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TWO-MICROPHONE ADAPTIVE ARRAY HEARING AIDS WITH MONAURAL AND BINAURAL OUTPUTS

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ABSTRACT

Traditional two-microphone array processors applied to hearing aids require binaural inputs and produce a monaural output with improved target-to-jammer ratio (TJR), but no binaural cues. Previous work with normal-hearing listeners has demonstrated the potential to obtain a tradeoff between localization and intelligibility with a *lowpass/highpass* system that splits the signal into two frequency bands, preserving binaural cues at low frequencies while improving TJR at high frequencies. In this work, the lowpass/highpass system and a conventional broadband array processor were evaluated with hearing-impaired subjects. The broadband array processor improved speech reception thresholds (SRTs) by an average of 10 dB relative to a binaural reference condition, while the lowpass/highpass method provided an average SRT improvement of 2 dB. Physical measurements show 20 dB of jammer attenuation over much of the frequency range where array processing is applied. These results suggest that hearing-impaired subjects rely on low frequency information more than normal-hearing listeners, and that effective array processing hearing aids must operate over the entire frequency range.

1. INTRODUCTION

Hearing-aid users commonly complain about the difficulty of understanding target speech in the presence of background noise, or jammers. Array processors using inputs from multiple microphones provide a promising approach for improving the target-to-jammer ratio (TJR) when the jammer sources arise from different directions than the target source. Adaptive array processors can potentially provide larger benefits than fixed processing methods because they utilize time-varying weights that respond to changes in the acoustic environment. Much effort has been given to the development of adaptive algorithms for microphone-array hearing aids [1], [2], [3], [4]. Previous work in our group [5], [6] has used physical measurements to demonstrate the potential of a modified generalized sidelobe canceller [7] (Fig. 1) for a number of microphone configurations in a variety of acoustic environments.

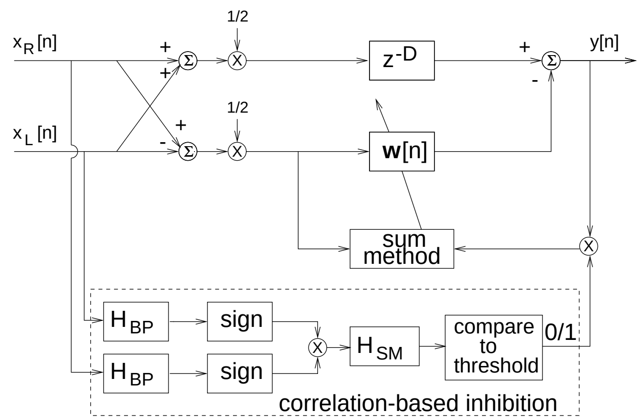

Figure 1: Block diagram of the modified generalized sidelobe canceller.

One drawback of traditional array processing methods is that they produce a single (monaural) output. Since binaural signals are well-known to provide directional information and a sense of auditory space, the application of traditional array processors to hearing aids results in a device that improves the intelligibility of the target speech while destroying a sense of auditory space. Depending on the listening environment,

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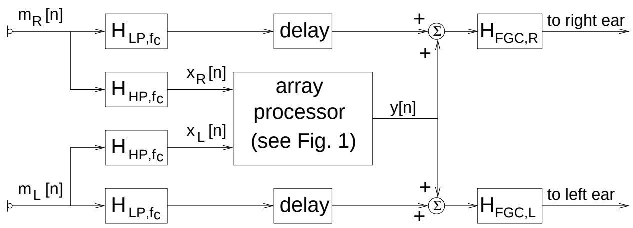

Figure 2: Block diagram of the two-microphone lowpass/highpass system for array processing with binaural outputs.

this may be unimportant, annoying, or even dangerous, as in the case of warning signals and alarms. This concern has motivated investigation of microphone-array hearing aids with binaural outputs [8], [9], [10], [11].

Welker et al. [9] proposed and evaluated a two-microphone adaptive array with binaural outputs based on the *lowpass/highpass* approach [8]. The goal is to provide a binaural output which improves intelligibility by increasing the TJR at high frequencies while providing a sense of auditory space by preserving binaural cues at low frequencies. This system (Fig. 2) splits each microphone signal into two frequency bands. The high frequency portion of the signal is the input to the adaptive array processor, which produces a monaural output. The low frequency portion of the signal is delayed and summed with the high-frequency result. This approach is motivated by the observations that the benefits of binaural listening (sound localization, improved speech reception and signal detection) are derived primarily from the low-frequency portion of the spectrum.

Tests with normal hearing listeners have shown that this system can improve intelligibility while maintaining the ability to localize (Fig. 3). The cutoff frequency between the two frequency bands (f_c) controls the tradeoff between binaural cues and jammer reduction. Lower cutoff frequencies result in jammer reduction over a wider frequency range, leading to higher intelligibility, at the expense of reduced binaural cues and poorer localization performance. Higher cutoff frequencies result in preserving binaural cues over a wider frequency range, leading to better localization performance, at the expense of reduced jammer reduction and lower intelligibility.

These encouraging initial results motivate the current study, which evaluated speech reception in noise by hearing-impaired listeners using a two-microphone adaptive array hearing aid based on the modified generalized sidelobe canceller. In particular, speech intelligibility benefits were measured with two system configurations: 1) array processing applied to the entire spectrum, and 2) array processing with binaural cues based on the lowpass/highpass approach.

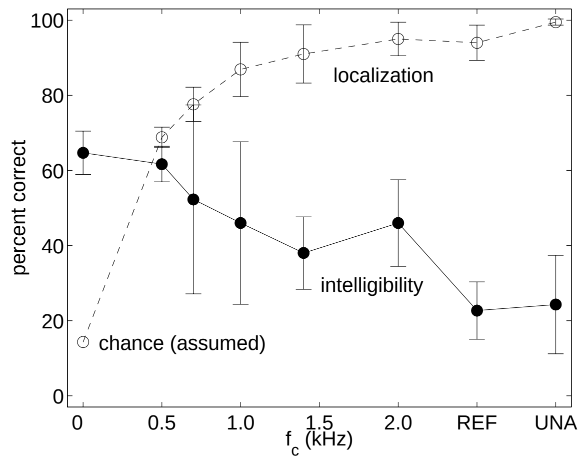

Figure 3: Tradeoff between intelligibility due to jammer reduction and localization due to preservation of binaural cues, as a function of cutoff frequency for normal hearing listeners. UNA denotes unaided listening. (Adapted from [9], ©1997, IEEE.)

2. METHODS

2.1. Hearing aid system

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 preamplified microphone signals were 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 algorithms described below. The analog output signals were presented to each ear 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.

2.2. Test environment

The experiments were performed in a double-wall soundproof room with internal dimensions of 2.7 x 2.5 x 2.0 m. The walls and ceiling were perforated metal panels with acoustically-absorptive foam backing. The floor was a solid metal panel covered with a carpet. The space is relatively nonreverberant, but not anechoic. The subject, wearing the microphones and receivers, sat in the center of the booth. Test stimuli were delivered from two Radio Shack Optimus Pro 7 loudspeakers placed at a distance of 1.0 meters and a height of 1.1 meters straight ahead of the subject and an azimuthal angle of 60° to the right. 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.

2.3. Hearing aid algorithms

This study considered three algorithms: a reference condition, which consisted of presenting the directional microphone signals binaurally (REF; $f_c = 8$ kHz), an array processor with a monaural output presented diotically (APM; $f_c = 0$), and an array processor with binaural outputs (APB; $f_c = 1$ kHz). In practice, the lowpass and highpass filters were only used for APB. They were implemented with 45-point FIR filters designed by the Parks-McClellan method.

For all three algorithms, the array processor was a modified generalized sidelobe canceller (Fig. 1) which used the LMS algorithm [12] to update the weights of the adaptive filter according to

$$\mathbf{w}[n+1] = \mathbf{w}[n] + \mu[n]y[n]\mathbf{x}[n], \qquad (1)$$

where $\mathbf{w}[n]$ is the L-element vector of adaptive filter weights, $\mu[n]$ is the time-varying adaptive step-size, $y[n]$ is the array processor output, and $\mathbf{x}[n]$ is the L-point vector of data in the adaptive filter's tapped delay line at time n. The adaptive step size was updated using the sum method [13]

$$\mu[n] = \frac{\alpha}{L(P_x[n] + P_y[n])} \qquad (2)$$

where α is a dimensionless constant and $P_x[n]$ and $P_y[n]$ are running estimates of the powers in $x[n]$ and $y[n]$, computed by smoothing the instantaneous power values with a first-order IIR filter with a time constant of 18 ms to equal the length of the adaptive filter. The parameter values were $L = 285$, $\alpha = 0.01$, and $D = 40$.

The modified generalized sidelobe canceller includes correlation-based inhibition to halt adaptation during intervals when the TJR is high. The bandpass filters pass frequencies between 500 and 1000 Hz, selected to minimize the correlation of off-axis sources for the array geometry. A running correlation measure is computed by smoothing the product of the signs of the bandpass signals. The correlation smoothing filter is a first-order IIR filter with 10-ms time constant. This correlation estimate is then compared to a threshold of zero. When the estimate is below threshold, the output signal is used to update the adaptive weights, otherwise it is replaced with zero to prevent adaptation.

The binaural (or diotic) signals produced by the algorithms were processed by frequency gain characteristics fitted to each ear's hearing loss ($H_{\mathrm{FGC,L}}$ and $H_{\mathrm{FGC,R}}$ in Fig. 2), implemented with 127-point linear-phase FIR filters. These filters were designed to provide a compromise between matching the insertion gains prescribed by NAL2 [14] and matching the insertion gains provided by the subject's personal hearing aids. The subject receives the strongest acceptable binaural frequency-gain characteristics with maximum gains at each frequency limited by the insertion gains of either their personal hearing aids or the NAL2 target.

Finally, the signals were limited to prevent the presentation of uncomfortably loud sounds. Simple digital clipping was applied to the output signal whenever the computed output value exceeded a threshold based on measurements of the subject's uncomfortable loudness levels [15].

2.4. Experimental Procedures

The target speech signal consisted of IEEE Harvard sentences [16] read by a male talker, delivered from the straight ahead (0°) loudspeaker at 65 dBSPL. The jammer source consisted of speech-shaped noise delivered from the 60° loudspeaker. The jammer source was activated one second before the onset of each target sentence, so the adaptive algorithms could at least partially converge. Speech reception thresholds (SRTs) were measured using an adaptive procedure that varied the jammer level to determine the TJR corresponding to 50% intelligibility, measured in terms of percent correct of keywords. For each algorithm, the SRT was measured at least twice and the results were averaged.

2.5. Subjects

Eight hearing-impaired subjects participated in this study. All subjects had bilaterally-symmetric sensorineural hearing loss[1] and at least four years experience wearing hearing aids binaurally. There were six males and two females, ranging from 26 to 77 years of age. One subject had a rising loss from moderately-severe to mild (GI); three subjects had flat losses that were moderately-severe (PG, AP) and severe (HK); four subjects had sloping losses from mild to moderately-severe (TM), mild to severe/profound (HB, WF), and severe to profound (JH).

[1] GI's right ear had a mild conductive component below 1 kHz and showed reduced compliance.

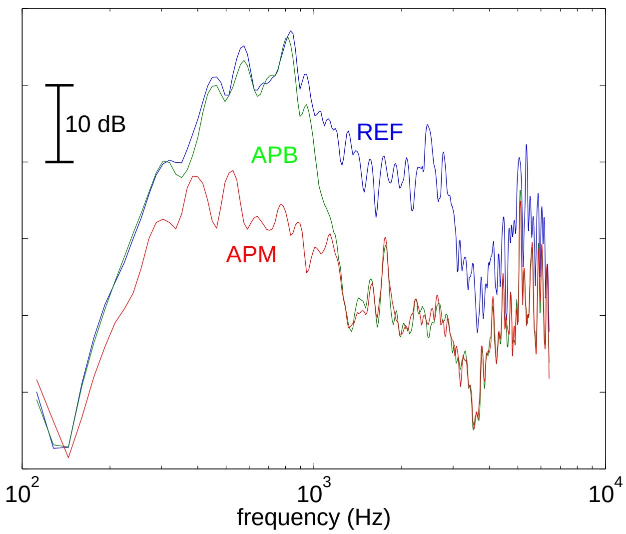

Figure 4: Jammer spectrum at the left ear for each of three algorithms described in the text.

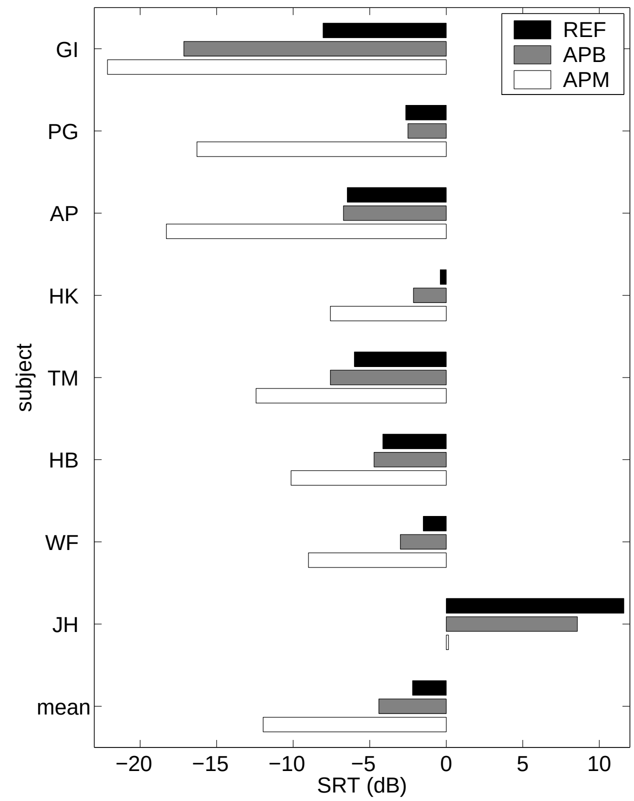

Figure 5: Average speech reception thresholds for individual subjects listening via each of three algorithms described in the text. Lower values (to the left) indicate better performance.

3. RESULTS

Figure 4 shows jammer spectra at the left ear (the side shadowed from the jammer source) for each algorithm in the case where no frequency gain characteristic is applied ($H_{\mathrm{FGC,L/R}}$ are unity). For the acoustic environment used in this study, array processing provides roughly 20 dB of jammer reduction over a broad range of frequencies. Figure 5 shows the average SRT for each subject with the three algorithms. Since the SRT reflects the TJR at which the target material was 50% intelligible, lower SRT values (to the left) indicate better performance. Averaged over subjects, the lowpass/highpass system (APB) improved SRTs by 2 dB relative to the reference condition (REF), while the broadband array processor (APM) improved SRTs by 10 dB relative to the reference condition. (Other results, not presented here, verified that both REF and APB preserved subjects' ability to localize sounds, as expected.)

4. DISCUSSION

The limited benefit provided by the lowpass/highpass system (APB) is surprising in light of previous results with normal hearing subjects [9]. This is most likely due to limitations on the ability of hearing-impaired subjects to extract useful information from audible speech cues in the high frequency region [17], [18]. These recently reported results contradict the conventional belief that increasing the audibility of target speech in any frequency region always leads to improved intelligibility. Future work could consider the possibility that hearing-impaired listeners might benefit from a lowpass/highpass system with a lower cutoff frequency (e.g. $f_c = 500$ Hz). However, the current results highlight the practical difficulty of preserving a sense of auditory space while improving intelligibility of speech in noise for hearing aid users, and support the use of a user-controlled switch to select between omnidirectional and directional listening modes.

The 10 dB improvement in SRT provided by the broadband array processor (APM) is plausible in light of the physical measurements and our understanding of the factors affecting speech intelligibility. Figure 4 shows that the jammer spectrum is reduced by 5 dB below 300 Hz, roughly 10 dB at 400 Hz, and 20 dB above 500 Hz. Future work will use the speech intelligibility index [19] to determine whether the current SRT results can be explained by accounting for each subjects' hearing thresholds and the relative contribution of different frequency bands to overall intelligibility.

The broadband array processor (APM) provides a substantial benefit for the favorable (relatively non-reverberant) acoustic environment used in this study. Furthermore, this represents a conservative measure of benefit, since other results (not reported here) revealed that the reference condition of binaural directional microphones improved SRTs by an average of 4 dB relative to binaural omnidirectional microphones under the current conditions. In other words, the actual benefit of broadband array processing in this environment is roughly 14 dB, relative to conventional hearing aids with omnidirectional microphones.

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