Scientific/Technical Articles

Section I. Digital Techniques in Acoustic Amplification

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Journal of Rehabilitation Research and Development
NOTICE TO CONTRIBUTORS

Purpose and Scope

The *Journal of Rehabilitation Research and Development*, published quarterly, is a scientific rehabilitation engineering, research and development publication in the multidisciplinary field cf disability rehabilitation. General priority areas are: Prosthetics and Orthotics; Spinal Cord Injury and Related Neurological Disorders; Communication, Sensory and Cognitive Aids; and, Gerontology. The *Journal* receives submissions from sources within the United States and throughout the world.

Only original scientific rehabilitation engineering papers will be accepted. Technical Notes describing preliminary techniques, procedures, or findings of original scientific research may also be submitted. Letters to the Editor are encouraged. Books for review may be sent by authors or publishers. The Editor will select reviewers.

Review Process

All scientific papers submitted to the *Journal* are subject to critical peer review by at least two referees, either editorial board members or ad hoc consultants, who have special expertise in a particular subject. To ensure objectivity, anonymity will be maintained between the author(s) and the referees. The final decision as to the paper's suitability for publication rests with the Editor of the *Journal*.

Originality

Authors must confirm that the contribution has not already been published by or submitted to another journal. The submission letter must be signed by all authors.

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Abstracts: An Abstract of 150 words or less must be provided with the submitted manuscript. It should give the factual essence of the article and be suitable for separate publication in index journals.

The style must be in the form of a structured abstract which outlines the Purpose of the Study, Subjects, Methods, Results, Conclusion, and Discussion.

Key Words: Three to ten key words, preferably terms from the Medical Subject headings from Index Medicus should be provided.

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A MESSAGE FROM THE SECRETARY

Jesse Brown, Secretary of DVA

Earlier this year, I was faced with a budget dilemma and was forced to take funds from some of our programs and shift them to other areas that were in "critical condition." Research was one of the areas that I reduced, and that was a very difficult decision to make. At the time, some people criticized my decision and deemed it to be the beginning of the end of research in VA. Nothing could be further from the truth.

My decision was not the setting of a precedent, and it in no way indicates a disregard for the research programs. Even as I made the decision, I was thinking of ways to restore those funds in future years. I firmly believe that VA research is just as important to our veterans as the medical care we provide them in our hospitals. In fact, clinical care as we know it would not exist without research. Research is where it all begins, and I am determined to keep our VA research efforts thriving.

The story of VA research successes is long and rewarding. The developments and break-throughs that have come out of that research over the years have helped to benefit *all* Americans, not just veterans. I have no intention of writing the final chapter in the book of VA research. That is why this *Journal* has been reestablished. This move signals our renewed commitment to our research endeavors and to those who devote their careers to conducting research.

After being wounded in Vietnam, I began the long and arduous process of hospitalization, therapy, and rehabilitation. I know that the success of my recovery was due to the efforts of the many compassionate men and women who tended to my wounds, guided me through physical therapy, and helped me in every way possible with my total rehabilitation. But I also know that they would not have the best treatment protocols, prosthetics, and therapy regimens if it were not for the efforts of the men and women in VA research.

As Secretary of Veterans Affairs Jesse Brown, I have an obligation of office to support our VA research programs. As Jesse Brown, disabled veteran, I have a personal reason to not only support those programs but to be thankful for what they meant to my treatment and recovery. That is why I am delighted to see the rebirth of the *Journal of Rehabilitation Research and Development*. It is my hope that this tool for your profession will help you carry VA research to new and even more rewarding levels.

Jesse Brown
Secretary of Veterans Affairs
Washington, DC 20001

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EDITORIAL

Much is said about quality in health care services by many organizations today. VA is proud of the hearing health services provided to our veterans, reflected in audiology and speech pathology clinics located in most of our medical centers. An important role for an organization like the Department of Veterans Affairs is to provide for links between applied research and actual benefit for these individuals. The veteran's needs for restored function and quality of life provide the focus of our mission. The Rehabilitation Research and Development Service is organized to provide such links, represented by merit-reviewed support for major studies in hearing aid and assistive device technology over the past decade.

In this special issue of the *Journal of Rehabilitation Research and Development*, we highlight the advances made since our last special publication on sensory aids for hearing impairment in 1987. With world-wide distribution of this journal, we expect broad interest in this topic on hearing loss, a problem which ranks as the third most prevalent chronic condition among people 65 years of age and older in the United States.

The Guest Editors for this special issue are Harry Levitt, Ph.D., and Allen E. Boysen, Ph.D., both renowned leaders in the field of audiology.

Harry Levitt, Ph.D.
Center for Research in Speech and Hearing Sciences,
Graduate School and University Center,
City University of New York

Allen E. Boysen, Ph.D.
Director, Audiology and Speech Pathology
Service
Department of Veterans Affairs

Harry Levitt obtained his Ph.D. in electrical engineering from the Imperial College of Science and Technology, London, in 1964. He then joined the Bell Laboratories where he did research on binaural hearing, adaptive testing in psychoacoustics, and digital processing of speech signals. During this time he became interested in the design, development, and evaluation of special telephones for hearing-impaired persons and other communication aids. In 1969, he joined the City University of New York, where he set up the Communication Sciences Laboratory. His research efforts since then have focused on applications of computer technology in speech and hearing sciences and, in particular, on the development of more effective sensory aids for hearing impairment using computer techniques. His many research contributions include the development of computer-assisted adaptive test procedures for use in speech and hearing, computerized visual tactile speech-training aids, computer speech synthesis for diagnosis of speech problems, a

pocket telecommunicator, a digital master hearing aid, digital signal-processing techniques for matching the speech signal to residual auditory function, noise reduction techniques for hearing aids, and computer synthesis of video speech signals for studies in lipreading. Dr. Levitt has received several regional and national awards for his work and was recently named Distinguished Professor at the City University of New York.

Allen E. Boysen, Ph.D., is the Director of Audiology and Speech Pathology Service, Department of Veterans Affairs. He holds a B.A. degree from the University of Iowa, an M.S. from Colorado State University, and a Ph.D. from the University of Oklahoma Health Sciences Center. His degrees are all in communication disorders. As a speech-language pathologist, Dr. Boysen devoted 15 years to clinical practice. He has also contributed to the design and development of multidisciplinary quality improvement systems for the Department of Veterans Affairs, and directed the agency's Rehabilitation Education Program, focused on VA-wide continuing education of professional staff. As Director of the largest program for clinical services in the nation, training and research in communication disorders, he has been involved in projects such as adaptive digital hearing aid development, cochlear implant center development, hearing aid consensus conferences, and an interagency agreement for hearing aid research with the National Institute on Deafness and Other Communication Disorders.

John W. Goldschmidt, M.D.
Director, Rehabilitation Research and Development Service
Veterans Health Services and Research Administration
Department of Veterans Affairs

TO THE READERSHIP

The *Journal of Rehabilitation Research and Development (JRRD)* is designed to serve the requirements of the scientific community for responsible reporting of original scientific data considered to have wide technological, clinical, and ultimately commercial applications.

Beginning with this issue of *JRRD,* Volume 30, No. 1, 1993, modifications have been made and descriptive material has been added. The Editorial Board of *JRRD* proposed these modifications and additions in order to further enhance the high quality of the *Journal* and to ensure that it continues to be the conservator of documentation of relevant research findings, speaking directly to the needs and interests of the veteran population and the private sector.

The following modifications and additions have been made:

• **Introduction of Structured Abstracts** to the authors' scientific pap rs which will present a concise summary of the essential facts as a means of explaining, in nontechnical terms, the purpose of the work, subjects (if any), procedures followed, the subsequent results, and the appropriate application for consideration by the veteran. A final explanatory sentence will state the clinical relevance of the information contained in the article. This information will be written by the author of the paper, in addition to the conventional abstract as written by the author and accepted by the peer review process as an integral part of the scientific paper.
• **Inclusion of Guest Editorials,** outlining the need for, and implications of, major research and development programs as they relate to the veteran.
• **Expansion of the Letters to the Editor Section.** Interested readers are encouraged to engage in an exchange of information through this Section. Letters should relate specifically to material published in *JRRD.* Authors of the article(s) in question will be asked to respond.

• **Expansion of the Clinical Reports Section.** This Section will report the *results of* rehabilitation technology clinical evaluations, final reports of R&D funded projects, surveys, and other subjects of interest to the clinician and veteran consumer in the Department of Veterans Affairs (VA) and elsewhere. By expanding this Section, we will bring clinical information to our readers from many sources, thus further assisting veterans in activities of independent and productive living.

It is our hope that the modifications and additions we have made will continue to enhance the quality of *JRRD,* reach out to the clinical community, and ultimately answer, through scientific research and development, the needs of the veterans we serve.

Tamara T. Sowell
Editor, *Journal of Rehabilitation Research and Development*
Rehabilitation Research and Development Service
Department of Veterans Affairs

GUEST EDITORIAL

Just over 5 years ago, in the Fall of 1987, the *Journal* published a special issue on Sensory Aids for Hearing Impairment. This was a time of great excitement in the field in that digital hearing aids had just been introduced and offered great promise for the future. Among the many advantages offered by this new techi.ology were more efficient methods of hearing aid prescription, powerful new methods of processing signals for improving speech intelligibility and reducing the effects of background noise, new measurement techniques, and improved diagnostic procedures. It is now five years later and the question to be asked is: Have digital hearing aids lived up to their promise? This special issue of the *Journal,* namely, *Part I: Advanced Hearing Aid Technology,* to be followed by a companion issue, *Part II: Clinical Evaluation of New Generation Hearing Aids,* in several months, addresses the question by providing concrete examples of recent positive results using this new technology. As shown in these papers, significant advances have been made in several areas. Although much of the early promise has yet to be fulfilled, the impact of digital technology on the field has already been profound. Hybrid analog/digital hearing aids providing superior performance (in comparison with older, conventional hearing aids) are now widely used, and significant improvements in cochlear implants have been obtained using digital signal processing techniques. Further advances are expected with the ongoing development of even more sophisticated digital chips. Veterans with hearing impairments have benefited substantially from recent advances in this area. It is our hope that the progress currently being made in applying advanced technology to clinical problems will continue with increasing momentum.

Harry Levitt, Ph.D.
Center for Research in Speech and Hearing Sciences
City University of New York

PREFACE

The Fall 1987 issue of the *Journal of Rehabilitation Research and Development* presented a comprehensive review of sensory aids for hearing impairment. The demand for this special issue exceeded expectations and a second, larger printing was required. This issue was also widely disseminated throughout the v orld. Five years later, we are revisiting the topic in order to update knowledge in this rapidly changing area of rehabilitation.

It is noteworthy that, compared to the 1987 publication, the number of contributions from Department of Veterans Affairs (VA)-funded investigators in this current special issue has increased substantially. This reflects a significant and sustained increase in the emphasis on clinical and technology-oriented research by the VA's Rehabilitation Research and Development Service. The *Journal of Rehabilitation Research and Development* has had a major impact on strengthening this emphasis.

VA can take pride in its contributions to research, development, and clinical applications involving sensory aids for hearing impairment. For example, as early as the 1970s, VA supported the development of a digital hearing aid. Today, programmable hearing aids with digital components have become a viable clinical alternative to conventional hearing aids. VA is still actively engaged in supporting research and development in the clinical application of these instruments.

The cochlear implant is another major development in the field that is also strongly supported by VA. To date, the largest clinical study on cochlear implants has just been completed as part of VA's Cooperative Studies Program. Furthermore, regional clinical cochlear implant centers have been established throughout the VA health care system. In the long run, preventive care is more effective than remediation, and VA has been supportive of this type of research. Several VA-supported projects in this area are also reported in this special issue.

In terms of delivery of services, VA is the world's largest dispenser of hearing aids, with over 60,000 of these instruments annually being fitted in over 140 clinics This system includes an annual, comprehensive, peer-based evaluation program for selecting and testing hearing aids. A component of this program is the electroacoustic testing of these hearing aids by the National Institute of Standards and Technology.

VA has joined forces with other organizations in sponsoring major conferences on amplification for hearing impairment. These activities have resulted in the dissemination of research findings, policy guidelines, and standards. Taken together, these contributions have influenced industry and the professions in ways that have benefited veterans as well as the general public.

During these times of austerity in public funding, VA has joined forces in sharing agreements with others to maximize treatment benefit for patients with outcome-oriented research and development. Numerous such arrangements for clinica services exist between VA and Department of Defense facilities throughout the United States. A major joint initiative for hearing aid research exists with the National Institute on Deafness and Other Communication Disorders.

This special issue reflects current investigations with signal processing for hearing aids, technological aids, cochlear implants, and problems with hearing measurements. VA has enjoyed a significant role in nurturing many of these investigations through a rigorous process of merit review, funding support, and transfer to clinical applications. We are indebted to all those within VA and private sector professions who have contributed to the success of this publication.

Allen E. Boysen, Ph.D.
Director, Audiology and Speech Pathology Service
Department of Veterans Affairs

GUIDE TO THE ISSUE by Harry Levitt, Ph.D.

The papers in this special issue have been divided into three groups. The first deals with various aspects of amplification using digital techniques that have shown promising results in either the clinic or the laboratory. The second group of papers deals with the specific problem of speech in noise and new methods of signal processing for noise reduction. The third group of papers provides examples of useful applications of digital technology in related areas, most notably in signal processing for cochlear implants, but also in developing improved methods of measurement.

SECTION I. DIGITAL TECHNIQUES IN ACOUSTIC AMPLIFICATION

The advantages of digital signal processing were clearly demonstrated in the experimental digital hearing aids that were first developed. A serious limitation of these early instruments, however, was that their size and power consumption were too large for a practical, wearable hearing aid. A useful compromise that reduces power consumption to a practical level for a wearable instrument while at the same time maintaining many of the advantages of digital techniques is that of combining analog and digital technology in order to minimize power consumption.

The most common form of hybrid analog/digital hearing aid is that in which the audio signal is processed by analog components (amplifiers, filters) under digital control. Virtually all of the programmable hearing aids that are currently available for clinical use are of this type. Another form of hybrid amplification is that in which the signal is represented by pulses of varying width (i.e., pulse- width modulation). Amplifiers of this type, known as Class D amplifiers, are very efficient with respect to power consumption and are being used increasingly for the power amplification stage in small in-the-ear (ITE) hearing aids where small size and low power consumption are of critical importance.

Given the limitations of hybrid analog/digital circuits of small size and very low power consumption, can a hybrid Class D amplifier provide the precision and flexibility needed for effective prescriptive fitting of ITE hearing aids? The paper by Sammeth, et al., addresses this issue by investigating the extent to which modern ITE hearing aids using hybrid Class D amplifiers are capable of achieving prescribed frequency-gain characteristics. The results show that modern ITE hearing aids using this hybrid form of amplification can achieve prescribed frequency-gain characteristics with significantly greater accuracy than conventional ITE hearing aids.

One of the many advantages offered by digital hearing aids is that of new methods of signal processing for improving acoustic amplification. One such possibility is the use of adaptive algorithms for reducing or canceling unwanted signals, such as noise or unstable feedback oscillations. The pair of papers by Engebretson and French-St. George describe the development and evaluation, respectively, of adaptive feedback equalization using an experimental wearable digital hearing aid. The results show that an additional 4 dB of gain could be achieved when adaptive feedback equalization was present, resulting in improved intelligibility.

Advanced signal processing techniques offer new opportunities for improving acoustic amplification, but it is important to know how to use this

GUIDE TO THE ISSUE *(continued)*

new technology in order to achieve practical benefits. In many respects, the limitation on further progress is not so much that of not being able to process acoustic signals as needed, but rather that of not knowing what form this processing should take. The paper by Posen, et al., reports on new ways of processing speech for severely hearing-impaired individuals using frequency lowering. This form of signal processing has yielded mixed results in the past, improved perception being obtained for some speech sounds, and degraded perception for other sounds. In the study by Posen, et al., frequency lowering was used selectively depending on the acoustic-phonetic structure of the speech signal. For example, frequency lowering was not used whenever the input signal was dominated by low-frequency components, as typically occurs during semi-vowel and nasal consonants. The results of this investigation show improved performance for stops, fricatives, and affricates without degrading the perception of nasals and semi-vowels.

The last paper in this group, by Kates, addresses the issue of signal processing from a broad perspective, that of developing a general approach to the optimization of hearing aid processing by computer simulation of the impaired auditory system. From this vantage point, practical methods of signal processing for hearing aids can be optimized, as described in the paper. This approach represents a new way of thinking about the problem and provides a useful framework for developing improved methods of signal processing.

SECTION II.
NEW METHODS OF
NOISE REDUCTION

One of the most pressing problems to be addressed in the development of advanced signal processing hearing aids is that of reducing background noise. Methods of signal processing for noise reduction can be subdivided into two categories, single-microphone and multi-microphone techniques. Single-microphone techniques have received the greatest attention, presumably because conventional hearing aids have typically used only one microphone. The problem of single-microphone noise reduction is particularly difficult, however, and there is a long history of unsuccessful attempts at improving speech intelligibility in noise using a single microphone. The paper by Baer, et al., breaks this mold in that improvements in intelligibility using a single-microphone technique have been obtained. The basic approach is to enhance major peaks in the spectrum by increasing the differences in level between peaks and adjacent valleys. This method of spectral enhancement was combined with amplitude compression to achieve improvements in intelligibility equivalent to a gain of roughly 4 dB in speech-to-noise ratio.

Improvements in speech-to-noise ratio can be achieved much more easily using multi-microphone techniques. This is demonstrated in the paper by Bilsen, et al., who obtained improvements in speech-to-noise ratio of 7 dB using fixed- microphone arrays on the frame and legs of a pair of eyeglasses. Relatively simple methods of signal processing were used that could be incorporated in a practical hearing aid using existing technology.

The third paper in this group, by Kollmeier, et al., describes a binaural signal processing system combined with multiband amplitude compression. The binaural signal processor amplifies sounds emanating from the front and

suppresses sounds coming from the sides, including reverberation. The multiband compression algorithm matches the dynamic range of the amplified signal to that of the impaired auditory system. The two methods of signal processing were found to provide significant improvements in both speech intelligibility and sound quality.

SECTION III.
DIGITAL TECHNIQUES
APPLIED TO
RELATED AREAS

Digital signal processing techniques have proven to be of value in a number of related applications. The paper by Dillier, et al., describes three separate applications of digital signal processing techniques that have been found to be useful for both hearing aids and cochlear implants. In the first application, a two-microphone noise reduction system has been implemented using an adaptive beam-forming algorithm in which a higher degree of directionality was achieved (i.e., the system behaves in much the same way as a highly directional microphone with concomitant improvements in the speech-to-noise ratio, provided the speech and noise come from different directions). In the second application, loudness distortions resulting from the hearing impairment are corrected using multiband amplitude compression. Although this form of compression is not new, its implementation in real-time using digital signal processing techniques based on relevant psychoacoustic data represents a useful step toward developing an improved multiband digital hearing aid. The third application describes an experimental evaluation of various signal stimulation strategies for use with multichannel cochlear implants. The results show that coding strategies involving the interleaving of pulsatile signals provide significant improvements and that coding strategies of this form for pitch information can be of use to a cochlear implant patient.

The next paper, by Wilson, et al., compared two approaches for representing speech information in a cochlear implant, a compressed analog representation and a digital representation using continuous interleaved sampling (CIS). The CIS procedure was found to yield substantial improvements over the compressed analog representation. The paper also reports on investigations for optimizing the parameters of the CIS procedure (e.g., pulse duration, pulse rate, interval between sequential pulses).

Methods of noise reduction developed for acoustic amplification systems can also be used with cochlear implants and other sensory aids. The paper by Weiss describes how the INTEL method of noise reduction has been implemented for use with the Nucleus-22 cochlear implant. The results of a computer simulation showed an effective improvement of about 7 dB in the speech-to-noise ratio. Subsequent experimental investigations with cochlear implant patients have obtained improvements in intelligibility corresponding to a 5 dB gain in intelligibility. These improvements are striking in that they were obtained with a single-microphone method of noise reduction. It is also interesting to note that previous evaluations of the INTEL method of noise reduction for acoustic amplification did not show any significant change in intelligibility, although improvements in overall sound quality were obtained.

The last paper in this special issue, by Larson, et al., deals with the application of digital technology to a difficult measurement problem, that of

GUIDE TO THE ISSUE *(continued)*

wideband measurement of the acoustic impedance of the ear. Measurements of this type are useful not only for diagnostic testing but also in developing a quantitative description of the acoustic coupling between a hearing aid and the impaired auditory system. This information, in turn, can be used in developing more effective prescriptive procedures for hearing aids. The measurement of acoustic impedance over a wide frequency range is not a trivial problem, and this paper reports on an automated technique for obtaining acoustic impedance measurements and the reliability of such measurements obtained in two different laboratories by two different investigators.

This special issue has focused on issues of signal processing and its application to hearing aids and cochlear implants. A companion special issue, to appear shortly, will address more general aspects and clinical implications of this new technology.

Clinical Relevance for the Veteran

SUMMARY OF SCIENTIFIC/TECHNICAL PAPERS

IN THIS ISSUE

by Harry Levitt, PhD, Guest Editor

Achieving Prescribed Gain/Frequency Responses with Advances in Hearing Aid Technology Sammeth, et al.
(p. 1)

Purpose of the Work. Technological limitations have, until recently, restricted the capability of in-the-ear hearing aids to match prescribed frequency-gain characteristics. New technology has recently become available which offers substantial improvements in this capability. The purpose of this study was to evaluate the extent to which this new technology is capable of matching prescribed frequency-gain characteristics for an in-the-ear hearing aid. **Subjects.** Sixty hearing aid users participated. Half of the subjects had been fitted with older generation in-the-ear hearing aids, the other half were fitted with modern in-the-ear instruments using the new technology. **Procedures.** Standard techniques were used in fitting all of the in-the-ear hearing aids. The quality of each fit was then evaluated by measuring the sound pressure levels generated in the ear canal by the hearing aids. **Results.** The older technology hearing aids were found to provide too much gain through the mid-frequencies, and too little gain in the high frequencies. In contrast, the newer technology hearing aids provided a much closer approximation to the prescribed gain across the frequency range. **Relevance to Veteran Population.** Large numbers of veterans are fitted with in-the-ear hearing aids each year. The prescriptive fitting of in-the-ear hearing aids can be improved substantially using the newer technology.

Properties of an Adaptive Feedback Equalization Algorithm Engebretson, et al. *(p. 8)*

Purpose of the Work. A common problem with hearing aids is that not all of the amplified audio signal reaches the eardrum. Some of it escapes and reaches the hearing-aid microphone and is amplified once again. This process, known as acoustic feedback, can cause whistling and other unstable behavior in the hearing aid. Digital signal processing techniques provide a means for reducing the effects of acoustic feedback in hearing aids. The purpose of this study was to develop and perform laboratory evaluations of one such method of feedback reduction. An evaluation of the technique with hearing-impaired subjects is reported in the companion paper in this issue, by French-St George, et al. (see p. 17). **Procedures.** The characteristics of the acoustic feedback signal are estimated using a known signal at the input to the hearing aid. An electrical signal with identical characteristics is then generated and subtracted from the signals being amplified, thereby canceling the feedback signal. Since the sound transmission characteristics of the feedback signal changes over time, the cancellation process also adapts over time providing continuous, adaptive feedback reduction. **Results.** Laboratory evaluations of the feedback reduction technique showed that the magnitude of the feedback signal could be reduced substantially. As a consequence, the output power of a hearing aid can be increased by a factor of 10 or more without causing the hearing aid to whistle. **Relevance to Veteran Population.** Many veterans using powerful hearing aids, or with earmolds that allow for significant acoustic feedback, are not receiving the amplification they require because of whistling and related problems. Hearing aids incorporating advanced signal-processing techniques for feedback reduction will alleviate this problem considerably.

Behavioral Assessment of Adaptive Feedback Equalization in a Digital Hearing Aid French-St. George, et al. *(p. 17)*

Purpose of the Work. In the companion paper by Engebretson, et al. in this issue (see p. 8), a method for acoustic feedback reduction was developed using digital signal processing techniques. The purpose of this study was to evaluate the technique with hearing-impaired subjects. **Subjects.** Nine hearing-impaired subjects (5 male and 4 female) having an average age of 63.4 years (range 39 to 76 years) participated in the study. **Procedures.** A wearable master digital hearing aid was used. The instrument was programmed to simulate the subject's own hearing aid both with and without acoustic feedback reduction. The subjects used the hearing aid for a range of conditions including speech at a low signal level both in quiet and in noise. **Results.** Hearing aid users are often limited by acoustic feedback when listening to speech in quiet at a low signal level. The feedback reduction technique allowed for significantly more gain to be used under these conditions without unstable acoustic feedback. The use of increased gain resulted in improved speech intelligibility. **Relevance to Veteran Population.** Many veterans who use hearing aids are not getting the gain they require because higher gain settings lead to unstable acoustic feedback (whistling). The signal pro-

cessing technique evaluated in this study would reduce this problem significantly.

Intelligibility of Frequency-Lowered Speech Produced by a Channel Vocoder Posen, et al. *(p. 26)*

Purpose of the Work. Frequency lowering is a form of signal processing designed to match the speech signal to the residual hearing capacity of a listener with a high-frequency hearing loss. The purpose of this study was to investigate new forms of frequency lowering for persons with no high-frequency hearing. **Subjects.** Two normal-hearing subjects participated. The test stimuli were processed to simulate a hearing loss in which no high-frequency speech cues are available. **Procedures.** The speech signal was subdivided into eight contiguous frequency bands covering the range from 1,000 Hz to 5,000 Hz. The variations in signal level in each band were reproduced on one of eight low frequency bands of noise. The eight low frequency noise bands covered the frequency range from 400 to 800 Hz, and had the same variations (modulations) in signal level as the eight high frequency speech bands. Speech recognition ability was then measured when only those frequencies of the speech below 800 Hz were present, and when the modulated noise bands were added to the low-frequency speech signal. In a modified version of the system, the noise bands were omitted for speech sounds with substantial energy in the low frequencies. **Results.** The addition of the modulated low-frequency bands of noise improved the recognition of speech sounds having significant information in the high-frequencies, but degraded the perception of speech sounds with mostly low-frequency energy. The modified system maintained the advantage for the high-frequency speech sounds without degrading the perception of the low-frequency speech sounds. **Relevance to Veteran Population.** The results of this research will facilitate the development of special-purpose signal processing hearing aids that will improve speech intelligibility for veterans with severe high-frequency hearing loss.

Toward a Theory of Optimal Hearing Aid Processing James M. Kates *(p. 39)*

Purpose of the Work. In most cases, a hearing loss is the result of damage to the inner ear. An ideal hearing aid for such losses would restore the functioning of the impaired ear to match that of a normal ear. As a first step towards developing such a hearing aid, a computer program was used to simulate the behavior of normal and hearing-impaired ears. The purpose of this study was to develop a theoretical model of the impaired auditory system in order to design hearing aids that would work more effectively with the various forms of hearing impairment. **Procedures.** A simplified theoretical model of the ear for a specific hearing impairment was developed in order to enable the development of a practical amplification system. The hearing-aid processing was designed to minimize the differences between normal and impaired ears to the extent possible. Processing examples consisting of several individual speech sounds presented for a flat hearing loss. **Results.** The results indicated that a 3-channel compression system, having frequency bands that change in response to the frequency content of the signal, and gains that are controlled by the peak signal level within each frequency region, will be close to the optimal solution. **Relevance to Veteran Population.** Hearing aids designed according to the procedures developed in this paper will be of greater benefit to hearing-impaired veterans.

Spectral Contrast Enhancement of Speech in Noise for Listeners with Sensorineural Hearing Impairment: Effects on Intelligibility, Quality and Response Times Baer, et al. *(p. 49)*

Purpose of the Work. Hearing aid users typically have great difficulty understanding speech when background noise is present. The purpose of this study was to evaluate a new method of processing speech signals in noise to improve both intelligibility and sound quality. **Subjects.** Five to eleven subjects participated in a series of four experiments. All subjects had a sensory (cochlear) hearing loss. **Procedures.** Major peaks in the frequency spectrum of the speech signal were enhanced. Large amounts of enhancement were found to reduce intelligibility, whereas moderate amounts of enhancement produced no significant change in intelligibility. Subjective judgments of intelligibility and sound quality showed a preference for moderate amounts of enhancement. Combining moderate amounts of enhancement with a reduction in the range of intensity variation in the speech signal produced a significant improvement in the intelligibility of speech-in-noise after some practice with the processed signals. The method of processing was also evaluated using response time measurements. This method of measurement was found to be a more sensitive indicator of relative improvement. **Results.** Both the intelligibility and sound quality of speech-in-noise can be improved by enhancing peaks in the frequency spectrum combined with a reduction in the range of intensity variations of the speech signal. **Relevance to Veteran Population.** Signal processing of the type described in this paper could be incorporated in the future development of advanced signal processing hearing aids. These hearing aids would provide improved intelligibility and overall sound quality for hearing-impaired veterans.

Development and Assessment of Two Fixed-Array Microphones for Use with Hearing Aids Bilsen, et al.
(p. 73)

Purpose of the Work. Hearing-impaired listeners often have great difficulty in situations involving several competing sources of sound, such as at a party. Speech understanding under these conditions can be improved using a microphone that focuses on the speech signal coming from a given direction. This paper investigated the use of an array of microphones to provide the directionality needed to improve the intelligibility of a desired speaker against a background of other competing sound sources. **Subjects.** Thirty normal-hearing and 45 hearing-impaired subjects participated. **Procedures.** A microphone array was developed which could be mounted on the frame and legs of a pair of spectacles. Computer simulations showed that this arrangement could provide an improvement of 10 dB in speech-to-noise ratio at the higher frequencies. Laboratory measurements on an artificial head showed improvements of about 7dB. Clinical measurements on both normal-hearing and hearing-impaired listeners showed improvements in speech reception threshold equivalent to a 7 dB reduction in background noise. **Results.** Microphone arrays that can be mounted on a pair of spectacles can provide significant improvements in speech-to-noise ratio with corresponding improvements in speech intelligibility when the speech and noise come from different directions. **Relevance to Veteran Population.** Veterans who wear hearing aids and who have difficulty understanding speech when several other sources of sound are present would benefit from the use of microphone arrays of the type described in this paper.

Real-Time Multiband Dynamic Compression and Noise Reduction for Binaural Hearing Aids Kollmeier, et al.
(p. 82)

Purpose of the Work. Binaural hearing aids have many potential advantages (over conventional monaural hearing aids) which have yet to be investigated. The purpose of this study was to develop and evaluate an advanced binaural signal-processing system in which a binaural noise reduction technique is combined with a technique for matching variations in signal level and the resulting changes in loudness to the auditory characteristics of the impaired ear. **Subjects.** Six adults with sensorineural impairments (i.e., impairments of the sense organ and/or its neural connections in the cochlea) were tested. **Procedures.** The signals reaching each ear were subdivided into 24 frequency bands corresponding to the critical bands of the ear. The range of variation in signal level in each band was matched to the available range of hearing in the impaired ear. In addition, an adjustment for the changes in loudness resulting from the impairment

was implemented. The binaural noise reduction system amplified sounds coming from the front, and suppressed sounds, including reverberation, coming from other directions. **Results.** Several versions of the processing technique were tested and, if the method of processing was optimized for each subject, improvements in both intelligibility and sound quality were obtained. The binaural noise reduction system was found to work effectively for most subjects for a certain range of signal-to-noise ratios. **Relevance to Veteran Population.** Binaural signal processing techniques offer a means for improving both the quality and intelligibility of speech-in-noise for the large number of veterans who use hearing aids.

Digital Signal Processing (DSP) Applications for Multiband Loudness Correction Digital Hearing Aids and Cochlear Implants Dillier, et al. *(p. 95)*

Purpose of the Work. Recent advances in the development of electronic chips for digital signal processing allow for the flexible implementation of a large variety of speech processing techniques that could be of value in wearable devices such as hearing aids and cochlear implants. This paper evaluates several practical speech processing techniques of this type that could be implemented on a single chip. **Subjects.** Nine normal-hearing and six hearing-impaired subjects participated in an experiment on noise reduction, 13 users of conventional hearing aids evaluated a new method of loudness correction for speech in quiet and in noise, and five cochlear implant users evaluated new methods of coding speech signals for cochlear implants. **Procedures.** Noise reduction was achieved using two microphones feeding a signal processor which adaptively adjusted the inputs to improve the speech-to-noise ratio. The signal processor for loudness correction adjusted the sound level in three adjoining frequency regions to approximate normal loudness perception for hearing-aid users. The speech processor for cochlear implant users recoded the speech signal into sequences of pulses that were delivered to different regions of the cochlea by means of the implant. Unlike earlier methods of speech coding for cochlear implants, several pulse sequences were interleaved to stimulate several regions of the cochlea sequentially rather than simultaneously. **Results.** The two-microphone noise reduction technique was found to improve speech intelligibility in noise subject to certain design constraints which need to be incorporated into any practical system. The signal processor for loudness correction resulted in improved intelligibility for those subjects with poor speech discrimination using their own hearing aids. The new method of coding for cochlear implants was found to improve the discrimination of various consonants in speech, as well as providing useful information on the pitch of the voice. **Relevance to Veteran Population.** Improved methods of signal processing of the type

evaluated in this paper could be incorporated in wearable units, such as hearing aids or cochlear implants, for use by either hard-of-hearing or deaf veterans, respectively.

Design and Evaluation of a Continuous Interleaved Sampling (CIS) Processing Strategy for Multichannel Cochlear Implants Wilson, et al. *(p. 110)*

Purpose of the Work. The coding of speech information for cochlear implants can take different forms, including the use of analog or pulsatile stimulation. In analog coding, the speech waveform is delivered directly to the electrodes of the cochlear implant. In pulsatile coding, the speech signal is represented by trains of pulses which are delivered to the electrodes. The purpose of this study was to investigate an improved version of the pulsatile method of stimulating a cochlear implant. An Ineraid cochlear implant was used in these studies. This device allowed for multichannel stimulation (i.e., several different frequency regions in the cochlea could be stimulated with different electrodes). **Subjects.** Nine deaf subjects who had been fitted with an Ineraid cochlear prosthesis participated in the study. Seven of the subjects were selected for their high levels of speech recognition with the Ineraid cochlear implant. Two were selected for their relatively poor performance. **Procedures.** An improved method of digital stimulation was developed in which the pulse trains to different frequency channels were interleaved to avoid the inter-channel interference that occurs with simultaneous stimulation of two or more frequency regions in the cochlea, as occurs with the analog method. Standard speech recognition tests were performed for both the analog and pulsatile methods of stimulation. **Results.** The new pulsatile method of stimulation resulted in significant improvements in speech recognition relative to the traditional analog method of stimulation. Whereas all of the subjects showed some improvement in speech recognition, the two low performing subjects showed by far the largest improvements on a relative (ratio) scale. **Relevance to Veteran Population.** The new pulsatile method of stimulation may provide substantial improvements in speech recognition for those veterans who have been fitted with a multi-channel cochlear implant.

Effects of Noise and Noise Reduction Processing on the Operation of the Nucleus-22 Cochlear Implant Processor Mark R. Weiss *(p. 117)*

Purpose of the Work. The Nucleus-22 implant system transmits electrical impulses to an array of electrodes that have been implanted in a damaged cochlea. These impulses encode characteristics of the sounds that are received by the implant system microphone and measured by the system's signal processor. When the characteristics accurately represent information bearing speech cues, the ability of the user of an implant to understand speech can be increased. However, when speech is accompanied by noise, the accuracy of the extracted information will be degraded and speech perception will decrease. The purpose of this study was to develop a method of noise reduction for cochlear implants. **Procedures.** Measurement errors for signals with and without noise reduction provided by the experimental method of processing were compared. The data were then analyzed to determine the reliability of the observed reduction in measurement errors. **Results.** The data showed a significant improvement in the accuracy with which essential speech parameters could be measured. As a consequence, cochlear implants of this type (using this mode of processing), function more effectively in noisy environments. **Relevance to the Veteran Population.** Veterans who have been fitted with this type of cochlear implant, if modified to include this type of noise reduction processor, would be better able to understand speech in a noisy environment.

Measurements of Acoustic Impedance at the Input to the Occluded Ear Canal Larson, et al. *(p. 129)*

Purpose of the Work. Measurement of the acoustic impedance of the ear has several useful clinical applications. Test instruments currently used for the clinical measurement of acoustic impedance are limited to only one or two frequencies over a narrow frequency range. This paper reports on the development and evaluation of a new technique for measuring acoustic impedance for many frequency components over a wide frequency range. **Subjects.** Thirty-five normal-hearing subjects at two locations were tested. Ages ranged from 25 years to 35 years. **Procedures.** Estimates of the length and diameter of the ear canal were obtained on each subject. Acoustic impedance measurements using the new instrument were then obtained for a prescribed position in the ear canal and also for two standardized acoustic cavities used for calibration. The acoustic impedance of the eardrum was then derived from these measurements and the estimated dimensions of the ear canal. Computer-based techniques were used to automate the method of measurement and subsequent calculations. **Results.** The measurement technique was found to be both precise and practical using the computerized system. The measurements obtained in the study provided normative data on the acoustic impedance of the ear for multiple frequencies over a wide frequency range. These normative data cover both the expected impedance values and possible range of variation for normal ears. **Pelevance to Veteran Population.** The measurement techniques developed and evaluated in this study will result in more effective clinical procedures for diagnosing hearing impairments in veterans and more effective methods of prescribing hearing aids.

Achieving prescribed gain/frequency responses with advances in hearing aid technology

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Abstract—Technological limitations have restricted the capability of older generation in-the-ear (ITE) hearing aids to closely match prescribed real ear gain/frequency responses. Newer technology, widely available in currently marketed ITE hearing aids, has considerably improved this capability. Data for 60 ears are presented comparing the real ear insertion gain (REIG) actually achieved to the target REIG, using ITE hearing aids having: 1) older generation narrow-band receivers, and amplifiers with single-pole-filter low frequency tone control and a class A amplifier output stage (n = 30), and 2) newer generation amplifiers with a two- or four-pole-filter low frequency tone control, and wide band receivers, containing a class D amplifier output stage (n = 30). With the newer technology ITE hearing aids, the means and ranges of deviation from target gain were reduced. Capability for achieving prescription REIG with ITE hearing aids can be further improved with multichannel amplifiers. Examples of the latter are shown for several difficult-to-fit audiograms.

Key words: *frequency responses, hearing aid technology, in-the-ear hearing aids, real ear gain.*

INTRODUCTION

Over the past 15 years, numerous prescription formulas for hearing aid fitting have been proposed, each specifying a desired gain/frequency response based on the audiologic test results of a given hearing-impaired patient (1,2). The majority of these formulas use audiometric thresholds and/or suprathreshold loudness judgments in their calculations, with the most common goal that of maximizing audibility of the speech spectrum or of amplifying the speech spectrum to "most comfortable" listening level. Given the current wide availability and use of probe microphone measurement systems, prescriptive fitting of the real ear insertion response (REIR) appears to have largely replaced the traditional, comparative speech approach (3), at least for initial selection of amplification parameters.

Unfortunately, a problem has arisen in the practical application of prescriptive approaches to hearing aid fitting. In-the-ear (ITE) hearing aids, which make up the largest percentage of hearing aids sold today (4), have been reported to be too inflexible to adequately match prescribed gain/frequency responses (5,6). The primary reason for these negative findings to date has been limitations in past hearing aid technology. Recent advances in technology, however, may make closer approximation to prescribed gain/frequency responses possible.

Past Limitations and New Advances in Technology

Two major limitations in hearing aid technology have been additive in causing the limited capability of many older technology single-channel ITE hearing aids to provide a good match to prescribed frequency response:

1. *Older-vintage hearing aid receivers had a primary frequency response peak in the 1600 to 2200 Hz frequency range and inadequate high frequency sensitivity.* The use of these receivers frequently resulted in two problems. First, a large dip was often seen in the REIR in the frequency range of the peak in the real ear unaided frequency response (REUR), usually at about 2800 Hz. Second, there was inadequate high frequency REIR above about 3000 Hz.

2. *Tone controls available had a limited range of variability.* Until recently, most ITE hearing aids had low frequency tone controls comprised of a single-pole highpass filter, primarily because of physical space limitations preventing the packaging of additional capacitors required for more filter poles. These low frequency tone controls were quite limited in their ability to vary the frequency response, having a slope of only 6 dB per octave. For example, it was possible to reduce gain at 500 Hz by only about 10 dB without changing the high frequency gain significantly.

Because of these limitations, a variety of methods were used in the attempt to fine-tune the frequency response of these ITE hearing aids to better match the prescription target. One technique used was to overfit the overall gain of the hearing aid in order to achieve the required high frequency gain and then to rely on varying the low frequency tone control to match the prescribed gain values at low and mid frequencies. However, since the low frequency tone controls had such a limited range, the resulting frequency response often still exceeded target gain in the low and mid frequencies. The desired high frequency insertion gain target was seldom achieved anyway, even with the highest possible overall gain provided. This was especially true for patients with steeply sloping audiograms and for those with hearing loss confined to the high frequencies.

Limited improvement was sometimes obtained with electronic shifting of the peak in the frequency response of the receiver to a higher frequency. This was accomplished by placing an appropriate-value capacitor across the receiver and/or by using a constant voltage rather than a constant-current amplifier output stage. Some hearing aid designers also turned to mechanical/acoustical techniques to increase high frequency gain relative to low and mid frequency gain. These included using a low-cut response microphone and a stepped bore earmold in conjunction with damping in the hearing aid receiver tubing (7). The stepped or flared earmold bore produces a miniature megaphone in order to match the high acoustic source impedance of the receiver to the low acoustic impedance of the ear canal at the high frequencies. When combined with the newer, wider-band receivers, this approach also achieved some improvement in frequency response fitting.

More recent ITE hearing aids that are on the market employ "active" tone controls, consisting of two- or even four-pole filters, having slopes of 12 dB/octave and 24 dB/octave, respectively. These higher-order filters provide much greater flexibility in varying the frequency response. It is now not uncommon to have low frequency tone controls with a 40 dB range at 500 Hz.

In the past, most ITE hearing aids utilized a class A power output stage in the amplifier. Although not directly related to the number of filter poles, hearing aids with a class A output stage are often associated with a narrow-band receiver response and limited tone control flexibility. Many ITE hearing aids currently being produced employ either a class D or class B output stage in the amplifier and a receiver with a wide-band frequency response. These newer hearing aids typically have a primary frequency response peak at about 3000 Hz and provide adequate gain in the high frequencies. Thus, many ITE hearing aids with good insertion frequency responses are associated with amplifiers that have class D and class B output stages.

METHOD

Comparison of Older Versus Newer Technology Hearing Aids

In order to illustrate the degree of improvement seen in frequency shaping flexibility with older versus newer single-channel technology, data are presented here for 30 ears fit with older-technology

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ITE hearing aids that used linear, class A amplifiers and relatively narrowband receivers, and 30 ears fit with newer-technology class D amplifiers with wider band receivers. The hearing aids with class A amplifiers (standard linear, from Argosy Electronics, Inc., Minneapolis, MN) used single-pole filters for the low-frequency tone control. Five of the class D amplifiers had low-frequency tone controls comprised of two-pole filters (Argosy Linear Plus®) and the remaining 25 were comprised of four-pole filters (Argosy Manhattan II®). Due to the small sample size for the two-pole class D amplifier, the data are presented as a group. Examination of individual data, however, revealed no apparent difference in results across the two class D devices.

These data were obtained retrospectively from patient files of veterans who received hearing aids at the Department of Veterans Affairs Medical Center in Nashville, TN, and represent, therefore, typical clinical accuracy in routine hearing aid fittings. The data shown for the older technology ITEs were collected in late 1989, and the data shown for the newer technology ITEs were collected in late 1991 by the same two audiologists. Shown in **Table 1** are the mean audiograms, and standard deviations, for the ears fit with newer versus older technology hearing aids. Audiometric configurations within each group ranged from mildly to severely sloping.

The following protocol was used for all fittings. For each ear, desired gain/frequency response was calculated using the NAL-R—revised National Acoustics Laboratories formula (8) and a custom ITE was ordered.[1] With an order, the manufacturer

Table 1.
Means and standard deviations of audiological thresholds for each group of ears.

		Frequency (Hz)						
		250	500	1000	2000	3000	4000	6000
Older	x̄	29	29	31	49	63	73	74
technology	(sd)	(14)	(15)	(16)	(20)	(19)	(16)	(17)
Newer	x̄	30	33	39	53	63	71	73
technology	(sd)	(13)	(14)	(14)	(13)	(19)	(19)	(21)

x̄ = mean
sd = standard deviation

was given only the earmold impression and the prescribed full-on 2 cc coupler gain values. Audiograms were not supplied. To provide maximum flexibility in fitting to prescribed values, each hearing aid was ordered with a tone control and variable venting inserts. In addition, most of the Manhattan II hearing aids had a bandpass trimpot. When a hearing aid arrived from the manufacturer, electroacoustic evaluation was accomplished first to ensure proper functioning (ANSI S3.22 - 1987), then the hearing aid was fit to the patient's ear using probe-microphone measurement of real ear insertion gain (REIG) with a Fonix model 6500 instrument. A broadband composite signal was presented at 70 dB SPL from a loudspeaker positioned 1 meter from the subject at a 45° azimuth. Trimpots and/or venting were adjusted as necessary to achieve the closest possible match to NAL-R prescribed real ear insertion gain (REIG) values at each frequency. For the data presented here, the REIG values obtained were compared to the insertion gain values that had been prescribed, in order to determine the degree of deviation from target.

RESULTS

The means of the 30 prescribed REIGs at each frequency (filled circles), and the means of the 30

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[1] It is notable that current prescriptive formula approaches were not specifically developed for use with adaptive-frequency-response (AFR) hearing aids such as the Argosy Manhattan II or K-Amp® (Etymotic Research, Inc., Elk Grove Village, IL). In fact, the prescribed frequency response is an attempt to provide the best compromise when a single, fixed frequency response must be used across multiple listening environments. Unfortunately, however, there is no established approach for the setting of electroacoustic parameters in AFR hearing aids. For our purposes, we chose to set the gain/frequency response of the AFR hearing aids to match prescribed values with a moderate speech-level input (70 dB SPL), on the assumption that adaptive processing will occur at higher and lower input levels.

An argument can also be made for specifying different frequency responses for AFR hearing aids than those prescribed with current formulas. Take, for example, an AFR hearing aid that functions in a manner similar to the Argosy Manhattan II; that is, low frequency gain is reduced as input level increases. The assumptions underlying this approach are that high-level, low frequency energy is more likely to be noise than speech, and that if excessive upward spread of masking occurs in a hearing-impaired ear, it will be reduced. It can be argued that more gain should be supplied in the low frequencies for this type of processor than is currently prescribed, particularly when fitting patients

with normal or near-normal thresholds in the lower frequencies. The rationale is that low frequency gain will be sufficiently reduced in high-level noise environments by the adaptive function to provide the above benefits, but that the greater low frequency gain supplied in low-level, quiet environments will actually enhance perceptual sound quality (10). It is clear that further research into the development of appropriate fitting techniques for AFR hearing aids is needed. At present, however, we would encourage the use of broadband stimuli presented at multiple input levels to more fully characterize the functioning of these devices (11). In addition, evaluation of speech recognition in quiet and in noise is important, and adequate follow-up is a crucial factor in assuring user success.

REIGs actually obtained for each fitting (open circles), are shown in **Figure 1a** for the older-vintage hearing aids, and in **Figure 2a** for the newer technology hearing aids. Data at 250 Hz were eliminated from the figures due to noise problems in the real ear measurements at this frequency. Shown in **Figure 1b** and **Figure 2b**, for the older and newer technology ITEs, respectively, are the means and standard deviations of individual deviations from target; that is, the differences between prescribed insertion gain and the insertion gain actually obtained with each hearing aid. A positive value indicates that the gain obtained was greater than that prescribed and a negative value indicates that the gain obtained was less than that prescribed. Linear regression lines and values are also shown in **Figure 1b** and **Figure 2b**.

Note that the older technology hearing aids (**Figure 1**) tended to provide too much gain through the mid frequencies and too little gain in the high frequencies relative to prescribed values, as reflected both in the mean data and in the downward slope of the regression line in **Figure 1b**. In fact, adequate gain was achieved at 4000 Hz in only two of the 30 ears, and too much gain resulted at 1500 Hz in 24 of the fittings.

In contrast, the newer technology hearing aids (**Figure 2**) provided, on average, closer approximations to prescribed REIG across the frequency range. In particular, the flatter regression line, and better fit to the mean data, indicates an improvement in the ability to achieve sufficient gain at 4000 Hz without excessive mid-frequency gain. Smaller standard deviations indicate the reduced spread in the data for newer technology hearing aids.

Another way to measure the central tendency of the data is to examine the number of fittings that fall within an acceptable degree of error from target REIG. We generally consider plus or minus 5 dB from target to be acceptable. **Table 2** lists, for older versus newer technology hearing aids, the percentage of fittings at each frequency in which the REIG actually fit was within ±5 dB of the prescribed REIG. Across mid and high frequencies, a substantially higher percentage of fittings fell within these guidelines for the newer technology ITEs than for the older technology ITEs, again indicating a large improvement in frequency response shaping capability.
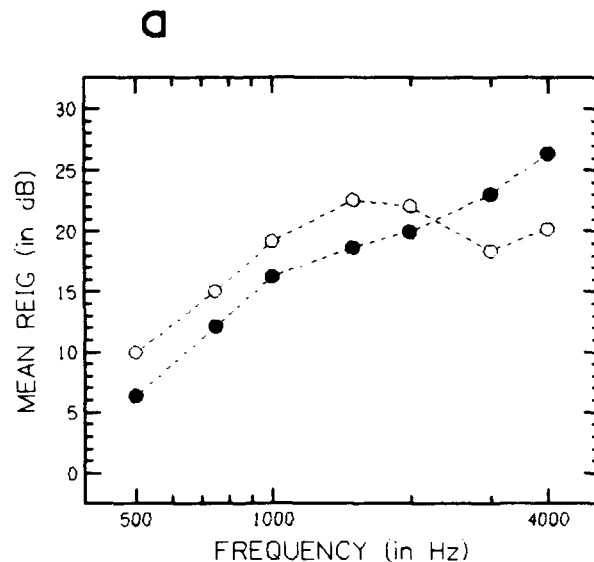
Figure 1a.
Closed circles represent the mean prescribed insertion gain/frequency response, and open circles represent the mean REIG actually achieved, for the 30 older technology ITEs.
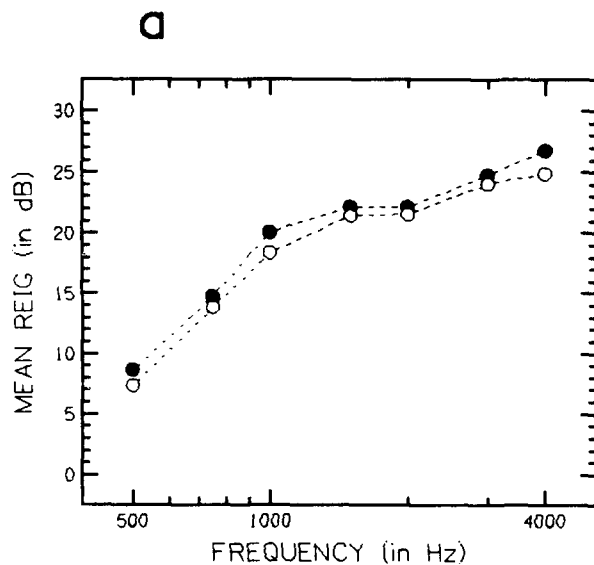
Figure 2a.
Closed circles represent the mean prescribed insertion gain/frequency response and open circles represent the mean REIG actually achieved, for the 30 newer technology ITEs.

Multi-Channel Amplifiers

Historically, most hearing aid amplifiers have been single-channel devices. In order to compensate for the insertion gain dip at 3000 Hz, some

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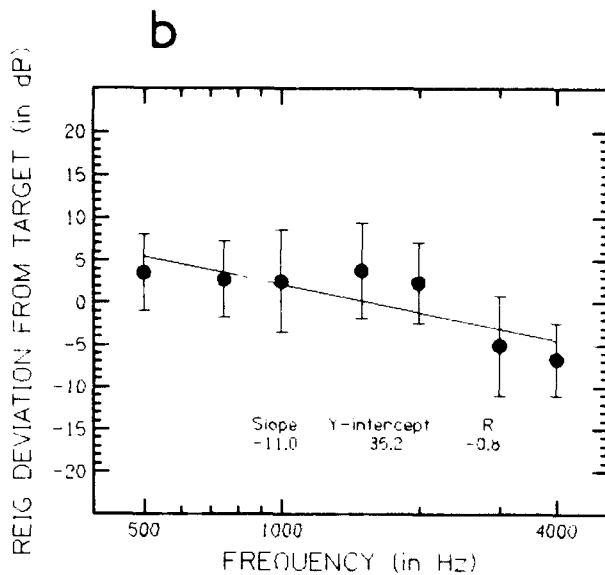
Figure 1b.
The means and standard deviations of individual deviations from target for each of the 30 older technology ITEs (i.e., the differences between prescribed insertion gain and the insertion gain actually obtained). A positive value indicates that the gain obtained was greater than that prescribed and a negative value indicates that the gain obtained was less than that prescribed. The linear regression line is also shown.
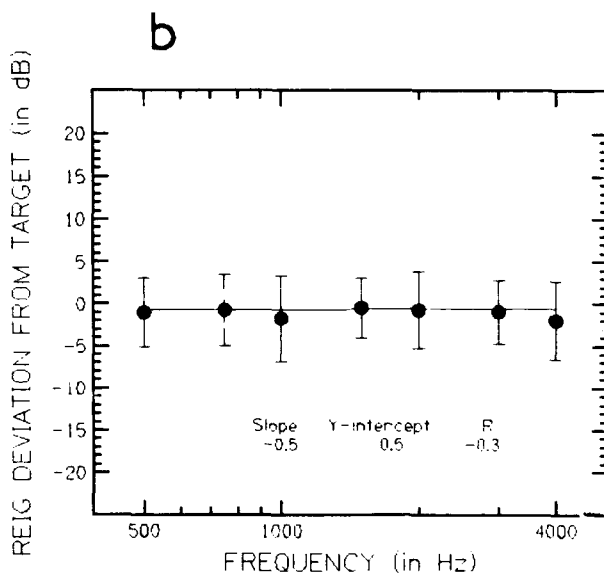
Figure 2b.
The means and standard deviations of individual deviations from target for each of the 30 newer technology ITEs. A positive value indicates that the gain obtained was greater than that prescribed and a negative value indicates that the gain obtained was less than that prescribed. The linear regression line is also shown.

Table 2.
Percentage of ears in each group with deviation from target REIG of plus or minus 5 dB or less.

| | Frequency (Hz) | | | | | | |
	500	750	1000	1500	2000	3000	4000
Older technology	67	67	57	70	67	53	40
Newer technology	67	80	77	93	73	83	80

manufacturers have successfully employed a second channel of amplification via an additional bandpass filter to add gain in the 3000 Hz frequency range. An example of this approach is shown in **Figure 3a**. The effect of the second channel on the HA-1 2 cc coupler frequency response is shown in **Figure 3b**. The effect on frequency response measurements in a real ear, obtained with an Acoustimed HA-2000, is shown in **Figure 3c**. A significant increase in mid and high frequency REIG is seen with counterclockwise rotation of the bandpass potentiometer in this ITE hearing aid.

One of the most recent technological advances in hearing aids has been incorporation of a graphic or parametric equalizer within the hearing aid amplifier. Multichannel amplifiers give much finer resolution in frequency response shaping than do single-channel amplifiers. With a graphic equalizer, gain is controllable in each frequency band. With a parametric equalizer, gain is controllable in each frequency band, but, in addition, the crossover frequencies of the bands can be shifted.

The ability to precisely match a prescribed gain/frequency response with multichannel devices greatly exceeds that of older single-channel devices. Three individual examples of matches between measured and NAL-R prescribed REIG with a 3-channel parametric equalizer ITE hearing aid (Argosy 3-Channel-Clock®) are shown in **Figure 4**. **Figure 4a** illustrates a match obtained for a severely sloping audiogram with normal hearing through 2000 Hz. This audiometric configuration is particularly difficult to fit with single-channel hearing aids because the actual insertion gain achieved will often be less than the target gain prescribed in the high frequencies, and greater than that prescribed in the mid frequencies. **Figure 4b** illustrates the match obtained for an audiogram with a rising configuration.
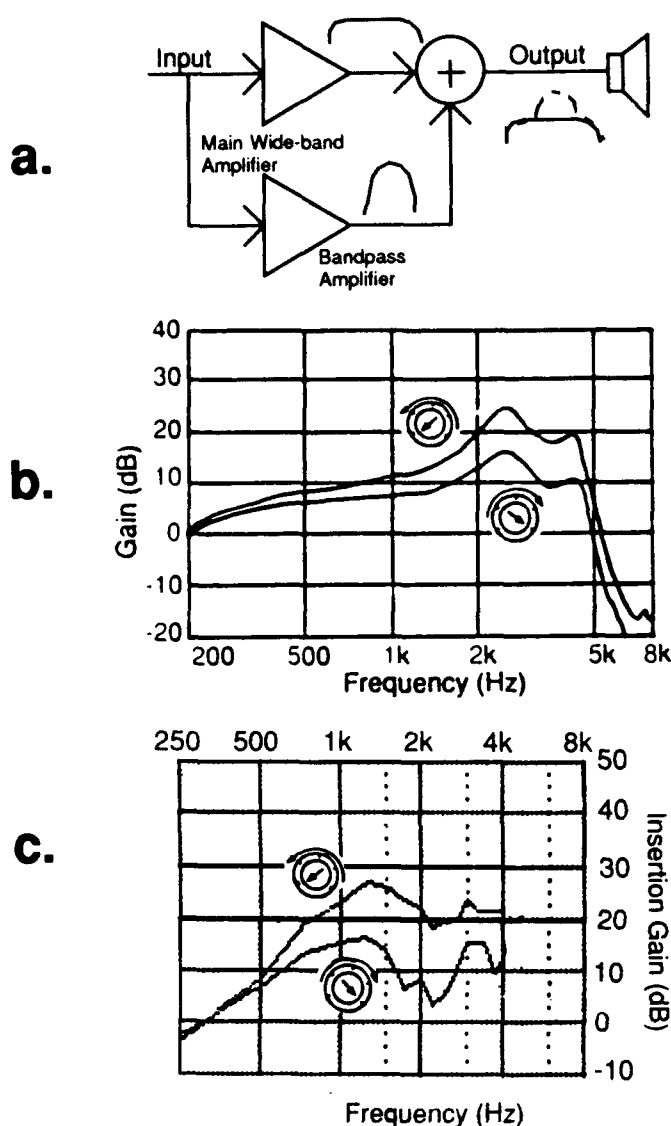
Figure 3.

Figure 3a illustrates an example (Argosy Manhattan II) of the use of an additional amplifier channel with bandpass filter, added to a single-channel amplifier to obtain an increase in gain in the mid to high frequency region. The effect on 2 cc coupler gain of adding the second channel is shown in Figure 3b, and with probe microphone measurement of REIG in Figure 3c.

Finally, **Figure 4c** illustrates a "reverse cookie-bite" configuration; that is, a region of normal hearing in the mid frequencies with hearing loss confined to higher and lower frequencies. For this latter audiogram, it is typically impossible to achieve a reasonable approximation to prescribed gain using single-channel ITE technology, and possible, but difficult, with acoustic earmold or earhook modifications in behind-the-ear hearing aids. Note the accuracy with which the gain/frequency response

matched the prescribed REIG for all three cases with the 3-channel parametric equalizer ITE hearing aid.

DISCUSSION

The data shown indicate substantial improvement in the capability of newer technology ITE hearing aids to achieve good approximations to prescribed gain/frequency responses. With two- or four-pole filters for low frequency tone controls, and wide-band receivers with class D amplifier output stages, adequate high frequency gain can be achieved without overamplification of the mid frequencies. Although the hearing aids used in this study were from Argosy Electronics, these technological innovations are also available from other manufacturers and we would expect to see comparable results. The flexibility of the three-band ITE hearing aid with parametric equalization is sufficient to supply an accurate match even with more unusual audiograms that have previously been quite difficult to fit.

This improved flexibility also provides greater opportunity for successful revision of a patient's gain/frequency response, if this is considered desirable after the initial fitting to prescribed values. Because prescriptive formulas are based on mean data for several parameters, the optimal frequency response for an individual patient may in fact differ from the prescribed target. A number of researchers have argued that a prescribed gain/frequency response should be considered only as a "ballpark" starting point in the fitting process, followed by adjustment based on evaluation of speech understanding ability and/or perceptual sound quality, and with follow-up regarding patient satisfaction (8,9). Modifications to the prescribed frequency response for a patient whose speech recognition performance is poorer than expected would typically be in the direction of increased high frequency gain relative to low frequency gain. The data presented here suggest that such modification will be achieved more easily with newer technology than it has been in the past.

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a.

b.

c.
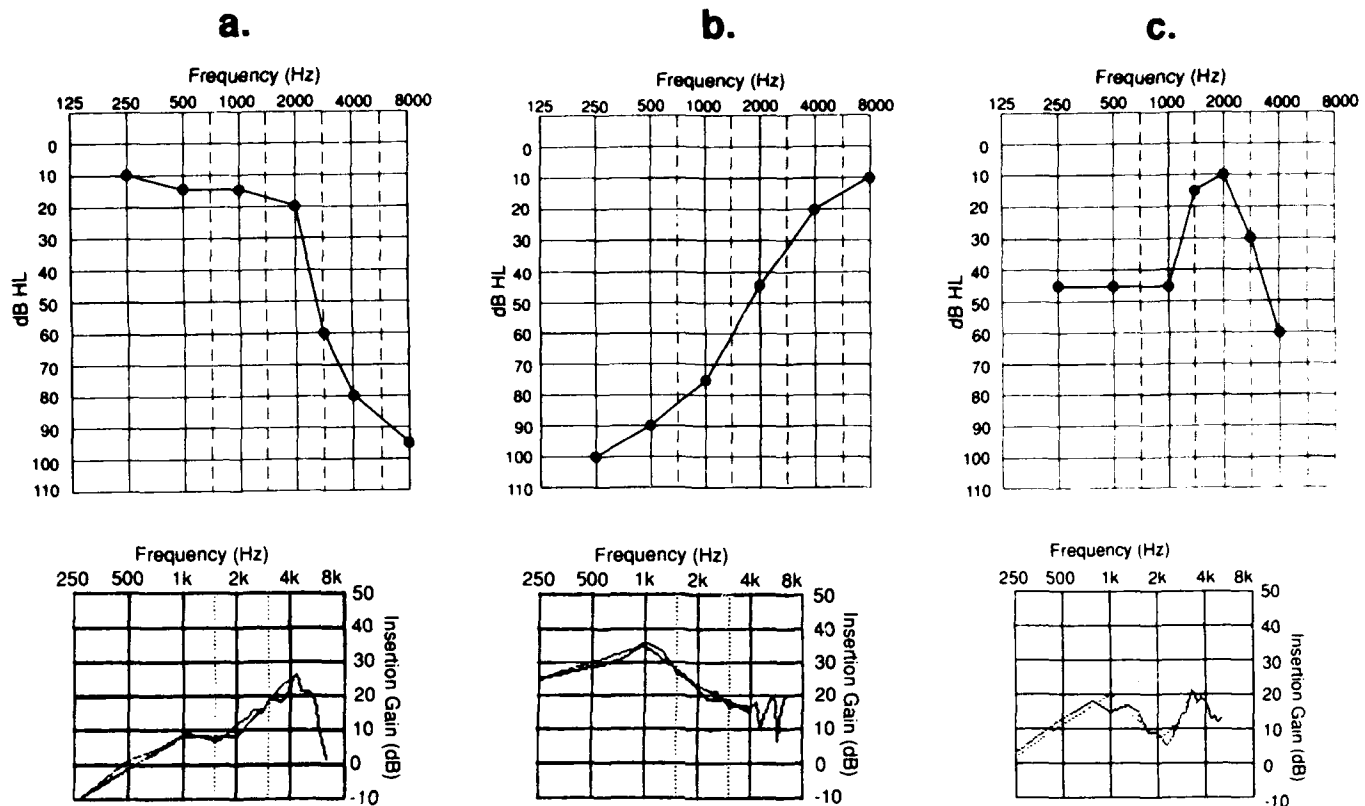
Figure 4.
Three examples of the match to prescribed REIG target obtained with a 3-channel ITE hearing aid with parametric equalization (Argosy 3-Channel-Clock). For each case, the audiogram is shown in the upper part of the figure, and REIG measurements in the lower part of the figure, with the smoother line representing the target insertion gain.

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Properties of an adaptive feedback equalization algorithm

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Editor's Note:

This is the first of two related papers in this issue on the subject of adaptive feedback equalization in digital hearing aids. This article describes an adaptive algorithm that estimates and tracks the characteristic of the hearing aid feedback path.

The second article, "Behavioral Assessment of Adaptive Feedback Equalization in a Digital Hearing Aid," by Marilyn French-St. George, et al., describes the results of behavioral testing of the algorithm in subjects with hearing impairment.

Abstract—This paper describes a new approach to feedback equalization for hearing aids. The method involves the use of an adaptive algorithm that estimates and tracks the characteristic of the hearing aid feedback path. The algorithm is described and the results of simulation studies and bench testing are presented.

Key words: *acoustic feedback, adaptive feedback equalization, feedback instability, new adaptive equalization algorithm.*

INTRODUCTION

Feedback instability is a problem with hearing aids that often results in performance degradation and reduced benefit for the listener with hearing impairment. Feedback instability reduces battery life. It limits the gain that can be prescribed.

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Furthermore, it is often a source of embarrassment for the wearer when the hearing aid breaks into a loud oscillation at inappropriate times. The problem is most serious with high-power hearing aids that develop high gains and with in-the-ear packages where acoustical and mechanical isolation between receiver and microphone is limited.

The theory of feedback instability is well understood (1). The basic concept of feedback is illustrated in **Figure 1a**. If the feedback signal, $Z(f)$, is out of phase with the input, $X(f)$, negative feedback results that is often used in the design of systems to stabilize them and make the systems less sensitive to component variation. If the feedback signal is in phase and greater in amplitude than the original input signal [the loop gain, $G(f)*A(f)$, is greater than 1], the system becomes regenerative and unstable. In the case of acoustic feedback, the phase of the feedback signal is a function of frequency. Therefore, the feedback signal will be in phase and out of phase at several frequencies within the bandwidth of the system, and if the loop gain is

Figure 1.

Fundamental models of feedback and feedback equalization. *a*. Basic hearing aid system with feedback. G(f) represents the desired hearing aid characteristic. A(f) represents the undesirable feedback path that can cause instability. *b*. Equalization filter E(f) added to cancel A(f). *c*. Equalization filter adapts so that the error signal, Z(f) is minimized in the least-mean-square sense.

in phase but at a gain that is slightly less than 1. In this case, the system will be stable but underdamped and will exhibit aberrant resonant peaks that are undesirable. Since many people who wear hearing aids adjust the volume control to just below the point where the hearing aid oscillates, it is likely that many are experiencing a highly resonant, underdamped characteristic.

The problem of hearing aid instability is difficult to solve. First of all, practical considerations are such that the acoustic output and input of the hearing aid cannot be well isolated. People often prefer small, in-the-ear hearing aids for reasons of aesthetics and convenience. Others prefer behind-the-ear hearing aids with earmolds that are vented or open to provide greater comfort. Tight-fitting, unvented earmolds are uncomfortable, collect moisture, and exhibit the occlusion effect wherein the wearer hears his/her own voice. With open earmolds, the magnitude of the feedback path is often close to unity and the phase varies on the order of 180 degrees per 1,000 Hz. Therefore, we are forced to deal with basically an unstable system when we try to achieve acoustic gains of 20 to 40 dB over a bandwidth of 8 kHz or more.

Egolf (2) has reviewed the acoustic feedback literature and describes a number of potential methods for stabilizing hearing aids and detecting instability. These methods include use of tunable notch filters, frequency shifting, phase shifting, and frequency modulation for stabilization and correlation and phase-lock-loop methods for detecting oscillation. Each of these methods seems to have a number of deficiencies as applied to hearing aids. The notch filter approach requires that the offending frequency of instability is known. However, if the frequency is known, improvements in gain of 7–17 dB can be achieved. The other three methods provide only modest improvement in gain margin (6 dB) and introduce perceptible distortion. A different approach that is more attractive is active equalization, which is shown in **Figure 1b**. The idea here is to simulate the feedback path of the hearing aid with an electronic filter, E(f), that is connected in parallel with the feedback path, A(f), to cancel the signal that feeds back at the input to the hearing aid, Z(f). This approach works well for time-invariant systems. Egolf reported that, with considerable fine-tuning, gain margins of 15 to 20 dB were achieved in the laboratory. However, since the

greater than 1 at these frequencies, the acoustic system will oscillate. Acoustic systems can oscillate at multiple frequencies, depending on the phase and amplitude relations of the loop gain with respect to frequency.

It is also possible for a feedback system to perform poorly but yet be stable in the sense of the above discussion. This can occur if the loop gain is

feedback path of the hearing aid is constantly changing, in order to get good cancellation the equalization filter must adapt to accommodate those changes, as shown in **Figure 1c**.

Studies of adaptive equalization filters with wearable hearing aid systems have been reported by Dyrlund and Bisgaard (3) and Engebretson et al. (4,5,6). Both groups of investigators have obtained similar results, that stable gain margins of hearing aids can be improved by 10 to 15 dB with adaptive equalization. Not only does adaptive equalization suppress oscillation, but it equalizes the under-damped, non-oscillatory feedback condition that often occurs at high gains and tends to degrade hearing aid performance.

The purpose of this paper is to describe the performance and limitations of one such adaptive equalization algorithm that has been implemented in digital form and that is the basis for the behavioral study reported in another paper in this issue (7). The algorithm is incorporated into a wearable version of the CID digital hearing aid (8) and differs in a number of ways from other implementations. First, the digital hearing aid uses logarithmic arithmetic to simplify the very large scale integrated (VLSI) circuitry. Second, the coefficients of the adaptive filter are implemented as up-down counters, which reduces the complexity of the VLSI circuitry further. The goal here is to achieve a circuit design of modest complexity that is practical to implement in the form of a small semiconductor chip that is compatible with the constraints of an ear-level hearing aid package without compromising the functionality of the digital hearing aid. Therefore, considerable attention has been given to developing an adaptive algorithm that is optimal with regard to perfor-mance, power consumption, and size.

The remainder of the paper is organized as follows. First, the feedback characteristics of a typical hearing aid are examined. Second, the model of feedback equalization is described. Simulation studies of the model are presented to demonstrate the performance of the binary, logarithmic adaptive algorithm. Third, results of bench tests using a KEMAR mannequin are presented. The relation between filter length and degree of equalization is examined. Results of tests of the system in subjects with hearing impairment are presented in the related article in this issue (7).

METHODS/RESULTS

Nature of Feedback Path

The feedback path of a hearing aid includes both mechanical and acoustical coupling between the receiver and microphone. However, the acoustic leakage path is the primary one. This is illustrated in the measurements of **Figure 2**, which were obtained with an experimental in-the-ear module containing a typical hearing aid receiver and microphone. The measurements were made on a KEMAR mannequin with a Zwislocki coupler using instrumentation amplifiers and phase meter. The receiver was excited with a 70 mV rms voltage drive, and the microphone preamplifier signal was measured with phase refer-enced to the receiver voltage. The condition shown in the figure is for a relatively loose-fitting ear module.

These results are typical. The feedback coupling is poor at low and high frequencies and one or more peaks representing resonances lie in between. The phase is approximately linear with respect to fre-quency and represents a propagation delay of about one-half millisecond, which is equivalent to an acoustic path length of 16 cm. Since the dimensions of the ear module are much smaller than 16 cm, the phase of the feedback characteristic is not domi-nated by acoustic delays. Instead, phase is primarily determined by the delay of the receiver. The amplitude of the feedback signal changes with tightness of fit between the ear module and ear. The

Measured Feedback Characteristic
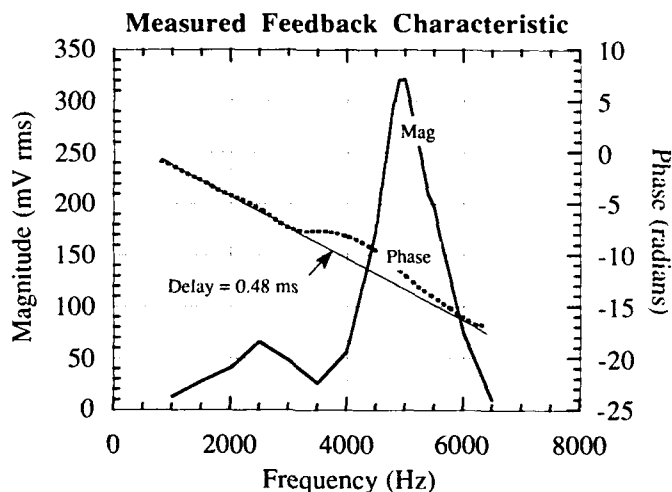
Figure 2.
Measured feedback characteristic of typical ear module. Typical are the multiple, low-Q resonances and large phase angles.

better the ∩l, the lower the amplitude. However, the phase remains about the same. Objects and surfaces in the vicinity of the ear modify the resonant peaks in both amplitude and frequency by creating standing waves. For example, the presence of a hat brim near the ear can increase the amplitude of the feedback signal and is known to cause hearing aids to oscillate.

The results described above are consistent with the models of Egolf et al. (9) and Kates (10). However, they differ somewhat. Egolf et al. analyzed an eyeglass-style hearing aid but did not incorporate a hearing aid receiver in the mathematical model or include an ear canal. They modeled the transfer characteristic from a point near the external opening of the vent of the earmold to the microphone port located on the eyeglass frame. Therefore, the primary delay in their model is acoustic and the phase shifts are considerably less than what we observed. Kates included a model of the receiver, ear canal, and vent in his simulation. However, the simulation exhibited a sharp, single resonance in the transfer function of the feedback path at about 7 kHz, which is in contrast with our observations of relatively low-Q, multiple resonances at much lower frequencies. The reason for pointing out these differences is that they have an impact on the choice of parameters of the equalization mode. For example, the low-Q resonances observed in our hearing aid system require a shorter equalization filter than the high-Q resonance of the Kates model to achieve the same degree of cancellation. We have observed, however, that greater delays will require longer filters.

Feedback Equalization Model

It appears that the feedback path, whether it be a leak around the earmold, through a vent, or both, can be modeled as a filter. This is shown in Figure 3 where the external feedback path, which is a composite of all sources of feedback, is represented by H_f. The feedback equalization filter, H_e, is connected between the output of the aid and the input. The output of the equalization filter is subtracted from the input signal to cancel the contribution from the external feedback path. The error signal, E, is the difference between the two paths and is used to adaptively adjust the coefficients to minimize this difference in a least-mean-
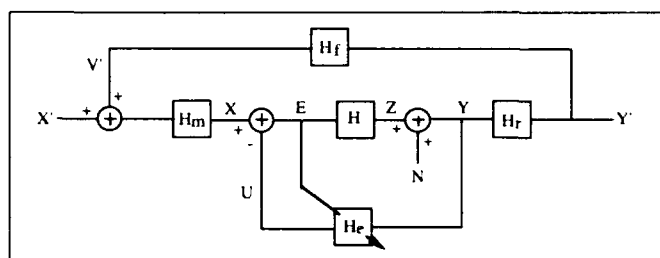
Figure 3.
Model of adaptive feedback equalization algorithm. H_m represents the transfer characteristic of the microphone, preamp, and analog-to-digital converter and H_r represents the transfer characteristic of the digital-to-analog converter, power amplifier, and receiver. The product, $H*H_m*H_r$, represents the desired acoustic transfer characteristic of the hearing aid.

square sense (11). A random noise source, N, which is uncorrelated with other signals in the system, is included to excite the system when signals are small. The noise source typically is set to a level that is low enough to be unobtrusive to the listener. Other sources of random noise also excite the system and serve the same purpose as is described below.

In the diagram in **Figure 3**, H_m and H_r represent the transfer characteristics of the microphone and receiver, respectively. H_m includes the analog-to-digital converter transfer characteristic and H_r includes the digital-to-analog converter transfer characteristic. H_f and H_e are as described above and H is the desired prescriptive frequency-gain function of the hearing aid. The signals X' and Y' represent the acoustic input and output of the hearing aid, respectively, and X and Y represent the digital equivalent of these signals, including the transfer characteristics of the microphone, preamplifier, ADC, DAC, power amplifier, and receiver.

The adaptive algorithm that is used to adjust the coefficients of H_e is based on the LMS algorithm (11). The expression

$$u(n) = \sum c_i y(n - i)$$

is a filtered version of the output signal, y(n), that is an estimate of the external feedback path. The cancellation error is:

$$e(n) = u(n) - v(n)$$

where v(n) represents the external feedback signal that the adaptive filter is attempting to cancel. The

mean of the squared error, $e^2(n)$, has a unique minimum with respect to the coefficients of the equalization filter, H_e. The coefficients are adaptively driven toward this minimum by using the gradient of the error to determine the direction of steepest descent. An algorithm for adjusting the coefficients that requires no direct calculation of the gradient and no multiplications is given by:

$$C_k(n+1) = C_k(n) + \lambda \, \text{sign}[y(n-k) \, e(n)]$$

where λ is a constant that is a power of 2. Therefore, updating the coefficients is equivalent to incrementing or decrementing the coefficient register, depending on the value of the sign function. The sign function is a simple exclusive-or function of the sign bits of $e(n)$ and $y(n - k)$. Typically, λ is chosen to be 1/64 the least significant bit of the coefficient by extending the coefficient registers six bits below the least significant bit. This corresponds to a value that is 1/128 dB. The added least significant six-bits accumulates an average that reduces the variance of the estimation. The resulting coefficient value can be considered to be a stochastic average that is related to the correlation between the cancellation error and the coefficient. Since the values of the coefficients are in log units, incrementing and decrementing them is equivalent to multiplying and dividing the coefficient values by a constant percent in the linear sense. Although it may not be obvious, this does not change the robustness and stability of the basic LMS method.

Simulation Study of Adaptive Algorithm

The behavior of the algorithm is illustrated in Figure 4 for a simulated open-loop condition (H = 0) For this illustration, the external feedback signal, $v(n)$, is derived from the model expression:

$$v(n) = 10 \, y(n-2) - 5 \, y(n-4)$$

where $y(n)$ is a pseudo-random sequence, $p(n)$, that also serves as the input to the adaptive filter. It can be seen that the coefficients follow a logarithmic path in adapting to the model values and that once the correct values are reached the algorithm randomly dithers around the least significant bits of the coefficients. The sign function simplification of the LMS algorithm results in a slow rate of adaptation. However, this is desirable in many applications. The simplified algorithm has other desirable characteristics. For example, since the sign function is either

zero (decrement the coefficient) or one (increment the coefficient), no dead zone occurs, as the error becomes exceedingly small, that will cause a coefficient tracking offset error. In addition, the coefficients can be updated in any order, singly or together, at any sampling rate up to the sampling rate of the system. Therefore, there are a number of possibilities for optimizing the implementation and for varying the rate of adaptation.

We were concerned initially that the algorithm would not converge properly if the system started in an oscillatory state. However, this is not a problem. Figure 4 also illustrates the behavior of the algorithm for an oscillatory closed-loop condition (H = 1). It can be seen that when the equalization filter is initialized to zero, the rms error quickly grows to a maximum as the system begins to oscillate. Although the time required for equalization is greater when the system is oscillatory, it can be seen that the coefficients eventually reach their desired values and the system becomes stabilized.

In either the open-loop or closed-loop case, once the final state of equalization is reached, the coefficient values dither randomly about the desired values with an error that is proportional to the coefficient value. This generates noise that is uniformly distributed across all frequencies and limits the degree of cancellation possible. The least significant bit of the coefficients is equivalent to 0.5 dB in our implementation, or 6 percent. Assuming that the coefficient error is uniformly distributed between ± 3 percent, the rms error is equal to about 0.9 percent. Therefore, the coefficient noise will be on the order of 40 dB below the signal level.

It has been mentioned above that a pseudo-random probe noise is inserted at the output of the hearing aid to serve as a common source of low-level sound for exciting the feedback path and the equalization filter. However, it has been found that other sources within the hearing aid generate an appropriate wide-band noise that serves the same purpose. Furthermore, these sources generate noise that is proportional to the signal level. Note that the Dyrlund and Bisgaard (3) implementation requires a separate circuit to adjust the level of the probe noise so that it is about 30 dB below the signal level. With log encoding, the quantizing noise is white and is 35 dB below the level of the encoded signal. Arithmetic roundoff noise and the dithering of the coefficients adds additional noise at levels about 40 dB below

Open Loop Model

Open Loop Model

Closed Loop Model
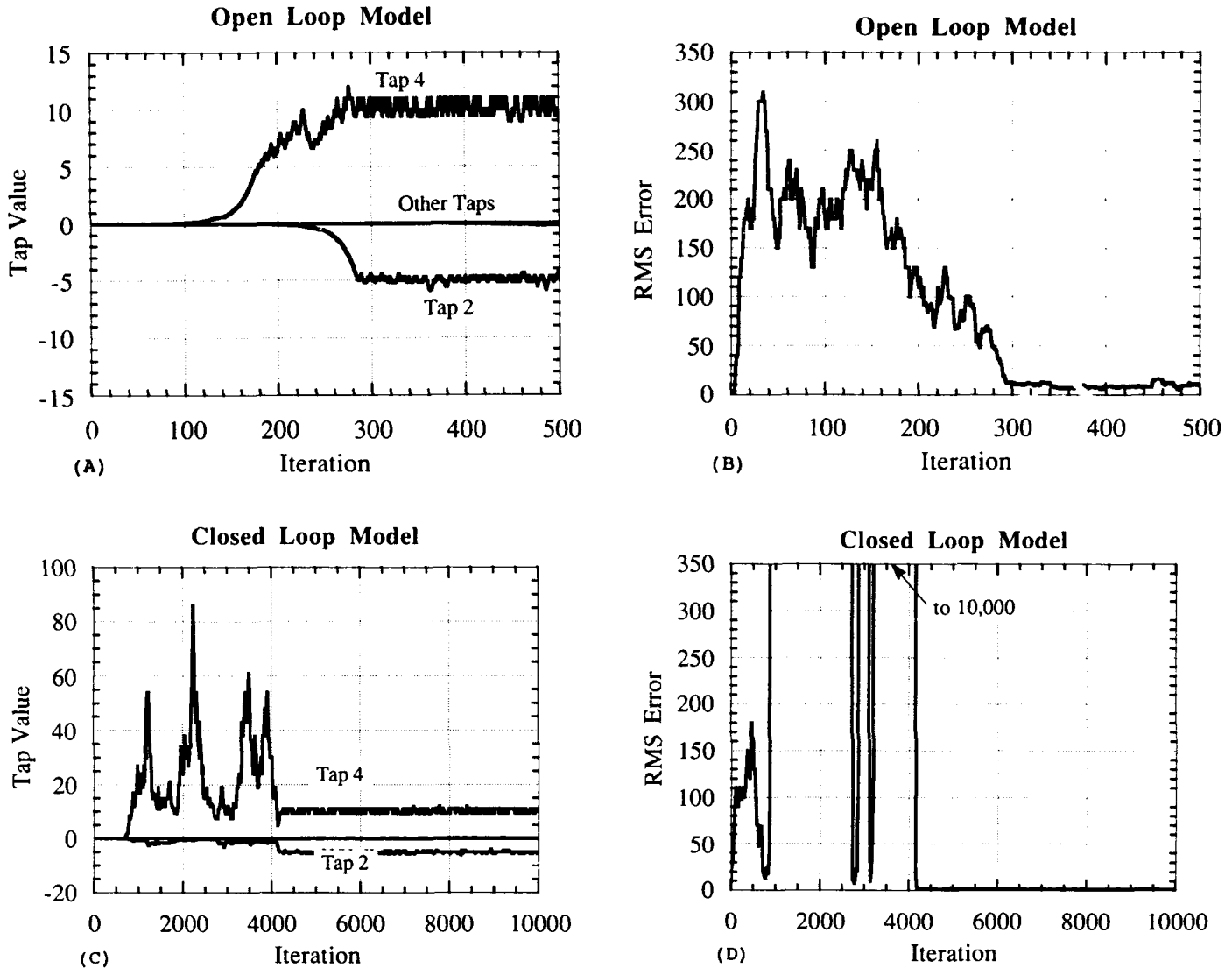
Closed Loop Model

Figure 4.
Simulation results of the log-binary-LMS adaptive algorithm for a simple model showing typical coefficient behavior for open-loop and closed-loop conditions. (a) Time course of coefficients for open-loop case; (b) time course of rms cancellation error for open-loop case; (c) time course of coefficients for closed-loop case; (d) time course of rms cancellation error for closed-loop case.

the signal level. The current adaptive algorithm takes advantage of these serendipitous sources of noise excitation.

There are several factors that limit the amount of additional stable gain that can be achieved with the feedback equalization algorithm. The first factor is the degree of cancellation that can be achieved. Because of the choice of log base and the number of bits used to represent the coefficient values, the coefficient estimation error is on the order of 0.9 percent. The rms error between the equalization filter and the feedback path is also 0.9 percent. Therefore, the maximum gain margin that can be expected will be about 40 dB due to this source of error alone. Another factor that limits gain margin is the presence in the system of narrow-band or periodic signals that have long autocorrelation functions. These types of signals will cause the adaptive algorithm to deviate from the estimate of the feedback path. This effect is reduced by delays through the digital filters of the hearing aid, which move the offending autocorrelation terms to later lag products. Also, because of the slow adaptation rates that are used (several seconds), only periodic external signals that persist will upset the equalized state of the system.
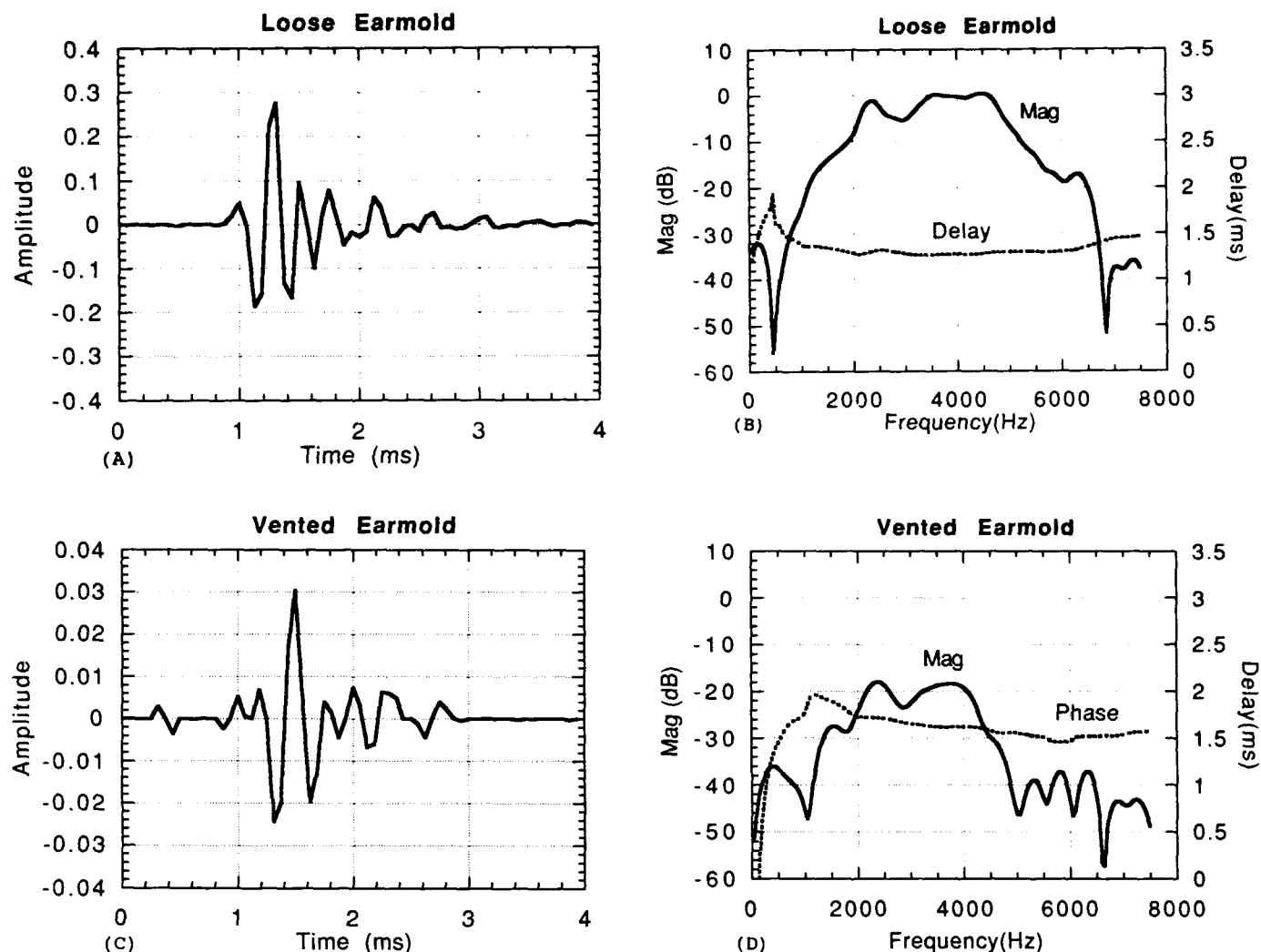
Figure 5.
Bench-test measurements of equalization filter after it has reached a steady state for two conditions of acoustic leakage. (*a*) Impulse response for loose earmold condition; (*b*) frequency response for loose earmold condition; (*c*) impulse response for tightly fitting earmold with vent condition; (*d*) frequency response for tightly fitting earmold with vent condition.

Bench Tests with KEMAR Mannequin

Bench testing of the adaptive algorithm on a KEMAR mannequin has been extensive. Typically, a Macintosh computer is used as a host system, and programming of the digital hearing aid is accomplished via a serial port. It is also possible to upload the equalization filter coefficients to the host computer via the serial port so that they can be observed. **Figure 5** illustrates the impulse response of the equalization filter for two conditions of acoustic leakage with a KEMAR mannequin test setup after a steady state of equalization has been reached. The impulse response **(Figure 5a)** represents an estimate of the external feedback characteristic of the hearing aid. The frequency response of the

equalization filter **(Figure 5b)** is obtained by taking the Fourier transform of the impulse response. The result in this figure can be compared with the direct measurement of the feedback characteristic of **Figure 2**. Results with a tighter fit and a vent are shown in **Figure 5c** and **Figure 5d**. Most of the delay before the start of the impulse responses in the figures (each tap corresponds to a 60 μs delay) is due to the delays through the receiver, ADC, and DAC.

The equalization filter has to be long enough to span the impulse response of the acoustic feedback path. The relationship between the length of the equalization filter and gain margin has also been studied and typical results are shown in **Figure 6**. Each curve represents the greatest gain that could be

Measured Acoustic Gain (output)
Feedback suppression active, 50 dB SPL pink noise input
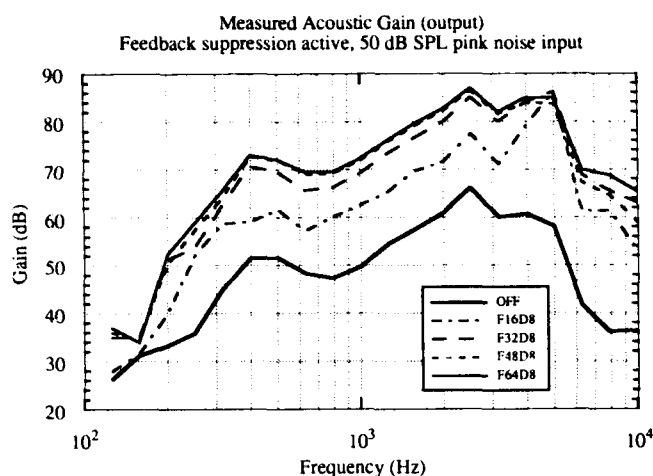
Figure 6.
Measured acoustic gain of equalized hearing aid as a function of filter length. The input to the adaptive filter was delayed by a fixed delay of eight samples. Maximum stable gain is achieved only for filters with greater than 48 taps. Including the delay, the total span of the equalizer path must be greater than 56 samples.

achieved without oscillation for an initial delay of 8 samples and an equalization filter with 16, 32, 48, and 64 taps. As can be seen, a total span including initial delay and filter of 56 samples is required to achieve a maximum stable gain. This corresponds to a delay of 3.36 ms, which can be compared with the impulse responses of **Figure 5**. It should be noted that the gain margin, which is the difference in achievable stable gain with and without feedback equalization, is about 20 dB. The adaptive behavior of the system at the limit of maximum achievable gain is shown in **Figure 7**. The curve represents the error between the external feedback path and the internal equalization filter. When starting from zero, the system requires about 1.5 seconds before approaching an equalized state and then an additional second while each of the coefficients reaches its final state. These results are not unlike the simulation studies described earlier in the paper.

DISCUSSION

The feedback equalization method described above appears to be a viable solution to the problem of hearing aid instability. The algorithm behaves robustly and is suitable for implementation in the

Adaptive Behavior at Limit of Gain Margin
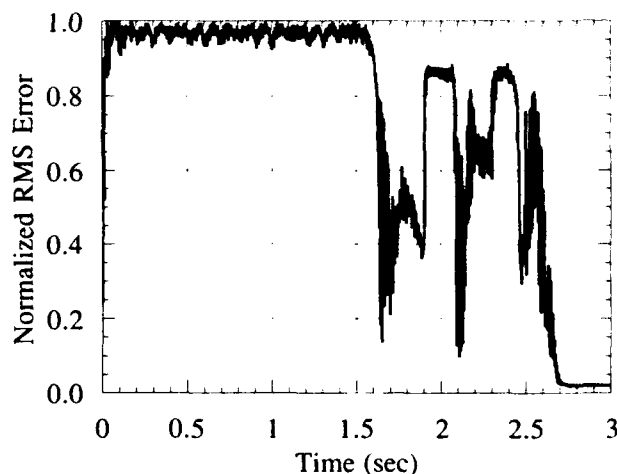
Figure 7.
Adaptive behavior of algorithm at the limit of maximum gain margin that could be achieved.

current generation of hearing aids. Additional usable gains of 10 to 15 dB can be achieved in practice, which corresponds to an additional population of hearing-impaired with 20-30 dB greater hearing loss that can be helped. In addition, open earmolds, which provide greater comfort, can be used more frequently with moderate hearing loss. We estimate that the algorithm can be implemented in the form of a small, low-voltage circuit that will require substantially less than 1 mW of power.

We recognize that this is one of the first attempts to apply principles of adaptive active cancellation to hearing aids. We hope that, as with other engineering endeavors, when more designers begin to apply their skills to the problem, improved algorithms will result that will extend performance and provide even greater benefit for the listener with hearing impairment.

ACKNOWLEDGMENTS

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Behavioral assessment of adaptive feedback equalization in a digital hearing aid

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Abstract—An evaluation was made of the efficacy of a digital feedback equalization algorithm employed by the Central Institute for the Deaf Wearable Adaptive Digital Hearing Aid. Three questions were addressed: 1) Does acoustic feedback limit gain adjustments made by hearing aid users? 2) Does feedback equalization permit users with hearing-impairment to select more gain without feedback? and, 3) If more gain is used when feedback equalization is active, does word identification performance improve? Nine subjects with hearing impairment participated in the study. Results suggest that listeners with hearing impairment are indeed limited by acoustic feedback when listening to soft speech (55 dB A) in quiet. The average listener used an additional 4 dB gain when feedback equalization was active. This additional gain resulted in an average 10 rationalized arcsine units (RAU) improvement in word identification score.

Key words: *acoustic feedback, digital feedback equalization algorithm, evaluation, wearable adaptive digital hearing aid.*

INTRODUCTION

The objective of this study was to evaluate the efficacy of a feedback equalization (FBE) algorithm implemented on the Central Institute for the Deaf's (CID) Wearable Adaptive Digital Hearing Aid (WADHA).

Acoustic feedback is a familiar problem to those working with people who use hearing aids.

Using a behind-the-ear (BTE) aid with a nonvented, well-fit earmold, Dyrlund and Lundh (1) reported that an average gain reduction of about 10 dB from prescribed gain was required to eliminate acoustical feedback. The limitations were greatest for frequencies above 1,000 Hz. Dyrlund (2) reported a limit on the hearing loss that could be managed without acoustical feedback for children with profound hearing impairment using BTE aids and nonvented acrylic earmolds. His guidelines for maximum hearing loss manageable before feedback show that for 1 kHz, maximum loss is 100 dB HL, and the upper intensity limit decreases as frequency increases. Grover and Martin (3) reported gains of about 50 dB before feedback oscillation using nonvented, acrylic earmolds and a BTE configuration. They compared the sound pressure level at the hearing aid microphone with that of a probe tube inserted into the ear canal. While many hearing losses can be accommodated with this amount of gain, the increased incidence of steeply sloping audiometric configurations gives rise to greater need for open or vented fittings, to provide adequate high frequency amplification without over-amplifying the low frequency region where hearing may be normal or near-normal.

The most common remedy for acoustical feedback is to simply reduce the gain of the hearing aid, which in turn reduces the intensity of the sound that may feedback to the microphone. This method is often employed by hearing instrument wearers. That is, many hearing aid wearers will adjust the volume control of their aids by turning the gain up until

audible feedback occurs, and then turn the gain down until feedback ceases. Although use of this method of gain adjustment is widespread, and even recommended by some dispensers, it is a clear indication that the gain that hearing instrument wearers use is limited by the acoustic feedback threshold. In addition, Cox (4) advised against using the threshold of feedback as the user gain level because of the effects on the frequency response of the hearing aid output. She noted that by setting the gain control to a position just below that which would cause audible feedback, there occurs suboscillatory feedback that results in the formation of erratic peaks in the frequency response of the hearing aid. Skinner (5) suggested that insertion gains should be 4–8 dB less than values at which audible feedback occurs to avoid the deleterious effects of suboscillatory feedback. Yanick (6) noted that hearing aid wearers who adjust hearing aid gain to the threshold of audible feedback may be receiving distorted speech (both spectrally and temporally) as a result of transient distortion. Preves (7), discussing this phenomenon, stated that when formant transitions are near the frequencies of the resonant response peaks of a hearing aid operated just below acoustical feedback oscillation, they may become severely distorted and detract from their perception by listeners with hearing impairment.

Consequently, an adaptive feedback equalization algorithm that suppresses acoustic feedback while maintaining an appropriate target gain function may provide significant benefit to many hearing aid wearers. Hearing aid users may then select gain levels that amplify signals to comfortable levels without feedback. Further, adaptive feedback equalization may smooth the erratic peaks that occur when listeners select volume control levels that result in suboscillatory feedback. The smoothing of response may also improve speech identification performance. In the present study, three questions were addressed: 1) Does acoustic feedback limit gain adjustments made by hearing aid users? 2) Does feedback equalization permit users with hearing impairment to select more gain without feedback? and, 3) If more gain is used when feedback equalization is active, does word identification performance improve?

METHOD

Subjects

Nine subjects with hearing impairment (5 male and 4 female) having a mean age of 63.4 years (range 39–76 years) participated in this study. Table 1 lists the pure-tone thresholds of the subjects. All nine subjects were experienced hearing aid wearers—five BTE and four in-the-ear (ITE). All subjects had sensorineural hearing losses, except Subject #1 who had a mixed hearing loss.

Wearable Adaptive Digital Hearing Aid

Figure 1 is a schematic of the digital hearing aid (DHA) used in this experiment. The DHA is a four-channel hearing aid. Each channel is specified by the first "Filter" in the Filter-Limit-Filter configuration. Gain shape and channel limit are specified in the "Limit" box. The final "Filter," shaped identically to the first filter, rejects any in-band harmonic distortion generated by the limiting process. Limiting in this evaluation was achieved using peak-clipping. The Noise Reduction feature was not active in this sequence of experiments. Details of the adaptive feedback equalization algorithm are described in a companion paper in this issue (8). In principle, the digital FBE algorithm suppresses feedback by adaptively canceling the feedback path. The process is illustrated schematically in Figure 2. The acoustical feedback pathway (H) represents sound energy that has leaked from a vent or around an earmold or hearing aid. The sound escapes from the ear canal (Y), and travels back to the hearing aid microphone (M). The input signal to the amplifier circuitry (G) is monitored at point (A). The output signal from (G) is monitored at point (B). A 1.8 ms delay is introduced at (G). Signal components at (B) that are correlated with those at (A) are considered to be derived from the feedback path (H). The coefficients of the FBE filter are subsequently adapted to reduce the correlation to zero. The FBE will adapt to the feedback signals whether the hearing aid starts out in a state of oscillation or begins to oscillate after the aid has been on for some time. The adaptive process initially may take 1–1.5 seconds to identify the feedback signals and adapt to them; however, any changes that occur in the

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Table 1.

Pure-tone air conduction thresholds (dB HL re: ANSI, 1969).

Subject #	Age Years	Sex	Ear	Frequency, Hz					
				250	500	1000	2000	4000	8000
1	57	F	R	80	75	75	80	80	95
			*L	60	70	65	70	70	90
2	66	M	R	15	10	55	55	60	70
			*L	15	15	60	60	55	55
3	69	M	R	20	15	50	80	90	85
			*L	15	15	45	60	70	100
4	68	F	R	60	65	75	65	65	105
			*L	40	60	80	65	65	80
5	76	M	R	10	10	20	40	65	90
			*L	10	20	25	50	65	90
6	39	F	R	40	60	80	80	75	105
			*L	35	50	75	75	75	95
7	65	M	R	30	45	55	55	55	65
			*L	35	40	60	55	55	55
8	71	F	R	35	50	55	75	95	NR
			*L	30	40	50	65	85	90
9	60	M	*R	45	55	70	70	70	100
			L	30	50	70	75	65	100

*ear tested
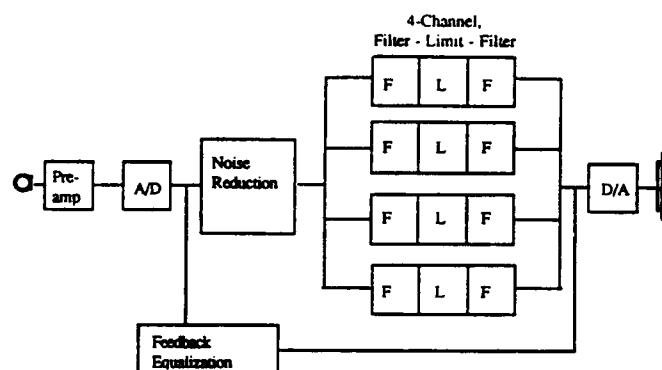
Figure 1.
Schematic illustration of the Digital Hearing Aid.

Figure 2.
Schematic illustration of feedback path and feedback equalization filter.

feedback pathway are adaptively canceled at a much faster rate (approximately 100 msec).

Test Room and Equipment

All listening tasks were carried out in a sound-treated room. Stimuli were presented from a loudspeaker located 1 m in front of the subject at 0°

azimuth. Signal levels were calibrated in dB A (slow) at the position corresponding to that of the center of the head of the listener if the listener were present. Real-ear probe-tube measurements were made at 45° azimuth using a second loudspeaker, also 1 m from

the subject. The real-ear measurement system consisted of a reference microphone clipped to the earlobe of the subject, and a soft cilastin probe-tube placed alongside the earmold in the external ear canal, within 5 mm of the tympanic membrane. Insertion depth of the probe tube was determined using the acoustic method as described by Gerling and Engman (9), which involves probe-tube insertion to a depth of 10 mm beyond the 6 kHz standing wave null. The subjects performed listening tasks aided monaurally, with the contralateral ear occluded with an EAR Noise Filter™ earplug (Cabot Corp., Indianapolis, IN).

Speech Stimuli

Subjects adjusted hearing aid gain control settings while listening to excerpts from the Connected Speech Test (CST) (10), which utilized a female talker. Word identification scores were determined using Pascoe High Frequency Word Lists (PHFWL), which utilized a male talker. Each PHFWL consists of 50 monosyllabic words containing a large proportion of high frequency consonants. Each word was presented with the carrier phrase, "Please write . . . .," with a 4-sec gap between presented words. Subjects wrote their responses on answer sheets, which were scored at a later time. In competing noise conditions, multitalker babble was presented at a 6 dB signal-to-babble ratio (SBR).

Determination of Gain Control Setting

Subjects adjusted the gain control of their own hearing aids (Part I) and a simulation of their own aids using the CID WADHA (Part II) in quiet and multitalker babble while listening to soft speech signals (55 dB A). Subjects were instructed to adjust the hearing aid gain control so that speech was perceived to be maximally intelligible.

In Part I, subjects were monaurally aided with one of their own hearing aids and earmolds. The subjects were instructed to adjust the gain control of their own hearing aid such that soft speech signal (CST paragraphs presented at 55 dB A) was perceived as most intelligible. Once the gain selection was made, the real-ear insertion response was obtained for input levels of 55, 70, and 85 dB sound pressure level (SPL). This procedure was performed for two listening conditions: speech-in-quiet and speech-in-babble. The order of experimental conditions was counter-balanced across subjects. The ' condition was replicated as a retest measure.

In Part II, the WADHA was configured to simulate the subject's own hearing aid by using target gain and saturation output values based on the real-ear measurements made in Part I. The real-ear insertion responses corresponding to the 55 dB A input composite noise signal were used to derive target gain values to be programmed into the WADHA. The real-ear insertion responses corresponding to the 85 dB A input composite noise signal supplied the values that were used to program the maximum power output of the WADHA configurations. The four WADHA memories were configured as follows:

Memory A: Simulation of subject's own aid in *quiet*, FBE **Off**.

Memory B: Simulation of subject's own aid in *quiet*, FBE **On**.

Memory C: Simulation of subject's own aid in *babble*, FBE **Off**.

Memory D: Simulation of subject's own aid in *noise*, FBE **On**.

The WADHA gain control was active during all listening tasks and measurements. The gain control has 16 steps of 2 dB. The target gain settings are achieved when the volume control is set at position "7." Once the WADHA was configured as above, the subjects made gain control adjustments for each of the four memories as before, and real-ear probe-tube measurements were made at the selected gain control settings using 55, 70, and 85 dB SPL composite noise input levels. The selected gain control settings were recorded for each memory. It should be noted that BTE hearing aid wearers retained their own earmolds for the WADHA evaluations while ITE hearing aid wearers were provided with custom silicone-shell earmolds. Fit measures obtained at 50 Hz intervals resulted in WADHA fit accuracy of 4.9 dB root mean square (rms) between 250 Hz and 6,000 Hz and 4.3 dB between 500 Hz and 2,500 Hz.

Word Identification Assessment

Percentage correct word identification was assessed using the PHFWL at a signal presentation level of 55 dB A. The listening conditions were varied randomly such that some of the trials required listening in quiet and others in a background of multitalker babble noise (+6 dB SBR).

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The hearing aid (either the subject's own or WARHA simulation) was adjusted to the volume control setting that was selected previously, for each listening condition. The subjects wrote their responses on answer sheets, which were scored at a later time.

RESULTS

Recall that the first objective was to determine whether acoustic feedback limited gain adjustments made by the subjects when listening to soft speech in quiet and soft speech in babble noise at 6 dB SBR. In Part I of this study, all subjects responded similarly to the initial instructions to adjust their own hearing aid such that the speech stimulus is perceived as maximally intelligible. All subjects turned up their hearing aid until audible feedback was produced and then turned down the gain somewhat. When subjects were fitted with the WADHA simulation of their own aid with FBE Off, similar adjustments of the gain control were observed. Thus, it appeared that acoustic feedback was limiting the usable gain range of all subjects in this study.

The second objective of this study was to determine whether FBE permits users to access additional gain when listening to soft speech in quiet and babble noise. Two comparisons addressed this issue. First, comparisons were made between FBE On and FBE Off WADHA conditions. This comparison provided data regarding the efficacy of FBE within the prototype, fully digital amplification condition. Second, comparisons were made between the WADHA FBE On and Own-Aid conditions. These latter comparisons provided guidelines regarding the magnitude of gain change listeners may need within existing conventional amplification.

Gain Changes (FBE On-FBE Off) Implemented Using WADHA. **Figure 3** and **Figure 4** illustrate the amount of additional gain subjects used in the FBE On condition compared with the FBE Off condition when listening to soft speech in quiet and soft speech in babble, respectively. The box plots illustrate the median (waist of box plot), interquartile range (ends of box), 10th and 90th percentile range (whiskers), and outlier points. In quiet, the median difference in gain between the FBE On and FBE Off conditions ranged from 0 dB at 6,000 Hz to 5 dB at
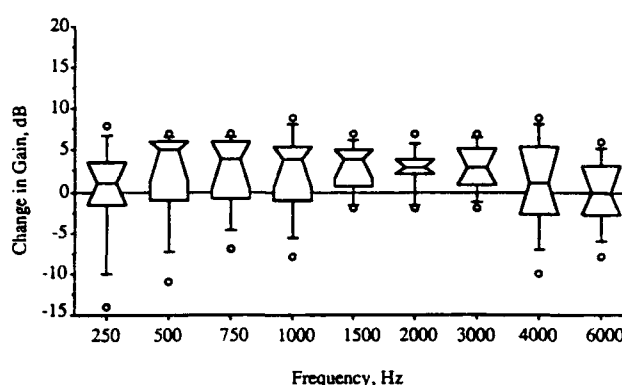
Figure 3.
Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and FBE Off conditions, selected by subjects listening to soft speech in quiet as a function of frequency. The median gain difference (FBE On-FBE Off) is represented by the waist of the box plot. The box extends beyond the median to the interquartile values. The straight-line "whiskers" extend to the 10th and 90th percentile values. Outliers ($\leq$ 10th or $\geq$ 90th percentile) are represented by circles.
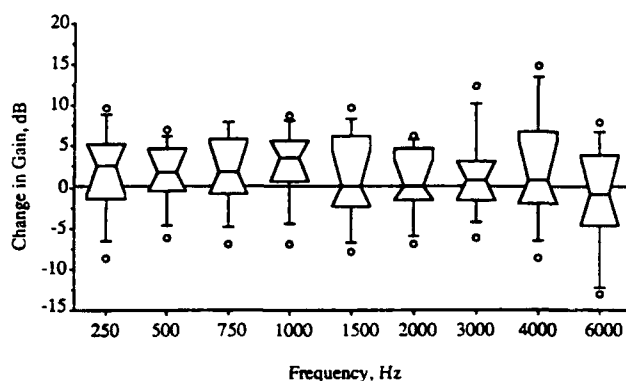
Figure 4.
Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and FBE Off conditions, selected by subjects listening to soft speech in babble as a function of frequency.

500 Hz. Over the major speech frequencies, the median gain adjustment was approximately 4 dB. In babble noise, the median difference in gain between the FBE On and FBE Off conditions ranged from -1 dB at 6,000 Hz to 4 dB at 1,000 Hz. Over the major speech frequencies, the median gain adjustment was varied between 0 dB and 2 dB. An example of the effects of FBE on the electroacoustic output of WADHA is illustrated in **Figure 5**. The WADHA real-ear frequency response in the FBE On
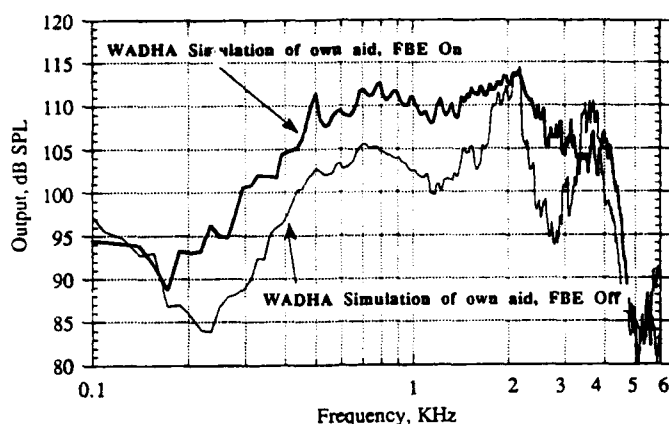
Figure 5.

Frequency response curves for WADHA simulations of own aid in FBE Off and FBE On conditions (Subject #1).

Figure 6.

Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and own aid conditions, selected by subjects listening to soft speech in quiet as a function of frequency.

and FBE Off conditions are shown for Subject 1. When listening to connected speech in quiet in the FBE Off condition, Subject 1 selected a WADHA volume control setting that produced audible feedback. The frequency response demonstrates large resonant peaks centered around 2,000 Hz and 3,500 Hz, and marked minima between 1,000 Hz and 2,000 Hz and close to 3,000 Hz. When FBE is activated, the resonant peaks are smoothed, and up to 7 dB of additional gain is available without audible acoustic feedback.

Gain Changes (FBE On-Own Aid) Implemented Using the WADHA Compared with Own Aid. **Figure 6** and **Figure 7** illustrate the amount of additional gain subjects used in the FBE On condition compared with their own aid when listening to soft speech in quiet and soft speech in babble, respectively. In quiet, the median difference in gain between the FBE On and FBE Off conditions ranged from 5 dB at 250 Hz, 500 Hz, and 3,000 Hz to 9 dB at 750 Hz. Over the major speech frequencies, the median gain adjustment was approximately 6 dB. In babble noise, the median difference in gain between the FBE On and FBE Off conditions ranged from 2 dB at 1,000 Hz and 3,000 Hz to 5 dB at 250 Hz and 6,000 Hz. Over the major speech frequencies, the median gain adjustment was varied between 0 dB and 3 dB.

The third objective of this study was to determine whether the additional gain used in the FBE On condition results in improved word identification performance when listening to soft speech in quiet

Figure 7.

Box plot distribution of averaged difference in insertion gain between the WADHA FBE On and own aid conditions, selected by subjects listening to soft speech in babble as a function of frequency.

and babble noise conditions. **Table 2** provides the word identification performance of the subjects when listening with their own hearing aid or a digital simulation of their own aid with the WADHA in quiet and +6 dB SBR. The scores reported have been transformed from percentage correct to rationalized arcsine units (RAU) to normalize the variability in the data (11). Examination of the group means reveals that subjects performed better in quiet than in babble (using either their own aid or a WADHA simulation of their own aid). Subjects also performed better in the FBE On condition compared with the FBE Off condition in quiet and in babble. **Figure 8** illustrates the linear regression of the word

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Table 2.
Word Identification Performance (RAU).

| Subject # | Subject's Own Aid | | Digital Simulation of Own Aid | | | |
| | Quiet | Noise* | Quiet | | Noise* | |
			FBE On	FBE Off	FBE On	FBE Off
1	72.80	89.10	91.90	77.00	89.10	66.80
2	62.90	53.60	81.50	68.70	59.20	57.30
3	53.60	51.80	68.70	64.80	53.60	48.20
4	77 00	66.80	83.90	79.20	81.50	68.70
5	89.10	77.00	107.20	98.40	77.00	70.70
6	53.60	50.00	59.20	57.30	44.50	46.40
7	68.70	59.20	63.90	74.90	70.70	77.00
8	53.60	42.70	91.90	72.80	51.80	55.50
9	53.60	55.50	64.80	48.20	57.30	55.50
Average	65.00	60.60	81.40	71.30	65.00	60.60

*Noise (babble) = +6 dB SBR

identification scores of the subjects using the WADHA simulation of subjects' own aid frequency response on the word identification scores obtained by subjects using their own hearing aid in the quiet listening condition. A small (mean difference = 6.27 RAU, t = 2.785, p = 0.019) but significant improvement in word identification score is observed when subjects' own aid scores are compared with the WADHA simulation FBE Off condition. This result is not entirely unexpected, as the majority of subjects were able to use additional gain without feedback under the WADHA simulation (FBE Off) compared with that available with their own aid. This was particularly noticeable in four subjects who used ITE hearing aids. An additional significant improvement in word identification performance is observed when the FBE On and FBE Off conditions are compared. A directional t-test for correlated samples was performed on the data, and revealed the group performance increase of 10.2 RAU observed in the FBE On condition to be significant (p = 0.0005; t = 5.066).

Figure 9 illustrates the linear regression of the word identification scores of the subjects using the WADHA simulation of subjects' own aid frequency response on the word identification scores obtained by subjects using their own hearing aids in the babble listening condition. No significant differences were observed in group mean performance.

DISCUSSION

The results from this study suggest that acoustic feedback restricts the volume control adjustments that hearing aid wearers can make to compensate for loss of audibility in soft speech environments. The current feedback equalization algorithm has been shown to produce gain margins of 15–20 dB in the laboratory (8). However, it should be remembered that these values are obtained under ideal testing situations using a mannequin that does not move. With real subjects who breathe, talk and chew, practical gain margins are likely to be less than 15–20 dB. Examination of Figure 6 and Figure 7 suggests that gain margins of up to 10 dB might be required to accommodate the additional gain needs of 75 percent of the subjects used in this study. It should be noted that none of the subjects in this study used a vented earmold. Thus, the needed gain margins in vented mold situations may start to encroach on the gain margin limits of the adaptive algorithm. Gatehouse (12), for example, has shown that gain must be reduced by 15–20 dB when a 2 mm parallel vent is introduced into an earmold connected to a BTE hearing aid with a forward-facing microphone. Gatehouse also observed that the listening environment influences the amount of gain that is available without audible feedback. For example, a BTE hearing aid with a forward-facing
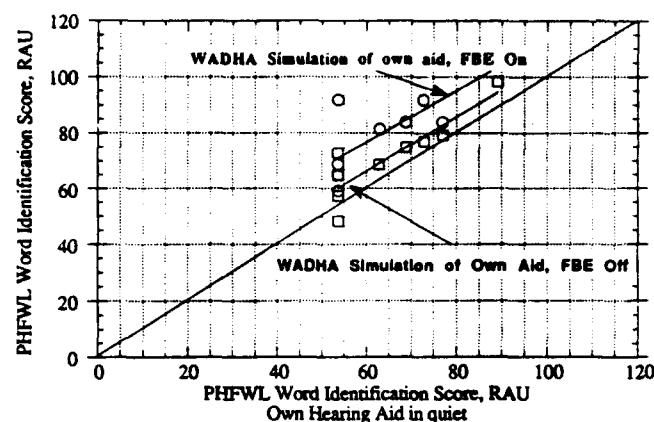
Figure 8.
Linear regression of PHFWL word identification scores (RAU) for WADHA simulations of own aid in quiet for FBE On and FBE Off conditions on PHFWL scores (RAU) for own aid in quiet.

$SCORE_{WADHA\ FBE\ On} = 22.027 + 0.91\ SCORE_{Own\ Aid},\ R^2 = 0.605$
$SCORE_{WADHA\ FBE\ Off} = 7.863 + 0.975\ SCORE_{Own\ Aid},\ R^2 = 0.776$
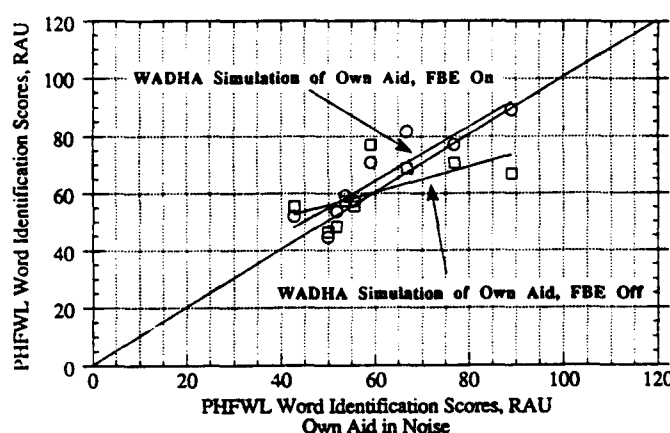
Figure 9.
Linear regression of PHFWL word identification scores (RAU) for WADHA simulations of own aid in babble for FBE On and FBE Off conditions on PHFWL scores (RAU) for own aid in babble.

$SCORE_{WADHA\ FBE\ On} = 7.71 + 0.94\ SCORE_{Own\ Aid},\ R^2 = 0.83$
$SCORE_{WADHA\ FBE\ Off} = 33.34 + 0.451\ SCORE_{Own\ Aid},\ R^2 = 0.393$

microphone could support 56.9 dB insertion gain without acoustic feedback in the presence of speech-shaped noise, 53.1 dB gain in the presence of a 250 Hz narrow-band noise, and 64.3 dB in the presence of a 2,000 Hz narrow-band noise. In this investigation, it was noted that subjects turned their own hearing aids up somewhat in the speech-in-babble condition. **Figure 10** illustrates the median gain
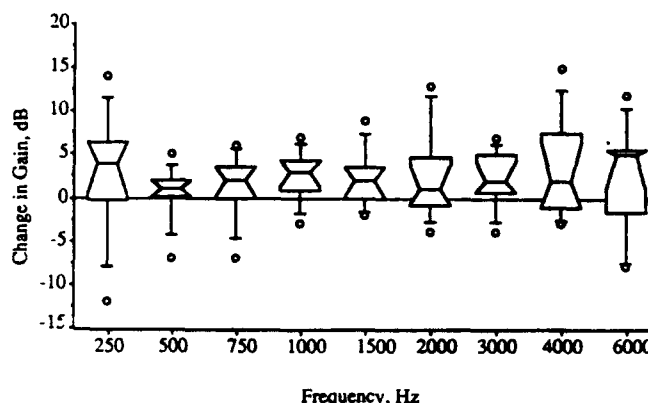
Figure 10.
Box plot distribution of averaged difference between insertion gain selected by subjects listening to soft speech in babble compared with soft speech in quiet as a function of frequency.

adjustments subjects made to their own hearing aids when listening to soft speech in quiet compared with listening to soft speech in babble. The gain adjustment resulted in a median increase in gain in babble of approximately 2 dB over the major speech frequencies. It is possible that the presence of the multitalker babble served to stabilize an otherwise unstable acoustic environment, thus permitting some additional gain in the speech in babble condition. Further work in this area is required.

The additional gain that subjects were able to achieve in the FBE On condition translated into significant improvements in word identification performance in quiet. The approximate 16 RAU improvement in mean score in the FBE On condition compared with the subjects' own aid condition suggests that if FBE were available in commercial aids, hearing aid wearers would derive perceptible improvement in word identification performance. The lack of significant improvement in word identification performance in babble is not surprising. The limiting factor in the babble condition was the SBR, which would have been largely unaffected by the linear gain changes demonstrated in this study.

In conclusion, subjects in this study were able to take advantage of additional gain without acoustic feedback provided by the FBE algorithm. The additional gain resulted in improved word identification performance in quiet. The gain adjustments required by these subjects appear to be within the capabilities of the technology.

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Intelligibility of frequency-lowered speech produced by a channel vocoder

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Abstract—Frequency lowering is a form of signal processing designed to match speech to the residual auditory capacity of a listener with a high frequency hearing loss. A vocoder-based frequency-lowering system similar to one studied by Lippmann was evaluated in the present study. In this system, speech levels in high frequency bands modulated one-third-octave bands of noise at low frequencies, which were then added to unprocessed speech. Results obtained with this system indicated, in agreement with Lippmann, that processing improved the recognition of stop, fricative, and affricate consonants when the listening bandwidth was restricted to 800 Hz. However, results also showed that processing degraded the perception of nasals and semivowels, consonants not included in Lippmann's study. Based on these results, the frequency-lowering system was modified so as to suppress the processing whenever low frequency components dominated the input signal. High and low frequency energies of an input signal were measured continuously in the modified system, and the decision to process or to leave the signal unaltered was based on their relative levels. Results indicated that the modified system maintained the processing advantage for stops, fricatives, and affricates without degrading the perception of nasals and semivowels. The results of the present study also indicated that training is an important consideration when evaluating frequency-lowering systems.

Key words: *frequency-lowering systems, speech intelligibility, vocoder evaluation.*

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INTRODUCTION

One general class of approaches toward lowering of the speech spectrum for individuals with high frequency hearing impairment is based on detecting high frequency information and recoding it through the use of low frequency signals which take advantage of the residual hearing of the listener. Two methods that have been employed to achieve such signal processing include frequency transposition and channel vocoding. In schemes employing frequency transposition (1,2), information in a specified high frequency band is shifted downward using amplitude modulation or nonlinear distortion. In schemes employing channel vocoding (3,4,5,6), the high frequency speech content is analyzed by a bank of filters whose output envelopes are used to control the amplitude of signals from low frequency synthesis filters. Braida, et al. (7) reviewed studies of transposition or vocoding prior to 1979 and concluded that, in general, the benefits to speech reception with these lowering schemes were generally small and restricted to a narrow class of speech sounds. Among the reasons offered by Braida, et al. for this lack of success were inappropriate selection of both the frequency range used to analyze high frequency information and the form of the recoded signals, as well as the choice of the level of the recoded signals relative to the normal low frequency speech components. For example, in many of the vocoder systems evaluated in the past, the analysis filter outputs controlled the levels of low frequency sinewaves, which were the sole signal presented to

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the listener (3,5,6). Not only do such systems fail to distinguish between voiced and unvoiced sounds, they also eliminate suprasegmental speech cues available in the low frequency speech range. Several recent studies employing either transposition or vocoding, however, have reported improved identification of fricative and affricate sounds over that obtained through low-pass filtering (8,9,10,11).

Velmans (9) described a transposer-based system in which high frequency information in the range of 4,000–8,000 Hz was shifted downward into the range of 0–4,000 Hz using a balanced modulator, and then combined with the linearly amplified signal. In evaluations of consonant reception by listeners with normal-hearing, conducted with or without frequency transposition, the speech signal was low-pass filtered to 900 Hz and combined with high-pass white noise. Consonant identification was superior for the transposer over low-pass filtering, both alone and in conjunction with speechreading (by roughly 13 percentage points without speechreading and 8 percentage points when speechreading was also available). Velmans and Marcuson (10) presented data which indicated that the device proved beneficial to impaired listeners whose residual hearing extended to 1,000 Hz (or beyond) but who had little or no residual hearing above 4,000 Hz. These listeners showed improvements in transposed consonant identification ranging from 11 to 30 percent, although identification of only those consonants with strong high frequency speech cues (/s, ʃ ,z, ʒ , tʃ ,dʒ,t/) was investigated. Further results obtained with a group of 25 students with hearing impairment in schools for the deaf (11) indicated that benefits for the transposition device were larger in children with severe to profound high frequency losses than in those with mild to moderate losses in this range. Again, evaluations were restricted to consonants with strong high frequency content.

In a vocoding system developed by Lippmann (8), speech sounds in the 1,000–8,000 Hz range were analyzed with a bank of band-pass filters whose outputs controlled the levels of low frequency bands of noise in the 400–800 Hz range. The noise signals were added to the original speech signal, which was low-pass filtered to 800 Hz. Preliminary discrimination tests conducted with a variety of choices for the analysis and synthesis filters led to the selection of a system with four analysis and four synthesis bands

for further study. Specifically, this system consisted of four analysis bands (two of which were two-third-octave wide and two of which were one-octave wide) and four synthesis bands composed of four one-third-octave bands of noise whose center frequencies ranged from 400 to 800 Hz. Consonant identification tests using CVC syllables composed from 16 consonants (C) and 6 vowels (V) were conducted on 7 subjects with normal hearing for both frequency-lowered speech and linearly amplified speech with an 800-Hz bandwidth. Overall percent-correct identification increased by 10 percentage points from 36 percent correct with linear amplification to 46 percent with frequency lowering. For individual subjects, increases ranged from 4 to 14 percentage points.

Several factors may underlie the positive results observed by Velmans and Lippmann relative to those reported in other studies of transposition and vocoding for frequency lowering (7). First, in both studies the recoded signals were presented at levels that did not significantly interfere with normal low, frequency speech sounds. Second, the frequency range selected to analyze high frequency information and the specific form of the recoded low frequency signals may have contributed to improved performance.

The current study was concerned with extending the investigation of the effects of vocoder-based frequency-lowered systems on the reception of consonants. In Experiment 1, the performance of listeners with normal hearing was evaluated using a system modeled after that described by Lippmann (8). In Experiment 2, a modified system designed to improve upon results obtained in Experiment 1, was evaluated. The results of the current study are compared with those of other relevant studies, and theoretical predictions are presented for performance through the frequency-lowered system in combination with speechreading.

EXPERIMENT 1: INITIAL EVALUATION OF A VOCODER-BASED FREQUENCY-LOWERING SYSTEM

Method

System Description. A block diagram of the system used in Experiment 1, modeled after that of Lippmann (8), is shown in **Figure 1**. This system
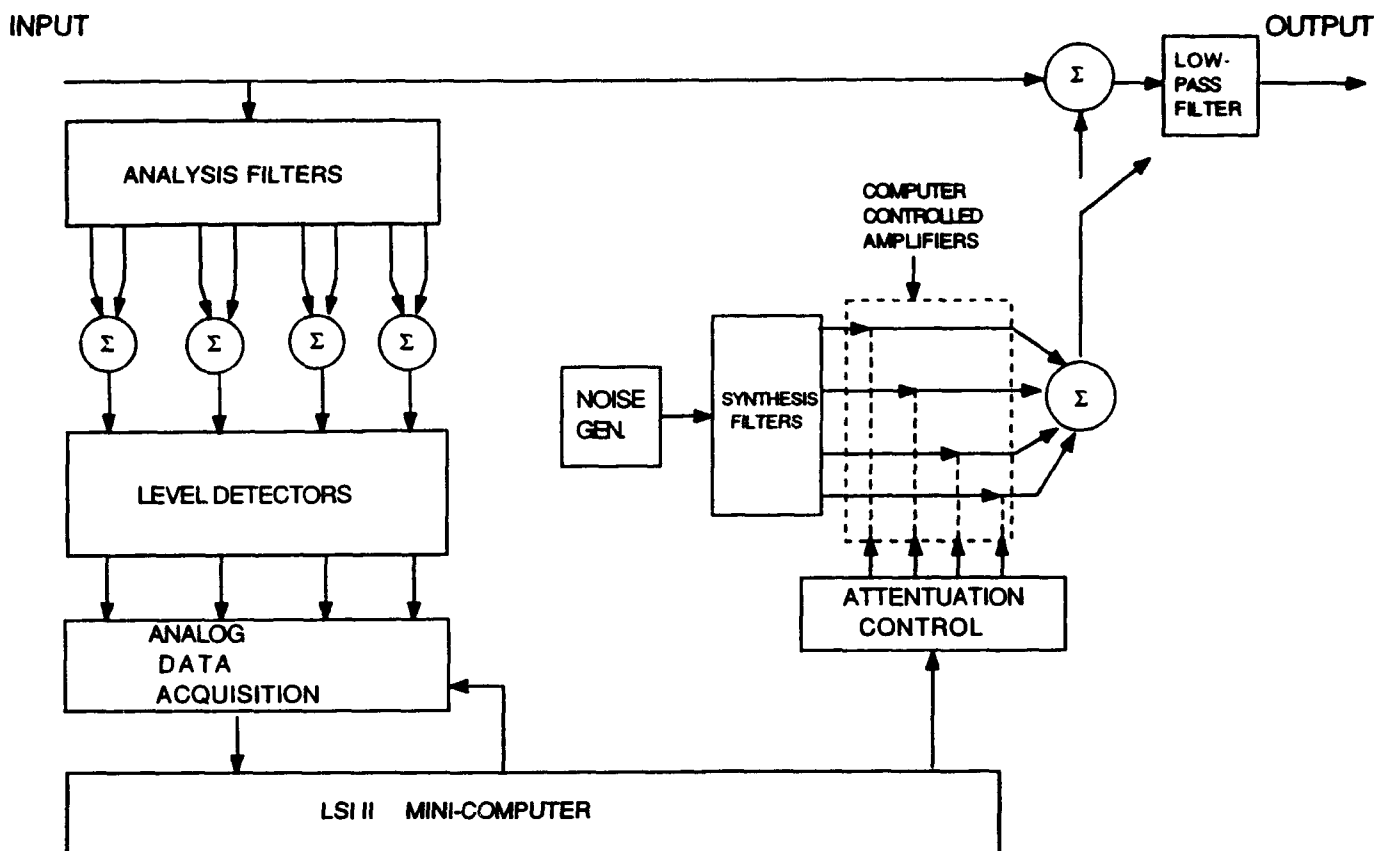
Figure 1.
Block diagram of the vocoder-based frequency-lowering system employed in Experiment 1.

made use of a computer-controlled multiband speech processor (12). High frequency speech information was analyzed by first passing speech through a bank of eight contiguous one-third-octave filters (General Radio 1925 Multifilter) with standard center frequencies in the range of 1.0 to 5.0 kHz (1.0, 1.25, 1.6, 2.0, 2.5, 3.15, 4.0, 5.0). The outputs of adjacent filters were combined using analog summing amplifiers to form four analysis bands with rejection rates of 48 dB/octave. The output levels of these bands were measured using rms level detectors that had logarithmic conversion, averaging times of 20 ms, and dynamic ranges of 65–70 dB. These levels were made available to an LSI-11 minicomputer by a multiplexed analog-to-digital converter, which had a 10 μs conversion time and a 12-bit capability. The sampling algorithm was based on that developed for controlling an amplitude compressor (12). The sampling periods ranged from 0.58 msec for the two lowest-frequency analysis

bands to 0.14 msec for the highest-frequency band. The sample values were used to determine the output levels of low frequency narrow-band noise signals, which were summed and added to the original speech signal. These noise-band signals were generated by passing wide-band noise through four contiguous one-third-octave filters (GR 1925) whose center frequencies ranged from 400 to 800 Hz. The high frequency analysis bands and the low frequency synthesis filters were monotonically related in that the lowest analysis channel controlled the lowest synthesis band, the second-lowest analysis channel controlled the second-lowest synthesis band, and so on. The levels of the noise bands were controlled by computer-controlled amplifiers that could vary gain with 0.5 dB resolution independently for each noise band. The output level of a noise band was linearly related to the output level of its analysis band in that a 1 dB increase in the signal level in an analysis band caused a 1 dB increase in the level of the correspond-

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ing low frequency noise-band signal. In addition, the levels of the noise bands were adjusted so that normal low frequency speech sounds were not significantly masked. Based on Lippmann's design as well as on subjective impressions from informal listening, the system was adjusted such that the 10 percent cumulative level of each noise band was 12 dB below the 10 percent cumulative level of speech measured using the same one-third-octave filter used to generate the noise signal.

The original speech signal combined with the noise signal was passed through two cascaded TTE miniature low-pass filters (series number J87E) to simulate a sharply sloping high frequency loss. The overall filter characteristics included 300 dB/octave rolloff, cutoff frequency of 800 Hz, and at least 80 dB attenuation in the stop band. These characteristics were chosen to provide an idealization of a hearing loss in which no high frequency speech cues are available. A toggle switch was included in the system to allow transmission of either unprocessed low-pass filtered speech or low-pass filtered speech with added low frequency noise-band signals.

Materials. The speech materials used for testing and training consisted of consonant-vowel (CV) nonsense syllables. The syllables were composed of 24 English consonants (stops /p t k b d g/; fricatives /f θ s ʃ v ð z ʒ /; affricates / tʃ dʒ /; semivowels /hw w l r j/; nasals /m n/; whisper /h/) with 3 vowels (/a i u/), resulting in a total of 72 syllables. Each syllable was spoken 3 times by each of 4 speakers, 2 male and 2 female, resulting in 864 tokens. Recordings were made in an anechoic chamber with the microphone situated 6 inches in front of the speaker's mouth. The recorded syllables were passed through a 4.5 kHz anti-aliasing filter with 140 dB/octave rolloff and converted to 12-bit digital samples at a sampling rate of 10 kHz. The digitized waveforms were normalized to equal rms levels and stored on a large disk memory. The waveforms were thus accessible for automatic computer presentation.

The stimuli were divided into two groups: a test set and a training set. Four tokens of each CV syllable were included in the test set used before and after training, one token spoken by each speaker. The materials used for training consisted of the remaining 576 tokens (2 tokens per speaker).

Subjects. The subjects were two normal-hearing students in their early 20s, one male (MP) and one

female (JR). The subjects had no more than 10 dB HL in the audiometric frequencies and were both native speakers of English with standard dialect.

Procedure. The subjects were trained and tested on the identification of frequency-lowered speech and low-pass filtered speech. On each trial a stimulus was selected at random from the available set and played over a high-quality audio-output channel. The signal was then passed through the processing system before being presented monaurally through TDH-39 headphones at a level of roughly 88 dB SPL.

For each pre- and posttraining test, each token from the test-token stimulus set was presented a total of 4 times, resulting in a total of 1,152 trials. Correct-answer feedback was not provided for either test. Training was achieved using the training-token stimulus set and stimulus-response procedure described above with correct-answer feedback. During a training session, each token from the stimulus set was presented twice, resulting in a total of 1,152 trials. When a subject responded incorrectly, the correct response was displayed visually and repeated aurally three times. Training sessions were continued until the learning curves of the subjects appeared to level off. The criterion for asymptotic performance was the observance of at least three consecutive training scores that were within three percentage points of each other. Each training session lasted approximately 2 hours. Subject MP began testing and training on frequency-lowered speech, while subject JR began the task with low-pass filtered speech.

Data Analysis. Confusion matrices were constructed from the pretraining and posttraining experimental runs for each subject under each experimental condition. The matrices were analyzed through calculations of percent-correct performance on various subsets of stimuli as well as through calculations of the percentage of unconditional information transfer (13) for the features described in **Table 1**, which were derived from definitions provided by Miller and Nicely (13) and Chomsky and Halle (14).

Results

Learning curves for each subject under low-pass filtering (circles) and the original vocoder-based lowering system (squares) are available in **Figure 2**. (Results from a third system, described in Experi-

Table 1.

Classification of the 24 consonants on a list of eight features derived from Miller and Nicely (13) and Chomsky and Halle (14).

	p	t	k	b	d	g	f	θ	s	ʃ	v	ð	z	ʒ	tʃ	dʒ	h	m	n	hw	w	r	j	l
Voicing	–	–	–	+	+	+	–	–	–	–	+	+	+	+	–	+	–	+	+	–	+	+	+	+
Nasal	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	+	+	–	–	–	–	–
Fricative	–	–	–	–	–	–	+	+	+	+	+	+	+	+	+	+	+	–	–	+	–	–	–	–
Affricate	–	–	–	–	–	–	–	–	–	–	–	–	–	–	+	+	–	–	–	–	–	–	–	–
Round	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	+	+	–	–	–
Vocalic	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	+	–	+
Duration	–	–	–	–	–	–	–	–	+	+	–	–	+	+	–	–	–	–	–	–	–	–	–	–
Place	1	2	3	1	2	3	1	2	2	3	1	2	2	3	3	3	3	1	2	3	1	2	3	2

Cumulative Number of Trials (in Thousands)
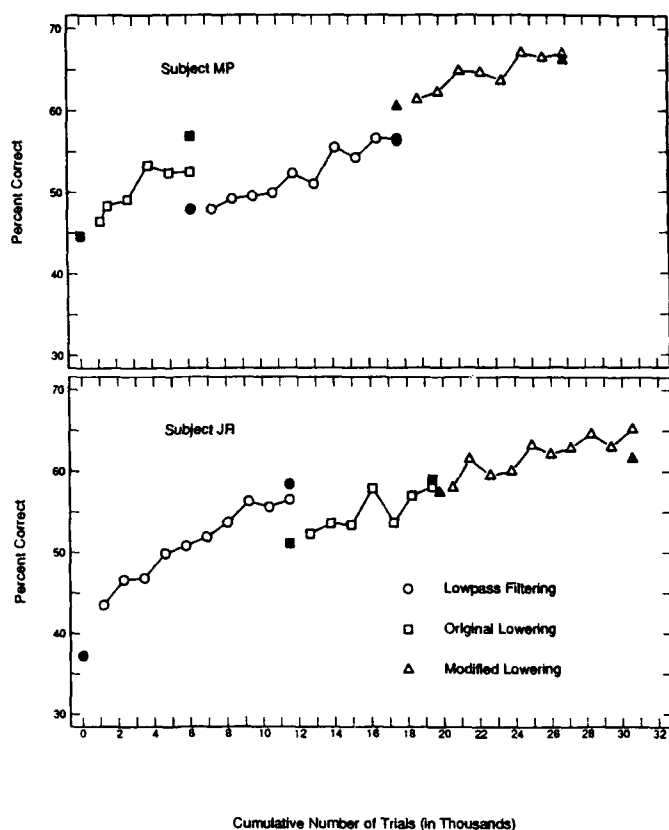
Figure 2.
Learning curves for each of three experimental conditions (low-pass filtering, original frequency-lowering system, and modified lowering system) for subject MP (top panel) and subject JR (bottom panel). Percent-correct score is plotted as a function of cumulative number of trials with feedback; filled symbols represent scores from test sessions and unfilled symbols, training sessions.

ment 2 below, are also included in the figure and will be discussed later in this article.) Filled symbols indicate results of pre- and posttraining test sessions (conducted without correct-answer feedback), while unfilled symbols indicate results from training sessions (in which trial-by-trial correct-answer feedback was provided). Posttraining test results of consonant identification using low-pass filtering and frequency lowering for subject MP showed gains of 8 and 12 percentage points, respectively, when compared with the pretraining test results. Similarly, subject JR showed improvements of 21 and 7 percentage points from training on low-pass filtered speech and frequency-lowered speech, respectively. Each subject achieved greater improvements with training for the system on which testing and training was begun (frequency lowering for subject MP and low-pass filtering for subject JR). Asymptotic performance on a given condition, however, appears not to depend on initial baseline performance: for example, both subjects leveled off at performance of roughly 56 percent correct for the lowering system, even though the pretraining score for subject MP was 47 percent compared with 37 percent for JR.

Posttraining scores obtained from each subject for these two conditions are shown in **Figure 3** for various groupings of consonants. The overall intelligibility of the full set of 24 consonants was not improved by the frequency-lowering system. Performance ranged from 56–58 percent correct across subjects and systems for the set of 24 consonants. When including only those consonants used by

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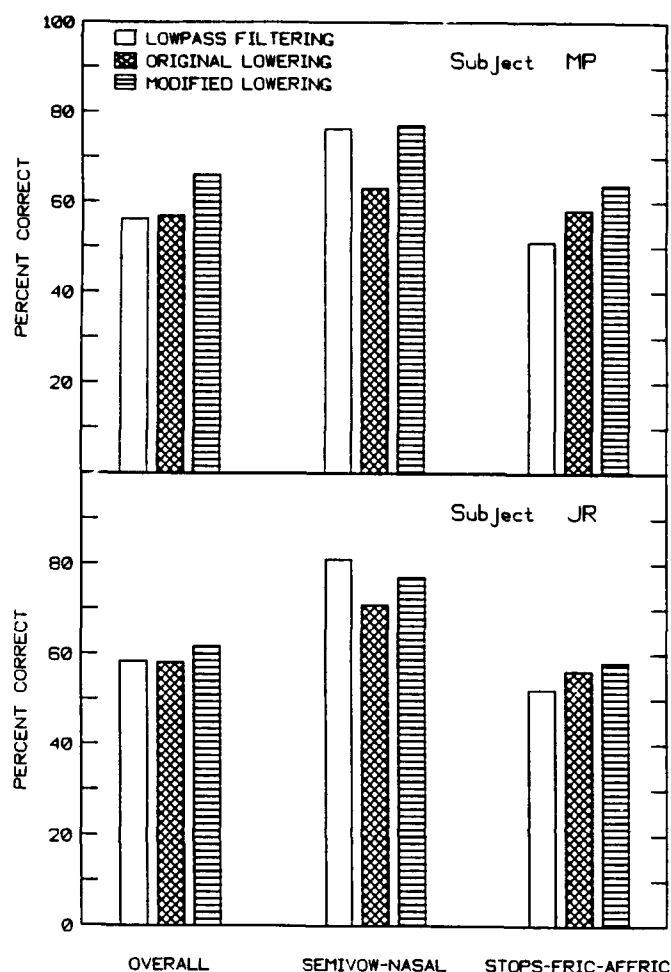
Figure 3.
Percent-correct score for each of three experimental conditions for subject MP (top panel) and subject JR (bottom panel). Scores are shown for the overall set of 24 consonants and for two subsets.

Lippmann (stops, fricatives, affricates), the percentage of consonants correctly identified by each subject was 4–7 percentage points higher with frequency lowering. This result is in close agreement with Lippmann's finding of a 4–14 percent range of improvement for seven normal-hearing listeners tested. When including only semivowels and nasals, correct identification scores for subject MP dropped 13 points from 76 percent with low-pass filtering to 63 percent with frequency lowering. Similarly, the scores of subject JR dropped 10 points, from 81 percent to 71 percent. It is apparent from tl confusion matrices that processing degraded the intelligibility of nasals and (particularly) semivowels

while improving the intelligibility of other sounds, in particular the fricatives and affricates. This result is confirmed by the analysis of unconditional informational transfer on individual features (displayed in **Figure 4**), which indicates that the features, affrication and duration, were better perceived under frequency lowering, while nasality and round were better perceived under low-pass filtering.

Identification of the vowels (/a,i,u/) was unaffected by the processing system. Each subject correctly identified 89–91 percent of the vowels presented with processing and low-pass filtering, and errors were based solely on confusions of /i/ and /u/.

EXPERIMENT 2: EVALUATIONS WITH A MODIFIED FREQUENCY-LOWERED SYSTEM

The results of Experiment 1 indicated that, while the frequency-lowering technique studied by Lippmann improved the recognition of stop, fricative, and affricate consonants, it also had the effect of degrading the perception of nasals and semivowels. Processing appeared to degrade consonantal signals characterized by low frequency energy. The processing of second and third formant vowel information may have interfered with the perception of low frequency cues of consonants in the CV syllables. Specifically, the lowered speech sounds of processed consonants may have been masked by the lowered speech sounds resulting from vowel processing. In an attempt to increase the intelligibility of frequency-lowered speech, the frequency-lowering system discussed in Experiment 1 was modified to suppress the processing when low frequency energy is predominant in the speech signal (as is the case for nasals, semivowels, and vowels) and to proceed with processing when high frequency energy predominates (as is the case for plosives, fricatives, and affricates).

Method

System Description. The modified system differs from that shown in **Figure 1** in that two additional signals were supplied to the minicomputer: high and low frequency energy of the input signal. These additional signals were formed by first passing speech through a bank of contiguous one-
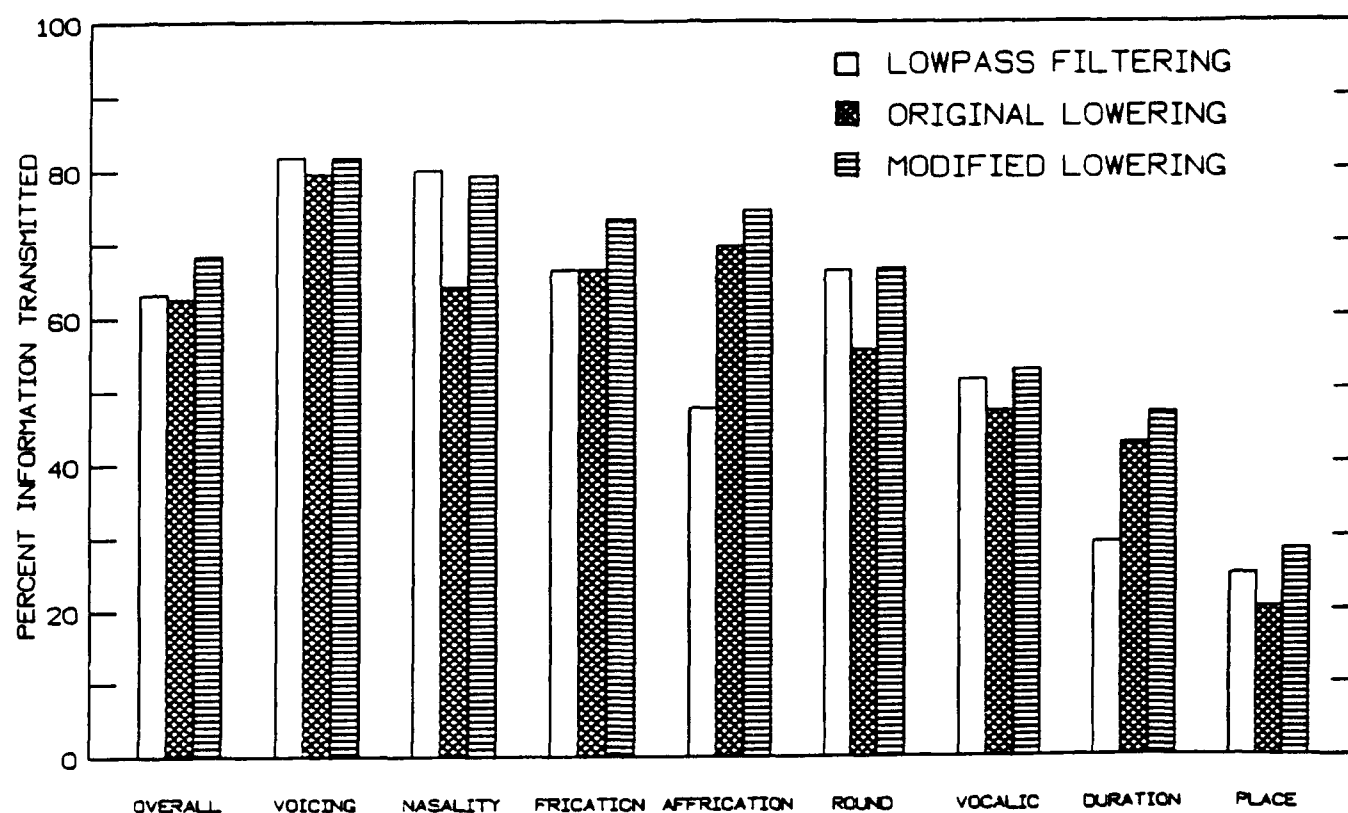
Figure 4.
Percentage of information transfer on the overall set of 24 consonants and on each of eight features for the three experimental conditions. Results were computed from confusion matrices combined across subjects MP and JR.

third octave filters with center frequencies of 125 to 5,000 Hz. The outputs of the filters with center frequencies of 125 to 1,250 Hz were combined using analog summing amplifiers to form the low frequency signal. Likewise, the high frequency signal was formed by summing the outputs of the filters with center frequencies of 1,600 to 5,000 Hz. (The selection of an appropriate corner frequency for defining low and high frequency components was based on samples of spectra of each consonant and vowel spoken by one male and one female talker.) The logarithms of the rms levels of the low and high frequency bands were determined, in addition to the output levels of the four analysis bands.

The low frequency noise was added to the original speech signal only when the power in the high frequency band exceeded that in the low frequency band less 3 dB. This threshold value was determined from subjective listening by choosing a value for which processing appeared to be sup-

pressed for most vowels, nasals, and semivowels and activated for fricatives and affricates. Thus, when the low frequency power was greater than the high frequency power plus 3 dB, processing was suppressed by setting each variable gain amplifier to a maximum attenuation. Otherwise, processing proceeded in the same manner as that used in the preliminary experiment except each analysis-band level was sampled only once and each noise-band attenuation set accordingly. After all variable gain amplifiers were set (to maximum attenuation or attenuation values determined from corresponding analysis-band levels), the program repeated the above cycle. If processing was performed on each cycle, each analysis-band level was sampled every 0.35 ms. Otherwise, each analysis band was sampled every 0.21 ms. Sampling each band at the same rate, as opposed to sampling higher-frequency bands more frequently (as was done in the preliminary experiment), had negligible effects on the process-

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ing. When processing occurred, the levels of the noise bands were controlled in the same manner as that described for the original system.

The same two subjects who participated in Experiment 1 were trained and tested on the identification of frequency-lowered speech produced by the modified system. Both subjects were aware of the modifications made to the system. The same stimulus sets and experimental procedure from the preliminary experiment were used. Results obtained were compared with the low-pass filtering results obtained previously and also to results obtained using the frequency-lowering system of Experiment 1.

Results

Learning curves for each subject for the modified lowering system (triangles) are shown, along with those for low-pass filtering (circles) and the original lowering system (squares), in **Figure 2**. Each subject's pretraining test score on the modified system was roughly equivalent to the posttraining score on the original system. Posttraining test results showed gains of 4 and 6 percentage points for subjects JR and MP respectively, when compared with the pretraining test results.

Posttraining test scores for various groupings of consonants are shown in **Figure 3** for each of the two subjects for each experimental condition. For subject MP, the overall percentage of consonants identified correctly using the modified system was 66 percent, an increase of 10 percentage points over low-pass filtering and an increase of 9 percentage points over frequency lowering using the original system. Subject JR identified 62 percent of the consonants correctly using the modified system, an increase of 4 percentage points over both systems of the preliminary experiment. Although we did not obtain repeated measures, the differences in scores between the two systems for each subject are larger than would be expected on the basis of Bernoulli fluctuations. At performance levels of roughly 60 percent correct and 1,000 trials, the standard deviation is 1.5 percentage points, and $2\sqrt{2}\sigma$ is approximately 4 percentage points.

As predicted, the perception of nasals and semivowels under the modified system was similar to that observed under low-pass filtering, due presumably to the selective processing achieved by the modified system. For subject MP, performance on

nasals and semivowels was equivalent for the modified system and for low-pass filtering, and was superior to the performance on identification of these consonants for the original system (an improvement of 14 percentage points). Subject JR also perceived nasals and semivowels better using the modified system over the original system (a 6 percentage-point improvement) but still perceived these consonants better with low-pass filtering (her identification score was 4 percentage points higher with low-pass filtering than with the best frequency-lowering system). When including only stop, fricative, and affricate consonants, the percentage of consonants correctly identified by each subject was greatest for the modified system. The improvement in perception of these consonants using the modified system over the original system (6 percentage points for subject MP and 2 percentage points for subject JR) offers some support for the notion that processed vowel formant information in the original system may have interfered with the perception of these consonants.

The identification of vowels for each subject was similar in all three systems tested. Each subject correctly identified 90 percent of the vowels, using the modified system, compared with 89-91 percent with the other two systems.

To compare consonant confusions across the three systems, confusion matrices were generated by combining the post-training test results for both subjects for each system tested. The matrices appear in **Table 2** for low-pass filtering, **Table 3** for frequency lowering for the original system, and **Table 4** for frequency lowering for the modified system. Overall, performance on the original frequency-lowering system and low-pass filtering were equivalent (57 percent correct consonant identification). Performance on the modified system (64 percent correct) was 7 percentage points higher than on the other two systems (a significant difference based on Bernoulli statistics). A comparison of the percentage of unconditional information transfer on individual features across the three systems tested is shown in **Figure 4**. The perception of each feature related to nasals and semivowels (voicing, nasality, round, vocalic, place) under the modified system was as good as that obtained with low-pass filtering. The modified frequency-lowering system maintained the large improvement in the amount of information

Table 2.
Confusion matrix compiled across subjects for post-training data on the low-pass filtering condition.

	p	t	k	b	d	g	f	θ	s	ʃ	v	ð	z	ʒ	tʃ	dʒ	h	m	n	hw	w	r	j	l	
p	64	24	7	—	—	—	—	—	—	—	—	—	—	—	—	—	1	—	—	—	—	—	—	—	96
t	23	45	24	—	—	—	—	—	—	—	—	—	—	—	4	—	—	—	—	—	—	—	—	—	96
k	11	25	48	—	—	—	—	—	—	—	—	—	—	—	12	—	—	—	—	—	—	—	—	—	96
b	—	—	—	57	30	6	—	—	—	—	—	2	—	—	—	1	—	—	—	—	—	—	—	—	96
d	—	—	—	7	62	27	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	96
g	—	—	—	6	20	60	—	—	—	—	—	—	—	—	—	9	—	—	—	—	—	—	1	—	96
f	—	—	—	—	—	—	24	29	29	7	—	—	—	—	—	—	7	—	—	—	—	—	—	—	96
θ	1	—	—	—	—	—	5	50	22	4	7	1	—	—	2	4	—	—	—	—	—	—	—	—	96
s	—	—	—	—	—	—	8	26	45	13	1	—	—	—	1	—	2	—	—	—	—	—	—	—	96
ʃ	—	—	—	—	—	—	—	7	38	40	—	—	1	2	1	—	6	—	—	1	—	—	—	—	96
v	—	—	—	—	—	1	2	3	—	—	36	25	15	5	1	6	—	—	—	—	—	1	1	—	96
ð	—	—	—	3	7	2	1	8	—	—	13	28	13	2	—	6	—	—	—	—	—	6	7	—	96
z	—	—	—	—	—	—	—	—	—	—	13	30	22	25	—	2	—	—	—	—	—	—	4	—	96
ʒ	—	—	—	—	—	—	—	—	—	—	8	24	17	37	—	1	—	—	—	—	—	—	5	4	96
tʃ	—	9	10	—	—	—	—	—	—	1	—	—	—	—	73	1	2	—	—	—	—	—	—	—	96
dʒ	—	3	3	—	—	14	—	1	—	—	—	1	—	—	16	58	—	—	—	—	—	—	—	—	96
h	2	1	—	—	—	—	1	3	1	2	—	—	—	—	4	2	73	—	—	6	1	—	—	—	96
m	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	36	53	—	—	2	1	4	96
n	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	8	78	—	—	4	—	6	96
hw	—	—	—	—	—	—	—	—	—	1	—	—	—	—	7	—	4	—	—	83	1	—	—	—	96
w	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	—	2	76	8	8	2	96
r	—	—	—	—	—	—	—	—	—	—	4	3	1	6	—	3	—	—	—	3	7	53	11	5	96
j	—	—	—	—	—	—	—	—	—	—	—	2	1	1	—	1	—	—	—	—	3	—	88	—	96
l	—	—	—	—	—	—	—	—	—	—	1	1	—	—	—	—	—	5	3	—	—	—	—	86	96
	101	107	92	73	119	110	41	127	135	68	83	117	70	78	121	94	95	49	134	95	88	74	126	107	2304

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Table 3.
Confusion matrix compiled across subjects for post-training data on the original lowering system.

	p	t	k	b	d	g	f	θ	s	ʃ	v	ð	z	ʒ	tʃ	dʒ	h	m	n	h	w	r	j	l	
p	48	16	25	–	–	1	–	2	1	–	–	–	–	–	–	–	3	–	–	–	–	–	–	–	96
t	28	55	9	–	–	–	–	–	–	–	–	–	–	–	3	1	–	–	–	–	–	–	–	–	96
k	30	34	24	–	–	–	–	–	–	–	–	–	–	–	6	1	1	–	–	–	–	–	–	–	96
b	–	–	–	74	12	9	–	–	–	–	–	–	1	–	–	–	–	–	–	–	–	–	–	–	96
d	–	–	–	13	56	27	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	96
g	–	–	–	16	9	68	–	–	–	–	–	–	2	–	–	–	–	–	–	–	–	–	1	–	96
f	–	1	–	–	–	–	28	15	26	3	–	–	–	–	–	–	21	–	–	2	–	–	–	–	96
θ	2	1	–	–	–	–	11	43	26	1	6	–	–	–	–	5	1	–	–	–	–	–	–	–	96
s	–	–	–	–	–	–	5	24	43	18	–	–	–	2	–	–	4	–	–	–	–	–	–	–	96
ʃ	–	–	–	–	–	–	–	1	18	72	–	–	–	3	1	–	1	–	–	–	–	–	–	–	96
v	–	–	–	–	1	–	8	–	–	–	41	29	4	5	–	–	1	–	–	2	–	3	2	–	96
ð	–	–	–	2	5	5	–	1	–	–	21	40	2	5	–	–	–	–	–	–	2	4	9	–	96
z	–	–	–	–	–	–	–	–	–	–	13	24	29	21	–	2	–	–	–	–	–	–	7	–	96
ʒ	–	–	–	–	–	–	–	–	1	–	7	10	15	60	–	1	–	–	–	–	–	–	2	–	96
tʃ	–	13	–	–	–	–	–	–	–	1	–	–	–	–	80	2	–	–	–	–	–	–	–	–	96
dʒ	–	3	–	–	2	6	–	–	–	–	–	–	–	–	12	73	–	–	–	–	–	–	–	–	96
h	2	–	–	–	–	–	3	4	1	1	1	–	–	–	–	–	71	–	–	8	4	1	–	–	96
m	–	–	–	–	–	–	–	–	–	1	–	2	–	–	–	–	–	35	47	–	1	2	–	8	96
n	–	–	–	–	–	–	–	–	–	–	1	3	–	–	–	–	–	15	54	–	–	3	2	18	96
hw	–	–	1	–	–	–	–	1	3	–	–	–	–	–	–	–	7	–	–	76	6	–	2	–	96
w	–	–	–		–	–	–	–	–	1	–	–	–	–	–	–	–	–	–	4	65	11	8	7	96
r	–	–	–	–	–	–	1	–	–	–	6	–	2	–	–	1	1	–	–	1	8	55	13	8	96
j	–	–	–	–	2	–	–	–	–	–	9	2	–	1	–	–	–	–	3	1	5	20	51	2	96
l	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	–	6	–	–	2	4	–	–	84	96
	110	123	59	105	87	116	55	92	119	97	106	113	52	97	102	86	111	56	104	94	93	103	97	127	2304

Table 4.
Confusion matrix compiled across subjects for posttraining data on the modified lowering system.

	p	t	k	b	d	g	f	θ	s	ʃ	v	ð	z	3	tʃ	dʒ	h	m	n	hw	w	r	j	l	
p	56	8	25	-	-	-	-	-	-	-	-	-	-	-	-	5	2	-	-	-	-	-	-	-	96
t	11	63	21	-	-	-	-	-	-	-	-	-	-	-	1	-	-	-	-	-	-	-	-	-	96
k	24	15	53	-	-	-	-	-	-	-	-	-	-	-	4	-	-	-	-	-	-	-	-	-	96
b	-	-	-	71	16	9	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	96
d	-	-	-	4	42	50	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	96
g	-	-	-	13	11	67	1	-	-	-	-	1	-	-	-	1	-	-	-	-	-	2	-	-	96
f	-	1	-	-	-	-	35	10	31	4	-	-	-	-	-	-	15	-	-	-	-	-	-	-	96
θ	-	-	-	-	1	1	12	38	29	2	4	1	-	-	-	6	2	-	-	-	-	-	-	-	96
s	-	-	-	-	-	-	3	12	53	25	-	-	1	1	1	-	-	-	-	-	-	-	-	-	96
ʃ	-	-	-	-	-	-	-	-	10	86	-	-	-	-	-	-	-	-	-	-	-	-	-	-	96
v	-	-	-	1	-	-	7	1	-	-	41	27	10	-	-	3	2	-	-	-	-	2	1	1	96
ð	-	-	-	3	6	5	1	4	-	-	27	26	9	1	-	-	-	-	-	-	-	3	11	-	96
z	-	-	-	-	-	-	-	-	-	-	13	24	38	15	-	-	-	-	-	-	-	-	6	-	96
3	-	-	-	-	-	-	-	-	-	2	-	8	16	57	-	13	-	-	-	-	-	-	-	-	96
tʃ	-	6	1	-	-	-	-	-	-	-	1	-	-	-	88	-	-	-	-	-	-	-	-	-	96
dʒ	-	2	-	-	-	1	-	-	-	-	-	-	-	1	10	82	-	-	-	-	-	-	-	-	96
h	1	-	1	-	-	-	8	-	1	1	-	-	-	-	-	1	81	-	-	2	-	-	-	-	96
m	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	42	49	-	-	3	1	1	96
n	-	-	-	-	-	-	-	-	-	-	-	1	-	-	-	-	-	12	71	-	-	3	-	9	96
hw	3	-	-	-	-	-	2	-	-	-	-	-	-	-	-	-	3	-	-	84	4	-	-	-	96
w	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	4	67	19	6	-	96
r	-	-	-	-	-	-	1	-	-	-	4	4	3	1	-	-	-	-	-	3	9	59	9	3	96
j	-	-	-	-	-	-	-	-	-	-	2	1	1	-	-	-	-	-	3	-	1	2	86	-	96
l	-	-	-	-	-	-	-	-	-	-	-	4	-	-	-	-	-	4	1	-	-	1	-	86	96
	95	95	101	92	76	133	70	65	124	121	91	97	78	76	104	111	105	58	124	93	81	92	122	100	2304

transferred under the original lowering system for the affrication and duration features and also showed a substantial gain in the amount of information transferred for the frication feature. Thus, the modified frequency-lowering system managed to maintain a significant processing advantage for fricatives and affricates without degrading the perception of the nasals and semivowels.

DISCUSSION

The modified frequency-lowering system performed significantly better than the Lippmann-based system evaluated in the preliminary experiment. Each subject found the nasals and semivowels to be more intelligible with the modified system, in addition to a number of the stop, fricative, and affricate consonants, as predicted earlier. Compared with low-pass filtering, the modified system performed equally well in handling nasals and semivowels, while improving the intelligibility of stop, fricative, and affricate consonants.

The effects of training on the reception of lowered speech were also studied by Reed, et al. (15) for frequency lowering accomplished through a pitch-invariant non-uniform compression of the

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Section I. Digital Techniques in Acoustic Amplification: Posen et al.

short-term spectral envelope (16). Using a training procedure similar to the one employed here, consonant identification was studied with a major subset of the same CV productions utilized for the present study. For speech lowered to bandwidths ranging from 1,000-1,250 Hz, posttraining scores of 60-67 percent correct were observed for two listeners with normal-hearing and one listener with high frequency sensorineural loss. The number of trials required to achieve asymptotic performance for the spectrally compressed speech, however, was nearly double that required for the vocoder-based processing. Reed, et al. (17) reported on the discriminability of consonants processed by the frequency-lowering system used by Hicks. Their results were similar to the results of the present study in that overall performance under frequency lowering was roughly comparable to that with low-pass filtering, but the perception of various consonants was different in the two systems tested; the perception of nasals and semivowels was better for filtering than for lowering, while the perception of fricatives and affricates was superior under lowering conditions than for low-pass filtering.

In the current study, evaluations of performance were limited to auditory presentation of low-pass-filtered or frequency-lowered speech alone, and did not include the more realistic condition of supplementing the auditory stimulus with lipreading information. An analysis of the confusions for the modified vocoder-based lowering system indicates that a high percentage of the errors (>75 percent) stems from place confusions within a given class of sounds. For example, a high error rate is observed among the three voiced stops as well as among the three unvoiced stops. It is highly likely that such confusions may be resolved through the addition of speechreading, for which place of articulation is better perceived than voicing or manner of articulation.

Braida (18) describes a model of audiovisual integration that can be used to estimate the identification scores that would be obtained in conjunction with speechreading for each type of processing. In the "prelabeling" model, identification in each modality is described in terms of a multidimensional Thurstonian process, and audiovisual identification is assumed to reflect the orthogonal composition of auditory and visual cues. Predicted audiovisual scores were derived by fitting auditory and visual confusion matrices using separate three-dimensional cue-spaces, and computing the highest value of accuracy corresponding to the six-dimensional audiovisual cue space. Predictions were derived for visual confusions reported by Busacco (19)—20 consonants, /a/ context—and Owens and Blazek (20)—23 consonants, /a/ and /u/ contexts, hearing-impaired listeners—combined with auditory confusions from each of the three systems tested in the present study. As expected, the predicted audiovisual scores were higher than both auditory and visual scores (roughly 95 percent correct for the /a/ context, 85 percent for /u/), and the range of scores was smaller than in the auditory case. Predicted AV scores for low-pass filtered materials were roughly equal to those for the unmodified processing. A small but significant advantage was observed for the modified processing, particularly under the /u/ context where predicted AV scores were four points higher than for low-pass filtering. These results suggest that the improvements in auditory performance obtained with the modified processing are likely to be seen in audiovisual consonant recognition tests, although the size of the improvement may be reduced.

SUMMARY AND CONCLUSIONS

The following points summarize the major findings of the present study:

1. Results obtained on two subjects with normal-hearing with a vocoder-based frequency-lowering system showed that the intelligibility of fricative, stop, and affricate consonants was improved by an average of 6 percentage points over low-pass filtering, while the intelligibility of semivowels and nasals was degraded by 12 percentage points on the average. The present study replicated results obtained by Lippmann (8), who used only fricative, stop, and affricate sounds and found an average improvement of 10 percentage points over low-pass filtering. In addition, the preliminary study indicated a degradation in performance for vowel-like consonants.

2. Based on the results obtained with the original frequency-lowering system, this system was modified so as to suppress the processing whenever low frequency components dominated the input signal. In this manner, it was

hoped that the modified system would maintain the processing advantage for stops, fricatives, and affricates while not degrading semivowels and nasals. This modification was achieved by measuring and comparing two additional signals, the high and low frequency energies of the input signal. When the low frequency energy exceeded the high frequency energy by a specified threshold amount, the processing was suppressed; otherwise, the system performed as before. Results indicated that the modified system further improved the intelligibility of stop, fricative, and affricate consonants by an average of 9 percentage points over low-pass filtering. In addition, the intelligibility of semivowels and nasals was similar for processing (with the modified system) and filtering.

3. The ability to use spectral and temporal high frequency cues available in the lowered speech was facilitated by training. Each subject improved his or her ability to perceive lowered speech with training and required an average of 15 hours of training to reach a stable performance level for each system tested.

ACKNOWLEDGMENT

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Toward a theory of optimal hearing aid processing

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Abstract—An ideal hearing aid for a peripheral hearing loss would process the incoming signal in order to give a perfect match between the cochlear outputs of the impaired ear and a reference normal ear. As a first step toward this objective, a model of the normal and impaired peripheral auditory system was used to derive the optimal hearing-aid processing filter based on a minimum mean-squared error criterion. The auditory model includes the compression and suppression effects of the cochlear mechanics and the sensitivity of the neural transduction process. Simplifying assumptions were then incorporated into the processing to yield a practical frequency-dependent adaptive gain system. Processing examples of several individual speech sounds are presented for a flat hearing loss, and the results indicate that a three-channel compression system with adjustable gains and band edges will be close to the optimal solution for this case.

Key words: *frequency dependent adaptive gain system, hearing aid, hearing aid processing filter, three channel compression system.*

INTRODUCTION

An ideal hearing aid for a peripheral hearing loss would process the incoming signal in order to give a perfect match between the cochlear outputs of the impaired ear and those of a reference normal ear. Implementing this ideal system would require access to the complete set of neural fibers in the impaired ear and to a corresponding set of outputs from an accurate simulated normal ear. The simulated normal outputs could then be substituted directly for the neural responses of the impaired ear. In designing a hearing aid, however, one can only indirectly influence the neural outputs by modifying the acoustic input to the ear. Hearing-aid processing thus represents a compromise in which the acoustic input is manipulated to produce an average improvement in the accuracy of the assumed neural responses in the impaired ear.

Even though the ideal solution is not feasible, one can still consider an optimal hearing aid that modifies the acoustic signal to produce the best possible match between the outputs of simulated normal and impaired ears. An advantage of deriving an optimal system is that it requires an explicit mathematical statement of the problem and of the criterion being used for the solution. Thus the *ad hoc* nature of conventional hearing-aid design is replaced by a more rigorous procedure. But since the objective is to match the outputs of the impaired and normal ears, this procedure will still have embedded within it a set of assumptions about auditory behavior and auditory impairment. Therefore, true optimality may not be achieved due to limitations in the understanding of the auditory system.

The most obvious feature of impaired hearing is the shift in auditory threshold. Since the hearing loss tends to vary with frequency, there have been attempts to compensate for the threshold shift with wide dynamic-range compression systems that have behavior that also varies with frequency. While there has been some limited benefit reported for

two-channel compression systems (1,2), multi-channel compression systems have not demonstrated any significant advantage when compared with broadband compression (3,4,5). It thus appears that building more complicated compression systems does not automatically lead to improvements in speech intelligibility.

Improvements in speech recognition or sound quality will require a processing system that better matches the output of the impaired ear to that of the normal ear. The derivation of an optimal processor will indicate the best possible signal processing for those aspects of the problem that can be improved. By concentrating on auditory function, rather than on the characteristics of speech, the resultant signal processing will be effective independent of the auditory stimulus; the structure of the optimal signal processing will place an upper bound on the complexity of compression algorithms that should be implemented in practical devices.

In the next section of the paper, the problem of designing a hearing aid is stated mathematically, and the general form of the solution is derived. Specific features of the auditory system are then described; compression, the sensitivity of the neural transduction process, and two-tone suppression are incorporated into the auditory model. These features lead to an approximate solution that has a simple digital implementation, and examples of processed speech sounds are presented for the new compression algorithm. The relationships between the optimal algorithm and other compression systems are then discussed.

METHOD

Optimal Filter

The hearing aid and impaired ear form the system for which the optimal filter is being designed, with the normal ear as the reference. A block diagram of this system is presented in **Figure 1**. The input signal is $X(k)$, where k is the frequency index at the output of a fast Fourier transform (FFT). This signal is processed by the auditory analysis filters $H_m(k)$ in the normal ear to produce the normal outputs $Y_m(k)$, where m is the channel index for the analysis filters. The filter function $H_m(k)$ includes the gain and frequency analysis of the
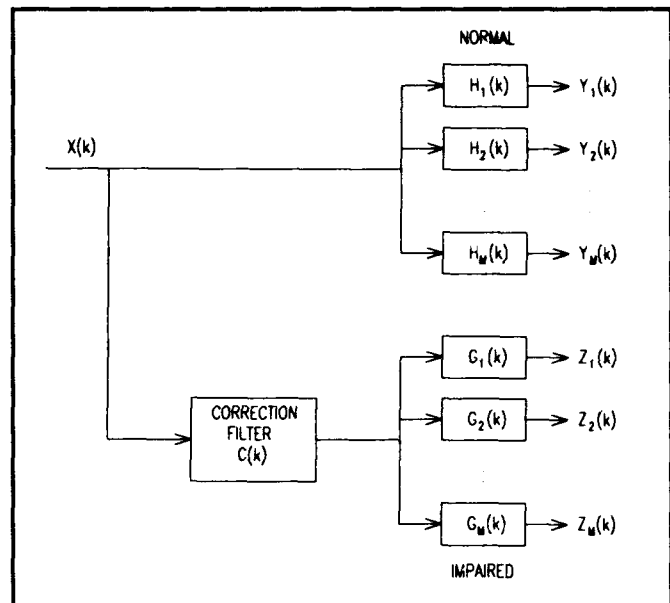

Figure 1.
Block diagram of the hearing-aid processing and the auditory system output.

cochlear mechanics and the sensitivity of the neural transduction process. The fine structure of the inner hair-cell neuron response is not included in this analysis, although it could be part of a more detailed solution that would attempt to match simulated neural firing patterns. The input signal $X(k)$ also goes through the hearing-aid correction filter $C(k)$, after which it is processed by the auditory analysis filters of the impaired ear, indicated by $G_m(k)$, to yield the impaired outputs $Z_m(k)$.

The design objective for the correction filter is the minimization of the differences between the outputs of the impaired ear and those of the normal ear. The criterion chosen is the mean-squared error between the sets of auditory outputs, summed across frequency for each analysis filter and averaged across all of the analysis filters that constitute the auditory model. The error is thus

$$E = \frac{1}{M} \sum_{m=1}^{M} \sum_{k=0}^{K-1} |Z_m(k) - Y_m(k)|^2 \qquad [1]$$

where M is the total number of auditory analysis filters in the model. Substituting the filtered inputs for the outputs yields

$$E = \frac{1}{M} \sum_{m=1}^{M} \sum_{k=0}^{K-1} |C(k)X(k)G_m(k) - X(k)H_m(k)|^2 \quad [2]$$

Setting to zero the partial derivative of the error with respect to the correction filter coefficients yields the optimal filter, given by

$$C(k) = \sum_{m=1}^{M} H_m(k)G_m^*(k) \Big/ \sum_{m=1}^{M} |G_m(k)|^2 \qquad [3]$$

The filters in the ear change in response to the signal level, so adjusting C(k) may cause additional changes in the auditory filters $G_m(k)$ that represent the impaired ear. Thus several iterations may be required to converge to the solution.

The behavior of the correction filter is essentially to remove the effects of the impaired ear and to substitute the effects of the normal ear. The exact nature of the optimal solution, however, will depend on the specific model of normal and impaired cochlear function, since this determines the characteristics of the auditory filters $H_m(k)$ and $G_m(k)$.

Auditory Model

The auditory model includes the effects of the mechanical behavior of the cochlear partition and the sensitivity of the neural transduction process. The purpose of the model is to represent the compression, gain, and frequency resolution in the normal and impaired ears. The model will be used to create a frequency-domain signal-processing system as shown in Figure 1, with a linear-phase correction filter C(k). The temporal features of the auditory processing, such as the variation of the auditory filter impulse-response duration with signal level (6), and the adaptation of the neural firing rate (7) are not included, even though they can be implemented in a more detailed model of auditory function (8). The features that are included in the simplified model presented here are those that will most strongly affect the average gain of the correction filter as a function of frequency.

Auditory Filters and Compression

The auditory analysis filters in the normal ear are narrow band-pass filters having high gain at low signal levels, and tend toward low-pass filters having low gain at high signal levels. The shape of the idealized filter used in the auditory model is approximated by the solid line in Figure 2 for a stimulus at auditory threshold, and is based on the tuning curves measured physiologically in mammals (9,10), and psychophysically in humans (11). The peak of
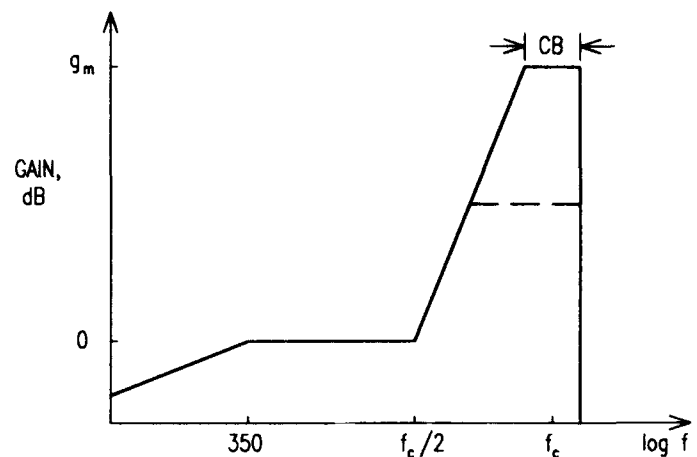

Figure 2.
Idealized auditory analysis filter shape for the condition of maximum filter gain g_m (solid line) and reduced gain due to increased signal level or outer haircell damage (dashed line). The filter bandwidth at auditory threshold is indicated by CB.

the filter response is at the characteristic frequency f_c, and the filter skirts decay to give a response of 0 dB at $f_c/2$ and decay even more rapidly above the characteristic frequency. Below 350 Hz, the response rolls off due to the transfer function of the middle ear (12).

The bandwidth of the auditory filters varies with frequency, being approximately proportional to the filter center frequency (constant-Q) at high frequencies and approaching constant bandwidth at low frequencies. The psychophysical correlate is the critical band (13,14), indicated by CB in **Figure 2**. The equivalent rectangular bandwidth of the critical band corresponds to a distance of approximately 0.9 mm along the cochlea at low signal levels (15). Additional variation of the filter characteristics with frequency, such as the reduced tip-to-tail ratio of the auditory filters at low frequencies (9) and the reduced relative gain of the auditory filters at low frequencies and at very high frequencies (10), have been ignored in this approximate cochlear model.

The maximum gain of the auditory filter is g_m, which in a healthy cochlea can approach 60 dB for a sinusoid at auditory threshold. Increasing the signal level results in a decrease of gain and a broadening of the auditory filters (16), until at high levels the gain can be reduced to a level close to 0 dB. In a cochlea with extensive outer hair-cell damage, the filter shape and gain is similar at all levels to that of the healthy cochlea at high signal levels (17). The

approximate change in filter shape and gain due to outer hair-cell damage or to an increase in signal level is indicated in **Figure 2** by the dashed line truncating the filter response at a gain below g_m while preserving the rest of the filter shape. Since the filter bandwidth increases with increasing signal level, a distance of 1.2 mm along the cochlear partition was used in the model to approximate the bandwidth at signal levels typical of speech.

To estimate the compression ratio in the healthy cochlea, assume that an input at 0 dB sound pressure level (SPL) receives 60 dB of gain, while an input at 100 dB SPL receives 0 dB of gain. The resultant compression ratio is 2.5:1. A severely impaired cochlea, on the other hand, would have 0 dB of gain at all input signal levels due to the outer hair-cell damage, resulting in a linear system. Total outer hair-cell damage results in a threshold shift of no more than 60 dB, since that is the maximum amount of gain provided by the cochlear mechanics. Hearing losses greater than 60 dB must therefore be accompanied by damage to the neural transduction mechanism, and support for this is provided by Liberman and Dodds (18), who showed that inner hair-cell damage results in a threshold shift but no apparent change in the mechanical behavior of the cochlea. Thus outer hair-cell damage, in this model of hearing loss, causes a loss of sensitivity and a reduction in the compression ratio, while inner hair-cell damage causes a linear shift in sensitivity.

The gain in the correction filter depends on the gain in the normal and impaired ears for the incoming acoustic signal. Let the signal level in a given auditory analysis band be x dB. The gain in the normal ear will then be

$$G_{norm} = 60 \frac{100 - x}{100} \text{ dB}, \ 0 \leq x \leq 100 \text{ dB} \quad [4]$$

given the auditory gain assumptions used above. The compressive gain in the impaired ear is assumed to be reduced proportionally by the amount of the outer hair-cell damage, with an additional linear reduction in gain due to the amount of inner hair-cell damage, giving a gain of

$$G_{imp} = 60 \frac{100 - x}{100} \frac{60 - L_o}{60} - L_i, \ 0 \leq x \leq 100 \text{ dB} \quad [5]$$

where L_o is the hearing loss in dB due to the outer hair-cell damage, $0 \leq L_o \leq 60$ dB, and L_i is the
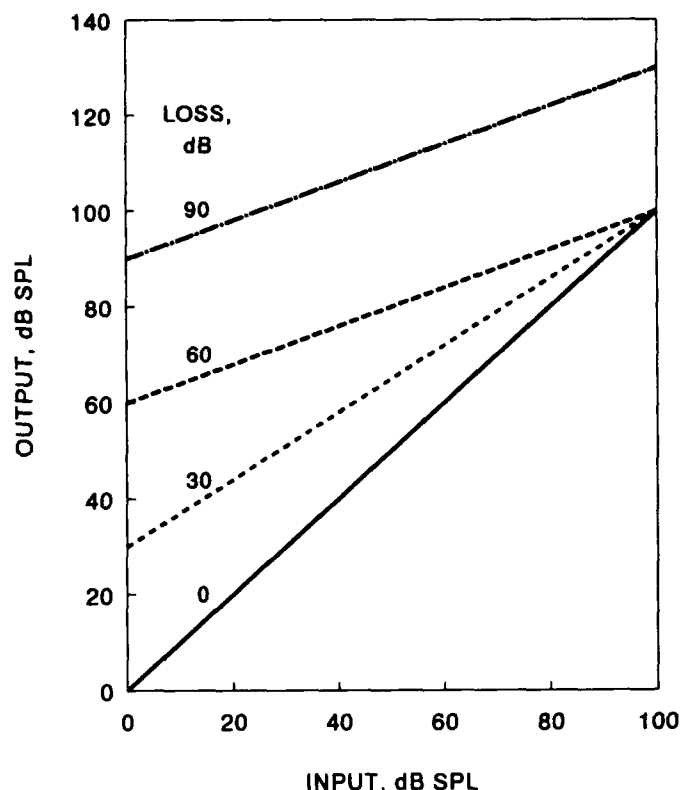

Figure 3.
Compression amplifier input/output curves within an auditory filter band for the model of hearing loss.

hearing loss in dB due to the inner hair-cell damage. The gain provided by the optimal correction filter, treating this band in isolation, is then given by $G_{norm} - G_{imp}$, and this leads to the family of processing input/output curves shown in **Figure 3**, where it has been assumed that a hearing loss of less than 60 dB is due exclusively to outer hair-cell damage and a hearing loss of greater than 60 dB is comprised of a 60 dB loss due to outer hair-cell damage with the remainder of the loss due to inner hair-cell damage. Hearing losses of less than 60 dB cause a change in the compression ratio, and losses greater than 60 dB add a linear shift in gain in addition to the maximum compression ratio of 2.5:1.

The compression model is consistent with fitting procedures, such as the half-gain rule, that have been developed for moderate hearing losses. For example, a narrow-band input at 50 dB SPL would get 30 dB of gain in the healthy cochlea given the compression action represented by Equation [4], while a 60 dB loss due exclusively to outer hair-cell

damage would result in 0 dB of gain for the same stimulus as indicated by Equation [5]. Thus, 30 dB of gain, or half the hearing loss, equalizes the levels that excite the inner hair cells within the auditory analysis band. For more severe losses, where inner hair-cell damage must also be assumed to exist, additional gain beyond the half-gain rule is needed according to the model given in Equation [5], and this has indeed been found to be the case in hearing-aid fittings (19).

Suppression

The second aspect of auditory function that strongly influences the gain in the cochlea is suppression. Physiological measurements of suppression in a normal ear (20,21,22) show that the neural response to a probe tone can be reduced by the simultaneous introduction of a second tone, termed the suppressor. The magnitude of the suppression is approximately linear, that is, increasing the suppressor amplitude by 10 dB will reduce the amplitude of the neural response to the probe to be approximately that of a probe having a 10 dB lower intensity. This effect, over a wide intensity range, depends only on the frequency and magnitude of the suppressor (21). At a stimulus level of 56 dB SPL, representative of speech one-third-octave band intensity levels, the effect of the suppressor has been shown in psychophysical experiments to extend over a frequency region from approximately one-half octave above to one octave below the probe frequency (23). Psychophysical effects for complex suppressors consisting of more than one sinusoid indicate that the suppression is dominated by the most intense sinusoidal component present in the complex (24,25).

The major features of the data cited in the above paragraph can be reproduced by a simple signal-processing model of suppression in the cochlea. When computing the compression gain for an auditory filter, the signal power within the filter is replaced by the maximum signal power observed over a frequency region extending approximately one-half octave above to one octave below the filter characteristic frequency. As a result of the signal-level substitution, the cochlear gain in the auditory model for any given frequency region will be determined by the most intense signal component within that region. Increasing the intensity of the strongest signal component will cause a reduction in

the system gain due to the compression, and this gain change will affect all of the less intense signal components within the frequency region. Thus both the frequency- and gain dependence of the cochlear response to a probe in the presence of a suppressor are incorporated into the auditory model.

Outer hair-cell damage reduces the suppression effects in impaired ears (26,27), and this behavior is reproduced qualitatively in the suppression model. Total outer hair-cell loss, for example, results in a linear system given the compression model, and no suppression will be observed because the gain is a constant independent of the signal. Intermediate amounts of outer hair-cell damage will cause an intermediate reduction of the suppression in the model; the largest amounts of suppression will be observed at the higher compression ratios associated with mild hearing losses, and reduced amounts of suppression will be observed at the lower compression ratios associated with more severe hearing losses.

More detailed suppression behavior in impaired ears is not incorporated into the model. In particular, destruction of outer hair cells in the frequency region of the suppressor will reduce the magnitude of the suppression, even when the cochlear behavior in the region of the probe tone appears to be normal (26,28). This effect will be most pronounced for a region of normal hearing bordered above and below by regions of impaired hearing. The assumption made in the suppression model is that this type of hearing loss rarely occurs, and that the reduction in suppression in the impaired ear can normally be described by the hearing loss within the auditory filter. Thus the additional processing complexity of incorporating the details of suppression behavior in the impaired ear is not warranted given the small difference anticipated in the hearing-aid gain function.

Simplified Algorithm

The optimal hearing-aid correction filter given by Equation [3] requires that the output of every auditory filter in the model be computed at every frequency in the FFT, leading to a substantial amount of computation. The amount of computation can be greatly reduced, with only a small sacrifice in accuracy, by replacing the auditory filter of **Figure 2** with an equivalent rectangular bandpass filter; the auditory filter output is then computed

over just the rectangular pass-band region, with the remaining frequencies set to zero. This simplification leads to a correction filter given by

$$C(k) = H_m(k)G_m^*(k) \, / \, |G_m(k)|^2 \qquad [6]$$

for frequencies k contained within auditory filter m. For a zero-phase hearing-aid system that ignores the auditory filter phase characteristics, the correction filter simplifies even further to yield

$$C(k) = H_m(k) \, / \, G_m(k) \qquad [7]$$

again for frequencies k contained within auditory filter m, and where the filters representing the normal and impaired ears now have a constant gain with zero phase shift within the pass-band and zero gain outside the pass-band. Since most of the auditory filter output power derives from signal components within the pass-band, the effect of the simplification on the optimal correction filter will be minimal, although there may be a slight increase in the low-frequency hearing-aid gain computed for the conditions of total outer hair-cell damage or very intense low frequency maskers since the simplified solution of Equation [7] ignores the effects of upward spread of excitation.

The correction filter given by Equation [7] is superficially similar to a multichannel system using wide dynamic-range compression. In most compression systems, however, the gain is computed independently in each channel (3). The suppression incorporated into the auditory model, on the other hand, links the computed gain values across channels to reduce the variation of system gain with frequency. Thus it is the auditory model that has the greatest influence on the simplified nearly optimal solution and provides its unique characteristics.

The simplified solution of Equation [7] was implemented in a frequency-domain processing system. The incoming speech is sampled at a 20 kHz rate, and divided into segments of 512 samples (25.6 msec) having a 50 percent overlap and weighted with a triangular (Bartlett) window. The spectrum is computed using the FFT, and the signal magnitude is computed in fixed auditory analysis bands corresponding to a distance of 1.2 mm along the cochlea and having an overlap of 0.6 mm; the variation of filter bandwidth with signal level is ignored in this simplified system. The correction filter gain is then computed in decibels for each analysis band, using as a reference signal level the

peak output of the auditory analysis bands over a region one octave below to one-half octave above the auditory band center frequency, as explained in the description of the suppression model. The gain calculations take into account the postulated outer and inner hair-cell damage for the impaired ear described in the compression model. The logarithmic gain values for each auditory analysis band are then interpolated across frequency, converted to linear gain values, and used to multiply the input spectrum to obtain the hearing-aid output using an overlap-add procedure.

Examples

The examples are individual phonemes excised from isolated consonant-vowel and vowel-consonant speech tokens produced by an adult male talker. The tokens are at a signal-to-noise ratio (SNR) of 25 dB, with the noise being multitalker babble from the SPIN test tape, and the root-mean-squared (rms) overall level of each speech token was adjusted to be 79 dB SPL. The impaired hearing was set to a flat 60 dB loss assumed to be due entirely to outer hair-cell damage, so the correction filter includes compression but not linear gain given the assumption of no inner hair-cell damage. Since the compression in the processing is wide dynamic range, a shift in the input speech level will cause a smaller compressed shift in the processed output level, but the shape of the hearing-aid frequency response will not change given the flat hearing loss.
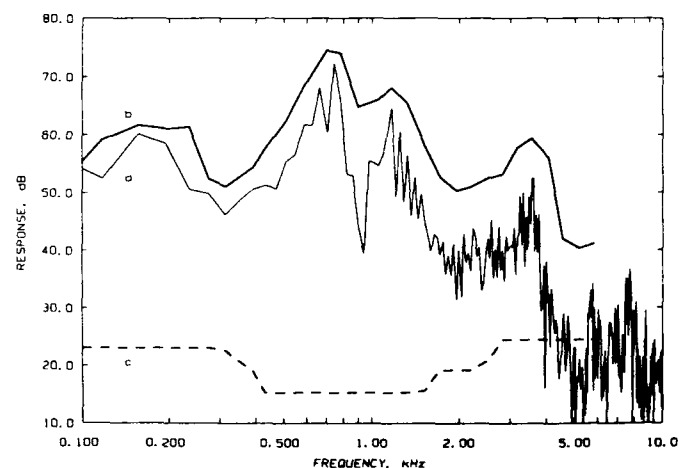

Figure 4.
/a/ in "ka": (a) speech spectrum, (b) spectrum in auditory analysis bands, and (c) computed hearing-aid gain for a flat 60 dB loss.

The first example is a steady-state portion of the vowel /a/ in "ka." The magnitude spectrum of the speech segment is shown in **Figure 4** as curve (*a*), along with the spectrum grouped in auditory analysis bands shown by curve (*b*) and the hearing-aid gain shown by curve (*c*). The vowel second formant at about 1,200 Hz is quite close to the first formant at about 750 Hz, so the first formant peak dominates the gain value computed in the region that contains both formants. The third formant is at about 2,800 Hz, which is more than an octave away from the first or second formants, and so the third formant controls its own distinct gain region. Since the level of the third formant is lower, it receives more gain than the first or second formants. Similarly, the low frequencies also form a region of constant gain determined by the peak at around 200 Hz.

The suppression model thus forms regions of constant gain around the major peaks of the spectrum. The hearing-aid gain function is much smoother than the speech spectrum, and can be represented by a set of plateaus separated by narrow transition regions. Formants that are close together will receive the same gain, thus preserving the details of the local spectral shape, while formants that are farther apart will receive separate gains that will amplify the weaker formant relative to the stronger, thus improving its expected detectability.

The remaining three examples are consonants that differ in their regions of primary spectral power density. The onset of the consonant /p/ in "pa" is shown in **Figure 5**. The spectrum, shown by curve (*a*), has most of its power at low frequencies, and the auditory band powers shown by curve (*b*) show the same pattern. The computed hearing-aid gain, shown by curve (*c*), is separated into a low-frequency region below 1,500 Hz and a high-frequency region above 2,500 Hz. The low-frequency gain is constant below about 1,200 Hz since the two low-frequency peaks in the auditory band spectrum have the same level. The high-frequency gain above 2,500 Hz reaches an asymptote determined by the background speech babble. As was noted for the vowel example of **Figure 4**, the hearing aid gain for the consonant also forms regions of constant gain due to the suppression model, with the gain within each region determined by the outer and inner hair-cell loss.

The second consonant example is just after the onset of /k/ in "ka." As shown in curve (*a*) of

Figure 6, this consonant has a concentration of power in the midfrequencies around 1,600 Hz. The higher mid-frequency signal level in the auditory analysis bands, shown by curve (*b*), results in the reduced hearing-aid gain shown by curve (*c*) for this frequency region. The peak level of the consonant determines the gain over an octave-and-a-half range, so the shape of the spectral peak is preserved. The gain to either side of the peak is essentially flat, being set by the local peaks at low and high frequencies, so the hearing-aid response is neatly divided into three gain regions.

The last consonant example is from a steady-state portion of /sh/ in "ish." This fricative has most of its power at the high frequencies, as shown
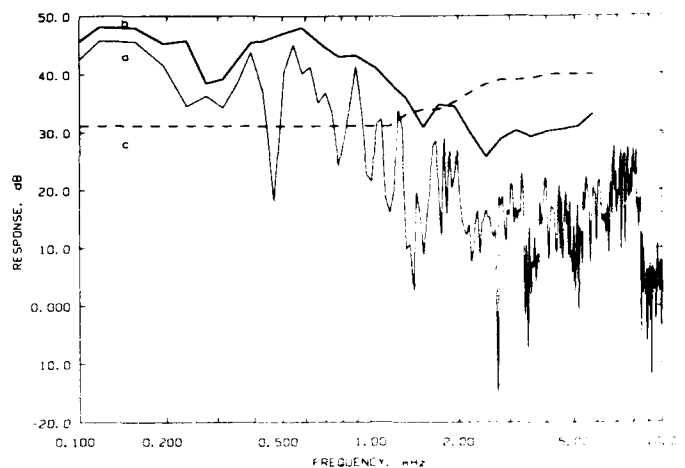

Figure 5.
/p/ in "pa": (*a*) speech spectrum, (*b*) spectrum in auditory analysis bands, and (*c*) computed hearing-aid gain for a flat 60 dB loss.
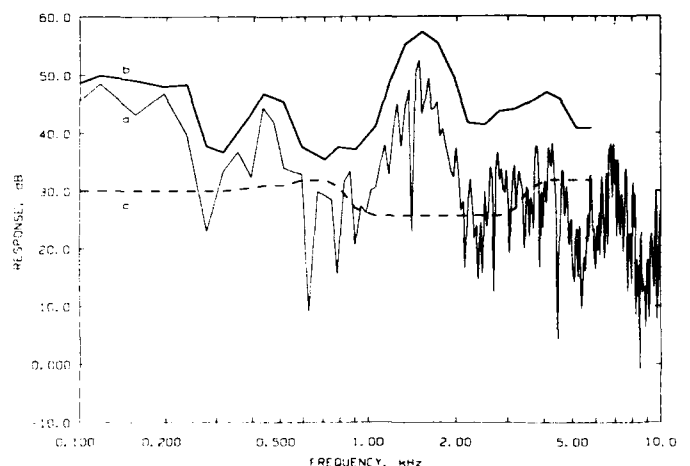

Figure 6.
/k/ in "ka": (*a*) speech spectrum, (*b*) spectrum in auditory analysis bands, and (*c*) computed hearing-aid gain for a flat 60 dB loss.

in **Figure 7** by the spectrum of curve (*a*) and the auditory analysis bands of curve (*b*). The hearing-aid gain, shown by curve (*c*), has a region of constant gain at high frequencies extending half an octave below and one octave above the high frequency peak at about 2,600 Hz. The low-frequency gain is essentially flat, and there is a broad transition band between the low-frequency and high-frequency regions. The complete outer hair-cell damage assumed in the examples has resulted in the maximum compression ratio of 2.5:1 being used in the system, so the overall spectral contrast has been reduced by the processing while the local variations and the shape and sidelobes of the spectral peak have been preserved.

DISCUSSION

The examples for a flat hearing loss show that the optimal hearing-aid processing is characterized by a limited number of regions of constant gain. This is a direct result of the auditory behavior assumed in the processing, since the suppression model causes each significant peak in the speech spectrum to control the gain over a region of one to one-and-one-half octaves. Thus the hearing-aid gain curves show two or at most three regions of constant gain, separated by gradual transition regions, and this basic pattern holds for all of the speech sounds studied. This processing behavior is illustrated schematically in **Figure 8**. The optimal hearing aid is approximated by a three-band system, where the
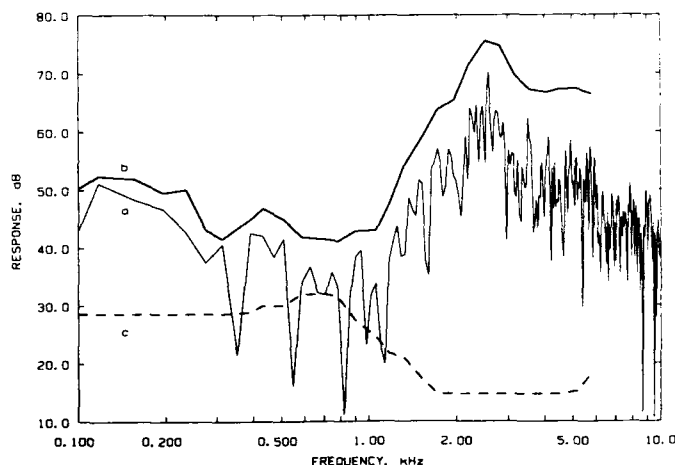
gain in each band and the location of the band edges describe the frequency response of the system for a flat loss. More than three bands are not needed due to the interaction of the speech spectrum peaks and the modeled effects of auditory suppression.

For a hearing loss that varies with frequency, the hearing aid response becomes more complicated. The curve of **Figure 8** now represents the regions of gain control related to the input speech signal, since the signal value used to compute the processing gain is determined by the speech peaks interacting with the suppression model. This control signal value is then used as the input to the gain computation based on the degree of outer and inner hair-cell loss at each frequency in the compression model. For a hearing loss that varies slowly with frequency, a three-band system with constant gains based on the central frequency in each band will still be a reasonable approximation. For a steeply sloping hearing loss, a level dependent frequency equalization curve will be required within each of the control bands indicated in **Figure 8**, and in this case the general FFT-based algorithm implemented in this paper or an equivalent multichannel system having cross-linked gains would be preferable.

The inclusion of suppression in the auditory model provides a system that reduces spectral contrast between broad, widely separated regions of the spectrum, but preserves the spectral shape within each region. The regions are determined by the input signal spectrum, while the gains within them are determined by the hearing loss. Thus the processing continuously adapts to both the signal spectrum and

Figure 7.
/sh/ in "ish": (*a*) speech spectrum, (*b*) spectrum in auditory analysis bands, and (*c*) computed hearing-aid gain for a flat 60 dB loss.
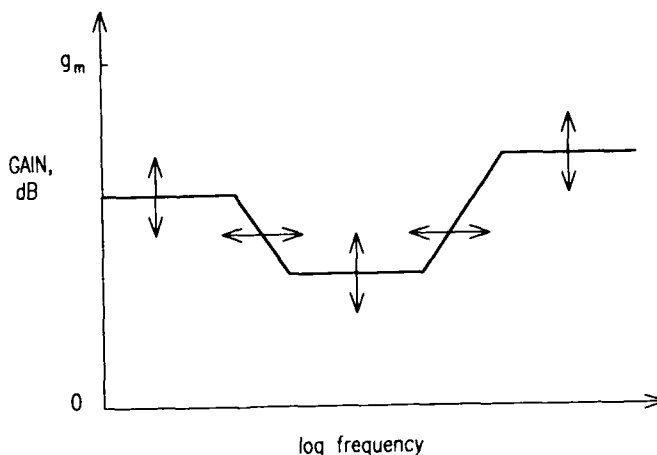

Figure 8.
Frequency response for a three-channel compression system that approximates the optimal hearing aid.

the auditory impairment. This system may offer benefits in comparison with two-channel compression systems having fixed filter crossover frequencies (2); a system having fixed filter band edges may provide inadequate amplification to a weaker speech sound located within the same frequency channel as a more intense sound, since the gain within the channel is determined by the total signal power. The optimal system, however, will provide additional gain for the weaker sound if its spectral power peak is sufficiently distant from that of the more intense sound.

The optimal system may also work better than many recent multichannel systems. Systems having independent gains in each channel (3) can cause wide differences in gain between nearby channels, and the concomitant flattening of the spectrum may reduce intelligibility. The broad regions of uniform gain in the optimum algorithm avoid this problem. The algorithm presented in this paper may also be more effective than compression systems that globally reduce the spectral contrast (4,5), since in the optimal processing the less intense regions of the spectrum receive additional gain, but there is no change in the spectral shape of a significant peak or in the relative amplitude of its sidelobes. The gains determined by the optimal processing algorithm indicate that only a modest amount of spectral modification is needed for the hearing-impaired; large spectral changes may increase rather than reduce the error between the auditory outputs in the impaired and normal ears that is used as the design criterion.

The relevance of the mathematically optimal solution is dependent on the accuracy of the optimization criteria and the assumptions built into the problem formulation and solution. Other optimization criteria, such as minimizing a mean-squared error formulation more representative of average neural firing rates, may yield better processing since the use of the linear outputs favors the higher signal levels. This change, however, would not cause any substantial differences in the simplified algorithm. Weighting the error would also influence the processing; the correction filter can match one auditory-filter output exactly or many outputs approximately, so a weighting function would apportion the accuracy of the match across auditory filter channels. It may also prove advantageous to provide a compression threshold in the processing algorithm in order to limit the gain for low-level sounds, thereby reducing the annoyance of some background noises.

The optimal processing system may also have applications beyond hearing aids, especially since no assumptions about speech or any other specific signal characteristics have been designed into the processing. Broadband compression for audio signals, for example, can be accomplished by setting a fixed compression ratio (e.g., 2:1) in the algorithm, which would be equivalent to selecting a mild-to-moderate flat loss due exclusively to outer hair-cell damage. Stereophonic signals would require selecting the same processing gain in both signal channels; one approach would be to control the gain with a composite spectrum made up of the larger of the left and right channel outputs in each auditory analysis band.

CONCLUSIONS

This paper has presented a new hearing-aid processing approach based on minimizing the mean-squared error between the outputs of simulated normal and impaired cochleas. An optimal solution was derived, and the salient features of the solution were preserved in a simplified frequency-domain algorithm having an efficient digital implementation. The resulting hearing-aid processing system is dependent solely on the characteristics of the modeled cochlea, and not on any assumptions about the nature of the speech signal. Thus the processing should work equally well on speech, music, and environmental sounds. Since the structure of the algorithm is clearly dependent on the optimization criterion and on the characteristics of the model used to represent normal and impaired hearing, inadequacies in the auditory model will limit the effectiveness of the signal processing. Even though the system is promising, it will still be imperfect due to the inability to completely model the auditory periphery and the effects of hearing loss.

The auditory model includes cochlear suppression effects, and this provides an important difference between the new algorithm and previous multichannel compression systems. Because of the modeled suppression, the hearing-aid gain is typically divided into two or three frequency regions, with the gain in each region governed by the most

intense spectral peak within the region. This results in a gain function that varies smoothly with frequency, increasing the relative gain of the less intense spectral regions while preserving the details of the spectral shape within each region. The speech examples for a flat hearing loss indicate that a three-channel system with variable channel gains and crossover frequencies can be designed to have performance close to that of the optimal system. Additional independent channels do not appear to be necessary in a hearing aid and may even be counterproductive, since additional spectral modifications may increase rather than reduce the error between the auditory outputs in the impaired and normal ears that is the hearing-aid design criterion.

ACKNOWLEDGMENT

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Spectral contrast enhancement of speech in noise for listeners with sensorineural hearing impairment: effects on intelligibility, quality, and response times

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Abstract—This paper describes a series of experiments evaluating the effects of digital processing of speech in noise so as to enhance spectral contrast, using subjects with cochlear hearing loss. The enhancement was carried out on a frequency scale related to the equivalent rectangular bandwidths (ERBs) of auditory filters in normally hearing subjects. The aim was to enhance major spectral prominences without enhancing fine-grain spectral features that would not be resolved by a normal ear. In experiment 1, the amount of enhancement and the bandwidth (in ERBs) of the enhancement processing were systematically varied. Large amounts of enhancement produced decreases in the intelligibility of speech in noise. Performance for moderate degrees of enhancement was generally similar to that for the control conditions, possibly because subjects did not have sufficient experience with the processed speech. In experiment 2, subjects judged the relative quality and intelligibility of speech in noise processed using a subset of the conditions of experiment 1. Generally, processing with a moderate degree of enhancement was preferred over the control condition, for both quality and intelligibility. Subjects varied in their preferences for high degrees of enhancement. Experiment 3 used a modified processing algorithm, with a moderate degree of spectral enhancement, and examined the effects of combining the enhancement with dynamic range compression. The intelligibility of speech in noise improved with practice, and, after a small

amount of practice, scores for the condition combining enhancement with a moderate degree of compression were found to be significantly higher than for the control condition. Experiment 4 used a subset of conditions from experiment 3, but performance was assessed using a sentence verification test that measured both intelligibility and response times. Scores on both measures were improved by spectral enhancement, and improved still more by enhancement combined with compression. The effects were statistically more robust for the response times. When expressed as equivalent changes in speech-to-noise ratio, the improvements were about twice as large for the response times as for the intelligibility scores. The overall effect of spectral enhancement combined with compression was equivalent to an improvement of speech-to-noise ratio by 4.2 dB.

Key words: *compression, hearing impairment, response times, spectral enhancement, speech intelligibility.*

INTRODUCTION

People with moderate sensorineural hearing impairment often complain of difficulty in understanding speech in noise. They can understand speech reasonably well in one-to-one conversation in a quiet room, but they have great difficulty when there is background noise or reverberation, or when more than one person is talking. This difficulty appears to be related to a variety of abnormalities in the perception of sound (1) and it persists even when

the speech is amplified sufficiently (by a hearing aid) to be well above the threshold for detection (1,2).

Reduced frequency selectivity is a well-documented abnormality that is associated with sensorineural hearing loss and which can affect speech perception in noise. Frequency selectivity refers to the ability of the ear to resolve a complex sound into its frequency components. This ability is often characterized by describing the ear as containing a bank of overlapping bandpass filters, known as the auditory filters (3). The characteristics of these filters for normally hearing people have been reasonably well established (4,5,6,7). Sensorineu.al hearing loss, and particularly cochlear hearing loss, is associated with broader-than-normal auditory filters, that is, reduced frequency selectivity (8,9). Several studies have shown that the ability to understand speech in noise is correlated with measures of auditory filter bandwidth, although the effects of filter bandwidth are difficult to separate from the effects of a simple loss of sensitivity to weak sounds, since the two are highly correlated (10,11,12). It seems likely that impaired frequency selectivity is at least partly responsible for reduced ability to hear speech in noise, although this causal link has not been universally accepted (13).

One mechanism by which impaired frequency selectivity could affect speech perception in noise involves the perception of spectral shape. The recognition of speech sounds requires a determination of their spectral shapes, especially the locations of spectral prominences (usually formants). One representation of spectral shape in the auditory system is called the excitation pattern. The excitation pattern of a given sound may be defined as the magnitude of the outputs of the auditory filters in response to that sound as a function of filter center frequency (4,6). The excitation pattern resembles a smoothed version of the spectrum. Broader auditory filters produce a more highly smoothed representation of the spectrum. If spectral features are not sufficiently prominent, they may be smoothed to such an extent that they become imperceptible. In one study where degree of spectral contrast was varied, the contrast (decibel [dB] difference between peaks and valleys in the spectrum) required for vowels to be identified was shown to be greater for impaired than for normal listeners (14). Adding a noise background to speech fills in the valleys between the spectral peaks and thus reduces their

prominence, exacerbating the problem of perceiving them for people with broadened auditory filters.

A second possible effect of reduced frequency selectivity on speech perception in noise is connected with the temporal patterns at the outputs of individual auditory filters. The perceived frequency of a given formant and/or the fundamental frequency of voicing may be partly determined by the time pattern at the outputs of the auditory filters tuned close to the formant frequency (15,16). Background noise disturbs this time pattern, which may lead to reduced accuracy in determining these frequencies. This effect would be greater in a person with reduced frequency selectivity, since broader filters generally pass more background noise.

If reduced frequency selectivity impairs speech perception, then enhancement of spectral contrasts might improve it for the hearing-impaired person. Either of the two mechanisms outlined above, one based on degradation of spectral shape and the other on degradation of temporal patterns, provides a rationale for performing spectral enhancement. If spectral features are smoothed by an impaired auditory system, then preprocessing the signal to enhance spectral contrasts can produce an excitation pattern that more nearly resembles the excitation pattern evoked by an unprocessed signal in a normal auditory system. The impaired auditory system can be thought of as convolving the spectrum with a smoothing function, and spectral contrast enhancement can be thought of as a partial deconvolution process. If temporal patterns are disturbed by the noise passing through a broadened auditory filter, then enhancing those portions of the spectrum where the signal-to-noise ratio is highest (the peaks) and suppressing those portions where it is lowest (the valleys) should minimize this effect.

Several authors have described attempts to improve speech intelligibility for the hearing impaired by enhancement of spectral features. Boers (17) processed a set of sentences so as to increase the level differences between peaks and valleys in the spectrum. Noise was added *after* the processing, and the effects of the processing were assessed by measuring the speech-to-noise ratio required for 50 percent of the words to be understood. Overall, the processing reduced intelligibility, although two impaired listeners did show a slight improvement with the processed signals. Even if it had systematically improved intelligibility, this kind of processing

would not be feasible with naturally occurring signals; with these the speech would already be contaminated with noise, and the processing would have to operate on the speech-plus-noise.

Summerfield, et al. (18), synthesized "whispered" speech sounds, and investigated the effect of narrowing the bandwidths of the formants (spectral resonances) used in synthesis. Narrowing these bandwidths led to both sharper spectral peaks and greater peak-to-valley ratios. However, it had only small effects on speech intelligibility; identification of consonants at the end of syllables tended to be slightly better for both normal and impaired listeners when the formant bandwidths were half their nominal normal values. Speech intelligibility in noise was not tested.

Simpson, et al. (19), described a method of digital signal processing of speech in noise so as to increase differences in level between peaks and valleys in the spectrum. Before spectral enhancement, the spectra were smoothed to eliminate minor peaks and ripples, using smoothing filters based on the properties of the auditory filters in normal ears. The enhancement was also done on a frequency scale related to the frequency resolution of normal ears (4). The enhancement procedure involved convolving the spectrum with a Difference-of-Gaussians (DoG) filter. This operation is similar to taking a smoothed second derivative of the spectrum. The spectral pattern obtained in this way was used to construct a gain function to enhance the original spectrum. The intelligibility of the speech in speech-shaped noise was measured using subjects with moderate sensorineural hearing loss. The results showed small but reasonably consistent improvements in speech intelligibility for the processed speech. The processing used by Simpson, et al. ran at about 200 times real time on a reasonably fast laboratory computer (Masscomp 5400 with floating-point accelerator).

Stone and Moore (20) described a speech-processing system similar to that used by Simpson, et al., but one that was simpler, and based on analog electronics running in real time, using a 16-channel band-pass filter bank. Each channel generated an "activity function" that was proportional to the magnitude of the signal envelope in that channel, averaged over a short period of time. A positively weighted activity function from the nth channel was combined with negatively weighted

functions from channels $n - 2, n - 1, n + 1$, and $n + 2$, giving a correction signal used to control the gain of the band-pass signal in the nth channel. Recombining the band-pass signals resulted in a signal with enhanced spectral contrast. Two different experiments were described, the first using the activity function as described, and the second using a nonlinear transform of the activity function. In both experiments, several different weighting patterns were used in calculating the correction signal. The intelligibility of speech in speech-shaped noise processed by the system was measured for subjects with moderate sensorineural hearing loss. In both experiments, no improvement in intelligibility was found. However, subjective ratings of the stimuli used in the second experiment indicated that some subjects judged the processed stimuli to have both higher quality and higher intelligibility than unprocessed stimuli.

Bunnell (21) described a method of digital signal processing to enhance spectral contrasts. Contrasts were enhanced mainly at middle frequencies, leaving high and low frequencies relatively unaffected. Unlike the processing used by Simpson, et al. (19), and by Stone and Moore (20), the enhancement was performed on a spectral envelope that was calculated with a linear frequency scale (using a cepstral smoothing technique) rather than a scale reflecting auditory frequency selectivity. Small improvements were found in the identification of stop consonants presented in quiet to subjects with sloping hearing losses. No measurements of the intelligibility of speech in noise were reported.

Several other authors have described methods of processing speech in noise aimed mainly at enhancing speech quality and/or intelligibility for normal listeners or as preprocessors for speech recognition devices. Lim (22) reviews work done prior to 1983. Many of the techniques that have been developed result in improvements of signal-to-noise ratio (SNR) without any improvement in intelligibility, and many have been plagued by artifacts such as the introduction of spurious sounds as a result of enhancing random spectral peaks. Cheng and O'Shaughnessy (23) described a method similar to that used by Simpson, et al. (19), but differing in several details. They reported an improvement in subjective quality for speech in white noise, based on informal tests with normal listeners. They used two alternative algorithms—one for

low-noise conditions where the improvement in SNR was modest but speech quality (naturalness) was retained or enhanced, and the other for high-noise conditions, where there was a large improvement in SNR but speech quality was degraded. No formal measurements of speech intelligibility were made.

Clarkson and Bahgat (24) filtered signals into several contiguous frequency bands and expanded the envelope in each band, so as to enhance spectral contrast. A measure of spectral variance was used to control the amount of expansion. Listening trials with a simplified real time system showed small, but reasonably consistent, improvements at 0-dB speech-to-noise ratio in a modified rhyme test.

In this paper, we describe a series of experiments aimed at further developing the technique of Simpson, et al. (19). Experiment 1 was a parametric study using processing similar to that described by Simpson, et al. The objective was to find optimum values of two of the parameters used in the processing. The intelligibility of speech in speech-shaped noise was measured for several different conditions involving spectral enhancement. Experiment 2 was carried out using a subset of the conditions from experiment 1, to determine whether the spectral enhancement produced improvements in subjective judgments of speech quality and intelligibility. Experiment 3 investigated the effect of combining spectral enhancement with amplitude compression, with a modified enhancement algorithm, again using measures of the intelligibility of speech in speech-shaped noise. Finally, experiment 4 used a subset of the conditions from experiment 3, but performance was evaluated in a test measuring both speech intelligibility and response time. Although the experiments were primarily concerned with the intelligibility and quality of speech in noise, informal listening tests were carried out using speech in quiet. In all cases, the quality of the processed speech was judged to be good, by both normal and hearing-impaired listeners.

EXPERIMENT 1

Method of Speech Enhancement

The technique used for spectral enhancement was similar to that described by Simpson, et al. (19), and involved manipulation of the short-term spectrum of the speech in noise. Sampled segments of the signal were windowed, smoothed, spectrally enhanced, and then resynthesized using the overlap-add technique (25). Each step is described below. The steps are also illustrated in **Figure 1**.

The speech in noise was low-pass filtered at 4 kHz (Fem EF16, 100 dB/oct slope) and sampled at a 10-kHz rate with 12-bit resolution using a Masscomp 5400 computer with EF12M analog-to-digital converter. A 12.8-ms segment of the signal was weighted with a 12.8-ms Hamming window; the segment was padded with 64 zeros at the start and 64 zeros at the end. A 256-point fast Fourier transform (FFT) of the windowed segment was calculated, giving 128 magnitude values and 128 phase values. The phase values were stored and subsequent operations were carried out only on the magnitude spectrum.

To avoid enhancing spectral details that would be undetectable even for a normal ear, the magnitude spectrum was transformed to an auditory excitation pattern, using the convolution procedure described by Moore and Glasberg (4). This involved calculating the output of an array of simulated auditory filters in response to the magnitude spectrum. Each side of each auditory filter is modeled as an intensity-weighting function, assumed to have the form of the rounded-exponential filter described by Patterson, et al. (26):

$$W(g) = (1 + pg)\exp(-pg), \qquad [1]$$

where g is the normalized distance from the center of the filter (distance from center frequency divided by center frequency, $\Delta f_c/f_c$) and p is a parameter determining the slope of the filter skirts. The value of p was assumed to be the same for the two sides of the filter. The equivalent rectangular bandwidth (ERB) of this filter is $4f_c/p$.

The ERBs of the auditory filters were assumed to increase with increasing center frequency, as described by Moore and Glasberg (4). As a result of this calculation, the original 128 magnitude values were replaced with 128 new values, representing a smoothed version of the original spectrum. The smoothing tended to remove minor irregularities in the spectrum, but to preserve peaks corresponding to major spectral prominences in the speech.

An enhancement function was derived from the excitation pattern by a process of convolution with a DoG function (on an ERB scale). This function is the sum of a positive Gaussian and a negative
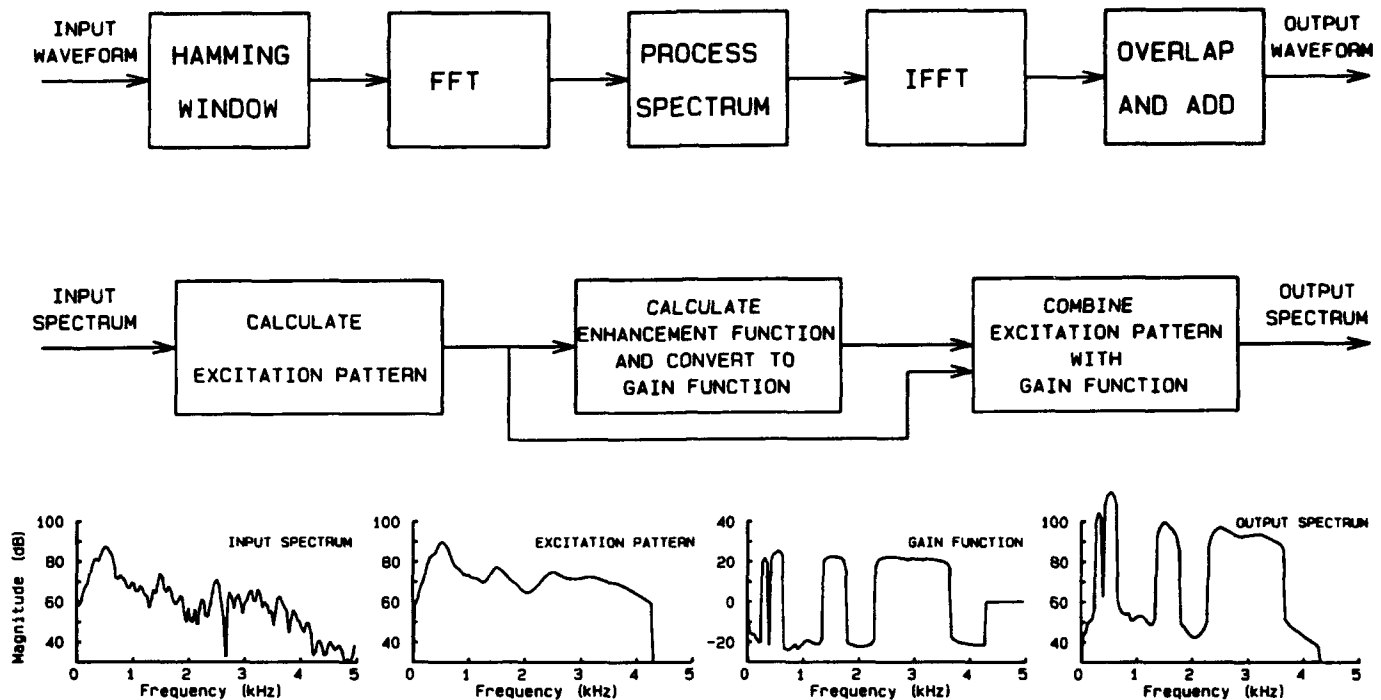
Figure 1.
Schematic diagram of the sequence of stages involved in the enhancement processing of Experiment 1. The top row shows all stages of the processing. The middle row shows the "process spectrum" stage in more detail. The bottom row shows an example of the spectral processing for a particular frame, for condition E3B2.

Gaussian that has twice the bandwidth of the positive Gaussian, as described by the following equation:

$$\text{DoG}(\Delta f) = (1/2\pi)^{1/2}[\exp\{-(\Delta f/b)^2/2\} - (1/2)\exp\{-(\Delta f/2b)^2/2\}], \qquad [2]$$

where Δf is the deviation from the center frequency, and b is a parameter determining the bandwidth of the DoG function. Note that the total area of this function, summed over positive and negative parts, is zero. In these experiments three values of b were used, chosen so that the width of the positive lobe (between the zero-crossing points) was either 0.5, 1.0, or 2.0 times the ERB of the auditory filter with the same center frequency (4). Thus, the width of the DoG function increased with increasing center frequency. The three bandwidths used will be referred to as B.5, B1, and B2.

The DoG function was centered on the frequency of each of the 128 magnitude values of the excitation pattern in turn. For a given center frequency of the DoG function, the value of the excitation pattern at each frequency (in linear power units) was multiplied by the value of the DoG

function at that same frequency, and the products obtained in this way were summed. The magnitude value of the excitation pattern at that center frequency was then replaced by that sum.

The enhancement function derived in this way was then used to modify the excitation pattern. At center frequencies where the enhancement function was positive, the excitation pattern was increased in magnitude; at center frequencies where the enhancement function was negative, the excitation pattern was decreased in magnitude. This was achieved in the following way. Let the absolute value of the enhancement function at a particular center frequency be denoted by abs(ENF) and the corresponding sign (positive or negative) of the enhancement function be denoted sign(ENF). The value of the enhancement function was converted to a decibel-like quantity by calculating

$$G = \log\{\text{abs(ENF)} + 1\} \times \text{sign(ENF)}. \qquad [3]$$

The value of abs(ENF) was generally large (in the thousands), but 1 was added to it to avoid the possibility of taking the logarithm of zero. The value of G was then scaled by a certain factor, E,

and added to the magnitude of the excitation pattern at that center frequency—the excitation level being expressed in decibels. The degree of enhancement of the spectrum was determined by the size of the factor E; values used were 0.3, 0.6, and 0.9, corresponding to small, medium, and large amounts of enhancement. These degrees of enhancement will be referred to as E3, E6, and E9, respectively.

The magnitude values from the enhanced excitation pattern, expressed in linear amplitude units, were then combined with the original phase values, and an inverse FFT was used to produce a 25.6-ms segment of spectrally enhanced speech in noise. This process was repeated every 6.4 ms, and the resultant overlapping segments were summed to give a complete processed waveform.

In summary, the processing had the effect of enhancing spectral contrast in the magnitude spectrum while preserving the phase spectrum. The processing was performed with three degrees of enhancement (E3, E6, and E9) and three values for the width of the DoG function (B.5, B1, and B2), giving nine experimental conditions in total. The condition E3B1 is similar to that used by Simpson, et al. (19). In addition, two control conditions were used. In one, the speech in noise was processed through all stages except those involving enhancement. Thus, the spectrum was smoothed in the conversion to the excitation pattern, but was otherwise unaltered; this corresponds to processing with the value of E set to 0. We refer to this condition as E0. In the second control condition, referred to as NULL, the speech in noise was passed through all stages except the conversion to the excitation pattern and the enhancement. The conversion to an excitation pattern has the effect of putting a high frequency emphasis on the spectrum; this happens because the ERB of the auditory filter increases with center frequency. Since the NULL condition did not involve conversion to an excitation pattern, the high frequency emphasis was obtained in this condition by increasing the power spectrum at a given frequency by an amount proportional to the ERB of the auditory filter at that center frequency. The overall level of the speech-plus-noise was equalized for all conditions.

Figure 2 snows an example of the spectra of stimuli processed using conditions NULL (top panel), E0 (middle panel), and E3B2 (bottom panel). The signal was a synthesized neutral vowel presented
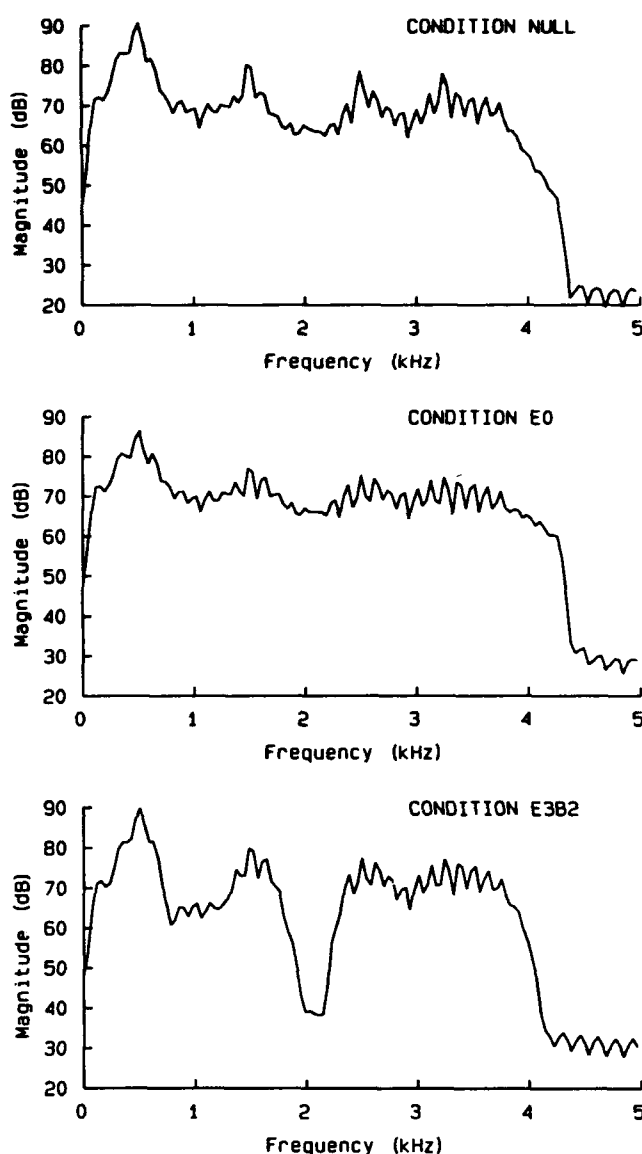

Figure 2.

Example of the effects of the processing used in Experiment 1, showing long-term-average spectra of a neutral vowel in noise, processed using the control condition (NULL-top panel), condition E0 (middle panel) and condition E3B2 (bottom panel).

in speech-shaped noise at a signal-to-noise ratio of 0 dB. The figure shows the long-term-average spectra of the processed stimuli, not the spectra of individual frames; the effects of the enhancement processing were generally more pronounced in the latter. Note how the spectral level between the formants, especially the second and third formants, is decreased by the processing.

Stimuli

The stimuli were the first 11 lists from the Adaptive Sentence Lists (ASL) (27) presented in a continuous background of noise with the same long-term-average spectrum as the sentences. Sentences were presented at 12-sec intervals, leaving ample time for subject responses. Most subjects were tested at a speech-to-noise ratio of 0 dB, both speech and noise levels being specified in terms of root-mean-square pressures. Subjects 4 and 6, who scored poorly at this speech-to-noise ratio, were tested using a ratio of +3 dB. The score was the number of key words identified (out of the 45 in each list). Stimuli were recorded on digital audio tape (DAT) and presented via a Quad amplifier and Monitor Audio MA4 loudspeaker.

Subjects

Eleven subjects were tested. All were diagnosed as having bilateral sensorineural hearing loss, probably of cochlear origin. Their audiograms and other relevant information are presented in Table 1. Most were experienced hearing aid users.

Experimental Design

A Latin Square design was used. All subjects were tested with the 11 ASL lists presented in the same, ascending order. Each subject was tested once in each of the 11 conditions, with the order of conditions counterbalanced across subjects. Thus, for each subject, a different list was used for each of the 11 conditions, and for each condition a different list was used for each of the 11 subjects.

Procedure

The subject sat in a sound-attenuating room facing the loudspeaker at a distance of 1.3 m. Seven of the subjects, those who normally wore hearing aids without any compression circuitry or other "signal processing," listened using their own hearing aids. Initially, they were asked to adjust the volume controls on their aids to the setting that they would use for normal conversation. Then, they were presented with ASL list 12 (i.e., not one of the 11 test lists) processed using condition NULL and the level was varied until they indicated that it was at their preferred listening level. Subject 3, who normally wore hearing aids incorporating compression, and subjects 2, 5, and 8, who did not normally use their aids, were tested unaided; the level of the

stimuli was adjusted to their preferred listening level. The adjustments were usually completed well before the list was completed. The remainder of list 12 was used as practice. In a few cases, list 13, also processed in condition NULL, was used for further practice.

Testing proper then started. Subjects were presented with 11 test lists, with a brief rest between each list. Subjects were told to repeat back as many words as they could, and to make a guess when they were not sure. They were told that the task would be quite difficult, and they were not expected to hear every word.

Results

The scores for each subject for each condition and the mean scores across subjects are shown in Table 2. The mean scores do not differ greatly across conditions, but tend to be lower for the conditions involving a high degree of enhancement (E9). To assess the significance of these effects, the data were subjected to an analysis of variance (ANOVA) with factor condition, with the data blocked across list number and subject (28). In this analysis, the proportions correct were transformed using the expression arcsine ($\sqrt{}$ proportion correct). This transform makes the scores follow a normal distribution more closely. The effect of condition was significant: $F(10,90) = 5.06$, $p < 0.001$. The GENSTAT package used gave estimates of the standard errors of the differences between the mean scores for the different conditions. These standard errors were used to assess the significance of the differences between means (28, p. 81). The mean score for condition E9B.5 was significantly lower ($p < 0.01$) than the mean scores for all other conditions. The score for condition E9B1 was significantly lower than the scores for conditions NULL ($p < 0.01$), E0 ($p < 0.01$), E3B2 ($p < 0.02$), and E6B1 ($p < 0.05$). Scores for the other conditions did not differ significantly.

Overall, these results are disappointing. In contrast to the results of Simpson, et al. (19), the processing did not improve speech intelligibility relative to the control conditions; and a high degree of processing led to a significant worsening of intelligibility. The processing condition giving the highest scores was one involving a moderate degree of enhancement, E3B2. If scores for this condition are compared with the mean scores for the two

Table 1.
Characteristics of the hearing impaired subjects used in the experiments.

| Subject | Age | Sex | Ear | Frequency in kHz | | | | | |
				0.25	0.5	1.0	2.0	4.0	8.0
1	74	M	L	35	30	35	40	55	55
			R	30	35	35	50	65	60
2	75	F	L	25	10	5	30	60	75
			R	15	10	5	35	65	80
3	67	M	L	60	55	55	60	65	75
			R	50	50	55	60	70	70
4	68	M	L	40	65	75	85	85	>100
			R	55	60	70	85	95	>100
5	72	M	L	30	20	35	55	95	80
			R	35	30	40	60	85	95
6	82	M	L	85	80	80	90	95	>100
			R	60	65	80	85	80	85
7	70	M	L	25	30	60	60	75	80
			R	10	10	60	55	65	80
8	78	M	L	40	50	55	60	80	>100
			R	25	25	40	45	70	80
9	69	F	L	35	35	45	45	65	85
			R	65	50	55	70	90	100
10	68	M	L	70	55	45	35	50	60
			R	30	35	45	40	40	50
11	73	F	L	45	50	50	50	55	55
			R	50	50	55	50	55	55
12	71	M	L	30	40	55	30	50	95
			R	30	40	45	30	30	60
13	62	M	L	45	50	65	60	40	70
			R	35	40	55	60	60	55
14	68	M	L	30	40	45	60	70	85
			R	20	35	45	50	55	75
15	72	F	L	35	45	50	55	60	70
			R	30	25	40	55	65	65
16	63	M	L	15	30	40	60	65	55
			R	20	25	40	70	70	60
17	64	M	L	25	15	20	40	65	70
			R	30	20	20	30	60	75
18	69	M	L	20	40	30	35	60	80
			R	15	30	40	40	55	75

Subjects 1-11 took part in Experiment 1. Subjects 1, 4, 7, 9, 10, and 11 took part in Experiment 2. Subjects 8-13 took part in Experiment 3. Subjects 14-18 took part in Experiment 4. Absolute thresholds are given in dB HL.

control conditions, NULL and E0, we find that seven subjects performed better with the processing and four performed more poorly.

There may be several reasons why Simpson, et al., found significant improvements in speech intelligibility with processed stimuli, whereas we did not. The first possibility is connected with tl. fact that Simpson, et al., used only one processing condition

Table 2.
Results of Experiment 1, showing the score for each subject in each condition (number of words correct out of 45) and the mean score for each condition.

Subject	Null	E0	E3B.5	E3B1	E3B2	Condition E6B.5	E6B1	E6B2	E9B.5	E9B1	E9B2
1	41	41	40	40	33	37	43	34	30	38	33
2	38	42	38	40	42	37	42	44	30	36	38
3	36	30	35	32	36	30	31	31	16	22	38
4	34	36	31	23	34	33	34	28	19	28	31
5	37	42	41	39	31	30	37	37	19	38	31
6	34	34	33	29	35	33	27	23	22	26	24
7	36	32	36	38	35	28	34	38	35	36	41
8	45	40	34	40	43	35	41	37	23	26	24
9	30	36	31	40	32	38	35	29	31	37	36
10	28	34	29	28	32	32	35	35	36	34	33
11	40	34	40	35	44	36	33	39	35	33	38
Mean	36.3	36.5	35.3	34.9	36.1	33.6	35.6	34.1	26.9	32.2	33.4

and one control condition. They gave subjects two practice lists (one control and one enhanced) and then tested subjects using six sentence lists for each condition. This gave subjects a reasonably large amount of experience with the processed stimuli. In contrast, each subject in our experiment listened to each condition only once, using a single sentence list. It may be that subjects require a more extended practice period to get a benefit from the processing. It should also be noted that the use of six lists per condition greatly reduces the inherent variability in the data compared with our use of a single list.

A second possible factor only became apparent after the main part of the experiment was completed. We discovered that the enhancement process had an undesired side effect; it tended to produce a high-frequency deemphasis. The spectral level of frequencies above 600 Hz tended to be 3-6 dB lower in the experimental conditions than in the control conditions. This may have offset any potential improvements in intelligibility produced by the enhancement process.

EXPERIMENT 2

Ratings of Speech Quality and Intelligibility

There have been several reports in the past of processing that improves the subjective quality of speech in noise without improving intelligibility (22). Previous work involving judgments of speech quality has mainly used normally hearing subjects, although Stone and Moore (20) reported such measurements for hearing-impaired subjects. Processing that improves speech quality without changing intelligibility may be useful as a means of making listening more pleasant and less effortful. Hence, we decided to investigate whether our processing led to any improvements in subjective speech quality.

Two tests were performed where six hearing-impaired subjects (subjects 1, 4, 7, 9, 10, and 11 from experiment 1) made pair-wise subjective comparisons between sentences in noise processed using conditions E0, E3B1, E3B2, E6B2, and E9B2 of experiment 1, rating them for sound quality in one set of tests and intelligibility in the other set. In addition, we used a condition resembling E3B1, but with the unprocessed signal added back to the processed signal. This had the effect of slightly reducing the amount of enhancement, but also of somewhat reducing the audibility of undesired side-effects of the processing, specifically a slight "gurgling" quality. This condition resembles the processing used by Simpson, et al. (19), and will be denoted by E3B1 + U.

For all 15 possible pairs of processing conditions, 10 pairs of sentences were compared. On a

given trial the same sentence was presented twice, the sentences differing only in the way they were processed. Five different sentences were used, taken from ASL list 1, chosen because performance in experiment 1 was especially poor for these sentences. They were edited so that each sentence was approximately centered in 3 sec of its masking noise. There was an interval of 500 ms between the noises for the two sentences in a pair. Following the end of the noise for the second sentence, there were 5.5 sec of silence during which th~ subject indicated which sentence had the higher quality or intelligibility. Five of the sentence pairs were presented as condition A followed by condition B, while the other five were presented as condition B followed by condition A. The order of presentation of the sentence pairs was randomized for both comparison of processing condition and order of presentation within each individual test. All editing was done digitally using a Masscomp 5400 computer system. Final stimuli were recorded on digital audio tape (Sony DTC 1000ES).

In the first test, subjects were asked to indicate which sentence in each pair had the higher sound quality in terms of pleasantness. In the second test, they were asked to indicate which sentence in each pair they felt was more intelligible. In addition to making a forced-choice decision, each subject was asked to make a confidence rating on each trial, by giving a number indicating how large the difference appeared to be.

For each processing pair, a distance metric was calculated by adding together the 10 (signed) confidence ratings. For each pair of conditions, AB, the sign was positive if B was selected and negative if A was selected. An analysis of the results showed that, for each subject, there was a high correlation between the number of selections and the distance metric; correlations ranged from 0.69 to 0.99, and were typically over 0.8. This indicates that the measures have a fairly high degree of reliability and internal consistency.

Results

The results are summarized in **Table 3** (quality judgments) and **Table 4** (intelligibility judgments). Each cell in each table shows the number of B selections (out of 10) with the distance score in parentheses. For example, in the quality judgments of subject 7, condition E3B1 was preferred over condition E0 nine times out of ten, with a distance metric of 5.0. It should be noted that the results showed a bias for the second sentence in a pair to be preferred over the first. This bias was controlled for by our procedure of balancing the order of processing conditions across pairs, but it probably had the effect of somewhat reducing the overall differences between pairs of conditions.

Two overall measures of the scores for each condition were also calculated. For the first, the preferences were summed across all five pairs of conditions involving a given condition. For example, the summed preference score for condition E0 was equal to the total number of times that condition was preferred over the other five conditions; the maximum value for this score is 50. For the second measure, the signed distance measures for a given condition were summed for all comparisons involving that condition. The signs were chosen so that a positive score would indicate an overall preference for that condition. These scores are also shown in **Table 3** and **Table 4**.

Consider first the quality judgments (**Table 3**). For subjects 7, 9, 10, and 11, conditions E3B1 and E3B1 + U were preferred over the control condition, E0. This is apparent both from the numbers of selections and from the distance scores. Preferences for the other processing conditions varied more across subjects. Subject 11 preferred enhanced speech over the control condition for all conditions involving enhancement. Her overall scores were lowest for the control condition, and highest for the conditions involving the greatest degree of enhancement (E6B2 and E9B2). For subjects 7 and 9, both the overall preference scores and the overall distance scores were highest for conditions E3B1 and E3B1 + U, and lowest for condition E9B2. For subjects 1 and 4, preferences were clearly lowest for the condition involving the greatest degree of enhancement (E9B2). For subject 10, preferences were less clear cut, but there was a consistent trend for the conditions involving enhancement to be preferred over the control condition. Overall, the results indicate that the quality of the enhanced speech in noise was generally preferred over that in

Table 3.

Results of Experiment 2 for the judgments of the quality of speech in noise.

Subject	A condition	E3B1	ENH	B condition E3B2	E6B2	E9B2	Global
1	E0	1 (−4)	5 (1)	3 (−2)	2 (−6)	0 (−11)	39 (22)
	E3B1		8 (1)	3 (−1)	1 (−5)	2 (−7)	27 (8)
	ENH			4 (−1)	0 (−5)	1 (−10)	38 (18)
	E3B2				3 (−4)	2 (−8)	25 (8)
	E6B2					0 (−8)	16 (−12)
	E9B2						5 (−44)
4	E0	4 (1)	7 (9)	6 (5)	3 (−7)	1 (−19)	29 (11)
	E3B1		3 (0)	5 (−2)	4 (2)	1 (−16)	31 (17)
	ENH			6 (2)	7 (5)	1 (−16)	26 (18)
	E3B2				5 (−4)	2 (−8)	30 (17)
	E6B2					3 (−5)	26 (50)
	E9B2						8 (−64)
7	E0	9 (5.0)	6 (2.5)	4 (−0.5)	2 (−4.0)	1 (−6.0)	28 (3)
	E3B1		5 (−0.5)	4 (−2.5)	2 (−3.5)	4 (−1.5)	34 (13)
	ENH			4 (−1.5)	4 (−2.5)	3 (−4.5)	30 (10.5)
	E3B2				2 (−5.)	3 (−3.0)	27 (3.5)
	E6B2					4 (−2.5)	15 (−12.5)
	E9B2						15 (−17.5)
9	E0	9 (17)	8 (13)	3 (−8)	3 (−8)	0 (−26)	27 (12)
	E3B1		4 (−2)	6 (4)	4 (−7)	0 (−28)	35 (50)
	ENH			2 (−11)	3 (−10)	0 (−25)	37 (57)
	E3B2				4 (−6)	1 (−18)	26 (9)
	E6B2					0 (−21)	24 (−10)
	E9B2						1 (−118)
10	E0	7 (4)	8 (4)	7 (3)	9 (7)	6 (3)	13 (−21)
	E3B1		4 (−1)	6 (2)	4 (2)	5 (−2)	28 (3)
	EN!I			4 (−2)	5 (0)	5 (−1)	28 (6)
	E3B2				6 (−1)	3 (−1)	28(5)
	E6B2					5 (0)	29 (8)
	E9B2						24 (−1)
11	E0	9 (13)	10 (18)	10 (21)	10 (18)	9 (16)	2 (−86)
	E3B1		6 (2)	5 (−3)	8 (11)	8 (12)	22 (−9)
	ENH			5 (2)	7 (7)	10 (19)	24 (−8)
	E3B2				9 (11)	9 (14)	22 (−5)
	E6B2					6 (2)	38 (45)
	E9B2						42 (63)
Mean	E0	65	73	55	48	28	46
	E3B1		50	48	38	33	59
	ENH			42	43	33	61
	E3B2				48	33	53
	E6B2					30	50
	E9B2						32

Preference scores (ranging from 0 to 10) indicate the number of times the B condition was preferred over the A condition. Distance scores (in parentheses) are positive if the B condition was preferred and negative if the A condition was preferred. The global scores indicate the total number of times (out of 50) that a given condition was preferred, and the total distance (in parentheses). The more positive the distance, the more that condition was preferred overall. For the means, only preferences are shown (as percentages).

Table 4.
Results of Experiment 2 for the judgments of intelligibility.

Subject	A condition	E3B1	ENH	B condition E3B2	E6B2	E9B2	Global
1	E0	6 (1)	4 (0)	7 (2)	5 (0)	4 (0)	24 (−3)
	E3B1		3 (−3)	6 (0)	3 (−1)	5 (1)	29 (4)
	ENH			7 (0)	2 (0)	6 (1)	22 (−4)
	E3B2				4 (−1)	5 (0)	31 (3)
	E6B2					7 (2)	17 (−4)
	E9B2						27 (4)
4	E0	9 (8)	6 (0)	8 (11)	8 (11)	9 (16)	10 (−46)
	E3B1		5 (−2)	5 (0)	5 (7)	4 (−3)	30 (6)
	ENH			4 (0)	5 (5)	5 (1)	27 (−8)
	E3B2				6 (1)	6 (3)	25 (7)
	E6B2					4 (1)	30 (23)
	E9B2						28 (18)
7	E0	7 (2.5)	7 (5.0)	9 (6.5)	9 (5.5)	7 (4.5)	11 (−24)
	E3B1		2 (−1.5)	4 (0.5)	6 (1.0)	7 (3.5)	28 (−1)
	ENH			7 (2.0)	7 (1.0)	6 (0.5)	19 (0)
	E3B2				7 (3.5)	6 (2.5)	27 (3)
	E6B2					8 (2.0)	31 (9)
	E9B2						34 (13)
9	E0	4 (−9)	5 (2)	4 (−6)	3 (−11)	1 (−18)	33 (42)
	E3B1		6 (3)	6 (0)	4 (−7)	1 (−19)	27 (14)
	ENH			3 (−11)	4 (−6)	1 (−24)	33 (46)
	E3B2				4 (−4)	2 (−21)	27 (8)
	E6B2					4 (−8)	21 (−20)
	E9B2						9 (−90)
10	E0	6 (1)	4 (0)	7 (2)	5 (0)	4 (0)	24 (−3)
	E3B1		3 (−3)	6 (0)	3 (−1)	5 (1)	29 (4)
	ENH			7 (0)	2 (0)	6 (1)	22 (−4)
	E3B2				4 (−1)	5 (0)	31 (3)
	E6B2					7 (2)	17 (−4)
	E9B2						27 (4)
11	E0	10 (13)	10 (11)	10 (15)	10 (17)	10 (20)	0 (−76)
	E3B1		6 (0)	5 (0)	8 (7)	9 (13)	22 (−7)
	ENH			7(2)	9 (8)	8 (9)	22 (−8)
	E3B2				10 (9)	9 (14)	23 (−6)
	E6B2					7 (2)	40 (39)
	E9B2						43 (58)
Mean	E0	70	60	75	67	58	34
	E3B1		42	53	48	52	55
	ENH			58	48	53	48
	E3B2				58	55	55
	E6B2					62	52
	E9B2						56

the control condition for moderate degrees of enhancement. As the degree of enhancement was increased, subject 10 showed little change in preference, subject 11 showed an increase in preference, and subjects 1, 4, 7, and 9 showed a decrease.

Consider now the intelligibility judgments (**Table 4**). For subjects 4, 7, and 11, all conditions involving enhanced speech in noise were judged to give higher intelligibility than the control condition, as indicated both by the numbers of selections and

by the distance scores. The overall preference and distance scores were lowest for the control condition. Subjects 1 and 10 did not show clear preferences for any condition, and their distance scores were all rather low. Subject 9 did not show clear preferences for conditions E3B1 and E3B1 + U relative to the control condition, but tended to prefer conditions E0, E3B1, and E3B1 + U over conditions involving large degrees of enhancement (E6B2 and E9B2).

In summary, the results of experiment 2 showed that, for judgments of both quality and intelligibility, speech in noise processed using a moderate degree of enhancement was generally preferred over the control condition. The results for higher degrees of enhancement varied across subjects. Subject 11 preferred the highest degree of enhancement both for quality and for intelligibility. For several other subjects, quality decreased for the highest degree of enhancement.

EXPERIMENT 3

Experiment 3 was similar to experiment 1, in that it involved measures of the intelligibility of speech in noise for several processing conditions. However, the processing differed from that used in experiment 1 in several ways. The first difference was in the way that the enhancement was performed. Instead of the DoG function, a function based on the difference between two rounded-exponential functions was used. This is equivalent to calculating two excitation patterns and taking the difference between them.

A second difference between experiments 1 and 3 was in the way that the enhancement signal was transformed into a gain function to modify the spectral shape of the signal. The transformation in experiment 3 was tailored to limit the maximum gain at any frequency to 20 dB (to avoid excessive increases in sound level) and was scaled so that, most of the time, the gain value was within reasonable limits. In addition, the enhancement function was applied to the original magnitude spectrum, rather than to the (normal) excitation pattern. This meant that only major spectral features were enhanced, but fine-grain spectral features were not smoothed in the conversion to an excitation pattern.

Finally, experiment 3 differed from experiment 1 by including conditions using fast-acting compression. This was done because the enhancement processing had the effect of expanding the dynamic range of the speech in noise. Potentially, this expansion could create problems for hearing-impaired subjects, who often have loudness recruitment and an associated reduction in usable dynamic range. The expansion of dynamic range produced by the enhancement processing might have offset the potential advantages to be gained from enhancement of spectral contrast. The compression used in experiment 3 was intended to compensate for the dynamic range expansion.

Method of Processing

Many of the stages in the processing were the same as used in experiment 1. Therefore, only the stages that were different will be described. The magnitude spectrum of a windowed sample of speech in noise was determined as before. An enhancement function was calculated by convolution of the power spectrum with the sum of a positive rounded-exponential function and a negative rounded-exponential function. For both the positive and negative functions, the ERB varied with center frequency according to equations described by Moore and Glasberg (4). The positive function had an ERB that was 0.5 times the "normal" value suggested by Moore and Glasberg, while the negative function had an ERB that was 2.0 times the normal value. The factors of 0.5 and 2.0 were chosen on the basis of informal listening tests. The sum of the two functions had a positive lobe whose width was approximately 0.67 ERB, which is intermediate between the B.5 and B1 values for the DoG function used in experiment 1. Each of the rounded exponentials was scaled, by dividing by its own ERB, so that the area under it was unity; thus, the area under the sum of the positive and negative rounded-exponentials was always zero.

The enhancement function will be designated D(f). It was converted to a gain function according to the following rules:

$$\text{Gain}(f) = 10^{K0.3D(f)} \qquad \text{for } D(f) \leq 0 \quad [4]$$

$$\text{Gain}(f) = 10 - (10 - 1)10^{-K0.3D(f)} \quad \text{for } D(f) \geq 0 \quad [5]$$

The resulting gain function was used to modify the original magnitude spectrum of the sample of speech

in noise by multiplying the magnitude value at each frequency by the value of the gain function at that frequency. The form of equation [5] was chosen so as to limit the maximum gain at any frequency to a factor of 10 (20 dB). The value of the constant K was chosen to give a degree of enhancement comparable to the E3 enhancement of experiment 1. We refer to this processing condition as ENH. Stimuli for the control condition, E0, were obtained by processing stimuli in the same way but with the constant K set to zero. Subsequent to the enhancement processing, the stimuli in condition ENH were digitally filtered (by adjusting the magnitude spectrum prior to calculating the inverse FFT) so that the long-term-average spectrum of the processed noise matched the long-term-average spectrum of the noise in the control condition.

Examples of spectra for stimuli processed using conditions E0 and ENH are given in **Figure 3**. The stimulus was the same neutral vowel in noise as used for **Figure 1**. Note that compared with the processing condition E3B2 of experiment 1 (lower panel in **Figure 1**), condition ENH gave rise to sharper spectral peaks associated with the formants, and a greater spectral valley between the third and fourth formants. The difference can be attributed to the fact that the enhancement function used in experiment 1 was applied to the (normal) excitation pattern, whereas the enhancement function in experiment 3 was applied to the original spectrum.

Four conditions using compression were also run. The compression was implemented using an algorithm described by Robinson and Huntington (29). It was based on the use of a 20-ms sliding rectangular window. The rms value of the waveform within the window was calculated for each position of the window, and that value was used to calculate a gain function applied to the waveform sample at the center of the window. The compression took two forms. The first gave a moderate amount of compression, used a compression ratio of 2 and a compression threshold 10 dB below the peak value of the speech plus noise. We refer to this condition as C10/2. The second used a greater amount of compression, with a compression ratio of 3 and a compression threshold 15 dB below the peak value of the speech plus noise. We refer to this condition as C15/3. The compression was applied both alone and following the enhancement processing. This gave two additional conditions involving both en-
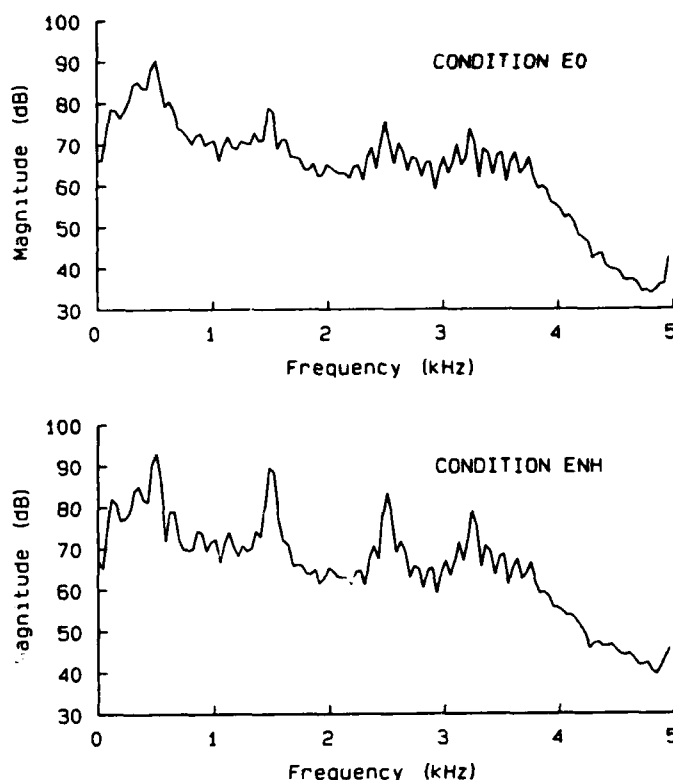
Figure 3.

Example of the effects of the processing used in Experiment 3 showing long-term-average spectra of a neutral vowel in noise, processed using the control condition (E0-top panel) and condition ENH (bottom panel).

hancement and compression, ENHC10/2 and ENHC15/3.

In summary, six conditions were tested: the control condition, E0; a condition involving enhancement alone, ENH; two conditions involving compression alone, C10/2 and C15/3; and two conditions involving both enhancement and compression, ENHC10/2 and ENHC15/3. The overall level of the speech-plus-noise was equalized for all conditions.

Stimuli

The stimuli were lists 13-18 from the Adaptive Sentence Lists presented in a background of noise with the same long-term-average spectrum as the sentences. All subjects were tested at a speech-to-noise ratio of 0 dB, both speech and noise levels being specified in terms of root-mean-square pressures. Subjects were tested without using their hearing aids. In order to compensate for the lack of

Table 5.
Results of Experiment 3, showing the score for each subject in each condition (number of words correct out of 45) for the first and second tests, and the mean score for each condition.

| Subject | Test | Condition | | | | | |
		E0	C10/2	C15/3	ENH	ENHC10/2	ENHC15/3
8	1	40	42	37	37	36	34
	2	44	45	42	43	44	38
9	1	41	43	32	41	35	32
	2	36	43	36	37	42	33
10	1	32	30	39	42	30	23
	2	35	42	43	41	41	27
11	1	41	40	30	38	41	31
	2	37	43	38	41	44	38
12	1	42	43	40	41	33	41
	2	42	36	36	38	44	39
13	1	19	23	21	30	26	23
	2	29	24	15	38	34	24
Mean	1	35.8	36.8	33.2	38.2	33.5	30.7
	2	37.2	38.8	35.0	39.7	41.5	33.2

aids, which usually give a high-frequency emphasis, the off-tape signals were passed through a spectrum shaping network that rolled off at 12 dB/octave below 200 Hz, was "flat" from 200 to 400 Hz, and rose smoothly to +2 dB at 600 Hz and +15 dB at 4 kHz. This form of spectral shaping is similar to that commonly used in commercial hearing aids.

The level of the replayed speech in noise was adjusted for each subject to the value that they found comfortable for everyday conversation in a domestic environment. Other aspects of the stimuli were the same as for experiment 1.

Subjects

Six subjects were tested, four of whom had been used in experiment 1. All were diagnosed as having bilateral sensorineural hearing loss, probably of cochlear origin. They are subjects 8-13 in Table 1. Most were experienced hearing aid users.

Experimental Design

A double Latin Square design was used. Each subject was tested once in each of the six conditions, with the order of testing of conditions counterbalanced across subjects. This was then repeated but with the order of testing "rotated" so that the order of conditions for a given subject was different for the two Latin Squares. In each Latin Square, one

ASL list was used for each subject and each condition.

Procedure

The procedure was essentially the same as for experiment 1. Subjects were given one practice list which was also used for adjusting the noise level.

Results

The raw scores are given in **Table 5**, which also shows the mean score across subjects for each condition for each of the two Latin Squares. Inspection of the data revealed a trend for performance to be better for the second Latin Square than for the first (i.e., there was a practice effect). Therefore, an ANOVA was conducted with factors condition and order of testing (first or second Latin Square) with the data blocked across subjects and lists (28). As for the data of experiment 1, the proportions correct were transformed using the expression arcsine($\sqrt{}$ proportion correct). The analysis revealed a significant effect of condition, $F(5,50) = 5.89$, $p < 0.001$, and order of testing $F(1,50) = 12.04$, $p = 0.001$. The interaction of condition and order of testing approached, but did not reach, significance, $F(5,50) = 1.79$, $p = 0.13$.

Considering the mean scores for both Latin Squares, the highest scores were obtained for condi-

tions ENH and C10/2 and the lowest for the conditions involving the greatest amounts of compression, C15/3 and ENHC15/3. *Post-hoc* tests, conducted as described earlier, showed that the mean scores for conditions ENH and C10/2 were significantly higher than the mean scores for conditions C15/3 and ENHC15/3 ($p < 0.01$ in all cases). The mean score for condition ENHC15/3 was also significantly lower than the mean score for the control condition, E0 ($p < 0.01$). Thus, a large amount of compression has deleterious effects. However, the scores for conditions E0, C10/2, ENH, and ENHC10/2 did not differ significantly from one another.

It seems reasonable to consider separately the scores for the second Latin Square, since there was evidence for improvements with practice. For the second Latin Square, the highest score overall was obtained for condition ENHC10/2, the condition involving both enhancement and a moderate degree of compression. The mean score for this condition (92 percent) was significantly greater than that for the control condition (82.6 percent) ($p < 0.05$). However, it was not significantly greater than the mean scores for conditions ENH (88 percent) and C10/2 (86.3 percent). The results of the second Latin Square for conditions E0, ENH, and ENHC10/2 are shown separately for each subject in **Figure 4**. For subject 8, the differences between conditions were limited by a ceiling effect; scores were close to perfect for all conditions. All of the other subjects scored better in condition ENHC10/2 than in condition E0.

In summary, the results of experiment 3 indicate that a large amount of compression, either used alone or in combination with spectral contrast enhancement, has deleterious effects on the intelligibility of speech in noise. The results showed clear effects of practice, suggesting that subjects may require time to get used to novel types of processing. The results for the second Latin Square (i.e., those obtained after a small amount of practice) indicated that the condition involving the combination of spectral contrast enhancement and a moderate amount of compression, ENHC10/2, gave a significantly higher mean score than the control condition.

Discussion

Although the results of experiment 3 suggest that the intelligibility of speech in noise may be
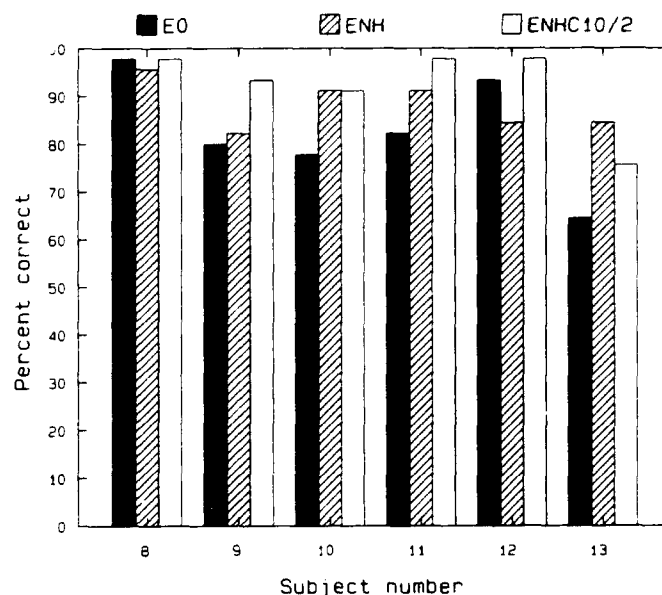
Fig re 4.
Results of Experiment 3 for the second Latin Square, for the control condition (E j, the condition involving enhancement (ENH), and the c.... tion involving enhancement combined with a moderate degree of compression (ENHC10/2).

improved by the enhancement of spectral contrasts, especially when combined with a moderate amount of compression, the effects were small. The small size of the effects probably arose partly from the lack of experience of the subjects with the processed stimuli; the results showed clear evidence of practice effects. This raises a dilemma. We wished to compare performance on several conditions, but we also wanted to avoid the possibility of subjects learning the sentence lists through repeated presentations. This latter requirement meant that it was not possible to give the subjects extensive practice on each condition.

A second factor that may have limited the size of the effects is related to the trade-off between accuracy and time/effort. Our subjects were effectively given as much time as they wanted to respond after each sentence had been presented. In difficult listening conditions, subjects may have devoted more effort and/or more time to the task of identifying each sentence. This would have resulted in reduced differences between conditions in comparison to the hypothetical situation where equal effort and/or time were devoted to all conditions.

It has previously been suggested that traditional speech intelligibility scores access only one component of disability and benefit, and that further information may be obtained by investigation of response times to speech stimuli (30,31,32,33). The response time aspects have previously been interpreted in terms of ease of listening. It may be argued that some or perhaps all of the benefits of spectral enhancement may accrue not from improvement of intelligibility, but rather from advantages to the listener in terms of the decreased difficulty (i.e., decreased effort required) in identifying the speech signal due to the sharper distinction of spectral cues. The availability of the sentence verification test, which yields measures of both speech intelligibility and response times, enabled this idea to be tested directly. A further advantage of this test is that, after an initial practice period, there is little evidence for improvements over time, and the materials themselves cannot be memorized. This made it possible to gather much more data for each subject and condition than in the earlier experiments.

EXPERIMENT 4

The Sentence Verification Test

The sentence verification test uses a closed vocabulary to construct four-word sentences from an overall vocabulary of 32 words. There are four alternatives for the first word in the sentence (LIZ, LYNNE, LEN, BEN), 12 alternatives for the second word (SOLD, SHOWED, STOLE, STORED, WORE, STITCHED, DROVE, CRASHED, CRACKED, CORKED, READ, TORE), 12 alternatives for the third word (FOUR, MORE, TWO, FEW, TWEED, CLOTH, FAST, SPORTS, GLASS, JAM, ROAD, STREET), and four alternatives for the fourth word (CAPS, CARS, JARS, MAPS). Of the 144 combinations of the second and third words, there are 82 for which there is at least one fourth word which makes the sentence unequivocally silly (nonsense) and at least one fourth word which makes the sentence unequivocally sensible (e.g., BEN SOLD STREET MAPS is sensible, while BEN SOLD STREET JARS is silly). Any combination of a fourth word with a second word–third word pair that may be considered equivocal with regard to sense/nonsense is not employed in the test. The eventual sentences require identification of the second, third, and fourth words in the sentence before a decision regarding the sense/nonsense of the sentence may be made.

The 32 words were stored as digitized waveform files which were isolated from sentences spoken by a single male talker and were concatenated to produce the desired sentences. During the construction of the test, care was taken to ensure that the intonation contours of the items, and other aspects, such as duration of voicing, were similar across items. This was done to remove extraneous cues not directly associated with the intelligibility of the individual word.

Following presentation of the sentence to the listener, the subject was asked to indicate whether the sentence was "silly" or "sensible" via a touch sensitive computer screen, and the response time for that decision (verification time) was recorded. This verification was followed by the identification component, for which four potential alternatives for the first word in the sentence, four for the second, four for the third, and four for the fourth were displayed on the touch sensitive computer screen. The subject was required to identify the components of the sentence. The test may be run either adaptively (yielding a signal-to-noise ratio for criterion performance) or at a fixed signal-to-noise ratio (yielding a percent correct score for the intelligibility component). The verification component of the test yields a median response time for all or a subset of the items for the cognitive decision concerning the sense/nonsense of the sentences. Evaluations of the within-session and between-session stability of the test for both normally hearing and hearing-impaired subjects has shown that there are no significant long-term learning effects associated with repeated administration of the closed vocabulary.

Processing of the Sentence Verification Test Items

Due to hardware constraints, the sentence verification test was available only in Glasgow. Hence, the stimuli to be processed were recorded in Glasgow, sent to Cambridge for processing, and then returned to Glasgow, using digital audio tape (DAT) as the recording medium. The 32 individual words constituting the vocabulary for the sentence verification test were each recorded at signal-to-noise ratios of 0, +3, +6, +9, and +12 dB, where the signal level was defined as the mean level of the speech peaks, and the noise level (shaped noise with

the same long-term spectrum as the single male speaker) was defined as the rms level. A 1,000 Hz sine wave was included at the beginning of the recording to provide a reference level. These recordings were then sent to Cambridge and processed as described earlier, using three of the processing conditions from experiment 3: the control condition, E0; the condition involving enhancement alone, ENH; and the condition involving both enhancement and a moderate degree of compression, ENHC10/2. The processed stimuli were subjected to the high-frequency emphasis described for experiment 3, before being recorded on DAT tape. Each condition was recorded on a separate tape. The tapes were then returned to Glasgow.

The 15 sets of the 32 words (three conditions by five signal-to-noise ratios) were each redigitized using a CED 1401 laboratory interface and stored as individual waveform files. These waveform files were concatenated during testing to produce the required sentences.

Test Conditions

Because the processing was done with the noise added to the speech, the test had to be administered at fixed signal-to-noise ratios. Each subject was tested both unaided and with the level of the speech-plus-noise adjusted to a comfortable value. According to the experimental design described below, the required condition and signal-to-noise ratio was identified and a total of 55 sentences were delivered to the subject via a Grason-Stadler GSI 16 Audiometer and a Goodmans B41 loudspeaker in a sound-treated room with the subject seated 2 m from the loudspeaker at 0° azimuth. The first five of these sentences were not scored, but were regarded as practice within each individual run. The remaining 50 sentences were used, giving a score out of 200 for the identification component of the test. For the verification component of the test, only those sentences that were correctly identified (each of the four constituent words in the sentence identified correctly) and verified (correctly labeled as being either silly or sensible) were used. The median of the response times for the verification process using this subset of sentences was then derived. Thus, each run of the sentence verification test yielded an identification score out of 200 (here expressed as percent correct) and a response time

(verification time) for the decision regarding the sense/nonsense of the sentence.

Subjects

The five subjects were all established users (at least 12 months) of a single post-aural BE10 series National Health Service hearing aid. The characteristics of the subjects are shown in **Table 1** (subjects 14-18). They all had broadly symmetric bilateral sensorineural losses of moderate degree, with greater losses at high frequencies than at low. All subjects had taken part in earlier experiments using the sentence verification test and were familiar with its form and configuration.

Experimental Design

The experiment consisted of five sessions for each subject, usually conducted at weekly intervals. Each session used seven complete runs of the sentence verification test as configured above. During each session, data were gathered for a pair of signal-to-noise ratios for each of the three processing conditions (E0, ENH, and ENHC10/2). An initial complete run for one of the signal-to-noise ratio/processing conditions was employed as practice, as previous experience with the sentence verification test suggested that optimal stability is achieved if this is done. The signal-to-noise ratios for each session were selected from a blocked design across subjects. Within each signal-to-noise ratio, the order of the three conditions was selected randomly. During the course of the five sessions, each subject was tested twice for each of the signal-to-noise ratios and each of the processing conditions.

Results

To show the overall form of the results, the mean of the two repetitions for each subject/signal-to-noise ratio/condition combination was taken and then the scores for the five subjects were averaged. The results are summarized in **Figure 5**. Error bars show 95 percent confidence limits. The figure shows the expected trend of increasing intelligibility and decreasing response times as the signal-to-noise ratio increases. For the identification component, there appear to be modest but consistent advantages at most signal-to-noise ratios for both of the processed conditions over the control condition (E0). For the
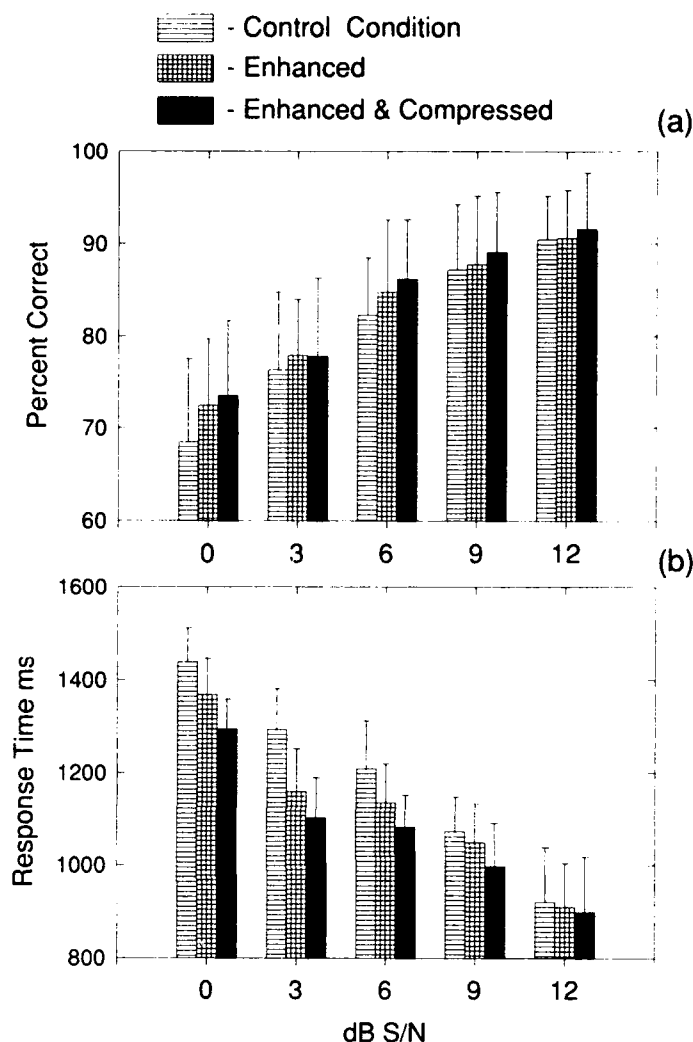
Figure 5.

Results of Experiment 4, showing the mean (and 95% confidence intervals for the mean) as a function of original signal-to-noise ratio and processing condition. Panel *(a)* shows scores for the identification component of the sentence verification test in terms of percent correct. Panel *(b)* shows the verification component in terms of response time.

response time component, the advantages of the processing conditions are larger, relative to the confidence limits, and there is a clear tendency for the processing condition involving both enhancement and compression (ENHC10/2) to give shorter response times than the condition involving enhancement alone (ENH).

The results of the five subjects were subjected to a repeated-measures ANOVA, using the GENSTAT package, with the following dependent variables: (i) percent correct score; (ii) arcsine

($\sqrt{}$ proportion correct)—this measure makes the scores follow a normal distribution more closely; (iii) response time; and, (iv) square root of response time—again, this measure makes the scores follow a normal distribution more closely.

The results for the transformed variables (ii) and (iv) were similar to those for the untransformed variables (i) and (iii), so the latter will be presented to facilitate interpretation. In the ANOVA, there were three within-subject factors. The independent variables were: (i) the signal-to-noise ratio (0, 3, 6, 9, and 12 dB); (ii) the condition (linear, enhanced, enhanced and compressed); and, (iii) replication (first and second replicate).

For the percent correct scores, there was a highly significant effect of signal-to-noise ratio [$F(4,16) = 97.1$, $p < 0.001$], as expected from **Figure 5**, and a significant effect of condition [$F(2,8) = 6.65$, $p < 0.02$]. The main effect of replicate was not significant, and none of the interactions was significant. The mean score for condition ENH was 1.76 percent greater than that for condition E0 (standard error $= 0.54$), and this difference was statistically significant ($p < 0.02$). The mean score for condition ENHC10/2 was 2.73 percent greater than that for condition E0, and again this difference was significant ($p < 0.001$). The mean difference between conditions ENH and ENHC10/2, 0.97 percent, was not significant.

The ANOVA for the response time component of the sentence verification test showed highly significant effects of signal-to-noise ratio [$F(4,16) = 333.6$, $p < 0.001$] and of condition [$F(2,8) = 31.4$, $p < 0.001$]. The main effect of replicate was not significant, but there was a significant interaction between signal-to-noise ratio and processing condition [$F(8,32) = 4.07$, $p < 0.002$], consistent with the greater effect of condition at low signal-to-noise ratios apparent in **Figure 5**. The mean response time for condition ENH was 62.8 ms less than that for condition E0 (standard error $= 10.1$ ms), and this difference was statistically significant ($p < 0.001$). The mean response time for condition ENHC10/2 was 113 ms less than that for condition E0, and again this difference was significant ($p < 0.001$). The mean difference between conditions ENH and ENHC10/2, 52.8 ms, was also significant ($p < 0.001$). Thus, the results show that there are significant advantages for the processed conditions

compared with the control condition, with combined enhancement and compression giving bigger advantages than enhancement alone. The advantages are statistically more robust for the response-time component of the test than for the identification component.

The magnitudes of the effects described above, especially the response times, are difficult to interpret because of the somewhat complex nature of the sentence verification test. One way of relating the effects to other, more familiar measures, is to convert the differences in percent correct scores or response times to equivalent changes in signal-to-masker ratio. The data in **Figure 5** indicate that both the percent correct scores and the response times are approximately linearly related to the signal-to-noise ratio, for signal-to-noise ratios between 0 and +6 dB. For the control condition, each 1-dB increment in signal-to-noise ratio produces a 2.3 percent change in the percent correct score and a 38.3-ms change in the response time. These relationships were used to transform the magnitudes of the differences between conditions into equivalent changes in signal-to-noise ratio in dB.

For the percent correct scores, the difference between conditions E0 and ENH was equivalent to a 0.8-dB change in signal-to-noise ratio, while the difference between conditions E0 and ENHC10/2 was equivalent to 1.2 dB. For the response times, the difference between conditions E0 and ENH was equivalent to a 1.6-dB change in signal-to-noise

ratio, while the difference between conditions E0 and ENHC10/2 was equivalent to 3.0 dB. Thus, the benefits of processing are approximately twice as large for the response-time component as for the identification component. If percent correct and response time can be regarded as subcomponents of an overall benefit from processing, then condition ENH gave an overall benefit of 2.4 dB compared with the control condition, and condition ENHC10/2 gave an overall advantage of 4.2 dB compared with the control condition.

The fully factorial, repeated-measures nature of the experimental design enabled individual differences to be investigated. For the identification scores, a general linear model (GLIM) analysis was conducted based on a logistic model, assuming that errors were distributed according to a binomial distribution. Here the proportion correct (PrC) is the dependent variable in an equation of the form:

$$\text{PrC} = 1/(1 + \exp(-(B0 + B1*X1 + B2*X2 + ...))) \quad [6]$$

where X1, X2, etc. are indices for specific values of the independent variables (signal-to-noise ratios, and processing conditions) and their interactions. The procedure produced estimates of the values of the parameters, B0, B1, B2, etc., referenced to a specific baseline, namely the mean for the control condition at 0 dB signal-to-noise ratio. B0 is the parameter estimate for the baseline itself. The effect of replicate was not significant and is not included in the analysis. The results are summarized in **Table 6**.

Table 6.
Summary of the results of the logistic regression analysis (GLIM) of the identification scores.

| Subject | Baseline | S/N ratio | | | | Condition | | Interaction |
		3 dB	6 dB	9 dB	12 dB	ENH	ENHC10/2	
14	0.315 (0.078)	0.282 (0.085)	0.691 (0.089)	0.836 (0.091)	1.234 (0.098)	0.169 (0.072)	0.182 (0.072)	N.S.
15	0.674 (0.083)	0.270 (0.089)	0.855 (0.098)	1.081 (0.103)	1.209 (0.106)	0.186 (0.080)	0.064 (0.078)	N.S.
16	0.816 (0.087)	0.442 (0.096)	0.627 (0.099)	1.157 (0.112)	1.549 (0.125)	−0.073 (0.084)	0.229 (0.088)	N.S.
17	1.050 (0.095)	0.327 (0.101)	1.001 (0.117)	1.260 (0.125)	2.041 (0.164)	0.116 (0.094)	0.277 (0.097)	$p < .01$
18	1.370 (0.104)	0.254 (0.112)	0.906 (0.130)	1.632 (0.164)	1.385 (0.151)	0.258 (0.105)	0.359 (0.108)	N.S.

Parameter estimates (and associated standard errors) are referenced to a baseline, namely the control condition with 0-dB signal-to-noise ratio. A parameter estimate for the baseline itself is given and an indication of the significance of the interaction of signal-to-noise ratio with condition.

To return to a percent correct score from a parameter estimate, the equation

$$\text{Percent correct} = 100/(1 + e^{-(\text{sum of estimates})})\ [7]$$

is used. Thus, for subject 14 the figure of 0.315 for the baseline (control condition at 0 dB signal-to-noise) is equivalent to a score of 57.8 percent. Using the properties of the logistic regression, the effect of a combination of factors may be assessed by simple addition of the parameter estimates. Thus the estimate associated with a signal-to-noise ratio of 3 dB in the control condition is 0.315 + 0.282 = 0.597 (equivalent to a percent correct score of 64.5 percent) while the estimate associated with a signal-to-noise ratio of 3 dB in the enhanced condition for subject 14 is 0.315 + 0.282 + 0.169 = 0.766 (equivalent to a percent correct score of 68.3 percent).

The data in **Table 6** suggest that the processing conditions give different effects for different subjects. For example, subject 16 showed no benefit for enhancement alone, but showed a clear benefit for enhancement with compression. In contrast, subject 15 showed a clear benefit for enhancement alone, and showed less benefit for enhancement with compression. Subjects 14, 17, and 18 showed some benefit from enhancement alone, and showed larger benefits from enhancement with compression, although the differences between conditions ENH and ENHC10/2 were not significant. The interaction of signal-to-noise ratio with processing condition was significant only for subject 17.

The results for the response time estimates were analyzed using an identical linear model but assuming that errors were normally distributed. The results for the five subjects are shown in **Table 7**. The pattern was similar to that for **Table 6**, though now there was a significant interaction between signal-to-noise ratio and condition for subjects 16 and 18. Overall, the effects of condition were more robust (as can be seen by comparing the parameter estimates with their associated standard errors). For subjects 14 and 15, the benefit of condition ENH was not significant, but the benefit of condition ENHC10/2 was significant. For subjects 16, 17, and 18, there were significant benefits in both conditions. The benefits tended to be larger in condition ENHC10/2 than in condition ENH, but the differences were not significant. It is noteworthy that the identification scores of subject 16 did not show a benefit for condition ENH, whereas the response-time scores did.

Although the experiment contained relatively small numbers of subjects, the nonhomogeneous pattern of results does suggest that, in future experiments, it would be worthwhile to investigate further the characteristics of individual subjects to try to find the predictors of benefit from enhancement.

GENERAL SUMMARY, DISCUSSION, AND CONCLUSIONS

The results of experiment 1 were disappointing, in that they failed to show any significant benefits of

Table 7.
Summary of the results of the logistic regression analysis (GLIM) for the response times.

| Subject | Baseline | S/N ratio | | | | Condition | | Interaction |
		3 dB	6 dB	9 dB	12 dB	ENH	ENHC10/2	
14	1357 (29.8)	− 212 (33.3)	− 248 (33.3)	− 350 (33.3)	− 513 (33.3)	− 37 (25.8)	− 111 (25.8)	N.S.
15	1469 (25.8)	− 160 (28.8)	− 213 (28.8)	− 293 (28.8)	− 433 (28.8)	− 20 (22.3)	− 85 (22.3)	N.S.
16	1383 (29.8)	− 160 (33.3)	− 220 (33.3)	− 310 (33.3)	− 403 (33.3)	− 71 (25.8)	− 83 (25.8)	p < .01
17	1535 (29.3)	− 165 (32.7)	− 198 (32.7)	− 343 (32.7)	− 432 (32.7)	− 117 (25.4)	− 165 (25.4)	N.S.
18	1378 (29.3)	− 217 (32.8)	− 248 (32.8)	− 355 (32.8)	− 512 (32.8)	− 69 (25.4)	− 121 (25.4)	p < .01

the enhancement processing, although the results did indicate that a large degree of spectral enhancement can have deleterious effects. In hindsight, the failure to find positive effects of the processing in experiment 1 probably can be attributed in part to the experimental design, which did not give subjects the opportunity to practice in the different conditions.

The results of experiment 2 showed that subjective ratings of both quality and intelligibility were affected by the processing, but the effects varied across subjects. For judgments of both quality and intelligibility, speech in noise processed using a moderate degree of enhancement was generally preferred over the control condition. The results for higher degrees of enhancement varied across subjects. Subject 11 preferred the highest degree of enhancement both for quality and intelligibility. For several other subjects, quality decreased for the highest degree of enhancement.

The results of experiment 3 indicated that a large amount of compression, either used alone or in combination with spectral contrast enhancement, had deleterious effects on the intelligibility of speech in noise. The results showed clear effects of practice, suggesting that subjects may require time to get used to novel types of processing. The results for the second Latin Square (i.e., those obtained after a small amount of practice) indicated that the condition involving the combination of spectral contrast enhancement and a moderate amount of compression, ENHC10/2, gave a significantly higher mean score than the control condition.

Taken together, the results of experiments 1, 2, and 3 indicate that high degrees of enhancement, or high degrees of compression, generally have deleterious effects. In other words, too much processing is a bad thing! However, the results of experiment 2 indicate that a moderate amount of spectral enhancement can lead to improved subjective ratings of quality and intelligibility, and the results of experiment 3 indicate that, after some practice, a moderate degree of spectral enhancement, combined with a moderate degree of compression, can give better results than those obtained with unprocessed speech.

Experiments 1 and 3 suffered from the problem that subjects were given rather little practice in each condition. This was forced upon us because, with the limited number of sentence lists available to us, it would not have been possible to give extensive practice without subjects memorizing the lists. The limited number of sentence lists created a second problem; it was impossible to gather a large amount of data for each condition. This meant that some of the effects observed were of marginal statistical significance. A third problem was that the measure used, the percent correct of words identified in short sentences, may not have been suitable for revealing all of the effects of the processing. Specifically, the measure probably did not tap the dimension of "ease of listening," which can be especially important in everyday situations involving decision making and selective attention.

The Sentence Verification Test used in experiment 4 was intended to overcome these problems. The test can be administered repeatedly without substantial learning effects, and it includes a measure of response time which is probably related to ease of listening. The results showed highly significant benefits of the processing, with spectral enhancement alone being superior to the control condition, and enhancement combined with compression being superior to enhancement alone. When expressed in terms of equivalent changes in signal-to-masker ratio, the benefits were about twice as great for the response time measures as for the identification scores, and they were also statistically more robust for the response time measures. This suggests that the major benefits of the processing may be in terms of increased ease of listening rather than in intelligibility.

The results of experiment 4 indicate that the improvement in the intelligibility score produced by processing alone was equivalent to a change in signal-to-noise ratio of about 0.8 dB, a relatively modest amount. The results of Simpson, et al. (19) for spectral processing with a similar degree of enhancement (although implemented using a somewhat different algorithm), showed typical improvements in intelligibility, relative to the control condition of about 7 percent. For the speech materials used by Simpson, et al., each 1-dB change in speech-to-noise ratio produces about an 11 percent change in intelligibility (34). Thus, the 7 percent change in intelligibility is equivalent to about a 0.6-dB change in speech-to-noise ratio. This is comparable to the 0.8-dB change found in experiment 4.

It should be emphasized that the overall effect of the processing found in experiment 4 was larger

than this. If the changes in intelligibility and in response times were both expressed in terms of equivalent change in speech-to-noise ratio, the net effect was an improvement (relative to the control condition) of 2.4 dB for enhancement alone, and 4.2 dB for enhancement combined with compression.

Experiments 1 and 3 used a Latin Square design, which makes it difficult to analyze the effects of individual differences. However, the results of experiment 2 showed clear evidence of individual differences in the judged pleasantness and intelligibility of the processed stimuli. Similarly, both the intelligibility measures and the response time measures of experiment 4 revealed clear individual differences. Further research is needed to clarify why these differences occur, and to establish whether they can be related to individual differences in psychoacoustic factors such as frequency selectivity.

ACKNOWLEDGMENTS

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Development and assessment of two fixed-array microphones for use with hearing aids

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Abstract—Hearing-impaired listeners often have great difficulty understanding speech in situations with background noise (e.g., meetings, parties). Conventional hearing aids offer insufficient directivity to significantly reduce background noise relative to the desired speech signal. Based on array techniques, microphone prototypes have been developed with strongly directional characteristics to be incorporated into the frame and the "temples" of a pair of eyeglasses. Particular emphasis was on optimization and electronic stability. Computer simulations show that a directivity index of more than 10 dB can be obtained at the higher frequencies. Simulations were verified with free-field measurements. To investigate the influence of the human head on directivity, two portable models were also tested with a KEMAR manikin. The measurements show that the two models give an improvement of the signal-to-noise ratio of approximately 7 dB in a diffuse background noise field compared with an omnidirectional microphone. For the clinical assessment of these microphone arrays in the diffuse noise field (simulating a cocktail party situation), the speech-reception threshold in noise for simple Dutch sentences was determined with a normal single omnidirectional microphone and with one of the microphone arrays. The results of monaural listening tests of 30 subjects with normal hearing and 45 subjects with hearing impairment show that the microphone arrays give a mean improvement of the speech reception threshold in noise of about 7 dB compared with an omnidirectional microphone.

Key words: *background noise, KEMAR manikin, microphone arrays, microphone eyeglass prototypes, omni-*

directional microphone, signal-to-noise ratios, speech reception.

INTRODUCTION

Many people have great difficulty understanding speech in surroundings with background noise and/or reverberation. This is especially a problem for the increasing number of elderly people and people with sensorineural impairment. Several investigations of speech intelligibility in noisy situations have demonstrated that subjects with sensorineural hearing loss may need a 5–15 dB higher signal-to-noise ratio than subjects with normal hearing (1). Every 4–5 dB improvement of the signal-to-noise ratio may raise the speech intelligibility by about 50 percent (2,3,4).

A directional hearing aid may reduce background noise relative to the desired speech signal. Until now, directional hearing aids consisted of a conventional hearing aid with a single cardioid microphone. Although Mueller (5) and Hillman (6) published studies showing a preference for a hearing aid with a directional cardioid microphone, the directional hearing aid has not yet enjoyed the widespread clinical acceptance that would be expected on theoretical grounds. In practice, the reduction of background noise with a cardioid microphone is not yet sufficient because the low directivity for higher frequencies (2,000–5,000 Hz) permits a maximum improvement of only about 2 dB (7,8).

Different solutions to further improve the directivity seem plausible on theoretical grounds. In *adaptive processing,* the processing of the signals from two or more microphones is continuously adjusted according to properties of the received sound signals and controlled by an adaptation mechanism that can be implemented on a signal processor (8,9,10,11).

In *fixed array processing,* a fixed configuration of a number of microphones offers the possibility of suppression of background noise while the desired speech signal in front of the user is transmitted undistorted. A high and robust directivity can be obtained when the length of the array is larger than the wavelength; the signal processing can be done with relatively simple analogue electronics. A desired sound source can be chosen by moving the principle direction of the array toward the source. A cosmetic disadvantage of a fixed array technique might be the array length. As a practical compromise, it was envisaged that the microphone array should be connected to, or built into, a pair of eyeglasses and should be used in combination with a conventional hearing aid. Therefore, the maximum array length is determined by the length of the eyeglass "temples" (i.e., the pieces that extend from the frames alongside the head and around the ears) or the width of the frame, which is approximately 10 cm and 14 cm, respectively.

The *directivity index* (DI) was accepted as a measure to differentiate between possible solutions. It was decided to optimize the DI within a frequency range of 500 Hz to 4,000 Hz. The shape of the directivity pattern was considered to be of secondary importance. Further, the new directional microphone is meant to be used monaurally. Profits of binaural fitting should be added by simply using two devices.

In the following sections, the results of computer simulations and measurements on different array configurations, optimization and stability of different array configurations, and listening tests with normal-hearing and hearing-impaired subjects will be summarized (12).

METHODS AND DISCUSSION

Simulations on Broadside and Endfire Array

For an application with microphones mounted on a pair of eyeglasses, we distinguish between two important groups of linear arrays characterized by the position of the microphones, viz. broadside arrays and endfire arrays.

In a *broadside array* the microphones are placed along the x-axis (alongside each other). The directivity pattern can be shown (13) to be given by

$$Q(\theta,\phi,\omega) = \sum_n D_n(\omega)A_n(\omega)e^{+jk_x n\Delta x} \qquad [1]$$

with θ and ϕ the angles of incidence, ω the frequency, n the microphone number, $D_n(\omega)$ the frequency-dependent directivity characteristic of the individual microphones, $A_n(\omega)$ the amplitude weighting of individual microphones, k_x equal to $k \cos \phi \sin \theta$ (k is the wave number), and Δx the distance between the microphones.

In an *endfire array* the microphones are placed along the z-axis (behind each other) and the phase correction should be $\tau_n = n\Delta z/c$ (c is sound velocity). Now, the directivity pattern is given by

$$Q(\theta,\phi,\omega) = \sum_n D_n(\omega)A_n(\omega)e^{-j\omega\tau_n} e^{jk_z n\Delta z} \qquad [2]$$

with $k_z = k \cos \theta$.

Considering the situation with the desired sound coming from the main direction of the array and the background noise distributed equally over all other directions, the directivity index $DI(\omega)$ is a proper measure to indicate the average attenuation of the background noise with respect to sound coming from the main direction (14); it is given by

$$DI(\omega) = 10 \log \frac{4\pi|Q(\theta,\phi,\omega)_{max}|^2}{\int_0^{2\pi}\int_0^{\pi}|Q(\theta,\phi,\omega)|^2 \sin \theta \, d\theta \, d\phi} \qquad [3]$$

For a broadside array of five microphones with total length $L = n\Delta x = 10$ cm assuming omnidirectional microphones [$D_n(\theta,\phi,\omega) = 1$] and uniform amplitude weighting ($A_n = 1$), the DI equals 4.9 dB at 4,000 Hz. An endfire array with similar parameters will have a DI = 7.6 dB.

A comparison of equations for the beam width of the broadside and endfire array shows that in the x-z plane the main beam of a broadside array is always narrower than that of an endfire array of the same size. Therefore, the use of a broadside array may be advantageous when a small beam width is wanted in one plane. However, the computed directivity indices show that an endfire array of

omnidirectional microphones is advantageous for diffuse noise suppression.

Amplitude weighting $[A_n(\omega)]$ of each microphone signal is equivalent with the application of a window function. The uniform weighting gives a small main lobe with relative high side lobe levels. A concave upward weighting results in a narrower beam width at the expense of having higher side lobe levels. The opposite effect can be obtained with a Cosine, Hanning or Dolph-Chebyshev window function. They reduce the side lobe levels at the cost of a broader main lobe and a lower DI. The broader main lobe is a result of the low amplitude weighting at both ends of the array, giving an array with a relatively shorter effective length. However, the uniform weighting and the concave upward weighting have the highest DI. Finally, it must be noted that the weighting functions can also be applied to an endfire array. The amplitude weighting is independent of the phase correction, but both can be used to shape the directivity pattern (12).

Using cardioid microphones $[D_n(\theta,\phi,\omega) \neq 1]$ can be very useful in array design to improve the directivity for the lower frequencies ($\lambda > L$), to suppress side lobes and/or unwanted main lobes. This is especially advantageous for low frequencies and for suppressing the backward lobe of the broadside array.

Figure 1 gives the directivity pattern and the directivity index at 4,000 Hz for both array configurations with five cardioid microphones in a free-field situation, for a broadside array with L = 14 cm and an endfire array with L = 10 cm. The cardioid microphones give a significant improvement of the DI at the lower frequencies. We may conclude that with one endfire array (L = 10 cm) or one broadside array (L = 14 cm) of cardioid microphones, a DI can be reached of at least 5 dB at the lower frequencies rising to more than 10 dB at 4 kHz.

Combinations can be made of one endfire and one broadside array, two endfire arrays with intermediate distance L = 14 cm, and a full configuration of two endfire arrays and one broadside array having the shape of one pair of eyeglasses. The directivity of the combined configurations was computed for a free-field situation and a simple summation of the (delayed) microphone signals giving one output signal (mono).The combined array configurations (mono, free-field) give an extra improvement of the DI between 2 and 3 dB.
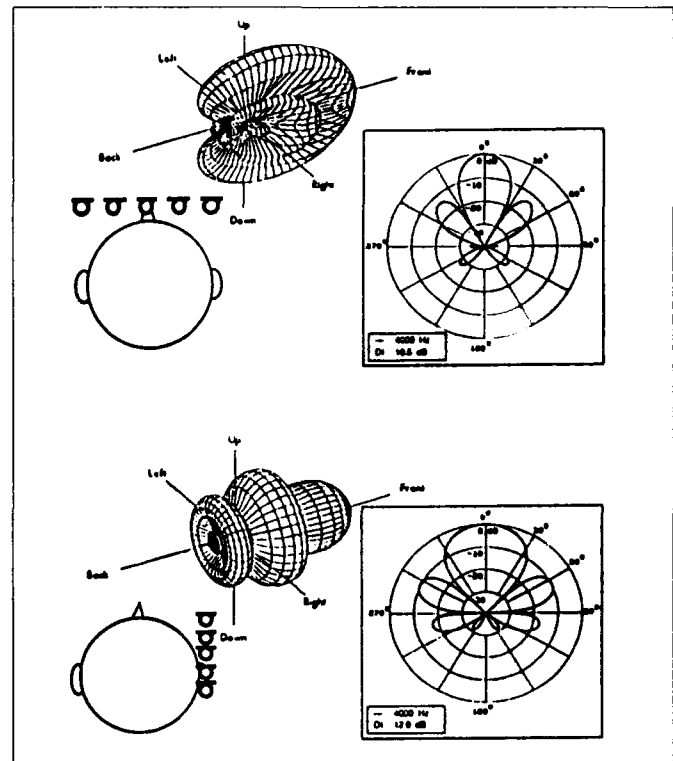
Figure 1.
Schematic representation of a broadside array and an endfire array, together with the 3-D directivity patterns and the pattern in the horizontal plane at 4,000 Hz, calculated for the case of five cardioid microphones and optimized array parameters.

A Numerical Approach to Optimization and Stability

For an array configuration to be optimally effective in the improvement of speech intelligibility in noise, it should have optimal directional characteristics. But an array configuration with a high directivity may not necessarily be a stable one. The high value of the DI may be reached for a special theoretical parameter choice that can be created in a laboratory situation but cannot be maintained in practice. Therefore, in a practical situation it is important to choose a stable array solution with a sufficient directivity, which is minimally influenced by intrinsic variabilities of microphones (amplitude and phase characteristics), amplitude weighting (amplifiers and resistor values), and/or time lag correction (delay elements).

Soede (12) used a comprehensive quasi-Newtonian algorithm (15) to study endfire, broadside, and Jacobi arrays. The algorithm searches for an unconstrained maximum of a function vector F

of parameters (represented by parameter vector x), where no mathematical derivatives of the function are required. The variables can be subjected to fixed lower and/or upper bounds. In this application the function vector F was defined by the equation for the DI. The optimization process was executed for single frequencies (i.e., 1,000, 2,000, 3,000, and 4,000 Hz) with respect to the amplitude weighting and the time lag correction of each cardioid microphone. The optimization processes showed that optimization of the parameter set at 4,000 Hz was sufficient for this application, giving a high DI for the lower frequencies too.

For the endfire array, optimization was done with a fixed overall length of 10 cm and for a changing number of microphones ranging from 2 to 17 with optimal time lags according to Hansen and Woodyard (16). It turned out that the DI at 4 kHz improves by 4 dB when five or more microphones are used instead of one. Using six or more microphones gives no further improvement. With respect to stability, it was shown that the influence of variations in amplitude weighting and delay times on the value of the DI of an optimized endfire array with five microphones is small. Therefore, it was concluded that an optimized endfire array with five microphones offers a stable and practical solution.

For the broadside array, optimization can be performed with respect to the position of the microphones as well as their amplitude weighting. Regarding the first, Ma (17) showed that the directivity of an array with variable microphone spacing will, at its optimum, be only a fraction of a decibel higher than an array with equidistant microphone spacing. Therefore, Soede (12) only paid attention to amplitude weighting with equidistant microphones. The optimization was done for a broadside array with a fixed width of 14 cm consisting of four, five, and six microphones. It was found that the profit of the optimization is very small. The directivity is mainly determined by the length of the array in relation to the wave length. Also, the spectrum of the optimized broadside array is hardly influenced by the optimization and is equal to the spectrum of one single cardioid microphone. A comparison of DIs for optimized configurations is presented in **Figure 2.**

With respect to stability, the influence of variations in the amplitude weighting appears to be very small as well. A difference of 0.2 dB between
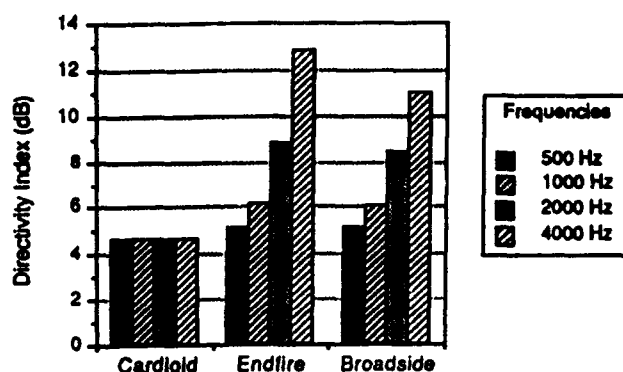
Figure 2.
The directivity index calculated for a cardioid microphone, the optimized broadside array and the optimized endfire array consisting of five cardioid microphones, at four different frequencies.

the DI of the uniform broadside array and the optimized array was reached by a variation in the amplitude weighting of more than 50 percent. Thus, variations with respect to sensitivity between the microphones used are negligible. Variations with respect to phase characteristics, on the other hand, reduce the directivity for a frequency of 4,000 Hz, but not below 6 dB. In summary, up to 2,000 Hz a broadside array appears very stable with respect to variations in individual microphone components— for further details see (12).

Directivity Measurements in Free Field and with KEMAR

For an assessment of the microphone arrays, a laboratory model of an endfire array and a broadside array was built (12). Because it was expected that the directivity of an array might be influenced by reflections and diffractions at the head, measurements were carried out in an anechoic chamber with the models placed in free-field conditions as well as in combination with an artificial head (KEMAR).

The laboratory model consisted of directional electret microphones (MICROTEL 61) with tube extensions to obtain a cardioid directivity pattern. The microphones were connected to movable little sockets. In the endfire configuration, each microphone is placed with its maximum sensitivity to $\theta = 0°$ (along the bar). The signal of each microphone is delayed relative to the first microphone signal using Panasonic MN3012 bucket-delays. The delay time of each microphone could be varied

independently of the other delays by variation of the clock frequency of the bucket-delay. Amplitude weighting is done with adjustable amplifiers. For the broadside array, the same laboratory model with delay times set to zero was used. Each microphone was placed in the broadside configuration with its maximum sensitivity to $\theta = 0°$ (perpendicular to the bar).

Directivity patterns were measured monochromatically in an anechoic chamber ($V = 1,000$ m³) with the microphone array mounted on a turntable and with a loudspeaker at a distance of 6.4 m. Data acquisition was carried out with a PC-controlled measurement system developed in-house. For practical reasons, the directivity patterns were measured in the horizontal plane only. For the endfire array, the DI was computed from the measured directivity pattern assuming the main beam at 0° and a cylinder symmetry along that main beam. For the broadside array, the DI was computed from the horizontal directivity pattern and corrected for the cardioid-like directivity pattern in the vertical plane.
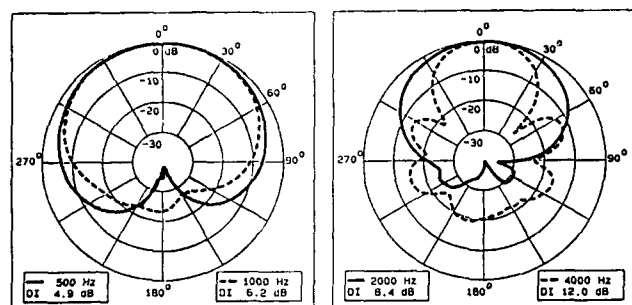
a. Free field measurement
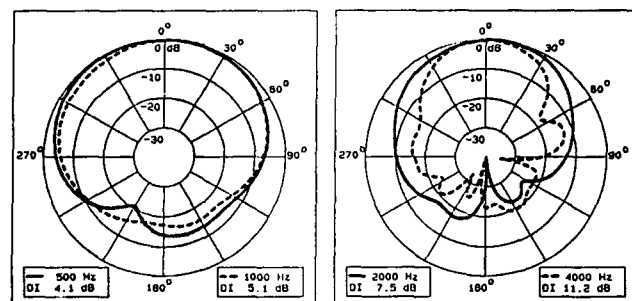
b. Measurement with KEMAR

Figure 3.
Measured polar diagrams of an optimized endfire array consisting of five cardioid microphones and an overall length of 10 cm, in free-field (a) and with KEMAR (b).

For an optimized endfire array with five cardioid microphones and a length of 10 cm, directivity patterns (polar diagrams) are presented in **Figure 3** for free-field conditions (a) and with KEMAR (b). Directivity indices were computed from these measurements with the restrictions mentioned in the previous paragraph. A comparison of the free-field directivity patterns and those measured with KEMAR show that, especially when the sound is coming from the right side of KEMAR, the influence of the head on the performance of the array is relatively small. The reduction of the estimated DI is less than 1 dB for all frequencies. The main beam is in the direction of 0° for all frequencies. Apparently, the directivity of the endfire array is hardly decreased by the addition of reflections or diffraction of the sound by the head.

For a broadside array with five cardioid microphones and total width of 14 cm, directivity patterns are given in **Figure 4** for free field conditions (a) and with KEMAR (b). A comparison shows that the influence of the head on the directivity patterns is again very small and is even advantageous for suppression of the backside lobes. The decrease in the DI at 500 Hz is less than 0.5 dB.

In summary, the estimated values of the DI computed from the polar patterns measured with the KEMAR manikin show that one cardioid microphone may give a mean improvement of 4 dB in comparison with one omnidirectional microphone. The estimated DI of the optimal endfire microphone array as well as the broadside microphone array ranges from 4 dB at 500 Hz to more than 10 dB at 4,000 Hz.

Directivity Measurements in an Artificial Diffuse Noise Field

Based on the results summarized in the previous sections, two portable array models were built that were suitable for psychophysical assessment with hearing impaired listeners: a portable endfire microphone array with a total length of 10 cm, and a portable broadside microphone array with a total width of 14 cm mounted on a pair of eyeglasses (12). Because cardioid microphones have a spectrum rising with 6 dB/octave, a correction filter for flattening was applied. The correction filter was designed in such a way that the spectrum in the front direction of both arrays was flat within ± 2 dB between 500 and 4,000 Hz.
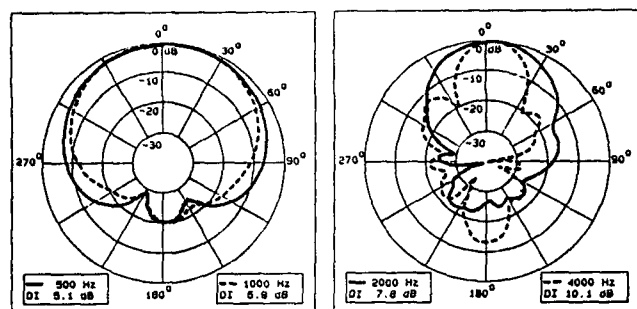
a. Free-field measurement
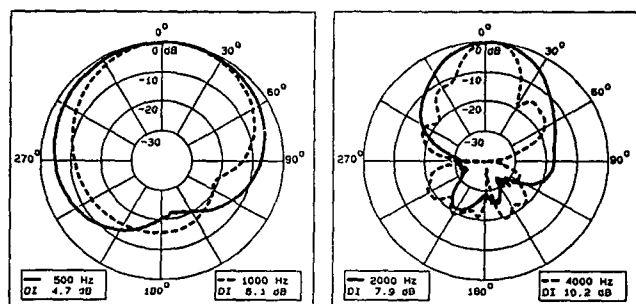
b. Measurement with array in front of the head of KEMAR

Figure 4.
Measured polar diagrams of an optimized broadside array consisting of five cardioid microphones and an overall width of 14 cm, in free-field (*a*) and with KEMAR (*b*).

An artificial diffuse sound field mimicking a cocktail party situation was realized with eight small loudspeakers positioned at the boundaries of an imaginary rectangular box ($2.0 \times 2.0 \times 1.70$ m) inside a sound-insulated booth (at the ENT department of the University Hospital, Rotterdam). Four loudspeakers were placed near the ceiling of the room at the edges of the rectangular box and the other ones were placed vertically (height 40 cm) at the corners of the cube. The eight loudspeakers were fed with eight independent noise sources, producing a spectrum equal to the long-time-average spectrum of speech. One loudspeaker was positioned at a height of 1.25 m in front of the listener (seated near the center of the box) and simulated the partner in a discussion during the listening test. The sound levels of the speech and the noise field could be varied

with an audiometer and an 8-channel attenuator developed in-house, respectively; both variables were controlled by a personal computer—for further details, see (12).

A KEMAR manikin was placed in the center of the experimental set-up facing the front loudspeaker (distance 1 m), and measurements were carried out with two behind-the-ear hearing aids, one with an omnidirectional microphone and one with a cardioid microphone, and then with the portable broadside and endfire microphone arrays. The hearing aids were connected to the right ear of KEMAR with an earmold (libby-horn with foam plug). Signals of the microphone arrays were measured via the behind-the-ear hearing aid using an induction-loop and the induction coil of the hearing aid (this equals the listening test conditions with the real subjects). With this set-up the attenuation of the diffuse noise field relative to the noise coming from the front direction was measured in one-third-octave bands.

The results of the measurements are reproduced in **Figure 5**. The attenuation is given as a function of the one-third-octave center frequency (400–5,000 Hz). The mean level is computed from the one-third-octave band levels with equal weights. The measurement with the hearing aid containing a normal omnidirectional microphone shows that the diffuse sound field is not attenuated. The mean value of -1 dB means an amplification of the diffuse sound field relative to the signal coming from the front. The hearing aid with one cardioid microphone attenuates the diffuse sound field for the lower frequencies with a mean of $+1.5$ dB, indicating an improvement of $+2.5$ dB compared with the omnidirectional hearing aid. This corresponds with everyday experience (5,6).

The measurements with the broadside and endfire microphone arrays show a strong attenuation, especially for the high frequencies, with mean values of $+6.0$ dB and $+5.8$ dB, respectively, thus indicating an improvement of 7.0 dB (broadside) and 6.8 dB (endfire) compared with the omnidirectional hearing aid. These results are about 1 dB lower in comparison with the DIs estimated from the KEMAR directivity patterns. This difference can be explained by a contribution of the sound of the front loudspeaker to the diffuse noise field due to (not negligible) reverberation in the soundproof booth.
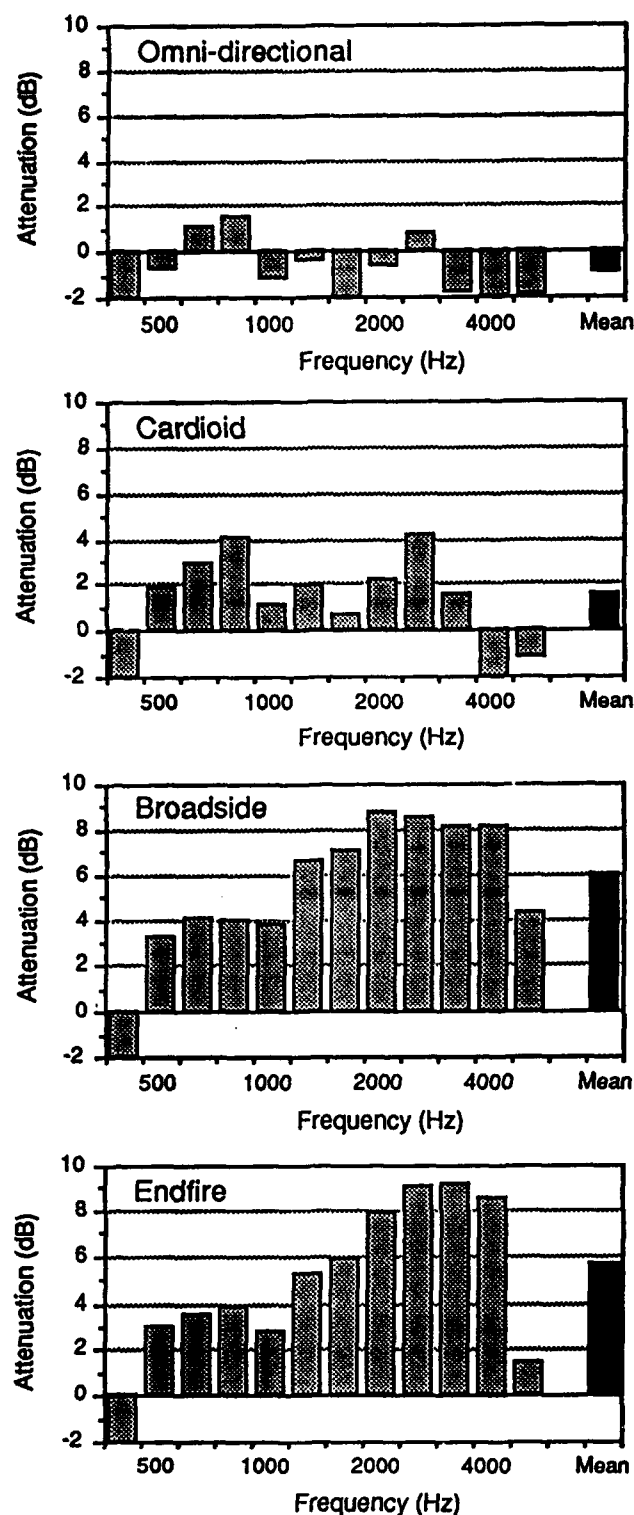
Psychophysical Assessment in a Cocktail-Party Simulation

Speech intelligibility with the arrays in a cocktail-party situation was determined using the artificial diffuse noise field as briefly described in the previous section. Using a simple up-down procedure, the 50 percent intelligibility level, the so-called speech reception threshold (SRT) was determined. The difference between the SRT in noise and the noise level was defined as the *critical speech-to-noise ratio* (S/N ratio). The speech material consisted of 10 lists of 13 short Dutch sentences, representative of everyday conversation (4). For the simulation of the background noise at a typical cocktail party, eight independent "speech noise" signals were used having a spectrum equal to the long-term-average spectrum of the sentences.

The monaural listening tests were carried out with 30 subjects with normal hearing and 45 subjects with hearing impairment. The group of 30 normal-hearing listeners, equally divided as to males and females, were mainly physics and medicine students with ages ranging from 19 to 37 years with a median age of 26.4 years. The hearing-impaired group consisted of 23 male and 22 female subjects, aged 36–90 years, with a median age of 68.8. The hearing-impaired listeners were asked for their cooperation while visiting the ENT department of the Rotterdam University Hospital for hearing-aid inspection. Cooperation was requested after the hearing-aid fitting received its final approval and when a discrimination score of at least 80 percent for monosyllables presented in quiet was found. Each listening test took about 15 minutes.

For the assessment of the microphone arrays with hearing-impaired subjects, an induction loop was used in combination with the subject's own individually fitted hearing aid—for experimental details see (12). If necessary, the other ear was occluded with a foam plug (E.A.R.-plug). Because we were primarily interested in comparing the omnidirectional microphone and the microphone arrays, most subjects with hearing impairment performed two listening tests under two conditions: with their own hearing aid in combination with an external omnidirectional microphone, and their own

Figure 5.
Attenuation of the artificial diffuse sound field, as measured in 1/3-octave bands with KEMAR and an omnidirectional, a cardioid, the broadside, and the endfire microphone. The mean level is computed from the 1/3-octave measurements with equal weights.

hearing aid with endfire or broadside microphone array. A subgroup of eight hearing-impaired listeners took the listening tests under all three conditions. The order of conditions (e.g., with and without microphone array) was varied to avoid effects of habituation and fatigue, and, moreover, to equalize small differences between the 10 lists of sentences.

For the group of normal-hearing subjects, the listening tests were carried out with changing subgroups for four conditions: own ear, hearing aid (Philips M47), broadside microphone array, and endfire microphone array via hearing aid. The normal-hearing subjects listened to the hearing-aid using a libby-horn with foam plug.

Figure 6 presents the averaged S/N ratios and intersubject standard deviation for the number of listening tests (n) per condition, for the normal-hearing group as well as for the hearing-impaired group. In addition, the values of the S/N ratios at overall levels of +20 and +30 dB and the difference between these values are given. The small differences between the values of the S/N ratios at +20 and +30 dB confirm the linearity of the SRTs as hypothesized by Plomp (3) and the reliability of the mean S/N ratios for most conditions. For the normal-hearing conditions, the standard deviations are less than 1.2 dB. For the hearing-impaired conditions, they are about 3 dB.

A comparison of the S/N ratios shows the following points:

1. The monaural S/N ratio of the normal-hearing group (listening with one good ear) equals -8.5 dB.
2. A hearing aid (Philips M47) decreased the S/N ratio of the normal-hearing listeners by 1.2 dB.
3. The S/N ratio of the normal-hearing group can be improved significantly using a microphone array, instead of an omnidirectional microphone.
4. The hearing-impaired group listening with the omnidirectional microphone has a S/N ratio of only -0.2 dB, with a large standard deviation of 3.4 dB.
5. The microphone arrays also give a significant improvement of the S/N ratios for the hearing-impaired group. The absolute values of the average S/N ratio obtained with the microphone arrays is comparable with the S/N ratio of the normal-hearing group listening with one good ear.

	Normal-hearing subjects				Hearing-impaired subjects		
	Own Ear	Hearing Aid	Broadside	Endfire	Hearing Aid	Broadside	Endfire
n	16*	14	7	6	45	26	27
+20	-8.3	-7.1	-11.3	-11.5	-0.4	-7.2	-6.5
+30	-8.8	-7.5	-12.3	-13.6	0.1	-7.0	-6.8
diff.	0.5	0.4	1.0	2.1	-0.5	-0.2	0.3
mean	-8.5	-7.3	-11.8	-12.6	-0.2	-7.1	-6.6
s.d.	1.0	1.2	0.8	0.9	3.4	3.0	3.0
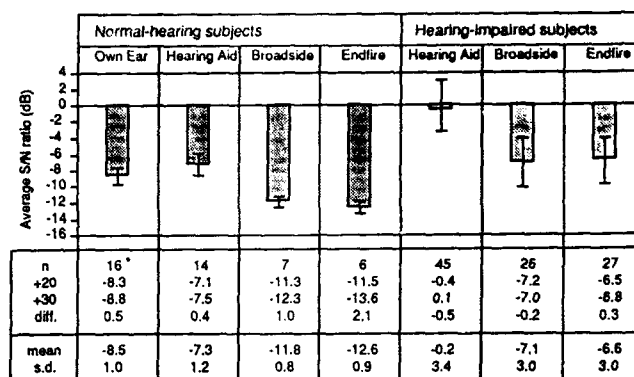
* 10 listening tests added for fixed noise levels of 55 and 65 dB.

Figure 6.
S/N ratios resulting from listening tests with normal-hearing and hearing-impaired subjects for different listening conditions.

On the average, the broadside microphone array gives an improvement of 7.0 dB with a standard deviation of 1.9 dB, and the endfire microphone array gives an improvement of 6.8 dB with a standard deviation of 2.1 dB. Most hearing-impaired listeners (41 of 45) obtain an improvement of at least 5 dB. The difference of 0.2 dB between the broadside and endfire microphone arrays was also found for the subgroup of eight subjects listening to both microphone arrays.

CONCLUSION

The following was noted as the result of the work described above:

• Computer simulations showed that it should be possible to construct a broadside or an endfire array with dimensions suitable for mounting on a pair of eyeglasses, with a stable and near-to-optimal directivity index using five cardioid microphones only.
• Directivity pattern and directivity index measured in free-field with and without KEMAR show only a slight degradation in the performance of the arrays due to reflection and diffraction of sound at the head of the order of 1 dB.
• KEMAR measurements in an artificial diffuse noise field show that the broadside microphone array and the endfire microphone array will attenuate the diffuse noise field ("speech noise") relative to sound coming from the front direction ("desired speech") and give a mean improvement of 7.0 dB and 6.8 dB, respectively.

• Speech intelligibility tests in a cocktail party situation (simulated by the diffuse noise field) with normal-hearing and hearing-impaired subjects showed that the microphone arrays improve the critical S/N ratio significantly: an omnidirectional hearing aid microphone broadside and endfire microphone array give a mean improvement of 7.0 dB and 6.8 dB, respectively. These results confirm the KEMAR measurements.

ACKNOWLEDGMENTS

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**Department of
Veterans Affairs**

Real-time multiband dynamic compression and noise reduction for binaural hearing aids

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Abstract—A multi-signal-processor set-up is introduced that is used for real-time implementation of digital hearing aid algorithms that operate on stereophonic (i.e., binaural) input signals and perform signal processing in the frequency domain. A multiband dynamic compression algorithm was implemented which operates in 24 critical band filter channels, allows for interaction between frequency bands and stereo channels, and is fitted to the hearing of the individual patient by a loudness scaling method. In addition, a binaural noise reduction algorithm was implemented that amplifies sound emanating from the front and suppresses lateral noise sources as well as reverberation. These algorithms were optimized with respect to their processing parameters and by minimizing the processing artifacts. Different versions of the algorithms were tested in six listeners with sensorineural hearing impairment using both subjective quality assessment methods and speech intelligibility measurements in different acoustical situations. For most subjects, linear frequency shaping was subjectively assessed to be negative, although it improved speech intelligibility in noise. Additional compression was assessed to be positive and did not deteriorate speech intelligibility as long as the processing parameters were fitted carefully. All noise reduction strategies employed here were subjectively assessed to be positive. Although the suppression of reverberation only slightly improved speech intelligibility, a combination of directional filtering and dereverberation provided a substantial improvement in speech intelligibility for most subjects and for a certain range of signal-to-noise ratios. The real-time implementation was very helpful in optimizing and testing the algorithms, and

the overall results indicate that carefully designed and fitted binaural hearing aids might be very beneficial for a large number of patients.

Key words: *binaural hearing aids, impaired loudness perception, multi-signal-processor, noise reduction, recruitment phenomenon, sensorineural hearing impairment, speech intelligibility.*

INTRODUCTION

The most common complaints of patients with sensorineural hearing impairment are their reduced ability to understand speech in a noisy environment and their impaired mapping between the sound-pressure level of natural acoustical signals and the perceived loudness of these signals. The impaired loudness perception is often associated with the so-called "recruitment phenomenon," (i.e., the inability of the patient to perceive any sound at low to moderate sound-pressure levels and a steep increase in perceived loudness if the level increases from moderate to high values). Therefore, dynamic compression circuits have traditionally been incorporated in hearing aids (1). They operate on the full input frequency range and/or in several independent frequency bands in order to account for the frequency dependence of the hearing dysfunction. In the literature, however, there has been controversy over the benefit of multichannel compression algorithms (especially if short time constants are involved) in comparison with linear or broadband compression systems (2,3,4).

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Unfortunately, due to the computational expense involved in multiband algorithms, only short speech samples have been used so far to evaluate these systems empirically and to compare their performance with other systems. In addition, most of the compression systems developed so far only operate monaurally (i.e., on the signal for one ear). Thus, the systems can distort the spatial auditory impression, which is primarily determined by binaural hearing (i.e., by listening with both ears). Therefore, a real-time binaural multiband-dynamic-compression algorithm is described and evaluated in this paper that incorporates interaction between both stereo channels to preserve interaural intensity cues. Adjustable interaction between frequency bands is also provided which allows for a parametric transition from a broadband (single-channel) system to a multiband system where all frequency channels are processed separately.

Binaural hearing also contributes significantly to the ability of normal listeners to suppress disturbing noise and to enhance the signal coming out of a desired direction (i.e., the so-called "cocktail party effect"). In addition, a reduction of the perceived reverberation and its negative effect on speech intelligibility is performed by normal listeners who are able to exploit binaural cues (e.g., interaural time and intensity differences) with sophisticated signal-processing strategies in the central auditory system (5). To restore the speech perception abilities of the impaired listener in noisy and reverberant environments, the evaluation and processing of interaural differences might therefore be performed by a "binaural" hearing aid using an intelligent processing scheme that operates on two input signals and provides one or two output signals. Several algorithms of this type have been proposed in the literature that were not necessarily intended for use in hearing aids (6,7,8,9,10,11,12). However, they tend to be very sensitive to small alterations in the acoustical transfer functions, require a high computational complexity, or introduce disturbing processing artifacts.

The directional filter algorithm proposed by the authors (13) minimizes these disadvantages since it is rather insensitive to changes in the acoustical transfer functions and exhibits a limited computational complexity. A real-time implementation is therefore possible, which helps to reduce the artifacts. In non-reverberant acoustical conditions, the algorithm

is successful in enhancing a "target speaker" in front of the listener with up to three interfering speakers distributed off the midline. When reverberation is added, however, the performance of the algorithm deteriorates due to processing artifacts. A combination with a scheme for suppressing reverberation is described here that also should extend to reverberant conditions the potential benefit obtainable from this algorithm.

In this paper, the implementation and first results with these algorithms on a multi-signal-processor set-up in real-time is described. After evaluating the binaural multiband-dynamic-compression algorithm, the combination of the directional filter with a dereverberation algorithm that operates on binaural input signals is evaluated. The real-time implementation facilitates the processing of large speech samples and allows for an interactive optimization of the processing parameters as well as an interactive fitting to the requirements of the individual patient.

METHOD

Hardware Set-Up

A block diagram of a hardware set-up is given in **Figure 1**. Three digital signal processors (AT&T WE DSP 32C), each a part of an Ariel PC-32 Digital Signal Processor (DSP) board in a PC-bus slot, are connected with serial high-speed interfaces. A stereo 16-bit A/D (analogue-to-digital) converter is serially connected to the first DSP, while a 16-bit stereo D/A (digital-to-analogue) converter is serially connected to the third DSP. The input microphone signals are either recorded with a dummy head or with miniature microphones located in the outer ear canal of an individual. These signals are amplified,
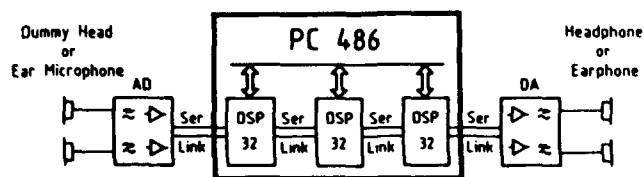

Figure 1.
Block diagram of the hardware set-up employing three Digital Signal Processor (DSP) chips with serial connections to external AD/DA converters.

low-pass filtered and converted to digital. The output signals of the D/A converters are low-pass filtered, amplified, and presented to the subject via headphones or insert earphones. An overlap-add technique (14) is implemented with the three DSPs. The first DSP divides the incoming time signal into overlapping segments, multiplies each time segment with a Hamming window, and extends the segment with additional zeroes before performing a 512-point fast Fourier transform (FFT). The second DSP processes the signals in the spectral domain while the third DSP performs the inverse Fourier transform and the overlapping addition of the filtered time segments in order to reconstruct the time signal.

The three DSP boards are housed in a PC-compatible 486 personal computer. A program library was developed that reflects the high specifications of a multiprocessing system with respect to the coordination of the processors, the data transfer protocol, and the debugging options. To retain the flexibility and the simple structure of the whole software, the high-level routines that structure the whole program system were written in "C" language. On the other hand, to provide an efficient real-time realization of certain routines, the computational intensive parts of the program were written in assembly language. To ensure an effective and time-saving data transfer between the processors, each processor operates on alternating DMA input and output buffers, which may be accessed while simultaneously processing the data from the other data buffers

Algorithms

Figure 2 gives the block diagram of the algorithm for multiband dynamic compression. Successive short-term spectra are calculated in both stereo channels using Hamming-windowed segments of 408 samples and an FFT length of 512 samples at an overlap rate of 0.5 (distance of successive frames: 204 samples; sample rate: 30 kHz). The subsequent processing is performed in the frequency domain. For each individual ear, linear frequency shaping is provided with a high spectral resolution by multiplying each FFT channel with a prescribed fixed value. In addition, a dynamic nonlinear weighting of the frequency channels is performed in 24 non-overlapping bands with a bandwidth according to the critical bandwidth of the ear (i.e., approximately
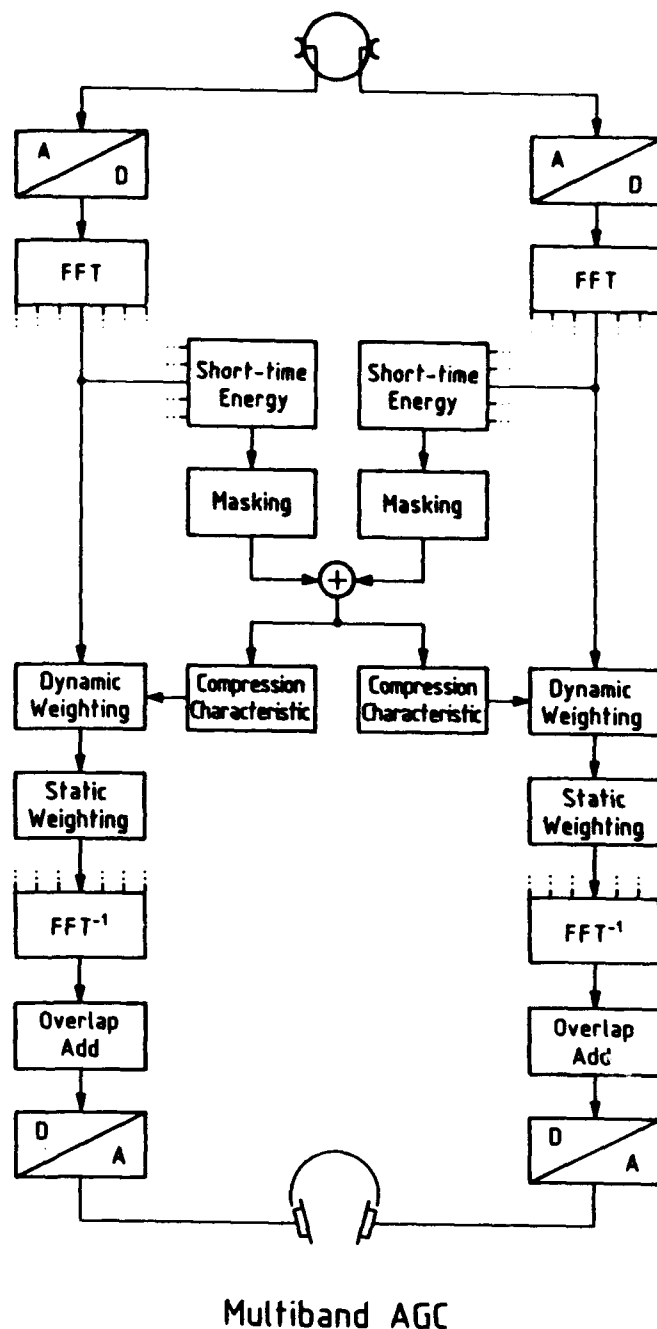
Multiband AGC

Figure 2.
Block diagram of the multiband compression algorithm.

100 Hz below 500 Hz center frequency and 0.2 × center frequency for frequencies above 500 Hz) (15). Thus, the nonlinear level adjustment is performed with less spectral resolution than the linear frequency shaping.

For each frequency band in each ear, a compression characteristic is prescribed that is computed

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as follows: The input energy for each frequency band is obtained by adding up the energies of all FFT channels belonging to the respective frequency band. This value is low-pass filtered with an exponential time window employing different time constants for increasing and decreasing instantaneous energy (i.e., "attack" and "release" time). Subsequently, the masking effect of the energy within a frequency band on adjacent frequency bands is taken into account. Upward spread of masking is realized by attaching ramps to each frequency band with 10 dB per bark toward higher frequencies. Similarly, downward spread of masking is realized by ramps with 25 dB per bark toward lower frequencies. In each band, the respective maximum out of the instantaneous energy within the band and the energy originating from the ramps of adjacent frequency bands is adopted as "effective" input level. Therefore, the level adjustments in the different frequency bands are linked together and the processing artifacts are reduced. The degree of this linkage may be altered by changing the slope values of the ramps between 0 dB per bark (broadband compression) and 50 dB per bark (multichannel compression). Finally, the "effective" energy values from the left and the right stereo channel are added in order to simulate the binaural loudness summation.

The fitting of the compression characteristic to the hearing loss of each patient can be explained by **Figure 3**, which outlines the result of a loudness scaling procedure. The dashed curves denote the level of a narrow-band noise as a function of its center frequency, which produces for normal listeners the loudness sensations "very soft," "comfortably loud," and "very loud." The solid lines denote the respective curves of a listener with high frequency hearing loss. For low frequencies, a relatively high dynamic range is retained, whereas at high frequencies only about 10 dB remains between the impression of "very soft" and "very loud." The aim of the algorithm is to restore the perceived loudness of the individual impaired listener in each frequency band as closely as possible to the perceived loudness of an average normal listener. Therefore, the amplification within each frequency band is adjusted for each "effective" input level to compensate the level difference between the loudness impression of the average normal listener (dashed curves) and the corresponding loudness
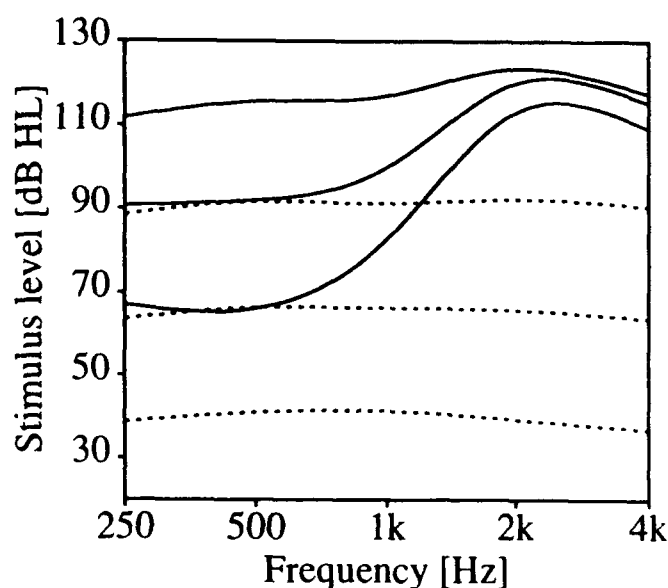
Figure 3.
Equal loudness category contours for subjects with normal hearing (- - - - -) and for one subject with sensorineural hearing impairment (--------). The three curves denote the level of a third-octave-filtered noise required to produce the loudness impression "very soft," "comfortable," and "very loud" as a function of frequency.

impression of the individual impaired listener (solid curves). This amplification is composed out of the (static) linear frequency shaping part (which does not depend on the input level) and the (dynamic) nonlinear compression part. The linear frequency shaping transforms the loudness sensation "comfortably loud" from the impaired listener into the corresponding sensation of the normal listener (i.e., it compensates for the level difference between the intermediate dashed and the intermediate solid curve in **Figure 3**). The nonlinear compression characteristic summarizes all input-level dependent deviations from this static amplification. For example, if the input level in a certain band equates the level belonging to the impression "comfortably loud," then the whole amplification is already provided by the linear frequency shaping part. Therefore, the dynamic compression part would set the amplification value to one. If the input level is higher, this value will decrease, whereas it will increase if the input level is lower. The attack and release time (i.e., decay of the impulse response to 1/e) were both set to 7 ms for all frequency bands and were not adjusted individually.
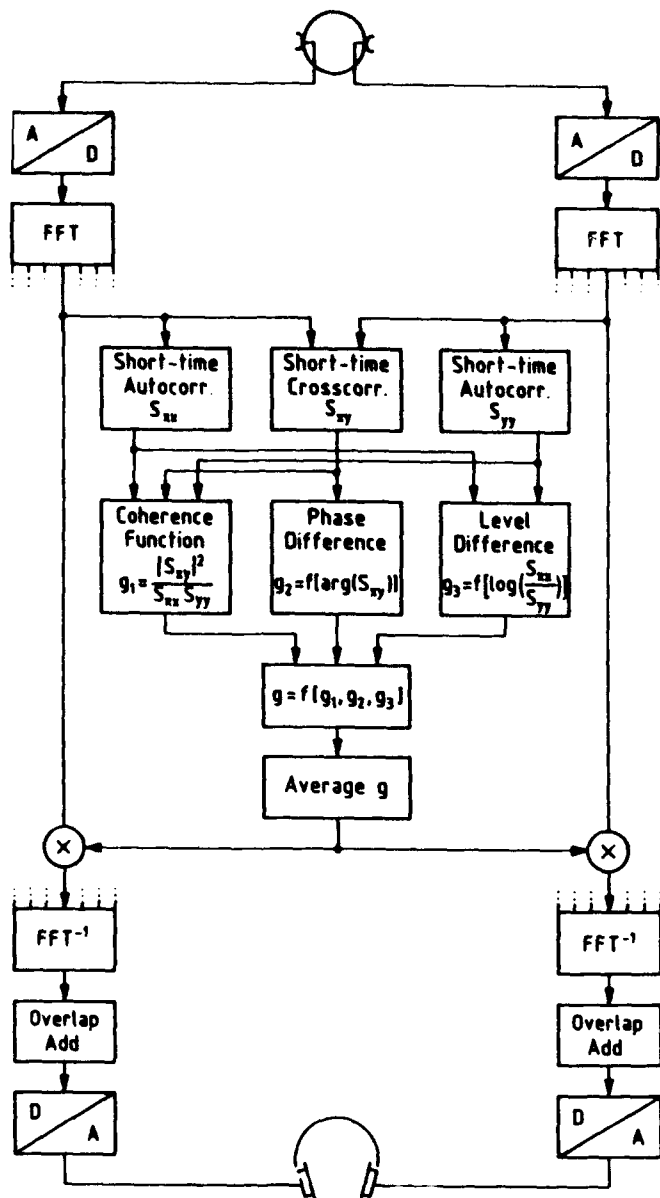
Suppression of Reverberation
and Lateral Noise Sources

Figure 4.
Block diagram of the algorithm for suppressing reverberation and lateral incident sound sources.

The algorithm for suppressing lateral noise sources and reverberation evaluates averaged interaural time and intensity differences to detect lateral incident sound components. It further evaluates the interaural coherence to detect reverberation processes in the input signals. Frequency bands

showing desired values of these interaural parameters (i.e., interaural time and intensity differences close to the desired "reference" values and interaural coherence close to 1) are passed through unchanged, whereas frequency bands with undesired values are attenuated. The lateral noise suppression part of the algorithm is a modification of the algorithm described by Kollmeier and Peissig (13) where instantaneous interaural phase and intensity differences were evaluated. In reverberant situations, however, these instantaneous values within each frequency band do not provide much information about the angle of incidence of a sound source located outside of the reverberation radius. In addition, the normal binaural system is capable of localizing sound sources even in extremely reverberant situations by, for example, evaluating the first wave front and detecting interaural time and level differences of the envelopes. Therefore, the current algorithm evaluates the phase of the short-term cross-correlation and the ratio of the short-term autocorrelation between each pair of frequency bands that are related to the phase and level differences of the input signals' envelopes, respectively. Thus, they should provide more reliable information about the angle of incidence in a reverberant room than the instantaneous interaural phase and intensity differences.

A block diagram of the algorithm is given in **Figure 4**. As above, the incoming signal is segmented, windowed, padded with zeros, Fourier-transformed and back-transformed after processing in the frequency domain. Within each frequency band, the short-term auto- and cross-correlation is computed for the left and the right stereo channel with an exponential weighting window as follows: If X and Y denote the complex output signals of the bandpass filters at the right and left stereo channel, respectively, n denotes the index of the time, and α denotes a coefficient between 0 and 1, we can write:

$$S_{xx}(n) = (1-\alpha)|X(n)|^2 + \alpha S_{xx}(n-1)$$

$$S_{yy}(n) = (1-\alpha)|Y(n)|^2 + \alpha S_{yy}(n-1)$$

$$S_{xy}(n) = (1-\alpha)X(n)Y^*(n) + \alpha S_{xy}(n-1)$$

From the values S_{xx}, S_{yy}, and S_{xy}, the interaural phase and level differences of the signal envelope and the interaural coherence are computed in each frequency band. The respective functions f_1 and f_2 are used to calculate weighting factors g_2 and g_3

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from these values. The shape of f_1 and f_2 determines both the range of incident angles for attenuation as well as the maximum attenuation within this region. They are obtained by measurements and may be optimized interactively later on. The weighting factor g_1 is directly given by the short-term coherence. By combining the weighting factors g_1, g_2, and g_3, the performance of the algorithm can be changed to suppress either reverberation or lateral sound sources or to perform a combination of both. In order to suppress processing artifacts, the final weighting factors g are averaged over adjacent frequency channels. If the processing parameters are adjusted properly, the algorithm yields very natural-sounding output signals and performs a satisfactory suppression of reverberation and lateral incident sounds.

Subjects

Six subjects with sensorineural hearing impairment, aged between 25 and 89 years with different degrees of high frequency hearing loss, participated in this study. All subjects were clinically examined to rule out a middle-ear dysfunction and to classify the hearing loss to be of cochlear origin with a positive recruitment phenomenon. The audiometric thresholds at 500 Hz and 4 kHz are given in **Table 1**. In addition, the binaural speech intelligibility threshold is provided, that is, the signal-to-noise ratio for

50 percent correct performance in a German mono-syllable rhyme test in speech-simulating, continuous noise (16). For a prescription of the dynamic compression algorithm, a loudness scaling method was performed with third-octave-bandpass-filtered noise. The subject's task was to associate each stimulus with a subjective loudness category ("very soft," "soft," "comfortable," "loud," "very loud") and to further subdivide each category into 10 subcategories. This procedure yields a loudness scale between 0 and 50 partitioning units (17,18). The residual dynamic range (i.e., the difference in level between the loudness categories "very loud" and "very soft") is also included in **Table 1** for each audiometric frequency and both ears.

Assessment Methods

To assess the subjective quality of different versions of the hearing aid algorithms, recorded materials from different acoustic situations were presented to the subjects with the respective processing condition. All materials were either dummy-head recorded using the "Göttingen" dummy-head or using stereophonic miniature microphones inserted in the outer ear-canal of a human listener. The subjects were allowed to listen to a combination of acoustic situation and processing scheme for as long as they desired. They were asked to assess the subjective transmission quality within a scale of five

Table 1.
Audiometric data and residual dynamic range derived from the loudness scaling experiment (in parentheses) for six impaired listeners. The binaural speech intelligibility threshold in noise obtained with a rhyme test and the individual sentence intelligibility scores for the evaluation of the multiband compression are also included.*

| Subject | Age | Sex | Hearing Loss (Residual Dynamic Range) Right (dB) | | Left (dB) | | Rhyme Test Threshold (dB) | Sentence Scores** (% correct) | | |
			500 Hz	4 kHz	500 Hz	4 kHz		unp.	lin.	comp.
JJ	25	M	55 (40)	105 (10)	60 (40)	105 (8)	3.0	35	38	24
JS	71	F	45 (35)	*** (15)	20 (50)	25 (50)	6.1	56	48	56
WH	68	M	45 (50)	75 (20)	50 (40)	80 (15)	2.0	36	48	0
RP	52	F	30 (40)	55 (20)	30 (20)	40 (20)	3.0	26	48	0
HS	89	M	55 (10)	*** (10)	40 (40)	60 (20)	2.6	23	36	29
WD	72	F	40 (40)	65 (40)	35 (50)	65 (30)	3.5	26	35	54

*For normal listeners, the average residual dynamic range is 50 dB and the speech intelligibility threshold in noise is −5 dB.
**Test conditions: unp. = unprocessed; lin. = linear frequency shaping without compression; comp. = linear frequency shaping with compression.
***No threshold measurable.

categories ("bad," "poor," "reasonable," "good," "excellent").

Speech intelligibility was measured for a subset of the acoustical situations mentioned above using an open German sentence test recorded on compact disc (19). The subject's task was to repeat the whole sentence, and the number of correctly repeated words was scored. A complete test consisted of 10 short sentences. For intelligibility measurements with the dynamic compression algorithm, a dummy-head recording of cafeteria noise was used as background noise, which was added to a dummy-head recording of the speech material alone at a fixed signal-to-noise ratio. For assessing the noise reduction and dereverberation algorithm, a dummy-head recording of the speech signal and the interfering noise was performed in a reverberant room with a reverberation time of 2 to 3 sec. The desired signals (i.e., running speech for quality judgments and test sentences for speech intelligibility test) were radiated with a loudspeaker directly in front of the dummy-head at a distance of 1.5 m. The interfering noise was running speech radiated 30° from the left of the midline. The speech level was always adjusted to match the most comfortable listening level for each individual subject.

RESULTS

Dynamic Compression Algorithm

For assessing the subjective quality of the dynamic compression algorithm, three dummy-head recordings of typical acoustical conditions were used: a sample of traffic noise, a loud doorbell presented in soft background noise, and a sample out of a string quartet by Schubert. All listening samples were recorded with stereophonic inserted ear-level microphones in real situations and were presented unprocessed, processed with linear frequency shaping alone, and with linear frequency shaping including compression. The sound samples were presented to the subjects with a Sennheiser HD 25 headphone. At the beginning of each session, an overall level adjustment of up to 10 dB was applied to match the average presentation level to the most comfortable listening level.

Figure 5 shows the differences in subjectively assessed transmission quality (expressed as grades
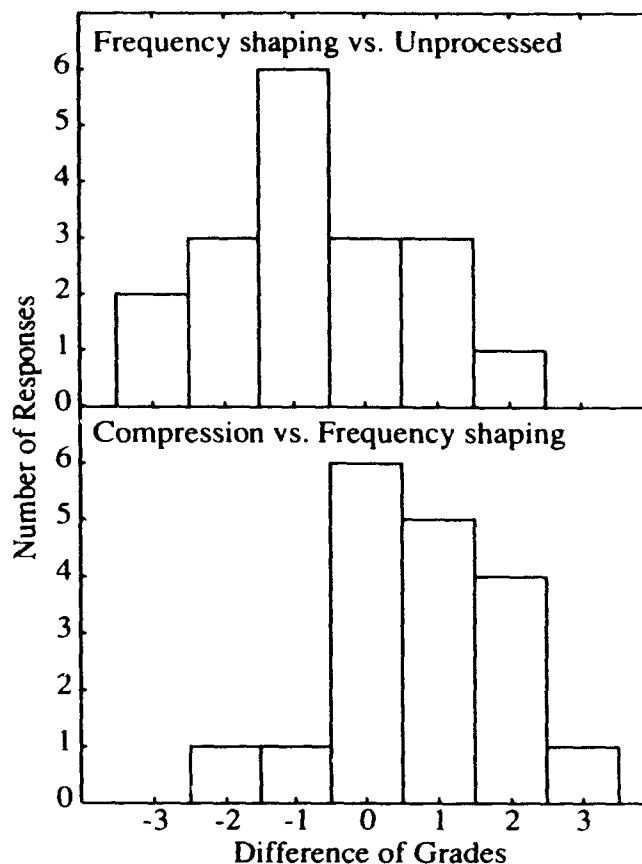
Figure 5.
Quality judgments of different versions of the compression algorithm for six impaired listeners and three different simulated acoustical situations. The upper panel gives the difference in grades between the condition with static linear frequency shaping and no processing. The lower panel gives the difference in grades between static linear frequency shaping plus compression versus shaping alone. Grades ranged from 1 ("bad") to 5 ("excellent").

ranging from 1, "very poor," to 5, "excellent") between the different processing conditions for all subjects and all three simulated acoustical situations. The upper panel of Figure 5 gives the score difference between linear frequency shaping and the unprocessed version. On the average, the unprocessed version is preferred. However, since the range of scores varies considerably, no significant advantage or disadvantage of linear frequency shaping versus unprocessed speech can be derived from these score differences. The subjects attributed their preference for the unprocessed condition to not being accustomed to high frequencies with their own hearing aid. Specifically, the loud doorbell caused

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uncomfortably loud sensations in the processed version, while the background noise was not audible at all. This effect was less prominent for the unprocessed version. The lower panel of **Figure 5** gives the difference in grades between linear frequency shaping, including compression versus linear frequency shaping. Obviously, the additional compression is judged to be positive due to the limitation of annoying acoustical components at high frequencies. This observation is quite unexpected for normal listeners who perceive a deterioration of transmission quality and an increase of processing artifacts caused by rapid dynamic compression. However, these artifacts appear to be inaudible for impaired listeners.

For measuring speech intelligibility, each subject was tested with two lists of ten sentences in each of the different processing conditions using cafeteria background noise. The average scores for each processing condition and each subject are included in **Table 1**. The difference in speech intelligibility score between the processed version with linear frequency shaping and the unprocessed version is given in the upper panel of **Figure 6**. On the average, intelligibility increases for linear frequency shaping. This effect is quite contrary to the assessed subjective preference of the unprocessed condition (*see* upper panel in **Figure 5**). However, the effect is rather small, since the interfering noise has approximately the same long-term spectrum as the speech signal. The lower panel of **Figure 6** gives the differences in speech intelligibility between the dynamic compression with linear frequency shaping versus linear frequency shaping alone. With few exceptions, intelligibility is increased by the addition of the dynamic compressor. These exceptions are caused by an erroneous fitting of the compressor characteristic for one subject; the loudness scaling yielded nearly the same level for the loudness categories "comfortable" and "very loud." Thus, the algorithm performs a clipping in all frequency channels, which nearly completely suppresses speech in the presence of an interfering noise and causes a drastic decrease in speech intelligibility.

Noise and Reverberation Suppression

To evaluate the performance of the algorithm to suppress lateral noise sources and reverberation, an acoustic situation was simulated by dummy-head recordings in a reverberant room employing one
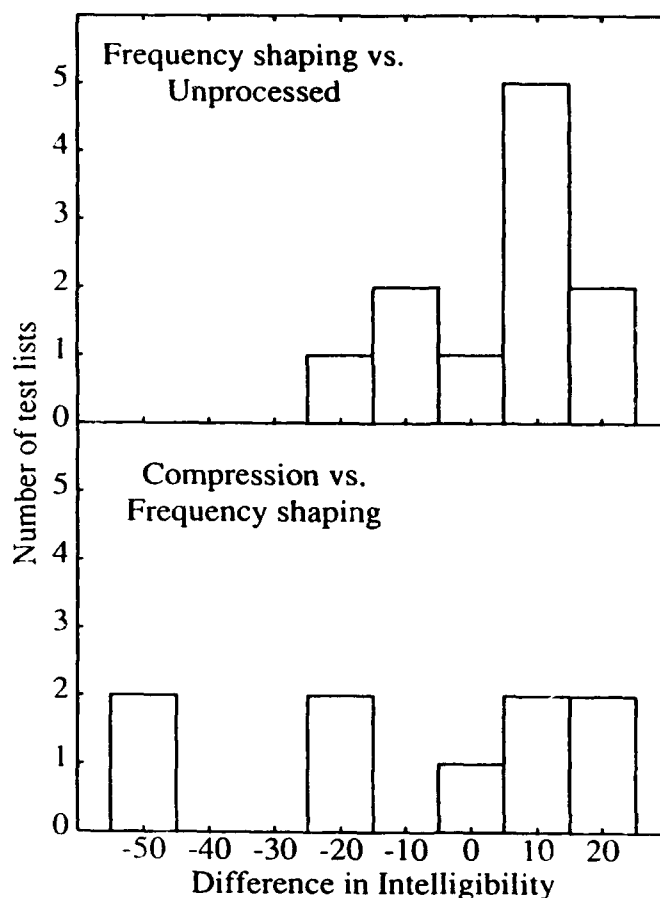
Figure 6.
Difference in speech intelligibility for static linear frequency shaping versus no processing (upper panel) and shaping plus compression versus shaping alone (lower panel). The score for each test list for each subject is counted for the three different processing conditions.

target speaker and one interfering speaker (see above). The signal-to-noise ratio was individually adjusted for each subject within a range of -5 dB to $+2$ dB in order to obtain a speech intelligibility of approximately 50 percent for the binaural unprocessed condition. **Figure 7** gives the difference in subjective assessment of the transmission quality between the dereverberation algorithm and the unprocessed condition (upper panel) and between the combination of dereverberation and directional filter as compared with the unprocessed condition. Note that linear frequency shaping without dynamic compression was provided in all conditions, including the reference situation. For the dereverberation algorithm, five out of six subjects graded the quality of the processed signal as better than the unproc-
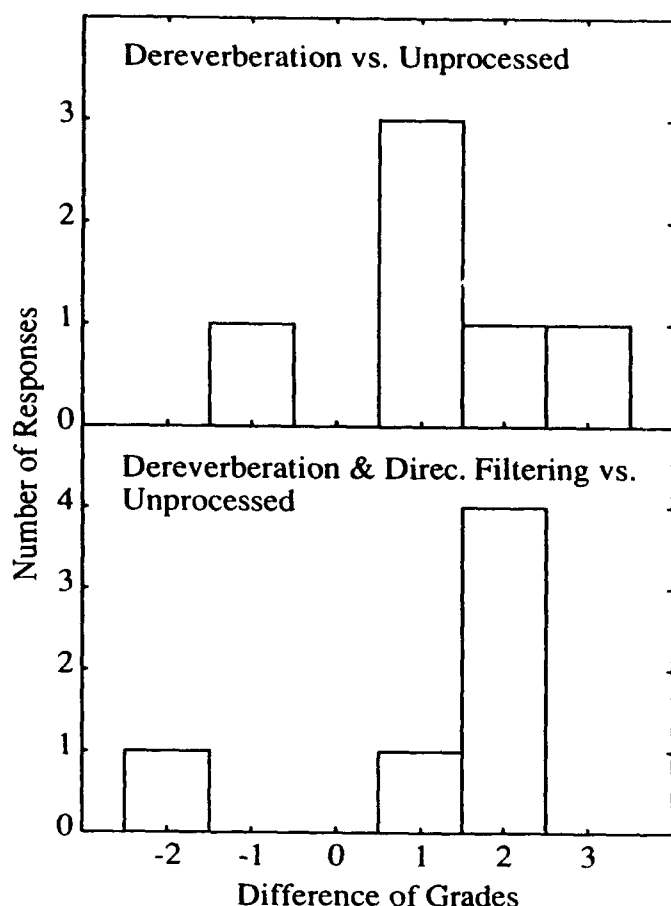
Figure 7.
Quality judgments of different versions of the interference suppression algorithm for six impaired listeners. The upper panel gives the difference in grades between the condition of suppression of reverberation and no processing. The lower panel gives the difference in grades for the combination of dereverberation and the suppression of lateral noise sources (i.e., directional filtering) versus no processing. Linear frequency shaping is always provided. Grades ranged from 1 ("bad") to 5 ("excellent").

essed material by at least one point. After the addition of the directional filter, four of six subjects reported an improvement of two grades as compared with the unprocessed version. Only one subject (JJ) reported better quality of the unprocessed version as compared with the dereverberation algorithm with and without additional directional filtering. This subject was the most severely impaired subject tested, and exhibited a very limited dynamic range (*see* **Table 1**). Apparently, the spectral changes introduced by the algorithms caused the speech signal to move out of this limited range.

Figure 8 gives the results of the speech intelligibility tests as the percentage of correctly repeated words. The first two bars for each subject give the results for the unprocessed, linear frequency shaped material, presented monaurally (first bar) or binaurally (second bar). Subject HS was only tested binaurally. Three out of five subjects exhibit a binaural gain in intelligibility compared with the monaural, unprocessed version. The binaural system of these subjects obviously manages to suppress parts of the interference caused by reverberation and interfering speech. However, subjects RP and JS exhibit a decrease in intelligibility if speech is also presented on the "worse" ear, indicating that the distorted internal representation of the input signals provided by this ear causes a "binaural confusion" rather than a binaural enhancement effect.

The third and fourth bar in **Figure 8** denote the intelligibility score for the dereverberation algorithm where the output signal is presented monaurally or binaurally to the subject, respectively. Compared with the linear shaped, unprocessed material (fourth bar versus second bar), a gain in speech intelligibility is obtained only for subject HS. This finding is consistent with a remark by Allen, et al. (6) that dereverberation algorithms tend to increase speech quality but not to improve speech intelligibility. However, after adding the directional filter, all subjects (except subject JJ) achieved a higher intelligibility for the monaural presentation than for the unprocessed version (fifth bar versus first bar). For the binaural presentation, however, no unambiguous conclusion can be drawn (cf. sixth bar versus second bar): three subjects (WH, WD, and JS) exhibited only a small change in intelligibility which is not significant. Only two subjects (RP and HS) obtained a significant gain in speech intelligibility of 25 percent with the combination of dereverberation and directional filtering.

The overall results from our subjects with various degrees of hearing impairment imply that the benefit obtainable for each individual listener from the preprocessing strategies described here depends on the hearing loss of the individual, the residual dynamic range in the high frequency region, and the signal-to-noise ratio of the test situation. Specifically, the two subjects with the smallest residual dynamic range at 4 kHz (subjects JJ and WH) exhibited the least benefit from the suppression of lateral noise sources and reverberation. This

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Section II. New Methods of Noise Reduction: Kollmeier et al.
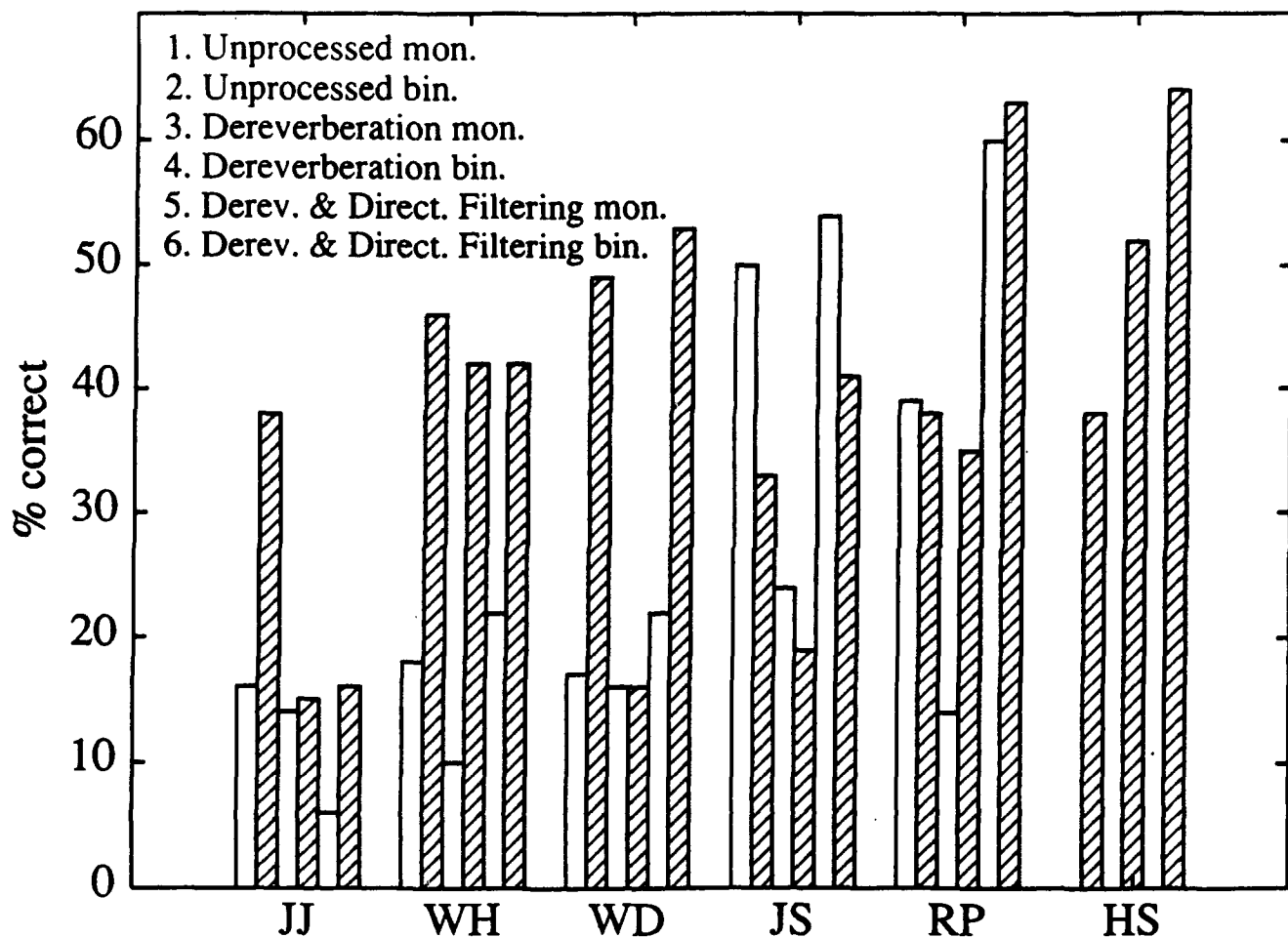
Figure 8.

Speech intelligibility results for different versions of the interference suppression algorithm for six impaired listeners. For each subject, scores were obtained for listening monaurally with the respective "better" ear and for listening binaurally (hatched). Three processing conditions were employed that all incorporated linear frequency shaping: *a*) unprocessed (columns 1 and 2); *b*) suppression of reverberation (columns 3 and 4); and, *c*) suppression of reverberation including suppression of lateral noise sources (columns 5 and 6).

effect might be due to the processing artifacts caused by suddenly switching on and off different frequency bands. They might be more distracting and disturbing if the remaining dynamic range is small. The subjects with the largest residual dynamic range at 4 kHz (WD and JS) were tested with the smallest signal-to-noise ratio of −2 dB. Their comparatively small gain in intelligibility provided by the algorithm might be explained by the unfavorable test condition, because the performance of the noise suppression algorithm decreases if the signal-to-noise ratio is decreased to values close to the speech reception threshold in noise of the normal listener.

DISCUSSION

Implementation of the Algorithms

The real-time implementation of the digital hearing aid algorithms proved to be very helpful in the developing and testing phase, where a number of processing parameters could interactively be adjusted in order to minimize the processing artifacts. For the dynamic compression algorithm, for example, musical tones and a perceivable roughness of the output signal occur if small time constants and no interactions between adjacent bands are involved. In addition, a small dynamic range of the output signal can only be achieved at the cost of

deteriorating the transmission quality for normal listeners. Fortunately, impaired listeners do not necessarily perceive these alterations as a degradation of speech quality.

The real-time implementation also enabled interactive changes of processing parameters while fitting the algorithms to the requirements of the individual patient. Although the parameters of the compression algorithm were primarily prescribed by the loudness scaling results, adjustments of the overall level of up to 10 dB were required to adjust the output level of the algorithm to the most comfortable listening level of the individual subjects. This difference between prescribed and perceived loudness is due primarily to the loudness summation in realistic broadband signals (such as speech) which is not accounted for by the original fitting method based on third-octave-band loudness scaling values. In our algorithm, only a rough estimate of broadband loudness summation is provided by accounting for upward spread of masking and downward spread of masking. Ideally, more precise ways of estimating the overall loudness for a broadband signal from its spectral contributions should be incorporated. Although quite accurate models of loudness perception have been developed on the basis of relational scales, such as the sone-scale (20), a quantitative model based on categorical loudness perception has yet not been developed (18).

A considerable disadvantage of the real-time system described here is the specialized software that had to be written for each of the signal processors and for the host processor. Although the flexibility and portability of the software was increased by programming the general structure in a high-level language (C language) and programming only time-critical parts in assembly language, the software is still processor-dependent and a migration toward more powerful DSP chips might be difficult. A further disadvantage of distributing the signal processing tasks over three DSP chips is the considerable delay between the input signal and the output signal, which amounted to approximately 50 ms in our case. This delay results from the transfer of blocks between the AD/DA converters and the three signal processors and from the overlap-add technique, which operates on successive time frames. Therefore, the use of the current system as a master hearing aid is limited, since the delay between auditory and visual input might already deteriorate the ability of the patients to use lip reading to aid their perception of speech.

Dynamic Compression Algorithm

One important feature of the implemented compression algorithm is the separate adjustment of the static, linear frequency shaping and the nonlinear dynamic compression. While the former is performed with the maximum frequency resolution of approximately 60 Hz, the latter is performed at a much broader frequency resolution that corresponds to the critical bandwidth of the ear. In addition, the effective frequency resolution for the nonlinear compression can be altered by using different slope values when accounting for upward and downward spread of masking. If these slopes are assumed to be very flat, all frequency channels are synchronized and a broadband compression will effectively result. The values used in our algorithm reflect approximate values for normal listeners in psychoacoustical experiments.

By assessing separately the effect of linear frequency shaping and dynamic compression, it could be demonstrated that linear frequency shaping was subjectively judged to deteriorate speech quality, although speech intelligibility in noise increased. The negative assessment is primarily due to the subjects being unaccustomed to a high gain at high frequencies in hearing aids. Therefore, additional compression is subjectively judged to improve the speech quality. In addition, speech intelligibility is not deteriorated by the additional compression if the processing parameters are carefully selected. These results are in agreement with studies that multiband dynamic compression does not significantly improve speech intelligibility (4,21,22), but are not consistent with Plomp's notion (3) that dynamic compression has a negative effect on speech intelligibility. However, the time constants employed here were relatively large and the cross-channel interaction provided comparatively smooth transfer functions. Hence, only a small detrimental effect of dynamic compression on speech intelligibility would have been expected on the basis of Plomp's arguments. Therefore, our data cannot be used to argue against Plomp's conclusions that small time constants and a large number of independent channels should not be employed for hearing aids.

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Noise and Reverberation Suppression

The algorithm for suppressing lateral noise sources and reverberation by exploiting binaural cues appears to operate quite efficiently even under adverse acoustical conditions (i.e., a reverberant environment). However, a trade-off exists between the potential of the algorithm to suppress interferences and its potential to preserve the quality of the transmitted speech (i.e., the absence of artifacts). High attenuation values of lateral sound sources imply large temporal and spectral fluctuations of the effective transfer function which inevitably produce processing artifacts. Hence, a realistic compromise between both specifications under different acoustical conditions has to be found empirically. This can be performed only if an interactive change of the processing parameters is possible, as in the real-time implementation described here.

Another important point is the performance of the algorithm as a function of the signal-to-noise ratio of the input signal: for high and intermediate signal-to-noise ratios, the algorithm operates quite well and yields virtually no artifacts. For low signal-to-noise ratios, however, the artifacts increase and no benefit is obtained from the algorithm as compared with the unprocessed situation, even for normal listeners. Therefore, the patients with moderate hearing loss who were tested at low signal-to-noise ratios obtained only a small benefit from the algorithm. However, patients with more severe hearing losses did profit from the algorithm at more favorable sig. al-to-noise ratios. In addition, it should be noted that the primary goal of the algorithms would be to enhance speech under conditions where normal listeners would not have difficulties understanding speech while impaired listeners would. In these situations, the signal-to-noise ratio is comparatively high and the algorithm would therefore be beneficial.

In conclusion, the algorithms presented here that are intended to be used in a "true binaural" hearing aid appear to have a large potential for aiding persons with hearing impairment. Specifically, the use of binaural information for suppressing reverberation and interfering noise appears promising. In addition, the real-time implementation of the algorithms is a salient tool for developing, testing, and assessing these algorithms. It is also a first step toward implementing these algorithms in future "intelligent" digital hearing aids.

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Digital signal processing (DSP) applications for multiband loudness correction digital hearing aids and cochlear implants

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Abstract—Single-chip digital signal processors (DSPs) allow the flexible implementation of a large variety of speech analysis, synthesis, and processing algorithms for the hearing impaired. A series of experiments was carried out to optimize parameters of the adaptive beamformer noise reduction algorithm and to evaluate its performance in realistic environments with normal-hearing and hearing-impaired subjects. An experimental DSP system has been used to implement a multiband loudness correction (MLC) algorithm for a digital hearing aid. Speech tests in quiet and noise with 13 users of conventional hearing aids demonstrated significant improvements in discrimination scores with the MLC algorithm. Various speech coding strategies for cochlear implants were implemented in real time on a DSP laboratory speech processor. Improved speech discrimination performance was achieved with high-rate stimulation. Hybrid strategies incorporating speech feature detectors and complex decision algorithms are currently being investigated.

Key words: *digital hearing aids, experimental digital signal processor, multiband loudness correction, noise reduction algorithm, speech coding strategies for cochlear implants.*

INTRODUCTION

New generations of fast and flexible single-chip digital signal processors (DSPs) allow the implemen-

tation of sophisticated and complex algorithms in real time which required large and expensive hardware only a few years ago. Signal processing algorithms such as nonlinear multiband loudness correction, speech feature contrast enhancement, adaptive noise reduction, speech encoding for cochlear implants, and many more offer new opportunities for the hard-of-hearing and the profoundly deaf. A recent review report on the status of speech-perception aids for hearing-impaired people came to the final conclusion that major improvements are within our grasp and that the next decade may yield aids that are substantially more effective than those now available (1).

This paper concentrates on three areas of DSP applications for the hearing impaired and investigates algorithms and procedures which have the potential of being effectively utilized and integrated into existing concepts of rehabilitation of auditory deficits in the near future. The first area concerns the problem of interfering noise and its reduction through an adaptive filter algorithm. The second area deals with the reduced dynamic range and distorted loudness perception of sensorineurally deaf persons and its compensation through a multiband loudness correction algorithm. The third area covers new processing strategies for profoundly deaf persons with cochlear implants. While the first DSP application does not rely on the individual hearing loss data of a subject, and thus could serve as a general preprocessing stage for any kind of auditory

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prosthesis, the second and third applications are intimately related to psychophysical measurements and individual fitting procedures. In the case of cochlear implant processing strategies, a number of technical and physiological constraints have to be considered. Although these three areas could be considered as three different projects that would need to be treated separately in more detail, there are many common aspects of signal processing and experimental evaluation. The use of similar tools and methods has been beneficial and fruitful throughout the theoretical and practical development of the respective projects. The complexity and interdisciplinary nature of the current auditory prostheses research problems requires collaboration and teamwork which, it is hoped, will lead to further practical solutions.

EVALUATION OF THE ADAPTIVE BEAMFORMER NOISE REDUCTION SCHEME FOR HEARING-IMPAIRED SUBJECTS

A major problem for hearing aid users is the reduced intelligibility of speech in noise. While the hearing aid is a great relief for conversations in quiet, its usefulness is often drastically reduced when background noise, especially from other speakers, is present. Several single-microphone noise reduction schemes have been proposed in the past which can improve sound quality but often failed to improve intelligibility, when tested with a normal-hearing or hearing-impaired subject in real life situations such as competing speakers as noise sources (2,3,4,5,6).

Multimicrophone noise reduction schemes, such as the adaptive beamformer (7,8,9), make use of directional information and are potentially very efficient in separating a desired target signal from intervening noise. The adaptive beamformer enhances acoustical signals emitted by sources coming from one direction (e.g., in front of a listener) while suppressing noise coming from other directions. Thus, the effect is similar to the use of directional microphones or microphone arrays. Because of the adaptive postprocessing of the microphone signals, the adaptive beamformer is able to combine the high directivity of microphone arrays with the convenience of using only two microphones placed in or just above the ears.

The aim of our investigation was to optimize the algorithm for realistic environments, implement a real-time version, and estimate its usefulness for future hearing aid applications.

Method

Parameter Optimization. Based on the work of Peterson, et al. (8), a Griffiths-Jim beamformer was chosen with two microphones placed at each ear. In the first stage, the sum and difference of the two microphone signals were calculated. The former contained mainly the desired signal, the latter mainly noise. A self-adaptive filter then tried to cancel out as much of the remaining noise in the desired signal path as possible.

While the system works quite well in anechoic rooms, its performance is not very satisfactory in reverberant conditions, because a part of the desired signal will also come from other than the front direction. In that case, the adaptive beamformer will mistake the desired signal for noise and try to cancel it as well. Therefore, it is necessary to detect the presence of a target signal and to stop the adaptation of the filter during these intervals (10,11). The method of target signal detection implemented in our system is based on the comparison of the variance of the sum (Sigma, Σ) and difference (Delta, Δ) signals which are easily obtained by squaring and lowpass-filtering the frontend signals of the Griffiths-Jim beamformer. An optimal threshold value of 0.6 ($\Sigma/[\Sigma + \Delta]$) was determined empirically through variations of the noise signal, filter parameters, and room acoustics. Other procedures have been considered as well but were either less efficient or computationally too expensive. We found that the performance of the adaptive beamformer can be notably increased by optimizing the adaptation inhibition.

Figure 1 shows measurements of the directivity characteristic with and without adaptation inhibition in a moderately reverberant room (RT = 0.4 sec). Because the microphones at the ears of a dummy head were omnidirectional, the beam not only pointed to the front (0°), but also to the back (180°) of the listener. It can be seen that the adaptation inhibition increased the difference between two sources located at an angle of 0° and 90° from around 6 dB to approximately 10 dB. The sharp profile of the beam with the adaptation inhibition present, when compared with the rather smeared
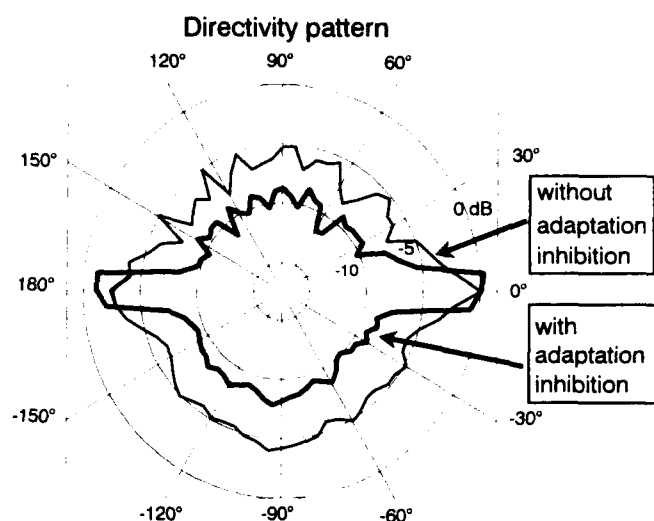
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Directivity pattern

Figure 1.
Effect of speech detection and adaptation inhibition on the directivity pattern of the adaptive beamformer. Front direction (nose): 0°, left ear: 90°.

pattern without adaptation inhibition, is a result of the target signal detection scheme which not only enhances the directivity of the adaptive beamformer, but is also responsible for the shape and opening angle of the beam.

The adaptive beamformer depends on a great variety of acoustic and design parameters. To analyze the sensitivity of these parameters, theoretical as well as experimental investigations were carried out based on adaptive filter and room acoustics theories. A computer program was written which prompts the user for room acoustic and design parameters, as well as directivity and position of a target and a noise source. The program then predicts the improvement in signal-to-noise ratio (SNR) which can be reached with a perfectly adapted beamformer. The predictions were experimentally verified. It is beyond the scope of this paper to discuss the theoretical analysis in detail or to describe the influence of all parameters investigated.

Three parameters were found to influence the performance of the adaptive beamformer most: the reverberation time of the room, the length of the adaptive filter, and the amount of delay in the desired signal path of the beamformer (12).

Figure 2 shows measurement data from an experimental setup, where white noise was generated by two loudspeakers located at a distance of 1 m

from a dummy head. The desired signal source was placed at 0°, the noise source at 45° to the right of the head. Thus, the two microphone signals picked up SNRs as indicated in **Figure 2** by squares (comparing the output of the beamformer with the signal of the microphone which is directly irradiated by the noise source) and triangles (comparing the output with the contralateral microphone signal).

The variation of reverberation time (**Figure 2,** left) was achieved by performing the measurements in a number of different rooms whose acoustic characteristics had been determined. It can be seen that a moderate amount of reverberation reduces the effectiveness of the beamformer to a few decibels.

The reverberation time of rooms where a noise reduction would be most advantageous can usually not be influenced. The design parameters of the adaptive beamformer however can be optimized. The most important of them is the length of the adaptive filter. When varying this parameter, it can be seen (**Figure 2,** right) that very short filters are able to switch to the microphone signal with the better SNR, while filters of at least about 500 taps in length are required to reach an additional 4 dB.

A computationally inexpensive way to optimize the adaptive beamformer is the selection of an optimal delay in the desired signal path. We found that for longer filters such as a 512 tap filter, the choice of a reasonable delay is important. It was also found that the optimum depends on the actual acoustical setup and will lie between 25 percent and 50 percent of the filter length.

Implementation. A real-time version of the adaptive beamformer was implemented on a PC-based, floating point digital signal processor (TMS320C30, Loughbourough Sound Images Ltd.). For the update of the adaptive filter, a least mean squares (LMS) adaptation algorithm (13) was employed. After evaluating several different adaptation algorithms, we found that this widely used and well-understood algorithm is still the best choice for our hearing aid application. An adaptation inhibition, as discussed above, was implemented.

With this system, filters of up to 500 taps in length at a sampling rate of 10,000 per second can be realized. For our implementation, a high-performance DSP had to be used whose battery power requirements are still excessive for a commercial hearing aid. The system is, however, flexible enough to allow an evaluation of the main aspects of the
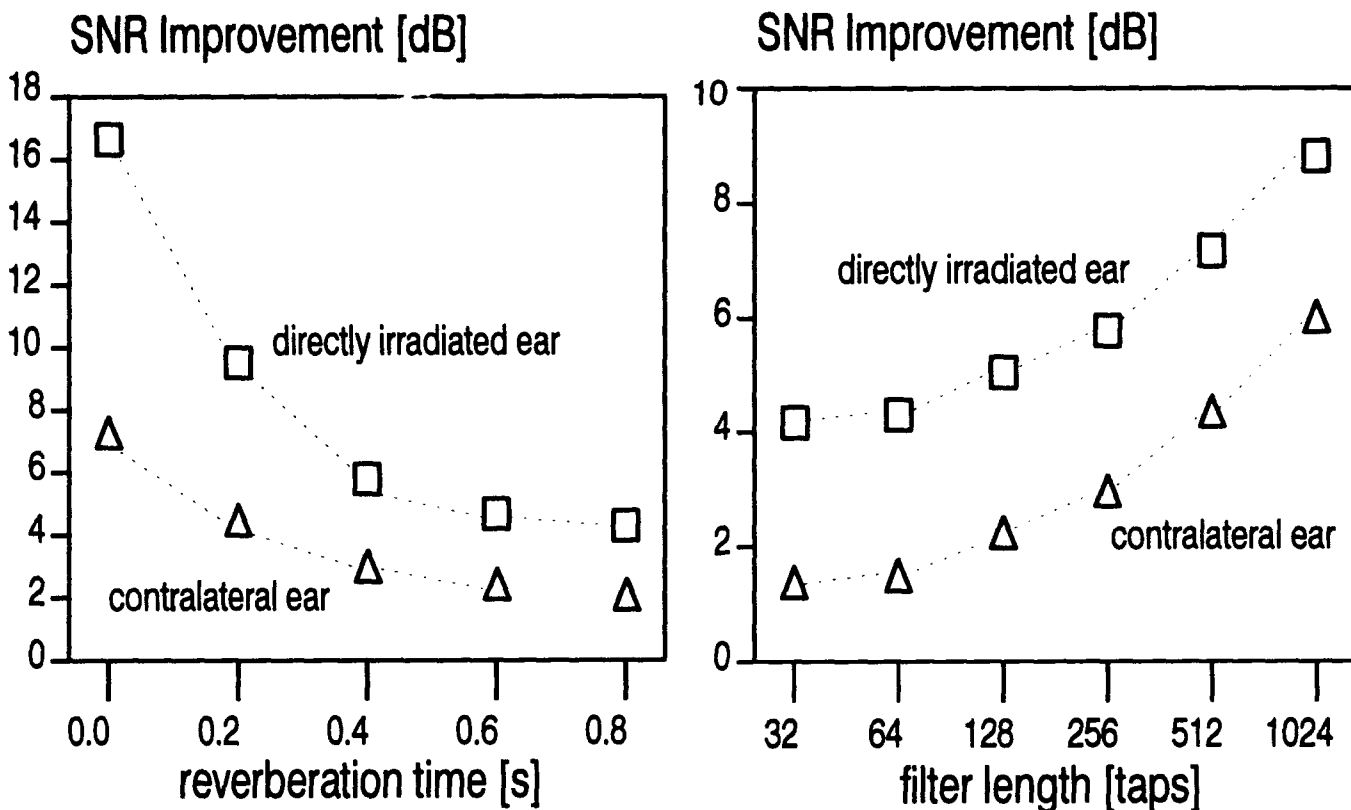
Figure 2.
SNR improvement as function of reverberation time (left) and filter length (right).

algorithm with the potential of future miniaturization.

Evaluation Experiments. To ensure that the adaptive beamformer really is a useful algorithm for hearing aid users, intelligibility tests were carried out with normal-hearing and hearing-impaired subjects. Speech test items were presented by a computer and consisted of two-syllable words with either the medial vowels or consonants forming phonological minimal pairs. Four response alternatives were displayed on a touch-sensitive computer display from which the subjects had to select a response by pointing to it with a finger. No feedback and no repetitions of test items were provided. Fifty or one hundred words per condition were presented and the number of correctly identified items were corrected for the chance level. Phoneme confusion matrices could be generated for further analysis of transmitted information (14).

Most of the test material was recorded in a test room with an average reverberation time of 0.4 sec. Speech was presented via a frontal loudspeaker at 70

dB sound pressure level (SPL). Speech-spectrum-shaped noise which was amplitude-modulated at a random rate of about 4 Hz was presented 45° to the right at different SNRs. A dummy head with two microphones located in the left and right conchas was used to pick up sounds. The distance between the dummy head and either of the two loudspeakers was 1 m. In the experiments with normal-hearing subjects, the sounds were presented directly via earphones monaurally or binaurally (unprocessed conditions) or binaurally after modification by the adaptive beamformer (processed condition). In the experiments with the hearing aid users, the unprocessed or processed sounds were presented via a loudspeaker in a sound-treated room.

This setup was thought to represent some important aspects of everyday listening situations: 1) the size and reverberation time of the room is typical for offices and living rooms in our surrounding; 2) a fair amount of head shadow is provided by the dummy head; and/or, 3) because of the integrated adaptation inhibition, the adaptation of the

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Normal Hearing Subjects (n=9)
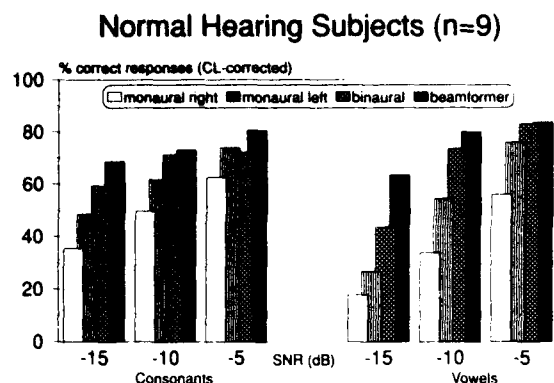
Figure 3.
Evaluation of the adaptive beamformer with normal hearing subjects (n = 9).

Hearing Impaired Subjects (n=6)
Speech Spectrum Noise
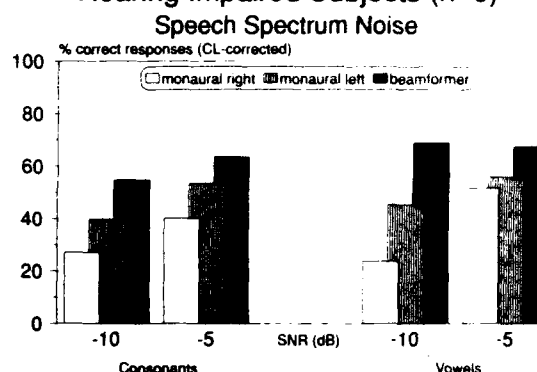
Hearing Impaired Subjects (n=6)
Cafeteria Noise
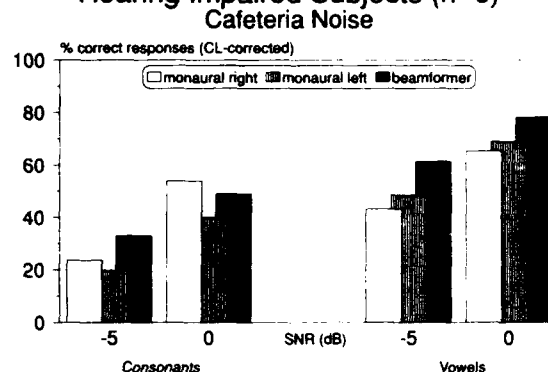
Figure 4.
Evaluation results of hearing-impaired subjects listening with their own hearing aid either to the unprocessed right or left ear signal or to the output of the adaptive beamformer. a) Speech spectrum shaped noise. b) With cafeteria noise.

beamformer did not have to be performed before the experiments were started. To include even more realistic situations such as moving and multiple sound sources, a part of the investigation was performed with stereophonic recordings of cafeteria noise.

Intelligibillity tests were first performed with nine normal-hearing subjects. As can be seen in **Figure 3**, intelligibility with the adaptive beamformer is highest compared with either one of the microphone signals or even with the binaural listening situation. This is true for all SNRs tested and consonant as well as vowel tests, the largest increase in intelligibility being achieved at low SNRs. Most of these differences were statistically significant.

The noise reduction scheme is primarily meant to be useful to hearing-impaired persons, and so it was also tested with six hearing-impaired volunteers, all of whom were regular users of conventional hearing aids **(Figure 4a)**. As for the normal-hearing subjects, the adaptive beamformer increased intelligibility, when compared with either one of the two unprocessed microphone signals.

The ability of the beamformer to cope with acoustically complex situations and several competing speakers as noise sources is documented by the increased intelligibility in the cafeteria setup **(Figure 4b)**. Note, however, that the differences between the two monaural conditions (directly irradiated and contralateral ear) could not be seen anymore, due to multiple sound sources from all directions. At 0 dB SNR, the beamformer did not provide any benefit for consonant recognition, which is probably due to erroneous behavior of the speech-detector algorithm

in some conditions. There was, however, still an improvement for vowel recognition which was statistically significant.

Conclusions

From our investigation, we conclude that three conditions are important for the design of an adaptive beamformer noise reduction for hearing aids: 1) an adaptation inhibition has to be provided; 2) filters of no less than approximately 500 taps should be used; and, 3) the delay should be set to a reasonable value, preferably somewhere between 25 and 50 percent of the filter length. When these requirements are met, the adaptive beamformer is able to improve intelligibility for normal-hearing as well as for hearing-impaired subjects, even in rooms with a realistic amount of reverberation and in complex acoustic situations, such as a cafeteria.

DIGITAL MULTIBAND LOUDNESS CORRECTION HEARING AID

In this section, an implementation of a new algorithm for a digital hearing aid is described which attempts to correct the loudness perception of hearing-impaired persons. A variety of multiband signal processing algorithms have been proposed and tested in the past (15,16,17,18,19,20,21,22). Some have failed, others showed large improvements in speech recognition scores. The aim of our study was the real-time implementation of an algorithm which restores normal loudness perception similar to the concept suggested by Villchur (23) and others.

A new aspect of our implementation concerns the close relation between psychoacoustic measurements and the ongoing analysis of the incoming signal to determine the required gains in the different frequency regions.

Methods

The digital master hearing aid consists of a DSP board for a 386-PC (Burr Brown PCI 20202C-1 with TMSC320C25) with analog-to-digital (A/D) and digital-to-analog (D/A) converters. Programmable filters and attenuators (custom-made PC-board) are used to prevent aliasing and to control the levels of the signals. In addition to the laboratory version, a wearable device was built which has been tested in preliminary field trials. The processing principle is similar to the frequency domain approach described by Levitt (24) and is illustrated in **Figure 5**.

Magnitude estimation procedures are used to determine loudness growth functions in eight frequency bands that are subsequently interpolated to obtain an estimate of the auditory field of a subject. Sinewave bursts of 8-10 different intensities within the hearing range are presented in random order. After each presentation, the subject judges the loudness of the sound on a continuous scale with annotated labels ranging from very soft to very loud. The responses are entered via a touch screen terminal. The whole procedure is repeated at eight different frequencies.

Figure 5 displays an example of such a magnitude estimation. It can be seen that the function of the hearing-impaired subject starts at a much higher intensity and is much steeper than the curve of the normal-hearing group. The difference between the two curves provides the intensity dependent hearing loss function at this frequency.

The sound signal is sampled at a rate of 10 kHz, segmented into windowed blocks of 12.8 ms and transformed via fast Fourier transform (FFT) into the frequency domain. The amplitude spectrum is modified according to preprogrammed gain tables. The modified spectrum is then transformed back into the time domain via an inverse FFT and reconstructed via an overlap-add procedure. The modification of the spectrum is done by multiplying each amplitude value with a gain factor from one of the gain tables. The selection of the tables is controlled by the amplitude values in the different frequency regions. Thus, the signal is nonlinearly amplified as a function of frequency and intensity.
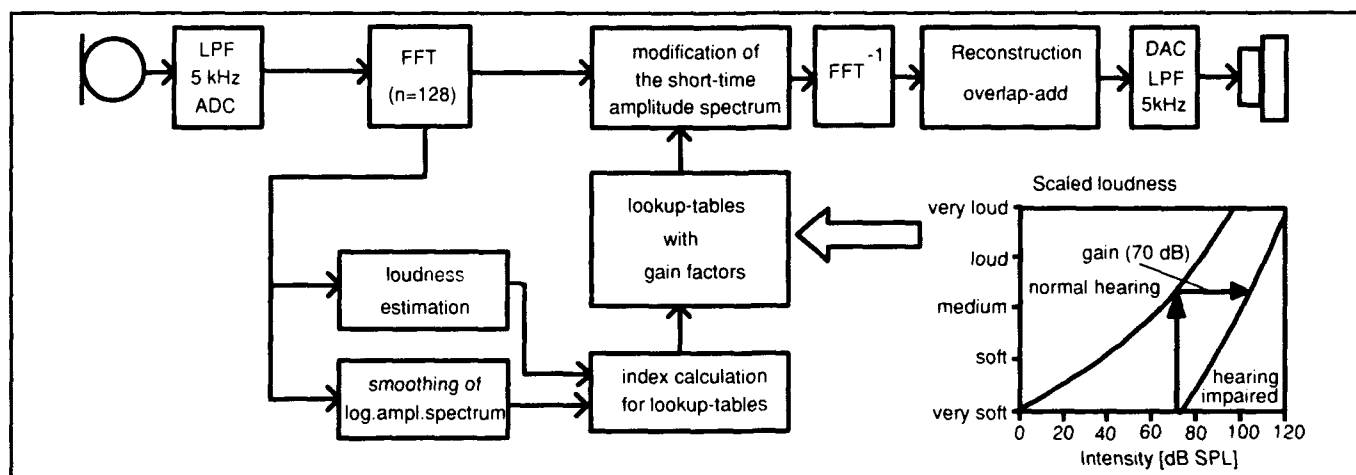
Figure 5.
Processing steps for digital multiband loudness correction algorithm.

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Gain tables are generated by interpolation of the measured hearing loss measurements over the whole frequency range. These functions describe how much gain is needed for every frequency to restore normal loudness perception. To prevent a spectrum flattening through independent modification of adjacent amplitudes, a smoothed spectrum is used to determine the gain factors. Thus, the fine structure of the spectrum will be preserved.

However, by determining the gain factors directly from the amplitude values, the reconstructed signal would generally become too loud, because the hearing loss functions were measured using narrow band signals and the incoming sounds consist mainly of broadband signals. The procedure was therefore modified to account for the loudness perception of complex sounds as follows (see **Figure 6** for an example of the processing steps): the frequency (f_{sin}) and amplitude (L_{inp}) of a sinewave signal are estimated which would produce a perceived loudness equal to that of the incoming complex signal. In this simple loudness estimation model, the intensity of this equivalent sinewave signal is determined as the total energy of the complex signal and its frequency as the frequency of the spectral gravity of the amplitude spectrum.

The spectral gravity was chosen because it always lies near the maximal energy concentration which is mainly responsible for loudness perception. The correct gain factors which will restore normal loudness perception (SL_N) are thus obtained by using the raised smoothed spectrum through the point of the equivalent sinewave signal.

Evaluation Results

The algorithm was tested with 13 hearing-impaired subjects in comparison with their own hearing aids under four different conditions in order to find out whether the scaling and fitting procedures were useful approaches for hard-of-hearing subjects and whether or how much the speech intelligibility could be improved with the multiband loudness correction (MLC) algorithm in quiet and noisy environments. The subjects had only minimal experience with the laboratory hearing aid, whereas they had used their own aids for more than a year. The function and fitting of the hearing aids were checked prior to the speech tests. Even though some hearing aids were equipped with an automatic gain control, the threshold of compression was set higher

than the levels used in the experiments. Subjects with moderate to severe flat sensorineural hearing loss were selected, which allowed simultaneous processing of the whole frequency range with sufficient numerical resolution. The same test procedure was used as in the previous section.

Figure 7 displays speech intelligibility scores for a consonant and vowel multiple choice rhyme test. At the higher intensity level (70 dB, open squares), which would correspond to an average speech communication level in quiet rooms, the scores with the subject's own hearing aids were rather high, especially in the vowel test. The difference between the new algorithm and the subject's own hearing aid was about 10 percent.

At reduced presentation levels (60 dB, open circles) which would be characteristic for a more difficult communication situation, with a speaker talking rather softly or from a more remote position, the scores with the conventional hearing aids became significantly lower for most subjects. The advantage of the loudness correction becomes more apparent. Almost all subjects achieved scores of from about 80 to 90 percent correct-item-identification, which was about as high as in the 70 dB condition. Subjects with very poor results with their own hearing aids profited most from the digital hearing aid.

In noisy conditions where the hearing aid users experience most communication problems at an SNR of 0 and − 5 dB (filled triangles and diamonds, respectively) the results were different from subject to subject. Most of them achieved higher scores with the digital hearing aid. Two subjects scored slightly worse with the DSP algorithm. One subject could not discriminate the words under this condition with her own hearing aid; with the MLC algorithm she obtained a score close to 50 percent.

Conclusions

The automated audiometric procedures to assess the frequency and intensity specific amount of hearing loss proved to be adequate for the calculation of processing parameters. The 13 subjects did not experience major problems in carrying out the measurements. All subjects who had not reached 100 percent intelligibility with their own hearing aids showed improved intelligibility of isolated words with the digital processing, especially at low levels. Some subjects also showed substantial discrimina-
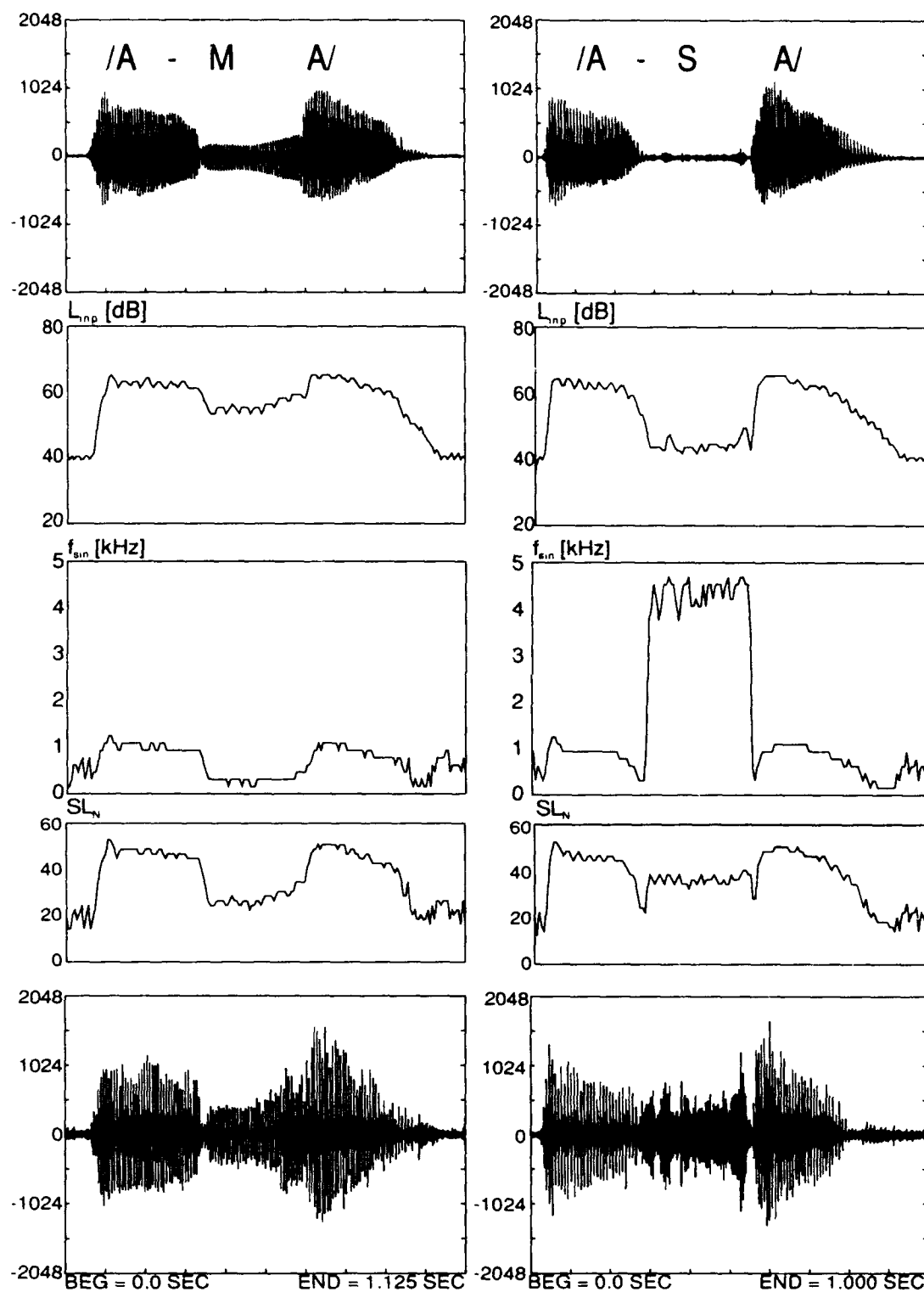
Figure 6.
Input time signals (top), input levels L_{inp} (corresponding to perceived loudness of narrow band signal), spectral gravity f_{sin}, equivalent scaled loudness SL_N, processed output signals (bottom). Speech tokens /A-MA/ (left) and /A-SA/ (right).
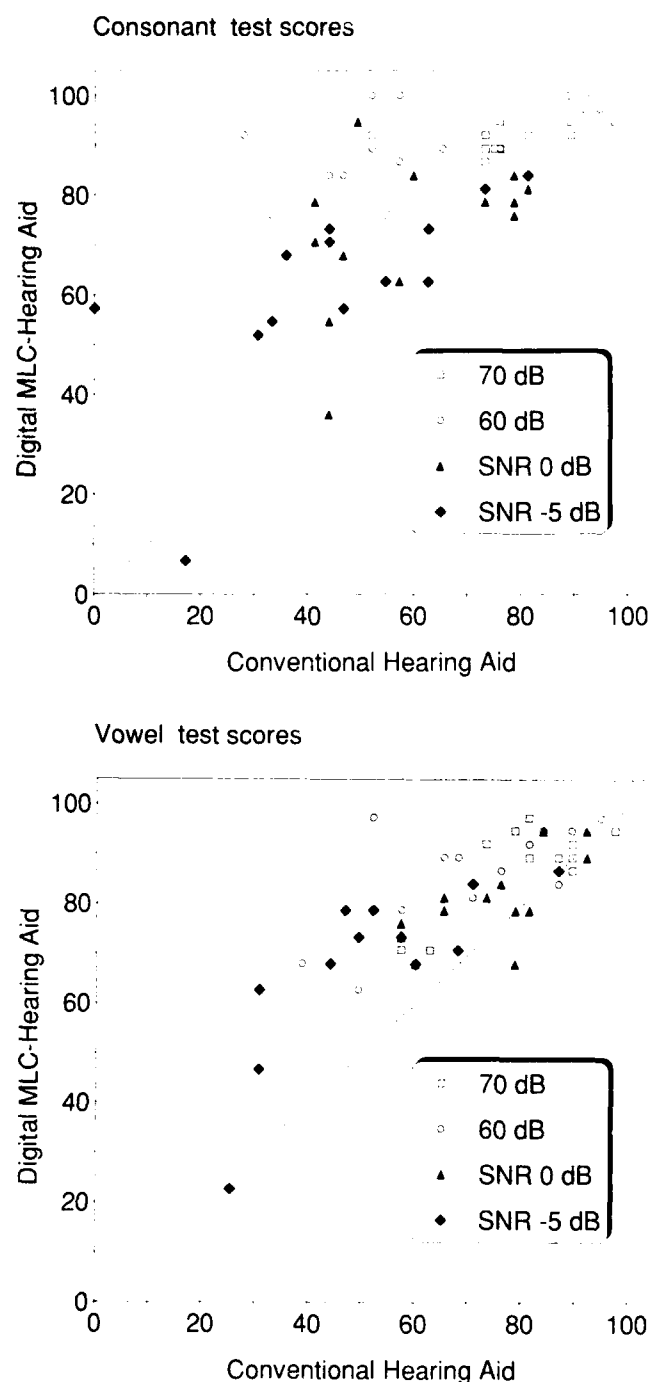
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Figure 7.
Comparison of speech intelligibility tests with own conventional hearing aid and digital MLC hearing aid for 13 subjects. Open symbols: tests in quiet; filled symbols: tests in speech spectrum shaped noise. a) Consonant test scores. b) Vowel test scores.

tion improvements in background noise, while others did not. A correlation between the hearing loss functions and the speech intelligibility scores was

not found. Other phenomena, such as reduced frequency and temporal resolution, might have to be considered in addition to loudness recruitment. It can be expected that fine tuning of processing parameters via, for example, a modified simplex procedure (25), or prolonged experience with a wearable signal processing hearing aid might further improve the performance of the MLC algorithm.

DIGITAL SIGNAL PROCESSING STRATEGIES FOR COCHLEAR IMPLANTS

Major research and development efforts to restore auditory sensations and speech recognition for profoundly deaf subjects have been devoted in recent years to signal processing strategies for cochlear implants. A number of technological and electrophysiological constraints imposed by the ʌnatomical and physiological conditions of the human auditory system have to be considered. One basic working hypothesis for cochlear implants is the idea that the natural firing pattern of the auditory nerve should be as closely approximated by electrical stimulation as possible. The central processor (the human brain) would then be able to utilize natural ("prewired" as well as learned) analysis modes for auditory perception. An alternative hypothesis is the Morse code idea, which is based on the assumption that the central processor is flexible and able to interpret any transmitted stimulus sequence after proper training and habituation.

Both hypotheses have never really been tested for practical reasons. On the one hand, it is not possible to reproduce the activity of 30,000 individual nerve fibers with current electrode technology. In fact, it is even questionable whether it is possible to reproduce the detailed activity of a single auditory nerve fiber via artificial stimulation. There are a number of fundamental physiological differences in firing patterns of acoustically versus electrically excited neurons which are hard to overcome (26). Spread of excitation within the cochlea and current summation are other major problems of most electrode configurations. On the other hand, the coding and transmission of spoken language requires a much larger communication channel bandwidth and more sophisticated processing than a Morse code for written text. Practical experiences with cochlear implants in the past indicate that some

natural relationships (such as growth of loudness and voice pitch variations) should be maintained in the encoding process. One might therefore conceive a third, more realistic, hypothesis as follows: Signal processing for cochlear implants should carefully select a subset of the total information contained in the sound signal and transform these elements into those physical stimulation parameters which can generate distinctive perceptions for the listener.

Many researchers have designed and evaluated different systems varying the number of electrodes and the amount of specific speech feature extraction and mapping transformations used (27). Recently, Wilson, et al. (28) reported astonishing improvements in speech test performance when they provided their subjects with high-rate pulsatile stimulation patterns rather than analog broadband signals. They attributed this effect partly to the decreased current summation obtained by nonsimultaneous stimulation of different electrodes (which might otherwise have stimulated in some measure the same nerve fibers and thus interacted in a nonlinear fashion) and partly to a fundamentally different, and possibly more natural, firing pattern due to an extremely high stimulation rate. Skinner, et al. (29) also found significantly higher scores on word and sentence tests in quiet and noise with a new multipeak digital speech coding strategy as compared with the formerly used F0F1F2-strategy of the Nucleus-WSP (wearable speech processor).

These results indicate the potential gains which may be obtained by optimizing signal processing schemes for existing implanted devices. The present study was conducted in order to explore new ideas and concepts of multichannel pulsatile speech encoding for users of the Clark/Nucleus cochlear prosthesis. Similar methods and tools can, however, be utilized to investigate alternative coding schemes for other implant systems.

Signal Processing Strategies

A cochlear implant digital speech processor (CIDSP) for the Nucleus 22-channel cochlear prosthesis was designed using a single-chip digital signal processor (TMS320C25, Texas Instruments) (30,31). For laboratory experiments, the CIDSP was incorporated in a general purpose computer which provided interactive parameter control, graphical display of input/output and buffers, and offline speech file processing facilities. The experiments described in this paper were all conducted using the laboratory version of CIDSP.

Speech signals were processed as follows: after analog low-pass filtering (5 kHz) and A/D conversion (10 kHz), preemphasis and Hanning windowing (12.8 ms, shifted by 6.4 ms or less per analysis frame) was applied and the power spectrum calculated via FFT; specified speech features such as formants and voice pitch were extracted and transformed according to the selected encoding strategy; finally, the stimulus parameters (electrode position, stimulation mode, and pulse amplitude and duration) were generated and transmitted via inductive coupling to the implanted receiver. In addition to the generation of stimulus parameters for the cochlear implant, an acoustic signal based on a perceptive model of auditory nerve stimulation was output simultaneously.

Two main processing strategies were implemented on this system: The first approach, Pitch Excited Sampler (PES), is based on the maximum peak channel vocoder concept whereby the time-averaged spectral energies of a number of frequency bands (approximately third-octave bands) are transformed into appropriate electrical stimulation parameters for up to 22 electrodes (Figure 8, top). The pulse rate at any given electrode is controlled by the voice pitch of the input speech signal. A pitch extractor algorithm calculates the autocorrelation function of a lowpass-filtered segment of the speech signal and searches for a peak within a specified time lag interval. A random pulse rate of about 150 to 250 Hz is used for unvoiced speech portions.

The second approach, Continuous Interleaved Sampler (CIS), uses a stimulation pulse rate that is independent of the fundamental frequency of the input signal. The algorithm continuously scans all frequency bands and samples their energy levels (Figure 8, middle and bottom). Since only one electrode can be stimulated at any instant of time, the rate of stimulation is limited by the required stimulus pulse widths (determined individually for each subject) and the time to transmit additional stimulus parameters. As the information about the electrode number, stimulation mode, and pulse amplitude and width is encoded by high frequency bursts (2.5 MHz) of different durations, the total transmission time for a specific stimulus depends on all of these parameters. This transmission time can
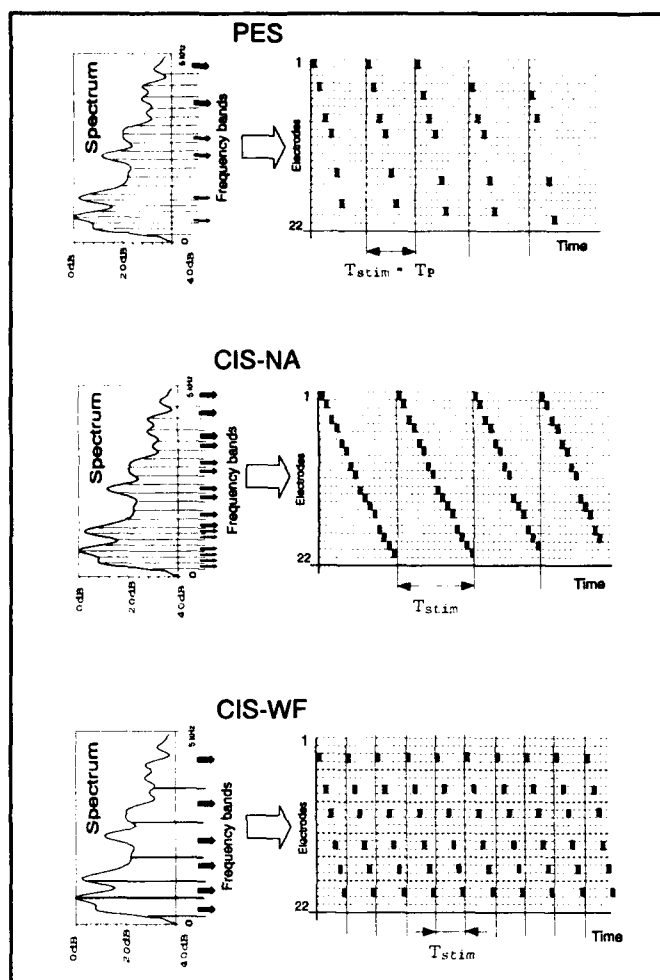
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Figure 8.
Three CI-DSP strategies: pitch excited sampler (PES), continuous interleaved sampler with narrow band analysis (CIS-NA), continuous interleaved sampler with wide band analysis and fixed tonotopic mapping (CIS-WF).

be minimized by choosing the shortest possible pulse width combined with the maximal amplitude.

In order to achieve the highest stimulation rates for those portions of the speech input signals which are assumed to be most important for intelligibility, several modifications of the basic CIS strategy were designed, of which only the two most promising (CIS-NA and CIS-WF) will be considered in the following. The analysis of the short time spectra was performed either for a large number of narrow frequency bands (corresponding directly to the number of available electrodes) or for a small number (typically six) of wide frequency bands analogous to the approach suggested by Wilson, et al. (28). The frequency bands were logarithmically

spaced from 200 to 5,000 Hz in both cases. Spectral energy within any of these frequency bands was mapped to stimulus amplitude at a selected electrode as follows: all narrow band analysis channels whose values exceeded a noise cut level (NCL) were used for CIS-NA, whereas all wide band analysis channels irrespective of NCL were mapped to preselected fixed electrodes for CIS-WF. Both schemes are supposed to minimize electrode interactions by preserving maximal spatial distances between subsequently stimulated electrodes. The first scheme (CIS-NA) emphasizes spectral resolution while the second (CIS-WF) optimizes fine temporal resolution. In both the PES and the CIS strategies, a high-frequency preemphasis was applied whenever a spectral gravity measure exceeded a preset threshold.

Subjects

Evaluation experiments have been conducted with five postlingually deaf adult (ages 26-50 years) cochlear implant users to date. All subjects were experienced users of their speech processors. The time since implantation ranged from 5 months (KW) to nearly 10 years (UT, single-channel extracochlear implantation in 1980, reimplanted after device failure in 1987) with good sentence identification (80-95 percent correct responses) and number recognition (40-95 percent correct responses) performance and minor open speech discrimination in monosyllabic word tests (5-20 percent correct responses, all tests presented via computer, hearing-alone) and limited use of the telephone. One subject (UT) still used the old wearable speech processor (WSP) which extracts only the first and second formant and thus stimulates only two electrodes per pitch period. The other four subjects used the new Nucleus Miniature Speech Processor (MSP) with the so-called multipeak strategy whereby, in addition to first and second formant information, two or three fixed electrodes may be stimulated to convey information contained in two or three higher frequency bands.

The same measurement procedure to determine thresholds of hearing (T-levels) and comfortable listening (C-levels) used for fitting the WSP or MSP was also used for the CIDSP strategies. Only minimal exposure to the new processing strategies was possible due to time restrictions. After about 5 to 10 minutes of listening to ongoing speech, 1 or 2 blocks of a 20-item 2-digit number test were carried out. There was no feedback given during the test

trials. All test items were presented by a second computer which also recorded the responses of the subjects entered via touch-screen terminal (for multiple-choice tests) or keyboard (numbers tests and monosyllable word tests). Speech signals were either presented via loudspeaker in a sound-treated room (when patients were tested with their WSPs) or processed by the CIDSP in real time and fed directly to the transmitting coil at the subject's head. Different speakers were used for the ongoing speech, the numbers test, and the actual speech tests, respectively.

Results and Discussion

Results of 12-consonant (/aCa/) and 8-vowel (/dV/) identification tests are shown in **Figure 9**. The average scores for consonant tests (**Figure 9a**) with the subject's own wearable speech processor were significantly lower than with the new CIDSP strategies. The pitch-synchronous coding (PES) resulted in worse performance compared with the coding without explicit pitch extraction (CIS-NA and CIS-WF). Vowel identification scores (**Figure 9b**), on the other hand, were not improved by modifications of the signal processing strategy.

The results for subject HS indicated that PES may provide better vowel identification for some subjects, while CIS was better for consonant identification for all tested subjects. It is possible that CIS is able to present the time-varying spectral information associated with the consonants better than can PES. Results from a male-female speaker identification test indicated that no speaker distinctions could be made using CIS, unlike PES which yielded very good speaker identification scores. This suggested that voice pitch information was well transmitted with PES but not at all with CIS.

Hybrids of the two strategies listed above were, therefore, developed with the aim of combining the respective abilities of PES and CIS in transmitting vowel and consonant information, as well as retaining PES' ability to transmit voice pitch information with the resultant hybrids. In one hybrid, the stimulation was switched between PES and CIS respectively, depending on whether the input speech signal was voiced or not. In another hybrid, for voiced portions of the speech, the lowest frequency active electrode was stimulated using PES while the remaining active electrodes were, at the same time, stimulated using CIS. For unvoiced portions, all
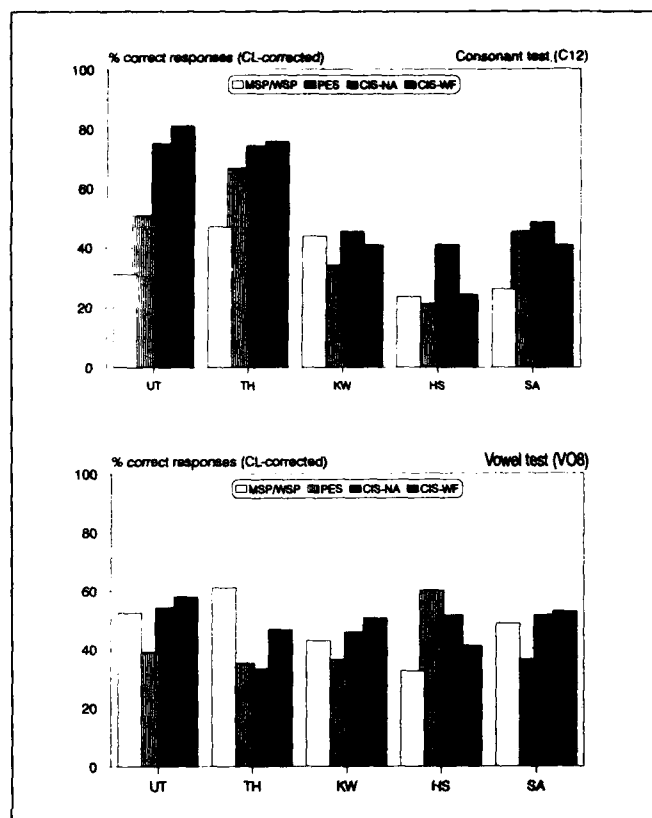
Figure 9.
Speech test results with five implantees, four processing conditions. a) Consonant test results. b) Vowel test results.

active electrodes used CIS stimulation. A third hybrid is a variation on the one above, using PES on the two lowest frequency active electrodes instead of only one.

Figure 10 shows the differences in electrode activation patterns between the Nucleus-MSP (**Figure 10a**) and the PES/CIS hybrid strategy (**Figure 10b**). The graphs were generated from direct measurements of the encoded high frequency transmission signals emitted by the induction coil of the speech processor. Using the subject's programming map, which determines the relationships between stimulus amplitude and perceived loudness, relative stimulus levels (ranging from 0 to 100 percent) were obtained which are mapped into a color (or black and white) scale for the purpose of illustration. The axes of the graphs have been arranged similar to a spectrogram, with time on the abscissa and frequency (tonotopical order of active electrodes) on the ordinate. A number of striking differences between the two coding schemes became apparent

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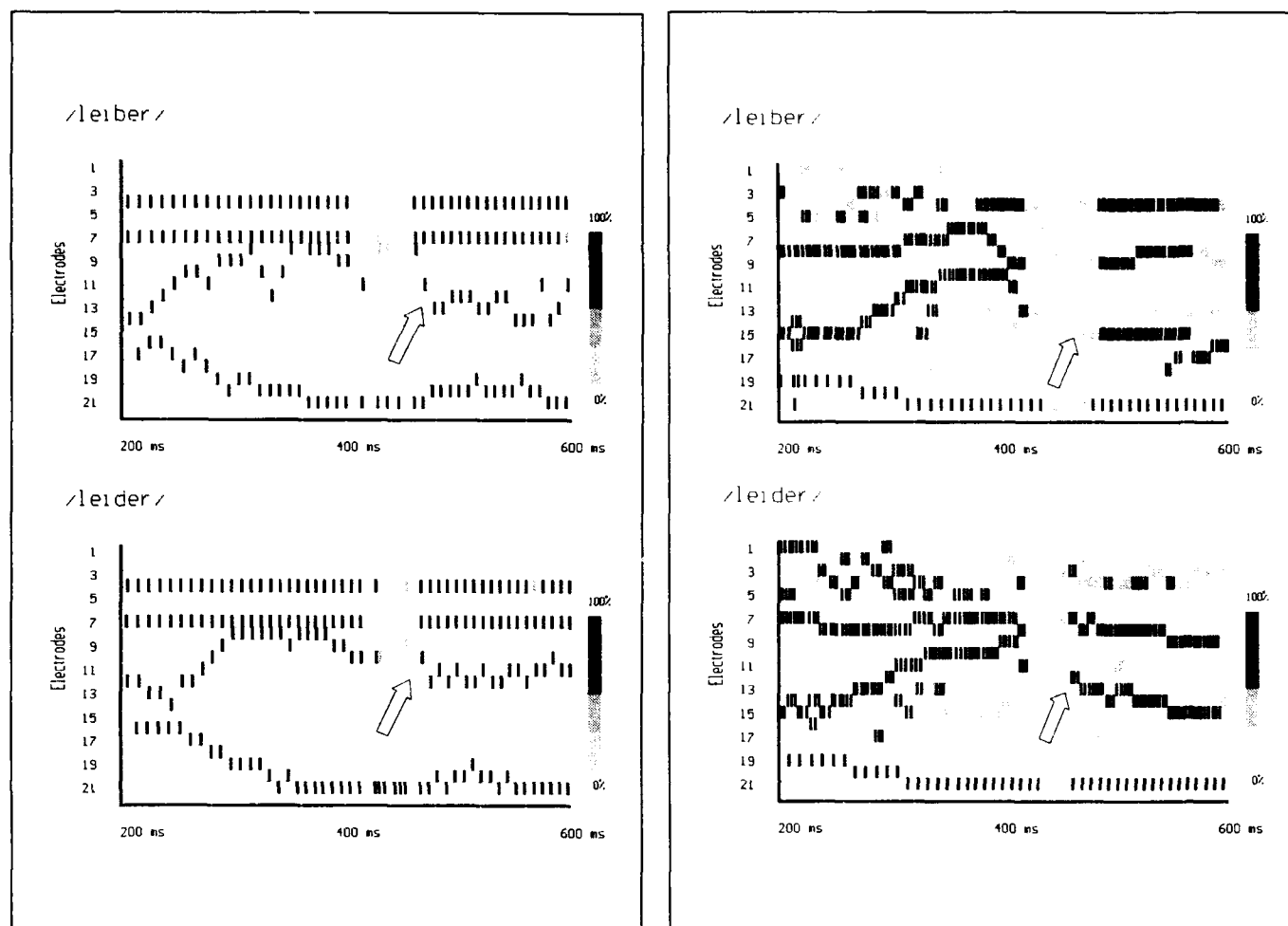
Figure 10.
"Electrodograms" of the phonological consonant minimal pair /leiber - leider/. a) Processed by the Nucleus-MSP. b) Processed by a new hybrid processing strategy where the lowest electrode is excited pitch-synchronously (PES-mode) and the higher electrodes at maximal rate (CIS-mode). The start of the voiced plosive sounds at about 450 ms is indicated by arrows.

from these displays. The first and second formant trajectories can be clearly seen in the MSP examples. Due to the smaller number of electrodes assigned to the first formant region in the case of the hybrid strategy, the variations in first formant frequency are less apparent and cover only three electrodes compared with six electrodes with the MSP. The rate of stimulation, however, is virtually identical for the lowest trajectory in both cases. The two highest stimulus trajectories in the MSP are assigned to fixed electrode numbers (4 and 7), whereas in the hybrid strategy, there is distinctly more variation visible in those areas than would be expected. The most striking difference between the MSP and the hybrid strategies, however, is the high-rate CIS

portion with rather large variations in stimulus levels across electrodes and over time, as opposed to the more uniform and scarce stimulus pattern for the MSP. It can be seen that a distinction between the voiced plosive phonemes /b/ and /d/ is virtually impossible for the MSP, whereas with the hybrid (or CIS) the second formant transition between about 460 and 490 msec becomes visible and might be utilized by the implantees.

Preliminary experiments with these hybrid coding strategies have indeed shown some improvement in consonant identification. The results suggested, however, that the presence of voicing within time-varying portions could interfere with the perception of the CIS encoded information. Implantees were

able to perform male-female speaker identifications quite well with all hybrids, indicating that the voice pitch information encoded into the PES stimulation within the hybrids can be perceived and used. This implies that by activating one or two low frequency electrodes using PES, it is possible to incorporate voice pitch information into an essentially CIS-like coding strategy.

Conclusions

The above speech test results are still preliminary due to the small number of subjects and test conditions. It is, however, very promising that new signal processing strategies can improve speech discrimination considerably during acute laboratory experiments. Consonant identification apparently may be enhanced by more detailed temporal information and specific speech feature transformations. Whether these improvements will pertain in the presence of interfering noise remains to be verified.

The experiments using hybrid coding strategies combining PES and CIS stimulation indicate that there is even more potential for improvement of the transmission of information regarding particular consonants. Further studies will have to be carried out to investigate in greater detail the perceptual interactions that arise between PES and CIS stimulation within the resultant stimuli. The results also show that voice pitch information, added to a CIS-like coding strategy by encoding it into one or two low frequency electrode(s), can be perceived and used by a cochlear implant user. Further optimization of these processing strategies preferably should be based on more specific data about loudness growth functions for individual electrodes or additional psychophysical measurements.

Although many aspects of speech encoding can be efficiently studied using a laboratory digital signal processor, it would be desirable to allow subjects more time for adjustment to a new coding strategy. Several days or weeks of habituation are sometimes required until a new mapping can be fully exploited. Thus, for scientific as well as practical purposes, the further miniaturization of wearable DSPs will be of great importance.

SUMMARY

The application of digital signal processors for the hearing impaired was demonstrated with three examples:

• Optimal parameters for a two-microphone adaptive beamformer noise reduction algorithm were determined by simulations and measurements for realistic environments. The experimental evaluation was carried out with normal-hearing and hearing-impaired subjects using a real time implementation on a floating point DSP. Results indicated the potential benefit of this method even in moderately reverberant rooms when a speech detection algorithm is included and an adequate filter length for the least mean squares (LMS) algorithm is used.
• A digital hearing aid with a multiband frequency domain modification of the short-time amplitude spectrum was fitted using loudness scaling measurements and evaluated with 13 hard-of-hearing subjects. The measurements that used narrow band signals were converted into gain factors via a simplified loudness estimation model based on the calculation of the spectral gravity point and the total energy within a short segment of the input signal. Significant improvements in speech recognition scores were found in quiet as well as in noisy conditions.
• A variety of high-rate stimulation algorithms were implemented on an experimental digital signal processor for the Nucleus multielectrode cochlear implant and compared with the performance of the patient's own wearable and miniature speech processor. It was found that consonant identification could be significantly improved with the new strategies in contrast to vowel identification, which remained essentially unchanged. Degradation in perceived sound quality due to loss of voice pitch information may be compensated by hybrid strategies.

ACKNOWLEDGMENTS

Design and evaluation of a continuous interleaved sampling (CIS) processing strategy for multichannel cochlear implants

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Abstract—Two approaches for representing speech information with multichannel cochlear prostheses are being compared in tests with implant patients. Included in these studies are the *compressed analog* (CA) approach of a standard clinical device and research processors utilizing *continuous interleaved sampling* (CIS). Initial studies have been completed with nine subjects, seven of whom were selected on the basis of excellent performance with the Ineraid clinical processor, and the remaining two for their relatively poor performance with the same device. The tests include open-set recognition of words and sentences. Every subject has obtained a higher score—or repeated a score of 100% correct—on every test when using a CIS processor. These results are discussed in terms of their implications for processor design.

Key words: *cochlear prosthesis, deafness, hearing, speech perception, speech processing.*

INTRODUCTION

Recent studies in our laboratory have focused on comparisons of *compressed analog* (CA) and *continuous interleaved sampling* (CIS) processors (1,2,3). Both use multiple channels of intracochlear electrical stimulation, and both represent waveforms or envelopes of speech input signals. No specific features of the input, such as the fundamental or formant frequencies, are extracted or explicitly represented. CA processors use continuous analog signals as stimuli, whereas CIS processors use pulses. The CA approach is used in the widely applied Ineraid device (4,5) and in the now-discontinued UCSF/Storz device, with some differences in details of processor implementation (6). Wearable devices capable of supporting the CIS approach are just becoming available for use in clinical settings.

To date, we have completed initial studies of nine subjects—seven of whom were selected for their high levels of speech recognition with the Ineraid CA processor, and two who were selected for their relatively poor performances with that processor. The "high performance" subjects were representative of the best results when any commercially available implant system is used (2). Equivalent studies have been begun but not yet completed with two additional patients in the "poor performance" group (7).

This paper will briefly review the previously published results for the seven subjects in the high performance group and present preliminary results for the first two subjects from the poor performance group.

PROCESSING STRATEGIES

The designs of CA and CIS processors are illustrated in **Figure 1** and **Figure 2**. In CA proces-

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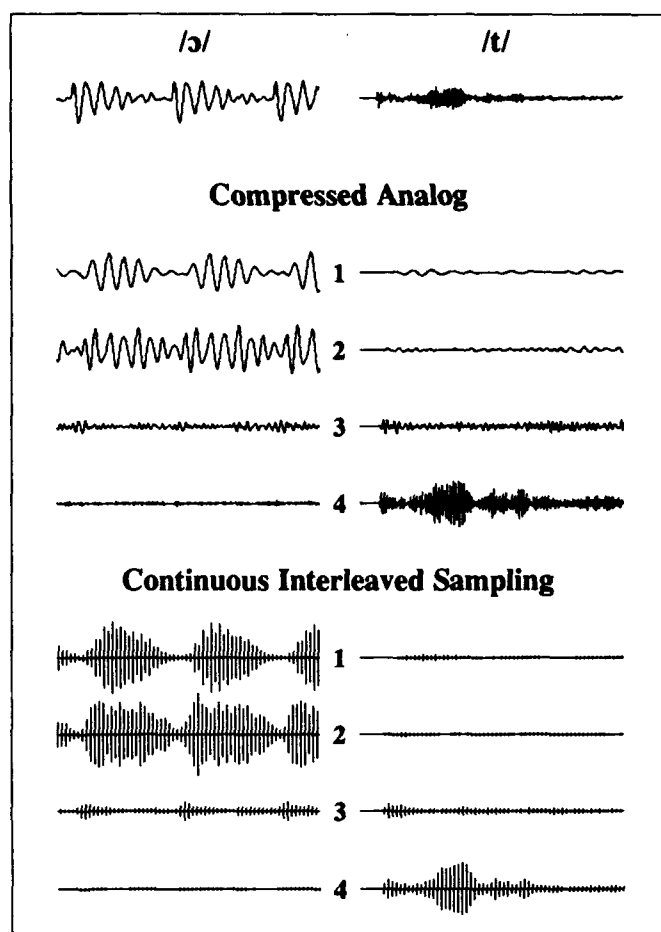Section III. Digital Techniques Applied to Related Areas: Wilson et al.

Figure 1.
Waveforms produced by simplified implementations of CA and CIS strategies. The top panel shows preemphasized (6dB/octave attenuation below 1.2 kHz) speech inputs. Inputs corresponding to a voiced speech sound ("aw") and an unvoiced speech sound ("t") are shown in the left and right columns, respectively. The duration of each trace is 25.4 ms. The remaining panels show stimulus waveforms for CA and CIS processors. The waveforms are numbered by channel, with channel 1 delivering its output to the apical-most electrode. To facilitate comparisons between strategies, only four channels of CIS stimulation are illustrated here. In general, five or six channels have been used for that strategy. The pulse amplitudes reflect the envelope of the bandpass output for each channel. In actual implementations the range of pulse amplitudes is compressed using a logarithmic or power-law transformation of the envelope signal. (From Wilson BS, et al. (2), with permission.)

Figure 2.
Expanded display of CIS waveforms. Pulse duration per phase ("d") and the period between pulses on each channel ("1/rate") are indicated. The sequence of stimulated channels is 4-3-2-1. The total duration of each trace is 3.3 ms. (From Wilson BS, et al. (2), with permission.)

sors, a microphone signal varying over a wide dynamic range is compressed or restricted to the narrow dynamic range of electrically evoked hearing (8,9) using an automatic gain control. The resul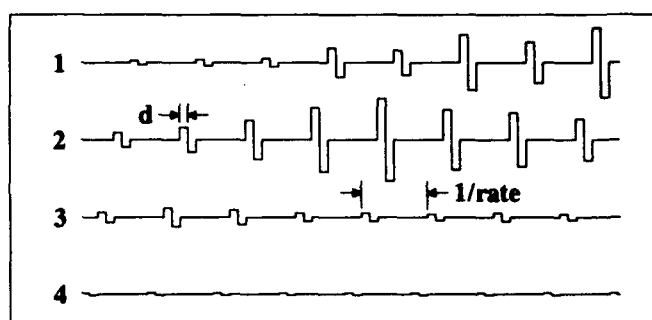ting signal is then filtered into four contiguous frequency bands for presentation to each of four electrodes. As shown in **Figure 1**, information about speech sounds is contained in the relative stimulus amplitudes among the four electrode channels and in the temporal details of the waveforms for each channel.

A concern associated with this method of presenting information is that substantial parts of it may not be perceived by implant patients (10). For example, most patients cannot perceive frequency changes in stimulus waveforms above about 300 Hz (11). Thus, many of the temporal details present in CA stimuli probably are not accessible to the typical user.

In addition, the simultaneous presentation of stimuli may produce significant interactions among channels through vector summation of the electric fields from each electrode (12). The resulting degradation of channel independence would be expected to reduce the salience of channel-related cues. That is, the neural response to stimuli from one electrode may be significantly distorted, or even counteracted, by coincident stimuli from other electrodes.

The CIS approach addresses the problem of such channel interactions through the use of interleaved nonsimultaneous stimuli (**Figure 2**). Trains of balanced biphasic pulses are delivered to each electrode with temporal offsets that eliminate any overlap across channels. The amplitudes of the pulses are derived from the envelopes of bandpass filter outputs. In contrast with the four-channel clinical CA processors, five or six bandpass filters (and channels of stimulation) have generally been

used in CIS systems to take advantage of additional implanted electrodes and reduced interactions among channels. The envelopes of the bandpass outputs are formed by rectification and lowpass filtering. Finally, the amplitude of each stimulus pulse is determined by a logarithmic or power-law transformation of the corresponding channel's envelope signal at that time. This transformation compresses each signal into the dynamic range appropriate for its channel.

A key feature of the CIS approach is its relatively high rate of stimulation on each channel. Other pulsatile strategies present sequences of interleaved pulses across electrodes at a rate equal to the estimated fundamental frequency during voiced speech and at a jittered or fixed (often higher) rate during unvoiced speech (13,14,15). Rates of stimulation on any one channel have rarely exceeded 300 pulses per second (pps). In contrast, the CIS strategy generally uses brief pulses and minimal delays, so that rapid variations in speech can be tracked by pulse amplitude variations. The rate of stimulation on each channel usually exceeds 800 pps and is constant during both voiced and unvoiced intervals. A constant high rate allows relatively high cutoff frequencies for the lowpass filters in the envelope detectors. With a stimulus rate of 800 pps, for instance, lowpass cutoffs can approach (but not exceed) 400 Hz without introducing aliasing errors in the sampling of the envelope signals at the time of each pulse. See Rabiner and Shafer for a complete discussion of aliasing and its consequences (16).

METHODS

Each of the nine subjects has been studied for a 1-week period during which: (a) basic psychophysical measures were obtained on thresholds and dynamic ranges for pulsatile stimuli; (b) various CIS processors (with different choices of processor parameters) were evaluated with preliminary tests of consonant identification; and, (c) performance with the best of the CIS processors and the clinical CA processor was documented with a broad spectrum of speech tests. Experience with the clinical processor exceeded one year of daily use for all subjects. In contrast, experience with the CIS processors was limited to no more than several hours before formal testing.

Tests

The comparison tests include open-set recognition of 50 one-syllable words from Northwestern University Auditory Test 6 (NU-6), 25 two-syllable words (spondees), 100 key words in the Central Institute for the Deaf (CID) sentences of everyday speech, and the final word in each of 50 sentences from the Speech Perception in Noise (SPIN) Test (here presented without noise). All tests are conducted with hearing alone, using single presentations of recorded material, and without feedback about correct or incorrect responses.

Processor Parameters

Each subject's own clinical device is used for the tests with the CA processor. As mentioned above, selection of parameters for the CIS processor is guided by preliminary tests of consonant identification. The standard four channels of stimulation are used for the clinical CA processors (4,5), whereas five or six channels have been used for the CIS processors. Additional parameters of the CIS processors are presented in **Table 1**. As indicated there, all CIS processors for the high performance subjects, SR2 to SR8, have had pulse durations of 102 μs/phase or less, zero delay between the sequential pulses on different channels, pulse rates of 817 pps or higher on each channel, and a cutoff frequency for the lowpass filters of 400 Hz or higher. The best processor for subject SR1 also fit this description, except that a delay of 172 μs was interposed between sequential pulses. The best processor for subject SR10 used long-duration pulses (167 μs/phase), paired with a relatively low rate of stimulation on each channel (500 pps) and a relatively low cutoff frequency for the lowpass filters (200 Hz).

Evaluation of Practice and Learning Effects

Because the tests with the CA processor preceded those with the selected CIS processor for each subject, we were concerned that practice or learning effects might favor the latter in comparisons of the two strategies. To evaluate this possibility, the CID and NU-6 tests were repeated with the CIS processor for five of the high performance subjects (subjects SR3, SR4, SR6, SR7, and SR8). A different recorded speaker and new lists of words and sentences were used. Practice or learning effects would be demonstrated by significant differences in the

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Table 1.
Parameters of CIS processors.

Subject	Channels	Pulse Duration (μs/phase)	Rate (pps)	Integrating Filter Cutoff (Hz)
SR2	6	55	1515	800
SR3	6	31	2688	400
SR4	6	63	1323	400
SR5	6	31	2688	800
SR6	6	102	817	400
SR7	5	34	2941	400
SR8	6	100	833	400
SR1	5	34	833	400
SR10	6	167	500	200

Parameters include number of channels, pulse duration, the rate of stimulation on each channel (Rate), and the cutoff frequency of the lowpass integrating filters for envelope detection (Integrating Filter Cutoff). The subjects are listed in the chronological order of their participation in the present studies. SR2 through SR8 are the "high performance" subjects while SR1 and SR10 belong to the "low performance" group. Additional processor parameters may be found in References 3 and 7.

test/retest scores. Nevertheless, no such differences were found (p > 0.6 for paired t comparisons of the CID scores; p > 0.2 for the NU-6 scores), and the scores from the first and second tests were averaged for all subsequent analyses.

RESULTS

The results from 1-week studies of each of the nine subjects are presented in **Table 2** and **Figure 3**. Scores for the high performance subjects are indicated by the light lines near the top of each panel in **Figure 3**, and scores for the two low performance subjects are indicated by the dark lines closer to the bottom of each panel. We note that low performance subject SR1 had participated in an earlier study not involving CIS processors (15). Results from his first week of testing with CIS processors are presented here. This is also true of high performance subject SR2, who has returned to the laboratory for many additional studies with various implementations of CIS processors (1). In those subsequent tests, SR2 has achieved even higher scores using a variety of six-channel CIS processors, with NU-6 percentages ranging from the high 80s to the low 90s.

As is evident from the figure, all nine subjects have scored higher, or repeated a score of 100

percent correct, on every test, when using a CIS processor. The average scores across subjects increased from 64 to 86 percent correct on the spondee test (p < 0.01), from 70 to 91 percent correct on the CID test (p < 0.02), from 39 to 76 percent correct on the SPIN test (p < 0.001), and from 34 to 54 percent correct on the NU-6 test (p < 0.0002).

Perhaps the most encouraging of these results are the improvements for the two low performance subjects. CA scores were low for SR1 and quite poor for SR10. Substitution of a CIS processor produced large gains in speech recognition for both subjects. Indeed, with the CIS processor SR1 has scores that fall within the ranges of CA processor scores that qualified subjects SR2 to SR8 as among the best performers with any clinical device.

Similarly, SR10 achieved relatively high scores with the CIS processor. The score on the spondee test increased from 0 to 56 percent correct, on the CID test from 1 to 55 percent correct, on the SPIN test from 0 to 26 percent correct, and on the NU-6 test from 0 to 14 percent correct. These increases were obtained with no more than several hours of aggregated experience with CIS processors, compared with more than a year of daily experience with the clinical CA processor.

Note that although these gains for SR10 are large, they are not atypical of results for the other subjects. His improvements follow the pattern of the

Table 2.
Individual results from the open-set tests.

Subject	Spondee		CID		SPIN		NU-6	
	CA	CIS	CA	CIS	CA	CIS	CA	CIS
SR2	92	96	100	100	78	96	56	80
SR3	52	96	66	98	14	92	34	58
SR4	68	76	93	95	28	70	34	40
SR5	100	100	97	100	94	100	70	80
SR6	72	92	73	99	36	74	30	49
SR7	80	100	99	100	66	98	38	71
SR8	68	100	80	100	36	94	38	66
SR1	40	60	25	70	2	30	6	32
SR10	0	56	1	55	0	26	0	14

other subjects, i.e., generally large gains in the scores of tests that are not limited by ceiling effects. The distinctive aspect of SR10's results is that he enjoys such gains even though he started at or near zero on all four tests. Thus, the relative improvements for SR10 are larger than those for any other subject in the series thus far.

DISCUSSION

The findings presented in this paper demonstrate that the use of CIS processors can produce large and immediate gains in speech recognition for a wide range of implant patients. Indeed, the sensitivity of some of the administered tests has been limited by ceiling (or saturation) effects: five of the seven high performance subjects scored 96 percent or higher for the spondee test using CIS processors; all seven scored 95 percent or higher for the CID test; and five scored 92 percent or higher for the SPIN test. Scores for the NU-6 test, although not approaching the ceiling, were still quite high. The 80 percent score achieved by two of the subjects corresponds to the middle of the range of scores obtained by people with mild-to-moderate hearing losses when taking the same test (17,18).

The improvements are even more striking when one considers the large disparity in experience with the two processors. At the time of our tests each subject had 1 to 5 years of daily experience with the

CA processor but only several hours over a few days with CIS. In previous studies involving within-subjects comparisons, such differences in experience have strongly favored the processor with the greatest duration of use (19,20,21).

Factors contributing to the performance of CIS processors might include: (a) reduction in channel interactions through the use of nonsimultaneous stimuli; (b) use of five or six channels instead of four; (c) representation of rapid envelope variations through the use of relatively high pulse rates; (d) preservation of amplitude cues with channel-by-channel compression; and, (e) the shape of the compression function.

An interesting aspect of the ongoing studies with low performance subjects, represented here by SR1 and SR10, is that the best CIS processors seem to involve parameters distinct from those of the best processors for subjects in the high performance group. The best processor for SR1 used short-duration pulses (34 μs/phase) presented at a relatively low rate (833 pps), and the best processor for SR10 used long-duration pulses (167 μs/phase) presented at an even lower rate (500 pps). The subjects in the high performance group, however, often obtained their best scores with processors tending to minimize pulse widths and maximize pulse rates (e.g., 31 μs/phase pulses presented at 2688 pps).

The use of such shorter pulses and higher rates allows representation of higher frequencies in the modulation waveform for each channel (i.e., the
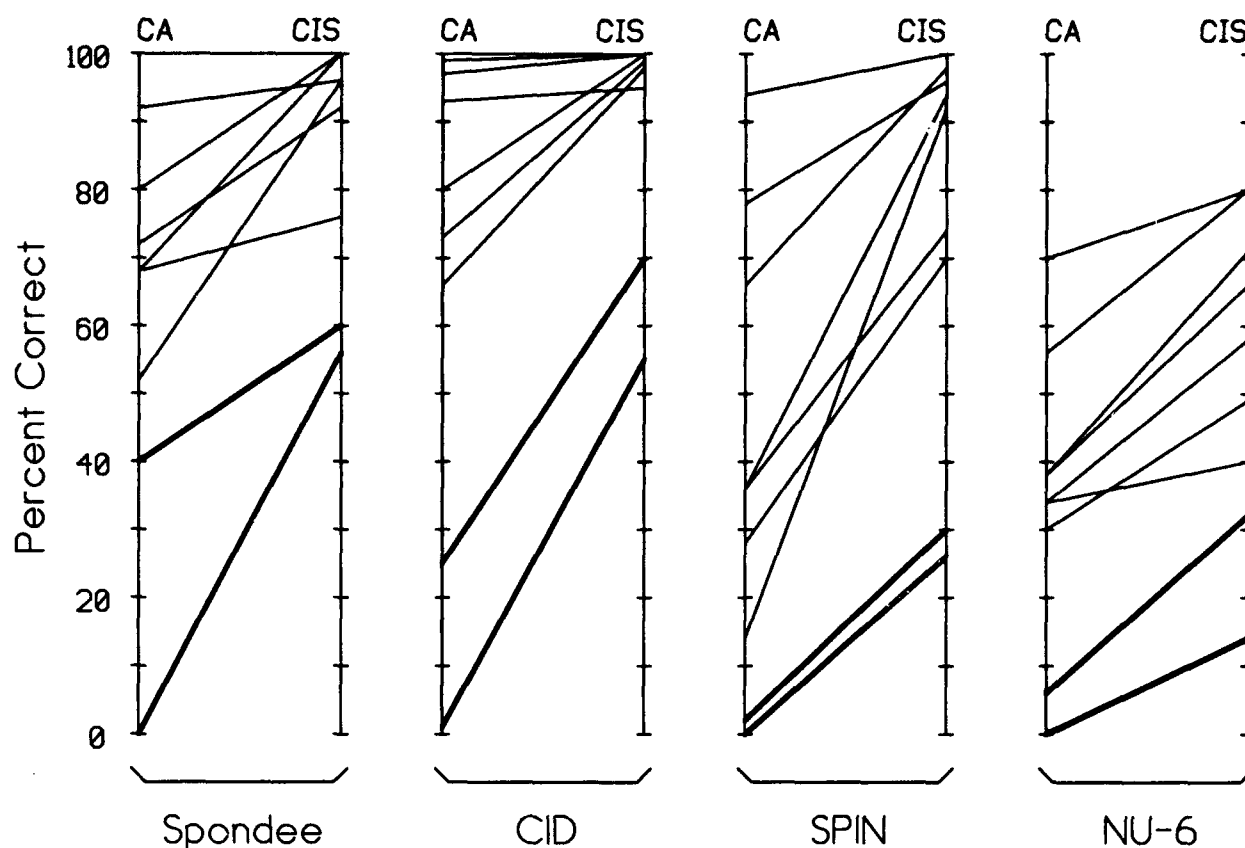
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Figure 3.
Speech recognition scores for CA and CIS processors. A line connects the CA and CIS scores for each subject. Light lines correspond to the seven subjects selected for their excellent performance with the clinical CA processor, whereas the heavier lines correspond to the two subjects selected for relatively poor performance.

cut-off frequency of the lowpass filter in the envelope detectors for each channel may be raised to one-half of the pulse rate without introducing aliasing effects). In addition, the dynamic range (DR) of electrical stimulation—from threshold to most comfortable loudness—is a strong function of pulse rate and a weaker function of pulse duration (11,22). Large increases in DR are generally found with increases in pulse rates from about 400 pps to 2500 pps. Smaller increases often (but not always) are observed with increases in pulse duration (at a fixed rate of stimulation) from roughly 50 μs/phase to higher values (e.g., out to 200 μs/phase for practical CIS designs).

For some patients, however, these advantages may be outweighed by other factors. For several subjects in our Ineraid series, for instance, we have observed that the salience of channel ranking can decline with decreases in pulse widths below 100 μs/phase. A favorable tradeoff for such subjects might involve the use of long-duration pulses (e.g., 100 μs/phase or greater) to preserve channel cues, while foregoing any additional DR obtainable with shorter pulses and higher rates of stimulation.

Another possible advantage of relatively low rates of stimulation is further reduction of channel interactions. Providing time between pulses on sequential channels can reduce the "temporal integration" component of channel interactions—a component produced by the accumulation of charge at neural membranes from sequential stimuli (12). Thus, use of time delays between short-duration pulses in the stimulation sequence across electrodes may reduce interactions. Alternatively, use of long-duration pulses with no time delay also might reduce temporal interactions in that a relatively long period still is realized between the excitatory phases of successive pulses.

Collectively, the present results indicate that: (a) the performance of at least some patients with poor clinical outcomes can be improved substantially with use of a CIS processor; (b) use of long-duration pulses produced large gains in speech test scores for one such subject; (c) use of short-duration pulses presented at a relatively low rate produced similar improvements in another such subject; and, (d) the optimal tradeoffs among pulse duration, pulse rate, interval between sequential pulses, and cutoff frequency of the lowpass filters seem to vary from patient to patient. Studies are underway to evaluate CIS processors for additional subjects in the low performance group and to investigate in detail the tradeoffs among processor parameters for subjects in both the low and high performance groups.

ACKNOWLEDGMENTS

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Effects of noise and noise reduction processing on the operation of the Nucleus-22 cochlear implant processor

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Abstract—Cochlear implants, like other types of auditory sensory aids, become increasingly ineffective with increasing ambient noise levels. One method of signal processing to reduce additive random wideband noise, the INTEL method, has been used to good effect as an input preprocessor for the Nucleus-22 cochlear implant. The implant's own signal processor estimates and encodes pitch frequency and the frequencies of Formants 1 and 2. The study reported here shows that additive noise results in substantial deviations in formant frequency estimates from those that would be observed in the absence of noise. When noisy speech is preprocessed by the INTEL method to reduce noise intensity, the deviations in the frequency estimates for Formant 2 are substantially reduced.

Key words: *cochlear implants, INTEL method, noise reduction, preprocessed noisy speech, signal processing.*

INTRODUCTION

Cochlear implants produce a perception of sound through electrical stimulation of auditory nerve fibers. Sounds received by the implant system microphone are transformed into patterns of electrical currents that are delivered either to a single electrode at the round window, or to one or more electrodes that are inserted into the cochlea itself. In general, two approaches are in use for converting audio signals into stimulus currents. In some devices the analog signals in one or more frequency bands are encoded, usually after some form of amplitude compression or frequency shaping has been imposed on them. Others encode information-bearing features of speech that have been extracted from the signals. Whichever coding method is used, when noise accompanies the received sounds the efficacy of the implant as an aid to speech perception is rapidly reduced. The loss of efficacy in noisy environments is similar to that experienced by users of amplification-type hearing aids and elicits similar complaints.

One of the most widely used multiple electrode cochlear implant systems is the Nucleus 22-channel device made by Cochlear, Pty. Ltd. (New South Wales, Australia). Its signal processor extracts intensity, voicing, pitch, and formant information from received sounds, encodes these features as pulsatile patterns, and transmits them to appropriate electrodes in the implant. The frequencies of the first two formants are estimated as the zero crossing rates for the dominant spectral peaks in two bands, which nominally are from 300 Hz to 1,000 Hz for Formant 1, and from 800 Hz to 4,000 Hz for Formant 2. Some noise immunity is provided in the two models of this implant that are in use. In the wearable speech processor (WSP III) model, stimulation is suppressed for sounds below a minimum average level. In the minispeech processor (MSP) model, the long-term average signal level is computed in each of several bands, and this constant value is subtracted

from the instantaneous signals in these bands. The band outputs are then recombined. While the frequency measurement and noise reduction methods are simple in comparison with more powerful procedures that are available, they apparently are adequate in relatively quiet environments. However, it has been reported that changes in user performance with the noise suppression system active are variable, and that users of the WSP III model tend not to activate noise reduction, while users of the MSP model vary in their use of it (1).

The effects of noise on speech perception are substantially different for sounds that are heard directly as compared with sounds received through cochlear implants. In the former case, it is primarily the masking and critical band characteristics of the ear that determine how noise will affect perception. In the latter case, noise-induced errors in the feature measurements will result in the generation of incorrect stimulation patterns, resulting in misperceptions of the encoded sounds. Such errors can be reduced by using more robust methods of extracting the selected speech features or, alternatively, by processing noisy speech so as to improve the signal-to-noise ratio. The latter approach was taken by Hochberg, et al. (2) who studied the effect of using the INTEL noise reduction method (3,4) to process noisy speech that was presented to users of the Nucleus-22 implant. Consonant-vowel-consonant (CVC) words drawn from isophonemic word lists (5) were presented to cochlear implantee subjects at signal-to-noise ratios in the range 25 dB to − 10 dB within a band from 0 to 4,500 Hz. Substantial improvements in perception accuracy were obtained for signals in additive speech spectrum noise at signal-to-noise ratios down to 0 dB. Significantly, the average shift in subject performance was reasonably consistent with the estimated reduction of noise in the processed signals.

In view of these results, it was of interest to examine the effect INTEL processing of the CVC words had on the formant measurement accuracy of the Nucleus-22 processor, and in particular to identify where and to what extent formant measurement errors were reduced. The next section of this paper describes the noise reduction process and the manner in which it was applied to the test stimuli in the experiments reported by Hochberg, et al. (2). Next, the method by which formant measurement errors were determined is discussed. This is followed by a presentation and a discussion of the results of the analyses. Finally, the implications for the design of cochlear implant systems are considered.

Noise Reduction Method

The INTEL noise reduction process is a "transformation subtraction" type of procedure. In this approach to noise reduction, a function that represents the noise is subtracted from transforms of the noisy speech signals. The modified signal transforms are then reconverted to time waveforms by use of an inverse of the original transformation procedure. In systems that use this approach, the noise reference function usually is generated from transforms of the input signals that ideally contain little or no speech energy. In multiple-channel noise reduction systems, the data that are needed to generate this function are obtained by use of one or more additional microphones that receive versions of the noise in which signal components are weak or absent, that is, adaptive noise cancelers (6). In single-channel systems, such as INTEL, the noise reference function is derived from the noisy speech signal itself, and so can be defined only when speech is absent or is negligible relative to the noise. Consequently, in such systems, the noise reference function cannot represent the instantaneous variations of the noise, but only their statistical characteristics. In INTEL, the noise reference function is based solely on the average transform of the noise.

INTEL operations (see **Figure 1**) are performed on successive overlapped segments of the input signal. Four Fourier transforms are computed during the processing of each segment. The first two of these are needed to generate a function on which the noise reduction operations are performed. First, the complex spectra of the signal are obtained and converted into magnitude and phase transforms. Then, an inverse transform of the square-root of the magnitude frequency spectrum is computed, and the magnitude of that transform is obtained. For convenience, the resulting function is referred to here as the period spectrum, since the locations of peaks in this function are linearly related to the periods of periodic components in the input signal.

To generate the noise reference function, the average period spectrum of the noise is computed, segmented at some period, Tc, into two regions, and scaled in each region. The value of Tc (usually Tc < 1 msec) and the scale factors used for each
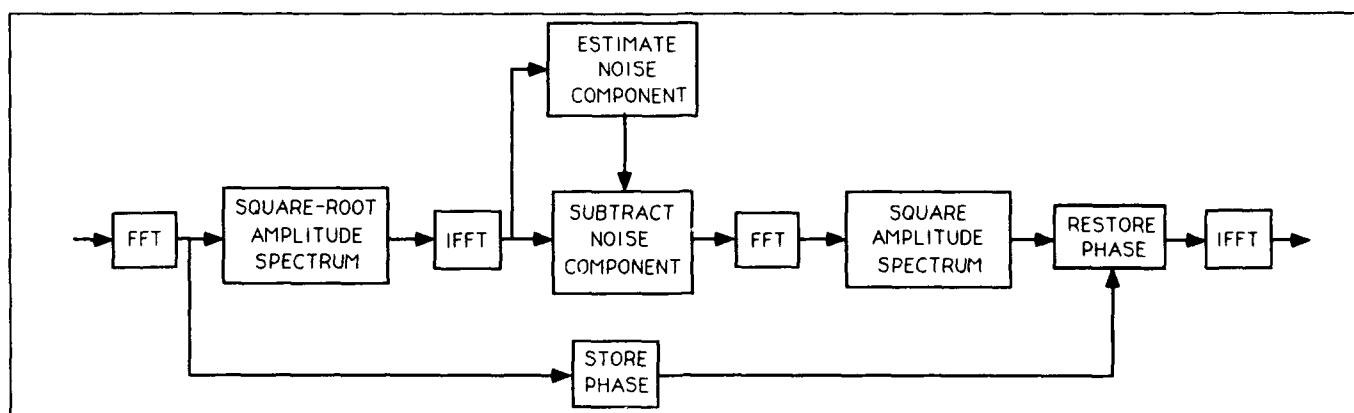
Figure 1.
INTEL method of noise reduction.

region are selected to maximize output signal quality for a given amount of noise reduction. When the noise reduction process is active, the noise reference function is subtracted from the period spectra of the input. Resulting differences that are negative are set to zero. Then the modified period spectra are converted by direct Fourier transformation to the frequency domain, squared to restore original amplitude relationships, and combined with the original phase frequency spectra to generate new complex spectra. These are transformed to the time domain to produce output signal segments which are combined by use of an overlap-add type procedure to generate a continuous output signal. The overlap-add procedure is used to reconstruct a signal or a processed version of a signal by summing the regenerated segments of the output that correspond to overlapped segments of the input (7).

The effects of the INTEL process on speech signals and on noise depend primarily on the values set for the noise-reference function scale factors. While the process reduces the level of noise, it also reduces the level of relatively weak speech sounds (e.g., weak fricatives or plosives). It also tends to weaken disproportionately the weaker components of speech sounds. These effects increase in proportion to the amount of noise reduction that is achieved. Generally, the larger the scale factors are made, the greater will be the reduction of noise and, concurrently, the greater will be the distortion in the regenerated speech. Where high levels of distortion are acceptable, as in the case of limited vocabulary speech, noise reductions of up to 14 dB can be

achieved, with scale factors usually in the range 0.6 to 0.85. With scale factors in the range 0.2 to 0.4, noise can be reduced by up to 4 dB with little perceptible distortion in the processed speech sounds.

In the experiments described in Hochberg, et al. (2) the INTEL process was set to reduce noise by 7 dB on average over the range of input signal-to-noise ratios. For this setting, the period spectrum low-band cutoff was 0.2 msec, and the low-band and high-band scale factors were 0.6 and 0.65. The amount of noise reduction was estimated by measuring the noise level before and after processing when speech was absent. Since the INTEL process is nonlinear, the measured reduction in noise level does not necessarily indicate that the average output signal-to-noise ratio was increased by 7 dB. However, it is noteworthy that the test results in Hochberg, et al. (2) indicate that the average shift between performance intensity (PI) functions for the processed and unprocessed stimuli, measured at the signal-to-noise ratio which obtained 50 percent of the maximum score, was 5 dB. This result is in reasonable agreement with the shift that would be expected for a signal-to-noise ratio enhancement of 7 dB.

A typical example of the noise reduction that was obtained for these settings of the INTEL process is presented in **Figure 2**. The spectrogram of the phrase "the word is ways," at a signal-to-noise ratio of 25 dB, is shown at the top. The middle spectrogram shows the result of adding speech-spectrum shaped noise at a level 10 dB below the

speech. At a signal-to-noise ratio of 10 dB, Formant 1 is partially obscured in regions where it is weak, and Formant 2 is almost completely obscured, especially during the /w/ glides. As seen in the lower spectrogram, after processing, Formant 2 is more apparent both during the transitions and in the regions of less rapid change.

METHODS

The formant frequency data required for this study were obtained by use of a Cochlear Pty., Ltd. simulator of the Nucleus-22 processor. The simulator, which ran in an IBM/AT-type personal computer, performed the signal conditioning, filtering, and formant measurement operations that are implemented in both the WSP III and the MSP model processors, and provided formant frequency measurements at intervals of 1 msec. Three sets of measurements were obtained. Two of these were at a signal-to-noise ratio of 10 dB, one set with noise reduction processing and the other without processing. Measurement errors for each test condition were determined relative to a reference set of data. Ideally, this set should consist of the true formant frequency values for the speech test material. However, the true formant values are unknown and the estimated formant frequency values obtained with the simulator in quiet were used.

The signal-to-noise ratios selected for these sets were chosen to permit the results of these analyses to be compared with data obtained in the experiments described in Hochberg, et al. (2). The signal-to-noise ratio of 25 dB used for the reference data set is the same as the maximum signal-to-noise ratio used in the cited experiments and was used to establish the upper performance limits for the tested subjects. The signal-to-noise ratio of 10 dB used for the test data sets is approximately at the middle of the test range used in the cited experiments.

The differences in the detectability of Formants 1 and 2 that are apparent in the middle and lower spectrograms of **Figure 2** are reflected in the performance of the Nucleus-22 formant frequency extractor. As can be seen in **Figure 3**, the measurements for Formant 1 at a signal-to-noise ratio of 25 dB form a reasonably smooth track for Formant 1 through the glide /w/ and the diphthong /eI/ in the test word "ways." Formant 2 values are tracked
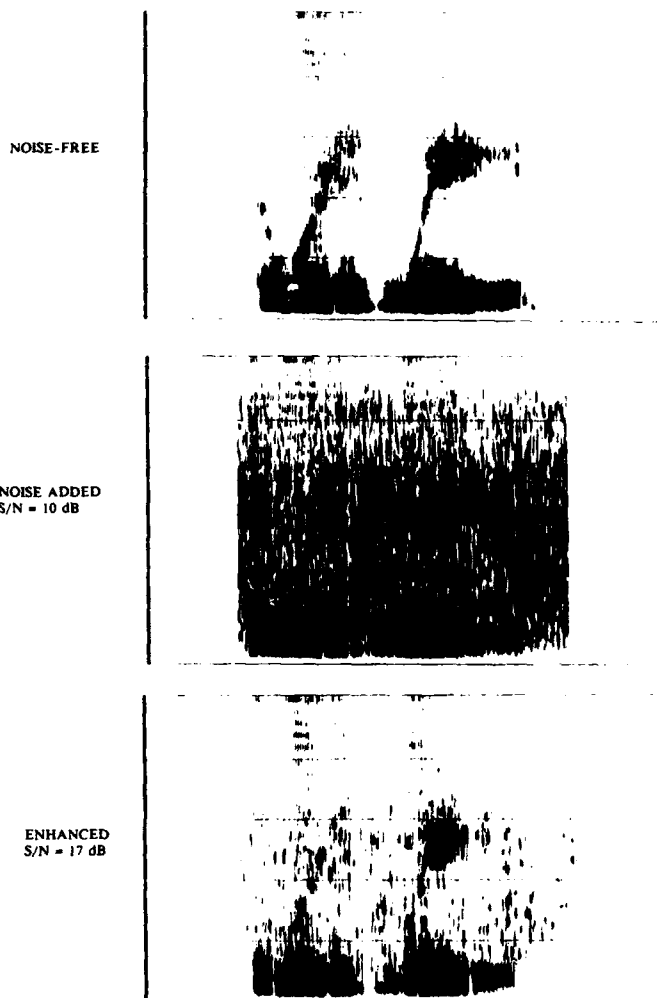

NOISE-FREE

NOISE ADDED
S/N = 10 dB

ENHANCED
S/N = 17 dB

Figure 2.
Spectrograms of "The Word is Ways" before and after INTEL processing.

well through the glide, but there appears to be some uncertainty during the diphthong. At a signal-to-noise ratio of 10 dB, the tracking of Formant 1 is still reasonably good, with some variability evident at the end of the diphthong. However, the first half of the Formant 2 transition during the glide is lost. After processing, the extraction of Formant 1 is essentially the same as before processing, but now the entire Formant 2 transition has been tracked.

In the experiments described in Hochberg, et al. (2) the test stimuli were drawn from recordings of 15 isophonemic CVC word lists as spoken by a female talker, with each list composed of 10 words. Speech-spectrum-shaped noise was added at a constant level, and the speech level was adjusted to
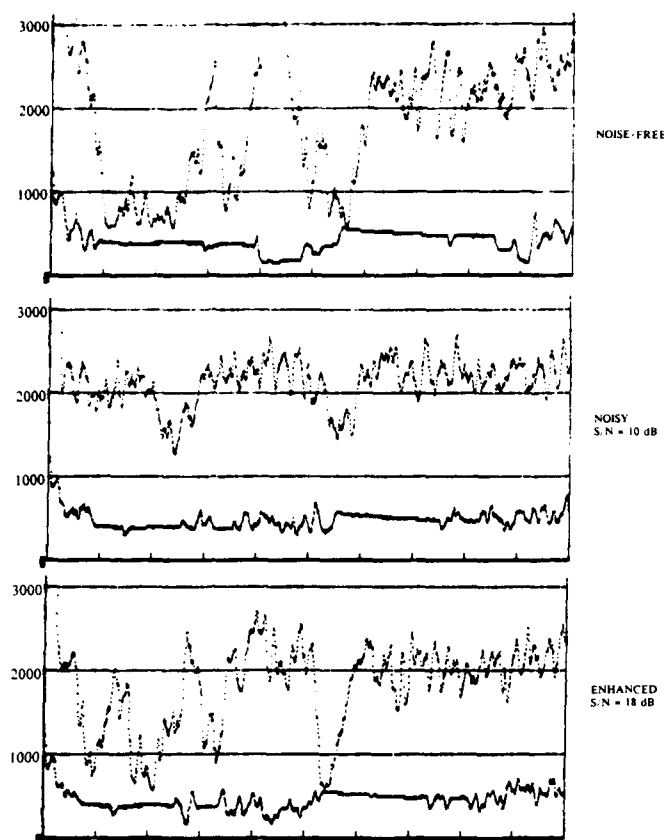
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Figure 3.
Estimated frequencies of Formants 1 and 2 for "Word is Ways" before and after INTEL processing.

achieve the desired signal-to-noise ratio. The formant frequency data that were used in the study described here were extracted from the recording of word list No. 5, in which the test words were: *bath, dig, five, hum, joke, noose, pot, reach, shell,* and *ways.* The intensity and spectral distribution of the noise and the settings of the INTEL process parameters were the same as in the cited experiments. Inputs to the simulator were 5-kHz low-pass filtered, sampled at a rate of 10 kHz, and converted to digital form with 12-bit accuracy.

For all but one of the test words in the reference set ("hum"), the measured frequencies of one or both of the formants changed rapidly either during the transition from the initial consonant to the vowel or during the consonant itself. It was expected that at any given signal-to-noise ratio, formant measurement errors would be greater during these rapid changes than during periods when the formant frequency was changing slowly or not at

all, and that this would be especially so when these changes occurred in regions where speech intensity is lower than it is during the vowel nucleus. Conversely, it was anticipated that an increase in signal-to-noise ratio would reduce errors in these regions more rapidly. To observe whether INTEL processed speech exhibited such an effect, the extracted formant tracks for each test token were divided into an initial region and a central region. The initial region was defined as beginning at the start of the transition to the central vowel and ending at the point at which the frequencies of both formants were within 10 percent of their estimated frequencies at the vowel nucleus. The central region encompassed the remainder of the vowel. The difference between the test and reference values, determined at 1 msec intervals, were used in the computation of three types of errors within each region: the average absolute error, the root mean square (rms) error, and the maximum absolute error. For the absolute error and rms errors the difference measures at each point were expressed as a percent of the reference value.

RESULTS

Formant Data

Figures 4a, 4b, 4c, 4d, 4e, 4f, 4g, 4h, 4i, and 4j present graphically the formant measurements made by the Nucleus-22 simulator for each of the test words under each of the study conditions. It is apparent that the reference data for Formant 2 exhibit considerable variability, both during the onset of the voiced speech sounds and during the relatively slowly changing segments of the central vowels. These variations indicate an intrinsic weakness in the method of estimating formant frequencies by counting zero crossings, particularly as applied to extracting vowel formants from speech with a high fundamental frequency. Comparisons of the reference condition measurements with formant frequency estimates obtained from spectrograms and from linear predictive coding (LPC) analyses also show occasional gross errors in average values for one or both formants. However, since these data were used to represent the normal "noise free" performance of the formant extractors, then for the purposes of these analyses the reference data were considered to be error free. Thus, an initial region
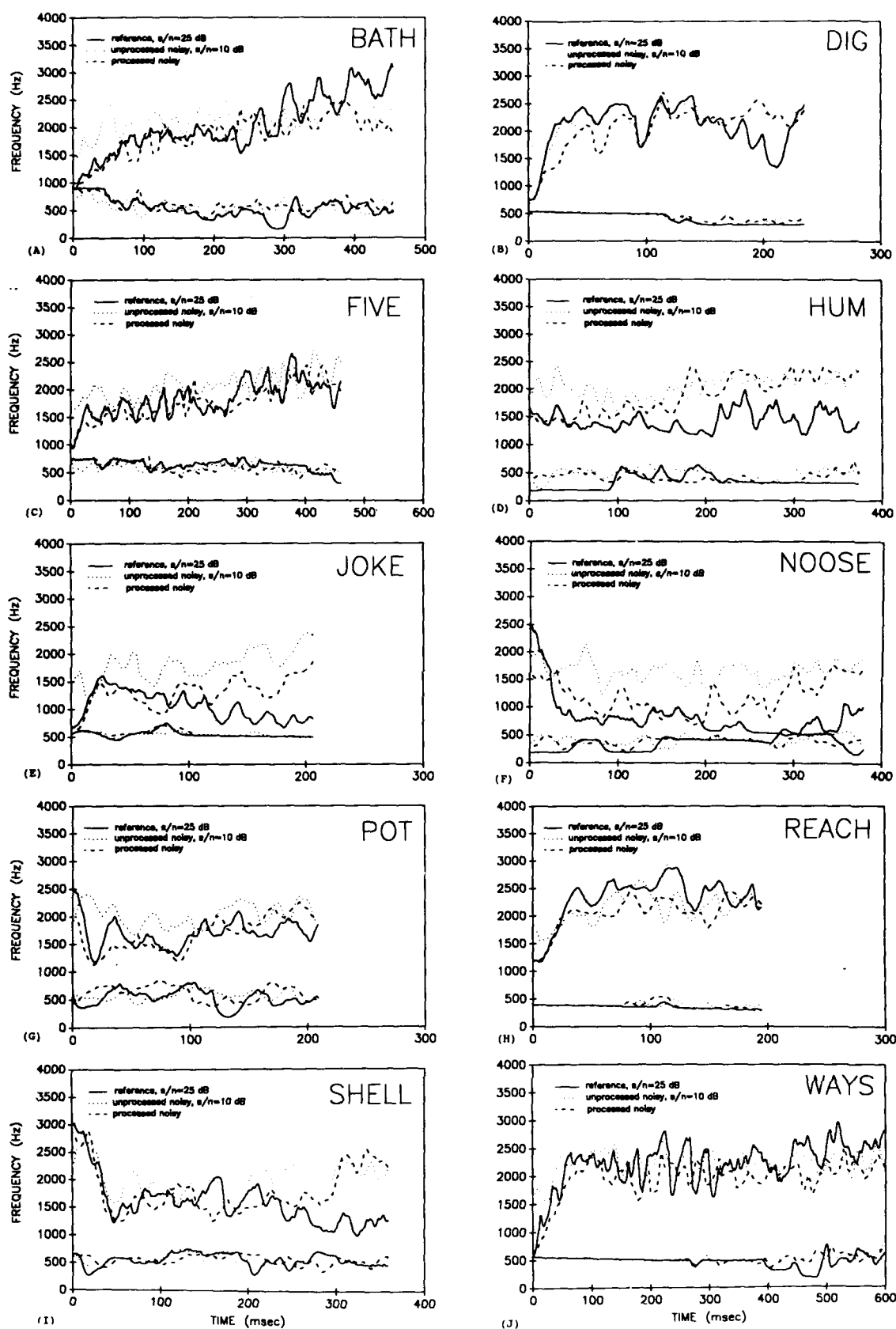
Figures 4a-4j.
Estimated frequencies of Formants 1 and 2 for 10 test words before and after INTEL processing.

was determined in accordance with the definition stated earlier even when the initial rapid variations in formant frequency did not correspond to events that could be observed in a spectrogram.

The formant tracks for the test conditions exhibit short-term variations similar to those seen for the reference condition. The general trends of the test and reference condition measurements are more evident in **Figures 5a, 5b, 5c, 5d, 5e, 5f, 5g, 5h, 5i,** and **5j,** which present the formant data fitted with tenth-order polynomials. During the first halves of the test, words the smoothed Formant 2 tracks for the test conditions tend to parallel the reference tracks, with the tracks for the processed data usually closer to the reference tracks than are the tracks for the unprocessed data. The deviation between the test condition tracks and the reference tracks tends to increase during the second halves of the words.

Error Analyses

The results of the error analyses are presented in **Table 1, Table 2,** and **Table 3**. Reduced error rates in the formant measurements made on the processed sounds are evident for both formants and for all error measures. Analyses of these data, using Wilcoxon matched-pairs signed-ranks statistics, show that the effect of noise reduction processing on Formant 1 was not significant ($p > 0.15$) in any of the error measures in either the initial or central region. On the other hand, the effect of processing is seen to be highly significant for Formant 2 in the initial region in all three error measures ($p < 0.01$) and significant in the central region ($p < 0.05$). The reductions in error rates for Formant 2 were also more substantial than they were for Formant 1. This is reflected in the computed average errors, where it is seen that the percent-average errors and the percent-rms errors for Formant 2 in the initial region are greater than they are for Formant 1 before processing, and that they are lower after processing. **Figure 5** shows the effect of processing on the mean error for each formant, test condition, and error type, with the reductions in mean error rates expressed as percentages of the mean error in the unprocessed condition. It is apparent that the greatest improvements were for Formant 2 in the initial regions of the test words.

It is of interest to compare these results with the pattern of reduction of phoneme identification errors. **Table 4, Table 5, Table 6** and **Table 7,** which are based on data obtained in the experiments described in Hochberg, et al. (2), present average percent recognition scores before and after noise reduction processing for each of the 30 phonemes used in the AB word lists at a signal-to-noise ratio of 10 dB. Increases in scores for stop consonants are highly significant, with scores increasing from an average of 17.6 percent before processing, to an average of 29.2 percent after processing. By comparison, the changes in the scores for fricative consonants are small (2 percentage points, on average) and were not statistically significant. The scores for nasal and liquid consonants also did not show a statistically significant change (6 percentage points).

DISCUSSION

The Nucleus-22 cochlear implant processor estimates the frequencies of Formants 1 and 2. When noise is added to speech, these estimates deviate substantially from those that would have been obtained in the absence of noise. INTEL processing of noisy inputs reduces the magnitude of these deviations for Formant 2. Formant 2 is very important for the perception of speech by users of this type of cochlear implant. The first version of the Nucleus-22 system encoded only pitch frequency and Formant 2 amplitude and frequency data (F0F2A2) as compared with pitch, Formant 1, and Formant 2 frequency values (F0F1F2) encoded in the current version. However, analyses of the amount of information transferred by each of these coding schemes (8) show that the original scheme transmitted, on average, 50 percent of vowel information and 40 percent of consonant information, as contrasted with 63 percent and 48 percent, respectively, for the current coding method. Thus, it appears that most of the useful information for vowel and consonant identification is conveyed by Formant 2. Therefore, it is highly probable that the reduction in the deviations of Formant 2 frequency resulting from the INTEL processing of speech at a signal-to-noise ratio of 10 dB is the reason for the observed increase in subject test scores at that signal-to-noise ratio, as reported by Hochberg, et al. (2). The data presented there show that the average percent phoneme recognition for 10 cochlear implant users shifted upward by an amount corre-
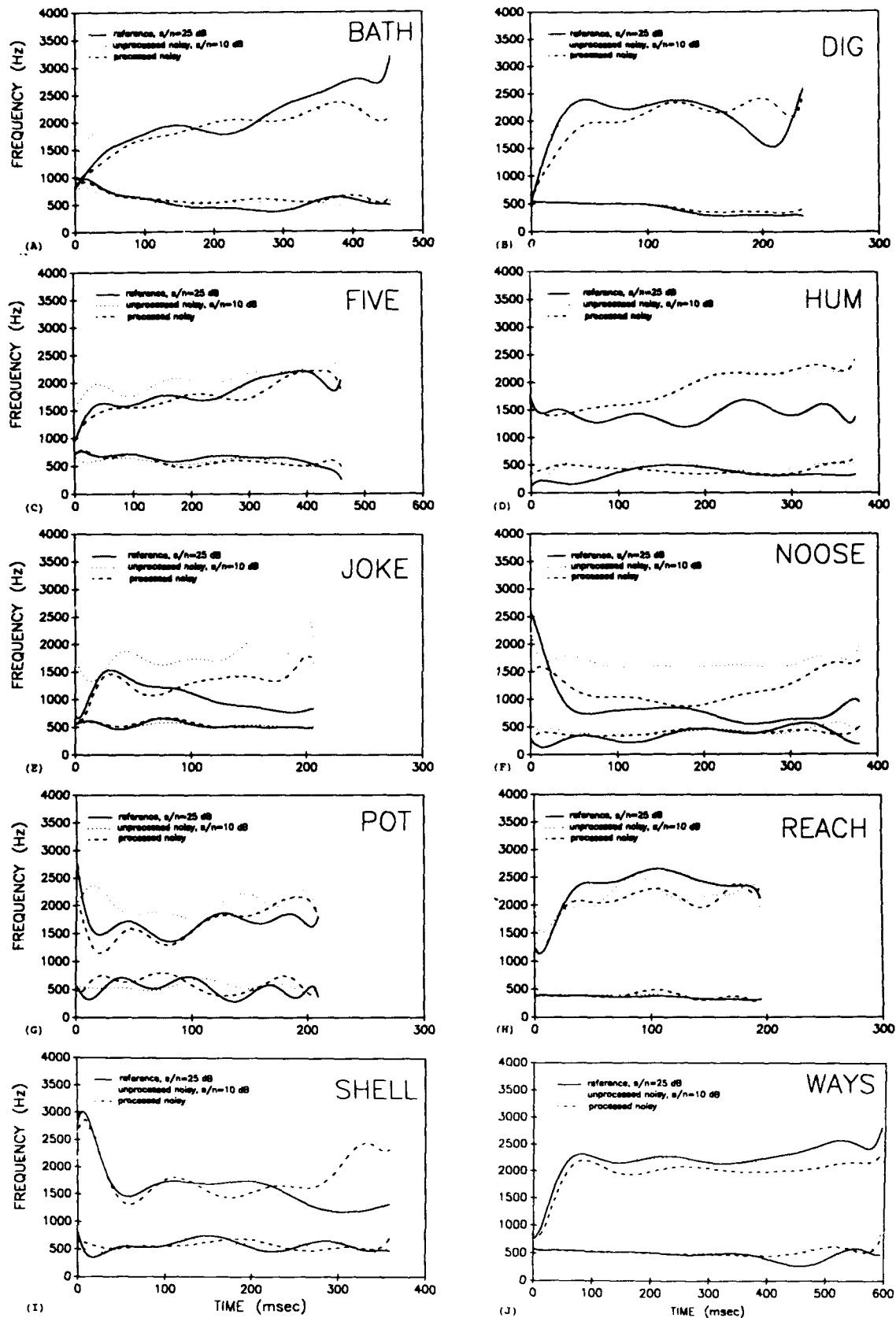
Figures 5a–5j.
Smoothed estimated frequencies of Formants 1 and 2 for 10 test words before and after INTEL processing.

Table 1.

Average percent absolute difference error in the measurement of Formant 1 and Formant 2 within the initial and central regions of each test word.

Word	Initial Region				Central Region			
	Formant 1		Formant 2		Formant 1		Formant 2	
	Unpr	Proc	Unpr	Proc	Unpr	Proc	Unpr	Proc
Bath	32	21	18	8	19	14	29	10
Dig	1	0	56	17	1	0	8	13
Five	22	2	48	4	13	15	17	19
Hum					44	26	46	23
Joke	2	1	62	10	5	3	88	45
Noose	182	109	53	13	33	47	87	57
Pot	40	45	18	18	31	41	21	13
Reach	1	1	52	15	18	11	10	11
Shell	21	45	9	9	17	18	37	31
Ways	1	1	44	16	11	7	9	13
Mean Error	34	25	40	12	19	18	35	24
LS	0.234		0.002		0.313		0.043	

Signal-to-noise ratio was 10 dB at the input to the INTEL noise suppression processor for both unprocessed (Unpr) and processed (Proc) conditions. Level of significance (LS) was computed by the Wilcoxon matched-pairs signed-ranks method.

Table 2.

Average percent rms difference error in the measurement of Formant 1 and Formant 2 within the initial and central regions of each test word.

Word	Initial Region				Central Region			
	Formant 1		Formant 2		Formant 1		Formant 2	
	Unpr	Proc	Unpr	Proc	Unpr	Proc	Unpr	Proc
Bath	42	25	20	9	25	20	35	12
Dig	1	0	77	21	2	1	11	15
Five	24	2	50	4	15	18	21	11
Hum					67	45	48	28
Joke	2	1	80	10	8	7	107	58
Noose	184	110	63	15	52	67	105	82
Pot	40	50	21	18	55	57	24	16
Reach	1	1	80	23	26	16	13	16
Shell	29	65	11	10	24	27	48	44
Ways	1	1	65	18	22	18	13	16
Mean Error	36	28	52	14	30	28	43	30
LS	0.234		0.002		0.278		0.042	

Signal-to-noise ratio was 10 dB at the input to the INTEL noise suppression processor for both unprocessed (Unpr) and processed (Proc) conditions. Level of significance (LS) was computed by the Wilcoxon matched-pairs signed-ranks method.

Table 3.
Maximum absolute difference error in the measurement of Formant 1 and Formant 2 within the initial and central regions of each test word.

| | Initial Region | | | | Central Region | | | |
| | Formant 1 | | Formant 2 | | Formant 1 | | Formant 2 | |
Word	Unpr	Proc	Unpr	Proc	Unpr	Proc	Unpr	Proc
Bath	243	143	577	353	320	279	915	740
Dig	11	4	1051	795	32	18	505	719
Five	243	41	637	85	223	273	887	557
Hum					380	278	1206	719
Joke	16	9	910	146	196	115	1607	1033
Noose	412	233	1019	585	307	370	1592	1444
Pot	202	287	733	667	395	436	751	653
Reach	10	10	1496	563	250	189	969	189
Shell	172	385	747	410	333	350	1209	1301
Ways	8	7	1042	542	302	255	910	930
Mean Error	146	124	912	461	274	256	1055	829
LS	0.156		0.002		0.216		0.032	

Signal-to-noise ratio was 10 dB at the input to the INTEL noise suppression processor for both unprocessed (Unpr) and processed (Proc) conditions. Level of significance (LS) was computed by the Wilcoxon matched-pairs signed-ranks method.

Table 4.
Percent identification of vowels, diphthongs, and glides by 10 subjects using the Nucleus-22 cochlear implant.

	i	I	e	a	ae	ˆ	u	eI	aI	ˆU	w
unprocessed	17	17	23	33	37	53	23	32	44	27	13
processed	42	42	35	32	45	42	52	35	61	48	32
difference	25	25	12	−1	8	−11	29	3	17	21	19
signed rank	9.5	9.5	5	−1	3	−4	11	2	6	8	7
tail probability		0.005									

Signal-to-noise ratio was 10 dB at the input to the INTEL noise reduction processor. Level of significance was computed by Wilcoxon matched-pairs signed-ranks method.

sponding to a 5-dB noise reduction when the noisy speech test materials were processed in the same manner as for this study (i.e., so as to improve the signal-to-noise ratio by an estimated 7 dB). While it is difficult to accurately measure the reduction of noise in the processed speech signals, the estimated reduction is consistent with the observed increase in subject performance.

This type of signal processing may also benefit other types of feature-extracting sensory aids in noisy conditions. The use of the INTEL method of noise reduction has been shown to improve the accuracy of spectrum-based speech recognition systems (9). Hence, it seems likely that the operation of cochlear implants that encode spectrum peaks or spectrum envelope characteristics also may be improved.

It is somewhat ironic to observe that cochlear implant users, who usually are severely to profoundly hearing-impaired, may in time be able to

Table 5.

Percent identification of nasals and liquids by 10 subjects using the Nucleus-22 cochlear implant.

	m	n	r	l
unprocessed	10	7	3	37
processed	16	29	10	26
difference	6	22	7	− 11
signed rank	1	4	2	− 3
tail probability		0.31		

Signal-to-noise ratio was 10 dB at the input to the INTEL noise reduction processor. Level of significance was computed by Wilcoxon matched-pairs signed-ranks method.

Table 6.

Percent identification of fricatives by 10 subjects using the Nucleus-22 cochlear implant.

	f	s	sh	th	h	v	z
unprocessed	17	11	47	3	17	7	16
processed	13	3	61	3	29	10	13
difference	− 4	− 8	14	0	12	3	− 3
signed rank	− 3	− 4	6		5	1.5	− 1.5
tail probability	0.38						

Signal-to-noise ratio was 10 dB at the input to the INTEL noise reduction processor. Level of significance was computed by Wilcoxon matched-pairs signed-ranks method.

Table 7.

Percent identification of stop consonants by 10 subjects using the Nucleus-22 cochlear implant.

	p	t	k	ch	b	d	g	dj
unprocessed	7	13	20	43	17	27	7	7
processed	23	35	31	55	26	32	3	29
difference	16	22	11	12	9	5	− 4	22
signed rank	6	7.5	4	5	3	2	− 1	7.5
tail probability	0.008							

Signal-to-noise ratio was 10 dB at the input to the INTEL noise reduction processor. Level of significance was computed by Wilcoxon matched-pairs signed-ranks method.

function better in noisy environments t. ... persons with mild to moderate hearing impairments. No single-channel method of attenuating wide-band random noise appears to be capable of consistently and/or substantially improving the intelligibility of noisy speech that is perceived through a conven- tional amplification system (10,11) whether by normal-hearing or by hearing-impaired listeners. On the other hand, the INTEL method is able to improve speech perception accuracy for cochlear implant users, and by an amount that corresponds to the apparent reduction in noise intensity. Because this

method is computationally intensive, there is little chance that it can be implemented in cochlear implant processors in the near future. However, similar benefits might be achieved by use of computationally less intensive transformation subtraction techniques, such as spectrum subtraction (12), which requires half the number of Fourier transform computations. If so, this would be a more practical approach to pursue.

ACKNOWLEDGMENTS

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Measurements of acoustic impedance at the input to the occluded ear canal

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Abstract—Multi-frequency (multi-component) acoustic impedance measurements may evolve into a sensitive technique for the remote detection of aural pathologies. Such data are also relevant to models used in hearing aid design and could be an asset to the hearing aid prescription and fitting process. This report describes the development and use of a broad-band procedure which acquires impedance data in 20 Hz intervals and describes a comparison of data collected at two sites by different investigators. Mean data were in excellent agreement, and an explanation for a single case of extreme normal variability is presented.

Key words: *acoustic impedance tests, electroacoustic impedance tests, hearing aids, middle ear, tympanometry.*

INTRODUCTION

There are several important clinical uses of acoustic impedance data. First, a clinical procedure known generically as "tympanometry" is commonly employed for the remote detection of middle-ear pathologies. In this discrete, low-frequency proce-

dure, which often yields only an estimate of the compliance component of the ear canal and eardrum, sound is monitored by a microphone in the ear canal at different levels of static air pressure. More recently, discrete, multi-frequency, multi-impedance component (i.e., reactance and resistance or susceptance and conductance) instrumentation has become commercially available. A more definitive estimate of middle-ear function should result from multi-frequency measurements. In fact, it is possible that the fine structure of acoustic impedance curves (i.e., at closely spaced frequency intervals) may yield a much more exacting and definitive appraisal of the status of the middle ear (1,2).

Second, shaping of the audio spectrum is an important focus of hearing aid prescription and fitting. Aural acoustic impedance data could be used as an integral part of a comprehensive, computer-based model (3,4,5) to select hearing-aid components for the individual patient, or, alternatively, acoustic impedance data could be used to "correct" hearing aid gain-by-frequency prescriptions. Unfortunately, even if manufacturers were able to obtain a precise gain-by-frequency specification (6) those protocols have inherent sources of error because they are based on audiometric measurements using supra-aural earphones and couplers representing the acoustic characteristics of the average ear (7). The specification of sound pressure level (SPL) in the individual ear canal (developed by a hearing aid) is fundamental to the realization of a precise frequen-

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cy-gain prescription. A principal determinant of SPL is the acoustic impedance of the ear canal itself. Objective evaluation of the effect of eardrum impedance on the SPL in the ear canals of normal and pathologic subjects has been lacking, perhaps due to the difficulties associated with making accurate measurements at higher frequencies, as well as with obtaining accurate estimates of ear canal dimensions.

Purpose

The purposes of this article are: 1) to describe the development of an automated, broad-band aural-acoustic impedance measurement system; 2) to report data, in fine structure, on a larger sample of normal subjects than has been heretofore reported in the literature; and, 3) to compare data collected on normal ears at two different laboratories by two different investigators using identical systems.

METHODS

Subject Selection

Data were acquired from 35 subjects in this study at two laboratories. The sample from Site 1 included 20 subjects, and the sample from Site 2 included 15 subjects. Subjects ranged in age from 20 years to 35 years. Each had normal hearing thresholds (equal to or better than 15 db hearing threshold level), normal ear canals, normal-appearing eardrums, no history of middle ear pathology, and normal otoadmittance findings (8).

Procedures

Following preselection measures, estimates of ear canal volume, diameter, and length were made on one ear of each subject. Measurements of aural acoustic impedance were made at a mid-location in the occluded canal on one ear of each subject. All ears were studied under a condition in which efforts were made to ensure that atmospheric pressure existed in the ear canal, after having checked for an hermetic seal. The details of the measurement procedures are described below.

Estimates of Ear-Canal Volume, Diameter, and Length

The computations for deriving acoustic impedance at the tympanic membrane require a knowledge of the length and diameter of the ear canal. After a pilot study in which a high degree of inter- and intra-examiner reliability was observed, the diameter, d, was measured directly using a calibrated and graduated set (0.1 cm intervals) of ear probes which were inserted deeply into the ear canal. The examiner rotated the probe in the ear canal and judged which probe best made contact without distending the ear canal.

An estimate of ear canal length was derived from the diameter measurement and a tympanometric estimate of the volume between the impedance probe tip and the eardrum by the equation

$$L = \frac{\pi d^2}{4v} \qquad [1]$$

where L is the length, v is the volume, and d is the diameter of the ear canal.

In conventional, low-frequency tympanometric measurements, the eardrum is "stiffened" by a high positive or negative static (dc) pressure (9). The assumptions underlying these measurements are that: 1) the impedance at the eardrum is driven to infinity by the static pressure; and, 2) an appreciable change in the length of the ear canal does not result from the dc pressure change. Shanks and Lilly (10) reported data which refute these assumptions, but they also provide data which can be used to correct tympanometric estimates to values obtained using a more rigorous measurement technique. Hence, in this study the eardrum was pressurized to -400 daPa; the volume was recorded in units of acoustic admittance ($|Y|$) using the conventional technique for a 220 Hz probe signal. This value was corrected by 13 percent, the error reported by Shanks and Lilly (10) to result from a 220 Hz measurement of $|Y|$ at -400 daPa, and entered into Equation [1].

Estimate of Ear-Canal Input Impedance (Z_L)

The method by which the magnitude $|Z_L|$, the reactance X_L, and the resistance R_L at the driving point (i.e., at a position 2 mm past the tip of an ear-insert sealed in the ear canal) is computed is based on a two-cavity or two-load method. This method, described by Beranek (11), is a variation of Thevinen's theorem which states that any one-port network of resistance elements and energy sources can be replaced by a series combination of an ideal voltage source, E_t, and a resistance, R_t, where E_t is

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the open-circuit voltage of the one-port, and R_t is the ratio of the open-circuit voltage to the short-circuit current.

The method has been used to determine either the source impedance of a transducer (12,13) or the input impedance of a load connected to a transducer (14,15). While the equation has many incarnations, Egolf and Leonard (13) presented it as:

$$Z_s = \frac{E' - E}{E/Z_1 - E'/Z_2}, \qquad [2]$$

where E and E' are the voltages (magnitude and phase) developed across each load, with the loads having known or calculable impedances of Z_1 and Z_2. Z_s is the impedance of the source transducer. While Arslan, Canavesio, and Ceruti (14) and Rabinowitz (15) used source impedance as a term in their equations, Egolf[1] demonstrated mathematically that this requirement can be bypassed and, hence, the real and imaginary parts of acoustic impedance at the input to the ear canal can be computed using the following equation:

$$\angle_L = j \left\{ \rho c(E_o/E_i)[(E_o''/E_i'') - (E_o'/E_i)] \right\} \qquad [3]$$
$$\left\{ S''(E_o''/E_i'')[(E_o'/E_i') - (E_o/E_i)]\tan(kL'') \right.$$
$$\left. - S'(E_o'/E_i')[(E_o''/E_i'') - (E_o/E_i)]\tan(kL') \right\}$$

where, as illustrated in **Figure 1**, E_o''/E_i'', E_o'/E_i', and E_o/E_i are the measured probe-assembly transfer functions when the assembly is coupled to the 2.0 cc cavity, the 0.5 cc cavity, and the outer ear canal, respectively. The quantities S″, L″, S′, and L′ are the cross-sectional areas and lengths of the larger and smaller cavities, respectively. The term, k, is the wavenumber $2\pi f/c$, where f is frequency and c is the speed of sound. Air density is represented by the symbol ρ, and j is the imaginary operator $\sqrt{-1}$. Some errors in acoustic measurements occur because, rather than attempting to reproduce the open- and short-circuit conditions, the method uses two cavities with impedances sufficiently different to allow stable calculation of the desired parameter.

Calculation of Acoustic Impedance at the Tympanic Membrane (Z_T)

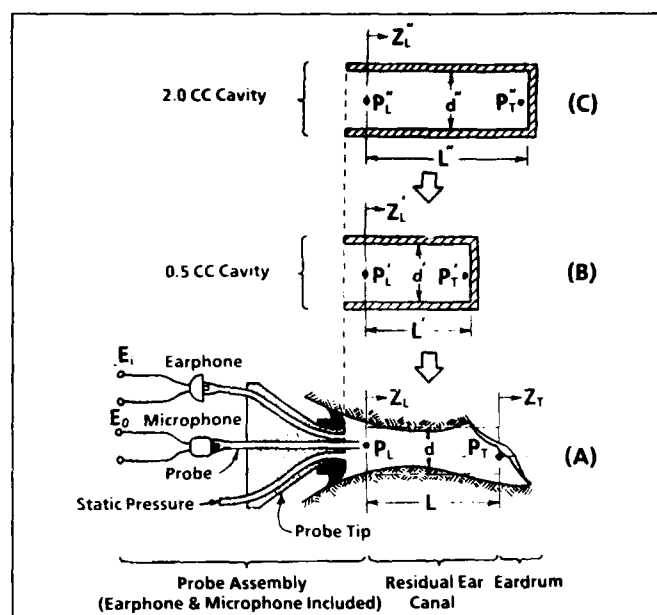The ear canal's diameter and length were estimated as discussed above. Calculation of imped

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[1] Egolf DP. A review of current technology dealing with the measurement of eardrum impedance/admittance. Unpublished lab report to V. Larson, VA Medical Center, Augusta, GA, March 1, 1988.
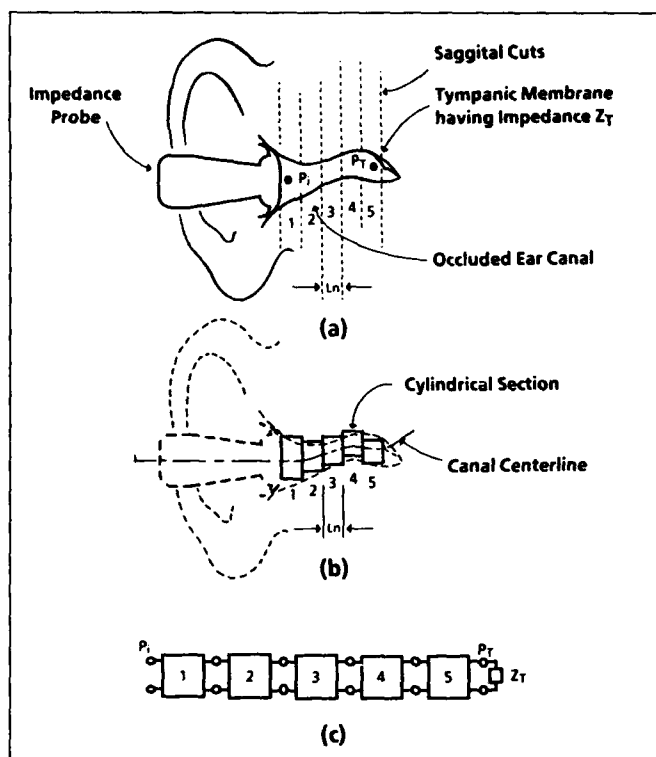

Figure 1.

Probe-tube assembly as sealed or mounted on A the ear canal, B a 0.5 cc cavity, and C a 2.0 cc cavity. (Reprinted, by permission, from *The Vanderbilt Hearing Aid Report II*, GA Studebaker, FH Bess, and LB Beck, editors. Parkland (MD): York Press, 1991.)

ance of the ear canal itself was accomplished in software using a distributed-parameter model described by Larson, Egolf, and Cooper (16) which treats the dimensions of the canal by a method similar to that of Kuhn (17). The algorithm calls for the residual ear canal to be sectioned via n hypothetical sagittal cuts as shown in **Figure 2a**, where each nth slice has a measured cross-sectional area S_n and thickness L_n, as shown in **Figure 2b**. In order to compute eardrum impedance (Z_T), the multiple-slice characterization of **Figure 2b** is represented by a serial connection of two-port electrical analog networks (see **Figure 2c**), each corresponding to one cylinder. Impedance at the tympanic membrane (Z_T) is then calculated by the equation

$$Z_T = \frac{Z_L D_S - B_S}{A_S - Z_L C_S}, \qquad [4]$$

where the terms in Equation [4] are

$$\begin{bmatrix} A_S B_S \\ C_S D_S \end{bmatrix} = \begin{bmatrix} A_1 B_1 \\ C_1 D_1 \end{bmatrix} \begin{bmatrix} A_2 B_2 \\ C_2 D_2 \end{bmatrix} \cdots \begin{bmatrix} A_n B_n \\ C_n D_n \end{bmatrix} \qquad [5]$$

and

Figure 2.

Diagram and illustration of a distributed parameter model for estimating the acoustic impedance of the ear canal. (Reprinted, by permission, from *The Vanderbilt Hearing Aid Report II*, GA Studebaker, FH Bess, and LB Beck, editors. Parkland (MD): York Press, 1991.)

$$A_n = D_n = \cos (kL_n),$$
$$B_n = j(\rho c/S_n) \sin (kL_n), \text{ and} \qquad [6]$$
$$C_n = j(S_n/\rho c) \sin (kL_n).$$

In this study, because of the lack of a method for estimating the dimensions of the ear canal, the dimensions for two equal slices were entered into Equation [5], essentially treating the ear canal as a circular cylinder terminated at right angles by the eardrum.

Instrumentation

Details of the measurement system and the technique on which it is based were reported previously (16). Briefly, however, the system used to make the impedance measurements uses a two-channel spectrum analyzer (Rapid Systems, Inc., Model 1200 with a Texas Instruments 320-10 signal processor) which is bidirectionally interfaced with an IBM-compatible computer. Locally developed soft-ware routines (Microsoft C, v. 1) pass parameters to assembly language routines in residence in the signal processor.

For all measurements, as illustrated in **Figure 1**, an impedance probe (Grason-Stadler, Model 1733 probe tip) was used to couple an earphone (Etymotic Research, Inc., Model ER-3A), a microphone (Etymotic Research, Inc., Model ER-7C), and a static air pressure pump and manometer to the ear canal or to either of the calibration cavities. The output of a waveform synthesizer (Quatech, Model WSB-10), a train of 100 μs clicks, drove the earphone. The electrical input to the earphone (E_i) and the output (E_o) from the probe microphone were routed through low-pass filters to the two inputs of the spectrum analyzer. At each site, preliminary measurements were made to insure that an overpressure of 400 daPa had no effect on the response of either the earphone or the probe microphone.

The data acquired from the two channels were used to compute cross- and auto-spectra which, in turn, were used to compute magnitude and phase transfer functions (E_o/E_i) for each of the three cavity measurements. These transfer functions, estimates of the magnitude squared coherence function, and calculations of Z_L (Equation [3]) are made in software by locally developed routines.

RESULTS OF IMPEDANCE MEASUREMENTS

Studies of Cylindrical Cavities

As reported by Larson, Egolf, and Cooper (16), a series of studies were conducted to validate measurements and computations with the system described herein. First, input impedance measurements were made on cylindrical tubes having rigid terminations which ranged in size from 0.8 cc volume to a tube having a length of 19.8 cm. Transfer function measurements were made as described in the previous sections and only data which approximated a 1.0 coherence function were accepted and passed to impedance calculation routines (Equation [3]). Comparisons were made of input impedance measurements of the tubes with data calculated using the following expression (11) for calculating the input impedance of a circular tube with a rigid termination:
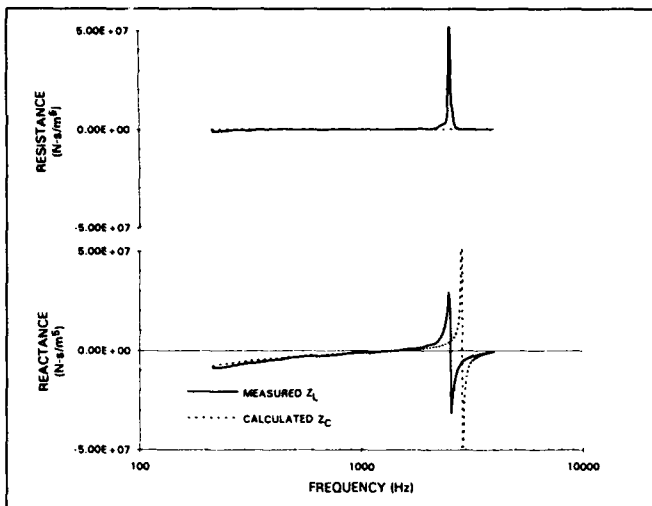
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Figure 3.

Resistance (upper curves) and reactance (lower curves) for measured and calculated values of a 1.65 inside diameter (i.d.) tube. (Redrawn, by permission, from *The Vanderbilt Hearing Aid Report II*, GA Studebaker, FH Bess, and LB Beck, editors. Parkland (MD): York Press, 1991.)

$$Z_c = R_c + jX_c,$$
$$R_c = 0, \qquad\qquad [7]$$
$$X_c = -j(\rho c/S_c)\ \cot kL_c,$$

where S_c is cross-sectional area and L_c is length. **Figure 3** shows one such comparison for a tube having the dimensions 6.0 cm (length) and 1.65 cm (diameter). Good agreement between the measured (i.e., using Equation [3]) and the computed data (i.e., via Equation [7]) was observed except for frequencies in the region of resonance. Note that the zero-crossing of the measured reactance curve occurs at a frequency which corresponds approximately to one-quarter wavelength: about 1,450 Hz for a 6.0 cm long tube. Note also a peak in the measured resistance (R_L) data at the frequency corresponding approximately to one-half wavelength. Ross (18) attributed such resistance peaks to mathematical anomalies (i.e, poles) that originate in Equation [3] when measurements are made on rigidly terminated cylindrical tubes, and hence provided sufficient justification to neglect them (16).

Acoustic Impedance Data and Interlaboratory Comparisons

Figure 4 presents mean Z_L data (i.e., from Equation [3]) measured on normal-hearing subjects
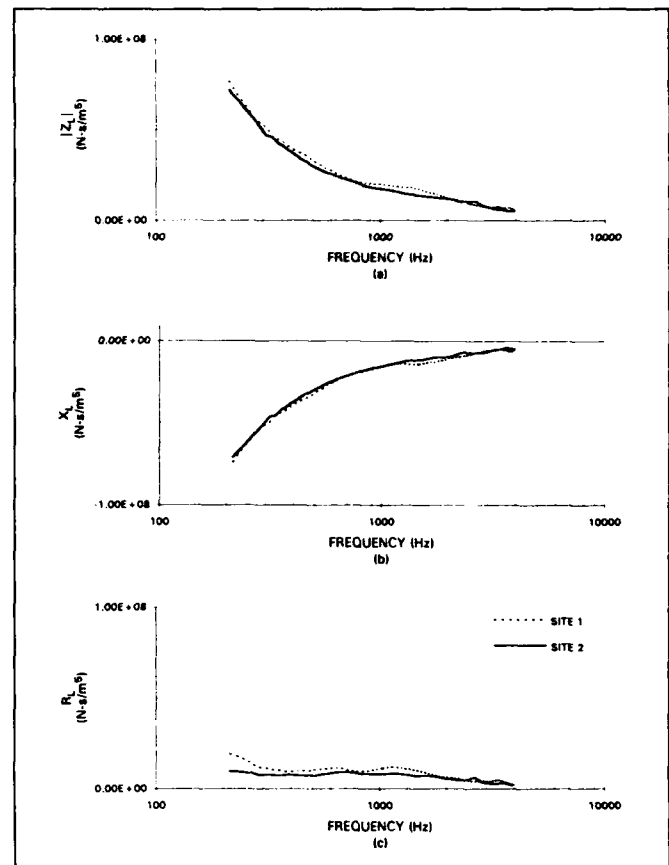
Figure 4.

Mean acoustic impedance (a) $|Z_L|$, (b) X_L, and (c) R_L for two sites.

at two sites. Illustrated in **Figures 4a, 4b,** and **4c** are data for $|Z_L|$, X_L, and R_L, respectively, at the input to the occluded ear canal. Experimenters at the two sites used different IBM-compatible computers, signal processing hardware, probe microphones, earphones, and calibration cavities. Transfer function measurements as well as the estimates of ear canal dimensions were made as described previously, and $|Z_L|$, X_L, and R_L were computed for the mid-canal measurement location using Equation [3]. Excellent agreement is apparent between the mean data for Z_L for the two sites.

Shown in **Figures 5a, 5b,** and **5c** are standard deviations associated with the means appearing in **Figure 4**. The standard deviations for the two sites are almost identical for frequencies below 1,500 Hz, but differences between the two sites emerge and reach large values in the 2,700 Hz region. In fact, a variance-ratio test in this frequency region
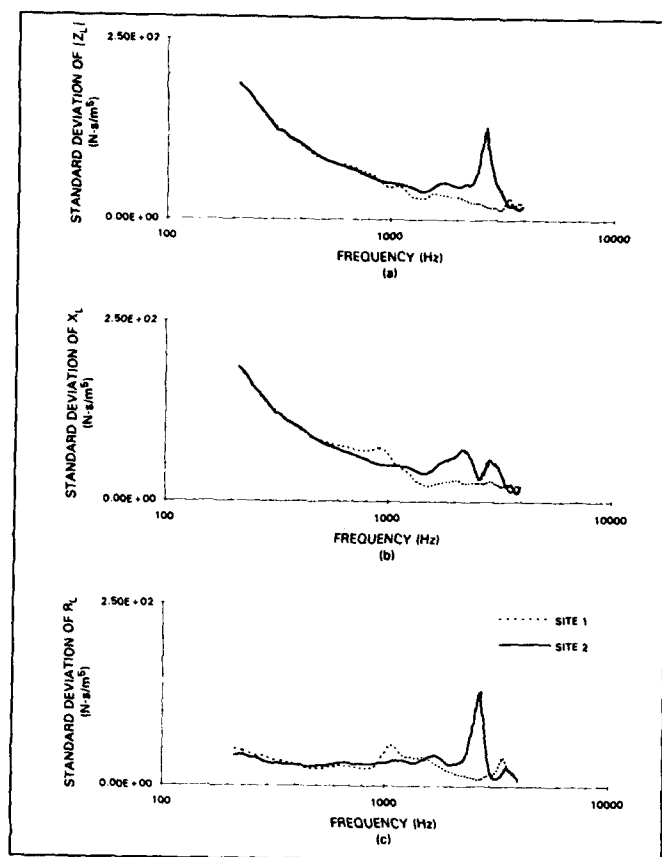
Figure 5.
Standard deviations for acoustic impedance (*a*) $|Z_L|$, (*b*) X_L, and (*c*) R_L for two sites.

Figure 6.
Comparison of mean and case acoustic impedance (*a*) $|Z_L|$, (*b*) X_L, and (*c*) R_L for two sites.

(p = 0.05) did not support pooling the mean data obtained from the two samples. An inspection of individual data, however, determined that the data for just one subject inflated the standard deviations of Site 2. **Figure 6a** shows $|Z_L|$ for this subject plotted with the mean data for Site 2. The peak in the 2,700 Hz region led us to question whether the data were in error in some regard or the subject was representative of a normal outlier.

Figures 6b and **6c** compare the data for the subject with the mean data for X_L and R_L, respectively. An inspection of the data suggests that the subject's data contain a resonance such as that observed in **Figure 3** for the 1.65 cm diameter by 6.0 cm long tube. The reactance curve of **Figure 6b** shows a typical resonance pattern in the 2,000 Hz to 3,000 Hz region. The subject's data also contained a resistive peak for the frequencies near resonance (see dashed curve of **Figure 6c**) similar to the peak in

Figure 3. Recall that Ross (18) discussed the resistive peak as being a mathematical anomaly when measurements are made in a cylindrical tube with a rigid termination, and hence provided a justification for ignoring the resistive data in the frequency region of resonance.

DISCUSSION

A knowledge of the acoustic conditions which produce "normal outliers" would be important to the interpretation of diagnostic acoustic impedance findings and clinical probe-tube measurements of ear canal sound-pressure levels, as well as to the application of acoustic impedance data to hearing aid design and fitting protocols.

Acoustically, ear canal resonances in the higher frequencies are related to the length of the ear canal

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and/or to the plane of measurement relative to the eardrum. Specifically, the frequencies corresponding to one-quarter and the first one-half wavelength, such as observed in the data of **Figure 3**, are predicted by

$$f = c/4L \quad \text{and by} \quad [8]$$

$$f = c/2L \quad [9]$$

where L is length of the occluded ear canal. Gilman and Dirks (19) demonstrated, in a mechanical simulation of eardrum impedance, that this location-dependent minimum in sound pressure predicted by Equations [8] and [9] is accurate for a purely resistive termination of the ear canal, but a correction to L must be made for terminations which are reactive, as is shown by

$$L = \lambda/4 + L_T \quad \text{where} \quad [10]$$

$$L_T = A/\pi(\lambda/4). \quad [11]$$

where A is the phase of the reflected wave relative to the incident and λ is wavelength. Equation [10] shows, then, that for L with a termination with no reactance, the quarter-wave minimum (and frequency location thereof) is predicted by Equations [8] and [9] but would be greater (longer effective distance from the eardrum) for a positi ι reactance and less (shorter effective distance from the eardrum) for a negative reactance. **Figure 7** shows a comparison of eardrum reactance X_T and resistance R_T (computed using Equation [4]) for the subject with the mean data. With reference to **Figure 7b**, the reactance for the individual case is clearly positive in the frequency region of resonance and, as shown in **Figure 7a**, resistance falls to zero. Apparently, the combination of ear-canal geometry and eardrum impedance are such that the frequencies of the one-quarter and one-half wavelengths fall within the bandwidth of measurements presented herein. For the other subjects, this phenomenon occurred at frequencies above 4,000 Hz.

SUMMAR $\underline{}$

Aural acoustic impedance measurements made at two sites by different investigators were in excellent agreement. An inspection of the intersubject variability suggested, however, that an occasional outlier may be expected when either the
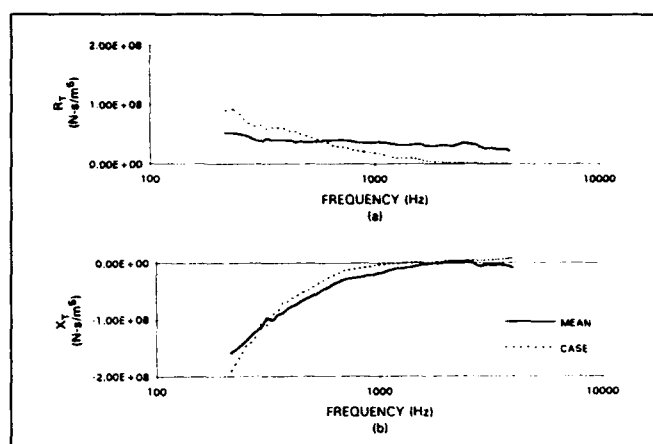

Figure 7.
Comparison of mean and case acoustic impedance (*a*) resistance, R_T, and (*b*) reactance, X_T, at the tympanic membrane.

actual distance between the plane of measurement and the eardrum or the effective length of the residual ear canal is such that the frequencies corresponding to one-quarter and one-half wave lengths are shifted downward to a frequency below 4,000 Hz. Studies of pathologic ears and studies in which aural pathologies are simulated are under way.

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CLINICAL REPORT

Efficiency of Dynamic Elastic Response Prosthetic Feet

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INTRODUCTION

Following years of accepting the Solid Ankle Cushion Heel (SACH) foot as the optimum compromise between durability and functional effectiveness, as well as being of reasonable cost, several new feet with dynamic elastic response qualities have been designed. The stimulus for these new designs is the recent development of materials which offer the potential to "store and release energy" in a manner that facilitates walking and running. Numerous new prosthetic feet have become available commercially. The effectiveness of these designs is not known, though each has its strong clinical advocates (1,2,3,4). It is claimed that these feet reduce the energy required for walking, and increase mobility. Four dynamic elastic response (DER) feet representative of this design were selected for study. The objectives of this project were to compare the efficiency of four DER prosthetic foot designs (Seattle, Flex-Foot, Carbon Copy II, Sten) to that of the traditional SACH foot; define the gait mechanics induced by each foot; and determine the relative effectiveness and cost/benefit ratio of these new feet for the dysvascular and traumatic amputee populations.

BACKGROUND

The principles of quantitated gait analysis and modern prosthetics were established by the Biomechanics Laboratory at the University of California at Berkeley. Following the fundamental studies of gait (6,7,8) attention was directed to improving prosthetic design. Pertinent to this proposal was their development of the patellar-tendon-bearing below-knee (BK) prosthesis and the SACH foot (9,10,11). By its structural durability and biomechanical soundness, the SACH foot readily replaced the earlier single-axis wood foot and has outperformed other designs until the present (12).

Limitations in SACH foot performance, however, are being documented. A survey of 179 veteran amputees (69 percent with BK amputation) identified excessive foot stiffness as a frequent problem: examination of 54 BK prosthetic feet substantiated this defect in 67 percent of those surveyed. Fatigue and heaviness of the prostheses were other common complaints. The size of the modern commercial foot heel cushion was about half of that originally designed (13). A static load response study indicated all of the SACH heel cushions were too hard (14).

Today, the SACH foot design is being challenged by new materials which provide controlled mobility by their capacity to "store and release energy." Functionally, they are being classed as DER feet.

At present, multiple designs are in regular clinical use (3,15). While they differ in material and structural design, each uniquely replaces the rigid SACH keel with a flexible segment that is proposed to replicate controlled "ankle" motion. Patients

describe a feeling of buoyancy when walking with these DER feet, and subjectively report using less energy compared to that expended when walking with the SACH foot.

Analysis of the "energy storing" prosthetic feet is just beginning. Czerniecki, et al., used kinematic and kinetic data obtained on two normal and one BK amputee to calculate energy storage and power output during running (16,17). They concluded that the Flex-Foot had the greatest (terminal stance) output power and the SACH had the least, with the Seattle foot midway between the two (16). Their kinematic calculations also indicated a high hip extensor power output in running that was reduced by the greater prosthetic ankle energy. The theoretical high hip extensor activity in terminal stance differs from the EMG evidence of normal running and walking, where hip extensor muscle action occurs only in early stance (18,19). This discrepancy between calculated requirements and actual functional performance indicates the need to combine EMG with kinematics and kinetics. The interplay between the multiple segments of the limb is both subtle and complex.

Wagner, et al., studied three subjects fitted with both a SACH foot and a Flex-Foot, and three others with only the Flex-Foot. Gait velocity and cadence were similar for both feet and less than normal, a finding consistent with the usual amputee limitation (20). The major difference in function was the greater ankle dorsiflexion (DF) in late stance by the Flex-Foot (20° vs. 11°). Terminal floor reaction forces were similar, with both foot types approaching that of the sound limb. A reduced terminal floor reaction force occurred in one subject with terminal stance knee hyperextension rather than flexion. Increased contralateral loading force was greater with both prosthetic feet. The authors concurred with others (20,21,22,23) that terminal stance force is a product of alignment rather than dynamic push-off. Their conclusion was that factors other than the terminal stance force must be considered in the investigation of energy storage and release. This is a major motivation for including dynamic electromyography (EMG), and relating the pattern of muscle action with the kinematics and kinetics of the amputee's gait.

These initial studies are useful pilot explorations, but meaningful comparisons must also accommodate the differences among patients. Conse-quently, a larger group of amputees must be analyzed in a single study. In addition, both the efficiency and mechanics of each prosthetic foot need to be addressed by direct energy cost analysis.

Many studies have assessed the energy cost of different amputation levels and types of pathology. Comparisons of their findings have been inconclu-sive, however, because each project assessed only one population, and techniques differed (24,25,26, 27,28,29). Also, the custom of measuring the rate of oxygen use failed to consider the reduction in physiological cost through gait velocity modifica-tion. This limitation is overcome by using energy cost per meter traveled as the measurement assessed (5,30). A series of amputee studies completed at Rancho Los Amigos Medical Center, which used a common testing and analysis system, demonstrated a significant correlation between the energy cost of walking, and both the level and etiology of amputa-tion (5,31,32).

SIGNIFICANCE

A BK amputation causes loss of plantar sensa-tion, free ankle and foot mobility, and selective muscular control of these joints. Current prosthetic replacement with a modern patellar-tendon-bearing BK socket and a SACH foot has only partially restored optimum walking ability (20,23,31). Even the young adult traumatic amputee, while expending energy at a 25 percent greater rate than normal walking, accomplished only 87 percent of the nor-mal velocity (32). Dysvascular amputees lack the necessary physiological vigor and strength and must slow their velocity to 47 percent of normal to maintain a normal rate of energy use (32). Amputees desiring to run must be extremely vigorous as major physical adaptations are needed to accommodate to the limitations of their prosthesis (3,15). Ramps and uneven ground are other daily experiences which present difficulty.

Below-knee has become the most common level of amputation following the advances in the diagno-sis of limb viability (33). With the knee preserved, the amputee has retained a potential for greater function. Realizing this potential, however, depends on optimum prosthetic support. Hence, the focus of this study was to characterize the gait of the amputee using different prosthetic feet. Achieving

this goal will provide the clinician with objective data to assist in prescribing a prosthetic limb that will give the BK amputee optimum gait.

METHOD

The energy cost and gait mechanics of four prosthetic feet with DER characteristics were compared to the functional qualities of the SACH foot.

Subjects

Seventeen BK amputees were studied. This group consisted of 10 amputees of traumatic origin and 7 subjects with amputations secondary to dysvascular disease. All subjects had a well-healed stump, had worn a prosthesis for at least 6 months, and were able to walk without an assistive device for 20 minutes without rest. All subjects were informed of the nature of the study and the time commitment involved. Each subject signed and received a copy of the Consent to Participate in an Experimental Project form, and the Bill of Rights of Human Subjects statement. All subjects were offered the prosthetic foot of their choice following completion of their participation in the study.

Prosthetic Management

Each subject was fitted with an endoskeletal prosthesis which allowed the interchange of foot components. In random order, the subjects were provided with four different feet of the DER type (Carbon Copy II, Flex-Foot, Seattle, Sten) and a SACH foot. Each foot was worn by the subject for approximately one month prior to kinesiological testing, to allow for adaptation to the prosthetic foot and ensure that it was fully functional for daily activities.

All prosthetic fabrication was completed at the Long Beach VA Medical Center. Anthropomorphic measurements included weight, height, limb length, and the measurements used for prosthetic fitting.

Instrumentation and Procedure

Functional testing was done at the Pathokinesiology Laboratory at Rancho Los Amigos Medical Center. This testing included energy cost, dynamic EMG, motion, joint moments (ankle, knee, and hip), and stride analysis.

Footswitches taped to the soles of the subjects' shoes were used to record the sequence of foot-floor contact. These compression closing sensors, located in the area of the heel, heads of the first and fifth metatarsal, and the great toe, respond to 3 psi. The footswitch data were used to identify the pattern of floor contact and the basic stride characteristics.

Dynamic EMG recorded the timing and relative intensity of muscle activity. Fine wire electrodes were inserted with a 25-gauge needle into the vastus lateralis, long head of biceps femoris, short head of biceps femoris, and gluteus maximus of the amputated limb being recorded. Electrode placement was confirmed by mild electrical stimulation through the inserted wires. The EMG signals were transmitted via FM-FM telemetry to the data acquisition computer which sampled the signals at 2,500 Hz. The amplitude of EMG recorded during a "maximal" manual muscle test was used to normalize the EMG recorded during gait.

Motion of the trunk, pelvis, hip, knee, and "ankle" was recorded by the Vicon motion analysis system. Reflective markers were taped to the lateral aspect of the body overlying the axis of the hip, knee, and "ankle" joints, and the pelvis and trunk utilizing designated anatomical landmarks. Positioning of the prosthetic ankle marker was estimated from the location of that joint on the intact limb.

Forceplate recordings identified the ground reaction forces during the stance phase of gait. The forceplate is obscured in the walkway, and the subject was not informed of its presence. The subjects were positioned at the beginning of the walkway so that their natural stride would place the desired foot on the forceplate. Trials were repeated until one stride had been obtained with the reference foot landing completely on the forceplate. The current system for determining the joint moments utilized an Apple IIe computer and video recording system for visual vector display. The moments occurring at each joint during gait were calculated utilizing the visual vector (resultant ground reaction force) and joint centers from the video motion system.

All measurements, other than energy cost, were performed during both free and fast velocity walking on a 10-meter walkway with the middle 6 meters designated for data collection. In addition, ascending and descending a 20-foot, 10 percent grade ramp, and 4 stairs (with a 6-inch rise) were evaluated. The middle stride on the stairs was used for analysis to exclude acceleration and deceleration. On

the ramp and stair trials, stride characteristics, EMG, and motion of the amputated limb were recorded.

Function of the subject's contralateral limb was assessed by motion, joint moment, forceplate, and stride analysis during free and fast-paced walking.

Energy cost was determined on a 60.5 meter circular outside track. The subject's expired air was collected in a modified Douglas bag for subsequent oxygen and carbon dioxide analysis. Heart rate, respiratory rate, and cadence (using a heel switch) were telemetered by a transmitter attached to the subject. All gas volumes were corrected to standard temperature, pressure, and humidity. Each test walk lasted 20 minutes with expired air collected and physiological parameters recorded at 5-minute intervals. During the first 5 minutes, the initial 3 minutes served as a warm-up to attain a steady state with data collected in the final 2 minutes. The subsequent data collection intervals began at 9, 14, and 19 minutes.

Data Processing

The EMG data were rectified and integrated over a 0.01 second interval. Baseline noise was removed utilizing data from a quiet resting run. All gait EMG data were normalized to values determined from a "maximal" muscle test and expressed as a percent of the muscle test. Multiple strides of EMG data recorded were combined into a single representative stride for each activity tested (free/fast level walking, ascending/descending stairs and ramps). The EMG was further processed to identify the on/off timing of each muscle and the relative intensity and duration of muscle activity occurring in the functional phases of the gait cycle during these activities. The timing of the phases in stance was identified from the footswitch signals. To facilitate comparisons between subjects, and to retain the specificity of stance or swing muscle activity, each of these phases was normalized to the average stance and swing percent of the gait cycle for each activity.

Motion data were collected at 0.02 second intervals and normalized to the average stance/swing phase percent for each trial. Multiple runs and strides were averaged to obtain representative joint motion for each activity.

The fore-aft, medial-lateral, and vertical ground reaction forces were collected at 0.016 second intervals and normalized to the average stance phase duration. All values were also normalized by the subject's bodyweight. This facilitated comparison among multiple subjects. The ground reaction forces were utilized to determine the center of pressure and the magnitude of the ground reaction vector used in calculating joint torques.

The velocity, stride length, cadence, single stance, and duration of single and double stance and swing were calculated from the footswitch data. The durations of heel, foot-flat, and forefoot contact were also determined from this information.

Energy cost was defined by two measures of oxygen use, rate (milliliters per minute) and net (ml per meter walked). Physiological demand was identified by the rate of oxygen use (O_2 ml/kg/min). Efficiency of walking was determined from the measurement of net oxygen use (O_2 ml/kg/m), which combines rate and gait velocity.

Data were collected on 17 subjects under 5 different foot conditions (85 testing sessions). To date, 70 of these data sets have been processed. Data processing for the traumatic group has been completed, while 60 percent of the dysvascular group data has been processed.

RESULTS

The results of the traumatic amputee group have been analyzed and summarized. These involve the loading of the sound limb during free walking, the functional demand of stair climbing in this population, and the comparison of energy cost among the different prosthetic feet. The results of each study are given below.

Below-Knee Amputee Gait with Dynamic Elastic Response Prosthetic Feet: A Pilot Study

Preliminary results of three traumatic and two dysvascular amputees were studied for any demon strated trends in producing optimum gait. Minimal differences were noted between the five feet. The Flex-Foot resulted in significantly greater ankle dorsiflexion (19.8 ± 3.3°; $p < 0.005$) and ankle joint torque (19.9 ± 7.5 a.u; $p < 0.005$) in terminal stance compared to the other feet. All feet resulted in similar forceplate values on the amputated side. EMG analysis revealed no differences in intensity or phasing among the five feet tested; however, all had

prolonged activity in stance compared to normals. Analysis of oxygen consumption revealed no significant differences.

Of the five prosthetic feet tested, only the Flex-Foot resulted in slight changes in gait dynamics. This difference, however, was not translated to an increase in velocity or energy expenditure. The results of these five subjects suggest that there are no apparent advantages of the DER feet.

Influence of Prosthetic Foot Design on Sound Limb Loading in Unilateral Below-Knee Amputees

Forceplate, motion, and stride characteristics were analyzed in the group of 10 traumatic amputees. When comparing loading of the sound limb among the various prosthetic feet tested, it was found that the Flex-Foot caused a significant reduction in vertical ground reaction force (109 percent BW), compared with the SACH foot (135 percent BW), Carbon Copy II (129 percent BW), Seattle (127.9 percent BW), and the Quantum (129.2 percent BW) ($p < 0.001$). Stride characteristics among the five feet were similar, with the only significant difference being a greater stride length with the Flex-Foot compared to the SACH and Quantum (1.5 m; $p < 0.05$). There were no significant differences in free-walking velocity between feet. Motion analysis revealed that the Flex-Foot achieved greater dorsiflexion in terminal stance compared to all other feet ($23.3°$; $p < 0.0001$). No other differences were evident with respect to the other joints of the sound or amputated limb. These results indicate that the Flex-Foot, by nature of its large arc of ankle dorsiflexion, reduced the need to use a heel rise for tibial progression. The subsequent minimization of the rise in the body center of gravity resulted in a lower vertical loading force on the sound limb, with the implication of decreased joint reaction forces on the sound limb.

Function of the Seattle Foot in Ascending and Descending Stairs

The gait characteristics of 10 BK amputees using the Seattle foot during stair ascent and descent were compared to a group of 14 nonamputee subjects. Stride characteristics revealed that the amputee group had a significantly lower rate of ambulation on the stairs than normal (1.6 vs. 1.8 stairs per second) ($p < 0.05$). The amputee group also had significant asymmetry between their two limbs in duration of stance with a shortened stance on the amputated limb (60 percent vs. 69 percent). The nonamputee subjects had equal stance times on both legs. Motion analysis recorded significant limitations in motion of the prosthetic ankle compared to the normal ankle during stair ambulation. During loading response, the prosthetic ankle lacked normal dorsiflexion (DF) ($7°$ vs. $14°$). In early pre-swing, the prosthetic ankle was in maximum DF while the normal ankle moved into plantarflexion (PF) ($5°$ DF vs. $15°$ PF). During initial swing, the prosthetic foot remained in DF, while the normal ankle further plantarflexed to $16°$. The hip and knee joint motion was similar to that displayed by the nonamputee subjects.

The limited DF range of the prosthetic foot-ankle assembly in the amputees compromised the rocker action at the ankle, which was compensated for by forward trunk-lean in stair ascent. The forward trunk-lean augmented the forward progression, but resulted in prolonged activity of the semimembranosis. The vastus lateralis had very high intensity of action during stair descent to control forward progression over the rigid ankle rocker. Future designs for prosthetic feet should consider the need for greater DF mobility, combined with stability, to allow easier stair ambulation for the person with a BK amputation.

Below-Knee Amputee Gait in Stair Ambulation: A Comparison of Stride Characteristics Using Five Different Prosthetic Feet

Stride characteristics of 10 traumatic BK amputees using five different prosthetic feet were evaluated to determine if any provide for increased performance during stair ascent and descent. Results indicate that the Flex-Foot and Carbon Copy II foot provided for a more symmetrical gait during the initial double limb support (IDLS) phase of stair ascent. The ratios of the amputated versus sound limb for the duration of IDLS were 1.03 and 1.05 for the Flex-Foot and the Carbon Copy II, respectively, compared to 1.3 for the SACH foot ($p < 0.05$). No other significant differences in stride characteristics were found between feet. These results indicate that none of the five feet tested were clinically more advantageous for the task of stair ambulation.

A Comparison of Energy Expenditure Between Five Different Prosthetic Feet

Seventeen male amputees (10 traumatic and 7 dysvascular) were evaluated for differences in energy expenditure while wearing five different prosthetic feet. No significant differences in heart rate, respiratory rate, cadence, respiratory quotient, and energy expenditure per distance traveled were noted between any of the prosthetic feet tested. However, there was a significant difference between the traumatic and dysvascular groups in velocity (82.3 vs. 64.0 m/min; $p < 0.05$); stride length (1.49 vs. 1.26 m; $p < 0.05$); resting heart rate (65.4 vs. 82.1 b/min; $p < 0.05$); and energy expenditure (13.7 vs. 17.7 ml O_2/min; $p < 0.005$).

These results indicate that prosthetic design does not contribute to energy conservation in this population. The dysvascular amputee demonstrates increased energy expenditure as a result of the limitations of physical conditioning associated with peripheral vascular disease.

FUTURE WORK PLANNED

Upon completion of the data processing for the dysvascular group, comparisons will be made to the traumatic group across all prosthetic feet tested. Of particular interest are the loading characteristics of the sound and amputated limb of the dysvascular amputee compared to its traumatic counterpart. In addition, the stride characteristics, EMG patterns, and joint motion of the dysvascular group will be analyzed for differences across the different prosthetic feet as well as between the two groups. Energy cost of the dysvascular amputee, as well as functional demand (joint torques), will also be analyzed.

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ABSTRACTS OF RECENT LITERATURE

by
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Abstracts are drawn primarily from the orthotics and prosthetics literature. Selections of articles were made from these journals:

American Journal of Physical Medicine and Rehabilitation
Assistive Technology
Journal of Medical Engineering Technology
Medizinisch Orthopädische Technik
Prosthetics and Orthotics International

PROSTHETICS, ORTHOTICS, AND RELATED TOPICS

Die Kineplastik nach Sauerbruch: Lanzeit-Ergebnisse [Sauerbruch's Cineplasty: Long Term Results]. Baumgartner R, reprinted from *Med Ortho Tech* 112(33):39, 1992.

The "Cineplasty" of Sauerbruch, first presented in 1916 and later on further developed from Lebsche, allows direct connection between the muscles of the stump and the prosthetic hand in a unique manner. Following a brief introduction to the working principle and earlier statistics, a report is presented on 25 follow-ups on patients that have worn their prostheses continuously for over 40 years at an average. Even compared with modern prostheses, the cineplasty has advantages (physiological movement, feedback) that would justify its reintroduction. Modernisation of the prosthesis' technology is, however, urgently needed.

NOTE: Abstracts from *Medizinisch Orthopädische Technik* by permission of Rene F. Baumgartner, MD, Westfälische Wilhems-Universität, Department of Prosthetics/Orthotics and Rehabilitation, Münster, Germany.

Erfahrungen mit der "Utah" Armprothese [Experience with the "Utah" Arm]. Miault D, et al., reprinted from *Med Ortho Tech* 112:17-19, 1992.

After a brief introduction of the system, the authors report their experience with 7 patients fitted with the UTAH-type upper limb prosthesis. The results over a shorter and longer period of time are analysed and discussed. They are most encouraging both from the technical and rehabilitation point of view. Prosthetic fitting and training however must take place in specialised centers.

Krankengymnastik zur Vorbereitung des Patienten fur eine Myoelecktrische Prothesenversorgung [Physical Therapy for Preparation of Patients for a Myoelectrically Controlled Prosthesis]. Reprinted from *Med Ortho Tech* 112:20-23, 1992.

Physical therapy plays a very important role in preparing patients for the fitting of myoelectrically controlled arm prostheses. The aim is to prepare the patients, physically and mentally, for the prosthesis in order to better the changes *sic* of success. A therapeutic programme is presented which covers as far as possible all the important factors.

Prothesenseligkeit: Armseligkeit, Pladoyer fur die Versorgung mit Keiner Prothese [Prosthesis Bliss-Wretchedness: An Entreaty for Treatment with No Upper Extremity Prosthesis]. Schuling S, Baumgartner R, reprinted from *Med Ortho Tech* 112:44-47, 1992.

Statistics on the acceptance of hand and arm prostheses show clearly that a large number of patients decline use of such devices since they are able to cope better without them. Even the most modern facilities do not alter this fact in the least. Hardly anything is written on this matter. The cause

of this refusal lies primarily in the natural inferiority of even the most sophisticated technologies when compared with the original, a wonderful work of nature. Therefore, it is of paramount importance to consider the alternative possibility of treating every hand and arm amputee without a prosthesis just as positively as all other means for the patient's rehabilitation.

Uber die Akzeptanz von Armprosthesen [On the Acceptance of Upper Extremity Prostheses]. Stinus H, Baumgartner R, Schuling S, reprinted from *Med Ortho Tech* 112:7-12, 1992.

Based on a retrospective study of 307 patients fitted with upper extremity prostheses, conducted by the Department of Technical Orthopaedics and Rehabilitation of the University Münster, criteria were determined that correlate with the success or failure of the treatment. The acceptance of the prosthesis depends on correct indication, competent workmanship, and also intensive physical and ergotherapeutic training in the use of the prostheses.

Passive arm prostheses show a higher degree of success than the powered ones.

Zur Orthesenversorgung der Halswirbelsaule [On Cervical Spinal Orthotics]. Baumgartner R, reprinted from *Med Oth Tech* 112:224-228, 1992.

The therapeutic effect of cervical orthotics stands in sharp contrast to cosmetic and psychological disadvantages. These disadvantages should be reduced as far as possible. With respect to this, the indication for such devices is first discussed, followed by a review of the loadable and non-loadable surfaces of the trunk, neck and head. The presentation closes with suggestions regarding information and guidance of the patients, and further measures to be observed in connection with the orthotic treatment.

Assessment of an Infra-Red Non-Contact Sensor for Routine Skin Temperature Monitoring: A Preliminary Study. Hershler C, et al., reprinted from *J Med Eng Tech* 16:117-122, 1992.

The accuracy and reproducibility of a new non-contact sensor for monitoring skin temperature was examined. Thirty measurements taken by the device were compared with those taken by a commonly used thermocouple contact sensor. The result was a very high correlation coefficient $(r - 0.9999)$. This accuracy was achieved with the probe held at an angle of 90° 1 cm from the skin. Changes in angle and distance were found to contribute to measurement error. Little difference was found between 39 pairs of measurements taken of the left and right sides of subjects. However, intra-subject variability was noted with respect to the dermatomal segmental fields. Inter-tester reliability analysis resulted in a correlation of $r = 0.937$ involving two independent testers and 26 pairs of measurements. These preliminary data will be used for power calculations to study further the device which we found to be simple to operate, portable, and practical for routine clinical use. This sensor may have applications in the diagnosis of nerve and vascular disorders and in prospective monitoring of skin conditions such as bony areas at risk of pressure ulcers.

Causes of Death of Lower Limb Amputees. Stewart CPU, Jain AS, reprinted from *Prosthet Orthot Int* 16:129-132, 1992.

A study was carried out on the cause of death of 100 lower limb amputees who had been admitted to the Dundee Limb Fitting Centre, Tayside, Scotland for prosthetic management or wheelchair training. A comprehensive database has been established in the Centre for 25 years and the database is updated regularly. The date of death was collected and recorded. One hundred sequential deaths were investigated to review the cause of their death and compare this with the recorded causes of death for the Tayside population for the year of study. Ninety three per cent had an amputation for vascular related causes, with 73% having a below-knee amputation and 17% above-knee. Heart disease was the most frequent recorded cause of death (51%) of the amputee whereas only 28.1% of the Tayside group died from this pathology $(p < 0.01)$. Carcinomatosis was reported as a cause of death in 14% of the amputees and 23.5% of the Tayside group. Cerebrovascular disease caused death in 6% of the amputees and in 12.3% of the Tayside group (both $p < 0.01$). These findings confirm earlier suggestions that vascular amputees die from heart disease more often than the general population.

Clinical Trials and Quality Control: Checkpoints in the Provision of Assistive Technology. Kohn JG, Mortola P, LeBlanc M, reprinted from *Assist Technol* 3:67-74, 1991.

Clinical trials and quality control measures are characterized by evaluation of assistive technology by users, and feedback to providers for the purpose of improving devices or service delivery. These processes recognize that consumer satisfaction is an important measure of device and service delivery effectiveness. In this article, types of clinical trials are reviewed, and both prospective and retrospective methods of quality control are presented. The authors take the position that rehabilitation engineering centers providing customized devices and adapted technology should implement quality control measures in order to improve services to their clients.

Comparing Three Head-Pointing Systems Using a Single Subject Design. Angelo J, Deterding C, Weisman J, reprinted from *Assist Technol* 3:43-49, 1991.

The keyboard is the most commonly used input method for interfacing with computers. When using a keyboard is not possible, alternative computer input methods are needed. Three methods using head control are: Head Master by Prentke Romich, Free Wheel by Pointer Systems, and LROP by Words+.

The purpose of this study was to compare these three methods for speed and accuracy using a single subject design for nine individuals with disabilities. Visual inspection of the data revealed that subjects obtained higher scores when using Head Master and LROP than Free Wheel. As a follow-up test, an anlaysis of variance test for repeated measures showed no difference between using Head Master and LROP but did show a significant difference between Head Master and Free Wheel, and LROP and Free Wheel.

Computer Algorithms to Characterize Individual Subject EMG Profiles During Gait. Bogey RA, Barnes LA, Perry J, reprinted from *Arch Phys Med Rehabil* 73:835-841, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

Three methods of precisely determining onset and cessation times of gait EMG were investigated. Subjects were 24 normal adults and 32 individuals with gait pathologies. Soleus muscle EMG during free speed level walking was obtained with fine wires, and was normalized by manual muscle test (%MMT). Linear envelopes were generated from the rectified, integrated EMG at each percent gait cycle (%GC) of each stride in individual gait trials. Three methods were used to generate EMG profiles for each tested subject. The ensemble average (EAV) was determined for each subject from the mean relative intensity of the linear envelopes. Low relative intensity or short duration EMG was removed from the ensemble average to create the intensity filtered average (IFA). The packet analysis method (PAC) created an EMG profile from the linear envelopes in successive strides whose respective centroid %GC locations were within $\pm 15\%GC$ of each other. Control values for onset and cessation times of individual gait trials were calculated after spurious outliers were removed. Mean onset and cessation times across subjects for control values and the experimental methods (EAV, IFA, and PAC) were calculated. Dunnett's test ($p < .05$) was performed to compare control and experimental groups in patient and normal trials. EVA differed from control values for onsets ($p < .01$), cessations ($p < .01$), and durations ($p < .01$) in both normal and patient trials. EAV differed from control values for onsets ($p < .01$), cessations ($p < .01$), and durations ($p < .01$) in both normal and patient trials. IFA and PAC had no significant differences from control value means. IFA was selected for clinical use as automatic analysis could be performed on all trials and a minimum number of decision rules were needed.

Design and Evaluation of an Instrument to Measure Microcirculatory Blood Flow and Oxygen Saturation Simultaneously. Dougherty G, Lowry J, reprinted from *J Med Eng Tech* 16:123-128, 1992.

There are compelling clinical advantages in being able to monitor, simultaneously and continuously, blood oxygen saturation and tissue perfusion from the same volume of tissue. We describe the design of

a new instrument capable of combining both functions. It uses a laser diode at 805 nm to give a blood flow index and oxygenation information at the wavelength, and pulsed LED at 660 nm to complete the estimation of oxygen saturation. By using a compensating photodetector these readings can be made relatively insensitive to sampling site. After calibrating the instrument on forehead sites it reliably read the oxygen saturation of sites on the thigh with a bias of $-0•2\%$ and a precision of $\pm 1•7\%$ relative to an IL 282 CO-oximeter.

Effects of Skin Blood Flow and Temperature on Skin-Electrode Impedance and Offset Potential: Measurements at Low Alternating Current Density. Smith DC, reprinted from *J Med Eng Tech* 16:112-116, 1992.

Skin-electrode impedance was determined at 100 Hz and 1 kHz between two disposable electrodes, 5 cm apart, at current densities <65 $\mu A•cm^{-2}$. Measurements were made on the volar skin of the forearm during cooling on cardiopulmonary bypass, and on the dorsum of the foot in the absence of skin blood flow during aortic aneurysm repair. Both the resistive and reactive components of the skin-electrode impedance showed an inverse linear relationship to temperature between 26 and 36°C. The magnitude of the impedance change was different for each patient studied; resistance changes ranged from $0•03$ to $23•2$ kΩ. $°C^{-1}$ at 100 Hz and from $0•03$ to $2•7$ kΩ. $°C^{-1}$ at 1 kHz, while reactance changes ranged from $0•4$ to $2•1$ kΩ. $°C^{-1}$ at 100 Hz and from $0•04$ to $0•18$ kΩ. $°C^{-1}$ at 1 kHz. Changes in skin-electrode impedance were not due to changes in skin blood flow. There was no consistent change in offset potential with temperature. Although the skin-electrode impedance increases as temperature falls, it is concluded that temperature effects at the skin-electrode interface are not responsible for the observed failure of evoked electromyography during clinical monitoring of neuromuscular function.

An Evoked Compound Electromyogram Simulator with External Microprocessor Control Facility. Smith DC, reprinted from *J Med Eng Tech* 16:129-132, 1992.

A circuit for an evoked electromyogram simulator is described, which produces a biphasic triangular waveform similar to the evoked compound action potential seen during clinical quantitative neuromuscular monitoring. The device can produce a fading train-of-four sequence, which can be controlled using a single externally-derived voltage. The simulator is useful for bench-testing of closed loop muscle relaxant administration systems, and for teaching aspects of neuromuscular monitoring in anaesthesia.

Extensor Carpi Radialis Recovery Predicted by Qualitative SEP and Clinical Examination in Quadriplegia. Jacobs SR, et al., reprinted from *Arch Phys Med Rehabil* 73:790-793, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

This prospective study examined the efficacy of the qualitative somatosensory evoked potential (SEP) and the initial clinical neurologic evaluation to predict motor power recovery of the extensor carpi radialis muscle (ECR). Twenty three C5-6 Frankel A-D spinal cord injured (SCI) subjects had SEPs of the median nerve (MN) and superficial radial nerve (SRN) performed within 72 hours to one week post injury. The MN and SRN cortical SEPs were qualitatively graded as either present or absent. Fifteen subjects whose initial ECR muscle strength was $\leq3/5$ and eight subjects whose muscle strength was $>3/5$ were followed up to 12 to 18 months post injury for improvement in ECR muscle strength. The subject's ECR strength was evaluated by manual muscle testing (MMT) at 72 hours, weekly for three weeks, monthly for three months, and then at six, 12, and 18 months. The pin sensation at the C-5 dermatome was also tested at the above intervals and graded as either present or absent. A one tail Fisher Exact test compared the presence or absence of the MN and SRN SEPs to the recovery of the ECR to 3/5. The same one tail test also compared the presence or absence of the 72 hour C-5 pin sensation and the 72 hour MMT to the ECR recovery. Among the 15 subjects with an initial MMT of $\leq3/5$, ten subjects had successful ECR recovery ($>3/5$); 5 did not. The C-5 pin sensation correctly predicted recovery in 12 of 15 patients ($p = .06$). The initial MMT also accurately predicted ECR recovery ($p < 0.007$). In no subject did the SEP predict ECR recovery where the MMT and or

sensory test did not predict recovery. The 72 hour clinical evaluation therefore predicted functional recovery of the ECR muscle whereas the qualitative SEP did not.

Factors Influencing Reintegration to Normal Living After Amputation. Nissen SJ, Newman WP, reprinted from *Arch Phys Med Rehabil* 73:548-551, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

This study identified factors affecting reintegration to normal living (RNL) after lower extremity amputation. A questionnaire was used to evaluate RNL at a veterans' medical center and private rehabilitation clinic. The patients were 42 elderly individuals (68±1.5 years). Eighty-eight percent were men and 76% had additional health problems. Unilateral below-knee amputations, unilateral above-knee amputations, and bilateral amputations accounted for 38%, 36%, and 26% of subjects, respectively. Eleven questions were asked to evaluate mobility, self-care, work, recreation, social activities (daily functioning), relationships, social self, and life events (perception of self). The median overall RNL score was 16 of 22 (range, 5 to 22). Poor reintegration occurred in community mobility, work, and recreation. Perception of self questions showed satisfactory reintegration. Examination of variables impacting reintegration showed only additional illness significantly reducing the RNL score. It was concluded that current rehabilitative efforts regarding home mobility and psychological adjustment are satisfactory. More attention to community mobility, recreation, and additional illnesses would improve RNL after amputation.

Functional Recovery in Young Stroke Patients. Adunsky A, et al., reprinted from *Arch Phys Med Rehabil* 73:859-862, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

Thirty young stroke patients were retrospectively assessed for levels of activities of daily living and of basic functional movements. Scores upon admission, discharge, and follow-up were compared in order to evaluate course of rehabilitation and functional outcome. Mean length of stay in the rehabilitation ward was 87 ± 17 days, and duration of follow-up was 31 ± 8 months. Multivariate analysis of covariance confirmed significant improvements during hospitalization, in transfer, standing, sitting and walking abilities (F = 3.5, $p < 0.02$), as well as in activities of daily living (F = 4.7, $p < 0.01$). Further improvement during the follow-up period was observed for standing and walking abilities (F = 10.2, $p < 0.001$) only. No fatalities occurred among the patients during the study period. Eighty-one percent of the patients resumed their previous or other jobs six months after discharge. We conclude that for young stroke patients admitted to a rehabilitation ward shortly after the event, prognosis in terms of survival and functional outcome is favorable, and independent of precipitation factor, age, sex, or side of weakness.

Functional Screening of Lower-Limb Amputees: A Role in Predicting Rehabilitation Outcome? Muecke L, et al., reprinted from *Arch Phys Med Rehabil* 73:851-858, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

The Functional Independence Measure (FIM), a single-score instrument used to measure independent functioning in six areas of basic self-care skills, was used to evaluate 68 patients following lower-limb amputation. Patients in a rehabilitation hospital were assessed with the FIM upon admission and discharge. Admission scores averaged 52.7, ranging from 25.2 to 70.0. Patients scoring in the lowest and highest quartiles were compared: no remarkable gender, ethnic, or age differences were evident. Persons with the lowest scores (ie, lowest functioning) had a higher prevalence of hypertension, coronary artery disease, and noninsulin-dependent diabetes mellitus. The success of rehabilitation in patients in the lower two quartiles upon admission was variable and not predicted well by the FIM. In contrast, predictability of rehabilitation success was high in patients functioning higher at admission, the majority achieving near-perfect scores by discharge. Length of hospitalization appeared to be largely unrelated to the net difference in FIM scores over the course of hospitalization.

Isokinetic Exercise System Modification for Short Below-the-Knee Residual Limbs. Marin R, et al., reprinted from *Arch Phys Med Rehabil* 73:883-885, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

The use of isokinetic exercise has been shown to be an effective way of strengthening debilitated muscles. In the below the knee amputee, significant quadriceps and hamstring muscle wasting has been documented. Although isokinetic strengthening of the debilitated knee extensors and flexors in the below the knee amputee would be beneficial, there are no fully described isokinetic equipment modifications in literature that would allow a short below the knee amputee to effectively use isokinetic equipment. This article describes such a modification.

Lower Extremity Amputation in Scleroderma. Reidy ME, Steen V, Nicholas JJ, reprinted from *Arch Phys Med Rehabil* 73:811-813, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

Scleroderma or Systemic Sclerosis (SSC) is a disorder characterized by fibrosis of the skin and multiple internal organs. The pathological lesion is a triad of small artery intimal proliferation, medial thinning and adventitial scarring. Autoamputation of fingers and toes is often seen, but only a few cases of limb amputation in scleroderma patients have been reported. The Pittsburgh Scleroderma databank includes 1,030 patients with SSC. Among these were seven patients who sustained lower limb amputation. There were four patients with the CREST variant of SSC, two with diffuse scleroderma, and one who had SSC/rheumatoid arthritis/polymyositis overlap who sustained limb amputation. Of the seven, three were male and five had a significant smoking history. Ages ranged from 46 to 71 years. All patients underwent amputation for nonhealing ulcerations. No problems with post-operative wound healing were seen. Pathologic changes typical of SSC in addition to atherosclerotic peripheral vascular disease were described in one case. Three patients were successfully fitted with prostheses and became independent ambulators. Four patients could not be fitted with prostheses.

No skin problems were reported related to prosthetic use. Our review demonstrates that SSC patients who undergo amputation can become successful prosthetic users and should be considered for prosthetic prescription.

Mathematical Modelling and Field Trials of an Inexpensive Endoskeletal Above-Knee Prosthesis. Mohan D, Sethi PK, Ravi R, reprinted from *Prosthet Orthot Int* 16:118-123, 1992.

The swing-phase motion of the shank of an above-knee prosthesis has been modelled mathematically. An inexpensive endoskeletal prosthesis was designed using the Jaipur foot and conduit pipes with a hinge joint for the knee. Results of field trials and the modelling indicate that a very simple above-knee prosthesis can give near normal gait at "normal" walking speeds on flat surfaces. The swing of the shank is most sensitive to the timing of toe-off.

A Pilot Study Comparing Mouse and Mouse-Emulating Interface Devices for Graphic Input. Kanny EM, Ansor DK, reprinted from *Assist Technol* 3:50-58, 1992.

Adaptive interface devices make it possible for individuals with physical disabilities to use micro-computers and thus perform many tasks that they would otherwise be unable to accomplish. Special equipment is available that purports to allow functional access to the computer for users with disabilities. As technology mores from purely keyboard applications to include graphic input, it will be necessary for assistive interface devices to support graphics as well as text entry. Headpointing systems that emulate the mouse in combination with on-screen keyboards are of particular interest to persons with severe physical impairment such as high level quadriplegia. Two such systems currently on the market are the HeadMaster and the FreeWheel.

The authors have conducted a pilot study comparing graphic input speed using the mouse and two head-pointing interface systems on the Macintosh computer. The study used a single subject design with six able-bodied subjects, to establish a baseline for comparison with persons with severe disabilities. Results of these preliminary data indicated that the HeadMaster was nearly as effective as the mouse

and that it was superior to the FreeWheel for graphics input. This pilot study, however, demonstrated several experimental design problems that need to be addressed to make the study more robust. It also demonstrated the need to include the evaluation of text input so that the effectiveness of the interface devices with text and graphic input could be compared.

Predicting Life Satisfaction Among Adults with Physical Disabilities. Kinney WB, Coyle CP, reprinted from *Arch Phys Med Rehabil*, 73:863-869, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

The perceptions of life satisfaction among adults with physical disabilities were examined in this research. Personal interviews were conducted with 790 adults who had a physical disability. Data were collected using the Center for Epidemiological Studies Depression Scale, Rosenberg's Self- esteem Scale, and the Life 3 Scale. Results from a stepwise multiple regression analysis ($n = 344$) indicated that leisure satisfaction was the most significant predictor of life satisfaction, explaining 42% of the variance in the life satisfaction scores for this population. An additional 11% of the variance in life satisfaction was explained by scores on financial status, self-esteem, health satisfaction, religious satisfaction, and marital status. Findings from this research highlight the role that leisure satisfaction plays in enhancing life satisfaction among individuals with physical disabilities. Furthermore, the findings suggest that leisure and life satisfaction levels are influenced by employment status and whether the disability was acquired. Discussion centers on the potential contribution that therapeutic recreation can have in the rehabilitation arena.

Recovery Following Complete Paraplegia. Waters RL, et al., reprinted from *Arch Phys Med Rehabil* 73:784- 789, 1992. (©1992 by the American Congress on Rehabilitation Medicine and the American Academy of Physical Medicine and Rehabilitation.)

Motor and sensory recovery were quantified by serial examinations prospectively performed on 148 persons with paraplegia. Of the 142 patients who remained complete injuries at follow-up, none with an initial neurologic level of injury (NLI) above T9 regained any lower extremity motor function at follow-up. Thirty-eight percent of patients with an initial NLI at or below T9 had some return of lower extremity motor function, primarily in the hip flexors and knee extensors. Twenty percent of the patients with an initial NLI at or below T12 regained sufficient hip flexor and knee extensor strength to reciprocally ambulate using conventional orthoses and crutches. Unlike motor function, recovery of light touch and sharp-dull discrimination was independent of the initial NLI. Six (4%) of the 148 patients demonstrated "late" conversion (more than 4 months after injury) from complete to incomplete spinal cord injury (SCI) status. Two of the 6 patients with an initial NLI at T12 and subsequent annual NLI at L1 and L2 reciprocally ambulated, and three of the 6 patients regained voluntary bladder/bowel control.

Rehabilitation After Amputation for Vascular Disease: A Follow-Up Study. DeLuccia N, et al., reprinted from *Prosthet Orthot Int* 16:124-128, 1992.

Rehabilitation of one hundred and twenty eight patients with lower limb amputation performed for vascular disease from 1979 to 1987 was assessed. Arteriosclerotic occlusive disease was the most frequent cause of amputation (85.9%). Sixty seven patients (52.3%) were diabetic. Early and late results were analysed. For long-term follow-up evaluation, Univariate method of Kaplan-Meyer product limit was employed. Multifactorial analysis was used to assess factors influencing mortality. On immediate evaluation of rehabilitation with a prosthesis 85.2% of patients were successfully fitted. On long term evaluation 47.8% of below-knee and 22.1% of above-knee amputees were alive and using the prosthesis full time at five years of follow-up ($p = 0.0026$). Opposite limb preservation at five years was 69.5% for diabetics and 90.2% for non-diabetics, respectively ($p = 0.0013$). Survival rate at five years was 42.4% for diabetics, and 85.0% for non-diabetics ($p = 0.0002$). On multifactorial analysis diabetic patients showed a risk of late mortality six times greater than non-diabetics. In conclusion rehabilitation after vascular amputation is feasible in a large number of patients, despite a limited life

span. Diabetes represents a major risk factor both for life and for the opposite limb. Knee preservation is an important factor for better rehabilitation.

A Review of Practices Among Information Resource Programs on Assistive Technology. Tractman LH, reprinted from *Assist Technol* 3:59-66, 1991.

There is presently a growing need for timely information on assistive technology products and services. Some exemplary assistive technology information resource programs exist, while others are just now being developed. A survey was used to review the practices of 24 information resource programs. The results, obtained from 15 responding programs, provide guidelines for those considering setting up such services of their own. Among the significant findings, 60% of the programs receive between 10 and 100 requests per month; over two-thirds of these come in by telephone, but over one-half must be answered by mail; the average time spent per request is 33 minutes; over one-half of all requests are for product literature, and over one-

half of all requests come from service providers. Some critical issues related to information delivery are presented for further investigation.

A Swedish Knee-Cage for Stabilizing Short Below-Knee Stumps. Isakov E, et al., reprinted from *Prosthet Orthot Int* 16:114-117, 1992.

Stump length is an important factor in attaining successful prosthetic rehabilitation in below-knee (BK) amputees. Stability of the stump-prosthesis complex is impaired in the case of a stump shorter than 10 cm. Thus, fitting a prosthesis to a BK amputee with a stump which is very short often requires the use of different prosthetic techniques. In this work, the authors suggest the use of a Swedish knee-cage attached to a conventional patellar-tendon-bearing prosthesis as an alternative solution in the case of a short BK stump. Objective evaluation was performed by an analysis of gait and the foot-ground reaction forces. The results obtained indicate an improvement in all the measured parameters resulting from the modified stump-prosthesis complex.

BOOK REVIEWS

by
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Professor Emeritus of Sensory Rehabilitation, New York University, and David Peikoff Chair in Deafness Studies, University of Alberta, Edmonton, Alberta, Canada
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An Annotated Bibliography on Visual Technology for Deaf and Hard of Hearing People, edited by Barbara M. Virvan. GRI Occasional Paper 92-1. Washington, DC: Gallaudet Research Institute, Gallaudet University, 1992, 60 pp.
by Jerome D. Schein, Ph.D.

The Technology Assessment Program continues to turn out broadly useful documents, and this one is no exception. It supplements an earlier version that covered the years 1976 to 1988. The present monograph brings together the widely scattered literature on visual aids for persons with impaired hearing, updating the coverage to 1991 (though a few references dated 1992 are included).

The 198 entries and annotations are organized under the following headings: Text Telecommunications, Captioning, Alerting and Alarm Systems, Speech Technology, Video Telephone, Face-to-Face Communication, Technology for Deaf-Blind People, Surveys and Needs Assessments, Tutorial Reviews and Overview Articles, Policy, Resource Guides, Newspapers and Newsletters, and Database. The contents of these sections include fugitive documents—mainly technical reports of research that easily can be missed by even assiduous literature searchers. Many such references should not be missed: for example, all but one of the entries in the section on alerting and alarm systems is either a conference proceeding or technical report that contains empirical findings and recommendations important to persons with impaired hearing and those responsible for the environments in which they work

and reside; the only article is from the *Fire Journal,* a publication rarely scanned by rehabilitationists.

The editor has made frequent use of the authors' abstracts, often a wise decision. Where annotations have been specially prepared, they are clear and concise. Anyone interested in this rapidly advancing technology will find this bibliography eliminates the need for searching databases for items prior to 1992.

Interpreting the Scores. A User's Guide to the 8th Edition Stanford Achievement Test for Educators of Deaf and Hard of Hearing Students, by Judith A. Holt, Carol B. Traxler, and Thomas E. Allen. GRI Technical Report 92-1. Washington, DC: Gallaudet Research Institute, Gallaudet University, 1992, 49 pp.
by Jerome D. Schein, Ph.D.

Psychologists often complain about the lack of psychometrics normed for special target groups, but seldom do much about it. For the past two decades, the Center for Assessment and Demographic Studies (CADS), rather than succumbing to the inadequacies of educational achievement tests for students with impaired hearing, has responded by adapting the popular Stanford Achievement Test (SAT). This effort has entailed much more than drawing stratified random samples of students with impaired hearing and analyzing their performances on the several forms of the SAT, though that alone would be worthy of praise. The SAT-HI (for SAT- "Hearing-Impaired" version) provides an innovative approach to assessment, beginning with a pretest to

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overcome the floor and ceiling effects that bedevil psychometrics applied to markedly deviant populations, and incorporating special scoring procedures to facilitate comparisons to the original SAT. Even more noteworthy is the rich accumulation of 20 years' data about this group's educational progress. With the added advantage of having a national sample of non-hearing-impaired persons for comparison on the same test, educators, rehabilitators and researchers have a valuable reference base.

The report under review supplements no less than five earlier publications about the eighth edition of SAT-HI alone. It contains detailed tables of normative data, as well as discussions of points relevant to researchers and educators alike. Those wishing to take full advantage of the instrument should, however, study the many earlier reports about SAT-HI's other seven editions. When stepping back to survey the full panoply of SAT-HI publications, one sees not only an outstanding exemplar for serving the educational and rehabilitation needs of students with impaired hearing, but also of other groups with disabilities. The authors of the technical report and those who have contributed to this long- term effort deserve more credit than they have received: first, for conceiving of it, and then for patiently tending to its day-by-day demands. Quite a departure from hit-and-run research that so often characterizes special-education and rehabilitation studies. CADS staff— past and present— merit the gratitude of all those who have profited and will continue to profit from their assiduous labors.

Academic Recovery After Head Injury, by Dinah Russell and Ankita Sharratt. Springfield, IL: Charles C. Thomas, 1992, 93 pp.
by Corwin Boake, Ph.D.

This brief book is intended as a how-to guide to help adult educators and college faculty members work with students who are survivors of head injury. The book's authors are a head-injury survivor (who also earned a master's degree in education) and a professional educator. The book adds to a small but important body of literature about head-injury survivors in higher education. The book's six brief chapters cover, in order, an introduction to the head-injury field, overviews of brain functions and cognitive deficits relevant to education, practical tips for working with students who have had head injuries, an account of one author's experience when she returned to college after head injury, and a concluding chapter that advocates using adult education as a means of accomplishing the same goals as cognitive rehabilitation programs. There are numerous quotations from students who have had head injuries, many of them commenting on being embarrassed by having brain injuries and school performance problems. Unfortunately, there are no descriptions of higher education programs that are specially designed for students with brain injuries. One example of such a program is Richland Community College in Dallas, Texas. Higher education programs could be a major method of making brain injury re-entry services accessible to many more persons with brain injury. Persons interested in brain injury programming in higher education will enjoy and profit from reading this brief book, but should also consult *The Head-Injured College Student,* by Cooper B. Holmes (Charles C. Thomas, Springfield, IL, 1988).

PUBLICATIONS OF INTEREST

This list of references offers *Journal* readers significant information on the availability of recent rehabilitation literature in various scientific, engineering, and clinical fields. The *Journal* provides this service in an effort to fill the need for a comprehensive and interdisciplinary indexing source for rehabilitation literature.

All entries are numbered so that multidisciplinary publications may be cross-referenced. They are indicated as *See also* at the end of the categories where applicable. A listing of the periodicals reviewed follows the references. In addition to the periodicals covered regularly, other publications will be included when determined to be of special interest to the rehabilitation community. To obtain reprints of a particular article or report, direct your request to the appropriate contact source listed in each citation.

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Acta Orthopaedica Scandinavica
Advances in Orthopaedic Surgery
American Annals of the Deaf
American Journal of Occupational Therapy
American Journal of Physical Medicine and
  Rehabilitation
American Journal of Sports Medicine
American Rehabilitation
Annals of Biomedical Engineering
AOPA Almanac (American Orthotic and Prosthetic
  Association)
Applied Optics
Archives of Physical Medicine and Rehabilitation
ASHA (American Speech and Hearing Association)
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  Cybernetics
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JAMA
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Medical Psychotherapy Yearbook
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Military Medicine
New England Journal of Medicine
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Scandinavian Journal of Rehabilitation Medicine
Science

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Sports 'N Spokes
Technical Aid to the Disabled Journal
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Topics in Geriatric Rehabilitation
VA Practitioner
Vanguard
Volta Review
Worklife

CALENDAR OF EVENTS

1993

August 24-26, 1993
15th Annual Meeting of the Thai Orthopaedic Association on "Axial Skeleton Injuries: Basic and Clinical Research for Fractures of the Spine and Spinal Cord Injuries," Bangkok, Thailand
Contact: Secretariat Office, Chusakdi Suwansirikul, MD, The Thai Orthopaedic Association, 94/2 Supavadee Tower, 100, Soi Mitanant, Nakornchaisri Road, Dusit, Bangkok 10300, Thailand; Tel: (02) 243 5834-6, ext. 1002; Fax: (02) 243 5837

September 15-17, 1993
Orthopaedica Belgica Annual Congress, Brussels, Belgium
Contact: Medicongress, Waalpoel 28, B-9960 Assenede, Belgium; Tel: 32-91 44 40 96; Fax: 32-91 44 40 10

October 7-8, 1993
Northwest Regional Conference of the Association for the Education and Rehabilitation of the Visually Handicapped, Boise, Idaho
Contact: Ken McCulloch; Tel: 208-338-3655*

October 11-15, 1993
37th Annual Meeting, Human Factors and Ergonomics Society, Seattle, WA
Contact: The Human Factors and Ergonomics Society, PO Box 1369, Santa Monica, CA 90406-1369; Tel: (310) 394-1811; Fax: (310) 394-2410*

October 12-16, 1993
American Orthotic and Prosthetic Association (AOPA), Annual National Assembly, Reno, NV
Contact: Annette Suriani, AOPA, 717 Pendleton St., Alexandria, VA 22314; Tel: (703) 836-7116

October 19-21, 1993
Biotechnica '93, Biotechnology Trade Fair, Hannover, Germany
Contact: Deutsche Messe AG, Messegelande, W-3000 Hannover 82, Germany; Fax: 49 (0)511 8932626

October 27-29, 1993
1st North American Regional Conference of Rehabilitation International, Atlanta, GA
Contact: Program Coordinator, North American Conference of Rehabilitation International, 45 Sheppard Ave. East, Suite 801, Toronto, Ontario, Canada, M2N 5W9; Tel: +1-416-250-7490; Fax: +1-416-229-1371

October 28-31, 1993
15th Annual International Conference of the IEEE Engineering in Medicine & Biology Society, San Diego, CA
Contact: Susan Blanchard, P.O. Box 2477, Durham, NC 27715; Tel: (919) 493-3225*

October 30-November 3, 1993
17th Symposium on Computer Applications in Medical Care (SCAMC), Baltimore, MD
Contact: The George Washington University Medical Center, Office of Continuing Education, 2300 K St. NW, Washington, DC 20037; Tel: (202) 994-8928

October 31-November 5, 1993
American Academy of Physical Medicine & Rehabilitation (AAPM&R), Annual Conference, Miami Beach, FL
Contact: AAPM&R, 122 South Michigan Ave., Suite 1300, Chicago, IL 60603; Tel: (312) 922-9366

November 19-22, 1993
American Speech-Language-Hearing Association (ASHA) Annual Convention, Anaheim, CA
Contact: Frances Johnston, ASHA, 10801 Rockville Pike, Rockville, MD 20852; Tel: (301) 897-5700

December 7-12, 1993
American Academy of Neurological and Orthopaedic Surgery: 17th Annual Convention, Las Vegas, NV
Contact: Dr. Michael R. Rask, 2320 Rancho Drive, Suite 108, Las Vegas, NV 89102-4592

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(no date yet), 1994
16th Annual International Conference of the IEEE/EMBS, Baltimore, MD
Contact: Dr. Joshua Tsitlik, Johns Hopkins School of Medicine, Rm 410 Traylor Bldg., 720 Rutland Ave., Baltimore, MD 21205

January 31-February 4, 1994
Visions in Mobility International Mobility Conference-7, Melbourne, Australia
Contact: Royal Guide Dogs Associations of Australia, Chandler Highway, Kew, Victoria 3101, Australia; Tel: +61-3-860-4444; Fax: +61-3-860-4500

March 15-20, 1994
American Academy of Orthotists and Prosthetists (AAOP), Annual Meeting and Scientific Symposium, Nashville, TN
Contact: Annette Suriani, AAOP, 1650 King St., Alexandria, VA 22314; Tel: (703) 836-7116

April 6-12, 1994
17th International Conference on Medical and Biological Engineering & 10th International Conference on Medical Physics, Rio de Janeiro, Brazil
Contact: Mr. OZ Roy, Sec Gen Intl. Union for Physical and Engineering Sciences in Medicine, c/o National Research Council, Room 307, Building M-50, Ottawa, Ontario, K1A OR8, Canada*

April 7-9, 1994
BME'94 International Conference on Biomedical Engineering, Hong Kong
Contact: BME'94 Conference Secretariat, c/o Rehabilitation Engineering Centre, Hong Kong Polytechnic, Hunghom, Kowloon, Hong Kong; Tel: 852-766-7683; Fax: 852-362-4365*

April 9-16, 1994
IRMA VII—Seventh World Congress of the International Rehabilitation Medicine Association: 25th Anniversary of IRMA, Washington, DC
Contact: IRMA VII, 875 Kings Hwy., West Deptford, NJ 08096

April 9-16, 1994
XIIth World Congress of the Rehabilitation Medicine Association, Washington, DC
Contact: Ms. D. Jo..s, 1333 Moursund, A-221, Houston, TX 77030

April 17-22, 1994
11th International Congress of the World Federation of Occupational Therapists, London, United Kingdom
Contact: Conference Associates and Services Ltd - WFOT, Congress House, 55 New Cavendish Street, London W1M 7RE, UK; Tel: 071-486-0531; Fax: 071-935-7559

May 31-June 2, 1994
Annual Meeting of International Medical Society of Paraplegia, Japan
Contact: Host Organizer, IMSOP 1994 Annual Meeting, Japan Organizing Committee, Orthopedic Department of Tokushima University 3, Kuramoto-cho, Tokushima-shi, 770, Japan; Tel: 0886-31-3111; Fax: 0886-33-0178

June 2-3, 1994
Improving the Quality of Physical Therapy, International Conference, 's-Hertogenbosch, The Netherlands
Contact: Mr. J. Dekker, PhD or Ms. E. Zoer, PO Box 1568, 3500 BN Utrecht, The Netherlands; Tel: 31-30-319946; Fax: 31-30-319290*

August 15-19, 1994
Rehabilitation Ergonomics, Toronto, Ontario, Canada
Contact: IEA '94 Secretariat, c/o JPdL Multimanagement Inc., Toronto Dominion Centre, 55 King Street West, Suite 2550, Toronto, ON, Canada M5K 1EZ; Tel: (416) 784-9396; Fax: (416) 784-0808*

August 21-26, 1994
17th International Conference on Medical and Biological Engineering and 10th International Conference on Medical Physics, Rio de Janeiro, Brazil

Contact: Conference Secretariat, Rua do Ouvidor, 60/414 Rio de Janeiro, CEP 20040, Brazil; Tel: +5521224 6080; Fax: +5521231 1492

September 4-9, 1994
6th European Regional Conference of Rehabilitation International, Budapest, Hungary
Contact: Rehabilitation Secretariat, ISM Limited, The Old Vicarage, Haley Hill, Halifax HX3 6DR, UK; Tel: 44(0)422 359 161; Fax: 44(0)422 355 604

October 11-15, 1994
American Orthotic and Prosthetic Association (AOPA), Annual National Assembly, Washington, DC
Contact: Annette Suriani, AOPA, 717 Pendleton St., Alexandria, VA 22314; Tel: (703) 836-7116

November 18-21, 1994
American Speech-Language-Hearing Association (ASHA), Annual Convention, New Orleans, LA
Contact: Frances Johnston, ASHA, 10801 Rockville Pike, Rockville, MD 20852; Tel: (301) 897-5700

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April 2-7, 1995
International Society for Prosthetics and Orthotics (ISPO), 1995 World Congress, Melbourne, Australia
Contact: ISPO, Australian National Member Society, Repatriation General Hospital, Banksia St., Heidelberg, 3081 Victoria, Australia; Tel: 61 (03) 499 6099

July 9-16, 1995
4th World Congress of Neuroscience, Kyoto, Japan
Contact: Host Organizer, Secretariat, 4th World Congress of Neuroscience, c/o International Communications, Inc., Kasho Building, 2-14-9, Nihonbashi, Chuo-ku, Tokyo 103, Japan; Tel: 03-3272-7981; Fax: 03-3273-2445

November 17-20, 1995
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LETTERS TO THE EDITOR

To the Editor:

Re: **Automated Fabrication of Mobility Aids (AFMA): Below-Knee CASD/CAM Testing and Evaluation Program Results***

San Francisco STAMP (Special Team for Amputations and Mobility Preservation) is enthusiastic about CAD/CAM, which is emerging as a valuable tool for prosthetists, and we anticipate that there will be a great deal of research performed to assess its efficacy in a variety of applications. Therefore we read with great interest the AFMA Final Report [Houston et al.] which appeared in the Fall 1992 *Journal of Rehabilitation Research and Development* (Vol 29, No. 4, p. 78). While this report did provide an interesting look at clinical application of computer-assisted technology, there are some gaps that need further examination before applying the results in a non-research program.

We would like to have seen a more detailed accounting of the inclusion and exclusion criteria for facilities, prosthetists, and patients participating in the study. Do the participating facilities represent the typical VA site, or were they selected by specific criteria? Also, since the report concludes that a new graduate and a seasoned prosthetist do not produce CAD/CAM limbs of comparable quality, it would be important to assess the range of clinical experience of prosthetists actually working in the VA system. Is the patient population of this report at all similar to the VA patient population? If not, this data may have very limited relevance for the over-all VA system. For example, the cause of amputation in 66% of the subjects was trauma and 31% peripheral vascular disease; on our inpatient service, those numbers would be closer to 1% and 95%, respectively. If the study had separated out and analyzed that subgroup of prosthetic users of less than five years, which was closer to our inpatient group (a large majority with dysvascular etiology), the results would have been far more useful. Unless DVA is proposing to fabricate permanent limbs for outpatient veterans, the total study population bears little resemblance to the vast majority of amputees we treat.

More disturbing, however, is that the report seems to be missing critical information regarding the actual logistics of the study. For example, which sites did remote

fabrication and which did on-site fabrication? A separate breakdown of patient satisfaction vis a vis fabrication site may be illuminating. It is commonly accepted among many of the researchers involved in this study that the results using the remote model were far less satisfactory than the on-site product. We would have liked to have seen the data analyzed and presented to either support or refute this perception. This is particularly important in that this and similar studies will likely guide the actual direction of CAD/CAM implementation within the VA, as well as prosthetic production throughout the public and private sector.

It is our fear that inappropriate application of this exciting technology may give CAD/CAM a bad name, especially given the difficulty the private sector has historically had making remote fabrication work profitably while producing a quality prosthesis. The issues raised above demand further study as a means of remote fabrication. Future research will be more generalizable if it selects a population that resembles the VA population and reports relevant methodological variations. Until then, it will not be appropriate to begin making decisions regarding implementation of CAD/CAM.

Sincerely,
Jerry Goldstone, MD, Director
David Kramer, PT, Coordinator
William Doyle, CP
Patricia Daley, PT
Special Team for Amputations and Mobility Preservation (STAMP)
San Francisco Dept. of Veterans Affairs Medical Center
4150 Clement St.
San Francisco, CA 94121

The Authors Reply:

Ernest Burgess, MD, replies:

I have reviewed the comments in the Letter to the Editor submitted by Jerry Goldstone, MD, and the STAMP Program at VAMC, San Francisco.

Dr. Goldstone's comments and critique are well stated. The AFMA Final Report which appeared in the Fall 1992 *Journal of Rehabilitation Research and Development* represented work completed more than 3 years prior to its appearance in the *Journal*. It represents a three-center study at distant geographical locations and in a sense is a preliminary report. The data submitted are entirely accurate as far as they go. Development and extension of AFMA techniques have proceeded at a very

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*Automated Fabrication of Mobility Aids (AFMA): Below-Knee CASD/CAM Testing and Evaluation Program Results. Houston VL, Burgess EM, Childress DS, Lehneis HR, Mason CP, Garbarini MA, LaBlanc KP, Boone DA, Chan RB, Harlan JH, Brncick MD. J Rehabil Res Dev 1992:29(4), 78-124.

rapid pace since the date of completion of the Houston, et al. report.

At this time, the AFMA system is essentially an accepted fact in the profession/industry. Further clinical experience with improvements both in software and fabrication equipment have rendered irrelevant the question as to whether or not automation for mobility aids will replace the existing labor-intensive craft nature of the standard techniques.

As of this date, approximately 105 private facilities in the United States and Europe have converted to AFMA technology. One of the three research centers involved in the AFMA Final Report has already fitted or supervised the fitting of 3,500 above-knee and below-knee amputees. The VA Prosthetics and Sensory Aids Service is installing the technology in 9 primary centers and 21 outreach networking centers during the current fiscal year. Training programs for prosthetists and technicians have been set up and are operating within the VA system and at independent universities. In addition to the many amputee-related benefits resulting from AFMA, the cost savings are large, even including capitalization of the equipment.

The challenge now for those of us doing research ar 1 development is to improve and extend automation to other mobility aids including, specifically: orthoses, seating for SCI and other patients, corrective insoles and footwear, etc. This R&D is underway and needs to be extended.

Dr. Goldstone and his colleagues at their excellent Service at VAMC San Francisco are in an excellent position to participate and contribute to this effort.

Sincerely,
Ernest M. Burgess, MD
Director and Principal Investigator
Prosthetics Research Study
Seattle, WA 98122

Vern L. Houston, PhD, CPO, replies:

In their letter, Dr. Goldstone, Messieurs Kramer and Doyle, and Ms. Daley observe that the National AFMA Program did not address all of the areas involved in the application of CAD/CAM technologies in Prosthetics. They are correct. It is hard to address every facet of every issue, especially in a field as broad as the application of CAD/CAM technologies in prosthetics. We were conscious of this, and tried to point out in the AFMA Report areas where further work is needed and issues where the Study data was too sparse to indicate more than statistical trends. One such area is the application of prosthetics CAD/CAM technologies in the post-surgical fitting and rehabilitation of new amputees. With the trend during the past decade to shift back to soft post-surgical dressings and delayed prosthetic fitting and patient ambulation, from the previously advocated regimen of immediate post-surgical rigid plaster dressings and aggressive rehabilitation therapy, with early weight-bearing and ambu-

lation, the issue of the optimum post-surgical treatment regimen and rehabilitation modality, especially for new peripheral vascular amputees, is again being debated. The potential impact of prosthetics CAD/CAM technologies in the post-surgical care and rehabilitation of new amputees, with prosthetics CAD/CAM systems' ability to accommodate precisely and rapidly residual limb geometric and volumetric changes, remains to be determined. Because the application of CAD/CAM technologies in clinical prosthetics settings was new at the time the Study was initiated, it was purposely decided to include only test subjects in the Study who had successfully worn a prosthesis without undue complications for a period of at least one year, until the clinical capabilities and limitations of prosthetics CAD/CAM systems were precisely established (Houston, et al. p. 80). Since the conclusion of the National AFMA Testing and Evaluation Program, researchers have obtained very promising initial results using second-generation prosthetics CAD/CAM systems (*viz.*, Shapemaker and CanfitPlus™) to design and manufacture temporary sockets and prostheses (Brncick, 1992; Wu and Krick, 1992). However, a definitive study with a statistically valid and representative sample population of new amputees remains to be performed.

In their letter, they question the protocol used in the Study for selection of the patients, prosthetists, and facilities that participated in the AFMA Program. This is well-documented in the report (pp. 80, 85, 90, and 91). For participation in the Study, patients were required to be unilateral below-knee amputees; 18 years of age or older; capable and willing to give their informed consent; to meet the prescription criteria for a PTB prosthesis; and to have been fit with and have worn a BK prosthesis without unusual problems for a period of at least one year prior to entry into the program. Prosthetists at outreach institutions/facilities were required to: be competent in the clinical practice of prosthetics; be able to provide one or more patients meeting the AFMA protocol given above; have sufficient interest in the program to collect the experimental data requested; and conscientiously fit and provide accurate feedback regarding the AFMA CAD/CAM sockets. These were the only criteria that were required for participation. As stated in the report, 198 patients and 66 prosthetists at 44 institutions and facilities participated in the first part of the Study, the Outreach Program, the principal goal of which was to introduce prosthetics CAD/CAM technologies to rehabilitation health care professionals throughout the United States. Of ⌐ ∵, 102 were patients and 30 were prosthetists at ⌐ DVA Medical Centers across the United States. In the third part of the Study, the comprehensive and carefully controlled clinical testing and evaluation trials, 90 patients and 34 prosthetists at 24 institutions and facilities participated through one of the three DVA AFMA Centers (the Prosthetics Research Study in Seattle, WA; Northwestern University/VA Lakeside Medical Center in Chicago, IL; and the New York DVA Medical Center/New York University Medical Center in New York, NY). Of these, 59 were patients and 13 were

prosthetists from 10 DVA Medical Centers. Whether those Medical Centers were representative of the "typical" VA facility, we do not know. They certainly were not atypical. We do not believe this to be important. Rather, we believe the pertinent issues, and the ones for which we expended considerable effort in the Study to try to ensure their validity, are: 1) the degree with which the test subject sample population "represented" the US veteran lower-limb amputee population; and to a lesser extent, the degree with which it represented the general amputee population in the United States; and 2) the degree to which the level of skill, training, and experience of the prosthetists participating in the Study represented that of the prosthetists in the United States who provide care for US veteran lower-limb amputees, and secondarily for lower-limb amputees in the general US patient population. As exact statistics on these parameters are not available, the degree to which the test subject and prosthetist sample populations match the respective US veteran, general lower-limb amputee population, and prosthetist populations in the United States cannot be precisely ascertained. This is the reason we described in considerable detail the characteristics of the test subject sample population enlisted in the comprehensive clinical trials in the third part of the Study (pp. 91-96). The fact that 66% of the test subjects in the comprehensive clinical trials in the third part of the Study were traumatic amputees and 31% were amputated because of complications from peripheral vascular disease (PVD), we believe to be reasonably representative of the US veteran lower-limb amputee population, and probably not all that different from the general lower-limb amputee population in the United States. Of those test subjects who underwent amputation in the past 5 years or less at the time of the Study, 80% were amputated because of PVD. Furthermore, of the test subjects whose amputations were due to PVD, 72% underwent amputation 5 or fewer years prior to their participation in the Study (see **Figure 1** below). The bias in the distribution of the test subjects with amputation due to PVD versus time since amputation, in large part reflects the fact that PVD is systemic, and impacts the general health and medical condition of patients. Exact statistics on the US veteran and general lower-limb amputee populations in the United States are not available, but these figures coincide with results from other studies conducted over the past decade (Dept.of Veterans Affairs, 1992; National Center for Health Statistics, 1990, 1992; National Institute on Disability and Rehabilitation Research, 1992).

In the comprehensive clinical testing and evaluation trials in the third part of the Study, a total of 34 prosthetists participated. Of these, 11 were American Board for Certification in Prosthetics and Orthotics (ABC) certified prosthetist-orthotists, 9 were ABC certified prosthetists, 4 were ABC certified orthotists, and 10 were not ABC certified. This is the only measure indicative of the participating prosthetists' level of training, skill, and experience we obtained. Some prosthetists learned how to use the CAD software and CAD/CAM
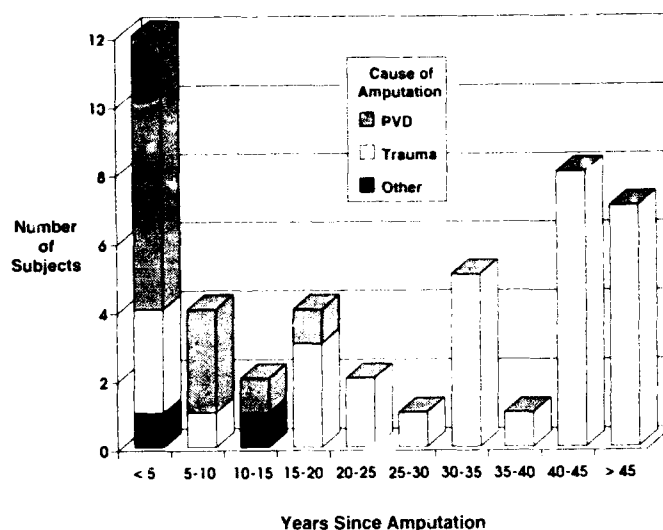

Figure 1.

AFMA test subjects' number of years since amputation categorized by the subjects' cause of amputation.

equipment more quickly than others, and some prosthetists became more adept in designing and manufacturing sockets with the CAD/CAM systems during the program than others. There was a reasonable correlation between the level of training and amount of experience of the participating prosthetists and their ultimate capabilities using the CAD/CAM systems, but not a one-to-one correspondence. This is also true in conventional prosthetics practice, so assessment of prosthetist training and experience does not necessarily reflect prosthetists' conventional or CAD/CAM design and manufacturing capabilities and skills. Technical, conceptual, and functional skills and capabilities are very difficult to ascertain accurately, and as such, it was agreed they were beyond the scope of the Study. They remain for others to assess.

With regard to the assertion that since the patient population (for which they provide care and service) in the DVAMC San Francisco, CA is composed of approximately "95% (peripheral vascular amputees) and 1% (traumatic amputees), the Study results, as reported, lack relevance, unless the DVA is proposing to fabricate permanent limbs for outpatient veterans." This is a bit parochial. As they are well aware, the DVA is responsible for the care and rehabilitation of all US veterans, not just those treated in DVA facilities. The application of plastic laminates in prosthetics, the patellar tendon bearing (PTB) BK socket, the total contact AK suction socket, and the SACH foot, to name a few items, were all developed under DVA funded research to improve the prosthetics care of US veterans, and secondarily the care of the general lower-limb amputee population in the United States. They were not developed just for use in DVA medical centers for the care of veteran inpatients. Similarly, it was the goal of the National AFMA Program to test the then current prosthetics CAD/CAM systems

and evaluate how these systems in particular, and CAD/CAM technologies in general, could impact the prosthetics care and rehabilitation of US veterans, whether this care and these services were provided at DVA facilities or at private, commercial facilities and institutions. In addition, it was the objective of the Study to determine those areas where further research and development were required to improve and enhance prosthetics CAD/CAM technologies to make them effective and efficient clinical prosthetics tools for the care and rehabilitation of U.S. veterans. We believe the DVA National AFMA Program was reasonably successful in meeting these goals. Admittedly, the results would probably be of greater use to Dr. Goldstone and other DVAMC rehabilitation health care professionals, if the Study had been confined to the specific cross-section of patients for which they provide care and service. In developing the protocol for the Study and in analysis of the Study results, we tried to make the Study and present the results we felt would have the greatest applicability for the greatest number of readers. If we fell short in this, then we must accept responsibility.

In analyzing the results of the Study, we compiled and analyzed over 400 parametric plots of the data. When we wrote the report summarizing the results, we necessarily had to limit what was presented. Some of the results in the report are given explicitly in terms of cause of amputation, and as such, are directly applicable to the prosthetics patient population at the DVAMC San Francisco. In particular, **Figures 36, 42, 45** (of the original article—pp. 107, 109, 110), show the work time (personnel hours) required to produce successfully fitting CAD/CAM sockets and prostheses. These figures show there was a trend that, on the average, the PVD subjects required less work to successfully fit than the traumatic amputee test subjects. However, much stronger correlations with ease/difficulty of successful CAD/CAM socket design and fit were found to exist between residual limb length, years since amputation, and subject activity level. Amputees with short residual limbs, regardless of whether they were amputated because of complications from PVD, trauma, or other causes, were generally more difficult to successfully design sockets for and fit. This is logical, as subjects with short residual limbs have far less surface area over which to distribute standing and ambulatory loads than do subjects with medium length or long residual limbs. In addition, the subjects who were long-term, chronic amputees (who were amputated 20 or more years prior to participating in the Study), and especially those that were very active (who used their prostheses for standing and/or walking in more than 75% of their daily activities), and who were very satisfied with the fit, comfort, and function of their previous, conventionally designed and manufactured prostheses, were generally very difficult to fit. As none of the Study's PVD amputee subjects had been amputated longer than 17 years, they were not in this category, and thus were generally easier to fit than the most active, chronic traumatic amputees. Other results, not included in the

report, given explicitly for the test subjects amputated because of PVD, are given in **Figures 2-5** below. These results should aid rehabilitation health care professionals at other DVAMCs and other institutions, in more directly extrapolating the Study results to their own prosthetics patient populations.

Figures 2 and 3 (below) show the pre-AFMA (conventionally designed and manufactured) and AFMA (CAD/CAM designed and manufactured) socket and prosthesis ratings of those test subjects amputated because of PVD. These results can be compared to those for the total test subject sample population given in **Figures 22 and 23** in the original report (pp. 96, 99). It is seen that more of the test subjects amputated because of PVD rated the fit, comfort, and function of their AFMA sockets and prostheses very-good-to-fair than rated their conventionally designed sockets and prostheses very-good-to-fair, and they generally rated the fit, comfort, and function of their AFMA sockets and prostheses slightly higher than did their traumatic and other amputee test subject counterparts. **Figure 4** below shows the number of check sockets required to obtain successful CAD/CAM socket designs and fits categorized by cause of amputation. **Figure 5** shows the total work time (personnel hours) required to design and manufacture successful CAD/CAM sockets and prostheses categorized by cause of amputation and by residual limb length. As noted above, the test subjects amputated because of PVD generally required fewer check sockets and less work to achieve successful CAD/CAM socket designs and fits for than did many of their traumatic amputee test subject counterparts. The test subjects with shorter residual limbs generally required more work to successfully design sockets for and fit than did those subjects with medium length and long residual limbs. The PVD amputee test subject sample population with short and long residual limbs was small, however, so further data should be
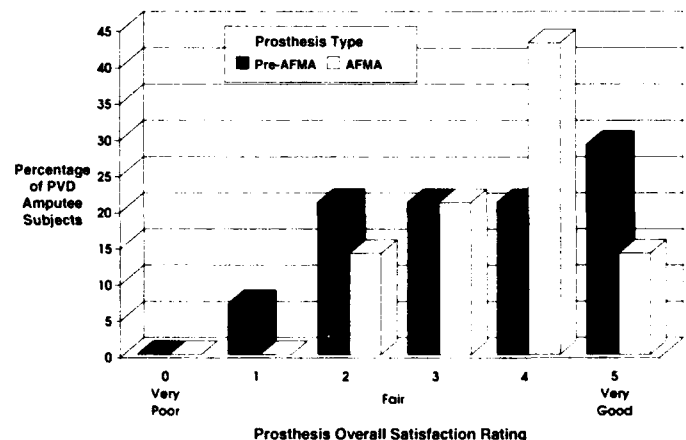

Figure 2.

AFMA PVD amputee test subjects' ratings of the degree of their overall satisfaction with their conventional, pre-AFMA prostheses and with their AFMA CASD/CAM prostheses.
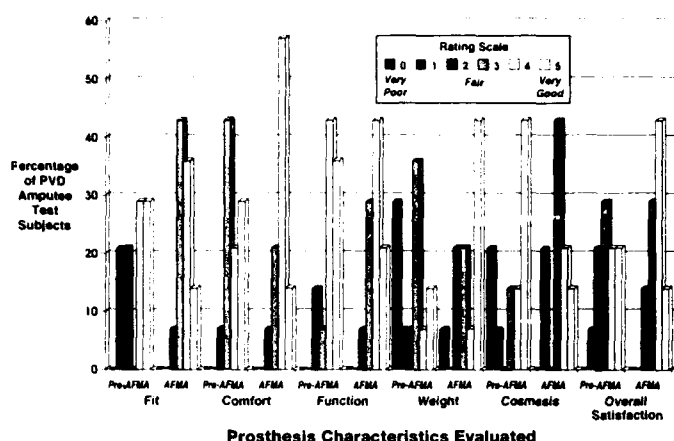
Figure 3.
AFMA PVD amputee test subjects' ratings of their conventional, pre-AFMA prostheses and their AFMA CASD/CAM prostheses.

compiled to confirm this trend. Hopefully these additional results will be of value to other rehabilitation health care professionals whose prosthetics patient populations consist principally of newer patients amputated because of PVD.

We are somewhat perplexed by their comment faulting the logistics of the Study. In particular, we fail to understand their "disturbance" that the Study results were not conditioned on and presented in terms of which facilities did onsite fabrication and which did not. Their comment that, "It is commonly accepted among many of the researchers involved in this Study that the results using the remote (fabrication) model were far less satisfactory than the onsite product," seems to confirm their misunderstanding of the objectives and scope of the AFMA Program. As stated in the report (p. 80), the objectives were: 1) to provide a limited introduction of
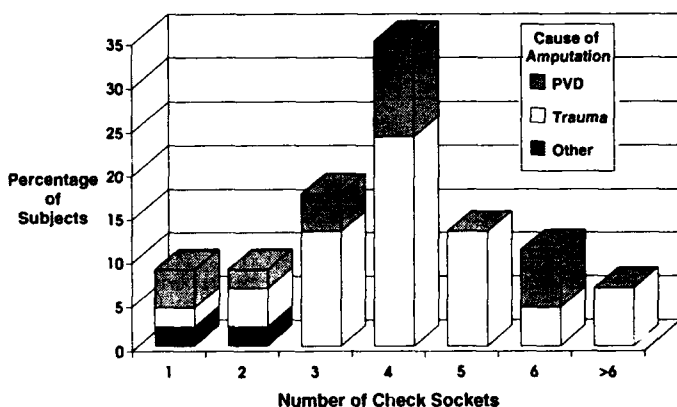

Figure 4.
Number of check sockets required to achieve successful socket designs and fittings with the UCL-BC CASD/CAM System, categorized by AFMA test subject cause of amputation.
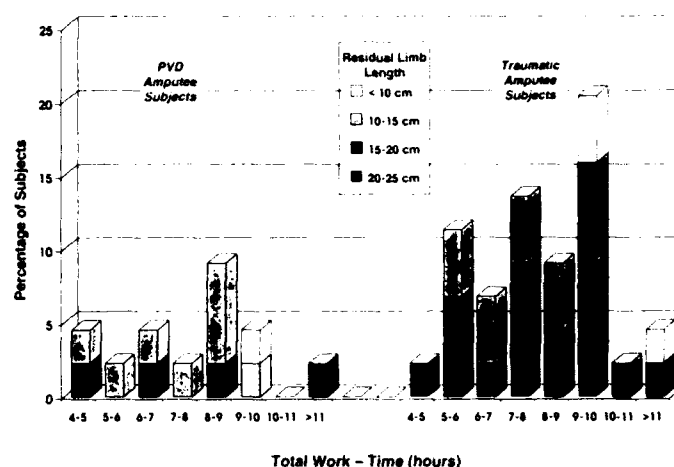

Total Work - Time (hours)

Figure 5.
Total work-time (personnel hours) required for successful CASD/CAM socket and prosthesis design, production, and fitting as a function of residual limb length, categorized by AFMA test subject cause of amputation.

CAD/CAM technologies to the rehabilitation health care field in the United States; 2) to evaluate the feasibility of utilizing CAD/CAM technologies in clinical prosthetics settings in the United States; 3) to developmentally test the UCL-BC and UBC-MERU BK CASD/CAM systems; and, 4) to obtain quantitative physiological, biomechanical, prosthetics, and engineering data for use in refinement of these systems and for development of new, enhanced, more efficacious and expedient CAD/CAM systems. To accomplish these objectives the Program was designed to consist of four parts. The first part was a clinical outreach program which was designed to introduce prosthetics CAD/CAM technologies to as many rehabilitation health care professionals and institutions/facilities across the United States as possible. In this part of the program CAD/CAM sockets were designed, manufactured, and modified on a remote "mail order" basis, with the local prosthetists evaluating, measuring, and casting the patients, providing design instructions to the prosthetists at the AFMA Centers, and subsequently fitting the resultant AFMA CAD/CAM sockets designed and manufactured at the AFMA Centers. This protocol was followed in the AFMA outreach program, except where the local participating prosthetists were in close enough physical proximity to one of the DVA AFMA Centers to come in and use the AFMA CAD/CAM software and equipment themselves. Only limited patient records and data were compiled during this phase of the Study, mainly for the use of the AFMA prosthetists designing the CAD/CAM sockets. The second part of the Study consisted of extensive laboratory testing of the CAD/CAM equipment and software to determine its capabilities, limitations, accuracies, repeatability, consistency, and maintenance requirements. The third part of the Study involved comprehensive, carefully controlled,

clinical testing and evaluation of the CAD/CAM software and equipment at each of the three AFMA Centers. Extensive medical, physiological, biomechanical, prosthetics, and engineering data were collected and compiled in a computerized database during this part of the program. Some prosthetists not on the research staffs at the AFMA Centers participated in this part of the Program. Almost all of the prosthetists involved in this phase of the Study used the AFMA CAD/CAM software to design CAD sockets for their own patients, and many of them used the CAM equipment to manufacture sockets for one or more of their patients as well. The fourth part of the Program consisted of compilation and analysis of the data compiled during the previous three parts of the program.

They seem to have confused the outreach program conducted during the first part of the Study and the comprehensive, carefully controlled, clinical testing and evaluation program conducted during the third part of the Study. Some of the results achieved during the outreach program during the first part of the Study were less than "ideal," and differed from the results subsequently obtained during the comprehensive clinical testing and evaluation trials conducted by the three AFMA Centers during the third part of the Study. However, there is no evidence that differences in rates of success and in the amount of work required to achieve successful socket designs and fits in the two parts of the Study have anything to do with the site of CAD socket fabrication. The extensive laboratory tests conducted during the second part of the Study established the accuracy of the CAM equipment to be less than 1 mm in radial and axial dimensions and less than one degree in angular dimension, and once the initial "glitches" in the CAM equipment were corrected, the repeatability and consistency of the equipment were shown to also be within these tolerances over a period of greater than 2 years. Thus, there could not have been any perceptible physical differences due to site of fabrication in the AFMA CAD/CAM sockets.

Because of the accuracy, repeatability, and consistency afforded by prosthetics CAM systems, as long as socket/prosthesis manufacture "turn-around time" is not increased, there is little difference if a CAD/CAM socket/prosthesis is manufactured by a technician in an onsite laboratory or in a laboratory several thousand miles away, except possibly for the sense of personal gratification and accomplishment obtained by the prosthetist standing around watching the CAM milling machine carve out a positive model of his/her CAD design, or watching the CAM thermoformer vacuum-form the negative socket model—time better spent attending to the needs of his/her patients, evaluating and designing sockets for new patients, or furthering his/her prosthetics and technical education. Because of the levels of accuracy, repeatability, consistency achievable by prosthetics CAM systems, and the demonstrated quality of their output, the only issues that remain to be assessed by whomever and wherever CAD/CAM systems are planned for implementation are: patient needs; types of

service demands; equipment utilization and capacities (productivity); support personnel requirements; and costs. These issues must be carefully considered. It is not cost-effective to install a complete CAD/CAM system at an institution or facility that only provides prosthetics care for one or two patients per quarter, especially if such equipment can be utilized at or near its capacity in a shared mode of operation. On the other hand, if a 2-hour or less turn-around time for prosthesis design and manufacture of intimately fitting sockets for removable, postsurgical prostheses is frequently required, it may be fully justifiable to install a complete suite of CAD/CAM equipment in an institution or facility, even if it is not utilized to its fullest capacity. As noted in the letter, these are very important issues that need to be considered in planning and implementing CAD/CAM systems. With the exception of establishing and demonstrating the productivities clinically achievable with prosthetics CAD/CAM systems, these issues were not addressed and were beyond the scope of the Study.

Differences in the degree of fit and levels of comfort and function that occurred between the CAD/CAM sockets designed and manufactured during the AFMA outreach program, and those designed and manufactured during the comprehensive, clinical testing and evaluation phase of the Study, can only be attributed to: 1) differences in training, experience, and skill of the prosthetists involved in the outreach phase, and the comprehensive clinical testing and evaluation phase of the Program; 2) inadequate or inaccurate patient information acquisition and/or errors in socket design (redesign) information transmission between the local outreach prosthetists measuring, fitting, and evaluating the test subjects and the AFMA prosthetists performing the CAD socket design; or 3) preconceived prejudices and psychological attitude differences between the subjects and/or prosthetists involved in these two parts of the Study.

Differences in prosthetist training, experience, skill, and technical and conceptual capabilities led to some differences in the results achieved with the CAD/CAM systems. At the lower end of the scale, analysis of the test results showed that if less knowledgeable, less experienced, and less skilled prosthetists followed the CAD/CAM systems' default design suggestions, the results they achieved tended to be better than the worst case conventionally designed and manufactured prostheses. This should hold even more so now, with the development from the DVA AFMA Program of improved and enhanced "second-generation" prosthetics CAD/CAM systems. The upper end of the spectrum should also be considerably improved with the development of second-generation CAD/CAM systems, such as Shapemaker, CanfitPlus™, and Screenform II, which provide a large armamentarium of effective design tools, so the most highly trained and skilled prosthetists are no longer limited in their capabilities, as they were in the first-generation CAD/CAM systems tested under the AFMA Program. The second cause of differences between the results achieved in the two parts of the Study—

problems in prosthetist information and measurement acquisition for patient characterization and transmission of information for CAD socket and prosthesis design/redesign—shall be difficult to avoid until the patient evaluation and measurement process, and the socket/prosthesis fitting and evaluation process with feedback for socket design refinement are completely quantified. These are the current areas of active prosthetics CAD/CAM research, and CAD systems solving these problems probably won't exist until sometime in the next 5 to 10 years. Until that time, aspects of the prosthetics CAD input, design, and fitting processes shall remain subjective, and thus be highly dependent upon the prosthetist evaluating and measuring the patient, the prosthetist designing the socket and prosthesis, and the prosthetist fitting and evaluating the socket and prosthesis and providing feedback for design refinement. If these are not one in the same person, then the potential for loss of information shall exist. The third potential cause of performance differences between the Study outreach and comprehensive clinical testing and evaluation programs—preconceived prejudices and psychological attitudes—can only be overcome with time. There are always individuals, be they patients, prosthetists, physicians, or therapists who are receptive to new and different ideas, devices, procedures, etc., and there are those who are resistive to change. This was the case when the VA began introducing plastics technology, the PTB BK socket, the total contact AK suction socket, and the SACH foot in the 1950s and 1960s. It is now very difficult, if not almost impossible, to find patients wearing and prosthetists who can successfully design and fabricate wood and leather, plug-fit sockets and prostheses with hand-crafted, articulated feet and ankles. Presumably it will be equally as difficult to find prosthetists utilizing conventional design and manufacturing techniques, and patients wearing conventionally designed and manufactured sockets and prostheses in another decade.

The effects each of these factors had upon the Study results is impossible to assess, as the data compiled are too sparse to enable statistically valid inferences to be made. The Study was not designed to evaluate: 1) differences in prosthetist socket/prosthesis design, fitting, and evaluation capabilities due to differences in training, experience, skill, and conceptual and technical abilities; 2) differences between remote and local prosthetics design practices; and, 3) effects of prosthetists' and patients' prejudices and psychological attitudes. Dr. Goldstone, et al. are correct in their assertion that additional studies are needed if these factors are to be evaluated.

Although we have no scientific or statistical proof, from a professional perspective, we would concur with them that the best prosthetics care may not be able to be provided for veteran amputees, and amputees in the general patient population in the United States, if a remote (centralized) CAD design structure is implemented. Many aspects of the patient evaluation, measurement, and prosthetics characterization process, and the patient/prosthesis fitting and evaluation process shall remain subjective, at least in the near future. As such, we believe that the same prosthetist(s) should evaluate, measure, design the socket/prosthesis for, and fit and evaluate a given patient. We do not agree with Dr. Goldstone et al. that on-site versus remote CAM makes any difference in the quality of prostheses that are produced. Local versus remote manufacture is an issue dependent upon type of service needed, productivity, and costs, not on quality. The reference they make in their letter to difficulties with the quality of prostheses from centralized fabrication facilities in the private sector, to our knowledge, have been problems attributable to remote design, not to remote manufacturing. Where established central fabrication facilities, staffed with well-trained, competent prosthetists and technicians, have had problems, the causes almost invariably have been due to insufficient and/or inaccurate patient measurements, poor or damaged patient residual limb casts, and/or failure to adequately communicate socket/prosthesis design specifications. The quality of manufacture has rarely been the issue. Such problems with patient and socket/prosthesis design information errors can be averted entirely in a CAD/CAM framework, as long as CAD socket/prosthesis design is performed locally by the prosthetist who evaluates, measures, and fits the patient. If reliable communications channels are utilized, and competent, well-trained technicians are employed to operate the CAM equipment, there will be no difference in the quality of the CAD/CAM prostheses produced, whether they are manufactured locally or remotely. CAM numerical control code is no different if the CAM equipment is located across the room in the local prosthetics laboratory or several thousand miles away in a centralized CAM facility.

Finally, we wish to note that the objective of the DVA CAD/CAM research is to develop a set of quantitative tools for prosthetists that will enable them to more effectively, efficiently, and expeditiously provide better fitting, more comfortable, and more functional prostheses for US veteran amputees, and secondarily for amputees in the general US patient population, regardless of whether this care is provided at DVA Medical Centers or in private institutions and facilities. Future studies addressing just the sector of the veteran population cared for at DVA facilities are certainly warranted as suggested, but the DVA has not in the past, nor should ever in the future, limit sponsored research to just the segment of the veteran population treated at DVA facilities. It is the responsibility of the DVA to ensure that all US veterans who have served their country receive the best possible prosthetics care and rehabilitation, irrespective of where they choose to obtain prosthetics care and services.

Sincerely,
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