ADAPTIVE GAIN PROCESSING TO IMPROVE FEEDBACK CANCELLATION IN DIGITAL HEARING AIDS

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ABSTRACT

Adaptive filters are commonly used to cancel acoustic feedback in hearing aids. The sound quality of hearing aids deteriorates as the hearing aid gain is increased. This paper presents a method to alter the gain function in digital hearing aids to provide additional amplification and better output sound quality. This approach employs a variable, frequency-dependent gain function that is lower at frequencies of the incoming signal where the information is perceptually insignificant. The increase in stable gain over traditional methods and the output sound quality were evaluated with a psychoacoustic experiment on normal-hearing listeners. The results indicate that the method of this paper provides more hearing aid gain and less distortion in the output sound quality than feedback cancelers with fixed gain functions.

Index Terms— Feedback, Hearing aids, Adaptive filters

1. INTRODUCTION

Hearing aids are used to compensate for the reduced audibility of hearing-impaired listeners by amplifying the incoming sounds. The maximum amplification possible in hearing aids is limited because of the acoustical coupling between the speaker and the microphones. State-of-the-art hearing aids use adaptive filters to estimate the acoustic feedback and cancel it in the digital domain. Several schemes for the adaptive feedback cancellation (AFC) have been investigated in the past [1, 2].

A block diagram of a typical digital hearing aid with adaptive feedback cancellation is shown in Figure 1. For simplicity of presentation, we have employed discrete-time signal representations throughout the paper. The input signal $v(n)$ is corrupted by the feedback signal $f(n)$ and is picked up by the microphone. An adaptive filter W is used to estimate the feedback signal using the reference signal $x(n)$ and the primary input signal $d(n)$ to produce the signal $y(n)$. The adaptive filter W estimates the combined response of the speaker, the microphone and the acoustic feedback. Often, the adaptive filter algorithm is implemented in some transform domain. It is desired that the signal $y(n)$ is close to the signal $f(n)$. The gain function G of the hearing aid is usually frequency dependent. The delay D is provided to reduce the bias in the estimate of the feedback signal [2, 1]. The output limiter (OL) limits the amplitude of the output signal $x(n)$ which is used here in place of a output compression limiter (OCL) for simplicity. The OCL reduces the gain of the system to mitigate clipping [2].

The adaptive feedback cancellation improves the output sound quality and provides an additional gain over the critical gain[1] for which the hearing aid is stable [2, 1]. The additional gain made possible by feedback cancellation is termed as added stable gain (ASG). When the amplification in a hearing aid is more than the limits of the added stable gain, the hearing aid becomes unstable or the quality of the signal degrades to below acceptable levels [1]. A major source of this loss of performance of the system is the presence of residual feedback components in the signal. If acoustical feedback components are reduced, the stability and the output sound quality of a hearing aid system can be further improved. Many researchers have proposed to change the characteristics of the signal in the forward path by changing its phase, shifting its frequency components or modifying its spectral magnitude with a notch filter to suppress the feedback and hence provide added stable gains [1, 3]. These methods modify the loop transfer function so that negative feedback is maintained in consecutive iterations. Although these methods keep the hearing aid stable, the output sound quality degrades as the signal characteristics are changed.

[1]Critical gain refers to the maximum amplification for which the output signal quality is acceptable without feedback cancellation.

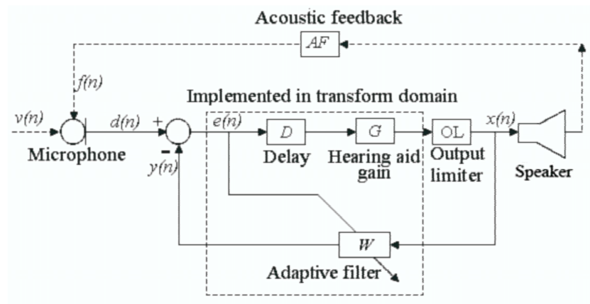

Fig. 1. Simplified block diagram of a digital hearing aid with adaptive feedback cancellation

In this paper, we present an approach that alters the hearing aid gain function to enhance the stability of the hearing aid. Our approach makes use of perceptually redundant components in speech to modify the gain function. Research in psychoacoustics has shown that humans can have difficulty hearing weak signals that fall in the frequency (frequency masking) or time vicinity (temporal masking) of stronger signals [4, 5]. Such components do not contribute to the understanding of speech regardless of whether they are amplified or not. The method presented in the paper also makes use of the fact that speech does not occupy the entire time-frequency space even during the voiced periods [6]. Typically, short(temporal) pauses appear between words and even between syllables. In addition, voiced intervals have spectral gaps between the harmonics of the pitch frequency. Amplifying speech during such temporal and frequency-domain gaps do not enhance the audibility of the signals. Such redundancies have been successfully used in the past in the area of coding and noise suppression [4].

The paper's method reduces the hearing aid gain for redundant speech components in time and frequency and keeps the prescribed gain for other components. Assuming that the models of perceptual redundancy are accurate, the changes in signal characteristics caused by reducing gain associated with those components will not audible to the listener. Furthermore, since this process reduces the feedback components, it makes the speech more intelligible. Finally, this scheme makes the hearing aid system less prone to instability because the average open loop gain is smaller than the gain for perceptually important components, and thus reduces the build up of acoustical signal in the closed loop [5, 1].

2. AN OVERVIEW OF THE HEARING AID MODEL

The algorithm of this paper performs adaptive feedback cancellation in the frequency domain. Among various frequency domain adaptive filter algorithms available we use the frequency-domain block LMS (FBLMS) adaptive filter [2] to cancel the feedback. The notations used in this paper are as follows. Let $x(k,n)$ defined as

$$x(k,n) = x(n + k\Delta K) \quad n = 1, 2, \cdots, L \tag{1}$$

represent the k^{th} frame of the input signal $x(n)$. In (1), L is the frame size in number of samples and ΔK represents the shift between successive frames. Let the discrete fourier transform (DFT) of the signal $x(k,n)$ be $X(k,m)$ where m represents the frequency bin $\omega_m = \frac{2\pi m}{L}$ radians/sample. We denote the vector of all frequency components $X(k,m)$ in frame k by $\mathbf{X}(k)$. The feedback path - the combined response of the speaker, microphone and the acoustic feedback shown in Figure 1 is modeled with an FIR filter containing N coefficients. Furthermore, signals are segmented into $L = 2N$-points vectors with N point overlap. The FBLMS algorithm for the k^{th} block is summarized in Table 1.

In Table 1, the variables $\mathbf{W}(k)$ and $\mathbf{g}(k)$ are the vector representations of the adaptive filter coefficients and the hearing aid gain respectively in the frequency domain. $d = D/N$ is the normalized delay. For ease of implementation we chose D to be an integer multiple of N. In most implementations, the hearing aid gain is fixed and frequency dependent. However, in the experimental comparisons presented later in this paper, the maximum allowed gain was set to be the same for all frequency bins *i. e.* $\mathbf{g}(k) = G$ for the conventional method. In the system of this paper the gain $g(k,m)$ may vary for frequencies in a frame.

3. ADAPTIVE GAIN PROCESSING

The new scheme utilizes the information about masking thresholds and speech presence/excitation. A gain control scheme that results in low artifacts and low distortions in the output signals is discussed later in the section.

3.1. Calculation of Masking Thresholds and Signal Presence

In this paper, we do not consider the contribution of temporal masking because it is usually difficult to quantify [5]. Calculation of the masking thresholds $T(k,m)$ for the k^{th} frame and m^{th} frequency bin involves defining critical bands on the power spectrum $P(k,m)$ of the speech signal. The power spectrum is calculated from the spectrum of the input signal to the speaker before amplification $Q(k-d,m)$ using speech pressure level (SPL) normalization [4]. The power normalization term is fixed at 90.302 dB.

Table 1. Adaptive Feedback Cancellation with FBLMS

Initialization

$\mathbf{S}(0)$	...	A vector with small positive constant
$\mathbf{W}(0)$	...	A vector with all zeros
μ_0	...	Suitable adaptation constant
β	...	An averaging constant close to 1

Iterations

$\mathbf{d}(k) = \left[\; d(kN) \quad d(kN+1) \quad \cdots \quad d((k+2)N-1) \;\right]$

$\mathbf{Y}(k) = \mathbf{W}(k) \bigotimes \mathbf{X}(k)$

$\mathbf{y}(k) =$ the last N elements of IFFT$(\mathbf{Y}(k))$

$\mathbf{e}(k) = \mathbf{d}(k) - \mathbf{y}(k)$

$\mathbf{E}(k) = \text{FFT}\left(\begin{bmatrix} \mathbf{0} \\ \mathbf{e}(k) \end{bmatrix}\right)$

$W(k+1,m) = W(k,m) + \mu_0 X(k,m) E^\star(k,m)/S(k,m)$ for $m = 0, \cdots, 2N-1$

$\mathbf{W}(k{+}1) = \text{FFT}\left(\begin{bmatrix} \text{the first N elements of IFFT } (\mathbf{W}(k+1)) \\ \mathbf{0} \end{bmatrix}\right)$

$Q(k,m) = E(k,m) + (-1)^m E(k-1,m)$ for $m = 0, \cdots, 2N-1$

$\mathbf{X}(k+1) = \mathbf{g}(k)\mathbf{Q}(k-d)$

$\mathbf{S}(k+1) = \beta\mathbf{S}(k) + (1-\beta)\mathbf{X}(k+1) \bigodot \mathbf{X}^\star(k+1)$

- $\bigodot$ denotes element-by-element multiplication
- $\mathbf{0}$ denotes column vector of length N
- $\star$ denotes complex conjugate

Subsequently, tonal and noise maskers are identified in each critical band which are above the hearing threshold [4, 7, 5]. If two or more maskers are close to each other in a critical band, only the strongest masker is kept and others are discarded. Details of masking models and estimation of the maskers can be found in [4, 5]. After identifying the maskers, the masking effects due to these maskers in their frequency bands and their neighboring bands are calculated using the spreading function that was derived from Zwickers data [7] as in [4]. Finally, the global masking threshold is calculated for each frequency bin by combining the individual masking thresholds of all the maskers identified in the previous steps.

Many methods to improve speech processing algorithms using the notion of temporal and spectral gaps in speech signals are available in the literature [6]. The simplest method to find such gaps uses an energy-based detector. Average energies $P_Q(k,m)$ and $P_{Q,min}(k,m)$ at each frequency bin are calculated with a single pole IIR filter

$$P_Q(k,m) = \gamma_1 P_Q(k-1,m) + (1-\gamma_1)|Q(k-d,m)|^2$$
$$P_{Q,min}(k,m) = \gamma_2 P_{Q,min}(k-1,m) + (1-\gamma_2)|Q(k-d,m)|^2$$

respectively. The averaging constants γ_1 and γ_2 are such that $0 < \gamma_1 < \gamma_2 < 1$. Consequently, the average energy $P_Q(k,m)$ is effectively based on fewer samples than $P_{Q,min}(k,m)$. If $P_Q(k,m)$ is sufficiently smaller (with the help of the parameter

δ) than $P_{Q,min}(k,m)$, we consider the frequency bin m to be not excited. Otherwise, the m^{th} frequency bin is assumed to be excited. The algorithm employs a smaller gain at unexcited frequencies than for excited frequency components. In all our work so far we have used $\delta = 0.8$. We use a variable $S_{AVL}(k,m)$ to indicate whether the m^{th} frequency bin of the k^{th} block is excited. For some δ, we define the parameter as

$$S_{AVL}(k,m) = \begin{cases} 0 & \text{if } P_Q(k,m) < \delta P_{Q,min}(k,m) \\ 1 & \text{if } P_Q(k,m) > \delta P_{Q,min}(k,m) \end{cases} \quad (2)$$

3.2. Gain Adjustment with $T(k,m)$ and $S_{AVL}(k,m)$

The adaptive gain processing algorithm reduces the hearing aid gain at frequencies where the instantaneous signal energy ($|Q(k-d,m)|^2$) is below the global masking threshold $T(k,m)$ or when the signal is determined to be unexcited at a frequency bin *i.e.*$S_{AVL}(k,m) = 0$. However, a large reduction in the gain may produce artifacts due to aliasing [2]. Consequently, the algorithm reduces the gain by no more than some preselected fraction η, where $0 < \eta < 1$ from frame to frame. Similarly, we also limit the minimum gain at a frequency to avoid unnatural artifacts in the output.

The algorithm for varying the gain $g(k,m)$ is summarized in Table 2 where G_{min} is the minimum permissible gain at any frequency.

Table 2. Adaptive Gain Processing

$$pr(k,m) = \begin{cases} 0 & \text{if } S_{AVL}(k,m) = 0 \text{ or } |Q(k-d,m)|^2 < T(k,m) \\ 1 & \text{else} \end{cases}$$

$$g(k,m) = pr(k,m)G + (1 - pr(k,m))\eta g(k-1,m)$$

$$g(k,m) = \max[g(k,m), G_{min}]$$

4. RESULTS AND DISCUSSION

This section presents the results from MATLAB simulations and real time implementations of the hearing aid algorithms to demonstrate the performance of the paper's approach and a classical scheme. A subjective evaluation of the output sound of the two methods from real time implementation is also be presented in this section.

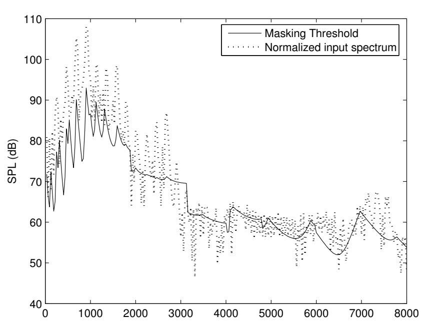

Fig. 2. Masking thresholds at various frequencies

In the first experiment, adaptive feedback cancellation was performed in MATLAB with an FBLMS algorithm with fixed gain and the method of this paper. The feedback path used in simulations was modeled with a 128-tap FIR filter. In addition, a random perturbation was added to the feedback coefficients to simulate real time changes. The random perturbation was such that the mean values of the coefficients do not change over time and the variance of the perturbation was 10^{-4} for each coefficient. The critical gain of the feedback path was approximately 11 dB. A delay of one block ($D = 8$ ms) at 16 KHz was used in the simulations. Other parameters employed were $N = 128$, $\mu_0 = 0.02$, $\beta = 0.99$, $\gamma_1 = 0.995$, $\gamma_2 = 0.85$, $\eta = 0.9$ and $G_{min} = 6$ dB. The input signals to the hearing aid were clean speech waveforms taken from the TIMIT database.

The global masking thresholds $T(k,m)$ for different frequency bins for one signal frame from a MATLAB simulation are shown in Figure 2. The hearing aid gain was 15 dB above the critical gain for this experiment. It can be seen from Figure 2 that there were many components below the masking threshold in this simulation.

To demonstrate the improvement in the feedback cancellation with adaptive gain processing over fixed gain systems, we define signal-to-feedback-ratio (SFR) for frame k as

$$\text{SFR} = 20\log\left(\frac{\sum_{i=(k-1)L+1}^{kL} v^2(i)}{\sum_{i=(k-1)L+1}^{kL} (e(i) - v(i))^2}\right) \quad (3)$$

In our case the clean speech is mostly contaminated by feedback, therefore, we can assume that higher values of SFR indicate more feedback cancellation. The average SFR for all frames after the adaptive filter converged was calculated for both methods at different gains. The results are tabulated in Table 3. As one would expect, the SFR values decrease for both schemes with increasing gain values. However, the adaptive gain processing system exhibited $1-2$ dB improvements in performance over fixed gain system. This is due to intermittent gain reductions done by the adaptive gain processing algorithm at redundant components of the input signal, which in turn reduces the feedback coupling.

Table 3. Signal-to-feedback-ratio (SFR) for two methods

Gain Above the CG (dB)	Average SFR in dBs	
	Fixed Gain	Adaptive Gain
2	12.33	14.41
6	10.06	11.87
10	7.32	8.73
15	2.42	3.43

The power spectra of the output produced with the two methods are shown in Figure 3. The spectra were estimated using the Welch method by dividing data into frames of 512 samples with 256 sample overlap. There are noticeable differences between the spectra of the output of the fixed gain system and the input speech at higher frequencies. The spectrum of the output of the adaptive gain processing scheme is closer to the input signal's spectrum especially at higher frequencies where the conventional method did not perform well.

In the next experiment, both algorithms were evaluated in real-time. A prototype inside-the-ear (ITE) hearing aid that can fit into the ear piece of the Knowles Electronics Manikin for Auditory Research (KEMAR) was used in the experiment. We used a standard EXPRESSfit hearing aid programming cable to drive and access microphones and the speaker of the hearing aid. The programming cable was connected to an interface board through an 8-pin mini DIN plug that provided the required power to the programming cable and amplified the received signal. The adaptive gain processing system and the feedback cancellation algorithm were implemented using an ADSP-21161N processor. With the above setup, output of the hearing aid system was recorded with a sound card for both schemes.

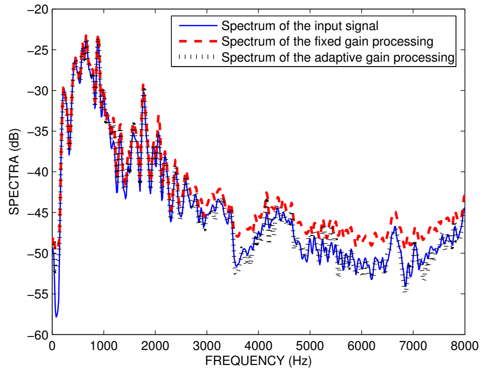

Fig. 3. Comparison of the output spectra of the two methods

The recorded outputs for both algorithms at different gains are shown in Figure 4. Figure 4a shows that the classical scheme became unstable at a gain of 10.6 dB above the critical gain. Figure 4b and 4c show that the adaptive gain processing produces stable output at gains of 10.6 dB and 12.3 dB above the critical gain.

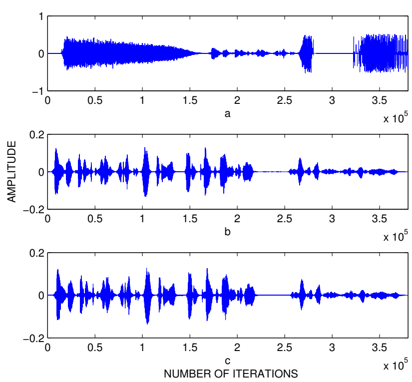

Fig. 4. Scaled output signals in the steady-state: (a) the fixed gain processing at 10.6 dB above the CG (b) the adaptive gain processing 10.6 dB above the CG (c) the adaptive gain processing at 12.3 dB above the CG

We performed an informal subjective evaluation of the recorded data with both schemes. The subjects evaluated the feedback canceled audio for the amount of residual feedback components and loudness perception. To assess the feedback components, the subjects were asked to characterize the amount of feedback components (whistling, ringing, howling) perceived in each sentence into one of the six classes enumerated in Table 4.

Loudness ratings refer to the volume of the words in each sentence. The subjects were asked to rate the loudness on a scale of $0 - 5$. 0 indicates that the sentence is inaudible, a 5 means that the sentence is uncomfortably loud and a 3 is the most comfortable level of sound. Five subjects participated in this procedure and they rated each system's output twice. During the test, recorded speech signals were played in a random order through a headphone in a quiet place.

The average subjective ratings for both the methods at different gains are summarized in Table 5. As can be seen from the table, the ratings for the feedback and the loudness obtained from the test at a gain 12.3 dB is approximately same as those for the fixed gain at 8 dB gain. The low perceptual ratings for the fixed gain processing at 10.6 and 12.3 dB along with the acceptable performance of the adaptive gain processing indicates the viability of the hearing aid system presented in this paper.

Table 4. Description of ratings to the subject

Ratings	Feedback	Loudness
0	Loud howling	Inaudible
1	Loud continuous whisting	Soft
2	Soft continuous whistling	Somewhat soft
3	Soft intermittent whistling	Comfortable
4	No audible feedback, acceptable quality	Somewhat loud
5	No audible feedback, good quality	Extremely loud

Table 5. Average ratings for the two schemes

Gain above CG (dB)		8	10	10.6	12.3
Fixed gain processing	Feedback	4.21	4.06	0.43	0
	Loudness	3.11	2.93	4.67	5
Adaptive gain processing	Feedback	4.23	4.18	4.06	4.11
	Loudness	2.98	3.07	3.08	3.02

5. CONCLUSION

A perceptually motivated feedback cancellation for digital hearing aids scheme is presented in this paper. MATLAB simulations and real-time experiments indicate that this scheme provides an additional stable gain over traditional approaches. Psychophysical experiments suggest that this paper's method also delivers perceptually better output sound quality. Further improvements in hearing aid performance are feasible by the incorporation of additional properties of the human auditory system.

6. REFERENCES

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