Evaluation of a Dual-Channel Full Dynamic Range Compression System for People with Sensorineural Hearing Loss

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ABSTRACT

This article describes an evaluation of an in the ear hearing aid, which applies fast-acting full dynamic range compression independently in two frequency bands. This can compensate for the loudness recruitment typically associated with sensorineural hearing loss. The crossover frequency between the two bands and the gain and compression ratio in each band are programmable to suit the individual patient. Twenty subjects with moderate sensorineural hearing loss were tested in a counterbalanced order using the aid programmed as a linear amplifier (condition L) and as a two-band compressor (condition C). All subjects were fitted binaurally. Subjects were also tested without hearing aids (condition U) and using the hearing aids that they normally wore (condition Own). Speech intelligibility was measured in quiet at three sound levels (50, 65, and 80 dB SPL), and speech reception thresholds (SRTs) in 12-talker babble were measured under monaurally and binaurally aided conditions, with the speech and babble both coincident and spatially separated. In condition C, speech intelligibility in quiet was high at all sound levels. Speech intelligibility at the two lower levels decreased in condition L, and decreased still further in conditions Own and U. Condition C gave, on average, better speech intelligibility in babble (lower SRTs) than conditions L, Own, or U. The advantage of condition C over condition L varied across subjects and was correlated with the dynamic range for tones at high frequencies; small dynamic ranges were associated with greater benefit from compression. A significant advantage for binaural aiding was found both when the speech and noise were spatially separated and when they were coincident. The binaural advantage was similar for the C and L conditions, indicating that the independent compression at the two ears did not adversely affect the use of binaural cues. Questionnaires on the subjects' experiences with the aids in everyday life indicated that they generally preferred condition C over condition L. (Ear Hear 13 5:349–370)

MOST PEOPLE WITH A sensorineural hearing loss have a smaller than normal dynamic range between the threshold for detecting sounds and the level at which sounds become uncomfortably loud. This is often described in terms of loudness recruitment; the rate of growth of loudness with increasing sound level is more rapid for an impaired ear than for a normal ear. This creates problems for the hearing-impaired person when listening to speech. In everyday situations, the overall level of speech may vary over a 35 dB range (Pearsons, Bennett, & Fidell, 1976). For the impaired listener, speech may only be both audible and comfortable over a small range of sound levels.

The most common way that this problem is dealt with by a hearing aid user is to adjust the volume control of the hearing aid as the listening situation changes. However, this can be difficult for elderly aid users, whose dexterity is often limited, and, for in the ear aids, the presence of a finger near the aid can induce acoustic feedback. Even for dextrous aid users with behind the ear aids, the need to adjust the volume control frequently can be irksome. It is desirable, therefore, to have a means of automatically adjusting the gain of an aid so that speech is always delivered within the range of levels where sounds are both clearly audible and comfortable, regardless of the input level.

Even when speech is presented at a constant average level, individual acoustic elements within the speech may vary over a range of at least 40 dB (Levitt, 1982). This range may exceed the dynamic range available to the listener, especially at high frequencies, where the range of sound levels over which sounds are both clearly heard and comfortably loud may be as small as 10 dB or even less. One way of dealing with this problem is to use fast-acting automatic gain control (AGC); with this, the gain changes rapidly from one speech sound to the next, so that weak sounds can be amplified more than intense sounds. This form of AGC is often called syllabic compression. The word compression is used

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because the dynamic range of speech is compressed into a smaller range of values.

Several authors (Laurence, Moore, & Glasberg, 1983; Villchur, 1973) have proposed that syllabic compression should be applied separately in two or more frequency bands. There are at least two reasons why this might be beneficial. First, the amount of hearing loss, and usually the amount of recruitment, often varies markedly with frequency; typically, both hearing loss and recruitment are greater at high than at low frequencies. Hence, the amount of compression needed varies with frequency, and this requires compression applied independently in more than one band. A second reason is that relatively weak high-frequency components in speech, which can be important for intelligibility, are often accompanied by, or follow rapidly after, relatively intense low-frequency components. The use of fast-acting AGC in two or more separate bands can ensure that these weak high-frequency components are always audible.

Research on the benefits of multiband compression has given somewhat conflicting results. Villchur (1973) reported distinct benefits for a two-band compression system in comparison to a linear amplification system, but his study was criticized (Lippmann, Braida, & Durlach, 1981) on the grounds that the linear system did not have sufficient high-frequency emphasis. Villchur's (1982) counter argument to this is that he could not use more high-frequency emphasis in the linear system because at high sound levels, this would have led to an unacceptable sound quality. In everyday life, impaired subjects do sometimes have to listen to sounds with high levels, and the maximum permissible high-frequency emphasis in linear amplification systems is limited by tolerance of such sounds.

Lippman et al (1981) evaluated a system incorporating 16-band syllabic compression. In several of their test conditions, they found no advantage of this system over one using linear amplification combined with frequency shaping. However, these conditions used speech material covering only a very limited dynamic range, which is not typical of everyday life. When they used speech material with a wider dynamic range, advantages were found for the compression system.

Adequate compensation for recruitment can probably be achieved with a small number of bands, say, two or three. Indeed, there may be a disadvantage in using a large number of bands, because this tends to reduce spectral contrasts in complex stimuli. In a multiband compression system with many bands, the differences in level between peaks and dips in the spectrum of the input are reduced. If the impaired ear has reduced frequency selectivity, as is often the case (Tyler, 1986), then the reduction in spectral contrast compounds the difficulties experienced in picking out the spectral features of complex sounds (Leek, Dorman, & Summerfield, 1987). To avoid this problem, the bands in a compression system should have bandwidths which are broad in comparison to the auditory filters of the impaired listener (Glasberg & Moore, 1986). In practice, this means that only a few bands should be used, probably no more than three.

Laurence et al (1983) described a hearing aid which incorporated two forms of AGC. Initially, a slow-acting AGC operated on the whole signal to compensate for variations in overall sound level from one situation to another. Then, the signal was split into two frequency bands, with fast-acting AGC in each band. The aid was shown to have significant advantages for the understanding of speech both in quiet and in background noise in comparison to a similar aid without AGC. The aid was also shown to give better intelligibility of speech in noise than a similar single-channel compression aid and unaided listening (Moore, 1987; Moore & Glasberg, 1986; Moore, Laurence, & Wright, 1985). It should be noted that these studies are among the few using wearable aids that the subjects could use in their everyday lives. This is important because if a person has had a hearing impairment for many years, it may take some time for them to learn to use the new cues provided by a compression system. More recently, Moore and Glasberg (1988), in a comparison of several forms of AGC, have shown that when the overall level of speech is maintained at a comfortable level by a slow-acting AGC system operating on the whole signal, the intelligibility of speech in noise can be improved using a two-band system with fast-acting compression in the high-frequency band only.

Although some studies have shown clear benefits of compression using a small number of bands, other studies have not shown such benefits, and there is still controversy over the interpretation of the results (for a review see Braida, Durlach, De Gennaro, Peterson, & Bustamante, 1982). One problem in evaluating and comparing different AGC systems is that these systems can differ in many ways. Among the possible parameters are: (1) the number of bands and the cutoff frequencies between bands; (2) the time constants of the compression circuits and whether these are fixed for each band or vary across bands; (3) the input-output function for each band; and (4) the overall frequency range covered and the smoothness of the overall system frequency response. In a review article, Plomp (1988) has argued that fast-acting compression in multichannel systems has deleterious effects, and that slow-acting AGC (long attack and recovery times) should be more effective. However, his arguments apply mainly to systems using a large number of bands and large amounts of compression. Villchur (1989) has presented counter arguments to those of Plomp (1988), and has pointed out that in practice the amount of compression used is generally considerably less than that assumed by Plomp in his arguments. Further discussion of the possible benefits of compression in hearing aids may be found in Moore (1990).

In this article, we present an evaluation of a two-channel system using full dynamic range fast-acting compression in each band. The system has been implemented as a wearable in the ear aid, the ReSound Corporation Personal Hearing System, model ED2. This means that subjects were able to wear their aids for some time before testing took place. It also allowed us to evaluate the effectiveness of the aids both in the laboratory and in everyday life by means of questionnaires.

DESCRIPTION OF THE AID

A block diagram of the ED2 aid is shown in Figure 1. The microphone signal is fed to a preamplifier and an AGC system. The AGC is of the compression limiting type; the compression threshold is 85 dB SPL, the attack time is 1 msec and the release time is 40 msec. The input-output function is illustrated in the bottom left panel of the figure. This AGC system is intended to provide protection from brief intense sounds, such as a door slamming, and to prevent circuit overload at high input sound levels. The signal is then split into high- and low-frequency bands using a high-pass filter with a slope of 18 dB/octave and a low-pass filter with a slope of 12 dB/octave (see panel at bottom). The crossover frequency between the two bands can be adjusted in 1/2-octave steps over the range 400 to 4700 Hz. Each band includes a compression circuit with a compression threshold of 45 dB SPL and with adjustable gain and compression ratio (see panel at bottom). In practice, the operation of each circuit is determined by specifying the gain for an input level of 50 dB (range 2–38 dB in the low band, specified at 500 Hz, and 2–44 dB in the high band, specified at 2 kHz) and the gain for an input level of 80 dB SPL (range 2–22 dB in the low band and 2–28 dB in the high band). The compressor circuits in the bands have attack times of 1 msec and release times of 50 to 100 msec, depending

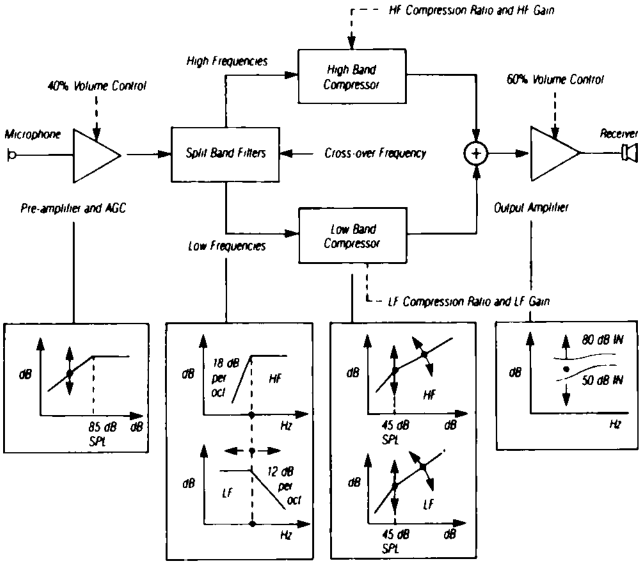


Figure 1. The top half of the figure shows a block diagram of the ED2, and the panels at the bottom illustrate schematically the operation of the different blocks.

on the program. The bands are then combined and fed to an output amplifier and receiver.

All adjustable parameters are remotely programmable. During fitting of the aid, it is programmed by the ReSound Digital Hearing System (DHS); more details are given below. Once the final program values have been determined, they are transferred into a pen-sized remote control unit which is used to turn on the aid and reload the program after a period when the aid has been turned off. The remote control can store two different programs (four programs with binaural aids). However, for the purpose of this trial, both program memories were loaded with the same program. The remote control can also be used to determine overall "volume"; the action of the volume control is split, so that 40% of the gain change occurs in the microphone preamplifier and 60% occurs in the output amplifier. The gains given above are specified at a "reference" volume which is set automatically when the aid is turned on using the remote control. The aids were fitted to the subjects with the volume at the reference setting and all tests were conducted with the volume at the reference setting. In addition, subjects were encouraged to use the aids at the reference setting as much as possible. With the volume set to maximum, gains are up to 8 dB higher than those specified above.

EXPERIMENTAL PLAN

The Choice of Comparison Conditions

One of the many difficulties in evaluating multiband compression systems is the choice of appropriate control or comparison conditions. Fortunately, the ED2 can be programmed as a linear amplifier (apart from the limiting compression operating at sound levels above 85 dB SPL), thus providing a control condition in which the aid was as similar as possible except for the presence or absence of two-channel compression. In a two-channel compression aid, the effective frequency response and gain vary with input level (as illustrated schematically in the lower right panel of Fig. 1). In a linear aid, the frequency response and gain are invariant with level (up to the saturation level or, in the case of the ED2, up to the threshold of the input AGC). To make the two aids as comparable as possible, the frequency response and gain have to be matched, but this can be done at only one level. We decided to match the linear aid (condition L) to the compression aid (condition C) by equating the gain and frequency response for an input level of 65 dB SPL. In practice, this was done by programming the aid so that the cutoff frequency between the two bands was the same for the C and L conditions, and the gain in each band in the L condition was the same as the mean gain at 50 and 80 dB SPL in the C condition. When programmed in this way, the gain and frequency response, as determined by real ear measurement with a 65 dB SPL psuedorandom noise as a test signal (see later for details), were very similar for the C and L conditions.

The level chosen for equating gains is toward the low end of the range of levels of speech encountered in everyday life, but is typical of normal conversational levels in a quiet room (Pearsons et al, 1976). However, because the subjects were required to use the aids in everyday life, it was necessary to check that speech and other environmental sounds were not unpleasant or uncomfortably loud with this gain/frequency response setting. After the linear aids had been programmed, a check was made using speech at 80 dB SPL (RMS) to make sure that the speech was not unpleasantly loud or of unpleasant quality. If this was the case, the gain in one or both of the bands was reduced by the minimum amount necessary to ensure that the speech at 80 dB SPL was acceptable; the criterion for acceptable was that the subject should be willing to listen to speech at this level for a period of at least 5 min and that the sound quality should not be judged excessively shrill. Thus, the gain and degree of high-frequency emphasis in the linear aid were chosen to be as close as possible to those used in the compression aid, subject to the proviso that sounds of moderately high level should be acceptable in quality and loudness. We consider this to be the fairest and most realistic way of comparing compression with linear amplification. Generally, the gain in condition C was less than that in condition L at high input sound levels and greater than that in condition L at low sound levels.

Wherever possible, the aids were fitted with vents. The vents were small and had little effect on the frequency responses of the aids. However, venting can lead to problems with acoustic feedback. It turned out that feedback was more of a problem in condition C than in condition L because the gain in condition C was greater at low sound levels; for a fixed vent size, the aids would sometimes oscillate in condition C but not in the condition L. Therefore, in condition C, the bore of the vent often had to be reduced relative to that used in condition L. Sometimes the vent had to be blocked completely in condition C. Note that this was done in order to achieve the matching of the C and L conditions described above.

Performance in the C and L conditions was also compared with that obtained in unaided listening. The rationale for this condition is provided by the observation that in comparison to unaided listening, most linear hearing aids do not improve the intelligibility of speech in noise at moderate to high noise levels (Duquesnoy & Plomp, 1983; Festen & Plomp, 1986; McAlister, 1990), although exceptions do occur (Moore et al, 1985). Comparison with unaided listening thus provides a useful baseline against which to evaluate the benefits of a hearing aid when listening in background noise.

Finally, the subjects who normally wore hearing aids (the majority) were tested using their own aids. It can be somewhat difficult to interpret results obtained in this way because many different types of aids were worn, and there is no guarantee that the aids were of the appropriate type or had been appropriately fitted. However, all of the subjects in this trial had been fitted by certified audiologists and stated that they were reasonably well satisfied with their own aids; most wore aids that were relatively new.

In summary, the following conditions were used: (1) listening using the ED2 programmed as a two-channel compression system (condition C); (2) listening using the ED2 programmed as a linear amplifier (condition L); (3) listening unaided in sound field (condition U); and (4) listening using the subjects' own aid(s) (condition Own).

For conditions C and L and Own, real ear measurements were conducted using a probe microphone in the ear canal, so as to check the gains achieved as a function of input frequency and level.

Performance in the different conditions was evaluated by measurements of the intelligibility of speech in quiet and in background noise (babble) and by asking the subjects to fill in questionnaires about their experiences in everyday life. All subjects were fitted binaurally.

Evaluation of the Benefits of Binaural Listening

When a person has a hearing loss in both ears, the use of binaural hearing aids can provide a number of benefits. Perhaps the most important and uncontroversial is that binaural aids can help to overcome problems associated with head-shadow effects. Consider as an example a situation where a person wears an aid only in the left ear. If that person is trying to listen to a talker on the right of the head, there may be considerable difficulty, especially if there is competing noise coming from the left. An aid in the right ear can overcome this problem. Essentially, it appears that subjects can attend to whichever ear gives the better speech to noise ratio. This benefit is likely to occur for almost any reasonable system of binaural aiding, and it was not specifically evaluated here.

A second potential benefit of binaural aiding is that listening with two ears can give better performance than listening with one, even when the signals at the two ears are highly correlated (Davis, Haggard, & Bell, 1990; Laurence et al, 1983). This aspect of benefit was evaluated in the present experiment by measuring speech reception thresholds (SRTs) in noise, with speech and noise coming from the same loudspeaker directly in front of the subject. Performance was compared for three conditions: binaurally aided, with the left ear only aided, and with the right ear only aided. In the last two cases, the aid in the nontest ear was turned off. It should be acknowledged that this is not representative of everyday listening, where an unaided ear would not normally be blocked. Hence, the results should be interpreted as indicating the benefit of binaural versus monaural listening, rather than binaural versus monaural aiding. However, separate tests on a small number of subjects with moderate to severe hearing losses have indicated that under the conditions used in these experiments, SRTs in noise when one ear is aided are hardly affected by blocking the unaided ear.

A third potential benefit of binaural amplification derives from the fact that subjects can sometimes use differences in the signal reaching the two ears to enhance signal detection or speech recognition. In particular, if the interaural amplitude or time of the primary signal differs from that of the interfering sound(s), performance may be improved relative to the case where the interaural amplitude and time of the signal and interference are identical. The most common practical situation where this occurs is when the person is directly facing a talker, giving essentially the same amplitude and time of the speech at the two ears, while the interference comes from one or more other directions. Binaural aids may enhance this effect (compared to monaural aids) by increasing the audibility of the stimuli at the two ears (i.e., ensuring that the stimuli in both ears are sufficiently above absolute threshold) and by giving more "balanced" stimuli than would be the case with a monaural aid. However, binaural aids may also reduce the effect, especially in comparison to unaided listening, by introducing spurious amplitude and/or phase differences of the speech at the two ears. This is potentially a serious problem for aids with AGC, because the independent AGC action at the two ears may introduce large interaural intensity differences for a speech signal coming from straight ahead.

We have tried to evaluate the importance of these spurious interaural intensity differences by setting up a situation which represents a worst case. The speech signal was presented via a loudspeaker directly in front of the observer. The interfering sound was 12-talker babble that was presented via two loudspeakers at +90° and −90° azimuth (i.e., on the interaural axis). The babble was independent at the two loudspeakers. Again, performance was compared for three conditions: binaurally aided, with the left ear only aided, and with the right ear only aided. If the AGC action was significantly influenced by the babble, then the fact that the babble reaching the two ears was partially independent would lead to a decorrelation of the speech signal at the two ears. This might be expected to produce a degradation in performance for condition C as compared to conditions L and U under conditions involving binaural aiding. Hence, the comparison of condition C with conditions L and U allows an assessment of the importance of this effect on speech intelligibility in babble.

Subject Selection

Subjects were selected to have the following characteristics: (1) A pure sensorineural loss (no conductive component) with clear signs of loudness recruitment, as determined by a measure of effective dynamic range (see below for details). (2) A hearing loss, averaged over the frequencies 1, 2, and 4 kHz, not less than 40 dB HL and not greater than 75 dB HL. (3) Subjects with normal hearing at low frequencies (hearing loss less than 20 dB) and a large loss at high frequencies were excluded. This was done because we expected that such subjects would rely primarily on their normal low-frequency hearing, and would show little difference in performance between the different conditions. (4) The difference in hearing level between the two ears, averaged over the frequencies 1, 2, and 4 kHz, was not greater than 20 dB. (5) Subjects were alert and able to understand and follow instructions. (6) Subjects were reasonably satisfied with their own hearing aids. Within these limitations, subjects were selected with a range of configurations of hearing loss (e.g., flat, shallow slope, steep slope) and a range of ages. Table 1 gives the absolute thresholds, ages of the subjects, and other relevant information. Initial audiometric evaluation was carried out using a Grason-Stadler GSI 16 audiometer with TDH 50P earphones.

Subjects were told that they would need to attend for a series of tests and that they would be reimbursed for travel expenses and paid a fee per session for taking part in the tests.

Experimental Design

Testing of the C and L conditions was done in a counterbalanced order; half the subjects were tested first in the C condition and half in the L condition. Evaluation of the first type of aid was done 1 and 2 weeks after fitting. After the 2-week period of experience with the first type of aid fitted, subjects were switched to the other aid type (i.e., C to L or L to C). Again, testing was done after 1 and 2 weeks. Testing in the U condition was done once at the start of the trial and once at the end. This was done to allow us to assess whether there were any long-term practice effects in identifying the test sentences. Testing in the Own condition was done after testing in the U condition. Each subject attended a series of test sessions whose structure was as follows.

Session 1

(1) Absolute thresholds were measured at 0.25, 0.5, 1, 2, 3, 4, 6, and 8 kHz. Highest comfortable levels (HCLs) were measured at 0.5, 1, 2, and 4 kHz. Instructions for establishing the HCL were as follows: "You will hear a tone which gets louder in small steps. I want you to indicate when the sound reaches the greatest loudness that you would be happy to listen to for a long time. The sound should not be uncomfortably loud. It should be at the upper end of the loudness range which is comfortable for you. The test may be repeated a number of times for any particular tone, and will be repeated for a number of different tones. In each case, indicate when the tone reaches the highest comfortable level for you. Do not wait until the sound becomes uncomfortably loud."

In previous work (Moore, 1987), we have found the HCL to give more consistent results than the more commonly used uncomfortable loudness level (UCL). Also, the HCL seems more relevant than the UCL for establishing the effective dynamic range available to the subject. A small range between the threshold and the

Table 1. Summary of the characteristics of the subjects used. The table shows the age of each subject, the absolute threshold in dB HL at eight frequencies, and the HCL in dB HL at four frequencies. For each ear of each subject, the upper of the two figures is the absolute threshold and the lower is the HCL. The table also shows the aid(s) normally worn by each subject (where applicable). ITE, in the ear; ITC, in the canal; BTE, behind the ear.

Subject	Age	Ear	Frequency (in kHz) 0.25	0.5	1.0	2.0	3.0	4.0	6.0	8.0	Own aid
1	66	R	30	20	40	50	40	25	20	35	Starkey Intra
		R		60	70	85		65			ITC
		L	20	25	45	60	45	25	35	70	Starkey Intra
		L		70	75	80		60			ITC
2	44	R	35	45	55	70	80	90	115	105	Argosy HS*
		R		80	95	105		120			ITE
		L	35	45	55	65	55	60	60	70	Argosy HS*
		L		80	85	85		85			ITE
3	75	R	40	45	40	45	35	35	55	70	
		R		65	75	75		75			
		L	40	40	35	45	45	45	50	75	Vanco CD*
		L		70	65	75		80			ITE
4	44	R	50	45	60	60	70	75	85	80	Telex 353C*
		R		95	95	90		105			BTE
		L	50	55	60	70	70	70	85	100	Telex 353C*
		L		95	95	100		105			BTE
5	79	R	35	35	40	40	55	55	75	90	
		R		70	85	80		100			
		L	50	40	40	50	55	60	70	80	Siemens 007*
		L		85	85	85		90			ITE
6	68	R	45	35	55	60	60	75	90	95	Phonic Ear 810*
		R		65	70	80		95			BTE
		L	40	30	50	55	60	65	95	90	
		L		75	75	80		105			
7	36	R	25	30	55	55	45	50	65	65	
		R		90	95	95		90			
		L	35	40	55	55	55	65	70	65	Dahlberg PCH
		L		90	95	95		100			ITC
8	23	R	35	40	55	45	40	40	60	60	Starkey Intra
		R		75	80	70		70			ITC
		L	30	30	35	45	45	40	55	70	
		L		70	65	70		70			
9	56	R	50	45	45	50	65	75	60	65	
		R		75	75	75		90			
		L	60	50	55	60	55	60	60	65	Starkey Intra II
		L		75	75	75		75			BTE
10	70	R	30	30	40	50	50	65	60	60	Starkey CE
		R		65	70	70		80			ITE
		L	30	20	25	40	55	60	65	65	
		L		55	65	70		80			
11	69	R	50	65	50	50	45	45	55	70	Nu Ear SS1*
		R		80	75	80		75			ITE
		L	45	60	55	60	50	55	60	65	
		L		85	80	90		80			
12	72	R	45	40	40	55	70	75	70	75	Oticon E19V
		R		85	85	80		90			ITE
		L	45	40	45	60	75	75	75	85	Zenith ZP9178*
		L		90	85	85		90			ITE
13	84	R	45	40	45	50	55	60	65	70	Siemens 268*
		R		65	75	75		80			BTE
		L	50	40	45	55	50	65	70	75	Siemens 007*
		L		75	75	80		85			ITE
14	61	R	45	20	65	60	50	50	55	80	Siemens 007*
		R		95	105	110		115			ITE
		L	50	55	65	60	55	60	50	65	Siemens 007*
		L		95	105	110		115			ITE
15	72	R	30	20	55	70	85	95	90	90	Oticon E27F
		R		95	100	105		120			BTE
		L	30	30	55	65	75	85	80	75	
		L		80	90	95		110			
16	78	R	20	35	50	55	65	60	50	55	Oticon I-22
		R		65	75	75		80			ITE
		L	40	40	50	55	65	65	55	70	Starkey CE
		L		70	75	80		80			ITE
17	76	R	40	45	45	45	55	65	80	70	
		R		75	75	80		85			
		L	55	55	45	45	55	55	75	85	
		L		75	70	80		75			
18	62	R	65	55	55	55	50	50	45	50	Oticon I-22
		R		85	90	95		100			ITE
		L	75	65	55	55	50	55	55	55	Oticon I-22
		L		95	95	100		100			ITE

Table 1. *Continued*

Subject	Age	Ear	0.25	0.5	1.0	2.0	3.0	4.0	6.0	8.0	Own aid
			Frequency (in kHz)								
19	82	R	35	35	25	45	80	90	75	75	
		R		85	60	75		105			
		L	40	45	30	55	80	95	100	110	Widex F8
		L		65	60	75		115			BTE
20	68	R	25	20	30	45	45	60	65	70	
		R		60	70	70		80			
		L	30	35	50	60	55	60	70	70	Oticon I-22
		L		65	60	70		75			ITE

[a] *Aids incorporating compression.*

HCL gives an initial indication of the presence of loudness recruitment.

(2) Ear impressions were taken.

(3) Subjects were tested in the U condition. This session lasted for a total time of about 70 min.

Session 2 For session 2, the hearing aids had been made, and the aids were ready for fitting to the subject. The initial procedures were aimed at determining an appropriate fitting for the C condition.

(1) A test was applied measuring the growth of loudness for 1/2-octave wide bands of noise centered at 500, 1000, 2000, and 4000 Hz (the Loudness Growth in Octave Bands or LGOB test) using the ReSound DHS with the DHS-CD1 compact disc. This test has been described in detail elsewhere (Pluvinage, 1989) and only a brief outline will be given here. It is a modification of a test developed by Allen and Jeng (1990). In an initial phase of the test, each center frequency is tested separately. On each trial, the noise band is presented as a series of three bursts, and the subject is required to indicate the loudness of the bursts by pressing one of seven buttons labeled "cannot hear," "very soft," "soft," "comfortable," "loud," "very loud," and "too loud." The sound level is varied randomly from trial to trial over the range from 30 to 110 dB SPL. In the second phase, all stimuli consistently eliciting responses of "can't hear" or "too loud" are eliminated, and a series of trials is presented in which both the center frequency and level of the noise bands are randomized. The step size in level is determined on the basis of the subject's dynamic range, varying from 8 dB for a normal dynamic range to 4 dB for a narrow one (<50 dB). Each frequency and level combination is presented at least twice (not successively). If different loudness judgments are given, the stimulus is repeated again later until the same judgment is given twice.

The results of the second phase are used to construct functions relating perceived loudness to level at each center frequency. These curves give a quantitative measure of loudness recruitment; the steeper the slope, the greater the recruitment. The test was applied separately for each ear. Some examples of growth of loudness curves are given in Figure 2. The four panels at the top show results for a subject with marked loudness recruitment; all of the curves have steep slopes, and at high levels, the loudness matches that which would occur in a normal ear. The bottom four panels show results for a subject with little or no loudness recruitment. Most of the subjects had results between these two extremes, a typical example being shown in the middle four panels. Usually, loudness recruitment was more marked at 2 and 4 kHz than at 0.5 and 1 kHz.

(2) The results of the LGOB test combined with the audiogram measures were used to determine appropriate gain values for programming the aid in the C condition for each ear. This was done automatically using the LGOB plus audiogram algorithm in the ReSound DHS. This algorithm works in the following way. The crossover frequency between the two bands is calculated from a weighted average of the four frequencies, 0.5, 1, 2, and 4 kHz, the weighting being the respective hearing loss at each frequency. The nearest standard crossover frequency is then located and used as the crossover value. The gains and compression ratios in the two bands are calculated from the slopes and intercepts of linear regression lines fitted to the loudness growth curves at each of the four frequencies (see Fig. 2). Compression ratios in the low and high bands are calculated as linear combinations of the four slope values weighted by constant multipliers. A reference loudness level is calculated from the "very soft" loudness level values at each frequency (shown as ratings of "1" in Fig. 2). The required gain at low levels (50 dB SPL input) is obtained as the difference between the calculated reference level and a predetermined normal reference level for that frequency. Insertion loss corrections are applied to each of the four gain values, and the 50 dB input gains are calculated as weighted averages of the gains at the four frequencies. The 80 dB input gains are calculated from the 50 dB input gains and the compression ratios in each band. If necessary, corrections are made when the gain values fall outside the possible range. Gains are reduced slightly when a binaural fitting is made.

Once the appropriate gain values had been determined, one of the aids was placed in the subject's ear. The gain values were then programmed into the aid from the DHS, using the ultrasonic link.

Although this procedure works quite well on average, individual subjects sometimes require slightly more or less gain. This may happen because the preferred loudness of speech may be different from the preferred loudness of bands of noise. Hence, a stage of "fine tuning" was also used, as described below. Although

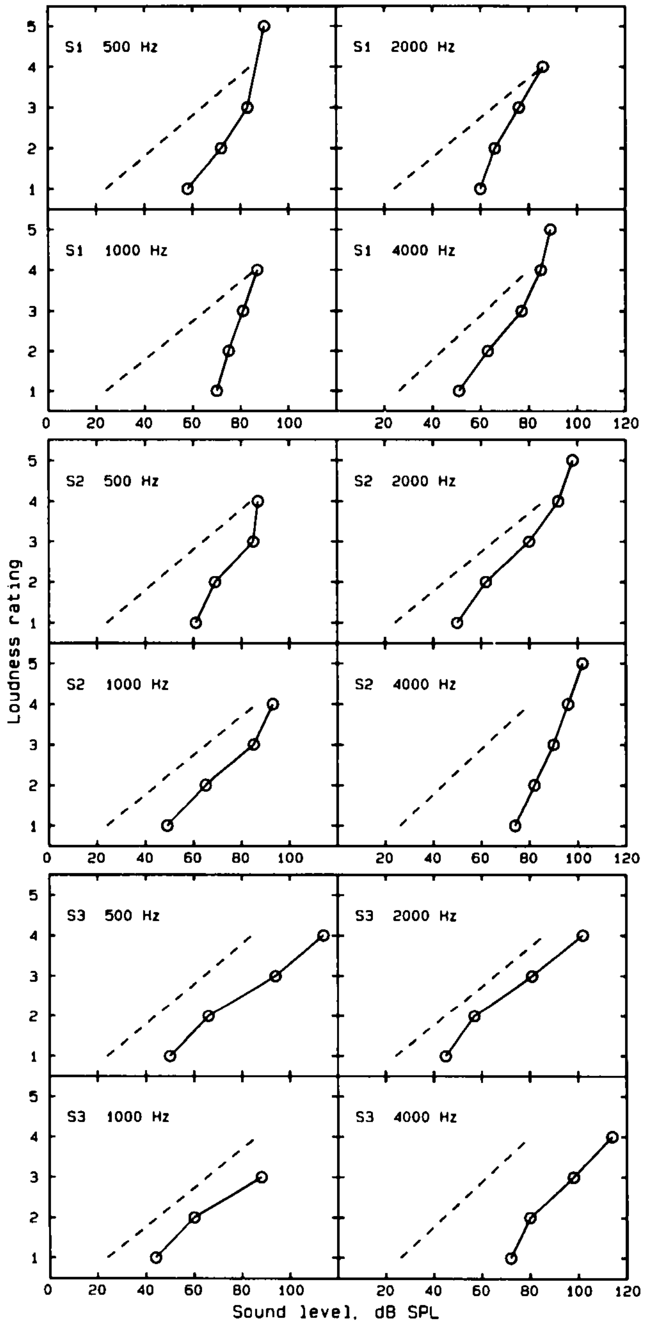


Figure 2. Examples of the results of the LGOB test. In each panel, typical results for normal subjects are shown by the dashed line and results for the subject under test are shown by the solid line and symbols. The top four panels show results at four test frequencies for a subject with marked loudness recruitment, and the bottom four panels show results for a subject with little or no recruitment, as measured by this test. The middle four panels show results for a subject with an intermediate amount of loudness recruitment.

the adjustments made at this stage were generally small, some subjects were quite definite that further adjustment was needed.

(3) Further adjustments to the high- and low-level gains and crossover frequency were made using high-level and low-level speech as stimuli. The aim was to make speech sound clear and comfortable over a wide range of sound levels from 50 to 80 dB SPL RMS. If speech at 80 dB SPL sounded too loud, the high-level gains in both bands were reduced or all gains were reduced. If it sounded only moderately loud, the high-level gains in both bands were increased slightly. If the speech sounded too "shrill" or "tinny," the high-level gain in the high band was reduced. If the speech sounded too "boomy," the high-level gain in the low band was reduced. If speech at 50 dB SPL could not be heard clearly, the low-level gains were increased. If weak consonants like "k" and "p" could not be heard, the low-level gain in the high band was increased. In general, the changes in gain from the values recommended by the fitting algorithm were small, usually less than 6 dB and averaging about 3 dB. Finally, a check was made that speech presented in speech babble at a 10 dB speech to babble ratio was of reasonable intelligibility. This procedure was then repeated using the second aid fitted to the other ear (with the aid for the first ear being turned off). Finally, the aids were fitted to both ears, and checks were made that the balance between the two ears was reasonable and the overall loudness was reasonable. If necessary, small adjustments were made. Test signals were produced by the ReSound DHS using the DHS-CD1 compact disc, and were presented using Realistic Minimus 3.5 loudspeakers placed on the desk in front of the subject at a distance of approximately 1 m.

(4) To obtain appropriate settings for the L condition, the device for each ear was initially set to have the same frequency-gain characteristic as for the C condition with an input level of 65 dB SPL. Speech was then presented at a level of 80 dB SPL RMS. If the subject reported that this speech was too loud or too "shrill," the gain in one or both of the bands was reduced by the minimum amount necessary to make the speech acceptable in loudness and quality.

(5) The aid was programmed with the settings appropriate for the starting condition for that subject (C or L).

This session took a total time of about 75 min.

Session 3 (1 week after session 2) The subject was tested in either the C condition or the L condition. The measures obtained are described below. To check that the gains actually achieved corresponded to those programmed in the ED2, real ear measurements were taken of the frequency response and gain of each aid for input levels of 50, 65, and 80 dB SPL. These measurements were made with a Fonix 6500 hearing aid test system using a pseudo-random noise as the test signal. Table 2 shows the mean and SD (across subjects) of the difference between the target gain and the actual gain in each band for both the linear and the compression conditions. Gains for the low-frequency band are specified at 0.6 kHz and gains for the high-frequency band are averaged over the range of 2 to 3 kHz. Overall, the

Table 2. Means and SDs (in parentheses) of the differences between the target (programmed) gains and the gains actually achieved. All values are based on 40 samples (2 ears of 20 subjects).

Condition	Input Level (dB SPL)	Low-Frequency Band	High-Frequency Band
Linear	50–80	−0.6 (3.5)	−1.7 (4.0)
Compression	50	−0.7 (3.7)	−1.5 (2.8)
Compression	65	0.0 (3.7)	−1.7 (3.0)
Compression	80	0.7 (3.7)	−1.9 (3.2)

differences between the target and actual gains are small, although there is a trend for the actual gains in the high-frequency band to be slightly lower than the target gains. This session lasted about 65 min.

Session 4 (1 week after session 3)

(1) The subject was tested again in the same condition as used in session 3.

(2) The subject was asked to fill in a questionnaire about experiences with the aids worn for the last 2 weeks. Details of the questionnaire are given later.

(3) The subject was fitted with the other type of aid (L if C previously, C if L previously).

This session lasted about 85 minutes.

Session 5 (1 week after session 4) The subject was tested in either the C condition or the L condition depending on which condition was fitted in session 4. Real ear measurements were conducted, measuring the frequency response and gain of each aid for input levels ranging from 50 to 80 dB SPL. This session lasted about 65 min.

Session 6 (1 week after session 5)

(1) The subject was tested again in the same condition as for session 5.

(2) The subject was asked to fill in a questionnaire about experiences with the aids worn for the last 2 weeks. This session lasted about 55 min.

At the end of this session, subjects were asked to spend the next week wearing their own hearing aids, if they normally wore them.

Session 7 (about 1 week after session 6)

(1) The subject was tested in the U condition.

(2) The subject was tested in the Own condition. For this condition, the gain of the aids was set in the following way. For listening to speech in quiet, part of a list was presented at a level of 80 dB and the subject was asked to adjust the volume control(s) until the speech sounded loud, but not uncomfortably so. The volume control(s) were left at this setting for all tests in quiet. For determining the SRTs in noise, the subject was presented with a sample of speech in noise at a +5 dB speech to noise ratio (noise at 65 dB SPL, speech at 70 dB SPL) and was asked to adjust the volume control(s) so that the speech sounded as clear as possible. Note that the volume control settings were optimized separately for listening in quiet and listening in noise. This was not the case for the C and L conditions, where the volume was left at the reference setting for all tests. This may have biased the results in favor of the Own condition.

(3) The subject was asked to fill in a questionnaire about experiences with own aid(s) (if normally worn) or with unaided listening (if aids were not normally worn). (4) Real ear measurements with the subjects' own aids were obtained. This session lasted about 70 min.

For a few subjects, acoustic feedback prevented the desired gain values from being reached when the aids were being fitted. This occurred especially in the C condition and was usually associated with an earmold with an imperfect seal to the ear canal. These subjects were initially tested using reduced gains, but were later recalled when new earmolds had been made, and were retested using gains closer to the desired values. Table 3 shows the final values of the programmed gains and cutoff frequencies used for each ear of each subject in the L and C conditions. For an input sound level of 65 dB SPL, the average gain in the low-frequency band was 0.55 dB lower in condition L than in condition C (SD 3.1 dB). The corresponding difference in the high-frequency band was 0.65 dB (SD 3.3 dB). Thus, we were reasonably close to our goal of equating gains in the C and L conditions for an input level of 65 dB SPL.

The real ear measurements provided a further check on gains. Insertion gains at frequencies of 500, 1000, 2000, 2500, 3200, 4000, and 5000 Hz were read off from the plotted curves. The upper panel of Figure 3 shows the mean insertion gain in the C condition for noise input levels of 50, 65, and 80 dB SPL. These curves show how the gain is reduced progressively with increasing input level. The lower panel of Figure 3 compares the mean insertion gains for the C, L, and Own conditions for an input level of 65 dB SPL. At this level, gains in the C and L conditions were not significantly different (using a t-test) at any frequency. Also, insertion gains were not significantly different for the C and Own conditions except at 5 kHz, where the mean gain was about 7 dB less in the Own condition. Thus, the overall frequency response and gain were rather similar for all three aided conditions with an input level of 65 dB SPL. The gain in the L condition was essentially invariant with level over the range of input levels from 50 to 80 dB. The mean gain in the Own condition was almost the same for input levels of 50 and 65 dB, but was reduced by about 5 dB at all frequencies for an input level of 80 dB, because some of the Own aids had compression circuity (see Table 1).

Measures of Speech Intelligibility

In a given test session, a subject was tested in only one condition: C, L, or U (except for the last session, where both U and Own were tested). The speech materials were the BKB sentences (Bench & Bamford, 1979) using the recording made by Cochlear Corporation with an American speaker. The following tests were conducted.

Table 3. Values of the programmed gains and cutoff frequencies (f_c, kHz) used for each ear of each subject in the C and L conditions. For the C condition, gains are specified for the low band with input levels of 50 (C_{lo50}) and 80 dB (C_{lo80}) and for the high band with input levels of 50 (C_{hi50}) and 80 dB (C_{hi80}). For the L condition, gains are independent of input level (up to 85 dB SPL) and are specified for the low band (L_{lo}) and the high band (L_{hi}). All levels are in dB SPL, and all gains are in dB.

Subject	Ear	f_c	C_{lo50}	C_{lo80}	C_{hi50}	C_{hi80}	L_{lo}	L_{hi}
1	R	0.9	12	6	16	4	6	6
	L	0.9	8	2	12	2	4	4
2	R	0.9	30	14	40	32	24	34
	L	0.9	32	20	36	22	26	30
3	R	1.3	12	4	18	6	4	8
	L	1.3	16	10	20	12	10	12
4	R	0.9	20	6	32	20	14	28
	L	0.9	26	10	30	22	18	26
5	R	1.3	6	4	18	2	8	12
	L	0.9	4	2	20	2	8	18
6	R	2.0	16	12	34	14	14	24
	L	2.0	14	10	34	18	12	26
7	R	1.3	16	4	30	12	8	14
	L	1.3	20	8	26	16	12	20
8	R	1.3	8	2	20	2	10	12
	L	1.3	8	2	22	2	4	10
9	R	1.3	16	8	32	12	6	16
	L	1.3	30	20	30	20	20	20
10	R	1.3	4	2	18	4	2	8
	L	1.3	2	2	18	6	2	10
11	R	0.6	14	10	26	14	12	20
	L	0.6	18	10	34	16	16	28
12	R	1.3	22	6	32	14	14	24
	L	1.3	20	6	30	14	12	22
13	R	0.9	10	4	24	16	8	20
	L	0.9	12	8	28	12	10	20
14	R	0.9	32	28	38	30	30	34
	L	0.9	30	26	38	30	28	34
15	R	1.3	18	4	40	26	14	32
	L	1.3	18	4	34	16	16	34
16	R	1.3	12	2	22	2	8	12
	L	1.3	16	2	26	6	10	16
17	R	1.3	12	2	18	2	6	10
	L	1.3	12	2	20	2	8	10
18	R	0.9	20	10	26	18	8	18
	L	0.9	20	12	30	22	6	20
19	R	1.3	12	4	22	4	8	16
	L	1.3	10	6	16	8	8	12
20	R	2.0	6	2	18	2	2	6
	L	2.0	12	2	20	4	4	12

Testing in Quiet One complete sentence list was presented at each of three sound levels, 80, 65, and 50 dB SPL. The score was the number of key words correctly identified; each list contains 50 key words. Testing was done binaurally aided (or unaided in the U condition).

Testing in Noise In these tests, the subject was required to identify sentences presented against a background of continuous babble. This was the 12-talker babble taken from the SPIN test (Bilger, Nuetzel, Rabinowitz, & Rzeckowski, 1984; Kalikow, Stevens, & Elliot, 1977). The noise from each loudspeaker had a level of 65 dB SPL at the point corresponding to the center of the subject's head, and the speech was adjusted in level using an adaptive procedure to determine the level at which 50% of the key words could be understood; this level will be called the SRT. Details of the adaptive procedure are given later. The following conditions were used: (1) Speech and babble coming from the same loudspeaker directly in front of the subject. (a) Listening binaurally aided (Bin). (b) Listening with only the left ear aided (aid in the other ear switched off) (LA). (c) Listening with only the right ear aided (aid in the other ear switched off) (RA). For the U condition, only binaural listening was used. (2) Speech coming from a loudspeaker directly in front of the subject, babble coming from two loudspeakers, one on each side of the subject (i.e., at +90° and −90° azimuth), with independent babble from the two loudspeakers and with the total noise level (measured at the point corresponding to the center of the subject's head) equal to 68 dB SPL. (a) Listening binaurally aided. (b) Listen-

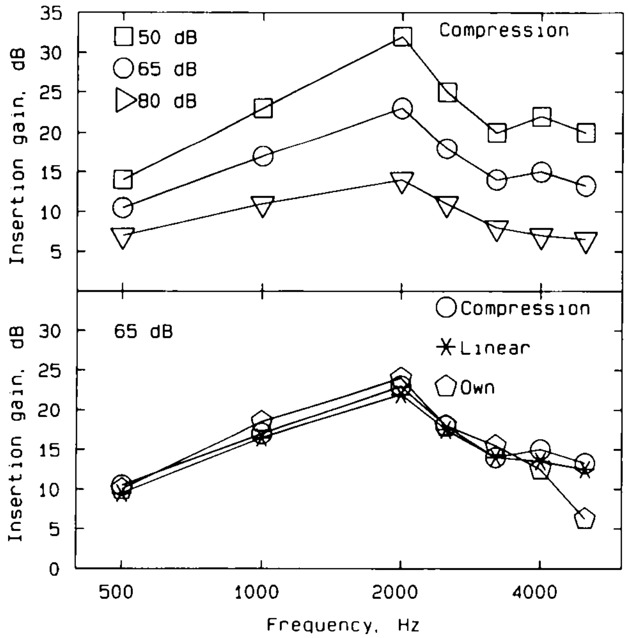


Figure 3. Average results of the real ear measurements obtained with the Fonix 6500 test system using a psuedo-random noise signal. The upper panel shows mean insertion gains in condition C, for input levels of 50 dB SPL (□), 65 dB SPL (O), and 80 dB SPL (∇). The lower panel shows mean insertion gains in conditions C (O), L (*), and Own (pentagons), all for an input level of 65 dB SPL.

ing with only the left ear aided (aid in the other ear switched off). (c) Listening with only the right ear aided (aid in the other ear switched off). For the U condition, only binaural listening was used.

In order to control for possible differences in overall difficulty between lists and to minimize the possibility that subjects would memorize individual lists, the lists were used in the order given in table 4.

This sequence had the following advantages: (1) In the C, L, and U conditions, a list was never repeated until a 2-week period had elapsed, with one complete intervening session. (2) The same lists were used for corresponding tests in the C and L conditions. For example, testing in quiet at 65 dB SPL was always done using lists 9 and 18. Thus, in the event that there were slight differences in difficulty across lists, we would still have a valid comparison of the C and L conditions. The final test using unaided listening was also conducted with lists corresponding to those used in the first test session for the C and L conditions. The lists used in the Own condition were chosen to avoid repeating any lists used in the U condition in the final test session.

The Adaptive Procedure and Method of Estimating SRTs

The adaptive procedure used to estimate SRTs in noise was conducted in the following way. A given test started with the speech level at 80 dB SPL RMS, which was usually sufficiently high to give good intelligibility. Each sentence contained three or four key words. Each time the subject scored two or three key words correct

Table 4. The numbers of the lists used in the different conditions. The following abbreviations are used: U1, unaided (first test, session 1); C1, compression (first test, session 3 or 5); C2, compression (second test, session 4 or 6); L1, linear (first test, session 3 or 5); L2, linear (second test, session 4 or 6); U2, unaided (second test, session 7); Q80, in quiet at 80 dB SPL; Q65, in quiet at 65 dB SPL; Q50, in quiet at 50 dB SPL; NCB, noise and speech coincident, binaurally aided; NCL, noise and speech coincident, left ear aided; NCR, noise and speech coincident, right ear aided; NSB, noise and speech separated, binaurally aided; NSL, noise and speech separated, left ear aided; NSR, noise and speech separated, right ear aided.

Condition	Q80	Q65	Q50	NCB	NCL	NCR	NSB	NSL	NSR
U1	3	4	5	6			7		
C1 and L1	8	9	10	11	12	13	14	15	16
C2 and L2	17	18	19	20	3	4	5	6	7
U2	8	9	10	11			14		
Own	20	3	4	5	12	13	6	15	16

out of three, or three or four out of four, the level of the speech was decreased. Each time the subject scored 1 or 0 key words correct, the level of the speech was increased. If the subject scored two out of four key words correct, the level was left unaltered. The transition from decreasing to increasing level and vice versa defines a turnaround. The speech level was varied in 5 dB steps until two turnarounds had occurred, after which the step size was decreased to 2 dB.

After a run had been completed, a histogram was compiled of the number of key words presented at each level visited during the adaptive procedure (excluding the results for the first sentence presented in each list) and of the number correct at each level. Probit analysis (Finney, 1971) was then used to fit a psychometric function to these scores. The analysis program gave estimates of the slope of the psychometric function, the level corresponding to 50% correct (the SRT), and a value of Chi-squared indicating the goodness of fit. Usually, the fits were very good, and in no case was the value of Chi-squared significant at the 0.05 level. In very few cases, it was not possible to apply probit analysis because there was only one level where performance was between 0 and 100%. In those cases, the SRT was estimated as the mean level at the turnaround points after the change to 2 dB steps.

After the results had been obtained, it became clear that the sentence lists used for determining the SRTs were not all equal in difficulty; the SRTs for a given condition were often consistently different for the two sentence lists used for that condition (as listed in Table 4). To allow for this, a "correction factor" was derived for each sentence list. Note that these correction factors were applied only to the SRTs in noise, and not to the measures of speech intelligibility in quiet. The SRTs for a given sentence list and a given condition were averaged across all subjects and across all aid types (e.g., C and L) used with that condition. If the average was different for the two sentence lists, then a correction factor of half that difference was derived. For example, the average SRT for list 4 in condition NCR was 2.8

Table 5. Correction factors in dB for the SRTs in babble for each list. The correction was added to the SRT.

List number	3	4	5	6	7	11	12	13	14	15	16	20
Correction factor	−1.7	−1.4	−0.8	−1.1	−0.9	−0.1	1.7	1.4	0.8	1.1	0.9	0.1

dB higher than the average SRT for list 13 in that same condition. Hence, the correction factor was −1.4 dB for list 4 and 1.4 dB for list 13. More generally, the correction factors were chosen so that after correction, the mean SRT obtained for a given condition was the same for both of the sentence lists used for that condition. The correction factors are given in Table 5.

Note that these correction factors were derived by comparing each list with only one other list; in each such case, each subject had listened to both lists in all conditions used with that list. Although this procedure does not guarantee complete equivalence across all lists, it seems preferable to using no correction at all. Note that the use of these correction factors has no effect on the difference in scores between the C and L conditions, because the same sentence lists were used for the C and L conditions.

In the remainder of this article, only the corrected SRTs will be reported. However, the general pattern of results would be the same for the uncorrected SRTs. Also, all effects reported as significant in the analyses of variance presented later were also significant when the analyses were based on uncorrected SRTs.

After applying the corrections, the RMS difference between repeats of the same test (e.g., C1 NCB versus C2 NCB) was 1.65 dB (without any corrections, the corresponding figure was 1.93 dB). This means that the SD of the estimates for a given condition is $1.65/\sqrt{2}$ = 1.17 dB. Because there were two estimates of the SRT for each condition, the SEM for each SRT is $1.17/\sqrt{2}$ = 0.83 dB. The corresponding SE with no corrections applied is 0.97 dB.

Equipment and Test Situation

The BKB sentences were supplied on cassette tapes by Cochlear Corporation. In pilot runs, it was found that the interval between sentences was too small to allow subjects to respond and to implement the adaptive procedure. Also, the overall level of the speech varied slightly from one cassette to another. To overcome these problems, the recordings were transferred to digital audio tape (DAT) using a Sony TCD-D10PRO DAT recorder. During the recording process, the overall level of the speech was carefully equalized across lists and an extra time interval was inserted between sentences, giving a resultant interval of about 7 sec. The BKB sentences were recorded on one channel of the DAT and the SPIN babble was recorded on the other channel.

During testing, the speech from the DAT player was routed through a Grason Stadler GSI 16 audiometer to a Tannoy PBM 6.5 loudspeaker in a single-walled sound-attenuating booth. The loudspeaker was directly in front of the subject at a distance of 1 m and was mounted on a stand so as to be at the same height as the subject's head. The level of the speech signal was controlled by the audiometer. For tests using babble coincident with the speech, the babble on the other channel of the DAT was routed through the second channel of the audiometer and mixed with the speech before being delivered to the loudspeaker. The level of the babble at the point corresponding to the center of the subject's head was 65 dB SPL RMS. For tests using babble spatially separated from the speech, the babble was derived from a cassette recording replayed via a Sony TC WR750 cassette deck, a Tascam 106 mixer/preamplifier, a Yamaha P2075 stereo power amplifier, and two Tannoy PBM 6.5 loudspeakers, again mounted on stands at head height. The babble on one channel of the recording was delayed by 30 sec relative to the babble on the other channel. Thus, the babble delivered by the two loudspeakers was uncorrelated at the two ears. The loudspeakers were positioned on either side of the listener at +90° and −90° azimuth. At the point corresponding to the center of the subject's head, each babble separately gave a level of 65 dB SPL. The two babble sources together gave an overall level of 68 dB SPL. Levels of the speech and the babble were calibrated using a Bruel and Kjaer 2230 integrating sound level meter set to RMS, slow, with linear frequency weighting.

RESULTS OF THE INTELLIGIBILITY TESTS

Speech Intelligibility in Quiet

Mean scores obtained in quiet are shown in Figure 4 for conditions U, Own (limited to those subjects who normally wore aid(s)), L, and C. Error bars indicate ± 1 SE. All subjects achieved high scores in the C condition at all levels tested. This was also true for some subjects in the L condition, but others showed a decrease in intelligibility at the lowest level tested. For every subject, scores at 50 dB SPL were lower for the L condition than for the C condition. In condition Own, scores tended to drop slightly at 65 dB SPL and markedly at 50 dB SPL. In condition U, scores were reasonably high at 80 dB SPL, but dropped markedly at 65 dB and still further at 50 dB.

An analysis of variance (ANOVA) was conducted on the scores for the C, L, and U conditions with the factors subject, condition (U, C, or L), and level. The two tests for each condition were treated as replications because there were no clear practice effects. All of the main factors and all of the two-way interactions were highly significant ($p < 0.001$). Of major interest here is the effect of condition [$F(2, 256) = 339.2$, $p < 0.001$]

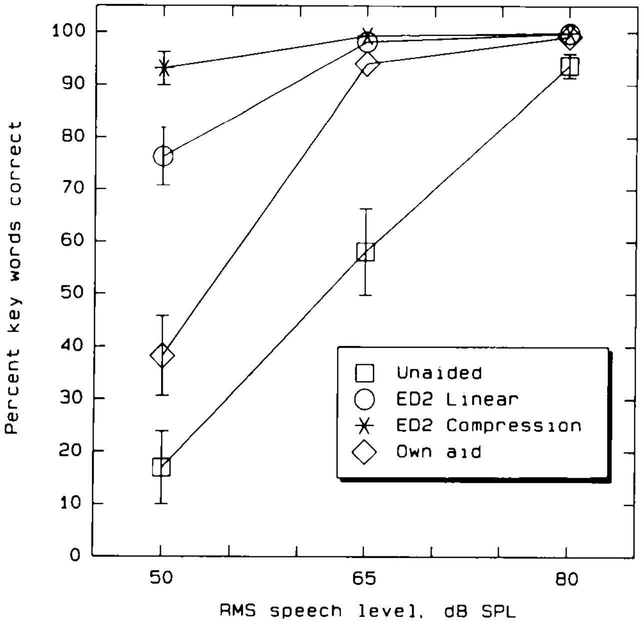


Figure 4. Mean results for speech intelligibility in quiet showing the percent key words correct as a function of the RMS level of the speech. Error bars show ±1 SE. Error bars are omitted where they would be smaller than the symbol used to plot the mean.

and the interaction of level and condition [$F(4, 256) = 77.6$, $p < 0.001$]. The analysis package used (GENSTAT) gave SEs of the differences between mean scores for the different conditions. These SEs were used to estimate the significance of the differences between means based on t-tests, with the degrees of freedom corresponding to the residual degrees of freedom in the ANOVA (Alvey, Galway, & Lane, 1982, p. 81). Condition C gave significantly higher scores than condition L [$t(256) = 3.47$, $p < 0.01$], and condition L gave significantly higher scores than condition U [$t(256) = 20.6$, $p < 0.001$]. The scores for condition C were significantly higher than those for condition L at 50 dB [$t(256) = 5.57$, $p < 0.01$], but not at the two higher levels.

Overall, the results clearly show the benefit of compression in allowing speech to be understood over a wide range of sound levels without any need to adjust the volume control. Two aspects of the results are somewhat surprising. The first is that performance in condition L remained relatively good for many subjects, even at the lowest level used. The second is that scores at the lowest level were markedly higher for condition L than for condition Own. These two aspects may be related. The two-band system used in the L condition, with adjustable cutoff frequency between the two bands and separate control of gain in each band, gives a great deal of flexibility in adjusting the frequency response to suit the individual subject. It seems likely that this allowed better fittings to be achieved than with the subjects' own aids. Although the mean insertion gain as a function of frequency was similar for the L and Own conditions (see Fig. 3), the plots for individual subjects often were markedly different between the L and Own conditions. A good fitting can lead to increased dynamic range both by ensuring adequate amplification at frequencies where it is needed most and by avoiding over-amplification at frequencies where not much gain is needed. It seems that for some of the subjects tested here, a rather wide dynamic range can be achieved for speech in quiet using linear amplification, provided the frequency-gain characteristic is appropriate. Other subjects, however, clearly benefit from compression, and understand speech at low levels better when compression is used.

The degree of benefit from compression was not predictable from the audiograms of the individual subjects. For example, the difference in score between the C and L conditions with an input speech level of 50 dB SPL [which we will denote Q50(C-L)] showed only weak nonsignificant correlations with the absolute threshold averaged over all audiometric frequencies ($r = 0.16$) or over the frequencies 2000, 3000, and 4000 Hz ($r = 0.16$). The value of Q50(C-L) was, however, highly correlated with the score obtained in condition L with an input level of 50 dB SPL ($r = -0.83$). In other words, the subjects who scored most poorly at the lowest speech level in condition L showed the largest improvements in condition C relative to condition L.

Speech Intelligibility in Noise

Slopes of the Psychometric Functions We start by considering the slopes of the functions relating percent correct intelligibility of speech in noise (babble) to speech to noise ratio; these slopes are relevant to the interpretation of the magnitude of changes in SRT in different conditions. As was described earlier, the SRT for a given subject, condition, and test session was estimated using probit analysis. The results did not show any systematic differences in the slopes of the probit functions across conditions. Therefore, to estimate the shape of the psychometric function for a given subject, the following procedure was adopted. The SRT obtained in a given run (i.e., for a single SRT estimate) was subtracted from each level used during that run. This had the effect of expressing all levels relative to the SRT. The data (the percent correct at each level relative to the SRT) were then collapsed across all runs for that subject, plotted as percent correct versus level, and a smooth curve drawn through the resulting points. A typical result is shown in Figure 5. The curve has roughly the form of a cumulative normal function, except that the asymptote is slightly less than 100%, reflecting the fact that subjects occasionally made mistakes even in quiet.

The slope of the psychometric function is rather steep. In the central part of the function (between 30 and 70%), the slope is about 19% per dB. The range of slopes across subjects was 16 to 22% per dB. Thus, each 1 dB change in speech to babble ratio gives about a 19% change in intelligibility. This slope value is similar to that found by Plomp and Mimpen (1979) using a series of Dutch sentences, but is somewhat greater than the slope of 11% per dB found by Laurence et al (1983) for the English recording of the BKB sentences presented in a background of speech-shaped noise. The steep slope means that quite small differences in SRT across conditions, say 1 dB, correspond to meaningful differences in intelligibility.

Comparison of SRTs for Binaural Listening in Condition U, L, and C To allow a comparison of overall performance in the U, L, and C conditions, the scores were averaged over the two binaural conditions (speech and noise coincident, and speech and noise separated). Because these SRTs are averaged across two conditions, the SE error of the estimates is about $0.83/\sqrt{2} = 0.59$ dB. There was some evidence of a practice effect in condition U; SRTs for the second test session in this condition were consistently lower than those for the first test session. This probably happened because the first test session in the U condition was also the very first test session of all, and subjects were unfamiliar with the test procedures. There was no evidence for practice effects in the L and C conditions. Therefore, results for conditions L and C were averaged over both test sessions, but results for condition U were averaged only over the final test session. The resulting SRTs are shown as speech to noise ratios in Figure 6. The horizontal line at 0 dB is provided as a visual aid; bars above that line indicate positive speech to noise ratios, and bars below the line indicate negative speech to noise ratios.

Several subjects show rather high SRTs (greater than 4 dB) in condition U, which can be attributed partly to their elevated thresholds for identifying speech in quiet. The asterisks above the filled bars identify those subjects whose score for speech in quiet at 65 dB SPL was 50% or less in condition U; for these subjects, the SRTs in quiet would have been greater than 65 dB, and so the SRT in 65 dB noise, expressed as a speech to noise ratio, could not have been less than 0 dB. Notice that there are several subjects (e.g., 7, 12, and 15) whose SRT in quiet would have been less than 65 dB SPL, yet whose SRT in noise was greater than 0 dB.

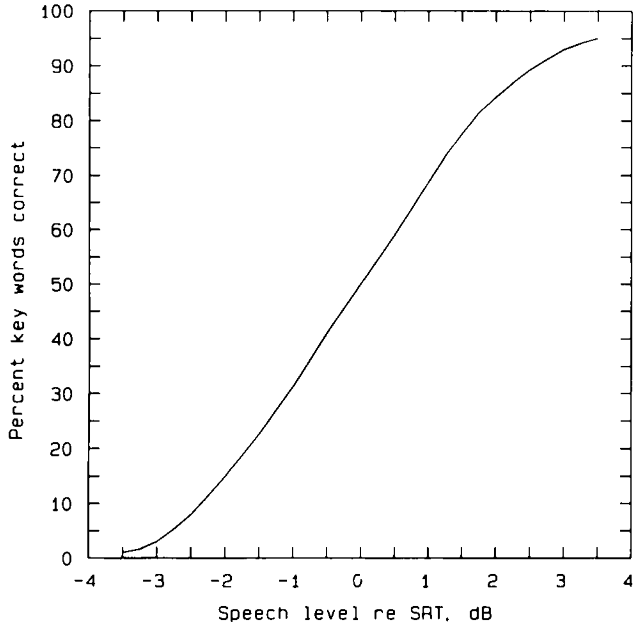


Figure 5. A typical psychometric function showing how the percent key words correct varies as a function of speech level (in noise) relative to the SRT.

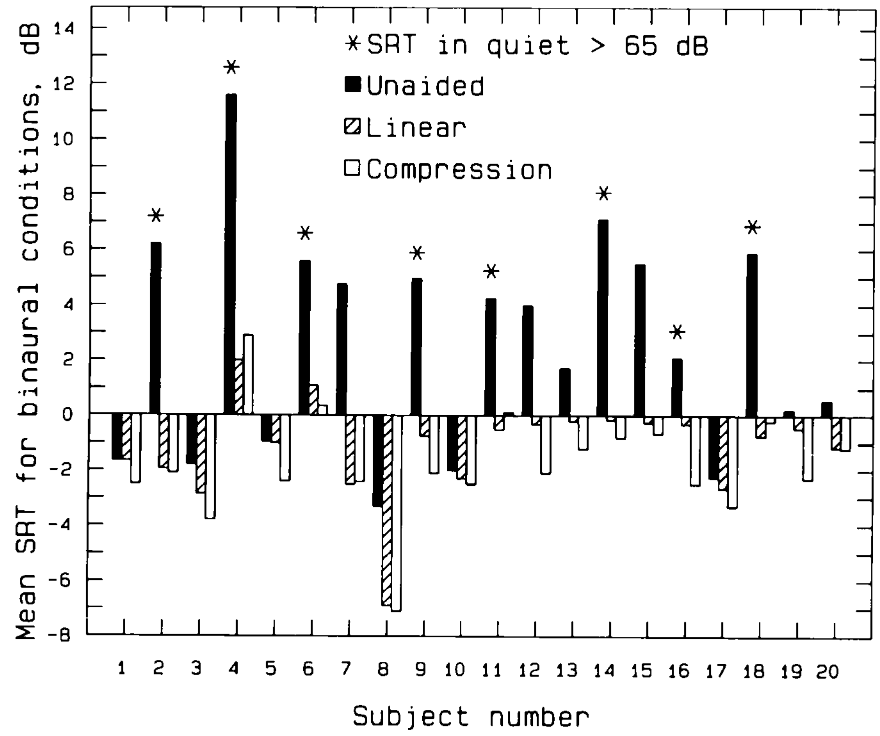


Figure 6. SRTs in noise, expressed as speech to noise ratios and averaged over the two conditions involving binaural listening (speech and noise coincident and speech and noise separated). For each subject, SRTs are shown for condition U (■), condition L (▨), and condition C (□). Asterisks indicate subjects who scored less than 50% correct when listening to speech in quiet at 65 dB SPL. For these subjects, SRTs in condition U were probably limited by their ability to understand speech in quiet.

The SRTs in condition L were mostly markedly below those in condition U, and the SRTs in condition C were lower still, indicating better performance. This occurred even for those subjects whose SRTs in noise were not limited by their SRTs in quiet. This result contrasts with that obtained by Duquesnoy and Plomp (1983) and Festen and Plomp (1986); they found that hearing aids did not generally improve the ability to understand speech in noise compared to unaided listening, except for subjects with moderate to severe losses when listening in relatively low levels of noise. Every one of our 20 subjects had a lower SRT (better intelligibility) in condition C than in condition U, including those whose performance was clearly not limited by the ability to hear speech in quiet. Thus, appropriate hearing aids can clearly improve the ability to understand speech in noise.

An ANOVA was conducted to assess the significance of these results. Factors in the analysis were subjects, conditions (U, L, or C), and noise location (coincident or separated). The analysis used the results from both test sessions for conditions L and C, but used only the results for the final test session in condition U; the analysis program was "told" that there were missing values for the replications in the U condition. This analysis showed the following: (1) There was a significant effect of condition (U, C, or L) [$F(2, 18) = 374$, $p < 0.001$]. All three conditions were significantly different from each other [$t(140) > 3.45$, $p < 0.01$]. Mean SRTs were 2.6 dB for condition U, −1.2 dB for condition L, and −1.8 dB for condition C. (2) There was a significant effect of subject [$F(19, 118) = 56.5$, $p < 0.001$] and a significant interaction between subject and condition [$F(38, 118) = 9.3$, $p < 0.001$]. This interaction indicates that subjects vary in the improvement produced by aiding. (3) There was no significant interaction between condition (U, C, or L) and noise location (coincident or separated) [$F(2, 118) = 0.32$, $p = 0.72$]. In other words, the binaural advantage occurring when the speech and noise were spatially separated was similar for the U, C, and L conditions. This indicates that the independent compression at the two ears did not adversely affect binaural processing. We will return to this point later.

In summary, the results show that binaural aiding gave significantly better speech intelligibility in noise than unaided listening. The average difference in SRT between conditions U and C was 4.4 dB, which corresponds to a substantial difference in intelligibility (about 80%) in babble at adverse speech to babble ratios. Every subject showed lower SRTs in condition C than in condition U. The results also show a significant advantage of condition C over condition L, although here the average difference was smaller (0.6 dB), and an advantage was not found for all subjects. Finally, the results show that the independent compression at the two ears does not impair the ability to use binaural cues under conditions where the speech and noise are spatially separated.

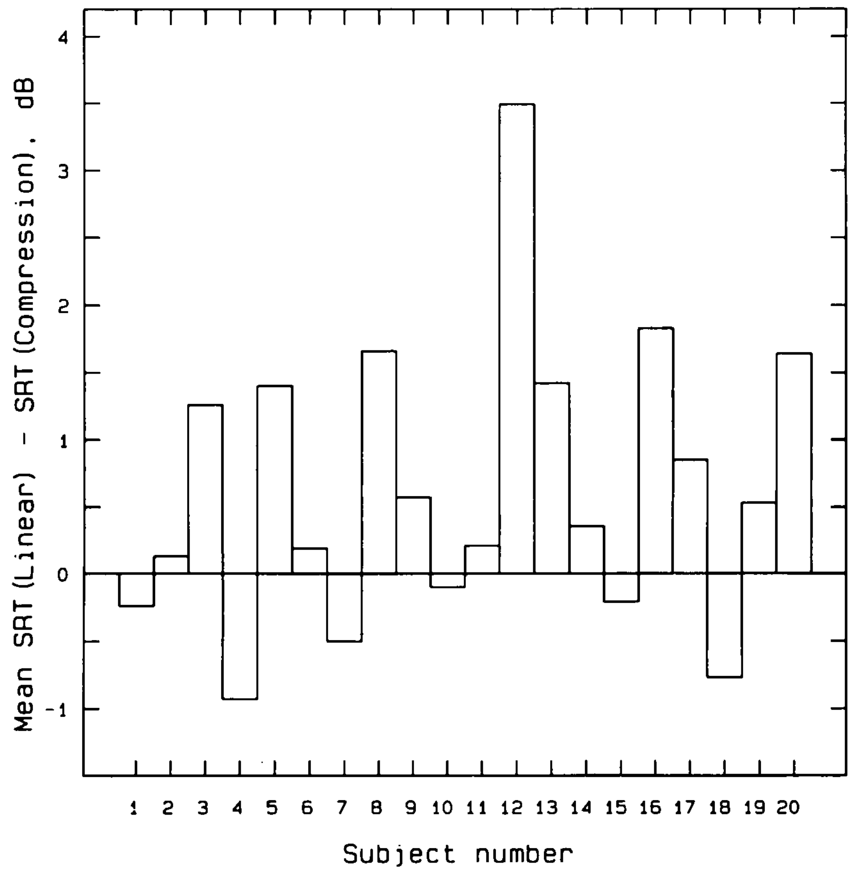
Comparison of Conditions L and C We consider next the differences between conditions L and C averaged over all listening conditions (binaural, left ear only aided, right ear only aided, speech and noise both coincident and separated). Figure 7 shows the difference between the mean SRT for condition L and for condition C separately for each subject. The differences are plotted so that positive numbers (above the horizontal line at 0 dB) indicate better performance in condition C and negative numbers indicate better performance in condition L.

It is clear that although most subjects showed better performance in condition C, there were marked individual differences. To try to get some insight into the nature of these individual differences, we calculated correlations between the scores shown in Figure 7 (i.e., the difference in mean SRT between the L and C conditions) and a variety of measures of dynamic range and loudness growth. The measures included the following: (1) The difference between the absolute threshold and the HCL separately for each frequency tested (0.5, 1.0, 2.0, and 4.0 kHz) and for various combinations across frequency. These data were obtained during the initial evaluation of each subject. (2) The slopes of the loudness growth functions obtained in the LGOB test, again separately for each frequency and for various combinations across frequency. (3) The difference between the absolute threshold and the UCL for 2 kHz sinusoids and for speech in conditions U, L, and C. These data were obtained after all sessions measuring speech intelligibility had been completed, using binaural listening in free-field conditions.

None of these measures showed a high correlation with the improvement produced by compression. The highest correlation ($r = 0.483$, $p < 0.05$) was with the difference between absolute threshold and the HCL averaged over the frequencies 2 and 4 kHz. We will refer to this measure of dynamic range as DR(2,4). The relationship of the difference in mean SRT between conditions L and C and DR(2,4) is shown in Figure 8.

There is a trend for subjects with small dynamic ranges to show large benefits of compression. Subjects with values of DR(2,4) greater than 27 dB tend to show small or no benefit from compression, whereas subjects with values of DR(2,4) less than 27 dB tend to show consistent advantages from compression. This is illustrated in Figure 9, which plots the mean SRT in condition C against the mean SRT in condition L; both SRTs are expressed as speech to noise ratios. Subjects with values of DR(2,4) greater than 27 dB are shown by asterisks, whereas those with values of DR(2,4) less than 27 dB are shown by circles. The diagonal line indicates where the points would lie if there were no difference between the L and C conditions. The asterisks tend to lie around the line, whereas the circles tend to fall below the line, indicating better performance in the C condition. Note that compression can produce improvements in intelligibility both for subjects who are performing rather well in condition L (i.e. have lower than average SRTs) and for subjects who are performing rather poorly.

Figure 7. The difference between the mean SRT in noise for condition L and condition C, shown separately for each subject. SRTs were averaged over all listening conditions. Results are plotted so that superior performance in condition C is indicated by bars above the horizontal line at 0 dB.



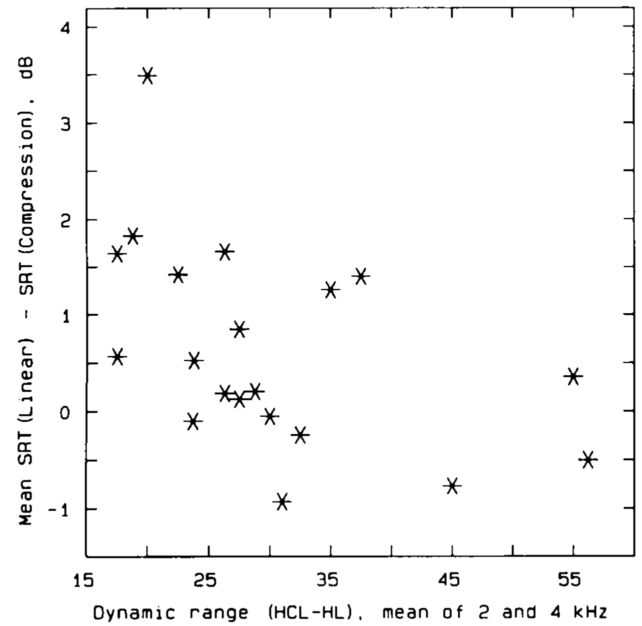


Figure 8. The difference between the mean SRT in noise for condition L and condition C, plotted as a function of the average dynamic range for tones at 2 and 4 kHz. Large advantages for condition C tend to be associated with small dynamic ranges.

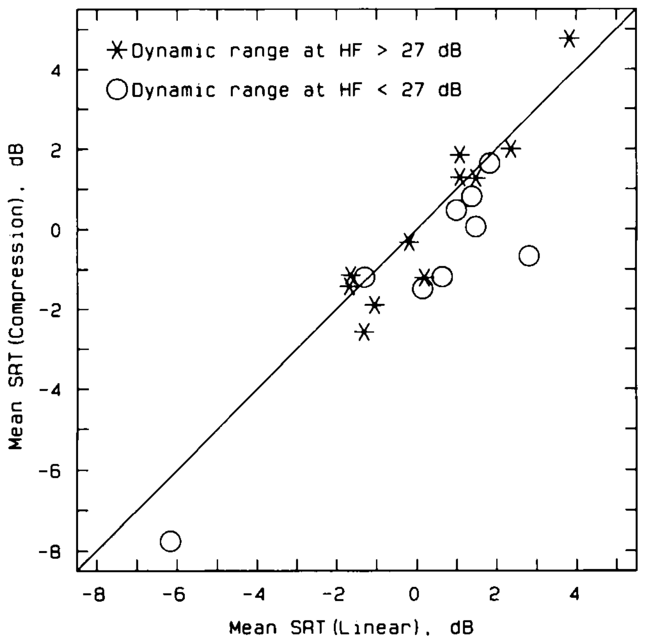


Figure 9. The mean SRT for condition C plotted against the mean SRT in condition L, separately for each subject; SRTs are expressed as speech to noise ratios. Points falling below the diagonal line indicate better performance in condition C. Subjects with dynamic ranges at high frequencies greater than 27 dB are indicated by (*), and those with dynamic ranges less than 27 dB are indicated by (○).

To assess the statistical significance of the differences between the L and C conditions, two ANOVAs were conducted. In the first, the results of all subjects were analyzed. In the second, the analysis was restricted to those subjects with values of DR(2,4) less than 27 dB. The following factors were used in both analyses: fitting condition (C or L), listening condition (Bin, LA, and RA, speech and babble coincident or separated), and subject. The results from the first and second test sessions for each fitting condition were treated as replications. The results of the first analysis are summarized below.

(1) There was a significant effect of fitting condition [$F(1, 335) = 30.2, p < 0.001$]. The mean SRT was 0.30 dB for condition L and −0.34 dB for condition C. In other words, condition C gave significantly better speech intelligibility in noise. (2) When the speech and noise were coincident, performance was better for binaural than for monaural aiding; averaged across conditions, SRTs were −1.33 dB for Bin, 0.57 dB for LA, and 0.52 dB for RA. The difference between binaural and monaural aiding, averaging 1.88 dB, was highly significant [$t(335) = 9.2, p < 0.001$]. (3) When the speech and noise were spatially separated, binaural aiding was better than monaural; averaged across conditions, SRTs were −1.66 dB for Bin, 0.93 dB for LA, and 0.87 dB for RA. The difference between binaural and monaural aiding, averaging 2.56 dB, was significant [$t(335) = 12.6, p < 0.001$]. (4) There was no significant interaction between fitting condition (C or L) and binaural advantage [$F(5, 335) = 0.32, p = 0.9$]. In other words, the extent of the binaural advantage was similar for the C and L conditions. As before, this indicates that the independent AGC action at the two ears does not reduce the binaural advantage of listening to speech in noncoincident noise. (5) There was a significant effect of subject [$F(19, 335) = 74.6, p < 0.001$] and a significant interaction between subject and fitting condition [$F(19, 335) = 4.23, p < 0.001$]. This confirms that the extent of the benefit of compression varies across subjects.

The results of the second ANOVA, restricted to subjects with values of DR(2,4) less than 27 dB, were generally similar to those just described, except that the average difference between fitting conditions was larger: the average SRT in condition L was 0.21 dB, whereas that in condition C was −1.04 dB, a difference of 1.25 dB. This is equivalent to an improvement in intelligibility in the C condition of about 24%.

In summary, the results show a significant advantage for two-channel compression over linear amplification. The size of the advantage varies across subjects. Subjects with dynamic ranges at high frequencies, DR(2,4), larger than 27 dB tend to show little benefit from compression, whereas subjects with values of DR(2,4) less than 27 dB tend to show larger advantages, averaging 1.25 dB. This is equivalent to an improvement in intelligibility of about 24%.

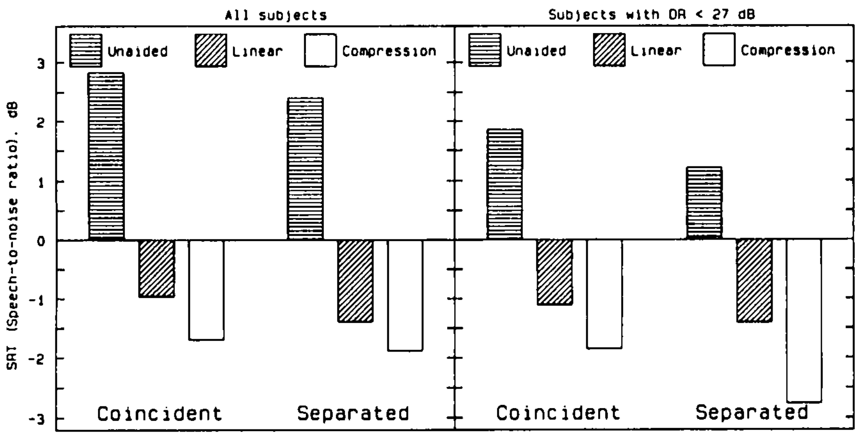
Comparison of Conditions C and Own Comparison of results for the C and Own conditions is complicated by the fact that all subjects were fitted binaurally in condition C, whereas only a minority of the subjects normally used binaural aids. To make the comparison as fair as possible, we decided to use the results only for those subjects who normally wore binaural aids and who had been wearing their own aids for the week before testing in condition Own; these were subjects 1, 2, 4, 12, 13, 14, 16, and 18. Furthermore, only the results for conditions involving binaural aiding were used. For the case where the speech and noise were coincident, the mean SRTs, expressed as speech to noise ratios, were −1.2 dB in condition C and +0.5 dB in condition Own. The mean difference in SRT, 1.7 dB, was significantly different from zero [$t(8) = 3.42, p < 0.01$]. Seven out of eight subjects had lower SRTs for condition C than for condition Own. For the case where the speech and noise were spatially separated, the mean SRTs were −0.9 dB for condition C and +1.1 dB for condition Own. Again, the mean difference in SRT, 2.0 dB, was significantly different from zero [$t(8) = 3.14, p < 0.02$]. Six out of eight subjects had lower SRTs in condition C than in condition Own. We may conclude that binaural aiding with two-channel compression aids gave better speech intelligibility in noise than binaural aiding with the subjects' own aids both when the speech and noise were coincident and when they were separated.

Advantages of Binaural Amplification In this section, we consider the advantages of binaural amplification for understanding speech in noise under conditions where the noise and speech are coincident and where they are separated. Figure 10 shows SRTs in the U, L, and C conditions (binaural aiding only) averaged across all subjects (left panel) and across those subjects with a value of DR(2,4) less than 27 dB; as before, all SRTs are expressed as speech to noise ratios. As noted before, performance was worst for unaided listening, and best for condition C. Thresholds were consistently slightly lower when the speech and noise were spatially separated than when they were coincident, and the size of this effect was similar for conditions U, L, and C. Thus, neither binaural linear aiding nor binaural compression aiding adversely affected binaural processing.

At first sight, it may appear surprising that the effect of spatial separation of the speech and noise was so small. However, it should be remembered that, in the case where the noise and speech were spatially separated, there was an independent noise source directly opposite each ear (i.e., at +90° and −90° azimuth). This would have produced a worse speech to noise ratio in the ear canal than for the case where the noise came from in front of the subject, which would have offset the expected binaural advantage. Note that the positions of the noises were deliberately chosen to avoid any advantages that might have occurred as a result of head shadow effects.

Figure 11 compares SRTs for binaural aiding and

Figure 10. Mean results for binaural listening conditions showing SRTs in noise as speech to noise ratios. Results are shown separately for condition U (□), condition L (□), and condition C (□). Results are shown averaged across all subjects (left panel) and across subjects with dynamic ranges at high frequencies less than 27 dB (right panel). In each panel, results with the speech and noise coincident are shown on the left, and results with the speech and noise separated are shown on the right.



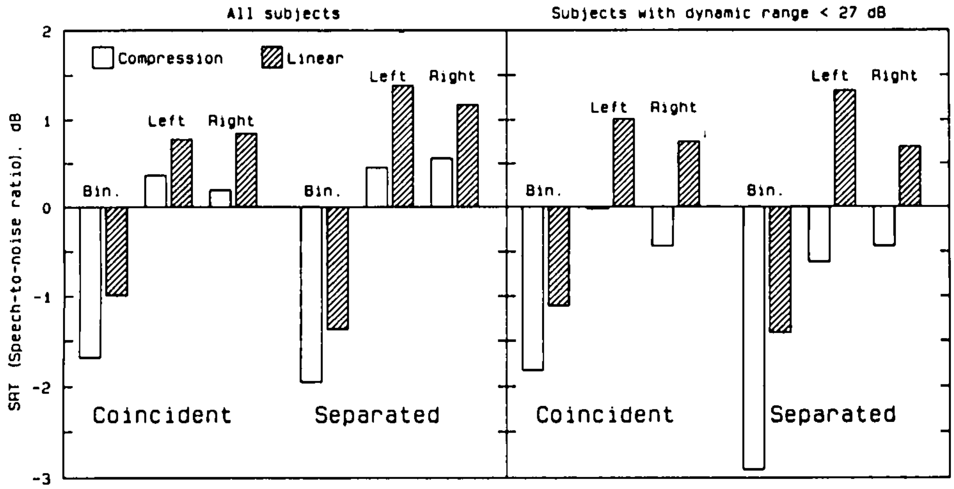


Figure 11. Mean SRTs in noise for the C (□) and L (□) conditions, shown separately for binaural aiding (Bin.) and left or right ear only aided. Otherwise, as Figure 10.

monaural aiding for conditions L and C. The left panel shows results averaged across all subjects, and the right panel shows results averaged across subjects with a value of DR(2,4) less than 27 dB. For both the L and C conditions, it is clear that binaural aiding gives markedly lower SRTs than monaural aiding. The advantage when the speech and noise were coincident (averaging 1.9 dB in the left panel and 1.8 dB in the right panel) reflects a diotic summation effect, because in this case, the speech and noise were similar at the two ears, except for any slight asymmetries produced by the aids. The advantage when the speech and noise were spatially separated (averaging 2.6 dB in the left panel and 2.4 dB in the right panel) probably reflects both diotic summation and binaural processing.

Festen and Plomp (1986) found lower SRTs in noise for binaural compared to monaural aiding only for subjects with relatively high absolute thresholds (average HL at 500, 1000, and 2000 Hz greater than 50 dB), whereas almost all of our subjects showed a binaural advantage, regardless of their absolute thresholds. Two factors may have contributed to this. First, the unaided ear of our subjects was blocked, which was not the case for the subjects of Festen and Plomp. Second, they used only a single competing noise source, whereas we used two noises, one on each side of the head. Laurence et al (1983) used a similar situation to ours (with the unaided ear blocked), and they found similar binaural advantages.

It is noteworthy that for all listening conditions shown in Figure 11, condition C gives lower SRTs than condition L. The advantage of compression is greater for the subjects with smaller dynamic ranges, reaching an average value in the latter case of 1.5 dB for binaural aiding with the speech and noise spatially separated. This is, of course, the usual situation in everyday life.

RESULTS OF THE QUESTIONNAIRES

As indicated earlier, at various points during the trial, subjects were asked to fill in questionnaires about their experiences with the aids in everyday life. The questionnaires administered at sessions 4 and 6 (conditions L and C) asked the subjects to indicate the extent to which they agreed or disagreed with each statement given below by circling a number between 1 and 7. Circling 1 meant strongly disagreeing and circling 7 meant strongly agreeing.

1. I easily understand speech in a quiet room with one person talking.
2. I easily understand speech when three or four people are talking at once.

3. I easily understand speech at meal times.
4. I easily understanding speech in a car.
5. I easily understand soft (quiet) speech.
6. Sounds never become unpleasantly loud.
7. I easily understand the TV and radio.
8. I do not need to adjust the volume controls to deal with different situations.
9. I find the overall sound quality pleasant.
10. My own voice sounds natural to me.
11. I like the appearance of the hearing aids. Note that the questions were all worded so that higher numbers indicated better preference/performance.

Note also that the final question was different in nature from the others. Because the appearance of the aids was identical in the L and C conditions, this question gave a useful check on the consistency of the subjects' responses.

The questionnaire administered at session 7 to subjects who had been wearing one or two aids (non-ReSound) during the past week (18 out of the 20 subjects) was essentially the same. The questionnaire administered at session 7 to subjects who did not wear hearing aids during the past week was similar except that questions 8 to 11 were omitted and the first sentence of the instructions was "Please think about your experiences when you do not wear hearing aids."

The mean scores are shown in Figure 12. There is a clear trend for the linear aids to be preferred over Own/no aids and for the compression aids to be preferred over both the linear and Own/no aids. An ANOVA was carried out on the scores for questions 1 to 10 (those relating to sound quality), with factors subjects, question number, and condition (C, L, or Own/no aid). The variance associated with the three-way interaction was used as an estimate of the residual variance. In a few cases, subjects did not respond to a particular question because they had not encountered that situation (e.g., listening to the radio or television). These cases were treated as missing values in the analysis. The same was done for the missing answers to questions 8 to 11 for the two subjects who had not worn their own aids. All of the main effects and two-way interactions were highly significant ($p < 0.001$). The scores for condition L were significantly higher than those for condition Own/no aid [$t(333) = 3.65$, $p < 0.001$], and the scores for condition C were significantly higher than those for condition L [$t(333) = 2.88$, $p < 0.01$]. Thus, the results of the questionnaires indicate significant advantages for compression over linear amplification in everyday listening situations.

An exception is in the case of judgments of the quality of the subjects' own voices. These judgments are influenced by the "occlusion effect," which can occur when the ear canal is blocked. Usually, this effect can be alleviated and the quality of the person's own voice made more natural by venting the aid; the larger the vent, the more natural voice quality becomes. However, as described earlier, in condition C the bore of the vent often had to be reduced relative to that used in condition L to avoid acoustic feedback. Sometimes, the vent had to be blocked completely in condition C. This can account for the fact that preferences for the quality of their own voices were slightly higher for the L than for

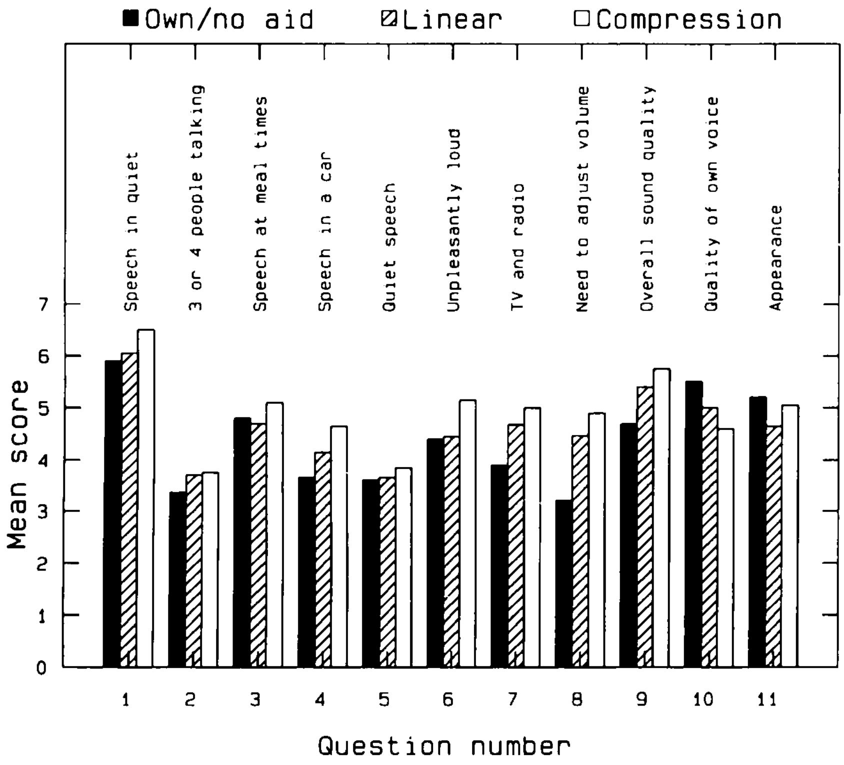


Figure 12. Results of the questionnaires averaged across subjects. Each question was worded so that higher numbers indicate greater preference.

the C condition. In the case of their own aids, several of the subjects wore behind the ear aids, which are less prone to feedback than in the ear aids, and which therefore allow larger vents to be used. This can account for the slightly greater preference for their own aids in judgments of the quality of their own voices.

Although most subjects were consistent in their judgments of the appearance of the aids used in conditions C and L, a few gave higher ratings for appearance in condition C. This appears to reflect a bias to prefer the appearance of the aid whose sound quality is preferred.

A few general comments should be made about the results of the questionnaires. First, the overall impression of condition C may have been adversely affected by the greater tendency for feedback to occur in that condition. Several of the subjects commented on this. In a few cases, the overall gain in condition C had to be reduced below the desired value to prevent feedback from occurring in some everyday situations (for example, eating, where jaw movements break the seal of the mold to the ear canal). The overall advantage of condition C occurred in spite of this difficulty.

A second comment is that we may have inadvertently misled some of the subjects in giving general instructions in the use of the aids. For the C and L conditions, subjects were asked to use the aids at the "reference" volume setting as much as possible; recall that this reference setting was always used in the test sessions. Some subjects interpreted this to mean that they should not use the volume control at all, even though they would have sometimes liked to make small adjustments. Thus, in response to question 8, about the use of the volume control, one subject gave a response of 1 for both the C and L conditions. Other subjects did not feel the need to adjust the volume control and gave a response of 7 for both the C and L conditions. Obviously, such responses reduce the average difference between the C and L conditions. Finally, some subjects were rather uncritical and gave responses of 7 to almost every question in all questionnaires.

In view of these points, it is not surprising that the average differences between the scores for the C, L, and Own/no aid conditions are rather small. Indeed, it is noteworthy that the advantage of condition C over condition L occurred consistently for questions 1 to 9 in spite of the factors described above. The advantages of condition C revealed in the laboratory tests of intelligibility in quiet and noise are confirmed by the results of the questionnaires.

GENERAL DISCUSSION

Comparison with Earlier Work

The results of this study are consistent with a growing body of evidence that multiband compression using a large number of bands gives small advantages or no advantage over linear amplification for listening to speech in noise (Braida et al, 1982; Lippmann et al, 1981; Plomp, 1988), whereas two-band systems can give significant advantages (Laurence et al, 1983; Moore, 1987; Moore & Glasberg, 1988; Ringdahl, Eriksson-Mangold, Israelsson, Lindkvist, & Mangold, 1990). Single-band, two-band, and multiband systems can all give advantages in quiet when speech material with a wide dynamic range is used (Bustamante & Braida, 1987; Laurence et al, 1983; Lippmann et al, 1981; Moore & Glasberg, 1986, 1988; Moore, Glasberg, & Stone, 1991). The relatively poor performance in noise found with systems using many bands can probably be explained by the spectral flattening (reduction of spectral contrasts) and the distortions in amplitude envelope introduced by such systems (Bustamante & Braida, 1987; Moore, 1990; Plomp, 1988).

It is instructive to compare the present results with earlier results obtained using two-band compression systems. The system evaluated by Moore and his colleagues (Laurence et al, 1983; Moore, 1987, 1990; Moore & Glasberg, 1988) combined two forms of compression. A slow-acting AGC system operating on the whole speech signal (referred to as front-end AGC) was used to compensate for variations in overall speech level from one situation to another. The signal was then split into two bands with fast-acting compression in each band. In both cases, compression limiting was used; the AGC amplifiers were linear up to a certain threshold level and compressed strongly beyond that. This contrasts with the present system using full dynamic range compression with adjustable compression ratio.

The results in quiet are similar for the present study and the studies of Moore and his colleagues (Laurence et al, 1983; Moore & Glasberg, 1986). Both compression systems allowed speech in quiet to be understood over a wide range of levels without any need to adjust the volume control on the aids. Thus, a wide dynamic range for speech in quiet can be achieved using two rather different compression systems.

For speech in noise, the system used by Moore and his colleagues gave advantages for compression similar to or slightly larger than those for the present system. For example, Laurence et al (1983) found that SRTs were, on average, 2.4 dB lower for the compression system than for a similar linear amplification system. For the speech and noise they used, this is equivalent to about a 26% improvement in intelligibility, similar to that found for the subjects in the present experiment with narrow dynamic ranges at high frequencies. Moore and Glasberg (1988) used an improved front-end AGC system, and applied fast-acting limiting compression in the high band only. They found a mean improvement in SRT of 4 dB for compression compared to linear amplification. In this case, the improvement was larger than that found in the present study. Moore and Glasberg also found that adding fast-acting compression limiting in the low-frequency band did not generally improve speech intelligibility in noise, and sometimes impaired it.

It is difficult to draw firm conclusions from comparisons across studies because there are many differences between studies other than the compression system used, for example, overall frequency response and range, the nature of the fitting procedure, the type of aid (behind the ear, in the ear, or laboratory prototype), the type of background noise, and the subjects used. Nevertheless, the results suggest that there may be advantages in using a slow-acting front-end AGC before the fast-acting compression in the channels. Further research is needed to establish whether this is the case. Research is also needed to establish whether the full dynamic range compression used in the present study is more or less effective than the compression limiting used in earlier studies by Moore and his colleagues.

The Fitting Procedure

Systems such as the ReSound ED2 have a great deal of flexibility in the way that frequency response shape, gains, and compression ratios can be chosen. Unfortunately, this flexibility also entails greater complexity in fitting the aid to a given patient, and in trying to determine whether an "optimum" fitting has been achieved. The present fitting system, based on the LGOB test and the audiogram shape, does seem to give reasonable results. However, the fitting parameters derived using this procedure often have to be modified somewhat before the patient is satisfied. Even after these modifications, some uncertainty remains as to whether the fitting is the best that can be achieved. One difficulty is that some patients are rather uncritical. After an initial fitting, a patient may comment that the aid sounds "perfect," yet, after modifications are made, they report that it "sounds even better!"

Most fitting procedures for hearing aids are based on measurements with very simple signals, such as sinewaves or bands of noise. Yet, such signals are far removed from the speech and other complex signals that the user wishes to hear in everyday life. Moore (1987) suggested that it may be better to use fitting procedures based on speech stimuli, and presented some pilot data lending weak support to that view. However, much work remains to be done to develop and evaluate such fitting procedures. Indeed the development of effective and clinically applicable fitting procedures is a major challenge which will be critical to the success of programmable hearing aids such as the ED2.

CONCLUSIONS

The main conclusions of this article are as follows: (1) A hearing aid incorporating two-band fast-acting dynamic range compression (the ReSound ED2, condition C) allowed speech in quiet to be understood over a wide range of sound levels without any need to adjust the controls of the aid. Performance in this respect was superior to that of the ED2 programmed as a linear aid (condition L), to the subjects' own aids (Condition Own), and to unaided listening (condition U). (2) For all subjects, the SRT in noise was lower (indicating better performance) for condition C than for condition U. For the majority of subjects, the SRT was lower in condition C than condition L, indicating that two-band compression improves the ability to understand speech in noise. The improvement was greatest for subjects with small dynamic ranges at high frequencies. For subjects who normally wore binaural aids, SRTs in noise were lower for condition C than for condition Own. (3) SRTs in noise were markedly lower for binaural aiding than for monaural aiding (with the unaided ear blocked). This was the case both when the speech and noise were coincident and when they were spatially separated, although the effect was slightly larger in the latter case. (4) The improvement produced by binaural aiding as compared to monaural aiding was similar for the C and L conditions. This indicates that the independent compression at the two ears did not adversely affect the processing of binaural cues. (5) The results of the questionnaires indicated that the C condition was preferred over the L and Own conditions in everyday listening situations.

REFERENCES

Allen JB and Jeng PS. Loudness growth in 1/2-octave bands (LGOB): a procedure for the assessment of loudness. J Acoust Soc Am 1990;88:745–753.

Alvey N, Galway N, and Lane P. An Introduction to GENSTAT. London: Academic Press, 1982.

Bench J and Bamford J. Speech-Hearing Tests and the Spoken Language of Hearing-Impaired Children. London: Academic, 1979.

Bilger RC, Nuetzel JM, Rabinowitz WM, and Rzeczkowski C. Standardization of a test of speech perception in noise. J Speech Hear Res 1984;27:32–48.

Braida LD, Durlach NI, De Gennaro SV, Peterson PM, and Bustamante DK. Review of recent research on multiband amplitude compression for the hearing impaired. In Studebaker GA and Bess FH, Eds. The Vanderbilt Hearing-Aid Report, Upper Darby, PA: Monographs in Contemporary Audiology, 1982:133–140.

Bustamante DK and Braida LD. Multiband compression limiting for hearing-impaired listeners. J Rehabil Res Dev 1987;24:149–160.

Davis A, Haggard M, and Bell I. Magnitude of diotic summation in speech-in-noise tasks: performance region and appropriate baseline. Br J Audiol 1990;24:11–16.

Duquesnoy AJ and Plomp R. The effect of a hearing-aid on the speech-reception threshold of hearing-impaired listeners in quiet and in noise. J Acoust Soc Am 1983;73:2166–2173.

Festen JM and Plomp R. Speech reception threshold in noise with one and two hearing aids. J Acoust Soc Am 1986;79:465–471.

Finney DJ. Probit Analysis. Cambridge: Cambridge University Press, 1971.

Glasberg BR and Moore BCJ. Auditory filter shapes in subjects with unilateral and bilateral cochlear impairments. J Acoust Soc Am 1986;79:1020–1033.

Kalikow DN, Stevens KN, and Elliot LL. Development of a test of speech intelligibility in noise using sentence materials with controlled word predictability. J Acoust Soc Am 1977;61:1337–1351.

Laurence RF, Moore BCJ, and Glasberg BR. A comparison of behind-the-ear high-fidelity linear aids and two-channel compression hearing aids in the laboratory and in everyday life. Br J Audiol 1983;17:31–48.

Leek MR, Dorman MF, and Summerfield Q. Minimum spectral contrast for vowel identification by normal-hearing and hearing-impaired listeners. J Acoust Soc Am 1987;81:148–154.

Levitt H. Speech discrimination ability in the hearing impaired: Spectrum considerations. In Studebaker GA and Bess FH, Eds. The Vanderbilt Hearing-Aid Report, Upper Darby, PA: Monographs in Contemporary Audiology, 1982:32–43.

Lippmann RP, Braida LD, and Durlach NI. Study of multi-channel amplitude compression and linear amplification for persons with sensorineural hearing loss. J Acoust Soc Am 1981;69:524–534.

McAlister PV. The effect of hearing aids on speech discrimination in noise by normal-hearing listeners. J Rehabil Res Dev 1990;27:33–42.

Moore BCJ. Design and evaluation of a two-channel compression hearing aid. J Rehabil Res Dev 1987;24:181–192.

Moore BCJ. How much do we gain by gain control in hearing aids? Acta Otolaryngol [Suppl] 1990;469:250–256.

Moore BCJ and Glasberg BR. A comparison of two-channel and single-channel compression hearing aids. Audiology 1986;25:210–226.

Moore BCJ and Glasberg BR. A comparison of four methods of implementing automatic gain control (AGC) in hearing aids. Br J Audiol 1988;22:93–104.

Moore BCJ, Glasberg BR, and Stone MA. Optimization of a slow-acting automatic gain control system for use in hearing aids. Br J Audiol 1991;25:171–182.

Moore BCJ, Laurence RF, and Wright D. Improvements in speech intelligibility in quiet and in noise produced by two-channel compression hearing aids. Br J Audiol 1985;19:175–187.

Pearsons KS, Bennett RL, and Fidell S. Speech Levels in Various Environments. Report No. 3281. Cambridge, MA: Bolt, Beranek and Newman, 1976.

Plomp R. The negative effect of amplitude compression in multichannel hearing aids in the light of the modulation-transfer function. J Acoust Soc Am 1988;83:2322–2327.

Plomp R and Mimpen AM. Improving the reliability of testing the speech reception threshold for sentences. Audiology 1979;18:43–53:28–29,32.

Pluvinage V. Clinical measurement of loudness growth. Hear Instrum 1989;39.

Ringdahl A, Eriksson-Mangold M, Israelsson B, Lindkvist A, and Mangold S. Clinical trials with a programmable hearing aid set for various listening environments. Br J Audiol 1990;24:235–242.

Tyler RS. Frequency resolution in hearing-impaired listeners. In Moore BCJ, Ed. Frequency Selectivity in Hearing, London: Academic Press, 1986:309–371.

Villchur E. Signal processing to improve speech intelligibility in perceptive deafness. J Acoust Soc Am 1973;53:1646–1657.

Villchur E. The evaluation of amplitude-compression processing for hearing aids. In Studebaker GA and Bess FH, Eds. The Vanderbilt Hearing-Aid Report, Upper Darby, PA: Monographs in Contemporary Audiology, 1982:141–143.

Villchur E. Comments on "The negative effect of amplitude compression in multichannel hearing aids in the light of the modulation-transfer function" [J. Acoust. Soc. Am. 83,2322–2327(1988)]. J Acoust Soc Am 1989;86:425–427.

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