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Design and evaluation of a two-channel compression hearing aid*

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Abstract—The design of a two-channel compression hearing aid for persons with moderate sensorineural hearing losses with recruitment is described. The aid applies slow-acting automatic gain control (AGC) to the whole signal, and then splits the signal into two bands, with separate fast-acting (syllabic) AGC in each band. Trials evaluating the aid have shown that it allows speech in quiet to be understood over a wide range of sound levels without any need to adjust the controls on the aid. It also gives speech intelligibility in noise superior to that allowed by a comparable linear (non-compression) aid, a comparable single-channel compression aid, and by unaided listening. Pilot experiments comparing two different methods for fitting the aid suggest that fitting using speech as the test signal is superior to fitting using narrow band tonal signals.

INTRODUCTION

Patients with sensorineural hearing loss suffer not just from a reduced ability to detect low-intensity sounds, but also from difficulties in the discrimination of sounds which are well above threshold, and therefore easily audible. Perhaps the most common complaint is of difficulty in understanding speech in noisy situations. Hearing aids have usually been found to be of limited benefit when listening in noise, and indeed often make the situation worse: e.g., Duquesnoy and Plomp (8); Welzl-Muller and Sattler (24); for reviews see Plomp (19,20); Duquesnoy and Plomp (8); and Laurence, Moore and Glasberg (12).

Several factors may contribute to the failure of hearing aids to improve the intelligibility of speech in noise. The first few sections of this paper review what are probably the more important ones.

The Problem of Dynamic Range

Hearing-impaired persons with loudness recruitment have a reduced dynamic range between threshold and discomfort. Furthermore, speech that is presented just above threshold is generally not intelligible. As a result, the effective dynamic range for speech (from threshold of intelligiblity to discomfort) is less than the dynamic range for tones. The dynamic range may also vary markedly with frequency. Hearing aids should process speech so that all of the important elements of the speech are above threshold but below the level producing discomfort. Although this requirement appears simple, it is not easily accomplished because of the complex way in which the short-term speech spectrum varies with time.

There are two reasons why reduced dynamic range creates difficulties for the hearing-impaired person. Firstly, the overall level of speech may vary over a range of at least 30 dB from one situation to another (18). Slow-acting automatic gain control (AGC) can be used to deal with this problem. Secondly, even for speech at a constant average level, the levels of individual acoustic elements of the speech may vary over a range of 30 dB (13). In general, the acoustic

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correlates of consonants are less intense than those of vowels, so that a hearing-impaired person may be able to hear the vowels, but not all of the consonants. To deal with this, fast-acting AGC or "syllabic" compression is needed, preferably acting independently in different frequency bands so as to allow for the variation with frequency of the listener's dynamic range.

Unfortunately, multiband compression introduces a series of problems of its own. Firstly, it is inherently nonlinear, introducing harmonic and intermodulation distortion. The distortion can become severe when the time constants of the compression circuitry are less than several periods of the lowest frequency being dealt with. Secondly, multiband compression produces a "smoothing" or flattening of the sound spectrum. This may make it more difficult for the listener to pick out the salient features of the spectrum such as the formant peaks, a difficulty which is compounded by the reduced frequency selectivity that accompanies cochlear hearing loss: e.g., Glasberg and Moore (9). Finally, multiband compression can introduce spurious changes in the spectrum and temporal envelope of the sound. There is good evidence that speech perception depends more on changes in the sound spectrum with time than on steady-state spectra: Summerfield and Assmann (21). Hence, even quite subtle changes introduced by the compression may have deleterious effects on speech perception.

These problems may account for the fact that multiband compression has not generally been found to give improved speech intelligibility in noise in comparison to linear amplification accompanied by frequency shaping: e.g., Abramovitz (1); Lippmann, Braida and Durlach (14); and Walker, Byrne and Dillon (23). The arguments presented above suggest that there may be distinct disadvantages in having too many channels in a multiband aid. The problems associated with spectral flattening and the introduction of spurious spectral changes can be minimized by using only a small number of channels, probably five or less in our view.

The Problem of Poor Suprathreshold Discrimination

Persons with sensorineural hearing losses have a reduced ability to discriminate sounds that are well above threshold. Deficits have been demonstrated for almost all stimulus dimensions that have been examined, with the exception of intensity discrimination. In particular, frequency selectivity, frequency discrimination, temporal resolution, and binaural processing are all impaired. These psychoacoustic deficits are a major cause of the difficulties of the person with impaired hearing in understanding speech, and they are generally not corrected by hearing aids. For reviews, see Dreschler and Plomp (5,7); Plomp (20); and Moore and Glasberg (16).

The Problem of Distortion in Hearing Aids

Hearing aids introduce various types of distortion into the signal (e.g., harmonic and inter-modulation distortion, limited frequency range, and irregular frequency response) and this has a deleterious effect on speech intelligibility. Although each type of distortion on its own may have only a small effect on speech intelligibility, taken together they can have a significant deleterious effect. Plomp (19,20) has suggested that the distortion introduced by a hearing aid (in addition to the distortion produced by the hearing impairment) can be characterized by a parameter S, in dB, which accounts for all the properties of a hearing aid which may affect the speech-to-noise ratio required for 50 percent sentence intelligibility. For high noise levels, S represents the amount by which this ratio is increased (i.e., performance is made worse) by wearing a hearing aid. Duquesnoy and Plomp (8) found that S was typically between 1 and 3 dB.

THE DESIGN OF A TWO-CHANNEL COMPRESSION HEARING AID

We have been working with a two-channel compression aid, built by R.F. Laurence, that was intended to overcome some of the difficulties described above, and in particular those difficulties associated with the limited dynamic range of most hearing-impaired persons. In the following paragraphs the design of the aid will be outlined. Letters inserted in brackets will indicate which of the three factors is alleviated by a given design feature: dynamic range [*DR*]; suprathreshold discrimination [*SD*]; and distortion [*DIST*]. A block diagram of the aid is given in **Figure 1**. In the following paragraph,

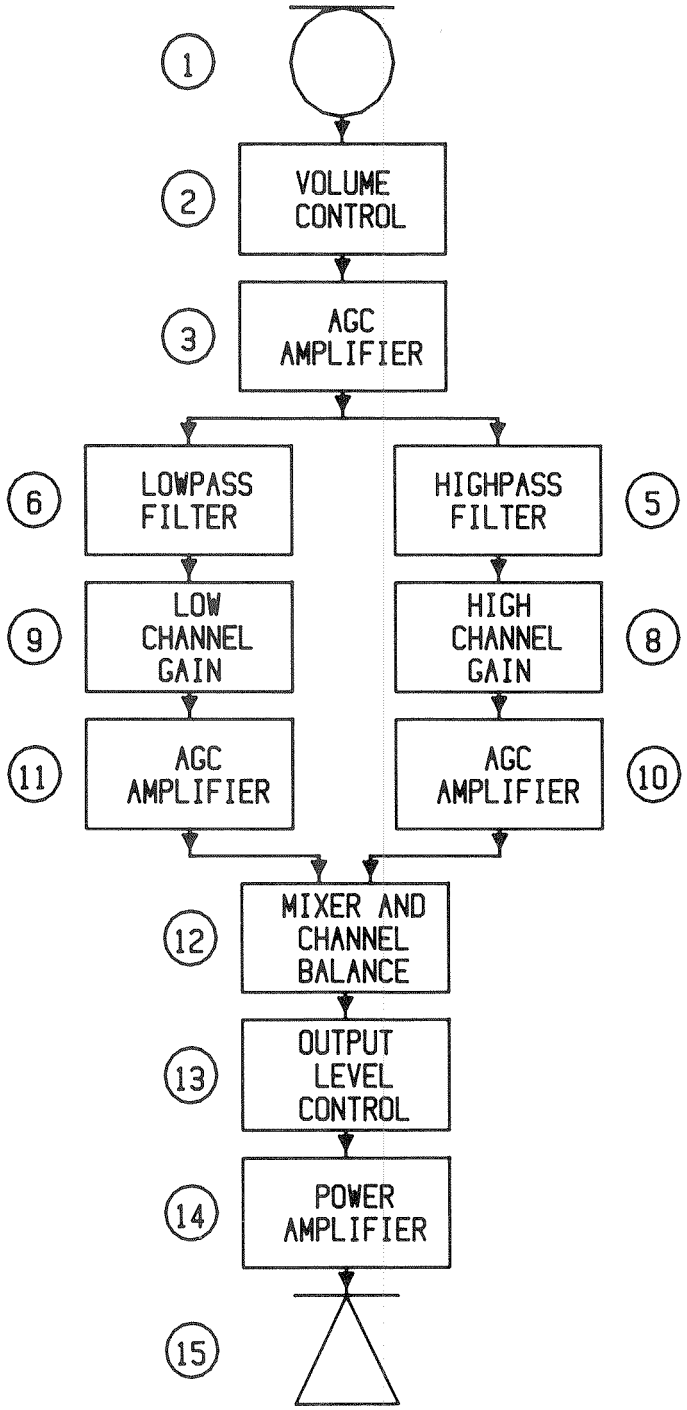


Figure 1.
A block diagram of the two-channel compression aid. Note that there are no *Blocks 4* and *7* in this diagram.

blocks within the figure are referred to by number.

Block 1 is a directional microphone which faces in a forward direction. When the wearer looks at the person they wish to hear, the speech from that person is enhanced relative to the background noise [*SD*]. *Block 2* is a user-accessible volume control normally used only to reduce distortion for high input sound levels [*DIST*]. *Block 3* is a relatively slow-acting automatic-gain-control (AGC) amplifier. The attack time is 5 ms and the recovery time is 300 ms (measured according to IEC 118-2:1979). The compression threshold is 75 dB SPL, and compression is applied strongly above that level (compression limiting). This amplifier is intended to compensate for variations in the overall level of speech from one situation to another, delivering the speech at a comfortable level regardless of input level [*DR*].

Blocks 5 and *6* are filters which split the speech into a band above 1500 Hz and a band below 1500 Hz. This allows the possibility of applying different amounts of compression at high and low frequencies, compensating for variations in dynamic range with frequency, and preventing intense low frequencies (mainly associated with vowels) from affecting the gain at high frequencies (mainly affecting the audibility of consonants) [*DR*]. *Blocks 8* and *9* are controls that adjust the gain and amount of compression in each channel to suit the individual patient. They can usually be set so all important components of speech are presented within the dynamic range between threshold and discomfort [*DR*].

Blocks 10 and *11* are fast-acting AGC amplifiers which compensate for differences in the levels of individual speech sounds within speech of a given average level. The attack and recovery times are 2 ms and 50 ms in the low-frequency channel and 2 ms and 10 ms in the high-frequency channel. These AGC amplifiers help to overcome forward-masking effects, and ensure that low-intensity consonants can be heard following high-intensity vowels [*DR* and *SD*]. They have to compress over only a limited range of levels, since the slow-acting AGC (*Block 3*) serves to keep the overall level of the speech within a relatively narrow range. This reduces harmonic and intermodulation distortion, and reduces spurious spectral changes introduced by the compression [*DIST*].

Block 12 is a mixer which controls the balance between the two channels. This is adjustable to suit the patient, and can compensate for individual differences in the shape of the threshold-of-discomfort curve as a function of frequency [*DR*]. *Block 13* is a potentiometer used to set the maximum output level of the aid. It is adjusted so that speech is easily

audible but not uncomfortably loud with high-level inputs [*DR*]. *Block 14* is a high-quality power amplifier that applies heavy damping to the receiver (*Block 15*) to reduce distortion and resonances introduced by the tubing/earmold/ear canal. Both the amplifier and receiver are wideband high-fidelity devices, and all circuitry is designed to minimize distortion [*DIST*]. The earhook incorporates an acoustic resistance to further reduce resonances, and the tubing takes the form of a "Libby Horn" to extend the high-frequency response [*DIST*].

The aid was built as a wearable behind-the-ear device. The controls at *Blocks 8, 9, 12,* and *13* allow the aid to be adjusted to suit a rather wide range of degree and slope of hearing loss. Furthermore, the adjustments can be made while the patient is actually wearing the aid. Hence, uncertainties as to what the appropriate insertion gain should be as a function of frequency, and as to what the insertion gain actually is, become largely irrelevant. Procedures for fitting the aid are discussed elsewhere in this paper.

EVALUATION OF THE TWO-CHANNEL COMPRESSION AID

In all of the trials conducted to evaluate the performance of the aid, subjects were given at least 2 weeks of experience wearing the aids in everyday life before formal testing began. All of the trials made use of the BKB sentence lists: Bench and Bamford (2). Each list contains 16 simple sentences which are scored by key word, with 3 or 4 key words per sentence and 50 per list. For assessing speech intelligibility in quiet, we measured the percentage correct of key words with the speech at a fixed level. For assessing speech intelligibility in noise, we estimated the level of the speech required to achieve 50 percent correct key words in a background noise of fixed level, using an adaptive procedure described in detail in Laurence, Moore and Glasberg (12). That level will be referred to as the Speech Reception Threshold (SRT)*. Speech levels will be specified as the levels of the peaks of the speech as read on a VU meter. The background noise had a spectrum shaped like the long-term average spectrum of speech. Noise levels will be specified as their root-mean-square (rms) values. We will present SRTs as speech-to-noise ratios (peak-to-rms) in dB.

All of our trials have shown that, for patients with moderate hearing losses, the aid allowed excellent understanding of speech in quiet over a wide range of sound levels, without any need to adjust the controls on the aid. While this is a notable benefit, and something that is difficult to achieve without AGC, our main concern is with the performance of the aid in noisy situations. Hence, this aspect will be emphasized in describing the results.

Comparison of the Two-Channel Compression Aid with a Matched Linear Aid

In this trial, the two-channel compression aid was compared with a linear aid which was similar in every respect except that the compression circuits were disabled: Laurence, Moore and Glasberg (12). For the linear aid, the two channels could be used to adjust the frequency response. The volume control [*Block 2*] was set to maximum and the output level control [*Block 13*] was set to give a comfortable listening level with an input speech level of 70 dB SPL. Eight subjects with bilateral moderate cochlear hearing losses were used; pure tone thresholds in the better ear, averaged across the frequencies 0.5, 1.0 and 2.0 kHz, ranged from 44 to 63 dB HL. All subjects were fitted binaurally, and they were tested using each ear separately (the aid in the unused ear being fitted but turned off) and using both ears together.

The speech was always presented via a loudspeaker directly in front of the subject. The noise was presented either via the same loudspeaker, at an RMS level of 65 dB SPL, or via two loudspeakers directly to the left and right of the listener's head. We will refer to these two conditions as coincident and separated, respectively. In the latter case two independent noise sources were used, and the level of each source was 62 dB SPL; the total noise level, measured in the absence of the listener at the point corresponding to the center of the listener's head, was 65 dB SPL.

Speech reception thresholds (SRTs) in noise, averaged across all conditions, are shown for each

*The term Speech Reception Threshold (SRT) is used differently in the United States and Europe. The European usage is as described above; it is the level at which 50 percent correct responses are obtained in a speech identification task. The American usage of SRT refers to the level at which spondaic words (bisyllabic words with stress on both syllables) are just detectable.

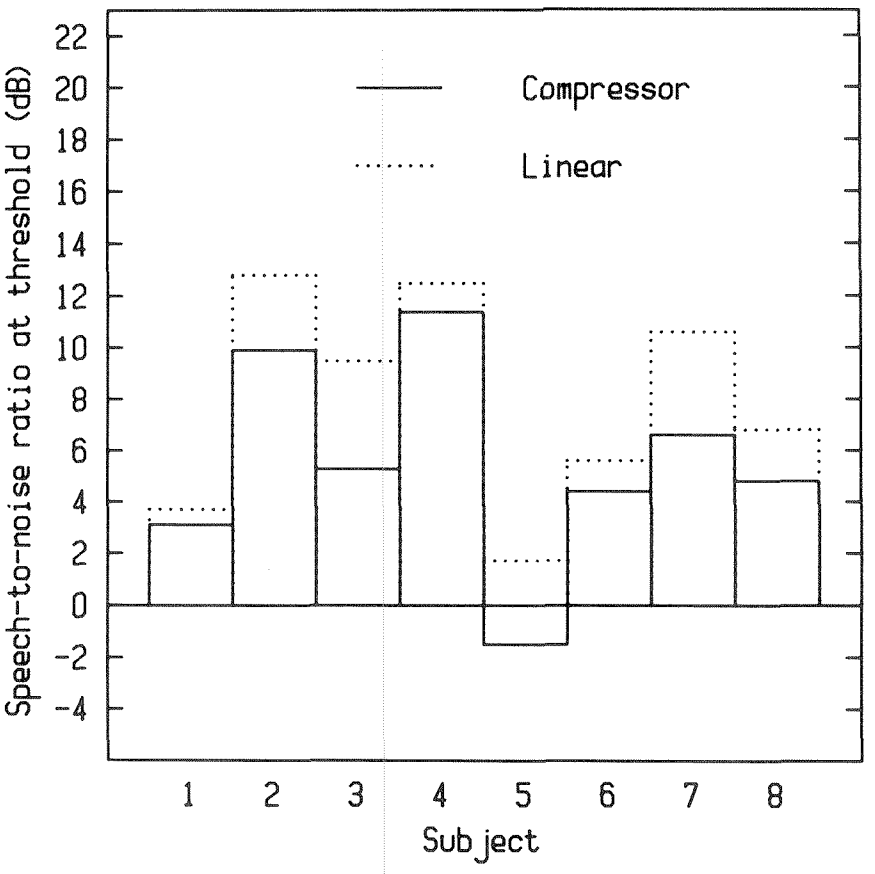


Figure 2.
Speech reception thresholds (SRTs) in noise at 65 dB SPL, expressed as speech-to-noise ratio in dB, for the two-channel compression aid (solid lines) and a matched linear aid (dotted lines), shown for each of eight subjects.

subject in **Figure 2**. Overall, the compressor aids gave significantly ($p<0.01$) better scores than the linear aids, by an average of 2.4 dB. Although this difference appears small, it is equivalent to quite a large change in intelligibility, since, for the materials we used, intelligibility changes by about 11 percent for each 1 dB change in speech-to-noise ratio (12). **Table 1** gives SRTs averaged across subjects for the coincident and separated conditions. SRTs were significantly ($p<.001$) lower (by an aveage of 4.5 dB) when the noise and speech were separated than when they were coincident. However, this difference was not greater when listening with both ears than when listening with the better ear. Thus the effect is probably not analogous to the binaural masking level difference; rather it can be attributed mainly to the directional microphones in the aids. Subjects did better when listening with two ears than with their better ear, but the improvement was a general one, not specific to conditions where the speech and noise were spatially separated.

As well as measuring the performance of the aids in laboratory conditions, we also gave our subjects questionnaires about their experience with the aids in everyday life. The compressor aids were preferred over the linear aids in most situations, but particularly for situations with a moderate amount of background noise (three or four people talking, street noise, and at meal times).

Table 1.
Speech reception thresholds (SRTs), averaged across subjects and expressed as speech-to-noise ratios in dB, for two types of aids (two-channel compression and linear) and two test conditions (speech and noise coincident and speech and noise spatially separated). Note that lower numbers indicate better performance.

	Masker location		
Aid	**Coincident**	**Separated**	**Mean**
Compressor	7.9	3.1	5.5
Linear	9.9	5.8	7.85
Mean	8.9	4.45	

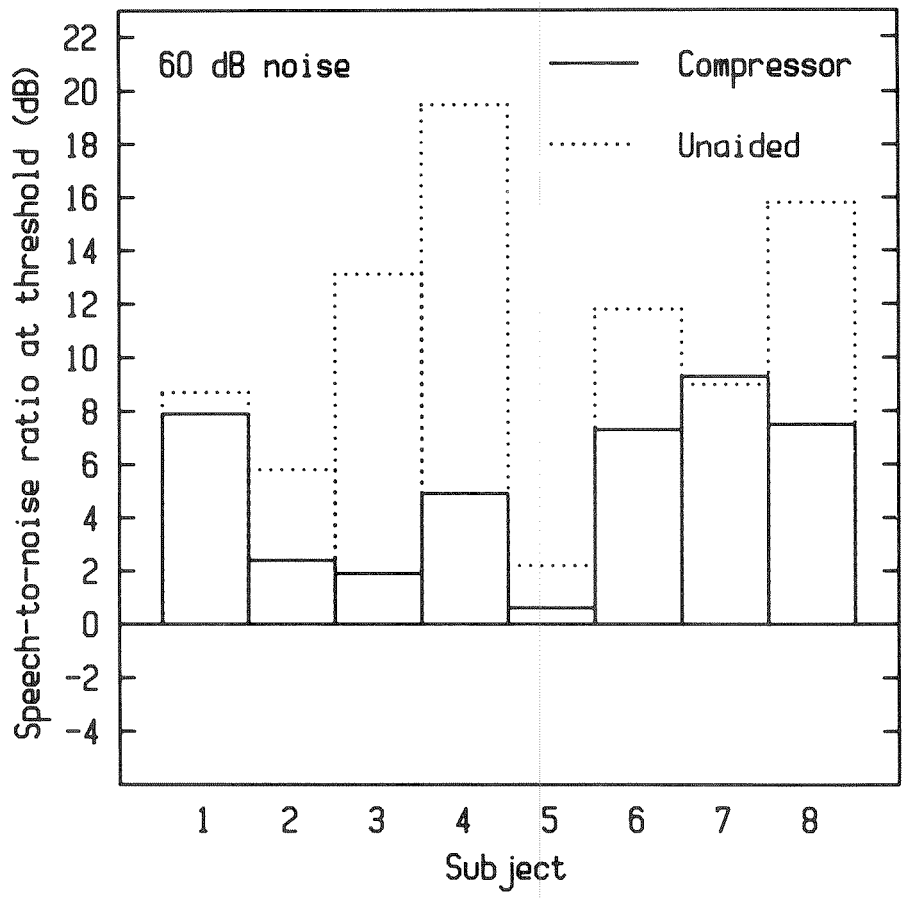


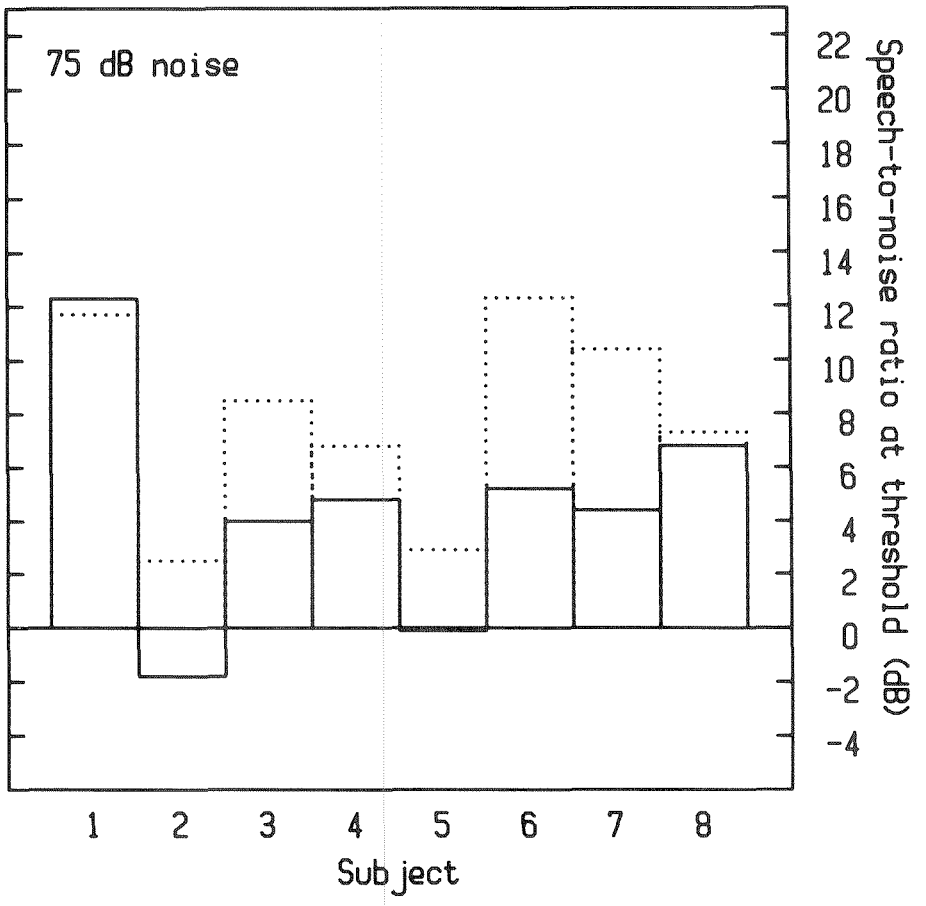
Figure 3. SRTs for the two-channel compression aid (solid lines) and unaided listening (dotted lines) for two noise levels: 60 dB SPL (left) and 75 dB SPL (right). Results for eight subjects are shown.

In summary, the two-channel compression aid gave better speech intelligibility in noise than a comparable linear aid, and was preferred for everyday listening. The directional microphone gave significant benefits under conditions where the speech and noise were spatially separated.

Comparison of the Two-Channel Aid with Unaided Listening

As mentioned earlier, hearing aids have not generally been found to improve the intelligibility of speech in noise when the noise level is sufficient to raise the SRT significantly above that measured in quiet. In our second study, we sought to determine whether the two-channel compression aid would produce improvements relative to unaided listening: Moore, Laurence and Wright (15). In this study, both speech and noise were delivered from a single loudspeaker directly in front of the subject, so we were not taking advantage of the properties of the directional microphones. Eight subjects with moderate bilateral sensorineural losses were used (different subjects from those in the first trial); absolute thresholds in the better ear, averaged for the frequencies 0.5, 1.0 and 2.0 kHz, ranged from 28 dB to 65 dB HL. SRTs were measured for unaided listening, using both ears, and listening binaurally aided with two-channel compression hearing aids. Two noise levels were used, 60 and 75 dB SPL. The higher noise level was sufficient to raise the SRT well above that measured in quiet for all subjects.

Figure 3 summarizes the results. Seven of the eight subjects gave a lower SRT (better performance) in the aided condition for at least one of the two noise levels. The mean improvement was 5.5 dB at the lower noise level and 3.5 dB at the higher noise level. The improvement was statistically significant ($p<0.01$) at both noise levels.



Comparison of the Two-Channel Compression Aid with a Single-Channel Compression Aid

In our third trial, performance was compared for the two-channel compression aid, a single-channel compression aid, and unaided listening: Moore and Glasberg (17). The single-channel aid incorporated slow-acting AGC operating on the whole speech signal. It also had two channels, but these were used only for frequency response shaping, and not for additional compression. In other respects the two aids were chosen to be as similar as possible. Again, eight subjects with moderate bilateral sensorineural losses were used (different subjects from those of the previous trials); absolute thresholds in the better ear, averaged over the frequencies 0.5, 1.0 and 2.0 kHz, ranged from 30 to 72 dB HL. The speech and noise were presented via a single loudspeaker directly in front of the subject, and noise levels of 60 and 75 dB SPL were used.

The SRTs for each condition are given for each subject in **Figure 4**. The two-channel compression aid gave significantly lower SRTs than both the single-channel aid ($p<0.01$) and unaided listening ($p<0.01$). The single-channel aid did not give significantly lower SRTs than unaided listening. A similar result was found by Dreschler, Eberhardt, and Melk (6). For the higher noise level, the SRT for the two-channel compression aid was, on average, 2.4 dB lower than that for unaided listening, and 2.9 dB lower than that for the single-channel aid.

Subjects were also given questionnaires asking about their experience with the aids in everyday listening situations. The two-channel compression aid was preferred in all of the listening situations mentioned in the questionnaires, particularly large differences being obtained for situations where difficulty was encountered (conversation in a party of a dozen or more people, conversation in a car and listening to the television). Overall, the results suggest that two-channel syllabic compression, combined with slow-acting AGC operating on the whole

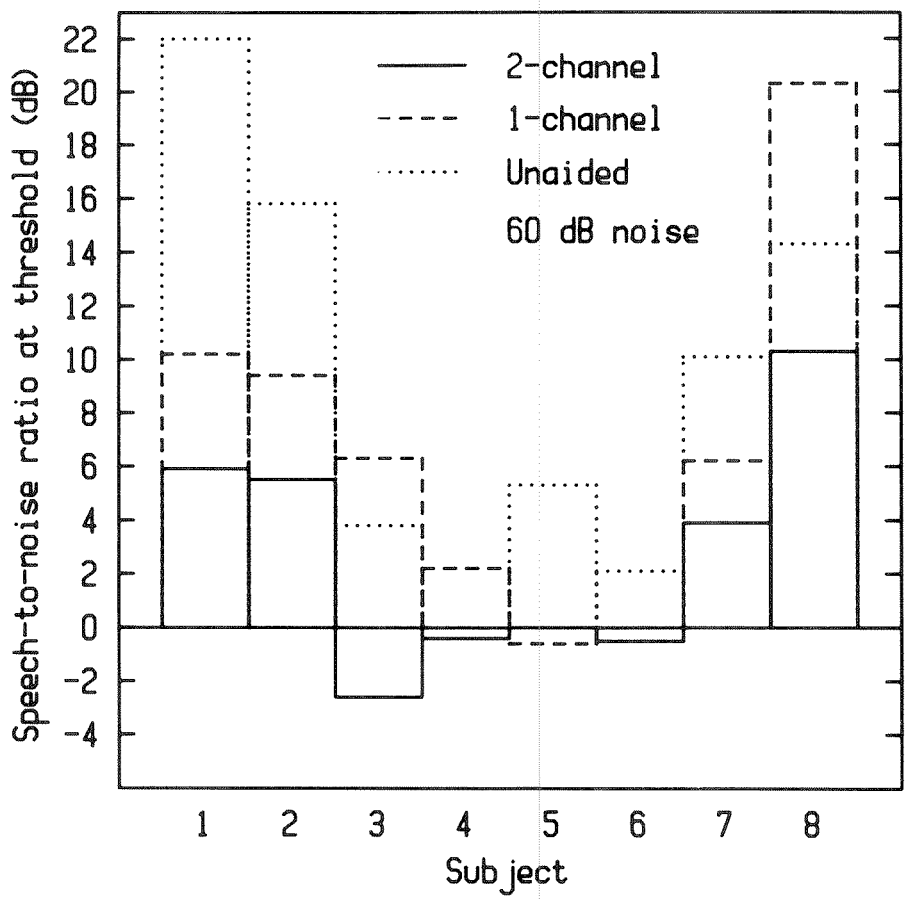


Figure 4.
SRTs for the two-channel aid (solid lines), a matched single-channel aid (dashed lines) and unaided listening (dotted lines) for two noise levels: 60 dB SPL (left) and 75 dB SPL (right). Results for eight subjects are shown.

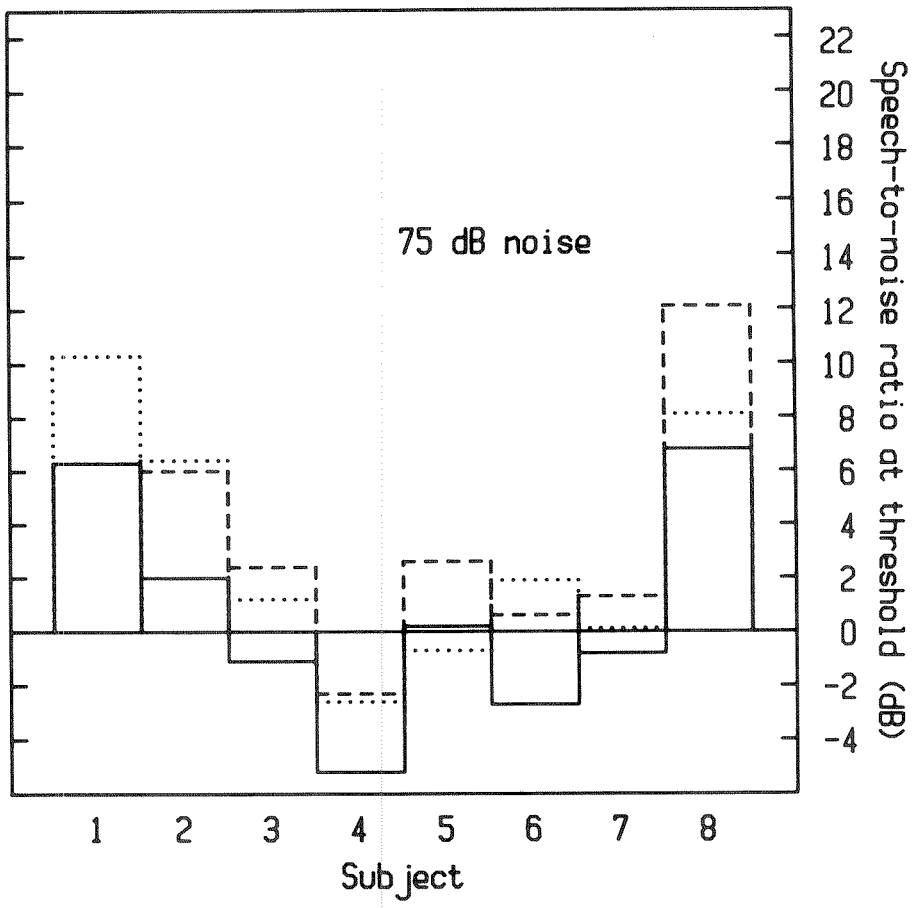
speech signal, allows better intelligibility of speech in noisy situations than single-channel slow-acting AGC alone.

Overview and Discussion of the Results

Our results have shown that, for the majority of subjects, the two-channel compression aid gives lower SRTs in noise than a comparable linear aid, a comparable single-channel aid, or unaided listening. The improvements are not large, typically being around 2 to 3 dB, but this is equivalent to about a 22 to 33 percent improvement in intelligibility in difficult listening situations. Even when listening binaurally aided, the SRTs of most of our subjects were higher than those of normally hearing subjects tested in the same conditions. Thus, while the two-channel compression aids can alleviate the problems experienced by the hearing impaired, they do not eliminate them.

The essence of our approach is that speech should be processed so that all of the important acoustic elements are easily audible, but below the level producing discomfort. This should be done in such a way that distortions of the speech are minimal; in particular, the aid should not reduce the salience of such features in the speech as formant frequencies, and it should not introduce spurious and potentially misleading acoustic cues. We believe that the combination of slow-acting AGC acting on the whole speech signal and fast-acting AGC in the individual channels is an effective way of achieving this goal. The use of the slow-acting AGC reduces the amount of compression needed in the individual channels, and so reduces spurious spectral and temporal distortions introduced by the compression. Almost as important is the care that has been taken to minimize harmonic and intermodulation distortion, and to ensure a smooth wideband frequency response.

The two-channel compression aid has been designed mainly to deal with the problem of reduced dynamic range, and only secondarily to deal with the problem of reduced suprathreshold discrimination. However, the directional microphone does



provide some assistance with the latter, by an amount equivalent to about a 3-dB improvement in SRT when the speech and noise are spatially separated. The fast-acting channel AGCs may also provide some relief from forward masking effects, which are often pronounced in the hearing impaired at comfortable listening levels: e.g., Moore, Laurence, and Wright (15); Glasberg, Moore, and Bacon (10). Finally, a feature not previously mentioned, the aid has been designed so that frequencies below about 200 Hz are severely attenuated. This reduces masking by intense low-frequency sounds such as are often encountered in the environment (e.g., in a car, in the subway, near air-conditioning vents). Hearing-impaired persons are often particularly susceptible to the upward spread of masking: e.g., de Boer and Bouwmeester (3); Glasberg and Moore (9); for a review see Tyler (22).

One important feature of the aid is that it can be adjusted to suit the individual patient. I turn now to a consideration of fitting procedures, and to a comparison of two different fitting methods.

FITTING PROCEDURES FOR THE TWO-CHANNEL AID

As the number of possible adjustments on a hearing aid increases, there is a corresponding increase in the need for a well-defined fitting procedure. We have found that, even though there are only four adjustable controls on the two-channel compression aid, devising and evaluating fitting procedures is by no means straightforward. The difficulties would be even greater for an aid with more than two channels. Similarly, the potential of digital hearing aids for adjustment to suit the individual patient will depend critically on the evolution of suitable fitting procedures, and this may well be a major factor limiting their usefulness.

We have conducted some preliminary experiments comparing two different fitting procedures for the two-channel compression aid. One procedure uses frequency-modulated (warble) tones as test stimuli, while the other uses running speech. The

former has the advantage of simplicity in administration and in the instructions to the subject. The latter requires slightly more difficult judgements of the subject, but has greater face validity because a major use of the aid will be in listening to speech. Both procedures aim to fit the aid in such a way that speech in quiet can be understood over the whole range of levels from 55 to 90 dB SPL without any need to adjust the controls on the aid. Almost all speech encountered in everyday life would fall within this range.

The two procedures are described below. Both procedures were conducted in a sound-treated (but not anechoic) room, the subject being seated 1.3 meters from a loudspeaker and facing the loudspeaker (Monitor Audio MA4). A master hearing aid was used, so as to make it easier to adjust the controls on the aid. The microphone and receiver were mounted on a headband. Sound levels are specified as the level at the position corresponding to the center of the listener's head, the listener having been removed from the sound field. For both procedures, the volume control [*Block 2*] was set to maximum (the position to which it would normally be set for everyday use), and the overall output level was controlled by *Block 13*. The channel gain controls [*Blocks 8 and 9*] were initially set to −10 dB re maximum gain. The channel balance control [*Block 12*] was set to the central position.

Fitting Using Warble Tones

The tones used had center frequencies of 750 Hz and 2500 Hz, chosen to be well within the passbands of the lower and upper channels, respectively. They were sinusoidally frequency modulated at a 10-Hz rate, with a modulation depth of 12.5 percent of the center frequency. The procedure was as follows:

a) Pulsed tones, 500 ms in duration (20 ms rise/fall times) with 1000-ms interpulse intervals, were presented in a continuous sequence, alternating between the frequencies 750 and 2500 Hz, and at a level of 85 dB SPL. The output level control [*Block 13*] was adjusted so that the sounds were loud, but not so loud that the subject would not be prepared to listen to them for a long period of time. The channel balance control [*Block 12*] was adjusted so that the two frequencies sounded equally loud. These adjustments serve to set the output level and frequency-response of the aid for high input levels.

b) Pulsed tones with a frequency of 750 Hz and a level of 45 dB SPL were presented in a continuous sequence. The low-frequency channel gain control [*Block 9*] was adjusted so tones were just audible.

c) Pulsed tones with a frequency of 2500 Hz and a level of 40 dB SPL were presented in a continuous sequence. The high-frequency channel gain control [*Block 8*] was adjusted so that the tones were just audible. The adjustments b and c serve to set the frequency-gain characteristic of the aid for low input levels. The sound levels of the test tones were chosen (on the basis of pilot trials) as giving settings which allowed speech at a level of 55 dB SPL to be just intelligible for most subjects.

d) Ideally steps a, b, and c should be repeated because the output level and channel balance can be affected by the channel gain controls. This was sometimes but not always done in our pilot trials.

Fitting Using Speech

The test stimulus was running speech with a constant average level recorded from a BBC Radio 4 news broadcast, using a male speaker. The fitting procedure was as follows:

a) The speech was presented at a peak level of 85 dB SPL, and the output level control [*Block 13*] was adjusted to the highest level at which the subject would be prepared to listen to speech for a long period of time.

b) The channel balance control [*Block 12*] was adjusted so that the speech sounded as clear and as natural as possible. The subjects were asked to indicate if the speech sounded too "hissy" or too "boomy" and the control was adjusted appropriately. Steps a and b serve to set the output level and frequency response of the aid for high input levels.

c) The recorded speech was presented at a peak level of 55 dB SPL and the channel gain controls [*Blocks 8 and 9*] were adjusted so that the speech could just be understood. When adjusting the low-frequency channel gain, the subject was asked to indicate when the vowel and nasal sounds (e.g., /m/ and /n/) were just clearly audible. When adjusting the high-frequency channel gain, the subject was asked to indicate when consonant sounds such as /t/, /f/, and /k/ could just be clearly heard.

d) Ideally steps a, b, and c should be repeated, for the reasons mentioned above, although this was not always done in our pilot trials.

Results of Pilot Trials

We have compared the two fitting procedures using 12 subjects with sensorineural hearing losses with recruitment. Only one ear of each subject was tested, the other ear being plugged with an EAR earplug and covered with a sound-attenuating muff. Once the aid had been fitted with a given procedure, performance was evaluated by measuring the SRT in quiet and in a speech-shaped background noise with a level of 60 dB SPL, using the BKB sentence lists. The order of evaluating the two procedures was counterbalanced across subjects.

The SRTs in quiet for the "speech" procedure had an average value of 49 dB SPL. Since the SRT is the level of speech required for 50 percent intelligibility, this means that we were close to achieving our goal of a high level of intelligibility for speech at a level of 55 dB SPL. However, for the "tones" procedure, the SRTs in quiet had an average value of 54 dB SPL, so that speech at 55 dB SPL would be only a little more than 50 percent intelligible. Thus, the "tones" procedure needs to be modified by using slightly lower signal levels in stages b and c. The differences in the SRTs in quiet are broadly consistent with the differences in the settings of the controls for the two procedures. For the "speech" procedure, the output level control [*Block 13*] was set, on average, about 2 dB lower than for the "tones" procedure. Also, the channel balance was set to give relatively more emphasis to the high frequencies for the "speech" procedure. However, the high- and low-frequency channel gain controls [*Blocks 8 and 9*] were set, on average, to give 8 and 11 dB more gain for the "speech" procedure. The gain for low-level inputs would thus have been about 6 to 9 dB higher for the "speech" procedure.

The SRTs in noise were lower (better) for the "speech" procedure for 9 of the 12 subjects, but the average difference of 1 dB just failed to reach statistical significance at the 5 percent level.

Clearly, much work remains to be done in refining and evaluating fitting procedures. Our pilot results suggest that the procedure using speech tends to give more satisfactory fittings than the procedure using warble tones. In experiments using linear amplification, Byrne (4) also found that frequency response selections based on the use of speech stimuli were more effective than those based on tones. However, our results were not conclusive, and modifications to the "tones" procedure might well produce better results. Many other procedures are possible. In particular, informal experiments suggest that, after initial fitting using the "speech" procedure, it may be useful to make minor adjustments to the channel gain controls while the subject is listening to running speech in background noise; the subject is asked whether the speech is more or less clear after each adjustment has been made, and further adjustments are made as appropriate. When subjects are fitted binaurally, they usually prefer a slightly lower setting of the output level control than for a monaural fitting.

GENERAL DISCUSSION AND CONCLUSIONS

Our results have shown that the two-channel compression aid has three main benefits for persons with moderate sensorineural hearing losses.

Firstly, it allows the understanding of speech in quiet over a wide range of sound levels, without any need to adjust the controls on the aid. This may be of particular importance to elderly people with poor manual dexterity.

Secondly, the aid usually improves the ability to understand speech in noise in comparison to unaided listening, or to a matched linear aid, or to a single-channel compression aid. The improvement is typically 2 to 3 dB in the speech-to-noise ratio required for threshold, which is equivalent to a 22 to 33 percent improvement in intelligibility for the test sentences which we used.

Finally, the aid can be adjusted to suit the individual patient, so as to deal with a wide range of degree and pattern of hearing loss. This eliminates many of the difficulties associated with the "prescriptive" fitting of hearing aids. Such fitting usually involves selecting one brand of aid from many available, using rules relating the required frequency-gain characteristic to the audiometric configuration of the patient. At present there is no general agreement about which rules give the best results (11) and, in any case, there is often considerable uncertainty about the real insertion gain provided by a particular sample of an aid on a particular patient. By adjusting the two-channel compression aid while the user is actually wearing it, and by using speech as the test stimulus, such difficulties are largely circumvented. The fitting procedure will ensure that speech heard through the

aid is at a comfortable level and has a reasonable balance between high and low frequencies.

Many questions remain unanswered. For example, we have yet to conduct systematic trials to determine the optimum time constants for the compression circuits, or to determine whether benefits would be obtained by having more channels. We do not know whether it is better to have compression which operates uniformly over a wide range of sound levels or to use compression limiting, whereby the aid is linear up to a certain level and compression operates strongly above that level (as in the present version of the aid). There are so many parameters which could be varied, in an aid of the type we have described, that it is essential to choose suitable values on the basis of "educated guesses" in order to make any progress at all. Similar (but worse) problems will face the designers of digital hearing aids; with such aids almost any kind of speech processing will be possible. The main difficulties will be in deciding what processing to do, in evaluating and comparing different schemes, and in devising fitting procedures that can be implemented in clinical practice.

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