OFFENDING FREQUENCY SUPPRESSION WITH A RESET ALGORITHM TO IMPROVE FEEDBACK CANCELLATION IN DIGITAL HEARING AIDS

Ashutosh Pandey and V. John Mathews

Department of Electrical & Computer Engineering
University of Utah, Salt Lake City, UT 84112, USA

ABSTRACT

Acoustic feedback limits the gain provided by hearing aids. Digital hearing aids identify acoustic feedback signals and cancel them continuously in a closed loop with an adaptive filter in the digital domain. This scheme facilitates larger hearing aid gain and improves the output sound quality of hearing aids. However, the output sound quality of hearing aids deteriorates as the hearing aid gain is increased. This paper automatically identifies and suppresses residual acoustical feedback components at frequencies that have the potential to drive the system to instability. The suppressed frequency components are monitored and the suppression is removed when such frequencies no longer pose a threat to drive the hearing aid system into instability. Experimental results obtained with real world hearing aid gain profiles using speech and music signals indicate that the method of this paper provides less distortion in the output sound quality than classical feedback cancelers enabling the use of more comfortable style hearing aids for patients with moderate to profound hearing loss.

Index Terms— Feedback, Hearing aids, Adaptive filters

1. INTRODUCTION

A hearing aid amplifies the incoming sound to make it audible for people with hearing loss. The maximum gain achievable in a hearing aid is limited by acoustic feedback. An adaptive filter is often used to continuously estimate the feedback path and cancel the acoustic feedback in hearing aids. Figure 1 shows the block diagram of a typical digital hearing aid with a single microphone, speaker and an adaptive feedback canceller implemented in the subband domain. Digital hearing aids use discrete signal samples of the microphone signal \( u(n) \) and the speaker signal \( x(n) \) to perform the necessary signal processing. In Figure 1, the prescribed amplification for a hearing-impaired listener is provided in the subband domain with gain values \( g(m) \). The delay \( d_1 \) is used to adjust the bias in the adaptive filter estimate. The broadband variable gain function \( g_c \) can be used to adjust the overall output sound level in changing acoustic environments. Adaptive feedback cancellation (AFC) improves the output sound quality and provides an additional gain over the critical gain for which the hearing aid is stable [1]. The additional gain made possible by feedback cancellation is termed as the 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].

In a recent paper [2], we presented an approach that reduces the forward path gain at frequencies in the signal that are more likely to drive the system to unstable behavior. We refer to the frequencies so identified as the offending frequencies (OF). If offending frequencies are detected accurately and gain is reduced in a narrowband frequency region around these frequencies, the gain reduction is not audible for a few offending frequencies [2, 3]. The gain is reduced with narrowband parametric equalization (EQ) filters.

\[ y(m) = \text{Microphone} \rightarrow \text{Delay} \rightarrow g(\text{Hearing aid Gain}) \rightarrow \text{Speaker} \]

Fig. 1. Simplified block diagram of a digital hearing aid with AFC.

In [2], offending frequency suppression (OFS) method uses a voice activity detector to accurately detect offending frequencies. While the method works well for speech signals, it is not as suitable for music signals. In this paper, we use additional characteristics of the acoustic feedback signal that is suitable for speech and music signals. Furthermore, the method of this paper tracks the changes in offending frequencies that is not done in [2] or traditional methods that employ similar approaches in other applications. If the changes in the offending frequencies are not tracked, it can result in unwanted gain reductions at several frequencies in the signal over long periods of time.

Traditionally, in public address (PA) systems, parametric EQ filters are reset (removed from the forward path) periodically and new offending frequencies are detected upon reset. However, this is not the most desired method because until the offending frequencies are reset, there may be unnecessary distortion in the system. This paper addresses this problem by developing a method that continuously monitor the adaptive feedback canceller coefficients to reset the parametric EQ filters when the offending frequencies change and no longer poses a threat to the stable operation of the hearing aid. Simulation results presented in this paper with real world hearing aid gain profiles, feedback path models and speech/music signals demonstrate substantial performance improvements with the method of this paper over prior art.

2. CLASSICAL METHOD FOR HEARING AIDS

For performance comparisons, we will use a subband-based system employing \( M \) subbands, which are created with oversampled generalized discrete Fourier transform (GDFT) filter banks [4]. Each subband component operates at \( L \leq M \) times lower sampling rate than the full sampling rate of the system. The adaptive filter coefficient
vector for the $i^{th}$ subband contains $N_s$ coefficients. The normalized least-mean-square (NLMS) adaptation is used to perform adaptive feedback cancellation (AFC) in each subband.

3. OFFENDING FREQUENCY SUPPRESSION (OFS)

A block diagram employing offending frequency suppression and the reset algorithm is shown in Figure 2. Detection of offending frequencies is performed by monitoring the energy of the microphone signal in the subband domain $u_i(n)$ and using adaptive notch filters tracking behavior on the microphone signal in the fullband $u(n)$. Specifically, the energy monitoring can distinguish feedback components from audio components because the acoustic feedback components are known to grow steadily in successive time intervals whereas the audio components behave otherwise. Additionally, our approach uses the variability of the notch frequency as the other measure to detect the offending frequencies as suggested in [2]. The offending frequency detection scheme is described in Section 3.1.

Fig. 2. Simplified block diagram of a digital hearing aid with OFS.

Upon detection of an offending frequency, a second-order IIR parametric EQ filter, that is specified by three parameters - the center frequency $f_p$, the depth of suppression $p < 1$ and quality factor $q$, is used to suppress the offending frequency [2]. The parameters $p$ and $q$ are fixed in our implementation whereas the parameter $f_p$ is derived from the adaptive notch filter. Calculation of the coefficients of the parametric EQ filter in the discrete-time domain for given $p$, $q$, $f_p$ and the sampling frequency $f_s$ is performed as described in [2]. Finally, the reset method that monitors adaptive filter coefficients is described in Section 3.2.

3.1. Detection of Offending Frequencies

The offending frequencies are detected independently for each subband. In this section, we explain offending frequency detection in the ideal frequency range of the $i^{th}$ subband - $[f_i^L, f_i^U]$, where $f_i^L = \frac{2\pi m}{M}$ and $f_i^U = \frac{2\pi (m+1)}{M}$, where $u_i(m)$ and $u_i'(m)$ are the lower and upper frequencies for band $i$.

Let the microphone signal energy in band $i$ at time $m$ be $P_i^s(m) = u_i^T(m)u_i(m)$, where $u_i(m) = [u_i(m) u_i(m-1) \cdots u_i(m-N_s+1)]^T$. The relative change in the microphone energy between two successive time intervals $P_i^\Delta(m)$ defined as

$$P_i^\Delta(m) = \frac{P_i^s(m) - P_i^s(m-1)}{P_i^s(m-1)}$$

along with the estimated microphone signal energy $P_i^m(m)$ and the estimated background noise signal power $P_i^b(m)$ are used by the counter $\gamma_i^z(m)$ to monitor the energy change for the $i^{th}$ subband at time $m$. Larger values of the counter $\gamma_i^z(m)$ makes the band $i$ more probable to contain an offending frequency.

The counter is incremented by $\Gamma_u > 0$ to indicate possible howling if the microphone signal $P_i^m(m)$ at time $m$ has sufficient energy (at least $T_b$ times larger than the background noise power $P_i^b(m)$

$$H_i^z(z) = \frac{1 - a_i(n)z^{-1}}{1 - \rho_i(n)z^{-1} + \rho_i^2 z^{-2}}$$

for band $i$ and it ($P_i^s(m)$) is greater than or decreased by a small amount, say $\nu_a$, compared to energy at time $m - 1 (P_i^s(m-1))$ i.e. $P_i^s(m) > 0$ or $P_i^b(m) < \nu_a < 0$. Increase in the microphone energy indicates sustained howling whereas small change in the energy may indicate early stages of howling [2]. On the other hand, the energy growth counter $\gamma_i^m(m)$ is reduced by an amount $\Gamma_r < 0$, if the relative change is smaller than a predetermined negative constant $\nu_t$ to indicate audio components. In other situations, where the relative change in energy $P_i^\Delta(m)$ lies between $\nu_t$ and $\nu_a$, the energy growth rate is modified with a number that is linear interpolation between $\Gamma_t$ and $\Gamma_u$. The amount of change in the energy growth rate value at time $m$ for a given $P_i^\Delta(m)$ is defined by a function $\Phi(P_i^\Delta(m))$

$$\Phi(P_i^\Delta(m)) = \begin{cases} \Gamma_u & ; P_i^\Delta(m) > \nu_u \\ \Gamma_t & ; P_i^\Delta(m) < \nu_t \\ \Gamma_r & ; \text{otherwise} \end{cases}$$

The complete energy growth rate calculation is described in Table 1. It is easy to see from Table 1 that if the relative change in energy is positive or close to zero in successive time indexes, the growth rate value $\gamma_i^m(m)$ grows. Otherwise, the growth rate value will tend to a minimum value $\gamma_m$. If the growth rate value $\gamma_i^m(m)$ exceeds a predetermined threshold $T_r$, one of the two criteria for band $i$ to have an offending frequency is fulfilled.

### Table 1. Detection of offending frequencies in the $i^{th}$ band.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Action</th>
</tr>
</thead>
<tbody>
<tr>
<td>$P_i^\Delta(m) &gt; \nu_u$</td>
<td>Increase $\gamma_i^m(m)$</td>
</tr>
<tr>
<td>$P_i^\Delta(m) &lt; \nu_t$</td>
<td>Reduce $\gamma_i^m(m)$</td>
</tr>
</tbody>
</table>

Adaptive notch filter update and tracking

$$s_i(n) = u_i(n) + \rho_i(n-1)s_i(n-1) - \rho_i^2 s_i(n-2)$$

$$z_i(n) = s_i(n) - a_i(n-1)z_i(n-1) + s_i(n-2)$$

$$P_i(n) = \lambda_i P_i(n-1) + (1 - \lambda_i)s_i^2(n-1)$$

$$a_i(n) = a_i(n-1) + \frac{\rho_i}{\lambda_i} \rho_i^{-1} s_i(n-1) - \frac{\rho_i}{\lambda_i}$$

$$a_m(n) = \lambda_m a_m(n-1) + (1 - \lambda_m)a_i(n)$$

$$\gamma_i^m(n) = \begin{cases} \gamma_i^m(n-1) + 1 & ; |a_i(n) - a_m(n)| < \delta_i \\ 0 & ; \text{otherwise} \end{cases}$$

Energy growth monitoring

$$P_i^\Delta(m) = \min(b_0P_i^m(m-1), P_i^m(m))$$

$$\gamma_i^z(m) = \begin{cases} \gamma_i^z(m-1) + \Phi(P_i^\Delta(m)) & ; P_i^\Delta(m) > T_bP_i^m(m) \\ 0 & ; \text{otherwise} \end{cases}$$

Offending frequency detection (when $n = Lm$)

The adaptive notch filter uses the microphone signal $u(n)$ for calculating the center frequency for each parametric EQ filter employed to suppress an offending frequency. In addition, the adaptive notch filters are also used to detect onset of instability along with the energy growth rate counters. We employ a second-order notch filter

$$H_i^s(z) = \frac{1 - a_i(n)z^{-1} + \rho_i^2 z^{-2}}{1 - 2\alpha_i(n)z^{-1} + \rho_i^2 z^{-2}}$$

for band $i$. The parameter $a_i(n)$ is constrained to adapt between $[2\cos(2\pi f_p^L), 2\cos(2\pi f_p^U)]$ to track the frequency range of the $i^{th}$ subband - $[f_i^L, f_i^U]$. The update equations for the adaptive notch filter realized in direct-form II are summarized in Table 1. In Table 1, $\lambda_a$ is a suitable averaging constant, $\alpha_s$ is the step size for adaptation and $\epsilon_s$ is a small positive constant to prevent singularities.

The second parameter we employ to detect onset of instability is the variability of the coefficients of the notch filter. The variability of the parameter $a_i(n)$ from the mean of its past values

$$\epsilon_s^2 = (a_i(n) - \mu_a)^2$$

and the variability of the center frequency

$$\epsilon_f^2 = (f_p(n) - \mu_f)^2$$

are calculated as the variance of the mean of their past values.

302
a_{\text{in}}(n)$ is monitored with a counter $\gamma_{\text{m}}^n(n)$ as listed in Table 1. If the parameter $a_{\text{in}}(n)$ does not vary significantly from its mean as determined by a pre-selected threshold $\delta_o$, the counter $\gamma_{\text{m}}^n(n)$ grows; otherwise it is reset to 0. If the counter $\gamma_{\text{m}}^n(n)$ gets larger than a pre-determined threshold $T_o$, and the energy growth rate value $\gamma_{\text{m}}^n(m)$ becomes larger than a pre-determined threshold $T_o$, the system determines that the hearing aid may go unstable (howl) at or around the notch frequency. In such event, a parametric EQ filter whose center frequency is derived from the current value of $a_{\text{in}}^n(n)$ as $\frac{1}{2}\pi \cos^{-1}(a_{\text{in}}^n(n)/2)$ is applied to the output signal and the counters $\gamma_{\text{m}}^n(n), \gamma_{\text{m}}^n(m)$ are reset to 0.

### 3.2. Resetting Offending Frequency Suppression (OFS) Filters

In practice, new offending frequencies appear because the feedback path of the hearing aid and the signal characteristics change [1]. We monitor each subband individually to track the changes in different frequency regions. The reset method calculates relative change in the current adaptive filter estimate from its older estimates. If the relative change is larger than a pre-determined threshold, the offending frequencies are reset - parametric EQ filters are removed.

<table>
<thead>
<tr>
<th>$L_i(m)$</th>
<th>$M_i(m)$</th>
<th>$D_i(m)$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\lambda_i L_i(m - 1) + (1 - \lambda_i)w_i(m)$</td>
<td>$\lambda_i M_i(m - 1) + (1 - \lambda_i)w_i(m)$</td>
<td>$\lambda_i L_i(m) - M_i(m)$</td>
</tr>
</tbody>
</table>

Specifically, the reset algorithm uses two measurements of the adaptive filter coefficients $w_i(m)$. For example, we calculate the average $\lambda_i L_i(m)$ of the adaptive filter coefficients for the $i^{th}$ band at time $m$ is estimated using a single pole IIR filter with averaging constant $\gamma_i$, $0 < \lambda_i < 1$. This is treated as a measure of the pre-stable path of the feedback path at time $m$. A short term average $M_i(m)$ of the adaptive filter coefficients is also included. The system assumes that the feedback path has changed for that band. In this event, any parametric EQ filters that fall in the frequency range $[f_1, f_2]$ are removed. The reset algorithm for the $i^{th}$ band at time $m$ is listed in Table 2. The normalized distance $\kappa_i(m)$ remains close to 0 if the feedback path is relatively stationary and increases in magnitude when there are changes in the feedback path. The variable $\gamma_i^m(m)$ counts the number of times the normalized distance has been more than a pre-determined threshold $\delta_o$ to trigger reset process.

### 4. RESULTS AND DISCUSSION

This section presents the results from MATLAB simulations of the hearing aid algorithms. The true feedback path was simulated using a 192-tap FIR filter in parallel with a homogenous quadratic nonlinearity. The nonlinearity simulates the nonlinear distortions in the in a hearing aid system. The harmonic signal strength was 40 dB below that of the output of the linear component. Coefficients for the linear component of the feedback paths were obtained from measurements of an inside-the-ear (ITE) hearing aid.

Two feedback paths were used in the experiments. In estimating the feedback paths, the ITE hearing aid was fitted into the earpiece of a Knowles Electronic Manikin for Acoustic Research (KEMAR) that was placed in a quiet location. A large white board on a stand was placed parallel to the face plate of the ITE hearing aid at different distances to create different feedback paths. The white board (reflective surface) was 100 cm and 5 cm away for the first and second feedback path, respectively.

The feedback canceller employed a linear FIR system model with 128 coefficients. This undermodeling and the model mismatch attempt to capture the practical situation. Parameters of the subband design were $M = 128, L = 16$ and $f_s = 16000$. The speech input signals to the hearing aid in experiments were 6 clean speech waveforms of length 80 seconds taken from the TIMIT database. Experiments where music signals were used, 6 music waveforms of length 80 seconds from the album “18 till I die” by Bryan Adam were chosen. The music signals were resampled at 16000 Hz for the experiments in this paper. Colored noise samples to model hardware noise, with the power spectral density reducing at the rate of 3 dB per octave as the frequency increases, were added to simulate a noisy signal with 40 dB signal-to-noise ratio.

![Fig. 3. Hearing loss profile and insertion gain of: (a) mild-gently sloping hearing loss (b) moderately-flat hearing loss (c) moderate-steeply sloping hearing loss; (d) profound-gently sloping hearing loss.](image)

To assess performance of the classical system and the method of this paper, four hearing aid profiles - mild-gently sloping loss, moderate-flat loss, moderate-steeply sloping loss and profoundly sloping loss and the corresponding insertion gain profiles - as shown in Figures 3a - 3d were used. The insertion gains for the hearing loss profiles used in this paper were obtained with the NAL-RP prescription [1] and are shown in Figures 3a - 3d.

The initial value of hearing aid gain was set to 20 dB below the target gain. Subsequently, the gain was slowly increased for 20 seconds at the rate of 1 dB/second to reach the target gain level. The adaptive feedback cancellation experiment ran for 80 seconds in each case. The last 10 seconds of the experiment were deemed as the steady state. The perceptual quality of the output speech from the steady state was measured in each simulation using the perceptual evaluation of speech quality (PESQ) measures [6]. PESQ provides perceptual quality rating of a speech segment between −0.5 and 4.5 and can be interpreted as follows. The highest score indicates that the speech signal is virtually identical to the clean speech segment. The PESQ scores between −0.5 and 1 indicate that the distortions in the speech signal are very high and the segment sound unacceptably annoying. The ratings of 3, 2 can be interpreted as “good quality”, “slightly annoying” and “annoying” respectively.

In the first experiment, we evaluated the offending frequency suppression scheme in terms of speed of detection and output sound
quality. The output sound quality was judged with the PESQ measure from the steady state output sound. The speed of the offending frequency suppression method was calculated with the time to recover from instability (TRI) measure. The TRI is an estimate of the time intervals during which howling occurs in the experiment. Howling occurrences were manually identified according to the method described in [3]. Let the time interval for the \( j \)th howling segment be \( \Delta t_j \), then the TRI is calculated as

\[
TRI = \frac{1}{N_h} \sum_{j=1}^{N_h} \Delta t_j
\]

(2)

where \( N_h \) is the number of offending frequencies detected. The parameters for detection of offending frequencies with adaptive notch filters and energy growth monitoring were \( \rho = 0.95, \alpha_a = 0.005, \lambda_a = 0.99, \epsilon_a = 10^{-5}, \lambda_m = 0.99, \delta_T = 0.05, T_a = 750, v_\nu = -0.01, \nu_\nu = -0.1, T_\Gamma = 1, \Gamma_\Gamma = -4, \gamma_\nu = -20, T_r = 75, \delta_r = 1.0003 \) and \( T_0 = 6 \). Parameters of a parametric EQ filter to suppress an offending frequency upon detection were \( P = 50.012 \) (\(-6 \text{ dB}\)) and \( q = 5 \). The maximum number of parametric EQ filters were set to 12 in order to limit perceptual distortions in the output sound. The parameters for the reset algorithm were \( \lambda_1 = 0.999, \lambda_2 = 0.7, \epsilon_r = 10^{-12}, \delta_o = 0.1 \) and \( T_o = 10 \).

Table 3. Measures for the OFS with and without reset algorithm.

<table>
<thead>
<tr>
<th>Measures</th>
<th>With reset</th>
<th>Without reset</th>
</tr>
</thead>
<tbody>
<tr>
<td>Case 1</td>
<td>1</td>
<td>3</td>
</tr>
<tr>
<td>Case 2</td>
<td>3</td>
<td>5</td>
</tr>
<tr>
<td>Case 3</td>
<td>3</td>
<td>8</td>
</tr>
</tbody>
</table>

In this simulation, profile 4 and the six speech input signals of 80 seconds were used. Three acoustic feedback cases were studied. In the first case, feedback path 1 was used for the whole 80 seconds. Feedback path 2 was used for the whole experiment in the second case. In the third case, the experiment started with feedback path 1 and after 40 seconds, it was switched to feedback path 2. For the three feedback cases discussed above, total number of howling occurrences (\( N_h \)), TRI and mean PESQ values were calculated and are listed in Table 3. Table 3 also lists the average number of offending frequencies that remained for the different feedback cases in the steady state (\( N_i \)) with and without the use of the reset algorithm.

The howling occurrences (\( N_h \)) in Table 3 indicate that ramping up gain to the target gain in the beginning and the feedback path change created several instances of howling. The number of howling occurrences were the most for the feedback case 3 because of the feedback path change. The effects of such occurrences were mitigated by the OFS method. The instability was detected in less than 0.4 second in all the experiments. The use of reset algorithm yielded in fewer number of parametric EQ filters in the steady state (\( N_i \)). However, this did not compromise the output sound quality as indicated by the PESQ values. On the contrary, use of the reset algorithm yielded slightly better output sound quality.

In the next experiment, the classical system was evaluated against the offending frequency suppression (OFS) with reset method on four hearing aid gain profiles using speech and music signals. Feedback path 2 was used for this experiment. The PESQ values for various profiles with all methods are summarized in Table 4. In Table 4, both methods provided good output sound quality for profiles 1 and 2 for both types (speech and music) of input signals. The classical method provided good output sound quality for profile 3 with speech signals, however, had soft whistling sounds in the output sound when music signals were input to the hearing aid system. On the other hand, the OFS method yielded good output sound quality for both types of input signals with hearing aid gain profile 3. Finally, the classical scheme had significant residual feedback components in the output sound for the hearing aid profile 4. Use of the OFS methods improved the output sound quality for profile 4 as well.

Table 4. PESQ values for various schemes.

<table>
<thead>
<tr>
<th>Performance Measure</th>
<th>Method</th>
<th>Signal</th>
<th>Profile</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PESQ</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Classical</td>
<td>Speech</td>
<td>3.9</td>
<td>3.9</td>
</tr>
<tr>
<td></td>
<td>Music</td>
<td>3.9</td>
<td>2.8</td>
</tr>
<tr>
<td>OFS</td>
<td>Speech</td>
<td>3.9</td>
<td>3.9</td>
</tr>
<tr>
<td></td>
<td>Music</td>
<td>3.9</td>
<td>3.6</td>
</tr>
</tbody>
</table>

It is interesting to note that the traditional system could not provide enough feedback cancellation in the case of profound hearing loss gain profile (profile 4) for an ITE hearing aid used in the experiment. Often, audiologist suggest the use of behind-the-ear hearing aids for patients with profound hearing loss [1]. The results in Table 4 indicates that the method of this paper will enable a patient with profound hearing loss to use an ITE hearing aid which is not possible otherwise. The method of this paper not only provides additional added stable gain but also enables patients to use more comfortable style hearing aids which is a big concern for many patients [1].

5. CONCLUSION

This paper presented an algorithm that modifies the forward path gain in a hearing aid to improve adaptive feedback cancellation efficiency. MATLAB simulations and evaluations suggest that this paper’s method delivers better output sound quality especially for patients with need for high hearing aid gains. The reset algorithm successfully eliminated unnecessary filters placed in the signal path during the transient periods. The reset algorithm developed here can also be extended for public address (PA) offending frequency suppression methods.

6. REFERENCES


Feedback path was derived from an ITE.