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A Digital Master Hearing Aid

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Abstract — The use of computer simulation in evaluating conventional and experimental hearing aids is described. Two illustrative examples are provided. The first involves the simulation of a conventional master hearing aid and its application in evaluating different adaptive strategies in the prescriptive fitting of hearing aids. The second example involves the simulation of an experimental hearing aid embodying modern digital signal-processing techniques for the reduction of background noise. A high-speed array processor is used in order to accomplish these simulations in real time.

INTRODUCTION

There are three basic approaches to the design of digital hearing aids. The first is to design instruments that essentially duplicate the operation of conventional (analog) hearing aids. The second approach is to develop digital hearing aids that embody the same basic principles as analog instruments, but make use of digital techniques to incorporate features that would have been difficult and/or impractical to achieve with conventional analog circuits. The third approach is to develop digital instruments that are conceptually very different from conventional analog hearing aids and which embody characteristics that, for all practical purposes, can be achieved only by advanced digital signal processing techniques.

An example of the first approach is a digital hearing aid consisting of a conventional hearing-aid microphone and preamplifier followed by an analog-to-digital converter, a digital filter, a digital-to-analog converter and a power amplifier driving a conventional hearing aid receiver. Prototype units of this type have already been developed; see Levitt and Sullivan, 1984 (1).

An example of the second approach is a digital hearing aid that can be programmed to best meet the needs of individual users. A fairly sophisticated digital filter may be required in order to obtain the optimum frequency-gain characteristic for each individual. Although programmable analog hearing aids have already been developed — see Mangold and Leijon, 1979 (2) — there are limitations to the range and flexibility of the analog filters that can be used in a practical system. Digital filters have fewer limitations. Among the important advantage of programmable instruments is that they allow a unified treatment of both the design and fitting of hearing aids: see Engebretson et al., 1983 (3), and Popelka and Engebretson, 1983 (4).

The third approach offers the greatest potential for significant improvements in aiding the hearing impaired. Innovative methods of signal processing that are currently being explored include

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frequency lowering, enhancement of perceptually important speech characteristics, and automatic noise reduction: e.g., Levitt, Pickett and Houde, 1980 (5). Experimental systems for investigating different forms of digital speech processing for the hearing impaired are already being developed in several research laboratories and it is only a matter of time before experimental hearing aids embodying advanced signal-processing techniques are developed.

In this context, with the application of advanced techniques a subject of wide and intense current interest, it is surprising that relatively little attention seems to have been directed toward the use of a computer to simulate hearing aids. Computer simulation is particularly well suited for the evaluation of innovative experimental hearing aids; e.g., digital hearing aids incorporating advanced signal processing techniques. Computer simulation may also be used to advantage in the simulation of conventional hearing aids for clinical applications, such as the prescriptive fitting of hearing aids using a master hearing aid, as described in Levitt in 1978 (6).

The purpose of this paper is to describe the application of computer-simulation techniques to the development of an extremely flexible, general-purpose master hearing aid. Two illustrative applications of the system are described. The first example involves the simulation of a conventional master hearing aid, but with the added feature that the system is computer controlled, thereby greatly facilitating the adjustment process. The second example involves the simulation of an experimental hearing aid in which modern digital signal-processing techniques are used to suppress background noise.

TECHNICAL DESCRIPTION

A block diagram of the basic system is shown in Figure 1. The simple diagram shows only the major components; actually the system is extremely flexible and allows many variations.

The input signals (typically but not necessarily audio signals) may be from a microphone, tape recorder, or FM receiver. The anti-aliasing filter limits the bandwidth of the signals processed by the system to a value compatible with the sampling rate of the digital system. This sampling rate is under computer control. Although a sampling rate as high as 125 kHz can be used, the time required to process signals in real time sets practical limits on the sampling rate. In general, the lower the sampling rate, the greater the amount of signal processing that can be achieved. A useful and practical compromise for this application has been a sampling rate of 12 kHz. A bandwidth of 6 kHz is theoretically usable with this sampling rate, but in practice a bandwidth of just over 5 kHz is used in order to allow for practical constraints such as roll-off in the anti-aliasing filters. This bandwidth is considered to be adequate for most hearing-aid simulations. In special applications, such as simulating "wideband" hearing aids, a sampling rate as high as 25 kHz can be used, but then the amount of signal processing is limited to little more than frequency shaping.

The signals from the antialiasing filter are digitized and fed into one or more storage buffers. A 12-bit analog-to-digital converter is used. A multiplexer is employed when simulations involve more than one input channel, as in the case of a binaural hearing aid.

The heart of the system is a MAP-300 array processor. This device allows high-speed vector operations and is well suited to performing extremely rapid Fourier transformations and convolutions. For example, a 1024-point fast Fourier transform (FFT), corresponding to 85.3 msec of speech at a sampling rate of 12 kHz, can be performed in less than 10 msecs. This capability allows much of the signal processing to be performed in the frequency domain, with many attendant advantages.

The array processor is controlled by a minicomputer (DEC LSI-11/23). The controlling computer feeds programs to the array processor and in this way controls what the array processor does. The controlling computer can also be used to monitor the subject's responses, analyze data, check the output of the array processor (for self-calibration and checking) and for the running of experiments. In the latter application, the controlling computer can be used to implement an adaptive test strategy whereby, depending on the subject's responses, the simulated hearing aid converges on the optimum parameter values for that individual user.

The limitations of the system are determined primarily by the operating time of the array processor and its high-speed memory. The current system (which can be expanded and speeded up by adding more high-speed memory) can perform at least eight FFT's per time window at a sampling rate of 12 kHz, and can store a filter impulse response of about 200 msec for realtime operation. For non-realtime applications, both the number of FFT's that can be implemented per time window, and the length of the system impulse response, are essentially unlimited.

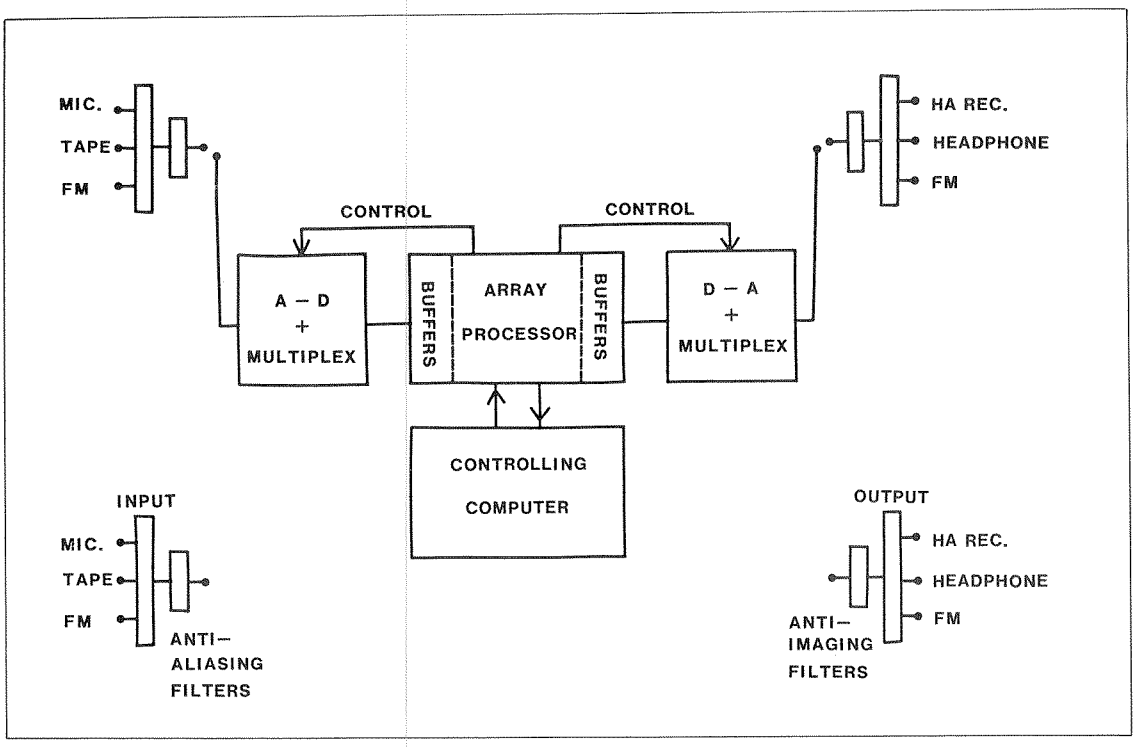

FIGURE 1
Components of Computerized System.

Note that in the realtime mode of operation, when using discrete time-windows, there is an inherent time delay between input and output equal to the duration of at least two time windows. If recursive filtering without buffering is used, the delay is reduced to that of the sampling period, e.g., roughly 80 microseconds at a 12-kHz sampling rate. In its current mode of operation, the discrete time-window approach is used. For the applications used to date, 256-point FFT's have been employed resulting in input/output delays of a little over 40 msec at the 12-kHz sampling rate.

The output of the array processor is stored temporarily in buffers, after which the signals are converted to analog form by a 12-bit digital-to-analog converter. A multiplexer is used for multichannel applications. In order to avoid the problem of transients between successive time windows, an overlap-add procedure is used in which two concurrent time waveforms using overlapping time windows are generated and then added, as described by Allen in 1979 (7).

The output of the analog-to-digital converter is fed to an appropriate acoustic delivery system (e.g., headphones, hearing-aid receiver, FM transmitter). A major concern regarding the clinical use of a desktop or rack-mounted master hearing aid is that the acoustic characteristics of a wearable personal hearing aid are inherently different from that of a non-wearable instrument. Not only are head and body baffle effects difficult to take into account, but there appears to be a complex interaction between head movements, which can be exquisitely fine, and associated auditory perceptions. In order to circumvent differences between wearable and nonwearable hearing aids, an FM signal transmission may be used because this can make the simulated hearing aid acoustically identical to that of a wearable personal hearing aid.

For applications requiring the simulation of wearable personal hearing aids, a conventional hearing-aid microphone and receiver have been mounted in a standard casing; thus far, a standard postauricular hearing-aid case has been used. The same approach can also be used for simulating an in-the-ear hearing aid that requires a custom-made mold. (For the postauricular simulation, the output of the hearing-aid receiver is led to a custom earmold using standard acoustic tubing.)

In a typical simulation, the electrical signal generated by the subject's wearable microphone is pre-amplified and led to a pocket-size FM transmitter whose transmitted signal is picked up by an FM receiver located at the input to the computer. The output of the computer is transmitted back to the subject to be picked up by a pocketsize FM receiver/

amplifier, the output of which is fed directly to the hearing-aid receiver.

For all practical purposes, the acoustic input and output stages of such a simulated hearing aid are identical to those of a conventional personal hearing aid. However, the electronic processing of the signals that takes place between microphone and output transducer has been transferred to the computer and back by means of the FM transmission system. (It is important to note that two independent transmitter-receiver systems, with non-interfering carriers, are used so that the hearing aid can transmit and receive at the same time.)

EXAMPLES OF SIMULATION

1. Conventional Master Hearing Aid

The potential of the computer-controlled hearing aid simulation system as a research and clinical tool is best illustrated by means of examples.

In the first example, the system is used to simulate a conventional master hearing aid under computer control. In addition to providing frequency-selective amplification, the system has been programmed to generate such test stimuli as tones or bands of noise, as needed. In this application the system serves as both a hearing aid and an audiometer.

A block diagram of the simulated master hearing aid (Fig. 2) shows that the input from each channel of this two-channel system is multiplexed and fed into one of four buffers, two buffers being used for each channel. The method of simulation uses the overlap-add procedure developed by Allen (7).

This system is in use in an experiment comparing different adaptive strategies for adjusting the frequency-gain characteristic of a master hearing aid. A general approach to the prescriptive fitting of frequency-gain characteristics is to begin with a first estimate of the optimum frequency-gain characteristic based on psychoacoustic considerations, followed by systematic adjustment of these characteristics to arrive at an improved estimate (6). Typically, a listener's judgment of relative speech intelligibility is used as the criterion for an "improved" estimate, although other criteria such as relative speech quality may also be used. The adjustment procedure is continued iteratively until no further improvements in the prescribed criterion can be obtained.

The specific implementation of this general procedure began in the current experiment with a first estimate based on the method developed by Pascoe (8). According to that technique the subject's threshold, loudness discomfort level, and several intermediate levels (very soft, soft, comfortable, loud and very loud) are obtained using third-octave bands of noise. The test stimuli are generated and controlled through the computer, using an adaptive

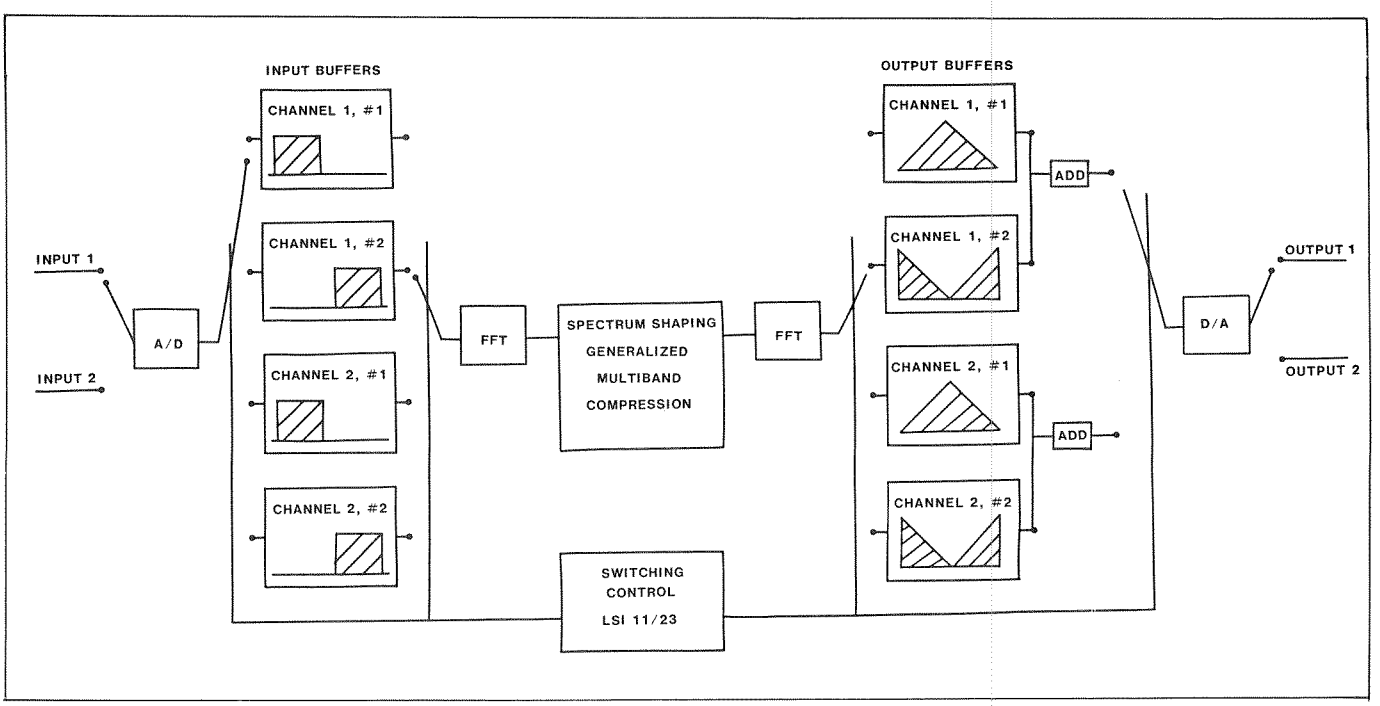

FIGURE 2
Block diagram of master hearing aid.

up-down strategy to converge efficiently on the levels indicated. The first estimate of the optimum frequency-gain characteristic is obtained by finding the frequency-gain curve that places the rms level of the speech stimulus at the average comfort level over the frequency range of interest, 200 to 5000 Hz. Subsequent estimates of the optimum frequency-gain characteristic are then obtained by systematically adjusting this frequency-gain characteristic and testing for improved intelligibility.

Methods of frequency-gain adjustment compared—An experiment has been performed to compare different methods of adjustment. One method, referred to as the "round-robin" procedure, was used in earlier investigations with a manually controlled master hearing aid and was described by Sullivan et al. in 1983 (9). A 3 × 3 matrix was formed with the central cell of the matrix corresponding to the first estimate. The rows of the matrix correspond to changes in low-frequency gain, the columns correspond to changes in high-frequency gain. Paired-comparison judgments of relative intelligibility were obtained for all possible pairs of cells in the matrix. The frequency-gain characteristic of the cell showing the largest number of "wins" in the paired comparisons became the second estimate of the optimum frequency-gain characteristic. Then the round-robin strategy was repeated using this second estimate as the central cell. Unless contraindicated by the data, it was usually possible to terminate testing after the second or third estimate of the frequency-gain characteristic had been obtained.

The second adaptive procedure used a double-elimination paired-comparison strategy of the type described by Montgomery in 1982 (10); this procedure is referred to as the "tournament" strategy. A 4 × 4 matrix of cells was formed by adding a row and column to the original 3 × 3 matrix. As before, rows of the matrix correspond to changes in low-frequency gain and columns correspond to changes in high-frequency gain. The tournament was then performed on all of the cells in the matrix using paired-comparison judgments of relative intelligibility. The frequency-gain characteristic corresponding to the winning cell became the second and final estimate of the optimum frequency-gain characteristic.

The third adaptive strategy used a variation of the simplex procedure (6) and is referred to as the "modified simplex" method. A two-way matrix of possible cells was formed using the same basic design as for the round-robin and tournament

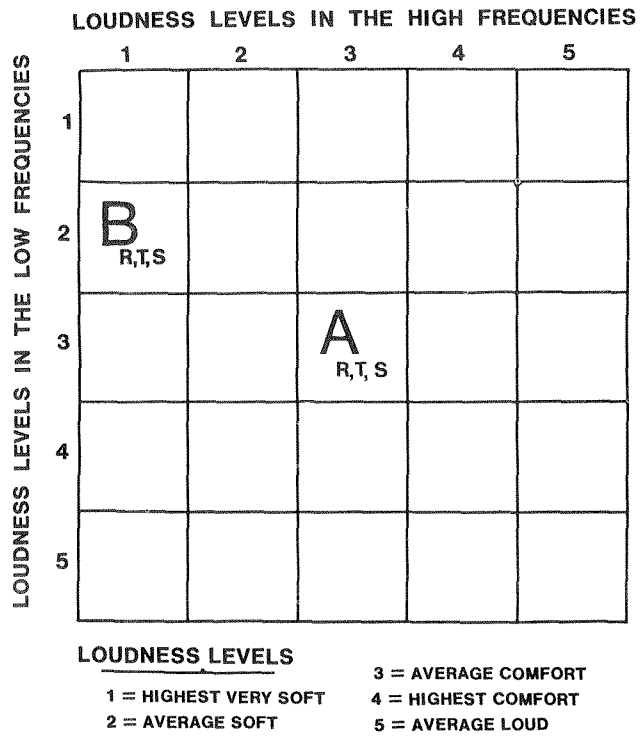

FIGURE 3
Typical data for comparison among adaptive strategies

strategies, i.e., rows correspond to changes in low-frequency gain, and columns to changes in high-frequency gain. An elemental L-shaped set of three adjacent cells was formed with the vertex corresponding to the first estimate of the optimum frequency-gain characteristic. Paired comparisons of relative intelligibility were then obtained between the vertex and the two adjacent cells. Depending on the results of these paired comparisons, a new L-shaped set was formed in an adjacent region away from less intelligible cells of the old L-shaped set. A sequence of L-shaped sets thus generated should converge on an improved estimate of the optimum frequency-gain characteristic. The criterion for convergence was three reversals in the direction of movement along each axis (i.e., across rows and columns of the large matrix of possible cells).

The results obtained for two typical subjects are shown in Figure 3. The diagram shows the set of possible cells centered on the first estimate of the optimum frequency-gain characteristic. The symbols A and B identify the two subjects; the suffixes R, T, S identify the round-robin (R), tournament (T) and modified simplex (S) strategies, respectively.

The location of each symbol identifies the cell corresponding to the final estimate of the optimum frequency-gain characteristic. In this case, for each

subject, all three strategies converged on the same cell. However, preliminary results on additional subjects indicate that this pattern does not necessarily hold, although the differences in the estimates obtained by different adjustment strategies are relatively small. Differences between estimates are to be expected when the optimum condition being sought is represented by a relatively shallow peak or ridge in the response surface.

An important difference between the two subjects is that for one subject, A, the final estimate of the optimum frequency-gain characteristic is the same as the initial estimate (i.e., the central cell in Figure 3 corresponds to the first estimate). For the second subject, B, the cell corresponding to the final estimate of the optimum frequency-gain characteristic differed from the initial estimate. (Subject B also consistently judged speech to be more intelligible for the final estimate of the optimum frequency-gain characteristic than for the initial estimate.)

All of the paired-comparison judgments of relative intelligibility were obtained by having the listener judge the intelligibility of a continuous discourse passage presented against a background of cafeteria noise. This is an extremely efficient method of testing, but the judgments are necessarily subjective. However, when a final check was performed using a more time-consuming objective measure of speech-recognition performance (Northwestern University Auditory Test #6), the scores supported the results obtained with the more efficient paired-comparison technique.

The relative efficiency of the three methods of adjustment is shown in Table 1. The round-robin strategy, which can be implemented manually without too much difficulty, was the least efficient of the three procedures both in terms of time taken and the average number of trials required. The most efficient strategy was the modified simplex procedure, which converged on the estimated optimum frequency-gain characteristic in a fraction of the time of the slower round-robin procedure. The modified simplex is a difficult technique to administer manually, however, and some degree of computerization is needed for its implementation. (Although it is not as efficient as the modified simplex procedure, the tournament strategy is also a relatively efficient technique, and it is also a relatively robust technique which can, if necessary, be administered manually using a conventional master hearing aid.)

TABLE 1
Comparison Between Adaptive Strategies*

	Subject A	Subject B
Round-Robin		
Number of trials	216	216
Time in minutes	97	125
Tournament		
Number of trials	93	93
Time in minutes	39	49
Modified Simplex		
Number of trials	18	15
Time in minutes	7	7

*All data are for three replications per paired comparison.

2. Simulating background-noise reduction—The computer system has also been used to simulate an experimental hearing aid that would be exceedingly difficult, if not impossible, to construct in analog form. The design goal is a hearing aid that will automatically reduce background noise. Background noise is particularly damaging to speech intelligibility for the hearing impaired, and systems for reducing background noise are thus of great interest.

The method of noise reduction that has been implemented is that developed by Weiss and Aschkenasy (11). Their method was chosen because of its potential power and also because it has achieved some practical success; it is currently used for specialized applications in which normal-hearing listeners are required to monitor speech-in-noise for long periods of time.

The input and output stages of the system are essentially the same as those shown in Figure 2, except that the system is monaural and that a triangular weighting function is applied to each input buffer. The central part of the system is shown in expanded form in Figure 4: after the first fast Fourier transform (FFT), the phase information is stored for later use while the square root of the amplitude spectrum is converted back to the time domain by an inverse FFT.

An estimate of the noise component in this cepstrum-like signal is obtained by time-weighted averaging over successive windows. This transformed noise spectrum is subtracted from the transformed speech-plus-noise spectrum. The result is an enhanced representation of the signal in which the noise content is greatly reduced. By inverse transformation, an enhanced amplitude spectrum is obtained which, with the addition of phase information, is converted back to conventional time-dependent signals. The output of the final FFT is led to the output buffers, as shown in Figure 2.

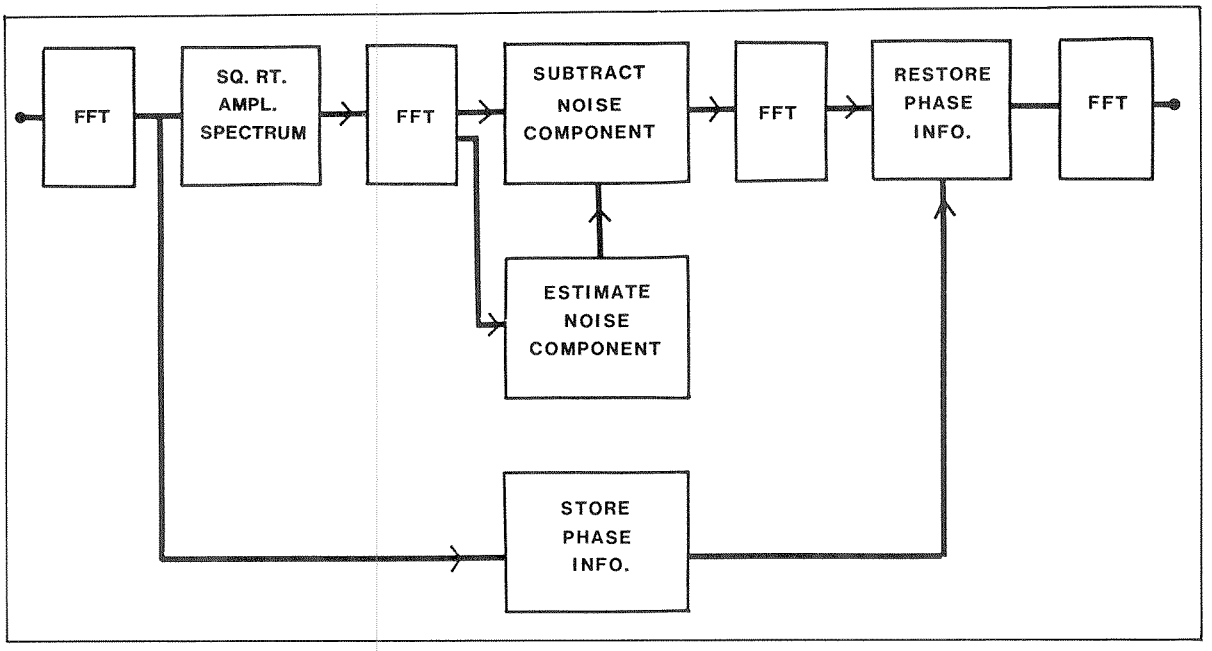

FIGURE 4
Block diagram of noise reduction hearing aid.

As before, the overlap-add technique is used to reduce transients.

A pilot study has been performed to evaluate the effectiveness of the noise-reduction process for normal-hearing listeners, in preparation for a study with hearing-impaired listeners. Ten normal-hearing listeners served as subjects. Hearing levels were within 15 dB of normal audiometric zero for all subjects. The subjects ranged in age from 16 to 35 years.

Ten subsets of the City University of New York Nonsense Syllable Test (NST), as described in 1978 by Levitt and Resnick (12), were chosen as the test stimulus for measurement of speech recognition. The NST consists of nonsense syllables (VC or CV construction) recorded by a male talker in the carrier phrase "You will mark —, please." Response foils include nonsense syllables which differ in manner and/or place of articulation, but not in voicing. The subject marks on his answer sheet which of the seven-to-nine possible syllables has been presented. The 10 subsets include voiced and voiceless consonants in final position in combination with the vowels / i /, / a /, or / u /, and voiced consonants in initial position with the vowel / a /. The nonsense syllable test is well suited to a repeated-measures experiment; the overall score on the test is highly reliable, as is the pattern of errors made by a given subject; see Dubno and Dirks, 1982, (13), and Dubno, Dirks, and Langhofer, 1982 (14).

Subjects were first familiarized with the format of the NST. Performance on the NST was then evaluated under three experimental conditions (i) the unprocessed speech-in-noise (+5 dB S/N), (ii) the processed signal, and (iii) unprocessed speech-in-noise with +17 dB S/N at the input. The latter was included as a reference condition, since the noise-reduction process increases the signal-to-noise ratio by approximately 12 dB. All testing was done at the subject's most-comfortable listening level (MCL). Presentation of the three experimental conditions was randomized.

Syllable-recognition scores on the nonsense-syllable test were calculated in proportions, converted to arcsine units, and a one-way analysis of variance was performed to determine the effect of experimental condition. The results of the analysis revealed that the factor of condition was highly significant [$F(2,18) = 178.6$, $p<.001$]. The Scheffe multiple-range test (used to test for differences between the three mean scores) revealed that only one of the means differed significantly from the others, that of the +17 dB S/N condition ($p<.01$). The mean scores for the +5 dB S/N unprocessed and for the processed conditions did not differ significantly. The failure of processing to improve syllable-recognition scores despite measurable improvements in signal-to-noise ratio, is in agreement with the findings of other investigators who have used digital noise reduction techniques with different speech materials and different types of noise: i.e., Lim and Oppenheim in 1979 (15).

While the overall scores on the nonsense syllable test did not change significantly as a function of processing, the pattern of errors did change and, in fact, showed certain improvements in perception after processing. For instance, place/manner confusions decreased with processing for the subset containing voiced final consonants following / u /. For voiced final consonants in combination with the vowel / i /, place and manner errors both decreased with processing. These improvements in the transmission of certain phonetic features are encouraging, since this may enhance the intelligibility of speech in context. Continued work is planned to evaluate the effectiveness of this type of processing for hearing-impaired listeners.

DISCUSSION

The preceding two examples show the power of computer simulation in two very different applications. The first example was clinically oriented and showed how computer simulation could be used to facilitate the prescriptive fitting of conventional hearing aids. The advantages of the computer system in this application are twofold. Because the hearing aids are simulated, it is not necessary to design and construct an experimental master hearing aid system. Further, the range and versatility of the computerized system far exceeds that which could be obtained with conventional master hearing aids. Secondly, the system is computer controlled, greatly simplifying the task of adjusting the master hearing aid. As a result, extremely efficient adaptive paired-comparison strategies can be used to converge on the estimated optimum electroacoustic characteristics.

In the example given, only the frequency-gain characteristic was adjusted, this being the thrust of much recent research. Using the method of computer simulation, it is a relatively simple matter to manipulate other electroacoustic characteristics such as frequency-dependent amplitude limiting, compression amplification, and other forms of acoustic processing.

The second illustrative example shows the level of sophistication that can be reached in real-time computer simulation of experimental hearing aids. In this case, the design and construction of a prototype noise-reduction unit would be a formidable task, and further changes in the design of the prototype which might follow from the results of the experimental evaluation would be likely to involve substantial additional effort in design and construction.

The programming of a realtime speech-processing system, although considerably less difficult than building hardware prototypes, is not a simple matter. By analogy, economy of programming effort can be achieved by developing the software for non-realtime simulations. This type of programming is not difficult, particularly if a high-level language geared to simulation of speech processing systems is available. If the non-realtime simulation appears promising, the decision may then be made to invest the additional effort needed to get the system to run in real time.

Although the experimental results obtained with the computer simulation of noise suppression were mixed (i.e., subjects preferred the processed signals but objective measures of intelligibility were lower), there is good reason to continue research in the area. Almost all of the previous research on noise suppression has been geared to the needs of normal-hearing listeners, according to Lim and Oppenheim, 1979 (15). Since the detrimental effect of noise on speech is greater for the hearing impaired than for normal-hearing listeners, any noise reduction that can be achieved is likely to be of greater benefit to the hearing impaired ■

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