

Subband Adaptive Filtering Applied to Acoustic Feedback Reduction in Hearing Aids

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Abstract

Acoustic feedback is a problem in hearing aids that contain a substantial amount of gain, hearing aids that are used in conjunction with vented or open molds, and in-the-car hearing aids. Acoustic feedback is both annoying and reduces the maximum usable gain of hearing-aid devices. This paper models the time-varying acoustic feedback path for hearing aids before performing a systematic evaluation of acoustic feedback reduction techniques through perceptual experiments. It is shown that by using a subband NLMS adaptive filtering approach, the same performance of wideband NLMS algorithms is obtained with a fraction of the complexity.

1 Introduction

A major complaint of hearing-aid users is acoustic feedback which is perceived as whistling or howling (at oscillation) or distortion (at sub-oscillatory intervals). This feedback occurs, typically at high gains, because of leakage from the receiver to the microphone.

Acoustic feedback suppression in hearing-aids is important since it can increase the maximum insertion gain of the aid. The ability to achieve target insertion gain leads to better utilization of the speech bandwidth and, hence, improved speech intelligibility for the hearing-aid user. The acoustic path transfer function can vary significantly depending on the acoustic environment [1]. Hence, effective acoustic feedback cancellers must be adaptive.

This paper proposes a subband adaptive filtering approach to reduce acoustic feedback in hearing aids and, therefore, allows increased gain while retaining reasonable sound quality. Using 4 bands, the proposed method achieves one-fourth the complexity of previously reported wideband implementations while retaining the same sound quality.

2 Previous Work

Both continuous and non-continuous adaptation schemes have been proposed to reduce acoustic feedback in hearing aids [1, 5, 6]. None of the proposed schemes uses a subband structure.

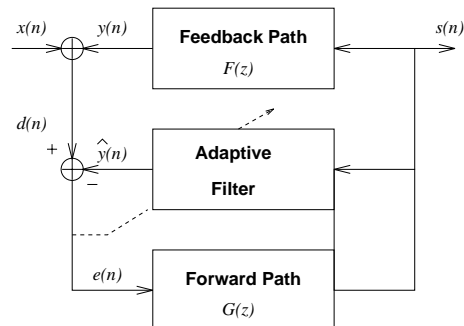


Figure 1: Continuous Adaptation Feedback Cancellation Scheme

An adaptive algorithm attempts to “identify” the hearing-aid feedback path. Once the hearing-aid feedback path has been identified, an estimate of its output is subtracted from the microphone signal. Thus, the hearing-aid user hears the amplified input signal with a reduction of the distortion caused by the feedback path. The basic structure for acoustic feedback cancellation is shown in Figure 1.

The difference between continuous and non-continuous algorithms is that the former has its weights continuously adapted, regardless of the input signal or hearing-aid stability. Non-continuous adaptation algorithms perform adaptation only during silent intervals or when instability is detected. When allowing adaptation during silent intervals, an up-to-date replica of the feedback path can be achieved without sacrificing speech quality. One problem associated with non-continuous adaptation is the need for tuning in order to define a threshold for silence or instability detection. Moreover, in a noisy environment silent intervals can be few and the performance of the algorithm degrades since it only adapts when instability occurs. In some algorithms, the closed loop in Figure 1 is interrupted when adaptation is performed. By interrupting the feedback loop, the algorithm can identify the feedback path with more accuracy and quickly. This, however, can be disturbing to the hearing-aid user since she/he is only listening to noise. It is also possible that the user will lose important acoustic information because of the interruption of the feedback loop. In this paper, we will focus on continuous adaptation algorithms.

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3 Measuring the Acousto-Electric Feedback Path

In order to evaluate the performance of adaptive-filtering algorithms, it is necessary to determine the Feedback Transfer Function ($F(z)$ in Fig. 1).

The acousto-electric feedback path transfer function (AEPTF) in hearing aids consists of the hearing-aid microphone and receiver responses, in addition to the response of the physical path between the receiver and the microphone.

Measurements of the AEPTF were taken in a sound insulated room on a KEMAR mannequin and three healthy-hearing, male subjects. All were fitted with vented in-the-ear (ITE) hearing-aid modules, with the processing plant of the hearing aid disabled. Wideband noise was delivered to the receiver of the hearing aid and recorded on one channel of a DAT (44.1 kHz). The output signal of the hearing-aid microphone was simultaneously recorded on the other channel of the DAT. Several measurements were taken with a metal plate near the ear, with the subject alternating between smiling and not smiling, and with a hand moving near the subject's ear. These measurements were designed to simulate typical situations in which hearing-aid users experience problems with feedback. It should be noted that this technique was not designed to measure the absolute attenuation of the feedback path since the sensitivity of the microphone and other transducers were not calibrated.

3.1 Modeling the Acousto-Electric Feedback Path

The AEPTF was modeled as a linear, IIR system and was assumed to be time invariant over intervals of 25msec. The recorded input and output signal pairs were read from the DAT, downsampled to 8kHz and windowed using 25msec windows with 50% overlap. An estimate of the feedback path transfer function was found for each window.

The Steiglitz-McBride algorithm [8] was chosen to estimate the poles and zeros of the feedback path transfer function. This algorithm resulted in smoother and more accurate estimates than DFT- or LPC- based techniques (see Figure 2).

Measurements taken with a metal plate close to the ear showed little or no change in the AEPTF when the plate was as close as 8 inches from the ear.

Changes in the AEPTF while the subject was alternating between smiling and not smiling were found to be subject-dependent. Some measurements showed a slight overall increase in acoustic feedback while the subject was smiling. In the extreme case, a large anti-resonance was found to occur near 5kHz as the mouth was in transition between smiling and not smiling. This anti-resonance results in a 27dB attenuation over a 300 msec interval.

In measurements taken with a hand moving near the ear, a resonance was found to move from 2 to 4 kHz over the course of 200 msec. As expected, the rates of change in the AEPTF were highly dependent on the rate and range of hand movement. Sample spectra of the AEPTF with a hand moving near a human ear

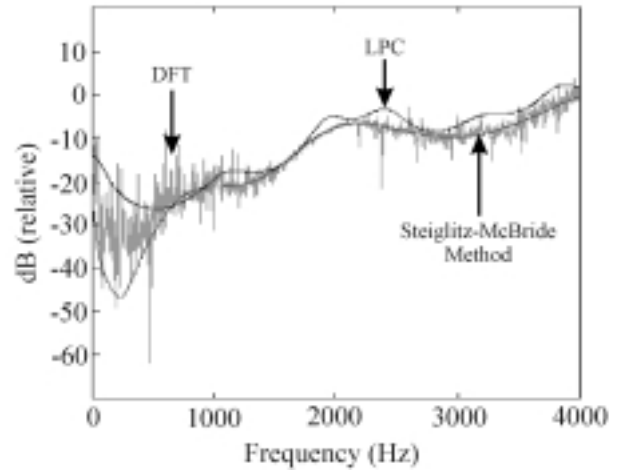


Figure 2: Comparison of Different Estimation Techniques in Modeling the AEPTF

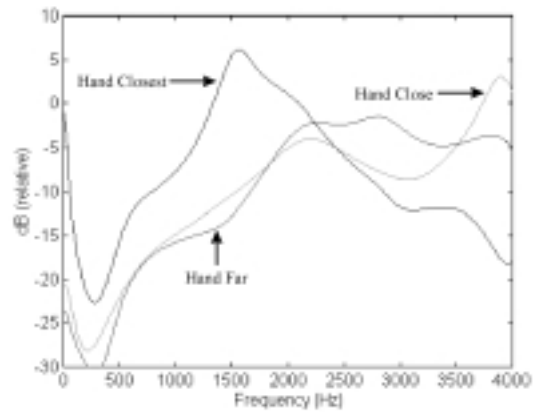


Figure 3: Short-Time Spectra of the AEPTF (Scaled to an Arbitrary Reference)

can be seen in Figure 3. These time-varying measurements were used in computer simulations of adaptive feedback cancellation schemes.

4 Comparison of Different Adaptive Filtering Techniques

Simulations were performed using different adaptive algorithms to evaluate their performance in reducing acoustic feedback when speech is used as the input to the hearing aid.

Initially, a wideband Normalized Least Mean Squares (NLMS) algorithm was implemented. Normalization was performed with respect to the signal ($s(n)$) and error ($e(n)$) power which were computed using an exponential window with a forgetting factor of 0.99. The algorithm produced reasonable sound quality and

was used as a baseline for comparison with other algorithms.

Another algorithm that was implemented was the Gradient Adaptive Lattice (GAL) [3]. Although its behavior was not inferior to that of the NLMS, it was abandoned due to its high complexity. An IIR adaptive algorithm based on the work of Steiglitz and McBride and presented in [7] was also implemented, but numerical instability was encountered when using this algorithm.

A subband NLMS adaptive scheme with four bands was evaluated. The adaptive filtering within each band is independent of the adaptation that takes place within other bands. The performance of this algorithm was comparable to that of the NLMS with a fraction of the complexity, hence this technique was explored further.

5 Continuous Adaptation Using a Subband NLMS Algorithm

The subband adaptive scheme is depicted in Figure 4. The analysis filter bank divides the input signal $s(n)$ into M bands (for simplicity the case of $M = 2$ is shown). Each band is then processed separately using the NLMS algorithm (filter coefficients in this case are updated using the error signals $e_0(n)$ and $e_1(n)$.) The adaptive filter outputs of the different bands are later combined to produce the wideband output signal.

The subband algorithm uses a version of the NLMS algorithm which is described by the following equations:

$$P_s(n) = \lambda P_s(n-1) + (1-\lambda)s^2(n-1) \quad (1)$$

$$P_e(n) = \lambda P_e(n-1) + (1-\lambda)e^2(n-1) \quad (2)$$

$$e(n) = d(n) - \mathbf{w}^T(n-1)\mathbf{s}(n-1) \quad (3)$$

$$\mathbf{w}(n) = \mathbf{w}(n-1) + \frac{2\alpha e(n)\mathbf{s}(n-1)}{N(P_s(n) + P_e(n))} \quad (4)$$

where

- $s(n)$: the input (speech) signal to the adaptive algorithm at time n
- $\mathbf{s}(n)$: the input signal vector defined as $[s(n), s(n-1), \dots, s(n-N+1)]$ where N is the size of the adaptive filter.
- $e(n)$: the error signal used by the adaptive algorithm to update the filter coefficients
- $P_s(n), P_e(n)$: estimates of the power of $s(n)$ and $e(n)$, respectively
- λ : the forgetting factor ($\lambda = 0.99$)
- $\frac{\alpha}{N(P_s(n) + P_e(n))}$: the NLMS adaptation step size (after [6])

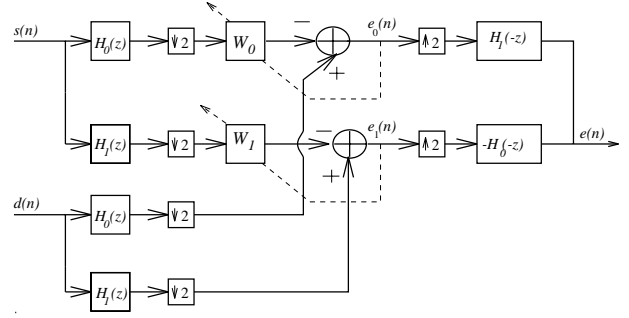


Figure 4: Structure for Subband Adaptive Filtering – Two Band Example

- $\mathbf{w}(n)$: a vector containing the adaptive filter coefficients
- $d(n)$: the reference signal -microphone output signal-

It should be noted that since the information within each band is bandlimited, decimation by a factor of M can be done [9] (assuming that the filters have narrow transition bands and sufficient stopband attenuation). Thus, the data rate per band is reduced and the complexity of the adaptive filters will be $1/M$ of that used in a wideband implementation. The complexity of the analysis and synthesis filters are not significant especially for VLSI implementations where the complexity of the algorithm will be dominated by that of the adaptive filters.

In Figure 4 the case of $M = 2$ (two bands) is depicted. In order to create a four band scheme, a two-level tree-structure is used. The same procedure can be used to produce any number of bands that is a power of 2. However, the tree structure is not recommended for a large number of bands since the overhead for analyzing the input signal becomes high. In that case, it might be more efficient to design the analysis filters directly.

It is expected that the subband NLMS algorithm yields better results than the conventional wideband approach since the overall spectral shape of the signal in each subband is “flattened” leading to a decrease in the eigenvalue spread within each band and, hence, faster convergence.

5.1 Choice of Filters

In subband schemes the choice of the analysis and synthesis filters is important. Ideally, it would be desirable, in the absence of any in-band processing, that the output of the subband scheme be a scaled and delayed version of the input signal. That is not easily accomplished and usually we are satisfied by having an overall transfer function which resembles that of an all-pass filter.

The two major design options for the filters are FIR and IIR. IIR filters can attain the same specs as FIR filters but with lower computational complexity. The main drawback of IIR filters is their non-linear phase response.

Both IIR and FIR tree-structure filters were evaluated. After conducting listening tests, it was clear that FIR filter banks produced better results, in terms of sound quality, than IIR filter banks. The degraded sound quality of the IIR implementation is most likely due to the corrupted phase of each band’s signal. When combined to construct the wideband hearing aid output signal, the phase of each band’s signal may be altered in an inconsistent manner resulting in a degradation in the output sound quality.

5.2 Cross Terms

Regardless of the type of filters used in the analysis and synthesis filter banks, it is desirable to have no aliasing between bands. One way to reduce aliasing is through using crossterm adaptive filters, as proposed in [2]; additional adaptive filters are used between bands to compensate for inter-band aliasing. Provided that the analysis filters have reasonably short transition bandwidths and reasonable stopband attenuation, crossterm filters are needed only between adjacent bands. Based on our simulations, however, the use of crossterm adaptive filters (order 10) did not improve sound quality and the added complexity was deemed unnecessary.

6 Computer Simulations of the Hearing Aid

Computer simulations with a model of a hearing aid, with input and output Automatic Gain Control (AGC), were carried out to compare the performance of the wideband and subband adaptive-filtering algorithms. The simulations used floating-point, double-arithmetric precision, and the initial adaptive filter taps were set to zero. Since the length of the impulse response of the feedback path was between 80-100 taps, the order of the adaptive filter for the wideband case was chosen to be 100 and for the 4-band case, it was 25 per band. The 32-tap analysis and synthesis QMF filters were designed using the coefficients provided in [4]. The value of α was chosen experimentally to be 0.0001; larger values of α resulted in a degradation in sound quality and higher excess error. The wideband NLMS approach was highly sensitive to variations in α whereas the subband approach showed a graceful change in performance as α changed.

The input AGC is used to protect the hearing aid from saturation while the output AGC is used to protect the hearing-aid user from loud sounds. The characteristics of the input and output AGCs are shown in Figure 5. The input AGC performs an initial amplification of the input signal; the signal may then be attenuated by a volume control that follows the input AGC. If, after attenuation, the signal is still high, the output AGC performs a final limiting of the signal before it reaches the output transducer (then to the ear canal.) In simulations it was found that the input and output AGCs remained in the linear region about 90 % of the time and most signal limiting was performed by the input AGC.

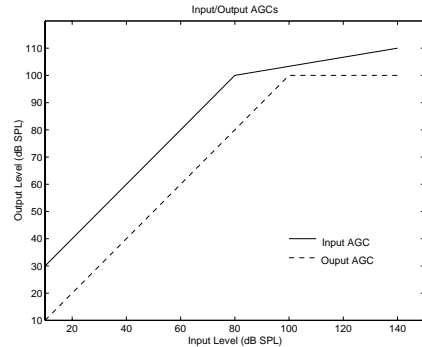


Figure 5: Input and Output AGC Characteristic

Table 1: Average Scores for Stationary and Time-Varying AEPTFs for the Subband (SUB) and Wideband (WB) Approaches

	Stationary Path		Time-Varying Path	
Gain=15 dB	Music	Speech	Music	Speech
SUB	4.3	3.7	2.3	3.2
WB	3.6	3.9	2.1	2.6

6.1 Subjective Evaluations

Listening tests were performed to evaluate the performance of the two algorithms in terms of sound quality. Three sentences from the TIMIT database and two music segments without lyrics were used in these tests. All sentences and music samples were downsampled to 8kHz.

Six native speakers of American English with normal hearing evaluated the sound quality of the hearing aid outputs. Four simulations were performed for each sentence or music segment with a hearing-aid gain of 15 dB for both the wideband and subband NLMS adaptive filters. The overall gain was manipulated by changing the gain in the forward path ($G(z)$ in Fig. 1). Models of the feedback path (both stationary and time-varying) were used in the simulations. The presentation level was maintained at around 70 dB SPL and the subjects rated the sound quality of the processed sounds on a scale from 1 (poor) to 5 (excellent).

Table 1 shows average scores for the wideband and subband NLMS approaches under different conditions; the results indicate that the subband approach performs in a similar fashion to the wideband approach with a reduction of 75 % in complexity. It should be noted that convergence plots (Figure 6) showed faster convergence but higher normalized steady-state error for the subband case when compared to the wideband implementation. The higher error may be due to inter-band aliasing and delay introduced by the analysis and synthesis filters; it seems, however, that the higher error did not affect sound quality especially in the time-varying feedback path case.

To examine whether or not the performance of the

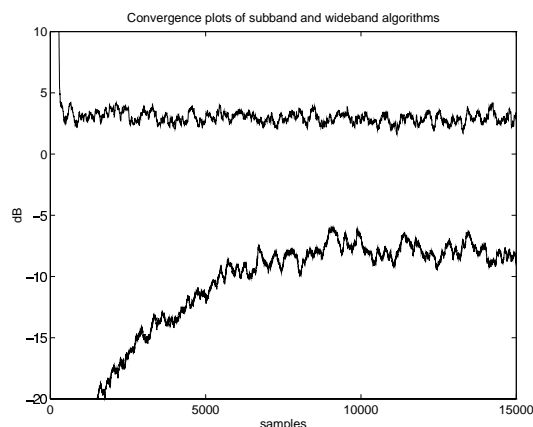


Figure 6: Convergence Plots of the Subband (top curve) and Wideband (bottom curve) Algorithms. The Curves Represent the Normalized Error Signal, Averaged Over 6 Tokens, When the Input is Speech-Shaped Noise and the Feedback Path is Stationary.

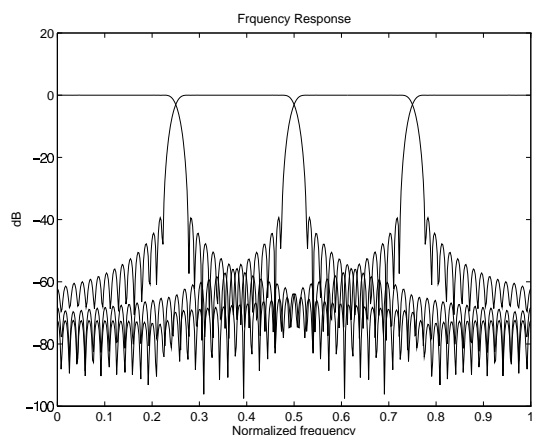


Figure 7: Magnitude response of the Cosine-Modulated Filters (128 taps/filter)

subband filtering approach can be further improved by using perfect-reconstruction filter banks, cosine-modulated filters were used; their magnitude response is shown in Figure 7. No improvement, however, in sound quality was observed.

7 Summary and Future Work

In this paper, a subband adaptive-filtering approach using NLMS is proposed to reduce acoustic feedback in hearing aids. The complexity of the algorithm is less than that of a wideband NLMS with no degradation in sound quality. Future work will include a systematic investigation of the effects of delay and aliasing on the algorithm's performance.

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