Hearing Aid Design Criteria

Les critères de conception des prothèses auditives

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Abstract

The modern hearing aid is an imperfect device. Practical implementations often fail to meet reasonable performance expectations, and adequate design criteria are often lacking for more advanced signal processing strategies. Performance limitations and design objectives will be reviewed for three areas of particular importance in hearing aids: linear amplification, dynamic range compression, and noise suppression.

Résumé

La prothèse auditive d'aujourd'hui est un appareil imparfait. Fréquemment, ses applications pratiques ne répondent pas à la performance attendue, et un modèle conceptuel décrivant des critères appropriés est souvent indisponible pour les stratégies plus avancées de traitement du signal. L'auteur examine les limites au niveau de la performance des appareils auditifs et les objectifs d'un modèle conceptuel pour trois aspects importants des aides auditives : l'amplification linéaire, le système de compression et la suppression du bruit.

Introduction

The modern hearing aid is an imperfect device. Limitations in circuit, transducer, and battery technology can make it difficult to achieve desired performance objectives. And the performance objectives may be inadequate or incomplete due to the inability to unambiguously specify what an improved device should actually do. Thus there are complaints about the limited benefit of hearing aids (Plomp, 1978), but improved devices and better algorithms are slow in coming.

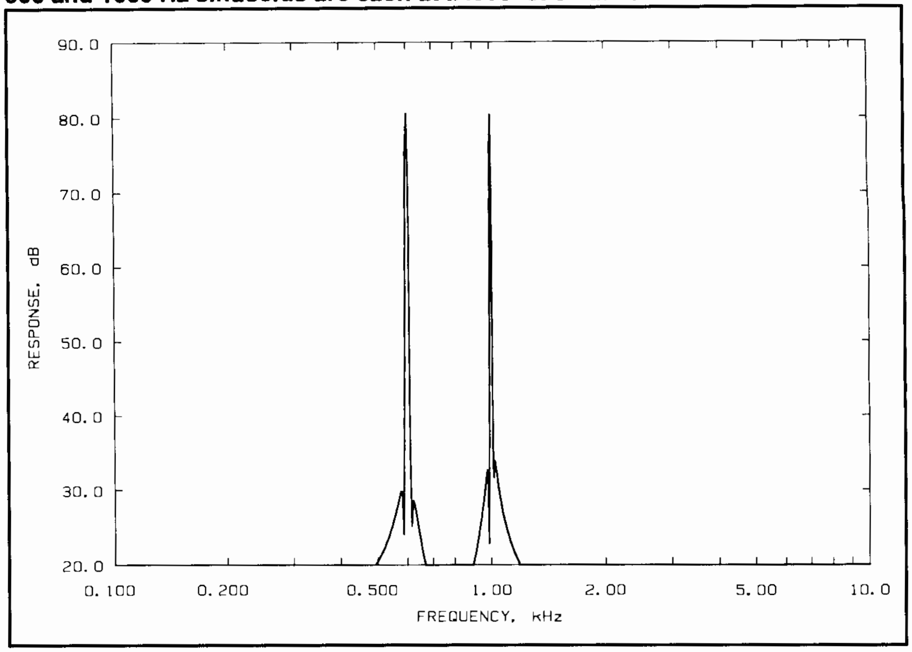
The design of a hearing aid represents a compromise between practical considerations and signal processing complexity. Small hearing aids leave little room for complex circuits or for large batteries. New processing technologies or improvements in existing approaches can, on the other hand, require larger cases for the more complicated circuitry and need batteries having much greater energy storage to provide the power. The hearing aid market today has a strong preference for small instruments; industry statistics (Cramer, 1991) indicate that in-the ear (ITE) and in-the-canal (ITC) instruments represent about 77% of the total market in North America. The major compromise is thus cosmetics versus function. New technology and processing improvements, if they carry a cosmetic disadvantage, will have to prove substantially better than existing devices if they are to achieve wide spread acceptance in the marketplace.

The problems with hearing instruments can be divided into two categories: those cases for which adequate design criteria exist but cannot be met, and those cases for which design criteria and effective processing algorithms do not exist. An example of the former class of problems would be insufficient gain at high frequencies to achieve the frequency response indicated by a fitting rule, and an example of the latter class would be improving speech intelligibility in broadband noise. Both types of problems will be discussed. In addition, three general classes of signal processing for hearing instruments — linear amplification, dynamic range compression, and noise suppression — will be discussed. These three areas were chosen to give an indication of both the practical problems in achieving clearly defined performance criteria and the difficulties of improving performance when the criteria are poorly defined. The intent is to identify weaknesses in existing technology, discuss the validity of design criteria and processing algorithms, and propose avenues for improvement.

Linear Amplification

The basic hearing aid circuit is the linear amplifier, and the simplest hearing aid consists of a microphone, amplifier, and receiver (output transducer). Additional shaping of the frequency response to match an individual audiogram can also be provided either through electronic circuitry or by modifications to the acoustic response, such as providing a vent in an earmold or ITE shell (Kates, 1988). In addition to being commonly prescribed on its own, the linear hearing aid also forms the fundamental building block for more advanced designs. Thus many of the problems associated with linear amplification will also affect other processing approaches when implemented in practical devices. Conversely, improvements in linear instruments will lead to improvements in all hearing aids.

Figure 1. A two-tone test signal for measuring intermodulation distortion. The 600 and 1000 Hz sinusoids are each at a level of 82 dB SPL.



Dynamic Range

The dynamic range of a hearing aid is bounded by noise on the bottom and amplifier saturation on the top. A typical hearing aid microphone has a noise level of about 20 dB SPL, which is comparable to that of the human ear (Killion, 1976). The addition of the hearing aid processing and amplification circuits gives input-referred noise levels of between 25 and 30 dB SPL. More complicated processing, such as a multi-channel filter bank, may generate higher input-referred noise levels due to the larger number of circuit components required. The equivalent hearing aid noise level, after amplification, is therefore about 10 dB higher than that of the normal unaided ear. This noise level tends to limit the maximum gain that a hearing aid user will select under quiet conditions because, in the absence of the masking provided by intense inputs, a user will reduce the gain in order to reduce the annoyance of the background noise. Thus a reduction in the circuit noise would encourage higher volume control settings in quiet, resulting in greater amplification of low-level speech sounds.

At the other extreme, amplifier saturation limits the maximum gain that can be achieved by the hearing aid. A typical hearing aid amplifier clips the signal when the peak input level exceeds about 85 dB SPL. A speech-like signal at an input of 70 dB SPL is therefore amplified cleanly, but a level of 80 dB SPL causes large amounts of distortion (Preves & Newton, 1989). Speech input at 65 to 70 dB SPL is typical of normal conversational levels (Cornelisse, Gagné, & Seewald, 1991), but the spectra of individual speech sounds can be as much as 15 dB higher when monitoring the talker's own voice at the ear canal (Medwetsky & Boothroyd, 1991). Thus the typical hearing aid does not have enough headroom to guarantee that the user's own voice will be amplified without distortion.

The available hearing aid dynamic range is thus about 55 dB from the noise floor to the clipping threshold. Selecting an amplifier with more gain, and turning down the volume control, will raise the clipping threshold, but will also raise the noise level by a similar amount. Thus a typical hearing aid, due to the compromises made in battery size and circuit design, can only handle half the dynamic range of a normal ear. Some progress is being made, however; the recently introduced class D amplifier (Carlson, 1988) can provide 10 to 20 dB more output at saturation than a class A amplifier having comparable gain (Fortune & Preves, 1992).

Distortion

Amplifier saturation most often takes the form of symmetric clipping (Armstrong, 1989). If a single sinusoid is input to the hearing aid, the clipping will generate harmonic distortion; for two or more simultaneous sinusoids, intermodulation (IM) distortion is the result. Consider, for example, an excitation consisting of sinusoids at 600 and 1000 Hz, each sinusoid at a level of 82 dB SPL. The spectrum for this signal is shown in Figure 1. This signal was processed by a simulated linear ITE hearing aid having a Knowles ED1842 microphone, a flat amplifier response with clipping for peaks greater than 85 dB SPL, and a Knowles ED1913 receiver (Kates, 1990a). The output spectrum is shown in Figure 2. A large number of distortion products are present at multiples of the 400 Hz spacing between the excitation tones.

Applying a high-frequency emphasis changes the distortion spectrum and its effects. A two-channel linear hearing aid was simulated having a crossover frequency between the two channels of 1500 Hz and gains of -20 dB in the low-fre-

Figure 2. Intermodulation distortion for a simulated hearing aid for the test signal in Figure 1. The amplifier clipping level was set to 85 dB SPL.

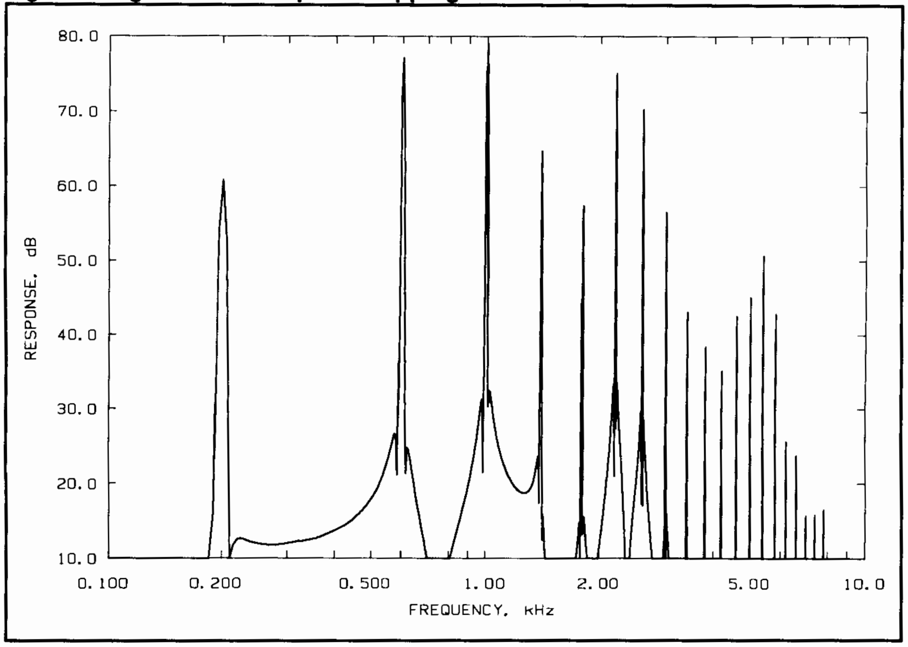
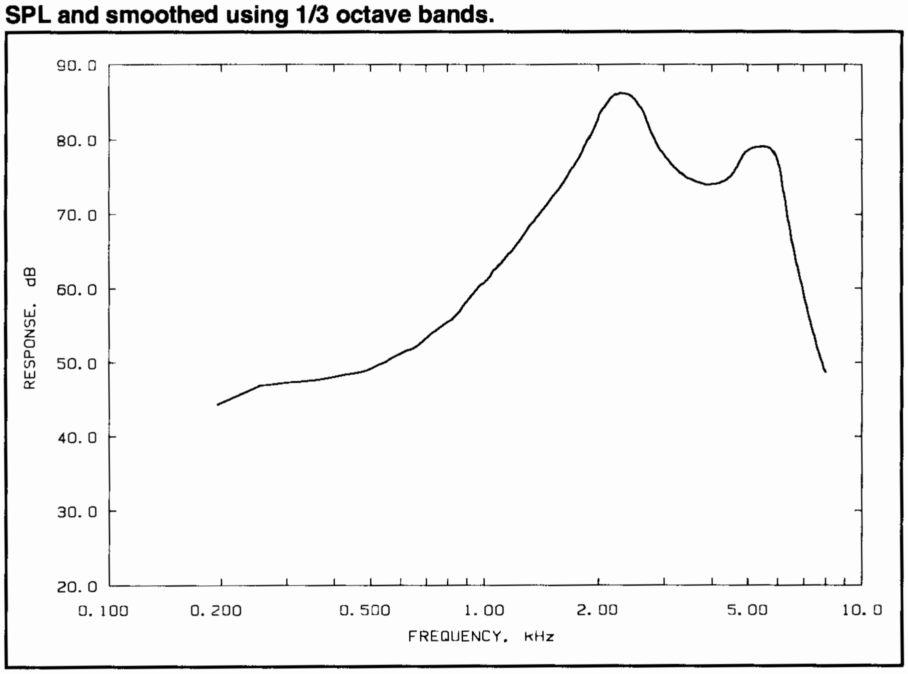


Figure 3. Frequency response of a simulated two-channel hearing aid having a crossover frequency of 1500 Hz, gains of -20 dB in the low-frequency channel and 0 dB in the high-frequency channel, and an amplifier clipping level of 85 dB SPL. The response was measured with speech-shaped noise at a level of 70 dB SPL and smoothed using 1/3 octave bands.

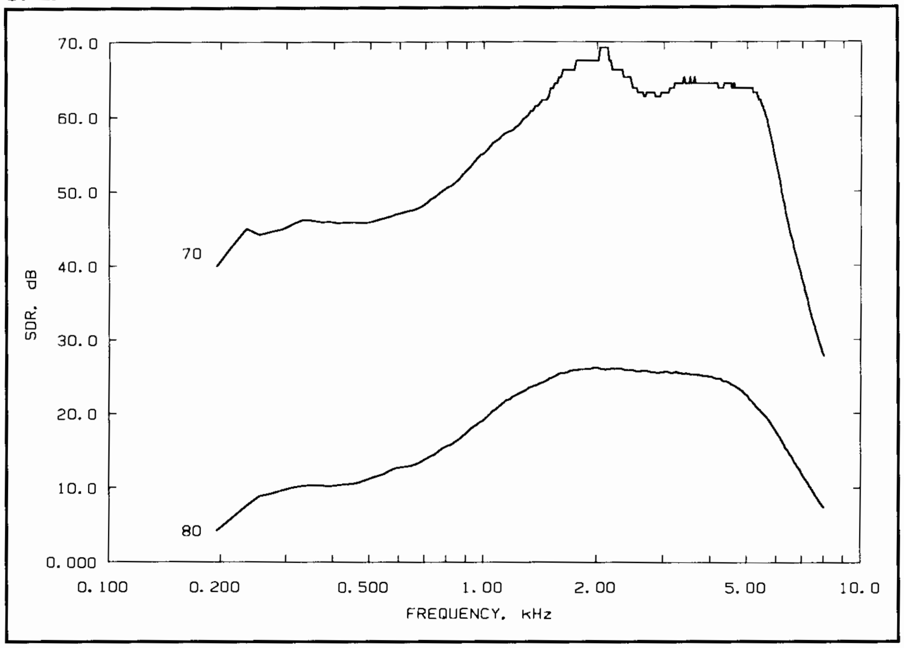


quency channel and 0 dB in the high-frequency channel. The amplifier clipping level was kept at 85 dB SPL. The frequency response of the simulated hearing aid, measured using speech-shaped noise and smoothed using one-third octave bands (Kates, 1990b), is shown in Figure 3. The signal-to-distortion ratio (SDR) in dB, derived from the unbiased coherence function for a speech-shaped noise excitation (Kates, 1992), is shown in Figure 4 for the same hearing aid. For an input at 70 dB SPL, representative of normal conversational levels, the SDR is better than 45 dB at all frequencies of interest. Increasing the signal level to 80 dB SPL causes a substantial increase in distortion, with the SDR lowest at low frequencies even though the amplified signal has most of its power above 1500 Hz. Thus the distortion products have a much broader spectral distribution than the excitation signal, and distortion from a high-frequency sound could mask less intense low-frequency speech components occurring at the same time.

Kates (1990b) has proposed a quantitative measure of the distortion effects formed along the lines of the Articulation Index (French & Steinberg, 1947; Kryter, 1962) as follows: Determine the SDR for each auditory critical band, limiting the value to a maximum of 30 dB and a minimum of 0 dB; sum the SDR values and divide by 30 times the number of critical bands to give a number between zero and one. This procedure produces a value of 0.999 for the 70 dB SPL input, 0.685 for the 80 dB SPL input, and 0.216 for an input at 90 dB SPL, so increased signal levels would be expected to yield reduced speech intelligibility.

The amount of distortion influences judgments made about hearing aid quality. Fortune and Preves (1992), for example, found that reduced coherence was related to a lower hearing aid amplifier saturation level and a lower loudness discomfort level (LDL), suggesting that LDL depends on the amount of distortion as well as on output power. In another study, a large majority of hearing aid users indicated that good sound quality was the most important property of hearing aids, with clarity being the most important sound quality factor (Hagerman & Gabrielsson, 1984). Thus reduced distortion would be expected to lead to improved speech intelligibility at high sound levels and greater user comfort and satisfaction.

Figure 4. SDR for the simulated two-channel hearing aid measured using the unbiased coherence function for speech-shaped noise at levels of 70 and 80 dB SPL.



system frequency response. One would also expect distortion to be increased in an instrument close to the onset of instability because the feedback oscillations will use up most of the available headroom. Mechanical vibrations from the receiver in a high power hearing aid can be reduced by combining the outputs of two receivers mounted back-to-back so as to cancel the mechanical moment; as much as 10 dB additional gain can be achieved before the onset of oscillation when this is done. But in most instruments, venting the BTE earmold or ITE shell establishes an acoustic feedback path that limits the maximum possible gain to about 40 dB (Kates, 1988) or even less for large vents or open fittings. Acoustic feedback problems are most severe at high frequencies because this is where a typical hearing aid has the highest gain. The criterion for effective feedback suppression is a useful increase in maximum gain, while preserving speech information and environmental awareness.

The traditional procedure for increasing the stability of the hearing aid is to reduce the gain at high frequencies (Ammitzboll, 1987), although phase shifting and notch filters also have been used (Egolf, 1982). Controlling feedback by modifying the system frequency response, however, means that the desired high-frequency response of the instrument must be sacrificed in order to obtain stability. A more effective technique is feedback cancellation, in which the feedback signal is estimated and subtracted from the microphone input.

Simulations and digital prototypes of feedback cancellation systems (Bustamante, Worrell, & Williamson, 1989; Kates, 1991; Engebretson, O'Connell, & Gong, 1991) indicate that increases in gain of between 6 and 17 dB can be achieved before the onset of oscillation with no loss of high-frequency response. However, there is a possibility that the feedback cancellation will also remove desired environmental signals, such as an alerting siren. Thus feedback problems can be greatly ameliorated without compromising the desired high-frequency gain, although other problems may remain, but the solution requires that digital signal processing be built into the hearing aid.

Bandwidth

The bandwidth of a hearing aid should be wide enough for good speech intelligibility and accurate reproduction of other sounds of interest to the user. French and Steinberg (1947) determined that a frequency range of 250-7000 Hz gave full speech intelligibility for normal hearing subjects, and more recent studies (Pavlovic, 1987) extend this range to 200-8000 Hz for nonsense syllables and about 100-10000 Hz for continuous discourse. For music, a frequency range of 60-8000 Hz reproduced over an experimental BTE was found to compare favorably with a wide range loudspeaker system, again using normal hearing listeners (Killion, 1988). Thus a reasonable objective is a 60-8000 Hz bandwidth.

Most hearing aids have adequate low-frequency but inadequate high-frequency response for optimal speech intelligibility. Increasing the high-frequency bandwidth would yield improved speech intelligibility, but only if the amplifier could cope with the increased power demands without undue distortion and if the system would remain stable in the presence of increased levels of acoustic and mechanical feedback. Thus increasing the hearing aid bandwidth, while desirable, must wait for other problems to be solved first.

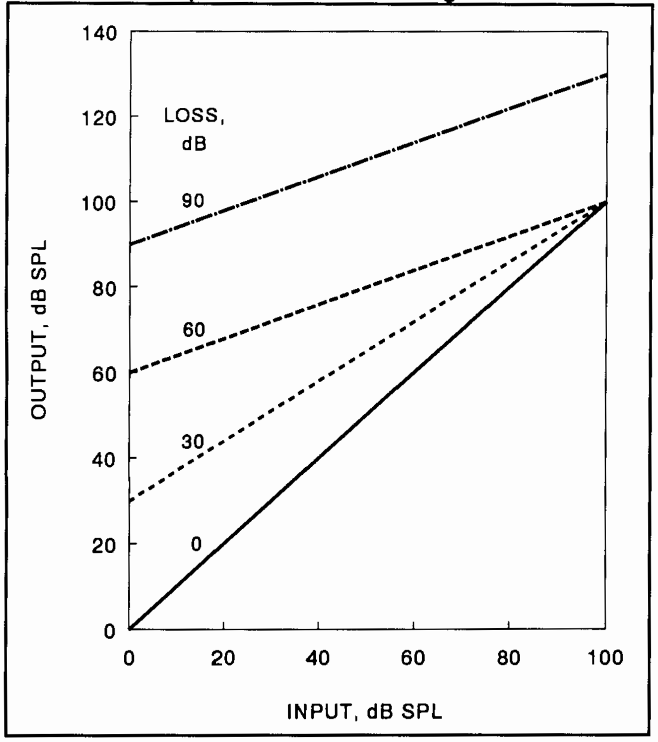
Feedback

Mechanical and acoustic feedback limits the maximum gain that can be achieved in most hearing aids and degrades the

Dynamic Range Compression

Dynamic range compression, or automatic gain control (AGC), is used for two different purposes in hearing aids. The first,

Figure 5. Proposed compression amplifier input/output curves for a simplified model of hearing loss.



and most prevalent use, is as a limiter to prevent overloading the amplifier circuits or the user's ear when an intense sound occurs. The second use is to match the dynamic range of speech and environmental sounds to the restricted dynamic range of the impaired listener, also termed recruitment compensation. These two uses imply different and even contradictory criteria for setting compression ratio and attack and release times. For reducing overload, one wants a rapid attack time so as to respond to a sudden intense sound, a high compression ratio to limit the maximum signal level, and a high compression threshold so as not to limit sounds that could otherwise be amplified without distortion. For recruitment compensation, on the other hand, one wants longer attack times and low compression ratios to minimize any deleterious effects of the compression on the speech envelope (Plomp, 1988; Boothroyd, Springer, Smith, & Schulman, 1988) and a low compression threshold so that speech sounds at any level of presentation can be perceived.

The optimum choice of compression ratio and gain for wide dynamic range compression has not yet been determined. Approximate values, however, can be estimated from considerations of auditory physiology. In a healthy cochlea, the active mechanism of the outer hair cells provides about 50-60 dB of gain for a sinusoid at auditory threshold (Kiang & Moxon, 1974). Increasing the signal level results in a reduction of gain and a broadening of the auditory filters (Johnstone, Patuzzi, & Yates, 1986) until at high levels the gain is reduced to about 0-10 dB. In a cochlea with extensive outer hair cell damage, the filter shape and gain is similar at all input levels to that of the healthy cochlea at high levels (Harrison, Aran, & Erre, 1981). As an approximation, assume that in the healthy cochlea an input of 0 dB SPL gets 60 dB of gain, while an input of 100 dB SPL gets 0 dB of gain, giving a compression ratio of 2.5:1. A severely impaired cochlea, on the other hand, has 0 dB of gain at all input levels resulting in a linear system. One could therefore argue that the highest compression ratio needed in a hearing aid, corresponding to complete outer hair cell damage, is 2.5:1 and that lesser amounts of damage would require correspondingly lower compression ratios.

Total outer hair cell damage results in a threshold shift of no more than 60 dB because that is the maximum amount of gain provided by the cochlear mechanics. Hearing losses greater than 60 dB therefore must be accompanied by damage to the neural transduction mechanism, and Liberman and Dodds (1984) have shown that inner hair cell damage results in a threshold shift but no apparent change in the mechanical behavior of the cochlea. Thus outer hair cell damage, in this model of hearing loss, causes a loss of sensitivity and a reduction in compression ratio, while inner hair cell damage causes a linear shift in sensitivity.

Fitting procedures for moderate losses, such as the half-gain rule, are really attempts to produce the same gain in the impaired ear for a typical speech stimulus as would occur in a normal ear for that stimulus. For example, a narrow-band input of 50 dB SPL would get about 30 dB of gain in the healthy cochlea given the compression action, while a 60 dB hearing loss due to outer hair cell damage would result in 0 dB of gain for the same stimulus; 30 dB of gain, or half the hearing loss, equalizes the levels that excite the inner hair cells. For more severe losses, in which inner hair cell damage also must be assumed to exist, additional gain is needed, as was indeed found to be the case by Byrne, Parkinson, and Newall (1990). Thus the family of hearing aid input/output curves would be as indicated in Figure 5, wherein the compression ratio is increased as the hearing loss increases up to 60 dB of loss, after which the compression ratio remains constant and the gain is increased.

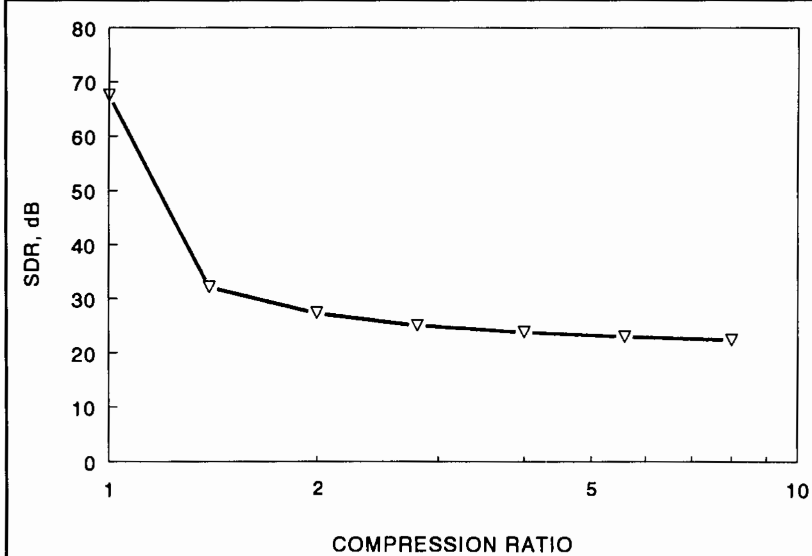
The choice of compression ratio and attack and release times will also effect the distortion of the hearing aid, especially for wide dynamic range compression in which the signal is nearly always above the compression threshold. The distortion in a simulated hearing aid having an idealized flat frequency response from 100 to 6000 Hz was measured using speech-shaped noise input at 70 dB SPL, and the distortion index was the SDR at 1000 Hz computed from the unbiased coherence function (Kates, 1992). The results for a compression ratio of 2:1 with a compression threshold of 50 dB SPL

Table 1. SDR in dB at 1000 Hz as a function of the ANSI attack and release times for a simulated AGC hearing aid having a flat response from 100 to 6000 Hz.

RELEASE ms	ATTACK ms 0	1	2	5	10
5	18.7	20.6	21.5	22.2	22.8
10	19.7	22.2	23.1	24.0	24.6
20	20.7	23.8	24.7	25.9	26.8
50	22.6	26.3	27.4	28.6	29.4
100	24.4	28.2	29.2	30.7	31.5
200	25.5	29.5	30.6	32.3	33.5

are presented in Table 1. The amount of distortion decreases as the attack and release times are increased; the SDR is approximately 30 dB for any combination of ANSI attack time greater than 2 ms and release time greater than 50 ms, so the distortion would not be expected to reduce speech intelligibility or significantly effect speech quality for time constants within this range. Increasing the compression ratio to 8:1 reduces the SDR by about 5 dB in every cell of Table 1, and the effect of varying the compression ratio on the distortion for an attack time of 2 ms and a release time of 50 ms is graphed in Figure 6. The higher compression ratios, given the 50 dB SPL compression threshold, reduce the SDR to a level at which speech intelligibility or quality may be compromised, but the distortion effects would need to be determined experimentally before any recommendations could be made for hearing aid design.

Figure 6. SDR at 1000 Hz as a function of compression ratio for an attack time of 2 ms and a release time of 50 ms for a simulated hearing aid having a flat response from 100 to 6000 Hz.



Increasing the number of channels in a compression system increases the need for effective design criteria and the associated fitting procedures. There is evidence that two-channel compression can offer small improvements in speech intelligibility over a single-channel system (Moore, 1987), but more complicated compression schemes have not demonstrated any significant advantage when compared with broadband compression (Bustamante & Braida, 1987; Levitt & Neuman, 1991). For two-channel systems, the criteria being used are to design independent channels having enough adjustable parameters (crossover frequency, compression ratio, and gain) so that the instrument can produce a wide range of frequency response and compression characteristics. This design approach, however, only serves to shift the burden to the fitting rule in trying to obtain optimum performance. Criteria for more complex systems would have to include minimizing unwanted phase and amplitude interactions that can occur in the filters used for frequency analysis/synthesis (Walker, Byrne, & Dillon, 1984) and that can give unwanted peaks or notches in the system frequency response. They should also include an awareness of non-linear interactions in the cochlea, such as two-tone suppression (Sachs & Kiang, 1968), in which the output from one region of the cochlea can be reduced by the presence of signal components at another frequency.

Noise Suppression

Improving speech intelligibility in noise has long been an important objective of hearing aid design. In cases in which the interference is concentrated in time (e.g., clicks) or frequency (e.g., pure tones) intelligibility in fact can be improved; clipping the signal in the former or using a notch filter in the latter case will reduce the noise level by a much greater amount than the speech. A much more difficult problem is to improve speech intelligibility in the presence of broadband noise. To this end two noise suppression approaches have been implemented in hearing aids, these being adaptive filters and directional microphones.

Adaptive Filters

Adaptive filters for reducing the low-frequency output of a hearing aid in the presence of noise have been commercially available for several years (Kates, 1986). While some instruments have been based on a two-channel approach having compression in the low-frequency channel in order to limit the amplification of intense low-frequency noise, it is more common to find a system using a high-pass (low-cut) filter having a slope of 6 or 12 dB per octave and having an automatically adjustable cutoff frequency. The adjustable high-pass filter reduces the low-frequency bandwidth of the hearing aid as the noise level increases. Tests of such systems have demonstrated that there is no net improvement in intelligibility when the volume control is kept in a fixed position (Kuk, Tyler, Stubbing, & Bertschy, 1989; Fabry & Van Tasell, 1990), but some improvement has been reported when the subjects are free to adjust the volume (Sigelman & Preves, 1987). This dependence on volume setting suggests that the major effect on intelligibility is actually a result of a reduction in distortion; the reduced gain at low frequencies allows for increased amplification at high frequencies before the amplifier saturates (Fabry, 1991).

The criterion proposed here for effective noise suppression systems is to maximize the Articulation Index (AI). Reducing the gain in any one critical band will not affect the signal-to-noise ratio (SNR) in that band, but may still increase the AI if the noise in that band, at the original level, was intense enough to mask speech in a nearby band. Masking effects extend primarily upward in frequency (Egan & Hake, 1950). Thus the gain in a critical band should be reduced if the masking effects of the noise in that band on sounds in a higher frequency band exceed the noise level in the higher frequency critical band, and the gain reduction should be enough only to make the out-of-band masking and in-band noise equal in the high-frequency band (Kates, 1989).

A specific example is a noise source (e.g., ventilation fans, highway traffic) having a spectrum that is proportional to 1/frequency. The auditory critical bands, below about 500 Hz, approach a relatively constant bandwidth (Moore & Glasberg, 1983). Thus at low frequencies, the noise spectrum in the ear for 1/frequency noise will increase in level at about 6 dB/octave as the frequency is decreased. A simple one pole high-pass filter, with a cutoff frequency at about 500 Hz, will thus equalize the noise levels in the critical bands, removing any masking effects and maximizing the AI. At frequencies above 500 Hz the critical bands are approximately constant, so the critical band spectrum for 1/frequency noise is more nearly flat and high-pass filtering will not reduce the noise masking effects significantly. Sharper filter slopes, or a higher cutoff frequency, will not lead to improved speech intelligibility given these assumptions because they will not increase the AI; this is consistent with the findings of Kuk et al. (1989) that subjects preferred an adaptive frequency response with a 6 dB/octave slope to one with a 12 dB/octave slope in daily use. Because many hearing aid microphones already have a one pole high-pass filter at about 300 Hz designed into the frequency response (Kates, 1990a), no additional processing is needed for 1/frequency noise. However, the effects of noises with relatively flat spectra at low frequencies, such as multi-talker speech babble, will not be reduced by this simple filter.

Directional Microphones

In many situations, the desired signal comes from a single well defined source, such as a person seated across the table, while the noise is generated by a large number of sources located throughout the area, such as other diners in a restaurant. Under these conditions the speech and the noise tend to have the same spectral distribution, so the AI-based approach discussed above will not be particularly effective. However, the spatial distributions differ, and the spatial characteristics can be exploited to reduce the noise level without any deleterious effects on the speech.

A directional microphone will improve the SNR by maintaining high gain in the direction of the desired source and reduced gain for sources coming from other directions. An ideal cardioid (heart-shaped) response will improve the SNR by 4.5 dB compared with an omnidirectional microphone for an on-axis sound source and an isotropic (diffuse) noise field (Olson, 1957). Measurements of an actual directional microphone mounted on the head, however, indicate that the advantage is only about 2.5 dB compared to an omnidirectional hearing aid microphone in a diffuse noise field (Soede, 1990). Larger benefits can be obtained under more constrained conditions; a relative improvement of 3-4 dB for the directional microphone was found when a sound source was positioned in front and a noise source behind the head in a reverberent room (Hawkins & Yacullo, 1984). Even though these improvements are small, directional microphones are the only practical technique that has demonstrated benefit in enhancing speech intelligibility in noise.

Greater improvements in the SNR require arrays that combine the outputs of several microphones rather than using just a single microphone element. The benefit of microphone arrays of the sort that can be built into an eyeglass frame, for example, is an improvement of 5-10 dB in SNR, with the greatest improvement at higher frequencies (Soede, 1990). Furthermore, the performance of both broadside arrays (across the front of the glasses) and endfire arrays (along the temple) does not appear to be affected by the head to any great extent. These arrays used directional microphone elements and fixed weights corresponding to classical delay-and-sum beamforming, so equivalent performance would be expected from a "shotgun" microphone having a similar length.

An important consideration in the design of a directional microphone or array is the assumed spectra of the speech and interfering noise. The AI can again provide guidance because the objective is to improve the SNR in those frequency bands wherein it will do the most good. If the noise is assumed to be concentrated at low frequencies, then the array directional pattern should be optimized for low-frequency performance. If the noise is assumed to have the same spectral distribution as the speech, then a broadband criterion should be adopted. The effectiveness of different designs can be compared by using the improvement in the AI as the performance metric.

Conclusions

The goal in hearing aid design is to produce an instrument that will ameliorate the problems of impaired hearing. The auditory system embodies a non-linear frequency analysis in which the filter gain and bandwidth depend on the signal level and frequency content, and impaired hearing actually involves a reduction of the degree of nonlinearity in the system. An ideal hearing aid would have to restore the cochlear compression and frequency analysis to match those of the normal ear, but this is impossible because we have access only to the auditory system acoustic input and not to the output from the cochlea. Thus hearing aid design is a series of compromises, first in achieving known objectives, such as low-distortion amplification within the confines of a cosmetically-acceptable package, and second in determining performance objectives for more effective processing algorithms given the complexities of the human auditory system.

Despite the desire for more sophisticated processing algorithms, much can be achieved in the evolutionary improvement of the conventional hearing aid. Quieter circuits and greater amplifier headroom are needed to give enough dynamic range to reproduce all speech sounds (conversational levels and the talker's own voice) without undue distortion. Wider bandwidth is needed for optimal speech intelligibility, and feedback reduction is needed to achieve the desired high-frequency gain. Compression circuits with low compression ratios and low compression thresholds will be more effective than today's compression limiting in matching the dynamic range of speech and music to that of the impaired ear, and rapid compression time constants will not be needed because an amplifier with adequate headroom will not clip on normal input stimulus levels. Directional microphones or small microphone arrays are needed, because these have been shown to constitute the only viable noise suppression approach. These design elements, taken together, will lead towards a high fidelity hearing aid, both to serve as an effective instrument in its own right and to become the platform for new algorithmic developments in the future.

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