Future Directions in Hearing Aid Research

Orientations futures de la recherche sur les prothèses auditives

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Abstract

Two broad avenues of research are addressed: (1) the development of more effective signal-processing techniques, and (2) the development of prescriptive procedures for programmable hearing aids with advanced signal-processing capabilities. Research on signal-processing techniques has focused on three inter-related areas: compression amplification, recoding of the speech signal (e.g., frequency lowering, feature extraction), and reduction of background noise. This trend is likely to continue, but with greater emphasis on adaptive methods of signal processing, such as automatic adjustment of the frequency-gain characteristic, and new methods of noise reduction. Of the various methods of noise reduction that have been explored, multi-microphone techniques appear more promising than the traditional single-microphone approach.

Research on prescriptive fitting procedures has, until recently, focused primarily on the frequency-gain characteristic. The development of prescriptive fitting procedures for compression amplification and other, more advanced signal processing systems, such as automatic frequency response hearing aids and multi-memory programmable instruments are beginning to receive increased attention. There is also increased interest in new methods of hearing aid assessment that take into account factors typical of everyday hearing aid use. An expanding research area with important implications for hearing aid prescription (and future hearing aid design) is the development of general theories for predicting speech intelligibility and overall quality of amplification. Recent research on extending the Articulation Index to include hearing impairment and the development of a similar index for specifying the audibility of non-linear distortion represents a major thrust towards improving our understanding of acoustic amplification for hearing impairment and, by so doing, providing a basic framework for future hearing aid research and development.

Résumé

L'auteur examine deux sujets de recherche : (1) la mise au point de techniques plus efficaces pour le traitement du signal; et (2) l'élaboration de procédures de prescription pour les prothèses auditives programmables qui comportent des fonctions perfectionnées au niveau du traitement du signal. La recherche sur les techniques de traitement du signal s'est concentrée plus spécifiquement sur trois domaines étroitement liés : l'amplification avec système de compression, le réencodage du signal sonore et la réduction du bruit ambiant. L'auteur anticipe que cette tendance devrait se maintenir et que l'accent portera davantage sur des méthodes adaptatives de traitement du signal, comme l'ajustement automatique du gain de fréquences, et sur de nouvelles méthodes de réduction du bruit. Parmi les diverses méthodes de réduction du bruit qui ont été étudiées, les techniques multi-microphoniques semblent plus prometteuses que la méthode conventionnelle à un microphone.

Jusqu' à tout récemment, la recherche sur les procédures prescriptives d'ajustement s'était surtout concentrée sur le gain de fréquences. La mise au point de procédures prescriptives d'ajustement pour l'amplification avec système de compression et d'autres systèmes plus perfectionnés de traitement du signal, comme des prothèses auditives avec réponse automatique aux fréquences et des instruments programmables multimémoires, commencent à retenir l'attention des chercheurs. On s'intéresse également de plus en plus à de nouvelles méthodes d'évaluation prothétique qui tiennent compte de facteurs propres à l'utilisation journalière des prothèses auditives. Un domaine de recherche de plus en plus important, qui comporte d'importantes répercussions pour les prescriptions de prothèses auditives (et la conception future des prothèses auditives), concerne l'élaboration de théories générale de l'amplification. Une recherche récente sur l'élargissement de l'index d'articulation de manière à inclure la déficience auditive ainsi que l'élaboration d'un index similaire pour déterminer l'audibilité de la distorsion non linéaire constituent des progrès importants pour améliorer notre compréhension de l'amplification acoustique pour la déficience auditive; ces développements fournissent également un cadre de base pour la recherche-développement future sur les prothèses auditives.

Background

Any attempt at predicting future trends must be accompanied by substantial caveats because the only prediction that can be made with confidence is that the unexpected is likely to occur. On the other hand, current trends have a momentum that is not easily altered in the near term. It is thus possible to make some useful projections for the near future, but long-term projections must be interpreted with caution. Having made the necessary disclaimers, current trends will be examined in order to identify worthwhile avenues of investigation for the near future. These projections will be garnished with a few long-term speculations.

A particularly important recent development has been the introduction of digitally controlled, programmable hearing aids. Almost every major hearing aid manufacturer has recently introduced a programmable instrument of some kind. Although a few far-sighted researchers anticipated this development from the outset, the rapidity with which programmable hearing aids have reached the marketplace has exceeded most expectations.

When the digital hearing aid was first developed, it was viewed primarily as a research tool and it was felt that it would be some time before a practical wearable digital hearing aid could be developed (Levitt, 1982a). Contrary to expectation, several major companies with no previous experience in hearing aid development, but with strong technological expertise, entered the field. Unburdened by conventional wisdom, they developed instruments that were new in both concept and design. Traditional hearing aid companies soon followed with competing designs.

All of the new programmable hearing aids offer a practical compromise between the conflicting demands of high power consumption (for advanced signal processing capabilities) and small size, including correspondingly small battery size, for user convenience and cosmetic acceptability. The nature of this compromise is typically that of an analog audio channel harnessed to a small digital controller. This arrangement incorporates many of the advantages provided by experimental, all-digital hearing aids. These advantages include programmability, precise computer controlled adjustment of electroacoustic characteristics, new options, such as multiple memories for changing the electroacoustic characteristics of the hearing aid at the touch of a button, and improved methods of signal processing.

The current generation of programmable hearing aids, however, does not have the advanced signal processing capabilities that characterized the experimental, all-digital hearing aids. These early experimental units included advanced forms of signal processing, such as spectral subtraction for noise reduction and generalized forms of frequency-dependent compression (Levitt, Neuman, Mills, & Schwander, 1986; Levitt, Neuman, & Sullivan, 1990).

An important recent technological advance is that a conventional hearing aid can now be made small enough to fit entirely in the ear canal. Because there is little to be gained cosmetically in making hearing aids even smaller than this, it seems likely that future engineering development will focus on problems other than further reductions in the size of the hearing aid. There is little doubt that the electronic chip industry will continue the ongoing process of reducing the size and increasing the complexity of integrated circuit chips. In the application of this technology to hearing aids, further engineering efforts are now more likely to focus on greater signal processing capabilities in the space available rather than further reductions in the size of the hearing aid. The recent introduction of programmable, multiband, in-the-ear hearing aids by several manufacturers is indicative of this change in focus.

The above developments have important implications for future hearing aid research. Two broad avenues of research that are particularly relevant to current engineering advances are: (1) the development of more effective signal processing techniques, and (2) the development of prescriptive fitting procedures for programmable hearing aids with advanced signal processing capabilities.

Advances in Signal Processing

Previous research on signal processing for hearing impairment has focused on three interrelated areas: (1) compression amplification; (2) recoding of the speech signal (e.g., frequency lowering, feature extraction); and (3) reduction of background noise. Whereas there has been significant research activity in each of these areas, the results obtained thus far have not shown any dramatic improvements in making speech easier to understand, although there have been other important improvements.

Compression Amplification

Much of the recent research on compression amplification has been on frequency-dependent forms of compression, such as multiband compression. Although early experimental evaluations of multiband compression showed promising results (Villchur, 1973; Yanick, 1976), subsequent experiments with a wide range of experimental multiband systems did not show the anticipated improvements in speech intelligibility (Abramovitz, 1980; Braida, Durlach, Lippman, Hicks, Rabinowitz, & Reed, 1979; DeGennaro, Braida, & Durlach, 1986; Walker, Byrne, & Dillon, 1984). It should be noted that the early experiments, which showed improved intelligibility, used a less than optimum control condition (linear amplification with uniform gain) in evaluating the experimental multiband compression system. On the other hand, most of the experiments that have not shown improvements in intelligibility for multiband compression used speech stimuli that did not vary significantly in level, thereby biasing the experiments against multiband compression (Villchur, 1982). The question as to whether multiband compression is superior to a well designed, properly prescribed, single-band compression amplifier remains unresolved and is worthy of further investigation. Although previous experiments have served to muddy the waters to some extent, they have nevertheless also provided some useful insights.

Some form of compression or limiting is clearly necessary for signals covering a dynamic range substantially greater than the listener's range of residual hearing. The dynamic range of the amplified signal can be reduced in several ways, such as by peak clipping, compression limiting, wide dynamic range compression, single or multiband compression, or some combination of the above (Preves, 1991). A very simple form of amplitude limiting is that of peak clipping. Peak clipping introduces substantial distortion, although much of the distortion occurs in the higher frequencies and may not be audible in cases of substantial high-frequency loss. Compression amplification is thus advantageous relative to clipping in that the dynamic range of the amplified signal can be reduced with relatively little distortion, but this advantage only holds for those conditions in which the dynamic range of the signal exceeds the available range of residual hearing and only for those subjects who are sensitive to clipping distortion.

The above caveats may help explain why compression amplification has not always shown superior results to conventional linear amplification with peak clipping, the clipping operation occurring only at very high sound levels in order to protect the subject. Even if it is accepted that compression amplification is superior to peak clipping (because of the non-linear distortion produced by peak clipping), there is considerable uncertainty as to which form of compression amplification should be used. Perhaps the most pressing question is whether multiband compression is superior to single-band compression. The most positive results obtained thus far with multiband compression amplification are those of Moore and his colleagues (Laurence, Moore, & Glasberg, 1983; Moore, 1987; Moore & Glasberg, 1986). These studies have shown that small improvements in intelligibility can be obtained with a well designed two-band compression amplifier. The variables of the compression system, however, have to be carefully adjusted so that the characteristics of the amplified signals are matched appropriately to the user's residual hearing.

The advantage of two-band compression over single-band compression applies primarily to situations in which there are large differences in the relative power of the acoustic signal between the low and high frequency bands. Such differences do occur, but they are limited in frequency of occurrence, and some subjects are more susceptible than others to the effects of improper matching of signal dynamic range to residual hearing range in these two frequency bands. The above caveat is even more limiting than that noted earlier for single-band compression. It may also help explain why so many experimental evaluations of two-band compression have not yielded the anticipated improvements over single-band compression despite the theoretical advantages of two-band compression. Similarly, it could be argued that the advantages of N-band compression (N much greater than 2) relative to two-band compression are restricted to an even smaller range of listening conditions and types of hearing impairment.

Individual differences are of great importance in the design and evaluation of multiband compression systems. As noted above, multiband compression amplification may be of benefit to only a subset of hearing impaired listeners. Recent experiments on new forms of amplitude compression provide some insights as to why individual differences should be so large with respect to the relative benefits provided by compression amplification. In orthogonal-polynomial compression, the short-term speech spectrum is approximated by a series of polynomials of increasing complexity (Levitt & Neuman, 1991). As the complexity of the polynomials is increased, the proportion of the short-term speech spectrum that is compressed into the residual hearing area is increased correspondingly. There is also a corresponding decrease in the amount of phonetic information conveyed by spectrum shape. For some subjects, the additional information provided by an increase in the proportion of the speech signal that is audible is more important than the loss of spectral shape information introduced by the compression process. For other subjects, the reverse appears to be the case.

In essence, Levitt and Neuman (1991) argued that the individual differences obtained in the evaluation of orthogonal polynomial compression resulted from differences in the balance between the gains provided by increased audibility and the losses resulting from the reduction in spectral shape information. Because the optimum compression parameters for a given individual depend on a compromise between two conflicting factors, small individual differences in the relative importance of each factor can result in large individual differences when the factors are balanced against each other. This situation is in some ways analogous to the relative sensitivity of a cancellation or null technique involving two equally large components. When one component is altered slightly relative to its own size, the magnitude of the cancellation term is changed substantially relative to its size.

The above interpretation provides a useful framework for addressing the problem of individual differences. Research on the nature of individual differences is not only important in furthering our understanding of compression amplification, but also critical for the development of effective prescriptive procedures and may also change our thinking with respect to how compression amplification should be evaluated experimentally. Thus far, most of the experimental evaluations of compression amplification have used speech intelligibility as the primary measure. Although improved speech intelligibility is an important criterion in evaluating hearing aids, it is not the only relevant criterion. Overall sound quality, ease of listening, comfort, and other subjective attributes are also important considerations.

The effects of multiband compression on the various attributes of speech are complex and deserving of further investigation, particularly for the case of speech in noise. In some cases, reduction in gain in one frequency region may produce an increase in both speech intelligibility and sound quality because of reduced upward spread of masking. In other cases, there may be a small reduction in intelligibility if some important frequency components of the speech signal are reduced in intensity, although overall sound quality may be improved because of the reduction in noise level.

There are many possible methods of amplitude compression, of which multiband compression forms an important subset. Advanced forms of amplitude compression, such as orthogonal-polynomial compression (Levitt & Neuman, 1991) and principal-component amplitude compression (Bustamante & Braida, 1987), are very useful as research tools, but are unlikely to be incorporated in a practical hearing aid. On the other hand, a relatively simple form of amplitude compression, which can be implemented in a practical hearing aid, is that of automatically adjusting the frequency-gain characteristic as a function of overall signal level (Killion, Staab, & Preves, 1990). Automatic frequency response adjustment can be quite effective in altering the dynamic range of amplified sound and may be viewed as a limiting form of multiband compression.

Skinner (1980), Sullivan, Levitt, Hwang, and Hennessey (1988), and others have shown that the optimum frequency-gain characteristic for speech varies as a function of signal level; that is, there is no single optimum frequency-gain characteristic. In light of this observation, the traditional design of hearing aids with a fixed frequency-gain characteristic may not be the best approach. In principle, it should be possible to adjust the gain and frequency response of a hearing aid adaptively so as to approximate the optimum frequency-gain characteristic as the speech signal varies in level. This is not too difficult a problem for speech in quiet; automatic estimation of the optimum frequency-gain characteristic in noise, however, is a considerably more difficult problem, as discussed in the next section.

Several types of AFR (Automatic Frequency Response) hearing aids have been developed (Goldberg, 1972; Graupe, Grosspietsch, & Basseas, 1987; Killion, 1990), and researchers are actively engaged in evaluating these instruments (see Fabry, 1991, and references cited therein) as well as exploring the characteristics of experimental AFR systems (van Dijkhuizen, Festen, & Plomp, 1989). Initial results from these ongoing evaluations indicate that, under appropriate conditions, simple forms of AFR can be of significant benefit.

An underlying objective in compression amplification is that of raising the level of the weaker sounds relative to that of the stronger (i.e., more intense) sounds. Because the weaker sounds are mostly consonants and the stronger sounds are mostly vowels, compression amplification can be viewed as a method of adjusting the consonant/vowel ratio. It has been shown that raising the level of the weaker consonants, for example, raising consonant levels to that of the neighboring vowels, can improve intelligibility (Gordon-Salant, 1986; Freyman & Nerbonne, 1989; Montgomery & Edge, 1988). These level changes are similar to those produced by wide dynamic range compression with short time constants. A study of the optimum C/V ratios for maximizing consonant identification, however, has shown that signal level is not the only factor affecting optimization of the C/V ratio (Kennedy & Levitt, 1990). Phonetic environment, spectral and temporal structure, and the configuration of the hearing loss are other important variables influencing the optimum C/V ratio. These findings suggest that a compression system that operates solely on relative signal levels is unlikely to maximize consonant intelligibility and that additional characteristics of the acoustic signal (including the acoustic-phonetic properties of neighbouring phonemes) in combination with residual hearing characteristics need to be taken into account.

Signal Recoding

A form of signal processing that has received attention intermittently over the years is that of signal recoding. A simple form of recoding is that of frequency lowering. Various forms of frequency lowering have been tried with mixed results. Most of the attempts involved radical recoding of the speech signal for persons with severe to profound hearing impairments. Typically, the high frequency components of speech are transposed downward to frequencies below about 1000 Hz. Experimental evaluations of these systems, for the most part, have not shown significant improvements in speech intelligibility (Ling, 1969). It has been argued that the transposed speech sounds may not be recognizable initially, but that it should be possible to learn these new sounds and that, with training, frequency lowering may be more effective than conventional amplification for severe high-frequency hearing impairments. There is some evidence to support this view, as demonstrated by Foust and Gengel (1973), Hicks, Braida, and Durlach (1983), and Reed, Schultz, Braida, and Durlach (1985), but there have been relatively few studies involving long-term auditory training with frequency lowered speech.

A form of frequency lowering that does not distort the speech signal quite as much is that of phonetically based transposition, in which only a small class of speech sounds is lowered in frequency (Guttman & Nelson, 1968). The voiceless fricatives are ideal candidates for this form of transposition because the acoustic cues distinguishing one fricative from another are conveyed almost entirely by the high frequency components of the signal (Johansson, 1966). These high frequency components are typically not audible, even with amplification, in cases of severe or profound hearing impairment. Further, the voiceless fricatives also contain few important low frequency cues, which could be masked by the superposition of transposed high frequency cues. Experimental investigations of fricative transposers have yielded positive results for both speech training and improved speech recognition (Johansson, 1966; Guttman, Levitt, & Bellefleur, 1970; Velmans, Marcuson, Grant, Kwiatkowski, & Rees, 1988). Phonetically-based transposition has been extended recently to include all consonants except those with significant low frequency content, such as glides and nasals (Posen, Reed, & Braida, to appear). The results indicate that, with training, significant improvements in consonant recognition can be obtained.

Another form of frequency lowering that has yielded positive results is that of reducing all frequencies proportionally by a small amount. Signal processing techniques have already been developed that can lower speech frequencies by 20 to 30 percent without radically distorting the speech signal. The perceptually most salient distortion is that the voice pitch is lowered substantially. A small child with a high pitched voice, for example, is made to sound like a large male with a deep bass voice. An experimental evaluation of moderate frequency lowering showed a small but significant improvement in speech intelligibility for persons with moderate hearing losses when listening to speech produced by a female talker (Mazor, Simon, Scheinberg, & Levitt, 1977). More advanced forms of frequency lowering using this approach do not alter the voice fundamental frequency but lower frequencies above the fundamental (Hicks, Braida, & Durlach, 1981).

The impetus for developing the technology for frequency lowering derived from the need to normalize the speech frequencies of deep sea divers who operate in a helium atmosphere. Frequency lowering is also used for reducing the frequency spectrum of speeded speech in talking books for the blind. Circuits for frequency lowering are thus already well advanced, and a BTE hearing aid incorporating both fricative transposition and moderate frequency lowering has recently been developed. Research on frequency lowering is continuing, but at a relatively slow pace because only a small number of research groups appear to be interested in this form of signal processing. Hopefully, the recent introduction of a BTE instrument that provides two promising forms of frequency lowering will spur renewed interest in the clinical applications of frequency lowering.

Another form of signal processing that appears promising for severe hearing impairments is that of exaggerating the acoustic cues conveying important speech information. Revoile, Holden-Pitt, Edward, and Pickett (1986), for example, have shown significant improvements in the identification of fricative consonants by increasing the intensity and duration of the frication component of each sound. Increased vowel duration is an important phonetic cue indicative of voicing in the following consonant (Raphael, 1972), and significant improvements in the perception of voicing have been obtained by enhancing this cue (Revoile, Holden-Pitt, Edward, & Pickett, 1986; Revoile, Holden-Pitt, Pickett, & Brandt, 1986).

Some degree of automatic speech feature extraction is needed in order to implement strategies of the above type for improving speech recognition. The state of the art in automatic speech recognition is such that advanced speech processing hearing aids of the above type may be feasible in the not too distant future. Uchanski, Delhorne, Dix, Braida, Reed, and Durlach (in press), for example, have shown that using current speech recognition technology, classes of speech sounds can be extracted with sufficient accuracy for practical cued speech applications. Pilot experiments by the author using a single word, automatic speech recognizer followed by speech synthesis with exaggerated phonetic cues showed that, in principle, improved intelligibility can be obtained using this approach. At present, however, even the most successful automatic speech recognition systems are limited in vocabulary size, require unacceptably long processing times (for this application), and, in many cases, also require a cooperative talker who will articulate each utterance one word at a time. These developments, nevertheless, open up exciting new avenues of investigation for the long-term future.

At present, two speech features that can be extracted automatically with reasonable reliability (at least in a quiet, non-reverberant environment) are frication and voicing. Speech feature hearing aids that depend on automatic extraction and processing of frication have already been developed in the form of fricative transposers. Speech training aids in which the voice fundamental frequency, Fo, is extracted and displayed either visually or tactually have also been developed (Levitt, Pickett, & Houde, 1980). Breeuwer and Plomp (1986), Boothroyd, Hnath-Chisolm, Hanin and Kishon-Rabin (1988), and others have shown that lipreading scores can be improved by presenting the voice fundamental frequency, Fo, as an auditory supplement. An experimental hearing aid that presents Fo in a perceptually salient way has been developed by Villchur and Killion (1976). The amplified acoustic signal is interrupted at a rate corresponding to Fo. Informal evaluations showed that severely to profoundly hearing impaired individuals could recognize changes in Fo using this device although they could not detect these changes using conventional amplification.

More recently, Rosen, Walliker, Fourcin, and Ball (1987) developed a feature extraction hearing aid in which Fo is presented auditorially using a time-varying sinusoidal signal with a frequency proportional to Fo. The range of variation of this sinusoidal signal is exaggerated so as to make the variations in Fo easier to recognize. Experimental evaluations of this experimental hearing aid showed significant improvements in lipreading ability. Speech feature hearing aids of this type appear promising for persons with very severe hearing impairments who are unable to benefit from the use of a conventional hearing aid.

Signal Processing for Noise Reduction

Methods of noise reduction for hearing aids can be subdivided into two broad categories: single-microphone techniques and multi-microphone techniques. Most of the research thus far has focused on single-microphone techniques, presumably because of the convenience of using only one microphone in a hearing aid. Signal processing for noise reduction using only one microphone, however, is a very difficult problem. Research on this topic has received considerable attention over the years because there are numerous practical applications for this form of noise reduction. Despite this effort, a satisfactory solution to the problem has yet to be achieved, although many different approaches have been tried with interesting results (Lim & Oppenheim, 1979; Lim, 1983). Because noise reduction requirements for hearing aids differ in some respects from the broader requirements of noise reduction for normal hearing people, research focusing on the special requirements of hearing impaired listeners and utilizing the general techniques already developed would appear to be the most fruitful avenue of investigation.

The simplest method of noise reduction for hearing aids is the use of a directional microphone. This approach has been tried but is not widely used even though significant improvements in intelligibility can be obtained. These improvements can be substantial when the speech and noise sources are spatially separate. A directional microphone is of limited value, however, in a highly reverberant room or when the speech and noise signals come from the same direction.The limited use of directional microphones in modern hearing aids, despite their measurable advantages, raises important questions regarding the viability of this approach. These questions are particularly relevant at the present time because many of the multi-microphone techniques that are currently being developed for noise reduction have characteristics that are very similar to those of a directional microphone.

Another relatively simple method of noise reduction is that of frequency filtering. A filter that attenuates those frequency bands in which the noise level exceeds that of the speech will lower the overall noise level without reducing speech intelligibility. This type of filtering is effective in improving overall sound quality. It may also improve intelligibility to a limited extent by reducing upward spread of masking. A difficulty in implementing this form of filtering is that the spectrum of the speech signal (and also, in some cases, the spectrum of the noise) varies over time. Adaptive time-varying filters have been developed that take the dynamic characteristics of speech and noise into account (Ono, Kanzaki, & Mizoi, 1983; Graupe, Grosspietsch, & Basseas, 1987). Unfortunately, there is no simple rule for optimizing the design of such filters and experimental evaluations of the time-varying filters that have been developed have yielded mixed results (Fabry, 1991; Rankovic, Freyman, & Zurek, 1992).

One reason for the mixed results is that investigators have not always used appropriate control conditions against which to evaluate the effect of the signal-processing scheme. Some methods of signal processing distort the signal for both the experimental and control conditions. If the effect of this distortion is less severe for the experimental condition, then it may appear that the signal processing technique is effective in improving intelligibility and/or sound quality (Van Tasell, Thomas, & Crain, 1992). A more appropriate control condition would be to bypass the experimental signal processor entirely for the control condition and use a fixed frequency-gain characteristic that maximizes intelligibility (or some other attribute, such as overall sound quality) for the speech and noise stimuli being considered.

An approach to filtering that has a useful theoretical foundation is that of short-term Wiener filtering. It is possible to maximize signal-to-noise ratio using a Wiener filter for the special case of signals and noise that are statistically stationary. Although maximizing signal-to-noise ratio does not necessarily maximize intelligibility or overall sound quality, it is a useful step in the right direction. Because speech is not a statistically stationary signal (e.g., the speech spectrum varies with time), a Wiener filter is not strictly applicable. The spectrum of speech, however, is reasonably stable over short periods of time, and it is possible to use an approximation to the Wiener filter over these short time intervals. Because the assumptions underlying the short-term Wiener filter are not strictly valid, this filter may improve but not necessarily maximize the speech-to-noise ratio.

Experimental evaluations of short-term Wiener filters with normal hearing listeners have not shown significant improvements in intelligibility (Lim & Oppenheim, 1979), but an experiment with hearing impaired listeners showed significant improvements for half of the subjects tested (Bakke, Neuman, & Toraskar, 1987). A possible explanation of this result is that the auditory system first divides the signals to be processed into critical bands. The widths of these critical bands in normal hearing listeners are relatively narrow, whereas for some hearing impaired individuals the critical bands can be quite broad. The gain in speech-to-noise ratio provided by a Wiener filter is relatively small for critical bands of narrow bandwidth and thus may not be of much help to normal hearing listeners, but may provide significant improvements for hearing impaired listeners with critical bandwidths that are much wider than normal. This hypothesis, or other hypotheses attempting to explain why only some hearing impaired listeners appear to benefit from this form of amplification, need to be evaluated.

A method of noise reduction that can also produce substantial improvements in speech-to-noise ratio is that of spectrum subtraction. In this procedure, the short-term spectrum of the incoming signal is obtained. This signal might consist of speech plus noise (S+N) or noise only (N). The latter would typically be obtained during a pause in the speech. A decision is made, based on the relative amplitude and harmonic structure of the signal, as to whether the short-term spectrum is of S+N or N only. If the signal is believed to consist of N only, the short-term spectrum is stored in memory as an estimate of the noise spectrum. If the short-term spectrum is believed to consist of S+N, then the most recent estimate of the noise spectrum is subtracted from the S+N spectrum resulting in a short-term spectrum with a much improved speech-to-noise ratio.

A variation of the spectrum subtraction technique is the method developed by Weiss and Aschkenasy (1981) in which the short-term spectra are subjected to an additional square root transformation. This modification, as well as other refinements of the spectrum subtraction technique, were arrived at empirically. Experimental evaluations of the Weiss-Aschkenasy technique (Levitt, Neuman, Mills, & Schwander, 1986) have shown significant improvements in judgments of overall sound quality, but no significant change in speech intelligibility. Similar results have been obtained for both normal hearing and hearing impaired listeners. As before, significant individual differences were observed.

The Weiss-Aschkenasy technique has found a useful practical application for those situations in which a listener is required to listen to speech in noise for long periods of time. Thus, although traditional experimental evaluations of the Weiss-Aschkenasy technique (e.g., conventional intelligibility tests) do not predict any advantage for the technique, in practice it has been found to be very useful, albeit for a rather specialized application. This observation raises concerns as to whether appropriate methods of measurement are being used in evaluating noise reduction systems. Essentially the same issue has been raised with respect to the evaluation of compression amplification and signal processing techniques for speech enhancement. This is an important area of research that deserves more attention.

A possible explanation for the large individual differences observed in the evaluation of noise reduction systems is that new forms of distortion (signal processing distortions) are frequently introduced in conjunction with the improvement in speech-to-noise ratio. For those subjects who are especially sensitive to the effects of noise, the reduction in noise level may be of greater importance than the signal processing distortions that are introduced. For other subjects, the reverse may be the case. An illustration of this problem is provided by Neuman and Schwander (1987) in their evaluation of a simple filtering procedure for reducing the masking effects of background noise. The frequency response of this filter (referred to as REDMASK for *Red*uced *Mask*ing) was adjusted individually for each hearing impaired subject such that the noise spectrum lay just above the threshold of audibility at all frequencies. Under these conditions, the noise is barely audible and not disturbing. The concomitant lowering of the speech signal, however, was disliked by many, but not all, of the hearing impaired subjects. As a consequence, those subjects who were very sensitive to the effects of background noise favored the REDMASK filter, while those with a high tolerance for noise did not. This situation is not unlike that observed in the assessment of orthogonal-polynomial compression in which a different balance between two opposing factors was obtained for each subject.

In contrast to the modest improvements obtained with single-microphone techniques, noise reduction techniques using two or more microphones have shown significant improvements. These include gains in intelligibility as well as overall sound quality. A relatively simple approach to the use of multiple microphones is to place several microphones along the frame of a pair of eyeglasses and to sum the outputs of these microphones (with appropriate weights and delays) using techniques commonly employed with sonar arrays. Microphone arrays of this type can be quite effective in improving speech-to-noise ratio because of the improved directional characteristics of the system. An experimental hearing aid using microphone arrays mounted on an eyeglass frame has been developed by Soede (1990) who reported improvements in speech-to-noise ratio on the order of approximately 7 dB for speech in a diffuse noise field, in comparison with an omnidirectional microphone. An additional advantage of this approach is that signal processing distortions are small in comparison with those typically obtained with single-microphone techniques.

A much greater degree of directionality can be obtained by using beam forming techniques that cancel sound coming from the direction of the noise source. A technique of this type, which appears particularly promising, has been developed by Peterson, Durlach, Rabinowitz, and Zurek (1987). A practical problem with beam forming techniques involving cancellation or nulling in a given direction is that the microphones and associated amplification channels must be matched extremely precisely in order to ensure that the cancellation operation is effective. This operation usually involves subtracting the outputs of two channels with relatively large signals, hence the need for precise matching. Adaptive signal processing techniques, however, can be used to optimize the accuracy of this subtraction procedure.

An adaptive method of noise cancellation that has been found to be extremely effective, but for a limited set of conditions, is that developed by Widrow et al. (1975). This technique requires that the noise be picked up by a separate microphone. The signal picked up by this microphone (the *reference* microphone) is passed through an adaptive filter producing an output N' which is then subtracted from the amplified output of the microphone picking up both speech and noise (S+N). The adaptive filter is adjusted systematically so as to maximize the difference, (S+N)-N'. Under ideal conditions, the adaptive filter adjusts the noise component N' to be identical to the noise in the S+N channel, thereby cancelling the noise completely.

The Widrow noise canceller as been applied to the problem of noise reduction in hearing aids by Chabries, Christiansen, Brey, Robinette, and Harris (1987). Substantial improvements in speech- to-noise ratio (exceeding 30 dB) have been obtained, but the technique is severely limited in practice, however, because of the need to place the reference microphone at or near the noise source. This is not always feasible. One approach to this problem is to mount both microphones on the head and to treat the microphone furthest from the speech source as the reference microphone. The adaptive noise canceller will not work perfectly under these conditions, but it can produce small improvements in speech-to-noise ratio, depending on the relative locations of the speech and noise sources.

In a variation of this approach, Weiss (1987) used two head- mounted microphones with different directional characteristics. An omnidirectional microphone picked up both speech and noise, while a second, directional microphone pointing towards the noise source picked up mostly noise. This approach does not require any spatial separation between the two microphones.The use of an adaptive canceller with two head-mounted microphones having different directional characteristics has been evaluated by Schwander and Levitt (1987). This study showed a significant improvement in speech-to-noise ratio and a corresponding improvement in intelligibility. Of particular interest was the effect of head movements on the cancellation procedure. Although head movements were found to reduce the gain in speech-to-noise ratio, hence reducing the corresponding improvement in intelligibility, the magnitude of the reduction due to head movements was small.

Multi-microphone techniques, however, have a major practical disadvantage. Linking two or more microphones to a single processor is inconvenient and may not be practical for many hearing aid applications. The use of an eyeglass frame for mounting the microphones represents one approach to the problem. Another approach is to use some form of wireless transmission (e.g., radio or infrared signals) to link the microphones to a central processor. This arrangement could be quite practical with non-wearable assistive listening devices. A third approach is to use the method described above in which microphones with different directional characteristics are used. In this case, the microphones can be placed close to each other (e.g., in the same hearing aid case), and adaptive signal processing used to optimize the overall directional characteristics of the system. The essence of this approach is to replace a fixed directional microphone with one having adaptive directional characteristics. A further refinement would be to have this adaptive system automatically focus on the speech source.

There appear to be several ways in which multi-microphone systems can be made to be more practical for hearing aid applications. This is a rich area for further investigation. It is important to bear in mind, however, that multi-microphone techniques typically have many of the limitations of conventional directional microphones; that is, the advantages of directionality are lost in a highly reverberant environment or in situations in which both speech and noise come from the same spatial location.

Prescriptive Fitting Procedures

Frequency-Gain Characteristics

Much of the recent research on prescriptive fitting procedures has focused on prediction of the optimum frequency-gain characteristic from audiological data. This research has led to several different approaches to the derivation of this frequency-gain characteristic (Berger, Hagberg, & Rane, 1978; Byrne & Dillon, 1986; Libby, 1986; McCandless & Lyregaard, 1983; Seewald, Zeliksko, Ramji, & Jamieson, 1991). These approaches provide very different prescriptions (Byrne, 1987), and there is no general agreement as to which approach should be used in practice.

The above situation is disconcerting in several respects. If there is an optimum frequency-gain characteristic, then it is important for the hearing aid dispenser to know how to prescribe that frequency-gain characteristic. Alternatively, it may be that the exact shape of the frequency-gain characteristic is not critical and that any one of a range of different frequency-gain characteristics could yield satisfactory results. If the latter situation is true, then perhaps too much effort has been spent in searching for *the* frequency-gain characteristic at the expense of neglecting other electroacoustic characteristics. A third possibility is that the exact shape of the frequency-gain characteristic is critical but that the optimum shape varies as a function of the signals being amplified.

It is important in discussing issues of optimization to specify the criteria that determine what is optimum. A frequency-gain characteristic that maximizes intelligibility, for example, may be optimum for intelligibility, but it may not necessarily be the optimum frequency-gain characteristic for overall sound quality. Sullivan, Levitt, Hwang, and Hennessey (1988) compared prescriptive fitting procedures representative of four major approaches to hearing aid prescription. These were an audiogram- based approach (Lybarger, 1978), an approach based on both the audiogram and the average speech spectrum (Byrne & Tonnison, 1976), a supra-threshold most-comfortable-loudness approach (Skinner, Pascoe, Miller, & Popelka, 1982), and an adaptive procedure (Levitt, Sullivan, Neuman, & Rubin-Spitz, 1987). The results showed that no single prescriptive procedure was uniformly better than any of the other procedures considered. Three different signal levels were used in this study, and it was found that the best frequency-gain characteristic varied as a function of signal level. The same result was obtained whether intelligibility or overall sound quality were used as the criterion for determining which was the best frequency-gain characteristic. It should be noted that the adaptive procedure was used to estimate the optimum frequency-gain characteristic at one signal level only and that the estimated frequency-gain characteristic was not altered for the other signal levels used in the experiment. The adaptive procedure yielded the best frequency-gain characteristic at the signal level for which it was adjusted, but not at other signal levels.

The above results support the view that there is an optimum frequency-gain characteristic, but that this optimum varies as a function of the signals being amplified. The experiment of Sullivan, Levitt, Hwang, and Hennessey (1988) varied signal level only. Other experiments have shown that the estimated optimum frequency-gain characteristic varies also as a function of the level and spectrum of the background noise (Van Tasell, Larsen, & Fabry, 1988). Data obtained by Levitt and Collins (1980) showed that the slope of the frequency-gain characteristic for maximizing intelligibility is steeper for nonsense syllables than for sentence length material. Further, in all of the above experiments, significant individual differences were obtained in estimating the optimum frequency-gain characteristics.

The above situation presents a very difficult challenge for hearing aid prescription which has not, as yet, been addressed adequately. Given a modern programmable hearing aid in which the frequency-gain characteristic can be programmed with a high degree of precision and given that practical procedures for adjusting this frequency-gain characteristic are available for maximizing a desirable attribute (e.g., intelligibility, overall sound quality) for a given acoustic environment, the dispenser is still faced with a difficult dilemma. Should the frequency-gain characteristic be optimized for the listening condition that the hearing aid user encounters most frequently in everyday life, even though that frequency-gain characteristic may not be a good choice for other listening conditions? Alternatively, should a frequency-gain characteristic be chosen that is not necessarily optimum for the most common listening condition, but is reasonably good for a wide range of different listening conditions? A compromise of some sort is obviously necessary. In practice, unfortunately, the selected frequency-gain characteristic is often the result of a very crude guess as to what is needed rather than a carefully determined compromise based on the different listening conditions that the hearing aid user will typically encounter.

Impact of New Technology

A very useful feature in many of the new programmable hearing aids is that of multiple memories allowing for different sets of electroacoustic characteristics to be selected at the touch of a button. This feature, in principle, will allow the hearing aid user to select an appropriate set of electroacoustic characteristics for several different acoustic environments. It is, however, necessary for the dispenser to program the electroacoustic characteristics to be stored in memory. For example, one memory may be programmed for speech in quiet, the other for speech in environmental noise, a third for a competing speech background, and so on. The prescription of these different sets of electroacoustic characteristics is not a trivial problem and represents a major challenge for further research in prescriptive fitting procedures. Note also that the problem is not limited to that of prescribing a set of frequency-gain characteristics but rather that of prescribing electroacoustic characteristics in general, including various forms of compression.

Current technological developments are also likely to add another dimension to this problem — that of automatic adaptive adjustment of electroacoustic characteristics. This is already the case with the K-Amp™ (Killion, 1990) as well as with various forms of adaptive filtering (Graupe, Grosspietsch, & Basseas, 1987) and adaptive compression™ (Gittles & Wilson, 1987). It is likely that this trend will continue and, given the large individual differences that have been observed in experimental evaluations of advanced signal processing schemes, the need for prescriptive fitting procedures will need to be developed for these adaptive methods of signal processing (Bentler, 1991).

Another difficult problem which needs to be addressed is that of acclimatization (Gatehouse, 1989). Long-term exposure to a given set of electroacoustic characteristics not only may result in a preference for those electroacoustic characteristics, but also the objective measures of performance (e.g., speech recognition scores) may also be higher for those electroacoustic characteristics. In a recent experiment, for example, Gatehouse (1992) found that new hearing aid users initially did not show differences in performance between amplification with a flat frequency response and amplification with frequency shaping. After long-term exposure (6 to 12 weeks) to a hearing aid with frequency shaping, however, significant improvements in intelligibility were obtained for amplification with frequency shaping.

The above observation supports the importance of and need for auditory training, but it also poses a difficult problem for the prescriptive fitting of hearing aids because the assessment of what is best for the subject depends not only on the electroacoustic variables being manipulated, but also on the subject's previous history of amplification. It is also necessary to distinguish between acclimatization, auditory deprivation effects, learning, and other effects. Silman, Gelfand, and Silverman (1984), for example, observed a decrement in speech identification scores in the unaided ear for long-term monaural hearing aid users. They ascribed this effect to auditory deprivation of late onset, although Gatehouse (1992) does not agree with this interpretation.

Acclimatization, learning, and other effects have also been found to be substantial among new hearing aid users (Cox & Alexander, 1992; Haggard, Foster, & Iredale, 1981; Scherr, Schwartz, & Montgomery, 1983). These observations cast doubt on the efficacy of a fitting procedure in which the prescription is based on evaluative data obtained when the hearing aid is first fitted. The above problem is not new, nor is it unique to modern signal processing hearing aids. It has, however, received renewed attention because programmable hearing aids provide a means for addressing the problem, for example, by reprogramming the hearing aid at regular intervals. In order to do this effectively, however, it is necessary to have a viable model of the adaptation process and the perception of sound by the impaired auditory system. There has been some research along these lines (Barfod, 1979), but much more needs to be done. Cox and Alexander (1992) provide some useful insights as to the nature of the problem in that they identified which aspects of hearing aid use were most susceptible to change over time. In particular, face-to-face communication with low background noise and communication in a noisy environment without visual cues showed the greatest long-term improvements in hearing aid benefit. These findings have important implications for the development of improved prescriptive procedures and the symbiotic relationship between prescriptive fitting and auditory training.

Programmable hearing aids also offer the means for addressing another troublesome problem — that of converting sound levels measured with a standard audiometer to sound levels generated in the ear canal by the hearing aid (Bentler & Pavlovic, 1989, 1992; Cox, 1979). In this application, the programmable hearing aid is used as a form audiometer during the fitting stage (Levitt, Sullivan, Neuman, & Rubin-Spitz, 1987).The hearing aid is first programmed to have as flat a frequency response as possible with mid-range gain and no compression. This is referred to as the *reference condition*. The hearing aid, in the reference condition and with the gain under program control, is used to measure the threshold of detection (DET), loudness growth function (LGF), and loudness discomfort level (LDL) for various test stimuli covering a range of frequencies (e.g., one-third octave bands of noise spaced at octave frequencies). The hearing aid is then prescribed using the values of DET, LGF, and LDL. (If an adaptive prescriptive procedure is employed then these measurements are used to obtain a first estimate of the optimum hearing aid setting.) Because all of the measurements are specified in terms of parameter values of the hearing aid, it is a relatively simple matter to program the hearing aid for this optimum setting. Also, because the hearing aid serves as its own sound delivery system during the measurement phase, there is no need to correct for any differences in acoustic coupling between transducer and eardrum. The value of DET obtained for the reference condition may be higher than the true auditory threshold because of internal noise in the hearing aid. The values obtained for DET thus should not be regarded as a measure of auditory sensitivity, but rather as an indicator of the lower bound of the available range of hearing for the subject wearing that hearing aid.

New Evaluative Tools

A particularly useful feature of many programmable hearing aids is the capability of switching rapidly from one set of electroacoustic characteristics to another. As a consequence, paired-comparison techniques can be used in evaluating different hearing aid settings and, more generally, in adaptively searching for the best set of electroacoustic characteristics for a given set of listening conditions. An example of this approach in the prescriptive fitting of hearing aids is provided by Neuman, Levitt, Mills, and Schwander (1987). More needs to be done, however, in modifying this technique for use with the programmable hearing aids that are currently available. Unlike the powerful digital master hearing aid used by Neuman, Levitt, Mills, and Schwander (1987), most programmable instruments are limited in terms of the variables that can be adjusted and their range of adjustment. The number of memories is also limited and paired-comparison procedures suitable for use under these constraints need to be developed and evaluated. Several researchers have begun to address the practical issues involved (Byrne, 1991; Kuk & Pape, 1992; Punch & Robb, 1992), but much more needs to be done given the complexity of the problem.

Paired-comparison techniques have many advantages. These include speed of testing, good test-retest reliability, sensitivity to small changes between experimental conditions, and the potential for using highly efficient adaptive procedures. Paired-comparison techniques are also not constrained to a single performance measure. Criteria such as overall sound quality, loudness, comfort, ease of listening, and other perceptual attributes have been used in addition to judgments of relative intelligibility. This is a very useful feature because there are many situations in which measurements of speech intelligibility do not differ significantly between two hearing aids, although other important attributes, such as sound quality, may show substantial differences. There is a limited, but very useful body of research on the perceptual attributes of hearing aids (for example, Gabrielson, Schenkman, & Hagerman, 1988; Punch, Montgomery, Schwartz, Walden, Prosek, & Howard, 1980). The paired-comparison technique used in conjunction with modern programmable hearing aids opens up new possibilities in terms of investigating the acoustic correlates of these attributes and also of providing a means for hearing aid prescription using performance criteria other than speech intelligibility.

An inherent limitation of paired-comparison judgments is their subjective nature. In the case of paired-comparison judgments of relative intelligibility, it is possible to compare these subjective judgments with objective measures of intelligibility. When differences in intelligibility are large, both objective and subjective measures of relative intelligibility have been found to yield essentially the same results, but when differences are small the two procedures yield slightly different results (Levitt, Sullivan, Neuman, & Rubin-Spitz, 1987; Sullivan, Levitt, Hwang, & Hennessey, 1988). Experimental evaluations of the differences between objective and subjective assessment of speech intelligibility are very revealing both in terms of identifying the factors affecting these judgments and in providing information on the subject's subjective biases (Byrne, 1991; Studebaker, 1991).

A related area of interest is that of self-report and questionnaire evaluations. There has been much recent activity in the development of these techniques and their application in evaluating hearing aid benefit (Cox & Gilmore, 1990; Demorest & Erdman, 1987; Tannahill, 1979; Walden, Demorest, & Hepler, 1984). The self-report approach, however, relies on subjective judgment, and there is concern regarding the reliability of these techniques. Data obtained by Cox, Alexander, and Gilmore (1991), and Cox and Alexander (1992) show good agreement between objective and subjective (self-report) measurements of hearing aid benefit for typical conditions of hearing aid use, except for face-to-face communication with high level background noise.

There is much scope for the development of objective evaluation procedures that are predictive of hearing aid benefit for conditions typical of everyday use, particularly for conditions involving face-to-face communication. Standardized lipreading tests could be used for this purpose, but these measurements would still not address the highly interactive nature of face-to-face communication. An innovative technique, which does take the interactive structure of human communication into account, is that of Continuous Discourse Tracking (DeFilippo & Scott, 1978). This technique can be used for both training and evaluation. Continuous Discourse Tracking (CDT) is widely used in the evaluation of cochlear implants and tactile aids, but has found relatively little application in hearing aid evaluation. One of the problems with CDT is that test-retest variability is high, including large inter-speaker differences (Schoepflin & Levitt, 1991). The major sources of variability in CDT can be reduced significantly by using video recordings of the speaker. This has been done using an interactive video disc system under computer control (Dempsey, Levitt, Josephson, & Porrazzo, 1992). The results showed significant reductions in test-retest variability as well as complete control of inter-speaker differences. A relatively simple approximation to the CDT procedure was used and more could be done in terms of improving the accuracy of the approximation. The use of computer controlled, interactive video has much to offer in terms of improving methods of hearing aid evaluation or for the evaluation and training of communication skills in general (Boothroyd, 1987, 1991).

Another measurement technique, which is readily implemented with modern computer-based systems, is that of measuring the subject's reaction time in a speech identification task (Gatehouse & Gordon, 1990; Pratt, 1981). Reaction times, for the most part, increase with increasing difficulty of the task; test items of low intelligibility will thus have longer reaction times. An exception to this rule occurs when the test item is so difficult that the subject immediately guesses at random, thus showing a small reaction time. It is important in using reaction-time measurements to guard against erroneous conclusions resulting from responses of this type. Another caveat is that individual differences in reaction time are very large and need to be taken into account. The technique is thus better suited for measuring relative rather than absolute performance (i.e., by measuring changes in reaction time within an observer rather than between observers).

A test procedure that is convenient to implement in terms of measuring reaction times is that of sentence verification. In this type of test the subject is simply required to verify whether the test sentence is true or false. In addition to providing reaction time data, the proportion of correct responses also provides a measure of how well the subject understood the test material. Further, the troublesome problem of very small reaction times occurring with very difficult test items (because of random guessing) is reduced considerably by considering only the correct responses when measuring reaction time. Two other features of sentence verification tests that are very appealing in terms of developing improved methods of hearing aid evaluation are: (1) the technique can be automated fairly easily with modern computer-based hearing aid systems, and (2) sets of sentences can be developed which show negligible learning effects even after repeated administrations (Levitt, 1984; Levitt & Neuman, 1990). In view of these advantages, greater attention should be given to the use of sentence verification techniques in hearing aid evaluation.

In summary, the current thrust in hearing aid evaluation is away from traditional speech intelligibility tests. This is partly because of the relative insensitivity of these tests to changes in hearing aid parameters as well as their relatively high test-retest variability. For tests involving sentence length material there is the additional complication of substantial learning effects on repeated administrations of the test (although, as noted above, this particular problem can be minimized using a sentence verification procedure with specially constructed sentence sets). A more important reason is that other relevant attributes, such as speech quality or ease of listening, are also important measures for hearing aid evaluation, which are not taken into account using traditional intelligibility tests.

A particularly important recent development, brought about by technological advances in the development of wearable master hearing aids, is that of field testing in which several different hearing aids (including competing commercial products) are simulated in a convenient wearable instrument (Cummins & Hecox, 1987). Field testing using wearable master hearing aids provides important information that cannot be obtained in the clinic or laboratory. Although evaluations of this type are difficult and expensive, they are of great value in assessing the practical benefits of new and experimental hearing aids.

General Theories

The problem of hearing aid prescription would be simplified considerably if a general theory were available for predicting the effects of acoustic amplification. This is a major research problem that has attracted the attention of researchers ever since hearing aids were first developed. A promising approach which has yielded useful predictions is that of modifying the Articulation Index for use with hearing impaired listeners (Dugal, Braida, & Durlach, 1980). The Articulation Index (AI) was first developed for predicting the intelligibility of speech over communication channels that were noisy and of limited bandwidth (French & Steinberg, 1947; Kryter, 1962). The AI procedure was developed for normal hearing listeners and although there were a few attempts at using the technique for hearing impaired listeners (e.g., Fletcher, 1952), it was not until Dugal, Braida, and Durlach (1980) modified the procedure that the AI technique received widespread attention as a means of predicting speech intelligibility in the hearing impaired. There are now several variations of the AI procedure for hearing impaired listeners (Dugal, Braida, & Durlach, 1980; Pavlovic, Studebaker, & Sherbecoe, 1986; Studebaker & Sherbecoe, 1991).

The underlying premise of the Articulation Index (AI) procedure is that speech intelligibility increases as the proportion of the spectrum that is audible is increased. Audibility in this context is determined by subdividing the speech signal into a set of contiguous frequency bands, each of which is assumed to contribute independently to intelligibility. Within each band, relative audibility is determined by the speech-to-noise ratio or speech-to-threshold ratio, whichever is lower. A maximum speech-to-noise (speech-to-threshold) ratio of 30 dB is allowed for each band because this is roughly the within-band dynamic range of the speech signal.

The various versions of the AI procedure that have been proposed differ in terms of the bandwidths of the contiguous frequency bands, the importance or weight assigned to each band in computing the AI, and the formulae used to take spread of masking into account. The adjustment for spread of masking is particularly important for high signal levels because at these levels the audibility of the weaker components of the speech signal is reduced (i.e., masked) by the more intense components. This effect can result in a performance-intensity function showing rollover; that is, intelligibility is increased as speech level is increased until a maximum is reached after which any further increase in speech level leads to a reduction in intelligibility. Rollover has been observed with hearing impaired listeners (Milner, Braida, Durlach, & Levitt, 1984) as well as with normal hearing individuals listening to speech in noise at high levels (Pollack & Pickett, 1958).

The AI procedure is consistent with the belief that the first step in prescriptive fitting of a hearing aid is to make the speech signal audible over a wide frequency range. However, making the speech signal audible, per se, is not necessarily sufficient to make the speech intelligible. Most forms of sensorineural hearing impairment also result in reduced resolution in processing suprathreshold signals. Reduced frequency resolution and reduced temporal resolution have often been measured in hearing impaired listeners in addition to elevated thresholds.

The AI procedure is well suited for taking reduced frequency resolution into account because the bandwidths used in computing the AI can be adjusted to match the critical bandwidths of the hearing impaired subject. Reduced temporal resolution as well as temporal spread-of-masking effects are not as easy to take into account. This is an issue that needs to be investigated further because, as shown by multivariate studies of the relative effects of reduced frequency and temporal resolution on intelligibility, poor temporal resolution shows a greater correlation with reduced intelligibility than poor frequency resolution (Levitt, 1982b).

An appealing aspect of the AI procedure is that the electroacoustic characteristics of a hearing aid that will maximize the AI can be determined fairly easily and, in principle, if intelligibility is a monotonic function of AI, then intelligibility should also be maximized. This has been tried (Levitt, Sullivan, Neuman, & Rubin-Spitz, 1987; Rankovic, 1991) but unfortunately, the conditions for maximizing AI involve relatively high signal levels and because of spread-of-masking effects and the inadequacy of the formulae used to correct for these effects, the monotonic relationship between the AI and intelligibility breaks down in the region where AI is maximized. Part of the reason for the inadequacy of existing formulae for correcting the AI for spread-of-masking effects is that experimental studies designed to validate the AI procedure typically have not used very high signal levels.

Research on improving the AI procedure for taking both temporal and spectral spread-of-masking effects into account is sorely needed. This research would not only substantially improve the accuracy of AI-based predictions of intelligibility at high signal levels, but also provide a practical means for designing hearing aids to maximize intelligibility by maximizing the AI. It is anticipated that substantial individual differences are likely to be encountered in this endeavor because the maximization of intelligibility will necessarily involve a compromise between two large, opposing factors — the increase in intelligibility produced by increasing the proportion of the speech spectrum that is amplified above the threshold of hearing, and the decrease in intelligibility produced by spread-of-masking effects resulting from the increase in signal level. The problem of large individual differences in establishing a compromise between conflicting effects is not unlike that noted earlier in optimizing parameter values in orthogonal-polynomial compression or, similarly, in optimizing parameter values in a noise reduction system.

In addition to a general theory for speech intelligibility, there is also a need for developing analogous theories for predicting sound quality from acoustic and audiological measurements. A first step in this direction would be a theory for predicting the audibility of non-linear distortion because one of the major factors responsible for reducing sound quality in hearing aids is that of perceptible distortion. A general measure for specifying hearing aid distortion, per se, would also be very useful.

Movement towards the development of a general measure of distortion for hearing aids has been spurred on by the growing use of digital signal processing techniques in audio systems. Digital signal processing techniques introduce new forms of non-linear distortion that cannot be handled using traditional methods of harmonic analyses. As a consequence, new measures of distortion are being developed to meet the challenge. A very useful measure of non-linear distortion is that obtained from coherence analysis. The coherence function, derived from the cross spectrum of the input and output signals of the system being measured, provides a measure of the non-linear distortion and/or noise generated by the system. Coherence analysis has been found to be a useful indicator of the non-linear distortion and internal noise generated by a hearing aid (Bareham, 1990; Dyrlund, 1989; Preves, 1990). Coherence analyzers also have been developed by the leading manufacturers of electroacoustic instruments, and the use of coherence analysis is growing, albeit slowly, because of the high cost of the instrumentation.

Another measure of non-linear distortion is that proposed by Williamson, Cummins, and Hecox (1987) in which a linear adaptive filter is programmed to simulate the transmission characteristics of the hearing aid. When the difference between the output of the hearing aid and the adaptive filter has been minimized, the adaptive filter represents a best-fit linear approximation to the hearing aid and the difference term represents an estimate of the non-linear distortion generated by the hearing aid. The results of this analysis are conceptually not very different from those obtained from coherence analysis which, in essence, also estimates an equivalent linear system. However, there may be important differences in the accuracy and precision of the two estimation procedures. Kates (1991) has identified several possible sources of error in coherence analysis. In particular, short-term coherence analysis using relatively small time windows can result in substantial errors of estimation. The accuracy and precision of the approach used by Williamson, Cummins, and Hecox (1987) depends on the adaptive algorithms used and the criterion for minimizing the distortion terms.

A measure of distortion that was proposed some time ago (Bumett, 1967) is based on the use of comb-filtered notched noise as the test signal. The reduction in the peak-to-valley ratio of the notched noise is used as the measure of distortion. The distortion term obtained from this analysis takes the form of a set of contiguous narrowband distortion components and is readily converted to an Articulation Index for predicting the effect of this distortion on intelligibility. The comb-filter or notched- noise approach to distortion measurement can be implemented using a modern personal computer, as demonstrated by Kates (1990) and Cudahy and Link (1991) in their development of a general purpose instrument for hearing aid measurement.

A more general approach to the measurement and specification of distortion is the Distortion Index proposed by Levitt, Cudahy, Jiang, Kennedy, and Link (1987b). This index is based on the concept of the Articulation Index except that in this case the distortion is treated as the signal and the amplified speech (or music) is treated as the background noise. The technique has been found to be effective in predicting the detectability of various forms of non-linear distortion (Levitt, Cudahy, Jiang, Kennedy, & Link, 1987; Cudahy & Kates, 1993). Of particular, interest for future research in this area is that the Distortion Index increases as the level of distortion is raised above the threshold of audibility and that under these suprathreshold distortion conditions the Distortion Index shows a negative correlation with ratings of sound quality (i.e., the larger the Distortion Index, the poorer the sound quality, Cudahy, 1992).

Research on generalized methods of specifying hearing aid distortion is both important and likely to result in significant advances given the rapid ongoing progress in measurement technology (Kates, 1991). Issues to be addressed are the accuracy and reliability of the various approaches that have been proposed and how the resulting measures relate to the perception of distortion by hearing aid users. It would be extremely valuable if these measures could be extended to address issues such as the prediction of sound quality from acoustic and audiological measurements.

Conclusions

Basic research and technological progress have exhibited a symbiotic relationship that has served both areas of endeavor extremely well. In some cases, basic research has provided the lead, resulting in major advances in technology. In other cases, technological innovation has leaped ahead providing researchers with new tools and new avenues of investigation.

Hearing aid research has benefited substantially from advances in technology. These advances, driven by the needs of the much larger computer and telecommunications industries, have leaped far ahead of current research efforts. Prescriptive fitting of hearing aids is a case in point. Until recently, a major focus of research in this area was that of developing formulae for predicting the optimum frequency-gain characteristic. There is still no general agreement as to which is the best method, but while the various schools of thought continued to refine their respective approaches, new hearing aids were being developed requiring prescriptive fitting procedures for a large number of electroacoustic variables in addition to (or instead of) the frequency-gain characteristic. Research on prescriptive fitting procedures for advanced, signal processing aids has a lot of catching up to do.

Whereas recent technological advances have introduced new dimensions to the problem of prescriptive fitting, this new technology has also introduced the means for addressing these problems. Computer controlled fitting procedures using highly efficient adaptive paired-comparison techniques is but one example of how advances in hearing aid technology have been helpful in improving methods of hearing aid prescription. This new technology has also served to address problems that have remained dormant until now, such as acclimatization to a less-than-optimum set of electroacoustic characteristics.

The technological innovations of the past decade have not always fulfilled their early promise. It is revealing to note that none of the highly sophisticated methods of signal processing that have been tried for amplitude compression, speech enhancement, or noise reduction have as yet yielded substantial improvements in speech intelligibility. Research on these advanced signal processing schemes, however, has provided significant new insights that have been used in the design of simpler signal processing systems. Several of these relatively simple systems have been quite effective in achieving their targeted goals (small improvements in intelligibility, large improvements in sound quality).

A common thread in almost all of the investigations of new processing schemes, simple or complex, is that substantial individual differences have been observed in almost all cases. A processing scheme that was found to work well for one hearing aid user was often unacceptable for another. The importance of individual differences should not be underestimated and underscores the need for developing appropriate methods for individualized prescriptive fitting of modern hearing aids. A possible reason for the large individual differences that have been observed in optimizing signal processing schemes is that the optimum setting often involves a compromise between two conflicting factors (e.g., between increasing the proportion of the short-term speech spectrum that is audible and the loss of spectral shape information or between the increase in overall sound audibility and the increased masking of important low-level components of speech). Small individual differences in the relative importance of either or both of these opposing factors can produce substantial individual differences in the optimum parameter values for the signal processing scheme.

A particularly important research need is that of a general theory for predicting speech intelligibility and other attributes of the amplified acoustic signal, such as overall sound quality. The Articulation Index, as modified for hearing impairment, represents a useful first step in this direction. The development of a generalized index for predicting the detectability of non-linear and other signal processing distortions represents another useful step in this direction. The need to understand how sound is processed by the impaired ear is perhaps the most important and rewarding challenge to be addressed. A fundamental understanding at this level is the key to solving the myriad of problems that have been raised.

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References

Abramovitz, R. (1980). Frequency shaping and multiband compression in hearing aids. *J. Commun. Disord., 13*, 483-488.

Bakke, M.H., Neuman, A.C., & Toraskar, J. (1987). The effect of Wiener filtering on consonant recognition in noise. *ASHA, 29*(10), 189 (abstract).

Bareham, J. (1990). Part 2: Hearing instrument measurements using dual channel signal analysis. *Hearing Instruments, 41*(1), 32-33.

Barfod, J. (1979). Speech perception processes and fitting of hearing aids. *Audiology, 18*, 430-441.

Bentler, R.A. (1991). Clinical implications and limitations of current noise reduction circuitry. In G.A. Studebaker, F.H. Bess, & L.B. Beck (Eds.), *The Vanderbilt hearing aid report II* (pp. 79-91). Parkton, MD: York Press.

Bentler, R.A., & Pavlovic, C.V. (1989). Transfer functions and correction factors used in hearing aid evaluation and research. *Ear and Hearing, 10*, 58-63.

Bentler, R.A., & Pavlovic, C.V. (1992). Addendum to "Transfer functions and correction factors used in hearing aid evaluation and research." *Ear and Hearing, 13*, 284-286.

Berger, K.W., Hagberg, E.N., & Rane, R.L. (1978). A method of hearing aid prescription. *Hearing Instruments, 29* (July), 12-13.

Boothroyd, A., Hnath-Chisolm, T., Hanin, L., & Kishon-Rabin, L. (1988). Voice fundamental frequency as an auditory supplement to the speechreading of sentences. *Ear and Hearing, 9*, 306-312.

Boothroyd, A. (1987). CASPER: Computer assisted speech perception evaluation and training. *Proceedings of the 10th Annual Conference on Rehabilitation Technology* (pp. 734-736). Washington, DC: Association for the Advancement of Rehabilitation Technology.

Boothroyd, A. (1991). *CASPER: A user-friendly system for computer- assisted speech perception testing and training.* (internal Report, available on request). Center for Research in Speech and Hearing Sciences, City University of New York, 33 West 42nd St., New York, NY 10036.

Braida, L.D., Durlach, N.I., Lippmann, R.P., Hicks, B.L., Rabinowitz, W.M., & Reed, C.M. (1979). *Hearing Aids-A review of past research of linear amplification, amplitude compression and frequency lowering. (ASHA Monograph No. 19).* Rockville, MD: American Speech-Language-Hearing Association.

Breeuwer M., & Plomp, R. (1986). Speechreading supplemented with auditorily-presented speech parameters. *J. Acoust. Soc. Am., 79*, 481-499.

Burnett, E.D. (1967). A new method for the measurement of nonlinear distortion using a random noise test signal. *Bulletin of Prosthetics Research, 10* (7), 76-92.

Bustamante, D.K., & Braida, L.D. (1987). Principal-component amplitude compression for the hearing impaired. *J. Acoust. Soc. Am., 82*, 1227-1242.

Byrne, D. (1987). Hearing aid selection formulae: Same or different? *Hearing Instruments, 38*(1), 5-11.

Byrne, D., & Tonisson, W. (1976). Selecting the gain of hearing aids for persons with sensorineural hearing impairment. *Scand. Audiol., 5*, 51-59.

Byrne, D., & Dillon, D. (1986). The National Acoustic Laboratories' (NAL) new procedure for selecting the gain and frequency response of a hearing aid. *Ear and Hearing, 7*, 25S- 26S.

Byrne, D. (1991). Evaluation measures of speech intelligibility and quality. Research and clinical application. In G.A. Studebaker, F.H. Bess, & L.B. Beck (Eds.), *The Vanderbilt hearing aid report II* (pp. 195-199). Parkton, MD: York Press.

Chabries, D.M., Christiansen, R.W., Brey, R.H., Robinette, M.S., & Harris, R.W. (1987). Application of adaptive digital signal processing to speech enhancement for the hearing impaired. *J. Rehab. Res. and Dev., 24*(4), 65-74.

Cox, R.M. (1979). Acoustic aspects of hearing aids - ear canal coupling systems. *Monographs in Contemporary Audiology, 1*(3).

Cox, R.M., & Alexander, G.C. (1992). Maturation of hearing aid benefit: Objective and subjective measurements. *Ear and Hearing, 13*, 131-141.

Cox, R.M., Alexander, G.C., & Gilmore, C. (1991). Objective and self-report measures of hearing aid benefit. In G.A. Studebaker, F.H. Bess, & L.B. Beck (Eds.) *The Vanderbilt hearing aid report II* (pp. 201-213). Parkton, MD: York Press.

Cox, R.M., & Gilmore, C. (1990). Development of the Profile of Hearing Aid Performance (PHAP). *J. Speech Hear. Res., 33*, 343-357.

Cudahy, E. (1992). Personal communication.

Cudahy, E., & Kates, J. (1993). Measuring the performance of modern hearing aids. In G.A. Studebaker & I. Hochberg (Eds.), *Acoustical factors affecting hearing aid performance* (2nd ed.) (pp. 51-64). Boston: Allyn and Bacon.

Cudahy, E., & Link, H.C. (1991). Towards implementation of a flexible hearing-aid measurement system. *J. Comm. Disord., 24*, 187-200.

Cummins, K.L., & Hecox, K.E. (1987). Ambulatory testing of digital hearing aid algorithms. *Proceedings of the 10th Annual Conference on Rehabilitation Technology* (pp. 398-400). Washington, DC: RESNA-Association for the Advancement of Rehabilitation Technology.

DeFilippo, C., & Scott, B. (1978). A method for training and evaluating the reception of ongoing speech. *J. Acoust. Soc. Am., 63*, 1186-1192.

DeGennaro, S.V., Braida, L.D., & Durlach, N.I. (1986). Multichannel syllabic compression for severely impaired listeners. *J. Rehabil. Res. and Dev., 23*(1), 17-24.

Demorest, M.E., & Erdman, S.A. (1987). Development of the Communication Profile for the Hearing Impaired. *J. Speech Hear. Disord., 52*, 129-143.

Dempsey, J., Levitt, H., & Josephson, J. (1992). Computer assisted tracking simulation (CATS). *J. Acoust. Soc. Am., 92*, 701-710.

Dreschler, W. (1988). Dynamic-range reduction by peak clipping or compression and its effects on phoneme perception in hearing- impaired listeners. *Scand. Audiol., 17*, 45-51.

Dugal, R.L., Braida, L.D., & Durlach, N.I. (1980). Implications of previous research for the selection of frequency-gain characteristics. In G.A. Studebaker & I. Hochberg (Eds.), *Acoustical factors affecting hearing aid performance and measurement* (pp. 379-403). Baltimore: University Park Press.

Dyrlund, O. (1989). Characterization of non-linear distortion in hearing aids using coherence analysis. A pilot study. *Scand. Audiol, 18*, 143-148.

Fabry, D.A. (1991). Programmable and automatic noise reduction in existing hearing aids. In G.A. Studebaker, F.H. Bess, & L.B. Beck (Eds.), *The Vanderbilt hearing aid report II* (pp. 65-78). Parkton, Md: York Press.

Fletcher, H. (1992). Perception of speech sounds by deafened persons, *J. Acoust. Soc. Am., 24*, 490-497.

French, N.R., & Steinberg, J.C. (1947). Factors governing the intelligibility of speech sounds. *J. Acoust. Soc. Am., 19*, 90-119.

Freyman, R., & Nerbonne, G. (1989). The importance of consonant-vowel intensity ratio in the intelligibility of voiceless consonants. *J. Speech Hear. Res., 35*, 524-535.

Foust, K.O., & Gengel, R.W. (1973). Speech discrimination by sensorineural hearing-impaired persons using a transposer hearing aid. *Scand. Audiol., 2*, 161-170.

Gabrielson, A., Schenkman, B., & Hagerman, B. (1988). The effects of different frequency responses on sound quality and speech intelligibility. *J. Speech Hear. Res., 31*, 166-177.

Gatehouse, S. (1989). Apparent auditory deprivation effects of late onset: The role of presentation level. *J. Acoust. Soc. Am., 86*, 2103-2106.

Gatehouse, S. (1992). The time course and magnitude of perceptual acclimatization to frequency responses: Evidence from monaural fitting of hearing aids. *J. Acoust. Soc. Am., 92*, 1258-1268.

Gatehouse, S., & Gordon, J. (1990). Response times to speech stimuli as measures of benefit from amplification. *Brit. J. Audiol., 24*, 63-68.

Gittles, T., & Wilson, F. (1987). Compression amplification with environment controlled release time. *Hearing Instruments, 38*(8), 39-41.

Goldberg, H. (1972). The utopian hearing aid: Current state of the art. *J. Audiol. Res., 12*, 331-335.

Gordon-Salant, S. (1986). Recognition of natural and time/intensity altered CVs by young and elderly subjects with normal hearing. *J. Acoust. Soc. Am., 80*, 1599-1607.

Graupe, D., Grosspietsch, J.K., & Basseas, S.P. (1987). A single-microphone-based self-adaptive filter of noise from speech and its performance evaluation. *J. Rehab. Res. and Devel., 24*(4), 119-126.

Guttman, N., & Nelson, J.R. (1968). An instrument that creates some artificial speech spectra for the severely hard of hearing. *Am. Ann. Deaf, 113*, 295-302.

Guttman, N., Levitt, H., & Bellefleur, P. A. (1970). Articulation training of the deaf using low-frequency surrogate fricatives. *J. Speech and Hear. Res., 13*, 19-29.

Haggard, M.P., Foster, J.R., & Iredale, F.E. (1981). Use and benefit of postaural aids in sensory hearing loss. *Scand. Audiol., 10*, 45-52.

Hicks, B.L., Braida, L.D., & Durlach, N.I. (1981). Pitch invariant frequency lowering with nonuniform spectral compression. *Proceedings of the IEEE International Conference on Acoustics, Speech, and Signal Processing* (pp. 121-124). Atlanta, GA.

Iwasaki, S. (1981). Automatic noise suppression in hearing aids. *Hearing Aid Journal, 40* (Dec.), 10-11, 40.

Johansson, B. (1966). The use of the transposer for the management of the deaf child. *Internat. Audiol., 5*, 362-373.

Kates, J.M. (1990) A test suite for hearing aid evaluation. *J. Rehab. Res. and Dev., 27*(3), 255-278.

Kates, J. (1991) New developments in hearing and measurements. In G.A. Studebaker, F.H. Bess, & L.B. Beck (Eds.), *The Vanderbilt hearing aid report II* (pp. 149-163). Parkton, MD: York Press.

Kennedy, E.S., & Levitt, H. (1990). Optimal C/V intensity ratio at MCL. *ASHA, 32*, 65.

Killion, M.C. (1990). A high fidelity hearing aid. *Hearing Instruments, 41*(8), 38-39.

Killion, M.C., Staab, W.J., & Preves, D.A. (1990). Classifying automatic signal processors. *Hearing Instruments, 41*(8), 24-26.

Kryter, K.D. (1962). Methods for the calculation and use of the Articulation Index. *J. Acoust. Soc. Am., 34*, 1689-1697.

Kuk, F.K., & Pape, N.M.C. (1992). The reliability of a modified simplex procedure in hearing-aid frequency-response selection. *J. Speech & Hearing Res., 35*, 3418-3429.

Laurence, R., Moore, B.C.J., & Glasberg, B. (1983). A comparison of behind-the-ear high fidelity linear hearing aids and two-channel compression aids in the laboratory and in everyday life. *British Journal of Audiology, 17*, 31-48.

Levitt, H. (1982a). An array-processor computer hearing aid. *ASHA, 24*(10), 805.

Levitt, H. (1982b). Speech discrimination ability in the hearing impaired: Spectrum considerations. In G. Studebaker & F. Bess (Eds.), *The Vanderbilt hearing aid report: State of the art- research needs* (pp. 32-43). Upper Darby, PA: Monographs in Contemporary Audiology.

Levitt, H., (1984). Procedural factors in speech recognition testing. *ASHA Reports, 14*, 57-64.

Levitt, H., & Collins, M.J. (1980). An experimental protocol for the prescriptive fitting of a wearable master hearing aid. In G.A. Studebaker & I. Hochberg (Eds.), *Acoustical factors affecting hearing aid performance* (pp. 323-339). Baltimore: University Park Press.

Levitt, H., Cudahy, E., Jiang, W.H., Kennedy, E., & Link, C. (1987). Towards a general measure of distortion. *J. Rehab. Res. and Dev., 24*(4), 283-292.

Levitt, H., & Neuman, A.C. (1990). A sentence test for hearing-aid evaluation. *Proceedings of the RESNA 13th Annual Conference* (pp. 271-272) Washington, DC: Association for the Advancement of Rehabilitative Technology.

Levitt, H., & Neuman, A. (1991). Evaluating orthogonal polynomial compression. *J. Acoust. Soc. Am., 90*, 241-252.

Levitt, H., Neuman, A., Mills, R., & Schwander, T. (1986). A digital master hearing aid. *J. Rehab. Res. and Dev., 23*(1), 79-87.

Levitt, H., Neuman, A., & Sullivan, J. (1990). Studies with digital hearing aids. *Acta Otolaryngology*, Suppl. *469*, 57-69.

Levitt, H., Pickett, J.M., & Houde, R.A. (Eds.). (1980). *Sensory aids for the hearing impaired.* New York: IEEE Press.

Levitt, H., Sullivan, J.A., Neuman, A.C., & Rubin-Spitz, J.A. (1987). Experiments with a programmable master hearing aid. *J. Rehab. Res. and Dev., 24*(4), 29-54.

Libby, E. (1986). The 1/3-2/3 insertion gain hearing aid selection guide. *Hearing Instruments, 37*, 27-29.

Lim, J.S., Oppenheim, A.V. (1979). Enhancement and bandwidth compression of noisy speech. *Proc. IEEE, 67*, 1586-1604.

Lim, J.S. (1983). *Speech enhancement.* Englewood, NJ: Prentice Hall.

Ling, D. (1969). Speech discrimination by profoundly deaf children using linear and coding amplifiers. *IEEE Trans. Audio. Electroacoust., AU-17*, 298-303.

Lybarger, S.F. (1978). Selective amplification - A review and evaluation. *J. Am. Aud. Soc., 3*, 258-266.

Mazor, M., Simon, H., Scheinberg, J., & Levitt, H. (1977). Moderate frequency compression for the moderately hearing impaired. *J. Acoust. Soc. Am., 62*, 1273-1278.

McCandless, G., & Lyregaard, P. (1983). Prescription of gain/output (POGO) for hearing aids. *Hearing Instruments, 34*, 16-21.

Milner, P., Braida, L.D., Durlach, N.I., & Levitt, H. (1984) Perception of filtered speech by hearing-impaired listeners. *ASHA Reports, 14*, 30-41.

Montgomery, A.A., & Edge, R.A. (1988). Evaluation of two speech enhancement techniques to improve intelligibility for hearing-impaired adults. *J. Speech & Hear. Res., 31*(3), 386-393.

Moore, B.C.J. (1987). Design and evaluation of a two-channel compression hearing aid. *J. Rehab. Res. and Dev., 24*(4), 181-192.

Moore, B.C.J., & Glasberg, B.R. (1986). A comparison of two-channel and single-channel compression aids. *Audiology, 25*, 210-226.

Neuman, A.C., Levitt, H., Mills, R., & Schwander, T. (1987). An evaluation of three adaptive hearing aid selection strategies. *J. Acoust. Soc. Am., 82*, 1967-1976.

Neuman, A.C., & Schwander, T.J. (1987). The effect of filtering on the intelligibility and quality of speech in noise. *J. Rehab. Res. and Dev., 24*(4), 127-134.

Ono, H., Kanzaki, J., & Mizoi, K. (1983). Clinical results of hearing aid with noise-level-controlled selective amplification. *Audiology, 22*, 494-515.

Pascoe, D.P. (1978). An approach to hearing-aid selection. *Hearing Instruments, 29*, 12-16, 36-37.

Pavlovic, C.V., Studebaker, G.A., & Sherbecoe, R.L. (1986). An articulation index based procedure for predicting the speech recognition performance of hearing-impaired individuals. *J. Acoust. Soc. Am., 80*, 50-57.

Peterson, P.M., Durlach, N.I., Rabinowitz, W.M., & Zurek, P.M. (1987). Multimicrophone adaptive beamforming for interference reduction in hearing aids. *J. Rehab. Res. and Dev., 24*(4), 103-110.

Pollack, I., & Pickett, J.M. (1958). Masking of speech by noise at high sound levels. *J. Acoust. Soc. Am., 30*, 127-130.

Posen, M.P., Reed, C.M., & Braida, L.D. (in press). The intelligibility of frequency-lowered speech produced by a channel vocoder. *J. Rehab. Res. and Dev.*

Pratt, R.L. (1981). On the use of reaction time as a measure of intelligibility. *Brit. J. Audiol., 15*, 253-255.

Preves, D. (1990). Expressing noise and distortion with the coherence measurement. *ASHA, 32*, 56-59.

Preves, D.A. (1991). Output limiting and speech enhancement. In G.A. Studebaker, F.H. Bess, & L.B. Beck (Eds.), *The Vanderbilt hearing aid report II.* Parkton, MD: York Press.

Punch, J., Montgomery, A., Schwartz, D., Walden, B., Prosek, R., & Howard, M. (1980). Multidimensional scaling of quality judgments of speech signals processed by hearing aids. *J. Acoust. Soc. Am., 68*, 458-466.

Punch, J.L., & Robb, R. (1992). Prescriptive hearing aid fitting by parameter adjustment and selection. *J. Am. Acad. Audiol., 3*, 94-100.

Rankovic, C.M. (1991). An application of the articulation index to hearing aid fitting. *J. Speech and Hearing Res., 34*, 391-402.

Rankovic, C.M., Freyman, R.L., & Zurek, P.M. (1992). Potential benefits of adaptive frequency-gain characteristics for speech reception in noise. *J. Acoust. Soc. Am., 91*(1), 354-362.

Raphael, L.J. (1972). Preceding vowel duration as a cue to the perception of the voicing characteristic of word final consonants in American English. *J. Acoust. Soc. Am., 51*, 1296-1303.

Reed, C.M., Hicks, B.L., Braida, L.D., & Durlach, N.I. (1983). Discrimination of speech processed by low-pass filtering and pitch-invariant frequency lowering. *J. Acoust. Soc. Am., 74*, 409-419.

Reed, C.M., Schultz, K.I., Braida, L.D., & Durlach, N.I. (1985). Discrimination and identification of frequency-lowered speech in listeners with high-frequency hearing impairment. *J. Acoust. Soc. Am., 78*, 2139-2141.

Revoile, S.G., Holden-Pitt, L., Edward, D., & Pickett J. (1986). Some rehabilitative considerations for future speech- processing hearing aids. *J. Rehab. Res. and Dev., 23*(1), 89-94.

Revoile, S.G., Holden-Pitt, L., Pickett, J., & Brandt, F. (1986). Speech cue enhancement for the hearing impaired: I. Altered vowel durations for perception of final fricative voicing. *J. Speech Hear. Res., 29*, 240-255.

Rosen, S ., Walliker, J.R., Fourcin, A., & Ball, V. (1987). A microprocessor-based acoustic hearing aid for the profoundly impaired listener. *J. Rehab. Res. and Dev., 24*(4), 239-260.

Scherr, C.K., Schwartz, D.M., & Montgomery, A.A. (1983). Follow-up survey of new hearing aid users. *J. Acad. Rehab. Audiol., 16* 202-209.

Schoepflin, J.R., & Levitt, H. (1991). Continuous discourse tracking: An analysis of the procedure. *J. Common. Disord., 24*, 237-249.

Schwander, T.J., & Levitt, H. (1987). Effect of two-microphone noise reduction on speech recognition by normal-hearing listeners. *J. Rehab. Res. and Dev., 24*(4), 87-92.

Seewald, R.C., Zelisko, D.L., Ramji, K.V., & Jamieson, D.B. (1991). *A computer-assisted implementation of the desired sensation level method for electroacoustic selection and fitting in children.* Child Amplification Laboratory, Hearing Health Care Research Unit, The University of Western Ontario, London, Ontario.

Silman, S., Gelfand, S.A., & Silverman, C.A. (1984). Late onset auditory deprivation: Effects of monaural versus binaural aids. *J. Acoust. Soc. Am., 76*, 1357-1362.

Skinner, M.W. (1980). Speech intelligibility in noise-induced hearing loss: Effects of high-frequency compensation. *J. Acoust. Soc. Am., 67*, 306-17.

Skinner, M.W., Pascoe, D.P., Miller, J.D., & Popelka, G.R. (1982). Measurements to determine the optimal placement of speech energy within the listener's auditory area: A basis for selecting amplification characteristics. In G.A. Studebaker & F.H. Bess (Eds.), *The Vanderbilt hearing aid report* (pp. 161-169). Upper Darby, PA: Monographs in Contemporary Audiology.

Soede, W. (1990). *Improvement of speech intelligibility in noise: Development and evaluation of a new directional hearing instrument based on array technology.* Unpublished Ph.D. thesis, Delft University of Technology, The Netherlands.

Studebaker, G.A. (1991). Measures of intelligibility and quality. In G.A. Studebaker, F.H. Bess, & L.B. Beck (Eds.), *The Vanderbilt hearing aid report II* (pp. 185-194). Parkton, MD: York Press.

Studebaker, G.A., & Sherbecoe, R.L. (1991). Frequency-importance and transfer functions for recorded CID W-22 word lists. *J. Speech Hear. Res., 34*, 427-438.

Sullivan, J.A., Levitt, H., Hwang, J., & Hennessey, A. (1988). An experimental comparison of four hearing aid prescription methods. *Ear and Hearing, 9*, 22-32.

Tannahill, J.C. (1979). The Hearing Handicap Scale as a measure of hearing aid benefit. *J. Speech Hear. Res., 44*, 91-99.

Uchanski, R.M., Delhorne, L.A., Dix, A.K., Braida, L.D., Reed, C.M., & Durlach, N.I. (in press). Automatic speech recognition to aid the hearing impaired: Prospects for the automatic generation of cued speech. *J. Rehav. Res. and Dev.*

van Dijkhuizen, J.N., Festen, J.M., & Plomp, R. (1989). The effect of varying the amplitude-frequency response on the masked speech-reception threshold of sentences for hearing impaired listeners. *J. Acoust. Soc. Am., 86*(2), 621-628.

Van Tasell, D.J., Larsen, S.Y., & Fabry, D.A. (1988). Effects of an adaptive filter hearing aid on speech recognition in noise by hearing-impaired subjects. *Ear and Hearing, 9*, 15-21.

Van Tasell, D.J., Thomas, R., & Crain, M.A. (1992). Noise reduction hearing aids: Release from masking and release from distortion. *Ear and Hearing, 13*, 114-121.

Velmans, M., Marcuson, M., Grant, J., Kwiatkowski, R., & Rees, R. (1988). *The use of frequency transposition in the language acquisition of sensory-neural deaf children.* Report of the Department of Psychology, Goldsmiths' College, University of London (1-115).

Villchur, E. (1973). Signal processing to improve speech intelligibility in perceptive deafness. *J. Acoust. Soc. Am., 53*, 1646-1657.

Villchur, E., & Killion, M.C. (1976). Auditory aid to deaf speakers monitoring fundamental voice frequencies. *J. Acoust. Soc. Am., 59*, (Suppl. 1), S69 (abstract).

Villchur, E. (1982). The evaluation of amplitude-compression processing for hearing aids. In F.H. Bess & G.A. Studebaker (Eds.), *The Vanderbilt hearing aid report* (141-143). Upper Darby, PA: Monographs in Contemporary Audiology.

Walden, B.E., Demorest, M.E., & Hepler, E.L. (1984). Self report approach to assessing benefit derived from amplification. *J. Speech Hear. Res., 27*, 49-56.

Walker, G., Byrne, D., & Dillon, H. (1984). The effects of multichannel compression/expansion amplification on the intelligibility of nonsense syllables in noise. *J. Acoust. Soc. Am., 73*(3), 746-757.

Weiss, M.R., & Aschkenasy, E. (1981). *Wideband speech enhancement, final technical report, (RADC-TR-81-53)*, Rome Air Development Center, Air Force Systems Command, NY: Griffiss Air Force Base.

Weiss, M. (1987). Use of an adaptive noise canceller as an input preprocessor for a hearing aid. *J. Rehab. Res. and Dev., 24*(4), 93-102.

Widrow, B., Glover, J.R., McCool, J.M., Kaunitz, J., Williams, C.S., Hearn, R.H., Zeidler, J.R., Dong, E., & Goodlin, R.C. (1975). Adaptive noise cancelling: Principles and application. *Proc. of the IEEE, 63*, 1692-1716.

Williamson, M.J., Cummins, K.L., & Hecox, K.E., (1987). Speech distortion measures for hearing aids. *J. Rehab. Res. and Dev., 24*(4), 277-282.

Yanick, P. (1976). Effects of signal processing on the intelligibility of speech in noise for subjects possessing sensorineural hearing loss. *J. Am. Auditory Soc., 1*, 229-238.