# Noise, Amplification, and Compression: Considerations of Three Main Issues in Hearing Aid Design

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**This paper deals with the following three topics: (1) interfering noise (voice babble, single competing speaker) as the main problem of many hearing-impaired listeners, (2) the amplitude-frequency response of the hearing aid, and (3) the benefit of frequency-dependent compression. Research by the author and his coworkers has shown that: (1) persons with impaired hearing typically need 3 to 6 dB higher speech-to-noise ratios than do normal-hearing listeners—a technically very difficult problem to solve; (2) within a relatively ample range, the speech-reception threshold in noise is independent of the amplitude-frequency response; and (3) the small time constants of syllabic compression deteriorate the speech signal. Multichannel amplification (2–4 channels) with automatic gain control for each channel is recommended, optimally adjusted to keep the (variable) speech signal within the impaired ear's limited dynamic range as well as to preserve the intensity differences of successive speech phonemes.**

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The great majority of papers on the design and fitting procedures of hearing aids in the last few decades essentially are dealing with the following three issues: the disturbing effects of interfering sounds, the optimal amplitude-frequency response, and the purpose of amplitude compression. These are the topics to be discussed in this paper, particularly in the light of the current research program in our department. I restrict myself to sensorineural hearing impairment up to a degree that speech communication *in quiet* is no problem if adequate amplification is provided. The questions involved are:

**(1) The Noise Problem** • Many hearing-impaired subjects have difficulties with speech recognition in the presence of other sounds: interfering noises or competing speakers. An appropriate measure for this loss is the number of decibels that their speech-reception threshold (SRT) in noise is higher than for persons with normal hearing. This difference has been labelled *hearing loss for speech in noise.* This topic is discussed first because it represents the main problem for the hearing impaired. A low SRT in noise should be the primary criterion for the optimal amplitude-frequency response as well as the compression characteristics.

**(2) The Amplitude-Frequency Response** • The fact that most hearing losses are frequency dependent implies that a simple, straightforward amplifier generally is not the best solution to deliver all speech-signal components at the optimal level within the reduced range between threshold and the loudness discomfort level; therefore, we have to know which amplitude-frequency response should be chosen and how critical this decision is.

**(3) The Compression Characteristics** • The considerably smaller amplitude range of the impaired ear between the *elevated* threshold of hearing and the usually *unchanged* loudness discomfort level introduces the question of whether and how variations in the speech signal have to be reduced in order to have optimal intelligibility under various acoustical conditions.

## The Noise Problem

After the publication of my first paper (Plomp, 1978) on the problems of the hearing impaired, I was criticized for my reference to "the limited benefit of hearing aids." In that article, I asked attention for the fact that, although as long ago as in 1935–1936 in an inquiry of the United States Health Service *stage 1* of hearing handicap was described as: "The individual has difficulty in understanding speech in church, at the theater, or in group conversation, but can hear speech at close range without any artificial assistance" (Beasley, 1940), the noise problem did not apparently bother most investigators involved in hearing aid research. The aim of my paper was to convince them that the traditional hearing aid as an amplifier does not improve the speech-to-noise (S/N) ratio and, therefore, falls short in meeting the main problem of hearing-impaired subjects.

However, the limited benefit terminology may have contributed to the considerably increased attention being given to the noise problem in the last

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10 to 15 years. As it appeared to be almost impossible to generalize the available data on the effect of hearing impairment on speech intelligibility in noise, I decided to focus my own research activities primarily on that question. A standard test was developed (Plomp & Mimpen, 1979) for measuring the SRT for sentences in noise under strictly defined conditions (viz.: SRT expressed in S/N ratio for 50% correct sentences; the noise spectrum equal to the long-term spectrum of the sentences; the speech level in Leq, based on the long-term sentence intensity). Each list contained 13 short sentences, presented successively in an up-down procedure. When the listener was able to repeat the entire sentence correctly, the next one was presented at a 2 dB lower level, and vice versa. The average level of the last 10 sentences was defined as the SRT. This test has been accepted by many investigators in the Netherlands and a similar approach in other languages has been introduced (Hagerman, 1982; MacLeod & Summerfield, 1990; M. Nilsson, S.D. Soli, & J. Sullivan, unpublished data, 1993).

As an illustration of a hearing aid's limitations, in Fig. 1 some SRT data, measured without and with hearing aid, are plotted (adapted from Duquesnoy & Plomp, 1983). The curves represent the best fit for the S/N ratio model proposed in my 1978 paper. They give the aided speech-reception threshold (ASRT) in dB(A) as a function of the parameters involved:

$$\mathrm{ASRT} = 10\ log\left\{10^{\frac{(L_o + A - G + D + S)}{10}} + 10^{\frac{(L_{NT} - \Delta L_{SN} + D + S)}{10}}\right\} \quad (1)$$

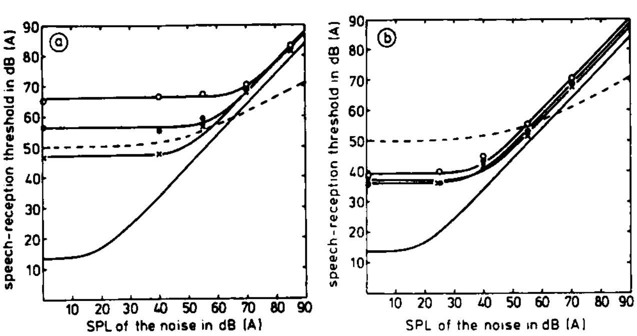


**Figure 1. Speech-reception threshold of sentences in noise as a function of noise level, without (panel a) and with (panel b) hearing aid. The data points represent average values for three groups of 10 hearing-impaired listeners with mean hearing levels for 500, 1000, and 2000 Hz of 36 to 42, 42 to 48, and 48 to 54 dB, respectively. The solid curves give the best fit of the points to the speech-to-noise model of Eq. 1, with the lower curve representing the threshold for normal-hearing listeners. The dashed curve illustrates how the average sound level of conversational speech (distance 1 m) rises automatically with noise level (Plomp, 1986).**

with $L_o$ = SRT for normal-hearing subjects in quiet, in dB(A)

$L_{NT}$ = sound-pressure level (SPL) of the noise (inclusive internal noise of the aid) in dB(A)

$-\Delta L_{SN}$ = SRT for normal-hearing subjects in noise, expressed in S/N ratio

$A + D$ = hearing loss for speech in quiet in dB

$D$ = hearing loss for speech in noise in dB

$G$ = gain of the hearing aid in dB

$S$ = properties of the hearing aid affecting the SRT in dB.

The diagrams illustrate that speech intelligibility in noise is determined by the S/N ratio: 10 dB more noise yields a 10-dB increase in SRT. Comparison of the two panels shows that the hearing aid is effective in amplifying the speech in quiet but does not improve speech intelligibility in noise (cf. Welzl-Müller & Sattler, 1984; Cox & Alexander, 1991). Generally, the linear and nonlinear distortion of the hearing aid, expressed in $S$, is negligible, demonstrating that it is an excellent amplifier. However, for steep audiograms with relatively large high-frequency losses, a hearing aid with a positive slope of its amplitude-frequency response, to be considered as a linear distortion of the signal, may help in bringing the high-frequency speech components above threshold, resulting in a small advantage described by a negative value of $S$ (Verschuure & van Benthem, 1992).

In a series of experiments, all with the standard test procedure mentioned above, the SRT as a function of noise level for normal and hearing-impaired subjects was studied (for a review, see Plomp, 1986). For Dutch sentences, the average monaural

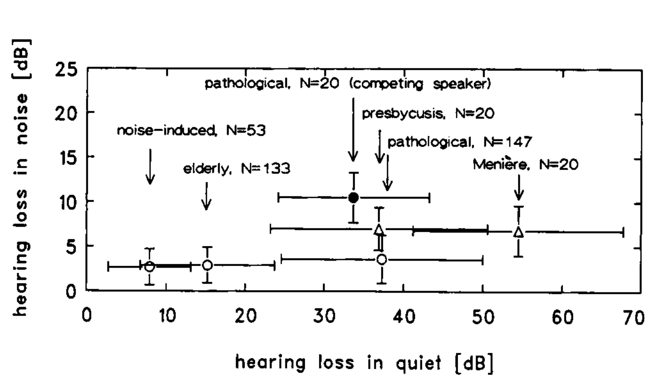


**Figure 2. Speech-reception threshold in noise as a function of the speech-reception threshold in quiet, both relative to the thresholds for normal-hearing listeners. The symbols give average values, with standard deviations, for various groups of hearing-impaired listeners. The data are adopted from Bosman (1989), triangles; Festen and Plomp (1990), solid circle; and Plomp (1986), open circles. The data were collected with the same sentence material and the same adaptive measuring procedure.**

SRT in quiet for normal-hearing listeners $L_0 = 16$ dB(A) and the SRT in noise $-\Delta L_{SN} = -5$ dB. Figure 2 summarizes data on the hearing loss for speech in noise $D$ versus the hearing loss in quiet $A + D$ for various groups of hearing-impaired subjects. Whereas my literature review of 1978 had yielded the rule-of-thumb that every 3 dB hearing loss for speech in quiet is, on the average, accompanied by 1 dB hearing loss for speech in noise, Fig. 2 gives a more differentiated picture. For instance, for a large clinical population of (potential) hearing aid users with an average hearing loss for speech in quiet of 37.3 dB, an average hearing loss in noise of 3.6 dB was found. The leftmost data point illustrates that persons with noise-induced hearing losses may have no problems in understanding speech in quiet (mean hearing loss 7.9 dB) but may have distinct problems in noise (mean hearing loss 2.8 dB). Smoorenburg (1992) has demonstrated that, even for relatively small dips induced by noise, the negative effect on speech intelligibility in noise cannot be eliminated by amplification. This illustrates that, generally, the hearing loss for speech in noise cannot be fully explained by inaudible speech components, as Zurek & Delhorne (1987), Humes, Espinoza-Varas & Watson (1988), and Dubno & Schaefer (1992) seem to suggest.

An exceptional condition is the case of a single competing speaker rather than the more or less steady-state condition of many simultaneous voices, investigated by Festen & Plomp (1990). Whereas the fluctuations in the interfering speech signal appear to help considerably for normal-hearing subjects, resulting in a 2DLSN of about 212 dB instead of 25 dB in speech noise, this 7 dB improvement is lost for hearing-impaired subjects (see the solid symbol in Fig. 2). This large difference implies that many hearing-impaired subjects have great difficulties even if only two persons are speaking at the same time. This holds particularly when the hearing-impaired person himself or herself is one of the competing speakers (Festen, 1990). This serious handicap in a one-to-one conversation has never been considered properly.

It cannot be repeated often enough that although, according to Figure 2, typical hearing losses for speech in noise are "only" 3 to 6 dB, these losses represent a very serious handicap for the hearing impaired. As the S/N ratio in many everyday situations with competing speakers is already nearly critical for normal-hearing listeners, the numerically small loss is sufficient to make speech communication very difficult or impossible for hearing-impaired listeners.

Table 1 indicates what a loss for speech in noise of 1, 3, or 6 dB means in actual situations. For example, in a critical situation in which a listener is able to understand, say, 90% of simple sentences, each increase of 1 dB in noise level (higher SRT) corresponds to a 15% decrease in the intelligibility score. Similarly, because of the interindividual differences in their speech level, in a critical situation, 8% fewer speakers can be understood for each increase of 1 dB in noise level. The last row indicates that in a situation in which 10% of elderly people has difficulties in following a speaker, this 1 dB more noise increases the percentage to 25% (cf. Plomp, 1986).

**Table 1. Effect of hearing loss for speech in noise at critical SRT values.**

| | Increase of SRT | | |
|---|---|---|---|
| | 1 dB | 3 dB | 6 dB |
| Intelligibility score for sentences | −15% | −50% | −80% |
| % of speakers understood | −8% | −25% | −50% |
| % of elderly persons handicapped[a] | 10–25% | 10–75% | 10–90% |

[a] *Values represent the effect of an extra increase in SRT for a situation in which 10% of the elderly are already unable to understand sentences correctly.*

Table 2 gives an idea of what has to be done to compensate for these hearing losses. When the listener is relatively near to the speaker, the S/N ratio can be improved with 1 dB by shortening the distance with 11% or reducing the reverberation time $T$ with 21%. When the speaker-listener distance is large, as in a hall or church, the first measure does not help, and $T$ has to be reduced by 25%. These figures illustrate that, to meet the problems of the handicapped, a considerable reduction of $T$ is required (for a more detailed discussion, see Plomp and Duquesnoy, 1980). The tables demonstrate that it is rather difficult to help the hearing impaired effectively through acoustical measures. Therefore, attempts have been made to improve the S/N ratio by signal processing in the hearing aid. I will come back to this point in the discussion.

## The Amplitude-Frequency Response

The question of which amplitude-frequency response is preferable is almost as old as the electronic

**Table 2. Acoustical measures required to compensate for the hearing losses in noise referred to in Table 1.**

| | Increase of SRT | | |
|---|---|---|---|
| | 1 dB | 3 dB | 6 dB |
| Distance speaker-listener | −11% | −29% | −50% |
| Reduction T, speaker in direct field | −21% | −50% | −75% |
| Reduction T, speaker in indirect field | −25% | −58% | −82% |

hearing aids themselves. Already half a century ago, the authors of the well-known Harvard report (Davis et al, 1947) studied this question and concluded that a positive slope of the amplitude-frequency response of about 6 dB/oct was optimal for their sample of hearing-impaired persons. Since that time an uninterrupted flow of papers has been devoted to this problem.

The primary requirement of the hearing aid is to bring the speech signal above threshold. In the case of a flat audiogram, uniform amplification over the speech range may be sufficient. However, hearing losses are generally frequency dependent, with greater hearing loss for high than for low frequencies. As was verified experimentally (eg, Skinner & Miller, 1983; Sullivan, et al, 1992), we need to bring the speech signal above threshold over its *entire* frequency range to get highest intelligibility. This suggests that in many cases the amplification factor should be frequency dependent, as the Harvard report already recommended. As the loudness discomfort level of impaired ears is approximately flat and comparable to normal hearing, the ear's dynamic range is not only strongly reduced but may differ considerably over the frequency range. These complications can explain why so many papers have been written about the selection of the optimal amplitude-frequency response of the hearing aid for each individual ear.

In the literature, we find a great number of methods for determining amplitude-frequency response based on thresholds for pure tones (eg, see Skinner et al, 1982; Byrne, 1986; Hamill & Barron, 1992). Usually, these rules are based on reasonable grounds; for example, a fixed ratio between hearing threshold and loudness discomfort level, or the most comfortable loudness level.

In my opinion, this large number of studies on the selection and fitting procedure of hearing aids leads to the following two conclusions. First, the everlasting search for a better amplitude-frequency response illustrates the fact that many subjects are not satisfied with their hearing aid(s); the explanation for this is likely to be the unavoidable noise problem rather than an inappropriate hearing aid as such. Second, the abundance of fitting procedures should be seen as evidence that the amplitude-frequency response is not as critical as the literature seems to suggest. This subject was addressed by my coworkers more recently; some of their results will be discussed below.

As was demonstrated above, the intelligibility of speech in noise is primarily determined by the S/N ratio. If this ratio were the exclusive criterion for all speech components, the precise shape of the amplitude-frequency response would not be critical as long as the components are audible and not too loud. One way to verify this is to investigate the effect of varying the slope of the amplitude-frequency response on the SRT in noise. This was studied by van Dijkhuizen, first for normal-hearing subjects and subsequently hearing-impaired subjects (van Dijkhuizen, Anema & Plomp, 1987; van Dijkhuizen, Festen & Plomp, 1989). She found that, over a wide range, the SRT in noise does not depend on the slope of the amplitude-frequency response as long as strongly negative slopes are avoided. This restriction is determined by upward spread of masking.

The question was studied more in detail by van Buuren (R.A. van Buuren, J.M. Festen, and R. Plomp, unpublished data, 1993), who adapted the spectra of both the speech and the noise carefully to each hearing-impaired listener's dynamic range. His group consisted of 26 moderately hearing-impaired subjects selected for their sloping audiograms. For each listener, the range from 5 dB above threshold to 5 dB below the loudness discomfort level for wide-band noise (shaped according to the levels found by measuring the discomfort level with 1/3 oct noise bands) was covered up with 25 different noise spectra as illustrated in Fig. 3. (Actually, the diagram represents the average spectra for the subjects involved.) For each of these 25 conditions, the SRT for sentences masked by the noise was

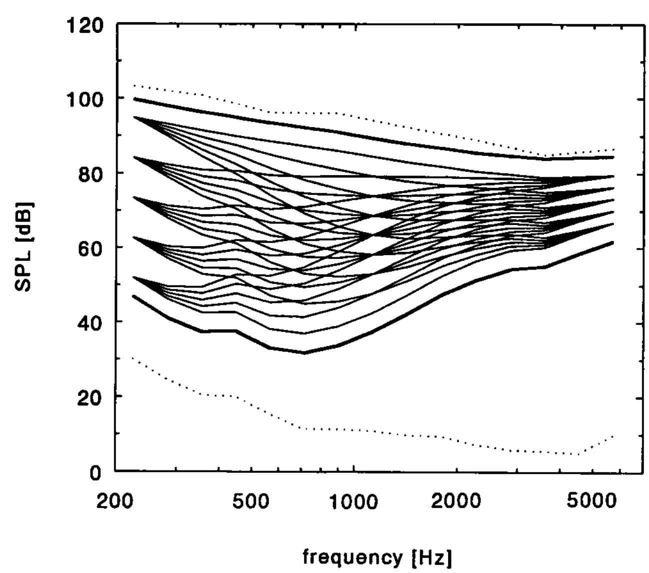


**Figure 3. Illustration of the 25 spectra of the noise applied for measuring the speech-reception threshold in noise (thin curves). In all conditions the long-term speech spectrum was shaped according to the noise spectrum. The dashed lines represent the hearing threshold and the loudness discomfort level in normal hearing, the heavy curves in impaired hearing; actually they give the average values for 26 moderately hearing-impaired listeners (R.A. van Buuren, J.M. Festen, and R. Plomp, unpublished data, 1993). All levels were measured with 1/3 oct noise bands.**

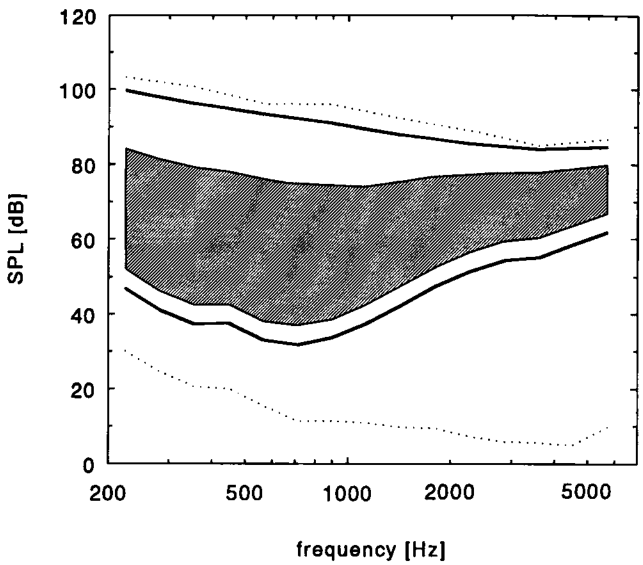


**Figure 4.** As in Fig. 3, but now the hatched area represents the spectra resulting in practically equal speech-reception thresholds in noise.

determined. The amplitude-frequency response used to obtain the noise spectra was also applied to the sentences so that their long-term average spectrum shape was equal to the noise. In this way, the potentially available area of each individual audiogram was scanned.

Figure 4 represents the same diagram, but now with the area hatched covering the spectra for which the SRT was practically the same (spread within 1.5 dB). It was verified that the hatched area, statistically based on the group of 26 listeners as a whole, can be considered as the *minimal* area for each individual subject. Comparison with Fig. 3 reveals that only spectra with strongly negative slopes and/or high levels needed a higher S/N ratio than were found for the great majority of conditions. Similar exercises in which the listener had to judge the 25 conditions in terms of comfort (sharpness, loudness, clearness, pleasantness) did not result in conflicting diagrams.

From these experiments we may conclude that, indeed, there is a relatively ample variety of amplitude-frequency responses, all satisfying the condition that the speech signal is above threshold over the entire frequency range, and over which the speech spectrum is allowed to vary without a detrimental effect on the SRT in noise. This holds for noises with the same spectrum as the long-term speech spectrum, as in the case for competing voices as the most frequent disturbing noise in everyday situations. This freedom means that the various standard hearing-aid selection and fitting procedures hold *as long as the speech level is the same as during fitting.* As this is not the case in everyday practice, the hearing aid needs a control system that keeps the speech spectrum continuously within the hatched area of Fig. 4. In other words, the hearing aid needs some sort of compression to be most effective in various acoustical conditions.

## The Compression Characteristics

The third problem may have a shorter history than the previous one, but it is nowadays the most controversial of the three. As we just saw, amplitude compression can be considered to be beneficial because it will help to keep the speech signal within the reduced range between threshold and the loudness discomfort level. It seems to me that, at least partly, the controversies have their origin in an unclear distinction between two sorts of variations of the speech signal one is listening to: differences in voice level or distance to the speaker, etc., which I will call *external* variations, and differences in the level of successive phonemes of the speech signal itself, which I will call *internal* variations. As these two types of variations differ considerably in their time constants, being large and small, respectively, each may require a different approach. Therefore, I will discuss them separately, beginning with the internal variations as the most controversial ones.

## Compression of the Internal Level Variations of Speech

The main argument for amplitude compression of variations inherent to the speech signal is *recruitment*: the loudness increases in the pathological ear much faster with sound-pressure level than in the normal ear. The idea is that, because of this phenomenon, faint speech elements are easily masked by stronger ones, so we need short-time compression to compensate for it. Usually, this is called *syllabic compression*, with compression times of less than 30 to 50 msec.

Before reporting some experimental results, the issue of whether recruitment compensation is desirable, adhered by several investigators (e.g., Villchur, 1974; Moore, 1991), should be discussed. Phenomenologically, recruitment is not pathological; it is also present near the absolute and masked threshold for normal-hearing listeners. Hellman & Zwislocki (1964) showed that, rather irrespective of the noise level, the loudness of a 1000-Hz tone in noise is almost restored at a sensation level of 100 dB. In their discussion (p. 1626), the authors found it "of considerable theoretical interest" to compare these results with the loudness increase in case of recruitment in impaired ears, studied by Miskolczy-Fodor (1960). They concluded properly that "the effect of sensorineural hearing loss on loudness seems to be

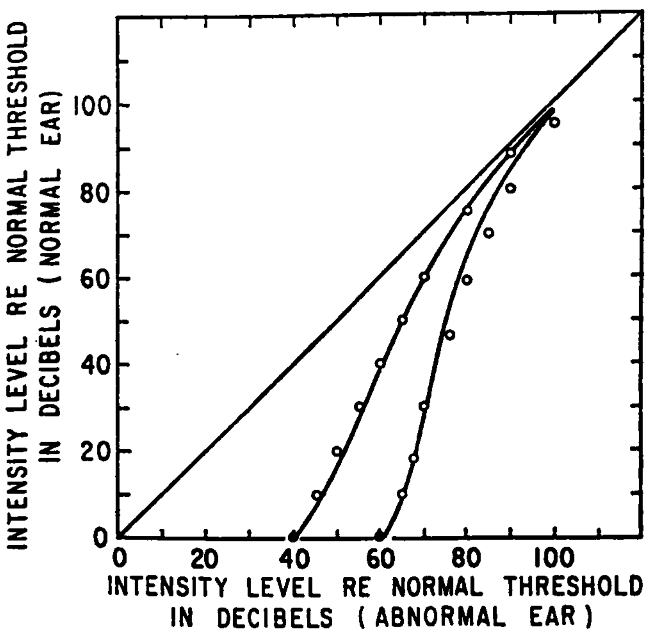


**Figure 5. Loudness-level curves of a partially masked tone, obtained by the method of adjustment, compared to loudness-balance data, collected by Miskolczy-Fodor for ears with sensorineural hearing loss exhibiting loudness recruitment (Hellman & Zwislocki, 1964).**

practically the same as that of a masking noise." This is illustrated in Fig. 5, in which the curves represent the masking-noise case and the symbols the case of sensorineural hearing loss (see also Hellman, 1988, with further experimental evidence).

It should be realized that recruitment is a positive property of the hearing process. It means that the loudness of a sound is largely independent of the presence of simultaneous sounds partially masking it. In other words, recruitment reduces the interaction between sounds, strengthening the identity of sounds. Consider, for example, the ability to identify different voices or instruments in musical compositions. In light of these facts, recruitment in the impaired ear should be considered positively, not to be destroyed by compression.

In 1967, Caraway and Carhart found that a three-channel compression system with small time constants did not yield higher speech recognition scores when the *input* levels of the compressors were adjusted to give the same sensation level at the ear (in that case, only *internal* variations were compressed), but could be very beneficial without this adjustment (now, also *external* variations were compressed). A lot of papers would not have been written if more attention had been paid to this study, with its clear distinction between internal and external variations. As the external level variations are very much slower than the internal level variations, we should not confuse them in the search for the most effective compressor time constants, as some investigators appear to do (for example, Villchur, 1989).

In view of this situation, it seemed necessary to develop new experimental evidence, in addition to the arguments in a previous paper (Plomp, 1988), to explain the negative rather than a positive effect of syllabic compression on speech perception. This was done recently by van Dijkhuizen (unpublished data, 1993) in a study of the effect of syllabic compression on the speech intelligibility in noise for normal and hearing-impaired subjects. Because we were especially interested in the effect of multichannel processing, she used 1, 2, 4, 8, or 16 frequency bands, into which the four-octave (250–4000 Hz) speech sounds were subdivided, as one parameter. The other parameter was compression ratio, equal to 1, 2, 4, and $\infty$ (ratio $n$ means that variations of $a$ dB were reduced to $a/n$ dB). The compression was applied on the speech and noise signals together. It started instantaneously, with a release time of 20 msec, at a level 30 dB below the average speech level. The speech signal in each frequency channel was always presented halfway between threshold and loudness discomfort level for all hearing-impaired subjects.

Figure 6 gives the average intelligibility scores of 16 normal-hearing listeners for sentences presented at an S/N ratio of 0 dB. Similarly, Fig. 7 gives the average intelligibility scores of 16 moderately im-

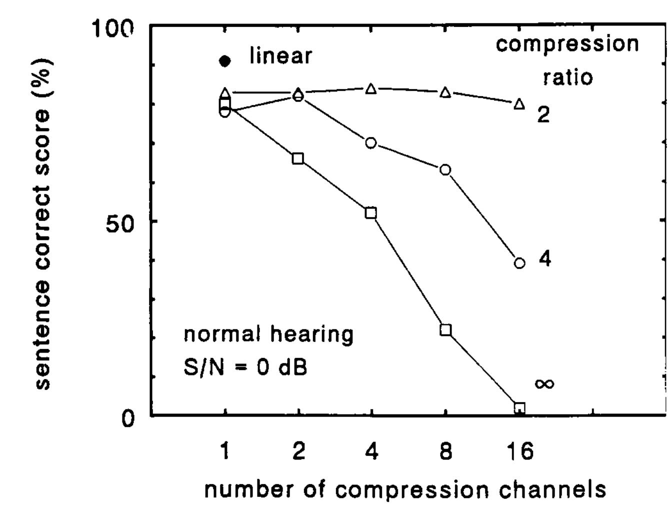


**Figure 6. Intelligibility score as a function of the number of channels in which the range 250 to 4000 Hz was divided, with compression ratio as the parameter. The data points give mean values for 10 normal-hearing listeners, with the sentences presented at an S/N ratio of 0 dB (J.N. van Dijkhuizen, unpublished data, 1993).**

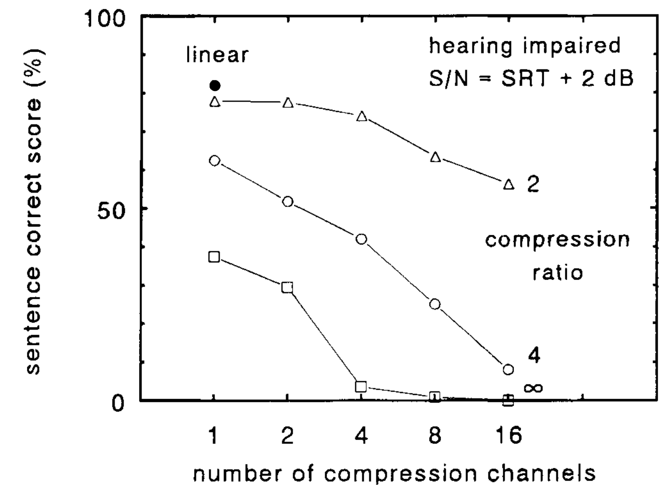


**Figure 7. As in Fig. 6, but now for 12 hearing-impaired listeners, with the speech spectrum adapted to the bisector of the (frequency-dependent) dynamic range (J.N. van Dijkhuizen, unpublished data, 1993).**

paired listeners at an S/N ratio individually adjusted to 2 dB above their SRT for wide-band speech. Comparable results, but with an unavoidable strong ceiling effect for the more favorable conditions, were obtained for sentences presented in quiet.

The results are quite unequivocal. In the first place they show that, *in none of the compression conditions* is the intelligibility score higher than for the linear condition. The score reduces monotonically as a function of the number of frequency channels into which the signal was divided. Apparently, 2 to 4 channels in case of a low compression ratio of 2 does not harm the signal very much, whereas a ratio of 4 is detrimental in all cases. We may conclude that, for a speech signal presented appropriately within the hatched area of Fig. 4, syllabic compression does not improve intelligibility. Of course, the negative effect will become smaller for increasing compression times, less effective in suppressing relatively strong phonemes than applied in the experiment.

This conclusion confirms my reasoning in a previous paper (Plomp, 1988) in which I used the modulation-transfer function to explain why the introduction of multichannel compression with small time constants has not been successful. That paper contains also further circumstantial evidence and arguments against syllabic compression which I will not repeat here.

The fact that short-time compression of *internal* speech variations does not improve intelligibility does not mean that this also holds for *external* speech variations. Therefore, let us now shift our attention to the latter aspect.

## Compression of the External Level Variations of Speech

The reduced dynamic range between threshold and the loudness discomfort level means that many hearing-impaired listeners are highly susceptible to differences in speech level. Normal-hearing listeners have no problems in understanding speech over a wide range from, say, 30 dB(A) up to 80 dB(A). For many hearing-impaired listeners with a hearing aid amplification optimally adjusted for the most common speech level of 55 to 60 dB(A), however, a level of 30 dB(A) will be below threshold, whereas a level of 80 dB(A) will be unacceptably loud. Hence, an amplification factor that adapts itself automatically to the sound level is very useful. (The problem of high-level impact sounds, as from slamming doors, I leave out of consideration; a peak limiter is the best remedy.)

Such a wide-band automatic gain control (AGC) as a compressor of external variations of the speech signal is most effective with flat audiograms. In that case a *moderate* compression ratio, equal over the entire speech-frequency range is sufficient to bring faint speech above threshold and to keep loud speech below the discomfort level. The situation is much less favorable, however, for frequency-dependent hearing losses. As in Fig. 4, the dynamic range is usually much larger at low than at high frequencies. Theoretically, AGC with a *large* compression ratio may seem the best solution as in that case the speech signal will always be presented at the optimal level. However, this is unacceptable from a practical point of view: the hearing-impaired person would be exposed continuously to the same sound level, whether there is a speech signal or not. Ideally, the gradations should be as similar as possible as in normal hearing: low-level sound should be just audible and high-level sounds should be loud. With reference to the hatched area of Fig. 4, with a larger dynamic range for low than for high frequencies, this means that the spectrum has to vary as a function of level: at low levels the speech spectrum should agree with the lower bound of the hatched area, at high levels with the upper bound. Otherwise, the high-frequency components may become either inaudible or too loud.

Frequency-dependent AGC can be realized by subdividing the frequency range into a number of channels, each with a compression ratio optimized to meet conditions such that the full dynamic range is used. To preserve the spectrotemporal variations of the speech signal as well as possible, the AGC should be controlled by relatively large time constants. However, this has the practical disadvantage that the system needs some time to adjust itself, so that the first words of a conversation may be too faint or too loud. The consequence of small time constants is that the speech spectrum varies continuously in time, and may reduce intelligibility, as in the case of syllabic compression discussed above. This effect increases with increasing number of frequency channels (see Fig. 7).

Current research by R. Drullman (unpublished data, 1993) shows that, independent of the number of frequency channels, the amplitude variations above 4 Hz have to be preserved untouched to keep the SRT of speech in noise at the same level as for the undisturbed signal. This suggests that frequency-dependent dynamic gain control will not reduce speech intelligibility if this control is slower than 4 Hz. This was verified more directly by van Dijkhuizen, both for normal and impaired hearing (van Dijkhuizen et al, 1987, 1989). In one of her experiments, she studied with hearing-impaired listeners the effect on the SRT in noise of a transition of the slope of the speech spectrum halfway through the sentence up to a range from +10 dB/oct and −10 dB/oct, and vice versa, relative to the bisector of the subject's dynamic range. The results showed that the SRT of speech in noise is practically constant when the slope of the amplitude-frequency response is varied with a transition time as short as 250 msec, as long as strong negative slopes are avoided.

## Discussion

The investigations reported above indicate the direction in which, in my opinion, the development of effective hearing aids should go. As the main problem of the hearing impaired is the difficulty of understanding speech in the presence of other (speech) sounds, it seems justified to give top priority to study how the S/N ratio can be improved.

Various ways have been explored, but without much success. Several hearing aid designs have been proposed and marketed with the pretense that they selectively amplify the speech signal. These hearing aids have an amplitude-frequency response that adapts automatically to the spectrum of the incoming sound. It is clear that in situations in which the speech signal is partly masked by strong low-frequency noise, intelligibility might improve by attenuating low-frequency sounds, with the positive effect of reduction of upward spread of masking. Laboratory experiments (van Dijkhuizen, Festen & Plomp, 1991; Rankovic, Freyman & Zurek, 1992) as well as clinical tests with commercially available hearing aids (e.g., Ono, Kanzaki & Mizoi, 1983; Stein & Dempesy-Hart, 1984; Van Tasell, Larsen & Fabry, 1988; Kiessling & Steffens, 1991) have demonstrated the potential benefits of multichannel amplification automatically suppressing strong, low-frequency sounds. However, the improvement of speech-reception threshold is modest. Expectations of the effectiveness in everyday listening situations have not been fulfilled. Because speech from competing talkers represents the most frequent "noise" condition, selective filtering will rarely, if ever, improve the S/N ratio. Hence, the disappointment of hearing-aid users is not surprising (cf. Tyler & Kuk, 1989; Kuk, Tyler & Mims, 1990; Sammeth & Ochs, 1991). Nevertheless, the modest effect of selective noise suppression should not be neglected.

Whereas the previous approach cannot discriminate voices, a much more sophisticated approach would be to separate the speech signal from other interfering voices on the basis of characteristic differences between their sound waves. A first attempt is to separate two voices by making use of differences in intonation, expressed in their fundamental frequencies $F_0$. Stubbs and Summerfield (1990, including also a review of earlier research) tried out two algorithms, but obtained no improvement in intelligibility for hearing-impaired listeners. It seems to me that this is a very problematic approach. First, the fact that a group of normal-hearing listeners had no problems in separating two equal-intensity voices from *different* talkers, whereas most of them had great difficulties in following each of two superimposed conversations recorded from the *same* talker (Festen & Plomp, 1990), indicates that the human hearing uses many more characteristic differences between voices than just intonation. Second, we should realize that our remarkable capacity to separate voices is strongly determined by cognitive factors; it will be highly difficult, or impossible, to simulate this in a hearing aid. Therefore, I am not optimistic about the perspective of separating voices instrumentally.

A third approach is to make use of directional differences between the speaker and the interfering sound source. Peterson (1987) studied a method in which the contribution of the noise is adaptively minimized by using two microphones. Such a system is quite effective in anechoic conditions, but is very susceptible to reverberation. In my opinion, a more promising approach is a highly directional device by means of an array of microphones as investigated by Soede (Soede, Bilsen & Berkhout, 1993). He developed practically applicable arrays giving an S/N ratio improvement of about 7 dB for hearing-impaired listeners in a typically everyday listening condition.

In view of the great difficulties of improving S/N ratio (see also Lim, 1986), much attention should be given to develop hearing aids that present the speech optimally to the impaired ear. This means that, irrespective of the acoustical conditions, the various frequency components of the speech signal have to be within the hatched range given in Fig. 4. Our finding that, as long as we remain within an ample range between threshold and the loudness discomfort level, speech intelligibility can be considered to be independent of the (long-term) speech spectrum, is much encouraging. The results of experiments in which a limited set of prescription rules were compared (Byrne, 1986, 1992; Sullivan et al, 1988; Leijon et al, 1991; Hamill & Barron, 1992) are consistent with this conclusion.

Multichannel hearing aids with frequency-dependent AGC seem to be most appropriate to meet this requirement. If well designed and fitted, they are the best guarantee to maintain optimal speech intelligibility under different conditions, without any need for the user to readjust the aid. By splitting the speech range into two or more frequency channels, the compression ratio can be adjusted to make optimal use of the individual's frequency-dependent dynamic range. The possibility of reducing low-frequency masking noise selectively is an additional advantage of multichannel hearing aids. We should not forget, however, that even the most favorable amplitude-frequency response is generally not able to eliminate the hearing loss for speech in noise (the D term in Eq. 1). This makes any prediction of speech intelligibility on the basis of the Articulation Index (cf. Kamm, Dirks & Carterette, 1982; Pavlovic, Studebaker & Sherbecoe, 1986; Humes et al, 1988; Dubno & Schaefer, 1992; Rankovic et al, 1992) problematic. Experimental results of Byrne (1992) support this view.

The results represented in Figs. 6 and 7 indicate that short-time compression does not improve speech intelligibility in noise and that the negative effect increases with the number of channels. This should be kept in mind in searching for the optimal specification of multichannel compression circuits. However, the detrimental effects of short compression times on speech intelligibility in noise are small as long as both the compression ratio and the number of channels are small. This can explain why, even with syllabic compression, carefully fitted two-channel amplifiers gave better results than wide-band amplification (cf. Laurence, Moore & Glasberg, 1983; Moore, Laurence & Wright, 1985; Moore & Glasberg, 1988; Benson, Clark & Johnson, 1992; Gordon-Salant & Sherlock, 1992; Moore et al, 1992), whereas the benefit of syllabic compression in single-band hearing aids appears to be questionable (Dreschler, Eberhardt & Melk, 1984; Peterson, Feeney & Yantis, 1990; Maré, Dreschler & Verschuure, 1992). The better chance of the two-channel aid to keep the speech signal above the hearing threshold apparently overcompensated the negative effect of the short compressor times.

To minimize the negative effects of compression, time constants should preferably be substantially larger than 20 msec. The experiments by van Dijkhuizen et al (1989) with a four-channel amplifier indicated that automatic gain control readjusting the amplitude-frequency response within 250 msec does not affect significantly speech intelligibility in noise. It seems reasonable that the best results are obtained with compressor constants not much lower than this value.

To control the complexity of the hearing aid, a minimum number of channels should be used. Although we have only experimental evidence with small time constants (Fig. 7), we may expect that, generally, a large number of channels is unfavorable for speech intelligibility. In view of the fact that most audiograms are rather smooth, we may assume that four channels is a reasonable maximum.

The optimal combination of number of frequency channels (2–4), compression ratio (not much larger than 2), and compression constants (say, between 100 and 250 msec) cannot be derived reliably from laboratory experiments as described in this paper, but has to be found out by means of clinical tests and field studies. We may expect that this optimum is not very critical, allowing designs with different combinations to be equally satisfactorily.

The optimum will also depend upon the success in developing circuits that keep the sound at a low level as long as no speech signal is present. A major problem of the application of compression in hearing aids is that the gain is maximal in a low-noise condition without a speech signal. This holds particularly for automatic systems in which the volume is not controlled manually by the user. Ways have to be found to reduce the annoyance of the permanent noise as much as possible. Experiments by van Dijkhuizen, Festen & Plomp (1993) with an algorithm reducing the gain as long as speech-like level fluctuations are absent have shown promising results.

Fortunately, since the introduction of subminiature digital circuitry, the implementation of these design directives in an effective hearing aid is not the bottleneck anymore.

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