained by the clinician. The impression can be considered to be a negative of the ear. There are two methods of earmold/earshell construction: the investment method and the computer-aided manufacture method.

Investment method

The impression is placed into liquid silicon or other material that cures around it to make a positive copy of the ear. This positive copy is called the ***investment***, and the finished mold or shell is made from this investment. A mold is made by filling the investment with liquid mold material and allowing it to harden, often accelerated by ultraviolet light. A hollow shell is made in the same way, except that most of the mold material is poured out before it hardens, leaving just a thin shell covering the inner surface of the investment. For ITE/ITC/CIC hearing aids, the manufacturer trims the shell to the desired size, inserts the electronic and mechanical parts, attaches the faceplate, and sends the complete hearing aid back to the clinician.

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[n] Addition-cured silicones appear to have superior stress-relaxation properties to condensation-cured silicones[1342], but data on this are limited.

Some practices are choosing to maximize their involvement in the supply of hearing aids by making molds and shells themselves. The earshells are constructed of plastic that is cured by exposure to a strong source of ultraviolet (UV) light. The electronic parts are purchased by the practice in bulk and come as a faceplate with pre-wired components. The receiver is on a flexible lead so that it can be suitably positioned before the faceplate is attached.

Computer-aided manufacture (CAM)

The starting point for CAM is a standard impression, which must be made of opaque material (e.g. silicon). Instead of making an investment, the manufacturer or the clinician scans the impression with a laser to produce a numerical representation of the three-dimensional surface, and hence of those parts of the otoplastic that contact the ear, in fine detail. For custom devices, the manufacturer technician, or an automated program, then "inserts" (in virtual reality on a computer screen) a computer representation of the hearing aid components (receiver, and faceplate with attached battery compartment, microphone(s) and switch or control). The fit can be confirmed, and the position of the outer surface of the shell determined. Generally, the shell is made as small as possible, while fitting in all the components without the receiver touching the other components.

With all dimensions of the shell now specified, the corresponding numerical values are transferred to a "plastic printer". The data are used to control lasers that cause a bath of light-sensitive liquid plastic to polymerize (i.e. solidify) in just those positions where the shell or earmold is to be formed. This process is called ***stereo lithography***.[306] An alternative printing process known as ***laser sintering*** builds up the shell or earmold by melting nylon powder, which then solidifies in place, in only those positions where solid material is needed. In both processes, the end product is formed one thin layer (a small fraction of a millimeter) at a time.[328]

The overall CAM process has rather specifically been dubbed CAMISHA (computer aided manufacturing of individual shells for hearing aids). Advantages of the CAMISHA process are:

- lower labor costs;
- improved ability to visualize the fit of the components within the shell before the final shell is produced, enabling the smallest size possible while still keeping the transducers apart;
- ability to "undo" the process while the virtual impression is being cut down in software to the size of the final product, without having to start all over again;
- ability to achieve a uniformly thick shell, with a consequent increase in available internal space and mechanical strength; and
- ability to remake the hearing aid from the same saved data should that be necessary.

Improved accuracy of reproduction, and hence fit of the finished product, has also been claimed, and this seems likely given that the location and thickness of build-up during manufacture can be controlled much more precisely than is possible with the wax-dipping approach that is used with non-CAMISHA manufacturing.[328]

There has been an expectation for nearly a decade that it will be possible to replace impression taking with a direct laser scan of the ear. The problem has been technically tougher than expected, but it does appear that a solution is imminent.

5.9.2 Materials for earmolds and earshells

Material for earmolds and earshells can be classified at a number of levels. Most simply, there are hard and soft materials. Within each of these categories, there are several base plastics. Within each base plastic there are many variations to the mixture that affect the physical properties.

Table 5.10 shows the most common base chemicals used, their range of hardness, and their disadvantages and advantages. The ease with which the materials can be modified refers to the ease experienced by clinicians using the tools commonly found in a clinic. Note that like the other materials, acrylic can be harder or softer. The softer varieties have decreased leakage,[1344] but lose some of the durability advantages of the hard acrylic. The higher leakage of hard plastic earmolds probably arises from their greater shrinking during curing and from the final polishing stage that these materials undergo to give them a pleasing appearance and greater ease of insertion.[1417] Otoplastics made from very soft materials may be more comfortable than those made from hard materials, but there is no research on this issue.

Otoplastics can contain more than one material. Most commonly this comprises a soft material in the canal stalk, or in just the deepest part of the canal stalk, com-

Table 5.10 Advantages, disadvantages and hardness of different materials used for earmolds and earshells (Source: Microsonic, GN Resound and Westone websites and catalogs).

Type of material	Hardness (Shore durometer scale A)	Advantages	Disadvantages
ACRYLIC (Poly-methyl-methacrylate) Hard acrylic, lucite Super-alerite, heat-cured acrylic Soft acrylic – see text	Hard (off scale)	• Little deterioration or shrinkage with time and use • Easy to grind, drill, re-tube, glue and buff • Smooth surface helps insertion and removal • Easy to clean	• Will not compress to insert past narrow areas in the canal • Leaks easily when the ear canal changes shape • Potential for injury when struck, especially if it shatters
ACRYLIC (Hydroxy-ethyl-methacrylate)	Hard (off scale)	Advantages and disadvantages as for poly-methyl-methacrylate, but used for ITE/ITC and CIC shells	
VINYL (Poly-vinyl-chloride) Rx, Polysheer, Polysheer II, Ultraflex, Superflex, Polyplus, Satin Soft Synth-a-flex II, Formaseal	40 - 50	• Comfortable when a tight fit is needed for high-gain hearing aids • Some vinyls (poly-ethyl-methacrylate) soften at body temperatures and harden at room temperatures, helping insertion	• Shrinks, hardens, and discolors with time, necessitating replacement approximately annually • Tubing is difficult to replace: removal is difficult and new tube needs toxic solvent or locking devices to retain it • Softer vinyls need a toxic solvent to polish them – cannot be worn for 24 hours
VINYL (Poly-ethyl-methacrylate) Vinylflex II, Vinylflex, Marveltex, Marvel Soft, Vinyl Flesh, Formula II, Flexible Plastic			
SILICONE (dimethyl-methyl-hydrogen-siloxane) M-2000, W-1, MSL-90, JB-1000, Softech, Soft Silicone, MDX	20-40	• Comfortable when a tight and/or long canal fit is needed, especially for the softer grades of silicone • Little shrinkage with time • Low incidence of allergic reactions	• Impossible to grind and buff; difficult to drill • Tubing cannot be glued – a mechanical tubing lock is required
SILICONE (poly-dimethyl-siloxane) Medi-Sil II, Mediflex, Emplex, Frosted Flex, Bio-por	50-70		
RUBBER (ethylene-propylene copolymer) Microlite, Excelite		• Soft, lightweight and floatable • Used for swim-plugs	
POLY-ETHYLENE	Hard (off scale)	• Extremely unlikely to produce an allergic reaction • Easy to grind, drill, glue and buff	• Will not compress to insert past narrow areas in the canal • Leaks easily when the ear canal changes shape • Noticeable plastic appearance

The hardness of otoplastics

The material property that most affects the comfort and acoustic performance of an otoplastic is its ***hardness***. Hardness is measured by noting how large an indentation occurs when a standard cone- or ball-shaped object is pushed into the material by a standard force. Sharp indentors and large forces are used for hard materials; blunt indentors and small forces for soft materials. The measuring tool is called a ***durometer***, and the resulting indentations are expressed as numbers between 0 and 100 on a ***Shore hardness scale***. Larger numbers represent harder materials. Each combination of indentor and force gives rise to a different scale. There are many such scales. Scale A is most suitable for the softer otoplastic materials and scale D is most suitable for the harder materials. For example, a reading of 90 on the A-scale is approximately equivalent to a reading of 39 on the D-scale.

Soft materials are intrinsically more flexible than hard materials. Greater flexibility makes earmolds easier to insert in a tortuous ear canal and may make them more comfortable when the ear canal changes shape. Soft materials also provide a better seal to the ear, perhaps because they are not polished during manufacture.[1417]

bined with a hard material in the more lateral parts of the otoplastic. The superior retention, feedback, and perhaps comfort properties of a soft material can thus be combined with the superior durability of the harder material surrounding the lateral parts of the hearing aid. A potential problem is that such mixtures may fracture at the plane where the two materials join.

The advantages of soft materials have to be weighed against the greater deterioration of soft materials with time. Note that a key requirement for comfort is to have some flexibility at the interface between the otoplastic and the ear. If the ear is sufficiently soft and flexible, which is common in older people, none may be required in the otoplastic. Consequently, the balance of advantages may swing in favor of hard materials for such people. Unfortunately, there is insufficient research for anyone to be dogmatic about the best earmold material in different situations.

One option (not commonly used) for ITE/ITC/CIC instruments is the use of a solid otoplastic instead of a hollow shell.[373] The material is a very soft, very flexible silicone of hardness 10-35 (Shore A scale) into which the hearing aid components are embedded. The soft material is bonded to a conventional hard acrylic faceplate. This approach may make it possible to achieve the advantages of a soft material in an adequately durable package, but durability is unlikely to be as great as for a hard-shell device.

The skin of some patients will react to an otoplastic. This may be caused by an ***allergic reaction*** to the specific material, or may be the result of ***prolonged occlusion***, no matter what material is used. A common cause of allergic reactions is that a small proportion of the original monomer did not cure into a polymer when the earmold was constructed.[1174] Potential solutions to the problem comprise:

- trying an otoplastic that has been heat-cured rather than cold-cured, as heat-curing decreases the proportion of uncured monomer;
- trying an otoplastic based on a different, low-allergenic chemical, such as silicone or polyethylene;[1676]
- gold-plating the otoplastic, but some people also have an allergic reaction to gold;[1356]
- referring the patient for a contact allergy test, or directly conducting one by taping samples of potential otoplastic material to the skin, so that the presence of a genuine allergic reaction, and the specific allergen, can be detected;[1174, 1676]
- trying a more open mold, if feasible (as a CROS fitting if necessary – see Section 17.1);
- alternating hearing aid use between ears; and
- when all else fails, using a bone conduction or bone-anchored hearing aid.

Note that the first four solutions will not help if the cause is occlusion, and the next two solutions will not help if the cause is an allergy. None of the first five solutions will help if the cause is physical pressure as a result of a poorly fitting otoplastic, but a simple physical modification of the otoplastic's shape may solve the problem. Note that impression material can also give rise to an allergic reaction.[1614]

5.9.3 Instant earmolds and hearing aids

The earmolds discussed so far are all the results of a two-stage process – an impression is taken and an earmold is made from the impression. An earmold is sometimes needed instantly – for a demonstration, as a temporary solution while waiting for a repair, because the patient is in a hurry, or to improve the efficiency of the overall assessment, fitting and follow-up by combining the assessment and fitting appointments. The following are several ways to achieve an instant otoplastic, the first of which is by far the most common:

- Use a hearing aid designed to be fitted with a stock canal fitting, typically comprising a thin, pre-bent tube (chosen from amongst two to four different lengths) to which an open or closed dome (chosen from amongst two to four different diameters) is attached.
- A temporary earmold can be formed from a foam plug with a tube through it, which can be coupled to an elbow and a tube. These provide a better seal (i.e. less leakage) than a conventional custom earmold,[612] and are more comfortable,[1361] but can be more difficult than a custom earmold to insert, and quickly become dirty.
- A custom earmold can be made in minutes by taking an impression using a two-part silicone material (e.g. *Insta-mold*™). The resulting impression, after some trimming, *is* the final earmold. A variation from the usual impression-taking technique is that the concha should be filled only to the degree required for the final earmold, and the lateral surface should be smoothed to the finished contour before the impression material hardens. Smoothing can be done with a finger or thumb that has been pre-wetted with a special lotion. Twisting and pushing a special punch through the impression can cut holes for tubing and a vent.
- A modular, prefabricated ITE, ITC, or CIC hearing aid of an appropriate size can be chosen. There are several innovations aimed at improving the comfort and fit of these hearing aids:
    - an articulated joint, in which the orientation of the canal stalk relative to the concha can be altered;
    - replaceable foam sleeves surrounding the canal stalk;
    - a controlled venting path inside the sleeve;
    - a disposable soft plastic sheath around a standard module; and
    - a hard modular shell combined with a soft, flange-like tip.

If either of the first two options is used, one must be aware that the acoustic performance, and especially the venting effect, may be very different from that which would normally occur with a custom earmold.

5.9.4 Modifying and repairing earmolds and earshells

The most common reasons for modifying or repairing earmolds and earshells are to:

- remove helix locks to ease insertion;
- shorten or taper canal stalks to ease insertion (but see Figure 5.20 and accompanying discussion);
- remove material from the inter-tragal ridge, conchal rim or the canal stalk to eliminate pressure points;
- widen or shorten vents to decrease occlusion;
- constrict vents to decrease feedback;
- thicken canal stalks to decrease feedback; and
- replace loose or hardened tubing.

Earmolds and earshells can be modified in the clinic with suitable tools and materials. For a BTE earmold, a hand-held motor tool is adequate, but to re-obtain the high luster that is usual on ITE/ITC/CIC hearing aids, buffing and polishing wheels are needed. Two such wheels can be mounted on one side of a ¼ Horsepower dental laboratory motor. Buffing compound, obtained from hearing aid manufacturers, is applied to the buffing wheel, but not to the polishing wheel. Drills and small burrs can be mounted on the other side of the motor, and this leaves both hands free to hold the hearing aids. If such a tool is not available, a relatively high luster can be obtained by applying a hypoallergenic clear lacquer. More detailed instructions on modifying shells can be found in an excellent series of practical articles by Curran (1990a, 1990b, 1991, 1992). For CAMISHA shells and molds made by laser sintering, the nylon material is easily melted, so a lower speed grinding tool should be use and less pressure should be applied.[328]

Modifying vents

Earmolds or earshells are made less occluding by enlarging the vent diameter, shortening the vent

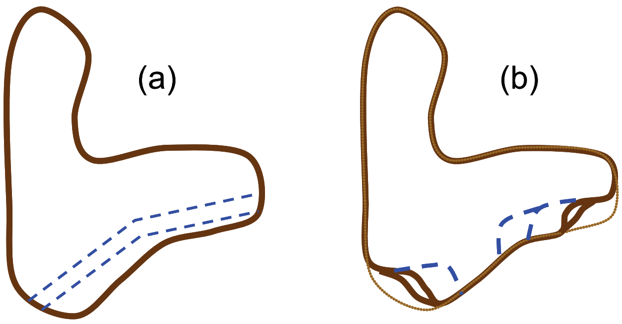


Figure 5.29 An unmodified vent (a) and a shortened vent (b). The dashed lines in (a) indicate the position of the vent. The dashed lines in (b) indicate potential further stages of shortening, and the dotted line indicates the original profile.

length, or a combination of both. Vent diameter is easily enlarged by drilling or grinding. Vent length is shortened by grinding away the mold or shell, from either end of the vent. Figure 5.29 shows how the medial end of the vent can be progressively cut away without affecting the sound bore.

Prior to modifying any custom earshell, view a strong light *through* it to identify the location of the components and to estimate the thickness of the shell walls. Also check that the shell contains a ***poured vent*** (also known as a ***molded vent***) encased in solid plastic, rather than just a vent made of tubing. The modification shown in Figure 5.29 can be performed only if the aid contains a solidly enclosed vent rather than a tube within a hollow shell. Shortening should be performed progressively, each time removing approximately 30% of the remaining vent length. If the vent is shortened so much that the hearing aid oscillates, the remaining vent can be partially filled with earshell build-up material. Tables 5.2 to 5.4 will give some guidance as to whether further shortening is likely to induce feedback oscillations. For hearing aids with vents made only from tubing, grinding trench vents in the canal stalk can increase the effective vent size.

Re-tubing earmolds

Replacing the tubing in a BTE earmold is commonly required and easy to perform. If necessary, ream out the existing hole with a drill bit, motorized reamer, and/or a pipe cleaner dipped in solvent to remove any old glue or debris. To facilitate insertion, the end of the new tubing should be cut at an acute angle. Unless the new tubing is an excessively tight fit in the earmold, the new tubing can be pushed into the mold, as shown in Figure 5.30a. If the point does not emerge at the lateral end, insert some fine-nosed pliers in the medial end and pull the tubing through. If the tubing fits too tightly to be pushed into the earmold, it can be bent back on itself and pulled through with a loop of fine wire, as shown in Figure 5.30b. Ensure that the lateral end of pre-bent tubing is pointing upwards *just in front* of the pinna. If the tubing points too far backwards, it will place excessive pressure on the front of the pinna when the hearing aid is worn. If it points too far forward, it will tend to pull the hearing aid off the ear.

If the earmold is made from acrylic, the tubing should be glued in. To introduce glue (Cyanoacrylate), bend the protruding tube at the medial end in each direction, and introduce glue into the cracks that open up around the perimeter. Apply glue completely around the perimeter of the tube so that there is no crack for cerumen to penetrate. Finally trim the excess tubing, and optionally withdraw the tubing by 1 or 2 mm. Make sure the glue is thoroughly dry before allowing the patient to handle or insert the earmold. If the tubing is a very loose fit, high viscosity (thick) cement can be used to fill the gap.

If the earmold is made from silicone, the tubing must be held in place mechanically. A collar is slid onto the

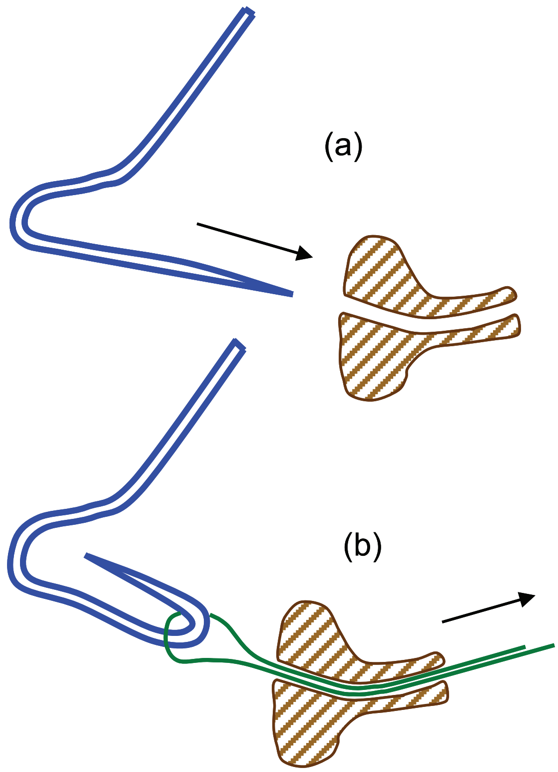


Figure 5.30 Insertion of tubing into an earmold by (a) pushing, or by (b) pulling with a loop of wire.

tubing prior to the tubing being inserted. Some collars are designed to slide in only one direction; these grip the tubing if the tubing is pulled out from the earmold, but this type may constrict the sound bore and decrease the high-frequency response. Other types slide more freely over the tubing and must be glued in place. After fixing the collar onto the tubing at the correct place, the tubing and collar are inserted until the collar is buried a few mm inside the mold material.

Full-length horns (as shown in Figure 5.23a) can be installed in the same manner, except that they are most easily pushed through from the medial end of the earmold. Partly inserted horns (as shown in Figure 5.23b) are preferably installed by pushing the horn in from the lateral end. Glue is applied around the circumference of the medial end of the horn prior to insertion. Provided the horn is inserted very quickly, the glue will act as a lubricant to help insertion.

Building up earshells

Material has to be added to earshells if a grinding operation breaks through the wall of the shell, thus exposing the inner cavity and the electronic components. Material is added by brushing plastic build-up material on to the earshell. A second reason for adding material to either a shell or a mold is to prevent feedback oscillations. If a hearing aid oscillates, despite the vent being plugged, and provided the feedback is not internal, the otoplastic must be made to fit more tightly within the ear, or else a new otoplastic must be made. (See Table 4.12 and 4.13 for diagnosis of feedback cause.) Adding build-up material in the region of the aperturic seal most effectively increases tightness. Build-up material should be applied to the canal stalk in the 6 mm medial to the ear canal entrance, and principally in the anterior-posterior dimension. That is, the narrower diameter of the canal stalk should be enlarged slightly. Different types of build-up material are required for hard versus soft earmolds. High-viscosity material is available for those applications where a thick build-up is required.

5.10 Concluding Comments

By keeping in mind the three key functions of earmolds and earshells listed at the start of this chapter, the clinician can easily understand the characteristics of new designs and styles as they are invented (or re-invented!). The three key questions are:

- How does the diameter of the sound bore change along its length from the receiver to the eardrum, and how long is the sound bore?
- How large (i.e. long and wide) is the vent path, or if there is no intentional opening, how well sealed is the mold or shell to the canal wall?
- How securely will the mold or shell remain in the ear, and how easily can the aid wearer get it in and out of the ear?

If the answers to each of these questions for any new design are compared to the answers for the most similar structure covered in this chapter, the characteristics of any new designs should be able to be predicted with reasonable accuracy.

The importance of selecting earmold and earshell characteristics to achieve a target real-ear gain has diminished with time. Hearing aids almost always have flexible tone controls, which are enabling target gains to be achieved with greater accuracy, assisted by increasingly effective feedback cancellation algorithms, although open fittings still limit the achievable high-frequency gain. For many patients, the freedom from occlusion and physical comfort of an open fitting combine to give higher satisfaction and a lower return rate than is achievable with close fittings,[629] even if some compromise in the high-frequency gain achievable has to be accepted. Active electroacoustic methods for reducing the occlusion effect may soon appear, which will remove the tension between eliminating occlusion and feedback simultaneously, while still achieving the prescribed high-frequency gain.

Despite these advances, selection of the vent size will remain important. The vent must be large enough to minimize negative effects of occlusion and maximize physical comfort on the one hand, but small enough to avoid feedback oscillation and maximize SNR improvement on the other hand. All of these hearing aid qualities are affected as one moves along the continuum from a fully closed to a fully open fitting, and these fitting fundamentals apply whether a custom earmold/earshell, or a modular canal fitting is selected. Equally, choosing a physical style that provides enough retention, combined with easy insertion and removal, will remain important for as long as hearing aids are taken on and off by the user.

CHAPTER 6

COMPRESSION SYSTEMS IN HEARING AIDS

Synopsis

The major role of compression is to decrease the range of sound levels in the environment to better match the dynamic range of a hearing-impaired person. The compressor that achieves this reduction may be most active at low, mid, or high sound levels. More commonly, it will vary its gain across a wide range of sound levels, in which case it is known as a wide dynamic range compressor. Compressors can be designed to react to a change in input levels within a few thousandths of a second, or their response can be made so gradual that they take many tens of seconds to fully react. These different compression speeds are best suited to different types of people.

The degree to which a compressor finally reacts as input level changes is best depicted on an input-output diagram or on an input-gain diagram. The compression threshold, which is the input level above which the compressor causes the gain to vary, is clearly visible on such diagrams. The compression ratio, which describes the variation in output level that corresponds to any variation in input level, is related to the slope of the lines on these diagrams. Simple compression systems can be classified as input-controlled, which means that the compressor is controlled by a signal prior to the hearing aid's volume control, or as output-controlled, which means that the compressor is controlled by a signal subsequent to the volume control. This classification is irrelevant for hearing aids with no volume control and inadequate for hearing aids with multiple, sequential compressors.

Compression systems have been used in hearing aids to achieve the following more specific aims, each of which requires different compression parameters. Output-controlled compression limiting can prevent the hearing aid ever causing loudness discomfort, or the signal being peak clipped. Fast-acting compression with a low compression threshold can be used to increase the audibility of the softer syllables of speech, whereas slow-acting compression will leave the relative intensities unchanged, but will alter the overall level of a speech signal. Compression applied with a medium compression threshold will make hearing aids more comfortable to wear in noisy places, without any of the advantages or disadvantages that occur when lower level sounds are compressed. Multichannel compression can be used to enable a hearing-impaired person to perceive sounds with the same loudness that would be perceived by a normal-hearing person listening to the same sounds. Alternatively, it can be used to maximize intelligibility, while making the overall loudness of sounds normal (rather than the loudness at each frequency). Compression can be used to decrease the disturbing effects of background noise by reducing gain most in those frequency regions where the SNR is poorest. Gain reduction of this type increases listening comfort and with some unusual noises may also increase intelligibility. Finally, compression can be applied by using the combination of compression parameters that patients are believed to prefer, irrespective of whether there is a theoretical rationale guiding the application. Although these rationales are different, they have various aspects in common. Furthermore, many of them can be combined within a single hearing aid.

Despite the complexity, the benefits of compression can be summarized simply, but accurately, as follows. Compression can make low-level speech more intelligible, by increasing gain, and hence audibility. Compression can make high-level sounds more comfortable and less distorted. In mid-level environments, compression offers little advantage relative to a well-fitted linear aid. Once the input level varies from this, of course, the advantages of compression become evident. Its major disadvantages are a greater likelihood of feedback oscillation, and excessive amplification of unwanted lower level background noises.

Compression is like motherhood – everyone agrees it is a good thing, but there is much disagreement about the best way to do it. This chapter describes the different ways that compression can be applied in hearing aids. All of the compression methods have some advantages over linear/peak clipping amplification. All also have some disadvantages.

6.1 Compression's Major Role: Reducing the Signal's Dynamic Range

The major role of compression is to decrease the dynamic range of signals in the environment so that all signals of interest can fit within the restricted dynamic range of a hearing-impaired person (see Section 1.1.2 and Figure 1.2 in particular). This means that intense sounds have to be amplified less than weak sounds. A compressor is an amplifier that automatically reduces its gain as the signal level somewhere within the hearing aid rises (Section 2.3.3). There are, however, many ways in which the gain can be varied to decrease the dynamic range of a signal.

Figure 6.1 shows three ways in which the amount of gain could change as the input level changes. In the left panel, gain starts reducing as soon as the input level rises above *weak*. By the time a moderate input level has been reached, the gain has been sufficiently decreased, and linear amplification can then be used for all higher input levels. The necessary squashing of the dynamic range of the signal has all been accomplished for low signal levels, so we could refer to this as low-level compression. This can be seen in the upper picture as the lower levels coming closer together after amplification than before, while the spacing of the upper levels is not affected by amplification. In the lower figure, the same squashing (i.e. compression) of levels appears as the decreased slope of the input-output (I-O) function for low-level signals, whereas the linear amplification of higher level signals appears as a 45° slope (see also Section 4.1.5).

The opposite approach can be seen in the right panel of Figure 6.1: low-level sounds are amplified linearly, but the inputs from moderate to intense sounds are squashed into a narrower range of outputs. In gen-

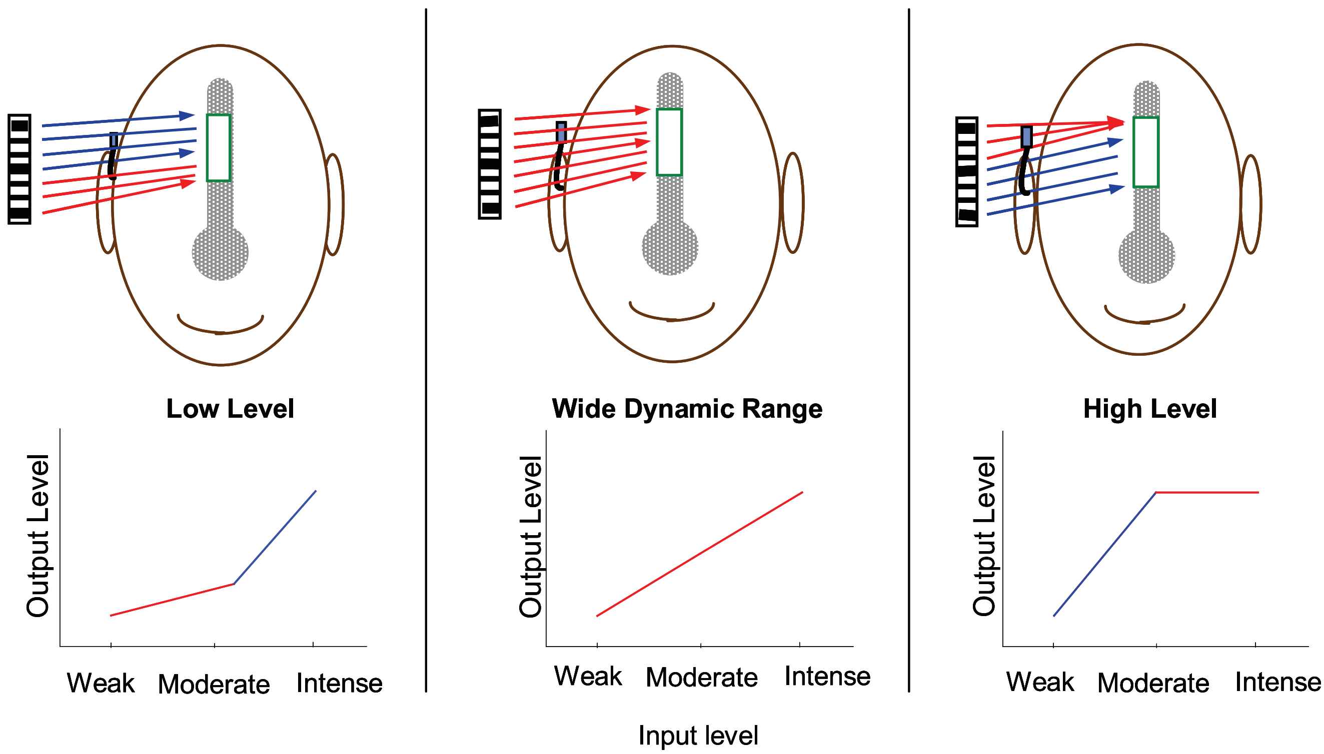


Figure 6.1 Three ways in which the dynamic range of signals can be reduced. In each case, the upper figure shows the spacing of different signal levels before amplification (the left end of the lines) and after amplification (the right end of the lines). The lower figure shows the same relationships, but as an input-output function. In each case compression is occurring in the red region.

eral, this could be referred to as high-level compression. In the case shown here, all high-level inputs are squashed into an extremely small range of outputs. This extreme case is called ***compression limiting***, because the output is not allowed to exceed a set limit.

The center panel represents a third way in which compression could decrease dynamic range. Compression is applied more gradually over a wide range of input levels, and we consequently call this ***wide dynamic range compression (WDRC)***. The overall reduction in dynamic range is the same as for the other two cases. The gradual reduction applies over such a wide range of input levels that there are no input levels for which the corresponding output levels are squashed closely together. Equivalently, the slope of the I-O curve is never close to horizontal.

It is interesting to note that there have been commercially successful hearing aids using each of the three compression strategies, despite the extreme differences among strategies. This is not to say that the differences do not matter, but perhaps the reduction in overall dynamic range of signals that they all achieve is more important than their differences. The relative advantages of different compression systems are discussed later in this chapter, but first we need to define some terms that describe how compressors operate.

6.2 Basic Characteristics of a Compressor

Although a hearing aid employing multiple compressors may operate in a complex manner, the operation of each compressor within the aid can be described with a few simple terms.

6.2.1 Dynamic compression characteristics: attack and release times

A compressor is intrinsically a dynamic device: its job is to change gain depending on changes in the signal level. Figure 6.2 shows an input waveform that rapidly increases, and then decreases in level. When the output level first rises, the detector starts to pass on the increased level to the compressor control circuit. As discussed in Section 2.3.3, the detector first has to convert the waveform to a smooth control signal. This involves ***rectification***[a] and then smoothing. A consequence of this smoothing is that following an increase in signal level, the detector output increases gradually to its new value. During the time taken for this to occur, the compressor is not aware of the full extent of the increased signal level, so it does not turn the gain down sufficiently to compensate for the increase. Consequently, the amplifier initially passes the increase without compression, until the compres-

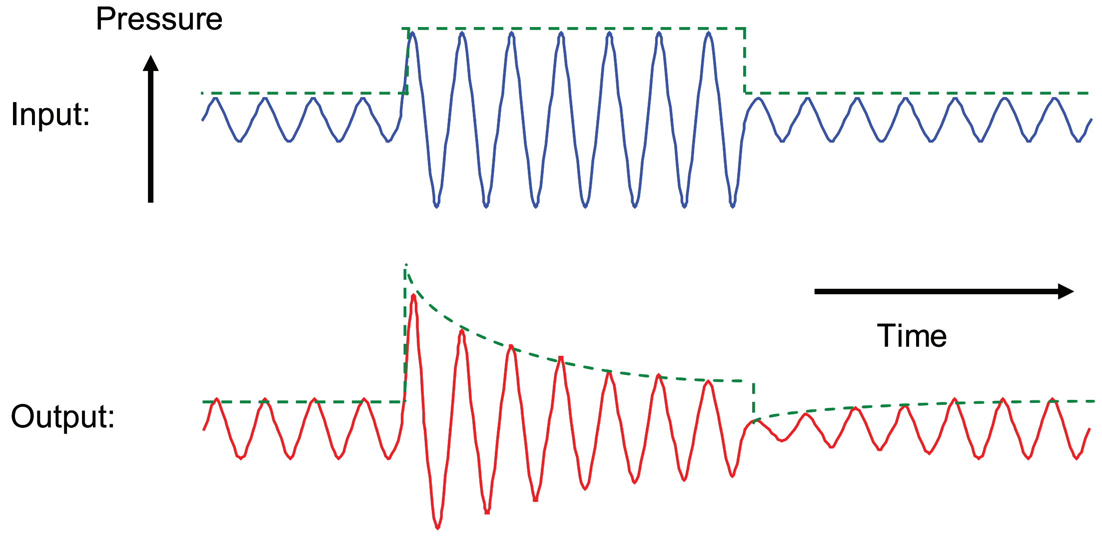


Figure 6.2 Waveforms that are input to a compressor and output from a compressor, showing the attack and release transitions that follow an increase and decrease, respectively, in signal level. The dotted line shows the envelope of the positive half of the signal.

[a] In full wave rectification, all negative values are converted to positive values of the same magnitude. In half-wave rectification, the negative values are simply ignored. In many hearing aids, the envelope for each frequency channel is calculated from the spectral estimates provided by a Fourier analysis. These estimates are summed across the desired frequency range and averaged across the desired time.

sor reacts to the new input level. The time taken for the compressor to react to an increase in signal level is referred to as the ***attack time***.

Because the output gradually approaches its final value, it has to be arbitrarily decided when the final value is reached. Attack time is defined as the time taken for the output to stabilize to within 2 dB (IEC 118-2) or 3 dB (ANSI S3.22) of its final level after the input to the hearing aid increases from 55 to 80 dB SPL (IEC118-2) or from 55 to 90 dB SPL (ANSI S3.22). Eventually, the compressor fully reacts to the increased signal level. That is, its gain has been decreased compared to its previous gain.

A similar event happens when the input signal decreases in level. Again, the detector progressively reacts to the new input level, so for a while the compressor amplifies the low-level signal with the gain that was appropriate to the high-level signal preceding it. The control signal decreases gradually, and consequently, the gain and output signal increase gradually. The ***release time*** is the time taken for the compressor to react to a decrease in input level.[b]

Although the attack and release times could be made to have extremely short values (even zero), the consequences are most undesirable. If the release time is too short compared to the period of the signal being amplified (e.g. 10 ms for a typical male voice), the gain will vary during each period, so the compressor will distort the waveform.[c] If the attack time is made extremely short, and the release time long, then distortion will be minimal. However, extremely brief sounds (like clicks) will cause the gain to decrease (because of the short attack time) and the gain will then stay low for a long time afterwards (because of the long release time). It is not necessary for gain to be decreased very much when a very brief click occurs, because very short sounds convey little loudness. It would certainly be undesirable for the gain to remain low long after the brief sound has gone. Attack times in hearing aids are commonly around 5 ms, but can be much longer, as we will see later. Release times are rarely less than 20 ms, and may be much longer.

The attack and release times have a major effect on how compressors affect the levels of the different syllables of speech. This is best understood in terms of the signal's ***envelope***, which is an imaginary line drawn through the extremities of a waveform. The envelope gives an indication of the level of a signal, without showing the fine structure of the waveform. It is particularly useful for showing the effect of compression because compressors intentionally change the envelope while leaving the fine structure almost unchanged. Figure 6.3 shows the envelope of a signal that alternates between two different intensities. An envelope similar to this shape would occur for someone saying the sound fafaf.[d] (Notice that there appear

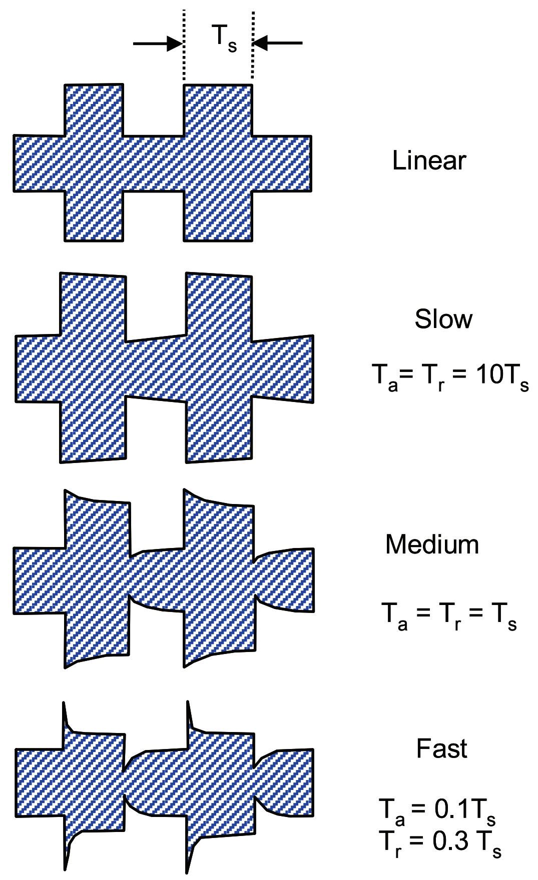


Figure 6.3 Envelopes for the output signal coming from a linear amplifier and compression amplifiers with different attack times (T_a), and release times (T_r) compared to the duration of each syllable (T_s) in the signal.

[b] Release time is defined as the time taken for the output signal to increase to within 2 dB (IEC 118-2) or within 4 dB (ANSI S3.22) of its final value following a decrease in input level from 80 to 55 dB SPL (IEC 118-2) or from 90 to 55 dB SPL (ANSI S3.22).

[c] In the extreme case of zero attack and release times and a high compression ratio, a compressor becomes a peak-clipper.

to be five phonemes in the envelope shown, but we will loosely refer to each as a "syllable".) The first envelope is the output that would occur for linear amplification.

The second envelope shows the output when the attack and release times are both much longer than the duration of a syllable. The compressor starts to turn the gain down or up when each new phoneme or syllable starts, but there is time for only a small gain change to occur before the syllable is finished. The compressor then starts to slowly establish a gain appropriate to the next syllable. Consequently, the gain is almost constant, so the envelope is almost the same as for linear amplification. Is such a compressor therefore doing anything to the signal? As long as the signal level fluctuates around these same values, apparently not. If the input level drops and remains low, however, such as might occur for a distant or softly spoken person, there would be time for the compressor to increase the gain to compensate for the decreased input level. This increased gain would apply to all the syllables at the low overall input level.

The third envelope shows the result when the attack and release times are about the same length as the syllables. The level of each syllable is continually changing as the compressor adjusts the gain. The fourth envelope shows the result when the attack and release times are much shorter than the syllables. The full effect of the compressor is applied during nearly all of each syllable, in this case removing most, but not all, of the intensity differences between the syllables. The brief portions of output at the start of each syllable when the compressor is adjusting are known as ***overshoot*** or ***undershoot transients***. In the case shown, the overshoots are shorter than the undershoots, implying that the attack time is less than the release time.

The overshoots are a distortion of the speech envelope that can change the identity of sounds, such as causing a fricative (e.g. z) to sound like an affricate (e.g. dj).[1681]

Although it is not shown in Figure 6.3, the shape of the envelope is little affected by compression provided *either* the attack time or the release time is significantly longer than the syllable duration. The syllables of speech are typically 100 to 200 ms long.

There is no necessity for a compressor to have a single attack and release time. In fact, there are good reasons why the release time, and possibly the attack time, should depend on the signal being amplified. Rapid attack and release is best for protecting the aid wearer against brief intense sounds. Unfortunately, a rapid increase in gain during the pauses in speech will cause greater gain to be applied to background noise than to the speech. As we shall see in Section 6.3.2, the desirability of rapid gain variations during speech itself is hotly debated.

Several hearing aids currently available have an ***adaptive release time***. Essentially, the release time is short (e.g. 20 ms) for brief intense sounds, but becomes longer (e.g. 1 s) as the duration of an intense sound increases. When an adaptive release time is combined with a short attack time, a brief intense sound will cause the gain to rapidly decrease and then rapidly increase when the intense sound ceases. This rapid action provides protection against excessive loudness for brief sounds without affecting the audibility of following sounds. Long intense sounds (or a succession of several intense sounds, such as syllables in high-level speech) will, however, cause the release time to automatically lengthen. This slow release means that the gain will not significantly increase during each brief pause between the syllables, or change from syllable to syllable.

Adaptive release times can be achieved by using a single detector with properties that vary with the signal, by controlling a compressor from multiple detectors, or by using multiple compressors in succession. All of these systems are able to provide protection against excessive loudness when brief intense signals occur, without causing rapid fluctuations in gain when high intensity speech occurs. Compressors with variable attack and release times have been used in the broadcast industry for decades,[1622] and are now common in hearing aids. One compressor using a combination of fast- and slow-acting detectors, when used at the input of a hearing aid, has been referred to as a ***dual front-end compressor***, and has been shown to have the advantages expected of adaptive release time compression.[1231]

The advent of digital hearing aids has opened some new possibilities for controlling compressors.

d Note that the complete envelope (comprising positive and negative parts) is shown in Figure 6.3. Because envelopes are usually approximately symmetrical, we often show only the positive part, as in Figures 6.2 and 6.8.

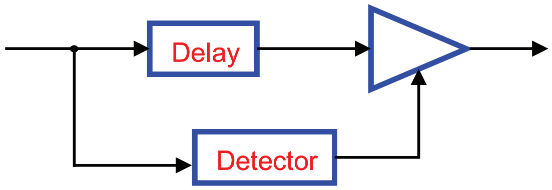


Figure 6.4 A block diagram of a feedforward, look-ahead compression control circuit.

Overshoots can be completely avoided if the compressor decreases its gain *before* the signal level increases. Does the compressor have a crystal ball? Effectively, yes! If the signal is delayed for a few milliseconds, the detector can fully react to the signal before the signal reaches the compression amplifier.[1516, 1622, 1853] An example of this, which can be called ***look-ahead compression***, is shown in Figure 6.4. This figure also shows that a compressor can operate with a ***feedforward*** control circuit instead of a ***feedback*** control circuit.

There is considerable scope for creativity in the dynamics of compression. One novel application attempts to interpret changing signal levels in terms of changes in acoustic events, and select from amongst three time constants (from very short to very long) the one considered most appropriate to the changed circumstances.[1347] In another novel application, the gain changes instantaneously each time the waveform passes through zero from negative to positive, and has the value needed to amplify the peak within the next few ms up to a pre-determined value, but the gain then remains constant until the next negative to positive zero crossing.[847] All brief segments of speech therefore end up with the same peak value, and the resulting decrease in dynamic range of the signal has been shown to increase speech intelligibility relative to a conventional compression hearing aid. Quality was not assessed.

One creative application of adaptive time constants already available in some commercial hearing aids (and cochlear implants) is ***adaptive dynamic range optimization*** (***ADRO***).[139] This algorithm is a compression-like technique that applies multiple time constants and several rules, in sequential order, to set the gain at each frequency so that:

1. gain is reduced to avoid the maximum level of the signal exceeding loudness discomfort level;
2. gain is reduced to avoid the upper levels of speech (90th percentile level) exceeding the comfortable range;
3. gain is increased to avoid the lower levels of speech (30th percentile level) becoming inaudible;
4. gain is never allowed to exceed a predetermined maximum value aimed at avoiding feedback oscillation and excessive amplification of background noise.

Rule 1 is applied with a short attack time of around 20 ms. Rules 2 and 3 are applied with very long attack and release times, respectively, of around 7 to 10 seconds. Application of these multiple rules affecting the gain at each instant in each narrow frequency region is described as fuzzy logic.[139] So long as all aims are being achieved, the gain remains constant so that linear amplification is provided. The algorithm combines the features of compression with those of adaptive noise reduction (Section 8.1).

Attack time and release time interact to affect the signal level at the output of a compression hearing aid. At one extreme, very short attack times combined with very long release times detect close to the peak level of the signal. At the other extreme, if the attack time and release times are equal, the level of the signal detected is closer to the mean level of the signal. As this level is much lower than the peak level, the compressor will "think" that the signal is weaker, and hence cause greater amplification than if peak levels controlled the compressor. Consequently, reducing release time without changing attack time will cause output levels to increase, typically by several dB.[517] The greater the compression ratio (see next section), the greater will be the effect of attack and release times on output levels.

6.2.2 Static compression characteristics

The attack and release times tell us how *quickly* a compressor operates; we need different terms to tell us by how *much* a compressor decreases the gain as level rises. When we measure and specify these gain changes we assume that the compressor has had time to fully react to variations in signal level. Consequently, the ***static characteristics*** are applicable to signals that are longer than the attack and release times.

The SPL above which the hearing aid begins compressing is referred to as the ***compression threshold***. Most hearing aids amplify linearly for input levels immediately below the compression threshold. Some continue with linear amplification down to infinitesimally small input levels, and many others have a region of expansion, as discussed in Section 4.1.5. Usually, we define the compression threshold as the *input* SPL at which compression commences, but in some circumstances it makes more sense to define it as the *output* SPL at which compression commences (see Section 6.2.3). As can be seen from the I-O diagram in Figure 6.5, the onset of compression can be very gradual. Measurement standards define compression threshold as the point at which the output deviates by 2 dB from the output that would have occurred had linear amplification continued to higher input levels.

Once the input is sufficiently intense that compression commences, the gain decreases with further increases in input level. The ***compression ratio*** describes, indirectly, how much the gain decreases. Compression ratio is defined as the change in input level needed to produce a 1 dB change in output level. With reference to Figure 6.5, it is equal to the ratio of $\Delta I/\Delta O$, and is therefore the inverse of the slope of the I-O curve.[e] If the slope of the I-O curve varies with input level, so too does the compression ratio. Compression ratios in the range of 1.5:1 to 3:1 are common in hearing aids with wide dynamic range compression.

In the linear part of the curve (below the compression threshold), every 1 dB increase in input level results in a 1 dB increase in output level. Consequently, the compression ratio of a linear amplifier is 1:1. The other extreme is compression limiting, such as that shown at the highest input levels in Figure 6.5. The slope of the I-O function here is close to zero, which means the compression ratio is very large. In practice, any compression ratio greater than about 8:1 would be considered to be compression limiting. Compression ratios can thus have any value greater than 1:1. Compression ratios less than 1:1 are also possible, but these correspond to dynamic range ***expanders*** rather than compressors (see Section 4.1.5).

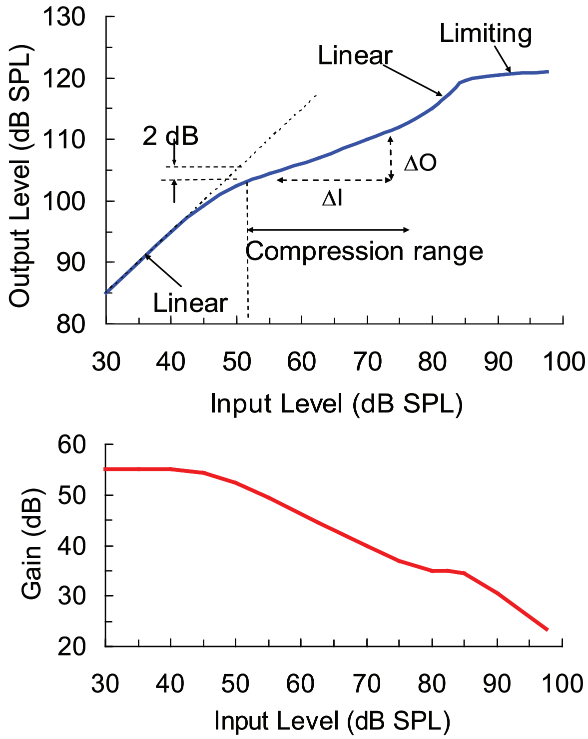


Figure 6.5 Upper: input-output diagram showing the definition of several static compression characteristics. Lower: the graph of gain versus input that corresponds to the I-O curve above it.

Important principle: The effects of user and fitter controls on compressor operation

- Any control that follows the sensing point (i.e. the feedback or feedforward point) in the signal chain does not affect the amount of compression, but does affect the final level of the compressed signal. Varying these controls causes the I-O curve to shift vertically.
- Any control that precedes the sensing point in the signal chain affects the compression threshold and hence the amount by which a signal is compressed. Varying these controls causes the I-O curve to shift horizontally.

This principle applies to the effects of tone controls as well as those of the volume control.

[e] The symbol Δ is pronounced "delta" and in mathematics stands for a small change in any quantity.

In the particular I-O function shown in Figure 6.5, there are four distinct regions, two of which correspond to linear amplification. To fully describe this curve, we thus need to specify four compression ratios (two of which are equal to 1:1) and three compression thresholds. Curves like this one might be designed to fulfill a particular purpose (e.g. loudness normalization, as will be described in Section 6.3.5). The range of inputs over which compression occurs is called the ***compression range***. In the I-O function in Figure 6.5 the first compression region has a compression threshold of 52 dB SPL, a compression ratio of 3:1, and a compression range of 30 dB. The second region has a compression threshold of 87 dB SPL, a compression ratio of 10:1, and a compression range of at least 15 dB.

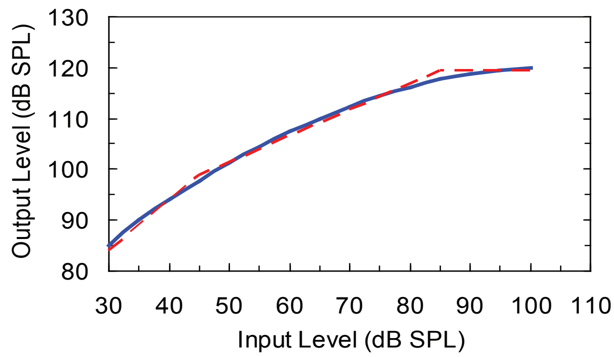


Figure 6.6 Input-output characteristics corresponding to curvilinear compression (blue solid line) and a fixed compression ratio combined with compression limiting (red dashed line).

An equally useful alternative to the I-O diagram is the input versus gain diagram shown in the lower half of Figure 6.5. Notice how the two curves show the same information:

- in the low-level linear segment, the gain is constant, so the gain-input curve is horizontal;
- in the 3:1 compression segment, the gain drops by 2/3 of a dB for every dB increase in input level;
- in the next linear segment, gain is constant;
- in the high-level compression limiting segment, gain drops by nearly 1 dB for every dB increase in input level; and
- for every input level, gain equals output SPL minus input SPL.

Sometimes, I-O curves do not comprise a number of straight lines, but are in fact curved, with the slope (and hence the compression ratio) changing continuously as the input level varies. These are called curvilinear compressors - they are no better or worse than compressors with different fixed compression ratios at different input level ranges, just more difficult to describe, except by drawing a picture, as shown in Figure 6.6. In this case, at every input level the curvilinear compressor produces much the same output level as the compressor comprising a fixed 2:1 compression ratio combined with compression limiting of high-level sounds.

We must remember that the static characteristics apply only to signals of long duration. As implied in Section 6.2.1, the hearing aid acts in an increasingly linear manner when the intensity fluctuations become increasingly rapid. That is, for rapidly changing signals, the effective compression ratio is less than the static compression ratio. The effective compression ratio is defined as the change in input level divided by the change in output level for a given signal containing high- and low-level components following each other in rapid succession. As can be seen in Figure 6.3, the effective compression ratio depends on the duration of the high- and low-level parts of the signal compared to the longer of the attack and release times.

Phonemes and syllables vary widely in duration, but it is sensible to ask what the effective compression ratio might be for a signal with a typical syllable duration of about 120 ms. Only when the attack and release times are much less than 120 ms will the effective compression ratio equal the static compression ratio. When the attack or release times are much greater than 120 ms, the effective compression ratio will be 1:1. That is, the hearing aid will amplify rapid fluctuations in speech linearly, although its gain will change when the overall level of speech changes. In between these extremes (in which the attack and release times of real hearing aids mostly fall), specification of the effective compression ratio is complex, but it will always be less than the static compression ratio.[1730, 1854] Both the static and the effective compression ratios are useful: the static ratio tells us how the long-term level of the output changes when the long-term level of the input changes. It tells us how the gain of the hearing aid changes when the acoustic environment changes. By contrast, the effective compression ratio tells us how the short-term level of the output changes when the short-term level of the input changes.[f] It tells us how the gain changes over time within a given acoustic environment.

6.2.3 Input and output control

As we have seen, compression commences at a certain SPL. How do things change if the user adjusts the volume control? The answer depends on the location of the volume control relative to the compressor within the signal chain. Figure 6.7 shows the block diagram of two hearing aids that differ in the relative location of the volume control and the compressor. Consider first the upper diagram, in which the compressor precedes the volume control. What effect does the volume control have on the operation of the compressor? Obviously none, because the compressor acts on the signal prior to the signal reaching the volume control. Consequently, compression commences at the same input SPL for all settings of the volume control. However, once the compressor has done its job (whether that be linear amplification or compression), the volume control determines the size of the output signal.

The I-O curves corresponding to different volume control adjustments are thus as shown in the upper I-O diagram in Figure 6.7. Because the compression is controlled at a point (labeled as F) on the input side of the volume control, this arrangement is referred to as input-controlled compression. It is also referred to as ***automatic gain control (input)*** or ***AGC$_i$***.

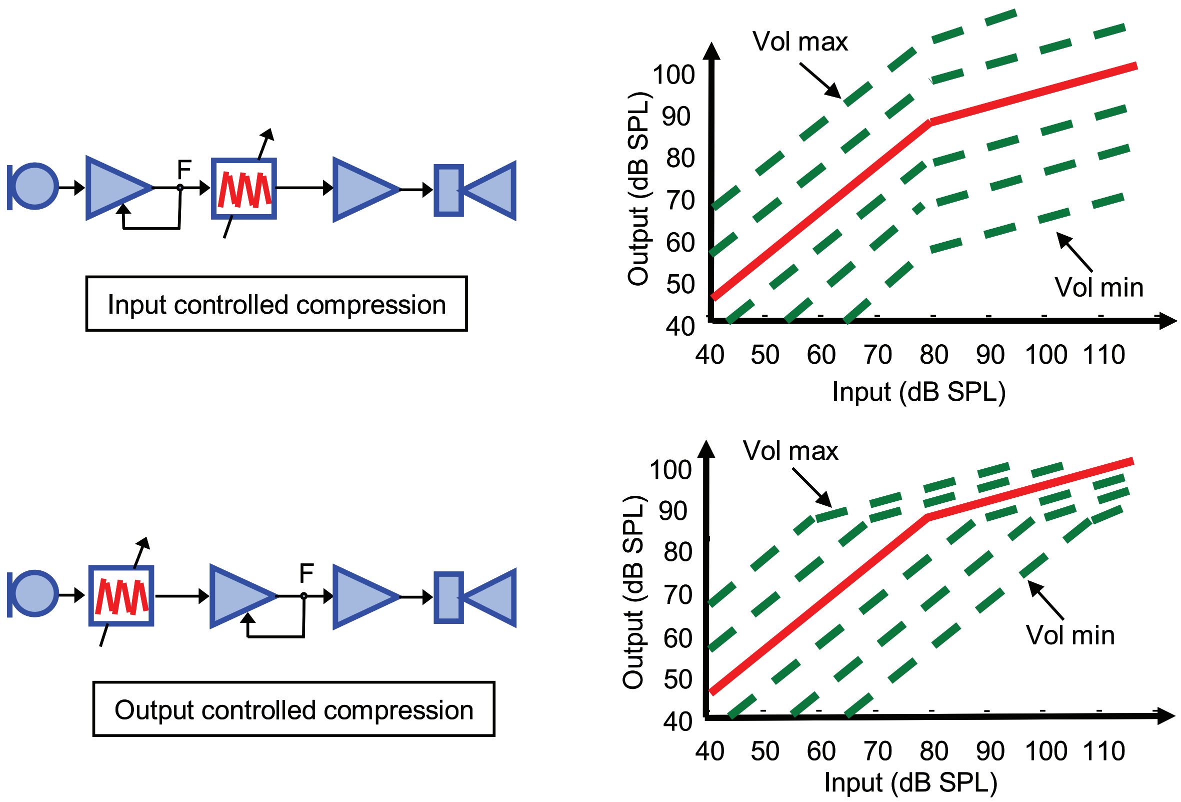


Figure 6.7 Input controlled compression and output controlled compression: their block diagrams and the I-O curves for each as the volume control is adjusted from maximum to minimum positions.

[f] Note that even the effective compression ratio does not tell the full story for a complex signal like speech. Although the dynamic range of the total signal level is compressed as described by the effective compression ratio, the dynamic range within a narrow frequency range is not compressed to the same degree as is the total broadband level.[1854] That is, the width of the speech banana (e.g. Figure 8.2) is not compressed to the extent that one would expect based on the effective compression ratio. This discrepancy occurs whenever the analysis bandwidth is less than the bandwidth of the signal passing through a compressor. The discrepancy is thus greatest for single-channel compressors and least for multichannel compressors with many channels.

Let us now compare AGC_i with the alternative arrangement, called ***output-controlled compression*** (***AGC_o***) shown in the lower half of Figure 6.7. In this case, the volume control affects the signal *before* the signal reaches the compressor. Suppose the input level was high enough for the compressor to be just in its compression region. (That is, the input level equals the input compression threshold.) If the volume control were now to be turned down, the amount of signal reaching the compressor would no longer be enough for compression to commence. Consequently, the compression threshold at the input has been increased by the amount of the gain reduction. This variation of compression threshold can be seen in the lower I-O diagram in Figure 6.7. A comparison of the I-O curves for AGC_i and AGC_o shows a basic principle: f*or AGC_i hearing aids, I-O curves move up and down as the volume control is varied, whereas for AGC_o hearing aids, they move left and right*. Equivalently, for AGC_i hearing aids, the compression threshold referred to the input is independent of the volume control position, whereas for AGC_o hearing aids, the compression threshold referred to the output is independent of the volume control setting.

Just as the position of the volume control relative to the sensing point determines the effect of the volume control on the compressor, so too the position of any fitter control determines its effect on compression. As discussed in Section 4.1.3, a compressor following a filter or tone control can partially undo the effect of the tone control on narrowband signals. Similarly, a tone control following a compressor can partially undo the effects of the compressor.[g] As the complexity of hearing aids has increased, the number of controls and the number of separate compressors within a hearing aid have both increased. It has thus become rather simplistic to describe compression hearing aids as being either AGC_i or AGC_o. For example, compression could be input-controlled with respect to the volume control, but not with respect to the tone controls. It is for this reason that block diagrams are essential if the operation of the hearing aid is to be understood.

With the increased use of wide dynamic range compression in hearing aids, it is common for hearing aids to not have a volume control. For these hearing aids, the main distinction between input and output control disappears.

6.2.4 Multichannel compression

Multichannel hearing aids split the incoming signal into different frequency bands, and each band of signal passes through a different amplification channel (Section 2.5.1 and Figure 2.2). In a multichannel compression hearing aid, each channel contains its own compressor. There are two basic reasons why we might want to compress different frequency regions by different amounts:

- the amount of compression varies with hearing loss, but hearing loss usually varies with frequency; and
- the amount of compression varies with signal level, but signals and noises in the environment have more energy in some frequency regions than in others.

Multichannel compression enables this variation of compression with frequency to be achieved. Across environments, the greatest amount of compression will occur if the compression ratio is high, and the compression threshold is low.

Even if neither the hearing loss nor the signal spectrum varies with frequency there is a theoretical argument (i.e. not yet substantiated) why we might want multiple channels of compression. In a single-channel compression hearing aid, when the compressor turns the gain down, signal components at all frequencies are decreased in level. It might not be appropriate to have signal components at one frequency being attenuated just because there is a strong signal at another frequency. Multichannel compression avoids this problem, although it can create other problems, as we will see in Section 6.5.

Although there are many ways in which compression can vary from one channel to the next, the degree of compression often either increases or decreases with frequency. A simple classification scheme describes this overall behavior. When the degree of compression is greater in the high-frequency channel(s) than in the low-frequency channel(s), there will be a greater high-frequency emphasis at low input levels than at high input levels. This characteristic has been labeled as a ***treble increase at low levels***, or a ***TILL*** response.[924] Conversely, when the degree of compression is greater in the low-frequency channel(s) than in

[g] If, for instance, a compression limiter removed all intensity differences, a filter or tone control that followed the compressor would create intensity differences that depended on the frequency or spectral shape of the signal.

the high-frequency channels, there will be less high-frequency emphasis at low input levels than at high input levels. This characteristic has been labeled as a ***bass increase at low levels***, or a ***BILL*** response.[924] Use of these terms has decreased over the last decade as compression in multi-channel hearing aids has become more complex.

6.3 Rationales for Use of Compression

This chapter so far has described how compressors work, but not what we would like them to do. The following sections outline several theoretical reasons why compressors should be included in hearing aids. There is no reason why only a single rationale should be used in any particular hearing aid, but as we will note, some combinations make more sense than others. Section 6.5 will attempt to summarize what we know about the advantages, disadvantages, and effectiveness of these forms of compression.

6.3.1 Avoiding discomfort, distortion and damage

The output of a hearing aid cannot be allowed to keep increasing in level as the input level to the hearing aid increases. There are two reasons why the maximum output must be limited in some way, and one reason why it should be limited using compression rather than peak clipping.

First, if excessively intense signals are presented to the hearing aid wearer, the resulting loudness will cause discomfort. Thus, the aid wearer's loudness discomfort level provides an upper limit to the hearing aid OSPL90.[445] Second, excessively intense signals may cause further damage to the aid wearer's residual hearing ability. As we will consider in more detail in Section 10.8, the OSPL90 may not be the most important factor to consider in avoiding damage, but it certainly is a factor.

These two reasons explain why the maximum output must be limited (i.e. why the hearing aid OSPL90 must be set appropriately), but this limiting could be achieved with either peak clipping or compression limiting. The reason for preferring compression limiting over peak clipping in nearly all cases (see Section 10.7.2 for exceptions) is that peak clipping creates distortion, as discussed in Sections 2.3.2 and 4.1.6. So too does compression limiting, but the waveform distortion created by peak clipping is far more objectionable than the envelope distortion created by compression limiting.

When compression limiting is used to control the OSPL90 of a hearing aid, it must be an output-controlled compressor, or else the OSPL90 will rise and fall with the position of the volume control. This would be unacceptable, as the user may increase the volume control position (and hence the OSPL90) when in a quiet place, only to have an intense unexpected signal occur. A high compression ratio is needed, so that the output SPL does not rise significantly for very intense input levels. The attack time must be short so that the gain decreases rapidly enough to prevent loudness discomfort. This gain reduction must be removed rapidly so that sounds following an intense sound are not overly attenuated, hence the release time must be either short or adaptive. As with all compressors, the attack and release times must not be so short that the compressor starts distorting the fine details of the waveform.

If a hearing aid does not include a compression limiter, peak clipping will occur once the input signal becomes sufficiently intense. If the hearing aid contains wide dynamic range compression, the input level needed to cause peak clipping may be so high that peak clipping seldom occurs. Despite this, all hearing aids will peak-clip once the input signal exceeds some value if for no other reason because the internal microphone pre-amplifier overloads. For good quality hearing aids, the input level at which peak clipping occurs is high enough not to matter; for others it may be so low that patients complain about the quality or intelligibility of amplified sound when they wear their hearing aids in noisy places or listen to nearby talkers.

Compression to control maximum output

- Output-controlled compression
- Compression ratio > 8:1
- Attack time <15 ms and release time between 20 and 100 ms or adaptive
- Compression threshold (referred to the output) low enough to avoid discomfort
- Single or multichannel

6.3.2 Reducing inter-syllabic and inter-phonemic intensity differences

The most intense speech sounds (some vowels) are about 30 dB more intense than the weakest sounds (some unvoiced consonants).[h] For people with very reduced dynamic ranges it may be difficult to achieve and maintain a volume control setting that makes the weakest sounds of speech sufficiently audible to be understood without the most intense sounds becoming excessively loud. Even when dynamic range is adequate to hear weak phonemes without intense ones being too loud, there is the potential for weaker parts of speech to be temporally masked by the stronger ones (see Section 1.1.4). Either a weak phoneme can be temporally masked by a stronger preceding phoneme, or the weak formants can be simultaneously masked by stronger formants.

A potential solution to both these problems is to include a fast-acting compressor that increases its gain during weak syllables or phonemes and decreases its gain during intense syllables or phonemes. Such compression, not surprisingly, is called ***syllabic compression*** or ***phonemic compression***. Figure 6.8(a) shows the envelope of the signal for a sentence spoken by a soft talker followed by the same sentence presented at a higher level (representing soft and intense speech). Part (a) of the figure also shows the envelope after amplification by such a compressor, and part (b) shows the gain applied by the compressor as it did its job. The dynamic range of the compressed output signal (the total range of the red curve) is much less than that of the linearly amplified signal (the total range of the blue curve) and hence of the input signal.

Compression to reduce inter-syllabic level differences

- Input-controlled compression
- Compression ratio > 1.5:1, but <3:1
- Attack time from 1 to 10 ms and release time from 10 to 50 ms
- Compression threshold <50 dB SPL
- Single or multichannel

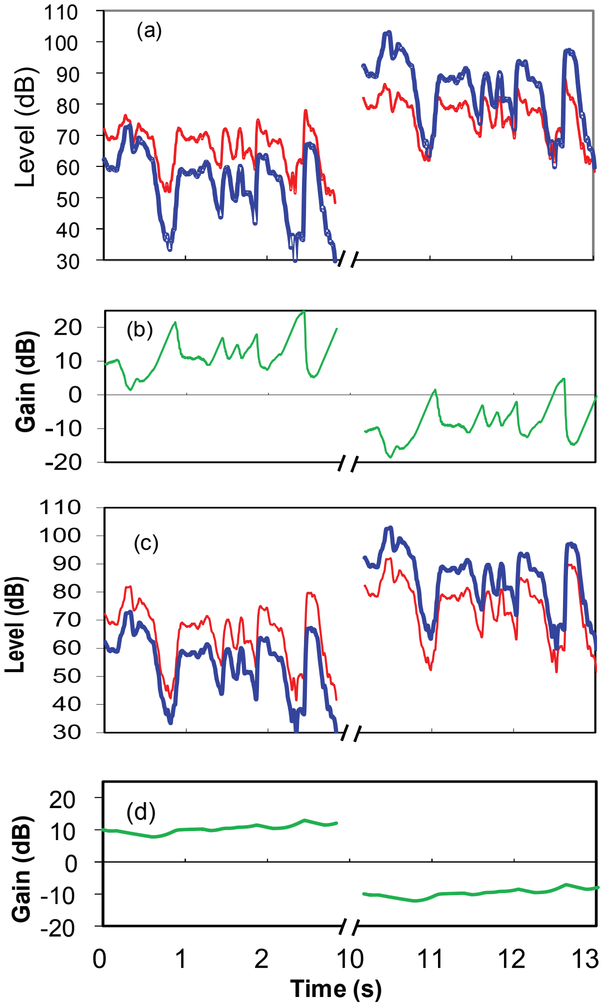


Figure 6.8 (a) Envelope of the signal *The yellow flower has a big bud* put into the hearing aid at one level (from 0 to 3 s) and then several seconds later at a 30 dB higher level (from 10 to 13 s). The thick blue curve shows the envelope for linear amplification and the thin red curve shows the envelope for a compressor with a 3:1 compression ratio, attack time of 20 ms, and release time of 200 ms.

Part (b) shows the gain applied by the compressor.

Part (c) shows the envelope for linear amplification and for compression when the attack and release times of the compressor were increased to 1000 and 2000 ms respectively.

The corresponding gain is shown in part (d).

h This intensity difference can be judged either from acoustical measurements of intensity, or by assessing how far speech has to be attenuated before different consonants become inaudible.[551, 892]

A potential problem is that fast compression alters the intensity relationships between different phonemes and syllables. This might seem like a strange thing to say, because altering the intensity relationships is the *aim* of the processing. However, if the hearing aid wearer uses the relative intensities of sounds to help identify them, altering relative intensities can decrease the intelligibility of some speech sounds, even if it increases their audibility.[808, 1436]

Another potential problem is the effect that compression has on brief weak sounds that follow closely after sustained intense sounds. Suppose a sound of higher than average level causes the gain to be lower than would be chosen for a linear amplifier. If the release time is longer than the gap between the intense and the weak sound, then the gain will still be decreased when the brief weak sound arrives. Consequently, such weak sounds will be *less* audible than they would be for linear amplification. Release times of 50 ms or less may be sufficiently short to eliminate this problem.

Yet another problem with fast-acting compression of any sort is that if the gain is fast enough to increase when a soft phoneme occurs, it is also fast enough to increase during pauses between words. Does this matter? If there is any background noise, it does. When noise is less intense than the speech, the compressor will increase its gain during the noise, and decrease it during the speech. Amplifying noise by a greater amount than speech is *not* a desirable feature in any hearing aid, but this disadvantage has to be weighed against the advantages of fast-acting compression.

For release times between about 100 ms and 3 s, the hearing aid wearer can hear noises grow louder following cessation of a preceding higher level sound. This phenomenon, where the loudness of one sound is clearly affected by the cessation and perhaps commencement of another sound, is referred to as ***pumping***. For release times shorter than about 50 ms, compression also amplifies background noise more during pauses than during speech sounds, but the loudness increase may occur too rapidly for the aid wearer to perceive a change in loudness due to compression.[i] Pumping is more evident for single-channel compression than multichannel compression, and is more evident when speech and background noise come from different directions.[915] Hearing aid wearers are unlikely to actually use the word pumping to describe its effects.

Compressors intended to decrease the intensity differences between syllables must have compression thresholds low enough for the compression to be active across the range of short-term input levels that apply to speech. They must have compression ratios high enough to significantly decrease dynamic range, but low enough to leave some intensity differences intact. Attack and release times have to be short enough that the gain can vary appreciably from one syllable or phoneme to the next, but not so short that they create significant amounts of distortion to the waveform. Phonemes are shorter than syllables, so phonemic compression requires attack and release times even shorter than syllabic compression. Because phonemic compression will amplify a consonant by a different (usually greater) amount than an adjoining vowel, it changes the consonant-to-vowel level ratio, as discussed in more detail in Section 8.4.

6.3.3 Reducing differences in long-term level

Although the fast-acting compressor discussed in the preceding section was intended to decrease inter-syllabic level differences, Figure 6.8 makes it clear that it had two effects. As well as changing the inter-syllabic relationships, the mean level difference between the soft and the intense speech has been decreased from 30 dB to 10 dB. An alternative use of compression is to decrease the longer-term dynamic range, but without changing the intensity relationships between syllables that follow each other closely in time. This is achieved by using attack and release times much longer than the typical duration of syllables.

Parts (c) and (d) of the figure show the envelope of the output signal and the gain applied to the compressor. There are several things to observe. First, notice that the gain now changes much less during each sentence than for the fast-acting compressor. Second, as a consequence, the intensity relationships between syllables at the output are almost identical to those at the input. Third, the desired goal has been achieved: the average level of the first sentence is now only 10 dB lower than that of the second sentence.

[i] The loudness of noise during speech sounds is normally less than in the pauses, even without compression, because speech sounds will partially mask the noise.

This type of compressor is often called an ***automatic volume control***. The term is appropriate, because the compressor varies the gain in very much the same way a person would adjust the volume control to partially compensate for differences in the incoming levels of sounds. Incoming levels may be high because the talker is close, because the talker has a naturally powerful voice, or because the talker has raised his or her voice above background noise. Incoming sounds of interest may not even be speech. People prefer sounds to be presented at different levels in different environments.[1649] (Life would be less interesting if some gently spoken words and the whistle of a steam train at close quarters were both to be heard at the same intensity and loudness.) Consequently, one would not want the automatic volume control to have too high a compression ratio. The optimum compression ratio depends on the dynamic range of the hearing aid wearer and the range of sounds to which the aid wearer would like to comfortably listen without having to adjust a manual volume control.

The biggest problem with slow-acting compressors is what happens when the input level varies suddenly. Suppose a person has for some time been listening to a softly spoken person in a quiet place. The hearing aid will react by turning up the gain appropriately. If a loud noise then occurs, or a loud talker joins the conversation, the new sound will be amplified with the high gain that was appropriate to the weaker talker. The output will thus be excessive and must be decreased with an appropriate limiter of some type, preferably a compression limiter. Sudden increases in level are very common: they will probably occur every time the aid wearer talks, because unless people are being very friendly, his or her mouth is probably closer to the hearing aid than is anybody else's.

The opposite problem, a sudden decrease in level, also occurs, but is not so easily fixed. If everyone at a gathering suddenly stops talking to hear what one person is saying, the wearer of an automatic volume control hearing aid may miss the important announcement if the hearing aid still has the gain appropriate to the higher input level that was present a moment before. This problem is minimized by having a release time no longer than that necessary to avoid rapid increases in gain during brief pauses in the conversation. Several multichannel hearing aids on the market use separate slow-acting compressors in each channel. Some hearing aids have level-dependent attack and release times, such that the gain changes more quickly for large changes in level (as will occur when the environment changes) than for the smaller changes in level that occur within continuous speech.

6.3.4 Increasing sound comfort

One might expect that a compression limiter would solve any problems caused by excessive loudness. While it is certainly true that setting the OSPL90 low enough will prevent loudness discomfort, people may not like the signal being close to discomfort level for a large proportion of the time. It may not be satisfactory to simply further decrease OSPL90, as this prevents any sounds from getting close to discomfort, and thus decreases the useable dynamic range by an even greater degree than does the person's hearing loss! One solution to this problem is to use, for higher level inputs, a form of compression that is more gradual than compression limiting.

Compression to decrease long-term level differences

- Input-controlled compression
- Compression ratio > 1.5:1, but < 4:1
- Attack time >100 ms and release time > 400 ms
- Compression threshold < 50 dB SPL
- Single or multichannel

Compression to increase comfort

- Input-controlled compression
- Compression ratio > 1.5:1, but < 4:1
- Attack time and release time unknown, possibly not important, but release time not too short
- Compression threshold approximately 65 dB SPL
- Single or multichannel

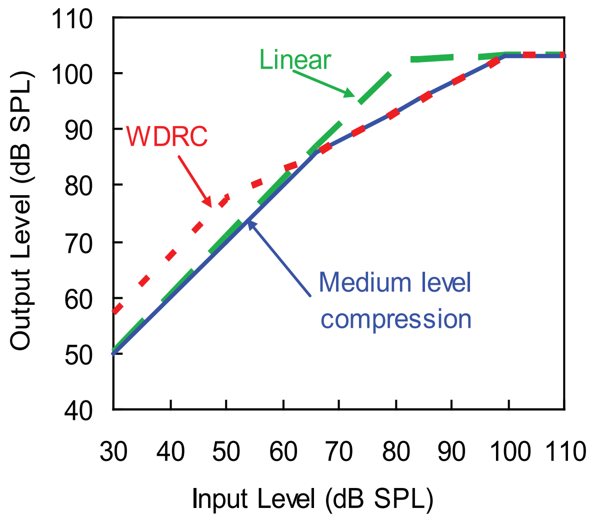


Figure 6.9 Input-output curves for medium level compression, wide dynamic range compression, and linear amplification, all combined with either compression limiting or peak clipping of high level signals.

Figure 6.9 shows the I-O diagram of a compressor that is activated only when the input SPL is at or above typical input levels. There is no agreed name for such compression, but it could be termed either ***medium-level compression*** or ***high-level compression***.[225] Two other I-O curves are shown for comparison. All three curves have the same output level for an input level of 65 dB SPL. The medium-level compressor provides the same gain for low-level signals as does the linear amplifier, and consequently does not boost soft signals like the wide dynamic range compressor does. Both the WDRC hearing aid and the medium-level compression hearing aid decrease their gain gradually once the input level rises above about 65 dB SPL, and so both increase comfort in noisy places. If, for instance, the OSPL90 of the hearing aid had been set close to, but below the hearing aid wearer's discomfort level, then for the linear aid, this maximum output level will be achieved whenever the input level is greater than 84 dB SPL. By contrast, the input level has to exceed 99 dB SPL before the output of the medium-level compressor or the wide dynamic range compressor reaches this level.

In summary, a gradual form of compression for medium- to high-level sounds will increase comfort in noise without conveying the advantages and disadvantages of gain increases for weak input sounds. Note that we do not necessarily expect the compressor to increase intelligibility in noisy places, but we do expect it to increase listening comfort. Of course, a reduction in output level can sometimes result in an increase in intelligibility, for reasons that are not fully understood. The poorer intelligibility at high levels of stimulation may possibly result from the increased spread of excitation (i.e. spread of masking) in the cochlea at high levels.

Before moving on to the next rationale, we should note that the first four rationales have all been discussed as though the hearing aid contained only a single compressor covering the entire frequency range. In fact, the comments made about each rationale apply equally well to each channel of a multichannel compression hearing aid. One can thus have:

- multichannel compression limiting - to achieve different OSPL90s at different frequencies;
- multichannel inter-syllabic intensity reduction - to decrease the intensity differences between syllables more in one frequency range than in another;[1859]
- multichannel automatic volume control - to slowly change the gain and shape of the frequency response as the long-term level and long-term spectral shape of the input signal varies; or
- multichannel comfort control - to decrease gain in noisy places more in some frequency regions than in others.

The justification for all of these is that because hearing characteristics (threshold, comfort level, discomfort level, dynamic range) vary with frequency, so too should a solution that in some way aims to match signals to these hearing characteristics.

6.3.5 Normalizing loudness

Probably the most common approach to deriving compression characteristics (though not necessarily the optimum approach) is to normalize the perception of loudness. As mentioned in Section 1.1.2, sensorineural hearing loss greatly affects loudness perception. The principle of loudness normalization is simple: for any input level and frequency, give the hearing aid the gain needed for the aid wearer to report the loudness to be the same as that which a person with normal hearing would report.

The required amount of gain at each input level can be deduced from a graph showing the loudness of sounds at different levels. Loudness can only be measured subjectively and there are several ways in which it can be measured. Currently, the most popular way is to ask the hearing-impaired person to rate loudness using one of several terms. This procedure is called ***categorical scaling of loudness***. The scales commonly have about seven different labels. In variations of the procedure, responses intermediate to the labels are also allowed.

Figure 6.10 shows a graph of loudness category versus SPL (referenced to SPL in a 2-cc coupler) for one hearing-impaired person and for an average normal-hearing person.[342] These graphs are often referred to as ***loudness growth curves***. Consider first the SPL needed to produce a rating of *very soft*. Whereas the normal-hearing people need only 23 dB SPL, the hearing-impaired person needs 66 dB SPL. The difference between these values, 43 dB, is of course, almost as large as the loss in hearing threshold, which we can think of as the hearing loss for extremely soft sounds. The difference is the insertion gain needed for an input level of 23 dB SPL if the hearing-impaired person is to rate the sound as being *very soft*. This comparison enables us to plot one point on the curve of insertion gain versus input SPL, as shown in part (b). If we know the input level (23 dB SPL) and the gain (43 dB), we can also specify the output level (66 dB SPL), so this also gives us one point on the I-O curve, as shown in part (c).[j] This process can be repeated for all the other loudness categories. For example, the hearing-impaired person needs 98 dB SPL for sounds to be *loud but OK*, which is 20 dB greater than the 78 dB SPL needed for normally hearing people. Consequently, an input level of 78 dB SPL requires an insertion gain of 20 dB and an output SPL of 98 dB SPL.

What can we conclude about the type of compression needed for loudness normalization? The greatest compression ratio is needed for low-level inputs. Consequently, we could refer to this as ***low-level compression***. The compression ratio becomes closer to 1:1 as the input level increases, but whether amplification actually becomes linear depends on whether the hearing-impaired and normal-hearing loudness functions ever become parallel.

Because the amount of compression needed depends on loudness perception, loudness perception depends

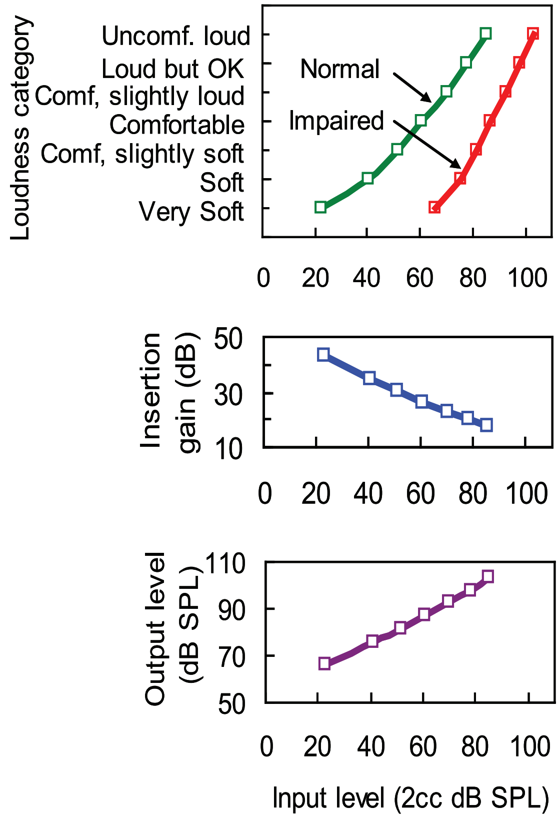


Figure 6.10 (a) Loudness growth curves for normal hearing people and a hearing impaired person with a 50 dB hearing loss. (b) Insertion gain needed for the impaired listener to receive a normal loudness sensation. (c) The corresponding I-O curve.

Compression to normalize loudness

- No volume control
- Compression ratio decreasing as input level increases
- Attack time and release time long or short
- Compression threshold as low as possible
- Different compression ratios needed for different frequencies

[j] Because the gain is an insertion gain, the 2-cc coupler output SPL will equal the input SPL, plus the insertion gain, plus the appropriate CORFIG (see Section 4.4.2). As CORFIG for an ITE at 1 kHz is close to zero, it is ignored in this example.

on hearing threshold loss, and threshold loss depends on frequency, it will not be surprising that the degree of compression for loudness normalization often varies markedly with frequency. Take for instance a person with near-normal low-frequency hearing and a high-frequency loss. Loudness perception is most different from normal for low-intensity, high-frequency sounds. Consequently, these sounds require the most gain and the greatest compression ratio. Except for people with flat hearing losses, loudness normalization will thus require the shape of the gain-frequency characteristic to change with input level. High-tone loss is the most common configuration. For such losses, as input level decreases, loudness normalization results in high-frequency gain increasing at a faster rate than low-frequency gain. Loudness normalization thus usually requires a TILL response (except of course for flat or reverse-slope losses).

The most common way of achieving loudness normalization is with separate compressors located in each channel of a multichannel hearing aid, such as that shown in Figure 2.2. Alternatively, the hearing aid may contain only two channels, and have a compressor in only the high-frequency channel. It is possible, however, to combine a compressor with a filter that alters its shape with input level, so that even a single-channel hearing aid can have a level-dependent frequency response. One well known early example of this is the K-Amp™, in which the gain and the corner frequency of a high pass filter simultaneously decrease as the input level increases, as shown in Figure 6.11.

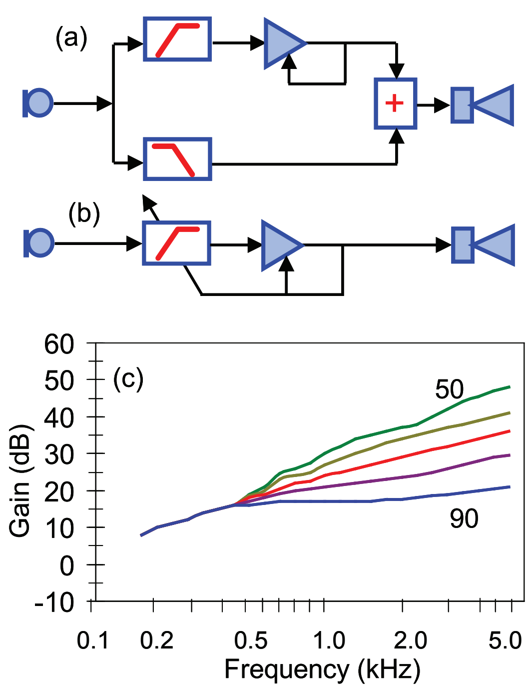


Figure 6.11 Block diagrams of (a) two-channel and (b) single channel processing schemes that can implement approximations of loudness normalisation, and (c) the resulting typical TILL gain-frequency response that increases in slope as the input level decreases from 90 to 50 dB SPL.

Compressors can attempt to normalize loudness for brief sounds by using fast attack and release times, or they can attempt to normalize only the average loudness in each frequency region, by using long attack and release times. Both types of hearing aids are available, but fast compression is a more complete form of loudness normalization. For example, it prevents the amplitude fluctuations in noise appearing so marked to patients with hearing loss that they misperceive troughs as complete gaps.[1230]

6.3.6 Maximizing intelligibility

Multichannel compression can be used to achieve, in each frequency region, the amount of audibility that maximizes intelligibility, subject to some constraint about the overall loudness. Such an approach will result in loudness *not* being normalized in any frequency region, although the overall loudness of broadband sounds may well be normalized. Further discussion of this rationale requires an understanding of several aspects of hearing aid prescription, and we will return to this topic in Chapter 10.

6.3.7 Reducing noise

The interfering effect of background noise is the single biggest problem faced by hearing aid wearers. Not surprisingly, compression is used to decrease the effects of noise. The assumptions behind this approach are as follows:

- Noise usually has a greater low-frequency emphasis than does speech (because of the combined effects of the nature of many noise sources, reverberation, diffraction around obstacles, and even distance).
- The low-frequency parts of speech are therefore the most likely to be masked so little speech information may be available at low frequencies.
- The low-frequency parts of noise may cause upward spread of masking and so mask the high-frequency parts of speech.

Important principle: the effect of compression on signal-to-noise ratio

- At a given instant in time, everything (i.e. signal and noise) passing through a compressor is amplified by the same amount, thus leaving SNR for simultaneous signal and noise unaffected at every frequency.
- Compressors (and filters) can, however, improve the overall SNR, although SNR at every frequency is unchanged (see Figure 6.12 for an example).
- Provided either the signal or noise entering a compressor is above compression threshold, and provided the attack and release times are short enough, signal and noise present at different times will be amplified by different amounts by a compressor. If the (non-simultaneous) SNR is positive, the compressor will decrease SNR. If the non-simultaneous SNR is negative, the compressor will increase SNR.
- Compression therefore moves the non-simultaneous SNR towards 0 dB, whereas expansion moves it away from 0 dB.

The actual change in SNR caused by compression depends on the compression ratio, attack and release times, number of channels, fluctuations in the noise, and, of course, the input SNR.[1307]

- The low-frequency parts of noise contribute most to the loudness of the noise.
- SNR generally decreases as the SPL in the environment decreases.[1399]

Consequently, if the low-frequency parts of the noise cause masking and excessive loudness, and the low-frequency parts of speech do not convey any useful information in noise, then comfort should be increased by decreasing low-frequency gain in high-level environments. Intelligibility may possibly be improved.

Figure 6.12 illustrates this. Part (a) shows the spectrum of a signal and the spectrum of a noise. The remainder of the argument is the same whether these are long-term spectra, averaged over a minute or more, or short-term spectra averaged over a few milliseconds. If we assume that information can be extracted from the speech spectrum whenever it exceeds the noise spectrum, then for these particular spectra, only information above 1 kHz is available to the listener. Furthermore, if there was as much upward spread of masking as that indicated by the uppermost dotted line, then nearly the entire spectrum of the speech will be masked by the noise. With or without upward spread of masking, the low-frequency region contributes no useful information in this environment. Despite that, it contributes enormously to loudness. If the level in the environment is much higher than a typical level of approximately 70 dB SPL, the loudness will be greater than is comfortable, and is unlikely to be welcomed by the aid wearer, particularly if it is dominated by unwanted low-frequency noise.

One solution to all of this is to decrease the gain of the troublesome low-frequency region, as shown in Figure 6.12(b). In this particular case, the gain at each frequency is proportional to the signal-to-noise ratio (SNR) at each frequency. (As we shall see in Section 8.1.1, this particular rule for altering the gain at each frequency is known as ***Wiener filtering*** and is commonly used in noise reduction algorithms.) The corresponding output spectrum is shown in Figure 6.12(c).

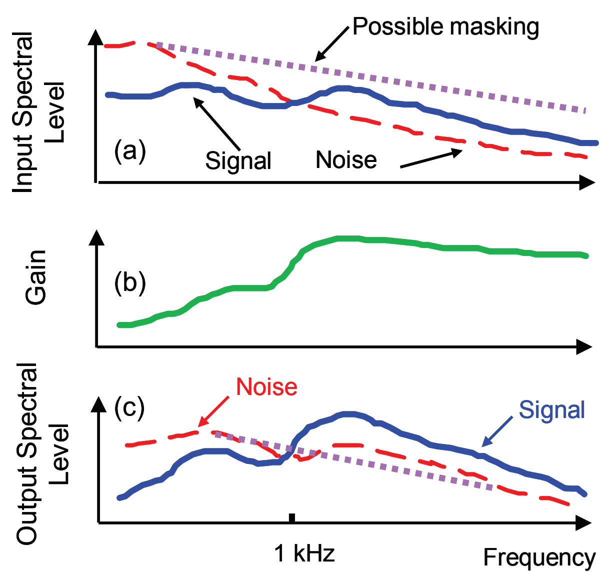


Figure 6.12 (a) Spectrum of the signal and noise input to a noise reduction hearing aid.(b) Gain applied to the signal and noise.(c) Spectrum of the signal and noise at the hearing aid output.

Notice that at every frequency, the SNR at the output is identical to that at the input, as it must be, because signal and noise both get amplified by whatever gain is present at that frequency. Consequently, information is still available only above 1 kHz. If the aid wearer can still extract information over only the same frequency region, then has this processing helped? Probably. First, the loudness and annoyance of the noise will have been greatly decreased. Second, there is now no chance that upward masking will further decrease the useful range of frequencies.

In summary, noise reduction should increase comfort (relative to a hearing aid with a fixed gain and frequency response). Noise reduction should result in an intelligibility increase only when the spectrum of the noise is markedly different from the spectrum of the signal (which is not commonly the case). Both of these expectations have been verified in practice.[453, 535, 1016, 1365] In those cases where the "noise" is actually one or more people talking nearby, and the signal is also somebody talking, the signal and noise will have similar spectra, and they cannot be separated by filtering (or currently within hearing aids by any electronic means other than directional microphones).

Critics of the noise-reduction rationale have argued that it may be best not to alter the frequency response shape electronically in noisy environments, but rather to present the full spectrum to the aid wearer.[912] This approach relies on the aid wearer's ear and brain being able to separate the signal from the noise, which of course is more difficult for people with hearing impairment because of their impaired frequency and temporal resolution.

Low-frequency compression may have an additional benefit. The aid wearer's own voice has a greater low-frequency emphasis, and a greater overall level, at the hearing aid microphone than the voices of other people. Consequently, if a hearing aid has been adjusted to give an optimal output in the middle of the wearer's dynamic range for typical speech from a conversational partner, the inclusion of compression for low frequencies will help give the aid wearer's own voice a more acceptable tonal quality than would occur for linear amplification.[1015]

The noise-reduction rationale can be implemented with slow attack and release times, in which case the frequency response would slowly change, depending on the long-term spectra of the signal and the noise. Alternatively, it can be implemented with fast attack and release times, in which case the frequency response would change rapidly depending on the short-term spectrum of the signal and the noise.

Reduction of low-level noise can also be accomplished through expansion as explained in Section 4.1.5. It might seem strange that both compression and expansion, which have the opposite effect on signals, can reduce noise. As explained in the accompanying box, compression improves the non-simultaneous SNR when the SNR is negative which often applies in extremely noisy situations, whereas expansion improves the non-simultaneous SNR when the SNR is positive, which is usually the situation for very low-level noise. Expansion is most effective, subjectively and objectively, when it has fast attack and release times.[1442, 1443] Noise is then reduced even in gaps in speech as short as 100 ms, but the gain increases quickly once speech re-commences so that the loss of audibility is minimized.

The expansion threshold and expansion ratio must, however, be carefully adjusted: too high causes expansion to reduce the audibility and intelligibility of weak phonemes in soft speech; too low and it is ineffective in reducing the loudness of low-level noise. The better the hearing thresholds of the aid wearer, the more likely it is that benefits of expansion will outweigh its disadvantages, though people with hearing loss anywhere in the mild or moderate range appreciate the noise reduction that expansion provides in quiet places.[1441] There may be scope to improve hearing aids by making the expansion threshold depend on the thresholds of the wearers, and/or by making it adaptive such that it depends on whether low-level speech is present.

Compression to decrease noise

- Gain reduction where SNR is worst (usually low frequencies)
- Sometimes approximated by compressing only the low frequencies
- Attack time and release time long or short
- Compression threshold medium
- Usually implemented by multichannel signal processing

BILL and TILL: complementary contradictions

For a person with a high tone loss, the loudness normalization and noise-reduction philosophies lead to contradictory conclusions about what to do. As level decreases, loudness normalization requires a steeper response achieved by high-frequency compression (i.e. TILL; Figure 6.11), while noise reduction usually requires a flatter response achieved by low-frequency compression (i.e. BILL; Figure 6.13). For a person with a high tone loss listening in low-frequency weighted noise, both rationales logically cannot be correct, though both arguments probably have some validity.

If loudness normalization and noise reduction were both implemented with fast-acting compression, the net result would be very similar to single-channel, wide dynamic range compression aimed at reducing the intensity differences between syllables. If both were implemented with slow-acting compression, the net result would be very similar to an automatic volume control.

What the two philosophies have in common is that averaged across frequencies, less gain is needed for high input levels than for low input levels.

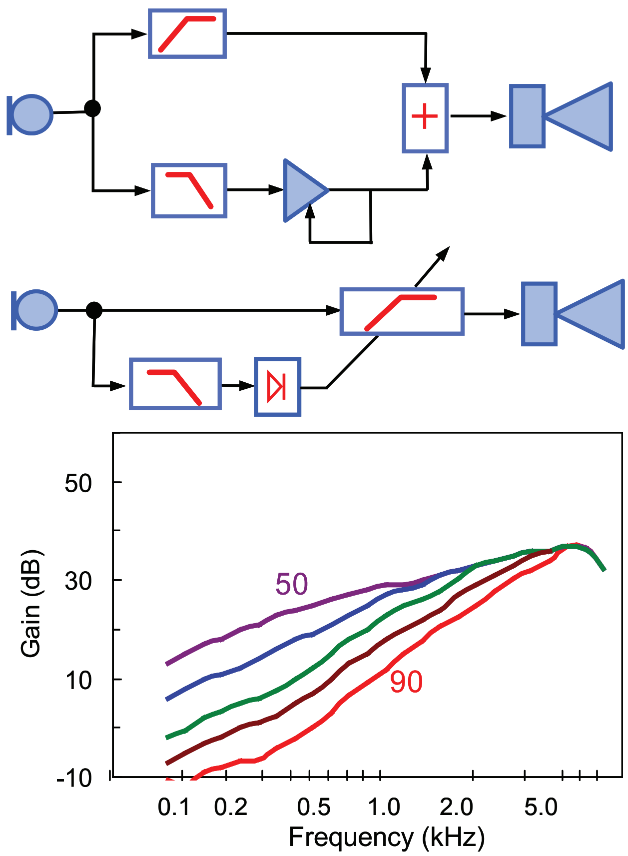


Figure 6.13 Block diagrams of two-channel and single-channel processing schemes that can implement simple noise reduction strategies, and the resulting BILL response that decreases in slope as the input level decreases from 90 to 50 dB SPL.

6.3.8 Empirical approaches

The previous seven rationales have all had an underlying theoretical rationale (and perhaps little empirical evidence supporting them). An alternative approach is to experimentally compare different forms of compression and choose the one that is preferred by the aid wearers or that gives the highest speech intelligibility or, hopefully, both. Unfortunately, there are insufficient studies to form a complete basis for selecting compression, and results are somewhat contradictory.

As examples, one study found that fast-acting compression was more valuable in the low frequencies than in the high frequencies, particularly for those people with the *widest* dynamic range in the low frequencies.[1084] Another study found that subjects with severe-profound loss preferred compression acting faster in the high frequencies than in the low frequencies[878], and a further study with severe-loss patients found that multi-channel WDRC was worse than linear amplification with compression limiting.[1679]

At our current stage of understanding, the empirical and theoretical approaches seem equally reasonable. On the one hand, our theoretical understanding of hearing impairment and the effects of different forms of compression with real world stimuli is incomplete. On the other hand, experiments can answer only the questions their design asks, and under only the conditions their design allows. Neither a theoretical nor an empirical approach to finding the best hearing aid processing should be accepted as being necessarily correct, or dismissed as being the wrong approach. Indeed, we will not be confident that we have the best form of processing for an individual until we have both a theoretical understanding of how the aid is helping *and* empirical evidence that the approach is better than any of several reasonable alternatives.

6.4 Combinations of Compressors in Hearing Aids

There is no reason why a hearing aid should contain a single compressor or be based on a single rationale. Some of the rationales would seem to combine particularly well. As but one of many possible examples, a hearing aid could combine:

- an input compression limiter to prevent very high-level input signals from overloading the circuitry in the rest of the hearing aid (several hearing aids include these);
- a slow-acting compressor to decrease the dynamic range associated with changes in long-term input level, or alternatively, a multichannel structure, with a slow-acting compressor in each channel; and
- a fast-acting output-controlled compression limiter to prevent the output from exceeding the required maximum output limit, without waveform distortion (many hearing aids have such a limiter).

As we have seen, several rationales require different amounts of compression in different frequency regions. The most straightforward way of achieving this is within a parallel structure, multichannel hearing aid as described in Section 6.2.4. When evaluating how many channels are needed or present in a hearing aid, it is wise not to overlook the effects of venting. As we have seen in Section 5.3.1, any vented or leaky hearing aid has a low-frequency, non-electronic, parallel channel. This will cause apparently single-channel hearing aids to act as though they are two-channel aids. Conversely, hearing aids that are apparently nonlinear in the low-frequency channel(s) may behave as though the low-frequency channel is linear if the vent-transmitted sound dominates the electronically modified sound.

There have been many attempts to classify compression systems into families of similar types. None of the systems does justice to the multitude of ways compression can be used. The most useful system is the TILL-BILL system referred to in Section 6.2.4. This system is simple but does not distinguish between compression that is most active at higher levels versus compression that is most active at lower levels (see Figure 6.1). Nor does it distinguish between fast-acting, slow-acting, and adaptive attack and release times. A more complex system[434] distinguishes between low and high-level compression. The only way to unambiguously describe a compression hearing aid is by a block diagram (or its verbal equivalent), information about compression speed, and either some I-O curves at different frequencies, or some gain-frequency responses at different input levels (preferably measured with a speech-like signal), or both.

A new combination of compressors is enabled by wireless-linked bilateral hearing aids. Totally independent compression in each hearing aid on opposite sides of the head reduces the normal interaural level differences presented to the ear, because the signal on the side closer to the source is stronger, and hence that hearing aid's compressor will select a lower gain than the compressor on the far side of the head. The resulting reduction in interaural level differences has the theoretical potential to reduce left-right localization accuracy. It has not, however, so far been shown to affect left-right localization, at least under controlled, anechoic, conditions, presumably because compression does not disturb interaural time differences.[887] Should it prove to be a problem in more reverberant conditions where interaural time differences may provide less reliable cues, coordinated bilateral compression of the two hearing aids should avoid the problem.

6.5 Benefits and Disadvantages of Different Compression Systems

In this section we will review the relative advantages and disadvantages of the different compression rationales. This is not a straightforward issue, because the advantage of a compression system or rationale depends on:

- the alternative to which it is being compared;
- the criterion used (intelligibility or quality);
- the signal level;
- the type of signal (e.g. speech, music, environmental sounds), the presence and type of noise, and the SNR,
- the frequency response shaping used (compression has a more beneficial effect on audibility if high-frequency emphasis has not been used); and
- the hearing loss characteristics of the research participants.

Compression affects the overall level of the output signal, so an important factor in any comparison is how

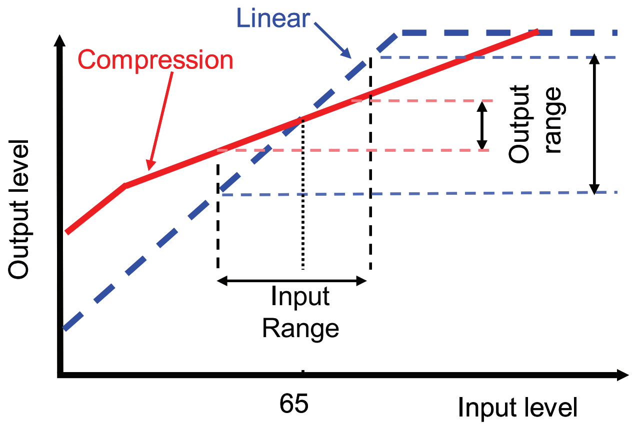


Figure 6.14 Input-output functions for two different hearing aids adjusted to have the same output for a 65 dB SPL input signal.

the volume control for each of the systems is adjusted. In this section, we will assume that any amplification systems being compared have been adjusted so that all have the same long-term output level when they are receiving an input signal with an average (65 dB SPL) long-term input level. For example, Figure 6.14 shows the I-O function for two different amplifiers, one linear, and one with a 2:1 compression ratio.

6.5.1 Compression relative to linear amplification

Table 6.1 shows the advantages and disadvantages we *expect* that each type of compression system should have.[434] Unfortunately, there is little experimental evidence to either support or refute some of these expectations. Fortunately, most of the advantages and disadvantages are inevitable consequences of the changes in gain and changes in output level that accompany compression. Suppose, for example, that a compression aid with a low compression threshold and a linear aid have the same gain for a moderate input level. When a low-level sound is input to both aids, the compression aid *will* have more gain, so its output *will* be more audible. On the downside, the compression aid *will* have a greater risk of feedback, which *will* cause a problem if there is enough leakage or a sufficiently large vent. Physical effects such as these are inevitable.

It is much harder to predict the effect of each compression rationale on intelligibility and comfort. Adding just about any sort of compressor will increase the range of input sounds that fall within a person's comfort range without use of the volume control. Unfortunately, there is no theoretical basis for us to predict how much compression is optimal. Hearing aid wearers will need to trade-off the increased loudness comfort and audibility against any extra amplification of background noise occurring in the gaps of speech, and against any adverse change in the quality of speech or other signals. Consequently, we need to look to empirical evidence to assess the advantages of different systems.

A comprehensive review[434] of the relative intelligibility of different compression systems lead to the following conclusions, which are also supported by more recent research, as indicated.

1. Limiting. To limit the maximum output of hearing aids, compression limiting should be used rather than peak clipping, except for hearing aids intended for people with the most profound losses.[693] For some people with moderate or severe losses, the compression limiter may offer no advantages over peak clipping, but it will not have any disadvantages. For others, the distortion in the peak clipper will be evident and the compression limiting will be preferred. There is thus no reason not to use compression limiting for everyone except those who will benefit from the additional SPL that can be generated with a peak clipping aid. These exceptions have so much hearing loss they should also be considered for cochlear implants.

2. Typical input levels. If a linear hearing aid is properly prescribed, and the aid wearer adjusts the volume control to get a comfortable loudness, there is no compelling evidence that any form of compression provides superior intelligibility. For speech that is already at an optimal level in the absence of compression, slow-acting compression does not affect the speech. Fast-acting WDRC will decrease the dynamic range of speech (see Figure 6.14), but will either not affect overall intelligibility,[1831, 1875] may very slightly decrease it,[396, 731, 1362, 1680] or may very slightly increase it.[1139] Certainly, the types of confusions are differ-

Practical advantages of medium or low compression thresholds over linear amplification

- Listening comfort is increased in noisy places
- Need for a volume control is decreased

Table 6.1. Summary of compression rationales, methods for implementation, and theoretically expected advantages and disadvantages. The advantages and disadvantages are based on the assumption that all systems produce the same output for speech at 65 dB SPL, as shown in Figure 6.14.

Rationale	Implementation	Expected Advantages (re linear amplification)	Expected Disadvantages (re linear amplification)
Discomfort, distortion and damage avoidance	Fast-acting wideband or multi-channel compression limiting	• No discomfort • Little distortion	• Less OSPL90 possible than with peak clipping
Reduction of inter-syllabic intensity differences	Fast-acting wideband or multichannel compression with low compression threshold	• Signal kept in audible range without using volume control for a wider range of overall levels and for soft and weak phonemes	• Decreased SNR for noises occurring within the gaps of speech • Increased chance of feedback • Useful intensity cues may be disrupted
Long-term dynamic range reduction	Slow-acting wideband or multichannel compression with low compression threshold	• Less need to vary volume control • No disruption of intensity cues to different phonemes	• Need further compression to avoid discomfort • Increased chance of feedback • Soft phonemes may still be inaudible, and loud sounds may fall outside most comfortable range
Comfort increase	Slow- or fast-acting wideband or multichannel compression with a medium compression threshold	• Increased comfort in noisy places without having to decrease the volume control	• Decreased SNR for noises occurring within the gaps of speech
Loudness normalization	Slow- or fast-acting multichannel compression, or adaptive high pass filter (frequency response typically steeper at low input levels)	• Signal kept in audible range without using volume control for a wider range of overall levels (and for soft and weak phonemes if compression is fast acting) • Normal tonal balance at all input levels	• Decreased SNR for noises occurring within the gaps of speech • Increased chance of feedback • Intensity cues may be disrupted (if compression is fast acting)
Noise reduction	Slow- or fast-acting compression in low-frequency band, or adaptive high pass filter (frequency response typically steeper at high input levels)	• Less masking and/or annoyance by low-frequency noise • Signal kept in audible range without using volume control for a wider range of overall levels (and for soft and weak phonemes if compression is fast acting)	• Signal attenuated as well as noise • Abnormal tonal balance • Intensity cues may be disrupted (if compression is fast acting) • Variation of the signal quality as the noise spectrum varies may be objectionable

ent for compression than for linear amplification, as one would expect given that compression increases audibility but reduces temporal and/or spectral contrasts.[476, 812] For example, multichannel compression makes it easier to identify the manner of articulation of consonant (e.g. plosives versus fricatives) but harder to identify the place of articulation.[1952]

3. Low-level inputs. However, as soon as the input sound is decreased (perhaps someone with a softer voice, or someone more distant, starts talking), any form of compression with a compression threshold less than the original input level can provide intelligibility superior to that of the linear aid.[396, 773, 811, 812] This occurs because of the greater gain and hence audibility provided by the compression aid for low-level inputs (see Figure 6.14).

4. High-level inputs. If the input level is then increased above the original level (and above the compression threshold), any form of compression will increase listening comfort.[397, 810, 845, 1341] Compression *may* also increase speech intelligibility because both excessively high presentation levels and peak clipping are harmful to intelligibility (see also Section 10.7.2).[811] The common use of WDRC has resulted in aid wearers complaining less about hearing aids amplifying environmental sounds to levels louder than they like.[834] All these advantages occur because of the lower gain (for high-level inputs) provided by the compression aid (see Figure 6.14).

These benefits (items 3 and 4) considerably decrease the need for a manual volume control, although they will not eliminate that need for all patients. People who have trouble manipulating a volume control will particularly appreciate the benefits. Both fast and slow acting compression provide these benefits. If the compression is fast enough (attack and release times less than about 1 s) the compressor will automatically provide the lower gains that are preferred whenever the aid wearer speaks.[1010] Similarly, if the compression is sufficiently fast acting, the hearing aid will be able to decrease the rapid and large variations in level that can occur in some music.[k] With linear amplification, music will often be too soft or too loud.

There is, however, a price to pay for these benefits of compression. Because the hearing aid is often unable to tell the difference between a weak sound that is wanted and a weak sound that is unwanted, it will turn up the gain whenever the sound remains weak long enough for the compressor to react. If this weak sound is actually background noise, the compression aid will sound noisier than the linear aid. (This disadvantage may be avoided if the hearing aid is successfully able to distinguish between wanted and unwanted sounds based on whether these sounds have speech-like characteristics.) In addition, whenever the compressor automatically increases gain, the aid becomes more likely to feed back. The choice of compression threshold is considered further in Section 10.3.5. Lastly, if the compression is fast acting, it will lessen some of the natural intensity differences between sounds. If the hearing-impaired person uses these intensity differences to differentiate sounds, the compression may make this task harder.

Intriguingly, for a small proportion of patients the disadvantages of compression seem to outweigh their advantages, as patients prefer and perform better with linear amplification.[598] These patients are most likely to have flat losses, wide dynamic ranges, and use their hearing aids in a restricted range of auditory environments.[598]

The degree of compression (i.e. how much the gain changes as the input level changes) increases as compression ratio increases and as compression threshold decreases. It is clear that compression has both beneficial and adverse effects. Consequently, it is not surprising that the balance of beneficial to adverse effects is maximized if the amount of compression is neither too small nor too large. If the compression ratio is too large, then sound quality and intelligibility both decrease, particularly when there is significant background noise.[149, 1314, 1527] As compression ratio increases, the non-simultaneous SNR within each compression channel (provided it is positive) is made progressively worse, as explained in the panel in Section 6.3.7. Though compression may thus degrade overall quality the most when noise is present, compression actually has the least physical effect on speech itself when noise is present. Noise reduces the dynamic range of the overall signal and hence reduces the amount by which gain changes, and hence the amount by which the speech is changed by the compressor.

[k] Classical music usually has a much bigger dynamic range than pop music, so there is a greater need for WDRC when listening to such music.

Compression is valuable in hearing aids because the dynamic range of hearing reduces as the degree of sensorineural hearing loss increases. One might therefore expect that those with the most loss and the least dynamic range would benefit the most from compression and would require the highest compression ratios, but this is not the case, at least for fast-acting compression. Patients with the greatest loss also have the most degraded frequency resolution (Section 1.1.3) and this makes them rely strongly on envelope temporal cues.[1525] It is these cues that are most strongly affected by fast acting compression. Accordingly, a number of studies have shown that for patients with severe or profound loss, less rather than more compression, or even no compression at all produces the highest intelligibility or sound quality.[82, 157, 410, 878, 1679] Even within the mild to moderate range of hearing loss, the benefit from fast-acting compression is paradoxically greatest for those with the least loss.[1362] Presumably, those with least loss have sufficient spectral resolution to take advantage of the increased audibility offered by fast compression, and are less reliant on the intensity cues that are degraded by fast compression.

Laboratory studies performed with all experimental conditions adjusted to the most comfortable listening level under-estimate the value of compression, however, as there is then little that compression can add, but much that it can take away. The value of compression in real-life is thus greater than such studies would imply, and is best assessed from real-life trials or from experiments in which the test stimuli are presented across a wide range of levels.

Given the undoubted negative effects of fast-acting, multi-channel compression with high compression ratios, what should be provided to patients with severe loss given their limited dynamic range, and their unquestioned need for good audibility across a range of listening environments? It would seem that unless the fitting includes at least some compression with time constants longer than typical syllables, these patients will frequently have to adjust their volume control, or will have to accept the adverse consequences of excessive fast-acting compression in order to obtain the benefits of compression. Initial research on combinations of fast- and slow-acting compression is positive,[1556] but more research in different listening situations is needed.

Whether in a research study, or in deciding whether to vary the compression characteristics for a particular patient, consideration must be given to the time taken (at least one month) for patients to acclimatize to any form of compression that is new to them.[878, 1009] This should be a consideration for all patients, but may be particularly important for patients with severe hearing loss.

6.5.2 The benefits of multichannel relative to single-channel compression

Relative to single-channel compression, multichannel compression can increase intelligibility because it increases the audibility of speech. (See the rationales in Section 6.3 for reasons why.) Unfortunately, fast-acting multichannel compression also decreases some of the essential differences between different phonemes. Multichannel compression, like single-channel compression, flattens the envelope across time, but does it more effectively because the flattening occurs independently in each of several frequency regions. Because compressors give less amplification to intense signals than to weak signals, fast-acting multichannel compressors also decrease the height of spectral peaks and raise the floor of spectral valleys. That is, unlike single-channel compression, they partially flatten spectral shapes. Spectral peaks and valleys give speech sounds much of their identity. This ***spectral flattening*** makes it harder for the aid wearer to identify the place of articulation of consonants,[410, 1068, 1070] and so offsets the positive effect of increased audibility.

Fast-acting single-channel compression also has disadvantages relative to multichannel compression. Most obviously, gain variations produced by the compressor affect all frequencies by the same amount, and this gain is largely determined by the strongest frequency components present in the combined speech and noise at any instant. Weak components that might already have little or no audibility can be attenuated just because a strong component is present at some other frequency.

Another obvious limitation is that for sloping hearing losses, the amount of compression has to be some compromise between that needed for frequencies where there is not much loss, and that needed for frequencies with more severe hearing loss. Less obviously, compression changes the amplitude of the background noise by just the same amount as it changes the amplitude of the speech signal, and this change is the same at all frequencies. Single-channel compression can thus cause signal and noise to have modulations in common (referred to as ***co-modulation***) and for these modulations to be consistent across frequency, which makes it harder for the aid wearer to distinguish speech from noise.[1724]

Considering that multichannel compression has both advantages and disadvantages, it is not surprising that some experiments have shown multichannel compression to be better than single-channel compression[907, 1225, 1226] and some have failed to show any advantage for multichannel compression.[1240, 1437, 1887] Multichannel compression decreases speech intelligibility for normal-hearing people, because it does not give them any increase in audibility, and hence there are no advantages to offset its negative effects.[477, 749, 1951] If high compression ratios (greater than 3:1) are used in a fast-acting multichannel compression aid, intelligibility is also decreased for hearing-impaired listeners.[206, 410, 478, 1437]

In the extreme case of many channels, short time constants, low compression thresholds, and infinite compression ratios, all sounds would have the same spectrum at the output, no matter how they differed at the input. Although no one would prescribe such compression, for patients with severe losses, and hence greatly reduced dynamic ranges, restoring normal audibility requires large compression ratios. If multichannel compression is applied to these patients in this way, it is detrimental to intelligibility.[478, 1070, 1951] For smaller compression ratios, the detrimental effects are smaller, so the positive effects (increased audibility) of increasing the number of channels can slightly outweigh the negative effects (spectral flattening).[1952]

Whether the positive effects of multiple channels of compression outweigh the negative effects depends on how much audibility is achieved in the reference condition. A net advantage for multichannel compression is thus least likely for sounds that in the single-channel condition are comfortably loud and have been amplified by an appropriate gain-frequency response shape. An advantage for multichannel compression over single-channel compression is most likely for very low and very high input levels, but these conditions have not been adequately investigated. (Multichannel compression enables different gain-frequency responses to be achieved at low versus high levels, thus increasing the range of sound levels over which good audibility and comfort can simultaneously be achieved, without using excessively high compression ratios.)

The advantages are, however, unlikely to be large for most patients. One extensive laboratory and field study[881] found that, overall, subjects slightly preferred single-channel compression to multichannel compression. For those with steeply sloping loss, two-channel compression was preferred to single-channel compression in real life. For all subject groups, speech scores obtained in the laboratory were insignificantly different for 1-, 2- and 4-channel systems. An overwhelming conclusion was that for most subjects, the choice of number of channels of compression was not an important issue.

There are many reasons other than the use of multichannel compression for choosing a multichannel hearing aid:

- A multichannel structure enables the gain-frequency response to be most easily and flexibly controlled.
- Effective strategies for noise suppression rely on being able to control the gain independently in different frequency regions. The more channels that are available, the better that noise with strong narrow-band components can be suppressed without significantly affecting speech quality and intelligibility.
- The pumping effects that can accompany expansion to suppress low-level noise are minimized with multichannel expansion.[1957]
- Some schemes for feedback management are based on a multichannel compression structure (see Section 8.2.1).

Overall, it seems probable that, for most patients, multichannel compression will not have marked advantages or disadvantages compared to single-channel compression. Overall, there are no strong reasons *not* to use multichannel compression provided compression ratios greater than 3:1 are avoided if the compression is fast-acting. A significant intelligibility advantage is more likely to be found for patients with a steeply sloping hearing loss. It may turn out that for other patients there will be intelligibility advantages at very low or high input levels, but further research is required before this could be confidently asserted. The full advantages of multichannel compression may not emerge until the aid wearer has had considerable listening experience.[1949]

Note that the discussion in this section compares multichannel versus single-channel compression. For low input levels, both multi- and single-channel compression offer intelligibility substantially greater than that available from linear amplification. For high input levels they both result in greater comfort.[1070]

6.5.3 Slow versus fast compression

Averaged across patients and environments, sound quality *may* be maximized by slow-acting compression, [597, 682, 1313] whereas intelligibility *may* be maximized by fast-acting compression.[597, 1846]

However, variations from this tentative and simple overall summary abound. The importance of release time, and the release time considered to give the best sound quality, vary with the sound being amplified, and from patient to patient.[356, 1315, 1846]

Some studies find that patients with low cognitive ability tend to obtain better speech intelligibility from slow-acting compression than from fast-acting compression, whereas patients with high cognitive ability tend to obtain better speech intelligibility from fast-acting compression.[598, 1086, 1747] Other research supports just the result for high-cognition patients[1538] whereas yet further research suggests that for patients with low cognitive ability, the optimal release time measured in laboratory tests depends on the degree of context within the speech material, and hence on the need to hear every word to understand the meaning.[356] No relationship was found, however, between cognition and preference for compression speed in real-life field trials.[356] Because there are considerable individual differences in the effect of compression on intelligibility, experiments with small numbers of subjects are likely to conclude that release time has no consistent significant effect on intelligibility.[1241, 1352]

The different time constants should, in principle, have the greatest effect when the noise fluctuates in level. Indeed, some research shows that the greater the degree of fluctuations in the background noise, the greater the benefit that fast-acting compression provides over slow-acting compression.[596] Other research finds no impact of modulations in the noise. [1,356] Fast-acting compression provides greater gain, and hence audibility, during gaps in background noise, which patients with higher cognitive ability can take advantage of.[596] Possibly those with lower cognitive ability are less able to distinguish lower level noise from lower level signal, both of which are amplified more by fast-acting compression than by slow-acting compression, or possibly they are less able to infer meaning from brief auditory glimpses of the speech.

The only commonalies between all the experiments in this area are that:

- some patients prefer and/or perform better with fast compression whereas others prefer and/or perform better with slow compression; and
- patients with high cognitive ability, especially when based on measures that require good working memory, perform better than those with low cognitive ability.[356, 1538]

Given the conflicting results, more research is needed on how preference is affected by cognition (measured in different ways) and other factors such as ability to use the fine structure of speech, the SNR, the degree of context in the speech, and the extent of listening experience with each type of compression.[356, 559, 1216, 1538] For the moment it is not possible with certainty to prescribe the optimum compression speed for a patient. Perhaps the speed, and the way it varies with the acoustic environment should be something that patients choose for themselves by training their hearing aids to their individual preferences (Section 8.5).

Slow versus fast need not be an either-or choice since it is possible to have multiple compressors in the same hearing aid, as described in Section 6.4.[1728] It seems likely that a combination of the two is optimal for all patients, but with the amount of fast-acting compression relative to slow-acting compression higher for hearing aid wearers in background noise that fluctuates markedly in level, in listening situations where speech varies rapidly in level, and perhaps for higher cognition patients. The reason for using both types of compression is that the combination enables audibility and loudness comfort to be maintained over a wide range of input levels without suffering the disadvantages that would occur if this was accomplished entirely by either method alone.

[1] Comparison of experimental results is confounded by differences in the SNR used in different experiments: Modulations in the noise have more effect on compression when the speech reception threshold occurs at very poor SNRs, which occur for easy speech material, whereas modulations in the signal have more effect on compression when speech reception threshold occurs at more positive SNRs, which occurs for harder, lower-context speech material.[1086]

6.6 Concluding Comments

We have long recognized the beneficial effects of compression on audibility and comfort once signal level departs from the typical input level for which linear amplification is optimally suited. Our knowledge of how best to apply compression has advanced considerably, however, in the last decade.

We now appreciate the increased likelihood that fast-acting compression will be more beneficial than slow-acting compression when high-cognition patients listen to speech in quiet or in noise with significant gaps.

We now better understand the negative effects of fast-acting compression in amplifying noise during gaps in the speech, in reducing or distorting envelope cues, and in reducing spectral cues when implemented in multi-channel systems. All of these changes decrease sound quality, with progressively worse effects as compression ratio is increased. None of these problems occur with very slow-acting compression, though simple slow-acting systems introduce their own problems by being too slow to react when the listening environment changes. The negative effect of compression (fast or slow) at low levels in inducing feedback oscillation has become less of an issue with the widespread availability of feedback cancellation, but is still a relevant consideration.

The benefit of compression, whether assessed by intelligibility or by quality measures, is unquestionably positive, both on average and for most patients, but is far from dramatic.[397, 597] While the advantages are worth having, the differences between compression versus linear hearing aids (provided the latter have a volume control) are small compared to the benefit that any hearing aid provides relative to unaided listening.[775, 1026, 1318, 1607]

Despite these advances, further research into compression is still needed. We need clever combinations of fast- and slow-acting compression that, by distinguishing speech from non-speech sounds, capture the advantages of each and the disadvantages of neither. We need evaluations of the benefits of such systems for a variety of patients under real-life, or simulated real-life conditions. Unfortunately, if the listening environment is highly controlled (or indeed, contrived) almost any amplification scheme can be made to look superior to another scheme by a judicious choice of the experimental conditions.

Compression has achieved a well-deserved place in hearing aids. The default hearing aid fitting should have fast-acting compression limiting, and wide dynamic range compression implemented with fast, slow, or preferably adaptive or dual time constants, and a compression ratio appropriate to the degree of hearing loss at each frequency.

CHAPTER 7

DIRECTIONAL MICROPHONES AND ARRAYS

Synopsis

Other than the use of a remote microphone located near the source, directional microphones (which work by sensing sound at two or more locations in space) are the most effective way to improve intelligibility in noisy environments.

Directivity is most commonly achieved in hearing aids with first-order subtractive directional microphones, in which the output of one omni-directional microphone is delayed and subtracted from the output of the other. This internal delay, relative to the physical spacing between the two microphone sound ports, largely determines the polar sensitivity pattern of these microphones. The head itself also affects the polar pattern. These subtractive directional microphones inherently cause a low-frequency cut in the frequency response, for which the hearing aid signal processing often compensates by a low-boost characteristic, but which also causes greater internal noise in the hearing aid. Split-band directivity, which combines a directional response for the high frequencies with an omni-directional response for the low frequencies, avoids this problem, but of course provides no noise reduction for the low frequencies. Irrespective of the frequency range over which the microphone is directional, the complete hearing aid fitting will have directivity only over the frequency range for which the gain of the amplified sound path exceeds that of the vent sound path. In open fittings, this will likely be only half the speech frequency range. Whether achieved by split-band processing, or by the acoustics of an open fitting, the resulting pattern of high-frequency directivity and low-frequency omni-directional processing simulates the directivity pattern of normal hearing.

Additive directional arrays create directivity by adding together the output of two or more omni-directional microphones. They do not create additional internal noise. To be effective, however, the microphones have to be separated by distances larger than a quarter of the sound's wavelength. They are therefore less suitable for hearing aids, but are suitable for accessories such as hand-held microphones.

These simple fixed subtractive and additive arrays have a fixed pattern of sensitivity versus direction of the incoming sound. Adaptive arrays, by contrast, have directional patterns that vary depending on the location, relative to the aid wearer, of background noises. Adaptive arrays automatically alter the way they combine the signals picked up by two or more microphones so as to have minimum sensitivity for sounds coming from the direction of dominant nearby noise sources. The multiple microphones that provide the input signals can be mounted on one side of the head or on both sides of the head.

The most sophisticated directional microphone arrays apply complex, frequency-dependent, adaptive-weights to the outputs of each omni microphone before combining them. Like all directional microphone arrays, complex adaptive arrays work most effectively in situations where there is a low level of reverberant sound.

Directional microphones are effective when either the target speech or the dominant (rearward) noise source(s) are closer to the aid wearer than the room's critical distance (at which the reverberant and direct sound fields have equal intensity). In the special case of a close frontal talker and many distant noise sources, the improvement in SNR will approximate the directivity index of the hearing aid averaged across frequency.

The disadvantages of directional microphones include insensitivity to wanted sounds from the sides or rear, increased internal noise if used in quiet places, reduced localization accuracy if the two hearing aids act in an uncoordinated manner, and increased sensitivity to wind noise. These disadvantages can be minimized by intelligent switching (automatically or manually) between directional and omni-directional modes, on the basis of noise levels and apparent SNR at the output of the omni and directional microphones.

All hearing aid wearers are candidates for directional microphones because all hearing aid wearers need a better SNR than people with normal hearing.

Adaptive directional arrays that combine (via a wireless link or a cable) the outputs of microphones on both sides of the head produce a super-directional response that should enable people with mild hearing loss to hear better than people with normal hearing in many social situations.

There are only two proven ways of increasing intelligibility above that obtainable with an appropriately adjusted conventional hearing aid delivering sound at a comfortable level. One way is to move the hearing aid microphone (or some auxiliary microphone) closer to the source. This increases the level of direct sound compared to reverberant sound and background noise, as discussed in Section 3.4. Unfortunately, moving closer to the source, or positioning a remote microphone near the source, is not always practical.

The other proven solution is to use some type of directional microphone. Directional microphones can be constructed from a single microphone with two entry ports or by combining the electrical outputs from two or more microphones, as explained in Section 2.2.4. A microphone or group of microphones with more than one entry port may be referred to as a ***directional microphone***, a ***microphone array*****,** a ***beamforming array***, or a ***beamformer***.

The first section of this chapter describes directional technology, the second how directivity is quantified, and the third describes the benefit for patients and the factors that affect the degree of benefit received. These sections depend on each other for a full understanding, so it may be necessary to read this chapter twice.

7.1 Directional Microphone Technology

7.1.1 First-order subtractive directional microphones

The directional microphones widely available in hearing aids are ***first-order subtractive*** directional microphones. This name is given because the output depends on a single subtraction of two signals. The block diagram of a first-order subtractive directional microphone is shown in Figure 7.1. With acoustic directional microphones, the subtraction occurs mechanically as sounds from each port press on opposite sides of the diaphragm, as shown in Figure 2.6. When two separate, omni-directional microphones are used, the electrical outputs of each microphone are subtracted after the output of the rearward microphone has been electrically delayed. The mechanism by which this delay-and-subtract process produces more sensitivity in the forward direction than in any other direction has already been covered in Section 2.2.4, which should be reviewed before proceeding further in this chapter. Whether mechanical or electronic subtraction is used, simple first-order subtractive directional microphones are linear devices that do not distort the signals they detect.

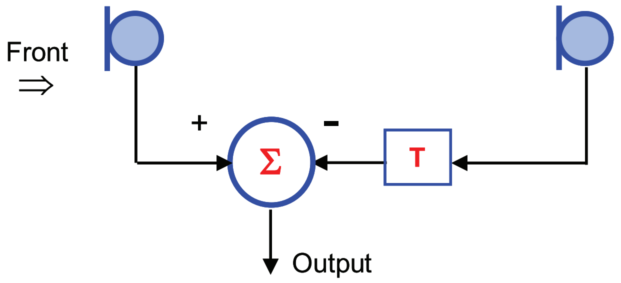


Figure 7.1 Block diagram of a subtractive directional microphone comprised of either a single microphone with two ports, or two separate microphones with one port each. The negative sign next to one of the inputs of the summer indicates that the two signals are subtracted.

Port spacing, internal delay and polar pattern

There are two design parameters that between them determine everything about the polar pattern and the gain-frequency response: the port spacing and the internal delay. The ***external delay*** (which sounds coming from directly in front or directly from the rear undergo) is calculated by dividing the port spacing by the speed of sound. The ***internal delay*** is the delay that is integral to a low-pass filter (for the acoustic type of directional microphone, see Figure 2.6) or it can be an electronic delay (for the ***dual-omni*** type of directional microphone).

The ratio of internal delay divided by the external delay, which we will call the ***delay ratio***, determines the shape of the ***sensitivity pattern***, also called the ***polar directivity pattern***, as shown in Figure 7.2. As the delay ratio decreases from 1.0 to 0, the shape moves from a ***cardioid*** through a ***super-cardioid*** to a ***hyper-cardioid*** and then to a ***figure-8***, and the sensitivity to sounds from the back grows (the secondary ***sensitivity lobe*** to the rear), but the sensitivity to sounds from the sides diminish. In the extreme case, which is referred to as a ***figure-8*** or ***bi-directional*** pattern, the front and rear lobes have the same sensitivity, but the microphone is completely insensitive to sounds coming from the sides.[a]

[a] If the aim is to make a forward-looking directional microphone that suppresses sounds from rearward directions, then delay ratios greater than 1 are useless, as there is no rearward direction for which the microphone is very insensitive.

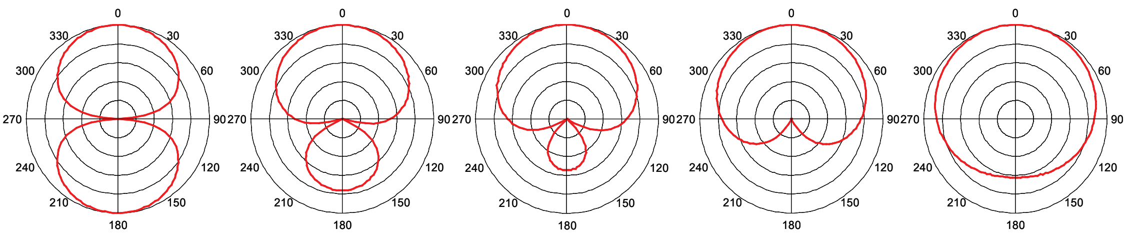
The directivity index (***DI***), introduced in Section 2.2.4 and explained in more detail in Section 7.2.1, quantifies frontal sensitivity relative to average sensitivity and is fundamentally important to our understanding of how much benefit directional microphones can provide. The unidirectional index (***UI***) is similar, but less useful; it describes the sensitivity averaged across all frontal directions (clockwise from 270° to 90°) relative to the sensitivity averaged across all rearward directions (clockwise from 90° to 270°).

Figure 7.2 shows the DI and UI measured in two and three spatial dimensions for each polar pattern, all measured in unobstructed space. For cardioids of different shapes, the highest three-dimensional DI (equal to 6.0 dB) is obtained with the hyper-cardioid. The highest two-dimensional DI (equal to 4.8 dB) is obtained for a pattern close to the super-cardioid. The highest 3D UI (11.6 dB) is obtained with a super-cardioid.

You will notice that Figure 7.2 does *not* show the front-back ratio, which as the name suggests is the ratio (or difference in decibels) of the response at 0° relative to the response at 180°. Although this figure is much quoted, it is almost meaningless, because the response at 180° is greatly affected by small changes of hearing aid position on the head, frequency, and directional pattern, and is therefore very unrepresentative of performance when noise comes from a range of rearward directions.[1499]

Each of these directional sensitivity patterns has a shape that does not vary with frequency until the frequency gets so high that the distance between the ports approaches half a wavelength. For a cardioid pattern for example, when the frequency equals the speed of sound divided by twice the port spacing, the microphone has zero sensitivity for sounds from the *front*. Consequently, the port spacing has to be kept less than about 12 mm if the pattern is to stay well behaved up to 8 kHz.

The port spacing cannot be made too small, however, or the microphone itself becomes too noisy. Recall that the microphone works by subtracting the pressures sensed at the two ports. The magnitude of this difference depends on the phase difference between the sound at the two ports, and hence how large the port spacing is compared to the wavelength of the sound wave (and on the level of the sound wave, of course). Small port spacings therefore decrease the microphone's sensitivity, but the internal noise generated by the microphone remains the same, and so becomes increasingly apparent by comparison with the signal. This will be evident to the aid wearer as the directional microphone sounding *noisier* in quiet situations if the two microphones are equalized to have the same gain-frequency response for frontal signals.



2D DI	3.0	4.6	4.8	4.3	3.0
3D DI	4.8	6.0	5.7	4.8	3.2
2D UI	0.0	7.4	12.1	11.0	5.6
3D UI	0.0	8.5	11.6	8.6	4.4
Delay ratio	0.000	0.333	0.577	1.00	2.00
Name	Figure 8	Hyper-cardioid	Super-cardioid	Cardioid	Useless

Figure 7.2 Theoretical polar sensitivity patterns of subtractive first-order directional microphones when not mounted on the head for various values of the ratio of internal delay to external delay. The table shows the 2D and 3D DIs, and the 2D and 3D unidirectional index, for each pattern. Each concentric circle represents a 5 dB change in sensitivity.

Also, for dual-omni directional microphones, mismatch errors (see below) become an increasingly greater problem as the port spacing decreases. As the mismatch errors and high internal noise caused by small port spacing are greatest at low frequencies, where the wavelength is largest, very small port spacings (like 5 mm) are most suitable for high-frequency hearing aids, such as those intended for open-canal fittings. The high internal noise associated with small port spacings is not a major problem in hearing aids that automatically switch to an omni-directional response in quiet listening situations, which most now do.

The neat polar patterns shown in Figure 7.2 occur only for hearing aids measured in isolation. Because the head and pinnae are acoustic barriers, they create directionality, even for an omni-directional microphone. Figure 7.3 shows the polar pattern for an omni-directional microphone mounted in a BTE placed behind the right ear of KEMAR. The hearing aid is maximally sensitive to sounds about 70° to the right of frontal. The polar pattern for a directional microphone with a free-air cardioid response, when placed behind the right ear of KEMAR is also shown in Figure 7.3. The on-head directivity pattern (solid line) reflects both the directivity of the microphone in free space (Figure 7.2; cardioid; attenuation of sounds from the rear) and the directivity of the head (Figure 7.3; dashed line; attenuation of sounds from the left).[b] The most sensitive direction is now about 40° to the right of frontal.

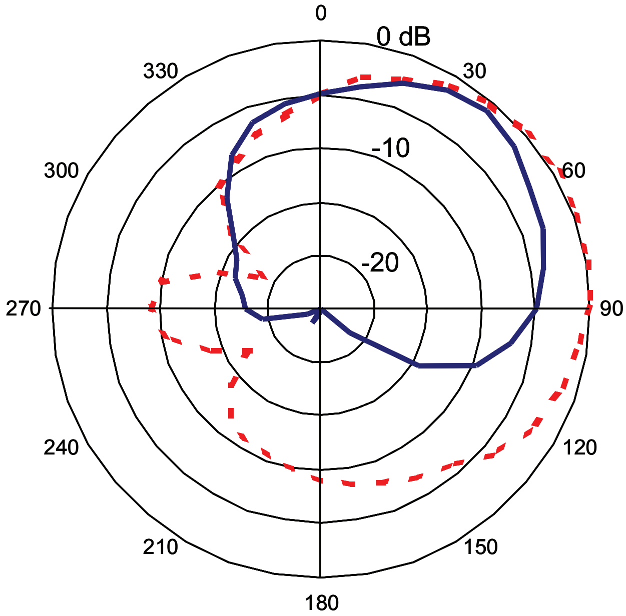


Figure 7.3 Polar directivity pattern, measured at 2 kHz, of an omni-directional microphone on the head (dashed line), and a cardioid directional microphone on the head (solid line), both in a BTE hearing aid. Data adapted from Knowles TB21. Each concentric circle represents a 5 dB change in sensitivity.

Because the head becomes more directional as frequency rises, the directivity pattern of directional hearing aids also varies with frequency when mounted on the head. Patterns are therefore needed for each frequency, or else a single pattern representing an average across frequency can be used. As was shown in Figure 2.8, first-order subtractive directional microphones typically have a 2D directivity index around 4 dB. When mounted on the head, 3D DI values are 1 to 3 dB less than the theoretical free-field values shown in Figure 7.2.[468]

The head, combined with the design of the hearing aid, can have another effect. It is common for the line joining the ports of a BTE to point upwards towards the front, which means that the direction of maximum sensitivity is also above the horizontal. Because first-order microphones are not exceptionally directional, this has little consequence provided the angle is less than 20° or perhaps even 30° (see Figure 7.2). The DI is, however, decreased if the angle of the BTE causes one of the ports to be obscured by the pinna.[1498] BTEs, and particularly micro-BTEs, should be positioned as far forward on the pinna as is comfortable for the patient to reduce the likelihood of producing an inappropriate directional pattern on the head.

Directional microphones can be used only in hearing aids large enough to accept the necessary port spacing. Currently, they are mostly used in BTE and ITE hearing aids. They are never likely to be effective in CIC aids, partly because there is not room to achieve the necessary port spacing, and partly because diffraction by the pinna creates a complex sound field near the faceplate of the hearing aid.

[b] The directivity pattern on the head can be approximately predicted by adding (in dB) the omni-directional pattern of the head position to the free-air pattern of the directional microphone.

Frequency response

Because wavelength progressively lengthens as frequency decreases, a fixed port spacing represents a smaller and smaller fraction of a wavelength as frequency decreases. Consequently, the sound pressure at the two ports at any instant becomes more similar, so the difference between the front and back port signals decreases. A subtractive directional microphone therefore has a low-frequency cut of 6 dB per octave in the gain-frequency response. An electronic filter can be used to boost the low-frequency gain and so compensate for this, but such a filter also boosts the internal microphone noise, which may then become audible and annoying. To avoid excessive noise, it is common for hearing aids to compensate only partially for the low-frequency cut. Even full compensation, such that the directional and omni-directional patterns have the same sensitivity for frontal sounds, will leave the directional pattern sounding quieter in most listening situations, as it will still have less sensitivity for sounds arriving from most other directions.

Compensation is most important for patients with more than 40 dB loss in the low frequencies, [1501] who rely on amplified low-frequency sound, and is least important for high-frequency, open-canal hearing aids where none of the low frequency sound arrives via the hearing aid anyway (Section 5.3.1).

Microphone sensitivity is maximized, and internal noise minimized, by using a large port spacing, but if this is made too large the frequency response for frontal sounds is adversely affected. Figure 7.4 shows the frequency response for microphones with a 20 mm and an 8 mm port spacing. The minimum in the high-frequency response occurs at 9 kHz for the larger port spacing, but at a frequency far above the bandwidth of the hearing aid for the smaller port spacing. It is also evident that the 8 mm port spacing has a lower sensitivity over most of the frequency range, as expected.

If the hearing aid has an excessively large port spacing, combining the delay with a low-pass filter will solve the problem of decreased frontal sensitivity at high frequencies, just as occurs within an acoustic directional microphone (as shown in Figure 2.6). This filtering causes the rear port (or microphone) to effectively close (or turn off) at high frequencies. There is then no problem with high-frequency frontal sensitivity, but neither is the microphone directional at these frequencies.

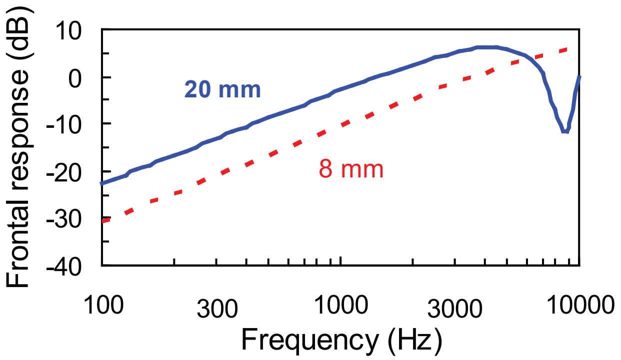


Figure 7.4 Frontal sensitivity of a two-port (or two-microphone) subtractive directional microphone relative to the sensitivity of an equivalent single-port microphone. The parameter shown is the port spacing. The internal delay needed to produce a cardioid polar response has been assumed.

Acoustic versus electronic subtraction

Hearing aids with directional microphones nearly always enable an omni-directional response to be selected, either by incorporating both a directional microphone and an omni-directional microphone, or by using two omni-directional microphones, the outputs of which are combined electronically.

This dual-omni approach has the advantage of enabling more sophisticated adaptive directionality (Section 7.1.3). It has the disadvantage that good directivity depends on the gain-frequency response of the two microphones being very well-matched to each other in gain and phase.

While it is straightforward for manufacturers to select pairs of well-matched microphones, the responses of the microphones change after manufacture, caused by the combined effects of age, humidity, temperature, vibration, and accumulation of debris in the microphone ports or in the pores of protective screens covering the microphone ports.[c, 305] The resulting mismatch in gain-frequency response causes a progressive deterioration in their matching, and hence directivity. For example, a sensitivity mismatch as small as 1 dB, or a phase mismatch as small as 5 degrees can decrease

c Protective screens may have to be changed regularly to prevent the sensitivity of the microphones they protect being adversely affected by accumulated fine debris.

the directivity index by one third at 1 kHz, with progressively much worse effects as frequency decreases below 1 kHz.[d]

The mismatch problem is minimized electronically in many hearing aids by the hearing aid continuously comparing the long-term average output of the two microphones. Given the close proximity of the two microphones, any difference in long-term output levels must be due to a difference in the microphones, so the output of one microphone can be corrected electronically prior to subtracting the two outputs, as shown in Figure 7.5. Hearing aids vary in the sophistication with which they make this correction.[305] The ideal is for the gain and phase to be corrected at all frequencies. Some older hearing aids make no correction at all.

Combining directional and omni-directional responses

Microphone mismatch also offers opportunities. Figure 7.6 shows how the free-air sensitivity pattern of the hyper-cardioid microphone changes as the relative sensitivity of the two constituent microphones varies at any low or mid frequency. Only a very small imbalance between the microphones removes the deep ***nulls*** (i.e. angles for which there is a larger attenuation), and further imbalance rapidly moves the pattern towards an omni-directional pattern. Intentional imbalances can therefore be used to smoothly change from a directional pattern to an omni-directional pattern as the listening situation varies. Several hearing aids automatically make such a smooth transition over a duration of several seconds to avoid the discontinuities in sound quality that would accompany a sudden switch from omni to directional or vice versa.

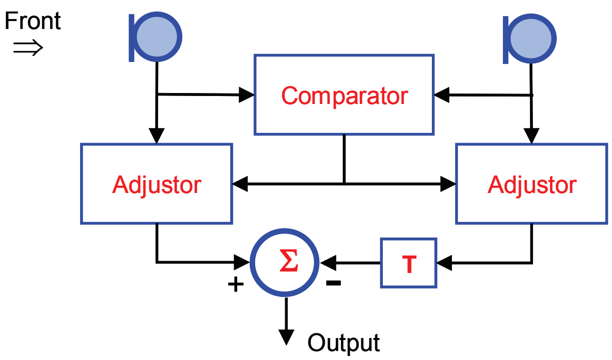


Figure 7.5 Block diagram of a subtractive directional microphone that automatically equalises the sensitivities of the individual omni-directional microphones.

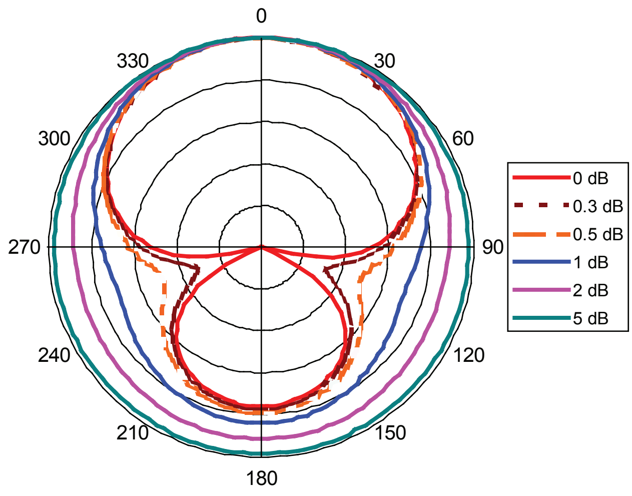


Figure 7.6 Variation of polar pattern with various degrees of sensitivity imbalance in the two microphones. Concentric circles differ in sensitivity by 5 dB.

Whether achieved by a smooth change, or by a sudden switch, it is common for hearing aids to automatically select a directional response in some situations and an omni-directional response in others. Although a hearing aid can never predict with absolute certainty what the wearer wants to listen to, if there is dominant talker in one direction, and a variety of sounds from other directions, it will likely be the dominant talker that will be the focus of the aid wearer's attention. The hearing aid can simultaneously process both a directional and omni-directional signal, and choose the one that has the higher apparent SNR. The types of information that hearing aids can take into account in determining that a directional response is most appropriate are:

- The overall sound level is high enough to indicate that voice levels are raised;
- The background sound level, measured during apparent gaps in the main signal, is greater than some amount, typically around 60 dB SPL, suggesting that there is noise in the environment, and also making it unlikely that the internal noise of a directional microphone will be audible;

[d] This example was calculated for the 3D DI of a hyper-cardioid response in free air with a port spacing of 10 mm.

- The output of the directional microphone has deeper envelope fluctuations, especially at the rates typical of speech (4 to 20 Hz) than the output of the omni-directional microphone, suggesting that there is a talker somewhere frontal of the aid wearer;
- The signal present is not characteristic of wind noise, as directional microphones are extremely sensitive to the chaotic, very localized sound fields caused by turbulence generated as wind passes the head, pinna, and hearing aid.[307] Wind noise can be automatically recognized by the low-frequency content at the two ports being poorly correlated, or, equivalently, by the low-frequency output of the (compensated) directional microphone being much greater than the low-frequency output of the omni-directional microphone.

As Figure 7.6 has already indicated, directional versus omni-directional is a graduated choice, not an all-or-nothing decision. In addition, some multi-channel hearing aids make the choice of directional versus omni-directional independently in different frequency channels. Frequently, the hearing aid will choose directional processing in the high-frequency channel(s) and omni-directional processing in the low-frequency channel(s) where the negative effects of internal noise and microphone mismatching are greater.

This frequency-dependent combination is referred to as ***split-band directivity***, or ***split-channel directivity***. It has the further advantage that the greater directivity of high frequencies compared to low frequencies mimics the directivity of the unaided human ear (Figure 2.8) and thus improves the ability of aid-wearers to localize sounds in the front-back direction.[883] Some manufacturers consider this mimicking such an advantage that the hearing aids preferentially adopt split-band directivity in many listening situations, or even make it permanently present. The disadvantage of split-channel directivity is that there is no increase in SNR for low-frequency sounds. All open-canal fittings with directional microphones effectively have split-channel directivity, irrespective of how the microphone senses low-frequency sounds.

There are some situations where the aid wearer wishes to hear a talker behind them, such as a driver listening to a back-seat passenger. In such situations, an omni-directional pattern will be better than a directional pattern, but a directional pattern pointing backward will be even better, at least for conversing with the passenger. This ***reverse cardioid***, or ***anti-cardioid*** response is easy to achieve: the signal from the front microphone is delayed and subtracted from the signal from the back microphone – the exact opposite of the frontwards-looking directional microphone shown in Figure 7.1. Some hearing aids now include these backward pointing patterns among those that they automatically select.

7.1.2 Additive directional arrays

The ***additive array*** works on a principle different from subtractive arrays. Instead of reducing sensitivity for sounds from all directions, but reducing it least from the front, the additive array produces the maximum possible sensitivity for sounds coming from the front, and less sensitivity for all other directions. Figure 7.7(a) shows one type of additive array that could be worn across the chest, on a headband across the top of the head,[647] or across the front of a spectacle frame. A simple, two-microphone array of this type is made when outputs from microphones on each side of the head are added, as in fact happens with a Bi-CROS hearing aid (Section 17.1.2).

When the target is directly in front of the aid wearer, then all microphones are equally distant from the

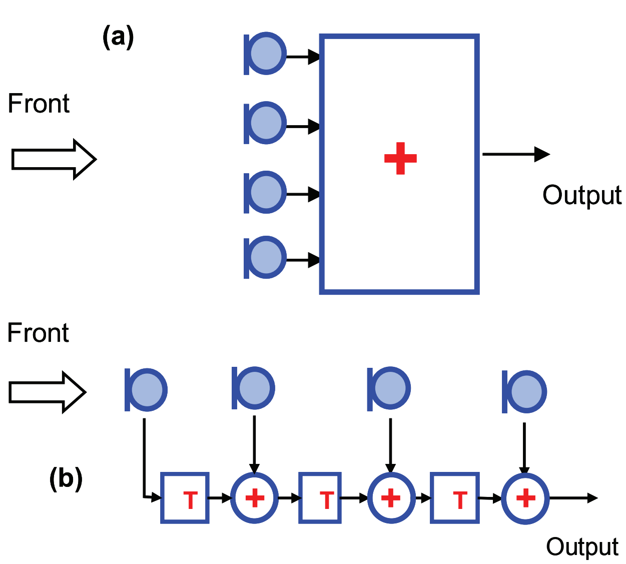


Figure 7.7 Block diagram of two additive directional arrays: (a) Broadside, and (b) End-fire, delay-and-add.

talker, and hence their output signals are in phase with each other. Simply adding the outputs together produces a signal that is *n* times larger than the output from any one microphone, where *n* is the number of microphones in the array. For any other direction in the horizontal plane, the sound wave reaches the microphones one after the other, causing phase differences between the output signals. Adding these phased outputs produces some degree of signal cancellation, such that the total output is less than for frontal signals, and so the array is directional. Giving some of the outputs more weight than others (rather than equally adding all outputs) will produce different directional patterns.[1672]

A different type of additive array, known as ***delay-and-add***, is shown in Figure 7.7(b). The output of each microphone is delayed by an amount *T* and is then added to the output from the next microphone in line. Consider what happens if the electrical delay *T* equals the time taken for sound to travel acoustically from one microphone port to the next. Sounds arriving from the front first reach microphone 1, and then continue on to microphone 2. The output from microphone 1, after being electrically delayed, reaches the adder at just the same time as the output from microphone 2 (whose input was acoustically delayed with respect to the first microphone). Consequently, the two signals combine perfectly in phase. The same process happens at the next adder, and then again at the next. The voltage of the final output is thus four times as great (corresponding to a 12 dB increase) as the voltage coming out of any one of the microphones.

But what happens if the sounds arrive from any other direction? When sounds arrive from the side, for example, they reach all four microphones simultaneously. Because of the electrical delays, the two signals going into each adder will no longer be in phase with each other. Consequently, they will not combine as constructively as do sounds from the front. The array must therefore be less sensitive for other directions than it is for sounds from the front.[e] Electronic delay-and-add processing is suitable for hand-held microphone accessories for hearing aids. The large boom microphones used in movie production are effectively delay-and-add directional processors, although achieved acoustically rather than electronically.

For either type of additive array, because signals from the frontal direction are always added together in phase, the array *decreases* internal microphone noise rather than *increasing* it as occurs for subtractive arrays. Small mismatches in microphone sensitivity have little effect on performance. Why then are they not more used?

Unfortunately, additive arrays are effective only for high-frequency sounds: i.e., those for which the length of the array is greater than, or comparable to, a quarter-wavelength. For sounds lower in frequency than this, all the microphone output signals will be approximately in phase no matter what direction the sound comes from. They are thus not effective for conventional styles of head-worn hearing aids. Even with a port spacing of 20 mm, for example, reasonable directivity would be achieved only for frequencies above 4 kHz.

Larger arrays are much more effective than first-order subtracting directional microphones, but are less cosmetically attractive, and so have been combined with headbands,[647] spectacles,[1672] and jewellery, to increase their acceptability. One novel arrangement combined several acoustic directional microphones into a delay-and-add array along the side-frames of a pair of spectacles.[1672, 1673] The five individual directional microphones, with their cardioid pattern, contributed directivity at low and mid frequencies and the delay-and-add array contributed directivity at mid and high frequencies. With a total array length of 100 mm the combination had a directivity index, averaged across frequency, of 7.5 dB. At a rate of 10% intelligibility per decibel of SNR improvement, this is sufficient to make the difference between understanding almost nothing and understanding almost everything (Section 7.3.1). A later version used individual microphones with a figure-8 rather than cardioid pattern. This narrowed the frontal lobe, which enabled just as high a DI, but with only 3 microphones in the array.[1094] The later version is available commercially as an accessory microphone that sends a magnetic signal to the hearing aid telecoil.

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e The array can be made even more directional by using electronic delays slightly longer than the corresponding acoustical delays between microphones. This is referred to as an ***oversteered array*** because as the internal delays are increased from zero, the direction of maximum sensitivity is *steered* from a direction perpendicular to the array through to a direction in line with the array.[334, 859]

End-fire and broadside arrays

Whether signals are combined in an additive or subtractive manner (or by filtering more complex than either of these methods), arrays can usefully be classified according to the physical arrangement of the microphones. Using terms borrowed from naval warfare, the microphones in an ***end-fire array*** are in a line pointing towards the direction of greatest sensitivity, whereas those in a ***broadside array*** are in a line perpendicular to the most sensitive direction. The simple additive array in Figure 7.7(a) is an example of a broadside array, whereas the delay-and-add array in Figure 7.7(b) and the subtractive directional microphone in Figure 7.1 are examples of end-fire arrays.

The directivity index of an array generally increases as the number of microphones and the array length increases. For a given number of microphones and array size, however, end-fire arrays are more directional than broadside arrays.[1690, f] The reason for this is easy to understand: The end-fire array has its maximum sensitivity in only one direction, whereas there are many directions in three-dimensional space for which a broad-side array is maximally sensitive (e.g. in front, above, below and behind).

7.1.3 Complex directional arrays

The arrays discussed so far are simple in two respects. First, the separate outputs are combined just by subtraction or addition, optionally with a delay. Second, they are ***fixed arrays,*** in that each array has a sensitivity pattern that never changes. This section describes several ways that arrays can be made more effective. ***Adaptive arrays*** change their directivity pattern in such a way as to minimize the pick-up of the dominant noise present at any time, and/or maximize the pick-up of signal coming from a target direction. Under most conditions, adaptive arrays are capable of greater noise suppression than fixed arrays.

Adaptive delay

The simplest form of adaptive array is the first-order subtractive array we have already met (Figure 7.1), but with the internal delay adaptively varied so that the array minimizes pick-up of sounds from the rear (Figure 7.8a). The internal delay, T, is automatically

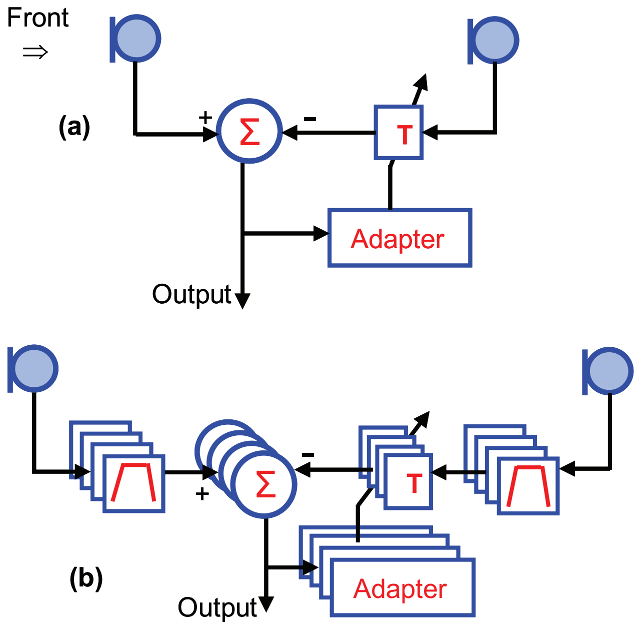

Figure 7.8 (a) A simple adaptive directional microphone with steerable nulls. (b) A multi-channel version where each microphone output is filtered into different frequency channels, enabling a different delay and polar pattern within each channel.

and continuously varied to the value that results in the least power in the output signal. Because the power of the output signal equals the power from the target (assumed to be in some frontal direction) plus the power from unwanted sounds (i.e. noise), the total power will be minimized when the power from the unwanted sound is minimized.

Examination of Figure 7.2 shows that as the internal delay and hence delay ratio varies, the direction of the sensitivity minima towards the rear also vary. In those cases where there is a single nearby interfering source in any rear direction (from 90 to 270°), the internal delay will adapt to the value that positions a minimum in the direction of this sound source.

Figure 7.9 shows an alternative way to achieve the family of polar patterns shown in Figure 7.2. This

[f] When arrays are placed on the head and body, their directivity is affected, and it will not necessarily be true that an end-fire array is superior to a broadside array of the same size.[649]

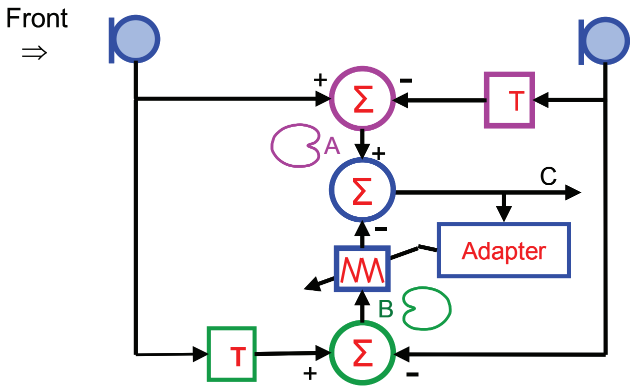

Figure 7.9 The Elko-Pong adaptive array, constructed from a forward-looking (signal A) and rearward-looking (signal B) cardioid, to produce an output (C) with polar pattern that varies in shape as the attenuation applied to signal B is varied.[515a]

algorithm, known as the Elko-Pong directional algorithm, has as its base the same forward-looking subtractive microphone shown in Figure 7.1.[515a] From this is subtracted an attenuated version of a rearward looking subtractive microphone. Each of the individual signals A and B has a cardioid pattern. Although it may not be obvious, as the attenuation of the rearward looking cardioid is varied from complete attenuation to no attenuation, the polar pattern of the final output C varies smoothly from a forward looking cardioid to a figure-8 pattern, including the intermediate patterns of super-cardioid and hyper-cardioid. The algorithm of Figure 7.9 tends to be used in hearing aids rather than that of Figure 7.8a because it is easier to automatically and smoothly vary an attenuation rather than it is to vary a time delay.

Some hearing aids are designed to adapt quickly (in around 10 ms), and some are designed to adapt slowly and smoothly (in a few seconds).[305] Quick adaptation allows the response to follow rapid head turns; slow adaptation avoids any disturbance of sound quality caused by rapid changes in the response. For any speed, the poorer the SNR, the more accurately will the adapter calculate the best value. If there is more than one nearby interfering noise, the adapter will come to some compromise value that attenuates both noises, but not as effectively as if each were the only noise present. An alternative adjustment strategy used in some hearing aids is to select the delay that results in the largest envelope modulation depth in the output signal, rather than the minimum power.[305]

One factor that limits the performance of simple adaptive arrays is that the neat, precisely positioned nulls shown in Figure 7.2 occur only for a directional microphone that is well away from all obstacles, such as a head. The head affects the polar pattern of the microphone (as shown in Figure 7.3) by different amounts at different frequencies, so the nulls occur at different angles for different frequencies. Consequently, it is not possible to simultaneously remove all frequency components of the noise. Solutions to this problem involve replacing the simple delay with a more complex circuit that delays different frequencies by different amounts. The version shown in Figure 7.8(b) filters each microphone output into four channels, and the adaptive process is carried out separately within each channel. In addition to allowing for frequency-dependent head diffraction, if different noise sources have their strongest power in different frequency regions, then each noise source can independently have its dominant frequency region attenuated without compromise.

Adaptive directional microphones remove noise more effectively than fixed directional microphones provided:

- there is a nearby noise source that dominates over other noise sources;
- the dominant noise maintains its direction long enough for the adaptive algorithm to lock-in on it;[111] and
- the dominant noise source is not, by coincidence, already aligned with the null in the directivity pattern of the fixed directional microphone.[1503]

The factor that most limits performance of adaptive-delay directional microphones is reverberation. If the noise source (even a single one) is considerably further away than the room's critical distance (Section 3.4), then noise power will reach the microphone evenly from all directions. All the adaptive algorithm can do is select the pattern that, averaged over all directions, most attenuates the noise. This is the hyper-cardioid pattern because that pattern has the highest 3D DI (Figure 7.2). Performance will thus be identical to that of a fixed hyper-cardioid directional microphone, and the SNR improvement will, at most, equal the DI of that pattern.

Directional microphone array terminology in review

- ***End-fire*** arrays have microphones arranged in a line that goes through the most sensitive direction, whereas ***broadside*** arrays are in a line that is perpendicular to the most sensitive direction.
- ***Additive*** arrays combine microphone outputs by addition, whereas ***subtractive*** arrays combine by subtraction.
- ***Fixed*** arrays have the same polar sensitivity pattern in all situations, whereas ***adaptive*** arrays have a sensitivity pattern that changes with the direction of the surrounding sources and, optionally, reverberation characteristics.

These three dichotomies are independent considerations except that broadside arrays are always additive.

Second-order arrays

If subtracting the outputs of two microphones can produce a directional pattern, then subtracting two signals, each of which is the output of a first-order subtractive directional microphone, can produce a super-directional pattern. The directional pattern of such a second-order subtractive process is shown in Figure 7.10. In this case, the first-order microphones had a cardioid response, producing a null at 180°, and the second-order subtraction had no additional delay, producing nulls at 90 and 270. This combination accounts for the good suppression around the entire rear hemisphere, including from the sides. The resulting 3D DI is 8.7 dB but there is a price to pay for this excellent DI: second-order processing increases problems with internal noise and susceptibility to microphone mismatching.

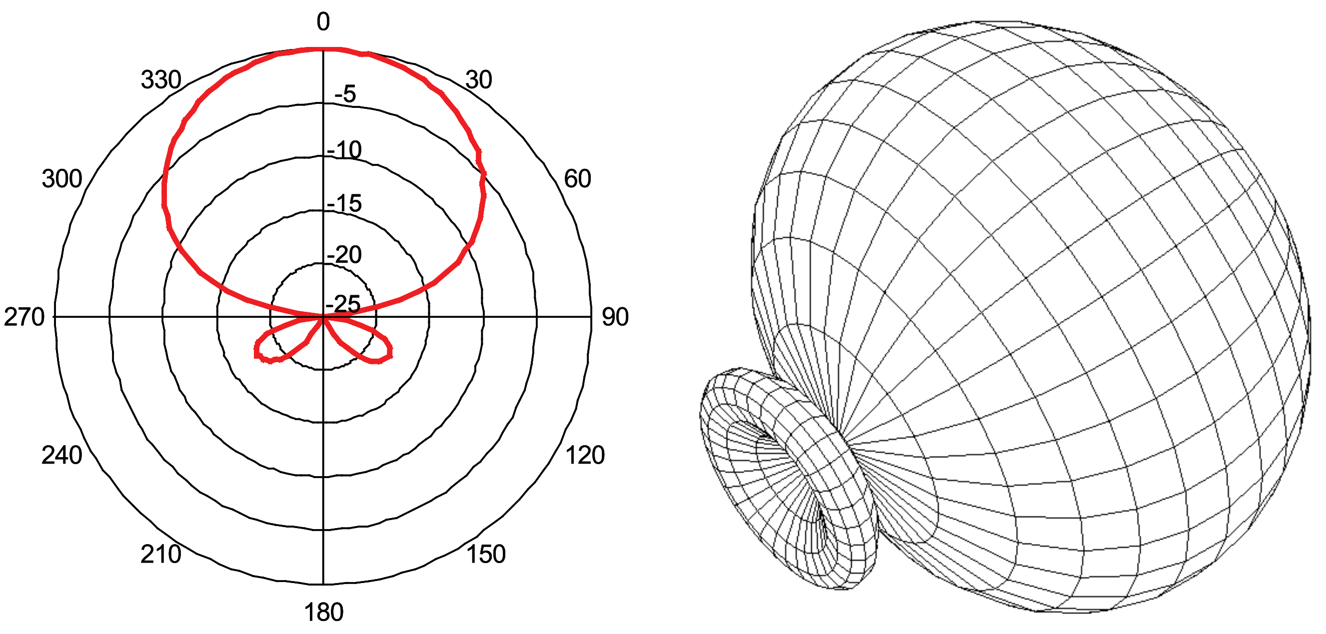

Figure 7.10 Directivity pattern (in two and three dimensions) of a second-order subtractive microphone.

Fitting a wide bandwidth, second-order subtractive array within even a large BTE would require a port spacing so small that internal microphone noise would be excessive. Second-order processing has, however, been incorporated into one BTE by restricting the second-order processing to the high frequencies where a small port spacing, internal noise, and microphone mismatching are less of a problem, and by using only three omni-directional microphones.[g] Conventional first-order directivity is retained for the low frequencies. Averaged across frequency, the resulting improvement in DI, and hence SNR, when head mounted, is a further 1 dB above first-order directivity.[100, 109, 1452]

A larger port spacing can be used in a hand-held directional microphone, and such devices are commercially available and effective in improving SNR.[1058]

Adaptive-weight arrays

The most sophisticated and effective arrays vary the gain and phase of each microphone output by different amounts at different frequencies, before adding all these filtered microphone outputs to produce the output signal. The fixed subtractive, fixed additive, variable delay, and second-order arrays discussed so far are all special (and simple) cases of general ***adaptive-weight*** arrays or beamformers.

A beamformer with only two microphones can produce a null in only one direction at a time at any frequency.[h] Consequently, if there are two noise sources with intensity in the same frequency region, they both cannot be removed. In general, a beamformer made from *n* omni-directional microphones can remove only *n*-1 different noise sources. It is important that adaptive beamformers be evaluated with multiple sources if a realistic assessment of their effectiveness is to be gained.

The basis of most sophisticated, adaptive, multi-microphone arrays is the ***Widrow Least Mean Squares (LMS)*** algorithm, with the structure shown in Figure 7.11.[1916] The top microphone picks up a mixture of signal plus noise. The bottom microphone is assumed to be positioned such that it picks up only (or mostly) noise. This microphone is referred to as the reference microphone. It is assumed that noise entering the two microphones comes from the same source, but reaches the two microphones by different paths, and thus has a different waveform at each microphone. If the noise at the reference microphone could be filtered to compensate for the difference in acoustic paths taken by the noise to the two microphones, this filtered noise could be subtracted from the mixture of signal and noise picked up by the main microphone. If the filtering and subtraction were perfectly carried out, the result would be speech alone.

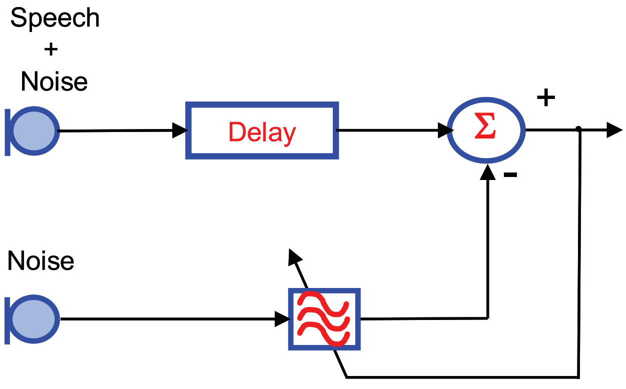

Figure 7.11 The Widrow Least Mean Squares adaptive noise reduction scheme, based on a reference microphone that picks up only the noise. The fixed delay compensates for the delay inherent in the adaptive filter.

The filter approximates the ideal shape by adaptively changing its response in such a way as to minimize the power of the output signal (hence the term LMS, as power is proportional to the square of the signal). Systems typically adapt to this response in less than

[g] If the front port of the pair forming the rear directional microphone array is located in the same place as the rear port of the pair forming the front directional microphone array, four microphones can be reduced to three.

[h] Actually, nulls can simultaneously occur in other directions, but the beamformer has no control over the directions of those other nulls once it has positioned the first null.

one second.[i] This adaptive filter can be thought of as applying complex weights (modification of gain and phase) to the noise reference, at each frequency, before subtracting the weighted noise signal from the main signal.

There are various methods, beyond the scope of this text to describe, for calculating what these weights should be. Some operate in the frequency domain so that the weights are separately determined in each frequency region. Others operate in the time domain, where the weights applied to past samples of each microphone output indirectly determine how the microphone outputs at each frequency are combined. Although both approaches are capable of converging to the same solution, approaches in the frequency domain appear to converge more rapidly, which is important when real-world signals are time-varying, and head movements can rapidly change the apparent direction of the sources.[1075]

The Widrow LMS system can increase SNR by 30 dB or more, provided a suitable reference signal is available. For head-worn hearing aids, however, it is not possible to position a single microphone so that it picks up only noise. If the reference microphone contains some signal, the filter will adapt to a shape that partially cancels both the signal and the noise. The situation is not hopeless, however.

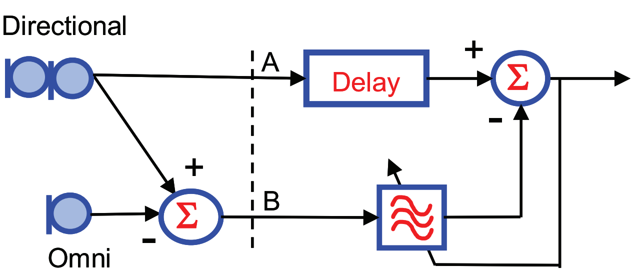

Figure 7.12 An adaptive noise canceller, where the noise reference signal is formed by subtracting the omni-directional microphone output from the directional microphone output.

Figure 7.12 shows one way in which a noise reference can be obtained in a fully head-worn hearing aid. A directional microphone provides (signal *A*) a combination of speech (from the front) and noise (from everywhere else). An omni-directional microphone (which can be one of the omni mics forming the directional microphone) also provides a combination of speech and noise, but with a poorer SNR than from the directional microphone. If the omni and directional microphones are equalized to have the same sensitivity and phase response for sounds from the front, then subtracting their outputs would remove any frontal signal, leaving just noise as signal *B*. The parts to the right of the dotted line, which are identical to the Widrow adaptive noise-reduction scheme shown in Figure 7.11, then perform the filtering and subtraction necessary to improve the SNR.

Adaptive filtering of the type just described, which is available in commercial hearing aids and cochlear implants, works extremely well under certain circumstances. In particular, where there is only one noise source, no reverberation, and a very poor SNR, the adaptive filter can change its characteristics so that the directivity pattern of the array has an almost perfect null in the direction of the noise, while keeping normal sensitivity in the target direction (usually straight ahead).[1690] Under these favorable circumstances, the SNR can be improved by as much as 30 dB.[1404] Unfortunately, there are unlikely to be any real-life circumstances where improvements this large can be expected.

Reverberation greatly decreases the effectiveness of adaptive arrays. Unless the wanted talker is very close, reverberation will cause significant speech energy to arrive from all directions. Consequently, the noise reference signal will contain speech as well as noise. This mixture makes it difficult for the filter to adapt, thus reducing the effectiveness of the noise canceling. The subtracter will also remove some of the speech as well as noise, and will thus affect speech quality. The beamformer can be modified in various ways to minimize, but not totally avoid, these difficulties. In one of these modifications, a speech/non-speech detector,

[i] The designer can select any adaptation time. If the adaptation is too quick, however, the filter will change excessively when the direction of the noise changes even slightly, and speech quality will deteriorate. If the adaptation is too slow, the filter will not be able to keep up with a changing source position, or movements of the hearing aid wearer's head.

(also known as ***voice activity detector***; ***VAD***) based on the more pulsatile nature and/or overall level of speech signals, is used to stop the adaptive filter from changing its response whenever speech is believed to be present so that noise alone determines the filter weights.[649, 691, 1621, 1833, 1834, 1843] Unfortunately, speech/non-speech detectors work least reliably just when SNR enhancement is most needed – at very poor SNRs.

In another modification, the noise canceller is preceded by another adaptive filter that is used to remove as much speech as possible from the noise.[1848] This other adaptive filter adapts only when speech *is* present. By creating such a reference signal that has as *little* speech in it as possible, the main adaptive filter has a much better chance of removing noise. A later version replaced the first adaptive filter with a fixed filter that made the system more stable in poor SNRs. For a single interfering source at 90° at a distance of 1m in a moderately reverberant room, the processing improved the speech reception threshold in noise by 11.3 dB.[1931]

The presence of reverberation also means that the echoes from a single sound will arrive at the hearing aid for some time after the direct wave arrives. These echoes can be removed only if the adaptive filter is sufficiently complex to store and combine sounds that arrived perhaps many hundreds of milliseconds before.[649] Such complex filters take longer to adapt.

As an example of the substantial benefits obtainable from adaptive filtering, and the limitation that reverberation can place on these benefits, Figure 7.13 shows results for a three- and a seven-microphone array mounted around the forehead.[745]

There are a number of ways in which the two (or more) different combinations of signal and noise needed for adaptive filtering can be obtained. What they have in common is that they must originate from microphones located at two (or more) points in space, whether these are on the same, or opposite sides (or front!) of the listener's head. One scheme uses a rearward-facing directional microphone to provide the noise reference and an omni-directional microphone to provide the mixture of speech and noise.[1908] These arrangements have the advantage that all the components can be mounted on one side of the head.

The techniques so far described adapt by minimizing the power at some point in the processing scheme.

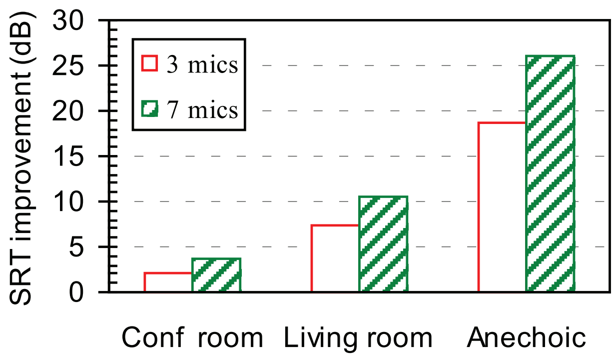

Figure 7.13 Improvement in speech reception threshold for an adaptive array relative to a single microphone. The experiment used frontal speech and a single noise masker at 45 degrees from the front in three simulated environments that differed in the amount of reverberant sound relative to the direct sound. From Hoffman et al (1994).

They have the aim of maximizing the ratio of a speech signal arriving from the front to signals arriving from other directions. These schemes must make assumptions about the location of the speech source (e.g. it is directly in front) and/or about the nature of the wanted signal (e.g. it is speech and therefore pulses in amplitude) and/or about the noise (e.g. it is continuous). A more general technique, referred to as ***Blind Channel Separation***, ***Blind Source Separation***, or ***Co-channel Separation*** is able to separate signals of any type arriving from different directions, as shown in Figure 7.14.[844, 1905] The only necessary assumption is that the original sources are statistically independent. This assumption will always be true if the signals originate from different talkers or other sources of noise. The adaptation process varies the filters in the direction needed to maximize the statistical independence of the output signals.

If the input to the Blind Source Separation processing scheme comprises n microphones, each picking up a different mixture of the original sources, the processing scheme can potentially separate up to n sources.

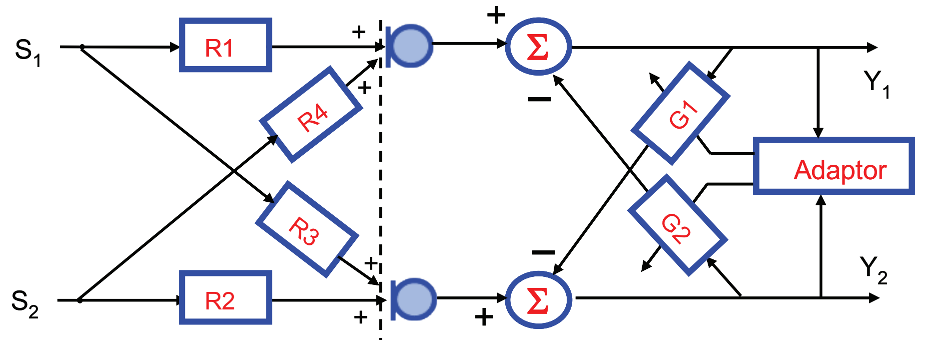

Figure 7.14 Blind source separation of two sources, S1 and S2, occurs when the two adaptive filters, G1 and G2, adapt to the response shapes that compensate for the room transmission characteristics, R1, R2, R3 and R4, from each source to each microphone. Note that everything to the right of the dotted line is in the hearing aid, whereas the blocks to the left are the transfer functions of the transmission paths within the room. When properly adapted, the response of G1 = R3/R1 and G2 = R4/R2. The output Y1 then does not contain any components of S2. The blocks G1 and G2 can alternatively be feed-forward blocks rather than feed-back blocks.

Which of the *n* possible outputs should finally be presented to the hearing aid wearer? To decide this, some assumption must be made, such as selecting the source with the highest level, or the source with the direction closest to the front. Like the other adaptive schemes, this scheme works much better in non-reverberant situations than in reverberant situations. In non-reverberant situations, the adaptive filters needed are much simpler, and hence adapt more quickly and accurately.

Adaptive versus fixed arrays

Where there are many noise sources, adaptive arrays cannot provide better performance than a fixed array, and can provide worse performance if the adaptive array has not been designed to realize its limitations! Any adaptive processing scheme that alters its amplification characteristics rapidly while speech is present is likely to introduce unpleasant artifacts to the speech. The hearing aid wearer perceives the rapidly changing response shape as added noises with a musical quality (understandably so, because narrowband components of the signal are suddenly being increased or decreased in level). The artifacts are minimized by slowing the rate at which the processing adapts and/or by not doing any adaptation when speech is estimated to be present, as explained earlier. Adaptive arrays can also inadvertently cancel the speech at some frequencies if the array is not pointed directly at the source. For most implementations, such ***mis-steering*** will occur whenever the listener is not looking directly at the source, but its effects can be minimized by statistical techniques.[745]

One advantage of adaptive arrays over fixed arrays concerns the accuracy with which they are constructed. Some fixed arrays (notably subtractive arrays) rely on the separate microphones having electroacoustic responses that are well matched to each other. Because adaptive arrays monitor the output signal, the adaptive filter can partially compensate for any mismatch between the microphones. A disadvantage of adaptive arrays is that they require more computations within the hearing aid, although this is becoming a less important issue with each passing year.

Although fixed and adaptive directional arrays have been contrasted in this chapter, they can be combined by using the outputs of two or more fixed arrays as the inputs of an adaptive array.[413, 961, 1848] Performance is generally superior to the performance of either array alone for nearby signals and jammers in non-reverberant situations. (The adaptive microphone literature often refers to unwanted noise sources as "jammers", by analogy with military radar systems.) Adaptive processing is unlikely to provide any additional benefits in reverberant situations in which the noise lies well beyond the critical distance.[649] Furthermore, when the SNR is good, adaptive processing can decrease the SNR if adaptation occurs when the speech is present.

7.1.4 Bilateral directivity

More often than not, people wear hearing aids on both sides of their head. When the target speech arrives from the front, and noise arrives equally from all directions, most if not all of the directional benefit can be provided by either hearing aid having a directional response.[106] If, however, the target speech or the noise are asymmetrical such that one hearing aid is exposed to a better SNR than the other, then it is important that the hearing aid with the better SNR be able to adapt its polar pattern if speech intelligibility is to be maximized.[1106]

This flexibility requires adaptive directional processing in each hearing aid, which poses minor problems, but major opportunities if the adaptive processing uses information from both sides of the head.

Synchronized bilateral directional microphones

The minor problem is that if each hearing aid independently determines if it should be directional or omnidirectional, independently determines what polar pattern should be adopted, or independently adapts with complex weights, then the two hearing aids will sometimes come to different decisions about what processing to apply.[j] This will occur because head diffraction will cause the SNR and direction-of-arrival to be different at each ear whenever either the target, or any other significant sound source, is anywhere other than directly ahead or directly behind.

Isn't this desirable? If the only aim is to maximize the SNR at each ear, and hence intelligibility of speech in the presence of nearby noises, then it is precisely the best outcome.[1106] Directional processing, however, alters the phase of signals, so if each directional microphone has a different pattern, the phase relationship between ears will be different at the hearing aid outputs than at the hearing aid inputs. Consequently, the inter-aural phase cues that help the aid wearer localize sounds will not be those caused by the head shape to which the aid wearer has become accustomed. Worse, the phase relationships will change whenever the directional patterns change, making it difficult or impossible for the wearer to become accustomed to the new phase relationships. This disturbance of inter-aural phase cues decreases localization accuracy in the horizontal plane.[887, 1837]

The disturbance to localization is minimized if the two microphone polar patterns match.[887] Some hearing aids use a wireless link between aids to determine a compromise response that both adopt. There has not been sufficient research to specify how closely the two patterns must agree, but it is clear that extreme variation, such as a cardioid response in one ear and an omni-directional response in the other ear does disturb localization, particularly if the pattern in each ear changes from time to time so that the aid wearer can never become accustomed to a new set of inter-aural cues.

Bilateral directional processing

The human head is an acoustic barrier of appreciable size (Section15.1.1), so it is no surprise that one hearing aid will often have access to signal and noise levels, phases, and SNRs that are vastly different from those at the other ear. These differences are precisely what are needed to form super-directional microphones that have the potential to offer hearing aid wearers *substantial* improvements in intelligibility in noisy places, even when there is moderate reverberation. Note that the term *bilateral directional processing* is being used to mean processing schemes that produce highly directional beams by combining inputs from both sides of the head, not just directional processing schemes where all the sounds presented to one ear are based on sounds picked up by microphones on that same side of the head.

An early example is the ***Griffiths-Jim beamformer*** shown in Figure 7.15.[654] The section to the right of the dashed line is the (by now hopefully familiar) Widrow LMS adaptive canceller. When the wanted signal is directly in front of the person, and there is little reverberation, there should be little wanted signal at point *B* in the lower chain, because the outputs from the two microphones for frontal signals should be equal and therefore cancel. The signal at point *A* can be recog-

[j] Because hearing aids average across time to determine whether to adopt a noise program, the two hearing aids are more likely to reach different decisions about which polar pattern to adopt than they are about whether to be directional or omnidirectional.

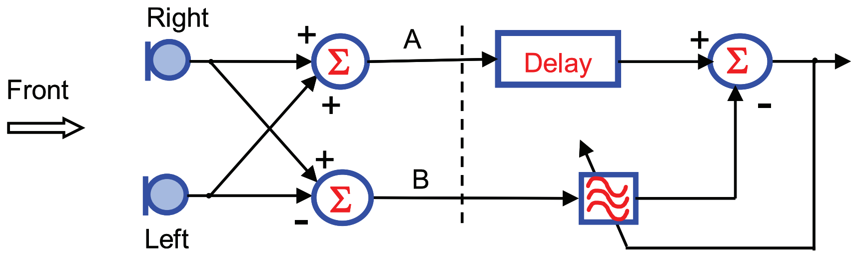

Figure 7.15 A Griffiths-Jim adaptive noise canceller, whereby the two microphone outputs from opposite sides of the head are added in the top chain but subtracted in the bottom chain.

nized as the output of a broadside additive array, and so already has a small amount of directivity.

Performance can be increased by using directional microphones on each ear, rather than omni-directional microphones. Several laboratory evaluations of this scheme have confirmed that it provides a large improvement in SNR when there is little reverberation, but diminishing benefit as the ratio of direct to reverberant sound decreases, and diminishing benefit when the SNR gets extremely poor, just as for monaural adaptive arrays.[1833, 1834, 1843]

As with monaural adaptive arrays, performance is best if filters that are intended to shape the noise (like the one in Figure 7.15) adapt only when the target speech is absent. This decision requires the processing to include a voice activity detector (VAD). Unfortunately it is difficult to make a VAD work accurately in poor SNRs, which is precisely when we most need noise reduction systems to work well. Similarly, filters that are intended to shape the speech target picked up by each microphone (not shown in Figure 7.15) should adapt only when speech is present, and it is difficult for a VAD to detect the weak parts of speech, for which SNR improvement is most needed.

There are several other processing schemes in the research literature where the aim is to attenuate any narrow frequency band for which the signals picked up at each ear do not show the phase difference and coherence appropriate to the direction of the target signal.[30, 147, 959] The target signal is usually assumed to be directly in front of the aid wearer. Such processing improves sound quality and intelligibility, and reduces the apparent amount of reverberation, as does all directional processing (Section 7.3.1).

There are two problems with bilateral beamformers that have only very recently been overcome:

- The beamformer produces a single output signal. If this signal is delivered unchanged to the left and right ears, the aid wearer will hear all sounds in the room as originating in the centre of the head, which gives the unwanted sounds more masking ability than if they are perceived as coming from different locations.[574] This clustering of sounds partially undoes the intelligibility increase that the beamformer can potentially offer.
- Aid wearers do not always look directly at the person to whom they are trying to listen, and if a highly directional main lobe misses the target, much of the advantage of high directivity will be lost, and the speech may even be distorted.

Figure 7.16 shows a bilateral processing scheme developed at the National Acoustic Laboratories that avoids these two problems.[1182] Directional microphones (or adaptive directional arrays) on each ear provide the input signals, which are combined by a beamformer that emphasizes sounds coming from a specified range of frontal directions (e.g. +45° to - 45°), but attenuates sounds outside this range. The beamformer also indicates the direction from which the dominant signal comes. The signal with the enhanced SNR is filtered to impart to it, separately for each of the left and right ears, the head-related transfer function (HRTF, Section 15.1.1) appropriate to the direction of the dominant frontal signal. This filtering makes the signal towards which the beam is pointed appear to come from its true direction.[75, 1182]

Unfortunately, because all other sounds are processed with the same HRTF, if this was all we presented to the aid wearer, every sound source would appear to come from this direction, which maximizes their potential for confusion.

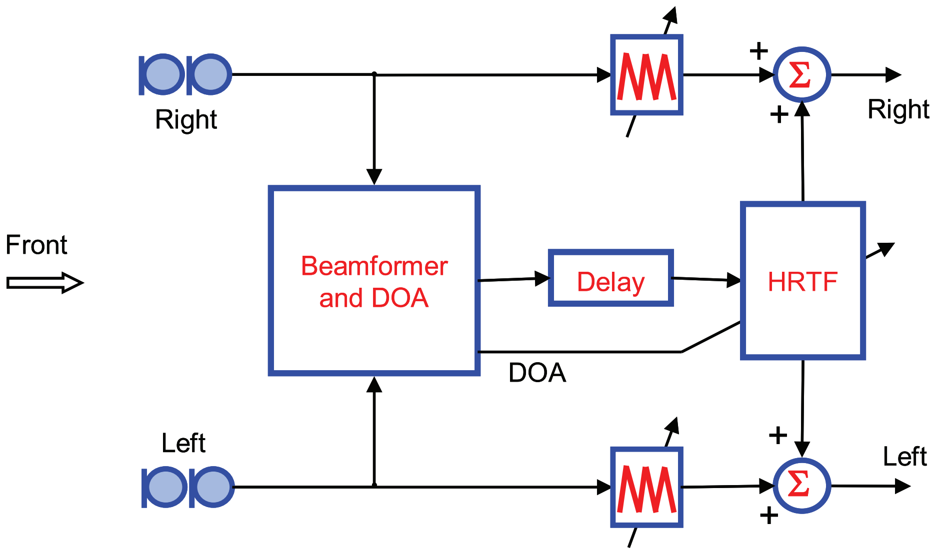

Figure 7.16 A bilateral processing scheme that improves SNR while preserving the apparent location of sources in the room.

DOA = direction of arrival.

HRTF = head-related transfer function.

The different spatial cues for each source can be re-introduced by adding in attenuated versions of the input signal. Unfortunately, this degrades the SNR improvement achieved by the beam, although the regained spatial cues may offset the loss of SNR they cause.[1836] The degradation can be minimized if the main signal is delayed by a few milliseconds before strongly attenuated versions of the original input signals are added in.[1182] The ***precedence effect*** then ensures that the attenuated signals dominate the perception of direction even if they are considerably weaker than the main signal. The main signal, however, because it is so much stronger, retains the high intelligibility of the beamformer. Preliminary laboratory investigation of this scheme indicates that, in many noisy listening situations, people with mild to moderate hearing loss will hear *better* than people with normal hearing listening unaided.[1183] Similar schemes, offering similar advantages, are likely to emerge from other research laboratories.

These complex bilateral processing schemes will become increasingly available over the next several years. Their availability has been waiting on the existence of a wide-bandwidth, low-power, bi-directional wireless link between the two hearing aids, and on very high-speed mathematical calculation ability within hearing aids. Steady advances in integrated circuit have just about removed both these impediments, and high-performance bilateral directional microphones should emerge within the next few years.

Simpler bilateral schemes that transfer signals wirelessly from one side of the head to the other are already available at the time of writing. Such schemes, for example, can pick up a microphone input from the side of the head nearest a communication partner and reproduce this signal in both ears of the hearing aid wearer. This is particularly beneficial in situations where the communication partner is in a predictable direction, such as inside a car. Alternatively they can add together the left and right signals to produce a slightly more directional frontal beam that is presented to both ears.

7.2 Quantifying Directivity

7.2.1 2D and 3D directivity index

Provided there is no strong noise source close to the aid wearer, reverberation causes noise to arrive approximately uniformly from all directions. If the target source is close to, and directly in front of, the aid wearer, then the microphone will pick up the target sound with its frontal sensitivity, but pick up the noise with its average sensitivity. The ratio of frontal sensitivity to average sensitivity is referred to as the directivity factor, which when converted to decibels becomes the directivity index (***DI***). The DI therefore tells us how the directional microphone affects the SNR compared to what a perfectly omni-directional microphone would produce.

3D-DI anechoic measurement: The sensitivity of the microphone, or complete hearing aid, is measured for many directions. Because the noise components arriving from different directions add together on the basis of their power, the different measurements must be combined by averaging the ***power sensitivity***[k] of the microphone for the different directions in three-dimensional space. If the directions sampled are at *n* equally spaced locations around a sphere, the three-dimensional DI (***3D-DI***) is calculated simply as:

$$3D\text{-}DI = 10\,\log_{10}\left(\frac{\textit{Frontal power sensitivity}}{\frac{1}{n}\sum \textit{power senstivity}}\right) \qquad \text{....7.1}$$

If the measurements are instead made at all combinations of evenly spaced azimuths and elevations, the upper and lower parts of the sphere are over-represented, and a slightly more complex formula is used to correct for this.[469] While either measurement would give us the 3D-DI that is so revealing about the effect of the microphone in many listening situations, both measurements are so time consuming and equipment intensive that they are virtually never performed except in rare research studies.[469] Instead, directivity index is measured in one of three ways.

3D-DI diffuse field measurement: A diffuse field is one in which sound power arrives equally from all directions. It is created with a small number of sources and takes advantage of reflecting surfaces and reverberation to spread power evenly throughout the room. It is nonetheless difficult to create a truly diffuse field. The SPL is measured with a reference omni-directional microphone and with the directional hearing aid being measured. Call these SPL values $Omni_{diffuse}$ and $Dir_{diffuse}$ respectively. Now the reference microphone and directional hearing aid are moved to a place where there are no significant reflections (an anechoic chamber, outdoors, or just very close to a source in an absorbent environment), and the SPL of a single frontal source is measured with both devices. Call these values $Omni_{frontal}$ and $Dir_{frontal}$. The diffuse-field DI, which is identical to the 3D-DI, is then calculated as:

$$3D\text{-}DI = (Dir_{frontal} - Dir_{diffuse}) - (Omni_{frontal} - Omni_{diffuse}) \qquad \text{... 7.2}$$

2D-DI anechoic measurement: The directional hearing aid is mounted on a turntable, and the sensitivity is measured for different azimuths. The measurement is performed in an anechoic chamber, or other place with no significant reflections, with a single source. Measurements such as these directly give polar diagrams like those in Figure 7.2. If the frontal sensitivity is divided by the average sensitivity for all the *n* azimuths measured around the circle, and the ratio is converted to decibels, the result is a DI, but one that tells us about directivity only in the horizontal plane. This 2D-DI is also called a ***planar DI.***[55] The calculation equation looks identical to equation 7.1, but the summation in the denominator now includes only the azimuths sampled around a circle, not the azimuths and elevations sampled around a sphere:

$$2D\text{-}DI = 10\,\log_{10}\left(\frac{\textit{Frontal power sensitivity}}{\frac{1}{n}\sum \textit{power sensitivity}}\right) \qquad \text{...7.3}$$

The 2D-DI is sometimes justified on the grounds that in many real-life situations there is a greater sound pressure arriving in the horizontal plane than from above or below. This is not a good assumption in reverberant environments.

3D-DI inferred from circular symmetry: The 2D-DI measurement is performed, but the averaging is performed in such a way as to estimate the effect of sounds arriving from above and below the horizontal plane. This estimation relies on the circular symmetry of end-fire microphones when measured suspended in free space: their sensitivity measured at, say, 30° to the left of the forward direction is the same as that measured 30° above, below or to the right of the forward direction. Trigonometry then shows that the 3D-DI can then be estimated by weighting each measurement in the average with $\pi \sin(\theta)/2$ where θ is the

[k] Power sensitivity is proportional to the square of pressure sensitivity. If a complete hearing aid is being measured, the output SPL is converted to a quantity proportional to power using: $power = 10^{SPL/10}$.

angle measured from the front.[1] ANSI S3.35 refers to the DI calculated in this manner as the ***planar directivity index***.

$$3D\text{-}DI_{estimated} = 10\log_{10}\left(\frac{Frontal\ power\ sensitivity}{\pi/2n\sum power\ sensitivity.\left|\sin(\theta)\right|}\right) \quad \ldots 7.4$$

The assumption of circular symmetry is poorly justified when the hearing aid is mounted on the head, because the combination of the head and a hearing aid on one side of it is far from symmetrical.

Whichever of these definitions is used as the basis of the measurement, the DI value obtained applies only to the condition in which the hearing aid is measured. If the hearing aid is suspended in free space, then it is a free-space DI; if it is mounted on a head (whether real or dummy), then it is a DI indicative of the hearing aid when in use. Because of the effects of the head or body, the values will not be the same, but microphones with a high DI in free space also have a high DI when worn on the head. As shown in Figure 7.3, the sensitivity pattern when worn on the head reflects both the sensitivity pattern of the head, and the sensitivity pattern of the microphone in free space.

If it is time-consuming to measure many directions in three-dimensional space (equation 7.1), difficult to produce a truly diffuse field (equation 7.2), and invalid to assume symmetry (equation 7.4), then how should DI be measured? Most commonly, researchers and industry just measure the 2D-DI (equation 7.3) and use it without further correction or comment. As Figure 7.2 shows, the 2D and 3D-DIs are similar in size for first-order subtractive microphones measured in free space. Larger differences between the 2D and 3D DIs (typically about 1 dB) emerge when the directional hearing aids are positioned on a head.[469] Hearing aids with a relatively high 2D-DI are nonetheless very likely to have a relatively high 3D-DI, so comparison of devices on the basis of their 2D-DIs is reasonable, even if theoretically not well grounded.

It is often observed that hearing aids worn on the head have their maximum sensitivity somewhat towards the side of the head on which they are worn (e.g. Figure 7.3), and that hearing aid wearers frequently do not fully turn to look at the talker. Consequently, it is sometimes argued that the *frontal sensitivity* in equations 7.2 and 7.3 should be replaced by sensitivity averaged over a range of directions (e.g. 0 to 30°) more likely to incorporate the direction of maximum sensitivity. This would produce DI values that made all head-worn microphones, directional and omnidirectional alike, *seem* more directional, but would add a further complexity to the types of DIs that already exist.

7.2.2 AI-DI

DI tells us how SNR is affected in diffuse noise environments, but what does this mean for speech if the hearing aid has different DI values at different frequencies, as is usually the case? Most commonly, directivity decreases as frequency decreases because vent-transmitted sound progressively dominates. Directivity may also decrease at very high frequencies if port spacing approaches half a wavelength at those frequencies.

The effect of directivity on speech intelligibility can be estimated by averaging DI across frequency, but because some frequencies contribute more to intelligibility than others, it is desirable to weight the DI at each frequency according to the importance of that frequency to intelligibility. A suitable set of weighting values is the importance function used in the ***Articulation Index*** (***AI***) method (a later variation of which is called the ***Speech Intelligibility Index***).[648] Another suitable importance function is available in the well-known count-the-dots AI calculation method.[918]

The weighted-average directivity index is thus known as the ***AI-DI***. It tells us how much the diffuse noise level would have to be increased if SNR at the hearing aid output were to remain the same when a directional microphone replaced an omni-directional microphone. Even a simple unweighted average of the DIs at each octave or third-octave frequency, sometimes referred to as ***DI-a,*** provides a good estimate of the SNR improvement that will be obtained in diffuse noise for a nearby frontal target.[1508]

1 There is only one direction in 3D space for which θ equals each of 0° and 180°, but there is an infinite number of directions for which θ equals 90°, hence the progressively greater weighting that sin(θ) gives as θ progresses from 0° towards 90°.

The SNR that results in a specified intelligibility, often 50%, is referred to as the ***speech reception threshold in noise*** (***SRT_n***). If a directional microphone increases the SNR by *x* dB, then the SRT_n should also increase (i.e. improve) by approximately *x* dB. This is true provided the AI-DI averages the SNR improvement across frequency with weightings appropriate to the speech material and characteristics of the hearing aid wearer (Section 7.3.5). The equivalence of SRT_n improvement and AI-DI also requires that at the SRT_n, the SNR not be greater than 15 dB at any frequency (because speech is already totally audible), and that the SNR not be poorer than -15 dB (because speech is so completely over-masked that a small increase in SNR may produce no benefit at all). If these conditions are not true, then AI-DI will over-estimate the benefit of the directional microphone in close-target, diffuse-noise listening situations. When the conditions do apply, every dB improvement in AI-DI, or in DI-a, should produce a 1 dB improvement in SRT_n, which is approximately equivalent to a 10 percentage point increase in speech understanding for high-redundancy sentence material.

Figure 7.17 shows an example where the directional microphone reduces noise by 5 dB at all frequencies, but the speech is made more audible by 5 dB only from 250 Hz to 2000 Hz. Below 200 Hz and above 3000 Hz, there is no change in the amount of speech audible (for different reasons in each frequency region), despite the 5 dB improvement in SNR.

Just as directional microphones directly alter SNR and SRT_n, they also directly alter acceptable noise level (ANL), the minimum SNR that people will subjectively accept while listening to speech.[570]

Omni-directional hearing aids have AI-DI values around -1 dB (for BTEs) and around 0 dB (for ITE/ITC/CICs). Directional hearing aids typically have AI-DI values from 3 to 5 dB. Because of this variation across hearing aids, it is important to examine the specifications of a directional hearing aid to find out how effective its directional microphone is when mounted on a head.

AI-DI is an enormously useful indicator of directional benefit but by no means indicates the benefit of the directional microphone in all situations. It reliably indicates how much the microphone improves the SNR of speech when the talker is very close to, and directly in front of, the aid wearer, and the noise comes uniformly from all directions.[m] When noise is close and comes from a single direction, directional microphones can suppress noise to a much greater, or lesser, extent than indicated by the AI-DI value, and it is this situation in which adaptive directional microphones can be much more effective than microphones with fixed sensitivity patterns.

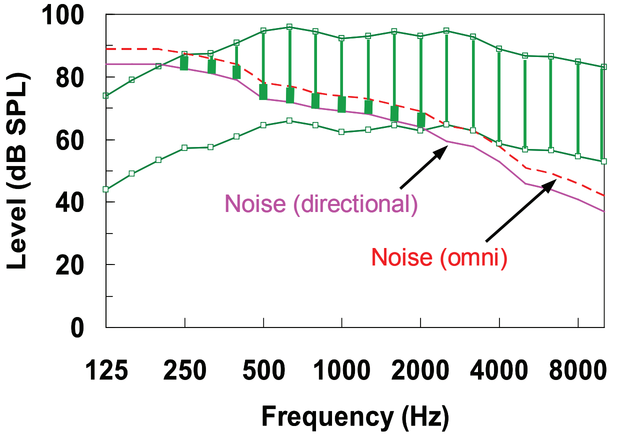

Figure 7.17 The long-term 1/3 octave speech spectrum maximum and minimum levels (green) at the output of a hearing aid, and the output noise spectrum for an omni-directional microphone (red, dashed) and a hypothetical directional microphone (pink, solid) that improves the SNR by 5 dB at all frequencies. The thinly shaded area shows the speech that is audible with the omni-directional microphone and the thick shading shows the additional speech made audible by the directional microphone.

7.3 Directional Benefit

In brief, the benefit provided by a directional microphone depends on the directivity of the hearing aid, the reverberation characteristics of the listening situation, the distance of the talker and noise sources,

[m] AI will be less precise an indicator of benefit if the DI varies markedly with frequency, and if the speech spectrum is markedly different from the noise spectrum. In this case, SNR benefit will be better predicted by DI averaged over just those frequencies for which SNR is neither so good that the entire speech dynamic range is audible, nor so poor that none of the speech dynamic range is audible above the noise.

and not at all on the speech material used to assess benefit. The directivity of the hearing aid depends on the openness of the fitting, the gain of the hearing aid, and its position within or behind the ear. With limited exceptions (Section 7.3.5), it does not depend on the characteristics of the hearing aid wearer. The SNR in the environment can affect the degree of benefit offered by adaptive directional microphones but not that offered by fixed arrays. These points are elaborated on below.

7.3.1 Impact of listening environment

Signal-to-noise ratio

Let us first define the benefit of a directional microphone as the amount by which noise has to be increased when the directional microphone replaces an omni-directional microphone if the intelligibility or apparent noise is to be the same as occurred when the omni-directional microphone was used. Defining benefit in terms of SNR improvement, rather than as a percentage increase in intelligibility, enables us to better understand what directional microphones can and can't do because it is SNR improvement that directional microphones potentially achieve.

Consequently, a fixed array will give the same improvement in SNR no matter what speech material is used to assess the benefit.[1932] By contrast, the effect on intelligibility score of a certain improvement in SNR depends on many other factors, such as the type of speech, the type of noise, the degree of hearing loss,[1508] and whether the score was close to 0% or 100% before the directional microphone was introduced.[1876] Figure 7.18 shows typical psychometric functions for two different types of speech material. Although the benefit of the directional microphone may *seem* to be different in each case, and at different SNRs, this is the case only if one thinks of benefit as the increase in percentage correct. In fact the omni-directional and directional psychometric functions differ by 3 dB in both cases and at all SNRs.

The benefit of adaptive arrays measured in dB can, however, vary with the baseline SNR, as the automatic adaption may become imprecise at very poor or very high SNRs.

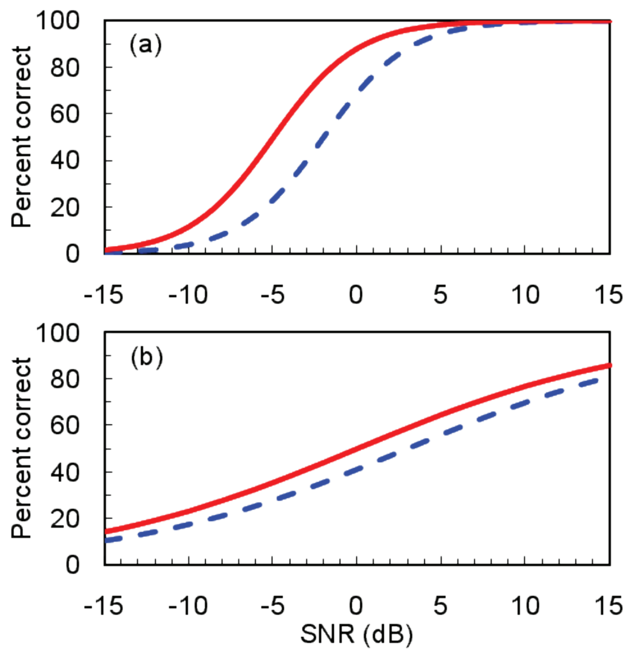

Figure 7.18 Typical psychometric functions for speech intelligibility for (a) high-context sentence material and (b) phonemically scored isolated words. In each case, the dashed blue line shows the function for an omni-directional microphone and the solid red line shows the function that would result if a directional microphone provided a 3 dB improvement in SNR.

Room acoustics and distance

By contrast with the non-effect of baseline SNR, room acoustics and listening distance determines almost everything about directional benefit. Figure 3.2 and accompanying text explained the concept of critical distance, the distance from a source at which the direct and reverberant field SPLs are equal (and the total SPL is therefore 3 dB greater than either component of the field). Indeed, nothing could be more critical to the effectiveness of directional microphones than how far away the target and noise sources are compared to the critical distance in any particular listening situation. To illustrate this, let us consider four extreme combinations of the target and masking noise(s) being much closer or much further away than the critical distance from the aid wearer.

Figure 7.19 shows four situations categorized by how far away are the target and distracting noise(s) relative to the critical distance. Directional microphones provide different degrees of benefit in each situation, and the relative advantage that an adaptive directional microphone provides over a fixed polar pattern also changes.

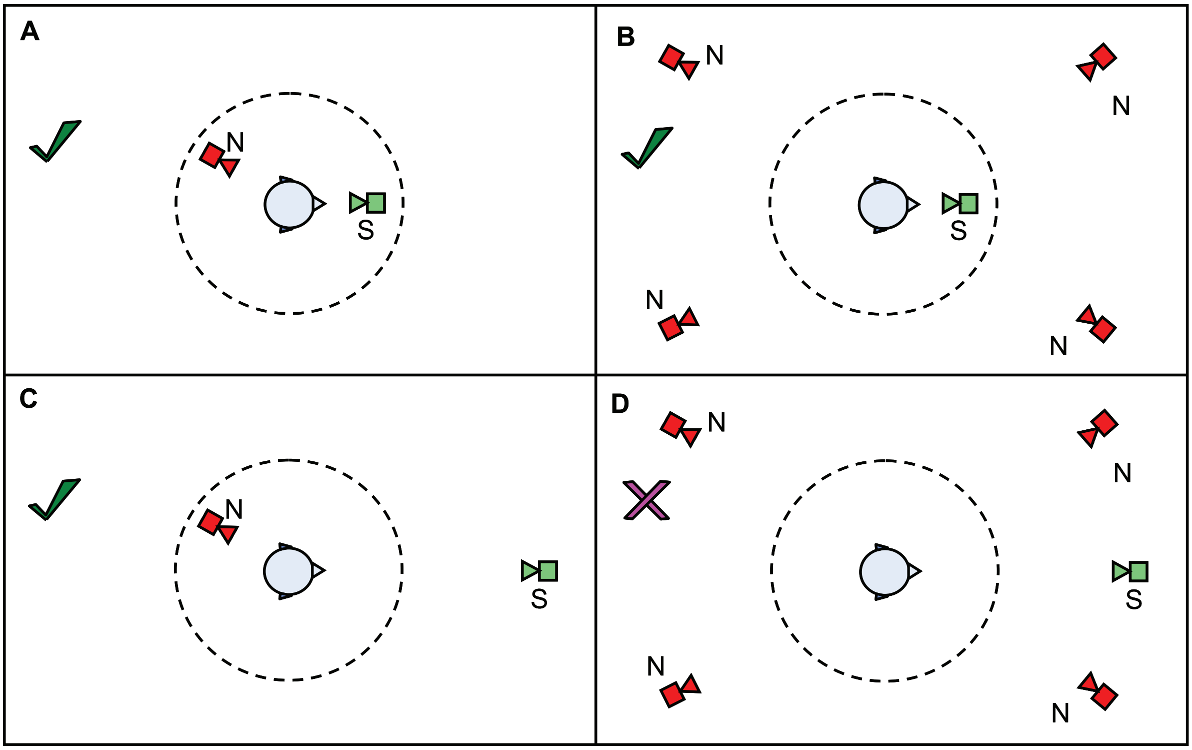

Figure 7.19 Four extreme situations in which the target and background noise source(s) are closer than, or much further away than, the critical distance from the listener. Regions within the critical distance are enclosed within the dashed line. Check marks show the situations in which directional microphones will provide benefit and the cross shows the situation in which directional microphones provide no benefit. S and N refer to the location of the target signal and competing noise, respectively.

A. Target and noise close. Any directional microphone with a pattern in the range cardioid to hypercardioid will provide benefit as the frontal sensitivity will be greater than the sensitivity for any angle to the rear. The benefit will be considerably greater than AI-DI. An adaptive directional microphone will very likely provide even more benefit, as it will adapt to produce a null in the direction of the single rearward (or sideward) noise source, which will only occasionally, by chance, be the case for a fixed pattern.[140, 1500] If there are multiple close noise sources, or if the direction of the dominant noise changes rapidly, then adaptive directivity is unlikely to have any significant advantage (but no disadvantage either) over fixed directivity.[111]

B. Target close, noises distant. The multiple noises far away will ensure that noise arrives approximately uniformly from all directions. As the target is detected with frontal sensitivity, and the noise is detected with the diffuse field sensitivity, a fixed array will produce benefit equal to its three-dimensional AI-DI. An adaptive array will provide benefit equal to the pattern that has the highest diffuse field DI possible for an array of its order. The observed benefit in SRT_n will equal AI-DI only if the target is much closer than the critical distance (otherwise part of the target sound will be picked up with the diffuse field sensitivity), and if the SNR at all frequencies is neither extremely negative nor extremely positive. In practice, SRT_n benefit will therefore often be a little less than the AI-DI.

C. Target distant, noise close. The target will be picked up with the microphone's diffuse field sensitivity, but the noise will be picked up with the microphone's rearward sensitivity, which is less. There will therefore be some small benefit for a fixed pattern, and very likely greater benefit for an adaptive directional microphone, for the same reason as in A.

D. Target distant, noises distant. Both the target and the noise will be picked up with the microphone's diffuse field sensitivity, so the directional microphone will provide no benefit. At best, if the target is not too far outside the critical distance, the directional microphone will increase the ratio of direct to reverberant power. Benefit, if any, will only be small unless the microphone is extremely directional, which is not possible with first-order directional microphones.

Although the critical distance has rightly taken central place in this discussion, nothing changes abruptly as the distance to the target or noise source crosses the critical distance. Rather, as the distance gradually increases from well inside the critical distance to well outside it, the degree of benefit offered by the directional microphone changes gradually from the benefit that applies for short distances to the benefit that applies for long distances. Some directional benefit can be expected even if the target is some way outside the critical distance and the noise very distant.[1753]

Another way of thinking about the benefit of directional microphones is that they effectively increase the critical distance, enabling people to receive a greater ratio of direct to reverberant sound, and hence a more intelligible signal. When a directional microphone is used, the formula for critical distance in Section 3.4 can be generalized to:

$$d_{c,\,effective} = 0.1\sqrt{\frac{Q_s Q_m V}{\pi T_{60}}} \qquad \ldots 7.5,$$

where Q_s is the directivity factor of the source,[n] and Q_m is the directivity factor of the directional microphone. Effectively, when there is reverberation, directional microphones increase the ratio of direct to reverberant sound, just as if the aid wearer had moved closer to the source. This is a much-overlooked benefit of directional microphones that applies even when there is no background noise. Of course, if the source is far outside the (normal) critical distance, then the reverberant field may be so much greater than the direct field that increasing the ratio with a directional microphone still leaves the direct field masked by the reverberation. Note that in highly reflective rooms, including some restaurants, the critical distance can be less than a metre.

The greater the DI of the directional microphone, the further it can *reach* outside the normal critical distance, and the greater the range of listening situations in which the directional microphone can provide benefit. Perverse as it may seem, although reverberation is the main factor that limits the benefits directional microphones can provide, reduction of reverberation is one of the benefits that directional microphones provide.

7.3.2 Objective benefit in the clinic and self reported benefit in real life

Evaluation of the benefits of directional microphones in real life (based on self-report) have indicated far less benefit than might be expected from speech intelligibility measurements in the clinic. The room acoustics considerations just discussed make this a very understandable combination of results.

Evaluation in the clinic. Evaluations in the clinic often have just a single noise source positioned somewhere to the rear, a target directly in front, and both are located about 1m from the listener in a very low reverberation environment (like situation A in Figure 7.19). All directional microphones will perform well in such environments and adaptive directional microphones will perform extremely well. Directional benefits in the range of 5 to 10 dB improvement in SNR are common.[41, 317, 1106] The unrealistic nature of a single, nearby noise source is well recognized by researchers, and many studies therefore use multiple noise sources, although mostly they are to the sides and rear, and mostly the test environment is a test booth with low reverberation. In such experiments, directional benefit in the range 2.5 to 5 dB improvement in SNR is common.[100, 106, 164, 317, 1466, 1502, 1504, 1874, 1876] Backward pointing directional patterns work just as well in such clinical tests as forward looking patterns, provided, of course that the target signal is to the rear.[1265]

Evaluation in real life. Many real life situations, however, are more like situation D, in which no benefit can be expected. When patients are asked to give

[n] The directivity factor of the source describes the extent to which a source radiates sounds straight ahead of it relative to the radiation averaged across all directions around the source. It is exactly analogous to the directivity factor of microphones, but is relevant to generating sound rather than sensing sound.

overall subjective reports of benefit, the many situations in which there are no benefits, and the situations in which directional microphones are disadvantageous, may well dominate patients' perceptions, even if the benefit in some proportion of situations is very worthwhile indeed. Directional benefits appear to occur in approximately one-third of real-life listening situations (but varying across studies from one-quarter to one-half).[140, 323, 1375, 1874] In at least one-third of situations,[1874] but perhaps even in the majority of situations,[1753] neither the omni-directional, nor the directional response is better than the other.

One-third of patients cannot tell the difference between directional and omni-directional responses in *any* situation they experience and leave the hearing aid in its default program rather than continue switching between omni-directional and directional patterns.[323, 990, 1375] The majority of patients can, however, detect situations in which either the directional microphone or the omni-directional pattern is better than the alternative. We know that these preferences are well founded, because the situations in which they most prefer the directional microphones are when the speech target is close and from the front, and when there is noise present and from the back or sides.[323, 1502, 1753, 1874]

Given the acoustical inevitability of a directional advantage when the talker is close and frontal and/or the dominant masker is close and rearward, it should be no surprise that the same speech intelligibility benefits are measured in the clinic for those who report benefit in real life as for those who report no benefit in real life.[322] The difference between the two groups either resides in either how often they are in situations where the physical increase in SNR will occur, and/or, in how capable people are of detecting the benefit provided by an improvement in SNR of a few decibels. (This is not as easy a task in real-world conditions as might be imagined, given that the levels and locations of the target talker and the masking sources are all likely to be changing while the aid wearer is comparing performance in the directional and omni-directional modes.)

Despite the difficulties that some patients have in reporting overall preferences when the advantages differ across listening situations, a meta-analysis of experiments examining self-reported benefit in real life indicated that directional microphones are perceived as beneficial.[104]

7.3.3 Interaction of directivity with other technologies

Directional processing occurs prior to any other signal processing in the hearing aid. It affects the SNR before the combined signal and noise are processed by compression, adaptive noise suppression, feedback cancellation, frequency shifting, or any other features, and the benefit it provides is independent of the benefit those other features provide, with one exception explained later in this section. The SNR advantage provided by directional microphones is additive with the advantage that bilateral hearing aids provide relative to a unilateral hearing aid.[705]

Compression can misleadingly give the appearance of undoing the advantage of a directional microphone. As a source is moved from the front to the rear, the directional microphone will attenuate its level. The compressor in the hearing aid will react to this lower level by increasing gain. In most real world situations, however, the noise that is of greatest concern is the noise that is present while the speech occurs. The SNR advantage enabled by the directional microphone will then still be preserved at the output of the compressor, because the compressor applies the same gain to whatever mixture of speech and noise the microphone is able to provide it with. Directional advantage will therefore be unaffected by compression.[1504]

The interaction of compression with directivity may have one unintended benefit. If a hearing aid is inadvertently switched to directional mode in a situation where the dominant sound comes from a wanted talker to the rear or sides, the compression will indeed partly undo the gain reduction that frontal directivity would otherwise apply. In other words, the disadvantages of directivity when it is inappropriately applied will be less than might be thought from the shape of the polar diagram.

An implication of the effect of compression on directivity when testing with one sound at a time is that to measure a directional polar pattern, or front-back ratio in a test box, it is necessary to either turn compression off, or use a test method that enables frontal and rearward signals to be presented simultaneously.[1498]

Similarly, the benefits of adaptive noise reduction systems in improving listening comfort (Section 8.1.2) are not affected by the improved SNR provided by a directional microphone.[164] It is not as if either a directional microphone or an adaptive noise reduction system is so effective at getting rid of noise that the other algorithm is no longer needed.

The one mechanism that can cause other signal processing features to affect directional benefit arises because of the vent path. If a feature, such as adaptive noise reduction, causes the hearing aid gain to reduce, then the frequency range over which the amplified sound path dominates the vent path also decreases. Consequently, the frequency range over which the directional microphone increases SNR also decreases. This does not mean that adaptive noise reduction should not be used.

Because directional microphones interact with the sound fields in the vicinity of the head, and are not affected by the processing that follows the directional microphone, all the aspects of directional microphones discussed in this chapter are just as relevant to their use in cochlear implants, where there is an even greater need for improving SNR than there is for hearing aids.

7.3.4 Disadvantages of directivity

Directional microphones have a range of disadvantages that in many situations outweigh their advantages. Most of these disadvantages have been alluded to earlier in the chapter, but are summarized here.

Target sounds to the rear or side: The major disadvantage of directional microphones is the inevitable consequence of their reason for existence – forward-looking directional microphone patterns attenuate sounds from the sides and rear, and if those sounds contain the target to which the hearing aid wearer wishes to listen, the directional microphone will create a disadvantage relative to an omni-directional pattern. If there is background noise, the directional microphone will decrease the SNR; if the target is in quiet, the directional microphone will decrease its level. If the source is already soft (e.g. less than 60 dB SPL), then decreasing the level will very likely decrease audibility, and intelligibility in the case of speech.[1000, 1502]

Low-cut response, or increased internal noise. Because of the inherent low cut of subtractive directional microphones, the directional response either has a low-frequency cut (if not fully compensated), greater internal noise (if the cut is fully compensated), or a lesser degree of both disadvantages (if partially compensated). This disadvantage is evident in listening situations with low levels of background noise, so automatic directional microphones invariably switch to omni-directional in these situations.

Reduced localization. Directional microphones reduce left-right localization accuracy if the two hearing aids adapt in an uncoordinated manner, but this minor effect must be weighed against the improved front-back localization that all directional microphones with a positive unidirectional index offer. Synchronized directivity in the two hearing aids obtains the benefit of directivity without any adverse effects on localization. Split-band directivity also helps avoid adversely affecting localization.

Wind noise. Directional microphones are more sensitive than omni microphones to wind noise.

Taken together, it is clearly not appropriate that hearing aids be full-time directional any more than that they be full-time omni-directional.[1000, 1460, 1502] The most common solution is for the hearing aid to switch between directional and omni-directional responses in different circumstances. This can be achieved by automatic switching, manual switching, or a combination of the two. Automatic switching is needed for the substantial proportion of people who are unwilling or unable to manually switch between directional and omni-directional as the listening situation varies. If automatic switching is used, then the decision must be based on more than just noise versus quiet, as it is common for the target signal to come from other than the front, even when background noise is present. As discussed in Section 7.1.1, automatic switching also enables different decisions to be taken in different frequency regions.

An alternative solution is for the hearing aid in one ear to be permanently set to directional and for the other hearing aid to be permanently set to omni-directional. Evaluation studies show that this combination is preferred to, and performs better than, permanent omni-directional responses in both ears.[106, 323a, 764a]

7.3.5 Candidates for directional microphones

Everyone who wears hearing aids can benefit from directional microphones. Every person, hearing-impaired and normal-hearing alike, experiences situations where the signal-to-noise ratio is sufficiently poor that understanding speech is difficult. It is just that the proportion of every-day listening situations for which such difficulties occur increases with patients'

SNR loss, and on average with the degree of hearing loss. In some of these difficult situations, either the target or the dominant masking noises will be sufficiently close that directional microphones, particularly adaptive directional microphones, will impart benefit. Whenever the physical location of targets and noise sources is such that the directional microphone objectively improves the SNR, then any listener, whether normal hearing, mildly hearing-impaired or profoundly hearing-impaired, and whether a new or experienced hearing aid user, will appreciate that improvement.[42, 989] Provided the microphone can be switched, or automatically switches, to omni-directional in those situations where the disadvantages of directional microphones outweigh their advantages, there is no reason not to make directional microphones available to all hearing aid wearers, even young children (Section 16.4.4).

The existence of directional benefit does not greatly depend on any characteristics of the hearing aid wearer.[1505] One caveat to this strong statement is that some patients have so much hearing loss at some frequencies they are unable to extract significant information from speech at those frequencies, even when the speech is amplified sufficiently to make it audible.[293] The potential benefit of a directional microphone relies on there being a significant frequency range for which the DI is greater than that of an omni-directional microphone, *and* for which speech is audible, *and* for which the hearing aid wearer is able to extract information from audible speech.

When DI is averaged across frequency to estimate benefit, the frequency range included should therefore include only the frequency range from which the aid wearer will be able to extract information. This will be a more restricted range for patients whose audiograms slope from near-normal hearing in the low frequencies to a severe or profound loss in the high frequencies than for other patients. Less dramatically, a patient with moderate low-frequency loss and severe high-frequency loss can likely extract more information from audible low-frequency speech regions than from equally audible high frequency speech regions (Section 10.3.4). High-frequency directivity will therefore be of less benefit than low-frequency directivity to that patient, implying that frequency weights different than those normally used to calculate the AI-DI would be more appropriate for predicting benefit in diffuse noise for such a patient.

Benefit for all patients does not imply that all patients will get the *same* degree of benefit from a particular directional technology. A patient with mild hearing loss in the low frequencies will likely have open-canal hearing aids, for which vent-transmitted sound may remove all DI below about 1000 Hz, and adversely affect the DI up to about 2000 Hz. The AI-DI for such a fitting will be much smaller than that for the same microphone technology in an occluded hearing aid fitted to a patient with more loss.[1001] The maximum degree of benefit in the open fitting will therefore be smaller, and so too will be the range of listening situations in which *any* benefit is detectable. Some benefit will still be measurable when the talker is close to the listener.[1826]

7.3.6 Evaluation of directional microphones in the clinic

It is inevitable that directional microphones will produce advantage in some listening situations, equivalence of performance in others, and disadvantage in yet others. Consequently, it is not possible to do a single behavioral assessment in the clinic that is indicative of "the" real-life benefit of the directional microphone. Any assessment made, such as by measuring the SRT_n in noise with the directional microphone and the omni-directional microphone, will just show the benefit in situations like the test situation. If this is in a sound-attenuating booth, which typically have a reverberation time of around 0.2 s, a critical distance of about 0.6 m, a distance to all loudspeakers of not much greater than this, and a small number of noise speakers (or perhaps just one), then a very substantial benefit of around 5 dB improvement in SNR will likely be measured, but in no way will this be typical of most real life situations.

What evaluation should the clinician do? There seems no point in doing anything other than ensuring that the directional microphone is working as expected, which can be achieved in some test boxes (Section 4.1.2). It is more important to counsel the patient about when to switch to directional if the hearing aid is manually switched, and to counsel the patient about controlling the listening situation so that the maximum benefits of directionality can be achieved. If a patient (or the clinician!) needs convincing that the directional microphone can ever help, then a speech test with a single noise speaker to the rear should provide a convincing demonstration.

7.4 Concluding comments

The greatest problem faced by hearing aid wearers is understanding speech in noisy places.[950, 1179] Unfortunately, hearing aid wearers communicate in noisy places more often than in quiet places.[1874] The best solution we have to offer is two hearing aids with directional microphones. Unfortunately, current directional microphones have very limited directionality. The benefit they offer (about 4 dB SNR improvement with close speech and diffuse noise for closed-ear hearing aids) is well worth having, but reverberation and distance often degrade benefit to a small fraction of this. Open fittings, and to a lesser extent, all vents in low-gain hearing aids, decrease the maximum attainable benefit obtainable even under ideal room acoustic conditions.

Clinical practice should be based on evidence wherever possible. There is no doubting the substantial efficacy of directional microphones when the acoustic conditions are conducive to benefit. By contrast the accumulated evidence makes it clear that their effectiveness, i.e. their real-world benefits, averaged across the range of situations encountered by hearing aid wearers, is very mild indeed; so mild that some hearing aid users cannot discern any differences between omni-directional and directional hearing aids. Self-report studies reassure us, however, that patients on average report benefit from directional microphones in just those situations where consideration of room acoustics would lead us to expect benefit. There seems no reason not to prescribe directional microphones for all clients, provided the hearing aids automatically switch (or can manually be switched) to omni-directional when the acoustics suggest that directional processing will be disadvantageous.

The automatic switching decision does not have to be extremely precise; in many if not most listening situations, neither the omni-directional not the directional response has any advantage over the other.[1753, 1874] Automatic switching will benefit even those patients who report hearing no difference between the programs, or who otherwise do not choose to switch between programs, as it is an acoustic inevitability[o] that a SNR benefit will occur when the talker location is close and frontal, and the dominant noise source is either close and rearward, or is distant. It is likely that there are more of these situations than those in which omni-directional processing is more advantageous.[1753] Provision of automatic directional microphones, with the option of disabling automatic processing for those patients who want maximum control, would therefore seem to offer more advantages to more people than having a default omni-directional program and a user-selectable "noise program" that included directional microphones.

The future emergence of high-performance bilateral directional processing is extremely exciting. Even with existing directional microphones, it is already true that directional microphones can enable hearing aid wearers with mild-moderate loss to hear as well in noise as normal-hearing people, but this occurs only in highly constrained listening situations.[109, 1106] Hearing aids are regarded as a visible sign of disability, and an indicator that the wearer will likely have difficulty understanding in noisy places. When the benefits of super-directional binaural arrays now being seen in research are attainable in wearable devices, hearing aids may come to be regarded as bionic devices that give the wearers super-human hearing.

Such a change of image will likely have a huge effect on the penetration rate of hearing aids. Even normal-hearing people may want them as hearing enhancers, a function they could easily combine with providing the audio interface to mobile phones, music players and hand-held portable computers (all of which are in any event merging). Reverberation and distance will still degrade performance of super-directional hearing aids, but directional benefits will be obtained across a much wider range of listening situations than is now the case.

Fitting high-directivity bilateral devices will be a major challenge to hearing aid designers and clinicians alike. When directivity is very high, not having it when it could benefit will cause a major loss, just as having it when it is inappropriate will cause major difficulties. The challenges for the future will be to find ways to make the hearing aid super-directional only when it is needed and to make the directional beam "look" where the hearing aid wearer wants to listen. The task of automatically tailoring directivity to each listening environment so that super-directional microphones live up to their potential will keep hearing aid engineers employed for many years.

[o] The degree of directional benefit is well predicted by measurement, with a directional and an omni-directional microphone, of the speech transmission index (STI), which is a totally objective, acoustic measurement.[1509]

CHAPTER 8

ADVANCED SIGNAL PROCESSING SCHEMES

Synopsis

Adaptive noise reduction schemes, such as Wiener Filtering and Spectral Subtraction, progressively decrease the gain within each frequency region as the SNR deteriorates. Although they generally improve sound comfort and the overall SNR, these schemes do not change the SNR in any narrow frequency band. Consequently, they do not generally improve intelligibility. Other types of noise reduction include wind noise reduction, achieved by a low-frequency cut, and transient or impulsive noise reduction, achieved by limiting the rate at which the waveform changes.

Feedback oscillation can be made less likely by several electronic means. One simple technique is to decrease the gain only for those frequencies and input levels at which oscillation is likely. A second technique is to modify the phase response of the hearing aid so that the phase shift needed for oscillation does not occur at any frequency for which there is enough gain to cause feedback oscillation. A third technique involves adding a controlled internal negative feedback path that continuously adapts to maintain the gain and phase response needed to cancel the accidental leakage around the earmold or shell. A final technique involves making the output frequency different from the input frequency. Often, a combination of these techniques is used.

High-frequency components of speech can be made more audible by lowering their frequency. This can be achieved by transposition: moving sections of the spectrum to lower frequencies and superimposing them on the spectrum already in the lower frequency range. Alternatively, frequency compression is used to compress a wide frequency range into a narrower (and lower) one. While both frequency transposition and frequency compression can guarantee audibility of high-frequency sounds, they do not necessarily guarantee better intelligibility, as the speech components shifted down in frequency may interfere with perception of the speech components that were originally dominant in this lower frequency range. The range of possible frequency lowering methods, frequency transformation maps, and gain characteristics is large. Finding the best combination is made more difficult by the time it takes people to adapt to a highly altered spectrum and by our present uncertainty over how best to evaluate success.

There are various theoretically appealing methods for enhancing features of speech that have been tried in research experiments. These include exaggerating the peaks and troughs in the spectrum of a speech sound, increasing the amount of amplification whenever a consonant occurs, increasing the amplitude at the onset of sounds, lengthening and shortening the duration of particular sounds, simplifying speech down to a few rapidly changing pure tones, and resynthesizing clean speech based on the output of an automated speech recognizer. On the evidence available so far, however, none of these techniques will produce a worthwhile increase in intelligibility compared to conventional amplification, so there is as yet little motivation to include the processing within commercial hearing aids.

Various other signal processing algorithms have already been implemented in commercial hearing aids, or could readily be implemented. Reverberant energy that does not overlap other speech sounds can be removed, giving a crisper sound quality. Hearing aids can automatically categorize the listening environment they are in, and select amplification characteristics that have been pre-programmed into the hearing aid for each type of environment. Their data-logging systems can record how often each environment is encountered, and how the user adjusts the hearing aids in each environment. Trainable hearing aids can learn from the adjustments the aid wearer makes, and infer how the aid wearer likes the hearing aid to be adjusted as the acoustics of the listening situation vary. Fine tuning is therefore carried out by the hearing aid and the client together, rather than by the clinician. Active occlusion reduction processing enables a hearing aid to sound like an open-canal hearing aid, despite the ear canal being completely blocked. In addition to cancelling the occlusion-induced sound, active occlusion reduction cancels the vent-transmitted sound, enabling directional microphones to work over the entire frequency range, increasing their efficacy.

This chapter describes several advanced processing schemes for hearing aids. Some of these have been available in commercial hearing aids for several years; some are likely to become available over the next few years.

8.1 Adaptive Noise Reduction

Improving speech intelligibility by removing noise, other than with the use of more than one microphone, is a very, very hard problem that has not really been solved. As Levitt (1997) has insightfully said:

> *Our understanding of this problem is so limited that we have not only been unsuccessful in finding a solution, but we do not even know whether it is possible to improve the intelligibility of speech in noise by any significant amount*.

Noise can, however, be reduced in such a manner that it improves listening comfort, and reduces listening effort, and there are several techniques that accomplish this. ***Adaptive noise reduction*** techniques have various synonyms, including ***noise suppression***, ***fine-scale noise cancelling***, ***single-microphone noise reduction***, and ***digital noise reduction***. The last of these terms has become most common, but is not very descriptive in an era where the processing in almost every hearing aid is digital.

8.1.1 Adaptive noise reduction technology

The direct aim of adaptive noise reduction is to provide less amplification to noise than to speech. This is achieved by identifying segments (frequency ranges, moments in time, or both) where noise is particularly intense relative to speech, and applying less amplification to these segments than to other segments where the SNR is better. If this aim is achieved, it is likely that the hearing aid wearer will find the noise less troublesome, but it is also likely that speech intelligibility will not be increased or decreased. The reasons why we might expect noise reduction to be beneficial have already been explained in the context of compression to reduce noise in Section 6.3.7, which should be reviewed before reading further. In brief, reducing the level of frequency regions and/or moments in time where noise is particularly intense will reduce the saliency of noise, and will minimize the likelihood that it masks speech segments at other frequencies and/or immediately later in time.[875] Reducing the gain in one frequency region, at one moment of time, will of course reduce the level of both speech and noise in that frequency region at that time. That is, the SNR in that segment will remain unchanged, which is the basic reason why noise reduction techniques usually leave intelligibility unchanged.

For noise reduction to be possible, the hearing aid has to detect speech, estimate the speech and noise levels at some points in time and across frequency, determine what the gain reduction (if any) for each frequency region should be, and determine how rapidly that gain reduction should change over time.

Detecting speech and noise

Speech sounds are created by successively opening and restricting the vocal tract, which modulates the level (and spectrum) of the periodic vibrations coming from the vocal cords and the random turbulence caused by air flow past restrictions in the vocal tract. This variation, or modulation, of the amplitude of speech is the primary characteristic used to detect the presence of speech. The amplitude (and hence the envelope) of speech thus increases and decreases with a frequency varying across the 3 to 6 Hz range, corresponding roughly to the rate at which syllables and words are produced.[a] As most noises do not fluctuate in this manner, speech is likely to be present whenever the ***envelope power*** in the 3 to 6 Hz range is sufficiently greater than the envelope power at other frequencies. When the speech power is at a maximum, the power is likely to be large simultaneously at many frequencies, so the same 3 to 6 Hz fluctuations should be detectable in more than one of the hearing aid channels. This ***co-modulation*** provides a second clue that speech is present.

A third clue comes from the fine structure of speech. Every time the vocal cords open, a burst of power across a wide frequency range occurs, and these bursts repeat at the pitch rate of the voice (from around 100 Hz for a male up to around 400 Hz for a child). The

---

[a] The envelope has frequency components extending down below 1 Hz and up above 40 Hz, but the power in the envelope becomes progressively weaker outside the dominant 3 to 6 Hz range. The maximum of the modulation spectrum is around 3 Hz for the lower frequency bands dominated by vowels and progressively moves up to around 6 Hz for the higher frequency bands dominated by consonants.[175]

presence of these bursts, at typical pitch rates, synchronized across multiple frequency channels within the hearing aid thus provides further confirmation that speech is indeed present.[305]

Different hearing aids have a ***speech/non-speech detector*** or ***voice activity detector*** (***VAD***) constructed using different combinations of these speech characteristics, or other more esoteric statistical characteristics of speech such as the distribution of amplitudes across time, or the rate at which the spectral shape changes with time.

Estimating speech and noise levels

The same modulations used to detect speech can also be used to estimate the SNR within each channel (i.e. frequency region). Figure 8.1 shows the envelope of a speech signal in the presence of noise. The eye quickly discerns that there is something present with a relatively constant level around 55 dB SPL during the gaps between each word. This is the steady background noise, whereas the peak levels of the envelope represent the maximum level of each speech sound. Electronic detectors track the maxima and minima of the envelope, and the difference between them is the ***modulation depth*** of the signal. These operations are carried out separately within each hearing aid channel, so the modulation depth in each frequency range is known.

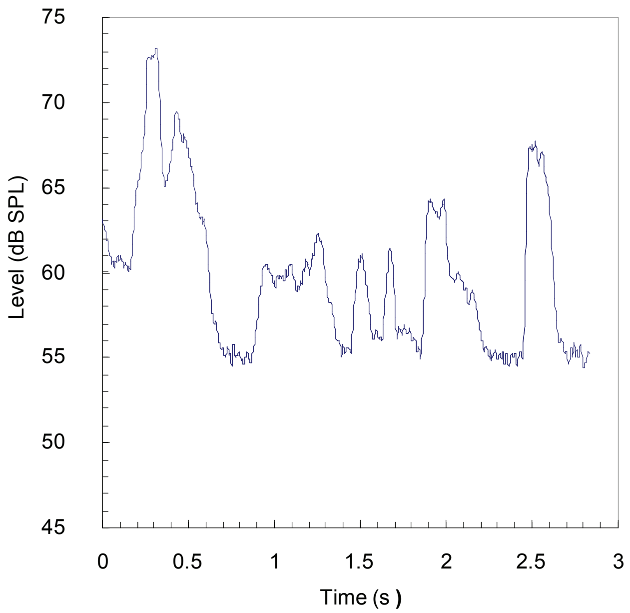

Figure 8.1 Envelope of the sentence, The yellow flower has a big bud, in the presence of background noise with a steady level of 55 dB SPL.

If the maxima and minima are averaged across several seconds, the long-term SNR can be estimated. If the near-instantaneous maxima are used, then the short-term SNR, which of course varies rapidly, can be estimated.

The SNR is smaller than the modulation depth (typically by about 10 dB), because the SNR is the difference between the average speech level and average noise level, whereas the modulation depth is the difference between the peaks of the envelope and the average level of the noise. The exact difference between modulation depth and SNR depends on the time constants with which the speech envelope is detected, but for any particular set of time constants, every 1 dB increase in modulation depth corresponds approximately to a 1 dB increase in SNR.

This modulation depth approach to estimating SNR works well when the wanted signal is a single talker and the noise is a continuous babble, or other noise with few fluctuations in level at the syllable or word rate. By contrast, the hearing aid is likely to make a very poor decision about SNR when the wanted signal has little fluctuations (like some music), and the noise has marked fluctuations (like a single nearby talker).

Information about the presence of speech and estimated SNR is also used by some hearing aids to automatically select the most appropriate microphone configuration (directional versus omni-directional, or the most appropriate polar pattern), as well as to control adaptive noise reduction.

Gain reduction algorithms

There are several ways to perform adaptive noise reduction, but most systems use a variety of either Wiener Filtering or Spectral Subtraction.

A ***Wiener Filter*** is a filter whose gain at each frequency depends in a particular way on the SNR at that frequency. Specifically, the gain equals the signal power divided by the sum of the signal power plus noise power:

$$W(f) = \frac{s(f)}{s(f) + n(f)} \qquad \ldots 8.1,$$

where $s(f)$ is the power spectrum of the signal and $n(f)$ is the power spectrum of the noise. It can be shown mathematically that of all possible filter shapes, the Wiener Filter makes the waveform at the filter output as similar as possible to the signal (without noise) at

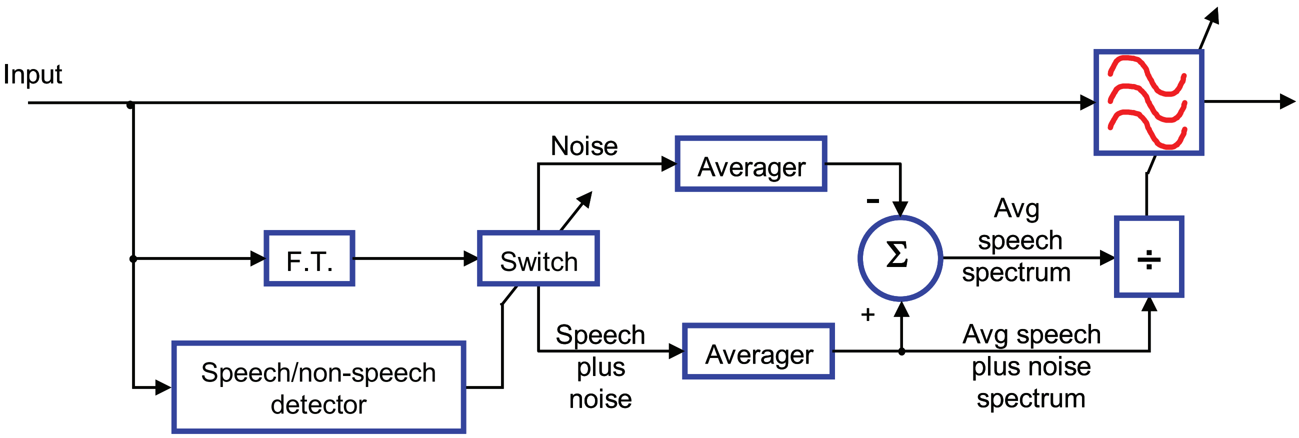

Figure 8.2 A Wiener Filter incorporating a Fourier Transform (F.T) to calculate the spectrum of the combined speech and noise. A speech/non-speech detector classifies the spectrum as noise or speech plus noise, and thus enables the average spectral power of the speech to be estimated.

the input. The problem with making such a filter is evident: how can the filter's gain be calculated when background noise prevents us from knowing the power of the signal at any frequency? All we know is the instantaneous power at each frequency of the signal plus noise, because that is what the microphone, whether omni-directional or directional, is delivering to the hearing aid signal processor.

The answer is that the signal power has to be estimated as described in the preceding sub-section. If we can estimate the spectral power (averaged over some preceding short or long time) of the noise when there is no speech, and also of the speech plus noise, then we can subtract the first of these from the second to estimate the power of the speech alone. Figure 8.2 shows the block diagram of a Wiener Filter that uses this principle. The filter is being controlled by a signal that reflects the SNR at each frequency.

The essential characteristic of the Wiener Filter is that gain is reduced as the SNR deteriorates. Equation 8.1 can easily be rearranged to show that W(f) depends explicitly on SNR:

$$W(f) = \frac{snr(f)}{snr(f)+1} \qquad \ldots 8.2,$$

Where *snr(f)* is the SNR at each frequency *f*, expressed as the ratio of signal power to noise power (rather than the decibel version we usually refer to).

Figure 8.3 shows two variations of gain with SNR. The curved solid line shows the degree of gain reduction calculated according to equation 8.2. The straight dashed line shows an approximation of the type often used in hearing aids. The exact position of this line varies from manufacturer to manufacturer, and varies as the noise reduction setting (e.g. "strong" or "moderate") is changed for a particular hearing aid. That is, there is variation in how much the gain is reduced at very poor SNRs (anywhere from 6 to 24 dB gain reduction), and variation in how good the SNR has to be before the full gain is applied.[305, 475]

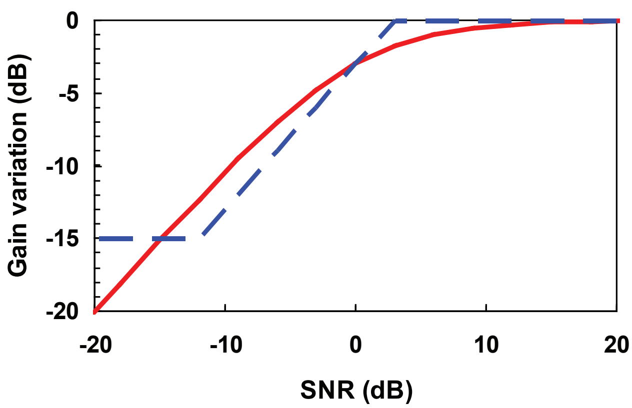

Figure 8.3 Attenuation provided by noise reduction systems for different SNRs. Solid line shows the Wiener Filter gain reduction of equation 8.2, and the dashed line shows the type of straight-line approximation used in many hearing aids.

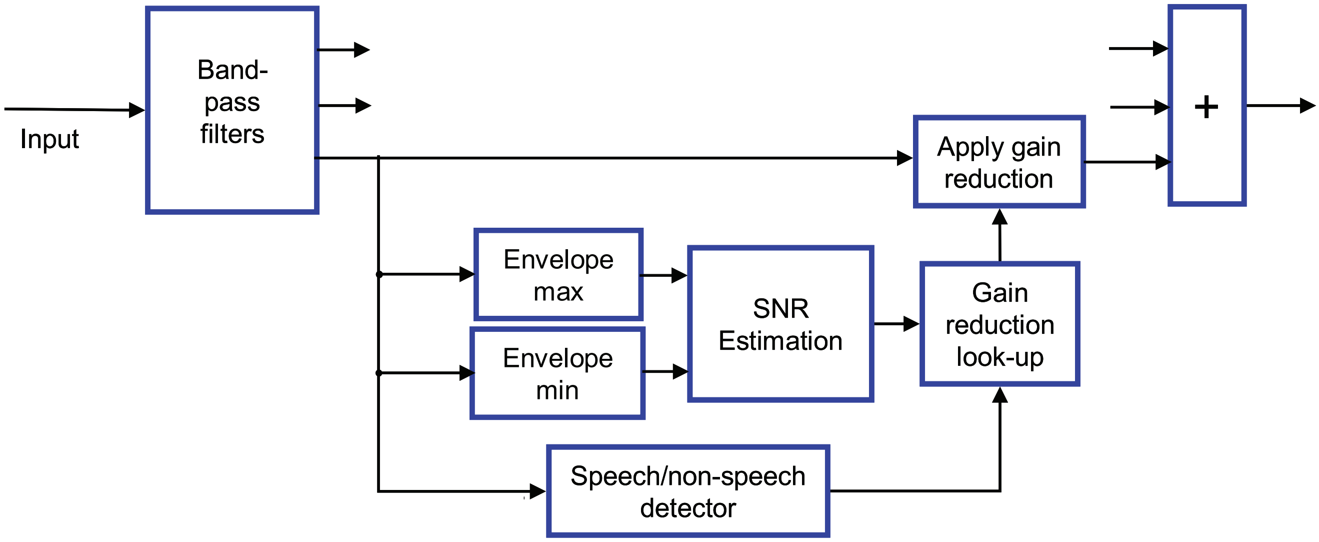

Figure 8.4 An adaptive noise reduction system that calculates the gain reduction based on the modulation of the envelope in each channel of the hearing aid. Processing is shown for just one channel. The speech/non-speech detector that determines whether adaptive noise reduction is enabled is likely to receive input from several channels.

The signal processing needed to accomplish adaptive noise reduction, using modulation depth to estimate SNR, is not complex, and an example is shown in Figure 8.4. Depending on the particular gain reduction rule used (e.g. the blue dashed line in Figure 8.3) the result may be very similar to Wiener filtering.

Unfortunately, there is insufficient research to guide exactly how much gain should be reduced at any SNR, which is why hearing aid engineers often design in a "strength" control and leave the choice to the clinician (who of course also has no way of knowing which is best, other than by trial and error). The optimal gain changes certainly are not insignificant. One experimental evaluation found that across a range of noises, hearing aid wearers, on average, preferred the shape of the gain-frequency response to change by 1 dB for every 2 dB variation in the shape of the SNR versus frequency contour.[867] There is, however, a variation across patients in the amount of noise reduction that is preferred.[1956]

It is important that gain reduction never be of sufficient degree that it decreases the amount of speech that is audible, or it will decrease speech intelligibility.[1482] In principle, the gain reduction should be most marked at the highest input levels, and least marked for the greatest hearing losses. This can be achieved by ensuring that attenuation is never greater than the amount that causes the noise level to be decreased to hearing threshold, which in turn is achieved by keeping the insertion gain (after noise reduction has been applied) greater than the value shown in equation 8.3.

$$REIG > MAF + HL - N \qquad \ldots 8.3,$$

where MAF is the normal minimum audible field, HL the hearing loss, and N the noise level measured in bands equivalent to the auditory filter bandwidth of the listener. As an approximation, 1/3 octave bandwidths can be used for mild loss, rising to octave bandwidths for severe hearing loss. Application of equation 8.3 will show that REIG often can usefully be reduced to negative values. Negative insertion gain (i.e. an earplug) is achievable in the high frequencies if earmolds or ear shells with small vents are used, but is not achievable in the low frequencies (where it most often will be called for) unless active noise cancelling, such as that provided by active occlusion reduction processing (Section 8.5), is also installed in the hearing aid.

In some cases (low or medium speech and noise levels, and severe hearing loss) achieving Equation 8.3 will require the noise "reduction" algorithm to *increase* gain at some frequencies. Some hearing aids already include gain increases at some frequencies when it is necessary to do so to maximize speech audibility.[1007]

Optimizing the gain-frequency response for each background noise via adaptive noise reduction is likely to be most important for people with the greatest amount of loss, because of the greater potential for spread of masking, and the greater difficulty in fitting sounds into a restricted dynamic range.[867] Varying gain reduction in a gradual manner as any one of SNR, hearing loss, and input level varies is a more sophisticated approach that achieves the benefits of adaptive noise reduction without decreasing the audibility of speech.[1007]

An alternative to Wiener Filtering is ***Spectral Subtraction***. The magnitude (i.e. the amplitude) of the noise spectrum is subtracted from the magnitude of the speech plus noise spectrum. If both magnitudes are known exactly, then the difference will be the magnitude of the speech spectrum alone. The obvious problem with this system is determining the spectrum of the noise, because the microphone picks up the speech and noise combined.

One solution is to estimate the noise spectrum currently present by averaging the noise spectrum that was present during some preceding moments in time, just as for Wiener filtering. Of course, we can measure this preceding spectrum only if we know when the noise was present by itself. Consequently, the Spectral Subtraction system also needs a speech/non-speech detector, as shown in Figure 8.5. Only the magnitudes of the speech plus noise signal are corrected by the processing. There is no known way to estimate what the phase of the speech alone should be, so the final inverse Fourier Transform usually uses the phase of the original speech plus noise. If the spectral subtraction process works perfectly, the magnitude of the output spectrum equals the magnitude of the speech spectrum alone, but the phase spectrum remains corrupted by the noise. This adversely affects the quality of the speech, but may be preferable to not removing any of the noise.

A greater problem for Spectral Subtraction (and for a Wiener Filter that uses the same method to estimate noise) is that the noise spectrum estimate is based on the noise characteristics during the preceding seconds (or fractions of a second). Unfortunately, just like speech, background noise can change its character entirely within a short time. In this case, both types of noise reducers are trying to remove a noise that is no longer present! Furthermore, they know nothing, and can therefore do nothing, about some new noise that has just commenced, or about noise that has just changed its character. Both types of system are most suitable for steady noises (technically called ***stationary noises***). These would include some machinery noise, air-conditioning noise, distant traffic noise, and to a lesser extent, the babble from a large number of other people. Lessening the effects of a single competing talker would be especially difficult with either Wiener Filtering or Spectral Subtraction.

Although Wiener Filtering and Spectral Subtraction may seem to work on different principles, they have similar effects on a noisy signal. Both decrease the gain most at those frequencies where the SNR is worst, and leave the signal unaltered when there is little noise present. In fact, for some implementations of each, they have mathematically identical effects on the input signal.[1066] The precise acoustic effects of each scheme depend on the implementation details.

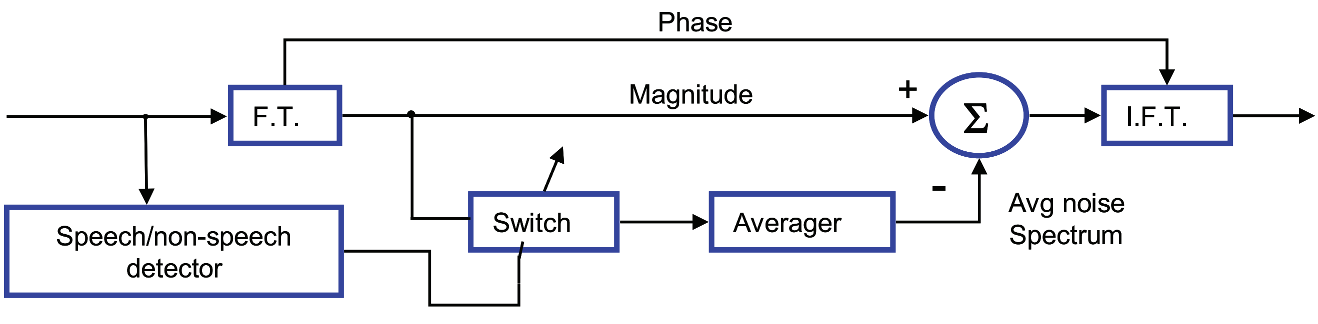

Figure 8.5 A Spectral Subtraction noise reduction system incorporating a Fourier Transform to calculate the power spectrum, a speech/non-speech detector to enable the average spectral power of the noise to be estimated, and an Inverse Fourier Transform to turn the corrected spectrum back into a waveform.

One important detail is the frequency resolution with which signal and noise power is determined. If a very narrow bandwidth is used, SNR will be greatest at the harmonic frequencies of sustained periodic sounds like vowels. The resulting filter shape will have a high gain at the frequency of each harmonic, and will provide a large attenuation mid-way between harmonics. Because of its alternating, spiky shape, such a filter is referred to as a ***comb filter***. Comb filters are very effective at removing noise, but during rapidly changing elements of speech, such as formant transitions, they are also likely to distort the speech signal. Comb filters can be generated by other techniques, such as basing the filter shape on the fundamental frequency of the speech.[1074] Unfortunately, the accuracy of fundamental-frequency extraction is adversely affected by noise, and consequently the comb filter passes inappropriate frequencies just when filtering is most needed.[1956a]

Another important detail is the frequency range over which gain reduction occurs. Some hearing aids allow gain to be reduced over the entire spectrum, others allow only low-frequency gain to be reduced, and yet others allow gain at any frequency other than the mid-frequencies to be reduced.[99] The rationale for not allowing mid-frequency gain to be reduced, or for limiting the amount by which it is reduced, is that the mid-frequencies contribute most strongly to intelligibility, so these hearing aids have less risk of inadvertently reducing the audibility of speech in the mid frequencies.

Noise reduction dynamics

Adaptive noise reduction systems can be designed to vary the gain reduction applied every few milliseconds, or take many seconds before they respond in any way to noise, and then to vary the gain gradually over a duration of several seconds. Both approaches have advantages and disadvantages.

We can define ***onset time*** as the time from when noise commences to when the gain has reduced to within 3 dB of final value.[99]This onset time can vary from a couple of seconds to more than 30 seconds.

Similarly, we can define ***offset time*** as the time from when noise ceases to when the gain has been restored to within 3 dB of the value it has in quiet. This offset time can vary from 5 ms to several seconds.[1263]

Spectral subtraction is inherently very fast acting, as the subtraction occurs separately for each brief segment of speech analyzed. Analysis frames in hearing aids are typically 4 to 8 ms long. Wiener Filtering, however, can also be implemented such that the gain in each frequency region varies every few ms, or the rate of gain variation can intentionally be slowed down so that it takes several seconds for the gain to change significantly. This slower approach is intended to react to changes in the listening environment.

The advantage of fast-acting noise reduction is that noise between words and syllables in the speech is reduced, not just noise in frequency regions where the noise dominates. The disadvantage of fast-acting noise reduction is that the rapid changes in gain can distort speech quality. Fast-acting noise reduction therefore has the greatest potential to improve speech comfort and even intelligibility, but runs the greatest risk of producing processing artifacts, especially if the speech detector mistakes lower level speech components for noise. One comparative evaluation has shown a preference for a system with a 16 second onset time over a system with a 4 second onset time.[102]

8.1.2 Adaptive noise reduction benefits

Adaptive noise reduction systems can be expected to improve the overall SNR when the levels of the signal and noise are measured objectively at the output of the hearing aid. Figure 6.12 provided an example where the SNR at the output will be much greater than that at the input. Unfortunately, this improved SNR does not generally result in any increase in intelligibility.[102, 164, 385, 1045, 1047, 1050, 1066, 1080, 1266, 1317, 1348, 1510, 1875, 1956]

The reasons underlying this have already been dealt with in Section 6.3.7. Essentially, if a hearing aid has a single microphone port, then when a noise and signal occur at the same time and at the same frequency there is no known way by which they can be separated. In principle, improving intelligibility in noise with only a single input relies on the signal and noise having components sufficiently different in frequency or in time to be separable by signal processing, but not by a person with impaired hearing without the processing. This *seems* like an achievable goal, especially for people with severe and profound hearing impairment, as these people have the most decreased frequency and temporal discrimination abilities, but finding a solution that provides a significant improvement in intelligibility has so far largely eluded researchers.

When noise is restricted to a narrow frequency region, adaptive noise reduction can lead to a substantial increase in intelligibility, as attenuation of this narrow frequency region can decrease the masking caused by the noise.[320, 1482, 1484, 1841] So far, just a few studies have shown a statistically significant increase in intelligibility (equivalent to around 1 dB improvement in SNR) from adaptive noise reduction for realistic, broadband noises.[b, 170, 1001]

Despite the general lack of speech intelligibility benefits from adaptive noise reduction (because SNR at each frequency is unaffected), adaptive noise processing is almost always preferred for comfort, aversiveness, ease of listening, quality or overall preference (because broadband SNR is improved).[102, 164, 385, 1266, 1510, 1956] When adaptive noise reduction is enabled, hearing aid wearers can accept an increase of several dB in background noise level at the hearing aid input, and hence a decrease in the input SNR, indicating that the processing really does decrease the perceived level of the noise relative to the speech.[385, 1266]

This advantage can be checked for an individual patient with the acceptable noise level (ANL) test (Section 9.1.5)[1296] or by adjusting the input SNR with the processing on until the sound quality matches sound quality with the processing off.[385] The amount by which noise reduction processing subjectively improves the SNR is greatest for steady noises and for noises with long-term spectra dissimilar to speech. The improvement is least for multi-talker babble, which of course has a spectrum very similar to that of speech. For one spectral subtraction system that actually caused a small decrease in speech intelligibility, the subjective improvement in SNR was 9 dB for traffic noise, and 4 dB for speech babble, irrespective of listening task (comfort, preference, noisiness).[385] The precise amounts doubtless depend on the specific noise reduction algorithm.

Even though we have no expectation of improved speech intelligibility, adaptive noise reduction should routinely be enabled whenever the hearing aid senses that significant levels of background noise are present, because a decrease in perceived noise is extremely likely. Improved ease of listening is likely, and it is a very worthwhile outcome. It may enable a hearing aid wearer to communicate for longer in a noisy, tiring situation, or may free up cognitive resources to cope with the many demands of communicating in noisy places. Adaptive noise reduction has been shown to enable improved performance in secondary tasks, such as memorizing, or responding quickly to an event, performed while listening to speech in noise. Presumably this increased ease of simultaneously doing other tasks is reflecting reduced listening effort.[1547] There is considerable scope for further research in this area.

Although adaptive noise reduction algorithms and adaptive microphone arrays have developed independently of each other, and have mostly been implemented as separate algorithms working sequentially within the hearing aid, hearing aid designers are starting make them work interactively. The advantages are that both estimate how much of the signal picked up by the microphone(s) is actually noise and both can make better decisions about how to process sounds if they have the benefit of the SNR improvement offered by the other.

8.1.3 Impulse noise reduction

Speech sounds are made by the human vocal tract. There are limits to how rapidly the vocal tract can change, and hence how rapidly the instantaneous pressure of a speech waveform can change. Some sound sources, such as a hammer hitting a nail, have no such limitations, so impulsive sounds can contain an extremely rapid variation of sound pressure with time.

A smart hearing aid can recognize when the pressure is changing too rapidly for the signal to be speech, and deliberately not reproduce the rapid rise and fall of such sounds. Such ***impulsive sound smoothing*** or ***transient loudness reduction*** will therefore reduce the loudness, and hence annoyance, of the impulse sound (without completely removing it), while having little or no net effect on any speech sound that occurs at the same time. While the impulsive smoothing algorithm certainly changes the speech waveform during the impulse sound, this part of the speech waveform would have been inaudible to the user without any signal processing, because the high intensity of the impulsive sound masks the speech sound for the duration of the impulsive sound. Consequently, impulsive sound smoothing gives increased loudness comfort, with no change in intelligibility.[885]

[b] Some other studies have also reported small improvements in intelligibility from adaptive noise reduction, but in each case the improvement has failed to reach statistical significance, and therefore cannot be relied upon.

8.2 Feedback Reduction

The cause of feedback oscillation has already been discussed in Section 4.7.1. In the following sections we will examine several electronic methods for addressing the problem of feedback. Any of the following methods can help, but none of them can banish feedback oscillations entirely. The need for carefully made impressions and earmolds or shells is unlikely to disappear! Feedback path cancellation has, however, made it much easier to simultaneously have the ear open enough to avoid occlusion problems, but closed enough to avoid feedback oscillation and achieve sufficient gain to make the hearing aid fitting worthwhile.

8.2.1 Feedback reduction by gain-frequency response control

For feedback oscillation of a specific frequency to occur, the gain from microphone inlet to ear canal must be greater than the attenuation from the ear canal back to the microphone at that frequency. Furthermore, at this same frequency, the phase shift around the entire loop must be close to an integral number of periods (Section 4.7.1). Not surprisingly, one way to avoid feedback oscillations is to decrease the gain at all those frequencies where these conditions are met. This can be done in several ways.

The simplest way is to turn the volume control or fitter gain control down below the point required by the patient. This is obviously unsatisfactory, as it will give the patient inadequate loudness, audibility, and intelligibility. A better alternative is to decrease the gain at only those frequencies where feedback oscillation is a possibility. This is most likely to be at or near the peaks of the gain-frequency response curve, so anything that decreases the gain at these peaks without reducing the gain elsewhere is likely to be beneficial. Acoustic damping in the sound tube meets this criterion particularly well (see Section 5.5). Unfortunately, it may not always be possible to damp the particular peaks causing the feedback oscillation without excessively decreasing the gain in the frequency region around some other resonances.

Multichannel hearing aids provide a more reliable way to decrease gain in only one frequency region. The degree of control over the gain-frequency response is extremely coarse, however, unless the hearing aid has many parallel channels. If there are only a few channels, gain may be decreased over an unnecessarily wide frequency range, again resulting in inadequate audibility. Figure 8.6 shows the gain-frequency response of a four-channel hearing aid that has had the gain decreased in one band to decrease the incidence of feedback.

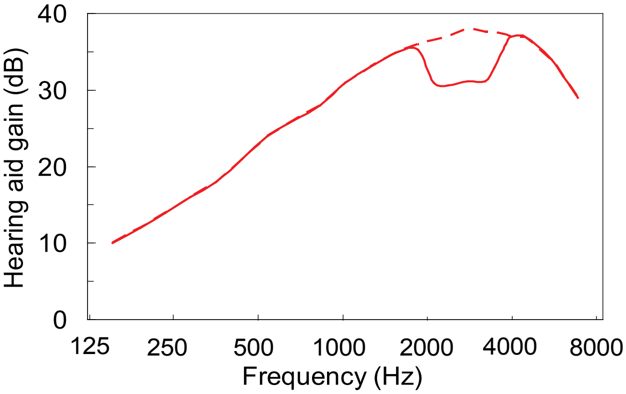

Figure 8.6 The gain-frequency response of a (hypothetical) four-channel hearing aid, where feedback oscillation has been avoided by decreasing the gain of the band from 2 kHz to 4 kHz (solid line) from the original response (dashed line).

Often, feedback occurs only when the volume control is increased above the aid wearer's usual setting, or at low input levels because of the effect of wide dynamic range compression on gain. In such cases it is necessary to decrease the gain in a narrow frequency region only under these specific conditions. Hearing aids can therefore avoid feedback oscillation by limiting the maximum gain that can be achieved in each frequency region. (The limit depends on the tightness of fit of the earmold or ear shell.) When feedback is not a problem, the full desired gain-frequency response is provided to the hearing aid wearer. When the overall gain is increased (either manually, or automatically by compression) the gain in frequency regions likely to cause oscillations can then be held down to a safe value.

The safe value can be determined by:

- the clinician at the time of fitting – this is achieved by the clinician selecting the maximum gain that just avoids oscillation, or by the clinician increasing the compression threshold (and hence the gain for low input levels) until oscillation ceases;[c]

- the fitting system at the time of fitting – this is achieved by performing an in-situ feedback test, in which the fitting system automatically raises the gain in each channel until it detects oscillation occurring;
- the hearing aid whenever the hearing aid is worn – this is achieved by the hearing aid reducing the gain in a channel whenever it detects oscillation occurring in that channel, but allowing the full prescribed amplification when oscillation is not present.

Digital filters (e.g. Figure 2.15) can provide even finer control of the gain-frequency response shape. Once the frequencies that can cause feedback oscillation are identified, narrow notches can be placed in the gain-frequency response around each of these frequencies. Public address systems have used this technique effectively for many years. It *seems* like this technique should do away with feedback oscillation altogether without excessively reducing audibility.

Unfortunately, the frequencies at which feedback occurs do not remain fixed over time. Remember that oscillations occur at the frequency with the correct phase response. If the earmold were to move a little in the ear, or the person were to move their jaw (and hence their temporo-mandibular joint and ear canal wall), wear a hat, put their hand near their ear, stand near a wall, or embrace a loved one, the characteristics of the leakage path can change, so the oscillation frequency can also change. This means that a notch is now needed at some other frequency. Oscillation could be totally prevented only if the notches are wide enough and numerous enough to cover all the possible oscillation frequencies, in which case there may be little gain left at any frequency.

Gain reduction, however implemented, represents a loss to the client. The degree of loss is minimized if the gain reduction is adaptive – that is, if it occurs only when oscillation is detected. Many hearing aids continually monitor their output to detect feedback oscillation, measure the oscillation frequency, and automatically adjust the gain-frequency response to prevent oscillation from continuing. Such automatic or adaptive gain reduction systems have been referred to as ***search and destroy*** feedback control.

Reasons for using electronic feedback control

An effective electronic feedback control is useful in the following circumstances:

- When more gain is needed. This is particularly useful for people with severe and profound losses, or people who would like a smaller hearing aid style than could otherwise be provided without feedback.
- When a more open earmold or earshell is needed. This is particularly useful for people with mild loss at low frequencies and severe loss at higher frequencies.

8.2.2 Feedback reduction by phase control

The previous section gave several methods by which the gain of the hearing aid at problem frequencies can be decreased. Some of these methods inadvertently vary the phase response of the amplifier.[d] Depending on luck, this phase variation may help prevent feedback oscillation, or may make it worse! In principle, the phase can be intentionally manipulated to reduce the likelihood of feedback oscillation. As we will see, the technique is not effective in digital hearing aids but it is worth understanding the technique to increase our knowledge of the feedback mechanism, and the principle behind active occlusion cancellation (Section 8.5).

The aim of phase control is to ensure that at any frequency where the gain is large enough to cause oscillations, the phase response around the loop causes the feedback to be negative rather than positive.

[c] Avoiding feedback by increasing compression threshold has the advantage of leaving intact the gain for high-level, and possibly mid-level, inputs.

[d] Electronic filters affect the phase response of a hearing aid, as well as affecting the gain response.

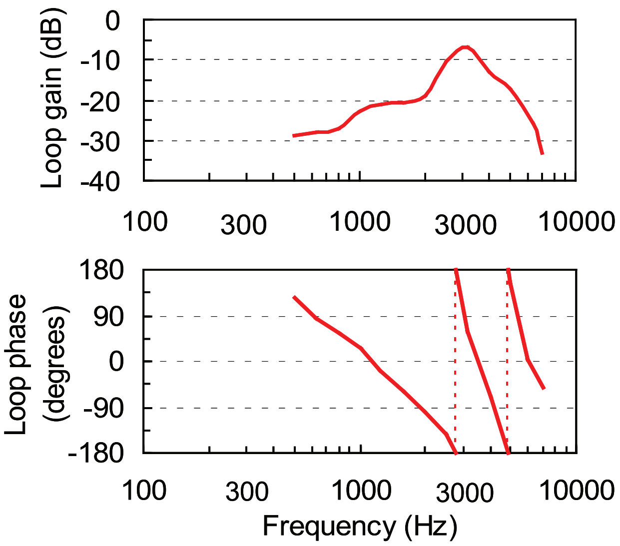

Figure 8.7 Gain-frequency and phase-frequency response of the complete feedback loop for an ITE hearing aid. Redrawn from Hellgren et al., (1999).

Figure 8.7 shows the gain-frequency response and phase-frequency response of the entire ***feedback loop*** for an analog ITE hearing aid. These responses are measured by breaking the connection between the amplifier and the receiver, and then injecting a test signal into the receiver. The test signal travels out of the receiver, back round the leakage path to the microphone, into the amplifier and finally back to the point that normally connects to the receiver. The magnitude and phase of the test signal reaching this point then shows the response of the entire feedback loop. It can be seen that phase is zero at the frequencies of 1200, 3500, and 6000 Hz. Feedback oscillation can therefore occur most easily at any one of these frequencies. Which one depends on the loop gain. For the volume control setting at which the measurement was performed, the loop gain is negative at all frequencies. If the volume control setting were to be increased, it is evident that of these potential feedback frequencies, the loop gain would first be positive at 3500 Hz. This, therefore, is the most problematic frequency.

If the phase response at 3500 Hz were to be modified so that it was closer to 180° instead of 0°, oscillation would no longer be possible at this frequency. Phase response can be manipulated by adding an ***all-pass filter***: a filter that has the same gain at all frequencies, but which affects the phase at some or all frequencies. Of course, once the first problem frequency has been tamed, (and the gain increased by a few dB) some other frequency will become the problem, and the phase must also be corrected here. The delay inherent in digital hearing aids means that phase changes very rapidly with frequency, making it impossible to correct the phase across a range of frequencies. Like the gain adjustment method, therefore, this method can thus allow additional gain to be achieved without feedback oscillation but only to a very limited degree.

Hearing aid designers previously used rudimentary control of phase in analog devices. Reversing the connections to the earphone (when possible) adds 180° to the phase response, which 50% of the time will allow a greater gain to be achieved without oscillation, at least for some settings of the tone controls. Unfortunately, the much longer delay in digital hearing aids causes phase to change very rapidly with frequency, which makes manipulating the phase an ineffective method of feedback control.[e] Fortunately, digital technology has also opened up a new and better technique.

8.2.3 Feedback reduction by feedback path cancellation

The most effective and widely used technique in digital hearing aids, ***feedback path cancellation***, intentionally creates a second feedback path, completely internal to the hearing aid. This internal path has just the right gain and phase response to cancel the external leakage path, as shown in Figure 8.8. That is, if at any frequency, the two feedback paths leak back the same amount of signal, and if these two signals have the same phase, they will sum to zero, and there is no net feedback. Without any feedback, there can be no oscillation.

This seems like a perfect solution, but like the other solutions, it can increase the ***maximum stable gain*** (***MSG***) only to a certain extent. This increase in MSG enabled by the feedback cancellation algorithm is referred to as ***added stable gain*** (***ASG***). The more closely the internal path matches the external leakage path, the greater the ASG. To achieve a high ASG in

[e] A delay of 5 ms between the input and output of a hearing aid, for example, produces an additional 360° of phase shift every 200 Hz, which is a much faster rate of change of phase with frequency than is shown in Figure 8.7.

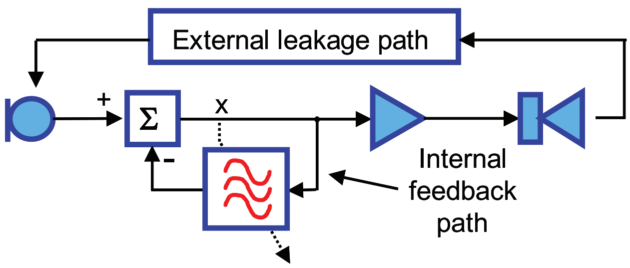

Figure 8.8 Internal feedback path added to cancel the effects of the external, unintentional leakage path. The filter adapts so that it minimizes the evidence of feedback at point x.

daily life, changes in the characteristics of the leakage path over time must be allowed for. There are two ways this can be achieved.

In the first method, no longer used, a test signal was injected, either with or without the amplifier chain being broken, to directly measure the characteristics of the external feedback path.[503] The scheme allowed the gain to be increased by approximately 10 dB before feedback commenced.[502] The scheme had the disadvantage that the test signal was audible to the wearer, except in the case of profound hearing loss.

In the method now used, the filter shown in Figure 8.8 automatically adapts in such a way as to minimize any signal that continues at a single frequency for more than a certain amount of time,[f] such as feedback oscillation, or low-level ringing caused by sub-oscillatory feedback.[518] A combination of methods is common. The hearing aid automatically measures the external leakage path during fitting, and hence initializes the internal cancelling path. Ongoing measurement of the output signal is then used to continuously fine-tune the internal path. Ongoing measurement is necessary if the algorithm is to cope with any of the events that normally cause a stable hearing aid to suddenly begin oscillating or to cause sub-oscillatory ringing. These include movement of the earmold in the ear canal, presence of a reflecting surface near the ear, or low levels of background sound that cause the hearing aid gain to increase.

A major advantage of feedback path cancellation over gain-frequency response control is that accurate feedback path cancellation does not cause *any* decrease in gain. Normally, if a hearing aid is close to oscillating, the positive feedback provided by the external leakage path increases gain (Section 4.7.2). The internal negative feedback path causes a corresponding decrease in gain and the combination of the two leaves the hearing aid with the same gain at all frequencies as when there is no feedback of any sort. A second advantage is that the cancellation can work even before the hearing aid continuously oscillates. Measurement of the output signal can detect and remove sub-oscillatory feedback or ringing (Section 4.7.2).

Feedback cancellation does have disadvantages. While the filter is adapting (which it should whenever the leakage path changes) the filter may distort speech quality during the second or so that it takes the filter to adapt. During this time, momentary audible oscillation may also occur. Another disadvantage is that the feedback canceller will also cancel other sustained periodic signals, such as somebody whistling or the sound of many musical instruments. Several enhancements have been added to some hearing aids to minimize the distortions that feedback cancellation can cause to musical sounds:

- Feedback cancelling algorithms can be disabled, or the rate of adaptation slowed down, when the music program for a hearing aid is selected.[g]
- The forward path of hearing aids can incorporate a small frequency shift, or a rapidly varying phase response, both of which help the hearing aid distinguish between internal feedback and external sounds.
- The hearing aid can compare the sound levels reaching each of the two microphone ports or, even more effectively, the two hearing aids on opposite sides of the head. External sounds will result in the level at the microphones being more similar than is the case for oscillation.

Feedback cancellation requires more calculations than gain reduction feedback control. More calculations

[f] Technically, the algorithm detects feedback oscillation when the output has a sustained prominent peak in its autocorrelation function, which occurs when there is a sustained tone included, even at a low level, in the output signal.

[g] The music will not be cancelled if the adaption time is considerably longer than the longest sustained note in the music.

require more battery current, although this limitation is becoming less important with each new generation of hearing aids.

Overall, the advantages of feedback path cancellation are substantial. Hearing aids are most likely to oscillate in quiet environments, where compression causes the highest gains. Unfortunately, quiet environments are precisely the environments where hearing aid wearers can least afford to reduce the hearing aid gain. Consequently, the additional gain enabled by feedback cancellation should translate into a real intelligibility advantage in quiet environments. Similarly, the close proximity of a telephone handset normally requires a significant gain reduction. Feedback cancellation reduces the amount by which gain has to be reduced, and hence improves speech intelligibility over the telephone.[1028]

Laboratory implementations of feedback path cancellation can achieve around 20 dB ASG,[518, 568] but commercially available hearing aids typically achieve less ASG (anywhere in the range 5 to 20 dB) and the amount varies widely across hearing aids and hearing aid wearers.[568, 657]

Some hearing aid designers prioritize achieving the greatest ASG possible, and risk the possibility of signal distortion, or a very slow response when the leakage path changes (during which few seconds the hearing aid will whistle). Others emphasize a fast response or minimal possibility of distortion, and hence do not achieve as much ASG. Room reverberation, with its very long delays, provides a limit on the ASG achievable.[855] Every time the patient moves within a room, even slightly, the feedback path changes, and so too must the internal filter if it is to accurately mimic, and cancel, the external leakage signal. Similarly, the ASG that is achievable without audible artifacts is affected by the presence of other adaptive signal processing in the hearing aid. The characteristics of the external feedback path change, for example, every time the characteristics of a directional microphone adapt. There are thus several criteria to consider other than ASG when evaluating the effectiveness of feedback cancellation systems.[993]

8.2.4 Feedback reduction by frequency shifting

Feedback oscillation occurs if a sound gets larger every time it goes around the feedback loop. What would happen if a sound came out of the amplifier at a different frequency to that which went in? Because the signal leaking back to the microphone would be at a different frequency from the original input, the two sounds could not remain continuously in phase with each other, and so could not build up in amplitude as effectively. Consequently, the likelihood of feedback would be considerably lessened.

As always, there are disadvantages. To achieve a large increase in gain without oscillation, a large frequency shift is needed.[h] This changes the quality of the output sound, and the pitch if applied to the low-frequency range. The method was developed for use in public address systems[1572] and has been briefly evaluated in hearing aids.[97] As the next section describes, there are now several frequency shifting algorithms available in hearing aids, so this method of feedback reduction is becoming more widely available.

8.2.5 Feedback reduction systems in combination

It is common for hearing aids to include more than one method of feedback control; indeed, it is desirable. A precise, but slow-acting feedback cancellation system can provide a large ASG, but may take 10 seconds to compute the new internal filter characteristics needed when the aid wearer puts on a hat. During this time, a fast-acting adaptive system can detect the oscillation, and reduce the gain at the appropriate frequency in less than a second.[1005] The aid wearer thus experiences little whistling, and the gain reduction is no longer needed once the slow-acting system completes its estimation of the altered external feedback path. Feedback cancellation algorithms can operate with more aggressive settings if the processing also incorporates frequency lowering.

One consequence of adaptive feedback reduction systems (whether by gain reduction, phase control, or feedback cancelling) is that when oscillation does

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h A frequency shift can be represented as a phase shift that changes with time. A phase-shifted signal can be represented as an in-phase sinusoid plus a second sinusoid phase-shifted by 90°. Thus, even a tone with altered frequency can be thought of as containing a component at the original frequency and phase, and it is this component that will lead to oscillation if the loop gain is high enough.

occur, the frequency of oscillation may rapidly change, a phenomenon referred to as ***frequency hopping***. This phenomenon occurs when the feedback reduction system lacks the flexibility to simultaneously deal with all the frequencies at which oscillation is possible. The system adapts to the frequencies actually oscillating, so the hearing aid oscillates at one of the other frequencies, which the system then adapts to, and so on. Experienced hearing aid wearers may need to be told what the warbling sound of this varying oscillation will be like if it occurs with their new hearing aid.

In conclusion, one of the major advantages of digital hearing aids has been the availability of effective methods of feedback reduction, especially feedback path cancellation. Feedback oscillation is consequently less of a problem now than it once was, although it is still commonly the case that feedback limits the ability to achieve the prescribed high-frequency gain, especially for open hearing aid fittings. Patients benefit from receiving more gain, having less occlusion, or being subject to the embarrassment of whistling hearing aids less often, or some combination of all three advantages. Evaluation of the effectiveness of frequency reduction schemes needs to consider the added stable gain that they enable, their effect on sound quality when they are operating, and the extent to which they are fooled by tonal sounds that the hearing aid is actually correctly amplifying. Excellent reviews of feedback reduction systems can be found in Agnew (1996) and Chung (2004).

8.3 Frequency Lowering

Most hearing-impaired people have a greater loss for high-frequency sounds than for low-frequency sounds. For some of these people, their high-frequency loss is so great that they cannot extract any useful information from the high-frequency parts of speech (Section 10.3.4). Because of the distortion associated with hearing loss (Section 1.1), this unfortunate situation can occur even if the speech is amplified sufficiently to be audible.[293, 747] Worse still, for some of these people excessive amplification of the high-frequency parts of speech can decrease their ability to recover useful information from the low and mid-frequency parts of the speech signal.[293, 747]

For such people to have any chance of accessing the information that exists only in the high-frequency parts of speech, the information must be moved down to some other frequency region where the person is more able to analyze sounds.[96, 829] This is the basis of ***frequency lowering*** hearing aids. An example of how frequency lowering might change the spectrum is shown in Figure 8.9. The term ***frequency shifting*** is also sometimes used, but as the shift is invariably downwards, *frequency lowering* is more specific.

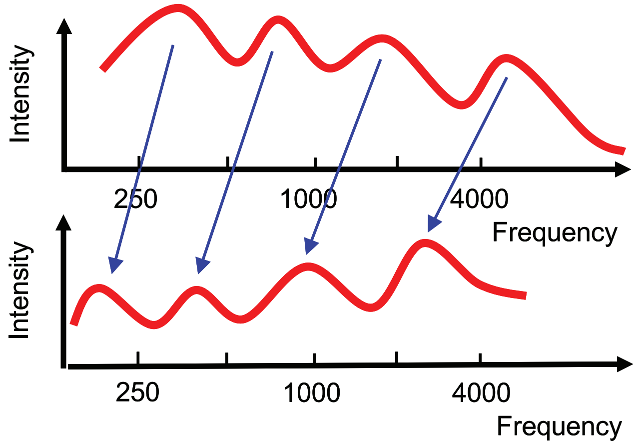

Figure 8.9 Input and output spectra for a frequency lowering scheme in which the output frequency equals half the input frequency. The amplifier also provides some high-frequency pre-emphasis. The arrows show the reduction in frequency of each formant.

The downward shifting can follow a number of mathematical rules and some of these rules can be achieved with a range of signal processing techniques.

8.3.1 Frequency lowering rules

A conceptually simple technique is to reduce all information in frequency by some constant number of Hz, which is referred to as ***frequency transposition***. For example, all information could be presented 2 kHz lower than it originally was. The problem with this approach is that input power from 0 to 2 kHz cannot be lowered in this way, and so remains in its original range. For sounds with significant energy below and above 2 kHz, such as voiced fricatives, the result may be confusing and ambiguous. For example, does an output component at 1 kHz originate from an input at 1 kHz or from an input at 3 kHz? For an input sound with energy over the whole range, the spectral shape of the output will be a complex mixture of the different input frequency ranges. Important features, like formants, originating in one frequency band may be obscured by speech components originating from the other frequency band. Nonetheless, many people with severe high-frequency hearing loss consider that transposition improves speech clarity for them.[1850] Perception of sounds with their dominant energy in the unshifted frequency range remains unaffected by the frequency lowering.[1517]

One way to minimize the problem of transposed sound sharing the same frequency range with unmodified sound is to apply transposition only when the input spectrum is dominated by high-frequency components, termed ***conditional frequency transposition***, or ***dynamic speech recoding***. Transposed energy will then be available for the sounds for which it is most needed, and will not produce adverse effects when low-frequency sounds are present.[1450] Conditional frequency transposition can increase the intelligibility of stops, fricatives and affricates without degrading the intelligibility of nasals and semi-vowels.[1450]

An alternative to transposition that has the advantage of no overlap in output spectra is ***frequency compression***. When the output frequency is a constant fraction of the input frequency, this is termed ***linear frequency compression***.[1804]

A variation on this is when the output frequency equals the input frequency raised to some power, which could be termed ***power frequency compression***. The term frequency compression is appropriate to both of these because any frequency range at the input is compressed into a smaller frequency range at the output. This term must not be confused with amplitude compression, as discussed throughout Chapter 6.

The frequency shifts shown in Figure 8.9 arise from linear frequency compression in which every output frequency equals half the corresponding input frequency. There is no ambiguity in frequency compression: every output frequency corresponds to only one input frequency.

Frequency compression produces undesirable effects if applied to the entire frequency range. Although linear frequency compression preserves the correct ratios between frequencies, which helps preserve the identity of vowel sounds and the voice-like quality of speech, fundamental frequency is decreased by the ***frequency compression ratio***. After transposition, young children may sound like female adults, and female adults may sound like males.[1804] Power frequency compression does not even preserve the frequency ratios, so all voices acquire an unnatural, inharmonic quality.

Because pitch cues reside in the low frequencies, below about 1.5 kHz, fundamental frequency will remain unchanged if frequency compression is applied only to frequencies above 1.5 kHz (or higher). Also, because harmonics above 1.5 kHz are not separately resolved by the auditory system, speech will retain its harmonic tonal quality even if these high-frequency “harmonics” are no longer actually harmonically related after frequency lowering. Lowering just the high frequencies is termed ***non-linear frequency compression***, as the amount of compression varies with frequency. The frequency above which input frequencies are lowered is variously called the ***transition frequency***, ***cut-off frequency***, ***frequency compression threshold***, or ***start frequency***. Non-linear frequency compression also has a one-to-one relationship between input and output frequency, and has been shown to improve intelligibility.[626, 1633] Evaluations have so far used speech in quiet or at a very high SNR.

Examples of each frequency mapping scheme so far discussed are shown in Figure 8.10. In each of the schemes shown, an input signal at 6 kHz is lowered to an output frequency of 4 kHz. The same mapping schemes are drawn with both linear and logarithmic frequency axes, as their appearance changes dramatically with the choice of axis. Be aware that there has not been uniform adoption of the terminology shown in Figure 8.10 and used in this section.

8.3.2 Frequency lowering techniques

Modulation

A simple and early technique for frequency lowering is distorting sound, such as with pronounced peak clipping. The resulting inter-modulation distortion products occur at frequencies far removed from the frequencies in the input signal, although the spread of frequencies occurs in an uncontrolled manner.[829]

A more sophisticated modulation method involves selecting a frequency range by filtering, multiplying (i.e. amplitude modulating, Section 3.6.1) a pure tone by the filtered signal, and selecting by filtering one of the sidebands created by the modulation. For example, multiplying the 4 to 8 kHz range by a 4 kHz pure tone moves the range down by 4 kHz (the frequency of the pure tone) to the range 0 to 4 kHz.[1850] Modulation is thus suitable for linear transposition, such as shown in Figure 8.10(a).

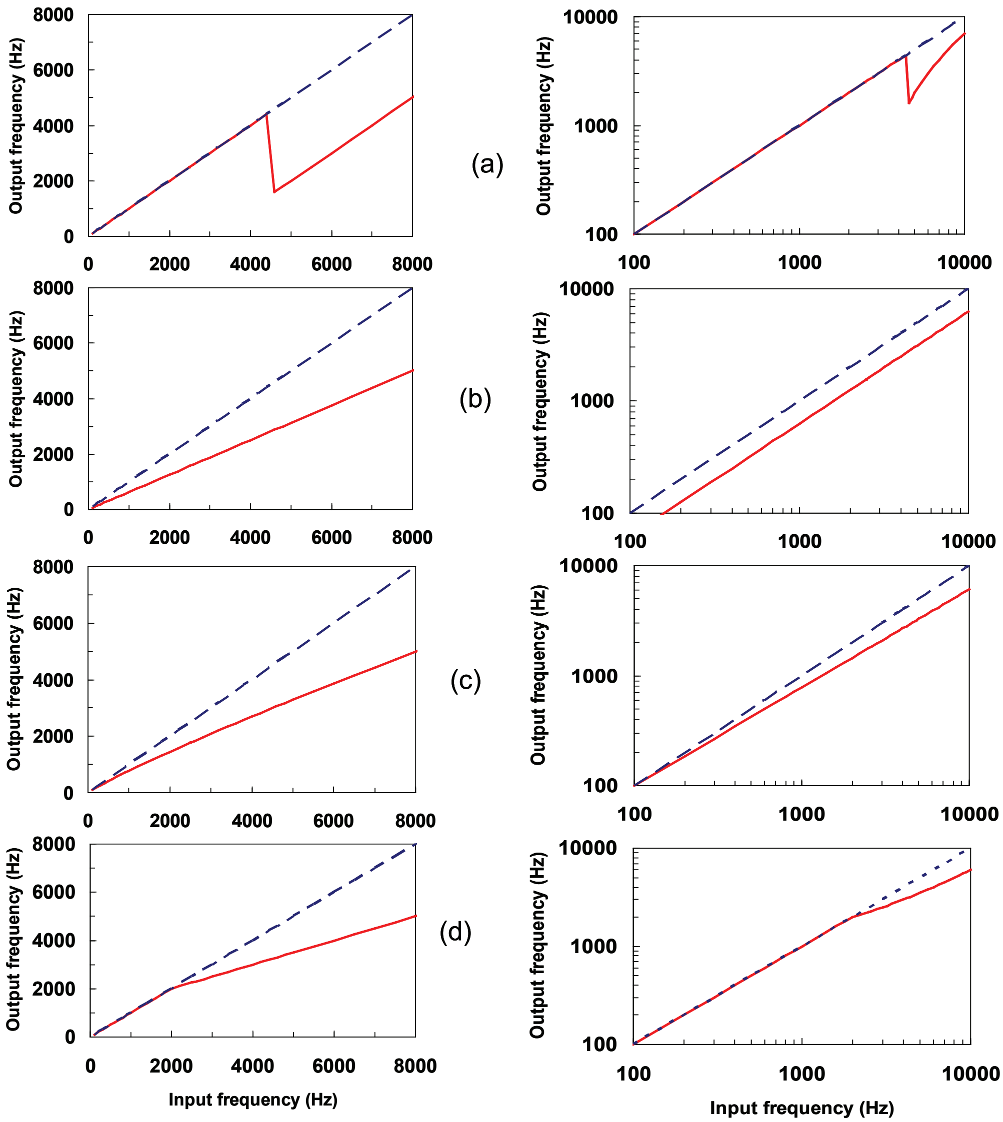

Figure 8.10 Relationship between input and output frequency in various frequency-lowering schemes. (a) transposition; (b) linear frequency compression; (c) power frequency compression; (d) non-linear frequency compression. Each scheme is plotted on linear frequency axes (left) and logarithmic frequency axes (right). In each case the diagonal dashed line shows no frequency lowering.

Slow playback

If a signal is played back at a slower rate than the one at which it was recorded, all frequencies are reduced by the same proportion (i.e. linear frequency compression). Slow playback is easily accomplished with digital processing. An obvious problem is that the processed signal, being slower, would get further and further behind reality unless something else is done. The solution is to delete sufficient sections of the signal such that the original and slowed-down signal have the same duration. Ideally, complete voice-pitch periods (segments ranging from around 2.5 ms for a child's voice to around 10 ms for a male voice) are deleted so that the waveform remains continuous and artifact free. Unfortunately, this is much easier to achieve accurately for speech in quiet than for speech under more typical conditions.

Speech vocoder

In a speech vocoder, speech is filtered into a bank of adjacent narrow bands, and the level within each band is detected. Speech can be re-synthesized by using

Fitting frequency lowering devices

- Commence by setting the hearing aid to lower every frequency for which hearing thresholds exceed about 80 dB HL.
- Some wearers may have an initial adverse reaction to sound quality, dependent on the degree of compression selected. If possible, facilitate the patient receiving an aural rehabilitation program during this time (Chapter 13) so that the patient can more systematically associate the new perceptions with the sounds of speech.[1003, 996]
- Fine tune the frequency lowering after the user has two weeks listening experience. Aim to use the highest start or cut-off frequency, and smallest transposition shift or frequency compression ratio, that enables /s/ and /ʃ/ to be detected and discriminated from each other. The adjustment can be guided by patient responses, live speech-mapping, or both.
- Always leave signal components below about 1.5 kHz at their original frequencies.

These guidelines, though based on practices used by researchers in this field, are not as yet based on extensive evidence.

these levels to modulate the level of narrow bands of noise, or pure tones, at the frequency of each original narrow-band filter. A ***frequency-lowering speech vocoder*** is constructed by making the frequency of the narrow bands of noise or pure tones lower than the centre of the bands from which the signals modulating them were derived.[1450] The frequency-lowering speech vocoder can thus achieve any mathematical relationship between output and input frequencies just by selecting the desired frequencies for the bands of noise and/or pure tones used for the re-synthesis. A frequency-lowering ***phase vocoder*** is a more sophisticated version that measures the phase as well as the amplitude of the signal falling within each channel. The rate of change of phase is then used to determine the precise frequency (not just which channel) with which the signal is re-synthesized at a lower frequency.[1633]

8.3.3 Commercially available frequency lowering schemes

The series of hearing aids produced by AVR Communications (TranSonic, ImpaCT) use slow playback to achieve linear frequency lowering. Frequency lowering is conditional in that only high-frequency dominated sounds have their frequency lowered.[406] Most commonly, the high-frequency dominated sounds will include all the unvoiced consonants and the low-frequency dominated sounds will include all the vowels.

The series of hearing aids produced by Widex (Inteo, Passion, Mind, Clear) lower frequencies in the first one or two octaves above the start frequency by transposition. The dominant spectral peak within this range is lowered by one octave, and the immediately surrounding spectral content is lowered by the same number of Hz as the dominant frequency.[1004]

The series of hearing aids produced by Phonak (Naida, Audeo, Nios, Exelia Art) achieve non-linear frequency compression by using the amplitude of each FFT analysis bin to re-synthesize the output at lower frequencies, in a manner similar to the processing described by Simpson et al. (2005).

8.3.4 Frequency lowering, speech intelligibility and candidacy

Numerous studies, summarized in Simpson (2009), have investigated the effect of different forms of frequency lowering on speech intelligibility. Simpson provides a comprehensive list of these studies. The overall picture is complex, with some studies showing group benefit, some showing no benefit, and many studies showing no group benefit, but significant benefit for just some individuals. Studies vary in the hearing loss configurations included, the type of processing, and the type of benefit measured.

Unequivocally, frequency lowering *can* improve speech intelligibility, at least in quiet, but there is as yet no method for predicting which individuals will benefit, nor which form or degree of lowering is most suited to them, nor how the lowering should be optimized to maximize benefit. Frequency lowering is much more likely to increase speech intelligibility than to decrease it, and those with the poorest speech intelligibility ability without frequency lowering probably have the most to gain.[66]

The difficulties in ensuring benefit are easy to appreciate. No matter how it is achieved, a spectral shape that normally occupies one bandwidth is squeezed into a smaller bandwidth. For the auditory system to extract information from this smaller bandwidth spectrum, the auditory system must have frequency analysis precision greater than is required for analyzing unprocessed speech. By contrast, the hearing aid wearer is very likely to have poorer than normal frequency resolution. Consequently, frequency lowering has the potential to decrease intelligibility as well as to increase it. Analyses of the consonant confusions made show that frequency lowering can simultaneously improve identification of some high-frequency phonemes while decreasing identification of other consonants,[1634] although the negative effects may reduce over time.[995]

For linear frequency compression, moderate lowering of frequency, where the output frequency is 20% less than the input frequency, seems more likely to produce benefit than greater amounts of lowering, where the disadvantages clearly outweigh the advantages.[1804]

Let us illustrate the considerations in matching frequency lowering to hearing loss characteristics with two examples. Hearing aid wearers with steeply sloping audiograms, like that shown with squares in Figure 8.11 were originally considered the ideal candidates for frequency lowering. To make an input bandwidth up to 6 kHz audible and useable, the band from 1 to 6 kHz has to be compressed into the band from 1 to 2 kHz. This compression (by a factor of 5) avoids compromising voice pitch information in the pitch range up to 1 kHz (ideally up to 1.5 kHz would be left unchanged), and avoids presenting lowered speech in the range above 2 kHz where the aid wearer has so much loss the ability to analyze audible information will be limited. The resulting concentration of spectral information may be so great that the lowered spectrum interferes with speech intelligibility as much as it helps it.[618, 1164, 1634] In addition, the steep variation of hearing threshold from 1 to 2 kHz may compromise the aid wearer's ability to analyze the compressed information within this range. Frequency

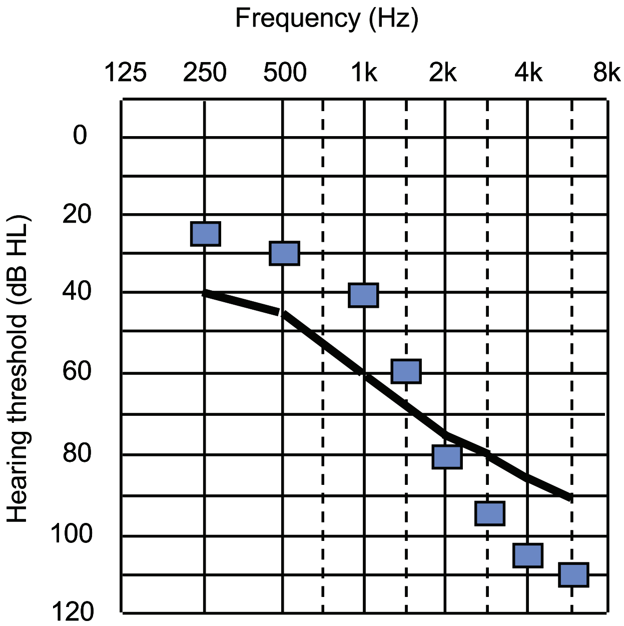

Figure 8.11 Two audiograms that might be considered for amplification with frequency lowering. Success is less likely for the steeper audiogram (rectangles) than for the shallower one (solid line).

transposition has, none-the-less, been shown to benefit children with audiograms not unlike those shown as the squares in Figure 8.11, especially after two months of experience.[66]

The second audiogram, shown as a solid line, slopes much more gradually, and likely provides useable hearing to a higher frequency. Lowering speech in those regions where loss is greater than 80 dB now requires the 1 to 6 kHz region to be compressed into the 1 to 3 kHz region, a compression factor of only 2.5. Results obtained with the same linear frequency compression method have been more positive for patients with gradually sloping audiograms than for patients sloping steeply to profound high-frequency loss.[1633, 1634]

Several issues so far preclude us from concluding the degree of benefit that different schemes offer, or the types of hearing loss characteristics to which they are best suited.

Adaptation

An obvious potential difficulty with frequency lowering schemes is that they make speech, and everything else, sound different. It thus takes people some time to become accustomed to, and maximally benefit from, this form of processing.[66, 996, 1450] It seems likely that children are more adaptable than adults at quickly making use of the altered cues that frequency lowering provides.[1632] It seems likely that greater listening experience will be needed for identification of sounds in continuous discourse than for simple differentiation of different consonants. It is possible that training will facilitate and accelerate adaptation to the new cues, but this is so far untested.

Frequency lowering parameters

One fitting strategy is to lower all frequencies for which hearing thresholds exceed some predetermined value.[1002, 1633] A similar strategy is to lower frequencies that occur deep within the dead region of the cochlea to a band around the lower edge of the dead region.[1517, 1518] It appears to be more beneficial to transpose down information much higher in frequency than the audible band than information that is in a band immediately higher in frequency than the audible band.[1857] That is, the lower the correlation between the transposed high-frequency signal and the unmodified low-frequency signal, the greater the additional information the transposed signal provides.[i] A third strategy is to lower the frequency range needed to ensure audibility of /s/ and /ʃ/ fricatives, but without so much compression that these sounds are confused with each other.[626, 1002]

Unfortunately, the range of possible transition frequencies, degree of lowering, mathematical functions relating input to output frequency, frequency-lowering methods, amplification provided to the frequency-lowered sound, and amplification provided to the unprocessed sound is so huge, it is virtually impossible to know that frequency lowering has ever been optimized for an individual.

Outcome measure

It is relatively easy to show that frequency lowering increases the ability to detect high-frequency consonants, especially /s/, and hence the differentiation (in English) of plural from singular nouns. This benefit does not imply that the high-frequency consonants detected will be discriminated from each other.[1487, 1517] Depending on the scheme and frequency lowering parameters, it may well be the case that frequency lowering interferes with the identification of some consonants and vowels. Different outcomes are therefore possible depending on the nature of the speech test given to evaluate benefit. The effects on intelligibility of different levels and types of background noise, including other talkers, for different frequency lowering schemes remain relatively uninvestigated. A comprehensive evaluation of frequency lowering should measure:[995]

- ability to *detect* high-frequency speech sounds and environmental sounds;[j]
- ability to *identify* a wide range of consonants, with a sufficiently high representation of fricatives spoken by a female talker (Section 16.4.1), but not restricted to high-frequency consonants;
- detection and identification in both quiet and in noise;
- impact on speech production;
- changes in these abilities over several months.

Confounding factors

Some devices simultaneously change another amplification characteristic, such as low-frequency gain or high-frequency gain, when frequency lowering occurs, and it may be difficult to know which change produced the positive or negative effects observed unless the effects of such changes are separately evaluated.[995, 1165, 1633]

One argument sometimes made against frequency lowering is that if the auditory cortex is deprived of the opportunity to process high-frequency sounds, its processing ability may degrade over time. This does not seem too likely because spread of excitation in the hearing-impaired cochlea will make it likely that all frequency regions still receive some stimulation. The problem would be further avoided if high-frequency regions of the cochlea were stimulated by even higher frequencies in the stimulus that are also lowered in frequency.

Frequency lowering is an extreme version of the frequency-shifting method of feedback reduction discussed in Section 8.2.4. A side-benefit, therefore, of frequency lowering is that feedback oscillation is much less likely, so greater stable gain can be achieved. Indeed, for wearers of high-gain hearing aids, it is difficult to assess how much of the benefit

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[i] This result is consistent with research showing moving narrow-band information upward or downward in frequency does not make speech any more intelligible than a reference condition where the information is not available at all.[1608]

[j] Kuk et al. (2010) contains a useful list of high-frequency environmental sounds.

provided by frequency lowering is due to this indirect benefit of enabling greater gain and hence audibility.

Hopefully, further research over the next decade will advance our knowledge concerning which patients benefit from frequency lowering, and how best to adjust it for each of these patients.

8.4 Speech Cue Enhancement

There are several experimental algorithms that, like frequency lowering, aim to modify speech itself to make it more intelligible for people with sensorineural hearing loss. Any acoustic feature of a speech sound that helps identify that sound, can, in principle, be detected and exaggerated to make recognition of that feature, and hence that sound, easier. Actually increasing intelligibility in a practical device, beyond that which frequency shaping and compression enables, has so far not been achieved.

Enhancement of spectral shape

There have been numerous attempts to detect the prominent spectral peaks of speech sounds (usually the formants), and provide them with greater amplification than is provided to the intervening spectral valleys.[27, 70, 71, 199, 457, 562, 1099, 1636, 1729, 1842] This is variously called ***spectral contrast enhancement*** or ***spectral sharpening***. The resulting formant structure is sharper, and the locations of formants on a spectrogram become better defined, as shown in Figure 8.12. Psychoacoustic experiments have shown that emphasizing small spectral peaks by surrounding them with deeper spectral valleys does help people with hearing impairment detect the original peaks.[426]

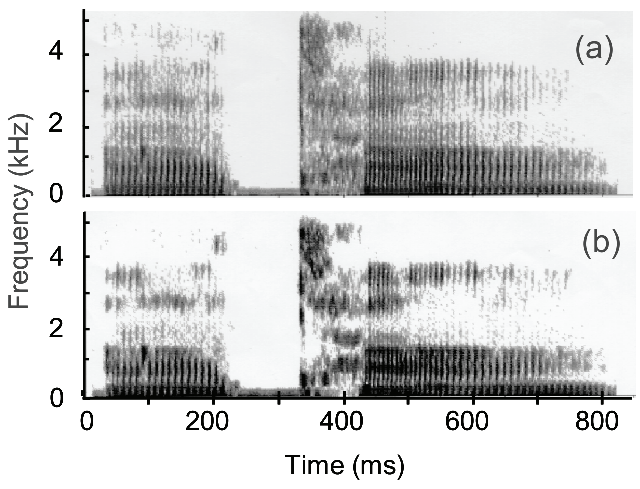

Figure 8.12 Spectrograms of the syllable /ata/ (a) unprocessed and (b) spectrally enhanced, showing more pronounced formants.[547]

Unfortunately, improvements in intelligibility have been small or non-existent. It is not as if the detection of spectral peaks is unimportant: when electronic processing is used to smear the shape of the spectrum across frequency, intelligibility in noise is adversely affected.[71, 1775, 1776] Presumably, the frequency resolution ability of the hearing-impaired test participants in some of the spectral enhancement experiments was so poor that the spectrum of enhanced speech passed on by the cochlea to the rest of the auditory system was just as smeared and indistinct as it was for unprocessed speech. Indeed, severely hearing-impaired subjects in one study could not detect *any* differences between processed and unprocessed stimuli, even in a paired-comparison listening task, and despite the differences being very evident to normal-hearing subjects.[547] Where differences are discernable, however, there is the potential for benefits to increase with listening experience.[70]

An extreme form of spectral enhancement is ***sinusoidal modeling***, in which the few most dominant spectral peaks are replaced by pure tones with the appropriate frequency, amplitude, and perhaps phase.[857, 1805] The method is very effective at reducing background noise, but has not yet been shown to increase intelligibility. Sinusoidal modeling can also be viewed as a form of speech simplification, as described below.

Enhancement of consonant-to-vowel ratio

The ratio of consonant level to vowel level is referred to as the ***consonant-to-vowel ratio***. This ratio is negative in unprocessed speech. The ratio can be increased (i.e. the consonant level made more similar to the vowel level) by increasing the amplification of consonants but not of vowels. Increasing the consonant-to-vowel ratio in this way has little or no effect on the loudness of speech,[1210] but increases speech intelligibility.[k, 575, 891, 1209, 1492] When the consonant-to-vowel ratio is increased by decreasing the vowel level, however, speech intelligibility does not generally improve,

[k] In the experiments with the most dramatic improvements in intelligibility, masking noise was added after the consonant was amplified.[640, 641, 661] Processing thus also improved the SNR for the consonant. This type of processing is not possible in a real hearing aid because the extra amplification added during a consonant would also amplify any noise that was present simultaneously.

suggesting that it is the absolute level of the consonants that is important, not their level relative to the vowels.[1543]

It is unquestionable that increasing the level of some consonants makes processed speech more intelligible than unprocessed speech. However, linear high-frequency emphasis and wide dynamic range compression with fast time constants each also increases the level of weak consonants relative to the level of vowels.[517, 729, 1462] It has not yet been established that particularly targeting consonants (which requires very complex processing) will give significantly better results than conventional hearing aid amplification comprising wide dynamic range compression and high-frequency emphasis. Another consideration is that all of these forms of processing should be advantageous only in those situations where the audibility of consonants is limited by the aid wearer's thresholds rather than by background noise.

Transient enhancement

Intensity enhancement has also been linked to the rate of change of intensity. Many consonants with a low level relative to adjoining vowels have rapid intensity changes (often with accompanying spectral changes) that must be perceived for the consonants to be correctly identified. There is the potential to increase intelligibility if the rapid variations in intensity can be made more prominent.

Increased prominence can be achieved by a circuit that automatically increases its gain whenever the intensity of the input signal is changing rapidly (such as during a plosive, or during the onset of an affricate), and decreases its gain whenever the intensity of the input signal is constant (such as during a vowel).[1185] The resulting speech is perceived as though all plosives have been articulated with great emphasis. Experiments have so far been unable to verify any improvement in intelligibility, relative to simple, linear high-frequency emphasis[428] or to more conventional compression[1755] (both of which also emphasize most of these same consonants). One potential problem is that consonants with a somewhat rapid onset followed by a steady level (e.g./ʃ/) can have their onset intensity enhanced to such an extent they sound like an affricate with a similar spectrum (e.g. /tʃ/).[428] The word *ship* hence becomes *chip*.

Transient enhancement may be particularly beneficial for hearing aid wearers with auditory neuropathy, as they have significantly reduced temporal resolution.[1960] This deficit presumably makes it difficult for people with this condition to identify the boundaries between phonemes, and hence the phonemes themselves, whereas transient enhancement highlights these boundaries. Experimental evaluations of transient enhancement using patients with auditory neuropathy, and normal hearing participants who receive sounds temporally smeared to simulate auditory neuropathy have been very positive.[1305, 1306] Unfortunately, the reference condition had neither high-frequency emphasis nor wide-dynamic range compression, so again it is unclear how much of the benefit would also have been obtained with these standard hearing aid characteristics.

Transient enhancement processing was implemented in one commercial hearing aid, which is no longer available, but doubtless the scheme will re-appear in another product. Transient enhancement processing has been beneficial in cochlear implants[614] although, here also, the benefit relative to simple compression is unknown.

Enhancement of duration

Another feature that has been modified is the length of vowels. Vowels preceding a voiced consonant tend to be longer than vowels that precede an unvoiced consonant. These differences in the preceding vowel length are one of the cues used by normal hearers to distinguish voiced from unvoiced consonants. Because duration perception is little affected by hearing loss, vowel length is a particularly important cue for people with hearing impairment, for whom spectral differences are less clear. [1493] Exaggeration of the natural differences in length enables hearing-impaired people to better perceive consonant voicing.[1493] Unfortunately, it is difficult to imagine how this processing could ever be done in real time (i.e. synchronously with the speech signal arriving). The decision to lengthen or shorten the vowel must be made before the vowel ends, and therefore before the final consonant has even reached the hearing aid. Montgomery & Edge (1988) increased consonant duration, although with little success in increasing speech intelligibility.

In an alternative approach to duration modification, an experimental hearing aid has been developed that lengthens vowels and transitions, so that hearing-impaired people have longer to recognize them.[1310] Such lengthening alone would make the output of the hearing aid progressively get further and further behind the input. Shortening some of the gaps between speech sounds solves this problem, and so the output "catches up" with the input. This approach increased intelligibility for a small proportion of hearing-impaired people,[1310] but decreased intelligibility for normal-hearing people with simulated losses.[1311] The negative impact of destroying synchronism between the visual and auditory signals that occurs with this system has not been evaluated to date.

Speech simplification

If profoundly hearing-impaired people are unable to perceive many of the complex cues in a speech signal, particularly when there is noise present, perhaps they will understand more if less information is presented. Simplification of the speech signal is the concept behind ***speech pattern processing***. At one extreme, the speech signal is replaced by a single pure tone pulsing on and off in time with the speech. The pure tone has a frequency equal to the fundamental frequency of the speech.[1526] Other features that have been extracted from the speech and presented in a simple manner include the amplitude of the speech envelope and the presence of voiceless excitation.[537] Speech identification is better with these additional features than with fundamental frequency alone.[538] Presentation of a simplified speech code also helps profoundly impaired people control the fundamental frequency of their own voice.[77]

The benefit obtained from speech simplification appears to be restricted to those profoundly impaired people with the least remaining frequency selectivity.[537] People who are likely to be candidates for speech simplification strategies are also likely to be candidates for cochlear implants, and can expect to perform better with the cochlear implant than with the speech simplification aid.

Enhancement by re-synthesis

An extreme example of using the special features of speech would be a hearing aid that recognized speech and then re-synthesized it in a clear, well-articulated, and noise-free way (and optionally in another language)! Of course, there are many problems with this. Just like hearing-impaired people, automatic speech recognizers do not perform well in noisy places, and have trouble with unusual accents. Also, the speech synthesizer would have to transfer many of the features of the real signal if the synthesized voice were to convey emotion, and were to sound like the person really talking. Because hearing-impaired people often use lip-reading cues, the automatic recognizer and synthesizer must output sounds within about 40 ms of the signal arriving.[1167, 1744] Given the current performance of speech recognizers, except under ideal conditions (a known talker in quiet), a hearing aid of this type still seems far away, even without universal language translation.

8.5 Other Signal Processing Schemes

Digital signal processing has made available a range of other schemes, some already available in commercial hearing aids, some still in research.

De-reverberation and echo reduction

Reverberation is highly detrimental to speech intelligibility, for normal-hearing and hearing-impaired people alike. The reverberation from each phoneme masks some of the power in the following phoneme(s), especially when the following phonemes are lower in level than the earlier phoneme. The reverberation following each sound makes it unclear where the offset of the sound is, making it harder for the listener to segment the speech.

Processing to markedly reduce reverberation, even when it is overlapped with other sounds, is technically possible. Unfortunately, it requires the processor to know the electroacoustic characteristics of all the signal paths from the talker to the listener, including the sum of all the reflected paths that cause the reverberation. This is unlikely to be feasible in real-world situations. Fortunately, reverberation that does not overlap a succeeding sound can be recognized by its characteristic gradual drop-off in intensity with time. Once recognized, its intensity can be reduced more rapidly than normal. This gives a perception of reduced reverberation and increased speech quality.

Environment classification

Most advanced hearing aids automatically classify the current listening environment into one of several pre-defined acoustic environments. These pre-defined environments usually include speech in quiet, speech in noise, noise alone, and music. Noise may be further classified into noise types, most notably wind noise versus others. Classification is based on numerous parameters, including overall level, spectral shape, depth and rate of modulation, and co-modulation across channels (Section 8.1.1).

The result of the classifier is used to automatically enable/disable other features in the hearing aid, including directional microphones, adaptive noise reduction, and wind noise reduction (low-frequency cut[1]). Enabling or disabling any feature usually changes sound quality (hopefully – that's why it is done!) and it is disconcerting to the hearing aid wearer for sound quality to change if the world around them has not changed substantially.

Environmental classification is far from an exact science, not least because the world is not neatly divided into mutually exclusive listening environments, but rather is a continuum in all respects. Consequently, the environmental classifiers take a cautious approach, analyzing the environment over many seconds or several tens of seconds, before pronouncing their conclusion, and even then, the resulting feature enabling/disabling is usually done gradually to avoid drawing unwanted attention to the change in hearing aid characteristics.

Alternatively, in some hearing aids if the classifier is uncertain as to which of several environments currently exists, amplification characteristics are set to values intermediate to the values that would be used for each of the potential environments. This approach really analyzes the world as a continuum, but achieves this indirectly by determining how well any real environment matches each of several idealized environments.

Automatic telephone detection

One listening situation of special importance is talking on the telephone. Hearing aids can easily detect if a strong external static magnetic field permeates the hearing aid. Ideally, this would result in the hearing aid automatically switching to telecoil every time the wearer holds a phone to the ear. Unfortunately, telephones do not always produce a sufficiently strong static magnetic field, and things other than telephones can produce strong fields.

Automatic detection can be made more reliable by a small circular magnet stuck onto the telephone.[1939] The ability of hearing aids to reliably recognize the close presence of a telephone is helped by a wireless link between the hearing aids (Section 3.2). In theory at least, the presence of a telephone could also be detected by a change in the feedback path.

Of course, room loops do not produce a static magnetic field and do not cause the acoustic feedback path to change, so to listen to a room loop through hearing aids in which the telecoil can only be selected automatically, a magnet must be suspended beside the hearing aid.[1939]

Data logging

Hearing aids have the capacity to store information about the listening situations they are used in (often categorized as described in the previous two paragraphs), how often the hearing aid was used in each situation, how the client adjusted the hearing aid in each situation, and how the hearing aid self-adjusted or activated signal processing features in each situation. Most critically, the total usage, and daily pattern of usage, of the hearing aid is revealed. The clinician can read all these data out at an appointment subsequent to hearing aid fitting, and use the data to help interpret the comments the client is making about the sound quality. The logged data may then assist the clinician in the sometimes difficult task of adjusting the hearing aid to increase the acceptability of the amplification to the client. Many clinicians show and explain relevant data-logging graphs to the client to elicit information from the client about particular experiences with the hearing aid.[1256] Some clinicians, however, do not like to divulge to their clients that their hearing aids are recording usage details.

Some of the ways that clinicians can use logged data include:

- If the data shows the client is consistently turning the volume control up or down in particular listening situations, the clinician can make this adjustment permanently.
- If the client has obvious difficulty manipulating the hearing aid, the logged data will indicate whether the client is actually using the hearing aid.
- If the client complains the batteries do not last long enough, the daily use patterns logged will show whether the client is turning the hearing aid off after removal, or possibly that the batteries really do have less capacity than their specifications indicate, perhaps due to being on the shelf for too long prior to use.

[1] Though not yet commercially available, future bilateral directional processing schemes will enable a greater degree of wind noise reduction than is possible with a low cut to the gain-frequency response. Such bilateral noise reduction systems may function even when the wind noise is sufficiently intense to cause one of the microphones to be saturated (i.e. overloaded) provided both are not overloaded at the same instant. Saturation can occur at wind speeds as low as 12 m/s.[1954]

- The appropriateness of the client's choice of listening programs can be checked by comparing the loged program usage against the situations in which the client reports wearing the hearing aids.
- Markedly different usage of the left and right hearing aids can provoke a discussion that may enable a problem with the adjustment or comfort of one of the hearing aids to be discerned.
- If some real-life situation is particularly troubling for a client, its acoustics can be discerned if the client selects that situation for recording in a short-term event log.

The software in some fitting systems automatically analyzes the downloaded data and then recommends specific changes to the fitting to address any problems identified.

Trainable hearing aids

Traditionally, hearing aids are fine-tuned by the clinician, in the clinic, in response to comments by the client about the sound quality. These comments may be based on limited experience listening to a restricted range of sounds (often just the clinician's voice). More commonly, they are based on listening experiences the client has had in the first few weeks of wearing the hearing aids.

There are two problems with this. First, if these are the client's first hearing aids, then the client's reactions to sounds may well change as he or she gains more experience with amplified sound. Second, if the clinician is to make appropriate adjustments to the hearing aid, the clinician has to infer the acoustic characteristics of the input sounds about which the client is complaining, interpret from the client's complaint what aspect of amplification has to change, know which electroacoustic parameters must be varied to solve the problem, and by how much they should be varied. Furthermore, if the client is happy with the sound quality, loudness, naturalness and clarity in other listening situations, the electroacoustic parameters affecting the hearing aid performance in these situations should be altered as little as possible. A paradox is that as hearing aid adjustment flexibility increases, so too does the number of ways to mis-adjust the hearing aid for any particular client. To further complicate the clinician's task, there may be no immediate feedback when a response is made: the client will likely not know if the listening problem has been solved until the client leaves the clinic and returns to the listening situation in which the problem occurred.

A solution to the problem is the ***trainable*** hearing aid, or ***self-learning*** hearing aid.[463, 1955] In the trainable hearing aid, the user adjusts a control or controls which not only cause the amplification to change, but which the hearing aid remembers for future use. In a simple trainable aid, the device just remembers and averages over time the position of the control that the user prefers. When the hearing aid is next turned on, the amplification characteristics are already those corresponding to the average of the amplification characteristics (e.g. gain) previously preferred by the client.

More sophisticated trainable aids, several of which are already available, remember not only the position of the control preferred by the client, but also some aspect of the acoustic environment in which the control was adjusted, as shown in Figure 8.13. After some training history has been built up, the hearing aid can then infer what position of the control(s) are preferred by the client in different situations. Effectively, the

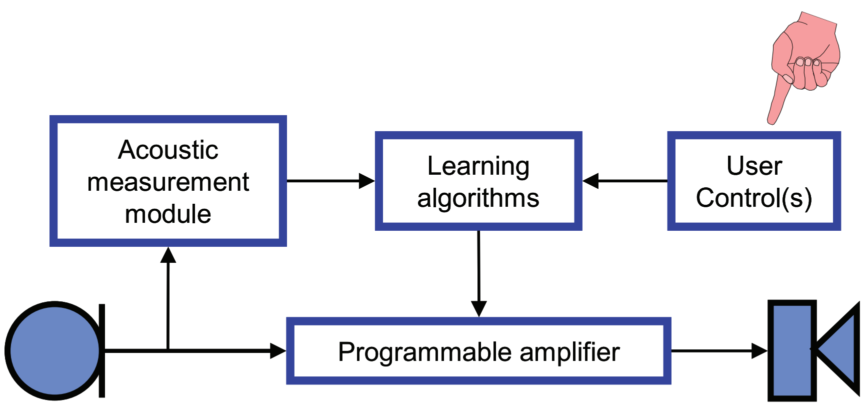

Figure 8.13 Block diagram of the trainable, or self-learning, hearing aid.

hearing aid automatically does what the clinician, and/or fitting software would otherwise do when inspecting the output of the data log referred to in the previous section.

Some hearing aids carry out the self-learning calculations and subsequent calculations separately in each of a small number of environmental categories (e.g. speech in quiet, speech in noise, noise only, music). The world of sound is not intrinsically categorized, however, and a better approach is for the hearing aid to use the training preferences to deduce the relationships between one or more amplification characteristics and one or more aspects of the environment.

As a simple example, suppose the control available to the client is a volume control, and the hearing aid measures just the SPL of the environment. After the client has adjusted the hearing aid on six occasions, the data available to the hearing aid might be those represented by the diamonds in Figure 8.14. On the basis of only these observations, the hearing aid can infer the relationship shown by the two-segment linear fit to the data. From this relationship, the hearing aid can deduce that the compression threshold is 55 dB SPL, the gain below compression threshold is 25 dB, and the compression ratio is 2:1. If the client also had access to a tone control that tilted the slope of the gain-frequency response, these same parameters could have been separately deduced for both the low-frequency channels and high-frequency channels of the hearing aid.

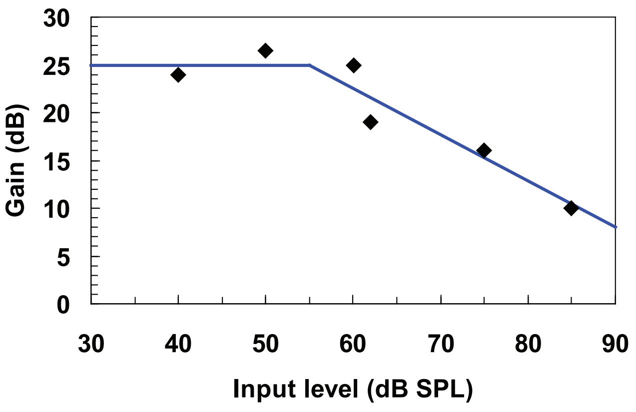

Figure 8.14 Data resulting from the client's adjustment of the hearing aid in different listening situations (diamonds), and the relationship the hearing aid has inferred between preferred gain and input level (solid line).

Is it feasible for the client to adjust more than one control? Experimental evidence is that clients can consistently adjust two or even three controls to achieve a preferred tonal quality and loudness,[474] although they prefer a maximum of two controls.[869] These can be provided on a remote control, or a single control on a hearing aid can alternate between the two functions on different occasions.[1955] Alternatively, the hearing aid can assign a different function to the control in different environments (e.g. degree of amplification in quiet environments, and control of adaptive noise reduction processing in a noisy environment).[1030]

The relationship between preferred amplification characteristics and the environment can likely be inferred more precisely if the hearing aid measures amplification characteristics more sophisticated than just the overall level. Examples include spectral shape, SNR, how SNR varies with frequency, and the direction from which the dominant speech signal arrives.

Active occlusion reduction

The aid-wearer's own voice creates excessive low-frequency sound in an occluded ear canal (Section 5.3.2). A signal processing solution to this problem is to sense the occlusion-induced sound in the ear canal, invert the sound pressure, and output this inverted sound wave through the receiver back into the ear canal.[1184] Because the original and re-introduced sound have the opposite polarity, they cancel each other (to the degree possible) resulting in a SPL much lower than either sound by itself. This is exactly the same principle used in the active noise reduction headsets used by many people on airplanes.

Figure 8.15 shows the essential ingredients: the additional microphone sensing the ear canal sound pressure, the internal negative feedback loop, and the various filters needed. Filters A and B are needed to ensure that the loop containing them and the two inward-looking transducers does indeed provide negative feedback throughout the frequency range for which there is appreciable gain around the loop.[m] The negative feedback loop reduces the level of *any* signal entering the loop. This is exactly what we need for occlusion-induced sounds entering the loop at the ear canal, but it is not what we want for sounds picked

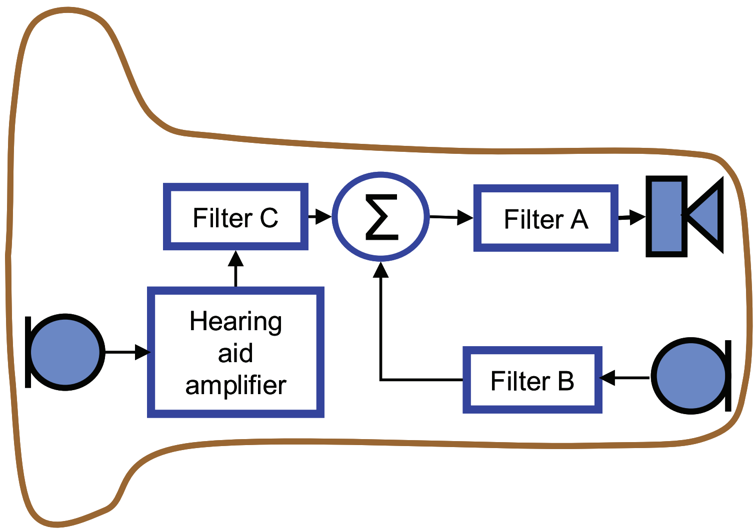

Figure 8.15 Active occlusion reduction system, shown inside an ITC hearing aid, but also able to be implemented as an RITC hearing aid.

up by the external microphone that are intended to be amplified. Filter C provides an amount of gain equal to the amount of attenuation that the loop causes, thus making the entire system transparent to the rest of the hearing aid. The system is able to attenuate occlusion sound within the frequency range of 80 Hz to 1 kHz, with a maximum attenuation of around 15 dB at the worst occlusion frequency of 300 Hz.[1184]

Active occlusion reduction has a number of advantages over passive occlusion reduction (i.e. a large vent or open fitting):

- The earmold can be closed or have a small vent, so leakage out is greatly reduced, which makes it possible to achieve greater high frequency gain without feedback oscillation occurring.
- The active system reduces any sound entering the ear canal, including sound that travels in through the vent. Consequently, amplified sound can become dominant over vent-transmitted sound down to a very low frequency. This means that directional microphones and adaptive noise reduction systems can work over the entire audio frequency range, rather than just the more restricted frequency range over which amplified sound usually dominates vent-transmitted sound. This is likely to be a considerable advantage because noise is often more dominant at low frequencies than at high frequencies.

Active occlusion reduction also has its disadvantages:

- The extra microphone requires additional space in the ear canal, making the hearing aid slightly bigger, and precluding its use in people with very small ear canals.
- The port for the extra microphone opens into the ear canal, creating a further place for wax and moisture to enter the hearing aid, and cause a fault. An effective wax barrier is thus essential.
- The additional signal processing, which must operate at a high rate to avoid delays around the feedback loop, uses additional battery current, thus shortening battery life. This issue becomes less important with each generation of integrated circuit design.

Own voice detection

We have seen that several signal processing schemes rely on estimation of the SNR to appropriately adjust the amplification characteristics. The close proximity of the hearing aid to the mouth and its location somewhat above and behind the mouth causes the input to have a particularly high level, and a pronounced

[m] Phase shifts in the receiver have the potential to cause the feedback to become positive feedback, which would make occlusion worse, or even cause the hearing aid to oscillate. Filters A and B prevent this from happening.

low-frequency dominance whenever the aid wearer talks. This can provide the signal processing algorithms with misleading information if their job is to optimally adjust the amplification to allow the wearer to hear and understand *other* people. Hearing aids can use the high level, low-frequency dominance, and equal input signals at both hearing aids (the mouth *is* located mid-way between the two ears!) to infer (but not precisely) whether the dominant speech signal is originating from the aid wearer or from another person. If the hearing aid has active occlusion reduction, the additional internal microphone provides a further information source that makes the task of own-voice detection easier and hence more accurate.

Self-checking hearing aids

An additional microphone sensing sound within the ear canal can also be used to check for proper performance of the hearing aid. A fault in the electronics, a receiver blocked with wax, or a very poorly fitting earmold will all cause the sound in the ear canal to differ from what should be there given the input sound and the amplification characteristics programmed into the hearing aid. An automatic comparison of the expected and actual sound characteristics can be used to trigger an audible warning to the client that the hearing aid is faulty and should be serviced.[n]

Predictive bandwidth extension

Speech sounds with significant energy present in the high frequencies (e.g. from 4 to 8 kHz) also tend to have significant energy in very high-frequency sounds, above 8 kHz. Hearing aids could use this correlation to artificially extend the bandwidth to 12 kHz, provided the receiver has the capability to output audible sound over this extended range. Improved speech intelligibility, naturalness, or localization would also depend on the aid wearer having sufficient remaining hearing ability to detect and analyze sounds in the extended frequency range. Such an algorithm would appear to be suitable to those with mild loss, at best.

Self-fitting hearing aids

If a completely automatic audiometer is built into a hearing aid, the hearing aid itself can measure the hearing thresholds of a patient, apply a prescription formula to calculate the appropriate real-ear response, and adjust the hearing aid to approximately match this response. This process, illustrated in Figure 8.16, mimics the process that a skilled clinician uses. Preliminary research indicates that both the automated threshold measurement and the automated hearing aid adjustment can potentially be as accurate as the same process carried out by a clinician.[879, 1355]

An obvious concern is that patients may do the threshold testing in a place sufficiently noisy to invalidate the thresholds. The hearing aid can use the normal hearing aid microphone to monitor noise levels, and even the noise spectrum, during the testing, and can advise the patient to move to a quieter place if the noise levels measured are too close to the thresholds obtained. Such a device, which would not require connection to a computer of other hardware to be fitted, would have major application in developing countries where the number of clinicians is grossly insufficient to meet demand. The patient can carry out fine-tuning if the hearing aid also incorporates a trainable algorithm.

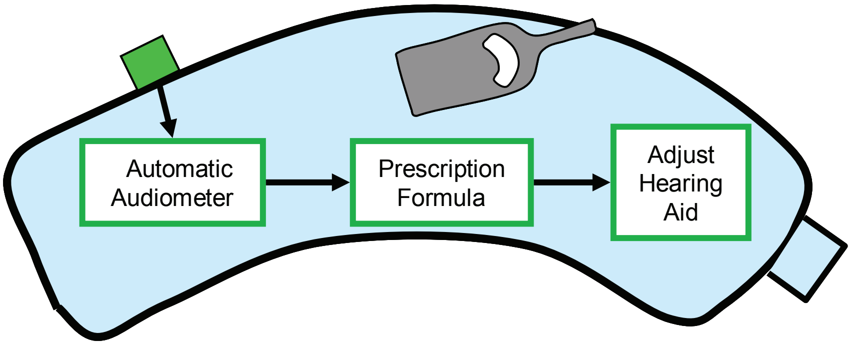

Fig 8.16 Key components of a self-fitting hearing aid, including a sound level meter to monitor the noise level in the environment during the automatic audiometry.

[n] Some hearing aids already include a self-check facility, and while this is useful, without a microphone sensing the sound level in the ear canal, the self-check cannot detect some problems occurring in the receiver, or any problems associated with the tubing or fit of the earmold or other canal fitting.

8.6 Concluding Comments

One of the major difficulties in evaluating novel processing schemes is allowing for the effects of familiarity and practice. If sound is markedly altered by the processing, it is likely that experimental subjects will need considerable listening experience, and perhaps even systematic training, before they are able to use the altered or new cues to identify speech sounds. Extensive listening experience is difficult to provide in the laboratory. Increasingly, it is possible to make wearable devices that process sounds in complex ways. Is it reasonable, however, to ask subjects to wear, every day, a hearing aid that produces strange sounds, before there is any evidence that the processing is beneficial?

One way to minimize this dilemma is to first test discrimination ability (the ability to differentiate contrasting sounds) after minimal practice. If the processing increases discrimination ability for at least some otherwise confusable sounds, then it may be more reasonable to proceed with extensive familiarization (perhaps just by using it during everyday life), training, and/or speech intelligibility (i.e. identification) testing.

Some of the processing algorithms reviewed in this chapter, such as frequency lowering, almost certainly have to be adjusted to best suit each patient. Further research investigating the effects of varying the frequency lowering parameters with a range of hearing loss characteristics would be particularly beneficial so that clinicians can optimize use of this technology, which increasingly is available in commercial devices. The optimal sensation levels of the frequency-lowered information have yet to be investigated. Such research would ideally measure confusion matrices so that the positive and negative effects of the frequency lowering on different speech cues, in quiet and in noise, could be discerned and the net benefit maximized.

Adaptive noise reduction should be useful for all patients in some listening situations, although the degree of attenuation provided in different situations should almost certainly vary with the degree of loss of the hearing aid wearer and level of the offending noise. Further research into how best to apply adaptive noise reduction, including the option of increasing gain at frequencies where the audibility of speech is limited by hearing thresholds rather than by noise,[1007] would therefore also be beneficial, despite long-standing adoption of adaptive noise reduction into virtually all advanced hearing aids.

How does the clinician evaluate the worth of new processing schemes as they become commercially available? Ideally, evidence-based decisions will be made.[340] These decisions will be based on research that show that with the new processing scheme, experimental participants obtained higher speech identification scores, and/or preferred the sound quality in their usual listening environments, compared to a reference amplification condition. The reference condition must, as a minimum, have a gain-frequency response appropriate to each subject, and some form of compression. It is essential that the experiment be blinded (so that the subjects do not know which is the new form of processing), and it is desirable that it be double-blinded (so that the experimenter cannot unconsciously influence the subjects). Unfortunately, all too often the devices are ready for sale long before any evidence evaluating their effectiveness is available.

Before leaving this discussion of signal processing algorithms, it is worth reviewing the effectiveness of signal processing algorithms (i.e. the different forms of compression discussed in Chapter 6, directional microphones discussed in Chapter 7, and the noise reduction, feedback reduction, transposition, and speech feature enhancement schemes discussed in this chapter) relative to other means for improving intelligibility. By far the best way to improve intelligibility is to remove all noise and reverberation from the signal before presenting it to the hearing-impaired person. The best way to do this is to put the microphone right next to the lips of the person talking and then use enough amplification, frequency shaping and compression to make the speech audible and comfortable at all frequencies. Consequently, FM or other wireless transmission systems, which position the microphone at the talker, still provide the greatest intelligibility improvement.

Another solution, far less effective but still worth having, is the use of directional microphones to decrease (but by no means remove) noise and reverberation. Intermediate to these will be the bilateral super-directional arrays, when commercially available. Any of these solutions can be combined with the more complex forms of signal processing (speech cue enhancement, frequency lowering, adaptive noise reduction), to potentially obtain greater benefit than any one processing scheme alone can provide.

Trainable hearing aids will likely have a big impact on clinical practice. First, the clinician may be freed of the task of fine-tuning the hearing aid in the weeks after fitting, as the hearing aid learns directly from the client what adjustments are needed. Second, if the clinician knows that the client will lead the hearing aid to the set of amplification characteristics that the client prefers, it does not seem sensible to spend expensive clinical time on achieving and verifying a prescription target that at best will be correct for the average person with hearing loss and lifestyle characteristics like those of the client. Rather, an approximation to the prescription target, such as that provided automatically by the fitting software, may be a perfectly acceptable starting point from which the client can fine-tune the hearing aid. The starting point must still be a reasonable fitting for the patient.[o] First, we want the patient's initial experiences to be sufficiently positive for the patient to continue use and training. Second, the time taken to reach the patient's preferred settings, and probably even the final settings, depend on the initial settings of the hearing aid.[474, 876, 1261] Of course, there will be many clients without the cognitive or physical ability, or motivation, to use a trainable hearing aid, for whom the fitting process will remain unchanged. However, approximately half of current hearing aid wearers are likely to be able to train their own hearing aids.[869]

The act of training a hearing aid to one's personal preferences may also produce greater emotional "ownership" of hearing aids among first-time aid wearers,. This possibility, and investigation of how types of controls, amplification characteristics, and acoustic environment measures should best relate all require research more intensive than has so far occurred.

[o] Unfortunately, not all automated "first-fit" hearing aid adjustments do produce a reasonably close match to prescription targets.[698]

CHAPTER 9

ASSESSING CANDIDACY FOR HEARING AIDS

Synopsis

Although the decision to try hearing aids is ultimately made by the patient, many patients will be in doubt as to whether they should acquire hearing aids and so will look to the clinician for a recommendation. This recommendation must take into account many factors other than pure tone thresholds.

Initial motivation to obtain hearing aids has been shown to be a key determinant of whether patients continue to use them. This motivation reflects the balance of all the advantages a patient expects hearing aids will provide offset by all the expected disadvantages, irrespective of whether all these positive and negative expectations are realistic. The advantages expected by the patient are affected by the degree of disability they feel they have. Disability includes how much difficulty the person has hearing in various situations, referred to as activity limitation, and the extent to which a person is unable to participate in activities because of the hearing loss, referred to as participation restriction. The advantages and disadvantages expected by the patient are affected by what the patient has been told about hearing aids by others. Disadvantages potentially include the impact on a patient's self-image of wearing hearing aids. The clinician must attempt to discover a patient's expectations and modify those that are unrealistically low or unrealistically high. Although hearing aids help in quiet and in noise, they help much more in quiet, so hearing aids are more likely to be valued and used if the patient needs help hearing in quiet places.

When a clinician encounters a hearing-impaired patient who does not want hearing aids, the clinician should find out whether this is because the patient is not aware of the loss and/or the difficulty that he has compared to others, or because the patient, although acknowledging the loss, does not wish to wear hearing aids. If the latter is true, the patient's reasons should also be discovered.

Difficulty managing a hearing aid can greatly affect use, so the clinician must consider likely manipulation difficulties when determining candidacy and aid type. People with tinnitus often find that hearing aid use diminishes their problems, so tinnitus positively affects candidacy. The presence of central processing disorders and extreme old age can both affect candidacy, but not in a manner sufficiently predictable to affect the clinician's recommendation. People who are not worried that hearing aids will stigmatize them are more likely to acquire them, and people who more readily accept the presence of noise while listening to speech are more likely to use them. Several personality characteristics also make hearing aid acquisition, use, and/or benefit more likely.

People with a severe to profound hearing loss are likely to receive more benefit from cochlear implants than from hearing aids. The most useful indicator of which device will be better for them is the speech score they receive for well fitted hearing aids after some years of becoming accustomed to them. For infants, this is not possible so the decision to implant has to be based primarily on aided or unaided hearing thresholds (as well as requiring no medical or psychological contra-indications). Cochlear implants and hearing aids generally provide complementary cues, whether they are worn in the same ear, or in opposite ears.

Vibrotactile or electrotactile aids are a worthwhile alternative for those with too much hearing loss to receive useful auditory stimulation from hearing aids, but who do not wish to receive a cochlear implant, or for whom a cochlear implant is not suitable on medical or psychological grounds. Training in integrating the tactile information with visual information is essential.

Hearing aids should not be withheld just because speech scores obtained under headphones fall below some arbitrarily determined criterion. There are, however, several audiological/medical indications that should cause hearing aid fitting to be delayed until the cause of the problems has been resolved.

A clinician therefore has to consider a large number of factors that may affect candidacy for hearing aids, none of which has such a strong effect that the remaining factors can be ignored.

To fit, or not to fit: that is the question. More precisely, when should the clinician encourage an uncertain patient to try hearing aids,[a] and when should the clinician advise the patient that hearing aids will probably not be beneficial, in the patient's current circumstances? The final decision about whether to try hearing aids will be the patient's (or that of a bossy close relative), but the clinician can greatly influence this decision if the patient is at all uncertain. The hard part for the clinician is knowing which way, and how strongly, to encourage.

Hearing loss is very common. Approximately 10% to 16% of an adult population will report that they have trouble hearing.[947, 1923] If we take an audiometric approach to defining hearing problems,[b] 16% of an adult population will have a four-frequency average hearing loss in the better ear of greater than 25 dB HL.[402, 1923] Of course, 25 dB HL in the better ear is an arbitrary criterion. If we increase the criterion by 10 dB to 35 dB HL, the proportion with a hearing loss of this degree or greater halves to 8%, and each further increase of 10 dB in the criterion up to 55 dB HL produces a further approximate halving of the population proportion with this degree of loss or greater.[402, 1923] Based on the audiograms of 30,000 people who obtained hearing aids, this pattern of the proportion halving for every 10 dB increase in the criterion continues up to 110 dB HL.[440]

Most people with hearing loss do not, however, acquire hearing aids. Various studies indicate that of those who consider they have a hearing loss, or who objectively have a hearing loss, only 14 to 31% own a hearing aid.[68, 269, 398, 724, 947, 1448, 1711, 1716] That is, approximately four out of five people with a hearing loss have not tried (at least recently tried) hearing aids. The ownership rate as a proportion of those with hearing loss (defined in different ways by different people) is referred to as the ***penetration rate***.

Not surprisingly, penetration rate increases with degree of hearing loss.[269, 398, 955] The penetration rate reaches 50% once the four-frequency average hearing loss in the better ear exceeds about 40 dB HL.[269, 398] There is conflicting information about whether penetration rate varies with age. When hearing loss is based on self-reported difficulty, older hearing-impaired people are much more likely than younger hearing-impaired people (with the same degree of self-rated hearing disability) to obtain hearing aids.[955] When hearing loss is based on audiometric thresholds, however, younger people and older people (with the same degree of pure-tone loss) are equally likely to obtain hearing aids.[269, 398] These apparently opposing conclusions are resolved if older people report less disability then younger people for the same audiometric thresholds, which does appear to be the case.[643, 767, 1093, 1658]

Hearing aid ownership also varies across countries; in developed countries typically ranging from a little over 2% of the population where the expense is mostly born by individuals, to a little under 4% where the expense is mostly borne by the government.[61, 84, 953, 1448, 1711] Ownership in developing countries is, sadly, many times smaller. Within a population, the proportion of those receiving hearing aids who actually use them does not depend on whether the hearing aids were free or paid for.[1711, 1819] While cost is undoubtedly a major reason why many people do not acquire hearing aids,[955] the low penetration rate in countries where hearing aids are free to the patient indicates that it is by no means the major reason.

Are people with hearing loss, but no hearing aids, all good candidates for hearing aids, if only they could be found and convinced to try them? Research suggests that when they are sought out and given the opportunity to try hearing aids, some of them certainly find hearing aids to be beneficial and continue to use them.[401, 1716] Some, however, decide that for them, the practical and psychological disadvantages outweigh the benefits. Many hearing-impaired people who have never worn hearing aids will understand speech (in certain listening situations) better if they do wear them,[399] but this does not mean they are candidates for hearing aids, if they consider they do not need them. Kochkin (1997) used self-report measures to estimate

[a] From here on, this book will refer to hearing aids in the plural, because more often than not, people will benefit more from two hearing aids than from one. A substantial minority of the hearing-impaired population will, however, prefer and/or benefit from only one aid. Chapter 15 will cover this issue in more detail.

[b] Although these self-report and audiologically defined prevalence rates are very similar, the two methods produce considerable differences in *which* individuals are considered to have a hearing loss.[1923] There is a variety of equally valid definitions of "hearing loss" which of course lead to different estimates of prevalence.[485]

that the number of people who could benefit from hearing aids is at least double the number who currently own them.

We should not be too optimistic about most of the four out of five hearing-impaired people who don't own hearing aids becoming candidates, at least with the current level of hearing aid performance. It is not difficult to find people with hearing loss who have not taken any steps towards obtaining hearing aids. Technology has increased the options for efficiently screening large numbers of people. Suitable methods include questionnaires, automated tests over the telephone, internet-based tests, and extremely inexpensive hand-held devices that emit preset sound levels.[324, 1162, 1656, 1933]

Even when found, most hearing-impaired people decline an invitation to try hearing aids.[663, 1528, 1835, 1933] Furthermore, even amongst those who have obtained hearing aids, a significant proportion don't use them at all.

Surveys of hearing aid owners have produced highly varying estimates of the proportion of owners who *never* use their hearing aids: 1%,[441, 1820] 3%,[122] 4%,[1693] 5%,[1869] 6%,[173] 8%,[447] 11%,[953] 12%,[1390] 20%,[171] 21%,[438] 24%,[269] 25%,[1089] and 29%.[1448] A further proportion of owners seldom use their hearing aids. There appears to be greater non-use found when the survey response rate is high (Section 14.6), when only new users are surveyed, when the survey is performed by someone other than the clinician who did the fitting, when entire populations are surveyed (rather than just those who received hearing aids in the last year or two), or when the hearing aids are free to the user *and* the fitting of hearing aids provides financial rewards to those involved in supplying and fitting them.

Until there is some large scale prospective experiment, we will not know what proportion of the population will use and benefit from hearing aids. Even then, the proportion will be true only for the technology level that exists at that time. It seems likely that a large proportion of those who have never sought hearing aids would not accept them if offered. Epidemiologists are not the only people uncertain about how many people need hearing aids. Every clinician can expect to see many patients who will be uncertain about whether hearing aids will help them.

There are two broad aspects to the question of whether an individual will benefit from hearing aids: is the person deaf enough, and is the person too deaf? As we will see, the first of these questions *cannot* be answered from the audiogram. The second question can be rephrased as *will the person benefit more from hearing aids, or from some other device such as a cochlear implant, a hybrid cochlear implant/hearing aid, a middle-ear implant, a bone-anchored implant, or even a tactile aid*? This chapter will identify some factors that should be considered when deciding whether to recommend hearing aids to a patient.

A purely quantitative approach to this decision is not possible at this stage because of the huge number of factors affecting the decision, and because we do not know how best to measure them nor weight them appropriately for each individual. There have been several attempts to predict benefit from hearing aids by statistically combining a wide range of factors easily measureable prior to fitting. These factors include hearing thresholds, age, self-reported hearing difficulty, education level, cognitive ability, visual ability, general health, hearing aid size, reaction to background noise, and expectations about hearing aids. Unfortunately, although various measures of success with hearing aids correlate with various potential predictors, individually and in combination, no combination predicts candidacy with enough accuracy to be reliable for an individual client.[272, 438, 767, 938, 1274, 1539]

A purely quantitative approach may never be possible and, given the importance of the interaction between patient and clinician, even this quantitatively inclined writer thinks that this may actually be a good thing. Every patient is unique. If you feel uncomfortable with uncertainty and fuzzy human-oriented decisions, you had better find new ways to cope! Other than for patients with moderate or moderately severe losses, there will often be some uncertainty about whether hearing aids will be the best option until they have been tried.

The first section of this chapter is particularly oriented towards adult patients. Although most of the factors are also relevant to children, the arguments and evidence are different. The chapter is, of course, directed towards patients who have never previously worn hearing aids, or who have not tried them for several years. For people who have tried and rejected well fitted, modern hearing aids, the candidacy question has probably been answered unless the patient's hearing loss, needs, or attitude changes, or until technology improves. It is, nevertheless, worth determining why the person was disappointed, and what it would take for the patient to consider that hearing aids are worthwhile. Options for amplification continue to expand rapidly.

While reading this chapter, you should keep in mind that when a hearing-impaired person walks into the clinic, the person has not been pre-ordained as a hearing aid candidate or non-candidate, and all the clinician has to do is determine which category applies. Rather, the patient will be best served if the clinician investigates if there are any factors preventing a person with a hearing loss from receiving benefit from hearing aids and, if so, what can be done to alter these factors.

9.1 Factors Affecting the Lower Limit of Aidable Hearing Loss

There have been many attempts to find the degree of pure tone hearing loss that would distinguish those who will benefit from hearing aids from those who will not. All such attempts have been spectacularly unsuccessful. At first sight this seems surprising. At the extremes, someone with a severe hearing loss will derive enormous benefit from hearing aids, and someone with normal hearing will derive no benefit.[c] Why then is it not possible to find a degree of hearing loss in between that differentiates those who will benefit from those who will not?

9.1.1 Attitude and motivation

The answer is that for these intermediate degrees of loss, other factors influence benefit more than does the audiogram. Arguably, the strongest of these other factors is the attitude of the person towards obtaining hearing aids and hence their motivation to do so. Of course, attitude and motivation are really the accumulated result of a variety of other positive and negative factors. These factors include:

- ***Acknowledgment of loss***: Does the patient realize (intellectually) and accept (emotionally) that his or her hearing mechanism is not normal? That is, does the patient fully acknowledge the presence of a ***hearing impairment***? When patients say that other people mumble, for example, this "cause" of their problem may reflect either a lack of *awareness* of the loss, or a lack of *willingness* to accept that there is a loss.
- ***Communication needs***: How often is the patient in a situation where he or she hears less clearly than is necessary to function effectively? Alternatively, how often is the patient in a situation where so much concentration is needed that fatigue quickly follows? That is, how much ***activity limitation***, previously known as ***hearing disability***, does the patient have? More importantly, how much hearing disability does the patient acknowledge?
- ***Consequences***: Does the hearing disability cause the patient to refrain from activities that he or she would otherwise like to do, or does it cause the patient to have negative feelings about life? That is, how much ***participation restriction***, previously known as ***hearing handicap***, does the patient have? More importantly, how restricted does the patient feel? Further discussion on the distinctions between impairment, disability, and handicap can be found in Hickson & Scarinci (2007).
- ***Self-image***: Does the patient consider that wearing hearing aids would make other people view him or her negatively in some way? Alternatively, does the thought of wearing hearing aids make the patient think about him or her self in a negative way? Some patients who accept they are having communication difficulties may prefer a self-image of social incompetence to one of defective hearing.[792] Their self-image may thus be better preserved by *not* acknowledging their loss.
- ***Expected benefit***: How beneficial does the patient believe hearing aids will be, on the basis of what the patient has been told by other hearing aid wearers, ex-hearing aid wearers, medical practitioners, or by observing other people who wear hearing aids?
- ***Fear or uncertainty:*** Does the patient anticipate having difficulty understanding how to operate hearing aids, or anticipate not having the dexterity to operate them? Does the patient equate a hearing loss with aging, reduced social competence, or even senility, and reject any tangible representation of deterioration, such as hearing aids?
- ***Costs***: How does the total of all the perceived costs (financial cost, inconvenience, effect on self image) compare to the perceived benefit

[c] Because a hearing aid microphone generates internal noise, a hearing aid can only make it *harder* for someone with normal hearing to detect soft sounds in a quiet place, no matter how much gain the hearing aid has.[908]

of wearing hearing aids?[565, 619] The benefit that counts is not some easily-measured increase in speech intelligibility in a specific situation, but rather what this increase leads to in terms of easier relationships with others, increased prospect of future employment, participation in activities, or decreased isolation, confusion and fatigue.

- ***Influence of others:*** Has the patient been encouraged or even coerced into seeking rehabilitation? Nearly half of all hearing aid candidates are positively influenced by their families to obtain hearing aids.[173, 486, 946, 1357] On the other hand, many people who have refrained from seeking help have done so on the advice of a

Determining motivation to obtain help

This panel outlines six tools, of varying length, that will provide insights into a patient's attitude towards wearing hearing aids. Many clinicians will prefer just to talk to the patient but, unless the conversation covers the main topics addressed in the following tools, insight into the patient's attitude and reasons for that attitude may not be obtained.

In an excellent article on evaluating a person for hearing aid candidacy, Saunders (1997) recommends asking two key questions to quickly determine motivation:

- *What prompted you to come for a hearing test?* Look for answers that relate to difficulties with hearing on one hand, or prompting from family or friends on the other.
- *What do you expect to gain from this visit?* The patient may either be hoping for proof that his or her hearing is normal, or may be seeking help with hearing, which may or may not involve hearing aids.

For those who already realize they have a hearing loss and need help, the second of these questions leads naturally to questions about the types of situations in which they need help (see next panel).

WANT. The Wishes and Needs Tool (WANT) comprises two other simple questions that reflect motivation, and which have been shown to relate to the success of the fitting:[438]

- *Overall, how much difficulty do you have hearing (when you are not wearing your hearing aids)?*
- *How interested are you in obtaining hearing aids?* This question is best asked after the client has been provided with information about his or her loss and the likely benefit of hearing aids.

HASP. The first subscale of the HASP (see Section 9.1.13) provides five questions on motivation, and the score obtained can be compared to percentile scores for a large group of patients.[801]

ALHQ. The Attitudes Towards Loss of Hearing Questionnaire (ALHQ) comprises 22 questions, which provides a score for each of:[1551]

- *Denial of hearing loss* – does the patient consider he or she has a problem?
- *Negative associations* – stigma-related concerns
- *Negative coping strategies* – fear and avoidance of communication
- *Manual dexterity and vision* – impacting on ease of handling small objects
- *Hearing-related esteem* – has hearing loss impacted on confidence

SPHA. The Self Perception of Hearing Ability (SPHA)[1383] asks patients: *On a scale from 1 to 10, 1 being the worst and 10 being the best, how would you rate your overall hearing ability?* Those who answered 6 or less were more likely than not to acquire hearing aids. Almost everyone who gives a rating of 1 or 2 acquires hearing aids and almost no one who gives a rating of 9 or 10 does.

HARQ. The Hearing Attitudes in Rehabilitation Questionnaire (HARQ)[183, 676] comprises 40 questions assessing attitudes towards impairment, loss-associated stigma, loss minimization, aid-associated stigma, acquiring hearing aids, pressure from others, and expectations.

medical practitioner or other health professional.[943, 955] Patients with long-standing, gradually acquired, high-frequency hearing losses often seem to be brought in by family members. These patients may not have had much opportunity to realize the extent of their loss – the gradual acquisition of the loss, the retention of good low-frequency hearing, and in some cases the minimal requirements to communicate with others, can understandably lead these patients to believe that their hearing is normal.

- ***Hearing impairment***: Finally, the physiological hearing impairment of the patient influences, but by no means determines, how much impairment, activity limitation, and participation restriction the patient *believes* he or she has. Degradation of frequency selectivity, temporal resolution and spatial processing ability are only partially correlated with the loss of sensitivity revealed by the audiogram. Despite this, the pure tone audiogram is a very good indicator of the overall degree of physiological impairment, is a moderate indicator of activity limitation, but is a poor indicator of participation restriction.

Arriving at the clinic at the behest of others is certainly no bar to successful hearing aid use: rather surprisingly, there appears to be no relationship between the original reason for seeking help (own initiative versus pressure from others) and the resulting use, satisfaction, or benefit that the hearing aids provide.[171, 592, 732, 1921]

Whatever the combination of factors that influences patients, unquestionably the patients' own acknowledgment of their hearing difficulties, and consequently their degree of motivation to do something about their problem, are strongly predictive of how much they subsequently wear their hearing aids.[188, 438, 525, 592, 730, 732, 1061, 1811]

Attitude varies on a continuum from being extremely opposed to obtaining hearing aids, to being extremely enthusiastic about obtaining them. Goldstein and Stephens (1981) present four attitude types drawn from this continuum and comment that negative attitudes may be frankly expressed or may require considerable clinical acumen to identify. The panel on the previous page shows some tools that can assist.

If attitude and motivation are so important, can they be altered? Even if a patient has unrealistic beliefs about every one of the above factors, his or her overall attitude towards obtaining hearing aids is likely to be a rational consequence of these beliefs.[955] If so, altering attitude is possible only if the underlying beliefs can be altered. This conceptualization of how health choices are made is known as the Health Belief Model.[464] Figure 9.1 shows a representation of this

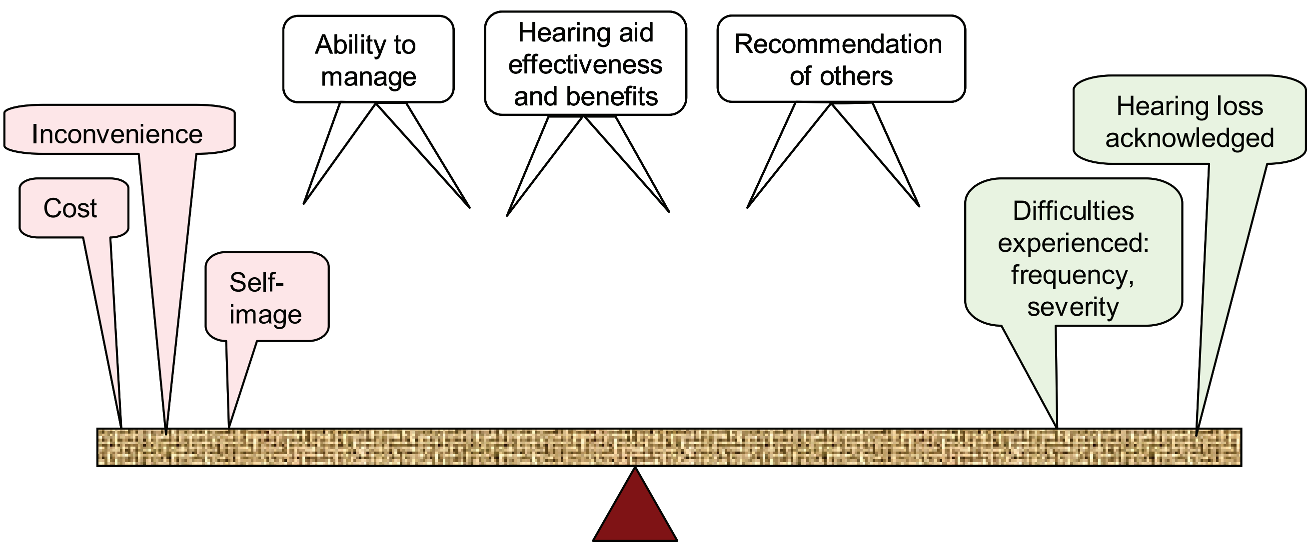

Figure 9.1 Visual representation of the Health Belief Model. Factors on the right may encourage a person to acquire hearing aids, whereas factors on the left are more likely to discourage take-up. Which way the balance tips may depend on the patient's views of the three factors in the middle.

model applied to hearing aid acquisition: depending on whether the patient's views on the three matters nearest the centre of the balance are positive or negative towards wearing hearing aids, and how much weight the patient puts on each positive and negative view, the decision to obtain hearing aids could easily swing in either direction.

Unfortunately, there is little research to indicate whether attitudes can be affected by appropriate counseling prior to aid fitting. Noble (1999) considers that counseling can lead to people accepting that their hearing loss is the cause of their difficulties, rather than ascribing the cause to the actions of others. Fortunately, we do know that time spent understanding the concerns of patients, and giving information and instructions to them before and after hearing aid fitting, will increase the amount that patients use their hearing aids.[188]

Some examples of how unrealistic and unhelpful beliefs of the patient can be challenged will be given in Section 9.1.14. First, we review the evidence for how a number of factors affect the likelihood of a person using, and/or receiving benefit from, hearing aids.

9.1.2 Pure tone loss and audiogram configuration

The benefit that people obtain from hearing aids, the proportion of hearing-impaired people who use hearing aids, and the number of hours per day that people use hearing aids, all increase with degree of pure tone hearing loss.[122, 269, 398, 533, 630, 1178, 1448, 1528, 1819] If one restricts the analysis to people with only mild and moderate hearing loss, however, degree of hearing loss is a very poor predictor of use or benefit.[185, 269, 438, 448, 452, 732, 788, 1895] Figure 9.2 shows that for mild losses, hearing aid ownership is not common, and amongst those who do obtain hearing aids, many discontinue using them.

The effect of hearing loss configuration on usage is unclear. One study has shown that people with flat audiograms use their hearing aids more than those with sloping audiograms,[1370] although other studies have shown no such effect.[52, 788] Disability appears to be more closely related to low-frequency hearing thresholds than to high-frequency thresholds, although both frequency ranges are important to good hearing.[126]

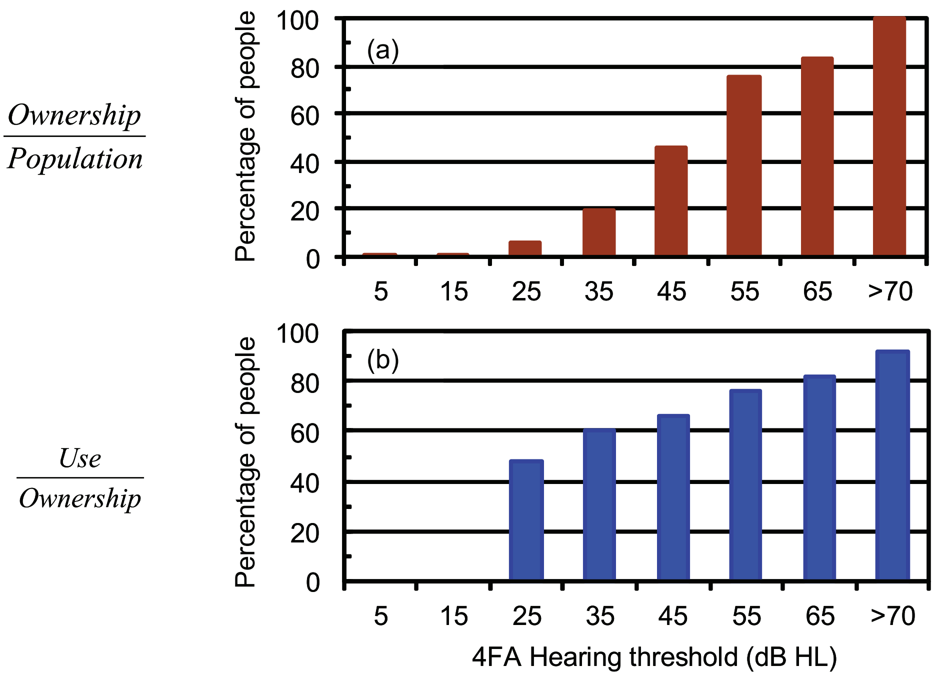

Figure 9.2 Part (a) shows the proportion of a population who have obtained hearing aids, by degree of hearing loss in the better ear. Part (b) shows those people who use their hearing aids as a proportion of those who have obtained hearing aids. The data are from a population survey of people over the age of 50 years in the Blue Mountains region of Australia.[692]

It has long been suggested, nevertheless, that people with three-frequency average losses (500, 1000, 2000 Hz) of less than 25, 30 or 35 dB HL will not benefit from hearing aids, but that people with greater losses will.[662, 947] Unfortunately, there are no data to suggest that pure tone loss is a reliable indicator of who will benefit, and plenty of data to suggest that it is not a reliable indicator, as the following examples show:

- Of 98 people with losses less than or equal to 20 dB HL at 500 and 1000 Hz, and less than or equal to 35 dB HL at 2000 Hz, 85% considered that after six months of use, the hearing aids were a worthwhile investment.[95] These people all had three-frequency average losses of 25 dB HL or less.
- A group of patients with normal or near-normal hearing up to and including 2 kHz received as much benefit from hearing aids as a group who on average had 52 dB of loss at 2 kHz.[1534]
- When hearing aids were offered to a group of patients with hearing loss, those willing to try them had no more hearing loss than those who declined. By contrast, the willing group had greater self-reported disability than those who declined.[1718]

If pure tone thresholds *are* used as a guide to candidacy, hearing thresholds in the ear with the larger pure tone loss should be used, as they appear to *better* predict hearing aid candidacy than loss in the better ear, at least for people with mild or moderate loss in both ears.[395] Haggard and Gatehouse (1993) point out that hearing in both ears should really be taken into account. For epidemiological purposes, they propose a two-part criterion for hearing aid candidature: four-frequency average loss greater than 35 dB in the better ear, or greater than 45 dB in the worse ear when combined with a difference of 15 to 35 dB between the ears. They caution, however, against using this to decide whether an individual should receive hearing aids.

Another problem with using hearing thresholds as a criterion for candidacy relates to the match of available technology to the impairment.[666] Suppose a patient has hearing thresholds of 0 dB HL up to and including 2 kHz, and a 25 dB loss from 3 to 8 kHz. Few people, including the author, would currently consider that such a person would benefit from hearing aids. But why not? The patient would have a small disability for soft speech or for medium-level speech masked by low-frequency noise. How would the small decrease in disability afforded by the aid, in limited circumstances, compare to the disadvantages

Special issues with a ski-slope hearing loss

- People with good low-frequency hearing are particularly likely to consider (inappropriately) that they do not have hearing problems, and to have been dragged in by relatives. This is especially likely if the hearing loss has been acquired gradually. If so, counseling to assess and modify attitude and motivation will be very important (see Section 9.1.14).
- The potential benefit from a hearing aid is least for extremely steep losses and where the high-frequency loss is greatest. The reason for this will be covered in Section 10.2.5, but relates to the diminished ability of an impaired ear to extract useful information from an audible signal when the loss becomes too great. The wider the frequency range where the loss is between 20 and 80 dB HL, the greater will be the benefit of the hearing aid. For all the reasons covered in this chapter, there is unlikely to be a single number (in octaves) that can predict whether hearing aids will be successful.
- The occlusion effect will be a problem that must be dealt with at fitting, almost certainly with an open fitting - until closed fitting, occlusion cancelling devices are available.
- A hearing aid that provides gain for only high frequencies may improve clarity, but have little or no effect on loudness. This lack of effect on loudness should be explained, and/or the clarity increase demonstrated with a speech test, otherwise the patient may believe that the hearing aid is ineffective.
- Further information about fitting people who have ski-slope hearing losses can be found in Harford & Curran (1997), and in Sullivan et al. (1992).

of wearing an aid? These disadvantages would potentially include a concern about appearance, cost, audibility of internal noise, the possibility of an occlusion effect or feedback oscillation, and the general nuisance value of wearing a prosthetic device.

Suppose the technology available enables high quality sound without any occlusion effect or feedback in a cosmetically acceptable package at a cost that is not too great for the patient. If the disadvantages are minor, the patient may consider hearing aids worthwhile even when they provide only a minor benefit. *In general, any audiometric criterion must take account of the technological solutions available*. The flexibility with which hearing aids can match unusual loss configurations is increasing rapidly.

Whether the loss is sensorineural or conductive affects the benefit provided by hearing aids. When listening unaided in low- and medium-level environments, a person with a conductive loss will have poorer speech recognition ability than someone with a sensorineural loss of the same degree.[254] The person with the conductive loss will, however, derive more speech recognition benefit from hearing aids than the person with the sensorineural loss. This greater benefit arises partly because for the person with a conductive loss, the unaided score is lower and partly because the aided score is higher. This better aided performance for the person with the conductive loss presumably results from the absence of the distortions that occur within and beyond the cochlea for people with sensorineural hearing loss (see Section 1.1). Surgical correction of conductive loss should have been considered before providing hearing aids to compensate for the loss.

In short, although pure tone hearing loss is seemingly obvious as a predictor of candidacy, it is unreliable as a sole indicator of who will benefit from hearing aids, except in the cases of normal hearing (no benefit) and severe hearing loss (substantial benefit). For all the people with losses in between, it is best to use hearing thresholds as a guide for further questioning.

Patients with a moderate hearing loss will not be able to hear parts of the speech signal in most listening situations. If they state that they experience no disability and therefore do not wish to obtain hearing aids, the reason should be investigated. Have they simply not noticed what they are missing due to the gradual onset of their hearing loss? Are they outwardly denying a disability they actually suspect they have? Have they structured their lifestyle and relationships to minimize the impact of the disability? If the latter, then are they happy with the changes to their life they have been forced to make? Conversely, if someone with almost normal hearing was desperately keen to obtain hearing aids, are the problems he or she is trying to solve consistent with the small amount of speech information that someone with such a loss would be missing? Are their expectations of hearing aids realistic? Clearly the person has needs, but are those needs of a type that can be met with an electroacoustic device?

9.1.3 Speech identification ability

People with poorer speech identification ability in quiet are, not surprisingly, more likely than people with higher speech identification ability to use hearing aids.[1448] High speech scores do not, however, prove that a person's hearing is too good to benefit from hearing aids.

Conversely, patients with the poorest speech reception thresholds in noise (i.e. high values of ***SRT_n***: the SNR needed to achieve some criterion intelligibility score) report the least benefit from amplification.[1880] Benefit decreases with age, however, and SRT_n loss increases with age, so it is possible that some other consequence of age is responsible for the apparent relationship between SRT_n and benefit.[1880]

Predicting candidacy of an individual from speech identification ability is not valid, because speech scores depend strongly on test conditions, such as the speech level, noise level, reverberation, spatial arrangement and distance of the sources, and difficulty of the speech material. Any conclusion that a person had no problems would be applicable only to the conditions under which the speech measurement was performed. Predicting whether a person could increase his or her understanding of speech in the wide range of circumstances that most people encounter would be a very daunting task.

A detailed literature review failed to produce any compelling evidence that speech identification ability prior to fitting, either in quiet or in noise, was related to the benefit or satisfaction received from amplification.[920]

9.1.4 Self reported disability

Not surprisingly, people who seek hearing aids are more likely to be aware of their hearing disability (i.e. activity limitation and/or participation restriction) than people who do not.[173, 717, 1019, 1448] Similarly, people who initially report the most disability are the most likely to report that they are helped by hearing aids.[95, 438, 852, 938, 1274] Those who most completely accept that they have a hearing loss use their hearing aids more than those who only begrudgingly accept that they have a loss.[823, 938] It is possible that some people who have not really accepted their need for amplification indicate their unwillingness by choosing to have a single hearing aid, despite having a bilateral hearing loss, although data on this issue are very limited.[616]

A possible approach to determining candidacy would be to administer a questionnaire that assesses disability while unaided. Unaided scores for the Abbreviated Profile of Hearing Aid Benefit (APHAB),[364] which assesses activity limitation, have been shown to be correlated with eventual reduction in activity limitation following rehabilitation, as well as with audiometric pure-tone thresholds.[349] Further details on the APHAB are given in Section 14.3.2. Similarly, unaided scores for the Hearing Handicap Inventory for the Elderly (HHIE),[1851] which assesses hearing participation restriction, have been shown to be correlated with eventual reduction in participation restriction.[d] Either could be used to help the patient decide if help is needed (see panel).

A low self-reported disability relative to the pure-tone loss is likely to reflect the person's inaccurate estimation of disability or unwillingness to acknowledge a disability, rather than an absence of problems.[1553] If there is an apparent mismatch between the amount of disability indicated by the questionnaire, and the degree of pure-tone hearing loss, further questioning of the patient may provide information that helps determine the appropriateness of fitting hearing aids. The situations described in some of the items on the questionnaire may provide a useful starting point for discussion. This may be particularly useful where the patient's response to any question was strongly different from that expected based on the measured hearing loss. Patients are likely to sum up their assessment of low disability with the simple statement "*My hearing is not bad enough to need hearing aids*".

A severe vision disability reduces the ability to speech-read (i.e. lip-read). When audition is not enough, such as in noisy places, or when hearing loss is severe, loss of vision is thus likely to increase hearing difficulty, and hence to increase the need for hearing aids.[523] Unfortunately, the same vision loss can also make it more difficult for the patient to manage the hearing aids.

9.1.5 Acceptance of noise

Hearing aids amplify every sound: excessive amplification of background noise is the almost-universal complaint that hearing aid wearers make about their hearing aids. When hearing aid benefit is measured using the APHAB measure, scores on the Aversiveness

Assessing the problem to be fixed: Self-report standardized questionnaires

If a patient is in some doubt about whether hearing aids are needed, administering a questionnaire to assess activity limitation (e.g. the unaided part of the APHAB) or to assess participation restriction (e.g. the HHIE) may be beneficial. Simply doing the questionnaire may help patients reflect on how much their hearing loss is impacting on their life. Scoring the questionnaire can add further information. Cox (1997) has suggested the following:

- Patients with relatively large problems hearing speech unaided and relatively few problems with intense sounds are likely to obtain significant benefit from hearing aids. This translates into unaided scores on the APHAB *Ease of Communication*, *Reverberation*, and *Background Noise* sub-scales of greater than 58, 75, and 74 respectively, combined with scores less than 24 on the *Aversiveness* scale.
- Patients with relatively few problems hearing speech unaided, but who find loud noise disconcerting, are particularly unlikely to benefit from a linear hearing aid.

[d] When a questionnaire score used to determine candidacy (e.g. a score representing a large unaided disability) is also used as part of the measure of benefit from rehabilitation (e.g. unaided disability minus aided disability), then the random error inherent in any measurement will create a correlation such that those with the greatest initial disability seem to receive the greatest benefit. An independent measure of benefit is therefore needed before claims about the predictive value of the unaided measure can be accepted.

scale indicate that environmental sounds bother people more when they are wearing their hearing aids than when listening unaided. Patients who are willing to listen to speech at very poor SNRs appear to be more likely to use their hearing aids than those who require a more positive SNR.[1295, 1296, 1301]

The test that has been developed to assess the SNR that people need for speech to be acceptable to them is called the ***Acceptable Noise Level*** (***ANL***) test. It is performed by first having the patient adjust speech to the Most Comfortable Level (MCL). Noise, such as speech babble, is then added and the patient is asked to adjust it to the highest level that he or she can accept or "put up with" while following the story being told in the original speech signal. The level selected is called the Background Noise Level (BNL).

ANL is defined as MCL minus BNL. It is the poorest SNR that is acceptable to the patient. Those with a small ANL (<7 dB) are likely to become full-time users of hearing aids (because they are willing to put up with amplified noise close in level to the signal of interest). Conversely, those with a large ANL (>13 dB) are likely to use hearing aids less, or not at all, because they find amplified noise objectionable in too many situations.[1296, 1301] Of course, there is a large grey region in the middle (ANL values between 7 and 13 dB) for which the acceptance of hearing aids is uncertain.

In these studies, full-time users were considered to be those who used hearing aids whenever they needed assistance with hearing, however few hours per day that was, and part-time users were those who used hearing aids for only part of the time that they needed assistance, however many hours per day that was. In some cases, "part-time users" wore their hearing aids for much longer each day than the "full-time users". ANL values are none-the-less also correlated to the number of hours per day that hearing aids are used.[1296]

ANLs can be obtained in the sound field or under headphones, with all frequencies amplified by the same amount, or with frequency-dependent amplification prescribed on the basis of the audiogram. The ANL values are somewhat affected by the competing signal used (speech-shaped noise or multi-talker babble), although patients with a relatively high or low ANL will be categorized the same way no matter which type of noise is used.[572] Aided ANLs have been reported as being the same as unaided ANLs in some studies,[1266, 1296] but as smaller than unaided ANLs in another study.[e, 19] Aided ANLs are particularly likely to be smaller than unaided ANLs when speech and noise are spatially separated,[19] and even more so when the speech is presented from the front and noise from the rear and the hearing aids have directional microphones,[570] or contain an adaptive noise reduction algorithm.[1266]

ANLs are not related to gender, hearing thresholds, loudness discomfort levels, acoustic reflex thresholds, or contralateral suppression of otoacoustic emissions, and are only weakly correlated to age and speech intelligibility scores in noise.[19, 687, 1296, 1299] The lack of gender difference in ANL occurs despite males having higher MCLs and BNLs than females,[1522] which is consistent with the higher gains prescribed to males by the NAL-NL2 formula (Section 10.4.6). ANL values are the same for normal-hearing adults and for children, despite MCL being higher for the adults than for the children.[1247] ANL values are decreased by stimulant medication for experimental subjects with attention deficit/hyperactivity disorder.[573]

Intriguingly, those with a low ANL (i.e. more accepting of noise) have lower-amplitude wave V peaks in their auditory brainstem responses (ABR) and Na-Pa peaks in their middle latency responses (MLR), but no difference in their wave I or wave III amplitudes.[688] This suggests that people who can accept more noise relative to speech have less-easily excited afferent (i.e. ascending) pathways in their upper brainstem, perhaps because their efferent (i.e. descending) pathways are more effective in inhibiting unattended sounds. This central origin of inter-patient differences in ANL is consistent with the finding that ANL values measured monotically (speech and noise in same ear) are correlated with, but not equal to, those measured dichotically (speech and noise in opposite ears).[687]

9.1.6 Listening environment, needs and expectations

Hearing aids provide much more benefit in some situations (e.g. listening to a softly spoken person in a quiet place) than in others (e.g. listening to a loudly spoken person in a noisy, reverberant place).[363, 441, 1159] The reason for this is easy to understand. Figure

[e] The effect of aiding on ANL is likely to depend on whether the additional frequency region made audible by aiding has a smaller or larger SNR than the frequency region audible in the unaided condition.

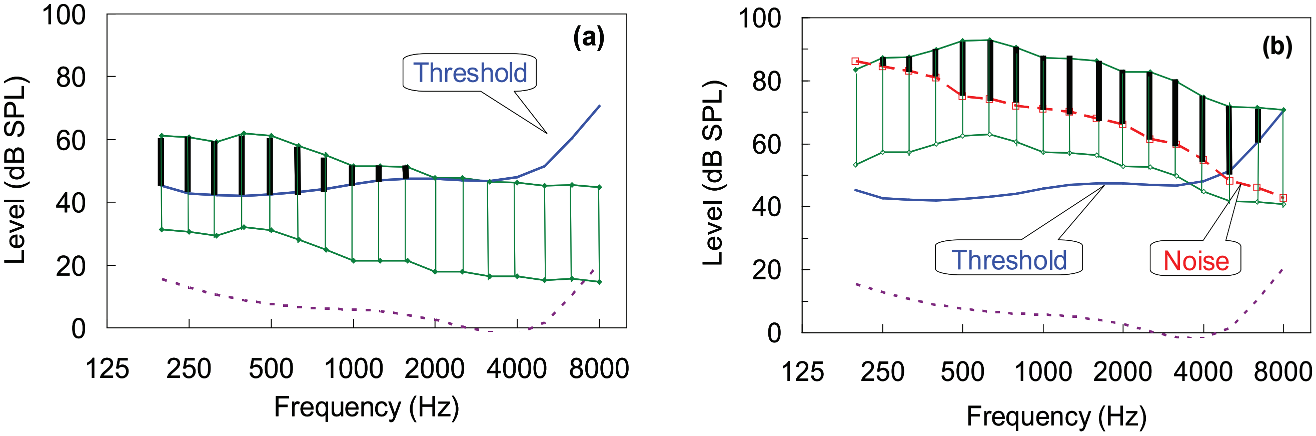

Figure 9.3 The long-term 1/3 octave speech spectrum for a) speech at 55 dB SPL in a quiet place, and b) speech at 85 dB SPL in a noisy place. Each speech spectrum includes the 30 dB dynamic range from the weakest useful elements of speech to the most intense elements (shown as the vertical lines). The portion of the speech range that is audible above noise and hearing thresholds is thickly shaded. The normal threshold of hearing is shown as the lower violet dotted line.

9.3(a) shows the long-term speech spectrum at an overall level of 55 dB SPL, in a quiet environment. It also shows the normal threshold of hearing and the thresholds corresponding to a person with a hearing loss that gradually increases from 30 dB HL at 250 Hz to 50 dB HL at 8 kHz. It is clear that much of the speech signal falls below the threshold of the hearing-impaired person. Figure 9.3(b) shows the spectrum of speech at 85 dB SPL in a noisy place where the background level is 80 dB SPL. Much of the speech will again be inaudible but, at most frequencies, audibility is limited by the background noise rather than by the person's hearing loss.

What will a hearing aid do in these two hypothetical but realistic situations? In the quiet situation, if the hearing aid has enough gain, the entire 30 dB range of speech could be made audible at every frequency, and intelligibility would increase dramatically. In the noisy situation, the situation is different for all frequencies less than 5 kHz. No amount of gain will make the speech more audible over this frequency range, because the hearing aid will amplify the noise just as much as it does the speech. Not surprisingly, hearing aid users report much more satisfaction with their hearing aids in quiet than in noise.[1179] Inadequate benefit in noise is the major reason for both lack of purchase of hearing aids and return of hearing aids after purchase.[955]

How does this affect candidacy for hearing aids? If the patient primarily needs to hear more clearly in places that are quiet and where speech is at a soft level, hearing aids are likely to be extremely useful. The greater the loss, and the older the patient, the more likely it is that improved intelligibility in quiet is indeed an important need.[1180] The greater the loss, the greater will be the likely benefit in quiet places.[363]

Conversely, the lower the loss, and the younger the patient, the more likely it is that improved intelligibility in noise is an important need.[1180] If the primary need is to hear better in *very* noisy places, hearing aids may disappoint, irrespective of the degree of the patient's hearing loss. Many real life situations lie between these extremes. Often, background noise will limit audibility for the lower frequencies, and the patient's thresholds will limit audibility for the higher frequencies. In such situations, hearing aids will provide more benefit than they do in very noisy places, but less benefit than they do in very quiet places. The greater the range of situations in which hearing aids benefit the patient, the greater the patient's satisfaction will be.[942, 945] Socially active patients use their hearing aids in a greater range of situations, and consequently report greater benefit and satisfaction.[1178]

As a rule, people who have not used hearing aids before expect that the aids will be as beneficial in noise as they are in quiet, although they subsequently report this is not the case.[1578] On average, patients' expectations about how much hearing aids will improve the clarity of speech in all situations is slightly higher than they subsequently find to be

Determining listening needs and expectations: the COSI™

The only way to find out where a patient needs to hear better is to ask! A systematic way to accomplish this is to use the ***Client Oriented Scale of Improvement*** (***COSI***).[448] This technique consists essentially of a blank form (see Section 14.4), on which the clinician records the situations that the patient nominates as being difficult. The situations to be recorded often emerge naturally while the patient's case history is taken. If not, the blank form will remind the clinician that something important is missing from the interview.

It is worth continuing to ask the patient for further examples of situations that are difficult as long as he or she is able to keep on nominating them. Five different situations, however, are probably enough to provide a focus for the rehabilitation program. The first two situations are likely to be particularly important, and patients will re-nominate these same situations if asked on a later occasion.[1250] As we will see in Chapter 14, the situations listed at the initial interview can later be used to assess the benefit of rehabilitation. To get full value out of this assessment, the initial needs should be recorded as specifically as possible. For example *hearing my granddaughter when she comes to visit* is more specific than *conversation at home*. Similarly, *understanding Sam and Lou at the club on Saturdays* is more specific than *hearing in noise*.

Determining the situations in which the patient is having problems leads naturally to determining the patient's expectations of hearing aids in each of these situations. Patients often arrive at the initial appointment with expectations that are unrealistically high (especially if the situation is noisy) or unrealistically low (especially if the situation is quiet).[982] In either case, the clinician should modify the patient's expectations so that the patient can make a realistic, personal overall assessment of the benefit of hearing aids.

McKenna (1987) has suggested a very specific way to determine expectations. Patients are asked how well they would need to hear in each situation for them to consider the rehabilitation worthwhile. If the expectation is unrealistic, the clinician and patient jointly negotiate a goal that the patient thinks is worthwhile and that the clinician thinks is achievable. An additional benefit is that both clinician and patient know that the rehabilitation program will be over once these goals have been reached, or when it is apparent that some of the goals can never be reached.

This goal setting approach was implemented in NAL hearing clinics throughout Australia and referred to as ***Goal Attainment Scaling*** (***GAS***).[451, 452] Some clinicians considered that the formal negotiation of realistic goals was useful, whereas other clinicians disliked it. The technique was replaced by the COSI technique. The ***Patient Expectation Worksheet***[1380] is essentially the same as GAS, except that both the level of performance desired by the patient, and the level of performance thought likely by the clinician are separately recorded.

Stephens (1999) recommends mailing information to patients prior to the first interview, and that patients be asked to think about help with hearing. This approach should facilitate COSI, though it is not an essential component of COSI.

true.[345, 1578] They also underestimate the negative aspects of hearing aids, such as feedback oscillation and the amplification of background noise.[133] Some studies have shown that those patients who arrive with the highest expectations of what hearing aids can do, eventually report the greatest use and benefit.[345, 823, 1554, 1963] A positive correlation between pre-fitting expectations and resulting benefit or satisfaction is not universally found, however.[1177, 1349] Furthermore, and not surprisingly, patients for whom actual performance falls short of their pre-fitting expectations (a discrepancy referred to as ***negative disconfirmation***) are more likely to report low satisfaction with their hearing aids.[1929] It seems important, therefore, that patients with *grossly* unrealistic expectations (in either direction) be identified prior to their deciding whether to try hearing aids, so that their expectations can be altered, and they can base their decision on the best possible information.[1554] This can be achieved by counseling, or by demonstration of the sound quality that is possible in different situations using recordings or computer simulations.[1554]

Two provisos should be added to the preceding discussion about different benefits in different environments. Anyone with a hearing loss will undoubtedly have more difficulty hearing in both very quiet and very noisy situations than would someone with normal hearing. Even if a patient initially reports difficulty in only one of these situations, it is worth questioning to find out if the other type of situation ever causes problems. If so, the difficulty experienced in the other situation, and the importance of this situation, may have a bearing on the likely overall benefit of hearing aids for this patient. As well as examining the type of situations in which a patient needs help with hearing, it is worth considering how many such situations the person regularly experiences and how often these situations are experienced. The more a person is in contact with other people (or would like to be if poor hearing was not a disincentive) the more likely it is that the advantages of hearing aids will outweigh their disadvantages.[1178] A hermit with a loud TV set and radio does not need hearing aids!

The second proviso is that if the hearing aids have a directional microphone, or dual microphones that can be selected to function as a directional microphone, significant benefit from the hearing aids *may* be experienced in even the noisiest places. When noise and the wanted signal are coming from different directions, hearing aids with directional microphones may allow the patient to communicate in poorer signal-to-noise ratios than would be possible without hearing aids. The directional microphone will decrease the amount of noise being perceived, and increase the SNR, but by an amount that depends hugely on the specific listening situation (Section 7.3.1). Technology thus directly affects benefit, and hence candidacy. People with mild hearing losses usually appreciate the physical comfort and freedom from occlusion that an open fitting gives. As explained in Section 7.3.5, this means that the hearing aid is not directional in the low frequencies. The directional microphone will help only for those frequencies high enough to be unaffected by the open ear canal, and low enough that it is noise limiting audibility rather than hearing thresholds. In the example shown in Figure 9.3(b), this may correspond only to the frequency range 1500 to 4000 Hz.

The amount of benefit that omni-directional and directional hearing aids provide to people with mild or moderate loss can be calculated using the Speech Intelligibility Index calculation method,[54] by assuming the speech levels that are typical in environments with different noise levels,[1399] and assuming a typical noise spectrum.[863] Figure 9.4 shows the calculated speech intelligibility for unaided listening, aided listening with both omni-directional and directional microphones, and normal hearing.[443] The hearing-impaired person has the greatest difficulty, both absolutely, and relative to normal hearers, in the noisiest places.[1178] Unfortunately, the hearing aid, whether in omni-directional or directional mode, helps least in the noisiest places.[1178] As input level increases, the gain of the hearing aid decreases, which decreases the frequency range over which the directional microphone can improve the SNR.[101] The hearing aid, most needed in noise, most helps in quiet! Although these are theoretical calculations, they rest on well-established methods, and the results are consistent with experimental results and previous data-based models.[1001, 1435]

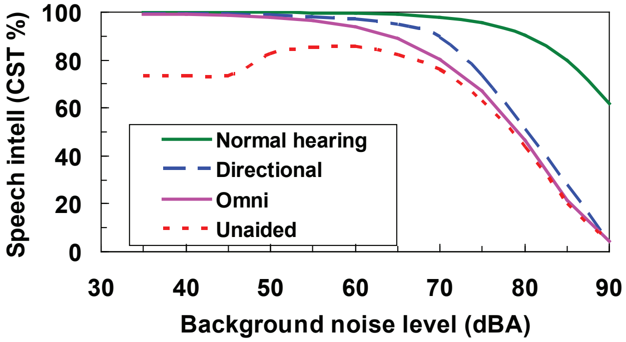

Figure 9.4 Predicted speech intelligibility (for Connected Speech Test) for a patient with a mild-moderate hearing loss (30 dB at 500 to 1000 Hz, sloping to 60 dB HL at 4 kHz), unaided and with hearing aids meeting the NAL-NL2 prescription. Microphone directivity is assumed to improve SNR by 3 dB over the frequency range for which insertion gain exceeds 3 dB.

9.1.7 Stigma and cosmetic concerns

Patients are often concerned about the appearance and/or visibility of hearing aids because of what they believe the hearing aids signify, either to themselves or to others, or both.

Perceived age

Many adults are concerned that they will be perceived as being older if they wear hearing aids. This is a very understandable concern: hearing aids are worn by a much bigger proportion of elderly people than by younger adults, and many middle-aged people do make the association between hearing aids and old age.[576] There have been several studies of whether hearing aids make people look older. When judgments are based on photographs, hearing aids do make adults look older, but the age difference is so small (less than one year) that the effect has no practical significance.[1269] In addition, the peers of elderly people do not view negatively those who wear hearing aids.[391, 532] Furthermore, young adults do not associate hearing aids with aging and do not consider that hearing aids are noticeable.[310]

Based on the finding that older people are more willing than younger people to adopt hearing aids, for the same self-reported disability, Kochkin concludes that stigma linking hearing aids with age is the major reason for the limited penetration rate of hearing aids.[955] Despite this, the research reviewed in the preceding paragraph suggests that ***external stigma*** (i.e. stigmatization by others) is not actually a strong phenomenon. Patients may come to realize this only after they have worn hearing aids. In one study, 26% of hearing aid candidates initially believed that other people would view them as being older if they wore hearing aids.[1754] After these people had worn hearing aids for six months, however, only 10% believed this to be the case. Consistent with this, there is less stigma attached to hearing aids than to hearing loss.[532]

Hearing aid wearers actually have a more positive self-image than hearing-impaired people who don't wear hearing aids,[689] although it is unclear whether hearing aids contribute positively to self-image, or whether people with a positive self-image (perhaps associated with a higher self-efficacy towards hearing aids) are more likely to acquire hearing aids. It seems reasonable to reassure candidates that they will *not* be viewed negatively if they wear hearing aids. Indeed, hearing aid wearers believe that their hearing loss is *less* likely to be noticed when they wear their hearing aids than when they do not.[1754]

Size matters

Some people would rather pay for a less obtrusive hearing aid than have a more obtrusive hearing aid for free,[190] and are willing to pay more for a CIC than for a larger style hearing aid.[10] When hearing-impaired people who have never worn hearing aids are shown photos of different hearing aid styles, the proportion who say they intend to purchase increases as the size of the hearing aid decreases.[944] Twice as many people say they are likely to purchase a CIC fitted below the ear canal entrance than would purchase an older style (i.e. large) BTE, price considerations excluded. Micro BTEs with thin-tube are, however, almost as invisible as CICs (Section 11.1).[830] Concern over appearance can apply to patients of any age, but older adult patients are less likely than younger patients to report dissatisfaction with the appearance of hearing aids.[721]

For many patients, choosing a suitably small and suitably colored device can usually overcome cosmetic concerns. (Unfortunately, the emphasis on small size in hearing aid advertising also reinforces the belief that hearing loss should be hidden from others.) Patients who need more gain or power, or who cannot manipulate an aid as small as a CIC, an ITC, or a micro BTE can be encouraged to reassess the importance of hiding their hearing loss relative to the benefits they could obtain from hearing aids (Section 9.1.14). Deeply-seated, long-term placed CICs provide a solution that has no manipulation difficulties or visibility.

Unspoken concerns

Some patients will consider that concern about appearance indicates vanity on their part, and so may not voice any comment about the appearance or visibility of hearing aids, even when asked. The clinician should thus consider that concern with appearance is possible, even where the patient appears to be unconcerned. People who are concerned about the appearance of their hearing aids are less likely to wear their hearing aids unless adequate counseling is provided.[188]

In summary, the presence of hearing aids will lead some people to view the aid wearer as being older, but the magnitude of this effect is small. The important issue is whether, and how strongly, the patient believes that hearing aids will affect his or her appearance. The patient's internal view of stigma, including the patient's image of his or her self as a hearing aid wearer, is what counts.

9.1.8 Manipulation and management

Operating hearing aids can be very difficult for many people and manipulation difficulties can preclude hearing aid use unless carefully managed by the clinician. Manipulation difficulties may be caused by poor flexibility of finger or arm joints or by low tactile sensitivity. Low cognitive functioning can also prevent patients from properly operating their hearing aids.[1089] The size of hearing aids can make inserting a battery, and operating the volume control and on-off switch difficult. Because the ears are out of sight, insertion of a hearing aid can be a difficult task to learn. If any one of these tasks is too complex, a patient may simply give up trying, and this is most likely to occur for patients with the poorest finger dexterity.[778] Difficulty inserting an earmold appears to be the major reason for ceasing to use hearing aids, at least for BTE hearing aids,[187] although part of the reason for the strong association is that lack of use causes poor apparent insertion skill.

Although management difficulty typically increases with each decade past 60 years of age,[422, 721, 804, 1387, 1717] there are so many counter-examples that age cannot reliably be used to predict how much trouble people will have handling their hearing aids. Furthermore, patients may continue to use their hearing aids even if they are having difficulty managing them. A majority of elderly patients, even those over the age of 90, reported using their aids regularly, despite the management difficulties many of them reported.[1089, 1387, 1389]

Because ease of management is so closely linked to patients' use of their hearing aids, it is a critical issue in aid selection and patient instruction.[90, 122, 565, 730, 778, 1017, 1176, 1911] The older a patient is, the more likely it is that he or she will regard ease of management as the most important aspect in hearing aid selection.[1176] For some patients, particularly for some in nursing homes, the inability to manage a hearing aid will be so great that hearing aids simply cannot be used. In these cases, use of one or more assistive listening devices (ALDs; e.g. infra-red or radio-frequency devices for TV listening, or body aid or hand-held devices with output through headphones to enable staff or visitors to talk to the patient) should instead be considered. The combination of larger controls and more obvious functionality can facilitate their use by the client, if necessary with the assistance of a carer.[1042] Section 11.1 contains some recommendations for choosing hearing aids that will help minimize management problems.

On the other hand, some perfectly capable patients initially may be fearful of not being able to manage hearing aids. Practical experience, with affirmation from the clinician, is usually all that is needed. The self-efficacy that a patient feels he or she has regarding management of a hearing aid can be assessed with a questionnaire (the Measure of Audiologic Rehabilitation Self-Efficacy for Hearing Aids; MARS-HA)[1911] if the clinician wishes to quantify the patient's beliefs in this area or diagnose which aspect(s) of self-efficacy are the problem.

Although low cognitive ability may contribute to difficulty in managing a device, low cognition may also make hearing aid usage more necessary. It is possible that people with higher level cognitive ability are more able to find ways to cope with hearing loss without using hearing aids. Difficulty in handling a hearing aid, whether caused by poor cognitive skills or poor fine motor skills, has obvious implications for device selection, as discussed in Chapter 11.

Checking for ability to manage the hearing aid

- Before finalizing the aid selection, hand the patient an aid of the style and size you are considering fitting.
- Assess the patient's reaction to the size of the aid.
- Show the patient how to change the battery or turn the aid on and off. Have the patient try it and assess how easily the task is learned. Initial failure does not mean that the task can never be learned, but early success is very reassuring to the patient and the clinician about this aspect of candidacy.

9.1.9 Age

Age (either old age or infancy) by itself does not directly affect candidacy for hearing aids.[938] It can, of course, affect several of the other factors already reviewed (manipulation difficulties, cosmetic preferences, hearing needs, hearing impairment) and so indirectly affect candidacy. Increasing age also increases the likelihood of an auditory processing disorder (Section 9.1.11) and/or need for greater SNR in noise (Section 9.1.3), both of which decrease the probability of hearing aids being beneficial.[707, 1880] Among adults who own hearing aids, daily use and benefit are probably less for the old-elderly than for the young-elderly,[1688, 1752, 1880] although some studies have found no relationship between age and use of hearing aids.[185, 767] Satisfaction with hearing aids decreases as age increases.[767]

Illness often accompanies advancing age. People with a hearing loss may regard other health problems as more pressing than their hearing problem, and may not feel able to deal with more than one problem at a time. This is understandable but unfortunate, as increased ease of communication may make it easier for them to deal with other problems they have. The prevalence of dementia also increases with age, but dementia, at least of mild and moderate degree, does not preclude use of hearing aids, and the fitting of hearing aids to people with dementia does not, on balance, increase the burden on carers.[32]

The younger people are when they are first fitted (e.g. under 70), the more likely they are to become regular users of their hearing aids.[25, 182, 187] Advanced age (e.g. over 80) makes it harder to learn the new tasks required to operate hearing aids, whereas these skills are more easily retained into advanced age if they have been learned at an earlier age.[955] Perversely, these younger people most likely to adapt well to hearing aids are least likely to try them.[182]

9.1.10 Personality

Several personality traits are associated with a greater likelihood of take-up of rehabilitation, and a greater degree of self-reported benefit from rehabilitation with hearing aids.

Internal locus of control: Patients who feel that they control the things that happen to them (rather than control by others or by random events) are more likely to acquire and use hearing aids than those who feel that things just happen to them (i.e. an external locus of control).[350, 588, 592] Closely related to an external locus of control is ***learned helplessness***.[1638] This phrase reflects a belief by some patients that, based on their experiences, they cannot positively affect their circumstances no matter what they do, so there is no point in doing anything. Counseling for such patients should presumably be aimed at helping them realize they *can* change how well they hear. Those who more strongly believe that their lives are controlled by others are also more adversely affected by loud sounds, but this applies whether they are aided or unaided.[348]

Extroversion: Patients with an extroverted (outward looking) personality report more benefit from amplification, and less activity limitation and participation restriction when aided, than patients with an introverted (inwards looking) personality.[348, 351]

Agreeableness: Patients who are more trusting, peaceable, sympathetic, helpful, and who believe that others will want to help them are more likely to obtain hearing aids, at least in hearing service systems where patients have to pay for their service and device. For patients who are more suspicious, shrewd and demanding, there may be a barrier to overcome associated with the patient's distrust of either the motives of the clinician seeking to help them, or of the likely effectiveness hearing aids because of negative stories they have heard from others.[350] For such people, trust-building will be an important part of the clinical encounter. Twenty-four percent of hearing-impaired people who have not obtained hearing aids reported that lack of trust of a physician or clinician influenced their decision to remain without hearing aids.[955] More agreeable patients are also subsequently more positive about the way they view themselves, and the way they believe others view them, as hearing aid wearers.[351] In this sense, they are indeed better hearing aid candidates.

Obsession: Patients who score highly on an obsession scale are likely to report less benefit and satisfaction from their hearing aids.[592]

Openness: Patients who are more open (variety-seeking, curious, insightful, broad-minded, analytical) are less likely to obtain hearing aids, perhaps because they use their open nature to reduce the problems their hearing loss causes in ways that do not require the use of hearing aids.[350] These patients can be congratulated on their resourcefulness, but reminded that hearing aids provide solutions that are complementary to those their good communication tactics are already providing. Less open patients may prefer to receive practical, problem-solving information directed at the specific situations in which they are experiencing listening difficulty.

Neuroticism: Patients who score more highly on a Neuroticism scale (prone to worry, experiencing frustration, discouragement, feelings of inferiority, and sensitivity to ridicule) are less likely to obtain hearing aids,[350] despite reporting a greater degree of disability (when unaided) arising from their hearing loss.[351] They also report greater disability than less neurotic patients when aided. Perhaps more than others, these people feel that hearing loss and hearing aids carry a stigma and so seek to avoid embarrassment and shame.[350] See Section 9.1.7 for information about stigma that might be helpful to these patients.

The association between benefit from rehabilitation and the presence of the above traits is far too weak to establish or preclude candidacy for hearing aids for any individual. If a trait is strongly evident, however, this can be one of the many factors taken into account in determining which information to provide to a patient and how to provide it, and again when the clinician is deciding on the final recommendation for or against candidacy. To a great extent, information readily obtainable from the patient prior to fitting (such as the disability they experience, their expectations about hearing aids, and their aversiveness to loud sounds) already reflects their personalities, so it is not productive to spend clinical time quantifying their personality.[351]

9.1.11 Central auditory processing disorders

As age increases, so too does the likelihood that hearing loss will involve a decrease in central auditory function.[631] The person (or ear) affected will be particularly susceptible to interference from competing signals. Several studies have shown that an auditory processing disorder (as evident on a dichotic speech test) diminishes the use, benefit, performance and/or satisfaction that hearing aids provide.[301, 623, 1686, 1688] Conversely, one case study showed that a person with central complications but normal pure tone thresholds in both ears benefited from a single hearing aid because it *decreased* adverse interactions (***binaural interference*** – see Section 15.4.2) between the ears.[1624] Another study showed no relationship at all.[983] Diverse conclusions about the impact of auditory processing disorders are not surprising considering the diverse types of auditory processing disorders that exist, and the diverse outcomes measures that can be assessed.

One form of central auditory processing disorder is almost guaranteed to be present in people with sensorineural hearing loss. Spatial processing disorder (see Section 1.1.6) appears to be a very significant contributor to the deficit in SNR experienced by hearing-impaired people.

The presence of an auditory processing disorder should therefore *not* prevent the clinician from fitting hearing aids. The presence of a central processing disorder may, however, help explain why some people report little benefit from hearing aids, and may impact on whether unilateral or bilateral aids are fitted.

Because central-processing deficits can appear and/or increase in magnitude as people age, it is possible for hearing aids to become less effective with time. When this occurs it is possible that people will complain of the hearing aid output becoming distorted, even though the hearing aid electroacoustic performance remains unchanged.[1689] There are few data on this issue, but the possibility of increasing central deficit, and decreasing hearing aid effectiveness should be borne in mind if previously satisfied hearing aid wearers indicate a growing dissatisfaction with their hearing aids. A more thorough review of the impact of central processing disorder on communication ability, and its implications for hearing aid candidacy, can be found in Stach, Loiselle & Jerger (1991).

Wireless systems (Section 3.6) provide a potential solution to the problems caused by central auditory processing disorders because of their ability to greatly attenuate unwanted signals and noise. A proportion of people with such deficits will use these systems regularly, despite the logistical difficulties associated with their use.[1687] A less effective but more convenient solution would be to use a directional microphone, either in the hearing aid or in the form of a highly directional hand-held microphone. If a person (adult or child) has normal pure tone thresholds, good speech discrimination ability in quiet, but unusually poor speech discrimination in noise - for whatever reason - consideration should be given to fitting a wireless system.

9.1.12 Tinnitus

Many people with hearing loss also have tinnitus. The amplification of external sounds can often relieve the adverse effects of tinnitus, including its psychological effects.[558, 710, 1590] Amplified sound can provide partial or even complete masking of the tinnitus, but one cannot assume that either of these will necessarily occur as, on average, the reduction in tinnitus-related problems offered by the hearing aid is relatively small.[1186, 1750, 1751] As tinnitus is often best masked by high-frequency sounds, open fittings can be an effective way to provide the amplification needed for both speech intelligibility and tinnitus masking. The combination of tinnitus retraining therapy, open-fit hearing aids during waking hours, and a noise maker ("sound enrichment") device near the bed during sleep has been shown to markedly reduce tinnitus problems for patients with mild, sloping hearing loss.[416]

It is becoming more common for hearing aids to include an optional controllable internal noise source so that tinnitus masking is not dependent on serendipitous internal noise or on amplification of noise in the environment.

The presence of tinnitus increases the likelihood that a person will accept hearing aids, presumably because of the masking that amplification provides, and should therefore be considered a positive factor when assessing hearing aid candidacy.[1718] Unfortunately, those with tinnitus are also likely to report higher aversiveness to sound when using hearing aids.[49] The use of hearing aids does not preclude other forms of treatment for tinnitus.

9.1.13 Factors in combination

As we have seen, a difficult audiogram (e.g. a ski-slope loss) does not rule out use of hearing aids. Neither does difficulty manipulating hearing aids, a belief that help is needed only in noisy places, poor speech discrimination, concern about the appearance of a hearing aid, arthritic fingers, nor a slightly hesitant attitude to trying hearing aids. However, a patient for whom all of these were true is less likely to find hearing aids useful than would a patient for whom there was only one of these difficulties to overcome. The clinician's job is to identify, for each patient, all the potential obstacles to success (see panel), overcome those that can be overcome (with technical solutions or by helping the patient modify his or her beliefs), and weigh up the remaining difficulties against the likely benefits that the patient will receive. A responsible recommendation for or against amplification can then be given.

The potential obstacles are usually identified through discussion with the patient. A supplementary approach is to ask the patient to complete a questionnaire. As well as saving clinician time, this may prompt the patient to reflect on his or her attitudes in advance of the assessment appointment, enabling more efficient use of time during the appointment. A suitable questionnaire with well developed statistics is the Hearing Aid Selection Profile (HASP).[801] Comparing the patient's score to the published percentile scores quickly shows if any of the domains assessed are likely to provide an obstacle to successful hearing aid use. If so, the corrective action may lie

Summary: Potential obstacles to the acquisition and use of hearing aids

1. little or no disability is perceived by the patient;
2. little or no handicap is experienced by the patient;
3. stigma, based on an association of hearing loss with old age, low social competence, or even mental disorder;
4. belief that hearing aids provide little help and/or a poor quality of sound;
5. passive acceptance of the inevitability that hearing loss, disability or handicap come with old age;[781]
6. attribution of problems to the actions of others and/or belief that the patient cannot do anything about the problems;
7. reinforcement of any of the above beliefs by friends, relatives, or health professionals;
8. difficulty manipulating small objects;
9. low cognitive functioning;
10. other health problems;
11. mild hearing loss, limiting the range of situations in which hearing aids increase audibility;
12. sensitivity to noise;
13. financial cost.

Note that for the first six items it is the belief that affects behavior, irrespective of whether the belief is well-founded. Experimentally, the first five beliefs have been found to distinguish those who use hearing aids from those who do not attempt to obtain them.[173]

in further counseling, the choice of hearing aid style and features to minimize the problem, or a joint decision with the patient not to proceed with hearing aid fitting at this time, along with the reasons why. Figure 9.5 shows the percentile scores, plus the results of one patient for whom success with hearing aids is likely to be limited by apprehension about technology, and difficulty in manipulating small objects. Such a result would have clear implications for the level of explanation given about hearing aids (greatly simplified), and the choice of user-operated features (none) on the hearing aid selected.

The clinician's job is also to give the patient all the information that he or she needs to make a well-informed decision about whether to proceed. What impression should the clinician convey if the clinician is uncertain whether the hearing aid will be useful? An obviously negative or uncertain attitude may well be a self-fulfilling prophecy if the patient is not given the confidence to persevere with the hearing aid during any initial difficulties. On the other hand, glowing predictions of wide-ranging substantial benefits without difficulties will be untrue, and will produce unrealistically high expectations that will make difficulties encountered seem all the greater.

An ethical, but encouraging, summary is that there will definitely be many situations in which the hearing aids will make speech easier to understand, but that there will be some limitations or difficulties to overcome, and there will be some situations in which the hearing aid provides little or no help. Furthermore, the patient is the only person who can decide whether the advantages outweigh the disadvantages, and this balance can be judged effectively only when hearing aids are worn.

9.1.14 Counseling the unwilling patient: some examples

Clinicians will frequently encounter patients that have so much pure tone loss that they *must* have trouble hearing clearly in a range of situations, yet the patient seems unwilling to try hearing aids. As an (all too frequent) example, suppose Mr X has been "brought" to the hearing clinic by Mrs X, who is sick of Mr X failing to follow conversations, and is annoyed by the TV volume setting insisted on by Mr X. (Their neighbors do not like the volume setting either!) Mr X says that he can understand most people most of the time, except for people who do not speak clearly. He can understand the TV just fine, and he came along only because his wife wanted him to. What does the clinician do next (assuming an audiogram has been obtained)?

Any attempt to immediately point out how much benefit hearing aids will give to someone with his degree of loss will probably be meaningless to Mr X, because he has not acknowledged that he has a problem and therefore does not need a solution. Such an attempt may cause Mr X to label the clinician as just another person telling him what to do. Even if he acquiesces

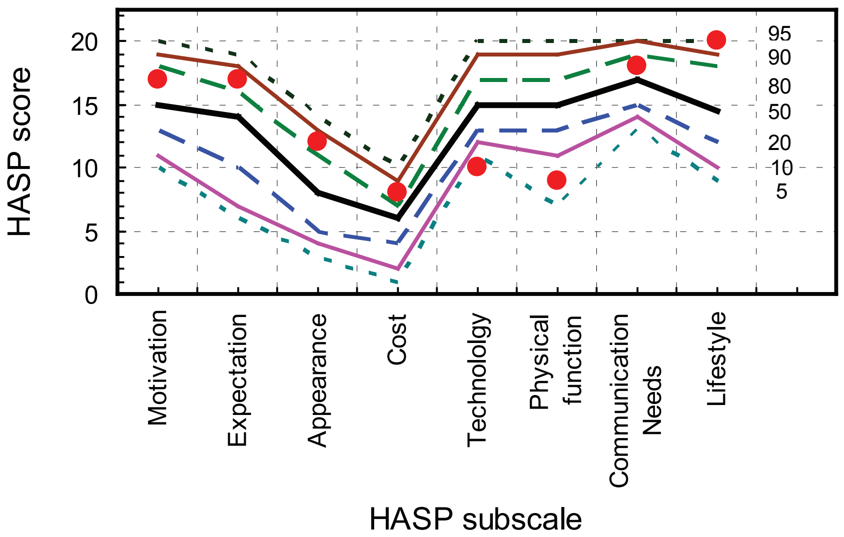

Figure 9.5 Scores for the 5th through 95th percentiles for the Hearing Aid Selection Profile (HASP). Scores for one hypothetical patient are shown as filled red circles.

to these two insistent people, he has plenty of later opportunities to sabotage the rehabilitation process, reassert control over his life, and declare that the hearing aids don't help.[183]

Mr X has reasons for his unwillingness and progress is unlikely unless the clinician can first find out those reasons. This knowledge can come only from Mr X. Obstacles can be of three types.

Unwillingness due to lack of awareness

Mr. X may simply not have noticed that he has more trouble than others in understanding speech, or that he needs the TV louder than anyone else. This is the easiest, but probably least common reason for the clinician to deal with. The following are some ways that the clinician can help Mr. X become more aware of his loss.

- Show Mr X his audiogram and explain it.
- Ask Mr X to talk about an occasion when he had trouble understanding conversation, and then reflect on whether anyone else present seemed to have the same trouble. (This is a good example of a patient-centered approach; the interaction revolves around the patient's own experience, not diagnostic and potentially meaningless information delivered to the patient by an expert clinician).
- Demonstrate Mr X's disability by having him repeat words presented in the sound field at a level where he cannot hear well, but where Mrs X, or any other person with normal hearing, can hear well.
- Immediately after presentation of each word or sentence, show on a card or computer screen the correct answer so that Mr X is immediately aware of how much he is mis-hearing.[1792]

Unwillingness to accept hearing loss

Mr X may have noticed that he has difficulty with conversation, but may not be willing to accept that his hearing has deteriorated. The course of action depends on Mr X's reasons for this.

Mr X may associate hearing loss with aging and/or senility. He may have seen hearing loss in someone significant to him whose health was deteriorating in some way, and may feel that hearing loss will be indicative of other sorts of deterioration in him too. Appropriate counseling would commence with giving Mr X basic information about how the hearing mechanism works, how it deteriorates, and how the state of the hearing mechanism is unconnected to mental functioning or other health issues. Good graphics or models of the hearing mechanism can help take Mr X's thoughts from diffuse concerns about his physical or mental state to more tangible rudimentary physiology.

Mr X may have noticed that he can hear well in some situations (e.g. moderate or loud speech in a quiet setting). He may have noticed that he can always hear speech, even if he cannot always understand it. He may also have heard a spouse say about him: *He can hear well when he needs to*. Based on any of these experiences he may conclude that his hearing is good.

- Appropriate counseling would commence with giving Mr X basic information about the partial loss of speech cues that accompanies hearing loss and listening in noisy places. Again, graphics, such as a transparent overlay of the speech range to superimpose on Mr X's audiogram, can be helpful. Figure 9.6 shows one such picture. The vertical overlay can be moved upwards to simulate softer speech. Some hearing aid manufacturers offer software tools that accomplish this, and also show

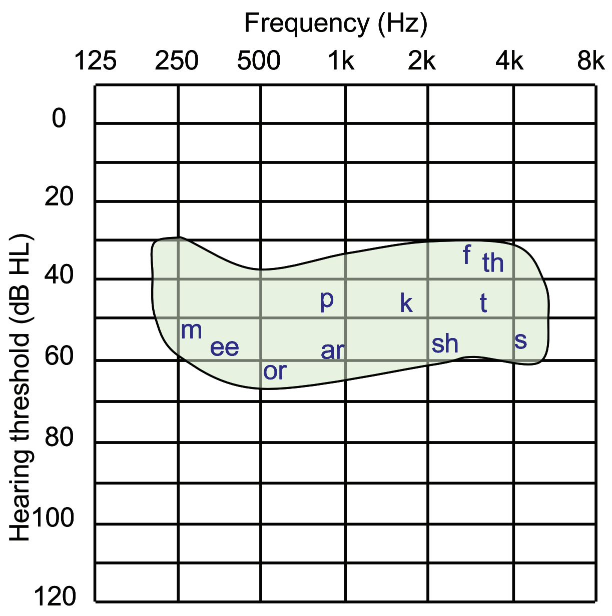

Figure 9.6 The speech spectrum, including a 30 dB dynamic range at each frequency, for speech at a long- term level of 65 dB SPL. The approximate locations of the spectral centre of a few speech sounds are indicated.

the increase in audibility that a hearing aid will provide. Live speech mapping (Section 4.5.7) can be used to show the same thing more dramatically. Mr X may develop confidence in the clinician if the clinician can nominate situations in which Mr X has had difficulty but has not ascribed the cause to his hearing mechanism. Such situations include listening from another room, listening to a softly spoken person in quiet, and listening in the presence of various background noises.

- The benefit of amplification can be demonstrated to Mr X by having him listen to a list of words played at a weak to moderate level, with and without a demonstration hearing aid. The demonstration hearing aid can be a BTE with a stock tube fitting, or an ITE/ITC mounted in a stethoclip, or a simulated hearing aid within the fitting software. There is no need to precisely prescribe the hearing aid's response, although its gain, power and response shape (including the effects of the stethoclip if used) should not be grossly inappropriate, or the demonstration will have an effect opposite to that intended!

Mr X may associate hearing loss with shame or guilt. In the mind of Mr X, accepting that he has a hearing loss may require him to accept a sense of shame for being defective, and/or a sense of guilt for being the cause of communication problems.[792] Involvement with other hearing-impaired people can help Mr X to establish his identity as a whole person with a hearing-impairment. Such involvement can also reinstate his sense of belonging and lead to the realization that difficult communication is the result of hearing loss rather than something for which he should blame himself. [792]

Unwillingness to try hearing aids and/or other rehabilitation activity

Mr X may acknowledge that he has a hearing loss, and may acknowledge that he has trouble hearing in some situations, but may still not want to do anything about it. Mr X may believe that the disadvantages of acquiring and wearing hearing aids outweigh the advantages. For Mr X to change his conclusion, he will first have to change his assessment of either the pluses or the minuses of wearing hearing aids. The clinician's emphasis should again be on understanding Mr X's views in an accepting manner. Only when they are understood can the clinician, or perhaps Mrs X, convincingly give information or suggest actions that may lead him to change his views.

Mr X may consider that hearing aids would not help much, perhaps based on what others have told him, or based on his observations of others:

- After acknowledging that hearing aids do not help some people much, the clinician could comment that often this is for correctable reasons, and that hearing aids have improved a lot in the last few years.
- Mr X could be asked to complete the unaided portion of the Abbreviated Profile of Hearing Aid Benefit (APHAB)[364] or the Hearing Handicap Inventory (HHIE).[1851] The poorer the score on either of these, the greater will be the benefit that he is, on average, most likely to experience (see Section 9.1.4), and this can be discussed with Mr X.
- Instead of a discussion about generalized benefit, Mr X could be asked about the difficulties he has in some specific situation that is important to him. He could be asked to imagine what it would be like to be in that situation and not have trouble understanding the conversation around him (provided that is a reasonable expectation in the circumstance).

Mr X may prefer his hearing impairment not to be visible to others, although he accepts that hearing aids may help. This is a difficult issue to deal with, as pressure by the clinician for the patient to reveal his loss to others may simply add one further stress to the life of the patient:

- If Mr X feels that revealing his loss would result in ridicule or embarrassment, he can be encouraged to reveal the loss to just one person and note the reaction (which is most unlikely to be negative).[f] He may then feel able to gradually expand the circle of people to whom he is willing to reveal his loss. If Mr X is willing, a group of hearing-impaired people organized by the clinician or through a self-help group provides a very supportive environment in which to make his loss known (see Section 13.14.4).

[f] The clinician should not overlook the possibility that in some cases, especially for working-age people, revelation of hearing loss may indeed lead to actual discrimination and disadvantage.[792]

- Mr X can be asked whether he thinks other people have already noticed that he misses or misunderstands things.
- Mr X can be asked if he really is happier with all the consequences of not wearing hearing aids than with the consequences of wearing them.
- The small size of hearing aids now available can be demonstrated.

Mr X may not think his problem is important enough to spend a significant amount of money to solve. He may have other reasons for not trying hearing aids that he is either unwilling or unable to articulate:

- Mrs X may be able to help Mr X understand all the consequences of his hearing loss. For example, it may be appropriate for Mrs X to say how she feels when Mr X fails to understand the conversation in group situations. Similarly, she could say how she feels about having to act as Mr X's interpreter, or spokesperson, in group situations.
- If there is no significant other person present, the clinician could relate the experiences of other families where one member has a hearing loss and ask if any of these experiences are relevant to Mr X's situation.
- In general, the clinician should make sure that Mr X has the best possible information relevant to his beliefs to help him come to an informed decision. To be balanced, this information is likely to include the very limited help that hearing aids give in very noisy situations.
- To help Mr X weight up the benefits versus the costs of trying hearing aids, he can be asked to write into each cell of ***The Box*** (see panel) those benefits and costs that apply to him. Just the act of completing these entries may give Mr X a changed perspective and if not, the entries provide a good basis for further discussion if any of the entries indicate beliefs that may not be well founded.

If, at the end of any pre-fitting counseling, the patient still considers that he or she either does not need hearing aids, or does not want hearing aids, the close association between attitude/motivation and subsequent benefit suggests that it would be unwise to in any way coerce the patient to try hearing aids. The negative experience that will probably result may make the patient less likely to return for help when the loss has deteriorated or when the patient otherwise decides that help is needed. A patient who turns his or her beliefs into actions by not wearing the hearing aids may also broadcast his unsatisfactory experience, and hence dissuade others from seeking help for their hearing.

Patients who at least accept that they have a loss may be willing to keep a log, for a specified period, recording any negative impacts that hearing has on quality of life. This may include situations where hearing loss made communication difficult and situations where hearing loss contributed to a patient withdrawing, physically or mentally, from some activity. Some patients may even be willing to attend a *Living with Hearing Loss* program, in which the impact of hearing loss on life is discussed.[976]

The Box: A counseling tool to weigh up benefits and costs of acquiring, or not acquiring, hearing aids.[794a]

Benefits of no action	Costs of no action
The potential benefits of taking action	The potential costs of taking action

For any of the three problems discussed in this section, a demonstration of instant-fit hearing aids may provide the patient with new first-hand information. The demonstration can be in the clinic for a few minutes, in the surrounding area for a few hours, or anywhere the patient likes to take them for a few days. The experience of hearing more clearly may be all that it takes for a patient to reassess his or her beliefs about hearing difficulties, hearing loss, or the balance of advantages versus disadvantages of hearing aids.

The appointment can close with the clinician acknowledging that the patient does not consider that hearing aids are currently necessary, but noting that people can change their assessment of this, either because hearing loss increases, or because their needs change. Open acknowledgment of the patient's current view that he or she does not believe there is a problem requiring action may be the best way to assist the patient to change that view over the coming months.[311] The patient can be encouraged to seek a reassessment in six or twelve months.[976]

Some readers may consider that there is an ethical issue in raising patients' awareness of problems. It seems a disservice to patients, however, *not* to probe for problems that are commonly associated with hearing loss, and which the clinician is able to assist if those problems are acknowledged by the patient.

9.2 The Upper Limit of Aidable Hearing Loss

Since the advent of cochlear implants, nobody is too deaf to benefit from a prosthetic device. For patients with severe or profound hearing loss, clinicians now have to consider various combinations, including:

- unilateral or bilateral cochlear implants;
- ***bimodal*** fitting, comprising a cochlear implant in one ear and a hearing aid in the other;
- ***hybrid*** fitting, comprising a cochlear implant and a hearing in one ear, combined with an implant, a hearing aid, another hybrid fitting, or nothing in the other ear;
- unilateral or bilateral hearing aids; or
- ***tactile*** hearing aid (in rare circumstances, as they are almost always outperformed by cochlear implants).

The question is not whether to recommend that the patient receive *any* device, but which *type* of device should be recommended. This section will briefly review the impact of poor speech identification scores on hearing aid candidacy and the relative performance of hearing aids, cochlear implants (including hybrid and bimodal devices), and tactile aids. A detailed description of cochlear implant and vibrotactile aid candidacy is beyond the scope of this book.

9.2.1 Poor speech identification ability

It is sometimes recommended that word recognition scores, obtained using headphones, of less than 50% indicate that hearing aid benefit will be limited to help with lip-reading, monitoring one's own voice, and detecting environmental sounds.[744] There are several reasons why speech identification scores obtained using headphones are not a good indicator of whether a person will benefit from hearing aids.[442]

A hearing aid does more than just amplify. It also re-shapes the speech spectrum relative to the flat frequency response that is available within an audiometer. Speech scores obtained with an individually prescribed hearing aid are usually greater than those obtained without frequency shaping, and often they are much greater.[304, 1405]

Even the maximum score possible with a flat frequency response may not be discovered during routine testing. Speech identification scores have an inherent random component. If the true score (i.e. one based on an extremely large number of items) was 50%, a score based on 50 items, for example, will be greater than 64% on 5% of occasions and will be less than 36% on 5% of occasions.[664, 1781] As the number of test items used decreases, the spread of scores from test to retest widens. Speech scores also depend on the level at which they are presented. The only way to be sure that the test is presented at the level giving the highest possible score is to test at several levels. Time constraints make it impossible to test at many levels with a large number of items per level. Reliability can be improved by testing at several levels, plotting the results, and drawing a smooth line through the resulting psychometric function. Despite this, considerable uncertainty is likely to remain over the maximum score.

Both of the above problems could potentially be rectified by spending a lot of time testing, and by doing the testing with an amplification system that has a gain-frequency response appropriate to the patient. A more fundamental problem is determining what cut-off score separates hearing aid candidates from non-candidates. It does not seem likely that any particular score could ever be shown to be valid. Some people with a profound loss wear hearing aids because they help with lip-reading, and because they give the aid wearer an awareness of sounds in their environment, which decreases stress, tension and insecurity.[531] These people may not be able to score anything on a speech test unless the material is particularly easy, such as a closed set test where the response choices differ in the number of syllables. In general, any cut-off value chosen would be highly dependent on the type of speech material.

A related problem is that the increase in intelligibility offered by hearing aids is highly dependent on the levels of speech and noise used. There is no logical reason why the intelligibility increase available in any situation should be predicted by the highest score obtained under headphones.

One should thus be very cautious in concluding, based on speech identification scores obtained with a flat frequency response under headphones, that hearing is too poor for a hearing aid to help in any situation. Not all authors or clinicians will agree with the preceding statement.

9.2.2 Hearing aids or cochlear implants?

Factors affecting CI performance

There are many factors that must be considered before a person can receive a cochlear implant (see panel on p.281). The underlying requirement is a reasonable expectation that the cochlear implant will provide speech identification ability superior to that which can be achieved with hearing aids. Because there is a wide range of speech identification performance achieved with cochlear implants, this expectation has to be viewed as a probability, rather than as a guaranteed outcome. Better implanted performance is likely for patients who:

- have had severe or profound hearing loss for the shortest time;[472, 638, 1814]
- are implanted as young as possible, and in the case of children, preferably before their first birthday.[294, 423] (Implant candidacy for children is further discussed in Chapter 16);
- had hearing at the time they were acquiring language;
- had the least hearing loss, and the highest speech intelligibility scores with hearing aids, before implantation; [472, 638, 1692]
- are motivated (or their families are motivated) to engage in rehabilitation activities.

None-the-less, considerable unexplained variability of outcomes remains.[138, 141, 142, 332, 585, 1807]

Predicting cochlear implant candidacy from hearing thresholds

There is also a wide range of speech identification performance across people with the same degree of hearing loss who wear hearing aids. If we attempt to predict from hearing thresholds alone whether a cochlear implant or hearing aids will provide the better outcome, there will be uncertainty about both the aided and the implanted performance, making the prediction very uncertain across a wide range of hearing thresholds.

We can none-the-less ask how much speech intelligibility is conferred, on average, by cochlear implants compared to that conferred by hearing aids for various degrees of hearing loss. Adults with cochlear implants typically have better speech identification ability than is typical of adults whose three- or four-frequency average hearing thresholds average 80 to 85 dB HL when wearing hearing aids.[617, 680, 1199] Of course, this means that the *average* cochlear implantee outperforms the *average* hearing aid wearer with this degree of loss, but as there is considerable overlap between the performance of the two groups, the superiority of the cochlear implant for this degree of loss by no means guarantees that *all* adults with an 80 to 85 dB hearing loss will improve speech intelligibility by obtaining an implant.

Predicting improvement from hearing aid speech scores

A much better prediction of implant benefit can be made if actual speech performance with hearing aids is already known. For adult patients with a long-standing hearing loss there will usually have been ample opportunity for the patient to have been fitted with hearing aids, for the hearing aids to be fine-tuned to get the best possible performance, and for speech identification ability to be measured. This rosy situation may not always apply, but even when it does not, a hearing aid trial can be performed prior to implantation. A commonly applied criterion for implantation is that for adults, open-set speech sentence scores in quiet with hearing aids should be less than 50% in the ear to be implanted.

Unfortunately, this simple criterion does not take into account the very important factor of how long the person has had profound hearing loss. Each line in Figure 9.7 shows the probability of an implant providing an open-set sentence speech intelligibility score greater than or equal to that obtained with hearing aids, as shown on the horizontal axis.[1813] As an example, someone who had a profound hearing loss for 10 years, and who now understands 33% of the words presented in an open-set sentence test while wearing hearing aids, has an 80% probability of an increased score with an implant.

The curves in Figure 9.7 may be unduly conservative (i.e. underestimating the likelihood of benefit) because:

- Implants have steadily improved in performance, whereas accumulated data like that in Figure 9.7 is based on the performance of people who received their implant up to 15 years before the analysis is performed.
- Many people now continue to wear a hearing aid in the opposite ear, and the two devices combined provide better performance than either device on its own (see Section 9.2.3).[281, 617]

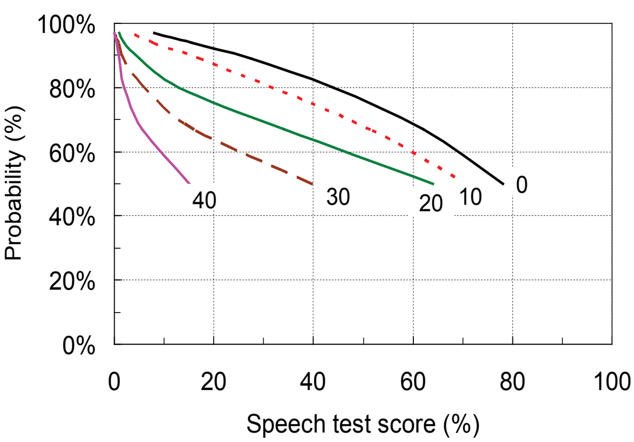

Figure 9.7 Probability of implanted adults exceeding the speech score for the BKB sentence test (from which the better known HINT test was derived) on the horizontal axis.[1813] Each line applies to a different duration of deafness prior to implantation, as indicated by the number of years shown next to each line.

- The statistical approach assumed that the implanted performance is independent of pre-implant hearing aid performance, whereas those who have the highest scores pre-implantation are likely to have the highest scores post-implantation.[332, 1813, 1975]

A less strict criterion for implantation is likely to be applied for a patient with a progressive hearing loss such that further deterioration in speech intelligibility is expected, particularly if progressive changes in the cochlea will make implantation more difficult with passing time.

A limitation of the preceding discussion is that it focuses solely on speech identification performance. It is not unusual for pre-lingually deafened adults to achieve open-set speech scores that are no higher than those they obtained prior to implantation. Despite this, many are successful implant users. They report that:[1974]

- they use the device regularly;
- they are satisfied with it;
- it helps them monitor their own voice;
- it facilitates their independence and employment; and
- it enables them to detect and recognize environmental sounds, which increases their feeling of security.[267, 1915]

9.2.3 Hearing aids and cochlear implants: bimodal and hybrid / electroacoustic stimulation

Cochlear implants are more effective at conveying mid- and high-frequency sounds than low-frequency sounds. They are also more effective at conveying information about spectral shape than conveying information about fine time pattern and pitch of sounds. Reasons for the failure to accurately represent pitch are not really understood, but may be related to the inability of individual electrodes to stimulate just low-frequency neurons. By contrast, hearing aids are usually more able to convey pitch and other low-frequency information than to convey higher frequency spectral information, perhaps because the best residual hearing is usually in the low frequencies. Consequently, hearing aids and implants provide information complementary to each other. This applies whether the hearing aid and implant are in the same ear or opposite ears.

Cochlear implant candidacy

The following considerations for cochlear implant candidacy reflect those in common use, and should be considered as an approximate guideline only. Criteria vary somewhat from country to country, from implant center to implant center and from implant company to implant company. Criteria are changing as implant performance increases,[392] as hybrid devices become more available, and as experience with implanting people with less hearing loss accumulates. As with hearing aid candidacy, many factors affect the likely outcome. The following must be considered in combination, not as separate criteria that individually enable or preclude implantation.

Adults and children

- No medical contraindications such as cochlear ossification, an absent cochlea, chronic middle-ear infection, or retrocochlear hearing loss. Etiology of the hearing loss will be a strong consideration in the decision to implant.

Adults

- A postlinguistic hearing loss. That is, a profound hearing loss occurred after the patient had acquired language aurally and was able to speak. Some exceptions are made for those who have been able to make adequate use of their residual hearing.
- A score of 50% or less on open-set sentence recognition 65 dB SPL (equivalent to approximately 60 dB SPL A-weighted, or 45 dB HL) in the ear to be implanted when optimally aided.
- A score of 60% or less on open-set sentence recognition when both ears are aided. This is approximately equivalent to a score of less than 30% for open-set monosyllabic words.[617]
- Motivated, emotionally stable patient, with realistic expectations, who is willing to attend for the required number of assessment, mapping, and training sessions.

Children

- Over the age of 6 months.
- For older children, the vocalization of varied sounds while communicating.
- Expectation of being in an aural and oral education program that includes appropriate rehabilitation.
- Insufficient access to speech cues via hearing aids. For example, aided thresholds above 2 kHz out of the 30 dB dynamic range of speech at 70 dB SPL overall level.
- Cooperative, motivated family (and patient, if old enough), with realistic expectations.

Bimodal stimulation

When the hearing aid and implant are in opposite ears, the combination is referred to as a ***bimodal*** fitting. In nearly all cases, the implant alone provides considerably better speech intelligibility than the hearing aid alone. Despite this, the combination of the two nearly always provides better speech intelligibility in noise than the implant alone.[278, 280, 281, 285, 291, 681, 1248, 1451] Just as with implants alone, the added benefit of the hearing aid increases for many months following implantation.[1087]

The major role of the hearing aid is in providing low-frequency information.[1205, 1473] Indeed it is possible that excessive mid- and high-frequency output from the hearing aid can decrease speech intelligibility by providing cues that conflict with the implant.[1205] This decrease is rarely the case if the hearing aid is prescribed for the ear to which it is fitted, and if the loudness provided by the hearing aid does not exceed the loudness provided by the implant.[280] Benefit is greatest when the wanted talker is positioned nearer the hearing aid ear and masking noise is positioned nearer the implanted ear, because of the higher SNR available to the hearing aid (see Section 15.3.1). The hearing aid then provides benefit not just because of the complementary frequency range over which it can provide information, but also because its microphone receives a clearer signal than does the implant's microphone.

In addition to increasing speech intelligibility, adding the hearing aid improves localization accuracy, on average.[278, 280, 281, 285, 291, 1451, 1808] Unfortunately, accuracy usually remains far from normal, and the degree of benefit varies markedly across recipients.[491, 1593] Given that each device type works best in different frequency regions, it is not surprising that localization ability remains poor, as the cochlear implant processing is likely to remove inter-aural timing cues that normal hearers rely on in the low frequencies, and the high degree of hearing loss in the aided ears is likely to render inaudible the inter-aural level differences that normal hearers rely on in the high frequencies. Certainly bimodal wearers are not able to use inter-aural time differences to improve speech intelligibility.[290]

The greater the speech intelligibility score when the patient is tested with just the hearing aid, the greater the benefit the hearing aid provides when it is used together with the cochlear implant.[1248, 1451, 1808] Because of this, we might expect that the degree of loss in the aided ear would affect the amount of benefit provided by the hearing aid. However, across the range of hearing thresholds in the aided ear so far evaluated, the benefit provided by the hearing aid appears to be independent of hearing loss.[281] Even when the hearing aid by itself provides virtually no speech intelligibility, the complementary low-frequency cues it provides can improve speech intelligibility in noise and melody recognition over that provided by the implant alone.[964] The benefit of adding the hearing aid can even be seen objectively in the shorter latency of evoked cortical responses for the two devices relative to the implant alone.[799, 1548]

Hybrid or electroacoustic stimulation

When a hearing aid and an implant are in the same ear, the combination is referred to as a ***hybrid*** that provides the wearer with ***electroacoustic stimulation***. In a hybrid device, signal components in the incoming sound above a certain cut-off frequency are presented via the implant and signal components below the cut-off frequencies are presented via the hearing aid.[g] Hybrid devices are most suitable for people with steeply sloping audiograms: mild or moderate loss in the low frequencies enables the ear to make good use of the low-frequency sounds delivered by the hearing aid, including the perception of pitch, whereas the severe or profound loss in the high frequencies is best served by the implant. One method for choosing the cut-off frequency is to select the frequency at which the audiogram equals about 70 dB HL. Because the implant does not have to convey low-frequency signal components, there is no need to deeply insert the implant electrode to reach the low-frequency part of the cochlea. Sometimes a short electrode array with only a few electrodes is used, in other cases a regular electrode array is used but only the electrodes close to the basal (high-frequency) end of the cochlea are stimulated. Speech intelligibility may be greatest when the hybrid device is designed so that each part of the cochlea receives the signals from the frequency region that it would in a normal-hearing ear.[86]

Hybrid devices have great promise for increasing the range of hearing losses for which implantation produces a better outcome than with hearing aids alone. The major limitation is that the act of drilling the hole in the cochlea and/or implanting the electrode can cause damage, which makes the low frequency hearing loss greater. The increase in hearing thresholds is typically only 5 to 15 dB.[583, 584, 897] Unfortunately, a much larger increase in hearing thresholds can sometimes occur, and in some cases can occur suddenly one or two years after the surgery. Given the possibility of low-frequency threshold degradation during or after implantation, the choice of electrode length represents a dilemma: a short electrode may minimize the chance of damaging the low-frequency part of the cochlea, but if the low-frequency thresholds are sufficiently degraded by implantation, then a long electrode has greater capacity to convey low-frequency information electrically.

One of the major advantages of electroacoustic stimulation is that the hearing aid component provides the

[g] Prescription procedures for hybrids have not yet been sufficiently researched. The optimal method for choosing the cut-off frequency is not yet known. Furthermore, in some cases, frequencies in a band around the cut-off frequencies are presented by both the implant and the hearing aid, in other cases by neither of them, though neither of these variations seems sensible.

Hybrid, or electroacoustic, stimulation in brief:

- A cochlear implant, with the electrode array inserted a short way into the cochlea, sends the high-frequency parts of speech via electrical signals to the high-frequency part of the cochlea.
- A hearing aid sends the low-frequency parts of speech via acoustic vibrations to the low-frequency part of the cochlea.

wearer with pitch perception better than any implant can so far provide. For this reason, or perhaps for other reasons, the hybrid also usually provides better speech intelligibility in noise than either device by itself could.[193, 583, 584, 898] The improved pitch perception offered by a hybrid relative to an implant alone certainly enables better perception of music.[615]

As with implants alone and bimodal fittings, speech intelligibility performance continues to improve for at least two years following implantation.[583] A hybrid device in one ear can be combined with a hearing aid, an implant, or another hybrid in the other ear.

9.2.4 Hearing aids or tactile aids?

Some patients are so deaf that they receive only vibratory information from hearing aids. They are able to receive time-intensity information, but little or no spectral information. For such people, it is likely that more speech information will be correctly perceived if a purpose-designed ***vibrotactile*** or ***electrotactile*** aid is used.[1431] These aids can encode more speech information into the sense of touch than is accidentally encoded by hearing aids designed to provide an acoustic stimulus. All commercially available tactile aids use vibration as the stimulus. An alternative used in several research studies is electrotactile stimulation, in which small electrical discharges stimulate nerve bundles under the skin. If the characteristics of the electrical discharge are appropriately adjusted, the stimulation will be felt as a tactile sensation.[333]

Although tactile aids can unquestionably provide information that supplements lip-reading, on average they provide far less information than a multichannel cochlear implant.[29, 257, 600, 601, 1198, 1514, 1641] This does not mean that every person with an implant will have better speech identification ability than every person with a tactile device, but the vast majority will.[1367]

Vibrotactile aids primarily provide supra-segmental cues, such as intonation and stress.[120] Even simple, single-stimulator vibrotactile aids can indicate whether consonants are voiced, and whether they are stops or continuants.[1429] These cues are not available by lip-reading alone.[1432] Multi-stimulator tactile aids can also provide more detailed information, such as the format frequencies of vowels.[143, 333] There is some evidence that perception of supra-segmental information may then be more difficult.[258, 1425] Such conclusions may, however, be very dependent on the design of the individual device and the extent of training.

Integrating the tactile information with the visual cues obtained from lip-reading is not easy. It is not unusual to initially find no improvement when tactile information is added to visual information[144] or even to find a decrease in score relative to visual perception alone.[120, 1098] Training has substantial effects on performance, however, and the new skills are maintained after the training is complete.[28, 1906] Training is also likely to benefit people with a severe or profound hearing loss who receive a cochlear implant or hearing aids, but the need for training is probably even greater for those who receive a tactile aid.

Although this brief discussion has focused on the benefits of tactile aids for speech recognition, tactile aids can also help people monitor their speech production, and this helps with speech production training.[581, 1368, 1428, 1907] Tactile aids also enable people to identify environmental sounds, particularly those with distinctive temporal patterns.[1486]

In summary, the clinician can be confident that a tactile device will provide information that supplements lip-reading and helps speech production. It is unwise to provide a tactile device without also ensuring that the patient will receive appropriate training for some weeks or months following provision of the device. If training can be provided, tactile devices should be recommended to any patient who receives little or no benefit from hearing aids and who is unwilling to be implanted with a cochlear implant or for whom a cochlear implant is unsuitable. It seems likely that multi-stimulator tactile devices will provide more information than single-stimulator devices, but may also require more training. More detailed information about tactile devices can be found in Plant & Spens (1995). Training procedures can be found in Plant (1994, 1996).

9.3 Medically Related Contra-indications to Hearing Aid Fitting

Any factor that would cause a clinician to refer the patient for medical assessment will temporarily, or in some cases permanently, halt the process of hearing aid fitting. These factors include:

- a hearing loss of sudden onset;
- a rapidly progressing hearing loss;
- pain in either ear;
- tinnitus of sudden recent onset, or unilateral tinnitus;
- unilateral or markedly asymmetrical hearing loss of unknown origin;
- vertigo (i.e. dizziness);
- headaches;
- conductive hearing loss of any origin;
- otitis externa or otitis media (i.e. infection in the external ear or middle ear and/or drainage);
- cerumen filling more than 25% of the cross-section of the ear canal (unless the clinician has been trained in cerumen removal), or a foreign body in the ear canal; or
- atresia (i.e. missing external ear) or deformity of the external ear.

Whether hearing aids are fitted to the patient after medical intervention will, of course, depend on the medical diagnosis, treatment, and outcome; the physician's recommendation if appropriate, and the patient's wishes.

9.4 Concluding Comments

This chapter systematically sets out the factors that affect candidacy for hearing aids. In the end, however, the decision about whether to recommend amplification requires a qualitative judgment by the clinician. The decision about whether to accept amplification requires a qualitative judgment by the patient. The clinician's task is to ensure that the patient is well informed about every factor relevant to the individual patient.

Hétu (1996) points out that reluctance to accept rehabilitation or even the existence of a hearing loss is not simply an irrational denial of something that is evident to everyone else. To many hearing-impaired people, accepting a hearing loss is equivalent to accepting a spoiled self-identity, and such acceptance may engender feelings of shame for being defective or guilt for being the cause of communication problems with their loved ones and peers. In many instances, it is less painful for a person to endure communication difficulties and social isolation than it is to view oneself as a hearing-impaired person. Helping the patient change these feelings is a more difficult task for the clinician than prescribing and adjusting hearing aids, but until the patient actually wants some form of hearing rehabilitation, there is little point in pursuing a technological solution. There is a great need for research into the effectiveness of different counselling methods for helping patients see the same need for improved communication, and feel the same willingness to act on it, that their communication partners can see and feel.

Candidacy also depends on the objective benefits that technology can provide. Hearing aids currently can provide some benefit in noisy places, but still leave the wearer with poorer than normal hearing ability in noise. Hearing aids are therefore a visible badge of disability: the wearer will likely not cope as well as a normal-hearing person in difficult listening situations. This limited effectiveness of hearing aids in noise may well lie behind the negative feelings that cause many hearing-impaired people to not even try hearing aids. Negative word-of-mouth reports about hearing aids is the major reason why 19% of non-adopters of hearing aids have never tried them.[955] When future hearing aids, incorporating binaural super-directional microphones (Section 7.1.4), enable better hearing in noise than normal hearers can achieve, it is possible that negative stereotypes about hearing aids will be replaced with positive ones. Many people who currently are not candidates for hearing aids may suddenly become candidates.

Technology improvement is not the only issue, of course. Candidacy for hearing aids also depends on their effectiveness relative to that of alternative coping mechanisms, whether these be socially enabling use of hearing tactics (see Chapter 12) or socially defeating mechanisms such as withdrawal and avoidance.

Much of this chapter is about issues that, together, determine a patient's needs, motivation and attitude. ANLs, by contrast, appear to operate in a different, and therefore complementary, domain. ANL takes only a few minutes to measure and appears to have the potential to assist with determining who is a candidate for hearing aids, and who might be a candidate if sufficiently effective noise reduction methods (wireless systems, or future highly directional microphones) can be used. The measurement of ANL and attitude/motivation has the potential to predict candidacy with useful accuracy, given the apparent importance of each. An evaluation of their combined effectiveness in predicting candidacy is greatly needed.

One aspect of hearing aid candidacy remains extremely uncertain. It is clear that bilaterally-impaired people with a cochlear implant in one ear should have some device in their other ear. It is so far unclear, and with few exceptions[1073, 1204] largely untested, whether that second device should be a hearing aid, a second implant, or a hybrid device. More precisely, we don't yet know the hearing loss characteristics for which each of these would be the preferred option. Certainly the benefit of a second implant is much less than that of the first.[1743]

CHAPTER 10

PRESCRIBING HEARING AID AMPLIFICATION

Synopsis

Amplification can be prescribed using a formula that links some characteristics of a person to the target amplification characteristics. Prescription formulae most commonly used are based on hearing thresholds, but some are based on supra-threshold loudness judgments.

Well-known procedures for linear hearing aids include POGO, NAL, and DSL. For all of these, gain can be prescribed based on hearing thresholds alone. These formulae all contain variations of the half-gain rule, but the variations are so different that the resulting prescriptions differ greatly, especially for people with a sloping hearing loss.

For non-linear hearing aids, all available prescription procedures include some aspect of normalizing the loudness of supra-threshold sounds. Several procedures (LGOB, IHAFF, DSLm[i/o] curvilinear, CAM-REST and FIG6) aim to normalize loudness at all frequencies, at least for sounds with levels above the compression threshold of the hearing aid. Other procedures vary from loudness normalization in some way. ScalAdapt decreases the loudness of low-frequency sounds; CAM2 and NAL-NL2 normalize only the overall loudness. CAM2 aims to equalize the contributions that different frequency regions make to loudness, whereas NAL-NL2 aims for the sensation levels across frequency that will maximize calculated speech intelligibility. As each of the formulae has been revised, their prescriptions have become more similar to each other, but marked differences still occur.

There are some issues related to prescription that are not yet resolved, although there is considerable information available about each issue. How much do patients' preferences and performance with hearing aids change following weeks, months, or years of experience with amplified sound? How loud (a perception, not a physical quantity) do patients like sound to be? Should tests of dead regions in the cochlea routinely be conducted? How severe does hearing loss need to be before it is considered unaidable? As signal level decreases, down to how low a level should gain keep increasing? How accurately must prescription targets be met? What is the best combination of fast and slow compression?

Maximum output (OSPL90) has to be prescribed so that loudness discomfort is prevented, but so that enough loudness can be obtained without the hearing aid becoming excessively saturated. In many procedures, the target OSPL90 is assumed to just equal LDL, in others it is predicted from threshold, in which case it may fall above or below an individual patient's LDL as measured in the clinic. For patients with mild to severe hearing loss, an acceptable sound quality is more likely if compression limiting controls maximum output than if peak clipping controls maximum output. Many patients with a profound loss, however, will benefit from the additional SPL that is achievable with a peak clipper.

People with conductive and mixed hearing loss require greater gain and OSPL90 than people with sensorineural loss of the same degree. For a variety of reasons, the gain needed to compensate for a conductive loss seems to be less than the amount of attenuation that the conductive loss causes in the middle ear. Consequently, the same is true of OSPL90.

Multi-memory hearing aids can have a different prescription for each memory. These alternatives can be prescribed as variations from the baseline response prescribed for the first memory. The variations are designed to optimize specific listening criteria or for listening to different types of signals, such as music. People who wear their hearing aids in many environments, have more than 55 dB high-frequency hearing loss, and require more than 0 dB low-frequency gain, are most likely to benefit from multiple memories.

Neither gain nor OSPL90 should be any higher than is necessary for a patient. Otherwise, a hearing aid may increase hearing loss because of the resulting high-level exposure to sound. The risk of temporary or permanent noise-induced loss is greatest for patients with a profound loss, and can be minimized by using non-linear amplification.

10.1 General Concepts Behind a Prescriptive Approach and a Brief History

Hearing losses vary widely in their degree, configuration, and type. Consequently, a hearing aid has to be selected, and its amplification characteristics have to be adjusted, to be appropriate for each hearing-impaired person. The only practical way to do this is by using a ***prescription procedure***. A prescriptive approach to hearing aid fitting is one in which some characteristics of the patient are measured, and the required amplification characteristics are calculated from them. Of course, this requires there to be some known (or assumed) relationship between the measured patient characteristics and the required amplification characteristics.[212] These required amplification characteristics are often referred to as the ***amplification target*** or ***prescription target***. The measured characteristics nearly always include hearing thresholds, and often these are the only characteristics measured.

A prescriptive approach may be contrasted with a (hypothetical) purely ***evaluative approach***. In such an approach, a number of hearing aids or response shapes would be chosen randomly, and then each tested on the hearing-impaired person to find the best one. Such an approach is totally impractical in its purest form because of the huge number of potential amplification characteristics that could be evaluated. Even in the 1950s and 1960s, when the systematic Carhart evaluation method (or a portion of it) was used to compare the performance of hearing aids on several criteria, the hearing aids to be evaluated were selected using a vaguely defined prescriptive approach.[248] Low gain, low power hearing aids, for example, would never be evaluated on someone with a profound hearing loss. In fact, all hearing aid selection and fitting invariably uses some combination of prescription and evaluation of the end result. A clinician may carefully prescribe, select, and adjust a hearing aid to meet some target, but it would be rare not to ask *how does that sound*? This question comprises a *very* rudimentary evaluation. If the answer is *terrible,* the clinician is bound to investigate further, and potentially alter the amplification characteristics away from the carefully matched prescription. More sophisticated methods for evaluation and fine-tuning are considered in Chapter 12.

Prescriptive selection procedures have a long history. As early as 1935, Knudsen and Jones proposed that the gain needed at each frequency was equal to the threshold loss at the same frequency minus a constant. This is often referred to as ***mirroring of the audiogram***, because the shape of the gain-frequency curve equals the inverse of the shape of the hearing loss. With mirroring procedures, every 1 dB increase in hearing loss requires 1 dB of additional gain to compensate. In sensorineural hearing loss, the gain needed to restore normal loudness perception is equal to the threshold loss only when the person is listening at threshold. For all higher levels, this amount of gain would be excessive, as can be seen in Figure 6.10. Mirroring thus leads to excessive gain, especially for those frequencies with the greatest hearing loss.

The next development was to base the gain needed on the person's ***Most Comfortable Level*** (***MCL***) rather than on their thresholds. Watson and Knudsen (1940) suggested that speech should be amplified sufficiently to make speech energy audible and comfortable. Their specific formula involved MCL but, surprisingly, did *not* take into account the variation of speech energy across frequency. Shortly after, Lybarger (1944) made a very important observation: averaged across frequency, the amount of gain chosen by people was approximately half the amount of threshold loss. This is known as the ***half-gain rule*** which, as we shall see, underlies several later prescriptive procedures.

These two ideas (raising speech to MCL and the half-gain rule) are really two sides of the same coin. For mild and moderate sensorineural loss, the threshold of discomfort is little different from normal, as shown in Figure 10.1. MCL is approximately half way between threshold and discomfort, so MCL increases by 0.5 dB for every 1 dB increase in hearing loss. This explains *why* gain needs to be approximately half the hearing loss. Of course, if the aim is to raise the level of *speech* to MCL, then we cannot predict how much gain is needed at each frequency unless we take into account the speech intensity at each frequency. Because the low-frequency components are more intense than the high-frequency components, the half-gain rule has to be modified. Either a little less low-frequency gain has to be given, or a little more high-frequency gain, or both. We will return to this with some specific examples in Section 10.2.1.

The half-gain rule has to be further modified for severe and profound losses. For hearing thresholds greater than 60 dB HL, discomfort thresholds are significantly above normal while MCL remains approximately midway between threshold and discomfort.

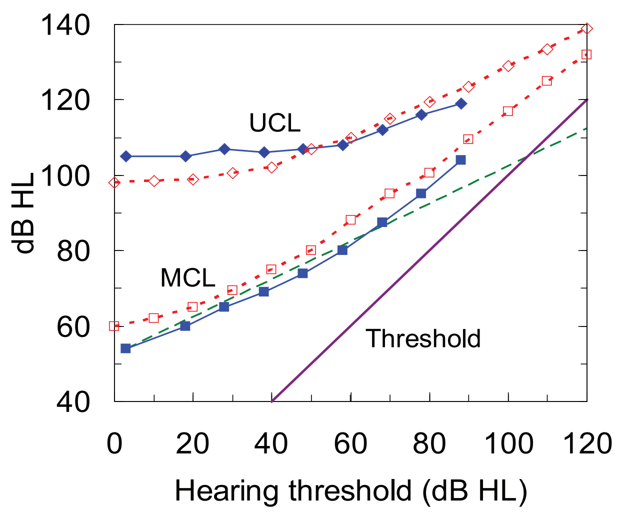

Figure 10.1 Uncomfortable listening level and most comfortable level for people with sensorineural hearing loss, averaged across 0.5, 1, 2, and 4 kHz. Data shown with filled blue symbols are from Schwartz et al. (1988) and those with open red symbols are from Pascoe (1988). The dashed green line has a slope of 0.5, illustrating the relationship between MCL and the half-gain rule.

This relationship means that MCL is elevated by more than half the hearing threshold loss. The gain, consequently, must be more than half of the hearing loss.

It can be seen that even 70 years ago it was recognized that there were two different auditory attributes that could provide a useful basis of prescription. One approach was to measure some supra-threshold loudness percept (such as MCL). The second was to measure hearing threshold. The link between these is made clear in some procedures: threshold and discomfort are measured, but are used to estimate MCL by assuming that MCL *bisects* the person's dynamic range.[1890]

The preceding discussion talked about "gain" as though it was the same at all input levels. Until the early 1990s, hearing aids did indeed provide the same gain over a wide range of input levels. As we shall see in Section 10.4, the dichotomy between basing gain prescription on threshold or on loudness perception survives through to the most recent procedures for non-linear hearing aids, in which gain decreases as input level rises. Procedures based on hearing threshold have been most popular, probably because threshold is easier and faster to measure and can be measured for infants and for people who have low cognitive ability, and the outcomes for threshold-based procedures have been at least as good as those for loudness-based procedures.

Gain prescription procedures based at least in part on loudness (MCL, discomfort, or entire loudness scales) include:

- Shapiro;[1609]
- CID (Central Institute for the Deaf);[1392, 1643]
- LGOB (Loudness Growth in half Octave Bands);[31, 1439]
- IHAFF/Contour (Independent Hearing Aid Fitting Forum);[361, 1829]
- ScalAdapt;[906] and
- DSL[i/o] (Desired Sensation Level Input-Output, curvilinear compression version).[327]

Gain prescription procedures based on threshold alone include:

- NAL (National Acoustic Laboratories),[236] NAL-R (Revised)[224] and NAL-RP (Revised-Profound);[234]
- Berger;[116]
- POGO and POGO II (Prescription of Gain and Output);[1163, 1579]
- FIG6;[919]
- CAMREST,[1214] CAMEQ,[1221, 1232] and CAM2;[1234]
- NAL-NL1 (NAL non-linear)[226, 435] and NAL-NL2.[446]

Some procedures have given the user the option of basing the gain-frequency prescription entirely on threshold, or on a combination of threshold and uncomfortable listening levels:

- MSU (Memphis State University);[360]
- DSL[i/o],[327] and DSLm[i/o].[1583]

Prescription of gain has received far more attention than the prescription of maximum output (OSPL90), despite the probable high importance of OSPL90 for linear hearing aids. For non-linear (compression) hearing aids, now almost universally used, the level at which hearing aids limit the maximum output is less important than for linear aids because some of the gain reduction that occurs when a linear hearing aid limits is instead provided by the more gradual form of compression that commences at lower input levels. In addition, OSPL90 can be considered to be just one

Understanding the nature of prescription procedures

The prescription procedures listed in this section vary in many ways other than whether they are based on threshold or loudness data. When confronted with a new procedure, there are four essential questions to ask:

1. On what type of patient data is the procedure based? Most commonly this will either be thresholds or the levels needed to achieve certain ratings of loudness;
2. What type of amplification characteristic is being prescribed? For a linear aid, this will most commonly be gain and/or the maximum output (OSPL90). For a non-linear aid, it will most commonly be the gain prescribed for several input levels, or some other characteristics derived from these gains, such as compression ratios at each frequency.
3. What is the aim of the selection procedure, and what relationships have been assumed in the link between the patient data and the amplification characteristics?
4. What evidence is there that the formula prescribes amplification that patients prefer and/or function well with, or even are willing to accept?

of the many output curves (each for a different input level) that can be specified for a non-linear hearing aid.

Prescribing the best response for a given patient *seems* like a simple problem, but prescription formulae have changed often during the several decades that this topic has been researched. Finding a simple relationship between hearing loss and gain has not been easy because:

- The optimum gain-frequency curve probably depends on the type of input signal, and its level and spectral shape, whereas much of the research has been carried out with linear hearing aids, which have not been able to provide this variation.
- The optimum gain-frequency curve *may* depend on things such as supra-threshold loudness perception and frequency resolution ability in a way that cannot be predicted from threshold (although no such relationships have as yet been established) and may depend on other unknown factors.
- The optimum gain-frequency curve for a person *may* depend on the nature of the auditory input to which the person has become accustomed during the preceding months or years.[474, 1581]
- For a particular person listening to speech at a particular time and input level, there may not even *be* a single optimum gain-frequency curve. Rather, the optimum may depend on whether the person wishes to maximize intelligibility, or comfort, or some other perceptual attribute of sound,[865] and on the type of material being listened to.[1013]

More thorough reviews of the early development of prescriptive procedures and evaluative procedures can be found in Byrne (1983) and in Hawkins (1984). Some of the more recent and popular procedures for gain and OSPL90 are described in more detail later in this chapter.

Prescription rules invariably involve some sort of formula. Once a prescription method has been chosen, the prescribed gains must be calculated. In the past, this was done with tables, slide rules, or a calculator. Universally now, the formulae are included within the software produced by each hearing aid manufacturer for the purpose of adjusting their hearing aids. Most of the major manufacturers enable the clinician to choose between one or more of the published procedures, or each manufacturer's own proprietary formula. There are, however, also standalone computer programs for the most recent and popular prescription procedures (NAL-NL1, NAL-NL2, DSL[i/o] and CAMEQ). Real-ear gain analyzers also include these better-known formulae, so the measured real-ear gain easily can be compared to the target gain-frequency curve it is meant to approximate.

The emphasis in this chapter will be on real-ear gain: both real-ear aided gain (REAG; Section 4.3) and real-ear insertion gain (REIG; Section 4.4). An REAG prescription specifies how much the SPL at the eardrum should exceed the SPL in the incoming field. Insertion gain, by contrast, describes how much more signal should be at the eardrum when the person is aided than when the person is unaided. Of course, either type of gain can be converted to the other by adding or subtracting the real-ear unaided gain (REUG) curve (Section 4.4). No matter how real-ear gain is calculated, it can be converted to a 2-cc coupler gain or ear-simulator gain, or to real-ear aided response (REAR) using the principles described in Sections 4.3.2 and 4.4.2 and the specific procedures described in Section 11.4.

10.2 Gain and Frequency Response Prescription for Linear Amplification

Linear hearing aids have the same gain-frequency curve for all input levels, until the output level is high enough to cause the aid to limit. The following three sections will present the concepts and calculation details, as applied to sensorineural hearing losses, for three procedures that were most commonly used. It is now rare to adjust hearing aids to a linear response, so these formulae are now rarely used, but the principles behind their development are just as relevant to nonlinear prescription formulae. In each of the formulae presented in the panels in this chapter, IG_i will represent the insertion gain at the *i'th* frequency, k_i represents an additive fitting constant at that frequency, and H_i represents the hearing threshold (in dB HL) at that frequency.

10.2.1 POGO

The original POGO (Prescription Of Gain and Output) procedure,[1163] was a straightforward application of the half-gain rule, with an additional low cut. The low cut was intended to decrease upward spread of masking from low-frequency ambient noise. The low cut could, of course, also be justified by the greater intensity of speech at low frequencies and by the lesser importance of speech information in the very low-frequency region. The amount of low-frequency cut specified was based on the originators' experience. Insertion gain at each frequency is equal to half the hearing loss at that frequency, plus a constant, as shown in the accompanying panel. The procedure was intended to be used only for hearing losses up to 80 dB HL.

POGO formulae

POGO formula

$IG_i = 0.5*H_i + k_i$

Freq	250	500	1k	2k	4k
k_i (dB)	-10	-5	0	0	0

POGO II formula

$IG_i = 0.5*H_i + k_i$, for $H_i \leq 65$

$IG_i = 0.5*H_i + k_i + 0.5*(H_i\text{-}65)$, for $H_i > 65$

In 1988, the procedure was extended to provide additional gain for people with severe and profound hearing losses.[1579] The revised procedure, known as POGO II, prescribes the same gain as POGO for losses less than 65 dB HL. For greater losses, however, gain increases by 1 dB for every 1 dB increase in hearing loss. The amount of additional gain prescribed in POGO II was based on an experimental observation that people with severe and profound hearing losses prefer to listen to speech at a low sensation level.[a] For sensation level to be held at a small but constant level as hearing threshold increases, the gain has to increase by the same amount that hearing loss increases.

10.2.2 NAL

The NAL (National Acoustic Laboratories of Australia) prescription formula has also been revised since it was first published in 1976.[236] From the outset, the aim of the NAL procedure has been to maximize speech intelligibility at the listening level preferred by the aid wearer. Intelligibility was assumed to be maximized when all bands of speech are perceived to have the same loudness. Does it matter if one frequency region is much louder than the rest? Yes! If speech is too loud the patient will turn down the volume control. Decreasing the volume control setting will also decrease the loudness of all other frequency regions,

[a] On average, the long-term rms 1/3-octave speech levels were only 7 dB above threshold.[1878]

NAL formulae

NAL-R formula

$$H_{3FA} = (H_{500}+H_{1k}+H_{2k})/3$$

$$X = 0.15\ H_{3FA}$$

$$IG_i = X + 0.31\ H_i + k_i$$

Freq (Hz)	250	500	1k	2k	3k	4k	6k
k_i (dB)	-17	-8	1	-1	-2	-2	-2

NAL-RP formula

$$X = 0.15\ H_{3FA} \qquad \text{for } H_{3FA} \le 60$$

$$X = 0.15\ H_{3FA} + 0.2\ (H_{3FA}-60) \qquad \text{for } H_{3FA} > 60$$

$$IG_i = X + 0.31\ H_i + k_i + PC$$

Values of the profound correction, PC (in dB), to use in the above formula, as a function of frequency and hearing threshold at 2 kHz.

	Frequency (Hz)						
$H_{2\ kHz}$	250	500	1k	2k	3k	4k	6k
≤ 90	0	0	0	0	0	0	0
95	4	3	0	-2	-2	-2	-2
100	6	4	0	-3	-3	-3	-3
105	8	5	0	-5	-5	-5	-5
110	11	7	0	-6	-6	-6	-6
115	13	8	0	-8	-8	-8	-8
120	15	9	0	-9	-9	-9	-9

which may then be at too low a level to contribute optimally to intelligibility. This logic can best be understood in the context of the Speech Intelligibility Index (SII) method of predicting speech intelligibility (see Figure 9.3 for an illustration of the portion of the speech spectrum audible above hearing threshold and background noise).[b]

The 1976 formula was derived as follows. Empirical observations indicated that preferred insertion gain at 1 kHz equaled 0.46 times the 1 kHz threshold (a minor variation of the half-gain rule).[229] It was assumed that at all frequencies an extra dB of loss required an extra 0.46 dB of gain. To deduce how much gain was needed at the other frequencies relative to 1 kHz, two additional sources of data were used. Gain at each frequency was adjusted by an amount that mirrors the shape of the long-term average speech spectrum (LTASS), so that less gain was given to those frequencies where the speech is most intense (the low frequencies). Finally, gain was adjusted so that for someone with normal hearing, speech was raised to MCL, which for normal hearers was estimated to be the 60-phon equal loudness contour. Although the

[b] If one used the Speech Intelligibility Index to derive the optimum frequency response, then frequency regions that contribute most to intelligibility would be made a little louder than those that contribute less. This is precisely the basis of the NAL-NL1 and NAL-NL2 prescription methods (see Section 10.4.6).

1976 procedure is no longer used, it is worth understanding the basis of the formula, as the concepts are still relevant. The *shape* of the gain-frequency curve is equal to the shape of the normal equal loudness curve, minus the shape of the speech spectrum, plus 0.46 times the shape of the hearing threshold curve. The *gain* at 1 kHz is equal to 0.46 times the hearing loss at 1 kHz. The NAL 1976 formula is extremely similar to the original POGO formula and to the Cambridge formula for linear hearing aids.[1228]

The type of gain prescribed by the NAL formula is insertion gain (or equivalently, in those ancient days, functional gain). The original publication also expressed the formula in terms of the coupler gain likely to be needed to achieve the target insertion gain. These coupler gain targets included a reserve gain of 15 dB so that the hearing aids could be measured in the coupler at their maximum volume control setting, but be worn at a mid volume control setting.

During the early 1980s, Byrne extensively evaluated the original NAL formula.[214, 215] This evaluation showed that the aim of the NAL procedure (equal loudness at all frequencies) was correct. Unfortunately, the formula did not achieve equal loudness, especially for people with steeply sloping losses. The evaluation data (and other published data) were used to relate the gain-frequency curve needed for equal loudness to the shape of the audiogram. This showed that the *shape* of the gain-frequency curve, measured in dB/octave, varied at only 0.31 times the *shape* of the audiogram. The revised formula, which became known as NAL-R, reflected this but retained the well-established half-gain rule (actually 0.46) for the three-frequency *average* gain.

A further series of experiments investigated the preferred gain and frequency response of adults and children with severe and profound hearing loss.[233, 234] Compared to the NAL-R prescriptions, these subjects required additional gain, and less high-frequency emphasis. For three-frequency average hearing thresholds above 60 dB HL, the required gain increased at 66% of the increase in hearing loss, rather than the 46% rate that applied to people with less loss.

The additional low-frequency emphasis (or equivalently, the decrease in high-frequency emphasis) needed to maximize speech intelligibility could best be estimated based on the hearing threshold at 2 kHz. The response slope required progressively *less* high-frequency emphasis as the threshold at 2 kHz increased beyond 90 dB HL. (Reasons for this are considered in Section 10.3.4.) The formula that implemented these changes became known as the NAL-RP (revised, profound) formula.

The NAL-RP procedure is based on measured speech intelligibility, and subjective preferences for quality and intelligibility, in quiet and in noise, for subjects with losses from mild to profound and, as we will see, is well supported by empirical evidence.

10.2.3 DSL

The Desired Sensation Level (DSL) formula aims to provide the aid user with an audible and comfortable signal in each frequency region.[1600, 1601] It differs from the NAL-RP and POGO procedures in at least three ways.

First, the target it prescribes is a real-ear aided gain rather than a real-ear insertion gain. Second, the DSL procedure has been particularly well integrated with measurement methods that are convenient for use with infants and young children, without the use of average correction factors. The procedure consistently refers all measured quantities to SPL in the ear canal so that the aided speech levels, hearing thresholds, and discomfort levels can be compared as accurately as possible.

Third, the DSL procedure does not attempt to make speech *equally* loud in each frequency region, although it does attempt to make it *comfortably* loud. For any degree of hearing loss, the procedure specifies a target (or desired) ***sensation level***, as shown in Figure 10.2.[c] As hearing thresholds increase, the target sensation level decreases. This is necessary because a person with a profound hearing loss has only a small dynamic range between threshold and discomfort.

The DSL sensation level targets were derived and revised as follows: [1594]

- For profound losses, the desired sensation levels are based on the sensation levels experimentally found to be optimal.[524, 1653]

[c] The sensation level of speech is defined as the short-term maximum rms level of a 1/3-octave band of speech minus the person's threshold at the center of the band. This is similar to the definition used in the Speech Intelligibility Index method.

DSL

The target Real-Ear Aided Gain values (in dB) used in DSL 4.0 as a function of threshold and frequency.

	Frequency (Hz)								
dB HL	**250**	**500**	**750**	**1000**	**1500**	**2000**	**3000**	**4000**	**6000**
0	0	2	3	3	5	12	16	14	8
5	3	4	5	5	8	15	18	17	11
10	5	6	7	8	10	17	20	19	14
15	7	8	10	10	13	19	23	21	17
20	9	11	12	13	15	22	25	24	20
25	12	13	14	15	18	24	28	27	23
30	14	15	17	18	20	27	30	29	26
35	17	18	19	21	23	30	33	32	29
40	20	20	22	24	26	33	36	35	32
45	22	23	25	27	29	36	39	38	36
50	25	26	28	30	32	39	42	41	39
55	29	29	31	33	35	42	45	45	43
60	32	32	34	36	38	46	48	48	46
65	36	35	37	40	42	49	52	51	50
70	39	38	40	43	45	52	55	55	54
75	43	42	43	46	48	56	59	58	58
80	47	45	47	50	52	59	62	62	61
85	51	48	50	53	55	63	66	65	65
90	55	52	54	57	59	66	69	69	69
95	59	55	57	60	62	70	73	73	
100	62	59	61	64	66	73	76	76	
105		62	64	68	70	77	80	80	
110		66	68	71	73	80	83	84	

From Seewald (private communication, by permission).

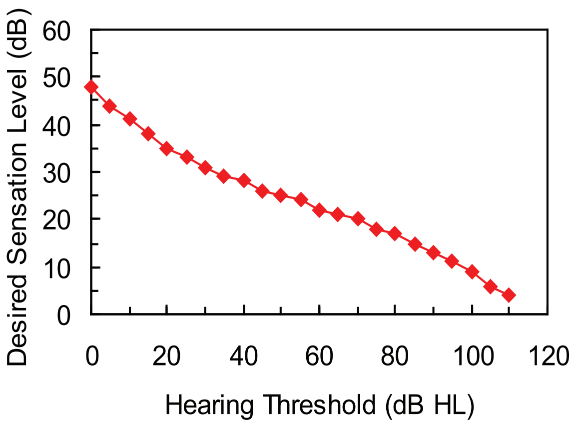

Figure 10.2 Sensation level targets for the Desired Sensation Level method as a function of hearing threshold, at 1 kHz. Values are very similar at other frequencies.

- For mild to severe losses, the sensation level targets for bands of speech are placed one standard deviation below the estimated MCLs for pure tones.[846, 1393]
- For normal hearing, the desired sensation levels are those that are experienced by people with normal hearing when listening unaided.

The DSL procedure uses desired sensation levels to calculate its target real-ear aided gain. At each frequency, REAG equals hearing threshold (in dB SPL at the eardrum), plus the desired sensation level, minus the short term maximum speech levels in the field for speech at an overall level of 70 dB SPL. Software (version 3) implementing the final linear version of DSL also performed many other calculations related to the implementation and verification of the prescription. These included allowing for the effects of different transducers used in the assessment of thresholds, allowing for individual RECD values, prescribing OSPL90, and graphically displaying measured and prescribed speech levels relative to threshold and discomfort.

10.2.4 Examples and comparisons: POGO II, NAL-RP and DSL

The three procedures use different formulae, are based on different principles and, not surprisingly, lead to markedly different prescriptions for many hearing losses. This section shows the target insertion gains prescribed by each procedure[d] for two different audiograms.

Moderate, flat loss. Figure 10.3 shows the audiogram and the insertion gain values prescribed by each of the three procedures. The NAL-RP procedure provides less gain than the other two procedures for both the low and high frequencies.

Moderate, steeply sloping loss. The prescriptions shown in Figure 10.4 are very similar up to 1 kHz, but both DSL and POGO-II prescribe average gains considerably higher than the NAL-RP prescription, and frequency response shapes considerably steeper than the NAL-RP prescription.

The differences between prescriptions are marked, and they matter. These differences among the prescriptions can be thought of as a difference in average

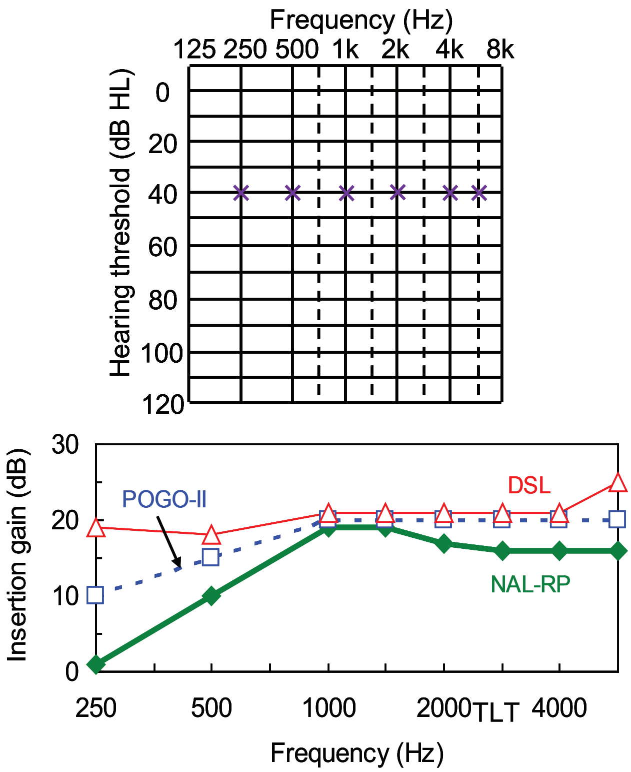

Figure 10.3 Audiogram of a moderate, flat sensorineural hearing loss and insertion gains prescribed by the DSL (triangles), POGO-II (squares) and the NAL-RP (diamonds) procedures.

gain plus a difference in the shape of the frequency response. For adult patients wearing a hearing aid with a volume control, inappropriately prescribed average gain is not a serious problem, because the patients will compensate for the error in prescription by adjusting their volume controls. Very young children and adults who cannot vary the volume control (for any reason) do not have this luxury, and the correct prescription of average gain is important. Patients usually cannot, however, alter the frequency response shape. Even patients able to alter a volume control will usually prefer not to have to adjust it. Given the differences in prescriptions that these procedures produce for these and other audiograms, it is likely that at least one of these procedures is not prescribing an optimal average gain or frequency response.

If the differences between the prescription targets are as great as shown in the preceding examples, why did

[d] DSL REAG targets have been converted to insertion gain by subtracting the adult average real-ear unaided gain curve (from Table 4.6, for 0° incidence, with no control microphone present).

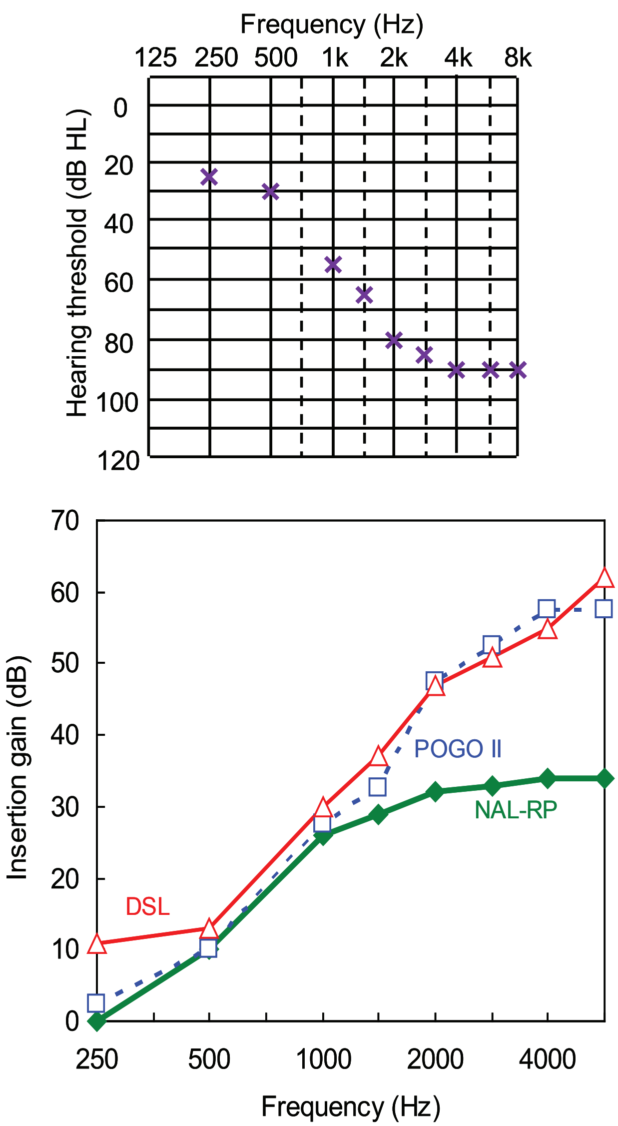

Figure 10.4 Same as Figure 10.3, but for a moderate, steeply sloping sensorineural hearing loss.

it not become obvious in clinical practice which, if any, of the three procedures was optimal? One reason is that with real hearing aids, patients may receive the same gain-frequency response no matter which prescription formula is used. When the NAL-RP method was used, it was common for the measured insertion gain at 3 and 4 kHz to be less than the target gain, because of the restricted range of gain slopes that most hearing aids have in the 2 to 4 kHz range, or because feedback oscillation limited the amount of high-frequency gain achievable. In such cases, substitution of an alternative formula that requires a high-frequency gain slope even steeper than the NAL-RP prescription does not result in a fitting with any more slope. Consequently, the large differences that exist between the procedures rarely emerged in clinical practice.

Because of its widespread adoption in clinical practice and research studies, the NAL-RP selection procedure has been more widely evaluated than any other procedure. Immediately after formulation of the NAL-R procedure, an experiment using adult subjects revealed that few people significantly preferred any of four alternative gain-frequency shapes, although the majority did not object to application of a high-frequency cut.[220] Averaged across frequency, the prescribed gain agreed closely with the preferred gain. Experiments using older children confirmed that, averaged across children with the same audiogram, the NAL-RP procedure neither underestimated nor overestimated the preferred response slope or the preferred average gain.[283, 1667] Other studies examining requirements for multi-memory hearing aids also provided support for the NAL-R or NAL-RP prescription.[864, 886]

Conversely, some studies that have closely examined audibility are sometimes interpreted as showing that other gain-frequency responses will lead to more intelligible speech. It is easily possible to provide a prescription that makes speech more audible than is achieved with the NAL-RP prescription, and this greater audibility directly leads to a higher SII value.[779, 1038, 1483] This is achieved simply by using more gain at some or all frequencies, and hence results in more loudness, which may not be appreciated by the patient. Conditions with a very high SII value (achieved by making speech highly audible at all frequencies) usually do not result in an intelligibility score commensurately higher than that obtained with the NAL-R prescription, and can sometimes result in a lower score and certainly in lower sound quality.[1038, 1483] If subjects are allowed to adjust the volume control to regain the loudness they prefer, then speech intelligibility scores are likely to be lower than for the NAL-RP prescription.[1738]

Meta-reviews of experiments evaluating the NAL-RP prescription indicate that while several studies have concluded that NAL-RP on average gives the amount of overall gain preferred by subjects, several other studies have concluded that people prefer *less* gain than NAL-RP prescribes.[318, 1255] Only in one study, and only for one subgroup of subjects, has a preference by adults for more gain than that prescribed by

NAL-RP been found.[e, 1013] On balance, and on average, it seems that the NAL-RP prescription prescribes too high a gain by a few dB with an appropriate frequency response shape.

There are three notes of caution to this otherwise positive verdict for the NAL-RP prescription. First, subjects used in research studies were not used to a high-frequency emphasis greater than that pre-

Theoretical conundrum: Should we preserve individual open-ear characteristics - insertion gain targets or real-ear aided gain targets?

With a REAG target, all people who have the same degree of hearing loss are prescribed the same gain from the free field to their eardrum, irrespective of the gain provided by their unaided ear canal. With an insertion gain target, all people who have the same degree of hearing loss are prescribed the same increase in SPL at their eardrum, relative to the SPL at their eardrum when listening unaided.

One might argue that the IG approach is better, because the job of a hearing aid is to provide more signal than a person gets when unaided, and this is precisely what insertion gain measures. Alternatively, one might argue that once a person puts on a hearing aid, what they used to receive at their eardrum when they were unaided is inconsequential!

Either type of procedure can be converted into the other by adding or subtracting an average REUG. For people with a REUG curve that is close to average, the type of gain target prescribed would then have little effect on the amplification prescribed (though the particular formula chosen may well do so). Consequently, it does not matter which prescription approach is used.[262, 1376]

A person whose REUG curve is a little different from average is more likely to prefer an amplification characteristic that incorporates his or her own REUG curve than an amplification characteristic that incorporates an average REUG curve - that is, an insertion gain prescription is preferable.[1376]

The impact of choosing a REIG versus a REAG target will be greatest for people whose REUG curves are most dissimilar from average. One such group comprises people whose external ears have been altered by surgery, especially mastoidectomy. Their enlarged ear canals cause a Helmholtz resonance in the 1 to 2 kHz range, instead of the expected wavelength resonance around 2.7 kHz. It certainly does not seem appropriate to fit a hearing aid in such a way that this unnatural resonance is maintained.[921] In other words, a REAG target is most appropriate for this group. The same argument applies to a second group of people who have a large perforation of the eardrum, and who hence have a REUG curve with two peaks separated by a valley.[1251] Adults with REUG characteristics greatly different from average prefer amplification characteristics based on an average REUG - that is, a REAG prescription is preferable.[1376]

A third group with unusual REUG characteristics is children under about three years of age. At birth, infants have an ear canal wavelength resonance around 6 kHz, decreasing gradually to adult values as their ear canals get longer.[984] It is unclear whether they obtain any advantages from having a high resonant frequency, or whether a high resonant frequency is just the inevitable consequence of a short ear canal, which in turn is the inevitable consequence of being born with a small head, which all mothers greatly appreciate. The small head argument seems more probable, and if so, a REAG target seems more appropriate for this group.

In summary, it seems more appropriate to adopt a REAG target for young children and for people with deformed or surgically altered ear canals, but an insertion gain target for all other patients. This split is convenient, because for young children it is easier to measure REAG (see Section 16.4.3), whereas for adults it is easier to measure insertion gain (because locating the probe microphone for accurate results is easier - see Section 4.4.1). For adults with REUG characteristics that are close to average, it is not important which type of real-ear gain is chosen as a target.

[e] The preferred response for subjects with moderate to severe flat loss had about 4 dB more gain than the NAL-R response from 250 to 1000 Hz.

Practical issues associated with the methods

- The NAL-RP formula (as with its NAL predecessors and successors) is based on maximizing speech intelligibility and sound quality in quiet and in noise. A convenient consequence of the procedure, however, is that for people with sloping losses, less high-frequency gain is required than for most other procedures. This makes targets easier to achieve, and makes feedback oscillation less likely.
- The measurement procedures developed as part of the DSL procedure are particularly advantageous with infants, as further discussed in Section 16.2. These excellent measurement methods can be applied to other prescription formula, as outlined in Section 16.4.3.

scribed by the NAL-RP procedure. (Often, not even the NAL-RP gain target at 4 kHz would have been achieved in the hearing aid they usually wore.) It is therefore possible that the subjects may have preferred and/or benefited from a greater high-frequency emphasis if they had enough experience with it prior to participating in the experiments. In the past, limitations with wearable hearing aids prevented the desired responses from being achieved. Multi-band hearing aids and effective feedback cancellation algorithms now make such an experiment possible.

Second, the NAL-RP response shape has not always been found to be optimal. Pleasantness is often found to be greater if less high-frequency emphasis than that prescribed by NAL-RP is used,[1038, 1125] even when the NAL-RP response subjectively and objectively provides greater speech intelligibility.[1125] Even the original validation of the NAL-R response indicated that many subjects were equally happy with the high-frequency cut response when the criterion was to choose the more pleasant response.[220]

Third, audibility of high-frequency sounds can sometimes be restricted by the high-frequency maximum output of the hearing aid, rather than by the high-frequency gain. It is conceivable that some additional high-frequency gain would be valuable if it could be achieved without saturating the hearing aid. This supposition does not seem likely, however, as some of the research showing the decreased value of high-frequency amplification for severe and profound hearing loss has *not* been constrained by high-frequency distortion.

10.3 Difficult Issues in Prescription

Before proceeding with a description of different approaches and formulae for non-linear hearing aid prescriptions, we will consider several issues, all of which affect non-linear prescriptions, and most of which have been informed by research using linear prescriptions.

10.3.1 Acclimatization and adaptation to gain and frequency response

Listening to amplified sound can produce gradual, long-term changes in the hearing abilities of patients. This general phenomenon is referred to as ***acclimatization***, although the same word is used in different ways. One meaning relates to the effect of listening experience on the ability to understand speech - this aspect of acclimatization is covered in Section 14.7.

Another change that takes place following an adult patient's first hearing aid fitting is that the amount of gain preferred by the patient may gradually increase over time. Because hearing losses usually occur gradually, the patient's auditory processing system becomes used to the lower levels of excitation that an impaired cochlea passes on to the central nervous system. When hearing aids are first worn, there is a sudden increase in output from the cochlea – possibly providing greater loudness than the patient is willing to accept. Over time, the auditory processing system readjusts to the increased cochlear output,[1363] so the patient prefers slightly more amplification, thereby enabling greater speech intelligibility in environments with low speech levels. This aspect of acclimatization is called ***adaptation to gain***.

Not surprisingly, the greater the hearing loss, the greater the increase in loudness when hearing aids are worn, the greater the adaptation to gain over the first few years of hearing aid use. For mild hearing losses, the change in loudness produced by the hearing aids is so small there is no measurable adaptation to gain.[883] For someone with a severe hearing loss receiving their first hearing aids (an uncommon experience in developed countries), it can be inferred that the preferred gain is likely to increase by nearly

10 dB over the next three years of aid use.[883] There are two consequences of this effect of hearing loss on adaptation. First, since most research studies are carried out on people with mild to moderate hearing loss, the amount of adaptation expected is small on average, and the difference may or may not be significant.[318, 1138] Second, any allowance for gain adaptation in hearing aid prescription formulae must depend on the degree of hearing loss. This applies whether the adaptation is included in the formula, and applied by the clinician (requiring an adjustment of the hearing aid after the patient gains some experience) or is built into the hearing aid as a signal processing scheme that automatically increases the gain over the weeks or months after fitting.

A further aspect of adaptation is that the shape of the gain-frequency response that is optimal for the patient (from the perspective of preference, or speech intelligibility, or both) may also change following experience with amplified sound. It would seem to follow from the hearing-loss-dependent adaptation described in the previous paragraph that a patient with a predominantly high-frequency hearing loss, who is prescribed a hearing aid with a marked high-frequency emphasis, would require more time to adapt to the large increase in audibility of the high frequencies than to the small increase in audibility of the low frequencies. Although such an inference is believed by many clinicians to be true, and may well be true, the research that has investigated changing frequency response preferences following fitting has not found any significant variation with time.[883, 1138]

10.3.2 Preferred loudness

As we will see, all the prescription procedures for non-linear hearing aids involve amplifying sounds so that they are as loud as, or no louder than, for a normal-hearing person listening to the same sound. It is, however, very much an assumption that hearing-impaired people would *like* to perceive sounds with normal loudness. Only a few experiments have investigated the loudness (which is a percept and difficult to measure precisely) that hearing-impaired people prefer when wearing hearing aids, and they have produced inconclusive results. When loudness is calculated using a loudness model adjusted to allow for hearing loss,[1227, 1229] hearing aid wearers seem to prefer *less* than normal loudness, both when measured in the laboratory and when wearing hearing aids in real life.[1647-1649] When the hearing aid wearers are asked to assign loudness categories to sounds while listening at the gains prescribed by NAL-NL1 (which aims to give overall loudness no greater than normal), they assign loudness ratings higher than normal.[1648, 1649] These results would be consistent with each other if the loudness model, which was used both to analyze the experimental results and as part of the derivation of NAL-NL1, underestimated the loudness of sounds for hearing-impaired people.

Even though we may be somewhat uncertain about the *loudness* that hearing-impaired people prefer, we know a great deal about the amount of *gain* they prefer. Because the NAL-RP and NAL-NL1 prescriptions have been used as the baseline response in so many experiments, we can be fairly confident about the gain that adults prefer relative to these two prescriptions, at least for typical input levels. Comprehensive reviews of experiments have shown that, on average, hearing-impaired adults prefer about 3 to 4 dB less gain than that prescribed by both NAL-RP and NAL-NL1.[318, 1255]

10.3.3 Dead regions

As introduced in Section 1.1.5, the primary responsibility for sending an electrical representation of sound to the brainstem falls on the inner hair cells (IHCs). When, in a particular region of the cochlea, there are no functioning IHCs and/or no auditory nerves to which they connect, that part of the cochlea is termed a ***dead region***. Consequently there will be a range of frequencies (those that normally resonate in that region of the cochlea) for which no part of the cochlea is dedicated to converting signal power into neural signals. Power in that frequency region, if sufficiently amplified, may still be detected in other parts of the cochlea, but the message sent to the brainstem will be confused, because those other parts of the cochlea will be sending information about the signal components in their own frequency region as well as the frequency components that would normally be conveyed by the dead region.

Intriguingly, people have better frequency discrimination for frequencies just inside dead regions than for slightly lower or higher frequencies. This is possibly caused by neurons in the auditory cortex that would normally be excited by the part of the cochlea that is no longer functioning finding that they have nothing to do, and reconnecting instead to the closest nerve fibers that *are* conveying signals.[935, 1166]

Despite this copious brain power being available to analyze the signals spilling over into the functioning frequency regions adjacent to a dead region, amplification within a dead region contributes less to speech intelligibility than when there are functioning IHCs. Two experiments have shown that extending amplification upward by more than about one octave past the lower limit of a high-frequency dead region did not seem to further improve speech intelligibility in quiet or in noise, although there was considerable individual variability in how far it was worth extending amplification.[f, 69, 1858] Another experiment confirmed that this same rule applied in noise, but that speech intelligibility in quiet was maximized if amplification extended over the entire speech frequency range![1108]

Choosing not to amplify too far into a dead region (where hearing loss and prescribed gain both tend to be large) can simplify the fitting by avoiding problems with feedback oscillation, and in some cases may slightly improve speech intelligibility.[69] Although much of the research into amplification within dead regions has focused on high-frequency dead regions, which are the most common type, dead regions can occur at low or mid frequencies, and can occur within a wide variety of audiogram configurations.[1215, 1454] On average, people with dead regions have poorer speech reception thresholds in noise than those who don't, even when the differences in their audiograms are relatively minor.[1454]

Regions can be considered effectively dead even when IHCs have some limited function. If the IHCs within a region required a basilar membrane vibration so great that sounds at their characteristic frequency are most easily detected at some other place in the cochlea, then that part of the cochlea is effectively dead: IHCs in the region may produce neural responses if the input is amplified enough, but neurons elsewhere in the cochlea will always be producing stronger neural firings to the same stimuli.[1215]

Dead regions can be detected by the use of psychoacoustic tuning curves and/or the TEN test.

Psychoacoustic tuning curves

A ***psychoacoustic tuning curve*** (***PTC***) is created by finding the softest level of narrow-band sounds at various frequencies that just masks a pure tone presented at a small sensation level, typically about 10 dB above its threshold in quiet. If the cochlea has functioning hair cells in the region tuned to the pure tone, then the masker that most easily masks the pure tone will have a centre frequency the same as that of the pure tone. If, however, that part of the cochlea has a dead region, then the person must be hearing the pure tone based on the neural signals created in some other part of the cochlea. (This form of neural stimulation is referred to as ***off-frequency listening***, or ***off-place listening***.[832, 1215]) If so, the masker that most easily masks the pure tone will be at that other frequency. Figure 10.5a shows the PTC for a signal frequency of 1 kHz in a cochlea with no evidence of a dead region. Each of the maskers shown is just able to mask the 1 kHz signal. The tip of the tuning curve is also at 1 kHz, because the signal could be masked by a masker at this frequency at a level lower than at any other masker frequency.

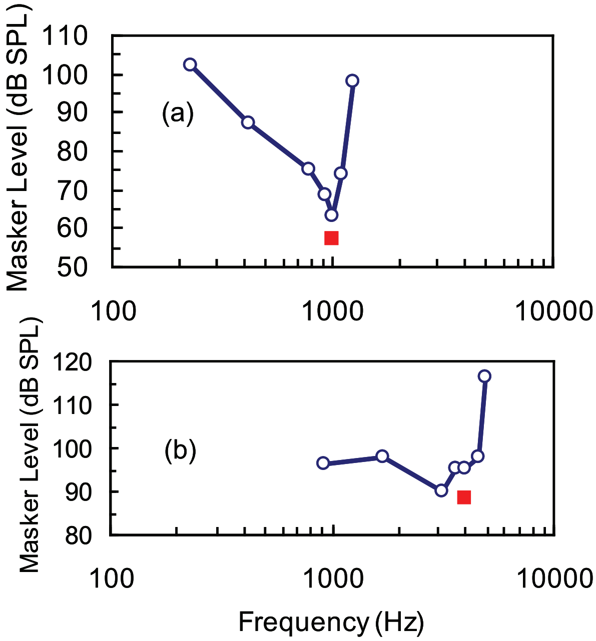

Figure 10.5 The psychoacoustic tuning curve for a person with (a) a threshold of 40 dB HL at 1000 Hz, and (b) a threshold of 70 dB HL at 4000 Hz. The square shows the frequency and level of the signal.

Figure 10.5b shows a PTC that indicates a dead region at 4 kHz. The 4 kHz signal is more easily masked by the masker at 3.15 kHz than by the masker at 4 kHz, indicating that the 4 kHz signal was probably being

[f] The two experiments recommended extending amplification up to 1.5 to 2.0 times the lower edge frequency of the dead region. Given the variability of the data upon which this rule is based, and the minimal impact of bandwidth on speech intelligibility, a simplification to one octave above the edge of the dead region seems justified.

perceived on the basis of neural signals created in the 3.15 kHz region of the cochlea.

Unfortunately, PTCs can produce inaccurate results. When a pure tone and a narrowband masking sound have similar (but not identical) frequencies, the two sounds can create beats or combination tones at much lower frequencies, which the patient may detect.[g, 218] Detection of these spurious sounds can create a tip in the PTC at or near the signal frequency, even if the signal frequency is within a dead region.[1215] PTCs comprise masked thresholds at several different masker frequencies for each signal frequency, and so require a longer time to measure than is usually available in the clinic. Even fast methods based on continuously sweeping the masker frequency require 10 minutes per frequency, plus familiarization and practice time.[936]

The TEN test

If a broadband masking sound is presented at the same time as the pure tone signal, then the threshold of the pure tone will be raised whenever the power of the masking sound in the frequency region immediately surrounding the signal is itself above threshold in quiet. If the pure tone is being detected at its normal place, then the masked threshold will be similar to the power of the broadband noise falling within the auditory filter (Section 1.2.1) surrounding that frequency. If, however, there are no functioning IHCs in that part of the cochlea, and the pure tone is actually being detected at some other place in the cochlea with better hearing sensitivity, what will be the effect of adding the broadband masking noise? The masked threshold of the pure tone will be much greater than the local power of the masking noise and also much greater than the threshold in quiet. Noise that produces equal masked thresholds at all frequencies for people with normal hearing has been termed ***threshold equalizing noise*** (***TEN***).[1235] Coincidentally, the minimum amount by which the masked threshold has to exceed the threshold in quiet and the minimum amount by which the masked threshold has to exceed the power of the masker within each frequency region before a dead region can be diagnosed are both 10 dB.[1235]

A later more convenient version of this clever, and cleverly named, TEN test produces equal masking in dB HL, rather than equal masking in dB SPL.[1233] This improvement means that the test can be applied by just re-measuring the audiogram in the presence of the TEN noise, and comparing the new masked thresholds to the original audiogram measured in quiet. A step size of 2 dB is recommended for both audiograms. The newer version of the noise also has a more restricted bandwidth and has noise with a low crest factor (Section 4.1.3), to decrease the likelihood of it producing loudness discomfort during the test. There are, nonetheless, sometimes still difficulties in applying the test. For subjects with profound or upper severe losses, loudness discomfort can still prevent a result from being obtained, the 10 dB criterion is approximate (the optimal criterion may depend on age or hearing loss), and the measured difference between the masked threshold and the threshold in quiet is subject to several dB of measurement error, especially if these thresholds are based on the usual 5 dB step size.[1215]

So is it necessary to test for dead regions before prescribing amplification? The jury is still out on this question for three reasons.

First, although some experiments have shown good agreement between dead regions diagnosed by PTCs and dead regions diagnosed by the TEN test,[785, 936, 1235] other experiments have shown a very imperfect relationship.[h, 292, 1746] It is likely, though, that this divergence has been caused by an inappropriate selection of stimuli for the PTCs.

Second, the value of testing for dead regions undoubtedly depends on what prescription would have been given in the absence of testing for dead regions.

[g] The use of narrowband noises, rather than pure tones, for the masker reduces, but does not entirely avoid, the problems caused by beats and combination tones.[1215]

[h] Agreement between the PTC result and the TEN test result was improved in one study if the criterion for a dead region in the TEN test was increased such that the level of the masked threshold relative to the masker has to be greater than 14 dB rather than greater than 10 dB.[1746] Unfortunately, another study found it necessary to reduce the criterion to 8 dB to make the two methods more consistent.[936] PTCs can produce misleading results if the bandwidth of the masking sound is too narrow, because the target sound and the masker can then produce beats and/or combination tones that are audible even if the target sound itself is masked.[933, 934]

Suppose one made the assumption that the benefit of achieving audibility within each frequency region did not depend on the degree of hearing loss within each region – one would then prescribe a large gain at every frequency for which the loss was severe. If some of the frequencies were within a dead region, a significant error would have been made – the amplification at best would be wasted, and at worst would cause feedback oscillation and reduce speech intelligibility.

Alternatively, suppose that the prescription was based on the assumption that the greater the loss in a frequency region, the smaller the contribution that region will make to intelligibility, no matter how much audibility is achieved. The frequency regions with the greatest loss would not then be prescribed as much gain as in the first example, perhaps not even sufficient to achieve audibility of speech at some levels. Learning that the frequency region has no functioning IHCs may not then significantly change the prescription.

Although there is not a 1:1 correspondence between degree of hearing loss and the presence of dead regions, it is certainly true that the likelihood of a dead region greatly increases as the degree of loss increases.[69, 292] At 4 kHz a dead region is more likely than not, once hearing threshold exceeds 70 dB HL.[3] Consequently, any prescription procedure that allows for the diminished usefulness of frequency regions as hearing loss increases already partly allows the impact that knowledge of dead regions would have on the prescription of amplification.

Third, even if the prescriptive procedure has not explicitly allowed for the reduced effectiveness of frequency regions with severe hearing loss, experienced audiologists are likely to over-rule the prescription and decide that some frequency regions have too much loss to be aidable. One study has shown that, rightly or wrongly, on the basis of the audiogram alone, experienced clinicians typically decide that hearing loss greater than about 90 dB HL is not aidable.[1745] When applied to audiograms for which the researcher (but not the clinicians) knew the extent of the dead regions, the resulting upper limit of amplification was little different from that predicted by limiting to nearly an octave above the edge of the dead region. Furthermore, the speech intelligibility resulting from the 90 dB rule was equal to or better than that resulting from the rule based on the edge of the dead regions.[1745]

Given the potential measurement problems, clinical time involved, and lack of certainty about how to use the result, most prescription formula do not require measurement of dead regions as a mandatory input to the prescription formula. The CAM2 prescription (Section 10.4.7) is, however, only intended to be applied at frequencies where there is no dead region. The impact of dead regions on amplification requirements is both plausible and supported by evidence, so it is possible that information about dead regions in different frequency regions will make its way into prescription formulae to a greater extent in the future.

10.3.4 Severe hearing loss, effective audibility and high-frequency amplification

As frequency progressively increases past 2 kHz, several things conspire to make life difficult for the clinician and the hearing aid engineer alike, not to mention the patient. The intensity of speech weakens and hearing loss increases (more often than not). Both of these mean that more gain is needed to achieve audibility. So what's the problem? Although speech information exists for frequencies up to approximately 10 kHz, the amount of information per 1/3 octave band decreases as frequency increases.[54] This decrease is compounded by the patient's decreasing ability to use the information, even if it is made audible, if hearing loss increases with frequency. Furthermore, the higher gain makes feedback oscillation more likely.

Even in the absence of high-frequency dead regions, how much high-frequency amplification is optimal? There is no simple answer to this badly formed question, or this topic would not be in the *Difficult Issues* section of this chapter! There are several connected issues:

- Up to what frequency should speech be made audible?
- Within this range, how should sensation level vary with frequency?
- How do the answers to both of these questions vary with input level and hearing loss?

Many experiments have evaluated how speech intelligibility and/or user preference change as bandwidth is increased from one upper frequency limit to a higher one. These studies have variously shown that there is value in extending amplification to: 1.8 kHz (for sound quality);[1740] 3 kHz;[833] 3.2 kHz;[43] 3.6 kHz;[19, 1289] 4 kHz;[1801] ≥4 kHz;[1440] 4.5 kHz (when non-individualized frequency responses are used);[747, 765] 5 kHz (for steeply sloping audiograms);[1506] ≥5.6 kHz;[765, 1635, 1802] ≥6 kHz;[763, 1642, 1740] ≥7.5 kHz (in the absence of dead regions, with spatial separation of sounds, for fricative detection, or for clarity but not pleasantness);[69, 577, 1223, 1224, 1858] and ≥9 kHz (for flat and gently sloping audiograms).[i, 1506] In some cases, speech intelligibility or preference actually decreased if bandwidth was extended past the optimal limit, but generally this occurred for just some individuals, not for the study group as a whole.

The results of these same experiments can be analyzed in another manner. Instead of recording the lowest bandwidth at which maximum performance was achieved, we can record the mean hearing loss of the subjects that corresponded to this upper frequency limit. On this basis, amplification should be extended until hearing thresholds reach approximately 75 dB HL. This is bound to be an under-estimate because in many of the experiments, the best upper limit was the highest out of all those compared.

There are many reasons why different experiments would suggest such different optimal upper frequency limits:

- The greater the high-frequency hearing loss, the smaller the value of an extended high-frequency response.[1506]
- The greater the unique information present in the individual talker's voice at high frequencies, the wider the optimal bandwidth will be. Audibility above 4 kHz appears to be more important for the perception of fricatives spoken by females than by males, for example.[1701, 1702] Inadequate audibility of very high-frequency sounds is likely the cause of the late development of fricative production in hearing-impaired infants.[1704]
- The shape of the noise spectrum relative to the speech spectrum will affect the value of a higher frequency limit. When SNR increases as frequency increases, there will be more value in extending the frequency limit than when SNR decreases as frequency increases.[833] Expressed differently, high-frequency speech cues become more important when low-frequency cues are not available, such as when they are masked by noise.[765] When music is used as the stimulus, the value of very high (and very low) frequency components increases.[1243] Similarly, when the test stimuli consist only of high-frequency fricatives, an extended high-frequency response will seem more valuable than when a range of speech sounds are used.[577]
- Spatial separation of speech and noise generally improves speech intelligibility, and high-frequency cues contribute to this improvement. Wider bandwidths are therefore likely to be more important when the target sound is spatially separated from competing sounds.[1223]
- The sensation level (or lack of it!) provided within the extended bandwidth will affect the outcome. Even if someone would prefer and benefit from a small sensation level extending to 8 kHz, if the choices offered are an 8-kHz bandwidth with a high-frequency sensation level so great that the high-frequency components dominate the overall loudness, versus a 4-kHz bandwidth with balanced loudness across frequency, the person may well choose, and/or perform better with, the more restricted bandwidth. Consistent with this, Horwitz (2008) showed that extending the bandwidth was valuable when applied to an individualized NAL-R gain-frequency response, but not when it was applied to an alternative response that provided greater amplification of the mid and high frequencies. In general, excessive presentation levels reduce intelligibility;[481, 482] and in particular, excessive high-frequency stimulation may mask information at lower frequencies (i.e. ***downward spread of masking***).[1640] When the hearing loss is severe, loudness increases markedly as the sensation level increases. Conversely, a hearing-impaired person will show no benefit for a wide bandwidth if the gain is insufficient to provide any audibility across that extended bandwidth.[564]
- Some experiments may have had a greater proportion of subjects with high-frequency dead regions. Much of the research on high-frequency amplification was performed prior to researchers being aware of the importance of dead regions.

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[i] Inclusion of the ≥ symbol indicates that the "optimal" upper frequency limit was the highest frequency limit investigated. The optimal limit may actually have been higher still.

- Higher bandwidths tend to be preferred by people with flat losses, and lower bandwidths by people with steeply sloping losses.[1506] This is understandable from two perspectives. First, those with the steeply sloping losses are most likely to have had high-frequency dead regions, and hence to have not benefited from high-frequency amplification. Second, even if none had dead regions, those with sloping losses are less able to extract information from high-frequency signals, where hearing loss is severe, than from low-frequency signals, where hearing loss is more moderate. Consequently, the value of high-frequency audibility relative to low-frequency audibility is much less for these patients.

The relative importance of different frequency regions for people with a wide variety of audiograms has been extensively investigated by low-pass and high-pass filtering speech with different cut-off frequencies.[292] By comparing measured speech intelligibility to intelligibility predicted on the basis of audibility and the importance of each frequency region (i.e. using the Speech Intelligibility Index), the ability of people with different degrees of hearing loss to use the information made audible to them can be computed.

Figure 10.6 shows the resulting maximum percentage of information that hearing-impaired people can extract, at least on average, relative to people with normal hearing. The ability to extract information decreases as hearing loss increases, and the maximum achievable proportion, even when good audibility has been achieved, drops to half when hearing thresholds reach 66 dB HL.[292] The same result was obtained for vowel-consonant-vowel test materials and for sentence test materials, and applied equally at all frequencies.

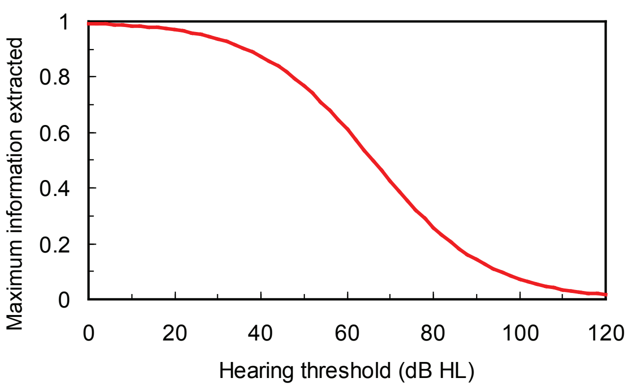

Figure 10.6 The proportion of information that hearing-impaired people with different degrees of hearing loss can extract when optimal audibility has been achieved.

The model that produced these results also indicated that at very low sensation levels, hearing impairment has minimal impact on the ability to extract information. Other studies also suggest that hearing-impaired people make excellent use of speech just above their thresholds.[747, 1701] As sensation level grows, however, normal hearers can extract more and more information whereas those with hearing impairment are limited to the proportion of information shown in Figure 10.6.[292] This graph is, of course, an average – some hearing-impaired people can extract more information than this, and some less. We might expect the latter to include a higher proportion of people with dead regions.

These results are consistent with those of several other research studies: although hearing impairment reduces the amount of information that can be extracted from audible speech, it does so in a way that is independent of frequency.[763-765] The ability to extract information does commonly decrease more for the high frequencies than for the low frequencies. This is because hearing thresholds are usually greater for the high frequencies, rather than because the same degree of hearing loss reduces information extraction more for the high frequencies than the low frequencies. Indeed, re-analysis of earlier data that had suggested a reduced ability of hearing-impaired people to use high-frequency information,[276, 293] but now using the same analysis methods that led to Figure 10.6, indicates that the reduced ability depends on hearing loss at each frequency, but not on frequency itself.

Figure 10.6 explains the research findings that led to the NAL-RP prescription (Section 10.2.2). As thresholds move into the severe-profound range, which typically occurs first at high frequencies, information extraction at high frequencies becomes progressively less effective, and it becomes less worthwhile achieving high-frequency audibility. As high-frequency audibility is reduced, so too is total loudness, so the low frequencies can be given greater audibility for the same total loudness, as also reflected in the "profound correction" values in the NAL-RP formula. Speech intelligibility is therefore greater than if more of the allowable loudness had been "wasted" on the less effective high frequencies.

High-frequency amplification in a nutshell

The following summary seems consistent with the research reviewed in this section.

- High-frequency information in speech is valuable out to around 10 kHz, particularly for fricative sounds, and particularly for listening to female talkers.
- The greater the hearing loss, the smaller the amount of information that people can extract from audible speech. The deficit relative to normal hearers increases as sensation level increases. This deficit applies equally at all frequencies, but will have adverse effects much more often in the high frequencies because hearing loss is most often largest in the high frequencies.
- *Excessive* high-frequency sensation level *can* decrease speech intelligibility, but most often just causes additional loudness and poor speech quality without impacting on speech intelligibility at all.
- Excessive high-frequency amplification arises from excessive gain, and hence sensation level, rather than excessive bandwidth. The bandwidth over which some audibility should be provided should always (with a possible exception in the case of known dead regions) be as wide as possible.
- Use a prescription procedure that attempts to provide a wide bandwidth and sensation levels that reflect both the importance of different frequency regions and the impact of hearing loss on the ability to extract information.

The decreased ability of the impaired ear to extract information from a signal even when it is audible has been referred to as ***hearing loss desensitization***.[293, 1396, 1737] The same concept is captured by referring to the ***effective audibility*** of the speech, rather than the physical audibility determined just by the sensation level of speech. The effective audibility of a speech signal is the sensation level of speech that would give a person with normal hearing the same intelligibility a hearing-impaired person receives from the speech signal.

If speech intelligibility is estimated using the Speech Intelligibility Index without allowing for hearing loss desensitization, intelligibility is over-predicted whenever thresholds at any frequency are severe or profound. Increasing the audibility of speech where hearing thresholds are largest may make things look good on a speech-o-gram (Section 4.5.7) and simple speech intelligibility index calculations may predict that speech should be more intelligible, but hearing loss desensitization means this is just not the case.[1483] Desensitization should not be confused with level distortion, as defined in the Speech Intelligibility Index model.[54] Even people with normal hearing are less able to recover information from a signal if they are forced to listen at high levels. People with a severe or profound hearing loss have no choice in this: either they listen at high SPLs or they do not hear anything! Desensitization is an additional difficulty faced by people who have hearing thresholds within the severe, or even upper moderate, range.

The need for a small, but positive, sensation level over the high-frequency range to maximize speech intelligibility brings with it some challenges. To retain a small sensation level as the input level changes over a wide range requires a very high compression ratio. If this is provided with fast-acting compression in a multi-channel hearing aid, the loss of spectral shape will decrease speech intelligibility relative to the intelligibility were the same audibility to be achieved without compression (Section 6.5.2). The only alternative appears to be to have at least some slow-acting compression in the hearing aid.[1556] Even so, unless extremely high compression ratios are used, achieving very-high-frequency audibility for soft input levels would seem to have more disadvantages than advantages. The effect on sound quality of having a small high-frequency sensation level over a wide range of input levels remains unknown, and requires further investigation.

10.3.5 Prescribing compression thresholds

As we saw in Section 6.2.2, the compression threshold is the input level above which the gain of the hearing aid starts reducing as the input level increases. Most hearing aids have multiple channels of compression, so we must be careful when describing compression thresholds. Compression threshold can be expressed as the overall level of a broadband signal with some particular spectral shape at which compression begins in some or all channels, or it can be expressed as the level of the signal within any one channel at which that channel enters compression. The greater the number of channels, the narrower in frequency each channel is, and the smaller the within-channel compression threshold will be relative to the overall, broadband compression threshold. As an example, the compression threshold prescribed by NAL-NL1 and NAL-NL2 is based on speech at an overall level of 52 dB SPL causing each channel to just enter compression. For an 18-channel hearing aid, with each channel 1/3 octave wide, this corresponds to a compression threshold of 46 dB SPL in the 250 Hz channel, and only 32 dB SPL in the 4 kHz channel.[j]

If loudness is to be completely normalized (Section 10.4) by a hearing aid, compression is needed for input levels from the threshold of normal hearing upwards. That is, compression threshold must be around 5 to 10 dB SPL! It is easy to see the impracticality of this. The gain for low-level sounds will equal the hearing loss. This requires that the mold/shell be much more tightly sealed than would be necessary for a linear aid with gain equal to only half the hearing loss or less. A likely result would be physical discomfort or adverse effects on the quality of the aid wearer's own voice (i.e., the occlusion effect; Section 5.3.2). Even if the high gain is achievable without feedback, it would not provide any benefit because there are almost no environments this quiet. High gain at low input levels that never occur in real life has been referred to as "empty gain".[919]

So how low should compression thresholds be? The pros and cons of a low threshold are easy to state, but the optimum value probably depends on other aspects of the compression. As compression threshold is lowered, more and more of the benefits of compression are obtained: the soft sounds of speech, and other soft sounds that people want to hear, are given greater audibility. Unfortunately, more and more of the disadvantages of compression are also obtained: softer noises that people would prefer not to hear are amplified more than higher level speech, and feedback oscillation is more likely. In particular, fast-acting compression with a very low compression threshold will amplify lower level background noise occurring in the gaps between speech sounds more than it amplifies the speech, making the speech seem noisier. On theoretical grounds then, the optimum compression threshold for a fast-acting compressor will be higher than the optimum for a slow-acting compressor.

The evidence base on which to prescribe compression threshold is inadequate. An evaluation of the preferred compression thresholds using a single-channel, fast-acting compression hearing aid with a 2:1 compression ratio, found that most people preferred to

Going in and out of compression: a non-issue

It is sometimes stated that compression thresholds should either be well below typical speech levels or well above them. Mid-level compression thresholds, it is argued, will adversely affect sound quality as the speech "goes in and out of compression." This argument either comes from a misunderstanding of compression, or reflects side effects that some compressors may once have had.

It is true that compression affects sound quality, but what is heard are the various effects of gain rapidly changing. The degree of the quality change depends on how much and how quickly the gain changes. For a fixed compression ratio and fixed attack and release times, the size of the gain change will be *greater* for speech that is totally within the compression region than for speech that is sometimes above compression threshold and sometimes below it.

An additional audible effect as the signal crosses compression threshold could occur only for badly designed compressors that generate a click as the compressor is activated. This is extremely unlikely with digital hearing aids.

[j] The calculation behind these numbers takes into account the dynamic behaviors of detectors in typical compressors (Section 6.2.1), the high crest factor of speech relative to the pure-tone signals with which compression thresholds are usually measured, and, most important of all, the spectral shape typical of speech.[227]

have wideband compression thresholds around 60 to 65 dB SPL.[81, 461] An evaluation using a very slow-acting multi-channel compression hearing aid found that narrowband compression thresholds of 20 dB SPL, within channels 1/3 octave wide, are strongly preferred over much higher narrowband thresholds of 50 dB SPL.[682]

The decision to base the NAL-NL prescriptions on a broadband compression threshold of 52 dB SPL is because it is near the bottom of the range of speech levels normally encountered by people.[1399] When used with slow-acting compression, however, values lower than this are almost certainly optimal. Given the lack of certainty about prescription of compression threshold for any prescription formula, extra emphasis should be given to evaluating the suitability of compression threshold after the aid wearer has had a chance to try the hearing aid in his or her usual environments (see Chapter 12).

It seems that less compression is appropriate for people with severe and profound loss than for people with moderate loss (Section 6.5.1). This could be achieved either by increasing compression threshold, or by decreasing compression ratio, as hearing thresholds increase. The latter approach is used in NAL-NL2. It appears that all prescription procedures prescribe compression thresholds that do not vary with hearing loss. That is, compression threshold is not a parameter that currently is individually prescribed based on any measured hearing characteristics.

10.3.6 Need for accuracy in prescription

How close to the prescription need a hearing aid be adjusted? This is the perennial question posed to prescription developers, and there is no precise answer. An oft-put argument is that because different prescriptions make such different recommendations, it can't matter how closely a hearing aid matches any one of them. This argument is illogical. If a particular set of amplification characteristics is truly best for a patient, another set of characteristics is not made equally good just because someone makes up a different prescription. Prescriptions, whether published or proprietary, certainly do differ: in average gain; response shape; compression ratio; and compression threshold.[868]

We expect prescriptions to achieve three main goals: to give the best speech intelligibility possible for that patient; to give an overall loudness that is acceptable to the patient; and to give a tonal quality preferred by the patient. Speech intelligibility is predominantly affected by the amount of signal audible above both hearing thresholds and background noise. In many situations it is background noise that determines the SNR across most or all of the frequency range. In these situations, for speech inputs at typical conversational levels, the gain-frequency response can be varied considerably (though with some limit) without affecting audibility at each frequency, and hence without affecting speech intelligibility. Consequently, the match to a prescription is then not critical for maximizing intelligibility, provided the alternatives result in the same audibility of the speech components that are not masked by the background noise.[26, 1832] Even when a range of prescriptions enable the same speech intelligibility, however, a much narrower range is likely to be preferred for clarity, pleasantness and tonal quality.[1832] It is common for people to consistently prefer one gain-frequency response over another when the root-mean-square differences between the responses, averaged across frequency, is more than 3 dB.[880, 882] Several experiments have shown that the greater the departure of the hearing aid gain from the NAL-RP prescription, the more poorly the effectiveness of the hearing aid was rated by research subjects.[82, 90, 375]

Most prescription procedures attempt to amplify speech with typical input levels (around 60 to 65 dB SPL) to give a comfortable overall loudness. The overall gain that achieves this is probably not highly critical, at least for people with mild or moderate hearing loss for whom there is still a reasonable dynamic range between threshold and discomfort in which to position the amplified speech. Lower and higher input levels cannot, however, also be positioned in the middle of the dynamic range (or the patient would have no perception of changing overall loudness in different environments). For these lower and higher input levels there is therefore less tolerance for error: if the gain is too low, soft speech will not have audibility adequate to enable speech perception; but if the gain is too high, loud speech (or other loud signals) will be uncomfortably loud. Achieving the right loudness, across a wide range of input levels, is therefore a demanding requirement for prescription procedures. Getting it right the first time will reduce the number of visits that patients have to make for hearing aid adjustments.

This author's impression is that if measured performance falls within 5 dB, from 250 Hz to 4 kHz, of the target values prescribed by a formula whose validity is supported by good empirical evidence, there is no point in spending time in further fine tuning. Even deviations up to 10 dB might be fine for some patients, but there is a greater likelihood that a better combination of loudness, quality and intelligibility could be achieved by improving the match to the target. The type of discrepancy must be considered. A gain 6 dB above target at one frequency accompanied by a gain 2 dB above target at the remaining octave frequencies is inconsequential if the hearing aid has a volume control, because a reduction in the volume control setting could bring all gains to within 2 dB of target. By contrast, a gain 5 dB above target at one frequency accompanied by a gain 5 dB below target at a frequency one octave away would be worth improving. This represents a discrepancy in the response slope of 10 dB/octave, which should be avoided if possible.

The match to prescription that is necessary will certainly be affected by the advent of trainable hearing aids (Section 8.5). If patients guide the fine-tuning of their hearing aids to their preferred response via training algorithms within the hearing aids, there is no point spending time on carefully adjusting the hearing aids to match a target. All that is needed is that the hearing aid be sufficiently close to an optimal response that the patient can get there via the trainable algorithm, and does not have an initial experience so bad that the hearing aid is rejected. Although the automatic adjustment to target provided by manufacturers' software often has poor accuracy,[1650] it may be sufficient when the patient is going to train the hearing aid. Unfortunately, not all patients may have the cognitive and manipulative abilities needed to make good use of a trainable aid.[869]

10.4 Gain, Frequency Response, and Input-output Functions for Non-linear Amplification

Non-linear prescription can be viewed as specifying the gain-frequency response for several input levels. Typically, both the average gain and the shape of the frequency response will vary with input level. Alternatively, the prescription can be viewed as specifying an input-output (I-O) curve for several frequencies. It is necessary to specify the I-O curve for at least as many frequencies as there are channels in a multi-channel hearing aid (such as the three-channel aid shown in Figure 2.2). In principle, if enough frequencies and levels are used, all the information in a set of I-O curves at different frequencies is also contained within a set of gain-frequency responses at different input levels. Both diagrams are useful: the required compression ratios and thresholds are most easily read from a set of I-O curves, and the required filter characteristics are most easily read from a set of gain-frequency responses. Filters are used to form the individual channels in a multi-channel aid, and can also be used to shape the frequency response within each of these channels. Alternatively, a filter providing different amounts of gain at different frequencies and levels can shape the signal in much the same way, but without forming separate channels and recombining their outputs.

If a comprehensive set of gain-frequency curves or I-O curves has been prescribed, compression ratios and compression thresholds at each frequency have effectively also been specified. Such prescriptions do not, however, reveal the compressor response times. Prescription of compression speed is discussed in Section 6.5.3.

The reader will do well to have absorbed the issues raised in the discussion about linear responses: they all apply to non-linear amplification, and many further issues arise. Our knowledge about required linear amplification characteristics also provides information useful for assessing non-linear prescriptions, particularly for typical mid-level inputs.[217] There is, for example, reasonably close agreement between the NAL-NL2 prescription for mid-level sounds and the NAL-RP prescription, despite the two being derived in a different manner. In fact, it has been suggested that the NAL-RP selection procedure can be applied to non-linear hearing aids for 65 or 70 dB SPL input levels.[1827] At higher or lower input levels, the response varies from this depending on the compression rationale chosen by the clinician or hearing aid designer. Such rationales might include noise reduction, or loudness normalization, with the typically opposing consequences outlined in Section 6.4. There seems no need for this approach, however, given that there are several explicitly non-linear prescriptions available.

The following sections discuss the rationale behind each of several published prescription procedures. As we will see, all aim to give the hearing-impaired person a loudness sensation in some way related to what a normal-hearing person listening to the same sound would perceive, although details vary considerably.

LGOB

Triple bursts of half-octave bands of noise are presented at random frequencies and levels between threshold and discomfort. Testing is performed at the octave frequencies from 250 Hz to 4 kHz. Patients rate their loudness using the following scale:

7. Too loud
6. Very loud
5. Loud
4. OK
3. Soft
2. Very soft
1. Not audible

The stimuli continue to be presented until the same response is obtained twice for each level.

IHAFF, Contour, and VIOLA

The Contour test is the loudness scaling procedure used with the IHAFF loudness normalization prescription.[352] Pulsed warble tones are presented in an ascending sequence from 5 dB above threshold until the patient indicates that the stimulus is uncomfortably loud. At each level, patients indicate which of the following seven categories best describes the loudness:

7. Uncomfortably loud
6. Loud, but O.K.
5. Comfortable, but slightly loud
4. Comfortable
3. Comfortable, but slightly soft
2. Soft
1. Very soft

The results of three or four ascending sequences are averaged. It takes approximately 5 minutes per frequency, per ear, for the test to be carried out.[361]

A software program called VIOLA (Visual Input/Output Locator Algorithm) simplifies the task of calculating the input-output curve, based on the Contour test results. At each frequency, an input-output curve with two compression thresholds and two compression ratios can be drawn. This is useful when prescribing for hearing aids with compression ratios that either increase (curvilinear compression) or decrease (low level compression) as input level increases.

The principles of loudness normalization were introduced in Section 6.3.5. The first three of the following procedures require the patients to give subjective estimates of the loudness of different sounds; the remainder prescribe gain based on hearing thresholds alone.

10.4.1 LGOB

The idea of using non-linear amplification to restore normal loudness perception has been around for at least 35 years.[1859] The first clinically practical procedure to accomplish loudness normalization, however, was the ***Loudness Growth in half-Octave Bands*** (***LGOB***) procedure.[31, 1439] In this procedure, the hearing-impaired patient categorizes the loudness of narrow bands of noise using a seven-point loudness scale. The average levels corresponding to each loudness category are then compared to the levels needed to produce the same categories for normal-hearing people. For each input level, the gain needed to normalize loudness is deduced, as was shown in Figure 6.10.

10.4.2 IHAFF/Contour

During the mid 1990s, a group of clinicians and researchers noted that there was an urgent need for a practical procedure that could be applied to any hearing aid with adjustable wide dynamic range compression.[361, 1829] The group was called the ***Independent Hearing Aid Fitting Forum*** (***IHAFF***), and the prescription they devised used loudness scaling to normalize loudness at each frequency. The particular loudness scaling procedure used is called the Contour Test (see panel).

In the IHAFF/Contour protocol, the VIOLA software program presents the results of the loudness normalization as three points on an input-output function at each frequency at which the loudness scaling is carried out. These three points show the output levels needed to normalize the loudness of 1/3-octave bands of speech, when the complete speech signal is at the levels needed for normal-hearing people to rate its loudness as *soft*, *average*, and *loud*, respectively.[361, 1829] The actual speech levels adopted, and the shape of

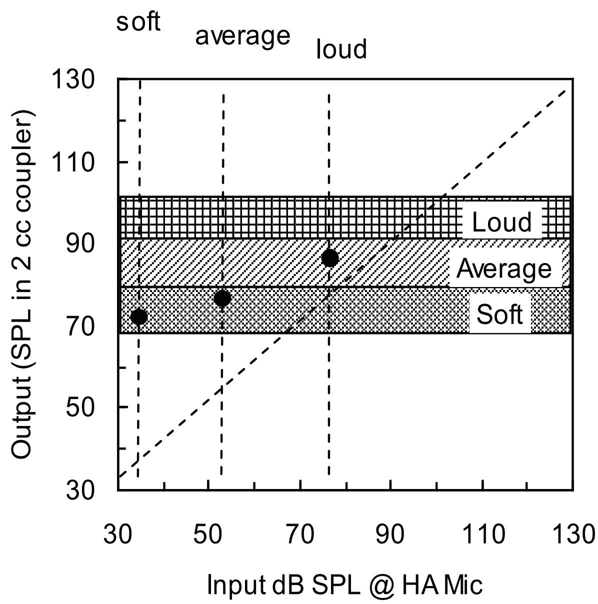

Figure 10.7 An example of the three-point I-O curve, for a frequency of 2 kHz, prescribed by the VIOLA software on the basis of the IHAFF procedure.

the speech spectrum assumed in the derivation of the procedure have no effect on the shape of the I-O curve prescribed, but rather determine which three points on the underlying continuous I-O curve are pinpointed as the targets. Figure 10.7 shows an example of an I-O target prescribed by VIOLA. To simplify adjustment of the hearing aid in a coupler, the output scale in this graph is expressed as SPL in a 2-cc coupler.

As with all loudness normalization procedures, it is not possible to read a compression threshold from the I-O target graph, because complete loudness normalization requires that compression be maintained down to input levels corresponding to the threshold of normal hearing. The IHAFF authors recommend, however, that the compression threshold be chosen so that soft speech just reaches the compression threshold. [1829]

10.4.3 ScalAdapt

For the loudness normalization procedures discussed in the two preceding sections, hearing aid prescription is a three-step process:

- the loudness scale for the patient is measured;
- at each level, the gain needed to normalize loudness is calculated; and
- the hearing aid is adjusted to match the gain target.

ScalAdapt is a clever one-step combination of these three steps.[906] The aid is pre-adjusted using an established threshold-based procedure. Loudness scaling, using an 11-point scale, is then performed while the patient is wearing the hearing aid. Instead of finding the loudness that corresponds to each input level, the clinician adaptively adjusts some characteristic of the hearing aid until the patient gives a desired loudness rating. This desired rating is the rating that would be given by a normal-hearing person listening unaided to an input of that level.

For instance, if a normal-hearing person would rate a sound of 60 dB SPL at a particular frequency as *comfortable,* then the gain of the hearing aid is adjusted until the hearing-impaired person also rates a 60 dB SPL sound at that frequency as *comfortable*. The hearing aid parameters are adjusted adaptively: if the loudness rating given is different from the target, gain or some compression parameter must be adjusted in the correct direction. The input levels used (and hence the target loudness categories), the order in which they are tested, and the amplification parameters adjusted, have to be appropriate to the filtering and compression characteristics that are adjustable on each aid. Otherwise, the adjustment made in one step may inadvertently undo the normalization for another input level that was achieved in a previous step. Kiessling et al. (1996) show how to apply the procedure to a three-channel hearing aid, but it should be possible to apply the concept to any hearing aid.

ScalAdapt

Double bursts of third-octave noise are presented at the center frequency of each channel of a multichannel hearing aid, while the patient is wearing the aid.

An appropriate parameter on the aid is adjusted adaptively until the patient gives two consecutive ratings that are within one category of the rating that an average normal-hearing listener would give for the same stimulus level.

The procedure is repeated for some combination of low, mid and high level stimuli that is appropriate to the adjustable parameters in the hearing aid being fitted.

Low frequencies are intentionally made softer than normal.

The procedure seems very efficient and direct: loudness measurements are concentrated around the loudness targets that are used in the prescription, and once the loudness scaling is finished, so is adjustment of the hearing aid. If loudness is then measured using a wide-band stimulus and it is found that loudness has not been normalized for this stimulus (because of loudness summation across bands), the hearing aid can be adjusted immediately using the same adaptive procedure. Adjustment of the hearing aid while the patient is wearing it and rating loudness means that there are no calibration difficulties arising from one transducer being used for loudness scaling and another for measurement of the aid response.

A problem with the procedure is that it may not be based on the correct rationale (see Section 10.4.8). Kiessling et al comment that in their experience loudness targets should depart from loudness normalization at the low frequencies. They say that complete loudness normalization creates excessive upward spread of masking, so they make low-frequency targets two loudness categories lower than those perceived by normal hearers.

10.4.4 FIG6

The ***FIG6*** procedure specifies how much gain is required to normalize loudness, at least for medium- and high-level input signals. Unlike the previous procedures, however, it is not based on individual measures of loudness. Rather, it uses loudness data averaged across a large number of people with similar degrees of threshold loss. This means that only hearing thresholds are needed to calculate the required gain.

FIG6 gets its name from Figure 6 of the article in which the underlying data were first outlined.[919] Gain is directly prescribed for each of the input levels 40, 65 and 95 dB SPL, and is inferred for other levels by interpolation.

For low level (40 dB SPL) input signals, the gain is prescribed on the basis that people with mild or moderate hearing loss should have aided thresholds 20 dB above normal hearing threshold. In most circumstances, it is not worth providing more gain than this, as background noise will prevent very soft sounds from being perceived, no matter how much gain is prescribed.[919] Except for the first 20 dB of hearing loss, every additional decibel of hearing threshold loss is therefore compensated by an extra decibel of gain. This rule is relaxed to a half-gain rule once the unaided threshold exceeds 60 dB HL because otherwise the high gains that result are likely to cause feedback oscillation.[913]

For typical (65 dB SPL) input signals, the amount of gain prescribed for any degree of threshold loss is equal to the average elevation of MCL for that threshold loss above MCL for normal hearing, using data published by Pascoe (1988). With this amount of insertion gain, narrow band sounds perceived as comfortable by a normal-hearing person will also be perceived as comfortable by the hearing-impaired aid wearer.

For high level (95 dB SPL) input signals, the gain is similarly prescribed to be equal to the boost in signal level needed to make sounds as loud for the hearing-impaired aid wearer as they are for the average normal-hearing listener, based on published average loudness data.[1070, 1097]

FIG6 formula

For 40 dB SPL input levels:

$IG_i = 0$	for $H_i < 20$ dB HL
$IG_i = H_i - 20$	for $20 \le H_i \le 60$ dB HL
$IG_i = 0.5*H_i + 10$	for $H_i > 60$ dB HL

For 65 dB SPL input levels:

$IG_i = 0$	for $H_i < 20$ dB HL
$IG_i = 0.6*(H_i - 20)$	for $20 \le H_i \le 60$ dB HL
$IG_i = 0.8*H_i - 23$	for $H_i > 60$ dB HL

For 95 dB SPL input levels:

$IG_i = 0$	for $H_i \le 40$ dB HL
$IG_i = 0.1*(H_i - 40)^{1.4}$	for $H_i > 40$ dB HL

Note that the data upon which these formulae were derived extended only to 80 dB HL, so application of the formulae to greater losses should be done with caution.

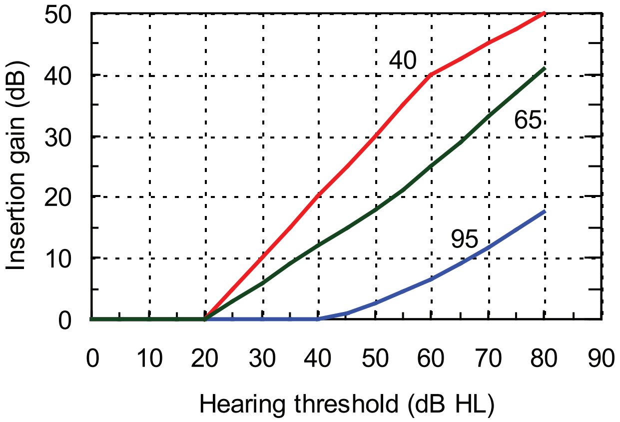

Figure 10.8 Insertion gain prescribed by the FIG6 method at any frequency as a function of hearing threshold, for each of the three input levels 40, 65, and 95 dB SPL.

Killion fitted a multi-line formula to the data referred to in the preceding paragraphs (see panel).[913] Because the gain needed to normalize loudness depends only on hearing threshold and input level, the same formula applies at all frequencies. The FIG6 procedure is easy to use, either with a calculator, or from a graph, as shown in Figure 10.8.

10.4.5 DSL[i/o] and DSLm[i/o]

The Desired Sensation Level (DSL) prescription for non-linear hearing aids has undergone several modifications in line with experience and new data since the first non-linear version appeared in 1995. The first non-linear version, called ***DSL[i/o]***, actually comprised two alternative procedures, each with its own underlying rationale.[327] One procedure was called DSL[i/o] linear, where *linear* means that the I-O curve is a straight line over a wide range of input levels. That is, the compression ratio is constant within the wide dynamic range compression region, as shown in Figure 10.9, and should not be confused with linear amplification. The DSL[i/o] linear procedure uses a compression ratio large enough to fit an extended dynamic range at each frequency into the dynamic range of the hearing-impaired person at the same frequency. This extended dynamic range is equal to the range from a normal-hearing person's threshold up to the hearing-impaired person's uncomfortable level. It thus prescribes a compression ratio greater than that required to normalize loudness.[k] Figure 10.9 shows the basic assumptions behind the DSL[i/o] linear formula. The upper limit of comfort for the hearing-impaired person can be estimated on the basis of threshold (using the Real-Ear Saturated Response recommendations from the DSL 3.1 method)[1600] or it can be individually measured for each patient.

The other original DSL procedure for non-linear hearing aids is more conventional in that it *is* aimed at normalizing loudness. This procedure is called DSL[i/o] curvilinear, because the I-O functions prescribed can be curved lines within the compression region. With this procedure, sounds at normal hearing threshold are amplified to the level of the hearing-impaired threshold, and sounds at the normal-hearing discomfort level are amplified to the hearing-impaired discom-

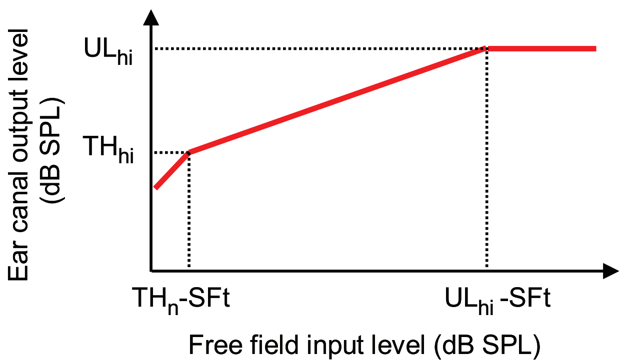

Figure 10.9 The DSL[i/o] method, showing which input levels are mapped to which output levels, using the terminology from Cornelisse et al. (1995). UL stands for upper level of comfortable listening, and TH stands for threshold, where both are expressed in dB SPL in the ear canal. The subscripts n and hi stand for normal and hearing impaired respectively. SFt is the sound field transform from free field SPL to ear canal SPL for the unaided ear for the frequency in question, and is synonymous with REUG.

[k] For people with a mild or moderate loss, average discomfort level is only slightly larger than the normal discomfort level, so the extended dynamic range is only slightly larger than the normal dynamic range. For people with a severe or profound loss, the difference is greater (see Figure 10.1 or Figure 10.15). The DSL 4.0 software allows the user to choose whether the prescription is based on mapping a normal or extended dynamic range into the impaired person's range.

fort level, or to an estimate of it. In between, however, the shape of the I-O curve depends on the rate at which loudness grows for the normal-hearing person relative to the rate at which it grows for the hearing-impaired person. These rates are characterized by the exponent to which stimulus level is raised in the equation that relates loudness category to stimulus level.

Both versions were implemented in DSL software version 4. Over time, the curvilinear procedure became the dominant method used, but because the DSL procedures were particularly aimed at children, it was not practical to measure individual loudness growth functions, so the exponent was fixed, and the prescription applied on the basis of threshold values alone.

DSL[i/o] was subsequently revised to a later version, ***DSLm[i/o]***, where the *m* stands for multi-stage, comprising: limiting at the highest levels, WDRC across the mid levels, linear amplification below compression threshold, and (optionally) expansion for very low input levels. DSLm[i/o] differs from DSL[i/o] in the following ways:[1583]

- Compression threshold is moved up from normal hearing threshold to considerably higher values, the narrow band levels of which increase from 30 to 70 dB SPL as hearing loss increases from mild to profound.
- The gain is reduced when there is low-level noise by further increasing the compression threshold.
- A maximum output that takes into account loudness summation across frequency and the crest factor of speech is specified for broadband signals like speech.
- A gain increase to allow for conductive hearing loss is added. The increase is achieved by increasing the predicted discomfort level by 25% of the conductive portion of the loss averaged across frequency (see Section 10.5). The gain increase is therefore greatest for high input levels, and zero for low input levels, compared to a sensorineural loss with the same air conduction thresholds.
- The gain prescribed for adults is reduced by 7 dB for mid-level inputs (but by less at higher input levels and by more at lower level inputs), hence also resulting in a lower compression ratio for adults than for children. This change is based on evidence that hearing-impaired adults have lower preferred listening levels than hearing-impaired children[1583] and prefer less gain than that prescribed by DSL[i/o].[26, 1069, 1220, 1385] Disconcertingly, research with normal-hearing people indicates that normal-hearing adults prefer listening levels *higher* than those preferred by normal-hearing children.[1247] The preferences expressed by hearing-impaired adults and children are likely driven by what they need to hear clearly, rather than just by the loudness they prefer when intelligibility is not an issue.
- The gain for binaural fittings is reduced by 3 dB.[1]

In summary, DSLm[i/o] is designed to normalize loudness at each frequency, except for:

- high input levels, where limiting prevents loudness discomfort;
- low input levels, on the grounds that the input is likely not speech; and
- adults, who prefer less gain than that predicted by the loudness normalization approach used.

Figure 10.10 shows an input output function for DSLm[i/o] compared to DSL[i/o] for a person with 50 dB loss at 1 kHz. DSLm[i/o] is incorporated within the fitting software of most manufacturers.

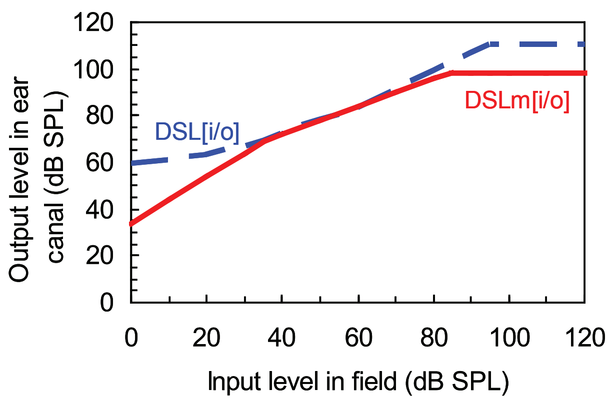

Figure 10.10 The I-O curve for a 50 dB HL hearing loss for the DSL[i/o] and DSLm[i/o] procedures, based on Scollie et al. (2005).

[1] In a subsequent software release, v5.0a, the binaural correction was removed for children's prescriptions but retained for adults.

10.4.6 NAL-NL1 and NAL-NL2

Unlike the preceding methods, the ***NAL-NL1*** (non-linear, version 1) and its revision (***NAL-NL2***) do not attempt to restore normal loudness at each frequency. The underlying rationale is to maximize speech intelligibility, subject to the overall loudness of speech at any level being no more than that perceived by a normal-hearing person.[435] To derive the gain-frequency response that achieves this at each input level, two theoretical models were used.

The first model was a modification of the SII method in which allowance was made for the effects of hearing loss desensitization, and for the effects of listening at high SPLs. Essentially, hearing loss not only decreases audibility, but also decreases the person's ability to recover useful information, even when the speech is made audible, as discussed in Section 10.3.4.

The second model was a method for calculating loudness, again allowing for the effects of sensorineural hearing loss.[1227, 1229] The only inputs required by both of these models are hearing thresholds, and the speech spectrum levels input to the ear after amplification.

For speech input at any level, gain at each frequency was systematically varied within a high-speed computer until the calculated speech intelligibility was maximized, but without the calculated loudness exceeding the loudness calculated for normal-hearing people listening to speech at the same level, as illustrated in Figure 10.11. This was repeated for hundreds of audiograms covering a wide range of degrees and configurations of sensorineural loss, and the optimized gains for each audiogram, for each input level, were found. Because this was a very time consuming process, even for a single audiogram at a single input level, an equation was fitted to the complete set of optimized gains. In the case of NAL-NL2 this equation was deduced by applying a neural network to the set of audiograms, input levels, and the optimized gains they produced. This equation thus summarizes all the optimizations and can be applied to any audiogram. It is available as part of a computer program

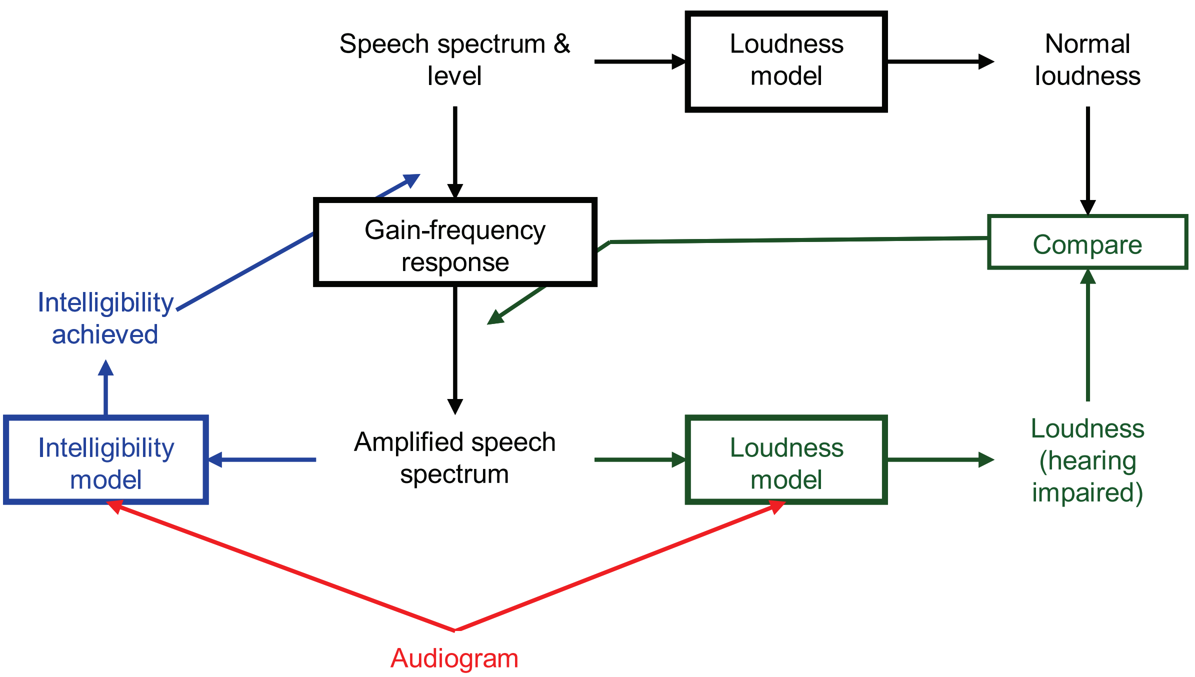

Figure 10.11 The NAL-NL derivation process. The loop on the left (shown in blue) varies the gain-frequency response to maximize speech intelligibility, but subject to the overall loudness not exceeding that for a normal-hearing person listening to the same speech, as calculated by the loop on the right (shown in green). We can think of this as the left loop turning up the gain one frequency at a time, but the right loop turning down the volume control whenever the overall loudness is greater than normal. Both the intelligibility model and the loudness model are adjusted to match the audiogram for which the gain-frequency response is being calculated.

NAL-NL2

The NAL-NL2 method is based on a complex equation that specifies insertion gain at each standard 1/3-octave frequency from 125 Hz to 8000 Hz. At each frequency, the gain depends not only on the hearing loss at that frequency, but also on the loss at all other frequencies.

Alternatively, the aid can be prescribed in terms of real-ear aided gain (REAG). REAG is deduced from insertion gain by adding the adult average REUG to the insertion gain target (Section 4.4)

The prescription can also be expressed as an I-O curve at any frequency, or as a coupler gain-frequency response. Because these are often measured with pure tones, the NAL non-linear software makes allowances for the crest factor and bandwidth differences between pure tones and speech signals. The prescription for pure tones, but not the prescription for broadband signals, therefore depends on the number of channels within the hearing aid.

In the case of coupler gain prescriptions, the acoustic effects of different venting configurations (including open fits) and tubing styles are allowed for. For multi-channel hearing aids, the NAL-nonlinear software also recommends crossover frequencies, compression thresholds, compression ratios, and gains for 50, 65 and 80 dB SPL input levels.

called NAL-nonlinear, and is included in the fitting software provided by the manufacturers of hearing aids and real-ear gain analyzers.

Although the principle guiding the derivation is one of maximizing intelligibility, for most hearing losses, all mid-frequency third-octave bands of speech turn out to have approximately equal loudness. As input level increases, the range of frequencies that are amplified to equal loudness increases. Equal loudness was the critical *assumption* behind the earlier NAL linear procedures, but in the NAL non-linear procedures, was a *consequence* of maximizing speech intelligibility.

Amplification requirements for people with mixed losses are worked out by applying the procedure to the *sensorineural* part of the loss (i.e. the bone conduction thresholds) and then adding gain equal to 75% of the conductive part of the loss (i.e. the air-bone gap).

NAL-NL2 differs from NAL-NL1 in a number of ways:

- A more recent loudness model is used.[1229]
- More extensive data on the ability of hearing-impaired people to extract information from audible speech is used to modify the SII model (Section 10.3.4).[292]
- Males are prescribed a slightly higher gain than females.[872]
- Experienced hearing aid wearers are prescribed a higher gain than new hearing aid wearers. The difference increases with the degree of hearing loss.[883]
- The optimized gains at high and low levels are adjusted so that excessively high compression ratios are not prescribed, particularly for people with upper severe and profound hearing losses, and particularly for fast-acting compression.[410, 878]
- The gain for children is adjusted to be 5 dB higher than for adults at mid input levels, and by more than this at low input levels and less at high input levels (Section 16.4.1).
- Binaural fittings are prescribed less gain than unilateral fittings. As with NAL-NL1 the difference in gain increases with input level, but the corrections in NAL-NL2 are smaller (Section 15.8).
- The derivation was carried out separately for non-tonal languages (like English) and tonal languages (like Mandarin). Tonal languages carry a bigger proportion of speech information in the lower frequencies, where pitch cues reside, so the optimal gain-frequency responses have slightly more gain for low frequencies relative to the high frequencies.

10.4.7 CAMREST, CAMEQ and CAMEQ2-HF

The same loudness models used to derive the NAL-NL1 and NAL-NL2 prescriptions have been applied by the inventors of the loudness models to a succession of prescription formulae. The first, known as the Cambridge formula, was applicable to linear hearing aids. It calculated the gain-frequency response needed to amplify 65 dB SPL speech so that the loudness of the signal components falling within each auditory filter band (referred to as ***specific loudness***) would be the same at all frequencies from 500 to 4000 Hz.[1228] This is essentially the same rationale as for the NAL and NAL-R prescriptions,[224] but was applied in a more sophisticated manner. This rationale has been shown to provide a reasonable basis for prescriptions, at least for mild and moderate hearing losses.[215] The resulting Cambridge formula was almost indistinguishable from the first (1976) NAL formula.[236]

This same approach was later applied to different input levels to produce a prescription for multi-channel non-linear hearing aids. An additional constraint was that the total loudness should equal the total loudness that would be perceived by normal hearers for speech at the same input level.[1221, 1232] The resulting formula was known as ***CAMEQ***, for ***Cambridge loudness equalization***. Different versions of the formula, with differing rationale for very low and very high input levels, are available depending on whether the hearing aid has low or high compression thresholds.

Changing the rationale to one of normalizing the pattern of specific loudness (i.e. the same loudness as normal at all frequencies) for speech at 65 and 85 dB SPL produced a further Cambridge formula, known as ***CAMREST*** (***Cambridge restoration of loudness***).[1214]

Finally, the same principles underlying CAMEQ were used to produce a later version that provided prescriptions up to 10 kHz.[1234] The formula, known as ***CAMEQ2-HF*** also provides a flat specific loudness pattern from 500 Hz to 4 kHz. Unlike the earlier version, however, it also provides a controlled amount of audibility from 6 kHz to 10 kHz. Specifically, the rms level of speech is amplified to the patients hearing threshold, unless the resulting partial audibility causes greater specific loudness than a normal hearer would experience. The mouthful CAMEQ2-HF was later simplified to ***CAM2***, and a gain reduction for new users was incorporated. It is available as a standalone computer program.

10.4.8 Comparison of procedures

We can group the non-linear prescriptions discussed in the preceding sections into four broad categories:

Loudness normalization achieved by loudness scaling: LGOB, IHAFF/Contour, and ScalAdapt all attempt to normalize loudness for narrowband sounds, one frequency at a time, and achieve this by directly eliciting loudness judgments from the patient.

Loudness normalization based on hearing thresholds: FIG6, DSL[i/o], DSLm[i/o], and CAMREST attempt to normalize loudness (over a wide range of input levels), and predict from hearing threshold the amount of gain needed to accomplish this.

Loudness equalization: CAMEQ and CAM2 attempt to equalize the loudness of speech across frequency, and limit the total loudness to be approximately that perceived by a normal hearer listening to the same sound. They predict from hearing threshold the amount of gain needed to accomplish this.

Intelligibility maximization: NAL-NL1 and NAL-NL2 attempt to maximize speech intelligibility, and limit total loudness to no greater than that perceived by a normal hearer listening to the same sound. The resulting loudness patterns often approximate loudness equalization, except at very low and very high frequencies, or when loss in a frequency region is severe.

The gain-frequency response needed to restore normal loudness typically has more gain at 500 Hz than those aiming to equalize loudness or maximize intelligibility. More 500 Hz gain is needed because normal hearers perceive the 500 Hz region as being louder than any lower or higher frequency regions.

Differences in prescriptions

Differences between the responses prescribed by different methods are evident for many hearing losses. As a first example, Figure 10.12 shows the insertion gains that would be prescribed by four non-linear procedures for a flat 40 dB hearing loss. The procedure based on loudness normalization with no regard to the input signal spectrum (FIG6) prescribes a completely flat insertion gain at all levels, corresponding to the flat hearing loss. DSLm[i/o] also aims to normalize loudness but takes into account the low-frequency emphasis of speech, and so prescribes a slightly rising response shape. CAM2, aimed at equalization of loudness has (surprisingly) a similar shape but higher

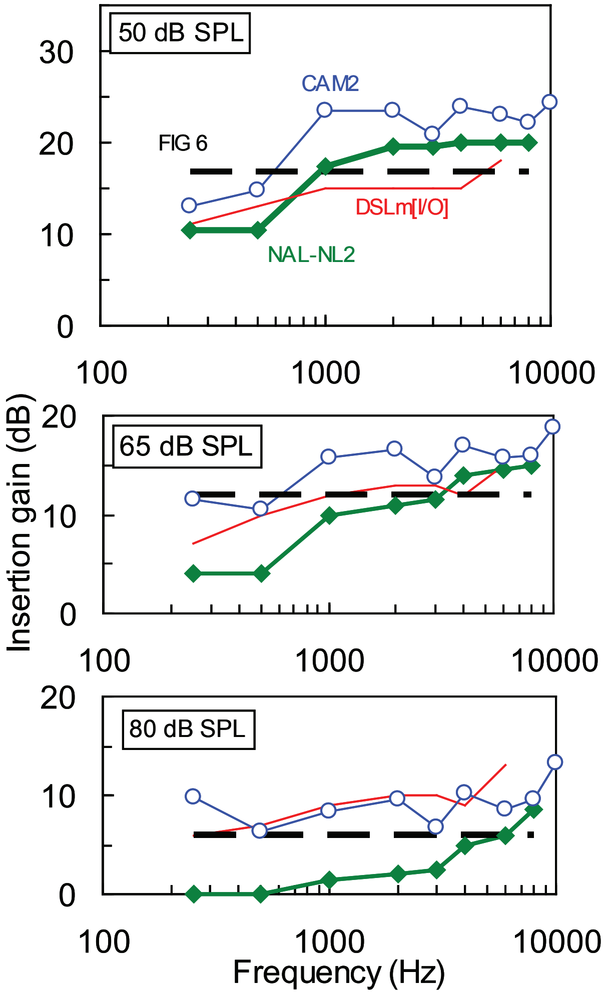

Figure 10.12 Insertion gain at input levels of 50, 65, and 80 dB SPL prescribed for an adult with a flat 40 dB hearing loss by four prescription procedures. A broadband signal with spectrum shape equal to the long-term spectrum of speech has been assumed.

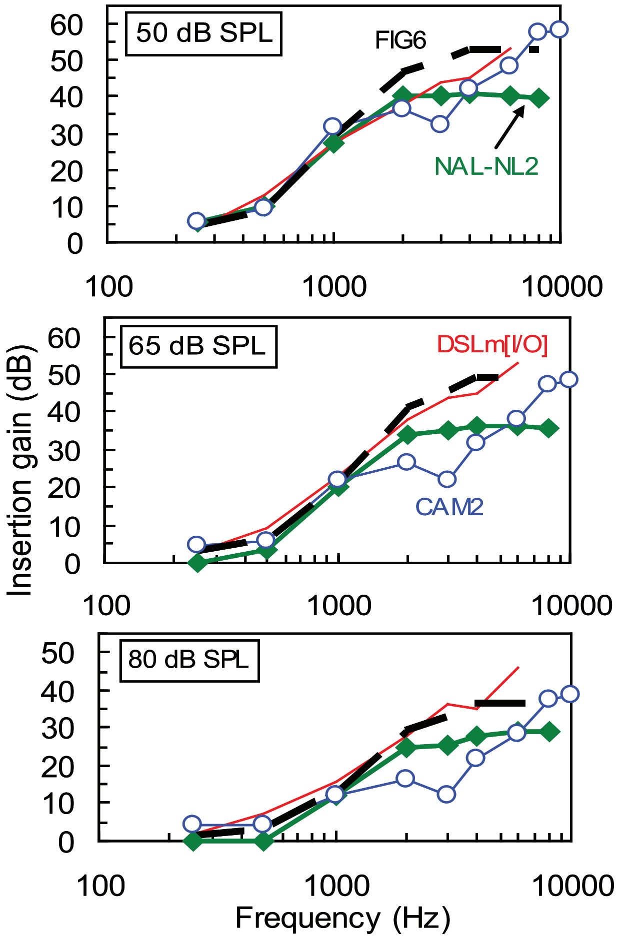

Figure 10.13 Same as Figure 10.12, except it is for the sloping hearing loss shown in Figure 10.4.

average gain. NAL-NL2, aimed at intelligibility maximization, prescribes the least gain for low-frequency signals, because of both the higher levels and the lower importance to intelligibility of low-frequency speech. Although a prescription for ScalAdapt could not be shown (as it is based on loudness judgments), we would expect greater similarity between NAL-NL1 and ScalAdapt, because ScalAdapt aims to make the loudness less than normal at 250 and 500 Hz.

Figure 10.13 shows a second example – the insertion gains prescribed for a steeply sloping loss. Across frequencies and levels, the prescriptions differ by amounts from 2 to 23 dB. CAM2 and NAL-NL2, which both rely on the same loudness model, provide less high-frequency audibility than the other two procedures, particularly at mid levels, because of the large contribution to loudness that even a small sensation level makes when there is a severe hearing loss, as in the high frequencies in this audiogram. In the case of NAL-NL2 the reduced high-frequency gain arises from the decreased ability of the ear to extract information at those frequencies. The amount of gain provided by NAL-NL2 is insufficient to restore normal loudness at these frequencies. The loudness contribution from this region is therefore decreased, so loudness can perhaps be increased more usefully at some other frequency.[m] For low input levels, NAL-NL2 may not attempt to achieve any audibility at the frequencies where loss is greatest and/or where speech has the least important cues to intelligibility, whereas CAM2 aims to preserve some audibility at all frequencies over a wide range of input levels.

Experimental comparisons and evaluations

Unfortunately, experimental evaluations of prescriptions typically are not available until several years after the prescriptions appear. While evaluations of NAL-NL1, DSL[i/o], and CAMEQ abound, evaluations of their successors NAL-NL2, DSLm[i/o] and CAM2 are scarce. Even the evaluations of the predecessor prescriptions are far from comprehensive. Most focus on whether the prescriptions produce an overall loudness that is too little, too much, or just right, often at different input levels, rather than the more difficult question of whether an alternative gain-frequency response shape would be better.

Various evaluations have shown that:

- NAL-NL1 prescribes an average gain, across frequency, that is a few dB more than adult subjects prefer.[872] (NAL-NL2 prescribes a lower overall gain than NAL-NL1).
- A prescription based on speech intelligibility maximization (NAL-NL1) is preferred to, and gives higher speech intelligibility in noise than, a prescription based on normalizing loudness (IHAFF).[880, 882]
- DSL[i/o] prescribes a higher average gain than NAL-NL1. Not surprisingly then, it prescribes more gain than is preferred by adult subjects.[26, 1069, 1220, 1385, 1699] (DSLm[i/o] prescribes a lower overall gain than DSL[i/o]).
- The range of input levels for which amplified sound remains within the comfortable range is greater for WDRC amplification fitted with DSL[i/o] than for linear amplification, and speech

Choosing a threshold-based or loudness-based procedure?

Arguments for threshold-based

- Fast;
- Usable with all patients;
- Loudness can be partially predicted from thresholds;
- No evidence that loudness (as opposed to audibility) is critically important, especially considering that the world has become louder since industrialization;
- Loudness normalization for narrow band test stimuli in a test booth may not achieve normal loudness for broadband stimuli in the real world;
- "Normal" loudness is ill defined, because it varies considerably across people and across measurement techniques.[512]

Arguments for loudness-based

- Individuals with the same audiogram can perceive different loudness for the same sound;
- Normal loudness is a worthwhile goal, as well as a means to achieving audibility and intelligibility;
- Accurate prescription of overall loudness is particularly important for hearing aids with no volume control.

---

m In hearing aid fitting, we can think of having a loudness budget. If we apply more gain than is needed at a frequency, we spend too much of our loudness budget at that frequency. This leaves less loudness (and hence audibility and intelligibility) available for all the other frequency regions. If we over-spend in total, the patient either turns the volume control down (and destroys what we have carefully provided at *every* frequency) or rejects the hearing aid as too loud if it does not have a volume control.

Complexities in truly normalizing the loudness of speech

Because hearing loss varies with frequency, loudness growth characteristics have to be measured with narrow-band signals. Differences between these test stimuli and speech can prevent loudness being normalized for speech, although it may be normalized for the test sounds.

- The differences in bandwidth complicate loudness normalization, unless the hearing-impaired person summates loudness across bandwidth in the same manner as does a normal-hearing person.
- Any difference in signal dynamics will create uncertainty in how to normalize. Should bands of speech and the test sounds be compared based on their rms levels, their maximum levels, or something else, and what effect will the compressor attack and release times have on this decision?

The magnitude of these bandwidth and dynamic factors is not well understood. As the need for normal loudness is unknown, the consequence of not achieving it for broadband speech stimuli is also unknown. If a fitting is based on individual loudness scaling, errors arising from differences in loudness between stimulus types can be minimized by a two-step approach, as carried out in the ScalAdapt procedure. After an initial scaling and adjustment using narrow-band stimuli, a final scaling and adjustment can be carried out at one or two levels using continuous discourse.

intelligibility is better preserved at low input levels.[773, 810,.811]

- CAMEQ prescribes about the overall gain preferred by adults.[26, 1220]
- CAM2 enables speech components above 5 kHz to be audible[577] and provides satisfactory ratings of loudness and sound quality in real life.[1222] Pleasantness is optimised with the amount of high-frequency gain prescribed by CAM2, or a little less.[1224]

Threshold versus supra-threshold measurements

A controversial issue is whether non-linear selection procedures should be based on hearing thresholds alone, or on supra-threshold loudness judgments. The FIG6, DSLm[i/o], NAL-NL2, and CAM2 procedures have the advantage of being quicker to use than procedures requiring loudness judgments as they require only hearing thresholds as the input data.[n] Loudness-based procedures are totally unsuitable for use with some elderly and very young patients. Plausible arguments for and against using supra-threshold loudness judgments to prescribe non-linear amplification can be advanced (see panel on page 317).

Apart from the issue of practicality, the decision about whether to use individual loudness scaling depends on the importance of individual differences in loudness perception. Although it is true that two people with the same hearing threshold may display different loudness growth curves, it is not apparent what the significance of this is. Two people with normal hearing can also have loudness growth curves that differ from each other. Perhaps if these two normal-hearing people were to obtain identical increases in threshold from identical cochlear damage, they would still perceive loudness differently from each other, just as they did before they had a hearing loss. If so, what would be the most appropriate "normal loudness" target for each of these people?[512]

Variations in loudness perception may partly be due to differences in how the instructions or loudness categories are interpreted, or to other random factors, rather than being totally due to fundamental differences in loudness perception. Comparisons of different scaling methods indicate that, even for a single person, the slope of the loudness growth curve varies greatly between scaling methods.[809, 906]

If loudness scaling is used, it should be as reliable and efficient as possible. Loudness scaling is certainly possible with many elderly patients, although one evaluation showed poorer test-retest reliability for subjects aged 60 to 79 years than for subjects aged 20-29 years.[903] There are at least three procedures that

[n] Discomfort levels are an optional input for DSL[i/o], and dead region measurement for CAM2.

use loudness scaling with a seven-point scale for the purposes of normalizing the loudness of narrow band stimuli (LGOB, IHAFF, and a proprietary method). The administration time and internal consistency of the loudness scaling part of these procedures have been quantitatively compared.[905] Whereas the IHAFF procedure required an average of 42 stimulus presentations per frequency, the LGOB procedure required an average of 18, and the proprietary method always required exactly 20. Based on the scatter of individual points around a straight line fitted to the loudness data, the LGOB procedure was less internally consistent than the other two. This decreased consistency is possibly because the LGOB procedure uses a randomized presentation of levels, whereas the IHAFF procedure uses an ascending test sequence and the proprietary procedure uses a partly randomized, partly ascending sequence. The proprietary method thus seemed to have best combination of consistency and efficiency.

Few hearing aids are, however, fitted using loudness scaling. This common choice to use threshold methods, probably made on the bases of convenience and time, is also consistent with the limited research evidence available. Wesselkamp et al (2001) showed no significant differences between normalization of loudness achieved through loudness scaling versus normalization of loudness achieved via a DSL[i/o] procedure. Keidser and Grant (2001) showed that intelligibility maximization, via NAL-NL1, produced significantly higher speech intelligibility, and was significantly preferred, relative to loudness normalization achieved via the IHAFF/Contour protocol.

10.5 Allowing for Conductive and Mixed Hearing Losses

Everything in this chapter so far has been applicable to sensorineural hearing loss. A conductive loss, or a conductive component in a mixed loss, comprises a frequency-dependent attenuation of sound in the middle ear. In pure conductive losses, hearing threshold, MCL, and LDL are all elevated by the same amount[1885] and this elevation equals the amount of attenuation occurring in the middle ear. In mixed losses, it seems reasonable to assume that the conductive component also causes all three quantities to increase by approximately the same amount. The size of the conductive component at each frequency is inferred from the size of the air-bone gap on the audiogram.

Given the above information, it might seem that to prescribe for someone with a conductive loss, insertion gain at each frequency should just equal the conductive loss at that frequency. This compensation would seem to result in a normal input to the cochlea, which itself is normal. Similarly, it might seem that a person with a mixed loss should be fitted by prescribing for the sensorineural portion of the loss, and then prescribing additional gain equal to the conductive portion.

Although these seem like reasonable deductions, they are probably not true, at least when implemented with current technology. It was long ago estimated that when a person with a mixed loss is fitted with a hearing aid, the average gain needed equals half the total loss, plus one quarter of the conductive component.[1095] A little arithmetic will show that this ***quarter-gain rule***

Otosclerotic hearing losses

Otosclerosis can affect bone-conduction thresholds because of stiffening or fixation of the stapes, even in the absence of sensorineural loss.[249, 792] To allow for this effect, bone conduction thresholds should be decreased by the amount shown in Table 10.1 before prescribing for either the sensorineural or the conductive parts of the loss.

Table 10.1 Corrections to be subtracted from bone conduction thresholds prior to calculation of the sensorineural and conductive portions of the loss.[249, 792] The values have been derived by averaging across studies. The 3 kHz figure has been interpolated.

	Correction (dB)					
Frequency	250	500	1000	2000	3000	4000
Correction	0	5	10	13	10	6

is equivalent to providing average gain equal to half the sensorineural loss plus *three quarters* of the conductive component. That is, the empirical observation is that compensation is needed for only 75% rather than 100% of the conductive component. There may be several reasons for this.

First, there is no point in providing additional gain if this gain causes the hearing aid to limit excessively, because limiting causes undesirable auditory effects (Section 10.7). Thus, the optimum gain will depend on the OSPL90 that can be achieved.[1884] If it is not possible to provide a high enough OSPL90 for conductive and mixed losses, then the optimal gain may also be less than theoretically expected. One way to express this is that because of device limitations, the dynamic range available to the listener has been decreased, even for a purely conductive loss.

Second, the acoustic reflex causes low-frequency sound entering the normal-hearing ear at high levels to be attenuated by the middle ear muscles. The acoustic reflex is usually absent in the case of conductive impairment.[1351] The hearing loss at low frequencies for high-level sounds is thus a little less than that for low-level sounds. Consequently, for low-frequency, high-level sounds, the gain required to provide a normal input to the cochlea is less than the elevation in hearing thresholds.

Third, it should not be assumed that normal is best. Normal-hearing people prefer other than a flat frequency response under some adverse listening conditions.[1467, 1886] Similarly, it is possible that they would prefer to hear some loud sounds at a lower than normal sensation level. It may be that even if a hearing aid has adequate OSPL90, people would prefer the gain compensating for the conductive component to be less than the attenuation caused by that conductive component, especially in noisy environments.

Although the discussion in this section has referred to the *proportion* of conductive loss that is compensated for with gain, the concept of a fixed proportion may not even be correct. It may be that the first 20 dB of loss can be ignored and the remainder fully compensated for. Alternatively, it may be appropriate to fully compensate for the conductive loss at low input levels (for which people presumably do not want to hear at a sensation level less than normal), but to decrease the compensation when the input level is high. In short, non-linear amplification might be appropriate for conductive losses, although the loss itself is essentially linear. If the dynamic range available to the aid wearer is less than normal (because OSPL90 is below discomfort), non-linear amplification may be just as appropriate for the person as it is for someone whose dynamic range has been decreased by a sensorineural hearing loss.

Given the uncertainty that still exists over how best to prescribe for conductive loss, it is fortunate that conductive components tend to be flat or gently sloping.[1883] The appropriate allowance for these losses is thus a gain increase of similar size at all frequencies, although not necessarily the same gain increase at all input levels. If the clinician provides the wrong amount of additional gain to compensate for the loss, the aid wearer can compensate by simply varying the volume control!

For the purposes of prescription, people with pure conductive losses can be considered to have mixed losses; the sensorineural part of their loss will just happen to equal zero. Note that, with many prescription procedures, the insertion gain provided for a flat loss of say, 10 dB is not a flat insertion gain. This is consistent with the observation that normal-hearing people do not necessarily prefer a gain of 0 dB at all frequencies.[1467, 1886]

Summary: Allowing for a conductive component when prescribing gain

- First, prescribe gain for the sensorineural part of the loss, using whichever procedure you select.
- Second, prescribe additional gain at each frequency equal to 75% of the conductive loss at that frequency. It is possible that for non-linear hearing aids, the additional gain prescribed for low-level signals should equal 100% of the conductive loss, and the additional gain for high-level signals should equal 50% of the conductive loss, but there are no research data on this issue.

10.6 Selecting Options for Multi-memory Hearing Aids

Most of this chapter so far is about selecting the gain-frequency response that best suits an individual. For some individuals, there is almost certainly no such single response. That is, in different circumstances, an individual might prefer different gain-frequency responses. Of course, non-linear hearing aids implicitly provide different gain-frequency responses for different input levels. Even beyond this adaptivity, however, the aid wearer might prefer different responses depending on the types of signals or noises in their immediate environment. Many hearing aids have multiple memories or programs (Section 3.3.2) that enable the hearing aid wearer to select the response that best suits each listening situation.

The need for multi-memory hearing aids has reduced over the last decade as hearing aids have taken more decisions on themselves. First, most hearing aids now have adaptive noise reduction (Section 8.1) that decreases the gain in any frequency region where the SNR is particularly poor. This practically eliminates the need to manually select a program that has reduced gain or different compression characteristics in a particular frequency region. Second, many hearing aids will automatically switch programs, or select/de-select processing features, based on their analysis of the environment they are in. This auto-switching will enable/disable features like directional microphones, adaptive noise reduction, telecoil input, feedback cancellation, and direct audio input. The "environment" of the hearing aid can include acoustic input (including the direction of sounds), wireless input (including from the other hearing aid), magnetic input, and direct electrical input.

No matter how smart hearing aids are, they can (so far) know only about external physical signals, not about the mood, goals, and attitude of the hearing aid wearer. That is, they know nothing about the ***listening criterion*** adopted at any time by the aid wearer. One use remaining for multi-memory hearing aids is to enable the aid wearer to accomplish different goals at different times – for example a program that provides a pleasant, not too loud, not too tiring sound for those occasions when all the aid wearer wishes to do is generally monitor the environment. This might contrast with a program that does everything possible to help the aid wearer catch every word in a difficult listening environment.

An under-researched potential application for multiple memories is adjusting the low- versus high-frequency emphasis depending on whether the aid wearer is in a position to speech-read (i.e. lip-read) the talker's face.[1873] The cues available from speech-reading primarily convey the place of articulation, which are mostly carried by the high-frequencies. This means that the low frequencies, which predominantly convey voicing and manner of articulation cues, become relatively more important when speech-reading is possible, than when the aid wearer is totally dependent on audition.[1873] The hearing aid cannot (yet) reliably know whether or not the talker is visible, so it is not feasible for the hearing aid to make this choice automatically.

Another use for multi-memory hearing aids is to help find the best single overall program for a patient, but we will discuss this application of these devices in Section 12.2.6.

Multi-memory hearing aids are also particularly valuable for people with a fluctuating hearing loss, such as those with Stage 2 Ménière's disease.[1825] Each of the two or more programs can be adjusted to match different degrees of hearing loss. A more extreme solution is to enable these patients to self-measure their hearing loss with a portable device and self-re-program their hearing aids whenever their thresholds change markedly.[1172] The self-fitting hearing aid (Section 8.5) would provide a much more convenient way to achieve this.

The most extensive multi-memory research has been carried out relative to a linear baseline response. A detailed review of these studies can be found in Keidser, Dillon & Byrne (1996). These studies helped provide the evidence base behind adaptive noise reduction, but now that it is widely available, the details are less relevant to clinical practice. The concept of a ***baseline response*** is still very relevant. The baseline response is individually selected for each hearing-impaired person, and will usually be included as one of the programs in the hearing aid. The alternative responses (whether manually or automatically selected) are then expressed as variations from the baseline response (e.g. a low-tone cut or selection of a signal processing feature). Two people with different hearing loss configurations will thus be prescribed different amplification characteristics in their baseline responses, even when they listen in the same acoustic environment, and with the same listening criterion.

10.6.1 Music programs

Music provides an obvious reason for a separate program. Music is characterized by a wide bandwidth, a long-term spectral shape different from that of speech and highly variable with time, a wide dynamic range (notably for classical music), very high rms levels and even higher peak levels, and a tonal quality that facilitates detection of small amounts of distortion. Musicians will invariably be exposed to high peak levels because of their close proximity to their instruments, but audiences are also commonly exposed to high sounds levels, with or without room amplification.

The high peak SPLs generated by musical instruments can easily overload the input stage of a hearing aid, producing marked non-linear distortion. It is common for hearing aids to overload when the input level exceeds about 95 dB SPL, but some may overload at even lower levels.[266] The resulting distortion for loud music with such a hearing aid causes a marked decline in the quality of music compared to a hearing aid that does not overload until the peak input level exceeds 105 dB SPL.[266] Unfortunately, there are currently no hearing aids that enable selection of a higher maximum input just by changing the program.[o] A procedure for testing the capability of a hearing aid to accept high input levels without distortion can be found in Chasin (2006).

A smooth gain-frequency response, free of marked peaks and troughs, is likely to be particularly beneficial for listening to music, as it would assist a musical instrument to maintain its identity as its pitch changes. It would be feasible to use a digital filter to provide this in a music program. The possible loss of average gain would not be a problem, given the high input levels characteristic of music.

Music programs probably should generally have the widest bandwidth that the hearing aid can provide, but extended high-frequency response for music perception appears not to be appropriate for people with steeply sloping hearing losses.[1506]

Several other amplification characteristics have been suggested for a music program, but there is as yet no research to validate these suggestions, and each may create its own disadvantages as well:[266]

- decreasing the amount of compression, either by increasing the compression threshold, or decreasing the compression ratio, or both;
- minimizing the variation of gain-frequency response with time, either by using a single processing channel, or by minimizing the variation of compression parameters across frequency;[p] and
- disabling adaptive noise reduction, and feedback cancellation algorithms.

There seems to be no problem in the music program keeping the same WDRC settings that are intended for speech in the normal program; indeed WDRC seems to be preferred over linear amplification (limited by either peak clipping or compression limiting) for listening to music.[397]

The following section outlines how the clinician can identify which patients are likely to benefit from hearing aids that have different programs in different environments.

10.6.2 Candidates for multi-memory hearing aids

Multi-memory hearing aids are not for everyone. Even before the availability of auto-switching hearing aids, estimates of the proportion of patients who will choose to use different programs in different listening conditions varied from 0% to 81%.[875] If a person wears hearing aids in only one situation (e.g. listening to television), it is most unlikely that a multi-memory hearing aid will be beneficial. If a person wears hearing aids in several situations, but the listening conditions are the same in all these situations, it is again unlikely that a multi-memory hearing aid will be beneficial. Patients who use their hearing aids in diverse listening conditions, and who are not happy with the performance of their hearing aids in at least some of those environments are most likely to appreciate and use multiple programs.[875, 1014]

There are two other less obvious issues related to candidacy for a multi-memory hearing aid. First, people

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[o] Most hearing aids are designed with an input overload limit around 95 dB SPL because doing so makes it technically easier to keep internal hearing aid noise from being audible.

[p] These first two suggestions overlook the intrinsic effect that a sloping hearing loss has on the apparent spectrum of the source as the input level or spectrum varies, even with a linear hearing aid.

with a high-frequency hearing loss (average of 2, 3 and 4 kHz) greater than 55 dB HL are more likely to use multiple programs than those who have less than 55 dB high-frequency loss.[874] The probable reasons for this are the restricted dynamic range and deteriorated frequency resolution of people with severe losses. When dynamic range is large, the sensation level in each frequency region is not critical. When the input spectrum varies in shape and level, the consequent variation in the output signal will have little effect on intelligibility or comfort. Conversely, for people with more loss, an inappropriate sensation level in any frequency region is likely to cause poor audibility, masking of one frequency region by another, or excessive loudness. Consequently, whenever the long-term input signal or noise spectrum varies, so too should the amplification characteristics. If an automatic hearing aid cannot adequately do this, there is a strong argument for using a multi-memory aid.

Second, people who have a target gain of close to 0 dB at 500 Hz in their baseline response are not good candidates for a multi-memory aid.[864] The reason for this may lie in the dual transmission paths of hearing aids worn by such people. Recall that a low-frequency gain of 0 dB can most easily be provided to people by simply allowing sound to enter the ear canal via the vent of the hearing aid (Section 5.3.1). When the vent provides the dominant transmission path, sound at low frequencies will not be affected by which program is selected (or even by whether the hearing aid is on or off). In such cases, the different programs can differ only in the high- and mid-frequency amplification and compression they provide. Altering the hearing aid's volume control thus alters amplification in a similar manner to selecting a new program. This similarity should decrease the advantage of multiple programs, as long as the hearing aid already has a volume control.

10.7 Prescribing OSPL90

As described in Section 4.1.4, the Output Sound Pressure Level for a 90 dB input level (OSPL90) is an estimate of the maximum SPL (measured in a coupler or ear simulator) that a hearing aid can put out. The equivalent term when measured in the real ear is the ***real-ear saturation response*** (***RESR***).

Surveys of satisfaction with hearing aids, or of reasons for return or non-use of hearing aids, elicit numerous complaints of loud sounds being too loud.[953] It is difficult to be sure whether these complaints refer to sounds of typical levels being too loud (excessive gain prescribed), to unwanted background sounds being too loud compared to wanted speech sounds (SNR loss by the patient), or to excessively high maximum output levels from the hearing aid, but presumably some are in the last category, and are therefore likely caused by excessively high OSPL90.

The OSPL90 may be quoted either at one particular frequency, at every frequency individually, at the frequency with the greatest OSPL90, or as the overall level of a broadband sound with a particular spectral shape, depending on the context. An inappropriately prescribed OSPL90 curve has more potential to make a hearing aid unusable than has an inappropriately prescribed gain curve, as we shall see in the following section. Despite this, there has been little research into the effectiveness of OSPL90 prescription procedures.

10.7.1 General principles: avoiding discomfort, damage and distortion

It is easy to describe a hearing aid with an appropriately prescribed OSPL90 curve:

- The hearing aid will never cause the aid wearer discomfort from excessive loudness. It is desirable for the hearing aid to sometimes make sounds loud, but if it makes sounds uncomfortably loud, especially for sudden sounds, the aid wearer will blame the hearing aid, and will be disinclined to wear it. Alternatively, the wearer may turn down the volume control, but this makes the aid less effective once the input sound decreases in level. When a normal-hearing person experiences loudness discomfort, it may be equally uncomfortable, but there is no prosthetic device to blame for the experience.
- The hearing aid will never create sounds intense enough to cause further damage to the residual hearing of the aid wearer. The possibility of damage is affected by more than OSPL90, so this issue is discussed separately in Section 10.8.
- OSPL90 will be no larger than is really *needed* by the aid wearer. If OSPL90 is larger than needed, the hearing aid probably could have been made with a smaller receiver, or battery, or both, without sacrificing battery life.

So far, these are all reasons for not making OSPL90 too high. If OSPL90 is too low, there will be several adverse consequences:

- Speech intelligibility may be decreased.
- Sound quality may be unacceptably poor, which is particularly likely if limiting occurs via peak clipping and the aid wearer has a mild or moderate hearing loss.
- The aid wearer will not be able to enjoy the full range of loudness sensations that normal-hearing people enjoy.
- The aid wearer may compensate for inadequate loudness or clarity by turning up the volume control. Unfortunately, this will further saturate the hearing aid. Consequently, loudness of the primary signal (e.g. speech) will not increase much, although loudness of lower level background noise during the gaps will increase. In the case of peak clipping, there will also be an increase in distortion.
- In extreme cases, the aid wearer will hear nothing within a frequency range if OSPL90 is less than threshold within this range.

In short, the optimum OSPL90 for a person must be low enough to avoid discomfort, damage, and wasted output, but must be high enough to avoid inadequate loudness, distortion, and removal of intensity cues to speech.

10.7.2 Type of limiting: compression or peak clipping

OSPL90 can be controlled by either compression limiting or peak clipping. As introduced in Sections 2.3.3 and 6.3.1, compression limiting generates less distortion than peak clipping (with distortion defined as the introduction of new, audible frequencies into the signal). Peak clipping is objectionable to people with normal hearing, most people with mild and moderate hearing loss, many people with severe loss, and some people with profound loss.[1700, 1734, 1769]

People with profound losses invariably have decreased frequency selectivity, so they are less able to detect the presence of distortion than are people with mild to severe loss. A peak-clipped waveform is at its extreme values for a greater proportion of the time than an unclipped signal, as shown in Figure 10.14. Consequently, its average power and hence its rms SPL will be greater. A peak clipping hearing aid can therefore always produce a greater OSPL90 than a compression limiting aid, for the same receiver and amplifier. When measured with pure tones, peak clipping aids can produce about 3 dB more output than compression limiting aids. Speech signals, however, have a higher crest factor, and the difference in OSPL90s increases to about 9 dB.[407] For some people with profound hearing loss, this increased OSPL90 more than compensates for the increased distortion in a peak clipper.

It has even been suggested that such distortion can be beneficial, in that intermodulation distortion will create low-frequency distortion products when a high-frequency sound enters the hearing aid. The aid wearer may thus be able to detect the presence of high-frequency sounds because of the low-frequency distortion they generate, much as would happen for a frequency lowering aid (Section 8.3). Evidence for this argument is scarce and conflicting, and it would seem better to use a frequency lowering scheme designed to provide this audibility if this is the aim.

10.7.3 OSPL90 prescription procedures

Although many OSPL90 selection procedures have been formulated, only one procedure (***NAL-SSPL***) has been systematically evaluated and shown to produce acceptable results.[1456, 1734] Consequently, that procedure will be described in some detail after a brief review of other procedures and some related issues. Most of the research has been carried out in the context of linear hearing aids. The implications

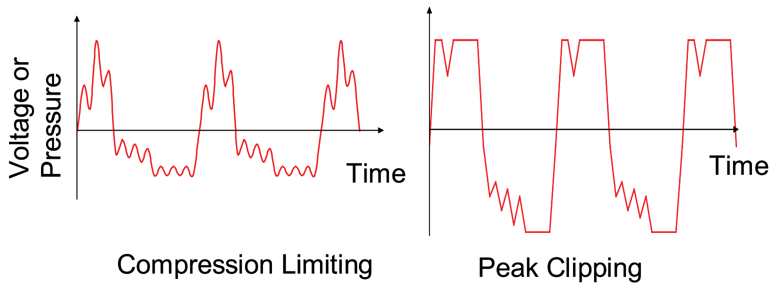

Figure 10.14 A speech waveform after passing through a peak clipper and a compression limiter, where both types of limiter can pass the same peak signal level without clipping.

Principles for prescribing compression limiting or peak clipping

- For people with mild or moderate loss do not use peak clipping. For a percentage of these patients it may not matter which you choose, but very, very, few will prefer peak clipping if they have had a chance to try both. Many, however, will prefer compression limiting.
- For people with severe loss, the choice is less critical; an increased proportion will not mind which you use, but again few will prefer peak clipping, so always prescribe compression limiting.
- For people with profound loss, use peak clipping if the patient has a history of preferring maximum volume control settings or complains that the aid does not make sounds loud enough. Otherwise, use compression limiting, although for most people, there may be little difference between the two options.
- For profoundly impaired children too young to indicate whether loudness is satisfactory, the choice is tricky. Such young children will not usually be fitted with the most powerful aids at their maximum OSPL90 setting. When OSPL90 is reduced below maximum, it seems most sensible to reduce it with compression limiting rather than peak clipping, in case the patient has residual hearing sufficiently good to use spectral cues in speech. (Peak clipping will degrade these cues more than multi-channel compression limiting, and much more than wideband compression limiting.)
- For any degree of loss, if the hearing aid has a form of compression that gradually reduces gain as input level rises above typical input levels (i.e. WDRC), the aid may rarely reach its output limit. If the compression ratio is high enough, the attack time is low enough, and the compression threshold is low enough, the type of limiting will not matter.

of how multi-channel channel compression affects the OSPL90 prescription will then be considered in Section 10.7.5.

Most procedures for selecting OSPL90 concentrate on avoiding discomfort. There are several procedures based on setting OSPL90 equal to, or just below, the ***loudness discomfort level*** (***LDL***) of the aid wearer.[q] The rationale behind these procedures is simply that OSPL90 set in this way should not lead to discomfort. Furthermore, setting OSPL90 as high as possible without causing discomfort should minimize the chances of OSPL90 being so low that it causes insufficient loudness or excessive saturation. The POGO procedure,[1163] for example, recommends that across frequency, the highest OSPL90 should be made equal to the average of the LDLs at 500, 1000, and 2000 Hz. To allow for the differences in calibration, POGO recommends that 2-cc coupler OSPL90 should be 4 dB below the three-frequency average LDL expressed in dB HL. Some experimental evidence (obtained with linear hearing aids) confirms that patients wearing hearing aids with OSPL90 exceeding their LDL values are more likely to complain of loudness discomfort than those for whom the OSPL90 is less than LDL.[445, 1284] By contrast, other experimental evidence suggests that there is a poor relationship between clinical measurement of LDL and complaints of loudness in real-world environments.[545, 866]

One problem with procedures based on individually measured LDL is that it is not straightforward to obtain a reliable and valid measure of LDL, although it is *possible* to achieve *reliability* similar to that of hearing threshold testing.[1258] The reliability of the values, and the mean values obtained after extensive testing, are both affected by the instructions given to patients and the psychophysical procedure used.[105, 609, 704] Hawkins et al (1987) considered that to get reliable results, patients should understand the purpose of the measurement, and that descriptive labels should be available above and below the target loudness. For some very old and very young patients, it may not be possible to measure LDL, although hearing threshold

[q] Terms synonymous with LDL are uncomfortable level (UL or UCL) or threshold of discomfort (TD). The term used does not matter, although whichever term is used, the instructions used to elicit it can greatly affect the value measured.

can be obtained. LDL increases following experience with hearing aids,[168, 1287] but it is *possible* that this is caused by increased familiarity with the clinical test rather than a change in the reaction to loud sounds in the real world.[228]

An alternative to measuring LDL is to predict it from threshold. A long-standing recommendation is that at each frequency, OSPL90 be set equal to 100 dB SPL plus a quarter of the hearing loss.[359, 1150] Again, differences in calibration are allowed for with suitable correction factors. Unfortunately, LDL cannot be predicted accurately from threshold. Several experiments have shown that although LDL, on average, increases as threshold increases, measured LDL may be up to 30 dB different from the predicted value.[105, 445, 454, 846, 1393, 1610] Of course, some of this apparent variability between people will be due to inaccuracies in the measurement of LDL, rather than to a real breakdown in the relationship between LDL and threshold.

A second difficulty with basing OSPL90 on LDL is that OSPL90 is expressed as dB SPL in a 2-cc coupler, whereas LDL is usually measured with headphones

Relating OSPL90 to LDL without using average correction factors

If LDL is measured as part of the prescription and fitting process, then there are four accurate ways to directly compare LDL and hearing aid maximum output:

1. Measure LDL with an insert transducer that is calibrated in a 2-cc coupler.[208, 358, 445, 695]
2. Measure the SPL of the LDL stimulus in the ear canal with a probe tube while LDL is being obtained.[700, 776, 1958]
3. Measure the individual's real ear to coupler difference (RECD) and use it to convert LDL, expressed as 2 cc coupler SPL, into LDL expressed as ear canal SPL.[1211]
4. Measure LDL using the hearing aid as the signal source, under the control of the fitting software.

In the first alternative, ER3A tube-phones can be used if an ITE/ITC/CIC aid is to be fitted. It is important that the tip of the insert phone be inserted to the same point in the ear canal as where the tip of the hearing aid will be located. This can be very difficult to judge if the aid is to be deeply seated. Because the SPL generated by the tube-phone will vary by 6 dB for every halving or doubling of effective ear canal volume, the errors with this approach should be acceptable for hearing aids that do not extend beyond about the second bend. For more deeply seated aids, the results should be viewed with caution.

For BTE fittings, the tube-phones can be connected to the tubing of the individual's earmold. This removes the problem of how far to insert the phones, and simultaneously allows for the tubing and venting characteristics of the individual earmold. If the tube-phones are calibrated in dB SPL in a HA2 2-cc coupler (complete with its 25 mm length of tubing), there are virtually no calibration errors involved in setting 2-cc OSPL90 equal to the measured LDL. (Of course, there may be considerable error in the LDL measurement itself.)

In the second alternative, SPL in the ear canal is monitored with a probe-tube microphone either before, during, or after the LDL measurement. Individual calibration errors largely disappear provided the Real-Ear Saturation Response (RESR) is also adjusted while ear canal SPL is being monitored.

The third alternative is really a combination of the previous two: LDL is measured with a transducer calibrated in a 2-cc coupler, and so is RECD, but the final result is expressed as ear canal SPL rather than coupler SPL.

The fourth alternative avoids all calibration issues as well as properly allowing for venting and tubing effects in the hearing aid.

All four approaches are equally accurate and individualized, but may differ in convenience and time efficiency, depending on the overall selection and verification strategy being used. The information in this panel should be taken as suggestions for how LDL *could* be compared to OSPL90, not that LDL *should* be measured.

calibrated in a 6-cc coupler. While appropriate average correction factors can be used, application of average corrections to individual people causes some error in the inferred value of LDL. There are several ways to obtain LDL without requiring an average correction factor to be used, as shown in the accompanying panel.

A third difficulty with basing OSPL90 on LDL is that OSPL90 has to be low enough to prevent discomfort for all possible sounds. Because loudness generally increases with stimulus bandwidth, broadband sounds may exceed LDL even if narrowband sounds lie below LDL at all frequencies.[r, 110, 1888] Even narrowband sounds when distorted by a saturated hearing aid (which effectively increases their bandwidth) can exceed LDL, despite an undistorted narrowband sound at the same frequency and SPL being below LDL.[561]

Finally, there is no logical reason why OSPL90 has to be as high as LDL. A few authors have suggested that we should think about a range of acceptable OSPL90s.[116, 1734] The maximum acceptable OSPL90 is equal to, or just below LDL. The minimum acceptable OSPL90 could be deduced either by assuming that people need sounds to be amplified at least 35 dB above their threshold[216] or by assuming that slightly loud speech should not cause the hearing aid to limit. The NAL-SSPL procedure adopts and quantifies this latter approach.[460] The minimum acceptable limit of OSPL90 has been assumed to be that which causes only a small amount of limiting when speech at a long-term rms level of 75 dB SPL is input into the hearing aid. To make this calculation, it was assumed that the hearing-impaired person uses the amount of gain predicted by the NAL-RP gain selection formula (which is appropriate for linear hearing aids, but leads to a higher OSPL90 than is needed for WDRC aids). The maximum acceptable OSPL90 was equated to LDL, which was estimated from hearing thresholds. The optimum OSPL90 for a person was assumed to lie midway between the two limits set by discomfort and saturation. In actual use, the mid-point is the only value used, so it is estimated directly from threshold (see panel).

As can be seen in Figure 10.15, there should be a wide range of acceptable OSPL90s for people with mild and moderate hearing loss. For people with severe and especially profound loss, however, the estimated maximum acceptable OSPL90 is less than the estimated minimum. That is, it may not be possible to have an OSPL90 setting that simultaneously avoids discomfort and saturation of the hearing aids, *at least for linear hearing aids*. Because both of the limits are only estimates, the procedure still places the optimum OSPL90 mid-way between the two.

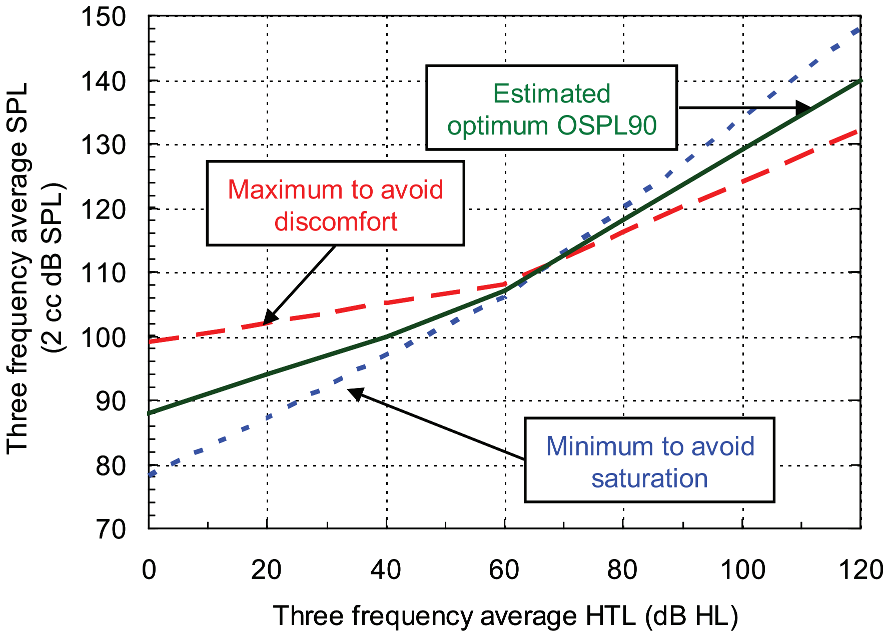

Figure 10.15 The NAL SSPL selection procedure, based on values midway between the OSPL90 needed to avoid discomfort and the OSPL90 needed to avoid excessive saturation.

r The degree of loudness summation across frequency decreases as hearing loss increases, so this is more of an issue for patients with mild and moderate loss than for patients with severe and profound loss.

The NAL-SSPL selection procedure

For hearing aids in which the shape of the OSPL90 curve cannot be controlled, three-frequency average OSPL90 (average of 500, 1000, and 2000 Hz thresholds) is prescribed on the basis of three-frequency average hearing loss using either Figure 10.15 or Table 10.2.

Table 10.2 The NAL-SSPL selection procedure, for prescription of three-frequency average OSPL90 in 2-cc coupler SPL.

3FA loss (dB HL)	3FA OSPL90	3FA (dB HL)	3FA OSPL90	3FA loss (dB HL)	3FA OSPL90	3FA loss (dB HL)	3FA OSPL90
0	89	30	98	60	107	90	123
5	90	35	99	65	109	95	126
10	92	40	101	70	112	100	128
15	93	45	102	75	115	105	131
20	95	50	104	80	118	110	134
25	96	55	105	85	120	115	136

These values can be translated to SPL in the ear canal by adding the RECD, averaged across people and across the three frequencies 500, 1000 and 2000 Hz. This average is 6 dB, so the NAL-SSPL procedure can be used as a real-ear selection procedure by adopting values 6 dB higher than those shown in Table 10.2 or Figure 10.15. These real-ear values apply no matter what type of hearing aid is used, or whether the person is an adult or an infant. Conversely, the 2-cc coupler SPL values are applicable only to BTE, ITE and ITC aids fitted to an ear of average adult size. A patient with a small residual ear canal volume will need less 2-cc OSPL90 than would an average adult fitted with an average length ITE or BTE aid, if they are both to receive the same real-ear target OSPL90. Methods for accomplishing this are given in Section 11.4.

An evaluation of the NAL-SSPL procedure using single-channel hearing aids showed that, on average, the procedure neither under-estimated nor over-estimated the OSPL90 found empirically to be best for the experimental subjects.[1734] For about 20% of the subjects, however, the OSPL90 prescribed by the procedure was outside the range of OSPL90 values found to be acceptable for each subject. A second evaluation, using two-channel hearing aids, also showed good agreement between the predicted OSPL90 and the OSPL90 empirically found to be optimal, although the prescription slightly underestimated the mid-point of the acceptable OSPL90 range in the high-frequency channel.[1456]

Both studies also carefully measured individual LDLs and evaluated whether knowledge of the individual LDL improved the precision with which the optimal OSPL90 could be predicted. Both studies came to the conclusion that if the threshold-based prescription was used, individual measurement of LDL did not significantly improve fitting accuracy. The clinical time saved is better utilized in evaluating the adequacy of the OSPL90 after fitting. Methods for evaluating OSPL90 are covered in Section 11.7.

As judged by the 60% of audiologists who measure LDL prior to aid fitting,[1147] this conclusion is not shared by all. If it were true that OSPL90 must be close to (and below) LDL, and that the wide range of LDL values measured for any degree of hearing loss is real (i.e. not due to different interpretation of instructions or other measurement error), then it would be reasonable to conclude that LDL must be measured on every patient prior to fitting the hearing aid. Evidence from the two studies reported above does not support this view on the importance of measuring LDL prior to aid fitting.

As we will see in Chapter 15, LDL for listening with two ears is lower than LDL for listening with one ear by some amount between 0 and 6 dB. As the amount is uncertain, and as the evaluations of the NAL-SSPL prescription have been performed with bilateral hearing aids, there is no bilateral correction within the prescription procedure, although arguably there should be a small correction.

10.7.4 Prescribing OSPL90 at different frequencies

While it has always been possible to design a hearing aid with an OSPL90-frequency response that could be varied independently of its gain-frequency response, this option has become more easily possible with multi-channel hearing aids.

Before prescribing OSPL90 for those aids where OSPL90 can be varied as a function of frequency, it is essential that the clinician identify whether the OSPL90 is being controlled independently within each channel of a multi-channel aid (e.g. Figure 10.16a), or whether the OSPL90 is controlled by a compressor or peak clipper that is operating on the whole bandwidth of the signal (e.g. Figure 10.16b). For hearing aids based on independent control in each of several channels, the effects of power and loudness summation must be allowed for.

Suppose, for example, that one channel of an aid was putting out a narrowband sound that by itself just failed to elicit LDL. What would happen if every channel simultaneously put out such a signal? First, the total SPL would be greater than the SPL of any channel by itself. Second, because the combined sound would have a bandwidth wider than any individual channel, the combined sound would be even louder than would be expected based on the increase in SPL. Consequently, the combined sound would easily elicit discomfort. The more channels there are, the greater the loudness summation. To compensate for this, the OSPL90 as measured by narrowband signals must be decreased relative to that needed for single channel limiting. Usually multi-channel aids will need this reduction in OSPL90 and single channel aids will not, but the hearing aid's block diagram should be examined carefully before deciding whether to make the reduction in OSPL90.

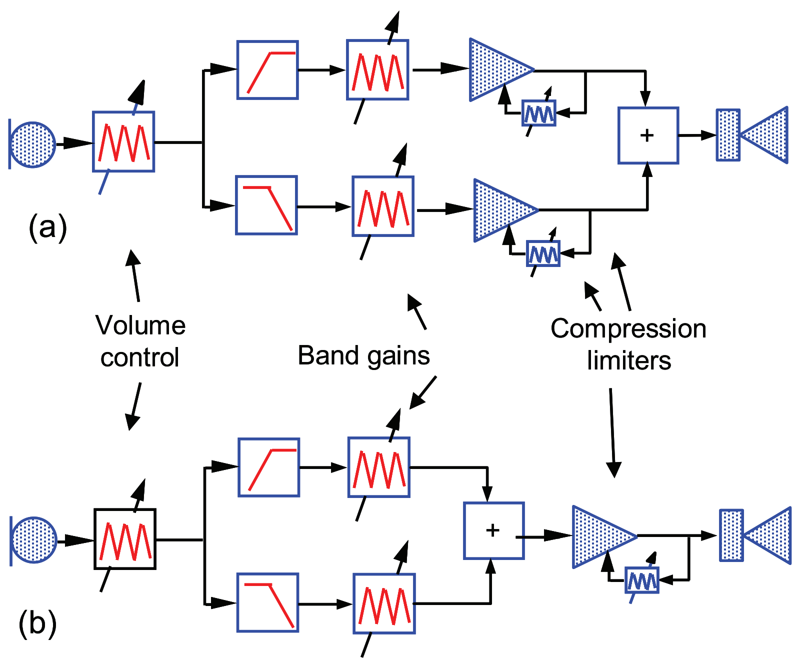

Figure 10.16 A multi-channel hearing aid in which limiting occurs (a) independently in each channel, and, (b) on the wideband signal after the channels have been recombined.

Theoretical background: Deriving a frequency-specific OSPL90 selection formula

First, discomfort levels are estimated from hearing threshold. Second, the minimum SPL at each frequency that is necessary to avoid saturation is estimated on the same basis as described in Dillon & Storey (1998). The only difference is that gains at each frequency are used rather than three-frequency average gain. Because the gain at each frequency depends on hearing threshold at other frequencies in the NAL-RP procedure, the gain corresponding to each degree of hearing loss was estimated using data from 700 audiograms.[1113]

These insertion gains were converted to real-ear aided gains, so that the resulting maximum output prescription would be in terms of the real-ear saturation response (RESR). The optimum RESR is then estimated to be the value mid-way between the minimum RESR to avoid saturation and the maximum RESR to avoid discomfort.

For hearing aids where the maximum output is limited independently in a number of channels, the RESR must be reduced for the reasons outlined in the text. The data of Bentler and Pavlovic (1989a), and of Bentler and Nelson (2001), indicate that to avoid discomfort with *n* equally loud sounds at different frequencies, their individual levels should be reduced by *4 + 13 log(n)*, relative to the LDL for any one sound presented in isolation. Similarly, the maximum output to avoid saturation need not be as great in each channel, because only a portion of the output power falls within each channel. If, after amplification, the power of a broadband sound was evenly distributed among *n* channels, the reduction in the minimum acceptable OSPL90 would therefore be *10 log(n)*.

Reduced RESR values for multi-channel hearing aids could therefore be calculated as the amount midway between the reduction in these maximum and minimum allowable values. However, speech does not continuously have a broad spectrum with power uniformly distributed across frequency. At any instant in time, loud speech is likely to drive only some of the hearing aid channels to their maximum values. The reduction in the optimum RESR, as shown in Table 10.4, is estimated at half this worse-case calculation.

Table 10.3 RESR (dB SPL) versus hearing threshold (dB HL). For hearing aids where the maximum output is limited independently in separate channels, the RESR (or OSPL90) should be decreased by the amount shown in Table 10.4.

HTL	250	500	1k	2k	4k
0	95	96	95	98	100
5	95	97	96	100	101
10	96	97	98	101	102
15	96	98	99	102	103
20	96	99	101	104	104
25	97	101	102	105	106
30	97	102	104	107	107
35	98	103	105	108	108
40	99	105	107	109	109
45	100	106	108	111	110
50	101	108	110	112	112
55	103	109	111	113	113
60	104	110	113	115	114
65	107	114	115	117	117
70	111	117	118	120	119
75	115	120	121	122	122
80	118	124	123	125	124
85	122	127	126	128	127
90	125	131	128	130	129
95	129	134	131	133	132
100	132	137	134	135	135
105	136	141	136	138	137
110	139	144	139	141	140
115	143	147	142	143	142
120	147	151	144	146	145

Table 10.4 Reduction that should be made to RESR values for multi-channel hearing aids with independent limiting in each channel.

Number of channels	Reduction (dB)
1	0
2	3
4	4
8	6
16	8

Prescribing frequency-specific maximum output

- At each frequency, look up from Table 10.3 the real-ear saturated response appropriate to the sensorineural part of the loss.
- If the hearing aid has independent limiting within each channel, reduce the values by the amounts shown in Table 10.4.
- For conductive or mixed losses, add 87.5% of the conductive part of the loss at each frequency to give the final RESR (see Section 10.7.6).
- To express the prescription in terms of 2-cc coupler SPL, subtract individual or average RECD values (Table 4.4) from the RESR values.

10.7.5 OSPL90 for non-linear hearing aids

With non-linear hearing aids (i.e. most hearing aids), gain decreases as input level increases, and accurate selection of OSPL90 becomes less critical than for linear aids. If the gain decreases sufficiently at high input levels, the hearing aid may never produce uncomfortable sounds for any input level that the aid wearer is likely to encounter. Killion (1995) argues that for people with mild or moderate hearing losses, limiting is not necessary at all provided the amount of gain is only that required to give normal perception of loudness. However, hearing aids cannot produce an unlimited output level, so *some* limit is necessary. People with normal hearing complain about loud sounds just as much as do hearing aid wearers wearing hearing aids without excessively high OSPL90.[871] The difference is that those with normal hearing will blame the sounds, whereas those wearing hearing aids may blame the hearing aids. Avoiding loudness discomfort, to the degree possible without compromising sound quality, therefore seems highly desirable.

It is fairly clear how non-linear gain would affect the OSPL90 that should be prescribed. With reference to Figure 10.15, non-linear gain will have no impact on the maximum to avoid discomfort, but the lower gain for high input levels will allow a lower minimum to avoid saturation. The estimated optimum OSPL90 will therefore decrease slightly. In the absence of a procedure specifically designed for non-linear hearing aids, a practical solution is to use the OSPL90 prescription for linear hearing aids outlined in the preceding sections.

10.7.6 OSPL90 for conductive and mixed losses

The general impact of conductive hearing loss on thresholds, discomfort, gain, and amplification requirements was discussed in Section 10.5. Suppose that for someone with a mixed loss, we have already prescribed gain and OSPL90 for the sensorineural component of the loss. The conductive part of the loss will affect the required OSPL90 in two ways. First,

Prescribing OSPL90 or RESR for conductive and mixed losses

1. If the person has otosclerosis, correct the bone conduction thresholds (see Table 10.1).
2. The sensorineural part of the loss is taken to be the bone conduction thresholds and the conductive loss is taken to be equal to the air-bone gap. It may be reasonable to smooth the air-bone gap across frequency, and it is often necessary to extrapolate the air-bone gap to lower and higher frequencies than can be measured.
3. Prescribe OSPL90 or RESR on the basis of the sensorineural part of the loss alone, using Figure 10.15 or Table 10.2 (three-frequency average OSPL90), or Table 10.3 (frequency-dependent RESR).
4. Increase OSPL90 and RESR by adding 0.875 times the conductive portion of the loss. (No, the procedure is not really that precise; adding 90 % of the air bone gap would be just fine.)

we will assume that additional gain, equal to 75% of the conductive loss, has been added at each frequency. This will increase by the same amount the minimum OSPL90 needed to avoid saturation. Second, discomfort level will increase by 100% of the conductive loss.

If we follow the rationale behind the NAL-SSPL selection procedure, the increase in optimum OSPL90 will be half way between the increase needed to avoid saturation and the increase needed to avoid discomfort. The required increase in OSPL90 to allow for the conductive loss is therefore equal to 87.5% of the conductive portion of the loss (i.e. the air-bone gap).

10.8 Excessive Amplification and Subsequent Hearing Loss

Hearing aids amplify sound. They therefore have the potential to cause a ***noise-induced hearing loss*** to someone who already has a hearing loss. Whether a hearing aid causes further loss depends on two factors.

First, a person's susceptibility to noise-induced loss partly depends on how much loss the person already has. A noise exposure that causes a certain permanent threshold shift to someone with normal-hearing will cause much less threshold shift to someone with a severe loss, for example. Essentially, people with hearing loss have already lost the most sensitive inner hair cells and their synapses and/or outer hair cell motors within the cochlea, and noise exposure has to be greater to damage the remaining inner and outer hair cells. Methods for calculating the degree of ***temporary threshold shift*** (***TTS***) and ***permanent threshold shift*** (***PTS***) that noise exposure causes to someone with normal hearing are well understood, at least in a statistical sense. The effect of prior hearing loss on subsequent damage can be calculated theoretically (see panel). For a known input level, the noise-induced loss caused by a hearing aid can be predicted as accurately as it can be measured.[1120]

The second factor affecting noise-induced loss is the ***daily noise dose*** experienced by the aid wearer. This dose depends on the levels at the output of the aid and the amount of time that these levels are maintained. Because the input level fluctuates with time, so too does the output level, and it is not obvious how to describe the output level as a single representative number. The mean value of the short term rms levels (each measured using the *fast* averaging time on a sound level meter) is believed to be the best way to represent a fluctuating level if one wishes to predict how much PTS or TTS will occur.[1120]

The output level of a hearing aid at any time depends on three things. First, the greater the gain, the greater will be the output level. Second, the greater the level of sound at the input to the aid, the greater will be the output level. Of course, both of these statements are true only when the output is less than the maximum output limit of the aid. Once the combination of input level and gain is sufficiently great to saturate the hearing aid, the output level is primarily determined by the OSPL90 of the hearing aid (or more precisely, by the RESR). For one group of school children studied by

Theoretical background: Predicting noise-induced loss for a hearing-impaired person

Empirical evidence has indicated that the hearing loss a hearing-impaired person will have after exposure to loud noise can be estimated from the sum of the loss experienced by a normal-hearing person, plus the pre-existing loss of the hearing-impaired person, when both losses are expressed as their equivalent excitation level in the cochlea. This transformation is referred to as the modified power law, and the steps are as follows:[780, 1115]

- the hearing-impaired person's initial loss is transformed to an equivalent internal excitation level;
- the noise-induced loss that a normal-hearing person would undergo is transformed to an internal excitation level;
- these excitation levels are added;
- this total excitation level is transformed back to an external sound level; and finally,
- this sound level represents the hearing threshold that the hearing-impaired person is likely to have after exposure to noise.

Macrae (1994b), the output of the hearing aid reached its maximum level so infrequently that *the noise dose was almost entirely determined by the combination of input level and gain, rather than by the OSPL90*. This finding is very important as it is often incorrectly assumed that the safety of a hearing aid is determined solely by its OSPL90. There is, however, some evidence that OSPL90 also affects safety, as reviewed in Macrae (1994a). This presumably happens only in those fittings where the maximum output of the aid is reached reasonably often.

PTS will grow towards a final value that is approximately equal to the asymptotic TTS.[s, 985, 1120] The rate at which PTS grows depends on the amount of noise exposure. First, if TTS exceeds a certain amount, referred to by Macrae (1994a) as the ***safety limit***, PTS will begin to accumulate rapidly, reaching its final value in less than 10 years.[21] For normal-hearing people, the safety limit is about 50 dB of TTS.[1118, 1894] The modified power law can be used to predict that this safety limit decreases dramatically as hearing loss increases, and is only 2 dB for a hearing threshold of 100 dB HL. Second, if TTS is much smaller than the safety limit, it may take many decades of noise exposure before PTS grows to its final amount.

TTS and PTS are real possibilities with hearing aid use. Using 15 dB more gain than that recommended by the NAL-RP procedure at 1 kHz, at least with a linear hearing aid, is enough to cause TTS of 3 dB (and hence probably the same amount of PTS) for anyone with initial hearing thresholds of 50 dB HL.[1119] This example assumes that the mean input level is 61 dB(A) SPL. If the mean input level is significantly higher than this, even a procedure as conservative in gain as the NAL-RP procedure can lead to TTS and PTS. Any TTS should be avoided if possible, even one as small as 3 dB. As well as being a precursor to PTS, TTS has immediate consequences. It will decrease the person's communication ability as soon as it occurs, because they will hear as if their hearing loss has been increased by this 3 dB whenever the TTS is present.[1120]

The risk and degree of hearing aid-induced loss increase as hearing loss increases, because people with more loss need more amplification. For hearing losses with a three-frequency average value (500, 1000, 2000 Hz) of less than 60 dB HL, hearing-aid-induced loss should not be a problem if gains similar to those recommended by the NAL-RP procedure are used.[1117] By contrast, once thresholds exceed

Practical steps: Avoiding hearing aid induced hearing loss

- Do not prescribe more gain or OSPL90 than is necessary for optimal intelligibility. This is particularly important for children too young to operate their volume control, or anyone fitted with a hearing aid that has no volume control.
- Advise the patient to avoid prolonged exposure to high noise levels.
- Prescribe a non-linear hearing aid in which the average gain decreases as input level rises from typical input levels to high input levels. (Gain may also vary as the input varies from low to typical input levels, but the aid's behaviour for low input levels is less likely to affect the likelihood of noise-induced loss.)
- Monitor hearing thresholds over time.
- Wherever doubt exists, check for temporary threshold shift by measuring hearing thresholds after 24 hours without a hearing aid in the test ear and then after 8 hours of hearing aid use. (For school children, measurements first thing on Monday morning and then late on Monday afternoon will be most convenient.) Where it is difficult to achieve enough sensation level without causing temporary threshold shift, consider advising the patient to alternate hearing aid use between the ears to allow the ears greater recovery time.

[s] The ***asymptotic TTS*** is the maximum amount of TTS that occurs when an ear is continuously exposed to noise. The amount of TTS grows exponentially with a time constant of about 2 to 3 hours.[1117, 1197] The TTS is therefore very close to its asymptotic value after about 6 hours of aid use.

about 100 dB HL, even the gains recommended by the NAL-RP procedure are likely to be unsafe.[1117] The result is a slow downward spiral of hearing, with each increase in hearing loss requiring an increase in gain, in turn causing increased noise exposure, and hence resulting in further hearing loss. The increments of hearing loss are gradual and small and take several years to develop, but for children and younger adults in particular, the concern is obvious. If a hearing aid is to provide a satisfactory sensation level for people with profound loss, however, it may be necessary to accept that some additional PTS will occur because of the hearing aid.[1116] Fortunately, this situation arises less often now as people with this degree of loss are likely to be candidates for a cochlear implant, at least as far as audiometric considerations are concerned (Section 9.2).

Finally, the safety calculations have all been performed with linear hearing aids, and the situation *should* be better with non-linear aids because their gain decreases as input level rises. Adaptive noise reduction, which decreases gain in noisy places, can also be regarded as a safety feature in hearing aids!

Given that one cannot be sure that a hearing aid will not exacerbate a hearing loss, it is important to determine whether such damage is occurring. TTS provides such a check. If hearing thresholds are measurably worse after a day's hearing aid use than after 24 hours without a hearing aid in the ear, TTS is occurring and PTS is very likely to follow unless something is done. Serial audiograms over several months or years can also be used to detect damage. Unfortunately, permanent damage must occur before it can be detected, and it is difficult to differentiate loss induced by the hearing aid from a loss that is progressing for some other reason. It is therefore better to detect excessive amplification by detecting TTS. If it is detected and corrected sufficiently quickly, permanent further elevation of hearing thresholds may be avoided, although IHC synapses may still be lost, potentially affecting the precision of supra-threshold discrimination abilities.

10.9 Concluding Comments

The impact of using an appropriate selection procedure should not be underestimated. When the first version of the NAL procedure was introduced to NAL hearing centers around Australia (replacing some vague combination of clinicians' judgment and evaluative procedures), the rate at which batteries were issued nationally increased by 51%.[1818, 1819] This increase in battery consumption was ascribed mostly to increased usage of hearing aids, because the number and type of hearing aids being issued remained unchanged. Some other changes to service delivery were also made, but these were considered to be less significant.

Some prescriptions in common use have not been mentioned in this chapter. Several manufacturers have a proprietary prescription procedure included in the fitting software for their own products. These procedures have not been reviewed in this chapter because the derivation, formula, and supporting evidence of these procedures are generally not available.

Depending on the prescription formula used, the clinician may need to look carefully at the prescription and compare it to the characteristics of the hearing loss. Most procedures prescribe increasing amounts of gain, without limit, at a particular frequency as hearing loss at that frequency increases. For someone with the audiogram (in both ears) shown by the diamonds in Figure 10.17, there would probably be no point in amplifying above about 2 kHz. For this person, amplified high-frequency sounds may contribute much to the overall loudness, but are unlikely to contribute significantly to speech intelligibility or quality. Furthermore, the large gain needed to achieve

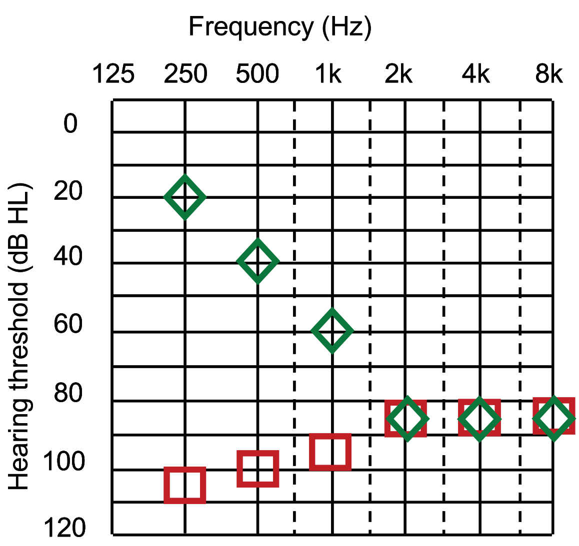

Figure 10.17 Two audiograms with identical losses from 2 to 8 kHz, but different upper frequency limits of aidable hearing.

audibility of speech above 2 kHz is likely to cause feedback oscillations, and so require a tighter earmold than would otherwise be needed. Where the gain prescribed is unlikely to result in any useful contributions to intelligibility at a particular frequency (for speech with typical input levels), the clinician may be well advised to further decrease the gain at those frequencies to minimize the chance of feedback oscillation.

For the person with the audiogram (in both ears) shown by the squares, however, it may be worth extending amplification out to 4 kHz or beyond, even though both people have identical thresholds at and above 2 kHz. This person will not be able to extract the full information present in speech in the lower frequency ranges, so the small additional amounts of information at 3 and 4 kHz may be worth having. The NAL-RP and NAL-NL2 procedures attempt to allow for these factors within their calculation formula, but other procedures do not.[t]

Similarly, some non-linear formulae may prescribe a compression ratio of 4 or much higher, even though in some cases the software implementing the formula gives a warning that such high compression ratios may not be optimal. Empirically, high compression ratios, especially for fast-acting multi-channel compression hearing aids, are associated with poorer speech intelligibility.[410, 1237, 1437] As mentioned in Chapter 6, fast-acting, multi-channel compression with high compression ratios will decrease the availability of spectral cues, and hence decrease speech intelligibility, even though it increases audibility.

The role of LDL testing as an input to hearing aid prescription is not completely clear. As we have seen, there is some evidence that the setting of OSPL90 relative to LDL is important, but that a prescription based on hearing thresholds predicts an OSPL90 within the range of acceptable OSPL90s for most patients, even for linear hearing aids. It also seems likely that the adoption of WDRC for all or most patients has made OSPL90 less critical than it once was. If so, the best use of clinical time is probably an evaluation of hearing aid maximum output after the prescribed hearing aid has been fit to the patient, rather than measurement of LDL prior to prescribing. There is room for further research on this topic.[866]

As one reflects on the prescription procedures described in this chapter, it is evident that they have been shaped almost exclusively by speech intelligibility and loudness considerations. There is much more to audition, such as localization, comfort, pleasantness and naturalness.[218] While it is becoming common to examine these aspects for different prescriptions, it has not yet been possible to quantitatively design these aspects into prescriptions. Another area that has received insufficient investigation is the effect that visual cues have on the optimum prescription. Lip-reading primarily conveys high-frequency cues, so the optimum emphasis for the hearing aid is likely to be more high-frequency weighted when speech reading is not possible than when it is.[1873]

Fortunately, the prescription is just the starting point for any new hearing aid fitting. In some (hopefully many) cases, it will also be the end point. Other cases will require fine tuning by the clinician. Increasingly, the patient will take responsibility for this via training (also called learning) algorithms in the hearing aid (Section 8.5).

Finally, new technology and new fitting procedures should continue to develop in tandem. Technological advances are of no use unless the resulting amplification characteristics can be appropriately matched to the needs of individual hearing-impaired people. With each new advance, it is necessary to ask whether the processing feature needs to be prescribed for each individual aid wearer, or whether it interacts only with the environment and can be either permanently enabled, or automatically enabled at appropriate times by the hearing aid.[901] The answer to this is not always clear. For example, the characteristics of adaptive noise reduction algorithms have so far largely *not* been individually prescribed. Most commonly, several degrees of gain reduction are made available, and clinicians can trial which of the available settings appears best for each client. This situation is analogous to the 1940s when each of a few different hearing aids would be experimentally compared. As discussed in Section 8.1.1, the optimal settings of adaptive noise reduction almost certainly depend on hearing thresholds relative to speech and noise levels, and should therefore be individually prescribed.

[t] For the two audiograms in Figure 10.17, NAL-NL2 does, in fact, recommend 6 to 9 dB greater amplification from 2 to 8 kHz for the audiogram shown with the squares than for the audiogram shown with the diamonds.

CHAPTER 11

SELECTING, ADJUSTING AND VERIFYING HEARING AIDS

Synopsis

The first decision to be made when a clinician and patient select a hearing aid is whether a CIC, ITC, ITE, BTE-RITE, BTE-RITA (with standard tubing and earmold, or thin tube and earmold or instant-fit dome), spectacle, body style, or a sub-variety of any of these, would be most suitable. For each style there are advantages relating to ease of insertion, ease of control manipulation, visibility, amount of gain, sensitivity to wind noise, directivity, reliability, telephone compatibility, ease of cleaning, avoidance of occlusion and feedback, ability to assess and fit in the same appointment, and cost. The weight given to each factor will vary greatly from patient to patient. The need for specific features, such as a volume control, a telecoil and switch, a direct audio input, and a directional microphone must be determined on an individual basis. These needs will also influence the style of hearing aid selected. BTEs have more advantages than the other styles for a majority of patients.

Next, signal-processing options appropriate to the needs of the patient must be selected. Compression limiting is a more appropriate form of limiting than peak clipping if it can provide a high enough maximum output. In addition to compression limiting, a low compression ratio, active over a wide range of input levels, is appropriate for most patients. This low-ratio compression will provide advantages whether it is single- or multichannel, and whether it is fast or slow acting. Multichannel compression will provide greater speech intelligibility and/or comfort for patients with moderately or steeply sloping hearing loss, and the multichannel structure facilitates other features such as adaptive noise suppression and feedback suppression. The comfort advantages of adaptive noise reduction are greatest for patients who wear their hearing aids in a range of environments and who also require amplification across a wide range of frequencies. These same considerations apply to multi-memory hearing aids, the only difference being that the patient, rather than the hearing aid, chooses the response variations. Feedback cancellation is most beneficial for patients with a severe or profound hearing loss, patients with a severe loss in the high frequencies but near-normal hearing in the low frequencies, any patients fit with open canal devices, and any patient who wishes to use the telephone without using telecoil input. This combination makes it useful for nearly every client. Frequency lowering is advantageous for some patients though it is not yet possible to predict which patients will benefit. Trainability enables patients to take responsibility for fine-tuning their hearing aids.

Hearing aid fitting software provides a first approximation to the prescribed gain-frequency response target. The software must appropriately allow for the acoustic configuration of the earmold shell or dome fitting. The approximation can be made even more accurate by incorporating the individual patient's real-ear to coupler difference (RECD) in the prescription. This increased accuracy in the pre-calculation is probably only worthwhile for hearing aids intended for infants, where measurement of the final real-ear gain is difficult. Any signal-processing scheme that requires adjustment for each patient must also be supported by an appropriate prescription method. Measurement of real-ear gain is necessary unless the hearing aid has been adjusted in the coupler using the individual's (or at least an age-appropriate) RECD.

Because it is not possible to prescribe OSPL90 with complete precision, the suitability of maximum output should be subjectively evaluated before the patient leaves the clinic. A variety of intense sounds should be presented to the patient to ensure that the hearing aid does not make sounds uncomfortably loud. Maximum output must, however, be great enough for the patient to experience intense sounds as being loud. This can be assessed by presenting speech signals at a high level and asking the patient to report their loudness.

This chapter uses much of the information in the preceding chapters to give step-by-step guidance on how to choose and adjust a hearing aid. First, we consider how to select a hearing aid style. Second, we consider how to select desired features. Finally, we discuss how to efficiently achieve the prescribed response. Selection of style and features is driven by consideration of the patient's needs and abilities. Needs can be assessed by informal questioning, but more systematic approaches may elicit a greater amount of useful information. Slightly more systematic techniques include the goal-setting component of the Client Oriented Scale of Improvement (COSI), Goal Attainment Scaling (GAS), Glasgow Hearing Aid Benefit Profile (GHABP) or the Patient Expectation Worksheet (PEW) – see panel in Section 9.1.6. More systematic again is the Hearing Demand, Ability and Need Profile, which determines the unmet need in a range of commonly experienced listening situations, including situations where an assistive listening device is likely to be beneficial.[1374]

However it is performed, some analysis of hearing needs is essential, whether the patient is a first-time wearer of hearing aids or an experienced wearer considering acquiring new hearing aids. Open-ended techniques that rely on patients volunteering places where they are having difficulty hearing and in which they would like to hear better seem more efficient and personal than techniques based on ratings of long lists of different listening situations, many of which may not be meaningful to the patient.

11.1 Selecting Hearing Aid Style: CIC, ITC, ITE, BTE, Spectacle Aid, or Body Aid

There are many factors to take into consideration when selecting hearing aid style. The relative advantages of the different styles are summarized in Table 11.1. Although spectacle and body aids are included in this review, these styles are now rarely used. In those few cases where spectacle aids are used, they are mostly implemented by attaching a spectacle

Table 11.1 Relative advantages of different hearing aid styles. Greater advantages relative to the other styles are indicated by a greater number of check marks. Some of the relative advantages indicated are based on the opinion of a small number of clinicians and the relative advantages shown may not apply to all brands and models or for all clients.

Factor	CIC	ITC	ITE	BTE/ mold	BTE/ dome/ RITA	BTE/ dome/ RITE	Spec-tacle	Body
Ease of insertion and removal	√√	√√	√		√	√	√	√
Ease of manipulating on-aid controls		√	√√	√√√	√√	√√	√√√	√√√
Invisibility	√√√	√√	√	√	√√	√√		
High gain and maximum output			√	√√		√	√√	√√√
Bandwidth and frequency response shape	√√√	√√√	√√√	√√	√√	√√√	√√	
Insensitivity to wind noise	√√√	√√	√√					
Directivity (for omni-directional mics)	√√√	√√	√					
Directivity (for directional mics)		√	√√√	√√√	√√	√√	√√√	
Reliability				√√√	√√√		√√√	√√√
Compatibility with telephones	√√	√	√	√√√	√√	√√	√√√	√
Ease of cleaning				√√√	√√√		√√√	√√√
Avoidance of occlusion		√	√√	√√	√√√	√√√	√√	√√
Avoidance of feedback			√	√√√			√√√	√√√
Same day assess and fit					√√√	√√√		
Cost		√	√	√√	√√	√√	√√	√√

adapter to a BTE hearing aid. The BTE /dome/RITA and BTE/dome/RITE styles assume micro BTEs with thin tubes and thin connecting wires respectively, both connected to instant-fit tips, loosely called domes.

Ease of management greatly affects the success of a hearing aid fitting.[90, 122, 187, 730] The older the patient, the more important ease of management becomes.[1180, 1181] The first two factors in the table, and in the description below, are thus extremely important for many patients.

Ease of insertion and removal

ITE, ITC and CIC hearing aids have been reported as the easiest to insert and remove because they comprise a single package, and they do not interfere with spectacles.[190, 1793, 1817] BTE hearing aids with no helix lock on the mold may be easier to insert than ITE hearing aids with a helix lock, however.[1188, 1717] Similarly, for some users, full-concha ITE hearing aids are harder to insert than ITC and CIC aids because of difficulty in inserting the helix lock (though more secure when inserted completely). CICs with removal strings are relatively easy to insert and remove.[1260] Deeply-seated, long-wear CICs do not present any insertion and removal difficulties for the patient simply because the clinician takes complete responsibility for insertion, and removal is needed only when the battery is near or at the end of its life.

Ease of on-aid user control manipulation

It is difficult for the aid wearer to manipulate a control on a CIC aid while it is in the ear, especially if the CIC is deeply inserted. Gain adjustment becomes easier if an extended flexible shaft is attached to the volume control, but this detracts from the cosmetic advantages of the CIC. The controls on body aids, spectacle aids, and larger BTE aids are easier to operate because they are larger and are more easily located by feeling alone. Add-on caps can often be ordered to increase the height and ease of use of a volume control for an ITE or ITC, although at some expense to their appearance. Ease of manipulation of volume controls is not an issue if automatic control of gain via compression is adequate for the patient, or if a remote control is available and acceptable to the patient. The choice of controls, and the closely related issue of the size and style of the hearing aid can be guided, in part, by asking the patient to adjust a small control, or insert a battery, and by observing his or her ease or difficulty in doing this.

Invisibility

CIC hearing aids have very low visibility,[830] and deeply-seated CIC hearing aids have complete invisibility. Small BTEs with thin tubes or connecting wires terminating in a dome inside the ear canal are almost as invisible, especially for patients with hair that obscures the hearing aid when viewed from the rear.[830] Other things being equal, RITE-style BTEs can have smaller cases than RITA-style BTEs, as the receiver does not need to fit within the case. However, by eliminating telecoils, user switches, and direct audio input, and by using size 10 batteries, many companies produce RITA-BTEs that also are extremely small. Some manufacturers disable the low-frequency channels of open-fit hearing aids to decrease battery current to facilitate the use of small batteries while retaining reasonable battery life.[1257] Removing amplification below about 1 kHz can be done only in hearing aids that will be applied exclusively as an open fitting.

High gain and maximum output

The further the aid microphone is from the ear canal entrance (which is where sound usually leaks from), the greater can be the gain without feedback.[a] The larger the receiver and battery, and hence the larger the hearing aid, the greater the OSPL90 can be, particularly for low frequencies. Open fitting BTEs, combined with effective feedback cancellation, can reportedly achieve high-frequency insertion gain of around 30 dB.[994, 1257] Application of the NAL-NL2 prescription rule in reverse shows that a 30 dB high-frequency insertion gain will match the prescription for high-frequency losses up to about 60 dB HL.

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[a] It sometimes is claimed that open-fit RITE devices are less prone to feedback oscillation than open-fit RITA devices, because of the greater separation of the microphone and receiver in the RITE. This is incorrect because, in both cases, the dominant source of signal leaking back to the microphone comes from the ear canal. This incorrect belief may have originated because in very-high-gain devices fitted with very tightly occluding earmolds (which micro-BTEs are not), gain is sometimes limited by feedback oscillation caused by the receiver directly coupling (mechanically or acoustically) to the microphone which lies very near it in the BTE case.[677, 1531] In this very-high-gain case, RITE BTEs may achieve more gain than RITA BTEs.

High-frequency losses up to about 80 dB could be fit if one was willing to match the prescription only for input levels of 65 dB or higher. That is, the device would be sub-optimal for soft input levels.

The dome-style (i.e. non-custom) canal pieces are unsuitable for achieving significant low-frequency gain. Open domes will not allow any low-frequency gain to be achieved at 250 Hz and greatly restrict the gain achievable at 500 Hz.[b] Some low-frequency gain is achievable with closed domes, especially those with double flanges, but unfortunately the amount of leakage around a closed dome is highly variable, so they are much less satisfactory than a custom mold if low-frequency gain is required.[1355, 1773] That is, although closed domes (or other shapes of instant-fit tips) are more closed than open domes, they are not as closed as can be achieved with custom molds. Low-frequency gain is required for low-frequency losses of 25 dB or greater if one wishes to match the NAL-NL2 prescription for all input levels above 50 dB SPL, and for losses of 30 dB or greater to match the prescription at 65 dB SPL and above.

Bandwidth and frequency response shape

Within the BTEs, the lack of tubing resonances in RITE hearing aids enables slightly higher gain and OSPL90 in the very high frequencies and a smoother response shape across the mid and high frequencies compared to the RITA style, especially if the RITA style has no damping. While electronic filters can remove the peaks and dips caused by tubing, this always comes at a cost of reduced OSPL90 when averaged across frequency.[c] The sound quality advantages arising from the inherently smoother response of the RITE style may be responsible for reported subjective preference for the RITE style,[39] but have to be traded off against the greater reliability of the RITA style arising from the receiver not being in the ear canal.

Thin-tube BTEs have slightly poorer high-frequency response than BTEs delivering their sound with wider #13 tubing, markedly so if the latter employs a high-frequency horn.

Insensitivity to wind noise

Most wind noise comes from turbulence created by the head and the pinna.[458] CIC hearing aids pick up less wind noise than the other aid types because the microphone is further from the turbulence-producing parts of the pinna and the head. CIC microphones are also protected from the direct flow of wind, although at particular angles, turbulence created by the tragus flows into the concha and is picked up by the CIC microphone, so even CICs are not immune to wind noise. BTE and spectacle hearing aids are strongly affected by turbulence created by the pinna. Any hearing aids with directional microphones are extra sensitive to wind noise.

Many hearing aids automatically provide a low-frequency cut when they detect wind noise, but of course this also cuts the low-frequency content of speech. Advanced binaural signal processing schemes, when commercially available, will provide substantial wind noise reduction with minimal reduction of speech cues.

Directivity

BTE, ITE and larger ITC hearing aids are the only styles of hearing aids big enough to contain a directional microphone, and are thus best able to suppress sounds coming from the side and rear of the head. Spectacle aids equipped with a multi-microphone array, and true binaural signal processing aids (Section 7.1.4) have potentially the best performance, but these microphone arrays have so far been limited to research studies and are not commercially available, except as accessory devices.[1182, 1672]

If only omni-directional microphones are considered, CIC aids have the best directivity, followed closely by ITC aids, because these aids make the greatest use of the sound collecting and sound attenuating properties of the head, pinna and concha. Micro BTEs sometimes sit so far behind the pinna that the directional microphone is less effective. Micro BTEs are often fitted as open-canal devices, which dramatically reduces directivity when averaged across frequency (Sections 5.3.4 and 7.3.5).

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b Note the 24 dB reduction in gain at 500 Hz for an open dome relative to a tightly sealed earmold shown in Table 5.1

c Effectively, electronic filters, just like acoustic dampers, can only decrease the OSPL at the resonant peaks, and cannot increase the OSPL at the resonant troughs.

Reliability

The greatest threats to reliability are moisture and cerumen. Hearing aids in which the receiver is located in the ear canal (i.e., CIC, ITC, ITE and BTE/RITE) are the least reliable, because cerumen and moisture limit the life of the receiver. Wax guards are useful for reducing wax ingress. Although receivers in RITE BTEs will break down more often than receivers in RITA BTEs a compensation is that they are easily replaced in the clinic rather than needing to be returned to the manufacturer. The next most unreliable parts are those that involve movement and/or electrical contact between moving surfaces, such as switches, volume controls and battery contacts. Nanocoating and waterproofing are improving reliability by reducing or eliminating moisture ingress.

Telephone compatibility

Hearing aids can pick up either the acoustic or the magnetic signals coming from telephone handsets. For non-micro BTE and spectacle aids, telecoil mode is easily selected and used. For body aids, the body-worn unit must be held near the phone handset, and this complicates usage. For ITE and ITC aids, a telecoil selector switch (whether just for the telecoil or functioning as a program switch) makes the faceplate more crowded and increases the difficulty of operating the controls, particularly if the hearing aid has a volume control. This difficulty can be overcome if the hearing aid has a remote control (but some patients consider that remote controls are inconvenient) or if the hearing aid automatically selects telecoil input when it encounters a strong magnetic signal.

For many hearing aids, the telephone can simply be placed over the ear so that the hearing aid amplifies the acoustic signal. This frees the patient from having to select telecoil mode, but is possible only if the proximity of the telephone does not cause feedback oscillations in the hearing aid. Feedback cancellation processing in the hearing aid and/or acoustic damping material placed over the telephone receiver helps avoid this problem.

Excellent compatibility is achieved if the hearing aid has a wireless receiver that receives signals from a streaming interface device, which in turn receives Bluetooth signals from the telephone (Section 3.11).

Stethoscope compatibility

Similarly, a hearing-impaired medical practitioner *may* be able to use a stethoscope while wearing a CIC without causing feedback oscillation, if the CIC does not have too much gain and/or leakage of sound past the shell.[d] Alternative solutions are:

- Remove the hearing aid, and use an amplified stethoscope.
- Use an amplified stethoscope for which the output device is a pair of supra-aural or circum-aural earphones, which are placed over the ears and hearing aids together.
- Couple the output of an amplified stethoscope to the direct audio input or telecoil input of the hearing aid, or to a streaming interface that wirelessly transmits to the hearing aids.

All of these solutions have some disadvantage and the best option depends on whether the hearing loss is sufficiently great that hearing aids are needed for conversation, and whether the hearing loss is sufficiently great that amplification (obtained somehow) is needed to properly hear the low-frequency and/or high-frequency sounds picked up by the stethoscope.

Ease of cleaning

For patients with chronic ear infections, ITE, ITC, CIC and BTE/RITE styles are unsuitable because the hearing aid cannot adequately be cleaned. BTE/RITA or spectacle hearing aids may be suitable, especially if they can be fitted with a large vent (including a completely open style). For these hearing aids, washing of the earmold and/or tubing is possible. Dust- and water-resistant coatings on hearing aid cases are making it easier to keep hearing aids clean and new-looking.

Avoidance of occlusion and feedback

Patients with near-normal low-frequency hearing combined with a severe high-frequency loss are difficult to fit accurately despite how common this hearing loss configuration is. The low-frequency thresholds require a large vent or an open fitting to minimize occlusion but the gain required at high frequencies to fully meet prescription targets may then cause feedback oscillation, even with feedback can-

[d] The hard tip on a stethoscope can be replaced with a soft silicone tip, available from earmold suppliers, that better couples the sound to the faceplate of the hearing aid.

cellation technology. These conflicting constraints are more easily met if the distance from the vent outlet to the microphone inlet is increased. The compromise is thus easier in BTE hearing aids than in ITE, ITC or CIC hearing aids. Although it is often claimed that occlusion is not a problem for CIC aids because the medial end of the aid is in the bony part of the canal, a deeply seated shell or earmold can be used with any hearing aid style, and is thus *not* a special advantage of CIC aids. Deeply seated devices, even with soft tips, are, however, more likely to result in discomfort and lower satisfaction levels.[e, 1882,]

Same-day assess and fit

BTE hearing aids that do not require a custom earmold can be provided to a patient in the same appointment that the patient's hearing and needs are assessed. If the patient is ready to proceed immediately to fitting following the hearing assessment, this arrangement is very efficient for the client and clinician alike, as one visit to the clinic is saved. A large-scale study showed that same-day assessment and fitting was possible for 81% of first-time hearing aid users, although feedback oscillation problems were reported for 28% of the same-day fittings.[1654] The proportion for whom non-custom earmolds were suitable decreased as both hearing thresholds and age increased, but over half of the patients aged over 85 years were none-the-less able to be fit with these devices. The clinician must, of course, keep one or more devices in stock to enable same-day fitting. The ear canal pieces used can be open or closed dome styles, or can make use of soft plastic foam disposable tips to achieve a fitting that is more acoustically closed than is generally possible with flexible silicon closed domes.

The major complication is determining before the client walks through the door whether a shorter appointment time appropriate to an assessment, or a longer appointment time appropriate to an assessment and a fitting should be allocated. If the client has already had a hearing screening, then the results may give some idea of the likelihood of progressing to a fitting,[1654] but if the attitude of the client towards hearing aids was also sought during the screening, this is likely to be a more reliable indicator (Sections 9.1.1 and 9.1.2).

Cost

It is common for CIC hearing aids to cost more than the other styles. They are more expensive to manufacture because fitting the components in is more labor intensive, and they have a higher return rate. BTEs cost less to produce than any custom products, and so the cost to patients should be less unless there are market distortions.

Battery size

A small hearing aid cannot accommodate a large battery. As battery size decreases, handling difficulties increase (for some patients) and battery life decreases (assuming no change in gain and maximum output).

Factors in combination

Each of the advantages of the different styles has to be weighted according to an individual patient's needs, wishes and capabilities: the capacity to provide a high gain, for example, is not an advantage if the patient does not need a high gain. It is therefore illogical to simply add the check marks against each style in Table 11.1. It is worth noting, however, that most hearing-impaired people are elderly, and that many elderly people have trouble manipulating small

Selecting to minimize management problems

The following choices are appropriate for patients who are expected to have difficulty manipulating hearing aids or their controls.

- Choose a hearing aid with wide dynamic range compression and no volume control.
- Choose the largest hearing aid style and battery size that the patient finds cosmetically acceptable, but choose a half-concha ITE rather than a BTE.
- For patients with good mental capabilities but poor physical manipulation ability, consider an aid with a remote control.
- For patients with poor vision, all controls and the battery compartment opening point have to be easily located tactually.

[e] The lower satisfaction with devices terminating in the bony canal was reported with non-custom, disposable devices, but is likely to also be true for custom devices.

objects, because of either diminished vision or diminished tactile sensitivity. There is a strong statistical connection between the degree of difficulty patients have in managing different hearing aid styles and the eventual satisfaction they report for each style.[90, 187]

Compromises will often be necessary. If a patient places high importance on invisibility of the hearing aid, the feel of the fitting within the ear canal, or on his or her own voice sounding normal, then a micro BTE with an open dome canal piece will meet these aims. If the patient has a 30 dB loss in the low frequencies, then the low-frequency gain achieved will be inadequate, and if he or she has an 80 dB loss in the high frequencies, then the high-frequency gain will be inadequate. As an unworn hearing aid provides no benefit, it is better to prioritize factors that enable the patient to accept the hearing aids, and get *some* benefit, over factors that *maximize* the benefit the hearing aids are capable of providing.

Note that the provision of less than optimal high-frequency gain to achieve physical comfort and good own-voice quality really is a compromise. There is no logical basis for saying that the target gain should be less for an open fitting than for a closed fitting applied to the same hearing loss, but there may well be sound reasons for accepting less than optimal gain.

Compromises may not always be in the direction of prioritizing comfort and appearance over performance. Although patients with mild flat or gently sloping loss will appreciate the absence of occlusion effect provided by an open fitting, their greatest need is for improved speech intelligibility in noise, rather than in quiet (Section 9.1.6). Intelligibility in noise will be maximized by a fitting that is directional over the widest possible frequency range, which is not facilitated by an open fitting.

Spectacle aids have a major logistical disadvantage not shown in Table 11.1: the eye specialist and ear specialist have to coordinate their activities to ensure that the spectacle adapter on the hearing aid matches the spectacle frame. Also, if either hearing aid or spectacles break down, both devices may be unavailable to the wearer until the repair is completed.

The reader will appreciate that the comparative advantages outlined in this section are generalities, and exceptions to these generalities will undoubtedly exist in individual models from individual manufacturers.

11.2 Selecting Hearing Aid Features

The following features should be considered before making a final choice of hearing aid model. Further signal processing features are discussed in Sections 8.4 and 8.5.

Volume control

All varieties of compression decrease the need for a volume control, although not necessarily to the same degree. Many patients are very pleased not to have to manipulate a volume control. Fortunately, many patients will not need a manual volume control if the hearing aid has wide dynamic range compression (WDRC) with an adequately high compression ratio and adequately low compression threshold. Unfortunately, there is an approximately equal number of patients using WDRC hearing aids who say that there are occasions on which they would have liked to turn their hearing aids up or down.[461, 937, 948, 1749, 1823] Of course, some of these patients would not be able to use a volume control, and some would, on balance, not choose to have one on the hearing aid if the need for one is infrequent.

Kochkin (2003) discusses the complex relationship between presence of a volume control and satisfaction with hearing aids, and concludes that there are three categories of people who benefit from having one:

- those for whom the WDRC does not achieve an acceptable loudness in some situations;
- those who psychologically strongly desire to control their hearing aids; and
- experienced aid wearers who are used to a volume control (who presumably also fall into one of the above two categories).

Volume controls can *create* problems for some patients if they are accidentally moved (less likely with a rocker switch than a rotary control) when the aid is being inserted or removed. There is no effective way to predict which patients are likely to need a control. A safe option is to order a manual control that can be electronically locked, unless:

- the patient is expected to have limited ability to manipulate a control;
- the patient has previously used, and been happy with, a fully automatic aid; or
- the hearing aid is so small that inclusion of a volume control is impractical.

Another way to provide some of the benefits of a volume control without actually fitting one is to use a push-button on the hearing aid to select from multiple memories. Each push advances the hearing aid to its next memory, and if these memories differ in gain (possibly along with other response changes) this feature can partly compensate for not having a volume control. If the aid wearer uses a remote control, there is little need for a volume control on the hearing aid.

Telecoil

A telecoil (Section 2.8) is essential for anyone with a severe or profound hearing loss. It is also likely to help people with a moderate loss use the telephone. People with mild loss can usually cope reasonably well with the telephone without their hearing aid, or by acoustic coupling to the hearing aid if it has effective feedback cancellation. People with all degrees of loss will appreciate the reduction in noise and reverberation that telecoils offer when used in conjunction with a room loop (Section 3.5).

The disadvantage of a telecoil is the increase in size needed to fit in the telecoil. If the presence of the telecoil makes it necessary to add a program switch, then a second disadvantage is the increased crowding of an ITE or ITC faceplate or micro BTE case. This increased crowding can make it hard for the patient to find and operate the correct control.

These disadvantages should be weighed against the substantial advantages. Most commonly, people with severe and profound losses have telecoils and people with mild losses do not. Automatic telecoil selection, especially by hearing aids that communicate across the head wirelessly so that they make good decisions about the proximity of a phone, enables a telecoil to be included without needing a selection switch on the hearing aid.

Direct audio input or wireless input

Direct audio input (Section 2.9) is particularly useful for:

- People who use a wireless transmission system that is electrically coupled to their hearing aids to improve SNR. Adults as well as children can benefit enormously from a wireless system, but the sensitivity of the FM system and the hearing aid when both are in use must be carefully adjusted by the clinician so that the combination gives the high SNR that it is capable of (Section 3.6).[153]
- People who use a hand-held directional microphone connected to the hearing aid via a cable. Most commonly it is people with severe or profound hearing loss who choose to use these devices. The increase in signal-to-noise ratio can be substantial. These microphones can provide directivity superior to that of head-worn microphones and often can be held closer to the source.
- People who watch TV in a noisy or reverberant place. A microphone placed near the TV, or a plug coupled to the TV audio output can be connected via a cable to the hearing aid. This can provide a substantial increase in signal-to-noise ratio and a substantial decrease in reverberation.

Increasingly, however, hearing aids are containing a wireless receiver of some sort to receive streaming from various devices producing audio signals (e.g. via Bluetooth; Section 3.11), which eliminates the need for a direct audio input connector in the hearing aid.

Directional microphones

Directional microphones can offer a substantial improvement in signal-to-noise ratio, as described in Chapter 7. Hearing aids can be ordered with directional microphones permanently selected, but most hearing aids with directional microphones can be switched, or automatically switch, between directional and omni-directional modes. The only reason for *not* choosing a switchable directional microphone is if the patient wants a low-visibility custom hearing aid (a CIC or an ITC). It is not possible to fit an effective directional microphone into these hearing aids, although larger ITC hearing aids can have directional microphones with limited directivity.

Directional microphones have the following disadvantages. It is for these reasons that a manually or automatically switched directional microphone is better than a permanently directional microphone:

- Directional microphones are even more prone than omni-directional microphones to wind noise. They can therefore disadvantage patients who spend a lot of time outdoors.
- In some circumstances, it is not possible for the patient to always look at the sound source. Examples include someone driving a car and listening to passengers, a pedestrian dodging traffic, and children in a classroom listening to those behind them. In such situations, speech and

environmental sounds may be clearer and more audible when an omni-directional microphone is used, unless the directional microphones can adapt so that their maximum sensitivity is in directions other than the front. Alternatively, an asymmetrical fitting (directional in only one ear) will avoid or minimize the disadvantages that would otherwise occur when sounds come from rearward directions. For a single hearing aid acting in isolation, the direction of maximum sensitivity can only be approximately frontal or approximately rearward. For true bilateral-processing hearing aids, maximum sensitivity can, in principle, be in any desired direction.

Note that in indoor situations where the talker is some distance away, the hearing-aid wearer will be well outside the direct field of the talker and may remain so even when using a directional microphone. If such situations provide the primary need for hearing aids, directional microphones (whether permanent or switchable) are not likely to offer any advantage (Section 7.3.1).

Two aspects of the patient's hearing loss may limit the range of frequencies over which directivity can be achieved, and hence limit the effectiveness of the directional microphone:

- The patient, especially one with a severe-to-profound hearing loss, may require a low- to mid-frequency response *considerably* flatter than can be provided by a hearing aid with a directional microphone. Achieving a flat response is rarely a problem, but greater gain and lower internal noise in the low frequencies can always be achieved with an omni-directional microphone than with a directional microphone. For these patients, directivity can be limited to the high frequencies – i.e. split-band directivity.
- The patient may require amplification only over a restricted frequency range (e.g. above 1500 Hz). The hearing aid will be directional only over the frequency range where amplified sound dominates over vent-transmitted sound.

It is sometimes suggested that a speech-in-noise test be used to determine the SNR deficit (i.e. a higher than normal value of SRT_n) before determining whether a patient needs directional microphones. However, *every* person with a hearing loss has trouble understanding speech when the SNR is sufficiently poor. Indeed, so does every person with normal hearing. As a directional microphone will improve the SNR of the signal passed from the microphone on to the rest of the hearing aid whenever the acoustics of the listening situation (distance, direction, reverberation) allow it, there is *no* result on a speech test that would indicate that a directional microphone is not needed. The only exception might be a patient who only ever listens in quiet situations. This is not to argue against measuring the SNR deficit of a patient as part of a needs assessment. Such a measurement may well indicate a deficit so large that a considerable deficit will remain even when directional microphones are being used. Such a result supports a recommendation to use a wireless system whenever the logistics of the situation allow it.

Compression limiting versus peak clipping

Peak clipping should be chosen in preference to compression limiting only for:

- Patients with a profound hearing loss who need the greatest possible OSPL90. If patients prefer the volume control to be turned to its highest setting, it is likely that they would benefit from more gain or more OSPL90, or both. Greater maximum output, especially for speech signals, is possible with peak clipping (Section 10.7.2).
- Patients who have to be fitted with a larger hearing aid to achieve the required OSPL90 in a compression limiting aid, but who do not want to wear a larger hearing aid. For example, it may be possible to achieve adequate OSPL90 with a peak-clipping ITC or with a compression-limiting BTE. Some patients will prefer the size and appearance of the ITC to the lower distortion of the BTE. For other patients, sound quality will be more important than appearance.

Remember that the choice of peak clipping versus compression limiting is less important for hearing aids that have wide dynamic range compression (see Section 10.7.5). Patients who are used to peak clipping may not initially like a changeover to compression limiting, although most appreciate the change within a few weeks.[407]

Wide dynamic range compression

There is ample evidence that WDRC with a low compression ratio should be available in all hearing aids (Section 6.5). While a small proportion of patients may not gain any advantage from WDRC relative to linear amplification, there is as yet no way to *reliably* predict which patients these are.[461] WDRC is, however, most likely to be advantageous for high cognition clients who need their hearing aids in a wide variety of communication situations and who have sloping hearing losses.[598] It seems safest to initially select some form of WDRC for everyone. For those with a profound hearing loss, relatively high compression thresholds and/or low compression ratios may be necessary (Section 6.5.1).

Multichannel compression

Evidence for additional benefit from more than one channel of compression is less clear (Section 6.5.2). It seems *likely*, however, that multichannel compression will provide some additional benefit for patients with a moderately or steeply sloping hearing loss, because a different degree of compression can be used in each channel. A possible criterion would be to use multichannel compression for any patient whose 2 kHz threshold exceeds the 500 Hz threshold by more than about 25 dB. These people are likely to benefit most from a TILL response aid (Section 6.2.4). Patients with a flat loss may have a weak preference for single-channel compression,[881] but further data on this issue are needed.

In practical terms, however, multiple channels are the most common way to implement adaptive noise suppression and adaptive microphone directivity, and it is likely that compression will also have been implemented in these same channels. Provided compression ratio is less than about 3:1 (or perhaps 2:1), multichannel compression is unlikely to be harmful for anyone.

Fast- or slow-acting compression

Some advanced multichannel hearing aids currently on the market have very fast-acting compression, others have very slow-acting compression, others have combinations of fast and slow, and others have programmable attack and release times. How can the right compressor speed be chosen for an individual patient? As yet there is *no* way of systematically prescribing which type of compressor is best for an individual patient. There are, however, indications that fast acting compression is best for subjects with a high level of cognitive functioning, as measured by their ability to identify target sequences amongst rapidly changing visual patterns, and who frequently need to use their hearing aids in communication situations where the sound level varies rapidly by large amounts.[598] Slow-acting compression is more likely to be beneficial for clients who wear their hearing aids in a range of environments that have different mean sound levels or in which sound levels change slowly.[598]

Most likely, both forms of compression should be present in the hearing aid so that the overall compression ratio can be increased while minimizing the disadvantages associated with a high compression ratio of either type. If either speed of compression dominates, and if the patient reports any of the disadvantages of fast or slow compression outlined in Sections 6.3.2 or 6.3.3 respectively, this would be an indication to decrease the compression ratio associated with this type of compression.

Adaptive noise reduction

Amplification schemes in which the gain is automatically decreased in those frequency regions that have the poorest signal-to-noise ratio are most likely to be appreciated by patients who wear hearing aids in a wide variety of noisy environments. The comfort advantages of these schemes should increase along with the variety of noise spectra that the hearing aid wearer encounters in everyday life.

The benefits of adaptive noise reduction will be far greater for patients who need amplification for all frequencies than for those who need amplification for only the high frequencies. If the low-frequency gain of a hearing aid is dominated by vent-transmitted sounds, the effects of adaptive noise reduction will be confined to higher frequencies, which will make the adaptive noise reduction less effective.

Multiple memories

The candidacy issues for multi-memory hearing aids are very similar to those for adaptive noise reduction. Patients are most likely to benefit from multi-memory amplification if they require amplification over a wide frequency range, regularly wear their hearing aids in acoustically diverse listening environments, and have a more severely restricted dynamic range in the high frequencies (Section 10.6.2). The reason for this similarity is that both multi-memory amplification and adaptive noise suppression aim to vary the amplification characteristics depending on the acoustic environment, and both can achieve useful gain variations only at those frequencies where the gain is greater than 0 dB.

In addition, however, multiple memories can be used to access the telecoil or to change the microphone directionality. Auto switching between memories (Section 8.5) depending on the environment is acceptable to many patients. The environment detectors are not perfect though, and some patients find it disconcerting when the program, and hence sound quality, changes in the absence of any distinct change in the environment.

Feedback management schemes

For people with a history of trouble with feedback oscillation, or people whose pure-tone thresholds suggest that feedback will be a problem, feedback management (preferably incorporating feedback cancellation) is worthwhile (Section 8.2.3). Feedback management processing will most benefit patients:

- with a profound hearing loss;
- with good low-frequency hearing combined with poor, but useable, high-frequency hearing;
- wearing open-canal hearing aids; or
- using the telephone on microphone setting.

If a hearing aid would otherwise oscillate or ring, then enabling feedback cancellation will be beneficial. If oscillation or ringing is not a problem, feedback cancellation will cause no adverse effects, with the possible exception of its effect on musical sounds.

Frequency lowering

It is unclear for which patients frequency lowering (i.e. frequency transposition or frequency compression) should be enabled (Section 8.3.4). However, provided it is limited to high-frequency sounds, and the extent of lowering is mild, there rarely seems to be any adverse effects, and at the least it enables higher gains to be achieved without feedback oscillation.[f] There seems to be little to lose in enabling it for patients for whom typical level speech cannot be made at least partly audible across the frequency range from around 3 to 6 kHz, or in patients with known dead regions, but the potential candidacy of frequency lowering is not limited to such people. Neither is it guaranteed that patients meeting these criteria will benefit from frequency lowering, especially in a noisy listening environment.

Trainability

Some patients will greatly appreciate the opportunity to train their hearing aid to their personal preferences in their actual listening situations. Others will not be capable of making the adjustments, either because of diminished manipulation ability or because their cognitive or perceptive abilities do not allow them to understand the concept of adjusting a control to the position that gives the best sound, or to hear the differences that variation of the control causes.

11.3 Hearing Aid Selection and Adjustment

After considering the features needed, the next step is to choose a manufacturer, model, prescription formula, and adjust the hearing to match the prescription target. The methods described here are most applicable for adult patients and children approximately six years of age and older. Much is also applicable

Automated and patient-controlled adjustment of hearing aids – you may have nothing to do!

The selection and adjustment process to achieve a reasonable match to the target is simple, provided the hearing aid manufacturer's fitting software includes a prescription procedure you are willing to use. Once you have connected the hearing aid (and if appropriate, specified the venting) the software will adjust the hearing aid to approximate the target response prescribed by the selected prescription procedure for the patient's audiogram.

If the fitting software is integrated with a real-ear measurement system, it will automatically measure the real-ear response and may adjust the hearing aid for you (Step 11) after the real-ear measurement, repeating automatically until the best possible match to the prescription is achieved. If the hearing aid processing includes trainability and the patient is able to use it, then the hearing aid and patient together will take care of most of the fine-tuning needed. Do not worry. Despite all this automation you are still needed because there are many people-oriented aspects of hearing aid fitting (see especially Chapters 9, 12, 13 and 14) for which your human skills are indispensable!

[f] Increased stable gain will be achievable provided the frequency lowering region includes the frequency at which oscillation occurs.

to younger children and infants, but Chapter 16 will recommend some more efficient variations for these younger people. Most prescriptive formula enable their prescriptions to be expressed as either REIG, REAG, REAR, or CG targets. There is no single right method to select and initially adjust a hearing aid. There are, however, some methods that take less time than others. This book recommends that REIG be used for adults and older children, and that REAG combined with CG be used for children under the age of six (see panel at the end of Section 10.2.4).

Figure 11.1 shows an overview of the twelve steps involved in the selection and adjustment of programmable hearing aids, prior to ascertaining the patient's opinion of sound quality. These steps are described in the following paragraphs. We will assume that the fitting tool is NOAH-based software on a personal computer (Section 3.3.1). The steps are similar if a dedicated programming device from a single manufacturer is used.

Step 1: Enter Audiometric data. As a minimum, audiometric data will include pure-tone thresholds for each ear to be fitted. Other audiometric data, not necessarily used for the hearing aid fitting, could include patient identifying data, discomfort levels, most comfortable levels, loudness scales, speech identification scores, acoustic reflex data, and tympanometric data. Within NOAH, these data will be entered in the ***Client Module*** and the ***Audiometry Module***. Most prescription procedures will use only the threshold information, but it is a good idea to enter enough information to unambiguously identify the patient at a later time. Having the audiometric information stored will save time during the fitting appointment. NOAH will store information about the hearing aids selected or adjusted for the patient in each session. This can provide a valuable history of the patient's fittings. Apart from its use in hearing aid fitting, the NOAH database can be used as the sole repository of patient information for the practice.

Step 2: Open manufacturer's software. There is no systematic way to choose a particular brand of hearing aid for a patient. The major manufacturers have such comprehensive ranges of hearing aids that most patients can be fit equally well with a product from any of them. Factors that may affect your decision of which manufacturer to consider first will include:

- a history of reliable and timely sales and after-sales service;

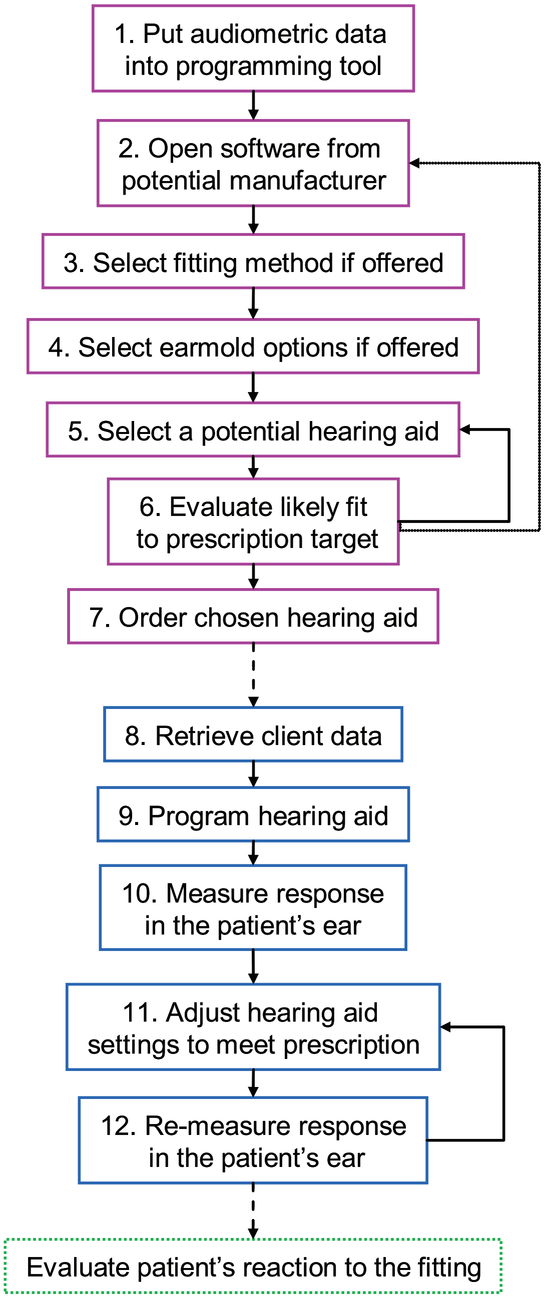

Figure 11.1 Twelve steps for selecting and adjusting programmable hearing aids.

- familiarity with the hearing aids and software of a particular manufacturer, especially if that manufacturer has a hearing aid that you have found to give good results for previous patients with similar needs and audiometric profiles;
- the availability of hearing aids with the combination of features required for the particular patient; and

- discounts applicable if the requisite number of hearing aids is purchased within a month from the same manufacturer.

Step 3: Select fitting method. Most manufacturers offer you a choice of generic prescription procedures (NAL-NL2, DSLm[i/o], CAM2, FIG6 etc). Others offer only a prescription procedure developed by the manufacturer, and some will offer both a proprietary and one or more generic prescriptions. See Sections 10.2.4 and 10.4.8 for a comparison of different procedures. With some prescriptions, the gain is affected by whether one or two hearing aids will be worn (Section 15.8) so also specify whether it is a bilateral or unilateral fitting.

Step 4: Select earmold or earshell options. Some manufacturers' software will automatically recommend a vent size (and for BTE hearing aids, a sound bore). Some software instead requires you to specify the earmold or earshell configuration. Other software makes no allowance for the earmold or earshell configuration. If you are able to specify the acoustic configuration, follow the procedure outlined in Section 5.7 to first determine what you should fit. It is important to specify the approximate vent size if a 2-mm or larger vent is used, or else the software may make large errors in calculating the coupler gain needed to achieve the real-ear gain target. This may result in you choosing an inappropriate hearing aid, and will almost certainly result in the software pre-adjusting the hearing aid to a tone control setting with an inappropriate low-frequency gain.

Step 5: Select a potential hearing aid. With most software, you make an initial specification of the hearing aids you would like to consider. How this is done varies between manufacturers. At one extreme you are required to specify which particular hearing aid style (e.g. ITE) and family of hearing aids, you are interested in. At another extreme, you indicate the style and features (e.g. telecoil, directional microphone, and volume control) you require, and the software will list a range of specific hearing aids that meet your requirements to varying degrees. You can then make a selection from this range.

Step 6: Evaluate likely match to the prescription. Once you have chosen a hearing aid, most software programs will indicate graphically how well the hearing aid should meet the prescription targets for your particular patient. The graphical display comprises a gain-frequency response or output-frequency at one or more input levels and/or input-output curves at one or more frequencies. It may be necessary to examine the match to the prescription for more than one hearing aid before making a final selection. If the match to the prescription is not sufficiently close, or if the hearing aids do not have the features you want, it may also be necessary to return to Step 2 and try the models from an alternative manufacturer.

Step 7: Order chosen hearing aid(s). The software will usually enable you to print out all the information necessary to order the hearing aids you have selected or, more commonly, will enable an order to be sent electronically to the manufacturer. The information required will vary from manufacturer to manufacturer and from hearing aid to hearing aid. For custom products the information required may include the following if the model number or type does not uniquely specify any of these features:

- battery orientation (toilet lid or swing-out);
- battery size;
- telecoil and switch;
- volume control add-on cap;
- removal handle;
- microphone directionality;
- vent diameter and adjustment options; or
- sound bore and earmold material (for an earmold).

Step 8: Retrieve patient data. When the hearing aids have arrived and the patient is about to be fitted, you can retrieve the patient data from NOAH.

Step 9: Program the hearing aid. The manufacturer's software, via the HiPro or NOAHLink interface, will likely make an initial adjustment of the hearing aid to approximate the prescription.

Step 10: Measure response in the patient's ear. The response of the hearing aid should be measured with a real-ear analyzer employing a probe microphone. The type of measurement to be done will depend on the nature of the prescription target: insertion gain, real-ear aided gain, real-ear SPL, or real-ear input-output curves. For nonlinear hearing aids, gain-frequency responses should be measured with a broadband stimulus, with speech-like dynamics if the hearing aid has adaptive noise reduction enabled (Section 4.1.3). Alternatively, stationary noise with a spectral shape equal to the long-term spectrum of speech can be used with adaptive noise reduction turned off.

It is most convenient if the results of the measurement appear on the same screen as the prescription target. This can be a screen within the manufacturer's software if the real-ear analyzer is able to send the measurement results to NOAH. Otherwise, the audiometric information or the prescription target should be imported or typed into the real-ear analyzer so that the prescription target can be displayed on its screen. It is possible to compare the measurement results on the analyzer with the prescription targets on a different computer, but the comparison is more time consuming. This comparison can be facilitated by drawing the targets on the analyzer screen with a whiteboard marker. (Make sure it's erasable or you'll need a new analyzer every week.)

Step 11: Adjust hearing aid settings to match prescription. If Step 10 reveals a significant discrepancy (Section 10.3.6) between the target and actual response, the hearing aid settings should be modified to minimize the discrepancy. If the real-ear analyzer is able to send the result to the manufacturer's software, via NOAH, then the manufacturer's software may make these adjustments (and perform the subsequent step 12) automatically for you.

Step 12: Re-measure the response in the patient's ear. Following each adjustment of the hearing aid, the response should be re-measured. Eventually, you will decide that the measured response is close enough to the target, as discussed in Section 10.3.6. After the prescription targets have been achieved with sufficient accuracy, the patient's reactions to all aspects of the sound quality have to be determined, but this is covered in Section 11.7 (for maximum output) and in Chapter 12 (for all other aspects).

11.4 Allowing for Individual Ear Size and Shape in the Coupler Prescription

Once the prescribed real-ear gain has been obtained, there is no need to consider the effects of variations in the size and acoustic properties of the patient's external ear. These effects are built into the real-ear gain that has been obtained. The coupler (or ear-simulator) response that is needed to achieve the prescribed real-ear gain, however, depends on the acoustics of the individual patient's ear. If our goal is to achieve a certain real-ear insertion gain (REIG), the individual patient's real-ear to coupler difference (RECD) and real-ear unaided gain (REUG) will affect the coupler gain required (see equation 4.15 or 4.17). If our goal is to achieve a certain real-ear aided gain (REAG), only RECD affects the coupler gain required (see equation 4.9).

Customizing the coupler prescription

If your prescription software prescribes coupler response, and you wish to modify this prescription to allow for the measured characteristics of an individual patient's ears, any of the following corrections can be used.

To accurately meet an REAG target:

$$\text{Custom coupler gain prescription} = \text{standard coupler gain prescription} + RECD_{average} - RECD_{individual} \quad \ldots\ 11.1$$

To accurately meet an insertion gain target:

$$\text{Custom coupler gain prescription} = \text{standard coupler gain prescription} + RECD_{average} - RECD_{individual} + REUG_{individual} - REUG_{average} \quad \ldots 11.2$$

To accurately meet a real-ear saturation response target:

$$\text{Custom coupler OSPL90 prescription} = \text{standard coupler OSPL90 prescription} + RECD_{average} - RECD_{individual} \quad \ldots\ 11.3$$

Average values for RECD (based on HA1 and HA2 couplers) and REUG can be found in Tables 4.4 and 4.7 respectively.

Consequently, a hearing aid that has been pre-adjusted in a coupler or ear simulator to match the prescription will most precisely match the real-ear target if these individual ear effects are known and incorporated into the coupler or ear simulator prescription. Some fitting software (DSLm[i/o], NAL-NL2) allows this to be done. To make this correction, measure the appropriate individual ear effect, and enter the data into the program at the appropriate place. If one wishes to allow for these effects and does not have access to these programs, any coupler prescription can be customized using equations 11.1 to 11.3. To apply these equations to an ear simulator, simply replace *coupler* with *ear simulator* in each equation.

As discussed in the panel within Section 10.2.4, it does not seem reasonable to preserve a person's REUG in a hearing aid fitting if that REUG is atypical as a result of surgery. This, however, is what happens if one adopts and achieves an insertion response target for that person. The solution is to use an REAG target rather than an insertion gain target. If the prescription method you use does not specifically give a REAG target, you must add an average REUG to the insertion gain target to convert it to an REAG target. Suitable average values for REUG can be found in Table 4.7.

Measurement of the individual's RECD, and optionally REUG, for use in calculation of prescription targets is worthwhile only in the following cases:

- when fitting a hearing aid with few controls (rarely the case with programmable hearing aids), especially if the average gain of the hearing aid being considered only marginally matches the prescribed average gain;
- when it is important to minimize the appointment time by accurately pre-programming the hearing aid before the patient arrives;
- when several hearing aids are to be pre-programmed prior to the fitting appointment to enable the patient to compare them at the fitting appointment; or
- for infants (RECD only), as further discussed in Chapter 16.

Other than in these circumstances, measurement of RECD or REUG is not worthwhile, because the hearing aid can quickly be adjusted to match the target REIG or REAG while the real-ear response is being measured with a probe microphone, whatever the individual's RECD and REAG are. Individual RECD responses tend to be parallel to the average RECD,[856] and so in most cases nothing more than a gain change is needed to compensate for the individual response.

11.5 Verifying and Achieving the Prescribed Real-ear Response

It is important to verify that the prescribed real-ear characteristics have indeed been obtained. Although the prescription will not be perfect for every client, a good prescription provides the best possible starting point from which fine-tuning can be carried out, and will minimize the number of patients for whom any fine tuning is necessary. Although most manufacturers' software will automatically make an initial adjustment of the hearing aid to approximate a prescription target, the accuracy of the match can be very poor, and can usually be improved by the clinician.[4, 698] It is therefore not adequate to "verify" the fitting just by looking at how closely the software predicts the real-ear response will match the target.

There are two basic measurement alternatives for adult patients, with the first being as well as more common, more intuitive, and more able to give realistic results when adaptive noise reduction algorithms are present.

- Real-ear gain can be measured at three input levels, such as 50, 65 and 80 dB SPL, and compared to the prescribed targets at those levels. If a reasonable match is obtained, it is extremely unlikely that there will be marked discrepancies at intermediate levels.
- Alternatively, the compression characteristics can be established with the aid of input-output curves. One curve should be used for each channel of a multichannel hearing aid. If these are correct, only a single gain-frequency response curve will then be needed. This should be done at a mid-level input, such as 65 dB SPL. Beware, however, that in some hearing aids the gain of a compressor is affected by the signal level in surrounding channels, so measurements made with narrow-band stimuli may differ in some way from measurements made with more realistic stimuli.

For infant patients, verification is better based on coupler gain, with the measurement results compared to coupler gain targets that have been calculated using the infant's individually measured or predicted RECD (Section 16.5).

Nonlinear hearing aids should be measured with a broadband stimulus (Sections 4.1.3 and 4.5.7). A problem can arise with some hearing aids that view a steady broadband test signal as a noise that should be attenuated! This problem can be overcome by using a more complex test signal (e.g. one with speech-like dynamics) so that the hearing aid treats it as a wanted signal. Alternatively, the noise reduction algorithm can be disabled.

Is there a role for behavioral testing to verify the gain-frequency characteristics? Possibly, and although tonal aided threshold testing can be used on patients of any age, the major application is for infants, so we will address this question in Chapter 16. If real-ear gain measurement is not available, then speech sounds can be used to evaluate the effectiveness of the hearing aid.[341] The Ling Six-Sound test consists of the patient repeating, with eyes shut, each of the sounds "mm", "oo", "ah", "ee", "sh", "ss" when uttered live voice by the clinician. The frequency spectra of these sounds range from very low-frequency dominated ("mm") to very high-frequency dominated ("ss"). As hearing aid prescription formula are aimed at making speech detectable, an evaluation of whether the patient can hear and recognize these speech sounds at a soft speaking level could be considered as a broad verification of the fitting.[g]

11.6 Verifying Signal Processing Features

There are no statistics on how often the signal processing features in hearing aids do not work as implied by the fitting software. There are certainly anecdotes of microphones being incorrectly connected such that directional microphones point backward, or of dual microphones being mismatched in sensitivity such that very little directivity is obtained. A cautious clinician may therefore want to verify that signal processing algorithms are performing broadly as expected, even if precise measurement of their performance is not possible. In some hearing aids, certain signal processing features are selected automatically by the hearing aid only in particular listening situations. To check such devices it is necessary to first input a sound for a time sufficiently long to cause the hearing aid to recognize the listening condition, and then to transition to having the signal processing feature enabled. This process may well take 30 seconds or even longer.

Directional microphones. The gain-frequency response of directional microphones can be evaluated in a test box with the hearing aid oriented towards, and then away from, the loudspeaker. The difference between the two gains, averaged across frequency, will approximately equal the front-back ratio (Section 7.1.1) of the hearing aid divided by the compression ratio. Even without a test box, a clinician can perform a simple listening test with the aid of a stethoclip. The hearing aid is positioned very close to a loudspeaker emitting continuous noise or multi-talker babble and then rotated so that the hearing aid is first facing towards, then away from, the loudspeaker. The noise should be louder when it faces towards the loudspeaker. Even more simply, the clinician can say "shhh" into the hearing aid while rotating it in front of the lips. A more time-consuming verification, but one that may also assist in counseling the patient, is performing a speech-in-noise test, with speech presented from the front and noise presented from the back, as discussed in Section 7.3.6.

Adaptive noise reduction. Adaptive noise reduction can be assessed by measuring the gain-frequency response with a steady (i.e. unmodulated) noise, and with a signal that has spectral characteristics similar to the unmodulated noise, but dynamics (i.e. modulation) typical of speech. Real speech is a suitable signal. When adaptive noise reduction is operating, there should be a lower gain for high-level unmodulated noise than for a modulated signal at the same long-term level. This reduced gain can be measured in a test box, or judged by the clinician listening to the hearing aid output through a stethoclip.[341] Note that hearing aids intentionally take from seconds to several tens of seconds to initiate adaptive noise reduction after noise is detected, and the gain reduction may then occur very gradually with time. Measurement in

[g] The Ling Sound Test would be better considered as evaluation of effectiveness, rather than as verification that a targeted response has been achieved. It is feasible that a prescribed response could be precisely achieved without all of the sounds being detectable when presented at soft levels, especially for a patient with very severe hearing loss. It is also feasible that, for any degree of loss, all the sounds could be perceived, even though the prescribed gain is much greater than optimal in one or more frequency regions.

the test box, or judgment about the loudness change, should not be made until the noise reduction algorithm has fully activated.

Frequency lowering. A major purpose of frequency lowering is to enable detection and recognition of fricatives. The effectiveness of frequency lowering settings adopted can be evaluated by determining whether the patient is able to hear, and differentiate, /s/ and /ʃ/ spoken by the clinician at normal vocal effort.[626] Failure to detect these sounds might indicate that frequencies have not been lowered sufficiently, or that gain in the frequency region to which they have been lowered is insufficient. Unfortunately, there is no simple test for indicating that frequency has been excessively lowered, or that gain in the lowered region is too large.

Impulse noise reduction. The easiest way to confirm the correct operation of impulse noise reduction is with a listening test, using an easily reproduced impulsive sound such as hitting a spoon against a cup.

11.7 Evaluating and Fine-tuning OSPL90

Hearing aid maximum output should be subjectively evaluated for all patients who are capable of indicating excessive or insufficient loudness. Because an excessive OSPL90 can cause an extremely negative first experience with hearing aids, evaluation of maximum output should be performed at the fitting appointment. Maximum output is evaluated by asking patients to judge the loudness of intense sounds presented in the clinic. Remember that any variations made to the acoustic coupling (vents, dampers, and sound bore profile) during the fitting will have the same effect on maximum output that they have on gain.

Note that the output level for a 90 dB SPL input may be controlled by the output limiting compressor or clipper, or may be controlled by the wide dynamic range compressor. In Figure 4.6(a), for example, OSPL90 is determined by the limiter. Were the output limiting level to be increased from 100 to 110 dB SPL though, the output level for an input of 90 dB SPL of about 104 dB SPL would be determined by the WDRC compressor. In this case, reduction of the limiting setting by 6 dB or less would have no effect on the OSPL90. Furthermore, OSPL90 would not increase no matter how much the limiting setting increased. Consequently, if evaluation of OSPL90 indicates that it should be changed, use a test box, a real-ear analyzer, or the manufacturer's software to determine which control will be most effective in achieving the desired change without affecting the gain for typical input levels.

Unfortunately, it is not clear that freedom from discomfort in the clinic guarantees freedom from discomfort in real life[866] or vice versa.[545] The appropriateness of maximum output therefore should be further evaluated at the first follow-up by asking patients about their reactions to intense sounds they have experienced in their home environments (see panel).

It would be unwise to assume that patients become used to sounds that they initially describe as too loud. There is conflicting evidence regarding an increase in loudness discomfort levels following experience with hearing aids.[168, 228, 1258, 1287] Even if loudness discomfort levels *do* increase significantly with aided listening experience, if the patient's initial experience of hearing aids is one of loudness discomfort, the hearing aids may be discarded long before any increase in loudness discomfort levels could occur.

Caution: Real-ear measurements at 90 dB SPL

If OSPL90 is too high, presentation of a 90 dB SPL input level could be an unpleasant initial hearing aid experience for the patient. Minimize this chance by:

- first running sweeps at 70 and 80 dB SPL;
- explaining that the sound should be loud, but should never cause discomfort; and
- reassuring the patient that you will stop the sweep immediately the patient so indicates.

Evaluating hearing aid maximum output

First, ensure that sounds do not cause discomfort.

1. Present several intense sounds to the patient. Include a pure tone or warble tone sweep at 80 or 90 dB SPL, supplemented by a few complex sounds with low- and high-frequency dominated spectra. These can be generated by the clinician; their exact level does not matter provided they are intense enough to saturate the hearing aids. Suitably high-level sounds can easily be made by hitting a cup with a spoon, clapping hands, rattling a metal can containing nuts and bolts, and by speaking loudly close to the person (or by using recorded speech). In each case explain to the patient that you are about to make sounds that *should be loud*, but not uncomfortable, and that you need to determine their loudness so that that you can properly adjust the hearing aids.
2. After you have explained what you are about to do, position yourself such that the patient can see you making the loud sound, so that it does not take him or her by surprise. Ask the patient to rate the loudness, using any of the loudness scales given in Section 10.4. Also carefully watch the patient's expression while the sound occurs. Follow up, with further questioning or instruction, any apparent contradiction between the rating given and the accompanying expression (e.g. a rating of only loud accompanied by a visible flinching or eye-blink, or a rating of uncomfortably loud said without any apparent concern). It should be possible for all the sounds (sweep, impulse sound, hand clapping, and speech) to produce a rating of loud but OK or very loud without any of them being rated as uncomfortably loud.
3. For bilateral fittings, these sounds should be made with both hearing aids turned on, and it may be necessary to also test each aid separately.
4. If the patient has already been wearing the hearing aids, ask if any sounds have been so loud that they were uncomfortable, have jarred the patient, have made the patient want to remove the hearing aids, or have given the patient a headache.

Second, ensure that maximum output is large enough.

1. Play speech (preferably continuous discourse) at approximately 80 dB SPL and ensure that speech is at least *loud* (this is actually done at the same time as step 1 above).
2. If the patient has already been wearing the hearing aids, ask if there have been any situations in which the level of background noise seems to increase markedly whenever someone stops talking, or in which things that should differ in loudness seem to be equally loud. (But beware, these are also symptoms of an excessively high compression ratio in a wide dynamic range compression system.)

11.8 Concluding Comments

The second half of this chapter describes procedures that enable the real-ear gain of a hearing aid to match a target and perform electro-acoustically as expected. It will be apparent that there is no single right way to do this. Some procedures, however, take more time than do others. It is worth reviewing one's procedures to ensure that no unnecessary or inefficient steps are being performed.

Finally, it is important to remember that matching a prescription target is only an intermediate goal within the whole rehabilitation process. The ultimate aim is for the hearing aid to provide the clearest possible speech combined with good sound quality and an acceptable loudness in a range of listening situations. Fine-tuning and troubleshooting methods to achieve this are covered in the next chapter.

CHAPTER 12

PROBLEM SOLVING AND FINE-TUNING

Synopsis

Many hearing aid fittings need to be fine-tuned, either electronically or physically, after the patient has had a week or two to try the hearing aids. When a patient has trouble managing hearing aids (inserting, removing, using the controls, changing the battery), re-instructing the patient may solve the problem. If not, the hearing aid should be physically modified, or if necessary, a different style chosen. Physical modification will also be necessary when a patient is suffering discomfort from the earmold, shell or case, or when the hearing aid works its way out of the ear.

Feedback oscillation has several potential solutions: reducing gain at selected frequencies; reducing the vent size; making a tighter earmold or shell; or changing the hearing aid to one that has more effective feedback canceling and management algorithms.

Complaints about the patient's own-voice quality are particularly common. The most common cause is physical blocking of the ear canal, so the best cure is to add a vent, or increase the size of an existing vent, including using an open-fitting. Where feedback oscillation precludes that, the earmold or shell can be remade with the canal stalk extended down to the bony canal, preferably using a soft material. Own-voice problems are sometimes caused, and cured, by electronic variation of the gain-frequency response for high-level sounds.

Complaints about the tonal quality of amplified sounds are fixed by changing the balance of low-, mid-, and high-frequency gain. The hard part is knowing when to ask the patient to persevere with a gain-frequency response in the expectation that it will eventually become the preferred response, and confer maximum benefit to the patient.

When a patient complains about the clarity or loudness of speech, or the loudness of background noise, he/she must be questioned particularly carefully so that the acoustic characteristics of the sounds causing the problems can be identified. The clinician's first aim is to identify whether it is the gain for low or high frequencies, and the gain for low, mid, or high levels, that should be adjusted. Only then can the appropriate hearing aid controls be adjusted.

In those cases where it is not clear which control should be adjusted, or by how much it should be adjusted, a systematic fine-tuning can be performed using one of two general methods. The first of these is paired comparisons, in which the patient is asked to choose between two amplification characteristics presented in quick succession. Multiple characteristics can be compared by arranging them in pairs. Paired comparisons can be used to adaptively fine-tune a hearing aid control if the settings compared in each trial are based on the patient's preference in the preceding trial.

The second general method for fine-tuning relies on the patient making an absolute rating of sound quality. The best amplification characteristic (out of those compared) is simply the characteristic that is given the highest rating by the patient. The absolute rating method can also be used to adaptively alter a chosen hearing aid control. This is achieved by deciding on a target rating (e.g. just right) and adjusting a control in the direction indicated by the patient's rating (e.g. too shrill, or too dull).

The paired comparisons and absolute rating methods are best carried out while the patient listens to continuous discourse speech material, or other sounds they are complaining about. Depending on the complaint being investigated, this can be supplemented with recordings of commonly encountered background noises. The paired comparisons method is more sensitive when the differences between the conditions are small.

Fine-tuning is usually carried out only for patients dissatisfied with the prescribed response, but can be used for all patients if desired.

This chapter describes techniques that help the clinician adjust hearing aids on the basis of the patient's comments and preferences. Such adjustment will sometimes involve moving the amplification characteristics *away* from the prescribed response that has so carefully been achieved. Some of the methods described in this chapter can be used when hearing aids are first provided, but more often they will be used after patients have worn their hearing aids for a week or two.

Although the chapter is focused on fine-tuning the amplification characteristics of hearing aids, the clinician should realize that sometimes listening and talking are all that is required, and/or all that can be done. People with a complaint want to be listened to, and sometimes active listening, plus the provision of additional information (see Chapters 9 and 13) can change a patient's expectations such that what was perceived as a major problem is now perceived as normal. That is, the clinician has to be able to fine-tune patients' expectations as well as hearing aids.

12.1 Solving Common Problems

After the hearing aid is fitted and the prescription target has been met as well as it can be, the time of problem solving begins. For some patients there will be many problems to overcome before they can gain significant benefit from their hearing aids; for others there may be none, particularly if the initial selection has wisely considered the patient's capabilities as well as his or wishes. This section describes several common problems and some potential solutions.

12.1.1 Management difficulties

Clients may have difficulty inserting the aid, removing it, switching it on and off, varying the volume control, or changing the battery. For any of these problems, the first thing to try is further training. The ultimate solution, in some cases, is to train a frequent caregiver to do the task instead of the patient. Training support staff in an aged-care facility, although necessary, may have only short-term effects unless there is some mechanism by which new staff members are trained as they take up duty.

Some more specific solutions are as follows. In every case, the first step is to closely observe the patient trying to perform the task, so that the clinician can identify precisely which part(s) of which operation(s) the patient is unable to perform.

Difficulty inserting an earmold or ear shell

Options include:

- If the patient picks up the hearing aid differently each time, or in an inappropriate manner, the patient has to be taught landmarks on the aid or earmold and a specific grip, and to have the procedure broken down into steps for them, which can even be written down using the patient's own words.
- If the patient is unable to insert the helix-lock of the earmold or ITE fully into the cymba portion of the concha, it may be necessary to remove the helix lock entirely.
- Similarly, if the patient is unable to get a BTE earmold under the anti-helix, part of the earmold's conchal rim (see Figure 5.3) may have to be removed, turning a skeleton into a semi-skeleton, for instance.
- If the earmold or shell is a tight or tortuous fit, a lubricant (water-based for safety of the ear) may have to be applied every time the aid is inserted until the patient is more practiced and/or the ear shape adapts to the aid. Alternatively, the earmold or shell (but not the aperturic seal) can be trimmed if feedback oscillation is not likely to result.
- The patient may have to pull the pinna upwards and outwards with the opposite hand, while inserting the hearing aid or earmold with the ipsilateral hand.

Difficulty locating or using a control

If re-training is not successful, the hearing aid may have to be modified or replaced:

- Volume controls on ITE and ITC hearing aids can be made more prominent with add-on caps.
- If the patient is confusing one hearing aid tactile feature (e.g. a program switch) with another feature (e.g. a volume control), one of the controls may have to be removed so that the more important control (whichever it may be for that patient) can be operated. Some controls can be removed cleanly with electrical wire cutters.
- The compression ratio can be increased, reducing or eliminating the need for a volume control, but potentially degrading sound quality.

- A remote control can be added, which might require fitting a different hearing aid model.

Difficulty removing a hearing aid

Options include:

- If the patient cannot grasp the hearing aid or earmold, a removal handle or line should be added, or a different hearing aid style used.
- If the patient can grasp the hearing aid but not remove it, and cannot be trained to use an appropriate twisting motion, parts of the earmold or shell will have to be removed, or a flexible dome-type ear fitting used instead.

Difficulty changing the battery

Options include:

- Coloring one side of the battery slot to lessen problems with battery reversal.
- Using a tool to open the battery compartment.
- Using a magnetic tool to hold the battery.
- Re-fitting with a hearing aid that has a bigger battery or a battery compartment that is easier to open or visualize.
- Teaching the patient to distinguish the positive side of the battery tactually rather than visually, or vice versa. The removable tab can help with either of these approaches.
- Fitting a hearing aid that has a built-in re-chargeable battery.

12.1.2 Earmold or earshell discomfort

Earmolds, earshells, and BTE hearing aids can all cause physical discomfort to the external ear if they apply excessive pressure at any point. The problem is diagnosed by asking the patient where it hurts, and by otoscopic or other visual examination of the affected area to look for inflammation. This diagnosis is easier if the patient wears the hearing aid for as long as he or she can reasonably stand the day before, or immediately before, the follow-up appointment. The usual solution to the problem is to grind away, and then polish, the area of the earmold or shell that causes the problem.

For CIC hearing aids, discomfort can also be caused by a hearing aid that is too loose. Discomfort can occur if the patient frequently pushes the hearing aid in further than it was intended to go, in an effort to retain it in the ear or to prevent feedback.[1153] Martin & Pirzanski (1998) make the useful analogy with shoes: they cause sore feet whether they are too large or too small. If this is the cause of discomfort, the problems should be viewed as one of poor retention (see next section).

For BTE hearing aids, an incorrectly cut tubing length can create excessive pressure, as shown in Figure 12.1. Pressure spots can arise if the patient has been wearing an earmold only partially inserted. Most commonly, this is because the helix lock has not been properly tucked in.

More generalized inflammation can be caused by an allergic reaction (Section 5.9.2), but this is more rare. The solution is to re-make the earmold or earshell with a different material, or to coat the earmold or earshell with material that the patient is hopefully not allergic to. Another problem, also rare, is that of the patient trying to use a hearing aid that has been made for his or her other ear, or that has been made for another person's ear! Such a mix-up can happen while the hearing aids or earmolds are being manufactured, or at any time subsequent to the fitting.

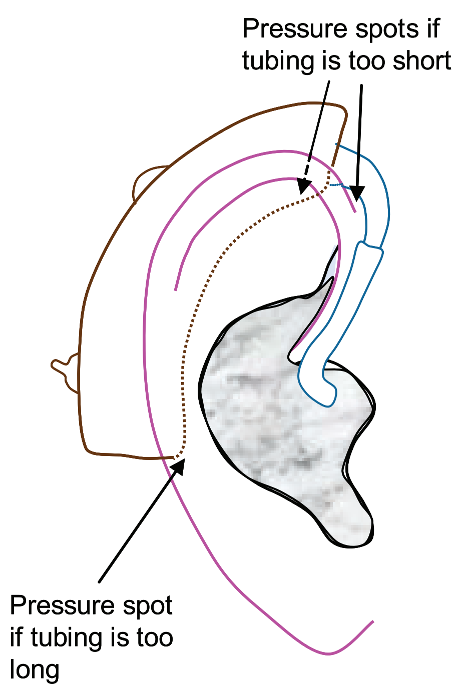

Figure 12.1 Excessive pressure caused by earmold tubing being cut too short or too long.

12.1.3 Poor earmold or earshell retention

Hearing aids, particularly CIC and ITC styles, can sometimes fall out of the ear. Movement of the patient's jaw when he/she is talking, yawning or chewing can move the ear canal walls sufficiently to *push* the hearing aid out of the ear. Solutions include:

- Remaking the hearing aid in a style that has better retention properties. For example, an ITC could be used instead of a CIC, or a low profile ITE could be used instead of an ITC.
- Remaking the earmold or shell with a longer canal portion, and/or with a helix lock. The latter, in the form of a thin rim, can even be added to CIC and ITC hearing aids.
- Remaking the earmold or shell, and taking the impression with medium viscosity material while the patient's jaw is open, so that the canal width is greater in the flexible part of the canal (Section 5.8.2). The impression should extend beyond the second bend, even if the hearing aid will not be inserted this deeply.[1420]

12.1.4 Own voice quality and occlusion

Any of the following descriptions by a hearing aid wearer about his or her own voice indicate that the spectrum of his or her own voice in the ear canal is inappropriate: *hollow; boomy; echoes*; *like speaking in a barrel, tunnel or well*; *like having a cold; or feeling plugged*. Because most people are not able to clearly describe different types of spectral emphasis (e.g. insufficient or excessive low-, mid- or high-frequency emphasis), all we can conclude from an adverse description of a patient's own voice is that there is something the patient does not like about the way it is being amplified. If the patient reports that other people's voices also do not sound good, that problem should be fixed first, as it may also solve the own-voice problem (Sections 12.1.6 and 12.2).

Assuming that other people's voices sound fine, the patient's own voice may sound unpleasant to the patient because of any of the following reasons, the first of which is the most likely.

The earmold or earshell is excessively blocking the ear canal

As discussed in Section 5.3.2, blocking the canal within the cartilaginous section will allow the walls of the canal to vibrate with respect to each other, and hence generate a high sound level in the residual part of the canal that they enclose. For low-frequency sounds, this causes the SPL at the eardrum to increase by up to 30 dB (but more typically by 15 dB) relative to that which would occur for an open ear canal. The problems can be diagnosed and solved, by:

- Increasing the area, and/or decreasing the length, of the vent (but review Section 5.3 to ensure that the way you change a vent significantly affects its acoustic properties). An extreme example of an open vent is an open dome fitting, which will certainly avoid any occlusion-induced build up of low-frequency sound.
- Making an earmold or earshell with a canal stalk long enough to extend into the bony part of the canal. Probable difficulties with this solution include increased difficulty with insertion and removal, and decreased comfort. These can be helped by constructing the tip (at least) of the earmold or shell out of soft material. The problem with this is that the life of the earmold or shell is likely to be decreased, because soft materials deteriorate in appearance and cleanliness more quickly than hard materials. Disposable soft tips are available. Invention of a very soft but non-porous material would be welcome! There is no point in extending the earmold or shell into the bony portion of the canal unless the extended section material makes good contact with the walls of the canal. In fact, failure to make good contact can increase the level of occlusion sound generated because it decreases the residual volume without suppressing the source of the vibration.
- Electronic cancellation of occlusion-generated sounds should become available in the near future (Section 8.5).

The hearing aid is distorting when the patient speaks

The proximity of the mouth to the ear causes the input level to the aid to be higher when the patient speaks than when other people speak, especially if the patient has a loud voice. The possibility of distortion being a cause of poor own-voice quality can be tested by letting the patient hear, and rate the quality of, another person's loud speech, using either the clinician's live voice or recorded speech as the signal. A presentation level of 80 to 85 dB SPL at the person's ear would be representative of own-voice levels. Alternatively, the distortion for high input levels (such as 85 dB SPL) can be measured in a test box. If distortion is a problem, the solution is to use a hearing aid that does not distort at high input levels (Section 10.6.1).

Although an aid wearer is likely to prefer less gain while listening to his or her own voice than when listening to other people,[1010] this should automatically occur in an aid with WDRC. The WDRC compressor will provide lower gain for the aid wearer's voice (typically around 80 dB SPL) than it does for the voice of other people (typically around 65 dB SPL). Apart from its effect on the aid wearer's own voice, a hearing aid with excessive distortion might also lead to complaints about music quality or the naturalness of other people's voices.[813] Adjustment possibilities to improve high-level own-voice distortion include increasing OSPL90 and varying the compression ratio of the WDRC compressor.

The hearing aid amplifier is excessively amplifying low-frequency sounds

The mouth radiates high-frequency sounds forward more than to the side. Also, low-frequency sounds travel around a barrier (the head) more readily than do high-frequency sounds. As a result, the spectrum of the aid wearer's voice near his or her own ear will be more heavily weighted to low frequencies than will anybody else's voice.[326] This bass boost occurs for everybody, but if the hearing aid is also excessively amplifying low frequencies, the combined effect can be a poor own-voice quality, even though the quality of other people's voices is not too bad.[1015] The problem is diagnosed by decreasing the low-frequency gain of the hearing aid for high-level sounds and seeing if the problem disappears, but this may conflict with the processing that is best for amplifying other sounds. Remember that if the hearing aid uses an open fitting, there is no point even thinking about low-frequency gain, because the hearing aid will have 0 dB gain for low frequencies, no matter how the hearing aid electronics are altered.

The patient has forgotten what his or her own voice should sound like

Because a frequency-dependent hearing loss affects the tonal quality of everything perceived by an unaided person, the new hearing aid wearer may have forgotten what his or her voice should sound like. This is a justification for the oft-repeated instruction *you will get used to the sound of your own voice*. In the author's opinion, this explanation is an unlikely reason for own voice complaints,[a] but there are no data on this question.

12.1.5 Feedback oscillation

Feedback oscillation may cause patients to report any of the following:

- The volume control cannot be increased to the desired level without whistling occurring.
- Whistling occurs whenever they chew, talk, wear a hat, or put their hand or a telephone near their ear.
- The hearing aid makes a brief ringing noise whenever certain sounds occur. This is the effect of sub-oscillatory feedback (Section 4.7.2) or of a feedback canceller operating.
- The hearing aid whistles when they are in a quiet place but stops when a noise occurs. This happens because WDRC causes the gain to increase in quiet places.
- The hearing aid appears to stop working or becomes weak or distorted. This observation could come from a person with severe or profound hearing loss at high frequencies who is unable to hear the feedback oscillation itself, but can hear the gain reduction that the oscillation causes.

As explained in Section 4.7, all of these problems indicate that too much sound is leaking from the ear canal to the microphone via some path. Tables 4.13 and 4.14 show how to diagnose the source of the leakage. Assuming that the hearing aid is not faulty, one of the following solutions should be tried. All have potential disadvantages, and all except the first three involve additional appointments and expense.

- Ensure that there are no excessive peaks in the real-ear aided gain (REAG) curve. If so, damp them, which is easiest, but also most necessary, for a BTE.
  Disadvantage: the peak in the REAG curve may be necessary to achieve the desired insertion gain curve.

[a] People who most complain about own-voice quality have good low-frequency hearing and poor high-frequency hearing. Such people would be used to hearing a treble-deficient voice when listening unaided, and yet the problem is usually solved by venting, which cuts the bass of the spectrum at their eardrum.

- If the hearing aid is vented, decrease the size of the vent with a vent insert or sealing material. Equivalently, change from an open dome to a closed dome canal fitting.
  Disadvantage: may cause or exacerbate the occlusion effect. The low-frequency gain of the hearing aid may need to be reduced to offset the increase in low-frequency gain that reducing the vent diameter will have caused.
- Decrease the high-frequency gain of the hearing aid, or for a multichannel nonlinear aid, decrease the high-frequency compression ratio or increase the high-frequency compression threshold in the relevant channel (Section 8.2.1). Some hearing aid fitting software can perform a test that identifies which channels are likely to be problematic.
  Disadvantages: may decrease intelligibility or sound quality, particularly for soft sounds.
- Re-make or re-coat the earmold or shell so that there is less leakage between the mold/shell and the walls of the canal. Use an open-jaw impression technique if a re-make is necessary.
  Disadvantages: additional time and expense, potential occlusion effect, potential earmold/shell discomfort, and uncertainty of outcome.
- Change to a hearing aid that has a more effective feedback cancelling algorithm (Section 8.2.3), or increase the strength of the algorithm in the current hearing aid if a choice is available. Large differences in effectiveness between the algorithms from different manufacturers have been reported, usually by the manufacturer who has the best system at any one time. Transitory feedback-like sounds can be caused by an over-active feedback reduction system. Alternatively, or additionally, change to a hearing aid with frequency lowering to achieve its feedback reduction advantages.
  Disadvantages: additional time and expense, and uncertainty about effectiveness until after the alternative hearing aid has been tried.
- Change to a hearing aid that better enables feedback to be managed by reducing the gain at those input levels and/or frequencies that are giving rise to feedback oscillation (Section 8.2.1).
  Disadvantages: additional time and expense, uncertainty about effectiveness until after the alternative hearing aid has been tried, and reduced intelligibility at low input levels caused by the reduced high-frequency gains for low input levels.

Clarifying the effects of changes

After making a change to a hearing aid's electroacoustics or physical fit, don't ask the question: *Is this better?* Many patients are just as eager to please the clinician as the clinician is to please the client, and may answer "yes" if they are uncertain about the effect of the change. Instead, ask the question in a more balanced manner, such as:

- Is this better or worse?
- Is this any different?
- I'm going to play two sounds; I want you to say which is better.

In effect, the clinician is performing a mini-experiment every time the hearing aid is fine-tuned, and as in any experiment, procedures are needed to ensure that the experimental results are reliable. Inappropriate fine-tuning may create the need for future appointments to solve the problems it creates.

It is possible to view as an advantage the ease with which the patient's assessment of the sound quality is affected by what they have been told or asked, rather than as a problem to work around. On average, patients who are told that one hearing aid is better than another one, subsequently report that to be the case, even if the hearing aids are identical.[108, 406a] Perhaps confident fine tuning by the clinician, combined with an affirmation of improvement by the client will sometimes be all that is needed to create a long-lasting perceived improvement. More research on the usefulness of placebo effects in achieving acceptance of hearing aids is needed, however, before we can rely on them.

Achieving enough high-frequency gain to meet the prescription target (and hence obtain good speech intelligibility), while having the vent large enough to avoid occlusion, and simultaneously avoiding feedback oscillation, is technically the most difficult part of hearing aid fitting. For people with normal low-frequency hearing and a substantial, but aidable high-frequency loss, compromise solutions are available only because of the growing effectiveness of feedback canceling algorithms.

12.1.6 Tonal quality

Patients may describe the quality of speech and other wanted sounds in a wide range of ways. (Reaction to noise and unwanted sounds will be considered in the next section.) Excessive high-frequency amplification or insufficient low-frequency amplification (compared to their preferred response) may be described as being *shrill, harsh, hissy, sharp, metallic*, or *tinny.* Excessive low-frequency amplification or insufficient high-frequency amplification may be described as *muffled*, *unclear, boomy or dull*. Patients are more likely to notice or adversely comment about excessive high-frequency emphasis than about insufficient high-frequency emphasis,[220, 867, 872] probably because most patients are used to having deficient high-frequency audibility when they are unaided. Solving these complaints by changing the balance of high- to low-frequency gain is complicated by three factors.

One complication is that an excessively peaky gain curve can produce similar comments, even if the overall balance of low- to high-frequency gain is optimal. The solution to this problem is not to let it develop in the first place. Hearing aid fitting and verification should have included measurement of real-ear gain, and this will have revealed a peaky response if it existed. A peaky response should be dealt with immediately through a suitable combination of filtering and damping. Changing the damping for standard tube BTE fittings is easy because the dampers can be added to the sound tubing and can be placed in most earhooks (Section 5.5). ITE, ITC and CIC receivers can also be damped, but this is most conveniently done at the time the hearing aid is manufactured.

The second complication is that the tonal quality may be unsatisfactory only for low-level, or only for high-level sounds, or may apply across all input levels. This is easily dealt with by appropriate questioning and choice of which controls to adjust (see the Troubleshooting Two-Step Panel).

The third complication is much harder to deal with. Suppose a patient complains about the shrill sound of a hearing aid that is providing the prescribed frequency response in a smooth manner. Is it because the patient knows better than the prescription procedure what is best, or is it because he or she has not yet become used to the high-frequency information that he or she has been deprived of for many years? Either of these could be true. People can take months to learn to fully use high-frequency information that they previously have not had.[591] This is known as the ***acclimatization effect*** (Section 14.7).

Initially, patients may choose an amplification characteristic that gives the greatest gain for the frequencies at which they have the least loss, presumably because they are most used to hearing sounds at these frequencies.[1897] Patients with a high-frequency loss may be a little more likely to prefer high-frequency emphasis four weeks after fitting than at the initial fitting.[1520] A longer-term study, however, found no change in preference for high-frequency gain after 24 weeks of listening experience,[1282] so one certainly cannot assume that patients will come to prefer a response they initially do not like.

A compromise is to provide patients with a response that is mid-way between the response they prefer and the response that is believed to be best for them. The aim is to enable patients to gradually get used to a new response without subjecting them to a sound quality with which they are unwilling to persevere. There is no research addressing whether this is the best management option, but it seems like a very reasonable approach. If patients wear their hearing aids every day, and have not changed their minds about what they prefer within a month, it seems reasonable to give them the response they prefer. The considerations are similar for patients who are used to a peak clipping hearing aid, and are changed over to compression limiting, or for patients who are used to linear amplification and are changed over to WDRC. There are numerous examples of people preferring a compression limiting hearing aid to a peak clipping aid after one to two weeks of use, even though they disliked the compression limiting aid at first.[407]

Trainable hearing aids provide a potential solution to this dilemma. Patients can directly train the degree of high frequency emphasis. If their preferred amplification characteristics do change following listening experience with more high frequency audibility than

Which hearing aid control will solve the problem? - The troubleshooting two-step

Choosing the correct hearing aid controls and adjusting them in the correct direction to solve a problem reported by the patient is potentially very difficult. Use the following approach to break the problem down to two more easily handled tasks.

1. Express the desired change simply. Question the patient until you understand the problem well enough to make a statement like: *I want to decrease the gain applied to low-frequency, high-level sounds.* In any particular case, the words *low-frequency* might be *high-frequency*, or may be absent altogether. That is, you may want to change the gain for high-level sounds at all frequencies. Similarly, the words *high-level* might be *mid-level*, or *low-level* or might be absent. Mentally picture the following table. Your sole aim is to determine which of the gains you need to adjust, and in which direction. This implies that you are also determining which gains are correct and should *not* be altered. If the patient cannot describe the problem and the types of sounds causing the problem well enough for you to know which gains you are targeting, you will have to present sounds in the clinic to gather further information. These sounds might be speech presented at a low-, medium-, or high-intensity, or background noises. Suitable noises can be found on various compact disks and in some fitting software. *Notice that this first step is the same no matter what hearing aid the patient is wearing, and no matter what controls or number of channels the hearing aid has.*

Low-frequency, high-level gain	High-frequency, high-level gain
Low-frequency, mid-level gain	High-frequency, mid-level gain
Low-frequency, low-level gain	High-frequency, low-level gain

2. Identify the controls and the direction of change needed. In this step you can forget the patient's complaint and concentrate on the complexities of the particular hearing aid. Unless the hearing aid controls are labeled *low-frequency, high-level gain* (etc.), you will need to understand how each control (e.g. compression ratio, compression threshold, gain, UCL offset) affects each of the gains in the table above. Unfortunately, this varies from hearing aid to hearing aid, so you will need to acquire this knowledge for each type of hearing aid that you use. The knowledge can be obtained from specification sheets, or by measuring a hearing aid in a test box, and altering each control in turn. It may be helpful to sketch some I-O curves showing how the curves alter as each control is varied. Figure 12.2, for example, shows two ways in which an I-O curve might alter as a compression ratio control is varied. In one aid the low-level gain remains constant, but in the other the high-level gain remains constant. Increasingly, fitting software has controls labeled like those in the table above, rather than in terms of compression ratio or compression threshold, so this second step does not need to be performed.

Example. Suppose that a patient complains that crockery noise is too loud, but that most other sounds are fine. In step 1 we deduce that we wish to decrease the high-frequency, high-level gain, but leave the mid-level gain the same at all frequencies (because conversational speech was comfortably loud with a tone quality that was neither hissy nor boomy). Suppose that the patient is wearing an eight-channel hearing aid, and that each channel has a selectable compression ratio, which applies to all levels above a fixed compression threshold of, say, 40 dB SPL. To alter the high-level gain while leaving the mid-level gain the same, we need to adjust compression ratio in the high-frequency channels. Depending on whether the I-O curve pivots around low-, mid- or high-level inputs as this control is adjusted, we may also need to adjust the overall gain for high-frequency sounds. Some compromises may be necessary. When we have achieved our aim of reducing the high-level gain but leaving the mid-level gain unchanged, we will find that we have inadvertently increased the low-level gain. Hopefully, this increased low-level gain will be acceptable to the patient. If not, some acceptable compromise between the low-, mid- and high- level gain applied to high frequencies will have to be found.

Some software adjusts the hearing directly based simply on the clinician selecting options like "traffic noise is too loud". This has the advantage that no one knows the effects of each hearing aid control better than the manufacturer. It has the disadvantage that the clinician has little or no knowledge of what they are doing to the hearing aid, and therefore what other sounds or situations might be adversely affected.

they had become used to without hearing aids, or with older, restricted bandwidth hearing aids, their gradually changing preferences will direct the hearing aid towards greater high-frequency emphasis. Because this training requires just the adjustment of a control by the patient while the patient is listening in his or her usual listening situation, the aid can be trained at any time, and can be re-trained months later if the patient's preferences change. Because responsibility for this aspect of fine-tuning has been transferred to the patient, which patients appreciate,[1580] no clinical time is required.

Some clinicians have reportedly been concerned that patients may always prefer a response that mimics their previous hearing aid, and hence train their aids away from a potentially better solution. It does not seem likely, however, that current preference is totally dependent on past experience, but future preference is unaffected by current experience. Some trainable aids may give the clinician the ability to determine the degree to which the patient is able to train the hearing aid away from the response that the clinician considers (rightly or wrongly) is best for the patient.

Another potential solution, available in some hearing aids, is for the gain to automatically, but very gradually, increase in the months following fitting. More generally, the shape of the gain-frequency response could also increase. Such an ***automatic adaptation*** hearing aid could be programmed to gradually increase its gain for low level sounds, and/or its degree of high-frequency emphasis, to provide a smooth transition towards what is believed to be best for the client, but starting from something that is closer to the high-frequency deficient, weak sound inaudibility to which they have become accustomed. There does not seem to be any reason why a trainable aid algorithm and an automatic acclimatization algorithm could not be combined within the same hearing aid.

12.1.7 Noise, clarity, and loudness

Many, many adverse comments by patients will mention noise and/or excessive or insufficient loudness. These can have a multitude of causes, and each cause has a different solution. Careful questioning of the patient is essential to make sure that patient and clinician are discussing the same problem before taking corrective action. Although excessive amplification of noise, inadequate speech clarity, and inappropriate loudness of wanted signals are different phenomena, they are combined in this section because solutions to one problem may affect the others. The total picture should be considered before taking any action.

It is essential to determine whether the patient is unhappy with the loudness of weak sounds, medium-level sounds, or intense sounds, and whether the offending sounds are sounds that the patient wants to hear, including speech, or other sounds. For sophisticated hearing aids that automatically change programs depending on the environment, it will be necessary to deduce which program the hearing aid is likely to be in when the adverse sound quality occurs. In the following, we will assume that the hearing aid has been adjusted so that typical, mid-level speech is at a comfortable level for the patient.

The hearing aid is noisy in quiet places

This complaint may be an indication that internal hearing aid noise is being amplified sufficiently to be audible in quiet places. It may also indicate that noises in the environment are being amplified and the patient has not realized that these are noises that are present and can be heard by people with normal hearing. Listen to the aid yourself, and note whether the noise level changes when you block the microphone port(s) with your finger or some putty. Your aim is to diagnose the noise source as being either internal or external to the hearing aid.

If the problem is amplification of low-level sounds in the environment, identify the noise source to the patient and explain that normal-hearing people can also hear these sounds and that they are part of the richness of life. Also explain that the sounds may become less loud and less noticeable as the patient becomes used to the sounds being there.[1262] Let the person know that the loudness of these sounds can be decreased if the patient really wants it, but that many people come to value being able to hear these sounds when they need to. Mueller and Powers (2001) recommend giving patients a handout that says: *You have to hear what you don't want to hear to know what you don't want to hear.* If complaints persist after further use, it will be necessary to decrease the gain the hearing aid provides for low-level sounds by raising the compression threshold or by decreasing the compression ratio. In the fitting software of many manufacturers, there are adjustment handles that affect the gain for low-level sounds, and possibly several of these controls, each affecting a different frequency region. Adjusting these will likely cause variations in output level like those shown in Figure 12.2(b), but may instead alter the compression threshold.

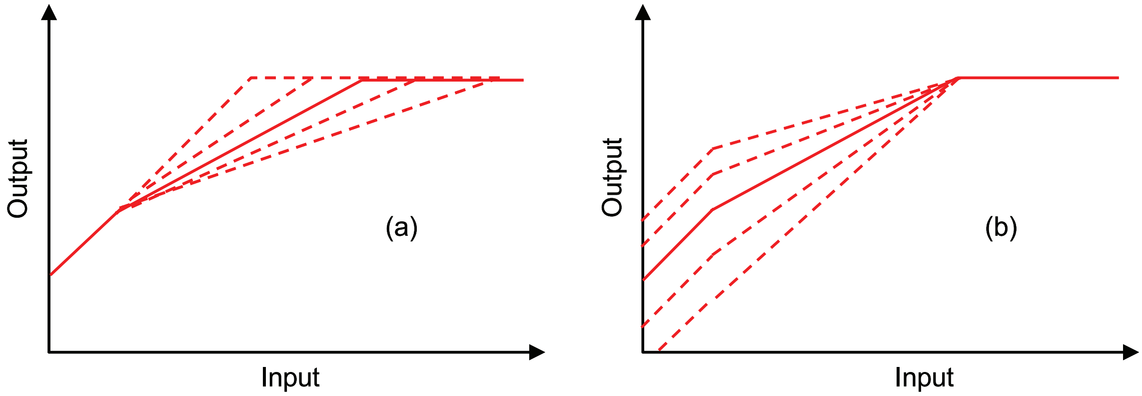

Figure 12.2 Variation of the I-O curve as the compression ratio control is varied for two different hearing aids.

If the problem is internal noise, ensure that the aid is within specification by measuring its noise in a test box. (Not a straightforward measurement – see Section 4.1.7). If the aid is within specifications, decrease the low-level gain (which may require the compression threshold to be increased) or by introducing low-level squelch (i.e. expansion) if it is available. The disadvantages are that the patient will not be able to hear wanted low-level sounds, like soft speech, as well as he or she now does, and that a more expensive hearing aid that has these features may be needed. Internal noise is most likely to be heard at the frequencies where a patient has hearing thresholds that are close to normal. Internal noise in the low frequencies will not be a problem for open-fit devices, as the venting effect of the open fitting also attenuates low-frequency noise at the output of the hearing aid. If the patient is concerned by tinnitus, then audible internal noise may be an advantage, not a disadvantage. Its level can be increased by increasing low-level gain.

Soft speech in quiet places cannot be understood

The solution to this problem is to provide more gain for low-level sounds. Potential difficulties include an increased likelihood of feedback and an increased likelihood that the hearing aid will amplify sounds that the person may rather not hear. It may be necessary to increase the gain in all channels, or it may be enough to increase the gain in only the low- or high-frequency channels. The frequency range requiring extra gain can be tested by having the patient comment on the audibility of speech sounds that rely on low-frequency cues (e.g., *moon*, *boom*) or that rely on high-frequency cues (e.g. /ʃ/, /s/). A low overall speech level of around 45 dB SPL should be used. Counseling, with the aim of lowering expectations, may be needed if the patient has a severe hearing loss and wishes to understand low-level speech without visual cues.

The hearing aid is sometimes too loud when noises occur

Comments similar to this require especially careful questioning.

If the noises ever get so uncomfortable that the patient <u>has</u> to immediately turn the volume control down or the hearing aid off, the OSPL90 of the hearing aid must be decreased. This can be done by electronic variation or by increasing the damping (in the case of a BTE), but the latter will also decrease the gain. The loudness may become uncomfortable only for sounds with significant high-frequency energy (like crockery or cutlery noise, paper rustling, brake squeals, or water flushing); it may occur only for low-frequency sounds (like traffic noise or a door slam); or it may occur for sounds with a wide range of spectral shapes. If the problems seem to stem from only one frequency region, it is necessary to decrease OSPL90 only in that region, assuming the hearing aid has that degree of flexibility. Exposing the patient to a pure tone sweep at 90 dB SPL, from a real-ear gain analyzer provides a rapid cross-check as to which frequency region is responsible for excessive loudness. See Section 11.7 for details on evaluating OSPL90. It may also be desirable to make the changes specified in the following point, if the hearing aid allows it.

If the patient <u>can</u> tolerate the noise, but would rather it was not so loud so often, a change to the input-output characteristics can improve the situation markedly. The compression ratio for input levels above about 65 dB SPL should be increased. This may require that the compression ratio also be increased for lower-level sounds. Ensure, however, that the output level remains comfortable for speech signals of around 65 dB SPL. In a multichannel hearing aid, the amount of compression could be increased in all channels. Starting from some reasonable prescription where different channels were prescribed different compression ratios, it is probably reasonable to increase all the compression ratios by the same percentage. For example, increase all compression ratios by 50%. If all the noises that are too loud are strongly high-frequency dominated, or strongly low-frequency dominated, it seems more reasonable to increase the compression ratio in only those channels causing the problem.

Explain to the patient that people with normal hearing find some sounds annoying and even uncomfortable, and that he/she may have become used to the decreased loudness that hearing loss causes. Hearing aids with appropriately adjusted OSPL90 and adaptive noise reduction algorithms should result in no more annoyance from loud sounds than that experienced by people with normal hearing.[1379] This level of annoyance is, however, greater than that to which the patient has probably become accustomed.

Background noise makes it hard to understand speech

If the primary complaint is not the *loudness* of the noise, but rather the effect the noise has on intelligibility, or the fatigue that is caused by trying to understand speech in the presence of the noise, the solution is different.

If the offending noise has a spectrum that is markedly different from that of speech, ensure that adaptive noise reduction is enabled, or if it is already enabled, increase its strength (an option from some manufacturers). Most hearing aids have adaptive noise reduction, but if the hearing aid does not have it, then instead change the compression and/or gain parameters as follows, preferably in a program that can be dedicated to noisy environments. If the offending noise is low-frequency weighted (traffic noise, most reverberation, some machinery noise), increasing the amount of low-frequency compression, or adding a simple low-frequency cut, will be helpful. If the noise is high-frequency weighted compared to speech (crockery or cutlery noise, paper rustling, brake squeals, water flushing or impact sounds), increasing the amount of high-frequency compression to achieve a TILL response, or a simple high-frequency cut will be helpful.

If the noise has a spectrum similar to that of speech, which unfortunately is usually the case, and the speech has a satisfactory loudness and tone quality, then intelligibility can be improved only by using an effective directional microphone, or by using a remote microphone with a wireless transmission system.

People in the distance are easier to understand than people nearby

The higher speech level from people nearby may be causing excess compression or even distortion. If this is the cause, then potential solutions are to increase the maximum output or change from peak clipping to compression limiting. Alternatively, the more distant people may be receiving excessive gain because of an excessively low compression threshold or an excessively high compression ratio, for which the solutions are obvious.[813] As a third alternative about which nothing can be done; some people have a voice that sounds like it could cut through steel, and they seem to be able to make themselves heard above others no matter where they are.

Noises levels rise and fall intermittently

Compression with release times from around 200 ms to 2 s can cause background noise levels to rise noticeably as the gain gradually rises during gaps within speech, or after brief impact sounds have forced the gain down. Technically this audible rise and fall in the noise level is called noise pumping, and some patients may even use this term.[813] The solution is to

Handling more than two channels

The procedures in this chapter concentrate on low versus high frequency, simply because it is relatively easy to identify low- versus high-frequency sounds. Many hearing aids have three or more channels. Any mid-frequency channels can be adjusted by amounts intermediate to the adjustments made to the extreme low- and high-frequency channels. For advanced hearing aids with many (up to 20) channels, there will likely be adjustment handles for just a few widely spaced frequencies and the software will appropriately interpolate or extrapolate settings for each of the remaining channels.

use either faster compression (so that the rise in level seems instantaneous) or slower compression (so that in brief gaps the rise in level is smaller, and in long gaps it occurs sufficiently slowly that the change is less noticeable).

12.2 Systematic Fine-tuning Procedures

The previous section described *which* hearing aid characteristics should be altered in response to specific adverse comments made by patients, but did not indicate by *how much* they should be varied. This section describes some systematic methods to improve on the initial prescription. Systematic fine-tuning procedures can be used for all patients (though they rarely are!), or can be used only to help solve problems as they arise. One advantage of routinely fine-tuning hearing aids is that no prescriptive procedure is perfect, and some patients are reticent to complain about sound quality no matter how bad it is. Furthermore, unless patients previously had hearing aids that provided better sound quality than their new hearing aids, they will not know that better sound is possible, and so will not know to complain. Even when the sound quality of a new hearing aid is very good, it is possible that some variation to the hearing aid characteristics could make it excellent.

Conversely, even though a patient complains about sound quality or clarity, there may be no setting or device that provides better sound, given the deficits in that patient's cochlea. Systematic fine-tuning can increase everyone's confidence that the best possible result has been achieved.

We will first review the basic methodology involved in performing paired comparisons, and absolute ratings of quality, and will then consider how these tools can be used to improve hearing aid fittings. They can be used to choose between responses that differ in any manner (Section 12.2.3) or to choose the best setting for any particular amplification control (Section 12.2.4).

12.2.1 Paired comparisons

Different response characteristics with similar perceptual effects can best be selected by allowing patients to choose between two alternative responses heard in quick succession. This process is referred to as ***paired comparisons***, and patients can simply be asked which of the two conditions they prefer. The ***response criterion*** can be made more explicit: patients can be asked which response they prefer on the basis of ***intelligibility***, ***comfort***, ***naturalness***, ***pleasantness***, minimizing ***annoyance*** of any noise present, or just about any other attribute of sound. In many cases, different response characteristics will be better for different response criteria.

Given the time pressures usually present in clinical settings, few clinicians will have time to use more than one criterion. If paired comparisons are being used to help address a specific problem, the criterion should be chosen to match that problem. For example, if the patient is complaining about the comfort of sound in noisy places, the criterion should be listening comfort. If the patient is complaining about the intelligibility of soft speech, the criterion should be intelligibility or clarity (to choose a more easily understood word). If in doubt as to the problem, or if the patient has several problems, the patient can simply be asked to choose the preferred response, with no specific criterion being mentioned.

Another key decision is what ***stimulus*** should be played to patients while they are choosing their preferred response characteristics. As with the response criterion, the stimulus chosen should be appropriate to the problem being addressed. If the patient is complaining about the effects of a certain type of background noise, there is little point in doing paired comparisons using speech material in quiet. Recordings of various noises are available on several compact discs and in manufacturer's software. If the problem relates to the clarity or tonal quality of speech, a speech stimulus has to be used. To administer the comparisons in the minimum time, speech should be continuously present. There thus seems little reason not to use ***continuous discourse*** as the primary stimulus. In many cases, patients will be complaining about the disturbing nature of noise when it occurs simultaneously with speech. It is then useful to be able to play continuous discourse combined with selected noises.

If you do not have access to recordings of a wide range of noises, the following set of five stimuli would allow you to assess hearing aid performance in such a way as to address many problems reported by patients.

- continuous discourse in quiet, with the ability to play it at 50, 65, and 80 dB SPL;
- continuous discourse in quiet by three quickly alternating talkers, speaking at levels of 55, 65 and 75 dB SPL respectively;

- continuous discourse at 80 dB SPL with a background noise containing high-frequency impact sounds (e.g. crockery noise) of 80 dB SPL;
- continuous discourse at 80 dB SPL with a background speech babble of 70 dB SPL; and
- continuous discourse at 80 dB SPL with a background noise dominated by low-frequency sound (e.g. traffic noise) of 80 dB SPL.

The stimuli are most useful if the speech and noise are recorded on separate channels, so that SNRs larger or smaller than those listed above can be selected when required.

The paired-comparison technique can be administered by programming the settings to be compared into different memories of the aid. The patient can then switch between memories as often as desired until he or she can say which (if either) is preferred. Patients usually take 10 to 30 seconds to make a judgment, although some may take a minute. The comparison is easiest (and quickest) for patients if the characteristics of any background noise do not change markedly during the comparison period.

Alternatively, the technique can be administered by the clinician changing the value of the parameter being investigated from time to time. The clinician indicates which response is currently selected by pointing to the letter ***A*** or ***B*** on a piece of card, or by holding up one or two fingers. The clinician can control the timing of each switch between programs, or the patient can indicate when a switch should occur. Comparisons can be made most accurately if the sounds being compared are not separated by more than one or two seconds of quiet. The value of paired comparisons is therefore limited if the responses being compared include adaptive features that take many seconds, or tens of seconds, to adapt to the input unless the software ensures that the responses are already adapted when the switch between programs occurs.

Finally, it is worth contrasting the paired-comparison technique to traditional measures of speech recognition in which the patient has to repeat or choose the syllable, word, or sentence that he or she perceived. Paired comparison judgments of intelligibility correlate with measured speech intelligibility, but when there are only small differences between speech identification performance with different hearing aids, paired comparisons provide a quicker and more reliable way to choose the best option out of several alternatives.[578, 1038, 1125, 1739] Furthermore, paired comparisons can assess aspects of the sound other than intelligibility. The disadvantage of paired comparisons is that it is not possible to discern what types of sounds the patient misperceives, but this is more of a limitation for research than for clinical practice.

12.2.2 Absolute rating of sound quality

One disadvantage of the paired-comparisons procedure is the procedure can never reveal how bad or good the sound quality is – just which of the amplification schemes compared is preferred, and potentially how much better it is than another scheme. Another disadvantage is that if there are many schemes to be compared, and even if some of these are expected to be much better or worse than the others, many comparisons of different pairs will still be needed to deduce which is the best response.

Both of these disadvantages can be overcome by obtaining ***absolute ratings of sound quality***. Patients can be presented with a simple scale like either of those shown in Figure 12.3. The labels marked on each scale are explained to the patient. Some patients will find the scale with numbers easier to use; others will like the one with words. Sounds are then presented, while the patient wears the hearing aid. The patient is asked to mark, or state the position, on the scale that corresponds to the sound quality just perceived. It is advisable to let the patient hear a selection, and

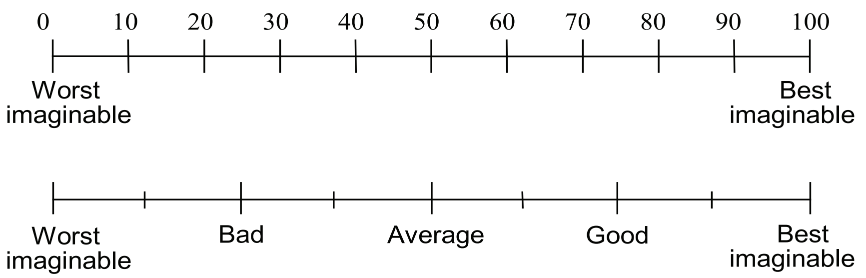

Figure 12.3 Two response scales used for obtaining absolute ratings of sound quality.

preferably the extremes, of the amplification conditions to be rated prior to obtaining the judgment for each condition. This makes it less likely that the patient will change his or her internal perceptual scale during the series of judgments.

For increased accuracy, each of the amplification conditions to be tested can be presented on multiple occasions, randomized with the other conditions being tested. Unfortunately, the time to do this is more likely to be found in a research setting than in a clinical setting. Presenting each condition several times is practical if there are less than ten conditions to be tested.

Absolute rating of sound quality is most useful when there are more than five or six conditions to be compared, or when it is expected that some of the conditions will be be much more acceptable than the others to the patient. Absolute ratings can then be used to weed out the amplification conditions that are rated poorly. If four or less of the conditions receive similar ratings, paired comparisons can then be used to select the most preferred condition.

It is common to have patients make absolute ratings about just one amplification condition at a time when they wear their hearing aids in real life. The judgments may be of tonal quality, or more generic judgments about hearing aid performance using self-report scales such as APHAB or COSI (Section 14.4). This is an insensitive and unreliable way to compare different hearing aid settings or different hearing aids.[1455] Their unsuitability is not surprising, as the comparison relies on patients adopting, and maintaining, over several weeks, the internal criteria used in making absolute ratings. Paired comparisons of two different amplification conditions, where the patient can switch back and forth between hearing aid settings in the same listening situation are much more sensitive and reliable.[1455]

Absolute judgments of different conditions made only a few seconds apart in the clinic are considerably more reliable and sensitive than speech intelligibility tests.[1472] Paired comparisons have been shown to be more sensitive than both absolute ratings and speech intelligibility tests in reliably choosing the better condition in the clinic, especially when the difference between the conditions is small.[506, 1739]

In principle, software-based fine tuning systems can combine the best of paired comparisons and absolute ratings. Several program labels can be displayed, and the patient can make an absolute rating after selecting and listening to each. If the previous ratings given are also displayed, the patient can explicitly consider whether to give each program a rating lower or higher than that of the other programs.

12.2.3 Systematic selection by paired comparisons

If there are just two different responses that one wishes to compare, the procedure described in Section 12.2.1 can be used to make that comparison. What if there are more, and we wish to find the best response? The best way to accomplish this depends on how many different responses we wish to compare, and whether we believe before we commence testing that one of the responses (e.g. the one prescribed by a trustworthy prescription procedure) is more likely to be preferred than the alternatives. In the following discussion we will assume that we have a ***baseline response***, which is the response that best matches some prescription. If we were not going to perform paired comparisons, this would be the procedure that we would fit to the patient because we believe it is *most likely* to be the best. There are at least three ways that we can organize the various comparisons to find the best response. Let us suppose that there are n responses to be compared.

Comparison to the baseline. Each response is paired, in turn, with the baseline response. Because we cannot have a great deal of confidence in the results of any individual trial, it will be necessary to compare each of the $n-1$ alternatives to the baseline response several times. Ideally, ten repetitions would be used, but more realistically, time might permit only four repetitions. This requires a total of $4(n-1)$ comparisons. For a baseline plus four alternative responses, this means 16 comparisons, which will take about eight minutes on average. If we have any *a priori* belief that the baseline response is right on average, we would not want to select one of the alternative responses unless it is chosen four times out of four in preference to the baseline.[b] If more than one of the responses is consistently preferred to the baseline, further comparisons

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[b] The consistency required before selecting an alternative to the baseline is a complicated balance involving the number of repetitions, the level of *a priori* confidence we have in the baseline, and the number of alternatives we are comparing to the baseline. Use the above recommendation unless you are confident about varying from it.

will be needed to choose among them. The chance of this happening increases with the number of alternatives. Because of this, and because of the total time needed, the comparison-to-the-baseline procedure is probably feasible only for five or less responses.

Round Robin. Each response is compared to each other, and the response that is preferred the most number of times is declared the winner and is permanently programmed into the patient's hearing aid. For reliable results, each response should be involved in about ten preference trials. If the procedure is to be carried out in 10 minutes, this limits the number of responses that can be used to about four. The Round Robin is particularly suitable for a small number of responses, especially where none of the responses is believed *a priori* likely to be better than the others. This can happen when we have already decided to put the baseline response into one hearing aid memory, and are using paired comparisons to decide what should go in the other memory or memories.

Tournament. The responses are organized in pairs, and each pair is compared, say three times. The winners are the ones that are preferred two or more times out of three. These winners advance to the next round, where they are again arranged into pairs. This continues, with half the contestants dropping out each round, until a single winner emerges (see Figure 12.4). The procedure requires between *2(n-1)* and *3(n-1)* comparisons (depending on the consistency of the answers). To be carried out in 10 minutes, the number of responses should be eight or less. It is also particularly well suited to situations in which there is no response thought likely to be better than the others. You will have to be well organized to keep track of which responses win and progress to each following round.

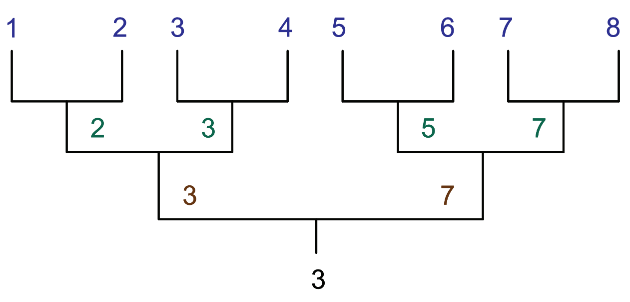

Figure 12.4 A tournament strategy for eight responses, with response number 3 being the eventual winner.

Irrespective of which of these three arrangements of pairs we use, how do we decide what the alternative responses should be? There is no set of rules for which parameters to vary, just a few guiding principles.

- The alternative responses should be potential winners. Do not waste time on a response that you strongly expect will be worse than the baseline response.
- Form alternative responses by varying the amplification parameter that you are least confident about prescribing accurately.
- Choose parameter values that are different enough to have clearly audible effects, with the stimulus you are using, but not so extreme that one or the other is very likely to be unsuitable.
- Alternative responses can vary from each other by only one amplification parameter (e.g. different values of compression ratio) or they can differ substantially by having different values for many amplification parameters.
- The alternatives should all be realistic amplification characteristics. If changing one variable (such as compression threshold) causes the gain applied for typical input levels to vary, it will be necessary to also change some other variable (such as overall gain) to compensate. There is no point in finding a winner under the conditions used for the paired-comparison trial if the patient would not be willing to use the program under typical conditions. (An exception would be if you were attempting to select a program for a second memory, to be used in specific circumstances, such as in very quiet or very noisy places.)

Irrespective of which strategy is used, each paired comparison judgment can be of three types:

- ***Forced choice***, in which the patients have to choose a response even if they consider they can't hear any difference between the two options. People underestimate their own ability to detect small differences and to reliably choose between two similar sounds. If enough repetitions are used, the consistency of the choices indicates whether the patient is making reliable choices or whether the patient is truly guessing.
- ***No-difference responses***, in which the patients can indicate if they can't hear any difference. In this option, no points are awarded for trials where no preference is indicated. Allowing a no-difference

response has been shown to increase the test-retest reliability of paired-comparison testing.[1467]

- ***Strength of preference***, in which the patients have to indicate not only which response is preferred, but how strongly they prefer it, such as *slightly better*, *better*, or *much better*. More points are awarded for strong preferences than for weak preferences. Preference strength judgments can allow, or not allow, no-difference responses.

There is a potential logistical difficulty with paired comparisons using multiple responses. The patient cannot make each comparison until the clinician has appropriately set all the controls for the two programs that the patient is about to compare. It is too time-consuming to change multiple control settings prior to each pair being compared. Thus, if it is necessary to alter more than one control to change from one response to the next, all the responses will have to be set up once at the start of the procedure, and stored somewhere. This can be inside the hearing aid (if it has enough memories) or inside the fitting software if it is configured to enable such comparisons. There is no reason why the number of memories in the fitting software has to be limited to the number in the hearing aid. The clinician should not attempt to compare more responses than the number that can be set up in advance and quickly recalled, with the exception of the special case considered in the following section.

Some manufacturers have made it particularly easy to administer paired-comparison testing by allowing the clinician or the patient to switch rapidly between many different amplification characteristics. These different characteristics may differ in only one amplification parameter or in many parameters. In one implementation, a very large number of alternatives are represented by different positions on a computer screen, such that characteristics differing only a little are represented near each other on the screen. The patient can use a mouse or touch screen to move around the space, and indicate which parts of the screen sound best.[9] The challenge is to ensure that a wide enough range of sounds is being listened to that the resulting program is good for a range of real-life situations.

12.2.4 Adaptive parameter adjustment by paired comparisons

A special application of paired comparisons is determining how much a single amplification parameter should be varied. We may strongly suspect, for instance, that we should increase the compression ratio, but by how much should it be increased? The hearing aid may enable a choice of many values above the one that appears to be unsuitable. Which should we choose? With such problems, our rationale is that some unknown setting of a control is best, and that some aspect of sound quality will deteriorate as the control is increased or decreased from this value.

The paired-comparison method is used ***adaptively*** to find the best setting of the control. We start by comparing two settings of the control. After each trial, the control setting that was not preferred is replaced with another value. The control is moved in the direction indicated by the winner of the most recent trial. Suppose a patient's difficulty in understanding soft speech led us to believe that a lower compression threshold in a single channel hearing aid would be better, and suppose that the hearing aid enables compression threshold to be set anywhere in the range from 30 to 70 dB SPL. Just as when we are finding an audiometric threshold, we must decide what the step size will be. Unlike threshold determination, there is no need to have different step sizes for ascending versus descending runs. In this example, we will assume that we use a step size of 10 dB.

Figure 12.5 shows the sequence of trials that might occur. Suppose the current compression threshold is 60 dB SPL and that the first comparison is with a lower compression threshold. The 50 dB threshold is

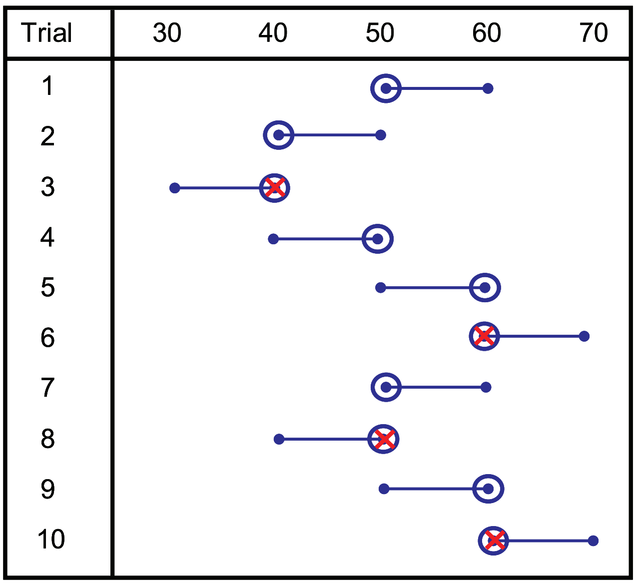

Figure 12.5 Worksheet for adaptive paired comparisons of different compression thresholds. Circles show the winner of each trial, and red crosses mark the reversals.

preferred (as indicated by the circle), so in the next trial, this is compared to an even lower threshold. You will notice that the winners of trials 3, 6, 8 and 10 are marked with a cross. These are the values of the winners when a ***reversal*** occurred. The comparisons are continued until four reversals have occurred. The values at these reversals are then averaged to give the final setting, in this case, 52.5 dB SPL. The more reversals that are used, the greater will be the precision of the procedure, but the longer it will take.[c]

The key to using this method is to choose the step size wisely. The step size must be large enough to make a perceptible difference to the sound quality. If the step size is too small, reversals will occur randomly, and the final answer will also be a random number unless a huge number of reversals is used. It is better to err on the side of making the step size too big. If it is too big, four reversals will be obtained with very few trials, and if it was considered worthwhile, further trials could be done with a smaller step size starting from the result found in the first series of trials.

If patients say they cannot hear any differences between the pairs of sounds then either:

- the step size is too small – increase it;
- the stimulus level or type is inappropriate to the task – change it; or
- the setting of the control is unimportant – cease the adjustment process and use the time for something more worthwhile.

If the adjustment being made affects a sound quality whose effects can be heard within a few seconds, such as degree of high-frequency emphasis, it may be more efficient to allow the patient to directly adjust the control to the preferred setting, rather than to present pairs of response alternatives as described in this section.

As was discussed in Section 12.2.3, one has to be aware of exactly how the hearing aid response is changing when any control is varied. If changing the compression threshold also changes the gain for mid-level sounds, each change of compression threshold has to be accompanied by an appropriate change in overall gain or compression ratio, such that the gain for mid-level sounds remains the same. This greatly increases the complexity of the adaptive paired-comparison task for the clinician, to the point where it is probably not worth the time required to do it. Fortunately, there are many occasions where a control can be varied without having to compensate for an unwanted effect by varying another control. The adaptive paired-comparison task is then a very efficient way to adjust a control on a hearing aid.

Although there are procedures, such as the Simplex procedure,[1316] for efficiently adjusting two parameters adaptively by paired comparisons, these are too time consuming for clinical use.

12.2.5 Adaptive fine-tuning by absolute rating of quality

Just as we can adaptively vary the hearing aid's response depending on which of two responses is preferred, we can adaptively vary the response depending on an absolute rating of loudness or sound quality. The University of Cambridge group has described,[1219] and later revised,[1238] a useful procedure for adjusting the gain-frequency response and compression characteristics of multichannel hearing aids. The goals of this procedure, referred to as ***Camadapt***, are that:

1. Intense speech, at 80 or 85 dB SPL, should be judged as ***loud;***
2. Weak speech, at 50 or 60 dB SPL, should be judged as ***quiet***;
3. Intense speech or music should have tonal quality that is preferred;
4. Weak speech or music should have tonal quality that is preferred.

The original version of the procedure uses intense speech at 85 and weak speech at 60 dB SPL for both loudness and tonal quality judgments. The tonal quality is adjusted by finding the degree of frequency response tilt that results in the intense speech being neither ***tinny*** nor ***boomy***, and the weak speech being neither ***shrill*** nor ***muffled***.[d]

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[c] If the reversal values are averaged as suggested, an even number of reversals should be used. An odd number of reversals *can* be used, but the midpoints between the reversals must be averaged, rather than the reversal values themselves, or the averaged value will be in error.

[d] The different adjectives for tonal quality used at the two levels are based on the words that test subjects most commonly use at each level when the low- to high-frequency balance is inappropriate.[1219]

The revised variation of the procedure uses intense speech at 80 and weak speech at 50 dB SPL for the loudness judgments. Tonal quality adjustments are made while patients listen to music from a three-piece jazz trio (piano, double bass and drums), also presented at 80 and 50 dB SPL. Rather than absolute judgments of tonal quality, patients are asked to make a paired comparison between gain-frequency responses that differ in the degree of tilt from low to high frequencies.

To facilitate the gain adjustment, a patient is shown the loudness scale shown in the first column of Table 12.1. In response to the patient's rating, the hearing aid is adjusted as shown in the remaining columns. To facilitate the frequency response shape adjustment, patients are:

- shown seven-point scales ranging from uncomfortably tinny to uncomfortably boomy, or uncomfortably shrill to uncomfortably muffled (for the speech-based procedure), or
- asked to choose the music they prefer, taking into account quality, clarity, and the balance between the bass and the cymbals (for the music-based procedure).

The procedure is complete when all four goals are achieved. If the procedure is to be carried out in the minimum possible time, it is important that adjustment of controls at one step does not undo the goals achieved at a previous step, otherwise the adjustments must be made iteratively. The procedure can be carried out manually, but it is most easily carried out if manufacturers include appropriate software to perform the adjustments within the fitting software provided for each specific hearing aid.

Although this procedure has been presented as a procedure for fine-tuning hearing aids, it can also be used as the primary adjustment method in the fitting process. The procedure takes the least amount of time (around 10 minutes per ear) if the hearing aid settings are as close to optimal as possible at the start of the procedure. Consequently, it seems most efficient to precede this procedure with a prescriptive procedure. A combination of a prescriptive procedure based on thresholds followed by the adaptive adjustment described in this section appears to be a reasonable way to adjust multichannel compression hearing aids. It certainly does not seem sensible to spend time carefully measuring loudness growth curves for the initial prescription, only to depart from the prescription during the adaptive procedure.

The procedure's goals seem very appropriate and the adjustment process uses efficient psychophysical procedures. There is some arbitrariness as to which input levels are used to represent intense and weak speech, and which particular loudness scale is used, both of which will somewhat affect the final settings. A reasonable way to match input levels with points on a loudness scale might be to choose the input levels that people with normal hearing assign to particular loudness categories, as happens in prescription by frequency-specific loudness normalization (Sections 10.4.1 and 10.4.2). Unfortunately, the normative levels to be used as targets are very dependent on the par-

Table 12.1 Gain adjustment for weak and intense speech, in response to the loudness ratings shown.

Loudness rating	Gain change at all frequencies for intense speech (presented at 80 or 85 dB SPL)	Gain change at all frequencies for weak speech (presented at 50 or 60 dB SPL)
7. Uncomfortably loud	-4 dB	-4 dB
6. Very loud	-2 dB	-4 dB
5. Loud	0 dB	-4 dB
4. Comfortable	2 dB	-2 dB
3. Quiet	4 dB	0 dB
2. Very quiet	4 dB	2 dB
1. Can't hear	4 dB	4 dB

ticular loudness scales and measurement procedures used to determine them. In general, patients prefer hearing aids adjusted using the adaptive procedure to a loudness normalization procedure, whether that procedure is based on individual loudness scaling or on thresholds alone.[1219] Averaged across patients, the gain-frequency responses resulting from the revised procedure, at least when applied to an open-fit device, are very similar to the CAMEQ prescription based on thresholds alone.[1238]

12.2.6 Fine-tuning at home with multi-memory or trainable hearing aids

There are two limitations to fine-tuning the hearing aid in the hearing clinic. One, of course, is clinical time which *someone* always has to pay for. The second is validity. It is possible to reliably choose, from a number of responses, the response that the patient prefers on some criterion while listening to some stimulus. The extent to which this indicates the best response in the real world depends on how good a job we have done of choosing the stimulus and asking the right question. Both multi-memory hearing aids and trainable hearing aids (and, to a lesser degree, hearing aids with data logging) make it possible to move the fine-tuning session out of the clinic into the patient's own environment (loosely called *at home*), largely overcoming these problems.

Suppose a patient has been fitted with a three-memory hearing aid, and the first memory has been programmed with amplification characteristics prescribed by some procedure. Because of the current state of knowledge, we may be unsure whether a lower or a higher compression threshold would be more beneficial. At the end of the appointment we program a lower compression threshold into one memory, and a higher compression threshold into another. The patient is instructed to try the alternatives, and return in a few weeks to tell us which memory (or memories) he or she prefers. At that stage, the unwanted program(s) are removed, and something more generally useful (like a low cut for listening in traffic noise) is put in their place.

This could be the end of the fine-tuning, or it *could* go on for as long as the patient and the clinician have the interest and the time. In some cases, the patients will indicate that they like one program in one environment and another program in other environments. In these cases, both programs can be retained, and the trial will have confirmed the usefulness of multiple memories for that patient, and given some information on how they should be programmed.

The limitations of continuing the fine-tuning at home are evident: only a few different responses can be compared, and each new set of comparisons takes a new appointment to establish. This is really only practical if a further appointment is already needed for other purposes or if the clinician is easily able to schedule one or more additional short appointments.

Performing the fine-tuning at home with the assistance of a trainable aid (Section 8.5) minimizes clinician time and ensures the validity of the test stimulus. More than one parameter can be adjusted (simultaneously or sequentially) and because the hearing aid automatically adopts the parameters that result from the training, no further appointments to lodge these parameters in the hearing aid memory are needed.

Data logging (Section 8.5) may also assist fine-tuning by giving the clinician insight into the patient's patterns of hearing aid adjustment outside the clinic (e.g. volume control adjustment made by the patient in different environments) which can guide the fine-tuning dialog.

As hearing aids continue their rapid advance through the digital age and into the wireless age, it is likely that there will always be amplification characteristics that we are not sure how to prescribe, and which we are not able to adjust empirically in the clinic in a reliable or valid way. A role for fine-tuning at home is therefore likely to continue, perhaps combined with remote adjustment of the hearing aid, or monitoring of the settings to which it adapts, via the telephone or Internet (i.e. ***tele-audiology***).

12.3 Concluding Comments: Fine-tuning in Perspective

Every clinician will have to fine-tune hearing aids in response to problems reported by patients. Doing this in the most efficient manner will minimize expense and frustration for clinician and patient alike. The two key skills are asking the questions that will help the clinician understand the problem as precisely as possible, and identifying which hearing aid controls should be altered to achieve the desired aim.

The extent to which systematic fine-tuning procedures should be used is a more difficult question. Some clinicians will consider that they do not have time to carry out the systematic procedures described in Section 12.2. It is possible that these procedures will not be used often in busy clinics, but five or ten minutes spent systematically fine-tuning a complex hearing aid may save considerable time in the long run. Targeted and systematic fine-tuning may prevent several return appointments from a client who is dissatisfied with the sound quality provided by expensive and very flexible hearing aids.

Until we know what proportion of patients will benefit from fine-tuning procedures after their hearing aid has been adjusted on the basis of a reliable prescription procedure, we will not know whether it is sensible to include a systematic fine-tuning component in every hearing aid fitting, or just to reserve this process for patients who report problems. Whether or not systematic procedures are used, hearing aids cannot be optimally adjusted unless time is allocated to listen to, and understand, patients' comments about amplified sound.

CHAPTER 13

PATIENT EDUCATION AND COUNSELING FOR HEARING AID WEARERS

Synopsis

People with a hearing impairment benefit from patient education and may benefit from communication training and counseling. These activities may be aimed at giving patients information about their hearing loss, developing skills needed to operate and care for their new hearing aids, improving listening skills, or changing patients' beliefs, feelings and behavior relating to their hearing and communication. Providing appropriate education and counseling increases the likelihood that hearing aids will be fully used and that residual communication difficulties will be minimized.

It is difficult to help patients understand the variety of hearing aid styles and performance features that may be suitable for them. The benefits and cost implications of each (including ongoing service costs, warranty, and trial periods) have to be presented in a suitably simple manner.

Once they start using their hearing aids, first-time hearing aid users experience a new world of amplified sound, and may benefit from guidance about how to gradually increase their range of listening experiences. The aim is to provide them with the best experiences first, and to avoid having them become overwhelmed by sound. Patients need to know that their brains may take some time to adapt to hearing parts of speech, and other sounds around them, that they have not heard for some time.

A major part of educating the new hearing aid user has nothing to do with hearing aids! A wide range of hearing tactics and strategies can help the hearing-impaired person understand more in difficult listening situations. The first group of hearing strategies requires the listener to look carefully at the talker and the surroundings. The second group requires the listener to alter the communication pattern in some way. The final group requires the listener to manipulate the environment to remove or minimize sources of difficulty. Patients will benefit if family members and/or other frequent conversation partners participate in education sessions on these topics.

Patients will more easily appreciate and learn this material if it can be taught in a patient-centered, individual problem-solving method, rather than as a set of rules disconnected from their everyday lives. Communication training comprises training in the use of these hearing strategies, plus practice in listening to speech (synthetic training) or to the basic sounds from which speech is built (analytic training), especially in difficult listening conditions. Increasingly, communication training is being provided in packages that patients can use on their computer or DVD at home.

Patients should be advised about protecting their remaining hearing, and be made aware of where they can obtain support (from peer groups or other professionals) beyond that which the clinician can provide. Hearing aids do not provide an adequate solution to all hearing problems, so patients must be made aware of other assistive listening devices that may help them.

Clinicians should be aware that different people learn in different ways. Consequently, the same material should be taught in different ways to different patients, and clinicians should develop the flexibility needed to accomplish this.

Clinicians must be flexible regarding how and when they present information and carry out other procedures. It is, nonetheless, useful to have in mind a standard program from which variations can be made as required. This chapter concludes with a list of activities that can be performed at each of the assessment, fitting, and follow-up appointments. The use of group follow-up appointments, in addition to an individual appointment, is strongly recommended.

A dictionary definition of counseling is: "giving advice, opinion or instruction to direct the judgment or conduct of another." At one extreme counseling can be aimed at changing how patients feel about their hearing loss and its consequences, and this may not involve the clinician *telling* the patient anything. Such activity is often called ***personal adjustment counseling***. At the other extreme, the term counseling has been used to describe giving factual information to a patient, such as detailed *instructions* about how to operate a hearing aid. Such a factual flow of information has been referred to as ***information counseling*** or ***content counseling***. In other health fields, giving factual information to the patient is known as ***patient education***. This latter term is becoming more common in audiology, and will be used in this chapter. The term ***counseling*** can then be reserved for transactions that are primarily aimed at modifying a patient's beliefs, emotions or behaviors.

Of course, many interactions have elements of both education and counseling. Advising a client about the severity and nature of his or her hearing loss, and the effect that hearing loss often has on people, is a factual flow of information, but may well result in the patient feeling different about him or her self.

This chapter focuses more on patient education, not because it is more important than counseling, but because it is more closely related to the title and purpose of the book, and because some counseling suggestions aimed at achieving acceptance of hearing loss and the need for rehabilitation have already been covered in Chapter 9. The stress and emotional difficulties that hearing loss often cause may, however, so dominate a patient's thoughts that it is not possible to effectively impart information, teach skills, or encourage him/her to commence any form of rehabilitation until these issues have been discerned, accepted, and discussed. In these situations, adjustment counseling must be provided, either by the clinician or via referral to another professional.

We can list several specific aims for education and counseling related to the provision of hearing aids:

1. Making sure the patient ***understands*** the nature of his or her hearing loss, its consequences, and treatment options (including both devices and procedures).
2. Helping the patient ***acknowledge*** that he or she has a hearing loss, and working through any consequential negative emotions that restrict enjoyment of life.
3. Helping the patient ***overcome obstacles*** that discourage him or her from engaging in any form of rehabilitation.
4. Instructing and encouraging the patient in the ***use of hearing aids***, or other assistive listening devices.
5. Helping the patient acquire additional ***communication skills*** in the form of listening and communication strategies. Some of these require personal adjustment by the patient, such as increased assertiveness.
6. Providing ***perceptual training*** in understanding speech. This training can comprise analytic and synthetic speech training, in either auditory, visual, or auditory-visual presentation modes.

The second and third of these points were covered in Chapter 9. In this chapter, we will assume that the clinician is faced with a patient who desires to improve his or her ability to hear. What information should the clinician give, what should the clinician ask the patient to do, and when is the most appropriate time to do each activity?

The following sections will describe the type of information that patients need to acquire. The chapter will conclude by outlining how this information can be structured into a series of appointments. Or rather, how it might be structured for some patients. If a patient does not absorb critical information, or learn critical skills the first time they are taught, the process must be repeated or varied later. The actual content of appointments must thus remain flexible.

Another reason for flexibility is that some patients will want to know as much as possible about everything. Others will just prefer to be told what to do, accompanied by the minimum possible explanations being given. Those who want to know more than you are saying will generally let you know by asking questions. Patients who understand less than you are saying may not tell you so. It is a good communication strategy on your part to intersperse your information-giving with questions that test whether the patient understands what you are saying. In addition, maintain good eye contact, and observe body language that will guide you on how to proceed.

Surprisingly, communication with patients is usually not complicated by their hearing loss! Face-to-face communication with a clearly-spoken clinician on a known topic, in a quiet, low-reverberation environment, does not usually pose much of a problem for people with a mild or even moderate loss. People with a severe or profound hearing loss probably already have hearing aids, and should of course, be encouraged to wear them during any appointments prior to re-fitting. When a person with severe or profound loss does not have hearing aids and needs temporary help, options include the talk-over facility of the audiometer, or a body aid or assistive listening device fitted with supra-aural or circum-aural headphones.

13.1 Understanding Hearing Loss

Most patients understandably want to know about their hearing capabilities and hearing loss. To give a balanced account of their hearing, the concepts of ***capability*** (i.e. the remaining hearing) and ***loss*** should both appear in your description. It will help patients to understand their loss, and be able to relate it to significant other people, if they are given a broad understanding of four different aspects of their hearing:

- The ***location*** of their loss (the outer ear, middle ear, inner ear, or the brain), with reference to a suitable wall chart or hand-out.

Evidence for the benefit of patient education and counseling

Patient education and counseling (loosely called counseling in this panel) affect the degree to which patients use their hearing aids. In one study using BTE hearing aids, a combination of pre- and post-fitting counseling increased usage from an average of 3.8 hours per day without significant counseling to 5.3 hours per day with counseling.[181] Another study showed that counseling two weeks after fitting increased usage from 3.9 hours per day to 6.3 hours per day.[1891] Patients are much more likely not to use their hearing aids at all if they do not receive adequate counseling. One major reason is that if patients are not sufficiently taught to insert an earmold or earshell, they will not be *able* to use their hearing aids.[187] Instruction in hearing aid use by volunteer helpers at the patient's home also increases aid use.[852]

Counseling unrelated to the use of hearing aids is also helpful. Patients who are given information about hearing loss, hearing strategies, and communication skills, in addition to being fitted with hearing aids, report less hearing disability[51, 1646] and/or handicap[12, 1646] than those who are given only hearing aids. Similarly, perceptual training can enable people to better understand speech.[1877] The increase seems, however, to come more from an increased use of context than from an increased ability to identify individual phonemes.[579, 1537]

There are studies reaching apparently conflicting conclusions about the value of counseling *prior* to aid fitting, although there is possibly no real contradiction. Brooks (1979) showed that a combination of pre-fitting counseling and post-fitting counseling, both carried out at home, will significantly increase daily usage and competence in manipulating hearing aids, and will significantly decrease hearing handicap. In this study, subjects in the control group were seen only twice. In the first visit, the audiogram was measured, and an ear impression taken. The remaining visit was for fitting the hearing aid. Given this minimal contact with the patient, it is not surprising that additional pre- and post-fitting counseling was effective. The pre-fitting counseling appeared to be comprehensive.[192] It covered assessment of communication difficulties, modification of attitude and expectations, discussion aimed at reversing many patients' tendencies to withdraw from social activities, provision of information about hearing loss and hearing strategies, and assessment of the need for post-fitting counseling.

By contrast, Norman, George & McCarthy (1994) found that pre-fitting counseling, with these same objectives, does not alone significantly increase satisfaction, usage, or benefit from the hearing aid. What is not clear is whether members of the control group were as bereft of any other counseling as were those in the other control group a decade earlier. Many of the things that should be discussed can probably be done just as effectively before or after fitting, though those involving hearing aid management and the effectiveness of hearing aids in different listening situations would benefit from the patient already having worn hearing aids. Issues related to attitude, motivation, and the choice of hearing aid must, however, be resolved prior to fitting.

- The ***degree*** of loss (mild, moderate, severe, profound), and ***configuration*** of loss (flat, sloping etc), with reference to their audiogram, and the prognosis as to how it is likely to change.
- The ***disability*** or ***activity limitation*** that is to be expected (inability to understand quiet speech or speech in noisy places) even when they consider that they can easily hear that speech is present. This can be made specific by referring to the situations in which they consider they have trouble hearing, to errors that they have actually made on speech discrimination tests, or to speech sounds for which they are likely to have particular difficulty (see Figure 9.6). Information about their hearing loss configuration and the influence that this has on speech discrimination may be instrumental in helping people accept that they do indeed have a hearing loss.
- The ***handicap*** or ***participation restriction*** that often results (withdrawal from activities, common emotional reactions, effects on other family members).

All of this has to be strongly tempered by the apparent interest and understanding that the patient is showing. Some patients will be stressed and/or depressed by the experience of having their hearing tested and their loss confirmed by a clinician. For this and other reasons, some patients will not absorb much information. Providing too much information will be counter-productive and may increase any stress they are undergoing. Close monitoring of the patient's reactions, verbal and non-verbal, is essential.

13.2 Acquiring a Hearing Aid

Discussion about the patient's hearing loss and its consequences leads naturally to a discussion about treatment options, which in most cases will include acquiring one or two hearing aids. After discussing whether or not the patient is likely to benefit from hearing aids (see Chapter 9), and assuming the patient wishes to acquire hearing aids, there are several issues to be discussed.

Hearing aid style

It seems wisest to outline the advantages of each style (See Section 11.1) that apply to that patient before asking the patient which style is preferred. Otherwise, the clinician may have to present information in the context of trying to change the patient's decision, rather than informing the patient to help him or her arrive at a decision. (Of course, it is possible that the patient will have already decided on a style before the appointment commenced.)

The patient should be allowed to physically handle each style. The patient should also be shown pictures

Handling preferences for an inappropriate style

What should you do when the patient wants a hearing aid style that you think is unlikely to be effective (for reasons of inadequate gain, power, or ease of manipulation)? Opinions vary, but here is one opinion:

- If you are reasonably sure that the hearing aid will be ineffective, politely tell the patient that you could not in good conscience fit him or her with a hearing aid that in your judgment will not be suitable, even though some other providers of hearing aids may be willing to do so.
- If you *think* that the hearing aid style is unsuitable, but have some doubt about your conclusion, and the patient is highly motivated to have that style, be specific about the reasons for your doubts. If the patient persists in his or her choice, fit hearing aids of the type the patient prefers. It may be worth documenting the patient's written acknowledgment that the type of hearing aids prescribed are not the ones that you most highly recommend.[1762]

The rationale behind the first suggestion is that your principled refusal to sell an ineffective hearing aid may cause the patient to change his or her preference. Even if it does not: you have saved the customer money, you have avoided the time wasted on fitting and returning an aid for credit, and you have avoided having an ex-patient walking around telling friends that the hearing aids you supplied were no good. The rationale for acceding to the patient's preference in the second scenario is that motivation is a very powerful contributor to success with hearing aids, and the documentation avoids future misunderstandings, both personal and legal.

of them when worn, or alternatively be shown what they look like when modeled in, or behind, the ear of the clinician. Thin-tube, or thin-wire BTEs are likely to be perceived more positively relative to other styles when viewed being worn, than when viewed in the hand or on the desk. Some clinicians wear one from the outset of the appointment, and point it out to the client when the discussion turns to the appearance of different styles.[830] One advantage of this style is that if the client is ready to purchase the hearing aid at the conclusion of the hearing assessment, it can be supplied immediately, potentially saving one appointment, and consequently lowering the cost of provision.

Hearing aid technology and cost

The clinician has to be highly aware of the advantages and disadvantages of different levels of technology, and this requires knowledge of complex technology plus the ability to assess the value of each performance feature to each patient. Even more difficult is the task of presenting this to the patient in a way he or she can understand. The patient must have this knowledge, however, to be able to make an informed decision about how sophisticated (and expensive) the hearing aids should be.

Depending on the features available in hearing aids at any time, and on the models and brands dispensed in the particular clinic, it is often possible to make a table that shows the models in order of increasing sophistication, price, and benefits to the patient. Information about style and technology can be combined in the form of a matrix with the different styles as rows and the different technologies as columns.[377] It is probably simpler, however, to separately consider style and technology wherever possible.

Most patients will be balancing cost against the advantages of the more complex technologies. To help them arrive at a decision you should ascertain whether any concern about cost is driven by a desire to limit the total cost, or by a desire to get the best value for money, irrespective of cost.[132] If the clinician knows of any forms of financial assistance that might be available to the patient, this is an appropriate time to provide this knowledge.

It may help patients place the cost in perspective if they are told the likely life of the hearing aids (perhaps three to six years). The total cost of the hearing aids, including batteries and average maintenance costs, can then be expressed as the average cost per day over the life of the hearing aids.[1762]

Responsibilities and rights

Patients have a right to know how much the hearing aid(s) and associated service will cost, what warranty period covers the hearing aids, what service plans are available (and their costs) and what ongoing costs they will have for batteries. They also should be told when payment is due, the period during which they can return their hearing aids, and the extent of refund available during this period.

Government legislation may guarantee patients a right to effective communication in public places. The details vary from country to country, but clinicians should provide patients with written material covering any rights they have in this regard, and details on how they should go about making the most of these rights (e.g. asking at a theatre ticket office for an assistive listening device).

13.3 Using Hearing Aids

Teaching patients how to insert a hearing aid, switch the aid on and off, operate the volume control, manipulate any other controls present, remove a hearing aid, and change the battery are essential parts of patient education. Unless these skills are mastered (by the aid wearer or by a helper if necessary) hearing aid use is not possible. Expect to spend longer, on average, for clients over the age of 80.[804] Techniques for teaching these skills vary with the type of the hearing aid. The techniques are similar to those used for re-teaching people who have not mastered these skills by the first follow-up appointment. Some suggestions have already been covered in Section 12.1.1.

Patients will also need to know approximately how long batteries will last. Some patients will appreciate having their own battery tester. If patients are highly reliant on their hearing aids, they will need to carry spare batteries, especially when the battery life is close to an end. A simple way for experienced hearing aid wearers to keep track of when this is likely to occur is to place the battery tag on a calendar on the day that new batteries are due.[1899]

As virtually all patients will want improved speech understanding in noise, and as the directional microphone is the major feature that will provide assistance in noise, patients should be taught two key things:

- how to activate it (e.g. by selecting a noise program), unless the hearing aid is fully automatic and makes its own decisions as to when it activates;

- the circumstances in which it is most effective (i.e. talker close and to the front and/or noise close and to the back – Section 7.3.1). This knowledge is important, as patients will often be able to position themselves to achieve at least one of these requirements. Without this information, they may or may not work out for themselves where the directional microphone is beneficial.[323, 1753] They are most unlikely to learn this by experience if they cannot toggle between directional and omni-directional modes.

The safety concerns related to battery ingestion should also be covered (see Section 16.10).

13.4 Adjusting to New Experiences with Sound and Hearing Aids

When people put on their hearing aids they receive an avalanche of sounds, usually in the high-frequency region, that they are not used to hearing. Many patients will make the transition to hearing aid use more easily if they build up their listening experience gradually,[a] commencing with quiet situations and wearing their hearing aids for only a short time each day.

There are several reasons why a gradual build-up can be useful. It is encouraging for patients to have the most positive experiences first. They can then build confidence in their hearing aids (and in themselves, by having some communication successes) while they are becoming accustomed to hearing more sound and to hearing sound with a new tonal quality. Patients will not instinctively know the situations in which hearing aids are most effective, and so should be guided by the clinician. In fact, they are likely to try the situations where they have the most difficulty (i.e. where there is a lot of background noise) which is where the hearing aids are least likely to be useful (Section 9.1.6). The following pages show the ***Situations To Experience and Practice (STEP)*** form that clinicians can hand to patients to guide them through the first weeks of hearing aid use.[b]

The second reason for a gradual build-up of listening experience is that a patient's attitude to hearing aid use may be positively affected if he or she commits to following a specified listening program. It is a well-accepted psychological principal that behavior can affect beliefs.[510, 541] In particular, if patients realize that success with a hearing aid is conditional on

Using the STEP form

- Explain to your patient the general principle of gradually stepping up daily listening experience, both in regards to hours per day and the noisiness of the situations encountered.
- Explain that the patient will have to re-learn how to recognize all the sounds that he or she will be hearing.
- Emphasize that like any other learning, this task will require some commitment and application by the patient.
- Tell patients that you are interested in their reaction to each situation, and that you would like them to record how helpful the hearing aid was, and any problems they experienced.
- Make sure that any situations in which the patient particularly needs to hear better are somewhere on the list. These will be evident if you have already carried out the initial phase of the COSI evaluation (Section 9.1.6). Specific situations either can be written in as examples of the standard situations, or can be specifically recorded in the final two blank spaces. In the case of the latter, you will need to indicate which of the standard situations should first be attempted.

[a] The need for gradual exposure is pronounced for linear/peak-clipping hearing aids, because of their capacity to frequently produce loud distorted sound. With modern hearing aids incorporating both WDRC and compression limiting, sounds need never be uncomfortable or distorted, and should only occasionally be very loud. The need for a graduated exposure may be less marked, but the benefit provided by hearing aids still varies across listening situations.

[b] The STEP form may be copied and distributed to patients. An enlarged version will be useful for patients with poor eyesight.

One step at a time
Situations To Experience and Practice

Welcome to some new experiences with sound. You will get the most out of your new hearing aids if you practice using them in certain situations around the home before you progress to situations that are more difficult. Also, do not wear your hearing aids for more than two hours per day for the first week unless you are finding them really comfortable in all respects. Make sure you use them for at least half an hour each day, however.

Try to wear your hearing aids in the following situations in roughly the order shown. Progress through the list as quickly or slowly as you are comfortable with. After you have tried your hearing aids in each situation, write down how helpful they were, and any problems that you encountered. Over the next few weeks, wear your hearing aids while you are:

1. Listening to one other person at home while you can see his or her face.

Comment: ______________________________

____

2. Listening to a TV or radio at home.

Comment: ______________________________

____

3. Walking around inside your home, trying to recognize any sounds you can hear.

Comment: ______________________________

____

4. Listening to one other person at home while you are not looking at their face.

Comment: ______________________________

____

5. Listening to music.

Comment: ______________________________

____

STEP (continued)

6. Listening to your own voice while you read aloud from a newspaper or book.

Comment: ____

7. Conversing with two or three people in a quiet place.

Comment: ____

8. Walking around outside, trying to recognize any sounds you can hear.

Comment: ____

9. Shopping or talking to another person in a noisy place.

Comment: ____

10. Conversing with two or three people in a noisy place.

Comment: ____

11. Conversing in a large gathering or at a noisy restaurant.

Comment: ____

12. Special situation: ……………………………………………………………

Comment: ____

13. Special situation: ……………………………………………………………

Comment: ____

Source: Dillon – Hearing Aids – Boomerang Press them using it in a certain way, they are likely to do so. In turn, the act of using their hearing aids may make them rationalize that the hearing aids are worthwhile. This belief then encourages further use. This is not a trick to convince patients to accept a worthless piece of apparatus; it is a technique to help patients overcome what can be a difficult time of adjustment so that they get the most from their hearing aids.

The third reason for a gradual adjustment is that it reinforces to the patient that listening situations *are* different. If hearing aids are found to be of no use in one situation, the patient may be less likely to generalize this conclusion to all situations, making statements like *These hearing aids are no use at all*.

There is a fourth reason that has nothing to do with sound. Earmolds and shells can cause discomfort and irritation when they are first worn, even if they fit well (just like new shoes). A gradual increase in daily usage allows the ear to become accustomed to them without pronounced irritation developing. This may be particularly important with any hearing aids that extend into the bony part of the canal.

The benefits of a step-by-step exposure can best be captured if a patient can discuss with the clinician his or her experiences during the first week or two of wearing the hearing aid. The clinician may use this information to demonstrate the need for hearing strategies (Section 13.6) or may adjust the hearing aids to provide increased benefits or decreased disadvantages. The STEP form includes space for patients to record their experiences with sound in each situation. These written comments provide an easy discussion prompter for patients who otherwise choose not to talk much about what they have experienced.[530]

The *Active Fitting* program available in Sweden combines gradual exposure to sound with several opportunities for patients to discuss with the clinician their experiences. It consists of five appointments interspersed with three periods of home use during which the patient completes a diary called *Try Your Hearing Aid*, and has been shown to increase hearing aid use and satisfaction with hearing aids.[530] The authors comment that the attitude of the clinician is critical: *patients will not regard the listening program as important unless the clinician appears to believe it is important*.

As always, expectations are important. Before new hearing aid wearers start wearing hearing aids, they need to know that they will be hearing background sounds that they have not heard for some years. People with normal hearing hear these sounds, or rather, they *can* hear these sounds, but learn to ignore them when they carry no meaning. The sound of a fan is meaningless except when we believe that we have turned off all the appliances in the house because we are leaving on holidays. Then the sound has great meaning. The clinician should explain to patients that it might take them some time to become so accustomed to hearing these background sounds that they can unconsciously recognize them and ignore them. This advice is particularly important for patients who will be wearing WDRC hearing aids with a low compression threshold and, consequently, a lot of gain for low-level sounds.

Similarly, people with normal hearing sometimes find sounds to be annoyingly or uncomfortably loud. Wearing a hearing aid seems inevitably to increase the number of sounds that are annoyingly loud, as judged by the increase measured on the aversiveness scale of the APHAB (Section 14.3.2). The increased annoyance is, however, similar to that experienced by people with normal hearing.[c, 1379] It may therefore assist new hearing aid wearers if they know in advance that they *will* likely be annoyed by loud sounds more often, and that this is something they will share with normal-hearing people. (Despite this knowledge, the clinician should still do everything possible to minimize the annoyance through the use of individually prescribed WDRC, OSPL90 prescription and evaluation, directional microphones, and adaptive noise reduction.)

Of course, telling patients that they will take time to get used to these sounds is not the same as guaranteeing that they *will* get used to them. When a patient complains about such sounds at a follow-up appointment, the clinician will have to choose between re-instructing about the normality of hearing these sounds, and adjusting the hearing aid so that the patient hears less of them (see Section 12.1.7).

Whether or not patients are gradually exposed to sound, they should be advised that it might take them some months to become used to the sounds provided

[c] These data were obtained with well-fitted hearing aids that had adaptive noise reduction enabled. Annoyance and aversiveness is likely to be greater than normal for hearing aids without adaptive noise reduction or with excessively high gain for higher sound levels.

by their hearing aids, and to receive maximum benefit from them. It is possible that new neural pathways have to form to allow this to happen, and this process has been referred to as ***brain rewiring*** and as ***acclimatization***.[595] If patients are advised in advance that such a process can occur, they may be less discouraged by any initial experiences where their hearing aids are not helpful. Section 14.7 contains further information about acclimatization.

Even after the patient has fully adjusted to the sounds made available by the hearing aid, the listening experience may be well short of the patient's expectations. Reduced performance of the auditory processing system for many elderly patients (Section 9.1.11) and distortions in the cochlea (Section 1.1.6) are likely to leave the patient with poorer sound quality and lower speech intelligibility in challenging situations than he or she is expecting. Without this information, the patient may expect to hear as well as he or she did before acquiring a hearing loss, and before any age-associated decline occurred in the auditory processing system. The patient should therefore be advised that he or she is likely to have more difficulty than people with normal hearing in understanding speech in challenging situations. Because less of the speech will be understood without conscious effort, communication in such situations will be harder work for the patient than for those with normal hearing, and expending the extra effort is likely to make the patient tired. The extent will vary from patient to patient. Of course, sound quality, speech intelligibility, and tiredness would be even worse without the hearing aids.

13.5 Care of Hearing Aids

Patients must be told how to care for their hearing aids. The accompanying panel shows a list of things to do and things to avoid. Although it contains some seemingly obvious statements, they will not be obvious to all patients and should therefore be stated. The list can be copied and provided to patients.[d] For individual patients, it may be necessary to vary the 3-mm dimension mentioned in the third point. The depth into a custom aid to which a cleaning implement can be inserted without damaging the receiver can often be ascertained by looking at the hearing aid with a bright light behind it.

Patients who frequently need to return their custom hearing aids for maintenance because of moisture build-up or cerumen build-up may benefit from stor-

Treating hearing aids kindly

Don't	Do
X Don't wash them.	√ Do wipe them regularly with a tissue and occasionally with a slightly damp sponge.
X Don't wear them in the shower, or the bath or the swimming pool, but if this happens by accident, don't dry the hearing aids in an oven or a microwave.	√ Do disconnect a BTE mold from the aid occasionally, and wash the mold in warm soapy water. The tubing may take a day or so to dry out unless you have a hand pumped air blower to dry it.
X Don't insert anything more than 3 mm up the hole in the end of the aid.	√ Do clean wax out of the tip, whenever it is present, with a brush, a loop, a pick, or by operating or changing an in-built wax guard.
X Don't spray them with hair spray.	√ Do store them overnight in their box or some other container.
X Don't leave them in the car in the sun.	√ Do remove the battery if you are going to store the hearing aid for more than one day.

[d] Some hearing aids are now sufficiently waterproofed that they can be worn in the shower. Delete as appropriate.

ing their hearing aids overnight in a de-humidifying environment. Storage devices are available that contain various combinations of heat, airflow, desiccant, deodorant, and germicidal electromagnetic radiation. The combined effects of heat and low humidity dry out any cerumen present, which can then more easily be removed with the usual cleaning methods. The battery should be removed prior to placing the hearing aid in the de-humidifier (to avoid drying the chemicals in the battery) and the battery compartment should be left open.

13.6 Hearing Strategies

Hearing strategies (also known as ***hearing tactics*** and ***listening strategies***) are methods that people can use to increase their understanding of speech.[1862, 1863] Patients can use hearing strategies separately from, or in conjunction with, hearing aids or assistive devices. Even people with normal hearing can use hearing strategies in difficult situations. Because hearing-impaired people have decreased ability to discriminate between sounds, they will certainly need to use hearing strategies if they are to function effectively in as many environments as possible. Moreover, when people have gone a long time without hearing properly, they develop maladaptive strategies and behaviors, like bluffing or monopolizing the conversation.

All hearing-impaired persons should therefore be taught constructive hearing strategies, and should receive some take-home material to remind them of the important points. Patients are likely to already know a few strategies, but there will be many more strategies of which they will not be aware.[542] People who have received even a brief instruction in hearing strategies report less disability and handicap than those who have not.[1892] In fact, effective use of hearing strategies may decrease disability and handicap sufficiently that the patient may not need to wear hearing aids in some situations.[48]

Hearing strategies can be grouped into three categories: those that involve ***observation***, those that involve ***manipulating social interactions***, and those that involve ***manipulating the physical environment***.[542] It is as well that multiple strategies are available, because in any given situation, some of the strategies will not be feasible for physical or social reasons.

13.6.1 Observing the talker and surroundings

Lip-reading

Considerable information can be gained from watching people's lips.[136] Most people, including those with normal hearing, probably watch lips naturally, or even unconsciously, in adverse listening situations. Some people, however, may not, and many may not make as much use of ***lip-reading*** as they could if they were made aware of its potential. It is therefore important to instruct patients about its value as part of any rehabilitation program. If time permits, the clinician can demonstrate its considerable value. A videotape of a "talking head" such as a newsreader can be played and the patient asked to follow by hearing alone, and then

Theoretical background: Why lip-reading is so valuable

The type of information obtained from lip-reading is especially useful to hearing-impaired people. The information most visible is the place of articulation, or constriction, of consonants (lips, teeth against lips, teeth, tongue against teeth, and several places further back inside the mouth). Place cues to speech are, however, the hardest cues for hearing-impaired people to perceive correctly via hearing alone. Lip-reading thus provides information that is complementary to audition.

For example, hearing may tell the person that the sound is a /p/, /t/, or /k/ (these are easily confused). Vision may tell the person that the sound is a /p/, /b/, or /m/, as these sounds look identical. The only possible conclusion when hearing and vision are combined is that the sound is a /p/. For either modality alone, there was only a 1 in 3 chance of correct perception (in the absence of contextual cues), but when the patient combines the two forms of perception the correct answer is assured. In this example, no error occurs. In general, however, errors will be greatly decreased compared to hearing alone, even if some errors remain.

Although lipreading ability deteriorates with age, the ability to integrate the visual cues perceived with the audio signal does not.[1674]

by hearing combined with vision. The difficulty of the task can be varied by adjusting the volume control on the TV monitor, such that the hearing-alone condition is not too easy.

To lip-read, the patient *has* to be able to see the talker's lips. This may involve moving, or asking the talker to remove his or her hands from in front of the face. If a patient considers that this would seem rude, some coaching will be necessary. The positive statement *I can understand you much better when I can see your lips clearly* may appear friendlier to both the patient and the talker than *would you move your hands from in front of your face please*, or *would you look at me when you talk please*. However, these more direct alternatives might sometimes be necessary.

Patients who normally place value on eye contact during conversation may be reassured to know that the person talking will not notice whether the listener is watching the talker's eyes or lips.

Non-verbal signals

It is not only a talker's lips that convey information. Facial gestures (e.g. smiling, frowning, surprise, quizzical looks, disgust) all convey the essence of the talker's message. If the patient understands the essential message, the words can more easily be filled in, or in some cases ignored. Bodily gestures or positions often also reinforce the message. The clinician should advise the patient about the richness of information available from the face and body of the talker. The combination of lip-reading, face-reading, and body-reading is often referred to as ***speech-reading***. All the *reading* terms are useful. It is important to point out to patients the various individual sources of information, as well as reinforcing that all of the information combined will contribute greatly to understanding speech.

Filling in gaps

Missing words can often be guessed based on the topic, the talker, facial expressions, or the physical surroundings. Some people are reluctant to guess, so it is appropriate to let the patient know that it is OK to miss words and to guess at meaning based on all the evidence available. When a patient becomes too uncertain as to the accuracy of his or her guesses, the patient can check with the talker about the interpretations that he or she is making by using the techniques described below. Some patients will need to be encouraged to guess more often; others will need to be encouraged to check more often.

13.6.2 Manipulating social interactions

All of the following strategies require hearing-impaired people to modify the way they interact with others. We learn the normal rules of communication from an early age. To the extent that hearing strategies require some variation from these rules, patients require practice and reassurance if they are going to be able to use them comfortably and naturally.

One hearing strategy used (consciously or unconsciously) by some hearing-impaired people is to talk all the time so that they rarely have to listen. If the clinician suspects that a particular patient has adopted this strategy, some tactful reminders about the adverse social consequences of this strategy, and the availability of alternative strategies that induce a more positive reaction from communication partners, would be appropriate.

Clear speaking

Some talkers are easy to understand, and anyone can more easily be understood when they speak more clearly.[1411] Consequently, clear speech is more resistant to noise and reverberation than is normal speech.[1397] Clear speech differs from conversational speech in several ways:[1412, 1652]

- speaking rate is lower, because speech sounds become longer when they are fully enunciated, and because people speaking clearly insert or lengthen the pauses between words. It is nonetheless possible to increase speech clarity without lowering the average rate, albeit with different modifications to speech production;[974, 975]
- vowels are fully formed, resulting in an enlarged range of format frequencies;
- stop bursts in word-final consonants are released;
- the relative intensity of stop consonants is greater;
- pitch range is increased.

Fortunately, people do not need a course in phonetics to become clear speakers. Clarity will improve even if people are just asked to speak more clearly, as if communicating in a difficult environment.[237] A more pronounced, and more sustained improvement in clarity may be obtained if the talker is instructed about speaking rate, articulation, pausing, stressing key words, and is given examples and feedback.[237] People apparently take only 10 to 15 minutes of practice to become proficient and once learned, the effect is maintained for weeks or months without further reinforcement.[1577] The effectiveness of clear speech cnnot be doubted; the only uncertainty is the extent to which family members will continue to speak this way in normal life.

Clinicians will not always have direct access to family members. To achieve and maintain clear speech amongst family members, the clinician can explain the principles to the patient, so that he or she can ask communication partners, in a non-threatening way, to speak more clearly. The patient can advise communication partners of his or her need for extra-clear speech. This approach avoids any implication that the speech of the communication partners is sub-standard in some way.

For aged listeners, part of the benefit from clear speech will arise from the slower rate of production, as it allows their brains more time to process what is being heard. Although most of the research into clear speech has been performed using English, the benefit of clear speech applies to at least some other languages (but almost certainly to all given the physiological processes involved in speech production).[1652]

Where there is background noise, increased speech intensity will also improve intelligibility, because it improves SNR. Most guides on hearing strategies advise that shouting is counter-productive, but in noisy circumstances it may be the only way to achieve adequate intelligibility, just as it is sometimes necessary for normal-hearing people to shout to each other. There is no point in a talker shouting when the SNR is good – if the listener needs more intensity the hearing aid's volume control (if present) can be increased. Shouting may be less of a problem for modern hearing aids with WDRC and/or compression limiting than it was for linear/peak-clipping hearing aids, but should be used only as a last resort.

Gaining the listener's attention

Because of the importance of speech-reading, a hearing-impaired person can hear best if he or she has the opportunity to speech-read right from the start of an utterance. This is possible only if the listener is looking at the talker right from the first word. Regular conversation partners can be asked (and trained) to gain the attention of the listener before talking. In adverse listening conditions this can be done by a touch, but in most circumstances it can be achieved just by saying the listener's name, then pausing, then talking. In structured groups, such as at a committee meeting, hearing-impaired people may find it hard to quickly identify who is talking, especially if their localization ability is impaired. The assistance of the chairperson can be sought to ensure that only one person talks at once, and that people talk only after the chairperson nominates them. The hearing-impaired person will then find it easier to follow people, aurally and visually, right from the start of their declarations (and everyone else may appreciate the orderly meeting that will result).

Knowing the topic

Knowledge of the topic makes it much easier for a person to correctly guess the words that are not heard or only partially heard. When a hearing-impaired person commences a conversation with others, particularly when joining into an existing conversation, his or her first task should be to find out the topic. Shy or unassuming patients will need considerable encouragement if they are to break into the conversations of others to ask what the topic is. An easier alternative is to take one friend aside and have that person state what the topic is. It may seem obvious, but patients need to understand the importance of knowing the topic, and be reminded that sometimes the only way they will gain this knowledge is to explicitly ask someone.

Repair strategies

Breakdowns in conversation are very common and when a listener has missed a key word or phrase, the listener can gain the missing information in a way that involves the minimum disruption to the ongoing conversation. Such breakdowns occur for all people irrespective of hearing status, although they occur more often for hearing-impaired people. Saying 'what' repeatedly is not always socially acceptable and is not as beneficial as some other strategies, such as:

- repeating back the words preceding the words not heard, with a questioning intonation, accompanied by a questioning facial expression;
- asking a specific question that indicates what was heard and what was not, e.g. W*hat sort of mood did you say he was in?*
- repeating back or re-phrasing what the listener thought he or she heard to confirm its correctness;
- asking the talker to say the last sentence or two in a different way;
- when all else fails, asking the talker to spell out a key word.

All of these techniques reassure the talker that he or she is mostly being understood, and minimize the time needed to gather the missing information.

Giving feedback

If the patient is constantly giving feedback to the talker (especially in one-to-one conversations), the talker will quickly learn to speak in a way that best gets a message through without needing further intervention by the listener. Feedback comprises smiles, nods, mmm's, yes's, aha's, frowns, and puzzled looks. Talkers will adapt by varying their talking speed, clarity, voice level, and complexity of expression. People *like* to be understood.

Disclosing the hearing loss

Finally, if the patient is willing to disclose that he or she has a hearing loss, talkers will make some adjustment, as described in the previous paragraph. It should be recognized, however, that it can be a very big step for some patients to disclose hearing loss to others. Clinicians may wish to discuss this with the patient and provide appropriate support and guidance.

13.6.3 Manipulating the environment

Lighting

Because observing the talker is essential for good intelligibility, good lighting is crucial in situations with adverse acoustics. The patient should be advised that it will sometimes be necessary to move or ask the talker to move. Situations that commonly cause problems are when the talker sits with his or her back to a window or lamp. The listener has the double disadvantage of looking into a bright light while trying to see the talker's dimly lit lips and face.

Positioning

The key to easy listening is position, position, and position! There are several reasons for this apart from vision. Close to the talker, signal levels are higher so the SNR is better. Similarly, the ratio of signal to reverberation is better. Both of these ratios are crucial to intelligibility,[1079] and both of these ratios deteriorate greatly when the talker and listener are in different rooms.

It is difficult for patients and their communication partners to overcome long-held habits of trying to communicate from another room. The extreme difficulties that such communication creates, even for normal-hearing people, should therefore be strongly pointed out to patients. Getting close to the talker applies whether the talker and listener are the only two people in a room, or whether there are a hundred other people. (It is tricky if they all want to be close to the talker.)

Another aspect of position is relevant to people who have a better side and a poorer side for listening.[e] The head is an effective obstacle for sound waves above about 1.5 kHz. This means that at the better ear, high-frequency sounds arriving from the same side are boosted, but high-frequency sounds arriving from the other side are attenuated (see Section 15.2.1). Consequently, the SNR at high frequencies is much better on the side of the head closer to the talker. For maximum intelligibility, patients should therefore orient themselves so that the talker is on the good side. If there is a dominant noise source, patients will obtain the highest intelligibility if they point their worse ear towards the noise. That is, the patient should be between the talker and the noise, at enough of an angle to benefit from these head baffle effects, but not so much that speech-reading is impossible.

Minimizing noise

Noise has such a disturbing effect that any reduction in noise level will enable easier understanding. Solutions include:

- turning the TV or radio off or down;
- closing a door; or
- moving to a quieter place to talk.

Minimizing reverberation

In the home, adding soft furnishings (thick curtains, well-padded lounge chairs, thick pile carpet) to a room will decrease reverberation and hence increase intelligibility. In other situations, patients should choose places with such furnishings for conversations whenever possible. The more absorbent the furnishings, and the bigger the room, the further apart the

[e] Better and poorer sides arise when people with symmetrical hearing loss wear only one hearing aid, or when people have asymmetrical speech identification ability.

listener and talker can be and still avoid the adverse effects of reverberation.[f]

Adjusting the source

When the source is an electronic appliance (a TV, a radio, a CD, a public address system) adjusting the tone control of the device may improve intelligibility and naturalness. If the listener is optimally aided, this may not help, but if the listener is not using hearing aids, or if the hearing aids are deficient in high-frequency gain, a treble boost in the electronic appliance should help intelligibility. For good music perception when listening through hearing aids, a bass boost (unless the device is an open fitting), and possibly a treble boost, may be helpful. There is very little research on this point from which to take guidance.

While all these modifications to the environment (lighting, position, noise, reverberation, and tonal quality) are easy to list, some may be difficult to achieve. Even when there is no physical constraint, the patient may feel that modifying the environment will result in some inconvenience to communication partners. A thorough discussion of these strategies will elicit how the patient regards this issue, and will enable patients to reflect on how their right to communicate can be balanced against the rights of others.

Hearing strategies in summary

- Watch the talker – lips, face, body
- Find out the topic
- Ask the talker to speak clearly
- Ask the talker to gain your attention
- Give frequent feedback
- Ask specific questions
- Guess meaning and repeat to confirm
- Get close to talker
- Get rid of noise
- Discuss clear speech with significant others

As a generality, the benefit to the patient of a modification far outweighs the disadvantage (if any exists) to communication partners. Indeed, environmental modifications that help people with hearing loss are more likely to help, than to interfere with, the enjoyment of others. It is important to encourage patients to experiment with different modifications and find out what works best for them.

13.6.4 Teaching hearing strategies

Hearing strategies can be taught in an abstract manner (similar to the preceding three sections) but preferably are taught to each patient in an individual, problem-solving way.[735, 978, 1893] A method that is bound to capture the attention of a patient is to identify a few problem situations that are important to him or her and devise a list of hearing strategies that are appropriate to each situation. For instance, a couple who have trouble hearing each other when they are watching the TV might decide that the following strategies are feasible: they will sit closer together, rearrange the lighting so that they can see each other's face, add some more soft furnishings to the room, and if necessary, acquire hearing aids with directional microphones. If all else fails they will use a remote control to turn down the TV sound before speaking. The situations listed on the patient's COSI form (see Sections 9.1.6 and 14.4) can provide a ready-made starting point.

Teaching hearing strategies using individual problems commences with the patient describing a situation to the clinician in as much detail as possible. The clinician then either suggests strategies or, if time permits, asks leading questions that help the patient work out strategies for him or her self. The latter approach, which is in line with adult learning principles, is more likely to result in the patient understanding, retaining, and being committed to the solution.

Hearing strategies are an ideal topic for group discussions (Section 13.14.4) because they enable a good venue for discussion of the social and relational implications of the strategies. Where hearing strategies involve gaining the cooperation of others, the group provides the opportunity for patients to practice asking for this cooperation.

[f] Close to the talker, the direct sound from the talker over-rides the fuzzy reverberant sounds, even if there is a lot of reverberant sound spread throughout the room. Sections 3.4 and 7.3.1 describe critical distance and other concepts related to reverberation.

13.7 Involving Families and Friends.

Although the discussion so far has concentrated on working with patients, both patients and their families can benefit if the families also participate in education and counseling. There are advantages to having a significant other person (SO), such as a spouse, parent, child, or friend, participate in all stages of these activities.

Candidacy

If, at the first appointment, the patient understates the difficulties he or she has with hearing, the SO can add an extra perspective, to the benefit of both clinician and patient. The patient's understatement may be matched by the SO's exaggeration or vice versa.

Understanding the disability caused by hearing loss

The SO may be dismissive of the patient's difficulties (*He can hear when he wants to*). If so, the patient will appreciate the clinician explaining and demonstrating to the SO how a high-frequency loss can make understanding difficult even though the presence of speech may be easily detected. A hearing loss simulation on CD or on some manufacturer's software will be helpful in showing the difficulties to a caring but non-understanding SO.[967, 1213]

It is emotionally beneficial for patients to know that those close to them understand the difficulties they are going through, even with their hearing aids. Patients commonly report (if asked) that *no one really understands*.[1018] If family members are able to appreciate the effort the patient may be expending in communicating, they may be able to structure their visits and conversations to be shorter when the listening environment is adverse.[979]

If the causes of adverse listening environments (noise, multiple talkers, reverberation, distance, accents, talking rate, unfamiliar topics, poor lighting, and rapid change of talkers) are explained, either the SO or the patient may be more able to recognize the cause when it occurs, and address it in some way.

Hearing strategies

Most of the hearing strategies covered in Section 13.6 require the cooperation of another person. If the other person has heard first-hand from the clinician what is required, the patient may find it much easier to gain appropriate cooperation, and new behavior patterns are more likely to be maintained.[613]

Learning to use hearing aids

The SO will witness the clinician instructing the patient to insert, remove, and operate the hearing aid. Unlike the patient, the SO can *see* the patient's ears, and can considerably assist the patient to learn these skills when they return home. For patients with very poor memory or dexterity, the clinician may elect to teach the SO, rather than the patient, how to use the hearing aids.

Follow-up questioning

The presence of a SO can keep the patient honest when he or she is asked about how much the hearing aids have been used and whether there have been any difficulties. Data logging also provides information about use, but only people can provide information about benefit or difficulties.

Overall encouragement

A SO can provide many forms of encouragement as the patient learns to use his or her hearing aids. A clinician will always tell a patient that the hearing aids can be adjusted if the sound quality initially provided is not acceptable. A SO who knows this can encourage the patient to return for adjustment rather than use inadequate sound quality as an excuse to give up.

Information retention

Patients retain only a proportion of the information given to them during appointments, which can be a stressful time for patients. One experiment indicated that patients retain 74% of the information presented at hearing aid orientation appointments.[1488] Having the SO present increases the chances that information will be retained by one or the other, and some will be passed on to the patient by the SO at a later time.

Third-party disability

The patient is not the only one in the family suffering from the problems caused by his or her hearing loss. The SO is likely to experience various adverse effects as a result of the patient's hearing loss; these have been referred to as ***third-party disability***.[1558] The adverse effects can take many forms:[1557, 1693]

- reduced participation in social activities because the patient does not wish to participate;
- reduced conversation between the couple, including distress at being misunderstood or not responded to, and frustration at having to repeat things;

- cessation of small, albeit interesting, comments made in passing, as the effort to make them understood by the patient is too tiring relative to their individual importance;
- cessation of whispered secrets or other intimacies in a social setting;
- annoyance at the high volume of TV or radio;
- a sense of burden, because the SO continually has to take responsibility for communication with other people, both in social settings and on the telephone. In social settings, the SO may feel the need to act as interpreter, or as controller of the conversation to protect their partner in various ways;
- annoyance at feeling they have to adapt more than their hearing-impaired partner;
- embarrassment over the patient responding inappropriately or withdrawing in social settings; or
- isolation, and a deteriorating relationship, as a consequence of all the other impacts.

Given the motivating effect that the SO can have on the patient to commence hearing rehabilitation,[1357] the supporting effect that the SO can have on rehabilitation itself,[1569] and the high interconnectedness of their problems, it seems a reasonable strategy for the clinician to provide advice and acknowledgment to the SO as well. The SO may directly benefit from being involved in the rehabilitation process.[793] It may be most economically feasible, as well as most efficacious, for this to happen primarily through involvement of the SOs, particularly spouses, in group rehabilitation programs (Section 13.14.4). Whether or not the SO is involved in the rehabilitation process, they are likely to experience reduced hardship as a result of the reduced disability and handicap experienced by the patient.[1693]

Although there are many advantages to having a SO present for all appointments, some patients will prefer to be seen on their own and this choice is theirs to make. Preliminary contact with the patient (telephone or mail) should simply make it clear that that the patient is encouraged to bring along a family member or friend to each appointment.

13.8 Auditory Training

Hearing loss typically restricts access to the high-frequency parts of speech. It seems likely that some occurs, to enable the power of the cortex to focus on those parts of speech that are still audible. If so, the reversal of this re-wiring that will need to occur if the patient is to gain maximum use of hearing aids that restore audibility may be facilitated by systematic training in understanding speech, especially in a low-frequency noise background.[1930]

Furthermore, patients, especially those with a severe or profound hearing loss, may have gradually restricted their activities and communication experiences over several years because of the difficulties imposed by their hearing loss. Many such people need to acquire new skills and gain confidence if they are to fully rejoin society. Patients may have come to believe that

Confidence building: success breeds success

A major goal of auditory training is to build confidence in those being trained. The key to achieving this is to ensure that the hearing-impaired people being trained achieve success in any tasks they are set. The aim is to show people what they *can* do, not reinforce what they *can't* do. Training should commence with tasks that the patient can definitely do, and rapidly increase in difficulty until the tasks are presenting a significant challenge, but are still able to be completed. There are many ways in which the material can be altered to control its difficulties:[588a]

- Syntactic and semantic context, and hence easiness, will decrease as the material progresses from familiar stories, through unfamiliar stories, paragraphs, sentences, phrases, and words, to individual syllables.
- Material can be presented in quiet, or background noise can be varied to include infrequent environmental sounds, white noise, multi-talker, or a single competing talker. The signal-to-noise ratio can also be varied.
- Situational context can be withheld or described, and the talker's face can be revealed or concealed.

they have lost, and will not regain, the ability to function in social situations. This belief is also expressed as reduced ***self-efficacy*** for communication.[979]

Facilitating both these types of changes is the role of ***auditory training***.[g] Auditory training can be categorized into two general types: analytic and synthetic training.

Analytic speech perception training is conducted by presenting speech to the patient, requiring the patient to identify the sounds or to indicate whether two sounds are the same or different, and then providing feedback as to the correct answer. Analytic training concentrates on developing the patient's ability to differentiate between syllable patterns and between phonemes in syllables or words. The aim is to help patients learn to use speech cues that should be audible to them, but which for some reason they are not using. A possible reason for lack of use is that the patient has only recently begun wearing amplification, and has not yet regained the ability to use the newly audible cues. For analytic speech training, speech material is usually presented one syllable or word at a time, so that the patient can focus on the characteristics of the sound being practiced. The material can, however, be presented in whole sentences. Analytic speech perception training is also called ***perceptual speech training***, and is routinely used to help children develop speech perception and production skills.

Synthetic communication training is conducted by presenting speech to patients in a natural manner, such as by conversing with them, or by having them listen to a story. In synthetic training, the emphasis is on the patient understanding the message, even if the patient does not correctly perceive every sound. The origin of *synthetic* in the name is that the listener has to synthesize (i.e. combine) any available pieces of information to correctly interpret the message. As the major part of the training, patients are taught to use any or all of the hearing strategies discussed in Section 13.6. In addition, patients are given practice at understanding speech in a context where they are given feedback about what they perceive and misperceive. Synthetic communication training is also called ***active listening training***.[977] This phrase implies that the listener frequently lets the talker know that the listener has understood the message (also called ***reflective listening***), but implies the use of other hearing strategies as well.

It should be apparent that there is considerable overlap between synthetic communication training and hearing strategies. There is also some overlap between synthetic and analytic training, (e.g. when sentence length material is used for training). Analytic and synthetic training can both be conducted using hearing alone or they can be supplemented with visual cues. If visual cues are excluded, however, so too are many hearing strategies that could otherwise be taught as part of the synthetic communication training.

Analytic and synthetic training differ in their aims. Analytic training is aimed at increasing patients' correct perception of the individual sounds of speech, using predominantly ***bottom-up*** auditory processes. Evidence as to whether this aim is achieved is contradictory.[980, 1760, 1877] An unsolved problem with analytic training is that we do not know how to ensure that the improved performance (that can usually be measured with the training material) generalizes to other sounds, talkers, or background noises, and can be sustained for extended periods (at least months) after training is completed.[202] Generalization *can* occur, and is sometimes even accompanied by changes in the auditory processing system as evidenced by changes in the electrophysiological responses to sounds, both in the auditory cortex[1797] and in the brainstem.[1644]

Synthetic training, by contrast, aims to alter patients' behavior when communicating, increase their confidence when engaging in communication, and strengthen their use of predominantly ***top-down*** processes when making use of incomplete information. It is unequivocally successful in achieving these aims.[980, 1760] Both forms of training are time consuming. Analytic training has the potential to be automated (see next section). Because synthetic training usually involves modifying human interaction, however, the potential for automation is less. Some parts of synthetic training can be carried out in small groups, which decreases the cost of providing the training. Extensive materials for analytic and synthetic training can be found in Plant (1994) and Plant (1996) respectively.

[g] Auditory training is a sub-set of ***aural rehabilitation***. The latter traditionally also includes hearing strategies, use of assistive listening devices and psycho-social counseling. Auditory training is also a subset of ***communication training***, which includes hearing strategies and working with communication partners.

It is feasible that in the future, analytic training will be supplemented by drugs that facilitate the breaking and making of neural connections, as has successfully been trialed with cochlear implant recipients.[1786]

13.9 Computer-Based Auditory Training at Home

While some might consider education or counseling by computer an anathema, many of the activities described so far in this chapter (particularly hearing strategies and auditory training) lend themselves to being carried out by the patient interacting with computer programs, or passively watching videos on computer or DVD/TV. If the patient can undergo these activities at home, it becomes economically feasible for him or her to undergo many hours or many tens of hours of education or counseling. Not surprisingly though, this method is more effective at imparting information or increasing auditory skills than in modifying emotional responses to hearing loss and the problems they create.[971] Computer-based training makes it relatively easy to provide repetition and reinforcement, give immediate feedback,[1281] adapt difficulty to keep the task challenging but achievable, engage the patient in active participation, and document and display progress to sustain motivation. All of these are thought of as important ingredients for successful training.

Computer-based auditory training at home appears to improve speech perception by an amount equivalent to a few dB improvement in SNR.[1051, 1765, 1930] This is a significant gain that should provide noticeable benefit in real life (provided, of course, that the measured benefit generalizes to real life). It is similar in magnitude to the benefit from directional microphones in circumstances conducive to directional benefit (Section 7.3.1).

It appears that those who have the greatest hearing impairment, disability and/or handicap gain the most, on average, from home-based auditory training.[718] Candidature for home-based auditory training is probably similar to that for hearing aids themselves: those with the most motivation to do it will be the most likely to complete the program and gain the benefit. Older patients are also more likely to complete the program, possibly because they have more time.[718] Premature termination of the training at home can be more common than not, and the clinician should use any motivational tactics he or she can think of to maximize the chance of the program being completed.[1766]

There are several computer-based auditory training programs now available:

- ***Listening and Communication Enhancement (LACE)***[1765] comprises tasks involving speech perception in babble and against competing talkers, time-compressed speech, and closure skills (deducing missing words in sentences from context). It also provides information on hearing strategies.
- ***Seeing and Hearing Speech***[1603] focuses on auditory-visual speech-reading training
- ***Conversation Made Easy***[1806] focuses on speech-reading training and hearing strategies.
- ***Read My Quips***[1046] focuses on speech-reading training in noise using humorous sayings to maintain motivation.

Although the clinician is not directly involved in the ongoing training, clinicians can expect to receive calls for technical support in the use of auditory training computer programs, unless the patients have recourse to technical support from some other source.

13.10 Avoiding Hearing Aid-Induced Hearing Loss

Patients must be told (but not unduly alarmed) that wearing a hearing aid increases their risk of acquiring further hearing loss because of additional noise exposure. Provided gain and OSPL90 are responsibly prescribed, and provided the patient does not have a profound hearing loss, the risk is very minor, especially if the hearing aid includes wide dynamic range compression covering at least the mid to high input levels (see Section 10.8). The patient, nevertheless, should be advised to avoid prolonged exposure to loud noise, and to wear hearing protection in very noisy places when intelligibility is not an issue. Interestingly, hearing aids will act as a form of hearing protection whenever the noise level is greater than the SSPL of the hearing aids.[h]

[h] This statement is approximately true. The mean aided level at the eardrum relative to the mean unaided level at the eardrum will depend on the individual RECD, the individual REUR, the spectrum of the ambient noise, the dynamics of the ambient noise, and the shape of the hearing aid input-output curve.

13.11 Assistive Listening Devices

At the assessment appointment, the clinician should consider whether one or more assistive listening devices (ALDs; see Section 3.10) rather than, or in addition to, hearing aids would meet the needs expressed by the patient.[i] The COSI goals will assist in identifying such a need. In the vast majority of cases, there will be at least one need that can be met only by hearing aids.[451] Often, however, it will be unclear at this early stage whether hearing aids will fully meet all the needs.

Consequently, it will often be the case that hearing aids are recommended, and a decision about other devices withheld, until the patient can evaluate how well the hearing aids meet all his or her needs. The need for assistive listening devices should therefore be reviewed at a follow-up appointment after aid fitting if they have not been discussed before this time.

In some cases, it may be evident from the outset that an ALD is all that is required. The most common example is a person whose most strongly felt need is for a device to make speech from the TV clearer. In such cases, an ALD plugged into the TV or with its microphone right against the TV loudspeaker will produce clearer speech than even the best hearing aids, at a fraction of the cost.

Patients provided with an ALD will need significant instruction in (and demonstration of) its use, including identifying situations in which they can be used, and where the microphone can be placed. ALDs can provide a huge improvement in speech understanding, but there are many potential impediments to successful use (technology aversion, cosmetic appearance, manipulation difficulties, physical discomfort, and not knowing how to use it), all of which require appropriate discussion to identify and resolve.[153, 1042] It will likely be necessary for the clinician to ensure that any audio signals an ALD delivers are appropriately matched in level to that provided by the hearing aid (Sections 3.6.4 and 3.11).[153, 1042]

13.12 Counseling Support

Counseling to help patients deal with the emotional consequences of hearing loss falls within the skill base expected of audiologists, and aspects are mentioned in a few places in this chapter and in Chapter 9. For some patients, however, needs will become evident that are beyond the ability of the clinician to deal with, or which cannot be dealt with in the time that can be made available to the patient. In these circumstances, the clinician can best serve the patient by referring the patient to a personal, relationship, or family counselor.

If the clinician feels that the patient may have emotional issues (e.g. impact of hearing loss on self-worth, self-concept or relationships) that are precluding or delaying successful rehabilitation, but which the patient seems unwilling to discuss, jointly completing a self-assessment handicap inventory (Sections 9.1.4 and 14.3) may provide a non-threatening way to get the patient talking.[519] Items that the patient gives strong answers to provide fertile grounds for further discussion – it does not matter whether the inventory is ever finished.

On a more practical level, the patient may also benefit from referral to:

- peer support groups;
- telephone relay services; and
- education services.

13.13 Interacting with Different Personality Styles

Everything in this chapter so far has dealt with the content of education and counseling, not the manner in which they are delivered. A clinician will tend to use styles of teaching and questioning with which he or she is most comfortable. Sweetow (1999a) gives a more thorough review of counseling styles and strategies. To be most effective, information has to be delivered in the way that each *patient* can most easily absorb, and in the way that is most likely to change the patient's attitude or behavior, if that is the clinician's intent.

An important distinction to be aware of is that some people learn most easily if they can *see* what is being taught, others most appreciate *hearing* a clear explanation, whereas others most easily absorb things by *doing* them. A problem arises if the clinician is able to teach things only one way, and the patient is able to learn them only in a different way. Flexibility of approach by the clinician is essential if knowledge and skills are to be imparted accurately and in the

i ALDs are increasingly being referred to as hearing assistance technology (HAT).

minimum possible time. This flexibility is the essence of patient-centered care, the benefits of which are well documented in other areas of health care.[1025]

People differ in how they see the world, and there are many ways to express these differences. One popular psychological profiling method is the ***Myers-Briggs Type Indicator*** (***MBTI***).[1293] The MBTI measures where people fall along each of four dimensions:

- **I**ntroversion-**E**xtraversion,
- **S**ensing-i**N**tuition,
- **T**hinking-**F**eeling, and
- **J**udging-**P**erceiving.

Depending on the dominant end of each of these dimensions, people are classified into one of 16 types, such as *ISTJ* or *ESFJ*, etc. The personality of each of the 16 types has been summarized, and linked to things such as the type of occupation likely to be chosen and enjoyed. Traynor and Buckles (1997) summarize these personality types, and suggest that knowing a patient's personality type may help clinicians adapt their approach to the patient appropriately. Such knowledge could help a clinician know a patient's preferred way of operating right from the first appointment, rather than gradually getting to know as the series of appointments progress.[1796]

For example, people with an *S* characteristic will respond to facts, whereas people with an *N* characteristic will like reasons and logical arguments for doing things. People with a personality type that includes an *EJ* combination are likely to blurt out any difficulties they face without first thinking them through, and will want to decide how to deal with them immediately. By contrast, those with an *IP* combination will need more encouragement to talk about difficulties if they do not consider they have had enough time to reflect on them and make sense of the difficulties themselves. Patients with an *EFJ* combination are particularly likely to need, and respond well to, encouragement and praise as they learn to use their hearing aids.

Another popular profiling method is the ***Merrill-Reid Social Style Inventory***. This inventory categorizes people according to their score on two dimensions. The first dimension contrasts fast-paced, assertive people (who prefer telling things to others) with slower-paced, cooperative people (who interact by asking things of others). The second dimension contrasts responsive, emotive people with cool, controlled people. The two dimensions enable people to be categorized into four groups: the assertive, controlled *driver;* the assertive, responsive *expressive;* the asking, controlled *analytic*; and the asking, responsive *amiable*.

Handling talkative patients

A dilemma confronting most clinicians is how much to let talkative patients talk. On the one hand, it is essential to find out what concerns patients in relation to their hearing. On the other hand, some patients talk incessantly about seemingly irrelevant things, and the clinician is acutely aware of how many essential things still have to be done or discussed in a limited time. It is useful to be direct with the patient in a positive way. Let them know for example, that you are extremely interested in how well they could hear at the party, rather than telling them (with verbal or non-verbal signals) that you are not interested in hearing what their grand-daughter wore to the party. The difficulty with steering conversation is that people will sometimes not tell you what is most on their mind until you have demonstrated by your acceptance of them that you will be interested in, and accepting of, the things they fear you may not like hearing.

The major tools available to the clinician are a compassionate nature and ***active listening*** skills. Active listening involves reflecting back to the patient the essence of their message, using either the same words used by the patient or different words that convey the same central message. If the essence of what the patient is trying to convey is a feeling, then so too must be the message reflected back.

If a patient continues to produce irrelevant small talk despite you continually bringing them back to your preferred topic, the patient may not be finding acceptance in your reactions and may hold back on things that are important to them. In such circumstances, patients often do not feel safe to say the thing that is troubling them most until they are half-way out of the door at the end of the appointment.[1092]

Interestingly, active listening is also an excellent way to encourage non-talkers to open up.

There are many other four-quadrant ways of categorizing the diversity of human interaction styles. No matter what system one uses to describe this diversity, patients will respond best, and come to a decision fastest about hearing aid acquisition, if the information is presented to them in the way that they prefer.[1936] This requires intuition, or direct questioning of the client, on the part of the clinician to recognize what the client prefers, and flexibility to enable the clinician to impart information and relate to the patient in this manner.

Adapting presentation style to the client's personality is the antithesis of having a standard patter that is delivered in the same way to each and every client. On the other hand, the clinician who varies too much outside of his or her personality comfort zone may appear to be not genuine. There is, nonetheless, a minimum set of information that we want most clients to take in and it is part of the clinician's job to work out how best to achieve this. Simple written summaries of options for the client (e.g. hearing aid versus ALD versus communication program) can be presented to the client to elicit the type of further information they would like to receive.[1023, 1024]

Should clinicians spend their time, and their patient's time, gathering data on personality (and presenting these results to the patient) before providing appropriate rehabilitation? In the future – probably not; right now – certainly not. The basic research to link personality type to attitude, motivation, disability, and the most effective means of communicating information about rehabilitation has not been done. Formally measuring personality is time consuming and would doubtless be viewed as irrelevant at best, and intrusive at worst, by many clients.

Clinicians should definitely be aware, however, that different patients respond best to different approaches and should seek to discover how each patient best operates. Every question a patient asks or statement a patient makes provides an insight into what type of information he or she prefers. If patients ask for *evidence* of the effectiveness of a high-priced hearing aid, do not give them an *explanation* of how it works, and vice versa. If patients say that something you are asking them to do does not *feel* right, find out why rather than giving them a logical, *thinking* argument about why the action you are proposing is the best for them. Similarly, people with a visual orientation may say that they *see* what you are saying. (Readers of this book with an *S* in their Myers-Briggs personality type will have particularly appreciated these concrete examples.)

Whatever style of interaction is used, it is essential that it be supplemented with provision of written information. There are too many important pieces of information for a patient to be able to take them all in during a few appointments. Anything that is essential for the patient to know should be provided in written form as well as discussed. Research in other fields of health has indicated that the proportion of information retained is increased markedly by supplementing words with pictures.[246, 771]

13.14 Structuring Appointments

The following sections list some activities that can usefully be performed in a service protocol nominally comprising three-appointments plus one remote follow-up appointment. This must not be interpreted rigidly. Some patients will need two assessment appointments before they are ready to acquire hearing aids and/or choose a style or performance level. Conversely, for some patients it will be possible to achieve assessment and fitting in a single appointment. Fittings that do not go smoothly for technical or human reasons may need two fitting appointments. Many people will need additional follow-up appointments because of special problems. Many people will not take in things the first time and the information may have to be repeated.

Individual idiosyncrasies will sometimes require variation from the following suggestions. Time constraints and difficulties encountered will often make it impossible to achieve all the items listed, and the clinician will have to judge what should be deleted, handled solely by providing written take-away information, or deferred to a later appointment. The clinician must have clear goals about what should be achieved, combined with high flexibility about how and when it is achieved, in part driven by the clinician's assessment of how much the client can take in at any time.

Furthermore, some clinicians will be more drawn to objective procedures and others will be more drawn to interacting with the patient. If the clinician avoids either type of activity in a pronounced way, however, this is likely to be detrimental to a good outcome.

Home visits

Brooks (1981) strongly recommends that at least one of the appointments should be carried out in the patient's own home. The advantages of this are:

- patients are more relaxed and more frank about their difficulties, both before and after fitting;
- the clinician can more precisely assess the needs for assistive listening devices; and
- the communication pattern between the patient and others at home can be assessed more easily, with a view to suggesting more effective communication strategies.

Unfortunately, such visits are very time consuming. Home visits may also make measurements (e.g. audiometry, real-ear gain) more difficult or impossible, although the availability of portable equipment has lessened this difficulty.

Vuorialho et al (2006) showed that home visits (performed 6 months after the initial fitting) to provide any additional help needed, resulted in an increased proportion of patients wearing their hearing aids, and increased quality of life (when measured 12 months after fitting, relative to a measurement made immediately before counseling was provided at the home visit).[1867] Despite the additional cost of the home visit, its cost-effectiveness (measured as cost per additional regular hearing aid wearer) was high relative to the cost of hearing aid fitting without the home visit.

13.14.1 The assessment appointment(s)

- Determine why the patient (or someone on the patient's behalf!) has initiated this appointment.
- Take history (family history of loss, etiology of loss, work history, noise exposure, tinnitus, dizziness, asymmetry, brief medical history including medications, referral source, flexibility and manipulation ability, vision).
- Determine hearing needs (e.g. via COSI).
- Perform otoscopic examination.
- Cerumen removal.
- Measure hearing, using whichever tests are appropriate to the individual patient.
- Explain test results and implications of loss.
- Determine expectations, and modify as necessary.
- Discuss rehabilitation options, including hearing aid advantages and limitations.
- Explain the likely program of fitting and follow-up, including the options of communication training and group appointments if applicable.
- Choose hearing aid style and performance features.
- Take ear impressions if needed.
- Provide a written report if appropriate.

13.14.2 The fitting appointment(s)

- Program/adjust the hearing aids if not already done.
- Modify the shell or earmolds for comfort and ease of insertion (if necessary), or select the appropriate length thin tube and appropriate diameter dome.
- Put hearing aids in, adjust volume for comfort, and leave on.
- Teach patient how to change battery, insert and remove hearing aids, differentiate left and right hearing aids, and operate the volume control and on/off switch or battery door. Mention the presence of the T-switch (if appropriate).
- Measure real-ear gain, and adjust hearing aids to meet prescription targets.
- Evaluate sound quality (including patient's own voice) and fine-tune if necessary.
- Evaluate maximum output and fine-tune if necessary.
- Teach patient how to care for hearing aids, including cerumen management.
- Demonstrate use of the hearing aid with the telephone, including operation of the T-switch if appropriate. Listening to a recorded message service is useful.

- Advise about the situations in which directional microphones will and will not be beneficial, particularly if the patient has to select a noise program to activate the directional microphone. (This will need to be deferred to a later appointment for patients who seem to be overloaded with more essential information at the fitting appointment.)
- Provide batteries and indicate expected battery life and cost.
- Advise patient about graduated use of hearing aids, including provision of the STEP form, and remind patient about likely magnitude of benefit in quiet and in noise. Inform the patient that at the next appointment you will be able to download information from the hearing aids about the use they are making of them, but will want to hear from them how useful they found the hearing aid in each situation.

The most efficient order for doing these things, particularly the timing of the real-ear gain measurement, is debatable. If evaluation of the patient's own voice requires the vent to be enlarged, or a closed dome to be replaced with an open dome, real-ear gain will be affected and the measurement will have to be repeated if it has already been done. Conversely, if real-ear measurement is delayed, patients may object to the sound quality *because* the response is far from target. Some clinicians leave the measurement until the follow-up appointment if they think they have adequately adjusted the hearing aid by using a test box or by viewing the simulated response on a computer screen. Deferment of real-ear gain testing is not recommended, however. As mentioned in Section 13.13, the level of detail and method of instruction must be varied to suit each patient.

New hearing aid wearers should not be asked about sound quality until they have had at least a few minutes experience listening with their new hearing aids. It is most efficient for them to hear something useful, like how to operate the hearing aids, while they are listening. A duplicate hearing aid facilitates this.

13.14.3 The follow-up appointment(s)

- Ask the patient about the degree of use, benefits, and problems related to the hearing aids.
- Confirm usage patterns with data logging where available.
- Ask about the volume control setting used (if appropriate), adequacy of loudness, sound quality, and intrusiveness of noise.
- Ask about problems with own voice quality, whistling, and loud noises.
- Fine-tune the amplification characteristics if so indicated.
- Ask the patient to remove the hearing aid, change the battery, insert the hearing aid, switch it on, and adjust the volume control to check on his or her ability to manage the hearing aid(s). (This can be done without it appearing to be an examination of the patient's ability.)
- Ask about ease of insertion and removal, ease of battery changing, and ease of volume control adjustment unless your observations have already convinced you that there are no problems with these operations.
- Examine ear canals for signs of irritation and ask about physical comfort.
- Ask about battery consumption or examine logged data (to check on reported use), provide information on battery life and battery tester (if appropriate).
- Note hearing aid condition and ask about cleaning.
- Ask how much the hearing aids have helped with the problems that originally led the patient to seek help, and how much difficulty remains (e.g. via COSI).
- Check on the ability of the patient to use the telephone (with or without a T-switch, as appropriate).
- Evaluate the need for assistive listening devices and provide appropriate information and demonstrations.
- Teach appropriate hearing strategies and provide written material for the patient to take away.
- Provide information about repairs, warranty, after-care, service charges, battery acquisition, and consumer support groups.
- Evaluate the need for additional follow-up appointments. If success with the hearing aid does not seem assured, schedule another appointment in one to four weeks. Otherwise, advise the patient to make a further appointment at any time in the future.

- At some stage from one to three months after the last appointment, perform a mail or telephone follow-up to evaluate benefit, use, satisfaction, and problems.[j] Schedule an additional appointment if any of these indicate there are problems that the clinician could solve.

It is essential to ask patients, on several occasions after fitting, if they are experiencing any problems with their hearing aids or their ease of communication. While one might expect that people experiencing problems would seek help, mostly they do not.[646, 850] In one survey conducted three months after fitting, 48% of patients reported having one or more problems with their hearing aids. Surprisingly, less than a quarter of those reporting a problem indicated in a questionnaire that they would like to make a further appointment with their clinician.[441] In another survey conducted 12 months after fitting, 86% of patients needed help with at least one problem.[1370] The people in this study had had an average of four visits to their clinician during the rehabilitation program associated with aid fitting.

One might argue that it is not the responsibility of the clinician to initiate contact to see if there are any problems. This is a dangerous argument unless we know why people so often do not initiate contact, and we do not know. It is also short-sighted: hearing aid owners who are in any way dissatisfied with their hearing aids are walking advertisements for why their friends should not seek rehabilitation, a situation which is to everybody's disadvantage. Non-use is, unfortunately, far too common (Chapter 9).

13.14.4 The power of groups

So far, the discussion in this chapter has assumed that the clinician is dealing with one patient at a time. There are many advantages to including one or more group appointments within an overall rehabilitation program for each patient. The group appointments supplement, rather than replace the individual appointments, although some of the things that are usually accomplished in individual appointments can be done just as well, or even better, in group appointments.

Patients most likely to benefit from a group, and most likely to be motivated to attend, are people newly fitted with hearing aids, people with severe or profound hearing loss, and/or people who feel strongly handicapped by their hearing loss. Experienced hearing aid wearers can, however, also obtain benefit,[1453] and activities focused on hearing tactics are just as useful to hearing-impaired people who do not wear hearing aids. It is beneficial if the regular communication partners of the hearing-impaired participants can also participate.[192, 735, 736, 1453] Abrahamson (1997) suggests that groups comprise at least three couples or at least five individuals, but that groups of up to 20 are possible if they can be comfortably seated.

Reasons for forming a group

There are many reasons why seeing several patients as a group is a good thing to do:

- Some activities (see panel) can be performed more efficiently in a group, which provides cost savings to the provider or the patient or both. Alternatively, for the same total cost, more rehabilitation activities can be accomplished, thus increasing rehabilitation effectiveness.
- Some patients find that participating in a group of people who are going through similar emotions and experiences is an extremely positive experience. It is not hard to understand why this might be so.
  - Group discussion can legitimize feelings, including the acceptance of hearing loss. Most people are relieved when they find that others have the same problems, reactions, and emotions when they are confronted with the same circumstances. Just knowing this can be liberating; people can then move on to deal with the circumstances and the emotions, instead of worrying about whether they *should* be feeling this way, or having these problems.
  - Other people in a group can sometimes analyze or put into words vague concerns that the patient already feels but does not understand and cannot enunciate. Again, this makes it easier to deal with the concerns.
  - People confronting the same problems often find different solutions. Seeing some alternative means of coping, and hearing first hand that they work, can raise new possibilities in the mind of a patient or his or her communication partner.

[j] The appointment can, of course, be in person if the patient and the clinician can afford the time and cost.

- Consumer groups appreciate the benefits that only groups can provide and have called for them to be available.[1530]
- When hearing aids do not provide the clarity of hearing in noise that a patient desires, the patient may attribute the cause of this to the hearing aid, rather than to his or her still-defective hearing mechanism. While the clinician can say that the hearing loss is the problem, having the patient hear other patients relate the same experience reinforces what the clinician is saying.[7] It can also be enlightening for spouses to hear that for other couples, hearing problems did not vanish the day that hearing aids were acquired.
- Because of all this, patients' rehabilitation becomes more successful and they become more satisfied.[978] A systematic review of experiments investigating the effects of group rehabilitation concluded that, in the short term, they result in decreased hearing handicap, better use of communication strategies, and increased use of hearing aids.[697] Evidence for long-term benefit is equivocal at this stage.[296, 697, 734, 736, 848] As there is a range of group activities that could be undertaken, a range of outcome types that could be measured, and severe biases in subject recruitment that make valid long-term measurements difficult, there is likely no simple yes-no answer concerning long-term benefit.
- The clinician will appreciate the variation in routine, and will gain greater insight into what it is like to have a hearing problem than is likely in individual appointments.

The value of a group may not be related to the reason the group is formed or to the apparent content of the group session. Ross (1987) recounts his experience of being in a lip-reading class. He concludes in retrospect that while the class did not increase his ability to lip-read, he received many benefits related to the "ancillary and unspoken factors intrinsic to the group experience". Experimental results back up this introspection well. In two studies, speech-reading training produced no change in speech-reading ability, but participants reported becoming more confident when conversing with others,[135] and the amount of hearing aid use increased with the extent of the group training[k] administered.[1083] This is not to say that the content of group training is necessarily irrelevant. Time spent on hearing strategies, including synthetic communication training tasks, seems likely to have more beneficial effects than time spent on analytic communication drills, for example.[980]

Any training aimed at encouraging patients to modify their behavior (including attempting to modify the communication behavior of those around them) is particularly well-suited to group appointments. Groups provide the opportunity for patients to try out new behavior patterns in a safe, supportive, encouraging environment. For example, if several people talk at once, or one person talks while looking away from the intended listener, the listener can practice requesting the talkers to modify their behavior. Polite assertiveness does not come easily to many people, and like any new skill, must be practiced and reinforced.

Groups are most commonly formed after the patients have received their hearing aids. They can, however, be formed before people have obtained hearing aids or before they have even thought about acquiring them. Many people who participate in such groups are likely to later acquire hearing aids, and apparently such people are extremely unlikely to return their hearing aids once they have obtained them.[467] In pre-fitting groups, there is the potential to involve successful users of hearing aids as "models". The presence of such people may allow those who doubt their ability to manage hearing aids to see that similar people (e.g. of the same age, gender, ethnicity, or socio-economic group) have been able to successfully adapt to hearing aid use. This, and the views of the successful hearing aid wearers, may provide considerable encouragement.

Problems in forming groups

If groups have all these advantages, why are they not more often used? There are several reasons:

- ***Appointment Logistics.*** It can be difficult to organize times that are mutually convenient for several patients. This is most difficult for a pre-fitting group. Only larger practices would have enough patients undergoing the same stage of

[k] It is possible that individual post-fitting training would also have been effective but, for financial reasons, lengthy individual training is rarely possible.

rehabilitation at the same time for there to be a reasonable number who can attend. Appointment logistics are less of an issue for post-fitting groups because the group appointment can be organized well in advance, at varying times after fitting, and can be scheduled at the end of the assessment appointment.

- ***Room.*** Running a group requires the clinic to have (or hire or borrow) a room big enough to hold everyone at once. The room should have good acoustics (i.e. low reverberation), particularly for pre-fitting groups. Even then, some participants may require assistive listening devices, such as FM systems.
- ***Uncertainty.*** Dealing with a group of people takes different skills (or perhaps just a different sort of confidence) than dealing with one person at a time. Some clinicians are reluctant to try it. A good introduction for uncertain clinicians is to pool patients with another clinician and to jointly run the group. Having two clinicians run the group is actually much easier, as there is ample time to observe and gather one's thoughts while the other clinician assumes responsibility.
- ***Perceived unwillingness.*** Some clinicians report that few of their patients are willing to participate in a group, whereas other clinicians report that most of their patients wish to participate.[1] The

Things to do in groups

- ***Explanation of hearing and hearing loss***: anatomy, frequency, intensity, the audiogram, effects on speech clarity.
- ***Consequences of hearing loss***: discussing and sharing the emotional and social consequences of hearing loss.
- ***Hearing aids and ALDs***: what a hearing aid is, its effectiveness in different situations (i.e. developing realistic expectations), explanation and demonstration of ALDs, explanation and demonstration of telecoil, care and maintenance, binaural and bilateral advantage.
- ***Hearing tactics and strategies:*** all of Section 13.6, including collaborative solving of problems volunteered by participants.

The first and third topics are more based on the clinician giving information. The emotional benefits of groups are most likely to emerge during discussion on the second and fourth topics.

The last topic is particularly suitable for groups, because the topic inherently involves interactions between people. This interaction enables participants without specialized knowledge to contribute perspectives that other members of the group will find useful. Discussion in a problem-solving context is believed more likely to lead to application of the ideas in real-life, compared to didactic, one-way teaching of the tactics.[1922]

Problem solving is the philosophy behind the ***Active Communication Education*** (***ACE***) group program, in which the patients learn to analyze and solve real-life communication problems. The program starts by identifying the communication difficulties (i.e. a needs analysis) of the participants. The group then devises potential solutions to the highest priority problems, and role plays any altered behavior that they would need to exhibit to implement the solution.[728, 733] A randomized controlled trial showed convincing benefits for hearing aid wearers, non-wearers, and significant others alike.[734, 736] Although this philosophy of helping the patients identify their own problems and discover their own solutions is the essential principle behind ACE, it is a productive aim to have for all counseling, whether group or individually based.[311] It is based on the principle that adults best learn things that interest them and are relevant to them, not just because some expert says they are important.

There are several references available that give further details on how to conduct a group rehabilitation session.[6, 7, 311, 733, 978, 979, 1041, 1208, 1898, 1920] Some of these references also contain useful handout materials.

1 In a particular case known to the author, two clinicians with opposite views on whether patients were willing to participate in a group worked in the same clinic seeing patients randomly selected from the same population.

willingness of patients to participate depends markedly on how the clinician goes about asking each patient. May & Upfold (1984) found that when the invitation was issued with enthusiasm, and the group appointment was presented as a normal part of the service structure, 87% of patients accepted the invitation, despite the median age being close to 80. The attendance rate amongst those who accepted was the same as occurred for individual appointments. Amongst those who would rather not participate will be many who are also reluctant to accept that they have a hearing loss.[792]

- ***Apparent cost.*** The time required for the group appointment has to be funded in some way. If one takes into account that time that is freed in individual appointments, and the increased satisfaction and decreased device return rate that is reported by those who conduct groups, it seems likely that group appointments save rather than cost money.[8] Consequently, it may even be cost-effective to provide a financial inducement (i.e. a discount) to patients who agree to attend group sessions, rather than consider charging them extra.[8] One study reported a much lower hearing aid return rate amongst those who participated in a post-fitting group than amongst those who did not. [m, 1350]

Structuring group activities

The topics for group discussion can be highly structured to follow a set curriculum. Alternatively, the group can have the single topic of how to improve difficult communication situations that confront the members of each group. Eventually, this will take in most hearing strategies and probably other topics as well. These alternatives can be combined, so that solving individually proposed difficult situations is just one of the set topics for the group.

Unless the sole aim of the group session is to impart knowledge from the clinician to the group members, the participants should sit in a circle to facilitate communication with each other, and to reinforce that exchange of information *among* group members is an essential part of the group's activities.

13.15 Concluding Comments

The approach to education and counseling described in this book is consistent with a ***rehabilitative model***, rather than a ***medical model***, of service delivery. In a rehabilitative model, patients actively participate in solving their own problems, rather than having their hearing diagnosed and then having some treatment done to them.[526, 1930a] In a rehabilitative model, any characteristic of the patient (i.e. not just hearing) is potentially able to affect the type of rehabilitation that the clinician chooses to carry out. In general, treatment via a rehabilitative model is believed to be more effective, and to result in greater patient compliance with treatment recommendations, than a medical model.[526] Because the patient and the clinician jointly assume responsibility for managing the problem, this approach is now being referred to across health services as collaborative self-management.

This chapter has provided many lists of things to say and do, and even a possible order in which to do them. It is essential, however, for the clinician to be continually tuned into the state of the patient. In education and counseling, listening is at least as important as instructing. The patient's comments, actions, and reactions (verbal and non-verbal) should dictate the clinician's behavior.[1762]

For example, the clinician must be skilled at differentiating between requests for information and requests for acceptance of an emotional reaction, and be able to respond appropriately. This skill can be taught to student clinicians.[521] If the clinician instead provides a content response, or worse, works rigidly through a standard, pre-established patter of questions and information giving, patients will assume that they are not being listened to.[519] In such circumstances they may be less than frank, or may appear to the clinician to be making irrelevant and repetitive comments. Competent clinicians must have communication and counseling skills that are as excellent as their technical skills if they are to best help their patients.

It can be difficult to know when the follow-up of patients should stop. Although each additional or lengthened appointment incurs costs which must be met by someone, there are certainly circumstances where the incremental cost of one additional appoint-

[m] It is not clear from this study whether people are more satisfied because they participate in group rehabilitation, or whether more highly motivated people are more likely to elect to participate. Both are likely to be true.

ment is far outweighed by the additional proportion of patients who use and benefit from hearing aids rather than become owners of permanently in-the-drawer hearing aids.[148, 1867] One understandable reason for this is that a significant proportion of patients fail to use hearing aids because of confusion over how to operate them.[1089]

The availability of computer-based auditory training programs that patients can use at home, at greatly reduced clinical cost, along with an increased understanding of what it takes to achieve neural change, promise to reverse the decline in auditory training that has occurred over previous decades for reasons of cost, lack of interest and lack of evidence.[981]

CHAPTER 14

ASSESSING THE OUTCOMES OF HEARING REHABILITATION

Synopsis

Clients and clinicians both benefit when the outcomes of the rehabilitation process (i.e. changes in the patients' lives) are measured in some way. Systematic measurement of outcomes can help clinicians learn which of their practices, procedures, and devices are achieving the intended aims. Some measures can also help determine how the rehabilitation program for individual patients should be structured and when it should be ended.

Outcome assessment can be based on an objective speech recognition test (the results of which depend hugely on the measurement conditions), or on a subjective self-report and/or the report of a significant other person. Speech test scores show the increase in the ability to understand speech in specific situations, whereas self-report measures more generally reflect the patient's views about the impact of rehabilitation. Many self-report measures have sub-scales so that outcomes can be separately assessed for different listening environments. Outcome measures can assess the domains of benefit, defined as a reduction in disability (comprising activity limitation and participation restriction), device usage, listening effort, quality of life, or the satisfaction that the patient feels.

Self-report measures that assess benefit can be grouped into various classes. First, patients can be asked to make a direct assessment of the benefit of rehabilitation. Alternatively, patients' views of their disability can be assessed both before and after the rehabilitation program. The change in score provides a measure of the effects of rehabilitation. Measures obtained both before and after rehabilitation provide a more complete view of disability status and change. These difference measures probably assess change less accurately than those that directly assess benefit because they involve subtracting two scores.

The second way in which self-report measures differ from each other is the extent to which the items are the same for all patients or are determined individually for each patient. Results can more easily be compared across patients if a standard set of items is used for all patients. When the items are individually selected for each patient, however, the questionnaires become shorter and can more easily be incorporated within interviews with the patient. They are also more relevant to each patient.

There are thus four types of self-report measures: standard questionnaires that directly assess benefit (e.g. HAPI); standard questionnaires that compare disability before and after rehabilitation (e.g. HHIE, APHAB); individualized questionnaires that directly assess benefit (e.g. COSI); and individualized questionnaires that compare disability before and after rehabilitation (e.g. GAS).

Self-report measures also commonly assess hearing aid usage (which can now also be measured objectively with data logging) and are the only viable way to assess satisfaction. Some measures contain questions that address only one domain (benefit, use, or satisfaction) whereas others address more than one domain. One comprehensive questionnaire (GHABP) addresses all three dimensions, contains standard and individualized measures, and assesses benefit both directly and by comparing disability before and after rehabilitation. A very simple and widely used questionnaire that assesses several domains with a single question each is the International Outcomes Inventory for Hearing Aids (IOI-HA).

Some questionnaires are designed to assess problems experienced with the hearing aid, although freedom from problems with the hearing aid is more properly viewed as a means to an end rather than a life-changing outcome.

While outcomes can be assessed any time after hearing aid fitting, the extent of benefit does not appear to stabilize until about 6 weeks after fitting. Hearing loss is associated with a decrease in many aspects of quality of life (such as increased depression) and use of hearing aids is associated with general improvements in health and quality of life. Causal relationships between these quantities are difficult to establish, however. Generic measures of health outcome are not efficient means by which a clinician can assess the outcomes of rehabilitation.

Lord Raleigh said that if we cannot measure something, we do not know much about it. Measuring the outcomes of a hearing aid fitting can give us invaluable insights into how the services and devices we have provided have affected the lives of our patients. As we review methods for measuring the effects of hearing rehabilitation, we should recognize that although hearing aids are the major components of most rehabilitation programs, they are rarely the *entire* program (see Chapter 13). Rehabilitation outcomes are thus likely to be affected by all aspects of the rehabilitation program. There are several reasons why clinicians might choose to measure rehabilitation outcomes:

- A clinician may want to determine if particular rehabilitation procedures, devices, or entire programs are more effective than others in helping his or her patients. *Rehabilitation procedures* include things as mundane as the way the clinician instructs a patient to insert an earmold, and as complex as helping a patient come to terms with the appropriateness of being assertive in a difficult listening situation.
- A clinician may want to determine whether he or she has sufficiently helped a patient. The answer may determine whether further appointments should be scheduled, and whether a change of tactics is required for this patient.
- A third-party provider of funds for health care (government or an insurance company) may make funding conditional on obtaining evidence that rehabilitation is beneficial to patients. Benefit may have to be substantiated for a hearing-impaired population as a whole, averaged across a sample, or for each individual.

14.1 Outcome Domains

What do we mean by "outcomes"? In general, *an outcome is something that changes in the life of the patient as a consequence of the services and devices provided to that patient by the clinician*. Some specific outcomes that we aim to achieve are:

Decreased activity limitation. We want patients to hear more of the sounds around them, and better understand speech in a range of situations. The World Health Organization previously described activity limitation as ***disability***.[a]

Decreased participation restriction. We want patients not to restrict the social, occupational and recreational activities they choose to participate in because of the difficulties caused by their hearing loss. The World Health Organization previously described such restrictions as ***handicap***.

Decreased listening effort. People with hearing loss find it tiring to communicate in many situations. We want hearing aids to decrease the effort that people have to make to understand. Hearing aids do achieve this, but listening effort is relatively unstudied,[678] even though it may be the only way to objectively demonstrate the advantage of adaptive noise reduction.[1547]

Decreased emotional consequences. Hearing loss commonly leads to a range of negative emotions (Section 14.8.2) and we hope that using hearing aids will decrease or eliminate these feelings.

Quality of life. Quality of life is a very general concept that is affected by many things, including ease of communication, so we expect that hearing aid use will improve overall quality of life.

Use. We want patients to use the devices we provide them in every situation in which they are having trouble hearing.

Satisfaction. Patients have contributed time and possibly money, and have probably undergone some emotional stress from participating in rehabilitation. We hope that they, and any family members involved, will feel satisfied with both the process and the results.

We will refer to the first five of these outcomes as ***benefits*** of rehabilitation. Hearing aid use should be regarded as an important means to an end rather than as a goal in itself. Satisfaction is affected by benefit, but also involves patient's expectations, monetary and psychological costs, problems encountered, and any communication difficulties that remain. As we shall see, all of these outcomes can be assessed through questionnaires and other self-report techniques. We will return to self-assessment in this chapter after examining how benefit can be assessed through speech identification testing.

[a] WHO now uses *disability* as a blanket to term to cover all of impairment, activity limitation, and participation restriction. It no longer defines the term *handicap*.

14.2 Speech Understanding Tests

The major reason hearing-impaired people seek help is to hear speech more clearly.[451] Speech tests are a direct and objective way to measure how much more clearly people can understand speech with their hearing aids than without them. There are many speech tests already developed from which to choose. These vary from tests that are very easy because they include a lot of context or have few highly contrasting response alternatives, through to tests that are very hard because there is little or no context available.

Speech test results can be made as repeatable as is necessary, simply by including enough items in the speech test.[664, 1781] Computer-based presentation and scoring techniques enable groups of words to be scored on a phoneme basis (rather than word scoring or sentence scoring) in a reasonable time.[603] Phoneme scoring maximizes the number of scored items per minute of testing time, which therefore maximizes the reliability of the speech score, given the testing time available. Speech tests consequently provide a ready means to assess the benefit of hearing aids. In particular, they assess the presumed reduction in activity limitation related to understanding speech.

14.2.1 Limitations of speech tests to assess benefits

Despite these considerable advantages, speech tests are neither *efficient* nor *sufficient* means of demonstrating the overall benefit that hearing aids provide to a patient. There is one major reason for this, which we will consider first, and several minor reasons.

Dependence on measurement conditions

The amount of benefit that hearing aids provide depends hugely on the acoustic environment and level of background noise (if any). The details and reasons for this have been covered in Section 9.1.6, but, in brief, hearing aids are most effective when signal levels are low and where, consequently, the patients' unaided thresholds limit audibility. Hearing aids are least effective in noisy places where audibility is limited by background noise. A hearing aid can thus be shown to provide a large amount of benefit, or very little benefit, depending on the target stimuli and competing sounds chosen for the test. The result cannot be a general indicator of benefit if the result depends on the measurement condition chosen by the clinician.

We cannot avoid the problem by measuring in several conditions and simply summing or averaging the results. Two patients with identical hearing losses and auditory processing capabilities, fitted with identical hearing aids, may have vastly different perspectives on how beneficial their hearing aids are. If one of them spends a lot of time in noisy, reverberant places, and the other spends a lot of time in quiet places listening to softly-spoken people, both will have excellent reasons for coming to opposite conclusions about the benefit their hearing aids provide. Not surprisingly, there is only moderate correlation between objective measures of benefit and self-reported benefit. Visual cues are rarely made available to the patient in clinical tests of speech understanding, but are often available in real life. This difference contributes to the difficulty of predicting real-life performance from clinical measurements.

Efficiency relative to other means of measurement

Hearing aids increase speech identification ability primarily by increasing audibility. The amount by which they increase audibility depends on the speech level and spectrum, background noise level and spectrum, the patient's threshold at each frequency, and the real-ear gain of the hearing aid at each frequency. These are all acoustic or electroacoustic variables. The ***Speech Intelligibility Index*** (***SII***) method allows us to combine them to predict aided speech intelligibility based on unaided intelligibility.[432] Figure 14.1 shows

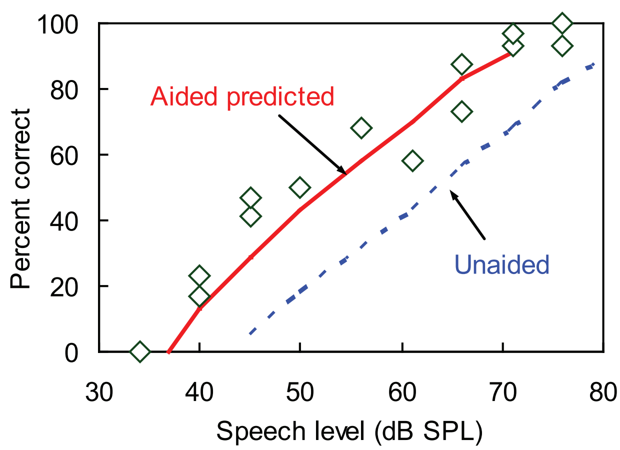

Figure 14.1 Data for one subject showing the aided speech performance (diamonds) and the aided performance that was predicted (solid red line). Predictions were based on the insertion gain, the background noise present, and the unaided performance (dotted blue line).

an example for one subject and one speech test. If we know the unaided speech performance-intensity function, we can predict the aided function based on the patient's thresholds and various electroacoustic measures. For patients with mild and moderate hearing loss, reasonably accurate predictions of speech intelligibility can be made using the SII even without knowing the individual's unaided speech intelligibility.[1124, 1125]

This is not to say that we *should* make such predictions, but that if such predictions *can* be made from electroacoustic measures, a comparison of aided and unaided speech performance is not really telling us anything new. It is faster and easier to measure an insertion gain curve than it is to measure unaided and aided speech understanding. Furthermore, the insertion gain curve is immediately useful: deficiencies relative to a prescription become evident, as do excessive peaks or troughs, so the corrective action necessary is obvious. Conventional speech tests can indicate that there is little benefit, but they do not indicate how we should change the hearing aid's characteristics to get a better result. If only aided speech understanding is measured, then a poor score (defined somehow!) could be caused by a poor hearing aid fitting, but equally by an auditory processing or cognitive problem in the patient.

Speech tests presented at a number of levels enable a ***performance-intensity function*** to be visualized (as in Figure 14.1). This function enables the clinician to determine the range of speech levels over which speech scores exceed some criterion score. An unduly narrow range would lead the clinician to question whether the hearing aid has enough compression. While the approach has potential, the difficulties are many: What is an acceptable speech score? Over how large a range of input levels should this score be achieved or exceeded? If the compression ratio of the hearing aid is increased, how will the clinician determine when the ratio is so large that speech quality has deteriorated? When this happens, how will the intelligibility advantages of the larger compression ratio be weighed against its quality disadvantages? In short, measuring a performance intensity function raises many important questions, but provides few answers.

Reliance on speech

Speech tests do not measure several potential benefits. Hearing aids can help people detect and recognize environmental sounds, and thus lead to a greater feeling of security by the aid wearer.[531] Hearing aids also help people monitor their own voice level and quality, especially for people with severe and profound hearing loss.

14.2.2 Role of speech testing in evaluating benefit

None of this is to suggest that speech tests have *no* role in assessing benefit. In fact, they are very useful for several things:

- If one can identify and simulate specific acoustic conditions, speech tests provide a clear assessment of how much the hearing aids change the person's ability to understand speech in this situation. The ability to store and quickly access a range of speech and noise materials on CD and computers has increased the feasibility of simulating, in the clinic, environments that are relevant to particular patients. In fact, simulating a known environment is relatively easy. Knowing *what* to simulate is difficult, because the conclusions reached are strongly dependent on having the right speech and noise spectra and levels, appropriate reverberation, and appropriate context in the speech material used.
- Identifying the types of speech sounds that are not well perceived is useful for evaluating the type of benefit that hearing aids provide. The ***Ling six-sound test*** /a, i, u, m, ʃ, and s/ for example, can be used to assess audibility and recognition of these sounds, which each have a relatively high intensity in at least one frequency region. The results apply only to speech presented at the overall level tested.[341]
- Speech tests can provide a convincing demonstration of benefit (in the condition tested) to a patient or to a family member. This can be worthwhile if either of them is not convinced that hearing aids can provide such benefit.
- Speech tests can demonstrate to patients and relatives the importance of visual cues to understanding. Clinicians could make more use of such demonstrations than commonly occurs.
- As we will see in Chapter 15, speech tests can be used to help decide whether a person should wear one or two hearing aids, or in which ear a single hearing aid should be worn.

- Speech tests can be used to predict how much difficulty a patient will have communicating in some specified environment while wearing hearing aids. This information can help the clinician decide if the patient needs some form of communication training or the provision of assistive listening devices or a cochlear implant (Section 9.2.2).
- If speech perception training is to be provided, speech tests can determine the level of training (speech feature, phonetic, supra-segmental etc) that should be offered.

In summary, speech identification tests are an excellent way to measure the benefit of hearing aids, provided one is interested only in the scores obtained in a specific environment, and provided this environment can adequately be simulated in the test room. If one wishes to assess benefit or communication effectiveness across many environments, or in environments that cannot accurately be specified or simulated, other measures of benefit have to be used.

14.3 Self-report Questionnaires for Assessing Benefit

14.3.1 Questionnaire methodology

Another way to assess benefit is to ask the patient, via a questionnaire, how beneficial the hearing aids are. Each item in a questionnaire asks patients to rate something about their hearing ability or ease of communication in some specific situation. A situation might be described in such a way as: *You are talking to a shop assistant in a busy store*. Additionally, simple pictures can help the patient identify each type of situation being described.[973] For each item there are a number of response alternatives and this number can vary from three (e.g. the Hearing Handicap Inventory for the Elderly)[1851] to eleven (e.g. the Gothenburg Profile).[1512] There are two ways we can use the information obtained from questionnaires to measure benefit:

Direct change measures. Patients are asked to directly estimate the degree of benefit their hearing aids provide in each designated situation. Response options are generally a set of words or phrases that vary evenly from some negative rating to some positive rating. Questionnaires that directly assess benefit need be answered only once, and this obviously has to be after the patient has received hearing aids and the associated rehabilitation activities. The ***Hearing Aid Performance Inventory*** is an example of this approach. As illustrated by the HAPI example in the panel below, a very specific communication situation is described and the patient is asked to note the "helpfulness" of the hearing aid in that situation. By doing so, a direct measure of relative change, in this case described as "helpfulness", is obtained. Each rating is given a score (e.g. 1 to 5 in this example), and the scores for all the items in the questionnaire are summed or averaged to produce the final measure of benefit.

State measures. The alternative to direct assessment of benefit is to ask patients how well they can hear in, or how much they avoid, the designated situations. In this case, they are asked twice: once for the

Examples of self-report items and response choices

Abbreviated Profile of Hearing Aid Benefit (APHAB)[364]

I have difficulty hearing a conversation when I'm with one of my family at home.

[*Always, Almost always, Generally, Half-the-time, Occasionally, Seldom, Never*]

Hearing Handicap Inventory for the Elderly (HHIE)[1851]

Does a hearing problem cause you to avoid groups of people?

[*Yes, Sometimes, No*]

Hearing Aid Performance Inventory (HAPI)[1872]

You are talking with the bank teller at the bank.

[*Hinders, No help, Very little help, Helpful, Very helpful*] unaided state and once for the aided state.[b] Response options are generally a set of words or phrases that vary evenly from extreme difficulty in hearing to no difficulty in hearing. In the ***unaided administration*** of the questionnaire, patients state how well they can hear when they are not wearing their hearing aids. This provides a ***baseline measure*** of hearing disability. Patients then answer all the questions a second time; the questionnaire is the same, but this time they state how well they hear while they are wearing their hearing aids (i.e. the ***aided administration***). Each questionnaire is scored and benefit is defined as the difference between the aided and unaided scores. The ***Abbreviated Profile of Hearing Aid Benefit*** and the ***Hearing Handicap Inventory of the Elderly*** (see panel on page 407) are examples of this approach.

Understanding the construction, reliability, validity and application of questionnaires: Factors, subscales, internal consistency, item-total correlation, test-retest differences and effect size

The items in questionnaires are often grouped into ***subscales***. The grouping may be based just on the apparent content of the item, or it may be done on a statistical basis by analyzing the results of a large number of patients and then grouping items using either ***factor analysis*** or ***principal components analysis***. These techniques examine how correlated the answers of each item are relative to each other item, and attempt to find some common underlying factors that are highly correlated to a number of items. These items are said to be ***loaded*** onto this common factor and it is these items that are grouped to form a subscale. Simply expressed, if a patient assigns a high rating to one item, the patient is also likely to assign a high rating to other items in that same subscale. The meaning of the factor, and hence of the subscale, is determined by examining the items that form the subscale and noting what content they have in common.

The reason for using multiple items in each subscale is to increase the accuracy of the subscale, and of the entire scale. One way to express accuracy is via the ***internal consistency*** or ***convergence*** of the subscale or scale. This characteristic is estimated by a statistic called ***Cronbach's alpha***. It is equal to the correlation that, on average, would be obtained between two randomly divided sets of items. If half the items in a scale give a very similar total score to that of the other half, we can be confident that neither half is producing random results. Consequently, when the two halves are recombined, the total scale also must produce a repeatable result. A good subscale will contain multiple items that look different, but in fact have a high internal consistency and thus provide multiple estimates of the same underlying phenomenon.

One way to produce such a scale is to weed out items that appear to be measuring something different from the rest of the items. This is achieved by calculating the correlation between the score for one item and the total score for all the remaining items. Items with a low ***item-total correlation*** are deleted. Generally, their content appears to contain concepts different from those of the remaining items.

The ***validity*** of scales and sub-scales involves several concepts:

- A measure has ***face validity*** if it subjectively appears to measure what it claims to measure.
- A measure has ***content validity*** if it includes sufficient items to cover the types of feelings or behaviors that would normally be affected in the condition the scale studies.
- A measure has ***convergent validity*** if it correlates highly with another measure that claims to measure the same attribute.
- A measure has ***criterion validity*** if it correlates adequately well with another measure that claims to measure a related (but not identical) construct. If the two measures are administered at the same time, this is also called ***concurrent validity***, and if the other measure is administered some time in the future, then the first measure has ***predictive validity***.

[b] Although we refer to these states (or conditions) as unaided and aided for convenience, the aided condition will probably be affected by things other than wearing hearing aids. The primary example of another difference between the two states is the counseling that the patient will have received prior to the aided administration.

Understanding the construction, reliability, validity and application of questionnaires: Factors, subscales, internal consistency, item-total correlation, test-retest differences and effect size (continued)

- A measure has ***discriminant validity*** if it is not too highly correlated with another measure that claims to measure something different (e.g. an activity subscale versus a participation subscale).

Two other desirable statistical properties are relevance and inter-patient variability. There is no point including an item if a large proportion of patients consider that the situation described is not relevant to them. One solution to this problem is discussed in Section 14.4. Alternatively, the questionnaire can include further questions that directly ask about the relevance of each item. Similarly, an item has little predictive value if nearly all patients answer it with the same rating.

As with any measure, the reliability of a questionnaire can be assessed by calculating the correlation between scores obtained with a group of patients versus the scores obtained on a retest using the same questionnaire and the same patients. The people chosen to be tested, however, heavily influence these test-retest correlations. The test-retest correlation of an audiometric threshold, for example, would be extremely poor if we only tested people with normal hearing: The apparent variations from test to retest would be comparable to the true variations between people in the sample. By contrast, if our sample included people with a range of threshold sensitivity from normal to profound impairment, the test-retest correlation would be extremely high. The audiometric test, however, is the same! Test-retest correlations of any measure should be treated with great caution as they apply only to the particular population tested.

A more robust and useful measure is the standard deviation of the test-retest differences. This indicates the spread of test-retest differences that can be expected due to chance variations. Two scores are significantly different at the 95% confidence level if they are separated by approximately two test-retest standard deviations or greater. Note that the test-retest differences expected for a questionnaire may depend on the method of administration as well as on the items. For one test, for example, the critical differences are almost twice as large when the client self-administers the test as when the clinician administers it.[1903]

Questionnaires are usually applied to understand the impact of a condition (e.g. hearing loss) or a treatment (e.g. hearing aids) on the abilities, activities or emotions asked about in the questionnaire. The effect of the condition or treatment is inferred by comparing the scores for two groups of people (with and without the condition or treatment), or for the same group of people before and after receiving the treatment. ***Significance testing*** of the difference in scores tells us how likely it is that there is any reliable connection between the condition/treatment and the outcomes asked about in the questionnaire, or whether the differences observed could be the result of chance alone. With large numbers of research participants, it is possible to have a statistically significant result, even if the link between the condition/treatment and the outcome is very weak indeed.

The ***effect size***, by contrast, tells us how large the effect of the condition or treatment is relative to the variation between participants in the aspect of life measured by the questionnaire. A useful measure of effect size, called ***Cohen's d***, is calculated as the ratio of the mean change in questionnaire scores as a result of the condition/treatment to the standard deviation of scores within each of the groups or within each of the measurement times. Standard deviations within different groups or measurement times are first pooled.[313] Values of d = 0.2, 0.5 and 0.8 are regarded as small, medium and large respectively. Effect size is not an invariant property of the questionnaire, nor of the condition/treatment, nor of the population, but rather shows the effect of a particular condition/treatment on a population having particular characteristics, for that particular questionnaire. Questionnaires are none-the-less loosely said to be sensitive or insensitive to particular treatments or conditions.

Further explanation of the psychometrics of self-report questionnaires can be found in an excellent paper by Demorest & Walden (1984).

The first administration can be before the patients receive their hearing aids or some weeks or months after they receive them. Both times of administration have pluses and minuses. If the unaided questionnaire is administered before provision of hearing aids, there has to be a delay of weeks or months before the patient has accumulated enough experience with the hearing aid for benefit to be assessed (Section 14.7). If the general mood of the patient is more negative or positive on the day of the second administration than on the first, this could affect the difference score markedly. Having the patients complete both questionnaires on the same day can solve this problem. Delayed administration of the unaided questionnaire, however, introduces a new problem: patients who are full-time hearing aid users in any of the situations addressed may have difficulty remembering how much trouble they had hearing unaided in those situations.

Either overall type of assessment method (direct change or state measures) can be used, but each type has its advantages and disadvantages:

- The aided versus unaided state method is probably a less accurate (and hence less sensitive) measure of benefit than direct change measures.[448, 593] When separate unaided and aided questionnaires are administered, the scores for each have to be subtracted to calculate benefit. Unfortunately, the random errors implicit in each score add,[c] so the result can have an error component comparable in size to the benefit measured.

- The direct change method does not reveal the full picture. A small benefit is not a problem if it occurs in a situation where the patient has little difficulty hearing. Conversely, a small benefit is a serious problem if it occurs in a situation where the patient has a lot of difficulty hearing. The direct change method, by itself, does not allow us to distinguish between these two extremely different cases.

- With state measures, some patients will underestimate, and some will overestimate their actual disability.[1550] Fortunately, their pessimism or optimism does not much affect their estimate of hearing aid benefit, as it applies equally to both the unaided and aided estimates of disability.[1552]

- In principle, the best compromise seems to be a combination of two types of questions. The first type directly assesses benefit. The second type assesses either the initial disability, or else the residual disability after the rehabilitation program has been completed. This situation can be approximated in the unaided versus aided approach if the patients are allowed, while they are doing the aided questionnaire, to see and change the answers they previously gave for the unaided questionnaire.[337] The intention is that the patients will mark the aided scale in such a way that their ratings, relative to those on the unaided scale, reflect their direct assessment of benefit. The degree to which patients can simultaneously respond in this absolute and relative manner, and the resulting effect on accuracy, has not been quantified.

Irrespective of which approach is used, the items on the questionnaire can be grouped into ***subscales*** that examine different aspects of benefit (see panel on Understanding Questionnaires). For example, one subscale could relate to listening in quiet situations, and another could relate to listening in noise.

Although this section has discussed the patient's own perception of the changes that occurred following rehabilitation, the concept can easily be generalized to reports by other people who have frequent contact with the patient. Most commonly this is the spouse[300, 1378] but can also be a friend, and for a child, it could be a parent or teacher. Hearing loss in one person generally has adverse effects on that person's frequent communication partners: frustration, annoyance, stress from being continually responsible for interactions with others as an intermediary, guilt over conversation neglect, excessively loud TV and radio, or restriction of social activities.[184]

Not surprisingly then, questionnaires have been developed to assess the effect of hearing loss or its rehabilitation from the perspective of the significant other person.[184, 1329, 1558, 1693] In principle, questionnaires for significant others can be designed to assess the other person's opinion of the disability or benefit from rehabilitation experienced by the patient, or the effect of the patient's disability or rehabilitation program on the significant other person. The latter is referred to as ***third-party disability***,[1557, 1559] and is a different concept

[c] This additivity assumes that the random fluctuations in the unaided answers are not correlated to the random fluctuations in the aided answers.

than viewing the views of the significant other as a proxy for the patient's views. These are different, but each is important in its own right.

For patients with dementia, outcomes measurement by a caregiver is the only viable option. One quantitative method comprises having the caregiver record and count the number of occasions on which the patient displays selected negative and/or positive behaviors following fitting of hearing aids.[496, 1377, 1378] It cannot be expected that behaviors affected by hearing loss will change overnight following the fitting of hearing aids. Durrant et al (2005) give examples of patients with Alzheimer's Disease displaying a gradual reduction in the number of daily adverse events (e.g. apparent forgetfulness, asking for repetition) in the days and weeks following hearing aid fitting.

14.3.2 Practical self-report measures

Unaided and aided questionnaires

If one wishes to separately measure the aided and unaided states, there are several well-standardized questionnaires from which to choose. With this approach, any questionnaire that assesses disability can be used to derive a benefit score. It is important, however, that any questionnaires used quantitatively have known psychometric properties (see previous panel). Some questionnaires that could be answered in a way that relates to either aided or unaided communication ability are shown in Table 14.1. These questionnaires are particularly suitable for adults; Section 16.6.4 shows some measures that are more suitable for children.

Table 14.1 Questionnaires assessing hearing disability.

	Questionnaire	Authors	Year	No of items
HHS	Hearing Handicap Scale	High et al.	1964	20
HMS	Hearing Measurement Scale	Noble & Atherley	1970	42
SHI	Social Hearing Handicap Index	Ewertsen & Birk-Nielson	1973	21
DS	Denver Scale of Communication Function	Alpiner et al.	1974	25
WISH	Weighted Index of Social Hearing Handicap	Brooks	1979	19
HPI	Hearing Performance Inventory	Giolas et al	1979	158
HHIE	Hearing Handicap Inventory for the Elderly	Ventry & Weinstein	1982	25
QDS	Quantified Denver Scale of Communication Function	Schow & Nerbonne	1980	20
RHPI	Revised Hearing Performance Inventory	Lamb, Owens & Schubert	1983	90
HHIE-S	Hearing Handicap Inventory for the Elderly - Screening	Ventry & Weinstein	1983	10
CPHI	Communication Profile for the Hearing Impaired	Demorest & Erdman	1987	145
PIPSL	Performance Inventory for Profound and Severe Loss	Owens & Raggio	1988	74
SAC	Self Assessment of Communication	Schow et al	1989	10
PHAP	Profile of Hearing Aid Performance	Cox & Gilmore	1990	66
OI	Oldenburg Inventory	Holube & Kollmeier	1991	21
PHAB	Profile of Hearing Aid Benefit	Cox et al.	1991	66
HHIA	Hearing Handicap Inventory for Adults	Newman et al.	1990	25
APHAB	Abbreviated profile of Hearing Aid Benefit	Cox & Alexander	1995	24
HCA	Hearing Coping Assessment	Andersson et al.	1995	21
AIADH	Amsterdam Inventory for Auditory Disability & Handicap	Kramer et al.	1995	60
GP	Gothenburg Profile	Ringdahl et al.	1998	20
GHABP	Glasgow Hearing Aid Benefit Profile	Gatehouse	1999	28-56
SSQ	Speech, Spatial and Qualities of Hearing Scale	Gatehouse & Noble	2004	50
QDS-m	Quantified Denver Scale – modified for significant others	Stark & Hickson	2004	20
EAR	Effectiveness of Auditory Rehabilitation	Yueh et al	2005	20
SOS-HEAR	Significant Other Scale for Hearing Disability	Scarinci, Worrall & Hickson	2009	27
SHQ	Spatial Hearing Questionnaire	Tyler, Perreau & Ji	2009	24

Applications of APHAB

- Some predictions can be made about likely benefit from hearing aids, as detailed in Section 9.1.4.
- Patients can be advised how much benefit they get relative to other users of hearing aids and/or how much difficulty they have, when aided and when unaided, relative to normal-hearing people. If they are in doubt about continuing with hearing aids, this may help them decide whether to keep their hearing aids, try other hearing aids, or discontinue attempts to wear hearing aids.
- The relative benefit provided by different types of hearing aids *could* be assessed using APHAB, but the large differences needed for statistical significance make APHAB (and other self-report measures) fairly insensitive for this purpose, except when averaged over a number of people.
- Particularly poor benefit scores in background noise can indicate the need for directional microphones or wireless transmission of signals.
- Large negative scores on the Aversiveness scale can indicate the need for a lower SSPL and/or a higher compression ratio for medium- to high-level signals.

Table 14.1 lists the APHAB measure because one half of the questionnaire assesses unaided ability and the other half assesses aided ability. The two halves are otherwise identical. These same comments apply to its longer parent questionnaire, the PHAB, but the shorter version is more practical for clinical use. The APHAB questionnaire is scored separately for its four subscales: *Ease of Communication (EC)*, *Reverberatio*n (*RV*), *Background Noise (BN)*, and *Aversiveness of Sounds (AV)*. The first three of these assess the increase in speech understanding in various everyday environments, and the last assesses negative reactions to more intense sounds. It is usual for this last measure to reveal a negative benefit. That is, patients find intense sounds more unpleasant when they are wearing their hearing aids than when they are not. A particularly poor score on this subscale indicates that the limiting and/or compression ratio for intense sounds should be reviewed.

A dilemma with all questionnaires incorporating subscales is that the most specific information is obtained if one examines each subscale score, whereas the most reliable information is obtained if one examines the total score (because of the greater number of items). Because the first three subscales in the APHAB are measuring something very different from the Aversiveness subscale, a good compromise is to combine the scores for the first three subscales into a single global measure of speech understanding benefit, and to keep the final subscale separate.[364] Table 14.2

Table 14.2 Statistics for the APHAB scale and subscales. Large numbers for the unaided and aided scores indicate a lot of difficulty in hearing, whereas large numbers for the benefit scores indicate that the hearing aid provides substantial benefit.[364]

Scale / Subscale	No of items	Median unaided problems	Median aided problems	Median benefit	Critical difference for aided score change (p=0.1)
Ease of Communication	6	65	16	41	22
Reverberation	6	81	33	39	18
Background Noise	6	81	37	35	22
Aversiveness of Sounds	6	17	60	-25	31
Global score (avg of EC+BN+RV)	18	73	31	37	14

Using APHAB

- The APHAB questionnaire can be downloaded[338] as either a printed form, or as a software program that can be used directly by the patient. This program can also be used to score the responses, irrespective of whether the patient uses the paper and pencil version or the computer version.
- Explain to the patient that the *Always* end of the scale sometimes means easy listening and sometimes means extreme difficulty in listening. This makes patients consider each item carefully, but unfortunately makes the questionnaire too difficult for some elderly patients to complete.
- Administer the unaided part of the scale before hearing aids are fitted.
- Administer the aided portion several weeks after fitting. Allow the patient to see, and change if they desire, the answers they previously gave for the unaided scale.
- Explain that it will help you to receive an honest account of how well, or poorly, the patient hears when aided, so that you can be sure the hearing aid is adjusted as well as possible.

The steps in this panel are based upon Cox (1997). Further instructions and applications can be found in that paper and at the web site referred to above.

shows some statistics for the APHAB subscales and for the global benefit combination. The critical difference is defined here as the difference needed between two administrations (e.g. with different devices) for a 10% probability of the difference occurring by chance alone.[d]

The Hearing Handicap Inventory for the Elderly (HHIE) has been used in numerous studies, including randomized controlled trials,[1946] to demonstrate and quantify the benefit associated with hearing aid fitting. After hearing aid fitting, handicap scores typically decrease by 20 to 30 points (on a 100-point scale). The HHIE has two sub-scales (Emotional and Social/situational) and each of these subscales contributes approximately equally to the decrease in measured handicap.

Recent developments in questionnaires have focused on situations where binaural auditory processing achieved by listening with two ears provides an advantage over listening with one ear (Chapter 15). The ***Speech, Spatial and Qualities of Hearing Scale*** (***SSQ***)[599] and the ***Spatial Hearing Questionnaire*** (***SHQ***)[1809] are both suitable for assessing the benefit of bilateral devices over unilateral devices.

Direct change questionnaires

If one wishes to use the direct assessment approach, there are fewer choices, as shown in Table 14.3. The ***SHAPI*** and ***SHAPIE*** questionnaires are shortened versions of the original HAPI questionnaire. In one evaluation of the sensitivity and reliability of different methods for assessing benefit,[448] scores obtained with the SHAPIE were better correlated with an overall consensus measure of benefit than were scores obtained using the HHIE or a modified version of the PHAB.[e] In another evaluation, the ***Effectiveness of Auditory Rehabilitation*** (***EAR***) Scale,[1945] was more sensitive to the effects of fitting new hearing aids than either the HHIE or APHAB measures. These differences are possibly because the SHAPIE and some questions in the EAR directly measure change, whereas the PHAB, APHAB and HHIE require the aided and unaided scores to be subtracted.

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[d] Critical differences for the stricter 5% probability level commonly used in research studies are about 1.2 times those shown in Table 14.2.

[e] Unfortunately, the PHAB was not administered in the manner now recommended for the APHAB, in which patients can see and modify their unaided answers while completing the aided scale. The sensitivity of APHAB administered in this way, relative to SHAPIE or other direct measures of benefit, is unknown.

Table 14.3 Questionnaires directly assessing improvement in hearing disability.

	Questionnaire	Authors	Year	No of items
HAPI	Hearing Aid Performance Inventory	Walden, Demorest & Hepler	1984	64
HAUQ	Hearing Aid Users' Questionnaire	Forster & Tomlin	1988	6
HAR	Hearing Aid Review	Brooks	1990	5
SHAPI	Shortened Hearing Aid Performance Inventory	Schum	1992, 1993	38
SHAPIE	Shortened Hearing Aid Performance Inventory for the Elderly	Dillon	1994	25
GHABP	Glasgow Hearing Aid Benefit Profile	Gatehouse	1999	28-56
EAR	Effectiveness of Auditory Rehabilitation	Yueh et al	2005	20

The two questionnaires in Table 14.3 with a very small number of items measuring benefit (HAUQ and HAR) both contain further questions related to other aspects of hearing aid use, as discussed later in this chapter. The EAR also contains questions covering a range of domains, including cosmetic appearance, reliability and convenience that few other questionnaires include.

14.4 Meeting Needs and Goals

Self-report questionnaires are extremely useful, and have been much used in recent years - particularly the APHAB, and increasingly, the SSQ. The EAR appears to have great potential for hearing aid evaluation. There are, however, four main problems with using them:

- Some patients do not like completing them, particularly if many of the items describe situations that the patients consider are irrelevant to them.
- Some clinicians do not like administering them, and if the questionnaires have been self-administered, some clinicians do not like scoring them.
- If the patients spend time in one or two situations in which they would particularly like to hear more clearly, standard questionnaires will give no insight as to what those situations are, nor what degree of improvement the rehabilitation has provided in those situations.
- Some patients, especially some elderly ones, have difficulty understanding complex questionnaires.

A solution to all three of these problems is to ask each patient to make up his or her own questionnaire. A patient is asked, in an open-ended manner, to list situations in which he or she is having trouble hearing and in which he or she would like to hear more clearly.[80] At the end of the rehabilitation process, the outcomes are assessed for each of these specific situations. Situations are thus never irrelevant to the patient, and the information obtained from the patient at the commencement of rehabilitation may help guide the way the rehabilitation program is organized, or the type of devices that are fitted. The disadvantage of this approach is that it is more difficult to compare results across patients or across populations.

We can liken this type of evaluation more to a structured interview than to administering a questionnaire. There are, however, some parallels with the questionnaire approach. Patients can be asked to directly estimate how much benefit they receive from their hearing aids (the direct assessment of benefit). Alternatively, they can be asked how well they hear in each situation when unaided, and then how well they hear when aided. As with the questionnaire methods, the benefit is the difference between the two scores.

Both of these methods have been used. McKenna (1987) applied a technique known as ***Goal Attainment Scaling*** (***GAS***) that previously had been used in the area of mental health (see Section 9.1.6).[930] At the initial interview, two pieces of information are collected for every listening situation: how well the patient initially hears in that situation, and how well he or she would need to hear in that situation if the rehabilitation is to be considered a success. The desired hearing ability is decided after some negotiation between the patient and the clinician. At the end of the rehabilitation program, the patient is asked how well he or she can now hear in each situation. This answer can be compared to both the initial hearing ability (to assess improvement) and to the desired ability (to assess whether any further rehabilitation should be attempted).

The method has advantages and disadvantages similar to the aided-unaided method for questionnaires: estimates of both initial and final disability are obtained, but the benefit measure may not be very accurate because it involves subtracting two potentially similar ratings. There is another disadvantage when it is used routinely. Some clinicians say they do not like administering the GAS at the initial interview, before they have established a good relationship with the patient, because of the tediousness of quantifying difficulty and establishing quantitative goals for each listening situation.[448]

These difficulties were overcome by devising the ***Client Oriented Scale of Improvement*** (***COSI***).[448] The broad idea is the same: the clinician identifies important individual listening situations at the initial interview, but does not do any quantifying at this stage. The quantitative part of the administration all occurs at the final appointment (see panel on Administering COSI). If the COSI results are expressed as a number, they can be compared to the scores obtained by a large sample of hearing- impaired people with predominantly mild and moderate hearing loss.

Figure 14.2 shows the proportion of patients who obtain different COSI change scores. It is apparent that many patients indicate a rating of "much better" (scored as 5.0) for all the listening situations they nominate. This skewing towards high scores makes the COSI unsuitable for detecting patients who report scores that are above average, but well suited to detecting patients who report abnormally low benefit.

Having the "items" nominated by the patient only partially solves the problem of some situations being more important than other situations. Although a patient presumably would never nominate a listening need that is irrelevant to that patient, the situations nominated may vary in importance because:

- the patient spends more time in some situations than in others;
- the patient has more difficulty in some situations than in others; and
- understanding everything that is being said is more important in some situations than in others.

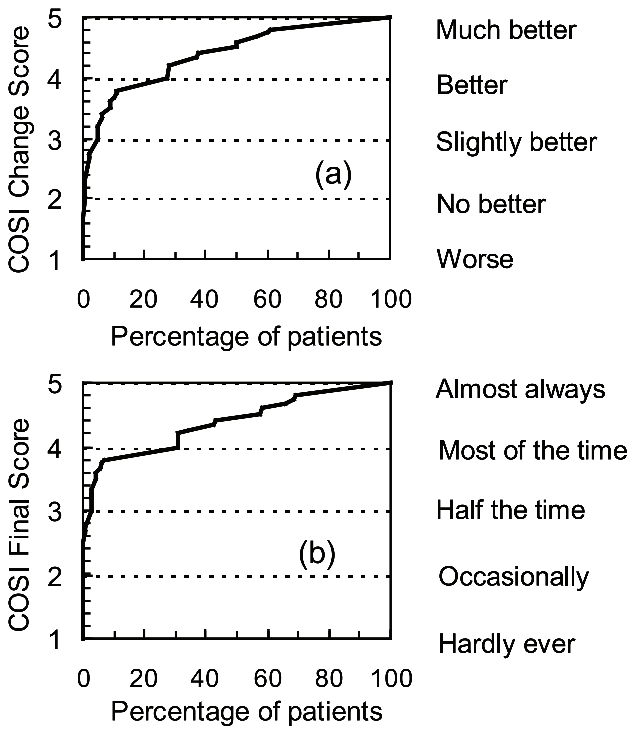


Figure 14. 2 Proportion of patients who obtain less than, or equal to, (a) the COSI change score, and (b) the COSI final score.

Of course, it is possible to ask the patient questions about these things. Dillon, James & Ginis (1997) asked each subject to rank the importance of the listening situations that each subject nominated, but this did not result in any further increase in the validity of the COSI scores.

Gatehouse (1994, 1999) devised the ***Glasgow Hearing Aid Benefit Profile*** (***GHABP***) which measures importance and relevance in a formal way. Patients are asked how often they are in each nominated situation, how difficult each situation is, and how much this restricts their participation. Following rehabilitation they are asked, for each situation, how often they used the hearing aid, how much the hearing aid helps them, how much difficulty they still have, and how satisfied they are with the hearing aid. This obviously provides a lot of information, but quantifying the answers to seven questions for every listening situation makes for a long questionnaire. The seven questions are applied to four standard situations and up to an additional four situations nominated by the patient. This gives a total of up to 56 items to be quantified. If the

NAL COSI
CLIENT ORIENTED SCALE OF IMPROVEMENT

Name : ____________________ **Category.** **New** ______
Audiologist : ____________________ **Return** ______
Date : **1. Needs Established** ______
2. Outcome Assessed ______

Degree of Change

Final Ability (with hearing aid)
Person can hear
10% 25% 50% 75% 95%

SPECIFIC NEEDS

Indicate Order of Significance

Specific Need	Worse	No Difference	Slightly Better	Better	Much Better	CATEGORY	Hardly Ever	Occasionally	Half the Time	Most of Time	Almost Always
☐											
☐											
☐											
☐											
☐											

Categories

1. Conversation with 1 or 2 in quiet
2. Conversation with 1 or 2 in noise
3. Conversation with group in quiet
4. Conversation with group in noise
5. Television/Radio @ normal volume
6. Familiar speaker on phone
7. Unfamiliar speaker on phone
8. Hearing phone ring from another room
9. Hear front door bell or knock
10. Hear traffic
11. Increased social contact
12. Feel embarrassed or stupid
13. Feeling left out
14. Feeling upset or angry
15. Church or meeting
16. Other

Administering COSI

- At the initial interview, identify and write down those specific situations in which the patient would like to hear more clearly. See panel in Section 9.1.6 for more details.
- If some of the listening situations require a different fitting strategy from others, it may be worth finding out what priority or significance the patient places on each of the needs expressed.
- When you believe that the rehabilitation program is completed, read back each of the situations, and for each situation ask the patient (a) how much *more* clearly they now hear in that situation, and (b) how well they can now hear in that situation. The response scales for each of these can be seen in the COSI form reproduced in this section. For indecisive patients, put a check mark on the line midway between the two categories over which they are undecided.
- Evaluate, with the help of the patient, whether the extent of rehabilitation is sufficient for both of you to consider that the program really is completed.
- If you wish to express the results in numerical form, assign 1 point to responses in the left-most column in each section, 2 points for responses in the second column, and so on up to 5 points for the right-most column. Average the number of points across the number of needs listed for that patient. The result will be two scores, each in the range 1 to 5. The first score will describe the benefit of rehabilitation, and the second score will describe the final listening ability of the patient, both averaged across their nominated situations. These can be compared to the normative data shown in Figure 14.2.

If you wish to compare the results for each listening situation to normative data for that situation,[441] categorize each of the needs into one of the 16 standard categories shown at the foot of the COSI form.

patient indicates that the situation is not experienced, the remaining questions about this situation are not asked, thus decreasing the number of items for that patient. The four standard situations are:

- listening to television with other people;
- conversing with one other in quiet;
- conversing in a busy street or shop; and
- conversing with several people in a group.

The GHABP is based on a longer version that had twelve, and later fourteen, standard situations plus the four individually nominated situations.[592] The research leading to the final version of the GHABP showed that including listening situations nominated by individual patients made the questionnaire more sensitive to differences in rehabilitation quality.[593] The GHABP appears in both Table 14.1 and 14.3 because it includes both direct-change and state questions. Humes et al (2009) suggest that GHABP scores (or other questionnaires) can usefully be reported as indicating below average, average, or above average results, rather than a quantitative result, which although more precise, may convey less meaning. Median scores are available in that publication and in Gatehouse (1999).

14.5 Assessing Usage, Problems, and Satisfaction

Use of hearing aids

The chapter so far has concentrated on benefits – a reduction in disability. As intimated earlier, there are other types of outcomes. That a patient wears the hearing aids provided to him or her is an outcome, although usage should be regarded as an important means rather than an end. If hearing aids are being worn we cannot deduce how much benefit they are providing, but if they are not being worn we can be sure they provide absolutely no benefit. Consequently, no use, or much less use than would be expected on the basis of the needs expressed by the patient, are useful indicators that something is wrong.

It is possible, however, for people to report substantial benefit even though they use their hearing aids for only a small amount of time each day or week. Many such people in this category are competent at using their hearing aids, thus supporting their claims that they regularly use and benefit from their aids, even though they may use them less than one hour per day.[1370]

As reviewed in Chapter 9, a significant proportion of people, with estimates ranging from 1% to 29%, completely cease using their hearing aids. A further significant proportion of patients use them so seldom (e.g. less than once per week) that it is hard to imagine that the hearing aids are providing significant benefit.

We can most simply find out how much patients use their hearing aid each day or week by asking them after they have had their hearing aids for a few weeks. We may not get a totally truthful answer. If patients have been well treated, they will not want to disappoint the clinician by saying they rarely wear their hearing aids. Also, they may see themselves as a failure if they have not learned to use their hearing aids regularly, and may not wish to disclose this fact to the clinician. Data logging hearing aids provide an easy way to objectively estimate hearing aid use. Several studies have used these or other objective measures to show that, on average, patients overestimate the amount they use their hearing aids when surveyed soon after the hearing aid fitting.[185, 667, 774, 1127, 1770] The amount of overestimation, however, is not so great as to make self-reported usage of no value, and the objective and self-reported uses are well correlated.

It may minimize exaggeration if patients are first asked to state in which situations they have found their hearing aids to be useful, and in which situations they have found it not worthwhile to wear them. (This also lets them know that stating they do not wear their hearing aids is an acceptable answer.) Following this they can be asked: *On an average day, for how many hours would you wear your hearing aids?* They can be asked to respond with either their estimate of the number of hours, or can be presented with some categories of use, such as:

- >8 hours per day;
- 4 to 8 hours per day;
- 1 to 4 hours per day;
- less than 1 hour per day;
- more than 1 hour per week, but less than 1 hour per day;
- less than 1 hour per week.

These are, in fact, the response choices offered in the ***Hearing Aid User's Questionnaire*** (***HAUQ***),[441, 560] and are sufficiently differentiated to identify patients who are using their hearing aids so little that the clinician will want to ask further questions to find out why. The ***Hearing Aid Review*** questionnaire contains three questions on use, as well as questions on benefit and satisfaction.[189]

Alternatively, for each of a range of situations, patients can be asked what proportion of the time they wear their hearing aids when they are in that situation. This is the approach taken in the GHABP.[593]

Detecting problems with the hearing aids

An absence of problems with the hearing aids (e.g. whistling, earmold discomfort) is obviously desirable. One could argue that a lack of problems caused by the hearing aid hardly qualifies as an outcome of rehabilitation. Avoiding problems is nevertheless an important means to ensuring usage and hence benefit. Not surprisingly, the extent of problems with hearing aids is negatively correlated with the benefits that patients get from hearing aids, the degree to which they use them, and with the satisfaction they express.[441, 949, 772a] Chapter 12 discussed methods for solving typical problems. However, before problems can be solved, they have to be detected. Detection is most likely if the clinician asks patients specifically about each of the typical problems that occur (see panel).

Problems to specifically ask each patient about

- Own voice quality
- Whistling
- Earmold/earshell discomfort
- Aid insertion and removal
- Operation of controls
- Battery changing
- Loudness discomfort
- Inappropriate loudness of speech
- Inappropriate loudness of background sounds
- Sound quality
- Telephone use
- Internal aid noise
- Loudness balance between ears

When the presence of problems is being assessed in person (as opposed to a phone or mail assessment) some of these problem areas can be assessed by observation of the patient instead of by asking.

Questionnaires to assess the presence of problems are reasonably easy to construct. Such questionnaires simply involve asking a series of questions such as: *Do you have any problems getting the hearing aid into your ear?* The precise wording is, however, important if patients are to interpret each question in the way intended by the clinician. Three questionnaires addressing problems with the hearing aid are the HAUQ,[441] the EAR,[1945] and the Hearing Aid Problems Checklist (HAPC).[772a] The patient's ability to use the hearing aid can be scored using the Practical Hearing Aid Skills Test (PHAST).[422]

Satisfaction with hearing aids

It has become very common to measure how satisfied patients are with their hearing aids and it is easy to see why this is important. Satisfaction probably expresses how happy the patients are with their hearing aids, taking into consideration how much (or little) the hearing aids help in different situations, how easy (or hard) they are to use, and the financial and psychological cost of obtaining and wearing hearing aids, in relation to the expectations that patients had about all of these things. Satisfaction has also been found to be related to experience, personality and attitude, usage and type of hearing aids, listening situations in which they are used, sound quality, and problems in hearing aid use.[351, 1928]

Considering the complexity of all these factors, it is surprising that it is sensible to ask a simple question: *How satisfied are you with your hearing aids?* The answers to this question are moderately correlated with much more complicated measures of benefit.[448] The accompanying panel shows two simple one-item questionnaires. The Hearing Aid Review includes a similar ten-point scale and was shown to have reasonable test-retest reliability.[189] A four-point scale is used in the HAUQ questionnaire, although the five-point version shown in the panel may be less affected by ceiling effects. There are, however, two problems with measuring *only* satisfaction as a summary of the effectiveness of the hearing aids.

The first problem is one of relativity: the patient does not know what the best available hearing aids would be like, or even what perfect hearing aids would be like. As Ross & Levitt (1997) have expressed the problem: "*Are you satisfied compared to what?*" Their summary succinctly states why achieving satisfaction should not be the major goal of a clinician: "*Hearing aids are still supposed to help people hear better and not just feel better.*"

Two simple satisfaction questionnaires

The two questionnaires below provide a simple measure of satisfaction. The second, a visual analog scale, may be more sensitive to small differences in satisfaction[448] but answers to the first may be more interpretable.

1. Overall, how satisfied are you with your hearing aid(s)?

    a) Extremely satisfied

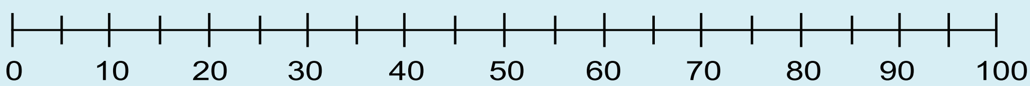
    b) Satisfied

    c) Neither satisfied nor dissatisfied

    d) Dissatisfied

    e) Extremely dissatisfied

2. On a scale of 0 to 100, how satisfied are you, overall, with your hearing aid(s)? A score of 0 means that you are not at all satisfied, and a score of 100 means that you are totally satisfied. Please state any number in between, (or mark the scale at the position) that corresponds to how satisfied you feel.



In studies where different hearing aids are compared sequentially, it is not unusual for subjects to say that a hearing aid is "perfect" and far better than anything they previously experienced, and then to subsequently say that some later hearing aid is even better! By contrast, some patients may have such high expectations of having their hearing restored to normal functioning (or better!) that even the best possible fitting, providing a substantial increase in hearing ability, may not make them very satisfied. A determinant of their satisfaction may be how well the clinician modified their expectations prior to fitting hearing aids.

This limitation does not apply when satisfaction is used in a relative way. If we examine which of two different hearing aid types a patient is more satisfied with, it does not matter at what level the patient sets his or her internal criterion as to what constitutes being satisfied. This is not a viable solution in clinical settings, because we rarely have the luxury of trying several different types of aids or sets of performance characteristics on patients over an extended period.

The second problem is that an overall rating of satisfaction does not tell you anything that is immediately useful! If a patient says that he or she is only slightly satisfied, the answer reveals nothing about the cause of that dissatisfaction. It is, however, worth measuring satisfaction routinely with at least a single item question. Any answer less than *extremely satisfied* (or its equivalent in the questionnaire you use) provides you with the opportunity to ask follow-on questions, the answers to which may reveal some causes of dissatisfaction that you can do something about. The follow-on questions can be open-ended, such as *Which aspects of the hearing aid are you least satisfied with?* and *Which aspects of the hearing aid are you most satisfied with?*

Satisfaction with hearing aids, though improving over the last decade, is far from perfect.[942, 953, 1791] Overall satisfaction is most related to the amount by which the hearing aids improve speech intelligibility in the situations experienced by the patients, but many other types of factors (e.g. clarity, naturalness, sound richness, fit and comfort, dispenser counseling, ease of use, warranty, value for money, match to expectations, reliability, battery life and personality of the patient) are also related to overall satisfaction.[351, 942, 953, 1928]

Cox & Alexander (1999) have provided us with a simple questionnaire to better understand the components of satisfaction. Structured interviews with patients indicated that the things affecting satisfaction could be grouped into six domains:

- cosmetics and self-image;
- sound quality and acoustics;
- benefit;
- comfort and ease of use;
- cost; and
- service.

Cox & Alexander devised 25 questions addressing these categories and, based on the questions' statistical properties (see panel in Section 14.3.1), they selected 15 items to form the ***Satisfaction with Amplification in Daily Life*** scale (***SADL***). Items in the SADL are grouped into four sub-scales:

- ***positive effect*** – comprising decreased communication disability, improved self-confidence, improved sound quality, and an overall assessment of worth;
- ***service and cost*** – comprising reliability, clinician competence, and cost;
- ***negative features*** – comprising reaction to background sounds, feedback, and the hearing aid's usefulness on the telephone; and
- ***personal image*** – comprising appearance and the apparent reaction of others.

The SADL thus provides a systematic way in which the reasons for low overall satisfaction can be discerned, with a view to correcting them. Interestingly, it achieves this without ever using the word satisfaction but it nonetheless correlates well with direct ratings of satisfaction obtained from a much longer questionnaire.[777a] An alternative, which produces results well correlated with the SADL, is a computer based method called the Dynamic Assessment of Hearing Aids (DAHA).[309] This method uses visual analog scales directly marked by the patient on a computer screen.

Of course, just asking the patient why he or she is dissatisfied is another option. However, Cox & Alexander comment that single negative statements by a patient in response to such open-ended questions might conceal some other concerns. When one sub-scale has a much lower score than the other sub-scales, the clinician can be more confident about the reasons for the dissatisfaction.

14.6 The International Outcomes Inventory for Hearing Aids (IOI-HA)

In 1999, an international group of scientists interested in outcomes measurement were locked up in a castle in Eriksholm, Denmark, and told that they would not be released until they came up with a short, simple outcomes measure that removed the need to invent a new questionnaire (and for students to master a new acronym) every time a new research study was undertaken. The scientists incarcerated were those most guilty of burdening the world with a rapidly increasing number of questionnaires.

The result of their deliberations was the ***International Outcomes Inventory for Hearing Aids*** (***IOI-HA***).[335] This questionnaire covers the domains of usage, benefit (activity limitation change), residual activity limitation, satisfaction, residual participation restriction, residual impact on others, and quality of life change. The seven questions, which have been translated into at least 21 languages,[355] are shown in the accompanying panel.

The IOI-HA was kept short (one question per domain) so that excessive length would never be an excuse for not using it. Scientists were asked to include it in research studies to facilitate comparisons of hearing aid effectiveness across studies and countries, even if the research also required additional study-specific questions to be asked.

Subsequent factor analysis of the IOI-HA reveals that the questions fall into two factors, which can be used to form two subscales by averaging the scores across the items in each factor.[346, 456, 725, 972, 1709, 1856] Items 2, 4 and 7 measure the change produced by the hearing aid and load into the first factor, along with item 1 measuring usage. Items 3, 5 and 6 measure residual difficulties experienced, and load into the second factor. It is easy to understand why the items reliably[f] form two

The International Outcomes Inventory for Hearing Aids (IOI-HA)

1. Think about how much you used your present hearing aids over the past two weeks. On an average day, how many hours did you use the hearing aids?

   [*None; < 1 hr/day; 1-4 hr/day; 4-8 hr/day; > 8 hr/day*]

2. Think about the situation where you most wanted to hear better, before you got your present hearing aids. Over the past two weeks, how much have the hearing aids helped in that situation?

   [*Not at all; slightly; moderately; quite a lot; very much better*]

3. Think again about the situation where you most wanted to hear better. When you use your present hearing aids, how much difficulty do you STILL have in that situation?

   [*Very much; quite a lot; moderate; slight; no difficulty*]

4. Considering everything, do you think your present hearing aids are worth the trouble?

   [*Not at all; slightly; moderately; quite a lot; very much worth it*]

5. Over the past two weeks, with your present hearing aids, how much have your hearing difficulties affected the things you can do?

   [*Very much; quite a lot; moderately; slightly; not at all affected*]

6. Over the past two weeks, with your present hearing aids, how much do you think other people were bothered by your hearing difficulties?

   [*Very much; quite a lot, moderately, slightly; not at all bothered*]

7. Considering everything, how much have your present hearing aids changed your enjoyment of life?

   [*Worse; no change; slightly better, quite a lot better; very much better*]

---

f In the Danish translation, item 5 does not correlate as expected with the other items and may give unreliable results.[1856]

factors when the IOI is applied to a sample of people with a range of hearing losses and self-reported difficulties. On average, patients with greater loss and difficulties use their hearing aids more,[1177] and find hearing aids more helpful than those with less loss and difficulties. However, even with their hearing aids, they experience greater residual difficulties than those with less loss. Consequently, subscale-1 scores increase with hearing loss, but subscale-2 scores decrease with hearing loss.[347, 456]

- Subscale 1 scores increase by 0.1 scale points for every 10 dB of hearing loss in the better ear.
- Subscale 2 scores decrease by 0.07 scale points for every 10 dB of hearing loss in the better ear.

Subscales 1 and 2 could loosely be thought of as measuring ***Overall benefit*** and ***Freedom from residual difficulties*** respectively.

The IOI-HA has now been used in many studies investigating hearing aid effectiveness. Figure 14.3 shows experimental results from several countries. Each question is scaled on a 1 to 5 scale, with 5 indicating the best possible hearing aid outcomes. Any comparisons between studies or countries must take into account differences between the populations on which the data were measured. Differences that affect results include hearing loss, age, response rate,[g] whether the patients are first-time or experienced hearing aid wearers,[439] and probably whether the questions are administered by the clinician who looked after the patient or by a third party. In Figure 14.3, all questionnaires were mailed out to patients, except for those of the highest scoring study (Stephens, 2002) where the survey was administered in person by the clinician who had looked after the patient. The second-highest scoring study (Heuemann et al 2005) had the lowest response rate of all the studies.

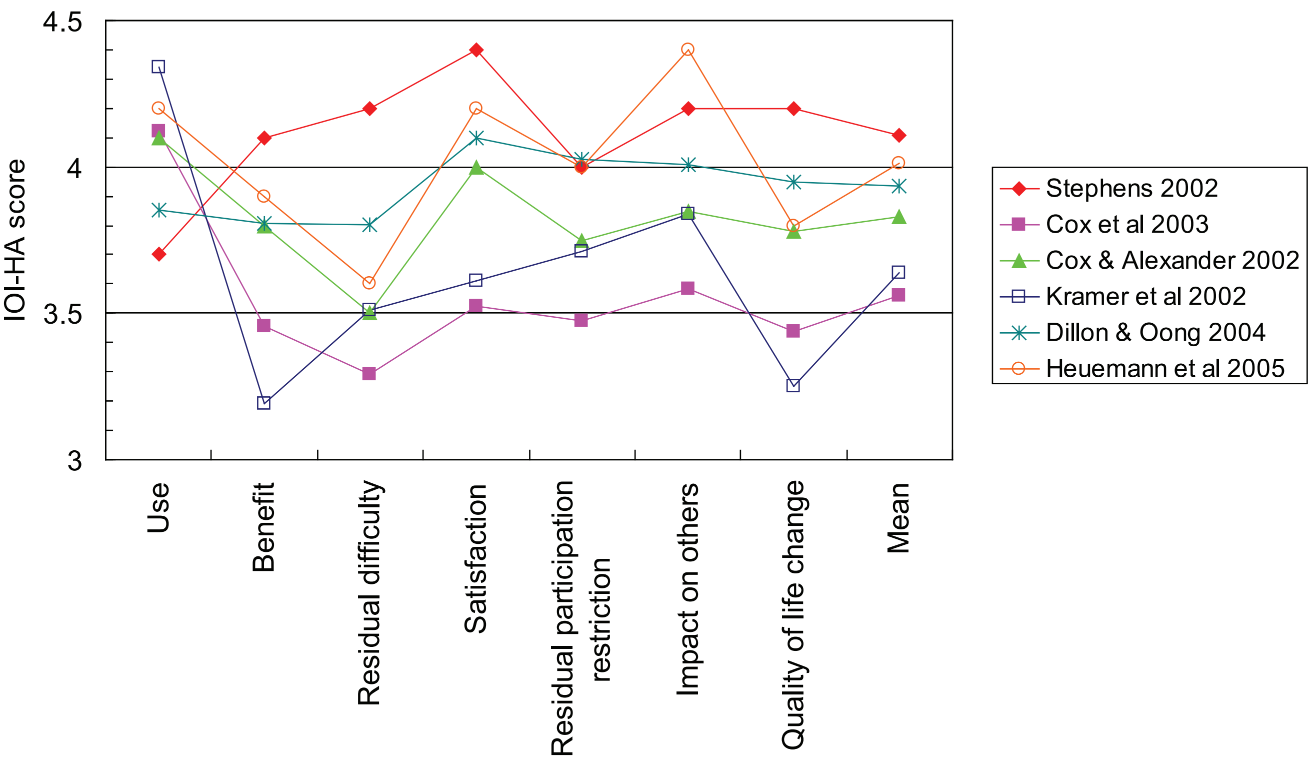


Figure 14.3. Data from several countries and studies on the IOI-HA. Responses for all questions are scaled from 1 to 5, with 5 representing the best outcomes.

[g] The proportion who use their hearing aids regularly (measured in item 1) decreases as the response rate to the survey increases. (That is, those who don't readily respond to surveys contain a much greater proportion of patients who have ceased wearing their hearing aids.) In one mail survey of 672 patients with a response rate of 47%, both responders and non-responders were contacted by phone to ascertain if hearing aids were being used. Of the responders, 6% said they were not using their hearing aids at all, but of the non-responders, 16% said they were not using them at all.[439] Similarly, a Swiss study of 14,285 patients indicated that hearing aids were never used by only 1% of responders but by 6% of non-responders.[122]

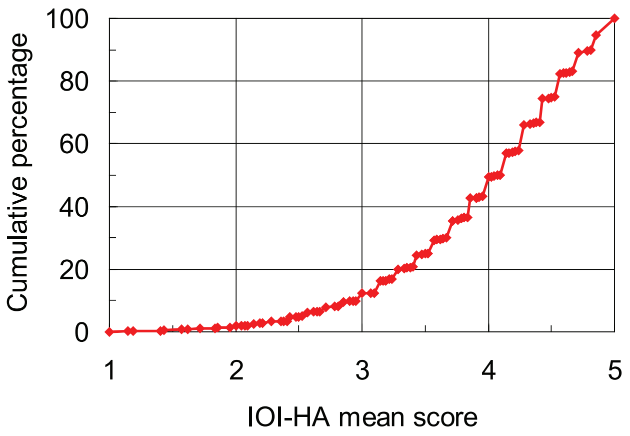


Figure 14.4 Percentage of people achieving less than or equal to the mean IOI-HA scores shown on the horizontal axis.

Figure 14.4 shows the distribution of mean IOI-HA scores observed in a sample of 2,379 patients wearing predominantly multi-channel compression hearing aids of various styles.[456] As the median score of 4.09 is some way below the ceiling score of 5.0 and well above the floor score of 1.0, the IOI-HA is suitable for detecting people with much better and much worse than typical outcomes.

The IOI-HA did not prevent the creation of more questionnaires. In fact, it has produced offspring![1329] The IOI-HA-SO measures the effect of hearing aids on significant others, and the IOI-AI measures the effect of alternative interventions, such as ALDs, hearing strategies, or surgery.

14.7 Changes in Outcomes with Time after Fitting

When should outcomes be assessed? The answer may *look* simple: at the end of the last appointment, but even if this is adopted, the timing of this last appointment relative to the fitting is still largely determined by the clinician. On the one hand, we want the outcomes evaluation to be soon after fitting, so that if the evaluation reveals that significant problems remain, these can be addressed as soon as possible. On the other hand, some problems might not emerge until the patients have become sufficiently experienced with their hearing aids. Similarly, patients may not appreciate the full advantages their hearing aids provide until they have become sufficiently experienced with their use.

Experimentally, all types of outcomes scores appear to change during at least the first few weeks after aid fitting. As long ago as 1939, Berry referred to increases in speech identification during the months after fitting as a ***process of adjustment***. Watson & Knudsen (1940) reported an increase in speech identification of 40 percentage points over the three months following aid fitting for one subject.[h] They referred to this process as ***accommodation***. The changes in speech identification ability that occur during the first few months following fitting are now referred to as ***acclimatization***.[i, 591, 766]

An extensive survey of experimental results indicated, however, that the increase averages only a few percentage points, and is therefore usually too small to be significant with individual patients.[1803] It seems likely that the amount of increase in speech intelligibility, and the time over which this occurs, is greater for unfamiliar forms of signal processing (e.g. multichannel compression, or frequency lowering) than for linear amplification,[1950] although the acclimatization period may still be as short as one month.[1009]

Self-reported outcomes also vary in the weeks and months after hearing aid fitting. Consistent with improved objectively-measured speech intelligibility, self-reported benefit, measured using APHAB scores, increases from two weeks post-fitting to three months post-fitting.[343]

It appears that there may be a halo (or honeymoon) effect on some outcomes measures a few weeks after fitting. There is higher satisfaction two weeks after fitting than twelve months after fitting.[777, 1171] The negative emotional aspects of disability (based on HHIE scores) are less marked three weeks after fitting than three months after fitting.[1129, 1771] Noble (1999) hypothesizes that shortly after fitting, tensions at home resulting from poor communication ability have been dissipated by the patient seeking rehabilita-

[h] This subject had a long-standing severe hearing loss, and was receiving a hearing aid for the first time, an unusual combination of circumstances these days in developed countries. It is likely that the magnitude of the acclimatization effect increases with the magnitude of the change in audibility that occurs when a hearing aid is worn.

[i] The word *acclimatization* is sometimes also applied to the increase in gain preferred by patients as they become accustomed to their hearing aids. This phenomenon, already discussed in Section 10.3.1, is referred to in this book as ***adaptation***.

tion and by the beneficial effects of the hearing aids on communication at home. As more time passes, the patient becomes aware of situations in which the hearing aids provide minimal benefit, so self-reported disability increases, although not to the level reported prior to rehabilitation, and satisfaction decreases.[1171] For patients who use their hearing aids more than 4 hrs per day, changes in outcome relative to 1 week after fitting are more likely to be positive, whereas for those who use their hearing aids less than 4 hrs per day, changes in outcomes with time are more likely to be negative.[1856]

When experienced hearing aid wearers are fitted with improved amplification characteristics, one might also expect a honeymoon period fuelled by expectations of better devices. Follow-up 1 year later, however, shows that benefits of the new hearing aids relative to the old hearing aids measured 4 weeks after fitting are sustained.[62]

A comparison of several studies examining changes in outcomes over time indicates that they settle down by about 6 weeks after fitting.

- Satisfaction, handicap reduction (HHIE scores), attainment of goals (GAS scores), and change in listening ability (COSI scores) all appear to be no different six weeks after fitting than they are three months after fitting.[448, 452]
- Handicap reduction (HHIE scores) and communication function (Denver Scale of Communication scores) are both little different six weeks after fitting than four months after fitting.[1273]
- Satisfaction, usage and aided speech understanding are little different 4 weeks after fitting than 6 months, 1 year, 2 years and 3 years after fitting. In the few cases where significant individual changes occur, they mostly show reduced satisfaction or benefit over the more extended time.[776a, 777] Individual patients with below- or above-average usage or satisfaction 4 weeks after fitting are likely to still be below- or above-average, respectively, 2 years after fitting.[777]
- Handicap reduction measured at three months is unchanged when it is measured at various later times up to one year after fitting.[1129, 1275, 1771]
- Similarly, both objectively measured benefit (speech identification ability) and subjectively measured benefit (PHAB scores) appear to be the same six weeks after fitting as they are one year after fitting.[1748]
- There is conflicting evidence regarding changes in hearing aid usage during the first year. Mulrow, Tuley & Aguilar (1992b) found a decrease in the number of hours per day that hearing aids were worn whereas Brooks (1981) found an increase.
- Various changes are possible over periods of several years. Usage may either remain stable four years after fitting,[722] or may decrease slightly, probably caused by patients making more refined decisions about the situations in which hearing aids are helpful.[1018] Some patients who initially make little use of their hearing aids in the first year can become more regular users 10 years later, presumably as their hearing deteriorates.[182]

Conclusion: changes in outcomes with time

Taken together, these studies imply that self-reported use and benefit is reasonably stable by six weeks after fitting, although small changes may occur for many months or years thereafter. The first few weeks following fitting does therefore not seem to be the best time for evaluation, although for practical reasons, it is often the most convenient time.

The time at which measurements stabilize depends not just on the measure but also on the personality of the patients. Cox (2003) showed that benefit decreased slightly from 3 weeks to 3 months for high-neuroticism patients, but did not change (and was much lower) for low-neuroticism patients. Satisfaction, by contrast, was equal and unchanging for both groups of patients.

In any situation where payment for a device and services is linked to demonstrating a favorable outcome, there would be great advantage in the patient having a 45- or 60-day trial rather than the commonly offered 30-day trial. Although most patients make up their minds about whether to continue with hearing aid use within two weeks, a significant minority has still not made a decision eight weeks after fitting.[1811] A period longer than 30 days is consistent with the time needed for benefits to stabilize. There is, however, no reason any single time has to be chosen for outcomes measurement:

- One measure (e.g. APHAB or COSI) focusing on benefit can be administered at the end of the regular appointments, which may well be within the 30-day period during which patients in some countries can return their hearing aids at no cost to themselves.
- Another measure (e.g. HAUQ or EAR) can be administered 6 to 12 weeks after fitting to ensure that any problems subsequently found are discovered and dealt with. The HAUQ and EAR also provide simple measures of benefit, and satisfaction, and the HAUQ also measures use. The Hearing Aid Review (HAR)[189] also measures benefit, satisfaction and use, but does not assess problems with the hearing aid. The last check can be performed by mail or by telephone to minimize the cost of administering it, and both HAUQ and HAR have been designed with this application in mind.

14.8 Impact of Hearing Loss and Hearing Aids on Health-Related Quality of Life

The effect of hearing aids can extend much further than just allowing people to better understand speech and identify other sounds, although these are the primary mechanisms that enable wider benefits to occur. These wider benefits of hearing aids can occur only because hearing loss causes, or is at least associated with, a decreased ***health-related quality of life***, as reviewed in Section 14.8.1 below. Other terms that may be used instead of health-related quality of life include ***subjective well being***, ***health state***, ***health status***, and ***well being***.[11] The effects that hearing aids may have on improving health-related quality of life are reviewed in Section 14.8.2.

14.8.1 Effect of hearing loss on health-related quality of life

Fortunately for people everywhere, no scientist has yet convinced an institutional ethics committee that it is worth doing a randomized controlled trial to see if giving people a hearing loss causes other health problems and/or decreased quality of life. We are therefore left with many observational studies that leave us with no doubt that hearing loss *is associated with* a wide range of adverse health issues. We then have to infer, on the grounds of what seems most likely given any statistical controls in the experiment, whether hearing loss has caused the adverse health issues, the adverse health issues have caused the hearing loss, some uncontrolled factor (e.g. age, if not corrected for, or poor cardio-vascular functioning) has caused both hearing loss and the adverse health issues measured, or some combination of these three causations are responsible.

Untreated hearing loss has been statistically associated with:

- depression;[56, 247, 1791] the prevalence of depressive symptoms (but not the prevalence of major depression) is further increased in those who have combined hearing and vision loss;[1088]
- increased social isolation, including decreased quantity and quality of interaction with others, psychological withdrawal, and increased loneliness;[56, 386, 1904] the isolation increases for those who also have significant vision loss;[802]
- decreased self-sufficiency;[56, 247]
- higher mortality rate;[56, 737]
- decreased cognitive function (even after allowing for age),[60 644, 1066b, 1812] including a greater likelihood of subsequently acquiring Alzheimer's disease for those with hearing loss than for those without hearing loss;[1066c]
- a general decrease in physical and psychosocial well-being, as measured by the ***Sickness Impact Profile*** (***SIP***);[127] and
- reduced access to other health services, which could lead to further health problems not directly related to hearing loss (investigated so far primarily for people with severe or profound hearing loss);[1810]

The decreased communication caused by hearing loss has also been asserted to lead to anger, anxiety, diminished safety, distress, embarrassment, exhaustion, fear of losing more hearing, insecurity, irritation, loneliness, paranoia, loss of group affiliation, loss of intimacy, restricted travel, sadness and unemployment.[635, 723, 1720, 1799]

14.8.2 Effect of hearing aids on health-related quality of life

The effect of treatment with hearing aids on health-related quality of life is of great interest in itself. It also provides a way to show which of the health issues listed in the previous section are *caused* by hearing loss. If people with hearing loss are randomly assigned either to a treatment group (who receive hearing aids and associated rehabilitative services including hearing tactics) or to a control group, then any differences in health-related quality of life between the two groups are very likely the result of the treatment partly reversing the original negative effects of the hearing loss.

Unfortunately, there are very few such randomized controlled trials of treatment with hearing aids. Mulrow et al (1990) showed that treatment with hearing aids decreased depression and improved social, emotional, communicative and cognitive function. These benefits were established when measured four months after hearing aid fitting and were still present when re-measured 12 months after fitting.[1273, 1275] Another randomized controlled trial, however, found that although hearing aids reduced disability, there was no reduction in depression when measured 6 months after hearing aid fitting.[1790]

An alternative to a randomized controlled trial that can also show a causative effect is a longitudinal study in which health-related quality of life of a group of people is measured before and after treatment with hearing aids. Crandell (1998) showed that hearing aid fitting led to improved general well-being as measured by the SIP. These improvements, measured 3 months after fitting, were sustained 6 months after fitting. Other indirect benefits that have been shown in longitudinal studies include improved working memory, sensory and social pleasure, social interaction, alertness, leisure activity, learning ability, and psychosocial well being. Hearing aid use also led to reduced anxiety, depression, and paranoia.[500, 837, 1036, 1494, 1693]

A weaker form of evidence for the effect of hearing aids on health-related quality of life is a quasi-longitudinal study in which hearing-impaired people, and/or members of their families, are asked to say, retrospectively, how the hearing-impaired person's life has changed as a consequence of receiving hearing aids. A large proportion of those wearing hearing aids, and an even larger proportion of their family members, report relationships, emotional feelings, participation, independence and life overall as being positively affected by wearing hearing aids.[956]

Finally, the weakest form of evidence for the beneficial effect of hearing aids are cross-sectional studies in which some health-related aspect of hearing aid wearers is compared to the same aspect for hearing-impaired people who do not wear hearing aids. Many such studies have shown that, on numerous health-related quality of life measures, hearing-impaired people with hearing aids fare better than hearing-impaired people without hearing aids. Those who use hearing aids report better mood, less depression, greater participation in social activities, warmer interpersonal relationships, greater self-sufficiency, greater satisfaction with life, more positive self-image, less self-perceived discrimination, greater emotional stability, greater control of their lives, better health, less anger/frustration, less paranoia, less anxiety, less self-criticism, less introversion (as assessed by others) and a greatly decreased mortality rate.[56, 172, 689, 956] Hearing aid wearers also have better cognitive functioning.[1066a]

Of course, significant correlations do not prove that hearing rehabilitation *causes* the improved health outcomes. It is also possible that better-adjusted and healthier people are more likely to seek rehabilitation. In particular, people with a major health issue who also have a hearing loss may well decide that they need to focus on the major health issue rather than attending to the hearing loss. The inevitable inclusion of such people in the non-hearing aid group will obviously lower the average health-related quality of life scores for those without hearing aids, but hearing loss may have had little effect on their scores.

Health policy makers have a great need to measure the effect of treatment by hearing aids on health-related quality of life in a way that enables their beneficial effects to be compared to those of treatments for other illnesses, diseases or injuries. Their effectiveness relative to the total cost of providing them enables rational decisions to be made about expenditure of public money on hearing aid treatments.

Numerous groups have attempted to devise generic measures that can quantify and compare the adverse effects of different health conditions and the effectiveness of different treatments. Generic measures other than the SIP already mentioned include:

- The ***Medical Outcomes Study Short Form 36*** (***SF36***) is a widely used measure assessing eight dimensions: physical functioning, impact of physical dysfunction, pain, general health, vitality, social functioning, emotional impact, and mental health.

- ***World Health Organization Disability Assessment Scale II*** (***WHO-DAS II***) measures health-related quality of life along six dimensions (communication, mobility, self-care, interpersonal, life activities, and participation in society). Of the six dimensions, the communication dimension is most affected by treatment with hearing aids.[1161]
- ***Health Utilities Index Mark 3*** (***HUI 3***) measures the domains of vision, hearing, speech, dexterity, ambulation, emotion, cognition and pain.
- The ***EuroQol*** uses the EuroQol five-dimension (EQ-5D) questionnaire to measure mobility, self-care, ability to undertake usual activities, pain/discomfort and anxiety/depression. The EuroQol methodology also includes a single ***visual analog scale*** on which the patients marks, on a 0 to 100 scale, how they feel.

Abrams et al (2005) give a more detailed review of these measures, and how they relate to hearing. Although the different generic measures all aim to give a ***utility value*** that describes what proportion of perfect health a person with a specific health issue has, and what proportion of perfect health a specific treatment restores, there is no single definition of health-related quality of life against which a questionnaire can be validated. Consequently, the different measures put different emphases on different types of impairments and on impairment versus activity limitation versus participation restriction.

The items in the most widely used measure, the SF36 and in a shortened version of it, the SF-6D, for example, appear to be more inspired by physical health issues than by communication disorders. Hearing loss thus appears less of a problem (i.e. utility is less affected) when measured with the SF questionnaires, or with the EQ-5D, than when measured with the HUI 3 which gives greater emphasis to communication.[85] Consequently we would also expect treatment with hearing aids to appear less valuable when measured by the SF or EQ-5D measures. Several studies have shown that hearing aids have little or no effect on overall health-related quality of life measured with the SF or EQ-5D questionnaires, despite having strongly beneficial effects when measured with hearing-specific measures of disability, such as the HIEE.[297, 755, 838, 1693, 1868]

In short, the relative rankings of different diseases or treatments depend on the measure chosen. The HUI 3 and WHO-DAS II measures, although still generic, appear to be more sensitive than the SF and EQ-5D measures to hearing loss and its treatment.[1161]

Another way to compare the value to patients of different types of treatments is to ask how much a patient would be prepared to pay to fix a health problem that they have or imagine having. The question can be asked before they have had a treatment, or, if they actually have the health issue, after they have received the treatment. Veterans asked after hearing aid fitting how much they would be willing to pay for their hearing aids (which were actually free to them) indicated an average amount of US$982 per hearing aid.[295] Across individuals, the amount nominated correlated with benefit measured with the APHAB questionnaire.

Despite the connection between hearing rehabilitation and general mental and physical health, so many other things impact on general health and well-being that it is not sensible for clinicians to assess the outcomes of hearing rehabilitation for an individual patient by measuring health-related quality of life.[124, 297, 1273] Generic measures intentionally take into account many aspects of a person's life, so they are too insensitive to reliably show the changes that hearing aids provide to individual patients. That is, hearing-specific quality of life measures give a much larger effect size.[297] When we attempt to describe the outcomes of hearing rehabilitation for a population, however, it would be wise not to overlook the beneficial effects of hearing aids on general health and well being. It would also be appropriate to mention to prospective hearing aid wearers that the benefits of hearing aids extend far beyond simply hearing better.[956]

There is a need for randomized controlled trials or longitudinal studies that enable us to better identify and quantify all the effects of hearing loss, and treatment with hearing aids, on general well-being. In particular, research must be done that enables the utility reduction caused by hearing loss to be validly compared to the utility reduction caused by other health issues, so that treatment effectiveness can properly be compared, and hearing rehabilitation funded commensurately with its importance to people. Even with the insensitivity of the EQ-5D questionnaire to hearing, treatment with hearing aids appears to offer a cost per QALY (see panel) that is competitive with the cost-benefit ratio of treatments for other health issues.[839]

Some terms used in the measurement of health-related quality of life

Utility describes what proportion of perfect health and well being a person has, with 1 representing perfect health and well being, and 0 representing death. Any health condition reduces utility by a certain amount. Any treatment (hopefully) increases utility by some amount. According to the HUI 3 system, a person with profound hearing loss, but otherwise in perfect health, would have a utility of 0.61.[11]

Quality Adjusted Life Years (***QALYs***) refer to the equivalent number of years of perfect health that a treatment offers to a person. For example, a treatment that increases utility by 0.2, applied to a person who is (statistically) expected to live for a further 10 years, provides the person with 2 QUALYs.

Time tradeoff, ***standard gamble*** and ***visual analog scales***, are the techniques used to calculate the utility decrement of health conditions or the utility added by treatments. For further details see Abrams, Chisolm and McArdle (2005)

14.9 Concluding Comments

Should clinicians routinely measure outcomes? If so, which measurement tools should a clinician routinely use? If a third party (e.g. an insurer) demands proof of benefit, then the clinician must use some formal measurement tool like the ones described in this chapter.

What if there is no external demand for a measurement of outcomes? "Outcomes" are multi-faceted, and there are good reasons why a clinician should find out if patients are using their hearing aids, are deriving benefit from them, are satisfied, and have any remaining problems that can be dealt with. These things *can* all be done to some extent without using any questionnaire or formal procedure. The use of a systematic procedure, however, makes it more likely that a clinician *will* check up these things, and will likely make the assessment more precise. Use of a defined measure also enables clinics to compare results across time, for individual patients, for the clinic as a whole, or against other clinics or health systems using the same measure.

Furthermore, the use of a formal procedure (whether it be APHAB, COSI, EAR, GHABP, HAUQ, HHIE, SADL, SHAPIE, SSQ or any of the other acronyms) may well teach or remind clinicians about the type of questions they should be asking. Clinicians may then continue to ask these questions even when they are not using a formal procedure. Each of the following questionnaires offers a perspective different from the others, but in many cases similar to other questionnaires:

- APHAB teaches that hearing aids will be more effective in some situations than in others, and may have adverse effects on intense sounds.
- COSI teaches about the importance of identifying and solving the specific hearing problems that caused the patient to seek help in the first place.
- GHABP gives a well-rounded picture of the disability, benefit, use, and satisfaction that patients experience in a range of situations.
- HAUQ teaches the importance of following up on whether all mechanical and electroacoustic aspects of the fitting have been adequately dealt with, and provides simple, single item measures of usage and satisfaction.
- HHIE shows us how much hearing aids and other rehabilitation have affected the lifestyle and emotions of the patient.
- SADL enables us to analyze the reasons why a patient may not be fully satisfied.
- SHAPIE is a relatively sensitive tool for examining the degree of disability reduction, but is probably more suitable for research studies or comparative evaluations than for routine clinical use.
- SSQ is sensitive to the benefit (localizing, understanding) of listening with two hearing aids instead of one.

In summary, whether we use a formal or an informal procedure, it is worth finding out *something* about each of:

- benefit, measured as change in disability (e.g. APHAB, COSI, EAR, GHABP, HAUQ, HHIE, or SHAPIE);
- aided performance (e.g. APHAB, COSI, EAR, GHABP, HHIE, or speech recognition tests);

- usage;
- problems with the hearing aids (e.g. EAR, HAUQ, HAPC);
- remaining difficulties (e.g. APHAB, COSI, GHABP, HHIE); and
- satisfaction (an overall rating, and if a low rating is obtained, following up with either informal questioning or the SADL).

An open-set format, where patients nominate the areas of difficulty they are experiencing, *and* a standardized format, where patients are asked about participation restriction in pre-specified areas, appear to provide complementary information.[1710]

The IOI-HA covers each of these domains except problems with the hearing aids, and additionally covers quality of life, residual effect on others, and quality of life change. Numerous studies have shown that measures in different domains are significantly, but imperfectly, correlated with each other.[772] This includes the least formal of the measures, the COSI, which correlates moderately with the IOI-HA.[1709] Imperfect correlation is, of course, partly caused by random measurement error in both measures, but a high rating on one type of outcome in conjunction with a low rating on another alerts the clinician that there may be further work to do on either the fitting, instructions to the patient, or discussion with the patient about his or her expectations relative to the performance achievable with available technology.[189]

The possibility of small changes in outcomes occurring up to around 6 weeks after fitting should not dissuade clinicians from measuring them earlier if that is the only viable option. Correlations between measures made soon or long after fitting are sufficiently high that we are unlikely to mistake broadly successful outcomes for broadly unsuccessful outcome, or vice versa, just by measuring them too early. Similarly, while the method of administration (paper versus phone versus face-to-face interview by clinician or other person) probably has statistically significant effects when applied to sufficiently large groups of patients,[1339, 1903] the effects are sufficiently small that clinicians should not be concerned about which method they use.

The fact that personality affects self-reported outcomes, summarized in Section 9.1.10, should not concern clinicians. It seems an unanswerable question as to whether personality affects the benefits that hearing aids deliver, or affects the way people fill in questionnaires. Both are likely, and it seems totally inappropriate to ignore what people tell us, just because their views are affected by their personalities. Conversations would be short if we applied such a philosophy generally in life.

Outcomes measures keep us grounded as to what we are, and are not, really achieving, from the perspective of the client. This perspective can easily be overlooked when each technological innovation is accompanied by so much hype about seemingly world-changing technology.[1386] It is worth remembering that simply labeling a hearing aid as advanced technology can cause slightly better outcomes to be reported.[108] It is difficult to decide if this finding more illustrates the sensitivity of self-report measures to a real placebo effect, or demonstrates the need to critically ask whether every self-report measure has been influenced by events or beliefs other than those that we think we are measuring.

CHAPTER 15

BINAURAL AND BILATERAL CONSIDERATIONS IN HEARING AID FITTING

Synopsis

Sensing sounds in two ears (binaural hearing) makes it possible for a person to locate the source of sounds and increases speech intelligibility in noisy situations. Wearing two hearing aids (a bilateral fitting) instead of one hearing aid (a unilateral fitting) increases the range of sound levels for which binaural hearing is possible. Bilateral fitting is thus more important when hearing loss is severe than when it is mild or moderate.

Accurate horizontal localization is possible because sounds reaching the two ears differ in level and in arrival time, and hence in phase. These cues are also present, but altered, when people wear hearing aids. Most hearing-impaired people, once they become used to the effect of their hearing aids on these cues, can localize sounds accurately to the left and right in the horizontal plane. Vertical localization and front-back localization, which are based on very high-frequency cues created by the pinna, are extremely adversely affected by hearing loss and are not significantly improved by hearing aids.

When speech and noise arrive from different directions, head diffraction causes the signal-to-noise ratio to be greater at one ear than at the other. Further, the auditory system can combine the different mixtures of speech and noise arriving at each ear to effectively remove some of the noise. This ability is known as binaural squelch. Even presenting identical sounds to the two ears provides a small improvement in speech intelligibility over listening with one ear, a phenomenon known as binaural redundancy.

Wearing a second hearing aid will improve speech intelligibility in noise whenever it causes speech to become audible in the previously unaided ear. Achieving audibility of speech in both ears is a pre-requisite to attending to the ear with the better signal-to-noise ratio, and to benefitting from binaural squelch and binaural redundancy. Bilateral fitting of hearing aids has several other advantages. These include improved sound quality, suppression of tinnitus in both ears, and greater convenience if one hearing aid breaks down or when one battery dies. A bilateral fitting may help prevent a problem sometimes associated with unilateral fittings: a unilateral fitting can lead to decreased speech processing ability in the unaided ear if this ear is deprived of auditory stimulation for too long, a phenomenon referred to as late-onset auditory deprivation.

The advantages of bilateral fittings also apply to patients with asymmetrical hearing thresholds. If such patients must receive a unilateral fitting, it may be generally advisable to fit the ear with thresholds closest to about 60 dB HL.

Bilateral fittings also have disadvantages: they cost more, are more susceptible to wind noise, and are more difficult for some elderly people to manage. Also, some people regard two hearing aids as an indication of severe hearing loss, and do not wish to be perceived in this way. For some people, binaural interference causes speech identification ability to be better when unilaterally aided than when bilaterally aided. The causes of this interference may lie in differences between the two cochleae, differences between the two hemispheres of the cortex, or distortions in transfer of information from one hemisphere of the cortex to the other.

Because of the variability associated with speech intelligibility testing, conditions have to be chosen carefully to reliably demonstrate bilateral advantage or detect binaural interference on an individual patient. To best demonstrate bilateral advantage, loudspeaker positions for speech and noise should be chosen to maximize the effects of head diffraction and binaural squelch. To best detect binaural interference, speech and noise should emanate from a single, frontal loudspeaker, so that the effects of head diffraction and binaural squelch are minimized. In either case, speech tests with steep performance-intensity functions should be used. A method for predicting whether individual patients will benefit more from a unilateral or bilateral fitting is urgently needed.

There are many advantages of listening with two ears instead of one. Using two ears enables a person to understand more when speech is heard in background noise or when there is reverberation. The ability to localize sounds is also highly dependent on being able to perceive sounds simultaneously in both ears. Loss of hearing in one ear will therefore leave a person with a considerable hearing deficit in many listening situations.[314] Similarly, when a person has a moderate or severe loss in both ears but wears a hearing aid in only one ear, a considerable deficit remains. A normal-hearing person can gain some appreciation of this by blocking one ear while trying to listen in a noisy and/or reverberant environment. Noises that were previously not noticed suddenly become apparent, if not prominent. It becomes very difficult to listen to, and understand, any target signal in the midst of all this noise. Sounds will also be very difficult to localize.

Despite the difficulties that arise from listening with only one ear, the question of how many hearing aids a person should have is not as simple as just fitting everybody with two hearing aids.

The purpose of this chapter is to outline the factors underlying the following decisions that the clinician has to make for every patient:

- Should one or two hearing aids be recommended?
- If the patient disagrees with the recommendation, how important is it to attempt to convince the patient to decide otherwise?
- If one hearing aid is provided, in which ear should it be worn?

A hearing aid in each ear used to be called a ***binaural fitting***, as contrasted to a ***monaural fitting***. This book will follow the terminology suggested by Noble & Byrne (1991) and refer to these conditions as a ***bilateral fitting*** and a ***unilateral fitting*** respectively. Appropriate terminology can help us appreciate the real situation:

- A person with a hearing aid in one ear is still able to hear many sounds in both ears (at least for people with mild and moderate losses) and thus hears many sounds binaurally.
- A person with hearing aids in both ears may not hear some sounds in one ear, and it is possible that sounds heard in one ear will interfere with sounds heard in the other.

These simple examples show that a unilateral fitting does not always imply monaural hearing, and a bilateral fitting does not necessarily imply that sounds in both ears will contribute positively towards perception. Consistent with these definitions, ***binaural advantage*** will be used in this chapter to mean the advantage of listening with two ears instead of one. ***Bilateral advantage*** will refer to the advantage of listening through two hearing aids rather than one.

The fitting of two hearing aids is now much more common than the fitting of one. In the USA in 2004, 82% of fittings were bilateral.[a, 61, 954] A decade earlier the figure was 65% and a decade before that it was only 25%.[946] Estimated bilateral fitting rates in 2004 varied from 10 to 75% across countries.[61]

This huge variation in bilateral fitting proportion across time and place reinforces that the proportion at any one time, in any one place, is simply a count of what *is* being done, not a prescription about what any clinician *should* be doing.

Some definitions

Binaural stimulation: Sounds are presented to (or perceived in) both ears.

Monaural stimulation: Sounds are presented to (or perceived in) one ear.

Bilateral fitting: Hearing aids are worn in both ears.

Unilateral fitting: A hearing aid is worn in one ear.

Diotic: Identical sounds are presented to both ears.

Dichotic: A different sound is presented to each ear.

[a] Arlinger (2006) reports the figure as 82% of *all* fittings, whereas Kochkin (2005) reports the figure as 86% of fittings to people with bilateral hearing loss.

Not everyone who receives two hearing aids wears them. In one survey of over 4000 patients, 48% of patients had been fitted bilaterally. Most of these (94%) reported, three months after the fitting, that they used hearing aids regularly. Of the bilaterally fitted, regular hearing aid users, however, 20% reported that they only used one hearing aid.[441]

There is overwhelming evidence that two hearing aids provide better performance than one for most people in some situations, and this evidence has been available for three decades. For comprehensive reviews of older literature see Byrne (1980, 1981) and Ross (1980) and for more recent self-report studies see Noble (2006). Nonetheless, bilateral hearing aids are not the best choice for all hearing-impaired persons, particularly when one takes cost, self-image, listening needs, aid management ability, and the possibility of binaural interference into account. Every clinician must therefore have a clear understanding of the benefits and limitations of two hearing aids relative to one.

The first two sections of this chapter will review the advantages and mechanisms of listening with two ears instead of one, for both normal-hearing and hearing-impaired people. We will then apply these concepts to clinical decisions about fitting one versus two hearing aids, and to testing bilateral advantage.

15.1 Binaural Effects in Localization

15.1.1 Localization cues in normal hearing

Localization of sounds can conveniently be discussed under the headings: horizontal localization, vertical localization, front-back differentiation, externalization, and distance perception. The last of these has not been as intensely studied as the others. Distance perception depends on perception of reverberation, echoes, overall level, and overall spectral shape.[1953a] The different dimensions of localization are enabled by two broad sets of cues: binaural cues that rely on differences between the two ears, and monaural cues that are present at either ear.

Horizontal (left-right) localization is made possible by differences in time and intensity between the two ears. As is evident from Figure 15.1, sounds will arrive at the ear closer to the source (the ***near ear***) before they arrive at the ear further away from the source (the ***far ear***). The resulting difference in arrival time at the two ears is called the ***interaural time difference***. It depends on the size of the head and the speed of sound. The interaural time difference is zero for frontally incident sound and increases to a maximum of about 0.7 ms for sounds coming from 90° with respect to the front, as shown in Figure 15.2.[b] Because any time delay leads to a phase delay, an interaural time difference results in an ***interaural phase difference***.

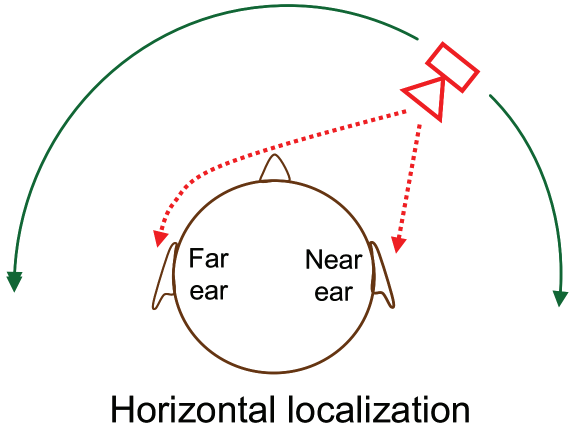


Figure 15.1 Variation of the source direction in the horizontal plane.

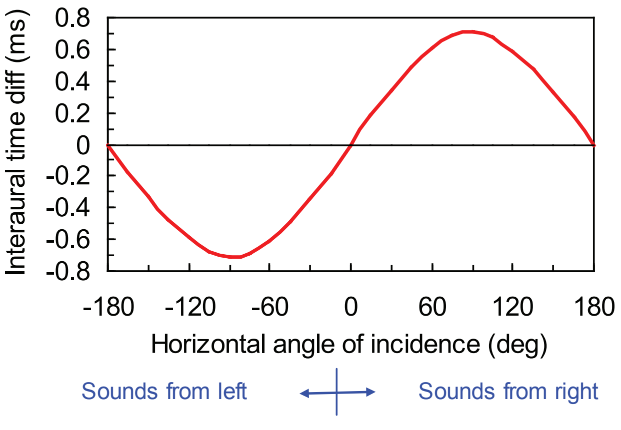


Figure 15.2 Interaural time difference for low-frequency sounds as a function of direction measured from directly in front. Data are the average of measurements on people and on a manikin.[987]

[b] Interaural time differences for high-frequency sounds are about two-thirds of the value for low-frequency sounds, because of the complexities of sound diffraction around the head.[986a]

As frequency increases, this phase cue becomes increasingly ambiguous for sounds that do not have rapid onsets or offsets. Once frequency is high enough for the time difference to cause more than half a cycle of phase shift for sounds coming from the side, multiple source directions will lead to the same interaural phase difference. This occurs for frequencies higher than about 700 Hz. Also, neural responses are highly synchronized to the sound waveform only for low-frequency sounds. Interaural phase difference is thus a strong cue only for low-frequency sounds. Interaural time differences are, however, also present in the ***envelope*** of sounds, and thus *can* be conveyed across the whole frequency range.[719] Time difference cues are nevertheless carried most efficiently by the low-frequency components of sounds, up to about 1500 Hz.[1217, 1942, 1944, 1967]

The head acts as an acoustic barrier and causes a level difference between the ears. ***Head diffraction*** produces an attenuation of sound on the far side of the head and this is usually referred to as ***head shadow***. Head diffraction also produces a boost on the near side of the head. Both effects have the greatest magnitude for high-frequency sounds (see Section 1.2.1). The resulting ***interaural level differences*** are thus much more pronounced at high frequencies.

Figure 15.3 shows how the interaural level difference varies with frequency for three source directions. Level difference cues are most valuable for high frequencies, above about 1500 Hz. Horizontal localization accuracy is, in fact, worst at 1500 Hz, presumably because neither the time nor level difference cues are entirely effective around this frequency.[1196] Horizontal localization accuracy is most precise for frequencies around 800 Hz and for sources directly in front. Under these conditions, people can detect differences in source direction as small as 1°, corresponding to an interaural time difference of only 10 μs.[1196] At 1500 Hz this rises to about 3°.

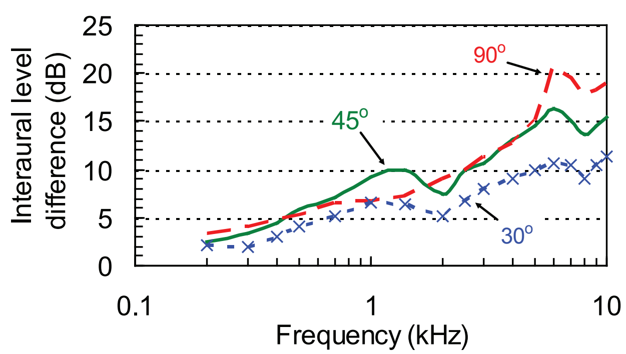


Figure 15.3 Interaural level difference for three source directions in the horizontal plane. Data are calculated from Shaw (1974). Interaural level differences are zero for frontally incident sound.

The relative importance of interaural time and level differences can be deduced from experiments in which headphones are used to present signals with conflicting time and level difference cues. There is a trading relationship between the cues, which confirms the dominance of time-difference cues for low-frequency sounds, and the dominance of level-difference cues for high-frequency sounds. For complex, broadband sounds, the time cues carried in the parts of the signal below 1500Hz appear to be dominant.[1545, 1919, 1967]

Because the ears are placed almost half way between the front and back of the head, for every frontal direction there is a backward direction that results in almost the same interaural level and time difference cues. ***Front-back differentiation*** is less well understood, but depends in part on spectral balance. The pinna boosts high-frequency sounds, principally in the 6 to 16 kHz range, when they arrive from the front, but attenuates them when they arrive from the rear.[1021, 1291, 1292]

Front-back confusions are the most common type of confusions for people with normal hearing, and their likelihood increases as adults age, even for people with good hearing (but possibly as a result of slight or mild hearing loss).[5, 1128] It is important to realize that front-back confusions do not just apply to sounds directly in front of or behind the listener. A location 30° to the right of front, for example, will most easily be confused with a direction 30° to the right of directly behind.

When the vertical plane is considered, there is an entire cone of directions (all subtending the same angle with an axis emerging from the ear canal) that are confused with each other. This cone of confusable directions, as shown in Figure 15.4, is referred to as the ***cone of confusion***. People with normal hearing resolve directions around the cone, somewhat imperfectly, by a combination of vertical localization and front-back localization.

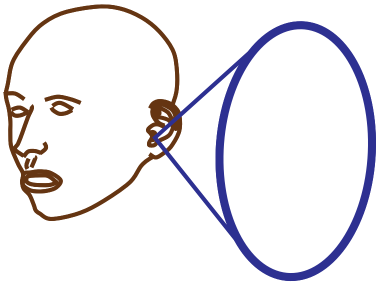


Figure 15.4 A cone of confusion for localization. Any direction for which sounds have to pass through the perimeter of the thick blue circle to reach the ear canal are easily confused with each other, particularly if the audibility of high-frequency sounds is restricted.

Sound during the first few milliseconds of a signal has a particularly strong influence on perceived direction, which helps us ignore the direction from which echoes or reflected sounds arrive.[145] This phenomenon is variously known as the ***precedence effect***, the law of the first wave-front, and the Haas effect.

In summary, accurate and easy horizontal localization in the left-right dimension is possible provided the low-frequency components of sounds are clearly audible in both ears. Audibility of high-frequency sounds assists left-right localization and enables front-back localization. Localization when there are background noises present is less accurate than in quiet, but is also less researched.

Vertical localization, even in the mid-sagittal (or medial) plane (Figure 15.5) where there are no interaural cues, is made possible by reflections and resonances that occur within the pinna prior to sound entering the ear canal.[207, 712, 1521, 1618, 1896] These reflections cause cancellations, and hence spectral peaks and notches, at frequencies that depend on the elevation of the source relative to the head. The resulting cues to localization are all above about 4 kHz (depending a little on the size of the individual pinna), because it is only in this high-frequency region that the wavelength of sound is small enough, compared to the size of the pinna, for the necessary reflections and resonances to occur.[1021] People can detect changes in vertical angle as small as 3°.[1402] Vertical localization is possible with only one ear, but performance improves slightly for listening with two ears, suggesting that the brain combines the two estimates made possible by the information at each ear.[746] The additional benefit provided by the second ear may be facilitated by asymmetries between the shapes of the two pinnae that affect the pick-up of high-frequency sounds.[253] Sources that are not in the mid-sagittal plane are localized by a combination of pinna effects and interaural time and intensity differences.

Externalization refers to the perception that a sound is originating from some point in space outside the head. For sounds to be externalized, the spectral shape of the sound at the two ears must have the features appropriate to the source direction.[495] The transformation from SPL in the undisturbed free field to SPL in the ear canal is called the ***head related transfer function*** (***HRTF***)**.** The HRTF is created by the acoustic barrier effects of the head and pinna and the direction-dependent resonances of the pinna. Appropriate reverberation, and variation of the HRTF as the listener moves his or her head, also contribute to a sense of externalization.[495, 1917] Although people become used to their individual HRTFs, it is possible to achieve good externalization with sounds on which someone else's HRTF has been imposed.[495] Thus, sounds that have been recorded through a dummy head and played back over headphones with a suitable flat frequency response can readily be externalized.

Head movements also help with all other aspects of localization, as they can be used to resolve ambiguous cues that can otherwise occur.[1401, 1889] Head movement also helps people with unilateral deafness to localize horizontally provided the sound lasts long enough to listen using several head orientations.

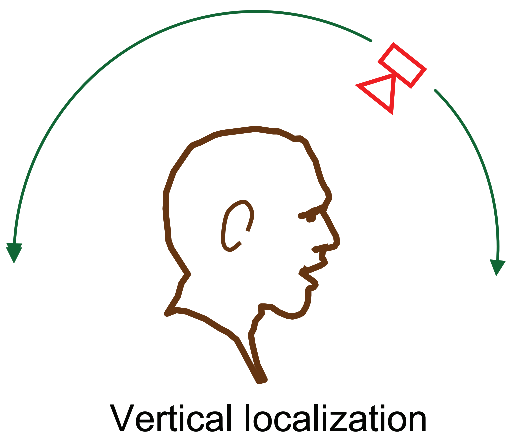


Figure 15.5 Variation of the source direction in the vertical plane.

Localization is important to all people, but none more so than blind people, who use auditory localization to re-construct inside their heads the world around them.

More detailed reviews of localization cues and neural processes are available.[252, 659, 1195, 1918]

15.1.2 Effects of hearing loss on localization

Patients do not often spontaneously complain about poor localization ability. When specifically questioned about localization, however, patients are likely to be aware of difficulties they face because of poor localization, particularly if they have a severe hearing loss.[230] Some researchers consider that impaired localization is one of the two biggest problems caused by hearing loss.[851] Self report measures show that patients with hearing loss report significantly greater problems than normal-hearing people with dynamic aspects of localization, particularly judging distance and movement.[599]

Impaired localization likely contributes to the biggest problem caused by hearing loss – listening in noise. Difficulty in following a conversation within a group of people may be exacerbated by being unable to quickly locate the person talking, particularly when the conversation switches rapidly between speakers.[230] Our sense of localization also helps us assign a separate identity to sounds that come from different directions.[83] Without this identification of different sound sources, multiple noises likely become a confusing general background, rather than a collection of individual sounds that can be perceived, and ignored if desired. As mentioned in Section 1.1.6, hearing-impaired people certainly need the SNR to be markedly higher than that needed by normal-hearing people when target speech and competing sounds come from different directions.[492, 604, 628, 1142, 1400] The impact of hearing loss on localization accuracy appears to be greater when there is background noise than when the target sound is presented in quiet.[129a, 1076]

When hearing aids are worn for the first time, localization is likely to be disrupted (see Section 15.3.2), possibly contributing to the common experience that hearing aids seem to amplify noise more than they do speech. The precise link between impaired localization and impaired speech intelligibility in spatially separated noise is yet to be fully investigated, but a comparison across studies shows that mild and moderate hearing loss have a greater effect on speech intelligibility in separated maskers[628] than they do on the ability to localize sounds presented one at a time.[232, 1335]

Poor localization ability can create a feeling of being isolated from the environment, potentially contributing to a feeling of anxiety.[529] Difficulty in locating environmental sounds can be inconvenient, or in some situations can put the hearing-impaired person in danger.

Hearing loss causes front-back confusions to increase markedly, almost to the point where patients can only guess at front versus back.[884] This marked increase is not surprising given the reliance of front-back discrimination on very high-frequency information. In real life, hearing-impaired people probably only achieve front-back discrimination by head turning. (When the head is turned to the right, for example, a directly frontal source causes sound to be louder and earlier in the left ear, but rearward sources cause the opposite change in interaural differences.)

As low-frequency (below 1500 Hz) sensorineural hearing loss increases, horizontal localization ability (ignoring front-back confusions) gradually deteriorates.[1334, 1335] Provided sounds are audible in both ears, however, very little deterioration occurs until low-frequency hearing loss exceeds about 50 dB HL.[232] In the low frequencies, many signals have their most intense components, hearing loss is usually least, and neural responses remain phase-locked to stimuli. Consequently, interaural time difference cues presumably remain available.

For sounds that have only high-frequency energy (e.g. bird calls), however, hearing impairment greatly decreases the ability to use interaural time difference cues but has less effect on the use of interaural level differences.[1290] The difficulties in localization reported by people with mild or moderate hearing loss must then primarily be caused by some sounds being inaudible (or at a very low sensation level) in at least one ear,[230] or be caused by front-back confusions. Audibility for localization is adequate if the sounds are more than 10 dB above threshold.[1133]

By contrast with sensorineural hearing loss, conductive hearing loss causes a marked reduction in localization ability.[494, 1334] As conductive loss increases, a greater proportion of the sound that activates the cochlea is carried by bone conduction rather than by air conduction and middle-ear transmission. Interaural attenuation is much lower for bone-conducted sound than for air-conducted sound, so interaural time and level differences at the cochlea will be much less than the corresponding differences at the eardrum.[1334, 1965] Even when the bone-conducted sound is weaker than the air-conducted sound reaching the cochlea, the combination of the sounds from the two pathways can greatly disturb the interaural phase differences of the combined signal in the cochlea.

When sound is attenuated in just one ear by wearing a single earplug, localization in the horizontal plane initially deteriorates greatly, as would be expected, because horizontal localization depends primarily on differences between the signals at the two ears. Over several days and weeks, people become accustomed to sound being softer in one ear than in the other, adjust their internal criteria, and can then again use interaural level differences to accurately localize sounds.[88, 555] Indeed, when an artificially induced attenuation in one ear is removed, it also takes some days for normal localization ability to return.[555] The ability to discriminate frontal from backward sources should survive occlusion of one ear right from the start, because this distinction is based on spectral shape, not interaural differences.

Other binaural phenomena also occur with asymmetrical sensation of sound. Binaural beats[c] can be perceived even when the sensation level in one ear is 50 dB greater than the sensation level in the other.[1787] Good horizontal localization ability should therefore be possible even for an asymmetrical hearing loss provided the sound is audible in both ears.

Vertical localization, by contrast, deteriorates markedly with hearing loss.[230, 232, 1476] In the high frequencies, most sounds have their least intensity, most people have their greatest hearing loss, and frequency resolution is usually most adversely affected. Consequently, high-frequency components of a signal will often not be audible. Even when high-frequency components are audible, listeners with sensorineural hearing loss may not have enough frequency selectivity to identify the frequencies at which the important peaks and troughs in the sound occur.[230] Vertical localization ability in the mid-sagittal plane is decreased only slightly by occlusion of one ear, provided the other ear has normal hearing.[711, 746, 1358] This is understandable because the necessary spectral cues will be available to the normal-hearing ear.

Hearing impairment adversely affects distance perception,[599, 1338] particularly when overall level cues are not available and the listener must rely on the ratio of reverberant to direct sound.[23] Hearing aids, even with wide dynamic range compression, do not appear to adversely affect distance perception.[22]

Section 15.3.2 will outline the advantages of bilateral versus unilateral hearing aids for localization. Byrne & Noble (1998) give an excellent and more detailed review of the effects of hearing loss and hearing aids on localization.

15.2 Binaural Effects in Detection and Recognition

In noisy and/or reverberant environments, people can understand speech much more accurately with two ears than with one. The ability to combine information at the two ears in order to listen to one person talking in the midst of many people talking at similar levels is often called the ***cocktail party effect***[d] (though the reader can investigate these effects personally by wearing one earplug at any sort of party they choose).[273, 1943]

There are three reasons why people can more easily understand speech with two ears than with one. The first of these arises from ***head diffraction*** effects and is a purely acoustic phenomenon. The second is referred to as ***binaural squelch*** and relies on the brain taking advantage of differences between the signals arriving at the two ears.[250] The third, which will be referred to as ***binaural redundancy***, also relies on the brain being able to combine signals arriving at the two ears, but does not require the signals at the two ears to be different.[690]

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[c] Binaural beats occur when there is a small frequency difference between the sounds presented to the two ears. See Moore (2012) for a more detailed discussion on this and other aspects of binaural interactions.

[d] Various terminologies exist. The difficulty of hearing in the midst of many talkers is referred to as the ***cocktail party problem***. The propensity of very familiar sounds, especially one's own name, to pop out from unattended speech is variously called the ***cocktail party effect*** and the ***own name effect***.

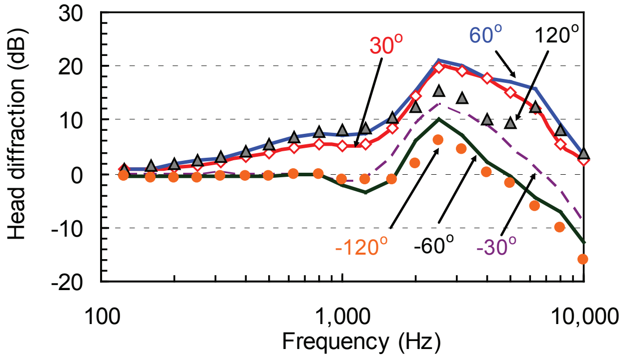

Figure 15.6 Head diffraction effects from the undisturbed field to the eardrum for five source directions in the horizontal plane, with positive angles representing sound arriving from the side of the ear in question. Data are from Shaw (1974).

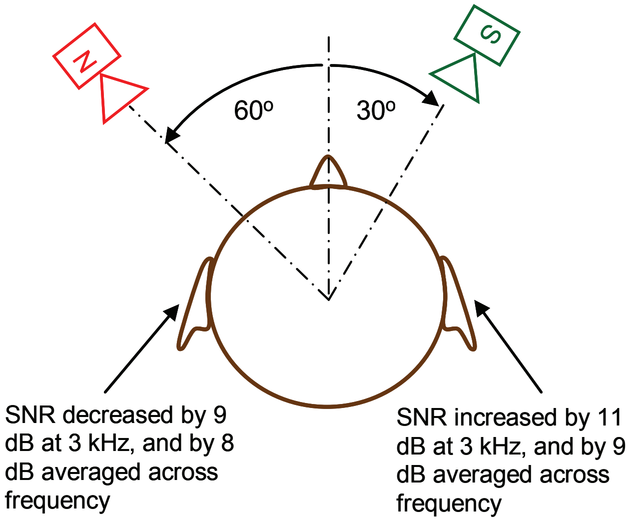

Figure 15.7 Effect of head diffraction on the SNR at each ear, relative to the SNR in the undisturbed field. The SNR at the right ear is thus 20 dB better than at the left ear at 3 kHz, and 17 dB better when averaged across frequency.

15.2.1 Head diffraction effects

Figure 15.6 shows head diffraction effects to the eardrum for sounds arriving at the head from five directions in the horizontal plane. (Note that around 3 kHz, diffraction effects are positive even for sounds arriving from the far side of the head. This boost occurs because all curves include the effects of the ear canal resonance.)

A person able to listen with both ears can benefit from head diffraction effects just by attending to the ear with the better SNR. The advantage arising from head diffraction is therefore commonly called the ***better-ear effect***. The effect on speech can be estimated by weighting the improvement in SNR at each frequency by the importance function used in the Speech Intelligibility Index.[648] For the orientations shown in Figure 15.7, for example, the weighted-average SNR will increase at the right ear by 9 dB and will decrease at the left ear by 8 dB. The ear nearer the speech will thus effectively have a SNR 9 dB higher than in the undisturbed field, and 17 dB higher than at the far ear.

By contrast, head diffraction will severely disadvantage a person who can hear in only one ear if the good ear is on the side of the noise[e] and opposite the side of the target speech.

Example: How head diffraction changes SNR

The following example shows how head diffraction (comprising head boost and head shadow) alters the SNR at each ear. Suppose that speech was arriving at 30° from the right and that noise was arriving at 60° from the left, both at close range, as shown in Figure 15.7. At 3 kHz, the ear canal and head will boost the speech in the right ear by 19 dB but, because of head shadow, the noise will be boosted by only 8 dB. The SNR will therefore be 11 dB *greater* at the right eardrum than it would be in the undisturbed field. In the left ear the opposite happens. The components of speech at 3 kHz are boosted by only 11 dB whereas noise is boosted by 20 dB. Consequently, the 3 kHz SNR at the left eardrum is 9 dB *worse* than in the undisturbed field.

[e] We will use *noise* to mean any sound that is not the target speech the listener is trying to understand. Noise might therefore actually be speech coming from a different (and less interesting) talker.

The magnitude of these head diffraction effects is very large – enough in some situations to make speech totally understandable at one ear and totally incomprehensible at the other. In many circumstances, the effects of head diffraction will be less than those indicated above. First, reverberation will diminish the differences in speech and noise levels arriving at each ear. This will be especially true when the listener is sufficiently far from the sources of both speech and noise that diffuse (e.g. reverberant) sounds dominate the direct sounds. In the extreme case of a listener in a reverberant room, far from both the speech and noise sources, head diffraction will have no effect on the SNR at either ear. Second, the effects of head diffraction on speech intelligibility will be small if the SNR is already so large over part of the frequency range that further improvements do not help. Nonetheless, head diffraction has a substantial effect on understanding speech in many real-life situations.

Head diffraction effects are a purely physical effect so, in a given situation, the SNR at each frequency will be affected in the same way for a hearing-impaired person as for a normal-hearing person. (There may be some differences when the hearing-impaired person is aided, because of microphone location effects.) For people with a steeply sloping high-frequency hearing loss, however, the benefit of head diffraction effects will be less than for normal-hearing persons.[479] Individuals with steeply sloping high-frequency losses will usually be more reliant on low-frequency cues, where head diffraction effects are less pronounced. Also, an improved SNR at high frequencies will not benefit a hearing-impaired person if the high-frequency components of speech are inaudible.[178, 479] This situation will often occur when the hearing-impaired person is unaided.

15.2.2 Binaural squelch in noise

The preceding section discussed how simply attending to the ear that is presented with the more favorable SNR can minimize the effects of noise. The brain and ears can, however, do better than this. The auditory system can *combine* the signals available to each cochlea to produce an internal, central representation of the target signal that effectively has a higher SNR than is available at either ear in isolation.

We can think of this process as the auditory system using the noise at the ear with the poorer SNR to partially remove the noise from the ear with the more favorable SNR. While the physiology responsible for this is unclear, the inputs and results are similar to those of the electronic adaptive noise-reduction schemes shown in Figures 7.10 and 7.11.

Suppose, for instance, that the noise were to arrive from directly in front, and thus have the same amplitude and phase at the two ears, as shown in Figure 15.8. The signal, in this case a pure tone, arrives from the right and therefore has greater amplitude at the right ear than at the left ear. If the brain were to subtract the total waveform at the left ear from the total waveform at the right ear, the resulting waveform would have no noise at all. Because the signal in the left ear is so much smaller (by about 10 dB in this example) than

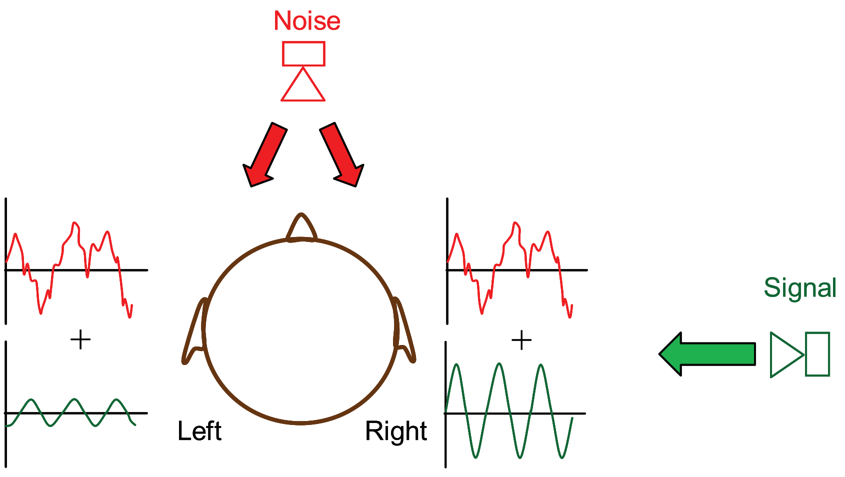

Figure 15.8 Waveforms at the left and right ears when noise arrives from directly in front and a signal (in this case a pure tone) arrives from one side.

the signal in the right ear, the difference between the two signals would be little different from the signal in the right ear.[f]

The auditory system cannot perfectly subtract the waveform at one ear from the waveform at the other. It can, however, make imperfect combinations of the waveforms at each ear to significantly decrease the effects of noise. Furthermore, the auditory system is very adaptable; it is not essential for either the noise or the signal to be in phase at the two ears for some noise suppression to occur. The amount of noise suppression that occurs when the signal has an interaural level or phase/time difference[g] that is different from that of the noise is called the ***binaural masking level difference*** (***BMLD*** or ***MLD***). It is also referred to as ***binaural release from masking***, ***binaural unmasking***, and ***binaural squelch***. When the task is understanding a speech signal, the improvement in SRT relative to diotic stimulation is referred to as the ***binaural intelligibility level difference*** (***BILD***). Both interaural level differences and interaural time differences contribute to BILD.[580]

MLD is most commonly measured clinically by presenting either a 500 Hz pure tone or spondee words to the two ears, in the presence of masking noise. In the baseline condition each of the signal and the noise are in phase at the two ears (the S_0N_0 condition). They are also presented in a phase-inverted condition in which the polarity of either the signal or the noise (but not both) is inverted in one ear relative to the other ear. As inversion is equivalent to a phase shift of 180° or π radians, the condition with the signal inverted is referred to as $S_\pi N_o$, and the condition with noise inverted is referred to as S_0N_π. MLD is calculated by subtracting the SNR at which the tone is detected, or the spondee understood, in the phase-inverted condition from the SNR at which the task can just be performed in the S_0N_0 condition.

The MLD for a 500 Hz signal is typically in the range 9 to 13 dB for normal-hearing adults, but the magnitude is affected by the type of masking noise, and decreases as frequency increases above 500 Hz.[480, 1764, 1925] This decrease with frequency is understandable, because the fine details of the waveform must be represented within the auditory system for the binaural interactions to occur, and the accuracy with which neural impulses are phase-locked to the waveform decreases as frequency increases.[1217] The MLD for spondees in speech-shaped random noise is smaller, at around 5 dB.[480]

In an excellent analysis, Zurek (1993a) has shown that the size of the MLD for pure tones can be used to predict the size of the BILD for speech (in random noise or multi-talker babble), thus suggesting that the same mechanism is responsible for both effects. Measurements on children with normal hearing thresholds but hearing difficulties in the classroom reveal, however, that it is possible to have abnormal BILD for speech in spatially separated competing speech maskers, despite having normal MLDs for pure tones in a noise masker.[242, 245]

The relationship (if any) between BILD measured with competing talkers at different azimuths and MLD measured by inverting the phase of signal or noise is therefore yet to be fully understood. One hypothesis is that bilateral advantage for spatially separated sources arises from the central nervous system selecting small segments of signal, separately at each frequency at each moment in time, from whichever ear has the better SNR, and then re-assembling a complete signal from these elements.

The amplitude and phase differences between the ears that are necessary for a BILD occur whenever the target speech comes from a direction different from the masking noise. The greater ease with which speech can be understood when speech and noise come from different angles is referred to as ***spatial release from masking*** (***SRM***). SRM and BILD are two different ways of viewing the same ability of the binaural hearing system to improve speech intelligibility when the interaural differences for the target are different from the interaural differences for the noise.

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f The signal in the resulting difference waveform may be slightly larger or smaller than the signal at the right ear, depending on the phase relationship between the signals in the right and left ears.

g Whenever an interaural time difference is present, so too is an interaural phase difference. A given azimuth produces approximately the same interaural time difference at all frequencies (Figure 15.2) whereas the interaural phase difference increases proportionally with frequency. Interaural time difference is thus a more invariant quantity within the auditory system.

The magnitude of SRM increases as the angle separating the speech from the noise increases.[1141] The largest SRMs occur when the listener knows where the target is, and when the "noise" is actually a competing talker or talkers with similar tonal qualities (or at least the same gender) as that of the target talker, because there are then few cues other than spatial cues to use in separating speech and noise.[57, 244, 895, 1141, 1337] When the talker and competing sounds are similar in nature, the increased difficulty in segregating the two (with or without the benefit that spatial separation provides) is referred to as ***informational masking***.[h]

BILD and SRM can be very large under certain conditions. For a frontal speech target being masked by two competing speech signals at -90° and +90°, with no reverberation, their magnitude is around 13 dB for young, normal-hearing adults.[240, 241, 1141] For different maskers and directions their size is smaller. For a continuous noise masker, Zurek (1993a) estimates that averaged over all directions from which noise can come, the BILD is about 2 dB, compared to simply attending to the ear with the better SNR. He also estimates that the benefits from head diffraction provide a further 3 dB advantage in SNR. The total average binaural advantage, relative to listening with a single, randomly chosen ear, is thus estimated at 5 dB.

The benefits resulting from head diffraction will be most effective for the high-frequency parts of speech, and the benefits resulting from interaural time differences will be most effective for the low-frequency parts of speech. Their relative importance when speech intelligibility is measured will thus differ depending on whether low-frequency cues dominate (as in a spondee test) or high-frequency cues dominate (as in a nonsense-syllable test).

BMLD, BILD and SRM also exist for hearing-impaired people, but their magnitudes are decreased.[494, 1142, 1364, 1691] The greatest reduction occurs for people with the greatest hearing loss[628, 674, 816, 1142] and for people with the greatest asymmetry of hearing loss.[816] This reduction is caused both by an inability to benefit from the SNR improvement offered by head diffraction (when high-frequency sounds fall below threshold) and by a reduction in the binaural interactions that enable binaural squelch.[178, 628] It is possible that hearing loss decreases the accuracy of the timing of neural impulses on which binaural squelch relies.

Some binaural squelch occurs even when loudness is not well balanced between the ears, provided sounds are audible in both ears.[178] BILD and SRM are entirely absent, however, when sounds are inaudible in one ear.[1141] Training appears to increase the ability of hearing-impaired listeners to take advantage of SRM when aided.[1309]

Binaural squelch also partially suppresses the adverse effects of reverberation,[957, 1206, 1297] and the combined effects of reverberation and noise.[705, 1206, 1297, 1298, 1300, 1433] Such a result is not surprising, as reverberation is similar in nature to background noise. Although reverberation is itself squelched by binaural effects, reverberation will decrease the extent by which noise is squelched, because it diminishes the interaural time and level differences of the speech and noise.[1104, 1141-1143, 1298, 1433]

15.2.3 Binaural redundancy

Binaural redundancy refers to the small advantage arising from listening with two ears even though identical combinations of signal and noise are presented to each ear.[403, 408, 1236, 1694] The phenomenon has also been referred to as ***diotic summation*****,** or ***duplication*****.**

Binaural advantage for diotic listening is not too surprising if one considers that hearing impairment can in many ways be simulated by an additive noise that produces appropriately elevated thresholds. We can think of binaural redundancy as the suppression of internal noise within each ear, or as an improvement in decision-making ability when the brain combines the two ostensibly identical signals sensed by each ear. It is as if the brain gets two "looks" at each sound. Binaural redundancy produces 1 to 2 dB improvement in SNR.[176, 366, 1104, 1433]

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[h] *Informational masking* is a term that has come into wide use but there is no single definition of it.[493] It can perhaps be thought of as that part of a masking effect that is *not* caused by components of the signal and masker occurring at the same time and frequency and arriving from the same direction. Informational masking occurs when the listener is uncertain which parts of the combined sound belong to the signal, and which parts belong to the masker, even though the separate parts are all audible. A contrast to informational masking is ***energetic masking***, where signal and masker occur at the same time and their components fall within the same auditory filter bands.

Binaural redundancy also results in better speech discrimination in quiet.[849, 1143] Even people with severe hearing loss[408] and people with an apparent central component to the hearing loss[849] can benefit from binaural redundancy (but see Section 15.4.2 for an important exception).

Binaural redundancy seems to require a lower level of binaural interaction than binaural squelch. In one experiment, hearing-impaired subjects who obtained a 3 dB binaural redundancy advantage could not distinguish a diotically presented stimulus from a dichotically presented stimulus, although normal-hearing subjects easily could.[i, 365]

One study with older children, all of whom had a congenital or pre-lingual bilateral profound loss, failed to find any bilateral advantage, even though the experimental conditions would have allowed binaural redundancy and perhaps binaural squelch to operate if present.[656] The impact of severe and profound congenital loss on the capacity of people to form binaural interactions has not been sufficiently studied to draw any general conclusions about it.

15.2.4 Binaural loudness summation

For a normal-hearing person, the loudness of a sound is greater if it is heard in two ears than if it is heard in only one ear. This loudness increase occurs for all levels although not to the same degree:[421, 668, 673, 1136, 1495]

- Near threshold, binaural summation of loudness is equivalent to increasing the level in one ear by 2 to 3 dB.[421, 1619]
- At a comfortable level, binaural summation of loudness is equivalent to a level change of 4 to 6 dB,[303] although some studies indicate that the difference is around 10 dB.[222, 668]
- At very high levels, binaural summation of loudness is equivalent to a level change of around 10 dB, although some studies indicate that the difference is only around 6 dB.[797, 1561]

Taken together these studies indicate that binaural summation of loudness increases from around 3 dB or lower near threshold to some value in the range 6 to 10 dB at high levels.

In these older studies, the increase in level to keep loudness constant when changing from binaural listening to monaural listening is about the same as the change in level needed to double loudness when listening via one ear.[j] If this doubling of loudness when the number of ears contributing to loudness doubles is more than a coincidence, and if the same principle applies to hearing-impaired people, then we would expect slightly less loudness summation (when expressed in dB of level change) for hearing-impaired people because of their steeper loudness growth curves. Unfortunately, loudness measurements are notoriously dependent on the measurement method, and more recent results indicate that listening with two ears is much less than twice as loud as (but is still louder than) listening with one ear if the sounds are heard in the sound field and/or come from a visible live talker rather than pre-recorded material presented via headphones.[353, 522]

Whatever the reasons, experimental data suggest that binaural loudness summation is equivalent to a smaller level change[k] for hearing-impaired people than the changes summarized in the preceding bullet points for normal-hearing people.[421, 673, 702, 1137] A reasonable estimate might be that the level difference between binaural stimulation and monaural stimulation for hearing-impaired people might be about 4 dB at mid levels, a little less at low levels, and a little more at high levels. Precision in this estimate is low because of the marked variation between individuals,[1137] the dependency on degree of loss, and the variation with measurement method. [522]

Note that a 3 dB change near threshold is the amount expected for the optimal combination of two physical detectors, each of which senses the same signal but which have their own, independent internal

[i] Binaural redundancy is one component of binaural squelch, if binaural squelch is defined as the improvement relative to monaural stimulation rather than to diotic stimulation. We can more clearly understand all the components of binaural advantage, however, if the reference condition for binaural squelch is diotic stimulation.

[j] Loudness grows more steeply with level at low SPLs than at high SPLs, so the change in dB needed to double loudness is much less near threshold than at higher SPLs.

[k] The comparison is affected by whether summation is compared at the same SPL, the same SL, or the same loudness category.

noise sources that determine the weakest signal they can detect. This improved ability to detect and rate the loudness of weak sounds is therefore just another aspect of the binaural redundancy advantage described in the previous section.

Although one might expect that binaural loudness summation might also apply to loudness discomfort level (LDL), experimental data suggest otherwise. Depending on the method by which LDL is measured, binaural summation decreases LDL by some amount between 0 and 6 dB.[107, 668, 702, 1712] It is possible that loudness discomfort is affect by factors others than the rated loudness. Binaural stimulation may allow sounds to be louder than for monaural stimulation without causing loudness discomfort!

By itself, the binaural summation of loudness leads to neither an increase nor a decrease in the benefits provided by bilateral hearing aids relative to a unilateral aid. There are, however, several implications of loudness summation for fitting procedures, and these will be considered in Section 15.8.

15.3 Advantages of Bilateral Fittings

The answer to a difficult questions (which the question *One hearing aid or two?* certainly is) often depends on how the question is asked. Quantifying the advantages (and disadvantages) of binaural fittings is no exception. For objective evaluations, the answer depends on the measurement set-up. For self-report evaluations, the answer depends on which aspects of auditory perception the questionnaire examines.

The following sections outline potential bilateral advantages and some empirical evidence concerning these advantages.

15.3.1 Speech intelligibility

In many situations, most people can understand speech more clearly with two hearing aids than with one. This occurs for the same three reasons that listening with two ears is better than listening with one. In essence, hearing aids make sounds audible, which can enable the same binaural mechanisms as in normal hearing to occur, although not necessarily to the same degree, and sometimes not at all.

Head diffraction effects

When speech and noise come from different directions, the SNR will be better at one ear than at the other.[1] If the ear with the better SNR is unaided, the patient may be unable to take advantage of this better SNR.[539]

The bilateral advantage arising from head diffraction will thus occur for those patients and situations where the high-frequency components of speech are inaudible when unaided. The bilateral advantage arising from head diffraction will be least for those whose high-frequency hearing loss is only mild and for those whose high-frequency hearing loss is so severe that the high frequencies make no contribution to intelligibility even when aided. Otherwise, the bilateral advantage for intelligibility caused by head diffraction will occur for all patients, because the SNR variations underlying the advantage depend only on head size. The benefit in intelligibility is, however, less than would be expected on the basis of the physical change in SNR for aid wearers with sloping high-frequency loss,[19, 479] presumably because they have become more used to relying on low-frequency cues (Section 10.3.4).

Although a unilateral hearing aid wearer can often position his or her self so that the aided ear is on the side of the head with the better SNR, this is not always possible, especially in dynamic situations where several people take turns at being the talker, such as at a dinner table.

Binaural squelch

The bilateral advantage arising from binaural squelch should occur for those patients and situations where the low-frequency components of speech and noise are not clearly audible in either ear unaided. Markides (1982a) estimates that an ear will contribute to squelch if speech is more than 20 dB above the speech detection threshold for that ear, and this estimate is consistent with more recent research.[1143] Binaural squelch therefore cannot lead to a large bilateral advantage if the low-frequency components of speech and noise are already well above threshold in both ears in the unilateral condition. The bilateral advantage arising from squelch should thus be small or non-existent for

[1] Either the speech or noise must be asymmetric relative to the midline of the listener if it is to produce a different SNR at each ear. Frontal speech, combined with either noise at 180° or noise at both -90° and +90°, for example, will result in equal SNR at the two ears.

those whose low-frequency hearing loss is only mild. Bilateral benefit from binaural squelch becomes more likely as hearing thresholds increase.

Because interaural level differences also contribute to binaural squelch,[580] and as interaural level differences are conveyed principally by the high frequencies, high-frequency thresholds may also slightly affect the extent to which hearing aids increase binaural squelch.

Binaural redundancy

Binaural redundancy can also occur only if sounds are audible in each ear. A bilateral advantage may therefore occur if sounds are audible in both ears when a patient is aided bilaterally but not when aided unilaterally. However, even with this proviso, we cannot assume that binaural redundancy will be present for all patients. In fact, for elderly patients, the opposite of binaural redundancy, ***binaural interference***, may well be present, as described in Section 15.4.2.

Experimental evidence

These theoretical expectations arising from diffraction, squelch and redundancy are consistent with laboratory measurements of speech intelligibility. Of 19 *laboratory studies* reviewed by Ross (1980), 15 showed a bilateral advantage for speech, and four showed no difference. None showed a unilateral advantage. Some studies are unfairly biased in favor of a bilateral advantage by plugging the unaided ear in the unilateral condition. Such studies really demonstrate the advantage of two ears (binaural advantage) rather than the advantage of two hearing aids (bilateral advantage). Other studies are unfairly biased against a bilateral advantage by using a single loudspeaker for signal and noise, as there are no spatial cues for binaural hearing to assist with. More recent laboratory studies continue to show a speech intelligibility in noise advantage for bilateral over unilateral aiding.[940, 1143]

Despite the bilateral advantage that laboratory studies show for listening in noise, many early surveys of bilateral use in *real-life* found that patients often used one (or no) hearing aid in noisy situations, even if they used two hearing aids in more favorable listening situations.[180, 186, 209, 308, 465, 1571]

It is unfortunate that where patients most need help, in very noisy places, speech and noise levels are high, so the second hearing aid is least likely to turn an inaudible signal into an audible signal (Section 9.1.6). That is, bilateral benefit is *least* likely in very noisy places.

These pessimistic conclusions about bilateral advantage in noisy places apply most strongly to patients with mild hearing loss. Patients with symmetric moderate, and especially severe hearing loss, show at least as much bilateral advantage in noisy places as in quiet places.[1134]

It is not surprising that patients removed one aid in noisy places in the 1970s and 1980s when many of these studies were performed. The hearing aids would commonly have been linear, peak clipping devices that produced highly distorted sound in noisy places, and were more cosmetically obtrusive and less physically comfortable than is usual now. Removing one or both hearing aids in such situations would seem

Summary: Bilateral advantage for speech intelligibility for different degrees of hearing loss

Bilateral advantage for speech intelligibility is greatest for those with severe hearing loss and least for those with mild loss:

- For people with a severe loss in an unaided ear there are unlikely to be any situations in which the unaided ear can contribute useful information at all frequencies.[539] The bilateral advantage for people with severe hearing loss applies even when there is some asymmetry between the ears.[408]
- For people with a mild loss in the unaided ear, signals are likely to be well above threshold in noisy places. Consequently, the ear will contribute to speech understanding even when it is unaided. When signal levels are low there is the potential for a bilateral advantage. However, in such circumstances SNR is often large, so intelligibility with a single hearing aid may be more than adequate. Consequently, the only situations in which there is a significant bilateral advantage are likely to be when speech is arriving from the unaided side, and/or noise is predominantly arriving from the aided side of the head.

to be well justified.[1302] People more strongly want to use two hearing aids if they are listening through high quality hearing aids than if listening through distorting ones, especially at high signal levels and when noise is present.[1302]

So what should happen with low-distortion, high-quality hearing aids? If noise levels are sufficiently high that noise, rather than elevated hearing thresholds, limits audibility at all useable frequencies, then simple amplification will not improve speech intelligibility. There will be no bilateral advantage over a unilateral fitting, just as there will be no unilateral advantage over unaided listening, as illustrated in Figures 9.3 and 9.4. If the acoustics of the situation favor the use of directional microphones, then a unilateral fitting may provide some benefit over unaided listening, but a second directional aid may not provide any further increase in speech intelligibility.[106]

There are at least two reasons why laboratory studies may produce more positive bilateral benefit concerning speech intelligibility in noise than we reach from real-life studies based on self report. First, the phrase *speech intelligibility in noise* is extremely vague. Laboratory studies mostly use typical levels of speech, in the range 60 to 70 dB SPL, with similar levels of noise. When patients report on their use of hearing aids in noisy places, they likely are referring to places with levels considerably higher than this. Second, laboratory studies often require the research subject to sit still, looking in a particular direction while the target speech occurs. In real life, people usually have the opportunity to orient themselves to maximize the SNR at their aided ear, as inconvenient as this sometimes may be.

Bilateral benefit appears to be as great in reverberant conditions as in anechoic conditions.[1352] The use and benefit of two hearing aids, relative to one, in real life is discussed further in Section 15.7

Effects of signal processing schemes on bilateral advantage

For those with a moderate or greater hearing loss, the speech advantage provided by bilateral hearing aids over a single aid is robust for most patients. It is additive to the advantages provided by directional microphones.[705] This is fortunate as directional microphones considerably alter the phase of sounds.[m] When an FM system provides the input (a monophonic signal) to hearing aids, we would not expect to gain any advantages from head diffraction or binaural squelch. We would, however, expect to retain the very small bilateral advantage arising from binaural redundancy. This small bilateral advantage is consistent with available evidence.[699, 1059]

There is a theoretical possibility that compression applied independently at the two ears could disturb the binaural squelch effect. Such disturbance is not likely because binaural squelch relies mainly on timing or phase differences, and compression does not affect these. Experimentally, a binaural[n] advantage is found irrespective of whether the hearing aids contain wide dynamic range compression or linear amplification.[1236] Nonetheless, it may not be advisable to use compressors with attack and release times that differ between the ears, because this could unnecessarily lead to potentially confusing level difference cues.

Similarly, it may not be advisable to use digital hearing aids with different processing algorithms in each ear as they may then have different processing delays and hence disturb interaural time differences. For patients with asymmetrical hearing loss it will usually be necessary to use different compression ratios and possibly different compression thresholds in each ear. There is no reason to refrain from such prescriptions as these differences arise from, and partially offset, the asymmetry in hearing thresholds that will already be disturbing the normal interaural sensation level.

15.3.2 Localization

The advantages of bilateral hearing aids to localization apply principally to people with a flat or gently sloping moderate hearing loss, and to those with a severe or profound loss. Bilateral advantage for localization is greater for weak sounds than for intense sounds. Based on our understanding of localization, and the psychoacoustic characteristics of hearing loss, it is easy to understand, in principle, how hearing aids affect localization.

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[m] Either the phase alterations are equal at each ear, leaving interaural time difference unaffected, or the bilateral advantage is arising primarily from head diffraction effects which do not require any particular interaural phase relationships.

[n] The term *binaural* is used here rather than *bilateral* because the non-test ear was occluded with a switched-off hearing aid.

Theoretical background: The effect of hearing aids on interaural time and phase cues

Hearing aids alter interaural time cues, and therefore phase cues as well. Within a hearing aid, tubing, transducers, and filters (i.e. tone controls) all create delays. Transducers delay sounds by different amounts at different frequencies. Digital processing can delay sounds by several milliseconds. These delays are very significant compared to interaural time difference cues created by the separation of the ears on either side of the head.

In the low- or mid-frequency region, sounds often arrive at the eardrum via two paths: an amplified path through the hearing aid; and a direct acoustic path through the vent and/or leakage around the mold or shell. Such multi-path transmission can drastically alter the phase response of the combined response at the eardrum, and hence can alter the interaural phase cues. Furthermore, small changes in the characteristics of either path can cause large changes in the phase response of the combined path. The characteristics of a leakage path will vary whenever the aid wearer moves his or her jaw. The amplified path varies its characteristics whenever a compressor or volume control causes the gain to change. It should not be surprising that inserting one or two hearing aids will cause localization to immediately deteriorate. Although the sense of localization is very adaptable (see other panel), it may not be possible to fully adapt to such a changing interaural phase difference.

With open-canal hearing aids, interaural time and phase cues are not likely to be affected by the hearing aid up to at least 500 Hz and, depending on the hearing aid gain, perhaps up to 1000 Hz.

Horizontal localization

Without audibility there can be no localization. Bilateral hearing aids will enable better localization than a unilateral hearing aid whenever the sound is inaudible in the unaided ear of a unilateral fitting. The localization advantages of bilateral hearing aids are thus significant for patients with a moderate or severe loss but not for those with a mild loss.[232] One study simultaneously measured localization and speech intelligibility in noise in unaided, unilaterally aided, and bilaterally aided conditions for subjects with four-frequency average loss averaging 45 dB HL.[940] Both intelligibility and localization were better in the bilateral condition than the unilateral condition, although localization with bilateral hearing aids was no better than when unaided. That is, unilateral aiding made localization worse than for unaided listening for people with moderate hearing loss. Unilateral aiding, as well as unaided asymmetrical hearing loss, result in asymmetrical confusions about the direction of a source.[1631]

For any degree of loss, the bilateral advantage for localization increases as the stimulus level decreases.[420] Patients with a moderate hearing loss may or may not report improved localization in real world situations when aided bilaterally,[1338, 1469] although such an advantage *must* be present for soft sounds. These considerations are the same as those determining when a second hearing aid will enable speech intelligibility to be improved by taking advantage of the SNR changes created by head diffraction (Section 15.3.1).

For patients with a mild hearing loss, localization may be worse when aided than when unaided.[230] This may be explained by:

- test subjects being unfamiliar with the hearing aids with which they were tested (bilateral when they are used to unilateral or vice versa, or being tested with a hearing aid that they have not yet gained experience with in real life);[1333]
- the hearing aid (other than CICs) obscuring anatomical features of the ear that provide cues to localization, or the microphone being positioned where the sound is not affected by these features; and
- the more complex low-frequency response when aided (see panel). Normal low-frequency cues can be maximally retained by using as open an earmold as possible.[232]

Differences in localization accuracy (excluding front-back confusions) between BTE, ITE and ITC hearing aids appear to be small,[130, 1033, 1366] especially when subjects have had time to become used to the localization cues their particular hearing aid or hearing aids provide.[232] This should not be surprising as bilateral fittings of any type are able to preserve interaural time and level differences.

Front-back confusions remain at least as common when wearing hearing aids, whether worn unilaterally or bilaterally, as when unaided.[884, 887, 940] This may be because the very high-frequency range in which the cues reside are not made sufficiently audible by hearing aids, the aid wearers lack sufficient frequency selectivity to make use of the cues that are audible, or the hearing aid microphone is situated where there are no front-back cues (e.g. an omni-directional microphone in a BTE above the pinna). The last of these is certainly not the only cause as although front-back errors are less with CIC, ITC or ITE hearing aids than with BTE hearing aids, these errors remain much larger than normal.[130, 383, 1800, 1912] Patients are more satisfied with sound localization when using CIC hearing aids than when using other aid types.[945, 1260]

Some improvement in front-back confusion is provided by directional microphones, particularly if they are directional in only the high frequencies.[o, 884, 887] High-frequency directivity imparts a greater high-frequency emphasis (i.e. ***brilliance***) to frontal sounds than to rearward sounds, which mimics the changes in sound quality that the pinna produces for people with normal hearing.

For patients with conductive hearing loss, hearing aids will likely produce a marked improvement in horizontal localization ability. This presumably occurs because, in addition to increasing audibility, hearing aids increase the proportion of sound delivered by air conduction, and hence increase the interaural time and level differences at the cochlea.[231] The material used in the earmold, or perhaps the tightness of the earmold, also affects the amount by which localization improves. Unfortunately, the mechanisms are not yet understood sufficiently well to say which materials work best for which patients.[231]

As mentioned in Section 2.4.5, the transducers, electronics and tubing of hearing aids create delays. Different frequencies are delayed by different amounts. Furthermore, insertion of the hearing aid takes away some of the frequency-specific delays caused by the shape of the external ear and ear canal. Consequently, the time response and hence phase response when a hearing aid is inserted are very different from those of the unaided ear to which patients have long been accustomed. Even more importantly, the phase response at the two ears when aided would not usually match each other, so the interaural time difference will also be abnormal. Not surprisingly, the disturbed interaural time cues greatly disrupt localization ability when hearing aids are first worn.

It is possible for the filters in hearing aids to include a frequency-dependent phase shift that, at the eardrum, approximately restores the phase response of the unaided ear (except for a constant delay at all frequencies that cannot be avoided).[929, 1845] As a consequence, interaural time differences are also approximately restored to normal.

Evaluation of these phase-preserving hearing aids showed that they greatly reduced the disturbance to localization that hearing aids otherwise cause when first worn.[473] Over the 16 weeks following fitting, localization accuracy improved for the phase-preserving hearing aids and even more so for the non-phase-preserving hearing aids. There was no significant difference in localization accuracy at 16 weeks, reflecting patients' ability to learn new cues to localization (see panel). Localization accuracy for both conditions approximated the accuracy when listening unaided.

Acclimatization: When can localization be measured?

Hearing aids almost always alter localization cues. If people are tested with a hearing aid type or fitting configuration (e.g. unilateral instead of bilateral) that they have not previously worn, localization performance deteriorates markedly.[1333] There is substantial evidence that people adapt to altered interaural time and intensity cues to localization.[228] Significant adaptation commences within a few hours and adaptation continues for a few days and to a lesser extent for a few weeks.[88, 555, 726, 803] Consequently, assessing localization ability (Section 14.5.4) at a fitting appointment is not feasible; assessment at a follow-up appointment within a typical hearing aid fitting and rehabilitation program is feasible although not usually necessary.

o Serendipitously, high-frequency directivity is the *only* type of directivity that is possible with open-canal hearing aids.

The phase matching also benefitted speech intelligibility in noise: speech intelligibility at 16 weeks was very slightly (but significantly) better for the phase-preserving hearing aids than the non-preserving aids.[473] Possibly binaural squelch can best operate when the interaural time differences are consistent across frequency. The initial improvements in localization and eventual improvements in speech intelligibility were consistent with higher self-report scores for the phase-preserving hearing aids than for the non-preserving aids, particularly for the spatial items on the Speech, Spatial & Qualities (SSQ) questionnaire.[p] It is unclear whether the localization and speech intelligibility benefits were caused by matching the phase response to the unaided ear, or from matching the phase response in the left ear to that in the right ear, since both were accomplished.

As signal processing schemes in hearing aids become more complex, it is possible that some schemes in the future will cause the phase response of hearing aids to vary with the signal, and to vary by different amounts in each ear. Any such schemes would adversely affect horizontal localization.[230] One such example in current hearing aids is adaptive directional patterns, which, if operating independently in each ear, changes the interaural phase differences, and hence affects horizontal localization.[887, 1837]

Compression operating independently in the two ears will certainly reduce interaural level differences, especially for short attack times.[q, 887, 1290] The adverse effect of independent compression on localization is very small for most sounds, however, as it does not affect interaural time cues. [887, 1290] Fast compression will affect localization of sounds that have energy only at high frequencies (e.g. bird calls), as hearing-impaired people are even more reliant than people with normal hearing on interaural level difference for such sounds.[1290]

Gain mismatch between the hearing aids will have a similar effect. If the patient adjusts the volume control of each hearing aid independently, and accidentally creates an imbalance between the two sides, the localization of high-frequency sounds is adversely affected.[870] There is thus some value in bilaterally linked hearing aids that force compression to change gain by the same amount in each ear, and that ensures the volume control has the same effect on both hearing aids. The advantages will be negligible for broadband sounds, because localization will be dominated by high-frequency interaural time difference cues.[870, 887]

Vertical localization

Once an ITE or a BTE earmold is inserted, the concha bowl shape that normally gives rise to vertical localization cues is removed. Consequently, vertical localization ability is almost totally destroyed. For a CIC hearing aid, vertical localization cues up to at least 10 kHz are present at the microphone inlet, because they are fully formed just inside the entrance to the ear canal.[1175, 1191, 1615] Whether the aid wearer is able to use these cues to localize sounds vertically is another matter. Because the cues are all above 5 kHz, localization is possible only if the hearing aid makes information audible over enough of the 5 to 12 kHz range, and if the aid wearer has frequency resolution sufficiently precise to identify spectral troughs within this range.

For ITC hearing aids, the situation should be intermediate to that for ITE and CIC hearing aids. Some of the concha remains open, but the sensing point (the hearing aid microphone inlet) will not be in the normal place (the ear canal entrance). Consequently, any vertical localization cues that remain will be drastically different from normal and the user will have to learn to use them, if indeed that is possible.

Experimental evidence confirms that BTE, ITE, and ITC hearing aids do not improve vertical localization, and can even make it worse.[130, 1332] It seems that normal-hearing people can localize vertically reasonably well when wearing a CIC with a bandwidth up to at least 8 kHz, but there is still some deterioration relative to normal, unaided listening.[383, 1637]

For those few patients who have near-normal high-frequency thresholds, an open earmold can preserve vertical localization, naturalness, and externalization.[235]

p The self-reported (SSQ) advantage of the phase-preserving hearing aids was of a worthwhile magnitude, but was not statistically significant, which is not surprising given the small number (7) of research participants.

q At the start of a sound or change in level of a sound, compression has no effect on interaural level differences. By the end of the compressor attack time, the interaural level difference equals the uncompressed interaural level difference divided by the compression ratio for sounds at that overall level. The perceptual effects are thus greater for shorter attack times.

15.3.3 Sound quality

Binaural listening provides sound quality superior to that of monaural listening. This advantage is found for a number of attributes, such as clarity, fullness, spaciousness, and overall quality.[76] People can generally make more discriminating judgments about sound when listening binaurally than when listening monaurally. For example, just noticeable differences for intensity are smaller[674, 675, 825, 1536] as are just noticeable differences for frequency.[825, 1413] The adverse effects of peak clipping are also more evident for binaural listening than for monaural listening.[1302] (This could be viewed as a disadvantage of bilateral fitting, but only if one is fitting poor quality hearing aids!) The adverse effects of reverberation on intelligibility and sound quality are less marked for binaural listening than for monaural listening.

In the experiments leading to these conclusions, however, the monaural results were obtained by presenting stimuli to only one ear. Consequently, we cannot confidently generalize the findings to the wearing of two hearing aids rather than one, except where the person has a sufficiently severe loss that unilateral aid use will indeed provide monaural stimulation. For people with mild hearing loss, there *may* nonetheless be quality advantages for bilateral aid use. Certainly patients often report that they appreciate the sound being balanced in loudness between the two ears.[165, 527]

15.3.4 Avoiding late-onset auditory deprivation.

After a person with a symmetrical hearing loss (referring both to pure-tone thresholds and to speech recognition ability) is fitted with a hearing aid in only one ear, the ability of the person to understand speech presented to the other ear may progressively deteriorate over the subsequent few years.[1625] The phenomenon is referred to as ***late-onset auditory deprivation*.** Late-onset auditory deprivation affects a significant minority of unilaterally aided patients.[783] In bilaterally aided patients its occurrence is rare and presumably has different origins.[783]

Auditory deprivation in the unaided ear has been demonstrated for children as well as for middle-aged and old adults.[607, 694, 782, 1567] In children, the positive effects of maturation offset the negative effects of deprivation in the unaided ear. Consequently, speech scores may actually increase over time in the unaided ear, but to a lesser extent than occurs in the aided ear.[694]

Along with the decrease in speech recognition in the unaided ear, there is sometimes a small increase in speech recognition in the aided ear.[151, 589] This aspect is referred to as ***acclimatization*** (see Section 14.7).

Both deprivation and acclimatization are consequences of ***plasticity*** in the auditory system. Neural reorganization occurs when the inputs to the auditory system are changed. Changes in cortical activity in response to sounds input to one ear at a time have been observed using functional magnetic response imaging (fMRI) in the 9 months following unilateral aid fitting.[790] More dramatically, following sudden sensorineural hearing loss in one ear, an increase in dominance of the hemisphere contralateral to the good ear has been demonstrated electrophysiologically soon after the sudden deafness occurred.[1444] There is also strong physiological evidence for plasticity in animals, as reviewed by Neuman (1996).

The exact time course of auditory deprivation is unknown, but the effect is observable in group data as soon as one year after fitting.[1627] It takes from seven months to five years for individual scores to significantly decrease.[605, 783, 784] Although performance decreases with time over the first few years, a further period of unilateral use does not seem to lead to a further decline in speech performance in the unaided ear.[608]

The magnitude of reduction varies across patients. One study using subjects with mild, moderate, and severe hearing losses showed a mean decrease in speech scores of 7% in the unaided ear compared to 3% in the aided ear.[608] Another study, for patients of unspecified degree of hearing loss, found *no* greater decrease in the unaided ear than in the aided ear.[1675] By contrast, a case report on one subject with profound loss revealed a decrease of 40%.[151] Another study using subjects with primarily conductive hearing loss with symmetrical hearing thresholds showed a mean difference in speech scores of 30% between the aided and unaided ears.[425]

Auditory deprivation effects are well documented in people with moderate to severe hearing loss.[1312] The effects also occur in ears with a three-frequency average loss as small as 35 dB.[608, 784] A comparison across studies suggests that the size of the deprivation effect increases with the degree of pure tone loss.[783] This is easy to understand as, the greater the loss, the greater is the disparity in the information made available to the brain by the aided and unaided ears.

Although the phenomenon affecting the unaided ear is referred to as *deprivation,* the phenomenon is more complex than a simple lack of stimulation, as the following observations imply:

- When hearing-impaired people with equal speech intelligibility in the two ears continue to live without any hearing aids, no decrease in speech intelligibility occurs (other than possibly a very gradual decrease associated with aging).[608]
- Prior to aiding, people with a small threshold difference between the ears often have markedly better speech recognition ability in the ear with the slightly better hearing thresholds than in the other ear.[759]
- Speech recognition for the impaired ear in a unilateral loss is likely to be much worse than for an impaired ear with the same pure tone thresholds in a bilateral, symmetrical loss.[759, 1244]
- A prospective study showed that speech recognition in the poorer ear of unaided patients with asymmetrical hearing loss deteriorated over a 2-year period. No decline occurred in the better ear, nor in the poorer ear of a similar group of patients whose poorer ear had been aided during the period.[1628]

These findings regarding auditory deprivation suggest that an ear may *become* strongly dominant for intelligibility because the other is initially only *marginally* inferior. This initial inferiority may be because of a slightly asymmetrical hearing loss or because only one ear has been aided. *It is as if the brain gives up attending to an ear that transmits a relatively poor signal when it has the option of attending to a better signal coming from the other ear.* [r]

Auditory inferiority, ***auditory inactivity,***[425] or ***lazy ear*** may perhaps be more accurate descriptions of the process than auditory deprivation. The *deprivation* term is, however, useful, because the root cause is an inadequate output from the cochlea. The extent of auditory deprivation in the unaided ear may therefore depend on the degree of loss in the aided ear. Auditory deprivation probably occurs only if the aided ear sends a signal sufficiently rich in information to cause the brain to cease to attend to the unaided ear.

Note that it is not necessary for the signal from the cochlea of the unaided ear to be distorted. Dieroff (1993) showed that speech recognition scores under headphones in the unaided ears of people with symmetric, essentially conductive hearing loss were on average 33 percentage points lower than scores from the normally aided ears of a matched group of subjects. This finding indicates that a simple attenuation of signals prior to the cochlea, relative to that occurring prior to the other cochlea, is enough to cause auditory deprivation.

Recovery of the unaided ear following bilateral amplification is possible, although it does not always occur.[151, 605, 1312, 1626, 1629] It can also be dramatic. In the case reported by Boothroyd (1993), commencement and later recommencement of bilateral aiding each led to increases in speech recognition performance in the previously unaided ear of around 40 percentage points.

When recovery does occur it may only be partial.[605, 784] Recovery is possible only after bilateral aiding for many months to several years. Individual case histories suggest that patients who have a rapid onset of deprivation effects also have a more rapid recovery, suggesting that some people have more plastic auditory systems than others, but this finding requires confirmation.[605] A substantial proportion of people simply give up trying to use the second hearing aid because the composite sound is poorer than when using the unilateral hearing aid to which they have become accustomed.[784]

Minimizing the risk of auditory deprivation

1. Prescribe and encourage the use of bilateral hearing aids.
2. If the patient prefers to wear only one hearing aid, encourage the patient to alternate aid use between the ears on a daily or weekly basis.[694].
3. Monitor speech discrimination ability annually for unilaterally aided patients.

All of these methods involve greater cost than a unilateral fitting with no ongoing monitoring.

[r] Even the concept of the relative deprivation of the two ears is overly simplistic: the unaided ear can actually have better speech discrimination when both ears are presented with speech at a low sensation level.[589] Presumably the brain is more used to receiving low-level signals from the unaided ear than from the aided ear, and is therefore better at dealing with them.

Non-speech sounds also reveal differences between the aided and unaided ears. Intensity discrimination is better in the aided ear than in the unaided ear for intense sounds, but is better in the unaided ear than the aided ear for weak sounds.[1519] That is, each ear appears to perform best for the sensation levels that it most commonly receives. For people fitted unilaterally, loudness discomfort levels appear to increase slightly over time in the aided ear, but decrease in the unaided ear.[679] Discomfort levels do not appear to change, however, following bilateral fitting.[679] These findings reinforce the conclusion that the way in which the ear and brain analyze sounds is affected by the nature of the sounds they are used to dealing with.

15.3.5 Suppression of tinnitus

The use of hearing aids can mask, or even suppress, tinnitus. Tinnitus is often bilateral. It is not surprising, therefore, that bilateral hearing aids are more effective than unilateral hearing aids at masking tinnitus.[191, 527] In one study, 66% of people with tinnitus reported that two hearing aids lessened the effects of tinnitus, compared to only 13% who found one hearing aid to be effective.[191] Occasionally, hearing aids can exacerbate tinnitus.

15.3.6 Miscellaneous advantages

One practical advantage to bilateral fittings is particularly important for people with a severe or profound hearing loss: when a hearing aid breaks down, people with two hearing aids still have one working aid.

While a patient can be loaned a hearing aid while the repair takes place, the ***loaner aid***:

- can be provided only for BTE hearing aids, or perhaps for modular ITC hearing aids;
- takes clinical time to fit;
- may have unfamiliar amplification characteristics and controls; and
- is available to the patient only after the patient arrives at the clinic.

A permanent, second, individually fitted hearing aid is a much better option.

Bilateral hearing aids need a little less gain than unilateral aids (see Section 15.8). The gain reduction needed to achieve some criterion loudness is less than the reduction (if any) in SSPL needed to avoid loudness discomfort. Consequently, bilateral hearing aids will be saturated less often and to a lesser degree than unilateral hearing aids, which should confer some sound quality advantages at high input levels. Also, high presentation levels adversely affect speech intelligibility (Sections 10.3.4 and 10.4.6), so the lower gain of a bilateral hearing aid may confer a slight advantage relative to a unilateral aid. Feedback will also be less of a problem if the hearing aid has less gain. All of these advantages of bilateral hearing aids stemming from a lower gain are relatively minor.

People with chronic effusion problems in their ears who own bilateral hearing aids can alternate use between the ears if a hearing aid in their ear canal exacerbates the problem. For such people, the earmold should be as open a style as possible.

Finally, unlike people with a single aid, those with two hearing aids can use two hearing aids, or one hearing aid in whichever ear they choose, in any situation they choose.

15.4 Disadvantages of Bilateral Fittings

15.4.1 Cost

Unless hearing aids are free to patients, the cost of the second hearing aid, and the batteries for it, will be a major disadvantage for many patients. When patients directly bear the costs of hearing aid provision, cost limits the uptake of hearing aids.[955] In service delivery systems in which the hearing aids are free to patients, the additional costs are borne by government or an insurance provider. The additional cost may be a significant issue for the funds provider, who may require clinicians to justify their actions if bilateral hearing aids are provided to a larger proportion of patients than the funds provider considers reasonable.

Expectations as to what is reasonable appear to be rising, in keeping with the growing proportion of people who acquire bilateral hearing aids. (This is, of course, a circular argument.) Fortunately, the trend is also in keeping with the considerable research that indicates additional benefits for bilateral compared to unilateral hearing aids. Irrespective of who pays for the provision of hearing aids, costs must always be balanced against the benefits outlined in Section 15.3.

It is unfortunate that in some delivery systems bilateral hearing aids cost patients twice as much as unilateral hearing aids. It does not take twice the work to fit two hearing aids as it does to fit one: many of the activities are the same whether the patient is provided with one or two hearing aids (assessment, providing information about hearing loss and hearing strategies, and much of the training in how to use the hearing aid or aids). Two hearing aids certainly do not provide twice the benefit of one.[1190] There is therefore no justification for a rehabilitation service incorporating a bilateral fitting to cost twice as much as one incorporating a unilateral fitting.

15.4.2 Binaural interference

An unknown proportion of elderly people have poorer speech discrimination when listening through headphones binaurally (diotically) than when listening monaurally,[33, 59] or when listening in the sound field with binaural hearing aids than when listening with a single hearing aid (with speech and noise sources directly in front or behind).[259, 1881] Either of these results (unless due to random measurement error) means that the signals to the brain from the poorer ear have interfered with the perception of speech that would otherwise have occurred on the basis of listening with just the better ear (or with just the better ear aided).

This counterproductive effect of the second ear is referred to as ***binaural interference***.[821] Some patients who experience binaural interference say that they can wear their hearing aid in the left ear, or the right ear, but not both, whereas others have a clear preference for which ear the hearing aid should be in.[259]

Several case studies displaying binaural interference have been published in which the patients had better speech reception thresholds in noise (SRT_n) with a hearing aid in one ear (more commonly the right ear) than for bilateral fitting or for a hearing aid in the other ear. The patients also had markedly better SRT_n, dichotic speech test scores, and/or electrophysiological responses in the ear with better aided performance than in the other ear.[20, 259, 302, 756, 821, 1879] As binaural middle latency electrophysiological responses were much weaker than the monaural response in the better ear, it appears that stimulation of the poorer ear was actually *suppressing* the response of the auditory system to stimulation in the better ear.[821]

It would be of great interest to know what proportion of patients displaying binaural interference also have marked asymmetry in their unilateral (or headphone-measured) SRT_n and/or dichotic digit scores and/or early, middle and late latency electrophysiological responses.

Binaural interference has been demonstrated in a 5-year old child with symmetrical hearing thresholds but markedly asymmetrical monaural speech intelligibility scores.[1567] This asymmetry was likely the result of auditory deprivation caused by the child wearing only one hearing aid for the previous 3 years. It is possible, but completely unproven, that many cases of binaural interference are caused by asymmetries in speech intelligibility which in turn are caused by auditory deprivation of various origins, including unilateral hearing aid fitting.[s, 821]

Prevalence of binaural interference

Although it has been estimated that perhaps 10% of elderly hearing-impaired people experience binaural interference,[33, 821, 1623] a more recent study indicated that the proportion may be much larger. Walden & Walden (2005) presented speech and noise from a single frontal loudspeaker to 28 sequential patients with mean age 75 (range 50 to 90 years) wearing bilateral and unilateral hearing aids. Relative to wearing two hearing aids the SRT_n values were, on average, 2 dB better when a hearing aid was worn in just the left ear and 3 dB better when worn in just the right ear. The SRT_n for bilateral fitting was, on average, 4 dB worse than the SNR when the hearing was in the better performing ear, and 1 dB worse than the SNR for the *poorer* performing ear. Both in this study,[1881] and in a much earlier study,[817] it appears that the degree of binaural interference increases with age.

These more recent results suggest that, for elderly patients, binaural interference in noise is the norm, and is more common than binaural redundancy. It is

[s] It is interesting to speculate on the relationship between the ear dominance that results from auditory deprivation and the ear dominance commonly found in dichotic speech testing. Even for people with normal hearing in both ears, speech identification ability is not necessarily symmetrical when a different speech stimulus is simultaneously presented to each ear. In this dichotic situation, superior speech identification is more common in the right ear than in the left.[20] The right ear also appears to be more resistant to the effects of auditory deprivation, although confirmation of this is required.[783] The effects and reversibility of unilateral deprivation, and its relationship to the normal differences between the way the left and right hemispheres process verbal and non-verbal signals,[1736, 424] remain exciting areas for future research.

hard to reconcile these results with either the ongoing use of bilateral hearing aids by the majority of people who obtain them (Section 15.7), or with laboratory research showing either a binaural advantage or that unilateral speech intelligibility is equal to bilateral speech intelligibility. Marrone, Mason & Kidd (2008) presented a slightly younger group of research subjects (mean age 70 years) with frontal target speech against a background of two competing talkers, either collocated with the target or separated with one competing talker on each side. In both spatial conditions, the mean scores showed neither a bilateral advantage nor a bilateral disadvantage relative to the unilateral SRT_n values. In quiet, speech reception thresholds were 2 dB better for bilateral fittings than for unilateral fittings (i.e. binaural redundancy).[1143] The following three factors may partly reconcile these findings.

First, the high prevalence of binaural interference found by Walden & Walden was measured with speech and noise spatially co-located, whereas they are usually spatially separate in real life. If a patient has only a small binaural interference effect (e.g. equivalent to a 1 dB decrement in SRT_n), it is likely to be outweighed by the head diffraction benefits that a bilateral fitting offers for spatially separated sources. Conversely, marked binaural interference when the same sounds are presented to both ears is likely to outweigh the SNR advantages caused by head diffraction effects and binaural squelch for spatially separated speech and noise (if the binaural squelch occurs at all for people who experience binaural interference).

Second, the measurements showing pronounced binaural interference were made in four-talker babble. Binaural listening provides greater spatial release from masking when there are fewer competing talkers (i.e. where there are gaps and where masking is more informational than energetic). The dependency of binaural interference on masker level and complexity is so far unknown.

Third, it has long been reported that many patients who normally use two hearing aids do remove one of them (or both!) in noisy places (Section 15.3.1).[1571] While the reason for this has been ascribed to distortion in hearing aids and to not needing bilateral hearing aids to achieve binaural listening when sound levels are high, it seems probable that binaural interference is also be a major contributor to this behavior.

Three causes of binaural interference have been proposed and each supported with some evidence.

Asymmetrical cochlear distortions

A binaural deficit can be simulated in normal-hearing people. When speech is artificially distorted in a different manner for each ear, to simulate different cochlear distortions, binaural speech scores are lower than monaural scores.[760, 1533] Intriguingly, asymmetrical distortion caused binaural interference only for adults, not for children.[1533] Applying the same type of distortion to each ear leaves binaural advantage intact. Subjects reported that when they were listening to the asymmetrically distorted signals, they attempted to attend selectively to the ear with the clearer signal, which supports the concept of binaural interference and deprivation facilitating each other.[760]

One mechanism that might give rise to destructive mixing of information from the two ears is a change in the tuning properties of one cochlea relative to another. Such changes may be caused by a loss of outer hair cell function. Hair cell dysfunction may be exacerbated if the brain's control of the cochlea via efferent nerve fibers is decreased.[1040, 1475, 1562] ***Diplacusis***, in which the person hears a different pitch in each ear, would be one possible psycho-acoustic consequence of differential retuning in the two ears. Diplacusis has indeed been shown to be an indicator of a lack of benefit from bilateral amplification.[1132, 1133]

Differential aging of the hemispheres

A second potential reason for interference derives from asymmetry in the cortex. It has long been suspected that in the elderly, right hemisphere functioning deteriorates faster with age than left hemisphere functioning.[635, 835] Each hemisphere is, of course, connected more strongly to the contralateral cochlea than to the ipsilateral cochlea. Consequently, when the signals originating in each cochlea are combined in either hemisphere, those that originated in the left ear and stimulated the more dysfunctional right hemisphere, may be distorted in some way relative to the neural signals in people with less physiological aging, and the combined signal is less clear than that of either ear alone. Certainly in the large majority of cases where speech intelligibility is markedly different in the two ears, the better performing ear is the right ear.

Inefficient transmission between the hemispheres

A third hypothesis also involves ear asymmetry. Because the left hemisphere is more specialized for speech perception, signals have to pass from the right hemisphere to the left hemisphere before language

Binaural interference in summary

- A proportion of patients will experience binaural interference when presented with identical signals at the two ears.
- This interference will cause many of these patients to prefer, and perform better with, one hearing aid than with two.
- This unilateral advantage may be because: the two cochleae generate neural signals sufficiently different that their outputs interfere with each other; the right hemisphere deteriorates more with age than the left hemisphere; or information is distorted as it transfers between the hemispheres of the cortex.

interpretation can occur. It is suspected that aging reduces the efficiency (causing delays or distortion) in the transfer of information between the hemispheres via the corpus callosum.[302, 637, 815] Note that this hypothesis, like the previous one, is consistent with the right ear advantage that is usually found in elderly people in dichotic speech perception tasks.[835] Evidence that it is disturbed inter-hemispheric transfer that causes interference, rather than reduced functioning in the right hemisphere, is that there are strongly asymmetrical event-related potentials (P_{300}) for right ear versus left ear targets in response to dichotic PB word pairs but that the asymmetry reverses for verbal versus non-verbal targets.

Other phenomena that are consistent with either the differential aging or hemispheric transfer hypotheses include:

- worse phonetically balanced word scores for the left ear than for the right, despite reasonably symmetrical pure tone thresholds and distortion-product otoacoustic emissions;[302]
- symmetrical ABR wave V latencies and amplitudes, indicating that the asymmetry occurs after the brainstem;[302]
- greater acclimatization (i.e. increase in speech discrimination ability) in the right ear than in the left ear following bilateral hearing aid fitting indicating reduced plasticity in either the corpus callosum or the right hemisphere.[1409]

15.4.3 Self-image

Even when hearing aids do not cost the patient any money, many patients will choose to obtain one rather than two hearing aids.[1754] Sometimes patients will say something like *I am not that deaf* to justify their choice of a unilateral fitting. (This was very common when only large BTE hearing aids were available.) Three different beliefs may underlie this statement:

- If patients associate a hearing aid with being old or deaf, they may associate two hearing aids with being very old or very deaf, which conflicts with how they see themselves and probably with reality. In some cases, friends and relatives will reinforce this negative assessment of bilateral hearing aids.[191] People who wear two hearing aids are more likely than those who wear one hearing aid to report that their hearing aids are noticeable to other people.[1754] On the positive side, they are also more likely to report that other people have noticed how much more alert they are when they wear their hearing aids.[1754] This last finding might be valuable to pass on to patients who do not wish to try two hearing aids because of cosmetic concerns.
- Patients may have an overly optimistic assessment of how well they will be able to hear with just one hearing aid.
- Patients with a mild loss may make an accurate assessment that the second hearing aid will not provide them with significant benefit in the short term.

15.4.4 Miscellaneous disadvantages

Wind noise

Even a light breeze can create noise in a hearing aid that is equivalent to the noise created by a 100-dB SPL sound at the *input* of a hearing aid.[458] The only thing worse than a hearing aid amplifying 100 dB SPL of unpleasant noise is two hearing aids amplifying this noise. Consequently, it is not surprising that for windy situations people do not rate bilateral hearing aids any more highly than unilateral hearing aids.[186] It is possible to orient one's head to minimize wind noise at the hearing aid microphone, but it is usually impossible to achieve this for two hearing aids at once. The problem is less for CIC hearing aids than for other hearing aids, but a substantial problem remains.

Aid management

Managing a hearing aid presents a major difficulty for some people with decreased physical or mental abilities. With two hearing aids, there is more scope for confusion with insertion, battery changing, on-off control, and volume control. For bilateral fittings, volume adjustment has two aspects – overall loudness and balance between the two ears. This is inherently more complex than managing a single volume control. Even some able-bodied and able-minded people report difficulty in balancing the two hearing aids during the first week or so of aid use, although the skill may eventually be attained.[527] Linking the gains of the hearing aids by wireless, so that the gain of both hearing aids is simultaneously varied with a single control solves this problem (Section 3.2).

Occlusion effect

If the hearing aid is sufficiently closed to create an occlusion effect (Section 5.3.2), then the occlusion is rated as more disturbing when hearing aids are in both ears than when only one is worn.[824]

15.5 Tests of Bilateral Advantage

There is a long history of failing to demonstrate a significant bilateral advantage in speech identification for any individual, even though there is usually no difficulty in demonstrating benefit for an experimental group as a whole with spatially separated speech and noise.[222, 713, 817, 1063, 1591]

There have been exceptions: Jerger, Darling & Florin (1994) demonstrated a significant bilateral advantage for seven out of ten subjects, using the Cued Listening Test. In this test, patients have to indicate every occurrence of a target word within continuous discourse coming from a loudspeaker on one side of the head, while different continuous discourse comes from a second loudspeaker on the other side of the head. Two important details in this experiment were that all subjects were experienced at using bilateral hearing aids, and that the test used 100 target words per amplification condition. For regular clinical practice, there does not seem to be much point in administering a lengthy test of bilateral advantage on patients who are already successfully using bilateral hearing aids.

There are not yet any tests of binaural functioning that can be given under headphones, prior to hearing aid fitting, to accurately predict whether bilateral amplification is likely to be better, the same, or worse than unilateral amplification for a patient.[164a] The reasons for this include:

- Interactions between the ears may be different if each ear is given a gain-frequency response appropriate to the hearing loss in that ear rather than the flat response that is most easily obtained using an audiometer;
- Headphone testing can most easily create monaural and binaural stimulation. For clinical applications, however, we wish to infer the relative performance of unilateral versus bilateral amplification.
- The nature of binaural interactions may change after some months, weeks, days, or even hours of listening experience with appropriately amplified sound in each ear;

Consequently, the tests to be outlined in this section are tests of bilateral and unilateral aided functioning, for people who have already been provided with two hearing aids, but for whom there is some doubt about the value of bilateral fitting relative to unilateral fitting.

Research should urgently be directed to developing tests that can be used *prior* to fitting to determine if a patient is likely to have adverse, rather than helpful, binaural interactions. There is the prospect of developing such tests in the future. Some contenders, all of which require considerable further research, include:

- The acceptable noise level test (ANL; see Section 9.1.5) may have the potential to predict bilateral or unilateral candidacy. On average, ANL values are the same for binaural listening and monaural listening, but some patients have much poorer ANL values when tested binaurally.[571]
- The Listening in Spatialized Noise Sentences test (LiSN-S[241]) creates a virtual auditory environment under headphones, and has recently been extended to include individual amplification similar to that provided by hearing aids.[628] This amplification can be applied to either ear or to both ears. This combination of features overcomes the problems in the first two bullet points above.
- The masking level difference (MLD) test or the binaural interaction component of the electrophysiological middle latency response, which have been shown to correlate to each other, but which probably only indicate binaural deficits arising in the brainstem or midbrain.[1037]

- Results on a dichotic test such as the dichotic digit test (DDT) or dichotic sentence identification test (DSI) may have some relationship with the ability to use the differing information present at the two ears.[93, 301]

Before giving recommendations for when and how aided testing should be performed (Section 15.5.3), two complicating issues will be discussed.

15.5.1 Bias in choosing the reference ear for the unilateral condition

If speech performance ability with two hearing aids is to be compared to a score obtained with one hearing aid, how should the ear be chosen for the unilateral condition? If we choose on the basis of the audiogram (e.g. the ear with the lesser average loss), on some practical grounds (like the person being right-handed), on a theoretical basis (right-ear advantage is much more common than left-ear advantage, especially in elderly patients),[260, 814, 820, 1881] or just choose randomly, there is a possibility that in fact the other ear has superior speech recognition ability. An apparent bilateral advantage may thus occur just because the bilateral condition includes the other (better) ear. This would cause a systematic bias in favor of bilateral hearing aids.

On the other hand, if both ears are measured in the unilateral condition, and the better of the two scores is chosen as the baseline, there is a systematic bias against bilateral hearing aids.[210, 223] This occurs because all speech scores have a random component, so the higher of the two unilateral scores is likely to be greater than the bilateral score, even when there is in fact no difference between the conditions. If there *is* a true underlying bilateral advantage, this statistical bias towards a unilateral fitting will decrease the chance of the bilateral advantage emerging during any particular test. The extent of the bias is substantial unless the speech test contains many items. For a 25 item test, for example, the bias towards the unilateral condition can be as high as 6%,[223] which is nearly as large as the expected bilateral advantage in some conditions.

Some methods for reducing bias when testing for bilateral advantage are:

1. Average the two unilateral scores and compare this average to the bilateral score.
2. Test the bilateral condition twice and compare the higher of the two scores to the higher of the two unilateral scores.
3. Subtract, from the higher of the unilateral scores, an amount that on average compensates for the bias.[223]
4. Use a large number of test items in each condition (like 100, but this is rarely possible because of time constraints).
5. Use a high-context test with a steep performance-intensity function, so that the true differences are as large as possible compared to the degree of bias (Section 15.5.2).
6. Test the unilateral condition for only one ear, but choose the ear for which the patient thinks speech is clearer, or that some other speech test has already shown to be the better ear.

None of these methods is perfect. When it seems likely that the two ears have identical speech performance ability (based on the audiogram and the patient's opinion), method 2 or 3 should be adopted. When there are strong grounds for believing that one ear is better than the other, method 6 should be adopted. Method 5 should always be adopted, but may not by itself sufficiently decrease bias.

15.5.2 The sensitivity of speech tests for assessing bilateral advantage

Is it possible to reliably test for bilateral advantage with individual patients using speech identification tests? Suppose we performed a speech test based on 50 scored items when the patient was bilaterally aided and when she or he was unilaterally aided. For scores in the range 30 to 70% correct, the standard deviation of test-retest differences will be 10%.[664, 1781] To achieve a significant difference at the 95% confidence level, the bilateral score will have to exceed the unilateral score by 20%. If the speech test comprises isolated words, the ***performance-intensity (P-I) function*** is likely to have a slope around 3% per dB of SNR.

The bilateral advantage will therefore have to be considerably more than 7 dB if it is to be reliably confirmed in the speech test. An advantage this large is possible only if the ear that is unilaterally aided is further from the speech and/or nearer to the noise. Were the unilateral score to be obtained with the hearing aid in the ear that receives the better SNR, the bilateral advantage is most unlikely to be significant.

The situation is more promising for speech tests with high context, and in which the masking noise has a similar spectrum to the speech. For such tests, the P-I function has a slope of at least 10% per dB of intensity or SNR.[952, 1236, 1438] Consequently, a bilateral advantage as small as 2 dB should sometimes be detectable with a list of 50 genuinely independent items.[t] We can hope to detect the presence of binaural squelch only if there are different combinations of signal and noise at the two ears. Such differences can arise only when the speech and noise are spatially separated, in which case the highly predictable head diffraction effects will also be present. If both head diffraction and squelch contribute towards bilateral advantage, we can be confident of an advantage much bigger than 2 dB. If only squelch contributes (i.e. the ear with the better SNR is chosen as the unilateral reference), we could not be so confident.

In summary, it seems that we can easily demonstrate a bilateral advantage for an individual patient if we:

- arrange the test situation so that head diffraction favors the bilateral condition relative to one of the unilateral conditions, and then use this more adverse unilateral condition as the reference unilateral condition; and
- use material with a steep P-I function.[u] Such materials include spondees and high-context sentence tests like the ***Bamford-Kowal-Bench*** (***BKB***) sentences,[94] ***Hearing In Noise Test*** (***HINT***),[1323] the ***Speech In Noise*** (***SIN***) test;[926] equi-intelligible Dutch sentences,[1438] ***Oldenburg (German) Sentences***[960] and matrix sentences (where a small number of words are randomly re-arranged into new sentences) available in several languages.[665, 1870]

The SNR has to be chosen so that scores for both the bilateral and unilateral conditions are obtained from the sloping part of the P-I functions. Because of the steepness of the P-I functions, this can most easily be achieved by adaptively varying the SNR after each sentence. If half, or less than half the words in the sentence are correct, increase the SNR; if more than half are correct, decrease the SNR. Increases in SNR should have the same step size as decreases. The average of the SNRs at four or so reversals provides an estimate of the SNR corresponding to 50% words correct. A greater number of reversals increases the accuracy of the estimate. Use an even number to track the mid-point of the SNRs in an unbiased manner. Alternatively, if only recordings with fixed SNRs are available, it will be necessary to obtain and plot scores for a few SNRs so that the position of the sloping part of the P-I curve can be estimated.

To maximize the chance of detecting bilateral advantage, testing level should be as low as is possible but realistic. The higher the test level used, the greater are the chances that the unaided ear will contribute useful information, so that both the unilateral and bilateral conditions will actually involve binaural listening. A speech level of 55 dB SPL seems like a suitable compromise in that it is lower than the level of typical speech, but high enough nonetheless to be encountered reasonably often in real life.

15.5.3 Role for speech tests in assessing bilateral advantage

Most patients receiving their first hearing aids are able to indicate a preference for bilateral hearing aids over a unilateral aid within the first few hours of trying hearing aids. This initial preference is indicative of their long-term acceptance of bilateral hearing aids.[527]

Furthermore, we have seen that the reliable detection of bilateral advantage requires head diffraction effects to favor the bilateral condition. There is no point in *routinely* testing for the bilateral advantage that arises from the head diffraction component of improved SNR. This advantage will occur, in some real life situations, for every patient who has a head and aidable hearing up to at least 1 kHz! Detection of a bilateral advantage caused in part by head diffraction effects does not indicate that any true binaural interactions are taking place. It does, however, indicate that the patient will gain more benefit from two hearing aids than from a single aid in at least some situations.[210]

There seem to be two reasons, demonstration to skeptics and detection of binaural interference, for measuring bilateral advantage when the circumstances of a particular patient so indicate. For the reasons in the preceding paragraph, it does not seem sensible to use

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[t] For high-context sentences, there will need to be more than 50 words but considerably less than 50 sentences to obtain the same statistical reliability that is obtained for 50 independent words.

[u] The benefit of using sentence material to evaluate hearing aids was pointed out at least 60 years ago.[552] The intrinsic relationship between P-I function slope and bilateral advantage (in percent) may explain the observation that individuals with the steepest P-I functions tend to obtain the greatest bilateral advantage.[1267]

clinical time to perform either of these measurements routinely.

Demonstration to skeptics

A speech test can be used to demonstrate bilateral advantage to people (either the patient or influential friends or relatives) who doubt the benefit of the second aid. The physical arrangement for these tests, as depicted in Figure 15.9, is designed to maximize bilateral advantage by capturing the benefits of head diffraction, squelch, and redundancy, but retain face validity by having speech in the frontal quadrant. An angle of 30º rather than 45º from the front can be used if desired, because binaural advantage increases sharply as angle increases from 0º to 30º from the front.[176]

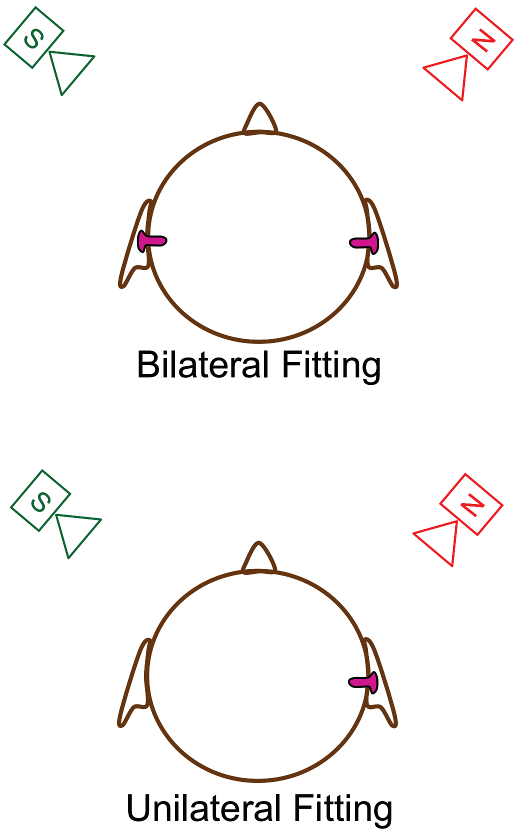

Figure 15.9 Test arrangement for demonstrating bilateral advantage, showing the location of the speech (S) and noise (N) loudspeakers. Speakers should be 0.5 m or more from the patient. For unilateral fittings to the left ear, the S and N sources should be reversed for both the bilateral and unilateral tests.

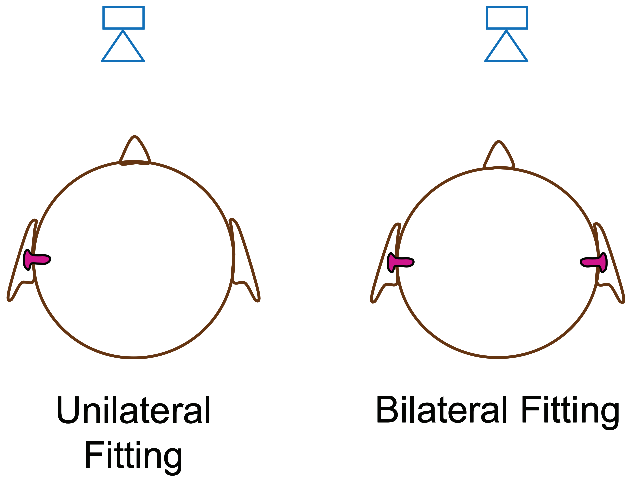

Figure 15.10 Test arrangement for detecting negative binaural interactions. Speech and noise both come from the same loudspeaker.

Detection of binaural interference

A speech test can be used to ensure that the patient is not someone for whom binaural stimulation is worse than monaural stimulation. For this purpose, we should minimize head diffraction effects, as their positive effect on intelligibility may partially cancel the negative binaural interactions that we are aiming to detect. Figure 15.10 shows a suitable test arrangement. The test can be performed to confirm complaints from patients who have tried two hearing aids and consider them worse than one aid. Also, if the test shown in Figure 15.9 failed to show the expected advantage, the test in Figure 15.10 could be used to investigate the reason.

Acclimatization effects

The need to confirm bilateral advantage, or to eliminate adverse binaural interactions, is probably greatest for patients who have become used to unilateral amplification. Unfortunately, the same factors that make initial acceptance of the second aid less likely also make it less likely that objective benefit can be demonstrated initially. The improvements in speech intelligibility that follow training with bilateral hearing aids illustrate the importance of listening experience.[1592]

Changes in the auditory system following hearing aid fitting are even evident at the brainstem level. Philibert et al (2005) observed altered auditory brainstem responses following binaural hearing aid fitting. Changes were greatest for the frequencies and amplitudes most affected by amplification.

15.5.4 Localization tests

Localization is not commonly tested in clinical situations, although such testing is simple to do (see panel). If localization testing is performed to compare one versus two hearing aids, it is important that the stimulus presentation level be as low as possible while still being realistic.[420] Otherwise, the patient may in fact be listening binaurally in both the unilateral and bilateral conditions.

Localization ability is important in its own right but, in principle, can be used to assess whether useful binaural interactions occur for an individual patient.[210] If the binaural interactions necessary for localization can occur, it is possible that the interactions necessary for binaural squelch or binaural redundancy will also be present.[420, 741, 1591] So far, however, localization has been shown to be only weakly related to bilateral advantage in speech identification.[1335, 1591] Self-reported localization ability is, however, correlated with ease of understanding speech, even when hearing loss is controlled for.[1338] In short, the precise relationship between localization and binaural intelligibility benefits is unclear!

15.6 Fitting Asymmetrical Hearing Losses

For a patient with hearing thresholds that are asymmetrical by more than about 30 dB, or speech discrimination scores that are asymmetrical by more than about 20%, the clinician will have to make the following decisions:

- Should a bilateral or unilateral fitting be recommended?
- If a unilateral fitting is recommended, should the better or worse ear (based on either pure tone thresholds or speech discrimination scores) be aided?
- Should an alternative such as some variety of CROS hearing aid, or FM system, be recommended?

These three questions are addressed in the following three sections.

15.6.1 Bilateral versus unilateral fittings for asymmetrical losses

Asymmetrical hearing loss may be defined on the basis of thresholds averaged across frequency, threshold shape, speech intelligibility testing, discomfort levels, or dynamic ranges. There have been many suggestions that patients with hearing losses that are asymmetrical by more than a certain degree on any of these criteria will not benefit from bilateral fittings.[115, 174, 403, 1132] It is understandable why binaural advantage *might* turn into binaural disadvantage for asymmetrical losses: as mentioned in Section 15.4.2, artificial asymmetrical distortion to the signal is sufficient to create binaural interference for normal-hearing people. People with permanent asymmetrical hearing loss may, however, acclimatize to this asymmetry.

It is true that the binaural redundancy component of binaural advantage diminishes as the average hearing

A simple localization test

Ask the patient to point to a (low intensity) noisemaker while wearing a blindfold, or while keeping his or her eyes closed. A correct response would be when the patient points to within approximately 20 degrees of the correct direction.

At least ten presentations should be given in each condition tested (e.g. unaided versus unilateral fitting, or unilateral versus bilateral fitting), to improve the reliability of the results. If each trial is scored as correct or incorrect, the significance of a difference in scores between conditions is assessed in the same way as for speech identification tests.[664, 1781] Test accuracy and sensitivity therefore increase with the number of trials used.

It is advisable to vary the stimulus from trial to trial, but present each stimulus the same number of times in each amplification condition tested. Otherwise, the patient may be able to localize using the spectral shape at the aided ear. This is a monaural cue, so the test results could not then be used to infer anything about binaural interactions or the ability to localize unfamiliar sounds.

thresholds of the two ears become more dissimilar.[713] The decrease is significant for a difference in four-frequency average hearing levels of 15 dB or more.[403, 594, 668]

Binaural squelch, however, occurs even when the poorer ear has thresholds 50 dB worse than the better ear[1794] or when sounds are greatly attenuated in one ear.[176, 1104]

The physical effects of head diffraction on SNR at each ear occur no matter what hearing loss the person has. A bilateral advantage arising from head diffraction should occur whenever the following are all true:

- the ear nearer the speech source and/or further from the noise has enough inherent speech identification ability under ideal conditions;
- the sound arrives from the unaided side in the unilateral reference condition; and,
- the sound is at a level lower than is optimal for the unaided ear.

In some listening conditions, intelligibility has been found to be maximized by either a bilateral fitting or a unilateral fitting to the *worse* ear.[594] In other listening situations, intelligibility has been found to be maximized by either a bilateral fitting or a unilateral fitting to the *better* ear.[594] The only solution common to the physical locations of speech and noise in all situations is therefore a bilateral fitting. Unfortunately, it may be that binaural interference also becomes more likely as speech recognition scores become more asymmetrical.

One survey showed that people with an asymmetric loss (defined according to the shape of the audiogram) were *more* likely than people with symmetric audiograms to use two hearing aids.[308] This is particularly understandable for some asymmetrical hearing loss profiles. For the hearing loss shown in Figure 15.11, for example, the patient has less loss for low-frequency sounds in the left ear, but less loss for high-frequency sounds in the right ear. We know that on average the ability to use audible information decreases as the degree of loss increases (Section 10.3.4). If the person whose (unusual) audiogram is shown in Figure 15.11 is to make maximum use of her residual hearing, it is essential that her left ear receive amplified sound in at least the low-frequency region, and that her right ear receive amplified sound in at least the high-frequency region. The brain is able to combine information at different frequencies sent to it by the two ears.[563] This variation of the better ear from frequency to frequency has been called the ***cross-over effect***.[221] This use of the term *cross-over effect* should not be confused with the transfer of sound from one side of the head to the other by bone conduction that can occur when masking one ear.

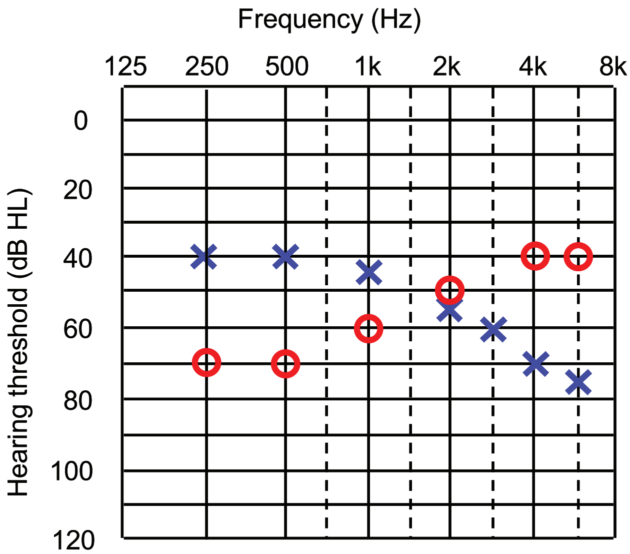

Figure 15.11 An audiogram (crosses for left ear, circles for right ear) for a person who is likely to benefit from the hearing aid cross-over effect if a bilateral fitting is provided.

The cross-over effect is an extreme example of the binaural redundancy advantage that occurs for normally hearing people – the same acoustical information may be presented to the two ears, but the sum of the information sent to the brain is greater than that which can be sent by either ear alone.

It is *possible* that excessive asymmetry in either thresholds or speech intelligibility precludes bilateral advantage, but it is unclear what constitutes "excessive". As an example, a person with a three-frequency average loss of 30 dB HL in the better ear and 80 dB HL in the poorer ear requires, on average, a SNR 8 dB higher in the poorer ear than in the better ear for the same speech intelligibility.[914] As shown in Section 15.2.1, the SNR at one ear, averaged across frequency, can be up to 17 dB better than the SNR at the other. When speech is on the side of the poorer ear and noise is on the side of the better ear, the poorer ear is therefore likely to provide considerably higher

Conclusion: asymmetrical hearing loss and bilateral fittings

If each ear, considered in isolation, has a loss that could effectively be aided then, in at least some situations, the person is likely to benefit from a bilateral fitting, irrespective of the degree of ear asymmetry. The greater the loss in the better ear, the greater will be the range of situations and sound levels for which a bilateral fitting will be better than a unilateral fitting to the worse ear.

intelligibility than the better ear.[v] Thus, even for an asymmetry as great as 50 dB, there is a theoretical basis for aiding both ears.

In short, there is no convincing evidence that an asymmetrical hearing loss precludes a patient from benefiting from a bilateral fitting. Furthermore there is much direct and indirect evidence that people with asymmetrical loss *will* benefit from a bilateral fitting in at least some situations.

15.6.2 Better ear versus poorer ear for unilateral fittings

If the patient prefers to have only one hearing aid, but has a hearing loss in both ears, which ear should you fit? Let us first consider two extreme examples to illustrate the two principles at work. For the audiogram shown in Figure 15.12, the loss in the left ear is so mild that only extremely weak sounds will be inaudible in that ear if the ear remains unaided. A hearing aid in the right ear, however, will improve audibility in that ear for many sounds. This will be valuable to the patient whenever head diffraction creates a better SNR in the right ear than in the left ear. Aiding the right (poorer) ear will also increase the likelihood that binaural squelch and binaural redundancy will operate, because many more sounds will be audible in both ears than if the better ear were to be aided.

The situation is very different for the audiogram in Figure 15.13. The right ear is capable of sending some signals to the brain, but the signal quality is likely to

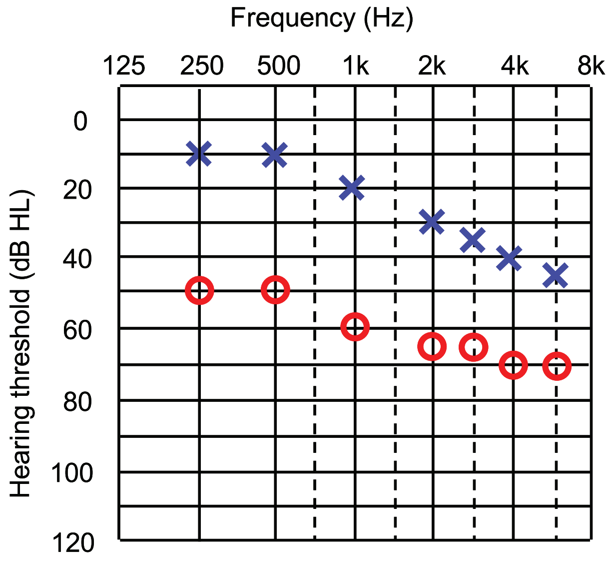

Figure 15.12 An audiogram where the poorer ear should be aided if the person chooses to have a unilateral fitting.

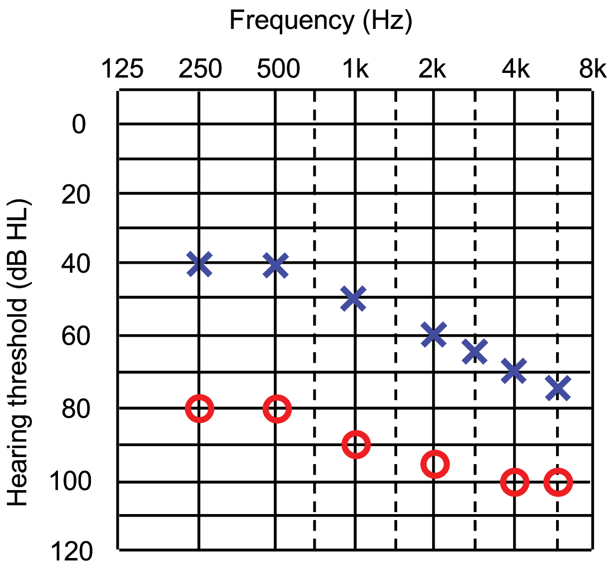

Figure 15.13 An audiogram where the better ear should be aided if the person chooses to have a unilateral fitting.

[v] This analysis overestimates the poorer ear advantage by some amount. A considerable part of the 17 dB advantage arises from high-frequency head diffraction effects. For flat and high-frequency hearing losses, the high-frequency regions contribute relatively little to intelligibility for severe hearing loss, so the poorer ear will not be able to take full advantage of the increase in SNR caused by head diffraction.

be grossly inferior to the quality of the signals sent by the left ear. The left ear can send its higher quality signals only when sounds are audible to it, and hence a hearing aid in the left ear will greatly improve the range of sounds over which it is able to operate. The person will thus be helped more by a hearing aid in the left (better) ear than in the right (worse) ear.

The same three factors are actually operating in the decisions about both audiograms:

1. Aiding the better ear maximizes the range of sounds *audible* to the person.
2. Aiding the poorer ear maximizes the range of sounds that will be audible in both ears. Consequently, aiding the poorer ear will maximize the likelihood of the person being able to use binaural interactions to assist understanding and localization. Depending on the losses in each ear, it may also maximize the likelihood of the sounds being audible at the ear that has the better SNR because of head diffraction effects.
3. The better ear, when aided, is able to send higher quality signals to the brain than the poorer ear, when aided.

For the audiogram shown in Figure 15.12, factor 2 is the most important, whereas for the audiogram shown in Figure 15.13, factor 3 is the most important. Note that for factor 1, "better ear" means the ear with the better pure tone thresholds, whereas for factor 3, "better ear" means the ear with the better aided speech discrimination scores. This is usually, but not necessarily, the ear with the better pure tone thresholds. In factor 2, both pure tone thresholds and speech discrimination ability are involved in the definition of the better ear.

For hearing losses where the decision about which ear to aid is less straightforward than these two examples, the same three factors operate. In many cases, it will not be obvious which of the factors should take precedence. Such cases are difficult because, in reality, all three factors are important, and the best option is actually to fit both ears unless there is evidence of binaural interference.

Swan, Browning & Gatehouse (1987) carried out an extensive study into the preferred side of fitting. They concluded that, on the basis of subjects' real-life experience over 20 weeks (comprising 10 weeks with hearing aids on each side), there was a strong overall preference for fitting the poorer ear. The poorer ear was preferred whether "poorer" was defined in terms of audiometric criteria or speech discrimination criteria.

In terms of audiometric thresholds, the favored ear was the one with the poorer 4FA thresholds and/or with the higher degree of slope. In terms of speech recognition ability, the poorer ear was the one with the greater half-peak level elevation (HPLE)[w] and/or with the lesser maximum discrimination score.

Counseling suggestion

When a patient wants only one hearing aid, resulting in a difficult decision as to which ear should be aided, advise the patient how a hearing aid in each ear will be advantageous:

- The better ear should be aided because you will then be able to hear a wider range of sounds, and you will be better able to understand speech when it comes from that side of the head.
- The poorer ear should be aided because otherwise speech will be **very** unclear when it comes from your poorer side. Furthermore, aiding that ear will help prevent further deterioration in the speech recognition ability of the poorer ear.

Hopefully, the patient will let you off the hook by agreeing to try hearing aids in *both* ears, which is the only way to meet the requirements of all three decision-making factors listed in the text. One possibility is to offer the second hearing aid at no charge for the first 30 days, so that the patient can assess the extra benefit before making any financial commitment to the second aid.[686]

[w] The HPLE is the level of speech needed to produce a score equal to half that of the maximum score achievable at any level.

Of those subjects who had a preference for hearing-related reasons, the speech-based criteria correctly predicted preferences for 87% of subjects, whereas the audiometric criteria correctly predicted preferences for 77% of subjects. Note that all subjects had 4FA thresholds less than 75 dB HL in both ears. The experimental finding should not be applied to patients with losses greater than this in the poorer ear, as in the example shown in Figure 15.13.

Swan et al. (1987) suggest that the reason most subjects preferred the fitting in their poorer ear is because of the large disadvantage they would suffer when speech arrives from the poorer side if they are aided only on the better side. Follow-up studies supported this suggestion.[1758, 1759] For frontal speech and noise, and for speech from the better side, people understand slightly more when the aid is in the better ear. When the speech comes from the poorer side, however, people understand *much* less when the aid is in the better ear than when it is in the worse ear. People may thus prefer a fitting on the worse side because it minimizes their disability in the most adverse situations they encounter, even if it is not optimal in other, easier situations.

The possibility of past and future auditory deprivation provides reasons why the poorer ear should be aided. Patients with asymmetrical pure tone thresholds usually also have poorer speech discrimination in the ear with poorer thresholds. As time progresses, the speech discrimination ability of the poorer ear deteriorates (because it is given less attention by the auditory system, as discussed in Section 15.3.4) even if the pure tone thresholds remain fairly constant.[1628]

Consequently, the poor speech intelligibility of the poorer ear at the time the patient is seen is partly a result of the damage to the cochlea, and partly a result of being less attended to by the auditory system since the asymmetrical hearing loss developed. The current speech intelligibility of the poorer ear is therefore a pessimistic estimate of its future ability. If the poorer ear is aided, its speech intelligibility ability is likely to improve in the year or so after aiding.[1628]

Conversely, if the poorer ear is not aided, its speech intelligibility ability may further deteriorate in the next few years.[1628] A hearing aid in the poorer ear appears to fully protect that ear against further auditory deprivation.[1628] Indeed, it seems possible (but has not been investigated) that fitting the better ear may exacerbate auditory deprivation effects in the poorer ear by increasing the asymmetry. Of course, fitting only the poorer ear may not be best for the patient in the short term, and the patient should be made aware of this.

Finally, it is worth noting that factors other than the inherent speech discrimination capacity of each ear can influence the ear of choice for unilateral fittings, whether the hearing loss is symmetrical or asymmetrical, as shown in the panel in Section 15.7.

Which ear to fit: a simple practical rule

Fit the ear that has the four-frequency average (4FA) threshold closer to 60 dB HL.

This rule is simple, practical, consistent with the three decision-making factors listed in the text, and appears to be a good description of how clinicians in the Netherlands make decisions on which ear to fit.[165] Unfortunately, none of these admirable features constitutes proof that it is an optimal rule.

15.6.3 Alternatives: FM and CROS

Several options should be considered for people with markedly asymmetrical hearing loss, including those with unilateral loss. Because a major symptom of the loss of full binaural function is the need for a SNR higher than that required by normal hearers, any hearing aid fitting that improves SNR is particularly worthwhile.

Chief amongst these is an FM or other type of wireless connection to a remote microphone near the source (Section 3.4). Directional microphones are another alternative. CROS hearing aids are a third alternative (see Section 17.1), and these can incorporate a directional microphone if desired. Not surprisingly, a wireless connection produces much better speech performance than either of the alternatives.[699, 893, 1815] A directional microphone is more effective than a CROS aid, but these solutions are not mutually exclusive.

15.7 Deciding on Bilateral versus Unilateral Fittings

How should a clinician approach the decision of whether to fit bilaterally or unilaterally? The research evidence suggests that in some listening situations (when speech and noise come from different directions), most hearing-impaired people will be better off with two hearing aids than with one. However, it also suggests that in at least some situations (when speech and noise both come from the same direction) a significant proportion of elderly hearing-impaired patients will be better off with one hearing aid than with two. If binaural interference is sufficiently strong, then there may be a much wider range of situations in which a unilateral fitting is best, but the relationship of bilateral advantage/disadvantage to spatial location of sounds for elderly hearing-impaired people has not been researched sufficiently to give guidance on this issue.

Overall, use of both hearing aids has been well researched. The proportion of those fitted bilaterally who preferred and/or continued to use two hearing aids varies markedly across studies, for example: 32%,[1571] 54%,[354] 55%,[1715] 66%,[941] 70%,[191] 76%,[447] 78%,[441, 840] 85%,[222] 90%,[527] and 93%.[164a] The size of these studies varied from 25 patients[1571] to 2127 patients.[441] The three studies substantially larger than the rest,[441, 447, 840] all based on follow-up of patients seen in routine clinical practice, concluded that 78%, 76% and 78% respectively of patients fitted bilaterally usually wore both hearing aids.

A range of reasons for preferring bilateral fittings are reported by patients:

- communication and clarity in noise and in competing sounds;[191, 354, 527, 941]
- communication in quiet places;[1571]
- sound quality;[165, 941]
- localization ability;[165, 527, 941, 1715]
- tinnitus;[191, 527] and
- hearing balance and comfort.[165, 354, 527]

Noble (2006) gives a more detailed review of studies that have evaluated preferences for unilateral and bilateral fittings in real life. Noble comments that improved listening in noise is not a prominent reason for preferring bilateral fittings, but rather localization and "overall better hearing".

Patients who choose a unilateral fitting typically cite convenience, comfort, sound quality, wind noise, self-consciousness, the other ear being too good or too bad, hearing their own voice, speech clarity, keeping one ear open for telephone use, and one hearing aid being sufficient, as reasons for choosing a single hearing aid. Several of the studies giving rise to these statistics are decades old and used linear, peak-clipping, inflexible hearing aids. We would expect the proportion that continues to use bilateral hearing aids would increase with the increasing sophistication (bandwidth, fidelity, compression, directionality, openness) of the hearing aid. Intriguingly, some of the studies with low bilateral take-up used 21st Century hearing aids, so rejection of a second hearing aid is certainly not something we can write off as a consequence of old technology.

We might expect the proportion of patients continuing to use two hearing aids to decrease with the proportion who actually receive bilateral fittings. (When only a relatively small proportion of patients receive two hearing aids, it is likely that they have self-selected to be those most highly motivated to wear two hearing aids, and a high bilateral wearing rate is to be expected. Conversely, when almost all patients leave the clinic with two hearing aids, there are bound to be many among them who do not really want to wear two, and they will ensure that they don't.)

Although one might expect that a recommendation for bilateral versus unilateral fitting could be based on the audiometric profile, this is true only in the extreme cases of normal hearing or profound hearing loss in one of the ears. Studies that have attempted to find differences between the audiometric profiles of those who choose a bilateral fitting compared to those who choose a unilateral fitting have been largely unsuccessful,[165, 354, 527, 939, 1756] although on average those who choose a bilateral fitting have very slightly more hearing loss and report more disability.[441, 1715]

In one large-scale study, of those who initially chose bilateral fittings, those who continued to wear two hearing aids had hearing thresholds only 1 dB greater (on average) than those who ceased using one of the hearing aids.[441] Asymmetry of hearing loss has also not emerged as a significant factor.[165, 1715, 1756] The inadequacy of hearing thresholds to determine bilateral candidacy parallels the considerations of who is a candidate for hearing aid fitting (Chapter 9).

Factors other than hearing thresholds apparently influence the choice to a greater degree than does the audiometric profile:

- Some patients will (inappropriately) equate two hearing aids with severe deafness or very old age (*I'm not that deaf/old!)* and prefer one aid over two on that basis.

- Age is also a factor: those over 75 years of age are less likely to accept bilateral hearing aids than those under 75 years.[1756] This parallels a reduced ability of older listeners to make use of binaural cues in some laboratory tests.[480]
- Patients who have better functioning central binaural processing, as evidenced by greater binaural summation of loudness, higher scores and greater right-ear advantage on dichotic speech tests, and better spatial perception (based on the spatial sub-scale of the SSQ questionnaire) are more likely to prefer two hearing aids.[354, 939]
- Patients who report the greatest difficulties unaided are most likely to prefer two hearing aids.[354]

Unfortunately, no measures have yet been found that can reliably predict the patient's choice. Because the final choice rests with the patient, let us look at the choice from the patient's perspective: what information does the patient need to make the decision that is best for him or her?

First, the patient would like to know which configuration will enable him or her to hear better. Except for a very small proportion of people with an extremely asymmetrical loss (Section 15.6.1) and for a probably small but actually unknown proportion of people who have sufficient binaural interference to outweigh the advantages that head diffraction provides (Section 15.4.2), this will be a bilateral fitting.

Second, the patient would like to know how *much* additional benefit will be obtained if he or she chooses to have two hearing aids rather than one. This is a much harder question because we do not know the answer with any precision. In general terms, we can say that:

- Most patients with a moderate or severe hearing loss in both ears will gain *substantially* more benefit from two hearing aids than from one.[185, 308, 408] The advantage in speech understanding, averaged across listening situations, may be equivalent to a 5 dB improvement in SNR if modeling is correct.[1966] This increase is enough to make the difference between understanding very little of a conversation compared to understanding most of a conversation. Bilateral advantage will be even greater than this when wanted sounds arrive from the unaided side in a unilateral fitting, provided the patient has aidable high-frequency hearing. Conversely, bilateral advantage will be less when wanted sounds arrive from the aided side in a unilateral fitting. Asymmetry in hearing will decrease the degree of bilateral advantage, but some small advantage may remain even when thresholds are asymmetrical by 50 dB. Bilateral hearing aids will also significantly increase the patient's ability to localize. This will be valuable in itself and will also indirectly increase speech understanding by helping the patient locate the person speaking in a group conversation.
- A patient with a bilateral loss that is mild in at least one ear will gain only a small and perhaps un-measurable benefit from two hearing aids compared to one. Bilateral advantage for intelligibility may be noticeable only in situations where the signal of interest is very soft and arrives from the unaided side of the head.
- A patient with a hearing loss between these extremes will experience a degree of benefit somewhere in between. The benefit is most likely to be found in complex, dynamic environments where there are multiple talkers or other sound sources, where attention frequently has to be switched, and/or where some of the sources are moving. Benefit is least likely to be found when there is a single talker in a noisy place, or for localization.[1336] In dynamic situations, the second hearing aid is likely to enable the aid wearer to cope with reduced listening effort.[1336]

Many patients will understandably have trouble deciding with certainty whether to acquire one or two hearing aids. A trial with two hearing aids for some weeks is extremely helpful. The following information may also be helpful.

- *There is a reasonable expectation of eventual superiority for two hearing aids.* Even in 1985, the vast majority of patients who tried bilateral hearing aids chose to wear both of them in at least some situations.[222] Averaged across a typical clinical patient caseload, and using the Glasgow Hearing Aid Benefit Profile, higher benefit is measured for bilateral fittings than for unilateral fittings, even after correction for age and hearing level differences between the groups.[x, 1654] This

[x] These data were obtained with instant-fit, non-custom earmolds, most of which were open fittings. The impact of open fittings and non-custom molds on the bilateral advantage is unknown.

finding does not necessarily mean that those patients who chose to receive one hearing aid would get greater benefit from two hearing aids.

- *Changing from unilateral to bilateral fitting is sometimes difficult, but sometimes not.* Many experienced unilateral aid wearers who acquire a second hearing aid consider that it takes them some months to adapt to the second aid, although most eventually find it to be very helpful.[191] They rate their listening ability in a wide range of situations much more highly than when they wear only one aid.[1134]
- *Technological advances may affect the preference for bilateral versus unilateral fittings*. Co-ordinated, wireless-linked processing of the two hearing aids may have already marginally improved bilateral fittings. Direction-dependent gain in the two ears may further improve them, and super-directivity will very likely further improve them (Section 7.1.4).

Where both ears are potentially aidable, and where the aided ear will receive sounds with a sensation level higher than that of the unaided ear, the clinician should advise the patient that speech recognition ability (but not pure tone thresholds) in the unaided ear may deteriorate if not aided. This concern should be expressed more strongly for a moderate or severe hearing loss than for a mild loss. The patient should be further advised that there is a reasonable possibility that the unaided ear will not recover any lost ability even if a second hearing aid is subsequently acquired.

Should two hearing aids be provided at the same time or sequentially?

A common strategy is to initially provide patients with one hearing aid, and subsequently (some weeks, or months later) fit a second one.[1849] This sequential approach has advantages:

- Patients can make a more gradual commitment to owning hearing aids, which can be useful if they initially associate two hearing aids with being very deaf or very old, or are unsure if the expense will be justified.
- Patients can acquire the necessary manipulation skills on one hearing aid at a time, which some patients may find less daunting.

The sequential approach also has disadvantages:

- The patient has to undergo two successive periods of adjustment. In the first period, the brain may learn to partially ignore the unaided ear (see Section 15.3.4), particularly if the loss is more than mild. For this reason, the unilateral period should be as short as possible, and preferably less than six months. In the second period, the brain must learn to use the information provided by the newly fitted ear, and to properly combine the signals coming from both ears.
- The number of visits and time taken to fit two hearing aids sequentially are greater than if they are fitted simultaneously, thus increasing the expense.
- The initial sensation of hearing being unbalanced between the two ears may induce rejection of hearing aids altogether.

There are no data on which approach is more effective, but the arguments for fitting the hearing aids simultaneously seem more compelling. When two hearing aids are fitted at the outset, it usually takes only a few hours or a few days for patients who are going to benefit from bilateral hearing aids to appreciate their advantages over a unilateral aid.[527]

Sequential fitting can be reserved for patients who are unwilling to try two hearing aids when they are first aided. There is no doubt that patients fitted sequentially can, and usually do, become successful users of bilateral hearing aids.[727]

If a patient rejects a second hearing aid, the clinician should document that it was recommended, the basis of the recommendation, and that the recommendation was rejected.

Certainly when patients have tried bilateral hearing aids and subsequently report that they hear better in noise when wearing only one, we should accept their view as valid. Such self-reports have readily been confirmed by measuring intelligibility in noise in the clinic for the bilateral fitting and the unilateral fitting on the preferred side.[259]

The provision of bilateral hearing aids is extremely important for hearing-impaired people with severe vision problems. Even small improvements in localization are likely to be extremely important because of the increased importance to such people of auditory perception.

Similar bilateral considerations also apply to cochlear implants and to bone-anchored hearing aids[1839] despite the much smaller interaural attenuation that exists for bone-conducted stimuli. Amazingly, the decision about unilateral versus bilateral fitting has to be made even for tactile aids. Bilateral vibrotactile aids, with head mounted microphones, enable localization of competing talkers, and are anecdotally reported to result in a greater externalization of sound than occurs for a single tactile aid.[1497]

15.8 Effect of Bilateral versus Unilateral Fitting on Electroacoustic Prescriptions

Because of binaural loudness summation (see Section 15.2.4), we would expect that bilaterally aided people will use less gain than unilaterally aided people with the same hearing loss. The gain difference should be about 4 dB for mid-level inputs. Because the extent of binaural loudness summation depends on level, the gain difference should be slightly greater for high input levels and slightly less for low input levels. This implies that a very slightly higher compression ratio is optimal for bilateral fittings. At any input level, the gain difference between bilateral and unilateral prescriptions should decrease as the degree of asymmetry increases. Corrections following these principles are included in the NAL-NL2 prescription procedure.

LDL appears to be only slightly lower, if at all, when people listen binaurally than when they listen monaurally (Section 15.2.4). From the perspective of avoiding discomfort, we would therefore not adjust SSPL differently for a bilateral fitting compared to a unilateral fitting.[696] Of course, discomfort is not the only thing that determines the target SSPL. We also wish to avoid excessive saturation in the hearing aid. Because we can prescribe about 4 dB less gain for mid-level inputs for a bilateral fitting, we can there-

Which ear to fit?

If the patient prefers to have a single hearing aid, then in addition to considering the symmetry of hearing thresholds, the following questions should be asked before determining which ear to fit:

- For which hand does the patient have the better dexterity?
- Which ear displays the higher speech discrimination in quiet, better speech reception threshold in noise, or significant advantage on a dichotic speech test?
- Are there complications (chronic otitis externa, suppurating otitis media, exostoses) with either ear canal?
- Is the patient routinely in a situation where the talker is always on the same side (most commonly this is as a passenger or driver in a car)?
- Does the patient prefer to use the telephone in one ear, and can he or she communicate on the phone unaided?

If none of the information available suggests which ear to fit, then choose the right ear, because of the right ear advantages found on many types of speech test.[20, 260, 814, 820, 1881] fore prescribe SSPL to be 4 dB lower than for a unilateral fitting, without changing the degree of saturation. Alternatively, we can leave SSPL unchanged, and consequently, the hearing aid will be less saturated for mid-level inputs.

Looked at differently, a bilateral fitting makes the selection of the correct SSPL less critical than it is for a unilateral fitting. This is particularly valuable for people with a severe or profound hearing loss where it can be difficult to achieve an SSPL low enough to avoid discomfort but high enough to avoid saturation. This is an indirect benefit of bilateral fittings.

It is theoretically possible that the optimum amplification characteristics for one ear depend on what is provided to the other ear. There are however, no compelling reasons why this should be true, and an experimental investigation of it failed to provide any evidence that it is true.[1469]

Overall, it seems reasonable to adjust each aid in isolation to obtain the best performance. It seems reasonable to allow for binaural effects by making sure that loudness at the two ears is balanced, and that overall loudness is acceptable. This is likely to require less gain for a bilateral fitting than for a unilateral fitting, especially at high input levels. There does not seem to be a strong case for having different SSPL values for bilateral fittings than for unilateral fittings, however.

15.9 Concluding Comments

Despite many years of research, there is much about bilateral hearing aid fitting that we do not yet understand. Part of the problem is that we do not have an adequate understanding of how the auditory system performs the binaural processing that it does. Further, a bilateral fitting does not necessarily imply that sounds will be above threshold in both ears, or that useful binaural interactions must occur. Similarly, many of the advantages of binaural hearing can occur with a unilateral fitting if the sound is sufficiently intense or the hearing loss is sufficiently mild. Head diffraction effects, however, are a reliable source of bilateral advantage in many situations.

Generally, therefore, the clinician should start from the assumption that two hearing aids will be more appropriate and look for reasons (of which there are several) why this may not be true for the individual patient, rather than the other way around. This is a conclusion that has also been reached by others who have surveyed the available evidence.[756, 904]

As a balance to this assumption of *bilateral unless indicated otherwise*, the advantage of bilateral over unilateral fittings for patients with mild and perhaps moderate hearing loss (i.e. most patients) seems to be, well, *mild*, and in a small proportion of cases, negative. Information to the client should convey this balanced message – a probable small advantage of bilateral over unilateral hearing aids, with the likelihood of an advantage increasing with the degree of the loss.

It is frustrating that after all this time we do not have any reliable method for predicting which individual patients would achieve greater speech clarity (averaged across a range of listening situations) with one versus two hearing aids, despite some serious attempts at devising such a method.[165] To start with, we need data showing what proportion of elderly hearing-impaired people shows each of three possible outcomes: binaural redundancy, no binaural interaction, and binaural interference (when measured with speech and noise both originating from the front). We then need data showing what proportion of patients has binaural squelch ability (when measured with frontal speech and symmetrical noise so that neither ear is presented with a better SNR than the other). We need to know the extent to which one of these abilities predicts the other. We need a test that measures these abilities and can conveniently be administered during the assessment appointment. Finally we need a study that shows the relationship between the predictive test outcomes and the relative real-world benefit of bilateral versus unilateral amplification, and how other non-speech factors have influenced the configuration that the patient considers is best.

Hopefully, a clinical tool with an appropriate evidence base will also remove the gulf that exists in some countries between beliefs about the high importance of bilateral hearing aids and actual low bilateral fitting rates.[498] It will also reduce the gulf that exists in other countries where there is a high bilateral *fitting* rate, but a much lower bilateral *wearing* rate.

Most of the research into localization benefits of bilateral versus unilateral hearing aids precedes both digital hearing aids and completely open fittings. Both of these developments are likely to impact on the outcome. Although there is no research on the issue, it is hard to imagine how patients could accurately localize with a full-bandwidth (i.e. closed canal) unilateral digital hearing aid. Such a hearing aid introduces a delay of several milliseconds to one ear, which must make it extremely difficult to use interaural time differences that are small fractions of a millisecond. If the fitting is open, however, then because both localization and binaural squelch principally rely on low-frequency interaural time difference, a bilateral fitting may not enable either of these binaural processing mechanisms any more than does a unilateral fitting.

Finally, when linked binaural hearing aids incorporating super-directional microphones become available, they will introduce an entirely new reason for fitting bilateral hearing aids: markedly improved listening ability in noise. For patients with binaural interference, it may even be worth wearing two hearing aids to obtain all the advantages in the microphone outputs that head diffraction can enable, but receive amplified sound in only one ear. Irrespective of the fitting configuration, research to measure the real-world benefits and disadvantages of bilateral fittings will need to start again. Researchers will have jobs for at least as long as clinicians do!

CHAPTER 16

SPECIAL HEARING AID ISSUES FOR CHILDREN

Synopsis

When a child is born with a hearing loss, early provision of hearing aids is essential if he or she is to learn to speak and listen with the greatest possible proficiency. Hearing aids should be provided by six months of age. If cochlear implants are a better option, these should be implanted by 12 months of age. Children with bilateral loss should receive bilateral hearing aids. There is some uncertainty over optimal treatment for children with unilateral loss, mild loss, or auditory neuropathy.

For the hearing aids to be optimally adjusted, frequency-specific hearing thresholds must be determined separately for each ear. No matter what type of transducer is used, the small size of a baby's ear complicates the interpretation of hearing threshold. This difficulty is overcome either by expressing threshold in dB SPL in the ear canal, or by expressing it as equivalent adult hearing threshold in dB HL.

BTE hearing aids are most likely to be provided, in conjunction with soft earmolds, until the child is at least 8 years old (and possibly much older). The hearing aid should contain features that will enable the child to receive the best possible signal. This is likely to include an audio input socket and/or telecoil, and/or internal wireless receiver, so that there is some means to receive wireless transmission. Ideally, the wireless device should be able to automatically attenuate the local microphone whenever the person wearing the transmitter talks.

To communicate effectively, normal-hearing children learning language need a better signal-to-noise ratio than do adults. They also understand speech less well than adults at very low sensation levels. These observations may lie behind the empirical finding that hearing-impaired children prefer more gain than adults with the same hearing loss. Compared to adults, they almost certainly do not need any more real-ear gain for high-level sounds, they probably prefer more gain for medium level sounds, and they almost certainly need more gain for low-level sounds. There is an even greater need for wide dynamic range compression in hearing aids for children too young to manipulate the volume control than there is for adults. Similarly, infants have an even greater need than adults for directional microphones and adaptive noise reduction systems. These algorithms also have potential disadvantages, but should be provided if the clinician has confidence in the automated manner in which the hearing aid selects them.

To achieve a certain real-ear gain, young children need less coupler gain than do adults, because children have smaller ear canals. An efficient way to allow for small ear canals is to measure real-ear to coupler difference before prescribing the hearing aid, and to calculate the coupler gain that will result in the target rear ear aided gain. A faster but less accurate way is to use age-appropriate values of real-ear to coupler difference. The maximum output that has been prescribed should be evaluated by observing the child when intense sounds are made and, for those over approximately six years of age, by assessing the loudness of these sounds.

Hearing aid fittings can be evaluated by speech testing (for those over three years of age), paired-comparison preference testing (for those over six years of age), and subjective reporting by the child, the parents, or the teachers (whether informally or using systematic methods like PEACH and MAIS). The audibility of speech can be estimated by calculating the articulation index (also known as speech intelligibility index) or assessed by measuring the presence, latency and perhaps morphology of the cortical responses elicited by speech sounds. The availability of speech to the child can be indirectly assessed by measuring the child's language development.

Effective amplification for young children is not possible without the support and understanding of parents. The audiologist must therefore inform and support the parents in a variety of ways. One way to provide ongoing habilitation is to base the service activities around goals determined jointly by the audiologist and the parents (and by the child when old enough).

Part of the information provided to parents includes safety aspects of amplification and hearing loss. Hazards include battery, earmold or hearing aid ingestion, excessive exposure to noise, physical impact, and failure to detect warning signals if amplification is not functioning correctly.

This chapter gives an overview of amplification for children, and particularly for infants. One chapter cannot do justice to the importance of getting fully functioning hearing aids on a child as early as possible. Much of the information about hearing aids in the other chapters, however, is also relevant to children. For extensive additional information specific to hearing-impaired children, the reader should refer to Seewald & Tharpe (2011).

16.1 Sensory Experience, Sensory Deprivation, and Candidacy for Hearing Aids

There are two reasons why it is important for a child to be fitted with hearing aids as early as possible. The first is to start improving the quality of life of the child and family. A year without hearing aids is a year in which the child has not been able to enjoy those interactions that require good hearing. The second reason is that early sound deprivation has permanent effects. Neural connections in the brain that allow speech to be understood are formed, based on the signals they receive from the cochlea.[986] Although neural connections can form or disappear at any stage of life,[655] these connections are most easily formed during the early years of life. The brain's opportunity to form connections during the first two or three years of life, and especially during the first six to twelve months, must not be missed if the child is to have the best possible auditory perception for the rest of his or her life.[294, 1200, 1941]

Providing habilitation early in life also maximizes expressive language ability.[1477] Children who receive amplification prior to six months of age also develop clearer speech than those who first receive amplification when they are older.[1135, 1941] In short, no age is too young to provide amplification, once it is clear that the child has a hearing loss, and once the degree of loss can be estimated.

The key ingredients for ensuring early and effective habilitation are:

- a universal newborn hearing screening system, preferably based on auditory brainstem response (ABR) testing for all children, or if based on otoacoustic emissions, then when all babies with neo-natal risk factors receive ABR screening;[a]
- seamless transfer to diagnostic testing;
- seamless transfer to habilitation services, resulting in hearing aid fitting and early educational intervention before 6 months of age (preferably earlier);
- implantation with at least one cochlear implant by 12 months of age for those infants for whom one or two cochlear implants are considered likely to provide speech perception superior to that which hearing aids alone can provide.

Delays in these steps, or instances of children with hearing loss falling out of the system between steps, frequently happen,[389, 1268, 1384, 1683] but they need not.[1821] One of the reasons for apparent failure to provide audiological services is the presence of unilateral hearing loss at the time of initial diagnosis.[1683] The need for habilitation during the pre-school years for children with unilateral loss is not yet known (see later).

16.1.1 Binaural stimulation

Early stimulation should be binaural. Some parts of the ascending auditory pathway (e.g. the superior olive and the inferior colliculus) combine and compare signals from the two cochleae, presumably to perform functions like localization and the binaural

Speech tests for babies

Although speech discrimination tests are commonly given only to older children, even babies and young children can do speech tests. Visual Reinforcement Speech-Sound Discrimination[986] or a variation of play audiometry[407a] can be used to determine whether they can distinguish between the different sounds of speech. It may be possible to apply these techniques, or one of the other techniques reviewed by Eisenberg et al (2005), to determine the effectiveness of amplification in general, and to detect any cases where bilateral amplification should not be provided.

[a] Screening systems in which the primary screen is based on OAE will miss many children with auditory neuropathy. Children with auditory neuropathy comprise around 10 to 15% of those born with hearing loss of moderate or greater degree, and around 7% of older children with hearing loss in the inner ear or auditory nerve.[1465, 1478, 1593a]

suppression of noise. These parts of the neural system can do their job, and probably learn how to do their job, only if both cochleae are sending out signals. For this reason, binaural stimulation during the first few years of life, whether provided by hearing aids, cochlear implants, or a combination of the two, appears to be essential for the neural development that enables binaural processing of sounds.[92, 1840] Not surprisingly, early hearing aid fitting increases the likelihood that children will become successful users of bilateral hearing aids when they are older.[1154] The advantages of bilateral hearing aids covered in Chapter 15 should apply to children just as much as to adults.

There may, however, be a minority of children who have such differences between the way each ear processes sound that better performance is obtained with a single hearing aid in the better ear only (Sections 15.4.2 and 15.6.1). How can this dilemma be resolved? The most conservative option is to aid both ears and continue to encourage use in both ears until it is clear that this is counterproductive. Evidence for withdrawing amplification from one ear would include:

- consistent and prolonged rejection of one hearing aid by the child after the clinician has made every effort to fine-tune the fitting for earmold comfort and loudness comfort;
- reports from the parent that the child functions better with one hearing aid during trial periods of a few days with only one hearing aid (see Section 16.6.4 for some subjective report tools to assist with this evaluation); or
- poorer speech test results when fitted bilaterally than when fitted unilaterally.

16.1.2 Unilateral loss

The effects of a unilateral hearing loss on a person's life are certainly not as pronounced as those of a bilateral loss. Several studies investigating those effects are summarized in Table 16.1, and it is clear that the conclusions are far from unanimous.

Table 16.1 Impact of unilateral loss. Red crosses show outcomes that were adversely affect by unilateral loss. Green check marks show outcomes that were not adversely affected.

Study	Participants	n	Recruitment	Speech understanding	Language development	Educational attainment	Psycho-social development	Vocational outcomes
Giolas & Wark (1967)	> 14 yo	20	Clinical	x			x	
Keller & Bundy (1980)	avg 12 yo	63	Population			x		
Stein (1983)	5-12 yo	19	Clinical		x	√	x	
Klee & Davis-Dansky (1986)	6-13 yo	25	Clinical		√	x		
Bess & Tharpe (1986)	6-18 yo	60	Clinical			x	x	
Bess et al (1986)	6-13 yo	25	Clinical	x				
Culbertson & Gilbert (1986)	6-13 yo	25	Clinical		x	x	x	
Colletti et al (1988)	30-55 yo	61	Clinical	x		√	√	√
Bovo et al (1988)	6-18 yo	30	Clinical	x		x	x	
Jensen et al (1989)	10-16 yo	30	Clinical	x				
Brookhouser et al (1991)	<19 yo	172	Clinical			x	x	
Ito (1998)	Uni students	305	Population			√		
Bess et al (1998)	8-15 yo	37	Population			x	x	
Kiese-Himmel (2002)	1-10 yo	31	Clinical		√			
Lieu et al (2010)	6-12 yo	148	Population		x	x		

The simple question *Does unilateral hearing loss matter?* does not have a simple answer for several reasons:

- The impact of unilateral loss probably depends on what outcome is measured. Unilateral loss certainly adversely affects the ease with which speech is understood in noise and reverberation. This increased difficulty may or may not lead to delayed language development, which may or may not lead to low educational attainment, poor psychosocial development (primarily behavior problems), and a vocational outcome less than that which would otherwise be obtained.
- The impact should, in principle, increase with the degree of loss in the impaired ear.
- While some educational outcomes (like having to repeat a grade or requiring special educational assistance) are highly visible, other impacts (like achieving average academic performance when the child is otherwise capable of excellent academic performance) may well go unnoticed.
- Experiments measuring the outcomes of children recruited from audiology or medical settings likely have an inherent bias towards finding that unilateral loss causes problems. The children studied are more likely to include those who have sought help *because* of poor educational progress than children with the same hearing loss who are having no problems.

Overall, it does seems likely that unilateral hearing loss adversely affects language development and educational attainment, a conclusion reached in other reviews of this topic[1064, 1778] and in a comprehensive workshop on its effects and management.[264a] The effect of unilateral loss on outcomes appears to be not sufficiently strong, however, that it is found in every study.

It would be surprising if a unilateral loss of severe or profound degree did *not* make it more difficult to acquire language or to learn generally in the classroom, as is the case with bilateral loss. The increased effort involved in listening when there is noise or reverberation will leave the child with fewer cognitive resources to deal with all the processes of learning and will fatigue the child more than his or her normal-hearing peers.

The existence of a problem caused by hearing loss in one ear does not imply that a hearing aid will necessarily be an effective solution. Certainly a significant proportion of children with unilateral hearing loss cease using their hearing aid.[400] Just as with adults with mild and moderate loss, reasons for non-use will be a complex mixture related to hearing difficulties, other compensatory strategies adopted, and psychosocial concerns. It does, none-the-less, seem reasonable to recommend aiding the impaired ear wherever possible, so that binaural processing mechanisms are given every chance to be as effective as possible in as many situations as possible.

Where there is a profound sensorineural loss on the impaired side, the only forms of device that are likely to be effective are a cochlear implant in the impaired ear, or wireless transmission from a microphone worn by the teacher to a receiver worn by the child, with the sound delivered to the better ear (Section 15.6.3). For lesser degrees of loss in the impaired ear, conventional hearing aids and CROS aids provide some benefit, but are much less effective than wireless transmission devices.[893] McKay (2010) provides a more comprehensive review of options, including bone anchored hearing aids (Section 17.3) and transcranial CROS aids (Section 17.1.4). Wireless transmission to the better ear will provide the clearest speech no matter what degree of loss is present in the impaired ear and no matter what type of device, if any, is fitted to it.

Even if hearing aids are not provided to children with unilateral loss in their first few years of life, their hearing status should regularly be checked. It is relatively common for the degree of hearing loss to progress during the first four years of life,[827] so it is possible that progression from unilateral loss to bilateral loss may also occur during this period.

Until better research data are available about the range of impacts of untreated unilateral loss (analyzed according to the degree of loss in the impaired ear), and the effectiveness of hearing aids and wireless transmission systems in minimizing the impact of the loss, decisions about aiding will very much have to be made on a case-by-case basis. These decisions are made in conjunction with the family, taking into account whether each child's educational or social progress appears to be impeded by the hearing loss.[1168] Strategies for minimizing the impact of the loss, especially those involving positioning talkers to the side with the good ear and providing enriched auditory/language exposure (e.g. conversation, singing, reading aloud, all with a good SNR) must be explained, irrespective of whether a hearing aid or other device is fitted.

16.1.3 Slight and mild hearing loss

The situation with slight (or minimal) and mild bilateral hearing loss is similar to that for unilateral loss in that while several studies suggest that mild bilateral loss adversely affects educational outcomes, a few find no effect.[123, 264a, 1778, 1871] There is also uncertainty as to whether fitting hearing aids improves outcomes, but fitting of hearing aids to children with mild loss is none-the-less extremely common.[1168]

Given the wide range of factors that affect educational outcomes, and the progressive effect that an increasing degree of hearing loss undoubtedly has, the question should really be expressed as: *For any particular degree of bilateral hearing loss, what is the probability of a child achieving below-average educational outcomes without hearing aids?* The answer will certainly smoothly increase from 50% for normal hearing (and no other disabilities) up to 100% for a severe hearing loss, but we lack the data to determine the degree of loss for which it departs significantly from 50%.

The associated question, for which we also don't have an answer, is: *For any particular degree of bilateral hearing loss, what is the probability that wearing hearing aids will improve educational outcomes?* This answer will increase from 0% for normal hearing to 100% for a severe hearing loss, but again we lack the data to be more specific.

Management options include conventional hearing aids, FM systems, classroom amplification and preferential seating.[1168] As with unilateral loss, whenever a bilateral loss is large enough to impact on educational outcomes, the greatest increase in speech perception and the greatest decrease in listening effort will be achieved in the classroom if an FM system is part of the solution.

16.1.4 Cochlear implantation

Numerous studies have shown that, on average, implanted children have performance (based on speech perception, speech production, or educational placement) markedly superior to aided children with average hearing thresholds greater than 90 dB HL.[58, 393, 548, 762, 1039, 1192, 1844] Consistent with this clear superiority of implantation for profound loss, several studies have shown that implanted children, on average, have speech perception and language acquisition similar to that of children with a pure tone average loss around 78 dB HL.[142, 508] Not surprisingly, about the same proportion of implanted children as aided children with a pure tone loss around 78 dB HL are also placed in mainstream schools.[58]

For children who receive implants, speech perception and production later in childhood are maximized if cochlear implantation in at least one ear is achieved as early as possible,[411, 1320, 1973] and preferably by around 12 months of age.[292, 423] Looking at this issue another way, children with hearing aids or cochlear implants increase their language-equivalent age on average by about one year with each passing year, although some will progress more slowly and some will progress faster than normal after implantation.[1940] That is, any deficit that exists at the time of implantation will, on average, not decrease with time. Consequently, the earlier that implantation occurs, the smaller will be the deficit at later ages.

The superior outcomes achieved when children are implanted early lead to cost savings when the lower educational costs are taken into account.[1574] Although late implantation (e.g. at 5 to 10 years of age) certainly does not maximize outcomes, it still provides benefit for children for whom hearing aids provide minimal benefit.[154, 1207]

Because the studies comparing children with hearing aids to children with cochlear implants have studied children who mostly received their implants well after their first birthday, it seems almost inevitable that these studies have underestimated the potential performance that children can obtain with cochlear implants. Early-implanted children with 78 dB HL pure tone average hearing loss will therefore likely achieve better performance, *on average*, with implants than with hearing aids.

Hearing aids interact with cochlear implants in three ways.

First, inadequate performance with hearing aids provides evidence that implantation is needed. Implantation is a big step, causing irreversible changes to the cochlea. Performance (reaction to sounds, understanding of speech, and production of speech-like sounds; Sections 16.6.1, 16.6.4 and 16.6.7) while wearing hearing aids is normally one of the factors considered before making the decision to implant. Electrophysiologically measured responses evoked in the cortex by speech sounds while wearing hearing aids can also be considered (Section 16.6.6). Although some measures of speech perception, speech production and language acquisition can be made within the first year of life, they cannot be measured as accurately as pure tone hearing thresholds can be estimated. Because of the importance of deciding on implantation well before 12 months of age, pure tone thresholds are the primary factor considered in whether implants or hearing aids will be better for infants.

Second, for most children, hearing aids should ***increase stimulation of the cortex*** prior to implantation, even if speech perception is poor. This early stimulation appears to increase the children's ability to use binaural cues later in life,[1840] possibly by enabling neural development during the first year of life when the auditory system appears to be most plastic. The longer the duration without effective stimulation of the auditory processing system within the first few years of life, the greater the likelihood that the auditory cortex or association cortex will be colonized by other sensory modalities, and the poorer that auditory performance will be when implantation eventually happens.[198, 1774] Unfortunately, early stimulation with hearing aids, although on average beneficial,[270] cannot be relied on to provide sufficient stimulation prior to implantation, hence the importance of early implantation.[b, 1320]

Third, after implantation in one ear, a hearing aid in the other ear provides ***complementary speech cues*** (Section 9.2.3) and continued stimulation of the contralateral pathways of the auditory system. This takes away the pressure to immediately implant the second ear.[1840] In fact, children who have had a period of bimodal stimulation, even if bilateral implantation subsequently occurs, appear to acquire more sophisticated language abilities than children who have only ever had one or two implants, possibly as a result of the greater access to fundamental frequency cues that hearing aids provide.[1325]

16.1.5 Auditory neuropathy spectrum disorder

There is great uncertainty over the most appropriate intervention for children with auditory neuropathy spectrum disorder, especially during the first nine or so months of life when it is difficult to determine how much language is being acquired by the child and when even the degree of pure-tone loss and audibility of speech sounds are difficult to estimate. All device options (no device, hearing aids, cochlear implants, FM systems) should be considered, as all can lead to satisfactory outcomes for different children.[118, 1480, 1479] There is a widespread belief and some supporting data to show that cochlear implantation is successful more often than hearing aids,[117, 118, 1485, 1959] but there is a paucity of studies investigating the proportion of children who gain comparable benefit from hearing aids fit early and with gain appropriate to the degree of loss.[1478]

Not surprisingly, there seems to be a close connection between the effectiveness of hearing aids for children with auditory neuropathy and the cortical activity elicited by speech sounds:

- Older children who obtain good speech intelligibility with hearing aids appear to have cortical responses to speech sounds, with normal morphology and latencies, whereas children who receive minimal benefit from hearing aids do not.[1481]
- For younger children, those children with the highest functional auditory abilities (as assessed by the parents) are more likely to have cortical responses present,[633] and for those cortical responses to have near-normal latency.[1611]

From basic principles, if the disorder is severe and the site of the faulty process is within the cochlea (likely within the inner hair cell or its synapses) or within the dendrites leading from the inner hair cells to the spiral ganglion cells, then a cochlear implant is likely to be successful (because the electrical stimulation bypasses these structures). Conversely, if the faulty physiological process or anatomical structure is within the auditory nerve between the spiral gan-

[b] In principle, the more profound the hearing loss, the less likely it is that hearing aid use prior to implantation will affect language outcomes later in life. The long-term effect of early hearing aid use before implantation is therefore likely to differ across children, and potentially across listening tasks.

glion cells and the brainstem, a cochlear implant is much less likely to be successful because the electrically stimulated nerve impulses will still have to pass through these faulty pathways. Cochlear implantation is a possibility even when hearing thresholds are in the mild to moderate range if speech understanding ability is significantly poorer than would be expected for an implanted child of the same age.

The following are some considerations that might be useful when deciding between options:

- If the child cannot hear the sounds of speech unaided, then either a hearing aid or a cochlear implant is certainly indicated as audibility is a pre-requisite for auditory speech perception.
- For older children, those who have robust cortical potentials evoked by sounds seem to obtain considerable benefit in speech understanding from hearing aids but those without cortical potentials seem not to benefit from hearing aids.[1481] Although it is sometimes recommended that low-gain hearing aids be fit to minimize the chance of causing noise-induced hearing loss,[c] there seems little point in fitting hearing aids with gain so low that speech sounds at typical conversational levels remain inaudible. The audibility of speech sounds can be determined behaviorally for children older than about 9 months, and by the presence of evoked cortical potentials for infants younger than this, although the lack of a cortical response does not necessarily imply a lack of audibility.
- As with other forms of hearing loss, children with ANSD need a better SNR than normal to understand speech. FM systems are therefore an important part of the solution to achieve good speech understanding when there are competing sounds, irrespective of whether hearing aids, cochlear implants, or neither, are also fitted.
- The auditory function of some children with auditory neuropathy improves (as assessed by both hearing thresholds and ABR waveforms) during the first year of life, particularly for babies with low birth weight and/or hyperbilirubinemia, which makes hearing aids a more conservative treatment option during this period.[65, 1122, 1465, 1485]

Pediatric clinicians will have to closely monitor developments in methods for managing babies with auditory neuropathy as the current state of knowledge of how to match treatments to children is very unsatisfactory. Although auditory neuropathy is far from the most common form of childhood hearing loss, it's too common to consider it a rare condition.

16.2 Assessment of Hearing Loss

16.2.1 Frequency-specific and ear-specific assessment

Methods for assessing the degree and type of hearing loss are not within the scope of this book. It is, however, essential that hearing loss be assessed as accurately as is possible within the time available. It is equally important that hearing aid fitting not be delayed just because some uncertainty remains over the degree or configuration of hearing loss.

The accuracy with which hearing loss can be assessed will vary with the age of the child, but the range of techniques now available will allow a reasonable assessment to be made at any age. The minimum requirements are estimated thresholds for one low frequency (preferably 500 Hz) and one high frequency (preferably 2 kHz), separately for each ear. Of course, it is better if thresholds can be estimated at more frequencies. If time and the behavior of the child prevent this, however, an appropriate fitting is more likely to be achieved with two reasonably accurate thresholds than with a greater number of inaccurate thresholds. As more audiological information is obtained during subsequent appointments, the hearing aid fitting can be fine-tuned. Typically, over time, more frequencies are added, electrophysiological (ABR or ASSR) thresholds are replaced by behavioral thresholds, and the accuracy of estimates is improved.

Although children's audiograms have a wider range of configurations than adults' audiograms, on average they are flatter than typical audiograms from adults.[1113, 1423, 1948] To assist in the temporary estimation of thresholds at frequencies not yet measured, Table 16.2 shows the statistics of audiogram slopes based on children's audiograms for 400 ears.[1113] The median values are extremely similar to the mean thresholds

[c] Many children with auditory neuropathy start life with otoacoustic emissions, which often disappear over time, whether or not hearing aids are worn.[1478]

Table 16.2: The distribution of audiogram slopes (dB/octave) in each octave for a random sample of 400 audiograms from children who wear hearing aids.[1113] Positive slopes indicate thresholds that become more severe as frequency increases.

Octave (Hz)	10th percentile	25th percentile	Median	75th percentile	90th percentile
250 to 500	-5	0	5	10	20
500 to 1000	-5	0	10	15	25
1000 to 2000	-10	0	5	15	30
2000 to 4000	-10	-5	0	10	20
4000 to 8000	-20	-10	0	10	18

from an unrelated sample of 227 children.[1423] If, for example, the only frequency-specific threshold available was at 2 kHz, based on the median slopes shown, the best guess for the remaining frequencies is that relative to the 2 kHz threshold, thresholds are the same at 4 and 8 kHz, and are 5, 15 and 20 dB better at 1000, 500 and 250 Hz respectively.

Although many infants have approximately symmetrical hearing losses, it is totally inappropriate to assume that they all do. For 180 aided children surveyed at random, the absolute value of the differences between thresholds in the left and right ears (at the same frequency) averaged only 8 dB but ranged up to 90 dB.[1113] At any frequency, hearing thresholds in the two ears differed by 20 dB or more for approximately 10% of the children. It is thus essential to obtain thresholds separately for each ear. Obtaining separate thresholds for each ear is most easily achieved by using insert earphones, which are more comfortable and more readily tolerated than supra-aural headphones. They also reduce the need for contralateral masking because of their much greater inter-aural attenuation. Insert earphones can also be calibrated more appropriately for small ears, as explained in the next section. The use of insert earphones is thus strongly recommended.

Frequency-specific assessment techniques that can be used with insert earphones include:

- tone-burst Auditory Brainstem Response (ABR);
- single or multi-frequency Auditory Steady State Evoked Potentials;
- distortion-product otoacoustic emissions, or click-evoked otoacoustic emissions, although these currently give only a generalized impression of high-frequency outer hair cell activity, rather than specific hearing thresholds.
- behavioral techniques such as Visual Reinforcement Audiometry (VRA; also referred to as Visual Reinforcement Orientation Audiometry - VROA), Tangible Reinforcement Operant Conditioning Audiometry (TROCA), Visual Reinforcement Operant Conditioning Audiometry (VROCA), and Play Audiometry.

The choice of techniques is determined by the age of the child and by the equipment available. To increase the accuracy and surety with which thresholds are known, more than one technique should be applied. For infants, these should preferably include one behavioral measure, at least one electrophysiological measure, and otoacoustic emissions to confirm a cochlear abnormality and to help differentiate sensorineural hearing loss from auditory neuropathy spectrum disorder.

Electrophysiological thresholds will initially be reported in dB nHL – the lowest stimulus level at which an electrophysiological response is reliably present, relative to the lowest stimulus level that adults with normal hearing can report hearing the same stimuli. These electrophysiological thresholds must be converted into predicted behavioral thresholds for the infants by the use of appropriate correction figures. This conversion may occur within the electrophysiological test equipment, may be done manually by the audiologist, or may occur within the hearing fitting software. In the latter case, both the NAL-NL2 and DSLm[i/o] software allow direct entry of the observed electrophysiological thresholds in dB nHL.

16.2.2 Small ears and calibration issues

Hearing thresholds may be determined using stimuli generated by a loudspeaker, supra-aural headphones, or insert earphones. In the last case, the insert phones may be connected to the child's ear by the standard foam earplugs, immittance tips, or the child's individual earmold. Even if the transducers have been calibrated so that an average adult with normal hearing has thresholds of 0 dB HL, the same will not be true for an infant with normal hearing. For example:

- A new-born baby will have an ear canal resonance nearer to 6 kHz than to the resonance of 2.7 kHz that applies to the average adult.[984] If stimuli are presented via a loudspeaker, supra-aural earphones or circumaural earphones, the baby's very high-frequency canal resonance will make the hearing thresholds seem better than they really are at 6 kHz (at least for the loudspeaker), but worse than they really are at 3 kHz (for all three transducer types).[1866]
- A baby will have a much smaller residual ear canal volume (the volume medial to the tip of the earmold) than an adult. If stimuli are presented via an insert earphone, a higher SPL will be present in the infant's ear than in the adult's ear at all frequencies.[1866] The infant will therefore appear to have less hearing loss than the adult at all frequencies, even if their middle ears and cochleae function equally effectively.

Large transducers that do not occlude the ear canal (loudspeakers, supra-aural earphones, and circumaural earphones) have a low acoustic impedance, so the characteristic of the ear that most affects SPL at the eardrum is the length of the canal. Small transducers that fill part of the ear canal (insert earphones) have a high acoustic impedance, so the characteristic of the ear that most affects SPL at the eardrum is the volume of the residual part of the ear canal. Both of these ear characteristics are very much smaller in infants than in adults.

Hearing thresholds (in dB HL) may thus appear to change during the first few years of a child's life, just because of changes in the size of the child's ears.[1212] There are two equally effective solutions to this problem. One is to express all thresholds in dB SPL in the ear canal. Expressing threshold in this way simplifies comparisons between threshold and hearing aid output expressed in the same manner. The second is to express thresholds as ***equivalent adult hearing level***. This is the hearing threshold level that an average adult would have if the adult has the same threshold in dB SPL at the eardrum as the child. Equivalent adult hearing level is synonymous with the term ***predicted hearing level*** used within the DSL software.[1599] Comparisons with hearing aid output are less straightforward (unless done within a computer program), but the familiar characterization of hearing loss in dB HL, i.e. departure from normal, is maintained.[d]

Both methods require threshold to be expressed in dB SPL in the ear canal. For the first method, no further calculations or conversions are necessary. For the second method, this ear canal threshold is converted to equivalent adult hearing level by subtracting the adult average REDD values shown in Table 16.3. Both methods are most conveniently carried out with the aid of suitable software (e.g. NAL-NL2 or DSLm[i/o] or their implementations in manufacturers' software).

Table 16.3 Useful correction factors for hearing thresholds. Reference equivalent threshold SPL values are those applicable to insert earphones calibrated in HA1 or HA2 style 2 cc couplers . Adult average REDD is the eardrum SPL corresponding to 0 dB HL for the average adult.[112] These REDD values can be subtracted from eardrum SPL at threshold if one wishes to convert thresholds from eardrum SPL to equivalent adult-average hearing level.

Correction Factor	Frequency (Hz)						
	250	500	1000	2000	3000	4000	6000
RETSPL HA1 2-cc	14.5	6.0	0.0	2.5	2.5	0.0	-2.5
RETSPL HA2 2-cc	14.0	5.5	0.0	3.0	3.5	5.5	2.0
$REDD_{average\ adult}$	16	12	10	16	15	13	16

[d] DSL software uses the first of these solutions (canal SPL) to address the problem of changing ear geometry, and NAL-NL2 software uses the second (equivalent adult hearing level).

Determining threshold in dB SPL in the ear canal

The best way to determine threshold in the ear canal is to use an individually measured real-ear to coupler difference (RECD; Sections 4.2 and 11.4). Thresholds expressed in dB HL obtained with an insert earphone are first converted to dB SPL in a 2-cc coupler by adding the reference equivalent threshold SPLs (RETSPLs) shown in Table 16.3. These coupler SPLs are then converted to real-ear SPL by adding the individually measured RECD values. Section 16.4.3 contains further information on measuring RECD for babies and small children.

Alternatively, the clinician can directly measure the difference between the audiometer setting (in dB HL) and the SPL in the real ear for each patient. This difference is known as the ***real-ear to dial difference*** (***REDD***) and can be obtained for any type of headphone.[1587] Figure 4.10 and equation 4.7 makes explicit the connection between RECD, REDD and RETSPL.

$$\text{Equivalent adult threshold (dB HL)} = \text{Ear canal threshold (dB SPL)} - \text{REDD}_{\text{adult average}}\text{ (dB)} \qquad \ldots\ 16.1$$

Note that even when equivalent adult hearing level is used, it is still possible for hearing thresholds to change over the first few years of life. Normal-hearing infants have elevated thresholds because of some combination of immaturity of the middle ear,[862] immaturity of axonal conduction[1245, 1246] and synaptic efficiency,[1445] inability to focus on a narrow frequency region containing the stimulus,[1909] and inattentiveness.[1909] It seems likely that these same changes would occur for infants with hearing loss so that thresholds improve over the first year or two of life. Conversely, some hearing losses are progressive,[827] and hence thresholds may also deteriorate.

16.2.3 Auditory processing disorders

Just as for adults, children with sensorineural hearing loss have spatial processing disorder, a particular type of auditory processing disorder that renders them less able to attend to target sounds coming from one direction by suppressing sounds coming from other directions.[289, 628] Children who have otitis media for more than half of their first six years are also likely to have a reduced ability to optimally combine the information present at each ear (that is, a binaural processing disorder) even after their hearing sensitivity returns to normal.[748] Assessment of the speech perception capabilities in noise of children with spatial processing disorder, irrespective of whether they have normal hearing sensitivity or sensorineural hearing loss, will underestimate the difficulties they face unless the speech test involves presentation of speech from a direction different from that of the competing signals.

Children with auditory processing disorders (probably of any type), and children with sensorineural hearing loss therefore need an SNR higher than that required by their normal-hearing peers in the classroom if they are to equally understand speech when there is background noise.[e] In both cases, wireless transmission from the teacher to the student is the most effective solution. For children with hearing loss, the received signal is amplified by the child's usual hearing aid. For children with normal peripheral hearing, optimal delivery to the child of the received wireless signal is less clear: both ear canals should remain open so that the child can retain normal perception of those close to him or her, but this allows background noise in at the usual level. Consequently, the signal received by wireless transmission must be presented at an amplified level so that at least some of its inherently greater SNR is preserved. A compromise is needed so that the level is great enough to provide a sufficiently improved SNR, but not so great that it risks discomfort or hearing damage caused by continuously listening to amplified sound through ears with normal hearing sensitivity.

[e] In the case of children with spatial processing disorder, training is available that allows the children to develop the binaural processing skills they have so far failed to develop.[243] A wireless system is no longer needed once the training is complete.

16.2.4 Miscellaneous issues in assessment

Vision loss in children born with hearing loss is extremely common, but the vision loss often goes undetected for a long time.[1322] Vision is, of course, even more important for children with hearing loss than for children with normal hearing. It is therefore important that when hearing loss is detected, the child's vision also be investigated. As with hearing, early intervention for congenital vision impairment is critical to obtaining the best outcomes. Because hearing is critical for localization and speech perception by deaf-blind children, the ability to select either directional microphones (for speech understanding) and omni-directional microphones (for localization) appears to be particularly important,[1777a] although much more research on this topic is needed.

Although the focus of this chapter is on the use of hearing aids by children with permanent sensorineural hearing loss, a conductive loss caused by long or repeated period of otitis media can be just as disabling. Children with cleft palate, even after repair, have a high incidence of otitis media with effusion caused by reduced efficiency of the Eustachian tube. With or without the fitting of ventilation tubes, hearing aids should be considered for these children.[1126]

16.3 Hearing Aid and Earmold Styles

16.3.1 Hearing aid style

The appearance and size of the hearing aid is likely to be important to teen-aged children, and to parents of children of all ages. Although some will choose the most brightly colored devices available, others will choose the most discrete. When worn by infants, even average-sized hearing aids look very big behind tiny ears. Parents will ultimately make the final decision about the size and style of hearing aids for young children. It is essential that they first understand the likely serious consequences of choosing a hearing aid that has inferior electroacoustic qualities, if such a choice is being considered. For example, the strong link between receiving an adequate signal during the first few years of life, and the development of good language and speech should be impressed on the parents. To achieve an adequate signal, the maximum output must be appropriate to the degree of loss, and this may dictate a hearing aid larger than parents would otherwise prefer.

Almost all hearing-impaired children are fitted with BTE hearing aids. These have the advantage over body aids that sound is picked up at head level instead of being affected by clothing noise and body baffle effects, especially if the infant is prone. They are also less likely to be covered in food or vomit.

Body aids should be considered only if there is some reason a BTE would be ineffective. This *might* include children who have additional disabilities that require their head to be supported if this head support would:

- muffle sound pick up by the BTE aid;
- frequently bump the BTE aid; or
- induce feedback oscillation in the BTE aid.[1777]

Practical tip: Securing the hearing aids

When children are active enough to lose their hearing aids but not old enough to make sure they keep them, hearing aids can be secured by a Huggie™ aid. This is a large loop that encircles the pinnae attached to two small loops that encircle the hearing aid. Alternatively, a fishing line can be tied around each earhook and secured by a safety pin to the collar at the back of the child's clothing. (A tip from the dad of an active child.) Commercial versions of this solution include *oto clips*, *dino clips, and Oliver clips.*

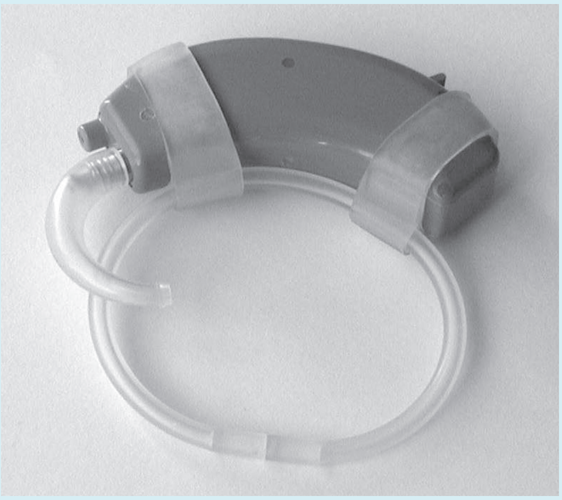

Figure 16.1 A Huggie Aid™ attached to a BTE hearing aid.

There is no fundamental reason why ITE hearing aids (or more generally ITC or CIC hearing aids) cannot be used for children of any age, but there are several strong practical disadvantages.

- It will likely be difficult to fit an ITE within a small ear.
- Small ears grow, rapidly at first, and then more slowly. If an ITE is worn too early, it will have to be replaced frequently as the ear grows. Replacing an ITE aid is much more expensive than replacing an earmold. A pediatric working group[125] considered that ear growth stabilizes sufficiently by the age of 8 to 10 years to consider ITE hearing aids, but with the availability of mini-BTE hearing aids combined with thin tubing, there is little cosmetic advantage in choosing ITE devices.
- All hearing-impaired children will benefit from the use of an FM or other wireless system in many circumstances. Some ITE hearing aids do not have the necessary audio input socket, wireless receiver, or possibly even telecoil. This deficiency makes it impossible to combine the signal-to-noise ratio (SNR) advantages of the wireless system with the individually prescribed electroacoustics of the hearing aid.
- There is a small risk factor with ITE hearing aids. Because the shell is a thin layer of hard plastic, breakage of the hearing aid while it is in the ear can create sharp edges that can lacerate the canal wall. This fracturing can happen to people of any age who receive a blow to the ear, but seems more likely to happen with a child's lifestyle. Ingestion of an ITE may also be more likely.
- It is difficult for a parent or teacher to visually identify whether an ITE is on or off, and where the volume control is set, although this is also true for many BTE hearing aids.

To maximize information transmitted to the child in noisy or reverberant situations, the child may need a wireless system immediately or within the life of the hearing aid. If so, it is necessary for the hearing aid to have a direct audio input connector, telecoil (used in conjunction with a neck loop - Section 3.11.1) or inbuilt wireless receiver.

16.3.2 Earmolds

Concerns similar to those for ITE shells also exist for the safety of hard earmolds for BTE hearing aids. Soft earmolds are used more commonly than hard earmolds, because of the greater risk of injury from a hard earmold if it is broken or pushed into the ear during active play. Soft earmolds are also less likely to cause discomfort or feedback oscillation.

As with hearing aids for adults, soft materials deteriorate more rapidly with time, but this is not an issue for young children because their earmolds will have to be replaced more often as the ear grows. In summary, either hard or soft materials can be used for earmolds, but for younger children, the advantages of soft materials nearly always outweigh their disadvantages.

The small size of a baby's ear can create three other problems. The angle at which the tubing protrudes from the earmold, combined with the close proximity of the earmold to the tip of the earhook can pull the hearing aid away from the surface of the head. The solution is for the tubing to bend more sharply, but often this cannot be done without kinking the tubing. A solution is to hollow out the center of the conchal part of the earmold, prior to inserting the tubing.[1345] A cross-section through the earmold is shown in Figure 16.2. (This process is most conveniently done by the earmold laboratory when the earmold is first made.) The hollowed area allows the tubing to commence its upward bend closer to the ear canal.

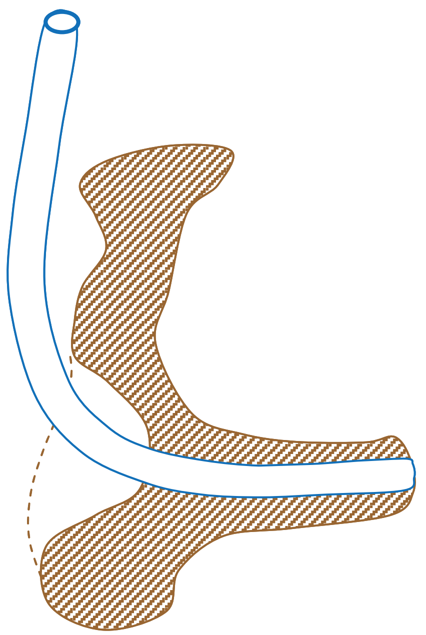

Figure 16.2 Cross section of a hollow-concha earmold.

The second problem created by a small ear is that it can be difficult to achieve even a 2-mm sound bore. The only solutions are careful drilling, plus terminating the sound tubing some distance from the medial tip of the earmold. If the tubing is constricted at any point, decreased high-frequency gain and maximum output are likely to result. Third, it can be difficult or impossible to use the acoustic modifications (horns and internal vents) that can be achieved in an adult-sized ear. Trench vents on the outside of the earmold can be added, however.

There appears to be no research on whether occluding the ear canal causes children's own voice to have an unacceptable quality, as occurs with adults. It is probable that it is less of a problem, because the fundamental frequency and formant frequencies of children's voices are considerably higher than those of adults. These are above the frequency (300 Hz) at which bone conduction causes the SPL to increase in adult ear canals by the greatest amount.

16.4 Prescribing Amplification for Children

16.4.1 Speech identification ability and amplification requirements

A fundamental question is whether children, and infants in particular, need amplification characteristics different from those needed by adults with the same degree of loss. An easy aspect of this question relates to the small size of infants' ear canals and the implications this has for the coupler gain needed. Children need less coupler gain and OSPL90 than adults if they are to receive the same gain and maximum output at the eardrum as adults. Procedures for dealing with this issue are described in Sections 11.4 and 16.4.3.

The more difficult part of the question is whether the *real-ear* gain and maximum output should be different for children than for adults. Let us first examine why children *might* have real-ear amplification requirements different from those of adults, then consider empirical evidence for differences, and conclude with practical recommendations.

When we adults hear speech, we use our knowledge of the language to fill in any sounds that are too weak for us to perceive directly. This occurs so naturally that we are not even aware that we are doing it unless the proportion of information that we can directly perceive is very small. A child still acquiring language is less able to fill in missing information.[1324] An infant presumably cannot do so at all.

Even when speech material contains no semantic, syntactic, or linguistic context, such as is the case for nonsense syllables, normal-hearing infants need the level to be 26 dB higher than that needed by adults to discriminate between syllables with the same accuracy.[1353] Similarly, 5-year old children need levels considerably higher than do older children if they are to achieve the same performance on a monosyllabic word test, even though the words are familiar to them.[213] Does this mean that we should prescribe more gain for hearing-impaired infants than for hearing-impaired adults with the same pure tone hearing loss? Possibly, but the data obtained with normal-hearing infants do not answer this question.

What the data directly tell us is that if we were prescribing amplification on the basis of achieving optimal speech scores, we would fit normal-hearing infants with a hearing aid that had a gain of approximately 26 dB for low-level sounds! There are two problems in applying this to hearing-impaired children. First, the data do not tell us how much gain children need at higher input levels. Second, the optimum gain for a particular input level is a delicate balance between choosing the gain and frequency response that optimizes intelligibility, but subject to the overall loudness of the sensation being acceptable to the listener in that listening situation. This balance may well swing more towards increased gain for those with little or no knowledge of language than it does for those with well-developed language skills.

It seems very likely that if there is a difference between what is best for adults and what is best for infants, this difference will be greater for low-level sounds than for high-level sounds. The loudness discomfort level for hearing-impaired children aged 7 to 14 years has been shown to be the same as for hearing-impaired adults with the same pure tone hearing losses.[860] Consequently, we would not want to prescribe any more high-level gain and OSPL90 for infants than we do for adults.

The previous discussion has been about overall gain and maximum output. Of course, we also have to decide how much gain should be applied at each frequency. That is, it is possible that the *shape* of the frequency response should be different from that which is optimal for adults. Some experts have conjectured that children need additional high-frequency gain, because high-frequency cues are often the most difficult for children to perceive. Other experts have argued that infants first need additional low-frequency gain because intonation and other supra-segmental cues seem to be important components of communication when language is first developing. Either of these conjectures could be right or wrong.

While it is possible that children need different response shapes as their knowledge of the language develops, we simply do not have the knowledge needed to link stages of language acquisition to optimal response shapes. Because important cues to speech reside in all frequency ranges, it seems unwise to make such an assumption without some evidence that it is true. The question is linked to the previous discussion about gain and loudness: every gain increase in one frequency region will be at the expense of amplification for the rest of the frequency range, unless the signal is also made louder overall.

Just as infants discriminate speech more poorly than adults at low absolute levels, infants also need an SNR 7 dB higher than that needed by adults to achieve the same discrimination of nonsense syllables.[1354] Similarly, to identify familiar words and sentences, five-year old children need an SNR 3 to 5 dB higher than that needed by older children or adults.[152, 645] The clearest implication from these findings is that there will be listening situations in which normal-hearing adults can discriminate between sounds, but in which normal-hearing infants and young children cannot. The situation is most unlikely to be any different for hearing-impaired adults and infants.

The implication of these findings for the SNR provided by hearing aids is much more straightforward than it is for gain and frequency response. There is usually no problem if SNR is improved more than is necessary, whereas if too much gain is applied, the result can be discomfort, decreased intelligibility, or both. (In some circumstances, improving SNR of one talker too much is a problem if accomplishing it stops the child from hearing his or her own voice or the voices of others nearby.)

Children will benefit from provision of wireless hearing aids in many situations, and these devices should be provided for every situation in which they can be practically used. Section 16.4.5 contains more details.

Unfortunately, there have been no studies directly comparing the optimal amplification for infants to that for adults, or to the prescriptions resulting from any selection procedure.[f] This is not surprising, as such studies would be extraordinarily difficult to do. Several studies[233, 283, 1667] have shown that the average gain preferred by children is the same as that preferred by adults with the same degree of loss, although the youngest age studied was 6 years. These studies also showed that the optimum shape of the frequency response could be predicted from the audiogram using the same rules that are appropriate for adults. On the other hand, Snik & Hombergen (1993) showed that older children prefer gains 7 dB higher than those preferred by adults with the same degree of loss. The gains used by the children were, however, extremely close to those prescribed by the NAL-RP method.

A comprehensive joint study carried out in Canada and Australia compared the preference of school-aged children for, and performance with, the NAL-NL1 and DSL[i/o] prescriptions.[286-288, 1581, 1585] Because of feedback oscillation and other limitations in the hearing aids, the often large difference between the prescriptions were not achievable, and the difference between the gain-frequency responses actually achievable for the two prescriptions primarily comprised a higher gain for DSL[i/o] than for NAL-NL1 (averaging 7 dB across frequency at mid input levels, a little more in low frequencies and at low levels and a little less at high frequencies and at high levels).

Objectively measured speech perception was equally good with both prescriptions. Nonsense syllable identification in quiet was close to ceiling performance and sentence reception threshold in noise was almost as good as for normal-hearing children. There was a tendency for children to prefer DSL[i/o] for low-level sounds, as it gave greater audibility, and a tendency for children to prefer NAL-NL1 for high-level sounds, or for listening in noisy places, as it gave greater listening comfort. There was also a tendency for children to prefer the prescription that they had prior experience with.

[f] There have been studies comparing prescribed amplification characteristics to what children younger than six years have been fitted with, but this has an illogical circularity if the fittings have been based on the same prescription procedure.

As a consequence of this collaborative experiment between the NAL and DSL research groups, the NAL prescription was revised so that the NAL-NL2 prescription for children had greater gain than NAL-NL1 at low and medium input levels.

Children's preference for greater gain than was provided by NAL-NL1 contrasts with the finding reviewed in Section 10.4.8 that adults prefer less gain than was provided by NAL-NL1, and much less than the greater gain provided by DSL[i/o]. *Consequently, it seems reasonable to conclude that, on average, children prefer higher real-ear gain than adults with the same degree of loss*. This empirical finding is consistent with the research already reviewed showing that young children need higher sensation levels than adults to achieve maximum speech perception. It is also consistent with research showing that normal-hearing children need higher SPL than adults to achieve the same loudness, particularly at low and medium input levels.[1604]

Given that for children the hearing loss is often congenital whereas for adults it is more often acquired, does the observed difference in gain preferences arise from the difference in age, or the difference in etiology of the hearing loss? Because there appears to be no significant difference in preferred gain between adults with congenital loss versus those with acquired loss, for the same degree of loss, it seems that the higher gain needed by children does arise because they are children, rather than from the different hearing loss etiology.[873] The age at which a child becomes an adult, from the perspective of preferred gain, is not known.

How can we translate the preceding information into practical guidelines for prescribing amplification? The finding that normal-hearing children, hearing-impaired children, and hearing-impaired adults all need a better SNR than normal-hearing adults to achieve the same performance has the clearest implications, and has already been discussed. Wireless systems and, to a lesser extent, directional microphones will help. It may be desirable for children to sometimes receive a noisy signal so that they can develop skills at perceiving speech in noise. Such opportunities are likely to occur without being especially arranged because use of a wireless system will not be possible for logistical reasons in many situations.

The question of how much real-ear gain is needed for children relative to adults is more difficult. We need to consider not just *gain*, but gain for high-level sounds, gain for medium-level sounds, and gain for low-level sounds.

High-level sounds (e.g. a group of children playing at 80 dB SPL). It seems unlikely that children will benefit from more gain for high-level sounds than that given to adults, given the impact of hearing aid limiting, loudness discomfort, and hearing loss desensitization. (Hearing loss desensitization refers to a person's decreased ability to extract useful information from an audible signal in any frequency region where the loss is severe or profound. The greater the sensation level, the greater its effects, as discussed in Section 10.3.4.) The close proximity that infants often have to the talker, and the consequently increased speech levels[1706] are further reasons against giving additional gain for high input levels. The likely frequent occurrence of high input levels suggests that compression to decrease gain as input level rises above about 70 dB SPL is even more important for infants than it is for adults.

Medium-level sounds (e.g. one person talking at 65 dB SPL). The weight of evidence is that children *prefer* more gain than do adults for medium-level sounds. On the little evidence available, it seems unlikely that higher gain provides greater speech intelligibility for mid level sounds, but it may reduce listening effort.

Low-level sounds (e.g. a person talking quietly, and/ or in the distance at 50 dB SPL). There is no doubt that children are less able than adults to make full use of low-level speech signals. It therefore seems very likely that the optimum low-level gain for children should be greater than for adults. Increasing the gain for low-level sounds may also increase the distance over which children can hear or overhear comments. Normal-hearing children enrich their language (whether parents approve or not) by overhearing comments that were not intended for them.[24, 1697] Increasing gain for low-level sounds also has few disadvantages: it is unlikely to cause such high sensation levels that it decreases speech intelligibility in the way it can do for high-level sounds, and is unlikely to cause the output levels to be so high that they will cause further hearing loss through over-exposure to sound.

A major complication in comparing children's requirements to that of adults for very low-level sounds (e.g. 40 dB SPL) is that we do not know how much gain should be prescribed for *adults* for such sounds! As discussed in Section 10.3.5, there is great uncertainty over what the compression threshold should be, and hence what the gain for very low-level sounds should be.

It seems desirable for children to have some experience receiving signals that are just above threshold. When adults are amplified in one ear only, their ability to process low-level information in that ear decreases in the months following hearing aid fitting.[589] We would not want to deny children the ability to process low-level signals by amplifying to such an extent that they get no practice listening to such signals. Low-level speech sounds may originate from either low vocal effort or a distant talker.

The inability of infants to alter a volume control, their decreased speech understanding at low input levels, and the likelihood of infants frequently receiving above-average input levels, all *mandate* the inclusion of wide dynamic range compression in hearing aids for infants. WDRC widens the range of sound levels in the environment that, after amplification, lie between threshold and discomfort.[810] With WDRC it is possible to amplify the range of sounds from soft speech to loud speech (approximately 55 dB SPL to 80 dB SPL) into the loudness range from *too soft*, or *a bit soft*, up to *too loud*.[1585] Without WDRC, the steep loudness growth that accompanies increasing sound level in people of any age with sensorineural hearing loss[1605] would result in speech being amplified outside this loudness range, or a highly saturated signal (with consequent distortion of speech cues) or both.

The greatest benefits of WDRC for speech perception will occur for low input levels (Section 6.5.1),[811] because WDRC increases audibility and, at low input levels, increasing audibility increases intelligibility. WDRC provides higher speech perception than linear amplification even for severe-profound hearing loss.[1140] Speech perception is invariably measured on older children, but there is no reason not to expect the same benefits for infants who, out of all patients, are least able to operate a volume control to compensate for changing speech levels.

The consequence of increasing the gain (relative to adults) more for low-level sounds than for high-level sounds is that a higher compression ratio will be prescribed for children than for adults. There is no direct evidence for children on how quickly compressors should change the gain as input varies. Because of the potentially high compression ratios, it seems safest for compressors to have release times longer (and perhaps much longer) than a few hundred milliseconds to minimize any increase of gain during the gaps between words in a signal of interest.[g] Infants learning speech presumably do not initially have the ability to recognize the amplified noise as being separate from the speech signal they are trying to unravel. The need for a very high compression ratio is reduced if the hearing aid has an adaptive noise reduction algorithm activated (Section 16.4.4), as the gain reduction provided by the noise reduction algorithm will in most cases be greatest when speech levels are highest.

The question of how frequency response shape should be altered for an infant or toddler relative to that prescribed for an adult is extremely difficult. There is no evidence to support an intentional variation from the shape prescribed for an adult with the same hearing loss. Unfortunately, there is also no evidence to show that the same response should be prescribed.

As outlined more generally in Section 10.3.4, the bandwidth over which speech is made audible should be as great as possible, ideally up to 10 kHz, although this is rarely achieved. A high upper limit of audibility ensures perception of unvoiced fricatives, particularly /s/, and the frication component of the voiced fricative /z/. For child and female talkers, these frication components may not have significant intensity below 6 kHz.[156, 1326] The sensation level of fricatives is lower for female talkers than for male talkers.[1702] Hearing-impaired children are less able than normal-hearing children to distinguish singular from plural nouns (in English) by detecting a final /s/ in the word, particularly for female talkers.[1702] Not surprisingly, the greater the hearing loss, the greater the inability to detect /s/, for male and female talkers alike.[1702]

It is almost certainly no coincidence that those sounds most difficult to make audible through hearing aids (high-frequency fricatives) are those for which children with hearing loss are most delayed in learning to produce.[1704] Obtaining audibility of high-frequency sounds is important not just for perception of other people, but also to enable children to monitor their own voice as they attempt to produce the sounds. Unfortunately, high-frequency sounds radiate from the mouth in a very directional manner, so despite the close proximity of a child's ear to his or her mouth, the level of high-frequency fricative sounds at the

g Hearing aids with an adaptive release time would also fulfill this requirement.

input to the hearing aid may be relatively low.[1424] Multi-channel WDRC partially undoes the low-frequency emphasis that the radiation pattern of the mouth causes, but children could potentially benefit from a special "own-voice amplification algorithm" that so far does not exist.

The greater the high-frequency hearing loss, the more important it is that sensation level in the high frequencies not be too great, as the greater the loss, the greater the rate at which loudness grows with sensation level, and the lower the value to speech perception that a high sensation levels provides.

The reader should recognize that the conclusions regarding frequency response emphasis and compression speed in this section are highly conjectural. Data to support or refute these conclusions do not exist. We do, however, have more evidence now than a decade ago about how much overall gain children with different degrees of hearing loss should have.

16.4.2 Threshold-based versus loudness-based procedures

The question of whether to base prescription on individually measured hearing thresholds or on individually measured supra-threshold loudness growth is easier to answer for children than it is for adults (and the younger the child, the easier the answer). There is no proven way to obtain loudness growth data from an infant, so there is no choice: we must base prescription on hearing thresholds alone. For older children, it is possible to measure loudness growth curves by representing different degrees of loudness pictorially. Unfortunately, children do not always interpret the pictures in the way that we expect.[1029] Categorical loudness rating is less reliable with children than with adults.[516] An alternative that has been successfully used with children as young as 4 years is to ask the children to match the loudness of a sound to the length of a line.[1604, 1605]

If children can reliably perform a loudness-rating task, the advantages and disadvantages of loudness scaling become similar to those for adults (Section 10.4.8). Of course, a loudness scale is no use unless one also has a validated prescription procedure that makes use of it. As discussed in Section 16.4.1, the balance between achieving some target loudness and achieving an output level that maximizes intelligibility may be different for children than it is for adults.

In summary, there are too many uncertainties to justify use of loudness scaling with children at the present time. It may be possible to derive some electrophysiological measures, such as ABR latency, ABR amplitude, acoustic reflex threshold, or DPOAE strength as surrogates for loudness measures.[394, 900, 1270] For the moment, this is a topic for research rather than clinical practice and, as for adults, there is the problem of what to do with the surrogate loudness measure once it has been obtained.

16.4.3 Allowing for small ear canals

REUG variation with age

As a child grows, so too does the length and volume of his or her ear canal. Ear canal length determines the resonant frequency of the unaided ear, and hence the frequency of the peak of the real-ear unaided gain (REUG) curve. The change in this peak frequency with increasing age is rapid. At birth, the peak of this curve is approximately 5 to 6 kHz but, on average, decreases to 3 kHz (only 10% above the average adult value) by the age of 2 to 3 years.[103, 419, 861, 984, 1914]

The REUG curve for a person directly affects the insertion gain received by that person. The panel in Section 10.2.4 discusses the appropriateness of taking an individual's REUG into account in hearing aid prescription, and concludes that it is not appropriate in the case of young children. Consequently, the prescription procedure recommended for infants in this book is based on real-ear aided gain (REAG) rather than insertion gain. Although the REUG characteristics of the individual therefore have no effect on the prescription, the frequency of the peak in the REUG curve does allow us to estimate the length of the ear canal of infants, and this is useful to know when placing probe tubes. Figure 16.3 shows the typical varia-

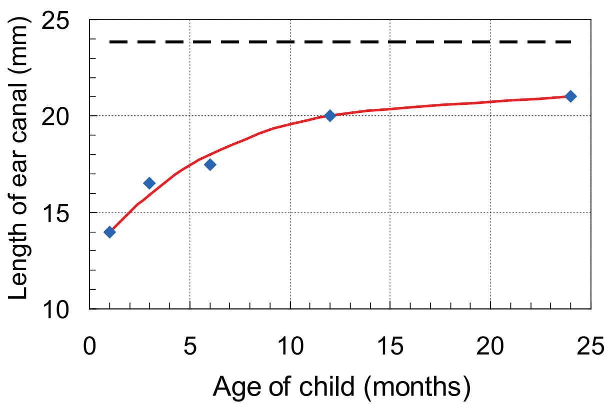

Figure 16.3 Typical length of the ear canal as a function of age. Solid line is a smooth curve fitted to the data and dashed line shows the average length of the adult ear canal.[861, 1542]

tion of ear canal length inferred from the peaks in REUG curves using a two-cylinder model of the ear canal and concha.[861]

RECD variation with age

The second change with age relates to the volume of the residual ear canal when aided. This volume, in conjunction with the input impedance of the middle ear, largely determines the real-ear to coupler difference (RECD). For babies, the impedance of the canal walls also affects RECD.[862] Changes in RECD occur most rapidly during the first year of a baby's life but continue for several years, gradually approaching adult values.

Figure 16.4 shows, for several ages, average RECD values for custom earmolds. They are also shown in tabular form in Table 16.4 for convenience. These curves for children are in broad agreement with those published by Bagatto et al (2005), except that the values in Figure 16.4 are approximately 3 dB higher from 4 to 6 kHz and 2 dB lower from 1 to 1.5 kHz.

These curves show:

- RECD broadly increases with frequency, partly because the effective volume of the ear decreases as frequency rises (Section 4.1.1), and partly because leakage past the earmold reduces SPL in the canal in the low frequencies as frequency decreases.
- RECD is greatest for the youngest ages because the residual volume is smallest for infants.
- RECD is smaller at 4 kHz than at 2 kHz for adults, because the reference HA2 2-cc coupler effectively contains an acoustic horn that increases high-frequency SPL in the coupler, whereas the custom earmolds on which these curves are based have constant diameter tubing.

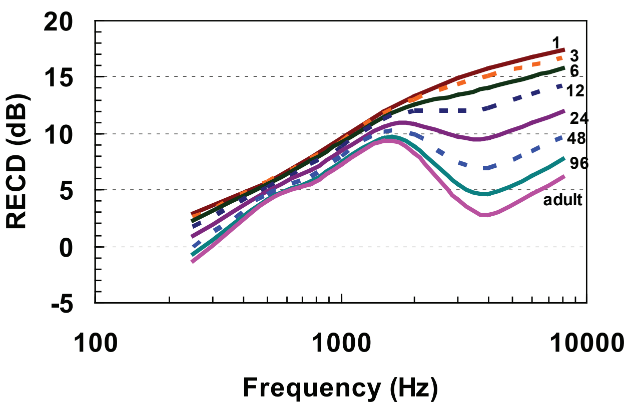

Figure 16.4 Average RECD for children aged from 1 to 96 months and for adults, as indicated for each curve. The RECD shows the SPL measured in the ear canal with an individual mold in place relative to the SPL measured in an HA2 2-cc coupler. Measurements on the children and adults extended only to 4 and 6 kHz respectively, so the higher frequency data shown are extrapolations.

Table 16.4 Average values of RECD for children of different ages. These values apply when the real-ear SPL has been measured using an individual earmold, and the coupler SPL has been measured using a HA2 2-cc coupler. Data have been derived by fitting mathematical functions across frequency and age to the data of 284 ears for hearing aid wearers from 1 month to 20 years of age (two-thirds of whom were under two years of age) measured in several NAL experiments, and the averaged data for 108 adults from the first five experiments cited in Table 4.3, and shown in row 2 of that Table. All data apply to unvented earmolds with typical leakage and constant 2 mm sound bores.

Age (months)	Frequency (Hz)						
	250	500	1000	2000	3000	4000	6000
1	3	6	10	13	15	16	17
3	3	6	9	13	14	15	16
6	2	5	9	13	13	14	15
12	2	5	9	12	12	12	13
24	1	5	8	11	10	10	11
48	0	5	8	10	7	7	8
96	-1	4	8	9	6	5	6
adult	-1	4	7	8	4	3	5

- RECD changes very little from 1 month to 3 months of age, probably because 1-month old infants have such tiny ears, their earmolds tend to be made very shallow, and possibly because they have more flexible canal walls which increases the effective acoustic volume medial to the earmold.

Values for individual children are, of course, scattered above and below the average values shown in Figure 16.4. This scatter has a normal distribution and standard deviations around 4 dB up to 3 kHz and 5 dB at 4 kHz. This means that for the data on which these curves were based, 5% of the children had RECDs more than 8 dB (or 10 dB at 4 kHz) from the predicted values. Although these data were collected by diverse, geographically separated clinicians, these clinicians received uniform training in hearing aid fitting and RECD measurement. A higher spread is likely if deeper or shallower earmolds than those normally used in Australian pediatric services were to be employed.

As explained in Section 11.4, the RECD values appropriate to an individual affect both the coupler gain and OSPL90 values needed to achieve a target REAG and real-ear saturation response (RESR). The accompanying panel shows the steps involved in prescribing and adjusting gain and OSPL90. Once RECD has been allowed for, there is no need for any further verification of real-ear gain. Adjustment of the hearing aids to the prescription targets is thus achieved while the hearing aid is in the test box, and only a single real-ear measurement (i.e. of RECD) has to be obtained from the child. Remember that the RECD values allow us to predict the SPL value created by the *hearing aid* in the ear canal. They do not predict the SPL that directly enters the canal via a vent or leakage paths (Section 5.3.1). Except for high-gain hearing aids where the vent-transmitted sound is inconsequential, the vent/leak-transmitted sound *must* be allowed for separately in the hearing aid fitting software that calculates the coupler gain needed.

Immittance tips

The discussion in this section has so far centered on RECD values of custom earmolds, and their application in adjusting hearing aid gain. As outlined in Section 16.2.2, RECD values also affect the amount of hearing loss that a child appears to have when measured with insert earphones. For infants, it is convenient to deliver sound to the ear canal using an immittance tip of diameter appropriate to the ear canal. The probe tube can be coupled to the immittance tip by wrapping both in plastic film (as used to cover food). The probe tube and immittance tip are then inserted together.[72, 74] Probe depth issues are the same as for custom molds, as shown in the panel. Average age-appropriate values are shown in Table 16.5.

Middle ear disorders

It will sometimes be necessary to fit hearing aids to a child with a ***ventilation tube*** (also known as ***pressure equalization tube***, ***grommet***, and ***tympanostomy tube***) in the eardrum or a perforation of the eardrum. Any opening in the eardrum has two effects on the hearing aid performance. First, for low-frequency sounds, it connects the middle ear cavity directly to the residual ear canal volume, without the intervening effect of the eardrum. On average, this decreases the RECD below about 1 kHz by about 10 dB.[1148, 1149] The hearing aid must therefore have additional gain and OSPL90 to achieve the same SPL that would be

Table 16.5 Average values of RECD for children of different ages. These values apply when the real-ear SPL has been measured using an immittance tip, and are based on the regression equations published by Bagatto et al (2005).

Age (months)	Frequency (Hz)						
	250	500	1000	2000	3000	4000	6000
1	0	4	8	10	12	16	17
3	0	4	8	10	11	15	16
6	0	4	8	9	10	14	14
12	0	4	8	8	8	12	11
24	0	4	8	5	5	6	5
>24	0	4	8	8	6	9	9

Should RECD be measured individually for each earmold and child or predicted from age-appropriate average data?

The almost universally recommended answer is that it should be measured for each child and each new earmold, but the decision is a close one.

In favor of individual measurement:

- Individual RECDs can be 10 dB (or even a little more) above or below average values; and if these values are accurate, the real-ear gain that results from using the predicted RECD values will be 10 dB greater than or less than the target gains.
- Faulty earmolds, such as those with a crimped section of tubing, will be detected by the RECD measurement.

In favor of using age-appropriate values:

- Many factors can cause incorrect RECD values to be measured. These factors include:
  - probe tubes creating additional leakage around the earmold;
  - probe tubes becoming blocked with wax or being excessively squashed between the earmold and the ear canal;
  - the probe tube not being inserted sufficiently past the end of the earmold, or sufficiently close to the eardrum;
  - probe tube microphones, or coupler microphones, being incorrectly calibrated; or
  - measurement of RECD using a configuration (coupler, transducer, coupling to the ear) different from that expected by the software into which the measured values are entered.
- If incorrect values are not recognized as being in error, adopting these values can create errors in gain larger than those that can result from using predicted values.
- Crimped tubing can be detected by visual inspection.

The better approach probably depends on the level of training that clinicians have with RECD measurement and its pitfalls, the clinical time and equipment available, and the certainty about the equipment's calibration.

Certainly, just ignoring the difference between infant RECD and adult RECD and prescribing as though the infant was an adult is totally inappropriate. If RECD is not measured, the age-appropriate values shown in Table 16.4 (for custom molds) or Table 16.5 (for immittance tips) must be used.

present at the eardrum with no ventilation tube or perforation present. This additional gain is automatically prescribed if the prescription is based on individually measured RECD. No additional gain or OSPL90 is needed at those frequencies where vent-transmitted sound causes the hearing aid to have a gain of 0 dB (Section 5.3.1).

Individual measurement of RECD does not allow for the second effect of any opening in the eardrum. Because the opening allows low-frequency sound to reach both sides of the eardrum, the sensitivity of the eardrum to low-frequency sounds is decreased, even if the SPL achieved in the ear canal is the same as for an intact eardrum. The loss of effective stimulation of the cochlea can be estimated from an acoustic model of a hearing aid, ear canal, and middle ear system. Such modeling indicates that, even when the SPL in the ear canal is held constant, adding a ventilation tube with an internal diameter of 1.3 mm and a length of 1.8 mm causes the input to the cochlea at 500 Hz to decrease by 12 dB.[312] The effects are less than this above 500 Hz, but grow rapidly below 500 Hz.

Otitis media with effusion, by contrast, causes RECD to increase by a few decibels because the increased stiffness of the eardrum and fluid mass behind the eardrum both act to greatly decrease the effective volume of the middle ear.[1149a] Although the effusion increases RECD and hence SPL in the ear canal, this increase is much less than the decreased transmission through the middle ear system.

Prescribing and adjusting hearing aids for infants

1. Obtain at least one low-frequency threshold and at least one high-frequency threshold for each ear.
2. Prescribe RESR, using the NAL-NL2 or DSLm[i/o] prescription software (prescriptions for RESR are similar). Alternatively, convert thresholds to equivalent adult average dB HL, as shown in Section 16.2.2, and use the RESR prescription shown in Tables 10.3 and 10.4.
3. Prescribe REAG, using the prescription procedure you have chosen. As you <u>*will*</u> be using a WDRC hearing aid, calculate the target gain for at least the input levels of 50 and 80 dB SPL.
4. Estimate the RECD based on the child's age from Table 16.3, or measure it as described in Section 4.2.2 and using the following information.
   - If the child is too active (most will be) to use the 6 kHz probe tone method of determining insertion depth (Section 4.3.1), insert the probe tube past the inter-tragal notch by 15 mm for babies from 6 to 12 months, by 20 mm for children from one to five years, and by 25 mm for older children.[1211] For babies 2 to 6 months old, insert the probe tube 11 mm past the ear canal entrance.[72] As can be deduced from Figure 16.3, this will result in the probe being 4 mm from the ear drum at two months of age and 7 mm from the drum at six months of age, for babies of average size for age. Either of these distances is sufficiently close to the drum to enable accurate measurements up to 5 kHz (Table 4.4).
   - These distances are approximate guidelines; the location of the probe tip relative to the eardrum during insertion should be observed by otoscopy and a smaller insertion depth should be used when required. In all cases the probe tube must extend by at least 2 mm (more in bigger ears) past the end of the earmold.[72, 74]
   - It is more important to avoid causing pain (and hence fear of the real-ear measurement equipment) than it is to obtain accurate measurements at and above 4 kHz, especially the first time a probe microphone is used.[1582]
   - Prior to measuring RECD, apply lubricating cream to the mold so that it can easily be inserted without disturbing the probe and so that feedback is minimized.[1914] Alternatively, if the canal is wide enough, the earmold can be ordered with an extra hole through which the probe tube is inserted. This makes it very easy to control the depth of insertion, and makes it less likely that the probe tube will increase leakage around the earmold.
   - If the child is old enough, allow him or her to hold and look into a mirror while you position the probe and earmold. This keeps the child still.[1595]
5. Calculate the target OSPL90 curve, by subtracting RECD from the target RESR.
6. Calculate the target Coupler Gain for 50 and 80 dB SPL input levels, by subtracting RECD from the target REAG.
   - Microphone location effects could also be allowed for, but as these are small for the BTE hearing aids that will invariably be fitted to infants, this correction can be overlooked.
   - At any frequency at which the REAG target is 0 dB or less, the coupler gain target can be ignored, provided 0 dB gain can be achieved by sound entering via the vent or leakage path (Figure 5.13). If you are relying on vent-transmitted sound to achieve the gain target, make sure that the mold has a vent!
7. Adjust the hearing aid in a test box to achieve a reasonable match to the OSPL90 and Coupler Gain targets.

Although you could easily do steps 2 to 6 manually, prescription is much easier if you use software specially prepared for the task. Both the NAL-NL2 and the DSLm[i/o] software programs, and the various manufacturers' implementations of them, will allow you to enter thresholds in a variety of forms. They will also allow you to enter individual values of RECD, but will use age-appropriate values if you do not enter individual RECD values. DSLm[i/o] generally prescribes more gain than NAL-NL2, particularly for patients with steeply sloping hearing loss.

16.4.4 Signal processing features

Each of the features in this section has been covered in detail in Chapters 7 and 8. The following discusses the applicability of these features to infants and young children.

Directional microphones

Switchable directional microphones are probably as useful for older children as they are for adults. Hearing aids permanently in directional mode are as unacceptable for infants and young children as they are for adults because of their disadvantages in many situations: increased pick-up of wind noise and slightly decreased sensitivity when a wanted sound (such as a warning sound, the noise of an approaching car, or comments by other children in a classroom or elsewhere) comes from behind or from the sides. It seems an unworkable solution to expect a parent or carer to switch between directional and omni-directional modes in different situations, especially when one realizes that it takes only a head-turn by the infant or child for the other mode to be optimal.

While simply never using directional microphones for infants might at first seem like an appropriately cautious approach, this solution means that the only feature in modern hearing aids that significantly improves SNR in noisy places would be unavailable to those who most need it - young children. As reviewed in Section 16.4.1, young children need a higher SNR than adults if speech is to be intelligible. Like adults, infants and young children will have the greatest difficulty understanding speech when it is partially masked by noise, so it is worth finding a way for them to gain the benefit of directional microphones if at all possible.

The change to the signal caused by directional microphones is a linear, low-distortion effect, similar to just changing the noise level. Thus, the magnitude of the benefit (or disadvantage) in decibels of SNR change and the impact of the environment on benefit should be no different for infants and young children than that experienced by older children or adults, as reviewed in Section 7.3, and as directly observed in children.[645] It's just that younger children, who are still learning language, more often than anyone else *need* the SNR to be improved.

It is important to understand that current directional microphones are not all *that* directional, particularly indoors where reverberation limits the disadvantages of directional microphones just as much as it limits their advantages. That is, just as directional microphones typically improve SNR by only around 2 to 3 dB when the wearer is looking in the general direction of the talker, they also decrease SNR by only around 2 to 3 dB when the wearer is looking away from the talker. Greater benefits, and presumably disadvantages, are observable if children are tested at close distances in artificial low-reverberation environments, such as test booths.[645]

Measurement of the looking behavior of children aged 11 to 78 months in natural listening situations (homes, play groups, playgrounds) showed that children looked at the dominant talker approximately 40% of the time the talker was speaking.[284] Surprisingly, this proportion was not affected by the age of the participants, nor by whether they had normal hearing or hearing loss.

Acoustic analysis in each individual listening situation showed that, relative to omni-directional microphones, directional microphones on average improved SNR by 2.4 dB when the children looked in the general direction of the talker and decreased SNR by 1.6 dB when they looked away. The overall "net benefit" that a directional microphone could provide can then be calculated by weighting its effect on SNR by the proportion of time it has this effect. The resulting net benefit averaged across listening situations was a 0.02 dB decrease in SNR – a change so small to be of no consequence. Furthermore, the effect of the directional microphone was assessed in the absence of any compression, which as outlined in Section 7.3.3, partly reverses the decrease in signal level caused by a directional microphone when a wanted talker is to the rear or sides.

This nil result suggests that infants and young children should routinely be fit with advanced directional microphones, and they should receive considerable benefit from them, for the following reasons:

- The experimental results were obtained on normal-hearing children, and children with hearing loss wearing omni-directional microphones. It is likely (but by no means proven) that children wearing directional microphones will notice that looking at the talker improves the clarity of the signal and will adapt their behavior to look at the talker more often than children wearing omni-directional microphones. A study of 4 to 17-year old children in the classroom indicated that hearing-impaired children were more likely than normal-hearing children to orient towards talkers, possibly to obtain visual cues to help understand the speech.[1507] It is also possible that suitably instructed carers will be able to engage the child's attention sufficiently to affect their look direction whenever they wish to communicate in adverse listening situations. Any orienting behavior by the child, even if aimed only at increasing their visual monitoring of a talker, facilitates directional advantage in situations that may at first appear to be disadvantageous for directional microphone use.

- Several hearing aids on the market simultaneously process the microphone outputs in directional and omni-directional modes, and automatically select the mode for which the output signal has the greater SNR. The SNR is estimated on the basis of the depth of the speech-like modulations in amplitude (Section 8.1.1). When the dominant talker is to the rear or sides of the child, such hearing aids will automatically select the omni-directional mode.
- Finally, even if neither the children wearing directional hearing aids nor their carers modified their behavior; even if automatic hearing aids made such extremely bad decisions that they randomly selected omni or directional modes (they don't); and even if compression never reduced the effect of directivity on backward or rearward dominant talkers (it does), the experimental worst-case result is that, averaged across listening situations, directional microphones would cause neither an increase nor a decrease in SNR.

One concern that is sometimes expressed about directional microphones is that they will interfere with localization ability. This concern is understandable as directional responses do create greater localization error in the left-right direction for high-frequency sounds, particularly when the hearing aids in each ear independently adapt the shape of their polar sensitivity patterns.[884, 1837] The concern is probably overemphasized, however, as directional responses have no effect on left-right localization accuracy for broadband sounds.[884] Furthermore, for hearing-impaired people (adults, but presumably also children) front-back errors are far more common than left-right errors, and the proportion of front-back errors with binaural directional microphones is less than or equal to that for binaural omni microphones.[884, 887] Hearing aids that are directional for high frequencies and omni-directional for low frequencies, and which therefore mimic the normal unaided directivity provided by the pinnae, certainly lead to fewer front-back confusions than omni-directional microphones (Section 15.1.2).[884, 887]

The evidence-guided recommendation of this author to fit automatic directional microphones to all children old enough to respond to sounds by looking at them is not shared by all experts in pediatric audiology. It has nonetheless been standard fitting practice throughout Australia since 2009 for all children old enough to support their own head and control their look direction (around 6 months of age). An important caveat is that invoking the directional mode should not unduly restrict the ability to achieve the low-frequency gain prescribed for the child. Depending on the rules the particular hearing aid uses to automatically engage the directional mode, this may preclude directional microphones for some children with severe or profound loss.[h]

Adaptive noise reduction

Adaptive noise reduction should have the same advantages for children as it does for adults: listening comfort should be increased, listening effort should be decreased, and speech intelligibility should be left unchanged. The first two expectations arise because adaptive noise reduction usually improves overall SNR, and the last expectation arises because it leaves SNR unchanged at every frequency (Section 8.1.2).

So far, there have been few direct evaluations of the impact of adaptive noise reduction on speech intelligibility or quality in children. Four studies have shown no effect of adaptive noise reduction on speech intelligibility in some or all conditions,[663a, 1131, 1421b, 1696] but with the following exceptions. Gustafson et al. (submitted) found that there was no significant increase in speech understanding for one hearing aid with adaptive noise reduction that, by objective measurement, improved overall SNR by 2 dB. A second algorithm, however, improved overall objectively measured SNR by 7 dB, and also significantly improved speech

[h] Loss of low-frequency audibility is minimized or eliminated if directivity is engaged only in higher second-level environments (where target gains are least) and if directivity is not engaged in the lowest frequency channel(s) (where the low-cut characteristic of directional microphones is the most difficult to compensate).

understanding. This second algorithm also resulted in improvements in response time (as a surrogate for listening effort) and perceived quality.[663a] In the second exception, Pittman (2011) found that adaptive noise reduction significantly increased speech understanding for children over 10 years of age, but had no significant effect for younger children.[1421b]

These are positive results because, at the very least, they indicate that children do not have to suffer any loss of speech understanding in order to achieve the decrease in apparent noise that adaptive noise reduction offers. The positive results are supported by two much less direct evaluations.

The first experiment examined the ability of adults to learn to discriminate unfamiliar words requiring speech sound contrasts that do not appear in the participants' native language.[1131] Adaptive noise reduction processing neither increased nor decreased the accuracy with which the unfamiliar contrasts could be discriminated.

The second experiment is a longitudinal study of children who have been fitted with hearing aids that contain adaptive noise reduction, as well as automatic directional microphones.[67] There is no control condition, so it is not possible to assess the effects of adaptive noise reduction separate from other benefits provided by the hearing aids. On average, the children acquired expressive and receptive language skills and speech production skills at greater than the normal rate. This suggests that it is unlikely that adaptive noise suppression has an adverse effect.

Of course, these experiments do not provide a blanket endorsement of all adaptive noise reduction algorithms for all noises and all children. The major concern is that if the noise reduction algorithm operates independently of the degree of hearing loss (which some do), it is possible that the adaptive system will reduce the gain sufficiently that speech cues are made inaudible. This is most likely to occur for lower input levels of speech, for children with greater degrees of hearing loss, and/or for algorithms that more aggressively reduce gain in any frequency region as the SNR in that region decreases.

This concern, combined with the good potential of adaptive noise reduction, leaves the clinician in a slightly difficult position: it is likely that adaptive noise reduction will improve the experience of listening for most if not all children. However, unless the noise reduction algorithm is linked to the audiogram such that gain is never reduced to the point where previously audible speech is rendered inaudible, or unless the noise reduction algorithm only activates at higher input levels, it is safest to select the mildest form of noise reduction available. These mild forms, which provide the least gain reduction at each frequency as the SNR decreases, also can provide only mild benefit.

The few remaining concerns about using adaptive noise reduction systems for all children should evaporate when all manufacturers configure their algorithms so that gain is never reduced by more than the amount needed to lower background noise in each frequency region to the child's hearing threshold in that same frequency region.

If technical information is not available, the depth of gain reduction can be measured in a test box by comparing the gain provided to a signal with speech-like dynamics and spectral shape to the gain provided to a steady noise with a speech-like spectrum at the same overall level.

Feedback cancellation

Feedback oscillation can adversely affect hearing aid amplification for anyone, but this is particularly likely for infants because there are often reflecting surfaces in close proximity to the infant's head, and because the ear canal rapidly changes shape and size over the first two years of life. Should feedback cancellation be used for infants and children too young to report on the sounds they hear? The following are some of the considerations:

- Feedback cancellation enables higher gain to be achieved than is possible without it, and so can directly improve the audibility of speech.
- Feedback cancellation can produce artifacts (i.e. distortion or extraneous sounds) when it is actively and aggressively preventing feedback oscillation. These sounds are not as bad as the feedback oscillation they are preventing, but unlike continuous oscillation, they may not be evident to anyone other than the child wearing the hearing aid.[i]

[i] Sub-oscillatory feedback (Section 4.7.2) is just as bad as the distortion that can be caused by feedback cancellation; its effects also may not be evident to anyone other than the hearing aid wearer.

- Feedback cancellation is not likely to produce artifacts if it is not needed to prevent oscillation, even though the feature is enabled.

The best situation is if the target gain can be achieved without needing feedback cancellation activated. The worst situation is frequent or continuous feedback oscillation. A reasonable compromise is to routinely activate feedback cancellation, but check at each appointment whether the hearing aid is stable without it, and use this information to guide whether new earmolds are needed.

If the hearing aid is only barely stable when the feedback cancellation is active, even with well-fitting earmolds, the high-frequency gain for low-level sounds may have to be reduced. It would be useful if the data logging feature in hearing aids indicated the extent to which the feedback cancellation feature had been relied on to achieve stability. Manufacturers take note!

Frequency lowering

A difficult issue is whether frequency-lowering hearing aids should be used for children too young to report whether this feature assists them to understand speech. On the one hand, knowledge of which patients will obtain benefit from frequency lowering, and how it should be adjusted for optimal benefit, has not been established for adults (Section 8.3) let alone for children. There is a small risk in training a young child's auditory system to use modified cues that may turn out to have been sub-optimal when further research becomes available.

On the other hand, many children with high-frequency hearing loss in the profound and upper severe range cannot have adequate access to high-frequency cues without frequency lowering or cochlear implantation. Unquestionably there are some children for whom frequency lowering can make the difference between hearing /s/ sounds and not hearing them. As discussed in Section 16.4.1, the spectral shape with which this sound is produced varies markedly between individuals in any event, so some artificial distortion of its spectrum does not seem critical. As a further advantage, permanently engaged frequency lowering is likely to provide some protection against feedback oscillation.

It therefore seems reasonable for clinicians to use a mild amount of frequency lowering if there is any reason to believe that the child cannot hear the high-frequency sounds of speech without it (Section 16.6). While there will be many children who do not receive any net benefit from frequency lowering, it is rare for it to be harmful to speech intelligibility, so its use is much more likely to increase speech perception ability than to decrease it. As with all aspects of hearing aids, performance and progress of the child should be carefully monitored. Because the processing deliberately alters the spectrum of sounds, children are likely to take several months to fully learn to recognize sounds and to discriminate between different high-frequency sounds.[627]

At least one test analyzer manufacturer offers a broadband test signal with a notch and narrow peak in its spectrum. Such signals are particularly valuable for fitting frequency lowering to children too young to give reliable feedback about sound quality. A display of the spectrum in the ear canal, plotted relative to hearing threshold, shows whether the lowering is sufficient to achieve audibility of speech in the region of the spectral peak.[625]

16.4.5 Assistive listening devices (ALDs)

Wireless systems and other assistive devices are covered in detail in Chapter 3. This section provides some additional comments related to their use by parents.

Frequency modulation (FM) and other types of wireless systems improve signal quality and hence intelligibility much more than any signal processing scheme located entirely within the hearing aid. Although FM systems will provide an improved signal for any hearing-impaired person, they are particularly relevant to children. This is partly because children need a higher SNR than adults, and partly because children spend a lot of their day in situations where there is one dominant talker who can conveniently wear a transmitter. More advanced FM systems enable two or more transmitters to be used so that if there is more than one talker in the room (e.g. mother and father, parents and siblings, grandparents) multiple microphone-transmitters can be used. This will enable the child to better follow conversations between other persons in the room, exposing them to more language input with a high signal quality.

In addition to their use in schools, FM systems can be used at home and elsewhere. FM systems will help a child of any age understand conversation during car travel, television watching, sports coaching, outings, etc.[1055, 1057] As in schools, a major problem to overcome is ensuring that the child receives a signal from the transmitter only when the words spoken by the person wearing the transmitter would normally be heard by a child in that situation, and that the local microphone has enough sensitivity to hear other sound sources and talkers near the child at all other times.[205, 1201] Speech-operated switching would thus be just as useful in the home or social situations as it would be at school.

Speech-operated switching is not a complete solution, however. Parents, teachers and care-givers still have to remember to switch the FM transmitter off whenever they are engaging in conversations that would not be helpful for the child to hear. It is inappropriate for the child to receive a transmission from someone else's conversation in another room while the child is engaging in some task requiring him or her to hear a nearby talker or requiring quiet concentration. This could perhaps be demonstrated to parents by putting a signal from some remote source in their ears while they attempt to converse with someone in front of them. This may reinforce the importance of the child always receiving the correct input signal.

It is worth ensuring that switching off the transmitter while leaving the receiver connected to the hearing aid does not cause any adverse effects. All FM systems now contain squelch, which prevents the receiver from injecting high-level white noise into the hearing aid when the receiver is not receiving a transmission. FM systems with speech operated switching will also ensure that the local (i.e. hearing aid) microphone resumes full sensitivity, but it is worth checking to ensure that this happens with the system you are using.

Body-level FM receivers are sometimes (infrequently) used, with the output either fed to an ear-level hearing aid, or directly to the ear via headphones. If the receiver has a local microphone, then the receiver should be worn at chest level rather than at belt level, as the latter will be shadowed by a desk when the child is seated.[1056] Body-level receivers with a local microphone should not be worn under clothing, even if that is less conspicuous, as clothing rubbing on the microphone port will create noise. Body-level receivers with self-contained hearing aids tend not to have the advanced compression circuitry available in personal BTE hearing aids. On the other hand, the greater separation of the microphone from the ear canal enables the hearing aid to have a higher gain without feedback oscillation. Furthermore, the larger controls are easier for children with fine-motor problems.[1056]

16.5 Verifying Real-Ear Performance

If the individual RECD procedure discussed in Section 16.4.3 has been carried out, and no mistakes have been made in using it to achieve the required coupler gain, there is no necessity to perform any further evaluation of real-ear gain. If time permits, and if the child is sufficiently cooperative, the REAG can be measured with probe-tube equipment to check that no errors have been made. Assuming you have, or your software has, correctly allowed for the effects of any vent or leak in the earmold, the results should be entirely consistent with the test-box adjustment of the hearing aid to the target calculated using the individual RECD measurement. If, for example, the gain was deficient at 4 kHz relative to the target when the aid was adjusted in the text box, the measured REAG should be deficient to the same degree relative to the REAG target. Any inconsistency indicates that an error was made in the RECD measurement, generation of the coupler gain target, test-box adjustment, or REAG measurement.

In the past, ***aided thresholds*** were commonly used to verify a fitting. These measurements should be viewed as a possible supplement to the RECD procedure or direct measurement of real-ear gain, *not* as an alternative to these electroacoustic measurements. As with adult testing, aided threshold determination is slower, less accurate, less detailed (in that only a few frequencies can realistically be measured), applies to only one input level (because of the impact of compression on gain), and may give invalid results if a threshold is masked by ambient noise (Section 4.6).

The following are some arguments that have been made in favor of measuring aided thresholds for verifying or validating hearing aid fittings. Only the first is sufficiently strong that it justifies measuring aided thresholds *instead* of real-ear gain.

- They can be used when measurement of real-ear gain is not possible, which is the case with bone-conduction hearing aids and cochlear implants.
- An aided threshold at the level expected, given the hearing aid coupler gain and unaided hearing threshold, provides further confirmation of the child's unaided thresholds. Calculating the expected aided thresholds is tricky, however, unless suitable software is available.

- In the case of profound hearing loss, aided thresholds at the expected levels confirm that the unaided thresholds were not based solely on vibratory sensations.
- The softest sound that a child can hear is meaningful. As is often said, although audibility does not guarantee intelligibility, lack of audibility does guarantee lack of intelligibility.
- They demonstrate to the parents that the child is capable of reacting to sound (and in some cases this may be the first time the parents have seen their child react to *any* sound). However, any sound can be used for this purpose.
- They can be used to predict whether soft speech will be audible in different frequency regions.[341] However, real-ear speech mapping with soft speech will show this more accurately and quickly.
- They demonstrate that the hearing aid maximum output exceeds the child's hearing threshold at each frequency tested. However, OSPL90 will always exceed threshold if it has been prescribed using either the DSL or NAL procedures and adjusted in a text box using actual or estimated RECD values.

If, for some reason, neither the RECD procedure nor real-ear gain measurement can be carried out (one of them should be!), aided threshold testing provides an inferior fallback means of verifying real-ear performance. The thresholds obtained should be compared to those prescribed by the selection procedure used.

16.6 Evaluating Aided Performance

Simply verifying that the prescribed response has been obtained is not enough (any more than it is for adults). The effectiveness of the hearing aids in providing auditory information to the child also has to be established. The appropriateness of the hearing aids (as well as their continued correct operation) must at first be checked frequently, perhaps every few weeks. The time between checks can increase to perhaps every 6 to 12 months once both audiologist and parents are convinced that everything that can be done, has been done.

There are several methods by which the effectiveness of amplification can be evaluated. Methods based on speech tests are absolute evaluations in that they indicate how well a particular amplification scheme works, against some criterion. These tend to be time consuming to perform, and are therefore not usually suitable for comparing multiple amplification schemes to see which is the most effective. The paired-comparison technique, by contrast, provides no information about how effective any scheme is, but can be used to select which of several schemes is preferred for clarity or other listening criteria.

16.6.1 Speech tests

Understanding speech is the major goal of hearing aid fitting, so measuring speech understanding is the most direct way to assess how well the goal has been achieved. In fact, it is the only way to assess achievement of the goal. That said, it is far from simple to know whether the performance of a child on a particular test is less than that child could be capable of with different amplification characteristics or an implant and, if so, how the hearing aid should be changed to improve performance.

Hearing aid performance can be assessed by presenting in the sound field any speech test that has a level of difficulty appropriate to the developmental level and vocabulary of the child. The report of a Pediatric Working Group[125] contains an extensive listing of tests commonly used with children. Features covered include the type of speech material (language level and phoneme, word or sentence construction), presentation method (recorded or live voice, and auditory or auditory-visual), the number of items and lists available, response choices (open set or closed set), the type of response (verbal, picture-pointing, or other action), and the range of ages and hearing losses considered to be appropriate. Other critical issues that need to be considered if the results are to be at all predictive of the child's speech understanding in real life are the SPL of the speech material and the SPL and type of any competing sounds.

Our goal is that children understand speech presented at a range of levels. Consequently, it is not sufficient to evaluate performance only in quiet and at typical speech levels (approximately 65 dB SPL or 45 dB HL sound field level). Performance should also be assessed at 55 dB SPL or 35 dB HL. This book cannot do justice to the complexities associated with determining test protocols for children of different (language and cognitive) ages, and the reader is referred to Madell (2008).

The difficulties associated with using a speech test to evaluate the effectiveness of a hearing aid are the same as those with adults (Section 14.2.1), plus any associated with retaining the child's cooperation for the duration of the test. Because children need receptive language ability, and for some tests expressive language ability, sufficient to successfully undertake the test, speech test results are affected not only by the child's current ability to understand speech, but also by the child's past experience with understanding speech that has resulted in his or her accumulated language acquisition. This dependence on language ability can be viewed as a limitation if the purpose is to evaluate the current hearing aids. Alternatively, the dependence on language can be viewed as a desirable feature if the purpose is to evaluate the success of the entire habilitation program so far, including the hearing aids.

If speech test performance is in any way unsatisfactory, performance should be measured for one ear at a time, just by turning off the opposite hearing aid but leaving it in the ear. As noted in Section 16.1, there may be some children with asymmetrical speech processing ability who understand speech better with one ear aided than with two, but this is a greatly under-researched topic, and even less is known about the prevalence of binaural interference in children, or its reversibility, than it is for adults (Section 15.4.2).

If responses to speech tests are not only scored to find a proportion correct, but are also analyzed to find the types of phoneme confusions being made, inferences can be made about which speech cues are not being perceived, and potentially the frequency region from which they arise. This detailed information may be useful for hearing aid fine tuning and for setting goals in intervention programs. Other potential actions arising from poorer than expected speech test results, whichever way the test is scored, include greater use of wireless assistive devices and consideration of cochlear implantation.

16.6.2 Paired-comparison tests

Although both the NAL-NL2 and the DSLm[i/o] prescription methods have been designed to provide the amplification that, on average, children prefer and perform best with, all children are not average, and some will prefer and/or perform better with responses different from that prescribed by either of these methods.[288]

Just as with adults (Section 12.2), children aged six or older, and at least some five-year olds, can indicate which of two hearing aid responses they prefer when they are presented with two alternatives in quick succession.[282, 505, 509] It appears that with children up to age ten, more reliable responses can be obtained if the material is presented via auditory-visual means rather than via auditory alone.[282] It is not clear if this is because the visual component maintains the children's interest or because it makes the continuous discourse easier to follow, thus preventing all the alternatives from being so difficult that the child cannot tell which one is better. The same frequency response shape is optimal in the auditory-visual mode as in the auditory mode.[282]

An alternative to paired-comparison tests is to have children give absolute or categorical ratings of sound quality (Section 12.2.2). They are not recommended, however, as they are less sensitive than paired comparisons at detecting differences in amplification effectiveness.[505]

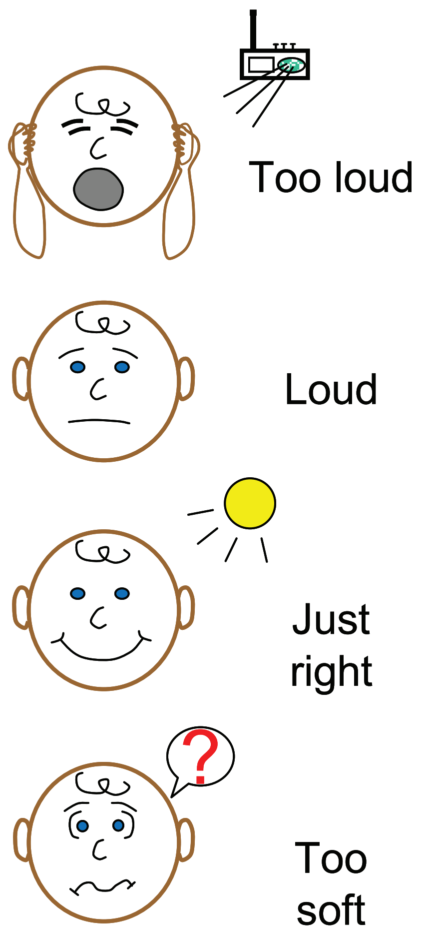

Figure 16.5 Verbal and pictorial loudness categories used for evaluation of loudness comfort and discomfort while wearing hearing aids.

16.6.3 Evaluation of discomfort

Maximum output must carefully be prescribed,[j] but even so, its suitability for the individual child must be evaluated. The most essential part of this evaluation is to ensure that the hearing aid does not cause loudness discomfort.

Several investigators have recommended the use of face icons to represent different loudness categories when a child's Loudness Discomfort Level (LDL) is measured.[211, 860, 1391] The technique appears to be suitable at least down to the age of seven.[860] Although this book does not recommend the routine *measurement* of LDL, even for adults (Section 11.7), loudness can be rated to ensure that the hearing aid can never cause loudness discomfort. Details are given in the accompanying panel and in Figure 16.5. The pictures have been adapted from the three pictures used by Byrne (1982) and the five pictures used by Kawell et al. (1988).[k]

By using a different technique, referred to as the ***tangible excess method*** the minimum age (or perhaps language level) at which LDL can be measured can be lowered slightly.[1109] This technique requires the clinician to first train the child to indicate *Stop* when excessive water, excessive weight, or an excessive number of small toys are placed into a container. The child is then asked to indicate *Stop* when the level of sound becomes excessive.

Evaluating maximum output in children seven years or older

The goal of this procedure is to ensure that hearing aid maximum output is sufficiently low that the output never causes loudness discomfort. The following instructions are adapted from Kawell, Kopun & Stelmachowicz (1988):

> "We're going to see how loud this hearing aid makes sounds. You will hear some whistles and I want you to tell me how loud the whistle is. (Go over the descriptor list shown in Figure 16.5, explaining each choice, starting with *Too soft.*) When the sounds are *Too loud*, this is where you want the hearing aid to stop and you do not want the sounds to be this loud. Now, for every whistle, tell me how loud it sounds."

When the child appears to have understood the instructions, and while he or she is wearing hearing aids at their normal volume control setting, present sounds from a loudspeaker, starting from approximately 65 dB SPL and increasing in 5 dB steps. It should be possible to increase level up to the highest level achievable (at least 85 dB SPL and preferably higher) without eliciting a response of *Too loud*. Use two or three successive ascending runs, starting from progressively higher levels. If it is not possible to elicit a response of *Loud* the maximum output is probably too low.

As with adults, one never knows how a patient interprets loudness descriptors. If a child gives a response of *Too loud* while appearing to be totally untroubled by a sound's loudness, it is worthwhile re-instructing or quizzing the child about how loud the sound really is (depending on the language ability of the child). Provided the child remains cooperative and attentive, ensure comfort for at least one low-frequency sound, one high-frequency sound, and one broadband sound (see Section 11.7). Evaluation can be done while the child is wearing two hearing aids, though if a response of *Too loud* is obtained it may be necessary to evaluate each hearing aid individually, especially in the case of asymmetrical hearing thresholds.

For children who do not appear to understand the verbal instructions or the pictorial analogy, a tactual analogy can be tried.[1391] Describe each of the degrees of sensation. Along with each description, make an appropriate face and squeeze the child's arm with an appropriate degree of firmness. For example, *Too soft* corresponds to a light squeeze, *Just right* to a comfortable squeeze, *A little bit loud* to a firm squeeze, and *Too loud* to a very firm (but not painful!) squeeze. (If you are charged with assault, re-read these instructions more carefully.) Following this training, ask the child to squeeze your arm to show you how loud each sound is.

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[j] Both the NAL-NL2 and DSLm[i/o] software prescribe maximum output in a way that allows for ear size.

[k] The top three categories from Kawell et al. (1988) have been combined into two, as the distinction between *Too loud* and *Hurts* is a difficult one for young children. This also makes the pictures more distinctly different.

For the future: objective indicators of discomfort

Because it is difficult to measure LDL reliably (even in some adults) there have been several attempts to estimate LDL on the basis of acoustic reflex threshold,[650, 1272] or on the basis of ABR latencies.[1783] Electrically evoked acoustic reflex thresholds have successfully been used to guide the programming of cochlear implants.[743] It remains to be experimentally determined whether such objective measures can be used to adjust OSPL90 more accurately than can be achieved based on pure tone thresholds (e.g. Table 10.4). There are two aspects to this question.

- Can the optimum OSPL90 at each frequency be predicted more accurately from objective measures than from pure tone thresholds?
- Can objective measures predict how OSPL90 should vary across frequency, even if the final overall adjustment of OSPL90 is based on subjective assessment?

Note that the question to be answered is how accurately optimum OSPL90 can be predicted, not how accurately LDL can be predicted. The latter question may, however, be more easily answered.

Even for children younger than seven (or five if the tangible-excess method is used), intense noises should be made while the child is looking at the person creating or controlling the noise. The child should be observed for any visible sign of discomfort as the noise is made. Take care that the physical movement of the tester, or puff of air from a hooter, involved in generating the noise is not by itself causing a blink response from a baby.

You should not be afraid to generate intense noises (preceded by a suitable warning) in front of the child and observe the child's response while he or she is wearing hearing aids. Outside the clinic, the child will frequently be exposed to intense noises (self-generated or generated by play-mates) and if the hearing aid maximum output is excessive, it is best to discover this as soon as possible.

16.6.4 Subjective report measures

Just as for adults, the performance of hearing aids can be evaluated by asking the child, parent, or teacher how much they help. Several tools are available, and Table 16.6 lists the characteristics of these tools. Stelmachowicz (1999) contains a more extensive description of parent and teacher report forms.

Reports by parents and children may be more deeply considered if they are obtained through a structured interview, in which examples of each functional ability asked about are sought, rather than simply asking for a rating for each ability via a questionnaire. A functional assessment scale that uses this approach is the ***Parent's Evaluation of Aural/Oral Performance of Children*** (***PEACH***). Extensive normative data, including information on test-retest reliability, inter-rater reliability, critical difference scores, internal consistency and sub-scale scores are available for children from 1 month to 4 years of age.[277, 1304] Because the scores plateau at around 4 years of age, and because children with hearing loss usually have auditory functional ability lower than average, the measure can also be used with school-aged hearing-impaired children. A form of the items more suitable for teachers (Teacher's Evaluation of Aural/Oral Performance of Children; TEACH) is also available.[1]

Scores on the PEACH/TEACH scales and the MAIS scale are significantly correlated with several other measures of hearing ability:

- MAIS scores correlate with monosyllabic word identification scores for children with cochlear implants.[1513]
- IT-MAIS scores correlate with speech babbling in infants.[931]
- When different amplification conditions are compared, the difference in PEACH scores (i.e. by parents) correlate with the difference in TEACH scores (i.e. by teachers) and with preference judgments obtained from the children as either overall preferences or paired-comparison judgments.[279, 286]

[1] Forms for PEACH and TEACH can be downloaded from www.outcomes.nal.gov.au

Table 16.6 Subjective outcome-assessment tools suitable for use with children arranged by minimum applicable age. The *type* column refers to whether the items are standard for all children or are individually devised each time the measure is used.

	Measure	Respondent	Age	Items	Type	Reference
FEW	Family Expectations Worksheet	Parent or child	>0	5	Indiv	Palmer & Mormer (1999)
COSI-C	Client Oriented Scale of Improvement for Children	Parent or Child	>0	5	Indiv	Lovelock (unpublished); NAL (2011)
PEACH	Parent's Evaluation of Aural/Oral Performance of Children	Parent	0.1 to 4 years	11	Stand	Ching & Hill (2007)
IT-MAIS	Infant-Toddler Meaningful Auditory Integration Scale	Parent	0.5 to 3 years	10	Stand	Zimmerman-Phillips et al (1997)
P-SIFTER	Preschool Screening Instrument for Targeting Educational Risk	Teacher	3-5 years	15	Stand	Anderson & Matkin (1996)
ABEL	Auditory Behavior in Everyday Life	Parent	3-14 years	24	Stand	Purdy et al (2002)
CHILD	Children's home inventory of listening difficulties	Parent or child	3-12 years	15	Stand	Anderson & Smaldino (2000)
MAIS	Meaningful Auditory Integration Scale	Parent	>5 years, profound loss	10	Stand	Robbins et al. (1991)
SIFTER	Screening Instrument for Targeting Educational Risk	Teacher	>5 years	15	Stand	Anderson (1989)
TOOL	Teacher Opinion and Observation List	Teacher	>5 years	4	Stand	Smaldino & Anderson (1997)
LIFE	Listening Inventories for Education	Teacher	>6 years	16	Stand	Smaldino & Anderson (1997)
		Child	>8 years	15	Stand	
HPIC	Hearing Performance Inventory for Children	Child	8-14 years	31	Stand	Kessler et al (1990)
APHAP-C	Abbreviated Profile of Hearing Aid Performance for Children	Parent or child	>10 years	24	Stand	Kopun & Stelmachowicz (1998)

- PEACH scores correlate with standardized measures of language acquisition.[275]
- PEACH scores correlate with objective measures of audibility based on evoked cortical responses in infants.[633]
- IT-MAIS scores correlate with the latency of evoked cortical responses in children with auditory neuropathy.[1611]

As is apparent from Table 16.6, there is a choice between measures that contain standard items versus measures where the items are created individually for each child, at each stage of development. The advantages of each approach are the same as for adults (Sections 14.3 and 14.4). As with adults, devising individual assessment items each time the measure is used can be an integral part of structuring an habilitation program. The ***Family Expectations Worksheet*** (***FEW***)[1381] uses the same concept as Goal Attainment Scaling (Sections 9.1.6 and 14.4): as well as individually defining the item to be measured, the degree of success desired by the child or parent and the degree of success considered likely by the audiologist are discussed, which provides an opportunity to establish realistic expectations. At some subsequent time, the degree to which the goal has been attained is rated.

Another individualized measure for children that can help direct the habilitation program is the ***COSI-C***,[1303] which is similar to the adult COSI (Client Oriented Scale of Improvement; Section 14.4). Differences are that the COSI-C form:

- includes space to record any strategies that will help achieve each goal;
- replaces the *change* and *final ability* five-point scales with a single four-point scale having the following alternatives: *No change, Small change, Significant change, Goal achieved*;
- specifies review date(s) at which progress towards each of the current goals will be assessed.

For either the FEW or the COSI-C, the actual listening goals can be established in a number of ways. One useful method is to consult the milestones of normal hearing listed in the ***Developmental Index of Audition and Listening*** (***DIAL***), shown in the panel.[1381] The ages shown against each listening activity are approximate, even for children with normal hearing. When setting goals, the current abilities of the child must be considered: it is inappropriate to select a goal (e.g. *uses telephone meaningfully*) if a less complex goal (e.g. *listens on telephone*) has not yet been accomplished, no matter how old the child is.[1381]

A second method is to ask the parents, at the outset of each interview:[1777]

- What are the most important results you hope to get from today's appointment?
- What are the chief problems that your child's hearing loss has caused you or your child?
- What milestones or changes have recently occurred, or are about to occur, in the life of your child (e.g. starting at a new school, joining a sporting team[m])?
- Is there anything that you fear will result from your child's hearing loss?

Parents may be able to give a more considered response if they are advised in advance of the appointment that their input into the child's program will be sought in this way. When choosing goals, it may also be useful to review the habilitation goals and strategies discussed in Section 16.8.

If any goals related to responses to sound are not achieved, the appropriateness of the prescription and the correct functioning of the hearing aids should both be reviewed especially carefully. In experiments examining the appropriateness of different prescription procedures, changes in auditory responsiveness following a change in frequency response have been observed within the first week after the change has been made.[274] Unfortunately, it is not possible to give any general guidelines as to how or when the amplification characteristics should be altered if the child has not made the progress expected.

16.6.5 Articulation Index or Speech Intelligibility Index (SII)

The Articulation Index (AI) or Speech Intelligibility Index (SII) methods[n] can be used to predict speech intelligibility for a specified type of speech (e.g. sentences) at a specified level in a noise background of specified level and spectral shape. The problems with using these methods with children are the same as with adults.

First, very high SII values can be achieved simply by applying enough gain at each frequency to make speech at that frequency highly audible. If this results in excessive loudness, or excessive saturation of the hearing aid, the result is not likely to be any more satisfactory for a child than it is for an adult.

Second, the conclusion reached about how much gain at each frequency is needed to maximize intelligibility depends strongly on how (or whether) hearing loss desensitization (Section 10.3.4) is allowed for in the calculation of SII. Many simple applications of SII make no allowance for hearing loss desensitization. Suppose such a simple SII calculation method were used to evaluate the fitting of a hearing aid to a child with low-frequency thresholds around 60 dB HL and high-frequency thresholds around 115 dB HL. The calculation implies that the best intelligibility would

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m Some specific goals for coping with sports activities, and strategies to achieve them, can be found in *Time Out! I didn't hear you* by Palmer, Butts, Lindley & Snyder, and can be downloaded from www.pitt.edu/~cvp.

n The SII and AI are essentially the same method, differing mainly in some of the numerical constants used.

Developmental Index of Audition and Listening (DIAL)

From Palmer & Mormer (1999), by permission

Age	Milestone
Infant	
0-28 days	Startle response; attends to music and voice; soothed by parent's voice; some will synchronize body movements to speech pattern; enjoys time in *en face* position; hears caregiver before being picked up.
1-4 months	Looks for sound source; associates sound with movement; enjoys parent's voice; attends to noise makers; imitates vowel sounds.
4-8 months	Uses toys/objects to make sounds; recognizes words; responds to verbal commands – bye bye; learning to recognize name; plays with noise makers; enjoys music; enjoys rhythm games.
8-12 months	Attends to TV; localizes to sounds/voices; enjoys rhymes and songs; understands NO; enjoys hiding game; responds to vocal games (e.g. so Big!!)
Toddler	
1 year	Dances to music; sees parent answer telephone/doorbell; answers to name calls; attends to books.
2 years	Listens on telephone; dances to music; listens to story in group; goes with parent to answer door; awakens to smoke detector; attends to travel activities and communication.
Preschool	
3 years	Talks and listens on telephone; sings with music; listens to books on tape; smoke detector means danger; enjoys taped books; attends to verbal warnings for safety.
4 years	Telephone play; attends movie theatre; dance/swim lessons/ watches TV/videos with family; neighborhood play.
Early School Age	
6-8 years	Uses telephone meaningfully; enjoys walkman/headphones; uses alarm clock independently; responds to smoke detector independently.
Late Elementary	
8-10 years	Uses television for entertainment & socializing; attends to radio; responds to sirens for street safety; participates in clubs and athletics; enjoys privacy in own room; enjoys computer/audio games; plays team sports.
Middle School	
10-14 years	Uses telephone as social vehicle; attends movies/plays; develops musical tastes; watches movies/TV with friends.
Older Adolescent	
14-18 years	Goes to dances; begins driving (e.g. needs to hear sirens/turn signal); participates in school groups/clubs; employment/ disability legislation.
18-22 years	Employment/career decisions; travels independently; listens in college halls/classrooms; participates in study groups/extra-curricular activities.

be achieved only when the high-frequency gain was large enough to make all the high-frequency speech components completely audible. This conclusion is not likely to be correct, even if it was technologically achievable. In addition, for a given SII value, children will have lower speech intelligibility than adults, as described in Section 16.4.1.

In short, the SII method is not a reliable way to choose between different amplification options. It does, however, provide a metric of what proportion of the information in speech is potentially available, and should be reasonably accurate for any child whose thresholds remain in the mild or moderate ranges at all frequencies. Remember, however, that the SII values refer to the importance-weighted proportion of speech information that is audible, not the proportion of speech items that will be correctly understood. The mathematical relationship between the SII value and the percentage of speech understood is called the ***transfer function***. Various transfer functions exist, and show that for the same SII value:

- speech intelligibility is higher for easy speech (e.g. digits or high context sentences) than for difficult speech (e.g. isolated words or nonsense syllables);[54]
- speech intelligibility is higher for adults than for children;[1584]
- speech intelligibility is higher for normal-hearing people than for hearing-impaired people.[293, 1396, 1584]

For any degree of hearing loss, a visual display showing how much of the speech spectrum (for soft, typical or loud speech) is made audible by the hearing aids can also be a useful counseling tool for parents.

The AI is calculated by some hearing aid fitting programs, and can easily be calculated using the well-known count-the-dots method in which hearing thresholds (aided or unaided) are superimposed over a representation on the audiogram of the speech spectrum (for 60 dB SPL speech in quiet). The AI value (in percent) equals the number of dots that have intensities greater than the thresholds, assuming that there is no noise present.

16.6.6 Evoked cortical responses

Evaluation of hearing aid effectiveness is most difficult during the first nine months of life when behavioral measurement techniques are most limited and least accurate. Unfortunately, this is precisely the time when accurate evaluation is most needed: there is least certainty about hearing thresholds; least certainty about the potential for audible information to lead to good speech understanding (markedly uncertain for children with auditory neuropathy); speech intelligibility is almost impossible to measure (at least within clinical time constraints); the brain is highly plastic and the auditory functions it learns to perform depend on it receiving auditory input;[1471] and urgent decisions have to be made about whether cochlear implantation should be considered (for children with more than a moderate loss – Section 16.1). One technique that can provide information about the reception of sound during this period is the measurement of ***cortical auditory evoked potentials*** (***CAEP***s).[437]

Evoked cortical responses have several advantages for evaluating the perception of sound by infants (or other people unable to reliably respond behaviorally, such as people with dementia or other cognitive problems, or those affected by strokes):

- Because CAEPs are generated from the auditory cortex, they test the complete auditory system. For children with auditory neuropathy, cortical responses can be observed even when there is not sufficient synchronized activity in the brainstem for an ABR to be observed. Furthermore, the cortical responses are more indicative than the absent ABRs of the behavioral thresholds that will be measurable when the child is older.[586, 633, 1398]
- The hearing aid wearer does not have to attend to the stimuli.
- Stimuli can be tone bursts or speech sounds. When a speech sounds elicits a cortical response, it directly confirms that the speech sound is being detected by the person, without need for any calculations or assumptions about how the bandwidth or duration of a complex signal affects its audibility relative to known pure-tone thresholds. Any phoneme can be used as the stimulus, and although the shape of the waveform varies slightly between stimuli, the broad shape is the same for all audible stimuli for the same person being tested, both for adults[18] and for infants.[18a]

We have known about cortical responses for over 70 years.[405] One of the factors that has held back their clinical take-up is the variability of response shape across age, and even across time within a particular measurement if the person being measured moves from alert to drowsy. Recently, a method for automatically detecting CAEPs that makes no assumption about the shape of the response has been devised. This method, which indicates the probability that the response is truly related to the stimulus, has been shown to be at least as accurate as judgements by expert clinicians for both adults[632] and infants.[261]

The availability of equipment incorporating this method, along with three speech stimuli representative of low-, mid- and high-frequency weighted speech sounds (*m*, *g* and *t*, respectively), make the measurement of CAEPs while infants wear their hearing aids (or cochlear implants) a viable evaluation technique. Although cortical testing is unfamiliar to most clinicians, the primary skill involved is no different from that needed for behavioral testing of older infants – holding the infant in a calm, awake state for as long as possible.[o]

Figure 16.6 shows a typical cortical response shape for an infant and an adult. The infant responses are characterized by a major positive peak, and a broad late negative peak.[p] The typical adult response includes a marked negative peak that occurs about 100 ms after the onset of the stimulus. This negative response gradually appears from around 9 to 10 years of age, as axons and dendrites in the most superficial layers of the auditory cortex increase markedly in density.[1446]

There are three important features of the cortical response (its presence, latency, and shape) that tell us about the transmission of sound through the peripheral and central auditory system, and about the effectiveness with which the sound is analyzed by the auditory system.

Presence of the cortical response. If the CAEP is present at an acceptable level of statistical confidence (e.g. $p<0.01$) then we can have a high degree of certainty that the child can detect that speech sound at the level presented. The greater the number of speech sounds at conversational level that evoke a CAEP, the greater is the likelihood that the infant has good

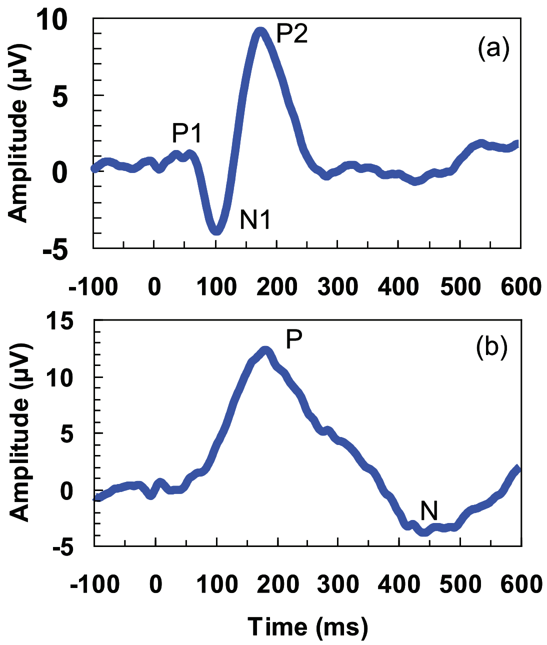

Figure 16.6 Typical waveforms of evoked responses at various times after the onset of the sound evoking the response for (a) adults and (b) infants.

functional auditory ability in daily life, as assessed by the PEACH questionnaire.[633] For children with auditory neuropathy, older children who have cortical responses have better understanding of speech, on average, than those for whom no cortical response is observable.[1481] This relationship is likely to be also true for children with sensorineural hearing loss, but this has not been investigated in older children.

Latency of the positive peak. As Figure 16.7 shows, when babies are first born, the positive peak has a latency of close to 300 ms. This latency normally reduces rapidly with age, but for babies with severe or profound hearing loss, the reduction in latency does not commence until the baby is first exposed to sound by fitting with hearing aids or by implantation.[87, 1613] If exposure to sound commences too late, the reduction in latency to normal values never occurs. There is increasing risk of this happening if the first exposure

o Cortical responses are also measurable in at least some stages of sleep, but as less is known about the nature of these responses, current recommendations are to measure infants while they are awake.

p Infant responses in some studies also show an earlier negative peak, around 200 ms after stimulus onset, which may perhaps be elicited when using intervals of several seconds between stimuli.

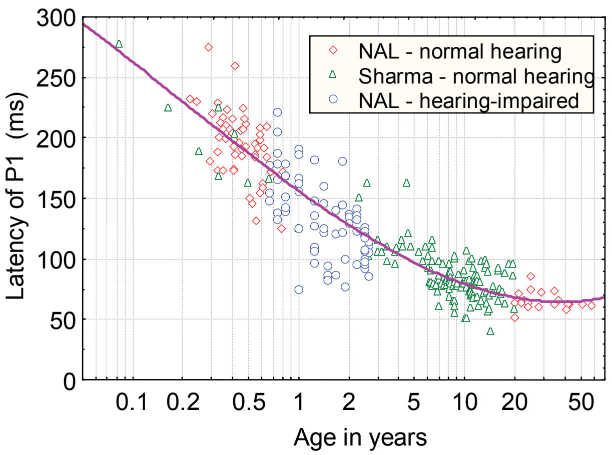

Figure 16.7 Latency of the first positive peak of the cortical response as a function of age.

to sound occurs after age 3 years and an inevitability if it occurs after 7 years.[471, 1612] The latency of the positive peak is also a strong indicator of functional ability, as assessed by the IT-MIAS. [1611] Overall, the latency of the positive peak can be viewed as a marker of the maturity of the auditory system, which for children with severe or profound loss, is very much affected by whether the devices with which they have been fit or implanted have been sufficient to make sounds audible.

Morphology. The shape of the evoked response is also an indicator of the maturity of the auditory system and appears to be related to the auditory functioning ability of the child.[1611] The relationship of morphology to hearing acuity has not yet been well investigated.

Measurement of CAEPs appears to be a very promising method of hearing aid evaluation, but there are some limitations that must be borne in mind.

- A small proportion of children do not have observable CAEPs until the stimulus is at a very high sensation level. The same applies to adults, and the reasons for the lack of cortical responses at low to moderate sensation levels are not yet understood. The implication, however, is that the absence of a cortical response does not absolutely imply that the sound has not been detected by the child.
- If children are too physically active, myogenic electrical noise (i.e. from muscle activity) will mask the CAEP unless the measurement proceeds for an unfeasibly long time. With current techniques, it normally takes 2 to 5 minutes to establish that a single sound is eliciting a response in an infant (and 20 to 60 seconds in an adult).
- If a single phoneme is used as the stimulus, the response indicates only that the sound has been detected, and the latency indicates the maturity of the auditory system in receiving such sounds. Detection and normal latency have positive implications for the ability to understand sound, but does not absolutely indicate recognition.

The discussion so far has been in relation to the CAEPs evoked by single speech sounds. When syllables are used as the stimulus, in which there is a natural transition from one phoneme to another, both the first phoneme and the second phoneme evoke a cortical response, indicating that the person has been able to hear the change.[1144, 1369] This research holds the promise of future objective tests of speech discrimination ability.

Measurement of CAEPs also has some role in parental counseling. At one extreme, parents sometimes find it difficult to accept that their new-born baby has hearing loss when there is nothing about their baby that is in any way observably different from a normal-hearing baby of the same age. The absence of cortical responses to conversational speech sounds provides additional evidence to them that there is indeed a problem that they need to address. At the other extreme, parents may not only accept that there is a problem; they can become pessimistic about the life that lies ahead for their baby. The appearance of normal brain-wave activity in response to conversational speech sounds when the baby is wearing hearing aids (and its absence when they are not)[q] can reassure them about the future communication ability of their baby, and reinforce to them the importance of the hearing aids being worn.

[q] Note that although some research has shown that low-gain hearing aids do not affect the amplitude of cortical responses,[134, 1681] this research has been carried out on normal-hearing adults, for whom the speech sounds were already audible unaided, and for whom internal noise in a hearing aid decreases the sensation level of speech, rather than increasing it.

16.6.7 Speech production and language acquisition

Because both speech production ability and language are normally acquired by hearing, the effectiveness of a child's hearing, including the effectiveness of the hearing aids, can be assessed by measuring the child's speech production ability and receptive and expressive language ability. The advantages of this approach are that these are important outcomes for the child, and there are both standardized and novel assessment tools available for children of different ages. Speech production can even be assessed during the first year of life when the child is only babbling.[1864]

The disadvantage of measuring speech production or language is that if the primary interest is in assessing whether the hearing aids are making speech optimally audible for that child (which is all the hearing aids can really do), these are very indirect measures. A result well within the normal range for children with normal hearing provides some reassurance about the entire habilitation program the child is experiencing, including the hearing aids.[r] A poor result could, however, be caused by many factors unrelated to the current hearing aids (including generalized developmental delay, insufficient exposure to language, low verbal intelligence, and the various cochlear distortions that inevitably accompany severe hearing loss). As the child's current speech production and language acquisition are the product of everything that has happened previously, even hearing aids worn in the past and long discarded bear some of the blame or credit for the current results obtained.

16.7 Helping Parents

For a normal-hearing parent, the world changes the day a hearing loss is diagnosed. For a while, a parent's hopes, aspirations, and beliefs about the child's future are overwhelmed by shock, disbelief, fear, and despair. The audiologist is likely to be the first person to bring to the parents the news they least want to hear, and so has a special responsibility for helping the parents through this time, both directly and by putting the parents in contact with others who can help. This section reviews some of the ways that the audiologist can help parents. The counseling skills needed by a pediatric audiologist go far beyond the scope of this book, however. Furthermore, the level of skills needed to best help parents at a time of high emotional shock have increased. The introduction of universal neonatal hearing screening means that more parents than ever are being confronted with an adverse diagnosis of their child's hearing, without having had any time to reflect on the possibility that their child might have a hearing loss.[1091]

It is essential for the future of the child that the parents receive every help possible. This is true in many ways, but most relevant for this book is that effective amplification for the child will need the active help of a parent. Amongst their many other tasks, parents are the only people on the spot to ensure that hearing aids continue to function and provide maximum benefit to the child.

Several researchers have interviewed or surveyed parents some months or years after their children were diagnosed as having hearing loss.[78, 321, 1091, 1535, 1540] The following list of attributes is based on this research into the things that parents found helpful or unhelpful about the way their audiologist assisted them during and after the time their child was identified as having a hearing impairment.

Parents want their audiologist to be:

Empathetic. The number one thing parents want is for their audiologist to be sensitive to the emotional shock they have received. This need is constant but is greatest at and soon after diagnosis, and whenever significant decisions (e.g. education type or device type) have to be made. Audiologists need to be great listeners. Audiologists must also be genuinely accepting of any *emotions* that parents reveal, although counter-productive *behavior* by the parents is a valid target for modification.[1090] There are no good or bad emotions. The stressful nature of dealing with the hearing impairment during the first months can be acknowledged, along with the reassurance that the stress and its impact on life quality greatly subside.[201]

Informative. Parents want to learn about the hearing loss and its implications for what their child will be able to do, what will change in the life of their family, and what hearing aids do. They want unbiased infor-

[r] Even a result exactly on normal average is not an absolute guarantee of an optimal hearing aid fitting. It could be the result of a substandard hearing aid fitting on a naturally gifted child who has been exposed to an excellent educational intervention program.

mation regarding devices and education, what other services are available and the means by which they can be financed.

Competent. Parents can tell if their audiologist is a competent specialist versus an audiologist who sees only an occasional child. When parents tell their audiologist what the child does and does not appear to be hearing, they want their audiologist to hear them, and to advise them expertly about devices and tactics.

Supportive. Parents do not want the audiologist to take over decision-making, but do want information and help in arriving at decisions. Parents want the audiologist to support their chosen course after they have taken a well-informed decision. Parental stress is high when consideration is being given to receiving one or two cochlear implants, particularly if the child has already achieved relatively good speech perception, speech production or language development with hearing aids, as there is more to risk losing with implantation.[200] Supportive audiologists will convey information without using any jargon. Jargon either confuses parents, or sends them the message that the audiologist is the sole expert in the room. This may be true of hearing aids, but when it comes to the child, the parents are the experts.

Patient. Parents need time to think through the implications of different management options for the life of the family. They need time to make decisions, time to experience the ramifications of those decisions, and time to change their minds if necessary. It can be difficult for the clinician to balance this need of the parents with the clinician's sense of urgency that the child receive high quality sound as early as possible. While early intervention is important, the variation of language outcomes with age of intervention is very gradual, so the clinician should not stress about a delay of several weeks if such a delay is helpful to the parents.

Positive. Parents need some hope and reassurance for the future, and the audiologist, having seen many other families cope well with hearing loss, is able to give that hope.

If parents are to help their child to the greatest degree, they have to acquire a lot of information as quickly as they are able. Even so, there is so much for parents to absorb; the audiologist has to impart knowledge gradually rather than in one or two sessions crammed with information. Apart from basic limitations on how quickly anyone can absorb new information, the type of information needed by parents changes as their child grows. The audiologist must appropriately combine explanation, demonstration, hands-on skill development with reinforcement, handouts, videos, and group discussion sessions to help the parents learn.[513] Getting the parents to practice a skill, and providing them with feedback, reinforces the importance of the skill as well as teaching the skill itself. Some topics relating to the hearing aids that should be covered include:

Essential: meeting other parents

Parents find an introduction to other parents of hearing-impaired children to be extremely helpful. Other parents, especially those with children whose hearing impairment has been diagnosed longer, can provide emotional support, understanding, and hope in a way that no one else can provide. Providing an introduction to other parents is considered by parents to be *the* most helpful thing an audiologist can do.[986]

Benefits and limitations. In the end, hearing aids can amplify only what comes into them. Background noise, reverberation, and distance greatly decrease the effectiveness of amplification. Audio demonstrations including simulations of hearing impairment are invaluable, as is a little information about the nature of speech and the types of distortions that may be occurring in their child's hearing (Section 1.1).

Hearing milestones. Expectations of what the child should be able to do, and approximately when these milestones are expected to occur should be explained to the parents. The parents are the people best placed to confirm the achievement of these milestones, or to raise an alarm if the milestones are delayed by an abnormal amount.

Care and use of hearing aids. Parents will need to know about cleaning the earmold and hearing aid, checking batteries, performing listening checks, putting the hearing aids on, setting the controls (if any), carrying out activities (like talking and playing) that promote their use, and avoiding hazards like moisture. They will need to be practiced enough that they can do all these things within the normal routine of the family, and without the hearing aids or the hearing loss becoming the focus of the family.[513]

Troubleshooting. When listening checks reveal a problem, parents need to know how to diagnose common faults like cracked or loose tubing, weak batteries, moisture, and internal noise. They need to know what they can fix themselves and what they need help for. Such skills can be taught by having available a few faulty hearing aids and asking the parents to diagnose the fault in each. Even when parents and teachers conscientiously check hearing aids, there can be a high incidence of non-functioning hearing aids at any given time. As soon as possible, children should be taught to monitor the status of their own hearing aids.[514]

Safety. See Section 16.10 for various aspects of safety with hearing aids, some of which should be communicated to parents.

The quality and speed of communication, and the ease with which new skills are imparted, depends on the quality of the relationship between the audiologist and the parents.[321, 513] At review appointments, parents may present general concerns, such that there is no clear way in which the concerns can be addressed. In many cases, the general concern can be solved by first identifying the specific problem that is causing the parents to have the general concern. When the specific problem is addressed, the general concern may disappear.[513] A general concern about the child *feeling different from his normal-hearing peers*, for example, may be precipitated by either the child or the parent being concerned about the large size of a hearing aid or assistive listening device. Parents should always be encouraged to contact the audiologist if problems or concerns arise between regular review appointments.

One thing parents want from a health care system is that the services available be seamless. They appreciate it when early and accurate identification is quickly followed by the provision of amplification. They also appreciate information that will help them make a confident decision about how and when their child should be educated. If the services provided are not well integrated with other services needed by the parents, the audiologist has an especially critical role in helping the parents negotiate the system(s) with minimum frustration and delay. Along the way, the family may need the intermittent services of general and specialist medical practitioners, geneticists, occupational therapists, psychologists, speech-language pathologists, and support teachers. The audiologist is more likely to be along for the whole journey. Some audiologists have the special privilege of helping the children of parents who were also once their child clients.

16.8 Hearing Habilitation Goals

We must not lose sight of the role of hearing aids; they are only a means to an end.[s] The real goal is that the child develops a high-level ability to listen and speak so that he or she will not be handicapped by the hearing loss. Effective hearing aid (or cochlear implant) fitting is one of the essential steps that must be taken to achieve this goal, as summarized in Figure 16.8.

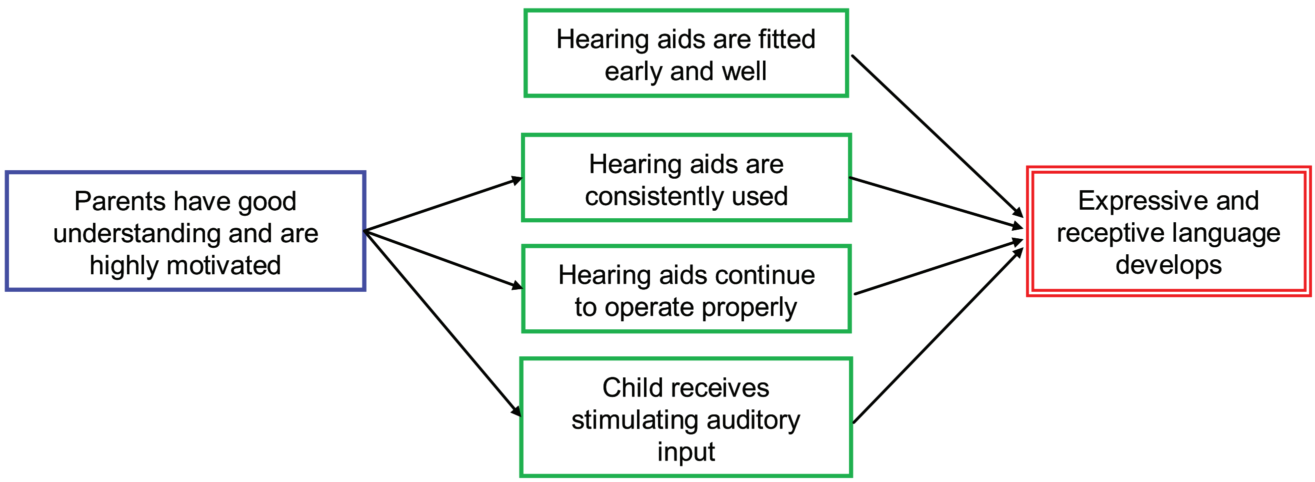

Figure 16.8 Goals of the hearing habilitation process, culminating in maximal development of language.

[s] There are other aspects of hearing habilitation in addition to hearing aids, such as participation in an early intervention program, but such things are outside the scope of this book.

For a child to maximize his or her mastery of language, hearing aids must consistently be worn, and they must be functioning correctly. The child must also be receiving rich, stimulating auditory input, and be engaging in meaningful listening activities. When the child is young, these three essential ingredients are most likely to occur if the parents have a good understanding of the nature of hearing loss and speech and of the importance of good quality auditory input to the development of language and literacy. This knowledge motivates parents to maximize effective hearing aid use.

Many of the initial steps in habilitation thus involve the parents more than the child. As the child grows, the clinician should help the child understand about his or her hearing loss, and encourage the child to be self-motivated to hear as well as possible in each situation.

Although the broad goal of good language development remains unchanged, the detailed goals and strategies that achieve this vary with the age of the child. The following sections list some goals and strategies that are appropriate to different ages, although many goals shown in one age category are also appropriate to later categories. The goals shown are intended as examples.[t] *They are far from comprehensive, and some will not be appropriate for some families.* To maximize the likelihood of goals being achieved, goals should be jointly developed by the audiologist, the parents, the habilitationist or teacher and, when old enough, the child. Many goals will be prompted by changes facing the child, such as progression to a new school or the commencement of new social or sporting activities.

16.8.1 Goals and strategies for infants

Goal: The child uses the hearing aids consistently

- Assist the parents to accept that their child has a hearing loss.
- Ensure that parents understand the close link between consistent high-quality auditory stimulation, brain development, and subsequent language and literacy development.
- Explain to parents why this type of hearing aid (nearly always a BTE) has been selected.
- Introduce parents to other parents whose children consistently use hearing aids and/or to parental support groups.
- Devise a behavior modification plan that links hearing aid use to some reward, such as book reading, food or attention.
- Discuss the child's daily routine to identify when aid use is practical and when it is not.
- Check that the hearing aid appears to be comfortable when the baby is sitting supported or lying down, and that feedback oscillation does not occur in these positions.
- Provide parents with information about hearing aid use in the event of ear infection with suppuration (e.g. earmold disinfecting, use at key listening times only).
- Encourage parents to record hearing aid use, and responses to sound and speech, in a communication diary, over a set period.
- If consistent use is not established, search for the reasons (e.g. parents not convinced of the need, problems with the hearing aids).
- Acknowledge that achieving consistent use is not easy, and that it often gets gradually easier to achieve over the first two years of life.[1202]

Goal: The hearing aids function properly

- Ensure that parents can operate the controls, can insert and remove the earmolds confidently, and that they understand what the hearing aid does.
- Provide parents with their own earmolds made with 300 mm of tubing, or with a stethoclip, so that they can do listening checks while holding the hearing aid in front of them and manipulating the controls.
- Demonstrate troubleshooting, including battery testing, the use of a puffer to dry earmolds, the causes of feedback oscillation, and the use of feedback oscillation as a quick check of hearing aid functioning (unless prevented by a feedback cancellation algorithm). Trouble-shooting can efficiently be taught in small workshops which, if followed by coffee also provides an easy way for parents to meet other parents.

[t] The goals and strategies are heavily based on "Goals for promoting hearing," an unpublished document by Karen Lovelock and Anne-Marie Phillips (pediatric audiologists in Australian Hearing), and I am grateful for permission to reproduce some of this material.

Goal: The child receives high quality auditory stimulation

- Explain to parents the effects of noise, distance, reverberation, and head position on the quality of sound received, and thus on the integrity of growth and development of auditory neural/brain pathways.
- If possible, *demonstrate* to parents the effects of noise, distance, reverberation, and hearing loss on sound quality.
- Reinforce to parents the need for regular, interesting and enhanced auditory stimulation.
- Discuss the use of FM systems and ensure they are adjusted in a manner appropriate to the hearing aids and situations in which they will be used.

Goal: The parents understand the education options

- Outline the basic education options in an unbiased and factual manner, especially those relating to early intervention. Ensure that parents know where to obtain more detailed information about each option. Direct and continuing liaison between the audiologist and any educational institution is necessary.
- Provide written information (this also applies to most other issues discussed).

Goal: The child reacts to sound

- Show the comparison of the amplified speech spectrum (measured or calculated from real-ear gain) to hearing thresholds, and explain what sort of sounds should be detectable and, if any, what sort should not. Alternatively, the same points can be made using an aided audiogram (see Figure 9.6).
- Ask parents to monitor whether the child reacts to louder environmental sounds, the voice of others, his or her own voice, and whether the child displays any preliminary turn-taking skills.

16.8.2 Goals and strategies for toddlers

Goal: Child accepts hearing aids

- Parents reinforce child when he or she indicates that a hearing aid is faulty or is feeding back.
- Parents reinforce child when he or she puts the hearing aids on or asks for them to be put on.
- Hearing aids are put on early each day, as part of the daily routine, such as when getting dressed.
- Parents, rather than toddler, decide when hearing aids are removed.
- Parents encourage child to look after hearing aids by putting them in the same specified place when they are not being worn.
- Parents encourage child to test hearing aids with own voice when they are first put on.
- Parents teach *hearing aids* along with *nose, feet, tummy, etc*.

Goal: Child develops listening skills and realizes benefits from hearing aids

- Parents draw attention to environmental sounds and reinforce child when he or she recognizes them.
- Parents and child play games that require the child to respond to sound.
- Parents select toys that have an appropriate auditory reward.
- Parents reward child for appropriate vocalizing and listening.
- Signing (if used) is accompanied by speech.

16.8.3 Goals and strategies for pre-schoolers

Goal: Child reports when a hearing aid is not working

- Play a game (with rewards) in which the child has to differentiate between a working and a non-working hearing aid.

Goal: Child manages hearing aids without help

- Practice with hearing aid insertion, on/off, and volume control and battery manipulation if appropriate. Reinforce with lots of praise!

Goal: Child displays appropriate communication skills

- Reinforce when child uses voice volume appropriate to situation (also useful for normal-hearing children to learn!), displays turn-taking skills, and visually attends to talker.

16.8.4 Goals and strategies for primary school children

Goal: Child can organize devices or environment to hear well in a range of situations

- Provide FM hearing aid (if not already provided) and instruct child in its use (battery changing/charging, connectors, use of controls).
- Provide a telephone coupler and TV listening device (if needed and if not already provided) and instruct child in their use. Note that an FM device can double as a TV listening device.
- Demonstrate to child the effects of distance, noise, and reverberation.
- Give child practice in identifying the source of communication difficulties and in using age-appropriate hearing strategies to alleviate problems.
- Instruct child in use of FM, T-switch, or listening position (close to source) to solve the problems of distance, noise, and reverberation.
- Show the child how to care for the hearing aids: earmold washing and drying, battery testing.

Goal: Child continues to accept the use of amplification devices

- Demonstrate benefit of device with simple speech test in the clinic.
- Ensure that child knows other hearing-impaired children.

Goal: Child understands about hearing loss in general and his or her hearing loss in particular

- Explain to the child (at an age-appropriate level) the cause of his or her hearing loss.
- Explain to the child the characteristics of his or her loss (e.g. better ear, better frequencies, and difficulty in separating sounds). Achieve an appropriate balance between the difficulties to be overcome and a positive view on what the child *can* do.

The goals and strategies described in this chapter are by no means a complete description of parental and child counseling. Other activities, such as referral for genetic counseling or other medical specialties, referral to speech therapist or occupational therapist, the impact of ear infections, and an explanation of the physiology, type and degree of hearing loss (and its permanency) all go beyond the scope of this book.

16.9 Teenagers and Cosmetic Concerns

Engaging with some teenagers can be challenging. One technique that has been found helpful is to have the teenager complete a self-assessment questionnaire dealing with communication difficulties in specific situations, impact on the teenager's social life, reactions of other people and feelings engendered in the teenager.[515] The teenager can also be invited to ask a significant friend to complete a complementary form of the questionnaire about how the hearing-impaired teenager *seems* to cope. A review of the hearing-impaired teenager's answers and a comparison with the perceptions of his or her friend (which may be very different) then provides good discussion material to assist the counseling process. The teenager is likely to find it easier to apparently talk about the questionnaire results than about his or her self, even though the former accomplishes the latter.[515]

Although it would seem that teenagers would be well able to recognize when their hearing aids were not functioning correctly, a purposeful theoretical and practical course on hearing aid functioning and maintenance has been found to materially increase the proportion of time that hearing aid faults were recognized and rectified.[1253]

Unfortunately (from the perspective of communication ability), some children reject any visible form of prosthetic device early in their teenage years. For them, at that stage of their life, the disadvantages of looking different from their peers presumably outweighs the communication advantages offered by their hearing aids, FM systems, or cochlear implants. Noble (1999) hypothesizes that there may be more to the rejection than simply avoiding a visible device. It may be that using an imperfect device in a world of noise and unpredictable signals, with poor results, poses more of a threat to their self-perceived ability to cope than does not using a device at all.

Whatever the reason may be, the audiologist should make sure that the teenager is aware of all the consequences of not using hearing aids. The audiologist should also teach or reinforce alternative strategies to reduce communication breakdown. It may also be important to reinforce the teenager's right to make his or her own decision, and to change that decision at any time. People commonly seek help once more in their late teenage years when the increasing seriousness of education and the demands of the work force, in conjunction with their growing confidence as young adults, change the balance of the equation for them. With the increasing connectivity of hearing aids, they can truthfully be portrayed as the auditory portal to phones, computers, portable music players, and any other electronic devices that teenagers view as fundamental to life, which might encourage continuous wearing through the teenage years.

16.10 Safety Issues

Parents should be advised about several aspects relating to the safety of their child, at some suitable time after the child has received his or her hearing aids.

Battery and hearing aid ingestion

Hearing aids are the major source of batteries ingested by children.[1071] Parents should be advised that new and used batteries *must* be kept away from young children. This particularly applies to children less than three years of age, but ingestion (deliberate or accidental) occurs in every age group.[u] Loose batteries are not the only danger. In a third of cases of hearing aid battery ingestion, the child removes the battery from his or her own hearing aid. Hearing aids for infants and toddlers *must* have tamper-resistant battery drawers. This is sometimes achieved by grinding away the ridge that is normally used to open the drawer, thus requiring that a tool be used to open it, but the effectiveness of this depends on the shape and tightness of the battery compartment. Specially designed locks are also available.

Parents should be advised to urgently seek medical attention if they believe a battery has been ingested. The major danger to the child is from chemical burning or choking if the battery becomes lodged in the esophagus, although this is most likely to occur with batteries larger than those used in hearing aids.[1071] Common medical practice is to confirm by X-ray this has not happened (whatever the battery size), and then have the parent confirm that the battery has passed through the digestive system. This usually takes 24 to 72 hours, but has been reported to be as short as 12 hours or as long as 14 days.[1072] Inducing vomiting is ineffective and potentially harmful as the battery can be ejected from the stomach only to become stuck in the esophagus.[1071] Chemical burns can also result from batteries inserted in the nose or ears or under plaster bandages.[1052]

As the size of hearing aids has shrunk there are now cases reported of entire hearing aids or earmolds being ingested. The same techniques used to stop a hearing aid from becoming lost (e.g. fishing line connecting the hearing aid to the child's clothing, also provides a safeguard against swallowing the hearing aid. A loose connection between the hearing aid and the earmold should be avoided.

Battery explosion

Parents should be advised that hearing aid batteries are not re-chargeable, and that they could explode if they are placed in a recharger. Similarly, they should not be disposed of in a fire or incinerator.

Noise-induced hearing loss

A hearing aid has the potential to exacerbate hearing loss by exposing the ear to high levels of noise. The audiologist can minimize this risk by prescribing gain and OSPL90 appropriate to the loss, and by selecting a hearing aid with low-ratio compression, rather than linear amplification, for at least mid- to high-level sounds (see Section 10.8).

Parents and older children should simply be advised not to increase the volume control setting above that recommended, except possibly in quiet environments. The need for volume control variation is minimized if a wide dynamic range compression aid is prescribed. If the child will be in a very noisy environment for extended periods of time, hearing protection should be worn.

Physical impact

The potential consequences of a blow to the head while wearing a hearing aid have already been discussed in Section 16.3. It is best that the child not wear a hearing aid during sports in which a blow to the head is possible. Such a blanket rule may make it impossible for the child to play sport at all and, if so, parents will have to balance the risks against all the consequences of not participating. The use of headgear that provides some physical protection but that nonetheless provides an open air-path to the hearing aid microphone inlet can be a solution. Soft earmolds should definitely be used.

[u] This includes adults who use their mouth as a third hand while changing batteries.[1071] Hearing aid batteries are small and slippery!

Warning sounds

Parents should be advised that one of the purposes of hearing is to provide warning of imminent danger. (Parents often provide the warning sound, of course.) The child will be best able to hear and understand these warnings if his or her hearing aids are being worn and are adjusted in the usual way. If two hearing aids have been provided, then two hearing aids should be routinely worn, or a child's ability to locate sources of danger may be decreased.

If the hearing aids have switchable directional microphones, the response should be switched to omnidirectional in any circumstance where warning sounds are likely to arrive from directions other than the front. (This is really a concern only in outdoor echo-free situations. Indoors, the extent of directionality is limited by the arrival of echoes from multiple directions.) For infants and toddlers, a gain appropriate to the detection of warning sounds is most likely to be achieved if the volume control is locked or disabled.

16.11 Concluding comments

Fitting a hearing aid to an infant is always an ongoing process rather than something that is carried out at one point in time. Achieving an optimal fitting, particularly for infants, is likely to remain a considerable challenge for some time. For no one else is it so important that the amplification be just right. Unfortunately, for no one else are we so unsure about what is best in principle. Furthermore, out of all patients, infants are least able to tell us what they like and dislike, and what works and does not work. Improved methods for evaluating the effectiveness of hearing aids for infants are urgently needed.

Even more than for adults, there is an ongoing knowledge gap between the availability of signal processing algorithms and research on their effectiveness for children in general and for infants in particular. Pediatric audiologists have, on the whole, long taken what was believed to be a conservative approach to this dilemma – "if you are not *sure* its right for children, don't use it". Unfortunately, this has led to many children being deprived of technological advances that would have been good for them. As an obvious example, no one needs WDRC as much as a child too young to operate a volume control, yet WDRC was not used in pediatric audiology until long after it was adopted with adult clients. The same (very understandable) desire to do no harm may now be preventing young children from achieving the benefits of fully automatic directional microphones and adaptive noise reduction in circumstances where these features can materially improve signal clarity.

As we will likely be faced with a succession of innovations at our fingertips, but with no pediatric research to lean on, perhaps the following principle might be appropriate. If a feature is designed to overcome a deficit that directly or indirectly arises because of some distortion in the cochlea, if its operation does not have to be controlled by the aid wearer, and if there is evidence of net benefit in adults, then it seems reasonable to adopt the feature for infants and young children. When the information in a signal is masked by noise or reverberation, or reduced by distortions in the cochlea, children still learning language (i.e. all children) will always perform more poorly than adults with enough listening experience and language competence to make maximum use of limited information. There is no research to suggest that any signal modification that makes understanding easier for an adult will make it more difficult for a child.

None of this takes away from the need for research into effectiveness with pediatric clients. Further research into signal processing algorithms is very much needed, especially into how adaptive noise reduction and frequency lowering should best be configured. The impact of these features on the rate at which novel words are learned[1703] would seem like a good metric for assessing the effect they have on ease of listening, particularly in the presence of background noise and reverberation.

CHAPTER 17

CROS, BONE-CONDUCTION, AND IMPLANTED HEARING AIDS

Synopsis

In the CROS (Contralateral Routing of Signals) family of hearing aids, hearing aid components on opposite sides of the head are wirelessly linked. Basic CROS aids are most suitable for people with unilateral loss. CROS aids consists of a microphone on the side of the head with a deaf ear, combined with an amplifier, receiver and open earmold or shell on the side with a normal-hearing ear. Adding a microphone to the side of the better ear converts it to a BICROS hearing aid, which is suitable for patients with loss in both ears. A transcranial CROS has all the components in one ear, but sends a signal across the head by bone conduction. CROS hearing aids must be carefully fitted to ensure the aid wearer receives, in a single cochlea, an appropriate balance of sounds reaching the two sides of the head.

Bone-conduction hearing aids output a mechanical vibration instead of an air-borne sound wave. They are most suited to people who, for medical or anatomical reasons, cannot wear a hearing aid that occludes the ear in any way, or for those who have a large conductive loss in either ear. For patients with normal external and middle ears, bone-conduction hearing aids cannot stimulate the cochlea as effectively as do air-conduction hearing aids because of the relative inefficiency of the bone conduction pathway. For patients with maximal conductive hearing losses, whether or not there is also a sensorineural loss, bone-conduction hearing aids can stimulate the cochlea as strongly as air-conduction hearing aids.

Prescriptions for air-conduction hearing aids can be converted into bone-conduction prescriptions by using available standards for the thresholds of hearing for air- and bone-conducted sound. Bone-conduction output is specified in terms of output force level instead of output sound pressure level, and in terms of acousto-mechanical sensitivity instead of gain. Disadvantages of non-implanted bone conduction hearing aids include their wearing comfort and the limited sensation level they can provide.

A commonly used form of bone-conduction hearing aid is the bone-anchored hearing aid, in which the vibrations are transmitted to the skull via an embedded titanium screw, thereby increasing stimulation of the cochlea by about 15 dB compared to a bone conductor applied to the skin. Bone-anchored hearing aids have successfully been used for patients with unilateral or bilateral conductive or mixed loss. They are also routinely being fitted to people with unilateral sensorineural loss, referred to as single-sided sensorineural deafness. The output levels they can achieve make them suitable for people with cochlear loss up to about 45 dB HL for head-worn devices and up to about 60 dB HL for body-worn devices. Bone-anchored hearing aids can provide greater cochlear stimulation than air conduction hearing aids for patients with air-bone gaps greater than about 30 dB.

A variety of other middle-ear implants have been researched, and several have been approved for routine use. Middle-ear implants may have only the output transducers surgically implanted, or may be combined with implanted microphones and batteries to form completely implanted hearing aids. Four types of output transducers have been used: magnets enclosed by a coil that rely on the inertial mass of the magnet, magnets mounted on the middle-ear chain driven by a remote coil, and piezoelectric or electromagnetic stimulators anchored to the mastoid bone and vibrating the middle-ear chain. Three types of microphones have been used: external microphones, microphones implanted under the skin on the scalp or in the ear canal, and transducers that are driven by the vibration of the middle-ear chain.

Several implanted hearing aids are now commercially available. For some clients, particularly those with mixed hearing loss, middle-ear implants may have advantages related to freedom from occlusion, amplification gain and bandwidth, stimulation level, and invisibility of the device. Fully implanted devices have additional advantages arising from there being no external parts. Candidacy criteria are still developing.

This chapter discusses several types of non-standard hearing aids that are used by only a small proportion of hearing-impaired people. A good understanding of these hearing aids is important. Otherwise, clinicians will not be able to make an optimal recommendation when they encounter patients for whom one of these hearing aids is the most appropriate device.

17.1 CROS Hearing Aids

In the vast majority of cases, people are fit either with a single hearing aid, or with two hearing aids that amplify sounds independently, or almost independently, of each other.[a] In some cases, it is better to fit people with a hearing aid system that combines components mounted at the two ears, and which thus requires the complete audio signal to be sent from one ear to the other via a cable or a wireless connection. This arrangement is known as ***Contralateral Routing of Signals*** (***CROS***).[684] As the following sections will show, there are several reasons for sending the audio signal from one side of the head to the other, and several variations of how the components on each side of the head are combined.

The major disadvantage of all types of CROS aids is that a connection must be made between the two sides of the head. The most common solution is wireless transmission, for which the major disadvantage is decreased battery life compared to self-contained hearing aids. Wireless CROS hearing aids look no different from other BTE or ITE hearing aids, but of course one of the hearing aids contains a wireless transmitter and the other contains a wireless receiver.

The longer-established solution is a cable run around the back of the head or along the frame of a spectacle aid. Cables are a nuisance and are not cosmetically attractive. Spectacle aids are also often unattractive, and are logistically difficult when repairs are made to either the glasses or the hearing aids.

17.1.1 Simple CROS aids

Basic considerations

Figure 17.1 shows the simplest CROS configuration. The microphone, mounted on the ear with the worse hearing, feeds its output to the amplifier and receiver mounted on the opposite side of the head. The separated microphone is referred to as a ***satellite microphone***. Any signal reaching the side of the head with the poorer ear will be amplified and heard in the better ear. The receiver is coupled to the ear using an open earmold, so that unamplified sound can also directly enter the better ear. The attenuation provided by earmolds with different vent sizes, including maximally open earmolds and shells, is indicated in Figure 5.13.

The major advantage of this arrangement is that sounds can be heard in the ear with the better residual hearing ability no matter which direction they come from. The head acts as a baffle for high-frequency sounds, boosting those sounds that come from the near side of the head and attenuating those that come from the far side. If the signal comes from one side of the listener and the predominant noise comes from the other side,

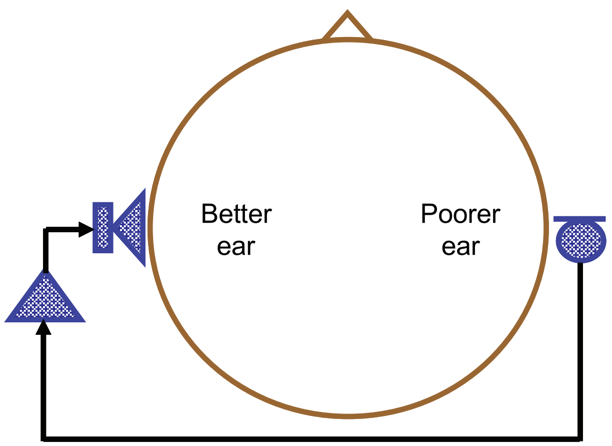

Figure 17.1 Block diagram of a CROS hearing aid system, viewed from above the head.

CROS aids: the essentials

For patients with a unilateral hearing loss:

- A CROS aid merely *transfers* sound from one side of the head to the other.[331, 528]
- A CROS fitting should not amplify sound, but if it does, the disadvantages are likely to outweigh the advantages.

[a] Many hearing aids share control information wirelessly between ears (Section 3.2); some newly developed aids send audio information to the other ear to facilitate additional forms of directivity (Section 7.1.4).

there will thus be a much better signal-to-noise ratio (SNR) at one ear than at the other (Section 15.2.1). In those cases where signal is arriving from the poorer side, the satellite microphone of the CROS aid will pick up this relatively clear signal. The electrical connection, amplifier, and receiver will transfer the signal to the better ear.

Because of head diffraction effects, a CROS hearing aid will always improve intelligibility in noise (relative to no hearing aid) when speech comes from the side of the poorer ear. These same head diffraction effects will, however, always cause sound amplified by the CROS aid to *decrease* intelligibility when speech comes from the side of the better ear. This disadvantage of a CROS aid when speech is on the normal-hearing side can be minimized by using no more gain than that recommended in the fitting procedure described in this section.

The second advantage of CROS aids is that the microphone and the receiver are well separated. Signal leaking from the receiver back to the microphone is greatly attenuated by having to pass around the head. The gain at which feedback oscillation occurs will therefore be *much* higher than would be the case if the receiver and the microphone were in close proximity.

Candidacy for simple CROS aids

Patients who may benefit from a CROS fitting are those with a ***unilateral hearing loss***, where the loss in the poorer ear is so great that aiding it will be of no benefit. The better ear should have normal hearing or at most a mild high-frequency hearing loss. Patients will particularly benefit if they frequently need to listen to signals arriving from the side of the head with the deaf ear. An example would be a taxi driver whose deaf ear is on the passenger side.

Patients who have near-normal low-frequency hearing and moderate or severe high-frequency hearing loss in one ear, combined with an unaidable loss in the other ear, may also benefit from a CROS aid if neither a conventional hearing aid nor a BICROS aid is satisfactory (Section 17.1.2). Such patients require open earmolds to avoid occlusion (Section 5.3.2) but also require substantial high-frequency gain. Feedback cancellation algorithms now enable most such people to be adequately fit with a conventional hearing aid, but if not, a CROS hearing aid will enable more high-frequency gain to be achieved without feedback oscillation occurring. CROS aids for such people should be fit using the method described for Stereo CROS aids (Section 17.1.3) rather than the method described in this section. Patients with bilateral hearing losses that could be aided in both ears are more likely to be satisfied with conventional hearing aids than with CROS aids.[606]

The effect of hearing loss in the better ear on the success of CROS fittings is contentious, probably because CROS aids have mostly been prescribed in an ill-defined manner. Gelfand (1979) found no relationship between loss and degree of use of CROS aids. Many authors have recommended that it is easier to achieve satisfactory sound quality with a CROS aid if the better ear has a mild high-frequency loss rather than normal hearing.[331, 685, 1468, 1828] This advice should be reviewed in the light of technological changes since this research was performed. Flexible tone controls enable the requisite gain, *and no more,* to be obtained in a smooth manner across a wide frequency range. With older hearing aids, one had to compromise between too much gain at some frequencies and inadequate gain at the rest.

If too much gain is used at any frequency, patients will complain about amplified internal noise, will be disadvantaged whenever speech is on the side of the better ear and, overall, will perform more poorly than with no hearing aid.[528, 1815] Too much gain in the CROS aid will effectively reverse the better and worse ears.[1078, 1133] *With a correctly balanced CROS aid, there will be neither a better side nor a poorer side when aided.* High success rates can be achieved, even for patients whose hearing is within normal limits in the better ear.[528, 739]

It may well be that people with normal hearing in one ear will elect not to use their hearing aid in very quiet environments (which is where internal hearing aid noise is most likely to be a problem). In environments with even moderate noise levels, however, external background noise will easily mask internal hearing aid noise.

Fitting procedure for CROS aids

The better ear receives the sounds that arrive directly at that ear mixed with an amplified version of the sounds that arrive at the poorer ear. When the sound source is on the better ear side, it is advantageous for sounds that enter the better ear directly to dominate this mixture. Conversely, when the sound source is on the poor ear side, it is advantageous for sounds picked up by the satellite microphone to dominate the mixture.

Domination by the side nearer the source is most likely to occur if the CROS aid is adjusted so that a frontally incident sound wave arrives in the ear canal with the same strength no matter which path it takes to the ear canal. That is, for frontally incident sounds, the real-ear aided gain (REAG) from free field input, via the satellite microphone, through the amplifier, to the ear canal of the better ear, should equal the real-ear unaided gain (REUG) of the better ear.[b] The corresponding coupler gain (CG) prescription can be calculated by rearranging Equation 4.9, and equating REAG to REUG:

CG = REUG – RECD – MLE – Vent effects
– Sound bore effects ….. 17.1,

where MLE describes the Microphone Location Effects and RECD is the Real-Ear to Coupler Difference.

Figure 17.2 shows 2-cc coupler gain prescriptions for a BTE and ITE hearing aid coupled to an open earmold or ear shell. These values are appropriate for an average adult. The values used for the terms in Equation 17.1 were taken from Table 4.3 (RECD), Table 4.5 (MLE), Table 4.6 (REUG), and Table 5.1 (Vent effects). The coupler gain prescribed for a BTE differs from that for the ITE in the low frequencies principally because there is more room to open the vent path in a BTE fitting than in an ITE fitting. In the high frequencies, the differences arise partly from the different MLE values and partly from the different RECD values for the two hearing aids, which in turn is the result of them being measured in different types of 2-cc couplers (HA1 for the ITE and HA2 for the BTE).

Over what frequency range should sounds effectively be transferred from one side of the head to the other? Interaural level differences, and hence interaural SNRs, are significant above about 500 Hz (Figure 15.3). From this perspective, it is desirable to have effective transfer of sounds extending at least as low as 500 Hz. On the other hand, the lower in frequency the transfer extends, the more likely it is that patients will complain about internal hearing aid noise. Internal noise is particularly likely to be a problem if a directional microphone is used. In noisy environments, internal noise is not a problem, and it would be preferable for the transfer of sounds to be effective down to 500 Hz. In quiet environments, it may be preferable for the transfer to be effective only for frequencies above 1500 Hz. CROS fittings may therefore be an ideal application for multi-memory hearing aids, although this is an untested proposition.

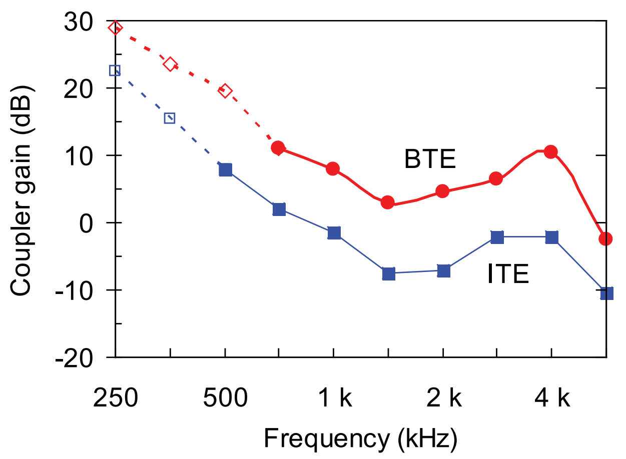

Figure 17.2 Coupler gain prescriptions for BTE and ITE hearing aids at used volume control setting for a CROS hearing aid. For the BTE hearing aid, a tube fitting is assumed. For the ITE hearing aid, a Janssen fitting is assumed. The gains shown are applicable to an average adult with no significant hearing loss in the good ear. The dashed lines in the low frequencies are not coupler gain prescriptions that should be achieved but represent upper limits that should not be exceeded.

The coupler gain prescription shown in Figure 17.2 provides a good starting point for a balanced CROS fitting. Unfortunately, one cannot directly verify that for frontally incident sound, the aided path provides the same gain-frequency response as the unaided path. The problem is that while the unaided path can be measured (by turning off the hearing aid) the aided path cannot be measured in isolation because the unaided path is always present. The response can, however, be indirectly verified by measuring the combined gain of the two paths for loudspeaker locations on both sides of the head, as described in the panel.

[b] This derivation of the coupler gain prescription assumes that the mold fitted to the good ear is sufficiently open not to have an appreciable effect on the real-ear unaided gain. The real-ear adjustment procedure shown in the panel does not require this assumption to be true.

Adjusting and verifying the CROS gain-frequency response

After the CROS aid has been pre-adjusted to approximate the coupler response given in Figure 17.2, its response can be more accurately adjusted using a real-ear gain analyzer as follows, and as shown in Figure 17.3.

Step 1 – Good-side response. With the hearing aid turned on, locate the speaker at 45° from the front, on the side of the good ear. Measure the response in the ear canal of the good ear. If the response does not approximate the usual real-ear unaided response of an ear with no mold or hearing aid, the mold is not sufficiently open to achieve a good CROS fitting.

Step 2 – Poor-side response. Move the speaker (or turn the patient) so that the speaker is at 45° on the side of the poor ear. Measure the response in the ear canal of the *good* ear.

Step 3 – Adjust the hearing aid. If the response measured in Step 2 does not match that measured in Step 1, adjust the hearing aid gain and frequency response, and repeat Step 2, until the poor-side response matches the good-side response. If a large adjustment has to be made, it may be necessary to start again from Step 1, because the good-side response is affected by the gain-frequency response of the hearing aid, although to a lesser degree than is the poor-side response. This interaction can be avoided by holding an earmuff over the ear (including the hearing aid or satellite microphone) on the side of the head away from the loudspeaker.

Step 4 – Check the frontal response. Position the speaker directly in front of the patient. Measure the real-ear aided gain. A smoothly rising response with a low-frequency gain of 0 dB and a maximum gain of 10 to 20 dB somewhere between 2 and 4 kHz should be obtained. If there is a pronounced dip at any frequency, it is possible that the amplified path is out of phase with the direct sound path at that frequency. The position and depth of such notches will vary from aid to aid, and may depend on the settings of the tone controls and the polarity with which the receiver is wired.

Note that when performing these measurements, the control (reference) microphone must either be moved to the side of the head nearest the speaker, or be switched off. With many brands of real-ear gain analyzers it is not possible to place the control microphone on the side of the head opposite to the probe microphone, so only the second of these options is a possibility.

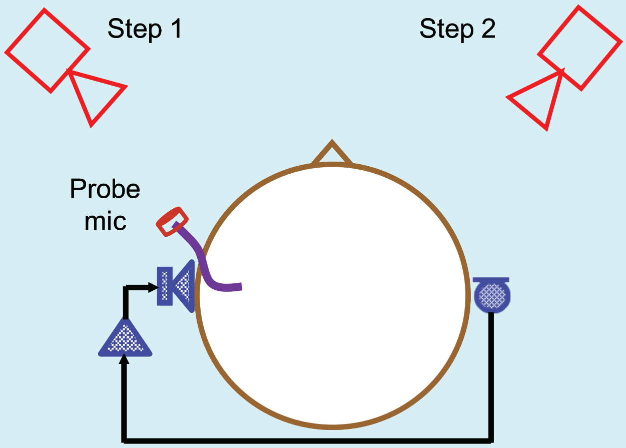

Figure 17.3 Test set-up for verifying and adjusting the gain-frequency response of a CROS hearing aid, showing the two different positions of the single test speaker.

17.1.2 Bilateral CROS (BICROS) aids

Basic considerations

If the better ear has a hearing loss, the patient is likely to benefit from amplification no matter which side of the head the wanted sound comes from. The only way to always pick up the clearer signal is to have a microphone mounted on each side of the head. If each of these microphones is connected to the same amplifier and receiver, as shown in Figure 17.4, the result is called a bilateral CROS, or ***BICROS*** hearing aid.[c]

Unfortunately, the less clear signal provided by the microphone on the side of the head further from the signal will always be added to the clearer signal from the closer microphone, thereby reducing the clarity of the clearer signal. Fortunately, some net benefit remains, because the final signal-to-noise ratio will always be better than that provided by a microphone on the head-shadowed side.

The BICROS system also works effectively for signals coming from directly in front of the listener. In this case, the wanted signal reaches the two microphones simultaneously and so the outputs of the two microphones are added together, in phase, before being amplified. Sounds coming from other directions reach the two microphones out of phase by different degrees. Consequently, the microphone outputs combine less effectively when they are added, and can even cancel each other completely for particular combinations of frequency and direction. Unfortunately, sounds from directly behind the person also arrive at the microphones in phase and are therefore also amplified with maximum gain.

Overall, however, the BICROS hearing aid works as a (weakly) directional microphone. The three-dimensional directivity index (Section 7.2.1) of a BTE BICROS system, when mounted on the head, increases from 1.5 dB at 500 Hz up to around 3 dB at 4 kHz.[312] The corresponding two-dimensional directivity indices are 2.4 dB at 500 Hz up to around 3 dB at 4 kHz. When each microphone is by itself directional, the combination of the two microphones is even more directional.[1458]

The BICROS system confers only a minor advantage in defeating feedback, because one of the two microphones is near the ear canal receiving the amplified sound, just as for a conventional hearing aid. (The maximum frontal high-frequency gain without feedback will be about 5 dB higher than for a conventional hearing aid because the satellite microphone will add to the total gain without increasing the risk of feedback.)

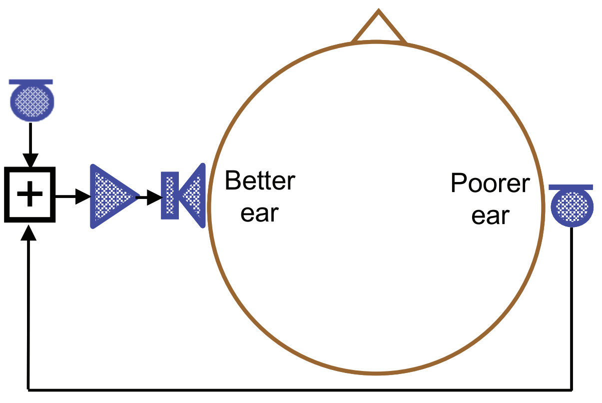

Figure 17.4 Block diagram of a BICROS hearing aid system.

Candidacy for BICROS aids

Patients can benefit from a BICROS hearing aid if they have an asymmetric bilateral hearing loss such that the poorer ear has too great a hearing loss to benefit from a hearing aid, or where amplification of the poorer ear adversely affects speech identification ability (see Section 15.4.2). Such patients, and even candidates for CROS hearing aids, may alternatively benefit from a BAHA (Section 17.3) or cochlear implant in the poorer ear, especially if that ear causes tinnitus.[63a, 1477a]

Fitting procedure for BICROS aids

Fitting a BICROS hearing aid is a combination of fitting a conventional unilateral hearing aid and fitting a CROS hearing aid. Provided the satellite microphone has the same sensitivity as the microphone in the hearing aid (this is usually the case), the necessary

[c] BICROS is usually an abbreviation of *Binaural* CROS, but in keeping with the terminology used in this book, *Bilateral* CROS seems more appropriate, as amplified sounds are heard in only one ear.

balancing of sensitivity between the two sides of the head is achieved without any action by the clinician.

The required gain-frequency response of the hearing aids can be prescribed in the same manner as for a unilateral hearing aid, using the hearing loss of the better ear as the basis of the prescription. No allowance should be made in the prescription for binaural listening. The response should be verified with both microphones in place and the speaker located directly in front of the patient. Averaged across frequency, the BICROS hearing aid has its maximum sensitivity for frontally incident sounds. For other source directions, the real-ear gain may show pronounced peaks and troughs because of the addition and cancellation effects referred to earlier. Real-ear gain should thus be measured only for 0º, and it is particularly important that there be no reflecting surfaces near the patient. If this cannot be achieved, the response should be verified with the satellite microphone disconnected.

17.1.3 Stereo CROS (CRIS-CROS) aids

A third type of CROS aid that can be found in publications is the ***stereo CROS***, as shown in Figure 17.5. This arrangement can be thought of as two separate CROS aids. The left microphone feeds the receiver on the right side, and the right microphone feeds the receiver on the left side. This solution was invented with the *aim* of achieving high gain, combined with an open fitting, in both ears. The hope was that because each microphone is separated by the head from the receiver to which it is directly connected, the feedback path would be weak, and a high gain would therefore be possible. Unfortunately, this seemingly clever idea overlooks a critical issue: each microphone is close to the receiver connected to the microphone on the opposite side of the head, so there is still a complete feedback path that does not involve sound propagating acoustically around the head. This can be appreciated by following the red arrows in Figure 17.5. As a consequence, the stereo CROS configuration has no advantage and should not be used.

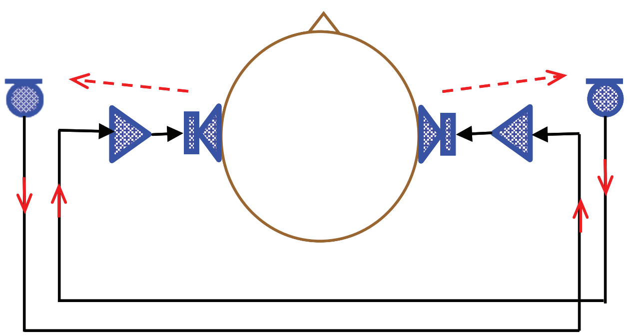

Figure 17.5 Block diagram of the stereo CROS hearing aid. The red arrows show a closed path around the components with the dashed lines representing the feedback leakage parts of the path.

17.1.4 Transcranial CROS aids

A ***transcranial CROS*** hearing aid (also known as a ***power CROS*** or ***internal CROS*** aid) transmits a signal from one side of the head to the other using bone-conducted sound.[1193, 1742, 1828] The arrangement is intended for a person with no useable hearing in one ear, but who has to listen to sounds arriving from that same side of the head. An ITE, ITC or CIC hearing aid is fitted to the non-functioning ear. Vibrations induced on that side of the head are coupled through the bones of the head to the cochlea on the opposite side of the head. To achieve the highest possible sensation level in the better ear, use the highest-powered hearing aid possible in the style chosen. Vibrations appear to get into the skull by two paths:[708]

- The hearing aid receiver creates a relatively intense SPL in the residual ear canal volume of the dead ear, and this vibrating air generates vibrations within the temporal bone.
- The hearing aid receiver vibrates the shell of the hearing aid, which in turn vibrates the canal wall.[536] To achieve a high sensation level in the better ear, the hearing aid should be deeply seated so that the case makes close contact with the bony portion of the ear canal (see Section 5.1). It is possible to use a CIC hearing aid in this way.[89]

Candidacy for transcranial CROS aids

The limits of effectiveness of transcranial aids and appropriate fitting methods are still being worked out. If the better ear has too much sensorineural hearing loss, the transcranial CROS fitting will not provide enough excitation to this ear. Also, the transcranial CROS can improve SNR significantly only if the level of sound reaching the better cochlea via the transcranial path is greater than the level that arrives by diffracting around the head to the eardrum of the better ear. Improved localization has been claimed for these devices, but it is difficult to see how this can occur if all sound is being perceived in a single cochlea (Section 15.1).

As vibrations are coupled from the hearing aid to the ear canal on the dead side in an almost accidental manner, there is presumably good scope for improvement in these devices by using output transducers, transducer mounting methods, and case designs that create a more effective source of vibration. See also the discussion on how BAHA achieves transcranial stimulation in a more controlled manner (Section 17.3.3).

Fitting procedure for transcranial CROS aids

A transcranial CROS aid has the same fitting goal (i.e. laterally balanced sensitivity) as a conventional CROS aid, but a different method of sound delivery. There is no sound pressure relevant to the fitting goal in either ear canal, so the fitting cannot be verified with a real-ear gain analyzer. The hearing aid should be adjusted so that sounds at all frequencies are as loud when they are presented from 45° on the poor side as when they are presented at 45° on the better side.

The accuracy of the transcranial fitting may be improved by placing a large earmuff over the ear and hearing aid opposite the loudspeaker, but this approach has not been experimentally verified. The extra isolation provided by the earmuff is most valuable at mid-frequencies where interaural level differences are not large enough for the contribution of the far ear to be insignificant compared to the near ear. Balancing can also be accomplished by achieving equal thresholds for presentation from each side, but in this case, the earmuff must not be used.[d]

17.2 Bone-conduction Hearing Aids

Bone-conduction hearing aids vibrate the structures within the cochlea without the sounds passing in the normal way through the middle ear. The output transducer is a vibrator known as a bone conductor (Section 2.11). Vibrations from the bone conductor have to be effectively coupled to the skull (and hence to the cochlea). To achieve adequate coupling, the bone conductor is usually mounted on one side of a headband, which uses spring tension to push the bone conductor against the head. Alternatively, the bone conductor can be mounted on the arm of a spectacle aid or can be strapped inside an elastic, fabric headband, or inside a cap. Apart from the output transducer, the remainder of the hearing aid is no different from a conventional (acoustic output) hearing aid. The hearing aid can be in a spectacle frame, in a BTE case or other small case mounted on the transducer headband, or in a body aid. Note that this section deals with non-implanted bone-conduction hearing aids. Implanted devices are considered in Section 17.3.

17.2.1 Applications of bone-conduction hearing aids

Bone-conduction aids are useful for four groups of people, almost all of whom have a conductive or mixed hearing loss.

The first group comprises people who, because of some medical condition, cannot wear a hearing aid that in any way occludes the outer ear. Typically this occurs when occlusion of the ear causes or exacerbates infections of the outer ear, or when the aid wearer has frequent infections of the middle ear combined with missing or perforated eardrums. Occlusion of the ear canal inhibits the ear drying out and can aggravate the infection.[1664] With a bone-conduction hearing aid, no part of the hearing aid obstructs the ear. An alternative that could be considered for some of these people is a BTE with a very open earmold, although it may not be possible to achieve enough gain with such open molds (Section 5.3.1). Also in this first group are some people who have undergone surgery using a canal-wall-down technique, for whom it can be difficult to obtain an adequate earmold.[1511]

A second group of candidates for bone-conduction hearing aids comprises those who have a congenitally malformed external ear (***microtia***), an absent external ear (***anotia***), an absent ear canal (***atresia***), an excessively narrow ear canal (***external auditory canal stenosis***) or a malformed middle ear. Vibration of the skull may be the only way to transmit sound to the cochlea. Most people with atresia have it in just one ear.[390]

A third group of people are those who have a large conductive hearing loss for any reason. Because the skull vibrations reach the cochlea without having to

[d] At first sight, it might seem that internal aid noise or external noise would invalidate the threshold by masking the signal arriving at the satellite microphone. It can, however, be shown that if the thresholds of the sounds incident from each side of the head are the same, the sensitivity will also be balanced for higher level sounds whether the thresholds were absolute thresholds or thresholds masked by internal noise or ambient noise. For this to be true, both paths must be operating, so an earmuff must not be used.

pass through the middle ear system in the usual manner, it may be possible to stimulate the cochlea more strongly with a bone-conduction hearing aid than with an air-conduction hearing aid (but see the next section).

The fourth group comprise people with unilateral sensorineural hearing loss which, within the bone conduction literature, is referred to as ***single-sided sensorineural deafness*** (***SSD***). This term usually implies that the degree of loss in the impaired ear is severe or profound. For people in this group, a bone conduction hearing aid worn on the deaf side functions as a transcranial CROS aid, as explained in Section 17.1.4.

People in the first three groups can be fit unilaterally or bilaterally. Bilateral fittings have two advantages. First, as with air-conduction hearing aids, each side of the head has its own microphone, so the SNR advantages created by head diffraction are always available. Second, binaural cues to localization will be available. Although clinicians learn during their training that there is little ***inter-aural attenuation*** (also referred to as ***trans-cranial attenuation***) from one side of the head to the other when stimulated by a bone vibrator on either side, there is sufficient attenuation to enable sound localization in the horizontal plane when bilateral bone conduction hearing aids are worn.[1103]

17.2.2 Bone-conduction hearing aid output capabilities

Because the output of a bone-conduction hearing aid is a mechanical vibration rather than a sound wave, these hearing aids can be measured electroacoustically only with equipment that measures vibration (IEC 373, ANSI S3.13). Furthermore, the amount of vibration they cause depends on the characteristics of the surface against which they are held. Consequently, bone-conduction hearing aids can be measured only when coupled to a ***mechanical coupler*** (that provides a mechanical impedance that matches that of the skull) and which incorporates a transducer to measure the applied force. The mechanical coupler is commonly called an ***artificial mastoid***, although it actually simulates the skull impedance at locations other than on the mastoid process.

Vibration is expressed in terms of the force (in Newtons or in µN) produced by the vibrator against the mass that represents the skull within the artificial mastoid. The vibratory force can be expressed in decibels relative to a ***reference force*** of 1 µN. The resulting number, equal to 20 times the logarithm of the actual force divided by the reference force, is then called the ***output force level*** (in dB). Because the input quantity (sound pressure) of a bone-conduction hearing aid is different from the output quantity (force), it is not sensible to talk about the *gain* of a bone-conduction hearing aid. Instead, we can talk about the ***acousto-mechanical sensitivity level***. Although this is a mouthful, it is simply equal to the output force level minus the input SPL, and is directly analogous to gain. It is the decibel equivalent of the output force divided by the input sound pressure.

Table 17.1 shows the maximum output force levels (***OFL90***) that can be produced at various frequencies by a typical high-powered BTE bone-conduction aid. The table also shows the force levels measured on an artificial mastoid when the same signal produces vibrations at threshold on the human mastoid for an average normal-hearing person (ISO 389-3). The final row of the table shows the sensation level that this bone-conduction hearing aid can therefore produce for a person with no hearing loss. Of course, patients with a sensorineural component to their hearing loss will receive sensation levels even lower than the sensation levels shown in the final row of Table 17.1.

Table 17.1 Maximum output force levels (OFL90) for a particular very high-powered BTE-style bone-conduction hearing aid, Reference Equivalent Threshold Force Levels (RETFL; ISO 389-3) for mastoid placement, and the resulting maximum sensation levels achievable for a person with normal bone-conduction thresholds (i.e. no sensorineural hearing loss). RETFL values given in ANSI S3.26 are very similar.

	Frequency (Hz)				
	250	**500**	**1000**	**2000**	**4000**
OFL90 (dB re 1µN)	107	122	122	119	104
RETFL (dB re 1µN)	67	58	42	31	35
Sensation level (dB)	40	64	80	88	69

How do these achievable sensation levels compare to those available with an air-conduction hearing aid? The answer depends on how much conductive loss is assumed. Table 17.2 shows the maximum acoustic levels that result when the same hearing aid referred to in Table 17.1 drives a receiver (inside the BTE case) rather than a bone conductor. These OSPL90 values can be subtracted from the thresholds of normal hearing referred to a 2-cc coupler (ISO 389-2) as shown in row 2. The resulting maximum sensation levels available to a person with normal hearing are shown in row 3. The sensation level available to someone with a conductive loss depends, of course, on the size of the loss. As an example, row 4 shows a large conductive loss, and the final row shows the resulting maximum sensation levels via air conduction for such a loss.

By comparing these with the values shown in the final row of Table 17.1, we can see that bone conduction provides a greater sensation level above 500 Hz, but air conduction provides a greater sensation level at 250 Hz. Of course, this conclusion depends directly on the degree of conductive loss assumed. As the conductive loss becomes smaller, the sensation level provided by the air-conduction aid increases. For conductive losses of 40 dB or less, this particular hearing aid provides more stimulation as an air-conduction aid than as a bone-conduction aid at all frequencies.

Based on this example, one would not select a (non-implanted) bone-conduction hearing aid with the sole aim of maximizing the input to the cochlea unless the patient's hearing loss has a conductive component of approximately 50 dB or greater.

The relative effectiveness of air- and bone-conduction hearing aids is not affected by the degree of any sensorineural loss, but is somewhat affected by the particular receivers and bone conductors used. A repeat of the calculations for an extremely powerful body-level bone hearing aid indicated that, for conductive hearing losses of 45 dB HL or less, it provided greater stimulation at all frequencies from 250 Hz to 6 kHz in air-conduction mode than in bone-conduction mode.

These examples also show that for a patient with a large conductive loss, achieving a high sensation level is not possible with either form of output transducer, particularly for low-frequency sounds. This deficiency is exacerbated if the patient also has a sensorineural loss.

17.2.3 Prescribing, adjusting and verifying electroacoustic characteristics for bone-conduction hearing aids

Few clinicians have access to an artificial mastoid. Consequently, bone-conduction hearing aids should initially be selected and adjusted based on the hearing aid specifications, followed by measurement of aided thresholds or the use of other subjective techniques.

Methods for prescribing the electroacoustic performance of air-conduction hearing aids are covered in Chapter 10. This section will describe how any such prescription can be converted into a prescription for a bone-conduction hearing aid. Suppose that some prescription formula has been used to deduce a target insertion gain, IG, for the *sensorineural* component of a mixed or conductive hearing loss (i.e. on the basis

Table 17.2. The first three rows show the maximum output of a sample air-conduction hearing aid (OSPL90), the Reference Equivalent Threshold Sound Pressure Level in a 2-cc coupler (RETSPL; ISO 389-2) and consequently, the sensation levels achievable with the hearing aid for a normal-hearing person. The final row shows the sensation level achievable for a person with the conductive loss shown in row 4.

	Frequency (Hz)				
	250	**500**	**1000**	**2000**	**4000**
OSPL90 (dB SPL, 2 cc)	128	130	137	132	127
RETSPL (dB SPL, 2cc)	14	5	0	3	5
Sensation level for normal hearing (dB)	114	125	137	129	122
Conductive hearing loss (dB HL)	60	60	60	60	60
Sensation level for maximum conductive loss (dB)	54	65	77	69	62

of the bone conduction thresholds), but that we wish to fit a bone-conduction hearing aid instead of an air-conduction aid. The acousto-mechanical sensitivity level, *A*, that on average results in a sensation level equal to that provided by an air-conduction hearing aid can be calculated from Equation 17.2:

$$A = IG + (RETFL - MAF) \qquad .....17.2,$$

where *RETFL* is the Reference Equivalent Threshold Force Level referred to an artificial mastoid (Table 17.1) and *MAF* is the Minimum Audible Field for normal hearing (ISO 226). Each of the quantities in Equation 17.2 may be different at different frequencies. The equation is easily understood: the bone conduction hearing aid acousto-mechanical sensitivity must be different from the insertion gain by the amount that, for normally hearing people, the force level at threshold exceeds the sound pressure level at threshold. No allowance has to be made for the conductive portion of the loss because the bone-conduction path bypasses the middle ear. For otosclerotic ears, the Carhart correction should be applied to the bone conduction thresholds prior to prescribing the hearing aid (Section 10.5).

Alternatively, one may have started from a prescription for the real-ear aided gain (REAG), rather than insertion gain, needed for the sensorineural part of the loss. The required acousto-mechanical sensitivity level, *A*, can be calculated from equation 17.3:

$$A = REAG + (RETFL - MAP) \qquad .....17.3,$$

where *MAP* is the Minimum Audible Pressure for normal threshold of hearing for air-conducted sound, referred to the average ear canal.

An analogous equation can be used to prescribe the maximum output for the bone conduction hearing aid, OFL90, in terms of the maximum output that would be prescribed for an acoustic hearing aid, OSPL90, for the same degree of sensorineural loss:

$$OFL90 = OSPL90 + (RETFL - RETSPL) \qquad .....17.4,$$

where *RETSPL* is the Reference Equivalent Threshold SPL (for normal hearing) in a 2-cc coupler (see Table 17.2). Table 17.3 gives suitable values for each of the terms in Equations 17.2 to 17.4.

Example of bone-conduction prescription

Suppose a subject with the audiogram shown in the first two rows of Table 17.4 is to be prescribed a bone-conduction aid using the NAL-NL2 procedure for gain (Section 10.2.2) and the NAL-SSPL procedure for OSPL90 (Section 10.7.3). Note that the IG shown in row 3 and the OSPL90 shown in row 6 are prescribed using only the sensorineural part of the hearing loss. If this table is used to construct a worksheet or spreadsheet, the correction figures in rows 4 and 7 will be the same for all patients. The acousto-mechanical sensitivity level shown in row 5 equals row 3 plus row 4. Similarly, OFL90 equals row 6 plus row 7.

Note that the values prescribed for OFL90 are considerably greater than the OFL90 values that are achievable with a high-powered BTE hearing aid (as shown

Table 17.3 Values for MAF (based on ISO 226), MAP (calculated as MAF plus REUG from Table 4.6), RETFL-MAF, RETFL-MAP, and RETFL-RETSPL. Similar computations can be made using the comparable ANSI standards.

	Frequency (Hz)				
	250	500	1000	2000	4000
MAF (dB SPL)	11	6	5	0	-4
MAP (dB SPL)	12	8	8	12	10
RETFL-MAF (dB)	56	52	37	30	39
RETFL-MAP (dB)	55	50	35	19	25
RETFL-RETSPL (dB)	53	52	42	28	30

in Table 17.1). Note also that the sensorineural portion of the loss in this example is particularly mild. *Consequently, the maximum output control of bone-conduction hearing aids can routinely be adjusted to give the greatest possible output, rather than being individually prescribed.* The adequacy of maximum output should be subjectively evaluated (Section 11.7) but there is little point in evaluating if the maximum output is excessive, as bone-conduction hearing aids will virtually never have sufficient output to cause loudness discomfort.

The gain-frequency response selection procedure defined by Equation 17.2 or 17.3, and illustrated in Table 17.4, should, however, be employed for each patient provided with a bone-conduction hearing aid. The target acousto-mechanical sensitivity level should be compared to the published specification for the hearing aid being considered. The appropriate tone control and gain settings can then be deduced. Manufacturers are encouraged to indicate the settings of the hearing aid programming adjustments that give an acousto-mechanical sensitivity equal to that shown in row 4 of Table 17.4. Adjustment for each patient then requires the controls to be changed so that the gain increases by the insertion gain desired for the sensorineural component of the patient's loss. Compression should be applied to at least the degree that is usual for that degree of loss.[e]

Two caveats must be applied to the prescription procedure and calculations of sensation level outlined in this chapter. Either of the following aspects of middle-ear function can cause the relationship between air-conducted and bone-conducted sounds to vary from that assumed in this chapter, and hence alter the prescription that is optimum for a patient.

- It has been assumed that the magnitudes of the conductive and sensorineural portions of the loss are known. Although it is common practice to determine these portions based on the air- and bone-conduction thresholds, it is well known that middle-ear disorders can elevate or suppress bone-conduction thresholds.[466] The Carhart notch (Section 10.5) for an otosclerotic loss is one example of this, but different effects, in both directions, occur for other types of conductive hearing loss.[466]
- High-level sounds from an air-conduction hearing aid pass through the middle ear, and are thus attenuated by the stapedius reflex. Prescriptive

Table 17.4 Calculation of the prescription for a bone-conduction hearing aid for the person whose air-conduction and bone-conduction thresholds are shown in the first two rows. The insertion gain (for a 65 dB SPL input level) and OSPL90 prescriptions, if an air-conduction aid were to be used for just the sensorineural part of the loss, are shown in rows 3 and 6. The correction figures in rows 4 and 7 are used to convert the prescriptions to bone-conduction specifications, in accordance with Equations 17.2 and 17.4.

		Frequency (Hz)				
		250	**500**	**1000**	**2000**	**4000**
1	AC (dB HL)	60	60	70	80	80
2	BC (dB HL)	10	10	20	30	30
3	Insertion gain for sensorineural loss (dB)	0	0	1	7	9
4	RETFL – MAF (dB)	56	52	37	30	39
5	Acousto-mechanical sensitivity level (dB)	56	52	38	37	48
6	OSPL90 for sensorineural loss (dB SPL)	98	92	94	100	105
7	RETFL – RETSPL (dB)	53	52	42	28	30
8	OFL90 (dB re 1 μN)	151	144	136	128	135

[e] The certain inability of the device to reach the OFL90 prescription means that the bone conduction hearing aid is likely to reach its maximum output at input levels lower than commonly occurs for air conduction hearing aids. The consequent degree of limiting can be reduced by applying WDRC with a compression ratio higher than usual.

formula, to the extent that they are influenced by average discomfort levels, account for the attenuation produced by the stapedius reflex. In a conductive hearing loss, however, the stapedius reflex generally does not affect middle-ear operation.[1351] Furthermore, if a reflex is present, it may *increase* the sensitivity for bone-conducted sounds.[256] In practice any effects of the acoustic reflex may be of no consequence, as they are unlikely to change the conclusion that the maximum output of bone-conduction hearing aids should be as high as is technically feasible.

How can the suitability of the electroacoustic performance of a bone-conduction hearing aid be evaluated? Certainly, we cannot directly measure insertion gain as we can with an acoustic hearing aid.

One possibility is to measure the aided sound-field thresholds. While this certainly indicates the softest sounds that will be audible to the aid wearer, it gives no indication of the sensation level that is achievable. Manufacturers could easily display this parameter within the fitting software, if the fitting software also enabled the clinician to indicate the output level at which the sound was just audible.

Subjective methods of verification, as outlined in Chapter 12, can always be used.

Correct operation of the hearing aid electronics (but not the bone vibration transducer) can be established by temporarily replacing the bone vibrator with a button earphone from a body aid, measuring the output in a 2 cc coupler, and comparing this measurement to the values measured using the same earphone attached to a bone conductor aid known to be in good working order. This measurement can be made at the time of fitting and stored for comparison with measurements made on the same aid if faulty operation is later suspected.

17.2.4 Disadvantages of bone-conduction hearing aids

Bone-conduction hearing aids have several disadvantages over air-conduction aids.

- The transducer has to be pushed against the head with at least as much force as the peak force imparted by the vibrator (or else the vibrator may bounce away from the head when it sends a signal). Continued use of conventional bone vibrators can cause hardened skin, permanent depressions in the skin, and pain. The reason for this is that because of the relatively small area that contacts the skin, the resulting application pressure exceeds the blood pressure inside capillaries in the skin. This applied pressure therefore causes the capillaries to collapse, depriving the tissue under the vibrator of the blood supply it needs to stay healthy.[1474] Fortunately, bone vibrators with larger contact areas will soon become available and will overcome this problem. These devices will also have lower distortion, and slightly higher output levels and bandwidths than conventional vibrators whose design has stayed unchanged for decades.
- The bone vibrator and the means to hold it against the skull are not small or discreet.
- The inter-aural attenuation for bone-conducted sounds is considerably less than for air-conducted sounds. Consequently, although it is possible to pick up different signals on each side of the head, it is not possible to deliver them independently to the respective cochleae. Binaural differences are thus smaller, but are still sufficient to allow localization and to take advantage of SNR differences between the ears made possible by head diffraction.[159]
- The inability to measure the output of the aid electroacoustically (without an artificial mastoid) makes it more difficult to check the functioning of these aids than is the case for air-conduction aids.
- The cable and plugs between the aid and the transducer can be unreliable with existing designs.
- The attenuation provided by the skin and the limitations of the transducer make it difficult to achieve an adequate low-frequency, and very high-frequency, response. As we have seen, the maximum output of bone-conduction aids is much less than optimal at these same frequencies and is less than optimal at all frequencies.
- The transducer and headband are easily dislodged.

Despite the considerable limitations of current designs, bone-conduction aids remain a better solution than air-conduction hearing aids for a small proportion of people with hearing loss, and newer designs are decreasing some of the above disadvantages. The bone-anchored hearing aid, described in the next section, removes the last three disadvantages and partially removes the first.

17.3 Bone-Anchored Hearing Aids

A ***bone-anchored hearing aid*** (***BAHA***) avoids many of the disadvantages of a bone-conduction aid. Like bone-conduction hearing aids, BAHAs also output a mechanical vibration, but transmit this vibration to the skull via a titanium screw embedded in the mastoid.[1784] Because the screw is titanium, the surrounding bone osseointegrates (i.e. bonds) to the screw. Most commonly, a head-worn BAHA is used, in which the microphone, amplifier and vibrator are contained within a single package. The package snaps onto an abutment that is screwed into the titanium screw fixture, as shown in Figure 17.6. This direct connection to the skull is called ***percutaneous coupling***. In the more powerful body-worn BAHA, only the vibrator is mounted on the head.

The BAHA's direct mechanical path to the skull enables vibrations to be more effectively and comfortably transmitted, because compression of the skin is avoided.[669] With a conventional bone-conduction hearing aid, most of the vibratory motion of the transducer is absorbed by the skin and subcutaneous soft tissues,[671] so above 600 Hz, a BAHA can provide around 10 to 15 dB greater stimulation of the skull than is possible by pressing the vibrator against the skin.[158, 670]

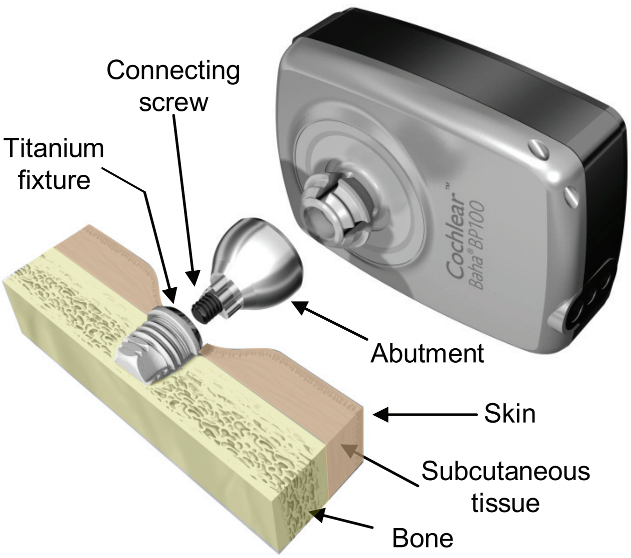

Figure 17.6 Bone-anchored hearing aid, showing its attachment through the skin to the bone. Used by permission from Cochlear Bone Anchored Solutions.

The surgery required to fit a BAHA is relatively minor; it can be done under local anesthesia on an outpatient basis.[263] In many countries, the BAHA has largely replaced traditional bone-conduction hearing aids for patients who have a permanent need for a bone-vibration hearing aid. The BAHA provides markedly greater physical comfort, is less visible, has greater output levels, and hence has better performance, than a conventional bone conduction hearing aid.[672, 1662] Not surprisingly, patients prefer it to conventional bone conduction hearing aids.[150, 194, 672, 1663, 1669] These preferences apply whether or not the device provides speech discrimination ability better than that provided by a bone-conduction aid.

The relative effectiveness of the BAHA and air-conduction hearing aids depends on circumstances.[1666] The greater the conductive loss of a patient, the more likely it is that a BAHA will be more effective than an air-conduction hearing aid.[1294] The BAHA is likely to provide greater sensation levels, and hence better performance, than air conduction hearing aids when the conductive component of the loss (i.e. the air-bone gap) is greater than about 30 to 35 dB.[412, 556, 1666, 1708]

Patients prone to ear infections also prefer BAHAs as they avoid the need for an earmold. The consequential reduction in ear infections is the most important benefit reported by these patients.[1294]

BAHAs appear to provide a satisfactory level of stimulation for patients with bone-conduction thresholds (average of 500, 1000, 2000 and 3000 Hz) up to about 45 dB HL for head-worn hearing aids and up to 60 dB HL for the most powerful body-worn aid.[13, 158, 160, 1666] Neither of these “limits” is absolute. BAHAs become more effective as the degree of sensorineural loss in the better ear decreases.[488] The better the bone conduction thresholds in the better ear, the better the aided thresholds will be after aiding, and the greater the sensation levels that the BAHA can provide.[1407] Consequently, speech intelligibility in quiet will also be greater, particularly for soft speech.[1407]

The effectiveness of a BAHA for a particular patient can be tested without implantation. In one method, a test rod is temporarily attached to the BAHA transducer, and the patient grips the test rod between his or her teeth (with the lips closed to prevent feedback) thereby bypassing the loss through the skin. Coupling to the skull via the permanent screw fixture is slightly more effective than the temporary coupling provided by the teeth, even more effective for the high frequencies.

More conveniently, but less effectively, the rod is held against the mastoid or frontal bone. A convenient temporary fitting is a steel headband that holds the BAHA against the side of the head, just like a conventional bone conductor hearing aid. This conventional method of application is called ***transcutaneous*** stimulation. It is recommended that patients trial this for a few weeks prior to surgery to ensure they have realistic expectations about the performance they will achieve.[f], [1666] Another temporary solution is to use a soft elastic band to hold the BAHA in place. This option is suitable for long-term use for children too young to receive the bone implant. With either type of temporary band, the BAHA provides about the same stimulation level as a conventional bone conductor hearing aid (which it is when used in this mode),[753] but lower stimulation than when it is implanted.[670, 1855] Because its surface area is greater than a conventional bone conduction aid, it is more comfortable to wear.

The types of hearing disorders for which a BAHA is potentially beneficial are the same as for non-implanted bone conductor hearing aids, as covered in Section 17.2.1. The following sections discuss the benefit of BAHAs for three important groups of patients. As we will see, it is common for the benefit subjectively reported to be greater than the benefit objectively measured. Although the first two sections are focused on conductive loss, there can be some sensorineural loss in the better cochlea.

Bone anchored hearing aids are now made by more than one company, and the term "Baha" is a trademark of one of these companies. More general terms also in use are ***bone-anchored implant*** and ***osseointegrated auditory implant***.[g] As the research on which this section is based was mostly performed with BAHAs, the original term *BAHA* will be used, but the principles explained in this section should be applicable to any hearing aid that transmits vibration to the skull by penetrating the skin. Most of the principles are also applicable to conventional bone conduction hearing aids applied across the skin, except that the achievable sensation levels and wearing comfort of existing bone conductors is less.

One newer device has been designed to vibrate the skull via the teeth – not for temporary testing – but as the permanent solution.[1447] The vibrator worn inside the mouth receives its signal from a microphone located in the ear canal, which means that cues to sound location provided by the pinna are available. Another advantage is that no surgery is needed.

17.3.1 BAHAs for unilateral conductive or mixed hearing loss

Studies into the effectiveness of unilateral BAHAs for unilateral conductive or mixed losses indicate inconsistent benefits based on performance measured in the clinic, but consistent self-reported real-life benefit. The following have been reported:

- Localization with BAHA may be slightly better than a patient with unilateral loss can achieve unaided,[754, 1665] or little different from unaided.[1463] (The latter result is not surprising, as the BAHA will stimulate both cochleae, so there will still be no interaural cues to localization.)
- Speech intelligibility with BAHA is likely to be better than unaided when the SNR is better on the impaired (aided) side of the head,[488, 754] and no different from unaided when the SNR is the same on both sides of the head.[488] Reassuringly though, even when noise is presented from the aided side of the head, the BAHA rarely makes speech intelligibility significantly worse.[488]
- Most recipients use their devices for most waking hours so, by inference, derive benefit from the device in ways that are not apparent from measurements made in the clinic.
- Most recipients subjectively report that the BAHA helps them,[488, 754, 1170, 1463, 1900, 1901] thus improving their quality of life.[64, 499]

Variation in results across studies is not surprising as benefit must be affected by the degree of both the conductive and the sensorineural components of the loss. As the degree of conductive loss (i.e. the air-bone gap) increases, unaided difficulty must increase, but the stimulation the BAHA provides to the cochlea is

[f] Performance will actually be better after aiding because of the higher sensation levels that will be possible with the more direct connection to the skull.

[g] These terms, as well as the word *Baha*, but not the acronym *BAHA*, help distinguish implanted bone-vibration devices from non-implanted hearing aids, which is important for medical reimbursement in some countries.

approximately unaffected.[h] Conversely, as the degree of sensorineural loss increases, the sensation level that the BAHA can provide decreases. Consequently, the benefit provided by the BAHA *must* increase with the degree of conductive loss and decrease with the degree of sensorineural loss. An additional issue concerns whether the loss is congenital or acquired. Patients with congenital unilateral conductive loss may not have acquired the binaural processing mechanisms needed to obtain good localization and binaural release from masking if implanted later in life.[1665]

17.3.2 Bilateral BAHAs for bilateral conductive or mixed hearing loss

Patients with bilateral conductive or mixed losses can benefit from bilateral BAHAs. Although each BAHA transmits vibrations to both cochleae, the cochlea ipsilateral to the BAHA receives greater stimulation than the cochlea contralateral to the BAHA and receives it around 200 µs earlier.[1707] The amount by which stimulation of the ipsilateral cochlea exceeds that at the contralateral cochlea, i.e. the trans-cranial attenuation, increases with frequency, and on average is around 10 dB in the high-frequencies, but varies greatly between patients.[162, 1346, 1707] Bilateral BAHAs can therefore provide some degree of dichotic stimulation. The resulting inter-aural difference cues will not be as strong as for normal hearing, or for air conduction hearing aids, because of the significant cross-stimulation relative to that which occurs for air-borne conduction of sound.

Nonetheless, the binaural cues produced are sufficient to improve localization relative to a single BAHA.[159, 1464, 1661, 1839]

Bilateral BAHAs provide better speech intelligibility in noise than unilateral BAHAs whenever the SNR at the unaided ear in the unilateral condition is greater than the SNR at the aided ear.[159, 497, 1661, 1839] Effectively, bilateral BAHAs ensure that if one side of the head has a better SNR than the other, the patient is never put in the position of having to rely on only the sounds coming from the side of the head with the poorer SNR.

As well as enabling the wearer to take advantage of head diffraction, the binaural cues are sufficiently strong to enable binaural squelch to occur. This is evidenced by bilateral BAHAs enabling a significant binaural masking level difference (Section 15.2.2) for low-frequency sounds.[159]

Bilateral BAHAs also enable lower (better) speech reception thresholds in quiet than is possible with a single BAHA.[159, 1464, 1661, 1839] This is presumably partly caused by each cochlea being stimulated by two BAHAs,[1707] and partly by central mechanisms benefiting from binaural redundancy, also referred to as binaural summation (Section 15.3.1).

Given these several advantages, it is not surprising that quality of life scores for bilateral BAHAs are greater than for unilateral BAHAs, although the difference is not large.[742] A single BAHA provides considerable benefit (relative to unaided) for people with bilateral conductive loss, irrespective of the location of the speech and noise.[488]

17.3.3 BAHAs for single-sided sensorineural deafness

For people with single-sided deafness (SSD; i.e. unilateral sensorineural loss), a BAHA can be mounted on the side of the head with the deaf ear so that it transmits vibrations through the skull to the cochlea on the opposite side (Section 17.1.4). The goal is that when the side of the head with the BAHA has the better SNR, this higher quality signal will be made available to the only functioning (or better functioning) cochlea.

The effect of the BAHA on speech intelligibility therefore varies greatly with the direction of arrival of the target speech and competing signals. When head diffraction causes the SNR to be better on the deaf side, the BAHA increases intelligibility,[488, 750, 751, 1067, 1947] although depending on the exact location of speech and noise sources, the effect may be too small to be measurable.[162] At distances from the sources significantly larger than the room's critical distance (Section 3.4) the SNR will be the same on both sides of the head, and the BAHA should not provide any benefit.

Conversely, when the SNR is better on the normal-hearing side, the BAHA decreases intelligibility because the signal it transmits to the normal-hearing cochlea is less clear than the signal that cochlea would otherwise receive.[488, 750, 1067] The net benefit to the

[h] This is an approximation, as the amount of vibration imparted to the cochlea is affected by the mechanical impedance of the middle ear system, as seen looking out from the cochlea, and this is differently affected by different conductive abnormalities.

patient thus depends on the magnitude of the advantage in some situations relative to the magnitude of the disadvantage in other situations. Based on objective measurements in the clinic, these advantages and disadvantages seem about equal, leading to no net benefit in SNR averaged across situations.

The benefits reported by wearers in real life do not, however, reflect these equally offsetting advantages and disadvantages.[488, 750, 1067, 1947] Possibly the reason is that the BAHA effectively removes the most disabling condition – when speech is on the impaired side and noise is on the side of the good ear. Although the BAHA also makes the easiest situation (speech on the good side and noise on the impaired side) more difficult when aided, communication may remain acceptably easy.

The beneficial self-reports may also partly result from self-selection bias. In one study where patients with SSD had the opportunity to use the BAHA with a temporary headband, the 63% who subsequently chose to acquire the BAHA reported significant benefit in real life. The remaining 37% reported that it made little difference to their communication ability.[963] Fortunately, the speech intelligibility in noise performance when implanted with a head-worn BAHA is well predicted by the same measurements made pre-implantation with the more powerful body-worn BAHA on a headband.[1659] The increased output power of the body-worn BAHA approximately compensates for the loss of vibration strength when it is applied through the skin via a headband.

The sole reason for using a hearing aid with single-sided deafness is to capture the benefits of head diffraction, and these benefits are greater for high-frequency sounds than for low-frequency sounds. Attenuating the frequencies below 1500 Hz, where head diffraction effects are weak, appears to retain most of the beneficial effects of BAHAs in this application, while reducing some of the disadvantage created by a BAHA when the SNR is better on the unaided side of the head.[1406]

Note that a BAHA will *not* improve localization for people with SSD.[488, 750, 963, 1067] All sounds will still be perceived through a single cochlea, so despite sounds being picked up on both sides of the head, no binaural cues are available to assist with localization. In fact, with a BAHA, head movements will produce less marked changes to the level and spectral shape perceived, so localization with the BAHA may be worse than without it, especially when the BAHA is first heard and the patient is unfamiliar with the new monaural spectral cues to localization that exist when the BAHA is worn.

The BAHA provides less benefit to patients with SSD than it does for patients with unilateral or bilateral conductive loss.[488, 1170] Both the latter groups obtain some benefit because the device enables the patient to perceive sounds arriving at both sides of the head. Only those with two functioning cochlea are enabled to hear sounds in both cochlea, thus enabling binaural processing mechanisms that remain unavailable to the first group. Patients with SSD do continue to use and report benefit from the BAHA a year or more after surgery.[751] Although the reasons for this are not completely apparent from objective data, it is becoming common practice to implant these patients with BAHAs.

17.3.4 Complications with BAHAs

There appears to be a very low incidence of problems associated with penetrating the skin.[1449] Complications comprise infection and inflammation surrounding the abutment and failure to osseointegrate. The rate of problems in both of these areas appear to be decreasing with later generation designs.[490]

It is critical that the patient, or the patient's carer, be instructed in regular but gentle cleaning of the skin and the abutment. It is also essential that patients avoid receiving a blow to the BAHA. In rare cases, fixtures in children have been knocked out by physical trauma. The major factor limiting application of the BAHA in children is that bone thickness and composition are not adequate for implantation until children are around three years of age.[1666]

17.4 Middle-ear Implantable Hearing Aids

A further alternative to air conduction or bone conduction hearing aids are ***middle-ear implant hearing aids***. These devices apply a mechanical vibration to the middle ear system or round window, bypassing the need for an acoustical output from the hearing aid. Consequentially, there is no need for an earmold. Middle-ear implants can be fully implanted with no external components, or partly implanted with the output transducer implanted but the microphone, battery, amplifier and transmitting device worn on the outside of the body.

If the microphone and battery are not implanted, then a signal has to be transferred from the external system to the vibrating transducer in some way. Most commonly, the amplified signal is sent inductively across the skin, from an external coil to an internal coil (just like in cochlear implants), and then on to the internal stimulator. Other systems use a coil external to the skin to create a fluctuating magnetic field inside the middle ear that acts directly on an implanted magnet.

Fully implanted hearing aids need an implanted battery which, with current technology, must be replaced every 3 to 10 years.

17.4.1 Output transducers

Middle-ear implants use either electromagnetic or piezoelectric output transducers, also called ***stimulators***, which cause vibrations in the middle ear system or cochlea in essentially four ways.

Floating electromagnetic transducers

The most researched, and clinically applied, transducer is an electromagnetic device that relies on the same inertial principle as the bone conductor, but which transmits the vibration via a more direct and sensitive sound path. The ***floating-mass transducer*** comprises a coil inside which a magnet is loosely suspended. When an alternating current is applied to the coil, the magnet and the coil move with respect to each other. Because the magnet has appreciable mass relative to the coil, its inertia restrains its motion, so the inertial force between the magnet and coil causes the coil to vibrate.

If the coil is attached to one of the middle-ear ossicles or to the round window, vibrations are transferred to the cochlea. The original and most common way of attaching it is to clip it onto the long process of incus.[1785] For patients for whom the ossicular chain is no longer intact, the transducer can be placed in the round-window niche (after widening) so that the coil presses against the round window.[315, 896, 1735] Alternatively, it can be clipped to the stapes or pressed against the footplate of the stapes (i.e. driving the oval window).[374, 569, 786]

Although the magnet has significant mass relative to the coil, their combined 25 mg mass is sufficiently small that attachment of the complete transducer to the incus usually has only a very small effect on hearing thresholds when it is not being electrically driven, once the ear has recovered from the surgery.[566, 1081, 1566, 1670, 1719, 1860] As the average change varies by about 5 dB between studies, surgical technique may influence the magnitude of change. Measurements on cadavers indicate that the effects are limited to the high frequencies, as would be expected for addition of a mass.[1308]

The floating-mass transducer is also referred to as a ***vibrating ossicular prosthesis*** (***VORP***).

Split electromagnetic transducers

Split electromagnetic transducers comprise a small permanent magnet vibrated by the magnetic field generated by a coil, but otherwise physically disconnected from the coil, or any other part of the implant. When a current passes through the coil, the magnet vibrates by an amount proportional to the variations in current through the coil. This is the same principle as used in loudspeakers and in receivers for hearing aids (Section 2.6) except that in the middle-ear implant transducer it is the magnet that moves, not the coil.

With most of the middle-ear implant systems, the magnet is firmly attached to some point in the ossicular chain, enabling vibrations to be transferred from the magnet directly to the middle-ear system. Mounting points for the magnet have included the tympanic membrane,[853] the incus,[567, 1130, 1785] the incudo-stapedial joint,[768, 770] and the round window.[1682] The coil that drives the magnet can either be external to the ear in a hearing aid case or custom shell[770, 853, 1682] or in the middle ear cavity.[1130]

Anchored electromagnetic transducers

In anchored electromagnetic transducers, either the permanent magnet or the coil is attached to the bone surrounding the middle ear cavity, and the remaining component is attached to some point in the ossicular chain. The vibrating force between the magnet and coil is thus transferred directly to the ossicular chain.[118a] In the Otologics stimulator, the body of the transducer is anchored within a cavity in the mastoid and transmits vibrations to the middle ear chain via a coupling link. Originally, the link was inserted into a hole cut by a laser into the incus,[854] but it can also couple to the round window.[1035] The coupling has minimal effect on hearing thresholds when the device is inactive.[806, 1145]

Anchored piezoelectric transducers

Piezoelectric transducers are based on a ceramic material that changes its shape when a voltage is applied to it. One end of the transducer is anchored to the bone surrounding the middle ear cavity and the free end connects to some point in the ossicular chain, thus transferring sound vibrations to the cochlea. In one implant system, the free end terminates in a thin plate that is sandwiched between the incus and the stapes by inserting it into the incudo-stapedial joint.[487] In another system, vibrations are coupled to the ossicular chain by a water-filled flexible tube terminating in a balloon tip.[787] Piezoelectric transducers inherently have a wide frequency response.

All stimulators must cope with the movements of the ossicular chain in response to changes in atmospheric pressure relative to static middle ear pressure (or vice versa). The movements associated with these pressure changes exceed by many orders of magnitude the amount of vibration associated with conduction of sound through the middle ear. There is intrinsically no restriction by the floating mass transducer (because there is only a single point of attachment) or by the split electromagnetic transducers (because the coil and the magnet are individually attached to different structures in the ear and have no rigid connection between them). With anchored transducers, freedom of movement for large slow changes has to be allowed for in the design.

17.4.2 Microphones

Microphones for middle-ear implants have (so far) been implemented in four different ways.

Externally located. Microphones for partially implanted hearing aids are located in the externally worn case, either behind the ear or against the scalp immediately lateral to the implanted electronics package. These microphones are identical to those used in conventional hearing aids, and may be directional or omni-directional.

Subcutaneous in the scalp. Microphones in hermetically sealed cases are positioned directly beneath the skin. The cases have extra large diaphragms that are designed to minimize the loss of signal strength caused by transmission of sound through the skin.

Subcutaneous in the ear canal. These microphones are also mounted in hermetically sealed cases. They have smaller diaphragms, which decreases their sensitivity, but because they are mounted under the skin in the bony part of the ear canal, their input is increased by the diffraction and resonance characteristics of the pinna and ear canal. They therefore acquire the directional properties of the normal ear.[415, 1101]

Transducer in the middle ear. These microphones are created by locating a hermetically sealed transducer inside the middle ear cavity, positioned so that it is driven by the natural vibration of one of the structures of the middle ear. One system available commercially uses the malleus or incus motion to vibrate a hermetically sealed piezoelectric transducer.[i, 268] The main benefit of middle ear microphones is that they take benefit of the directional and resonance properties of the pinna, concha and ear canal. On the other hand the surgery is more complex than placing a subcutaneous microphone.

The last three methods are suitable for fully implanted hearing aids. Those using microphones under the skin have to be designed to minimize the loss of sensitivity that is caused by detecting sounds after they have passed through the skin, rather than sensed while they are still air-borne sounds.[415] While sensitivity can be increased electronically, the low microphone sensitivity is accompanied by increased internal noise, which is not so easily rectified. Another issue that designers of implanted microphones have to deal with is their tendency to pick up body-borne noises such as those caused by walking, breathing, chewing, touching the head surface, or even just blood flowing through arteries and veins.

A second problem is caused by the desire to keep the ossicular chain intact in case implanted hearing aid use is discontinued. An intact ossicular chain implies that if the output transducer is coupled to the ossicular chain, the microphone cannot be, or feedback oscillation will occur, even for very low gains. This is unfortunate, because the tympanic membrane is perfectly placed to convert air-borne vibrations to mechanical vibrations that could be detected by a suitable sensor. The eardrum is so perfectly placed to initiate vibrations, some designers have opted to use it (or the malleus) but to disarticulate the incus from the stapes.[268]

17.4.3 Complete systems

There are three complete middle-ear implants and one other hearing aid with an implanted component that, at the time of writing, are approved for implantation in at least some countries. These systems use some of the components described in the previous section.

[i] Piezolectric crystals, like most transducers, are reciprocal devices. Just as a voltage applied to the crystal causes motion, vibration of the crystal cause a voltage to be generated.

Vibrant SoundBridge

The Vibrant SoundBridge by Med-El (previously by Symphonix Devices) is a partially-implanted hearing aid. The externally worn audio processor contains the microphone, amplifier and battery. It inductively sends an alternating magnetic signal across the skin to a receiving coil, which produces a corresponding alternating electrical signal. This is conveyed to the floating mass transducer, which is attached to the incus, or placed against the round window,[315] or less often to some other location in the middle ear system.[896, 1081] Stable performance over periods in excess of 5 years have been reported.[1252] Figure 17.7 shows the location of the key components (microphone and amplifier, receiver, and floating mass transducer) in this system.

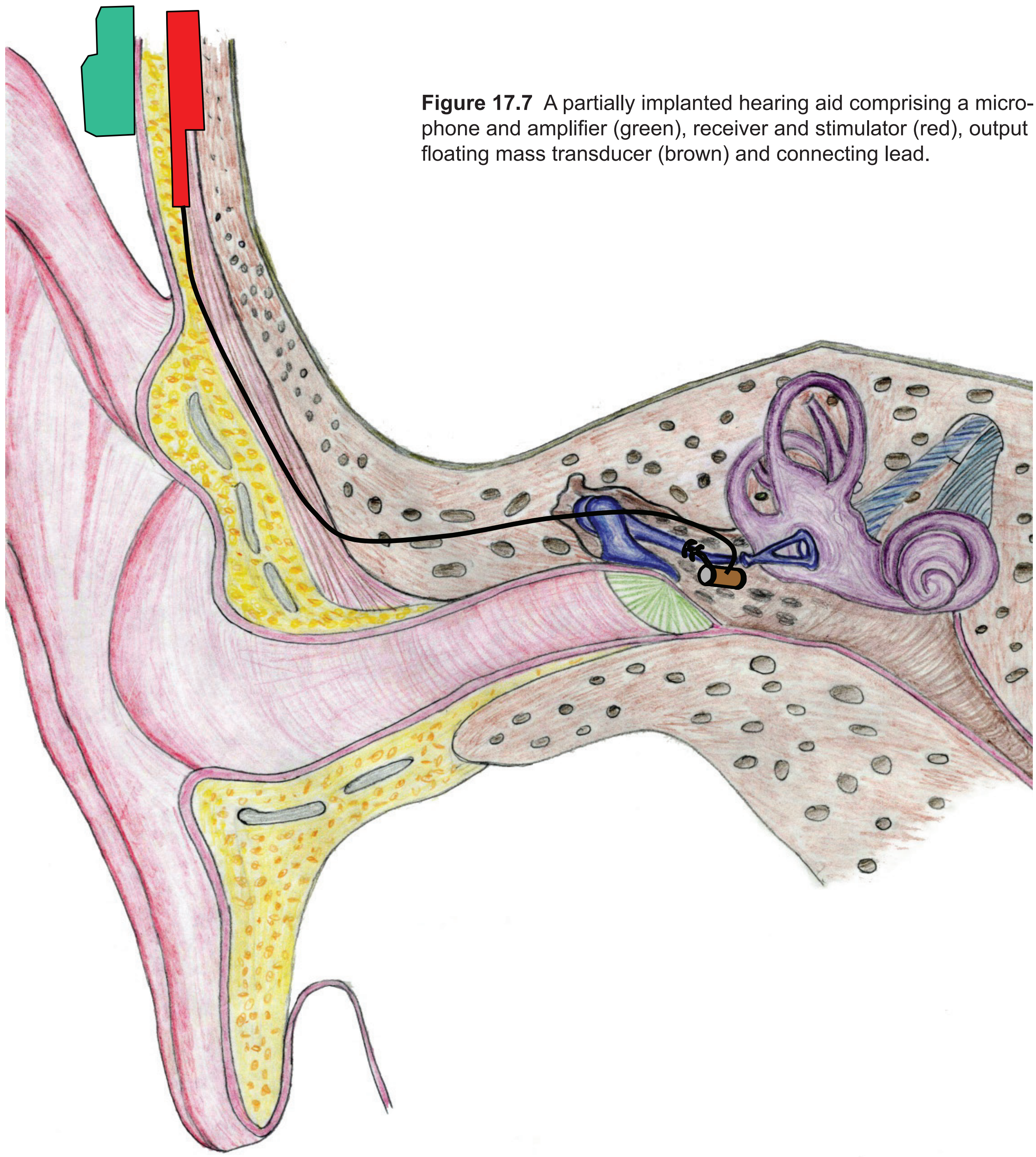

Figure 17.7 A partially implanted hearing aid comprising a microphone and amplifier (green), receiver and stimulator (red), output floating mass transducer (brown) and connecting lead.

Carina

The Carina by Otologics is a fully-implanted hearing aid. The microphone is mounted under the skin on the mastoid near the main package, which contains the electronics and the rechargeable battery. The body of the electromagnetic stimulator, referred to as the ***middle ear transducer*** (***MET***) ossicular stimulator, is anchored in the bone of the mastoid and connects via a ***partial ossicular replacement prosthesis*** (***PORPS***) or ***total ossicular replacement prosthesis*** (***TORPS***) to the incus, stapes, oval window, or round window, depending on the state of the individual middle ear.[805, 854, 1035, 1145, 1796a]

Aided thresholds are around 30 dB HL,[1035] presumably limited by the equivalent internal noise of the fully implanted microphone. The hearing aid is programmed via magnetic induction transmission. A remote control uses the same transmission method to enable volume adjustment by the wearer. An earlier version was driven by an externally worn microphone, amplifier and battery, with the amplified signal sent inductively across the skin.[806]

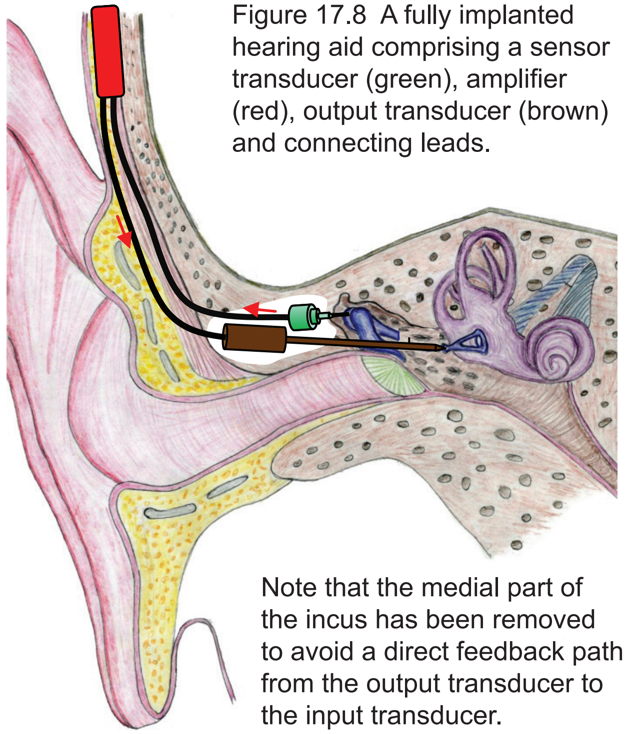

Figure 17.8 A fully implanted hearing aid comprising a sensor transducer (green), amplifier (red), output transducer (brown) and connecting leads.

Note that the medial part of the incus has been removed to avoid a direct feedback path from the output transducer to the input transducer.

Esteem

The Esteem by Envoy Medical (previously by St Croix Medical) is a fully-implanted hearing aid. The microphone comprises a piezoelectric transducer driven by the malleus or the short process of the incus, which of course are driven by the eardrum. Its output is amplified by an electronics package mounted under the skin behind the ear, and the amplified signal drives a second piezoelectric transducer, this one functioning as a stimulator, that vibrates the head of the stapes via a connecting rod. To avoid mechanical feedback oscillation, the ossicles are disarticulated by removing the most medial 1 to 2 mm of the incus. The drive is so efficient that the battery is non-rechargeable and must be replaced, by very minor surgery, approximately every 5 years.[268] Figure 17.8 shows the three major components (sensor transducer, amplifier, and output transducer) in a fully-implanted hearing aid based on these principles.

Retro-X

The Retro-X hearing aid, although not a middle-ear implant, does require a component to be surgically implanted. A hollow titanium tube is inserted between the back of the pinna and the ear canal. The hearing aid receiver, mounted in a BTE case, sends the output signal down the tube. The ear canal therefore remains entirely open.[587] A disadvantage is that the microphone, which is omnidirectional, is shielded by the pinna to a greater degree than is usual in BTE hearing aids, resulting in greater sensitivity to the rear than to the front for high-frequency sounds.

Other systems that have been used in research studies include:

The ***Maxum*** partly-implanted hearing aid (previously called Soundtec) combines an externally worn microphone, amplifier, and battery, with a driving coil embedded in an earmold and deeply placed in the ear canal, and a magnet inserted into the incudostapedial joint.[770, 1523] An average functional gain of 26 dB has been reported.[1630]

The ***totally-implanted cochlear amplifier*** (***TICA***), previously developed by Implex, combines a microphone under the skin in the posterior osseous ear canal, an amplifier and rechargeable battery in the mastoid, and a piezoelectric stimulator anchored in the mastoid driving a connecting rod that terminates in a laser-cut hole in the incus.[1962] Alternatively, it can drive the stapes head or oval window.[1100]

The ***Rion device E-type*** partly-implanted hearing aid, combines an external microphone, amplifier and battery, transcutaneous inductive coupling, and an implanted piezoelectric stimulator that drives the stapes.[1934] Long-term use of over 10 years has been achieved in some of the patients receiving it, albeit with some reduction of output level over time.[1935]

The ***direct acoustic cochlear stimulator*** (***DACS*** - also known as ***Codacs™***) by Cochlear, and the ***Ingenia*** by Phonak, combine an external microphone, amplifier and battery, transcutaneous inductive coupling, and an implanted electromagnetic stimulator that drives the perilymph of the cochlea via a stapes prosthesis vibrating through a hole in the oval window.[694a] Note that as the implanted components directly drive the fluid of the cochlea, these devices are better classified as ***direct acoustic cochlear implants***.

Several of the systems described in this section are reviewed in more detail by Haynes et al (2009).

17.4.4 Candidacy and benefits

Although middle ear implants were originally implanted only in patients with normal middle-ear function, the group of patients who can gain the most from middle-ear implants appear to be those with some problem in the external or middle ear or mixed losses with a significant conductive portion. This includes:

People unable to wear an earmold: either because it exacerbates infections in the external ear, or because of malformation of the ear canal or pinna, whether congenital or the result of disease or surgery. Just as with BAHAs, those patients for whom conventional hearing aids (with an earmold) cause or exacerbate infections in the ear canal report the greatest benefit from middle-ear implants.[1660]

People with dysfunctional middle ear systems that cannot be corrected by surgery. Vibration of the round window is becoming increasingly common.[315] This stimulation method has the advantage that it is viable even when there are no ossicles, or when the stapes' footplate has become immobile because of otosclerosis.

Because movement of the middle ear ossicles is very small (120 dB SPL at the eardrum causes only 1 µm movement at the stapes at mid frequencies and even less at higher frequencies),[639] and because middle-ear implants directly drive the mechanical vibration path to the cochlea, they can stimulate the cochlea more strongly than BAHAs or other bone conduction hearing aids. Floating mass transducers can drive the cochlea at levels equivalent to 110 dB SPL for air-conducted sound in a normal ear,[582] and anchored stimulators are able to drive the cochlea at levels equivalent to 135 dB SPL[854] or 145 dB SPL.[1961] They are thus more suitable than any bone-conduction devices for people with mixed losses that include a sensorineural loss of moderate or greater degree.

The benefits claimed, and in some cases demonstrated, for middle-ear implants, relative to conventional hearing aids, include:

- a more extended high-frequency response (to 10 kHz for some systems), and lower non-linear distortion (less than 1% total harmonic distortion for some systems), and hence greater signal clarity;[582, 854, 1961]
- greater speech intelligibility in quiet or in noise, presumably as a consequence of the higher gain and bandwidth, and lower distortion as reported in the preceding dot point;[769, 770, 1789, 1822]
- no obstruction of the ear canal (and hence no occlusion effect);
- for fully-implanted devices, unrestricted use in hot, dusty or wet environments, combined with invisibility, and no need for any handling of the device; and
- increased gain before feedback oscillation occurs, particularly for the high frequencies.[566, 769, 770, 1768, 1789, 1798] Increased gain is an advantage if it is not otherwise possible to achieve the gain prescribed for a patient, and should lead to higher speech intelligibility in quiet for weak input levels. Note that feedback oscillation can, in principle, occur even if the output is mechanical vibration rather than air-borne sound, because the vibrating middle-ear system will cause the tympanic membrane to radiate sound, an attenuated version of which will be picked up by the hearing aid microphone, wherever it is. The greatest gains therefore seem possible for patients with disarticulated middle-ear systems.

Just as with BAHAs, it is common for patients to report benefit relative to conventional hearing aids even when none can be shown with objective measurements in the clinic.[566, 1081, 1523] The reasons for this are currently unclear.

Because criteria for who would benefit more from an implanted aid than a conventional hearing aid are still changing, it is difficult to be certain about the proportion of the hearing-impaired population for whom the extra costs and risks of surgery outweigh the benefits. One estimate based on applying multiple criteria to a very large database of patient details has indicated that considerably less than 1% of hearing-impaired people are candidates[842] but the situation may change markedly in the future.

It is feasible to use a middle-ear implant in one ear and a conventional hearing aid in the other.[1541] The precise effect of the combination on localization and perception of spatially separated signals will likely depend on the relative processing delays and the sensation levels achieved by the two hearing aids.

17.4.5 Complications with middle-ear implants

Although the operation is regarded as minor, middle-ear implants are not always successful. Revision surgery is sometimes necessary, and permanent alteration of taste sensation can occur.[1566] Excessively tight attachment of the floating mass transducer to the incus can cause necrosis of the incus.[374] Conversely, if an implanted component supported by the ossicles is too loosely attached, it can move and create sensations unrelated to any external sound, although this problem is minimized with suitable attachment methods and surgical techniques.[1630] Implanted components can become dislodged or extruded, and if the particular implant requires components to be positioned deeply in the ear canal by the wearer, this can be sufficiently difficult that patients give up.[264]

A potential limitation of any implanted device that includes a magnet is that it may preclude the patient from undergoing magnetic resonance imaging (MRI) scanning should this be required to investigate other health concerns any time after implantation. The applied magnetic field may move or twist the internal magnet, potentially damaging the anatomical structures to which the magnet is attached, and may demagnetize the magnet. The internal magnet will also distort the MRI image in the area around it.

Despite these concerns, the magnet in the Soundtec middle-ear implant has been shown to be safe when scanned in a low field strength (0.3 T) MRI scanner,[501] and the magnet in the Vibrant Soundbridge has been shown not to become demagnetized, nor to damage surrounding tissues when scanned in a medium field strength (1.5T) MRI scanner, although scanning did re-position the transducer slightly in some cases.[1788] Piezoelectric transducers do not contain magnetic parts and should intrinsically be safer in MRI scanners.

17.5 Concluding Comments

Hearing aids in the CROS family are not extensively used, even for many patients with hearing losses for whom a CROS fitting would appear to have some advantages. The rapid development of technology may affect this situation in the near future, and there may be some merging of CROS amplification concepts with other signal processing ideas. For example, when miniature bi-directional wireless links between the ears are widely available, it will be possible for the sound presented to each ear to be a desired combination of the sound picked-up by microphones on each side of the head, without the inconvenience of any cables. Such a combination could range from a simple linear addition of sounds, as in a BICROS fitting, to processing that adaptively reduces noise and reinserts cues to localization, as is done in processing for virtual reality.

For both bone-anchored hearing aids and middle-ear implants, it is common to judge their effectiveness in restoring hearing by the aided thresholds that are achieved. This is an extremely inadequate and potentially misleading measure. Provided the physical separation of the microphone from vibrating surfaces driven by the output is sufficient to avoid feedback oscillation, aided thresholds can be reduced just by adding more electronic gain between the microphone and the output transducer. If, however, the maximum output of the device results in a sensation level only 10 dB above threshold, then low aided thresholds means that a wide range of sounds in the environment have to be crammed into a very restricted dynamic range of hearing, which degrades signal quality.

There is no single metric that captures the adequacy of the fitting, but the maximum achievable sensation level (in the absence of compression) is an important metric that should be considered when evaluating the appropriateness of devices. Whenever the maximum sensation level is less than the inherent dynamic range of the patient (i.e. discomfort level minus threshold), then the hearing aid is not enabling the patient to make full use of his or her remaining hearing. It would therefore be valuable if manufacturers of implanted devices included in the fitting software a display of the maximum undistorted output level, relative to the output level needed to reach threshold.

There is a great need for studies of the relative effectiveness of (properly fitted) CROS hearing aids, conventional open-fit hearing aids, BAHAs, fully implanted middle-ear implants, and partly-implanted middle ear implants. These studies must be carried out for well-defined groups of patients categorized according to their degree of sensorineural loss, degree of conductive loss, unilateral versus bilateral hearing disorder, and unilateral versus bilateral fitting. When more comfortable and effective non-implanted bone-conductor hearing aids become available, these should be added to the mix. One might expect that for patients with bilateral conductive or mixed loss, bilateral middle-ear implants will provide more benefit than bilateral bone conduction hearing aids of any type, as the two cochlea will be independently stimulated by the signals from their respective sides of the head, giving stronger binaural cues.

The outcome variables measured must include performance measures concerning speech intelligibility in spatially distributed noise, localization ability, sound quality at different input levels, acceptability of internal noise, effect on external ear infections or irritation, and subjective factors related to comfort, convenience, and ease of manipulation. This will be no easy task, as it is usually a requirement of surgical implantation that a non-implanted hearing aid first be trialed and found to be unsuitable on some grounds, which makes it difficult for the patient and experimenter to avoid experimental bias, however unintended.[j]

Until careful studies by disinterested experimenters are conducted, it will be difficult to know which device is preferable for which patients, although implanted hearing aids of some type clearly have a very important role for some types of hearing loss. They are likely to be particularly beneficial for patients with losses that include a significant conductive component that cannot be fixed by surgery with passive prostheses, for patients unable to wear any component in their ear canal, and for patients who put the highest premium on having no externally-worn prosthesis.

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[j] This requirement for lack of success with a conventional aid could perhaps be re-considered in the context of a randomised controlled trial, particularly for implanted devices that have been shown to have no effect on passive hearing ability with the device turned off. Surgery to modify the middle ear with a passive middle-ear prostheses is routinely carried out without first requiring that a conventional hearing aid be tried and found wanting.

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[References in square brackets are not in English]

INDEX AND GLOSSARY OF ACRONYMS

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