Hearing Aid Device Development

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Project Staff:
Mr. Andrew Brughera, Ms. Lorraine A. Delhorne, Dr. Joseph G. Desloge, Dr. Julie E. Greenberg, Dr. Patrick M. Zurek

The overall objective of work under this contract has been to evaluate promising signal processing algorithms for hearing aids under realistic conditions. Recent work has aimed at: 1) evaluating the behavioral and acoustic performance of several feedback-cancellation algorithms; 2) assessing the improvements to speech reception in the presence of interfering noise provided by processing signals from an array of four microphones worn on a headband; and 3) developing a predictive model of cancellation performance achievable with two ear-level microphones.

1. LABORATORY EVALUATION OF FEEDBACK REDUCTION ALGORITHMS

Three adaptive feedback reduction algorithms plus a reference algorithm were implemented in a laboratory-based digital hearing aid system and evaluated with dynamic feedback paths and hearing-impaired subjects. The evaluation included measurements of maximum stable gain and subjective quality ratings.

The reference condition (**REF**) consisted of three common elements that were also used in all of the experimental feedback reduction algorithms. These three common elements are: 1) a frequency-dependent amplification appropriate for each subject's hearing loss; 2) an adjustable broadband gain controlled by the experimenter; and 3) compression limiting. Compression limiting protects the subject from uncomfortably loud sounds by restricting the maximum level of the hearing aid output.

The Continuous, No Noise (**CNN**) algorithm attempts to estimate the acoustic feedback path by continuously adapting on the ambient signals.

The Open loop, Noise when Oscillation detected (**ONO**) algorithm employs a detector to determine when strong oscillation occurs. When this condition is met, the normal hearing aid forward path is broken and a broadband probe noise is presented to the hearing aid receiver for a predetermined interval. The adaptive filter that estimates the feedback path is only updated during the interval when the probe noise is active.

The Open loop, Noise when Quiet or oscillation detected (**ONQ**) algorithm is similar to the **ONO** algorithm, but in addition to the oscillation detector, it also employs a quiet-interval detector. The idea is that it may be possible to prevent oscillation before it occurs by identifying quiet intervals due to brief pauses in the desired speech signal and then injecting a low-level probe noise and updating the adaptive filter weights during those intervals.

The hearing aid system used in this study consisted of microphones, preamplifier circuits, digital signal processing boards installed in a personal computer, and hearing aid receivers. For each ear, two Knowles EK-3024 omnidirectional microphones were mounted in a behind-the-ear (BTE) hearing aid shell with 12 mm spacing. The microphone signals were preamplified by 36 dB and then presented to the analog inputs of two DSP-96 boards (Ariel Corporation). Each of these boards contains 16-bit stereo A/D and D/A and a Motorola 96002 processor. The boards were programmed to sample the omnidirectional microphone signals at 16 kHz, digitally combine the two microphone signals from each ear to simulate a simple directional microphone, and implement the feedback reduction algorithms described above. The analog output signal was presented to the subject via a Knowles receiver (either ED-1932 or CI-2748, depending on the subject's hearing loss) embedded in a custom full shell in-the-ear (ITE) module with a 1.5 mm diameter vent.

Measurements of both the maximum stable gain provided by the algorithm under test and the listener's ratings of pleasantness and intelligibility were made at various gain settings. Seven subjects participated in this study. All subjects had bilaterally-symmetric sensorineural hearing loss and at least four years experience wearing hearing aids binaurally.

Overall, for the subjects and ear/vent combinations considered in this study, the **CNN** algorithm provided the largest increases in maximum stable gain relative to the reference algorithm. For subjects with moderately-severe losses below 1 kHz, the **CNN** algorithm provided almost 13 dB of added stable gain, while the **ONO** and **ONQ** algorithms provided 7-8 dB of added gain. Subjects with milder losses below 1 kHz received an average gain increase of 3 dB from the **CNN** algorithm, and no significant benefit from the **ONO** and **ONQ** algorithms.

Analysis of variance results indicate that relative to the reference algorithm, the **CNN** and **ONO** algorithms had no significant effect on subjective intelligibility and pleasantness ratings. Only the **ONQ** algorithm had a small, but significant, detrimental effect on pleasantness at 6 dB below maximum stable gain. In addition, the lack of an algorithm effect indicated that overall the subjects found all algorithms equally pleasant and equally intelligible: 1) at their preferred gain setting, and 2) at various levels relative to the maximum stable gain for that algorithm. In the case of the intelligibility ratings, the lack of an algorithm effect at gain levels relative to the maximum stable gain for that algorithm is presumably because the subjects generally did not need additional gain for the stimuli (60 dB SPL speech in quiet) presented in this study. However, it is likely that the additional gain will improve intelligibility in other listening situations.

Based on the results of this study, the **CNN** algorithm is the logical choice for feedback reduction in hearing aids. The **CNN** algorithm provides the highest maximum stable gain values, and informal subject comments about the objectionable nature of probe noise support its choice over the **ONO** and **ONQ** algorithms. Depending on the particular fitting, the **CNN** algorithm provides hearing aid users with moderately-severe losses at low frequencies additional stable gain margins of 5--25 dB, with no effect on speech quality. Subjects with mild-to-moderate losses at low frequencies generally obtained additional gain margins of a few dB, with no effect on speech quality. Regardless of the severity of their hearing loss, hearing-aid users will also benefit from more comfortable fittings due to looser earmolds and larger vents. Additional benefits may be provided by the **CNN** algorithm for situations not considered in this study. For instance, subjects with mild-to-moderate losses may obtain larger increases in stable gain in situations causing stronger feedback paths, such as the placement of an object near the ear, which were not considered in this study. Furthermore, although the **CNN** algorithm did not demonstrate any effect on subjective intelligibility and pleasantness for the comparisons and listening situation considered in this study, it is expected that the ability to provide additional gain without "ringing" will enhance pleasantness and intelligibility in some real-world listening situations.

2. LABORATORY EVALUATION OF ADAPTIVE-BEAMFORMING ALGORITHMS FOR HEADBAND HEARING AIDS

Several array processing algorithms were studied with an array of four directional microphones mounted in a broadside orientation across a headband in the "On The Head'" (OTH) configuration. The evaluation was performed to assess the relative benefits of four experimental algorithms plus a reference condition for speech reception by experienced hearing aid users. Performance metrics included both speech reception thresholds and subject preferences measured under a variety of noise conditions.

Five algorithms were implemented with the OTH array:

REF - The reference condition in which left and right signals are taken from the nearby outermost microphones on the headband;
SM4 - A fixed-processing algorithm in which the four microphone signals are summed;
GJI - An adaptive algorithm based on the modified Griffiths-Jim beamformer (Greenberg and Zurek, 1992). The primary signal consists of the sum of the four microphone signals, while the reference signal is the difference between the two innermost microphone signals. This system uses only one adaptive filter;

GJ4 - An adaptive algorithm based on the modified Griffiths-Jim beamformer (Greenberg and Zurek, 1992). The primary signal consists of the sum of the four microphone signals, while three reference signals are formed from unique pairwise difference between the microphone signals. This system uses three adaptive filters;
LNS - An adaptive algorithm that first estimates the locations of interference sources and then steers frequency-dependent nulls in those directions (Desloge, 1998).

The hearing aid system used in this study consisted of microphones, preamplifier circuits, digital signal processing boards installed in a personal computer, and hearing aid receivers. Subjects wore the OTH array, a headband holding four forward-pointing cardioid microphones (Knowles EL-3085) with 6 cm spacing. The microphone signals were pre-amplified by 36 dB and then presented to the analog inputs of two DSP-96 boards (Ariel Corporation). Each of these boards contains 16-bit stereo A/D and D/A (programmed to operate at 16 kHz) and a Motorola 96002 processor. The boards were programmed to perform the array processing algorithms described above, and then apply frequency-dependent amplification appropriate for each subject's hearing loss. The analog output signal was presented to the subject via a Knowles receiver (either ED-1932 or CI-2748, depending on the subject's hearing loss) embedded in a custom full shell in-the-ear (ITE) module with a 1.5 mm diameter vent. A vent plug was used to reduce the vent to 0.5 mm when possible. However, for some subjects, the desired gains could only be provided without feedback when a closed plug was used to completely occlude the vent.

The experiments were performed in a double-wall soundproof room with internal dimensions of 2.7 x 2.5 x 2.0 m. The subject, wearing the microphones and receivers, sat in the center of the booth, with cables connecting the transducers to a preamplifier. Test stimuli were delivered from an array of seven loudspeakers (Radio Shack Optimus Pro 7) placed at a distance of 1.0 meters and a height of 1.1 meters in the directions 0, +/-45, +/-90, +/-135 degrees. The experimenter, the computer containing the DSP boards, and the test equipment were located outside the booth. The subject used a hand-held terminal and an intercom to communicate with the experimenter and to record responses.

The desired (target) speech signal was delivered from straight-ahead of the listener. the 0 degree loudspeaker. Interfering (jammer) sources were delivered from one or more off-axis loudspeakers, depending on the noise condition:

The algorithms were tested for speech intelligibility and ease of listening under the various noise conditions. Nine subjects participated in the speech intelligibility study, and twelve subjects (the same nine plus three more) participated in the ease of listening study. All subjects had bilaterally-symmetric sensorineural hearing loss and at least several years experience wearing hearing aids binaurally.

In general, the results of both the speech intelligibility and ease of listening experiments showed that in the relatively non-reverberant acoustic environment used in this study, all three adaptive processing algorithms provided significant benefits over either the reference condition or a fixed array processor. Although one might expect the **GJ4** algorithm (which uses three adaptive filters) to outperform the **GJI** algorithm (which uses only one adaptive filter) in the presence of multiple jammers (NS3 and NS4), in fact the **GJ4** algorithm performed worse in both experiments. The **GJ4** algorithm is more susceptible to target cancellation resulting from inequalities in the transfer functions between the straight ahead target source and the microphones. These inequalities can be caused by misalignment of the array to the straight ahead target and/or variations in the microphones themselves. The difference between the **GJ4** and **GJI** algorithms can be attributed in part to the fact that the modifications to the Griffiths-Jim beamformer were developed to reduce target cancellation specifically for a system with two microphones and one adaptive filter, and are apparently not as effective when more microphones and adaptive filters are used. In terms of objective speech intelligibility measures, the **LNS** algorithm is superior to **GJI**. However, it appears that subjects tend to prefer **GJI** over **LNS**. This may be because of artifacts introduced by the **LNS** processing which do not interfere with intelligibility, but which subjects nevertheless judge to be detrimental to speech quality.

3. ANALYSIS OF ADAPTIVE ARRAY CANCELLATION PERFORMANCE

Over the past fifteen years, several research groups have explored the use of microphone arrays for noise reduction in hearing aids. The most powerful of these algorithms reduce noise by adaptively searching for filter parameters that result in a cancellation of noise components when the filtered microphone signals are summed. Adaptive systems have been shown to reduce noise, without affecting the desired signal from the straight ahead direction, by tens of decibels in favorable conditions, much more than can be achieved by fixed-processing systems.

The principal limitations of adaptive cancellation systems come from the need for the noise sources to be both limited in number and very directional. The limitation on number comes from the fact that a microphone array system with M microphones can control M-1 independent nulls while preserving the target. If there are M or more noise sources, then noise-reduction performance degrades. It is often argued that this limitation is not severe in practice because, usually, one sound source will be dominant in any frequency band.

A more important limitation on adaptive cancellation performance comes from the reverberation in the environment. Past work has shown empirically that the performance of adaptive cancellation systems degrade as the degree of reverberation increases (Greenberg and Zurek, 1992). While in principle the solution for this problem is to increase the temporal length of the adaptive filter, in practice there will be limits (from implementation imperfections and from non-stationarity of the environment) on effective filter length.

A basic concern in the application of adaptive systems to hearing aids, in light of these limitations, has been whether the benefits from adaptive cancellation will be large and frequent enough in everyday environments to justify the complexity and expense of the system. In order to address this concern it is necessary to have a better understanding of the factors that control adaptive cancellation performance in reverberant environments. This project has been an attempt to provide that understanding with a simple model in which system parameters and environmental acoustic parameters are taken into account in deriving predictions of cancellation performance.

The noise cancellation algorithm used in this work is the one we have used in the past, the Griffiths-Jim adaptive beamformer (Greenberg and Zurek, 1992). A prediction for the degree of cancellation achieved by this system can be readily determined. Using a frequency domain analysis, the output noise spectrum and degree of cancellation can be expressed, as a function of frequency, by a relatively simple closed-form expression involving the two microphones' power spectra and their cross-spectrum.

The predictions of this theory have been tested against an actual (time-domain) adaptive beamformer operating in non-real-time with simulated noise environments. The quality of these predictions has been excellent in all cases for which the direct-to-reverberant ratio is very high (i.e., very low reverberation). When reverberation is appreciable the actual obtained cancellation is always *better* than predicted. The source of this unusual discrepancy has been shown to be a bias in the estimation of cross-spectra between signals having substantial delay between them. When the inter-signal delay is a sizeable fraction of the analysis window, the true cross-spectrum (or coherence) is underestimated; this leads directly to the underestimate of noise cancellation. In short, cross-spectral coherence is not a good measure of the degree of potential cancellation that can be achieved between two signals when there is substantial delay between them.

Since our aim is to relate parameters of the acoustic environment, the array configuration, and the processing system to cancellation performance, the theory just described, which expresses cancellation performance in terms of input signal characteristics, is not sufficient. What we are developing is a model that merges that prediction for the direct component with a prediction for cancellation of the reverberant component as characterized by room-acoustics parameters (direct-to-reverberant ratio, D/R, and reverberation time, T_{60}). This model is based on the conceptual notion that the adaptive filter is capable of canceling all of the direct component plus any part of the reverberant impulse response that falls within its temporal scope. It is assumed that all of the direct component is cancelled just as it would be with no reverberation present, and that the portion of the idealized parametric reverberant response contained within the length of the adaptive filter is also cancelled. This leads to predicted cancellation of the direct component that exhibits the expected frequency-dependent variations, but no such frequency-dependence for the cancellation of the reverberant component.

This model is currently being tested with simulated noise environments. It is difficult to test it with recordings made in real environments because of the difficulty in measuring D/R. For the purposes for which it is to be used, however, the level of verification should be adequate. Assuming the tests are successful, the model will be used to generate predictions for a variety of everyday environments based on estimated room-acoustics parameters.

Having a model available to predict the performance of a cancellation system will help the hearing-aid research and development community understand the dependence of performance on acoustic-environment parameters. This increased understanding will encourage the development of microphone-array hearing aids and assistive devices for applications and circumstances where they will be most beneficial.

REFERENCES

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