HOWLING SUPPRESSION IN HEARING AIDS USING LEAST-SQUARES ESTIMATION AND PERCEPTUALLY MOTIVATED GAIN CONTROL

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ABSTRACT

Howling is a significant problem even in digital hearing aids equipped with adaptive feedback cancellation. Among the many causes of howling is the inability of the adaptive filter to track rapid changes in the feedback path. Many systems use howling detectors to detect the start of howling and reduce the hearing aid gain for several seconds to avoid prolonged howling. Unfortunately the inadequate speech pressure levels (SPL) during times when the gain is reduced causes loss of information and reduced intelligibility of speech signals arriving at the patient's ears. This paper presents a new method that switches to a least-squares adaptation scheme with linear complexity at the onset of howling. The method adapts to the altered feedback path quickly and allows the patient to not lose perceivable information. The complexity of the least-squares estimate is reduced by reformulating the least-squares estimate into a Toeplitz system and solving it with a direct Toeplitz solver. In addition, the gain function is changed immediately after howling detection in such a way that the system operates in a stable manner and the distortions caused are not perceived because of temporal masking. Simulation results comparing with a conventional method is presented in the paper to demonstrate the superior howling suppression capabilities of the method.

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, which is present mostly because of a vent that provides patients comfort from the acoustic pressure difference at the ear drum. 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 equipped with an adaptive feedback canceller. It consists of a microphone, a speaker, a broad-band gain function G and an output compression limiter (OCL) in the forward path. The adaptive filter W estimates the feedback path H that comprises of the characteristics of the microphone M, the speaker S and the acoustic feedback path AF. The OCL attempts to ensure that the output level of the hearing aid is comfortable for the patient.

The adaptive feedback cancellation scheme improves the output sound quality of hearing aids significantly and provides added stability to hearing aids [1, 2, 3]. However, this type of closed loop hearing aid systems is susceptible to unstable behavior that results in howling when the feedback path changes suddenly or rapidly. This problem is annoying and occurs often in daily routine whenever a reflective

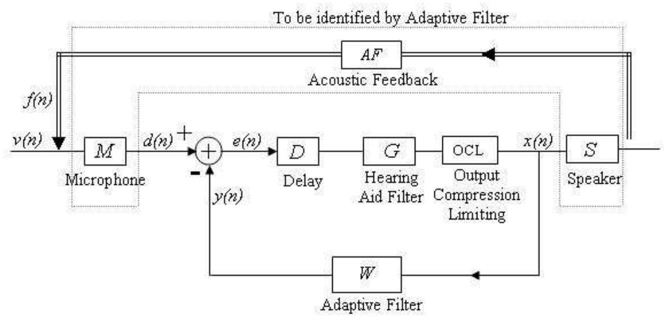

Fig. 1. Block diagram of a typical digital hearing aid

surface such as a telephone receiver is brought near the face plate of the hearing aid, and in many other ways. The majority of the feedback canceller systems employ gradient adaptive algorithms. They have relatively low computational complexity and therefore can be implemented within the small chip areas available in hearing aids. Unfortunately gradient adaptive filters have slow convergence characteristics and may not be able to track the high rate of build up of the output signal during times of sudden or rapid changes in the feedback path, thus giving rise to howling behavior [1]. One approach to tackling this problem is to use a howling detector to sense the start of howling. In order to avoid prolonged howling, the gain function of the hearing aid is reduced when howling is detected. Subsequently, the gain is increased slowly while the adaptive filter estimates the altered feedback path and the hearing aid system is stable [1]. The problem with this approach is that the gain has to be increased gradually over several seconds to keep the hearing aid system stable and produce an output with low distortion. This may cause the patient to miss some information because of inadequate sound pressure levels at the eardrum while ramping up the gain function. It is highly desirable to develop an adaptive filter with fast convergence and low computational complexity to suppress howling in hearing aids.

It is well known that least-squares adaptive filters converge faster than gradient based algorithms in general [4, 5]. However, even the most efficient least-squares algorithms [4, 5] have much higher computational complexity than most gradient adaptive filters. In our approach, we use the least-squares method to obtain an initial estimate of the altered feedback path immediately after howling is detected and switch to a gradient algorithm after a pre-determined number of iterations. The computational complexity of the least-square adaptive filter is comparable to that of the gradient algorithm because we make use of the efficiencies available during the initialization of the

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LS estimation process. The least-squares problem $Ax = B$, where A is an $N \times N$ known Toeplitz matrix, B is an $N \times 1$ known vector and x is an $N \times 1$ unknown vector can be solved in $O(N^2)$ operations using a direct Toeplitz solver [6]. These operations are spread over N iterations in our approach to maintain linear complexity of the adaptive filter. Feedback cancellation is not performed during the LS adaptation process immediately after howling is detected. During the transition period, the hearing aid gain is reduced and increased periodically so as to prevent the hearing aid from becoming unstable. The gain is reduced for short durations so that the user will not perceive a loss of information because of the post temporal masking effect [7].

The rest of the paper is organized as follows. An overview of the feedback canceller for hearing aids is presented in Section 2. This section also describes a howling detection algorithm. The new howling suppression algorithm is detailed in Section 3. In Section 4, the performance for the new algorithm is evaluated and compared with a competing structure using MATLAB simulations. We make the concluding remarks in Section 5.

2. AN OVERVIEW OF DIGITAL HEARING AID SYSTEMS

Figure 1, shows the block diagram of a typical digital hearing aid. The input signal to the speaker $x(n)$ and the output of the microphone $d(n)$ are used to estimate the feedback path H with an adaptive filter W. In this work, we approximate the feedback path with a linear impulse response with N coefficients. In what follows, we denote the coefficient vector as $\mathbf{w}(n)$. Among the many gradient adaptive filters available to us, we chose the normalized least-mean-square algorithm (NLMS) in this work. The update equations for the NLMS adaptation for estimating the feedback path are given in Table 1. In the update equations, α is a small positive constant that controls the adaptation speed of the system and ϵ is another small positive constant designed to prevent the denominator of (4) from going to zero [4]. The parameter D is a fixed delay value and is provided to reduce the bias in the adaptive filters coefficients [1, 2, 3].

Table 1. Update equations for an adaptive feedback canceller using NLMS

$$x(n) = Ge(n-D-1) \quad (1)$$

$$\mathbf{x}(n) = \begin{bmatrix} x(n) & x(n-1) & \cdots & x(n-N+1) \end{bmatrix}^T \quad (2)$$

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

$$\mu(n) = \frac{\alpha}{\|\mathbf{x}(n)\|^2 + \epsilon} \quad (4)$$

$$\mathbf{w}(n+1) = \mathbf{w}(n) + \mu(n)e(n)\mathbf{x}(n) \quad (5)$$

We have assumed a broadband adaptive filter and a constant gain for the hearing aid. In most commercially available systems, the forward path contains a filter bank and the gain in each band may differ from those in others. Extension of the algorithm in Table 1 to a multirate implementation is conceptually straightforward. However, we do not discuss this approach here to keep the presentation at a simple level.

The design of the howling detector used in our work is based on a simple correlation analysis of the error signal $e(n)$ and its delayed version $e(n-D)$. If the adaptive filter matches the feedback path closely, we expect the error signal $e(n)$ and its delayed version $e(n-D)$ to be relatively uncorrelated. When the feedback path is suddenly changed or the adaptive filter is unable to track the feedback path, the feedback signal $f(n)$ is not cancelled, implying that the feedback signal $f(n)$ is present in the error signal $e(n)$. When howling occurs, $f(n)$ typically has dominant sinusoidal components making $e(n)$ and $e(n-D)$ relatively more correlated. Consequently, the cross-correlation between $e(n)$ and $e(n-D)$ may be used as a marker to detect the onset of howling. We use a correlation factor defined as

$$\rho = \frac{\left| \sum_{i=0}^{L} e(n-i)e(n-D-i) \right|}{\sum_{i=0}^{L} |e(n-i)e(n-D-i)|} \quad (6)$$

and computed over a segment of length L for this purpose.

3. THE NEW HOWLING SUPPRESSION ALGORITHM

The basic idea of howling suppression is as follows. The hearing aid employs an NLMS adaptive filter for estimating and canceling the feedback path. The system is also equipped with the howling detector described in Section 2. The adaptive filter coefficients stops adapting and coefficients are reset to zero as soon as howling is detected. For N samples after howling detection, we do not update the coefficients of the filter so that the dominant spectral components created by the howling activity do not affect the new updates. The least-squares algorithm described in the next subsection is employed to adapt the coefficients for the next $N + d - 1$ samples, where $d << N$. The gain function of the hearing aid is varied during the transition period in a manner that would allow the overall system to behave in a stable manner and at the same time allow the patient to mask the distortion caused by the variations. This process is described in Section 3.2. After the transition period of $2N+d-1$ samples after howling detection, the coefficients obtained using the LS estimate are copied to the adaptive filter coefficients and the NLMS adaptation is resumed.

3.1. Least Squares Estimation After Howling Detection

Let the data matrix $\mathbf{X}(n)$ and the desired response vector $\mathbf{d}(n)$ be defined as

$$\mathbf{X}(n) = \begin{bmatrix} \mathbf{x}(1) & \mathbf{x}(2) & \cdots & \mathbf{x}(n) \end{bmatrix} \quad (7)$$

$$\mathbf{d}(n) = \begin{bmatrix} d(1) & d(2) & \cdots & d(n) \end{bmatrix}^T \quad (8)$$

where $\mathbf{x}(n)$ is defined in (2). It is well known that the optimal least-squares coefficients vector $\hat{\mathbf{w}}(n)$ that minimizes $\| \mathbf{d}(n) - \mathbf{X}^T(n)\hat{\mathbf{w}}(n) \|^2$ is given by

$$\hat{\mathbf{w}}(n) = \Psi^{-1}(n)\theta(n) \quad (9)$$

where $\Psi(n) = \mathbf{X}(n)\mathbf{X}^{\mathbf{T}}(n)$ and $\theta(n) = \mathbf{X}(n)\mathbf{d}(n)$. In what follows, we assume that $\Psi(n)$ is invertible when $n = N$. The data matrix $\mathbf{X}(N)$ is a square matrix that will also be invertible in this case. The extension of the results to the case when $\Psi(n)$ is a singular matrix is not difficult. At time $n = N$, we can write

$$\begin{aligned}\hat{\mathbf{w}}(n) &= \Psi^{-1}(n)\theta(n)\\ &= (\mathbf{X}(n)\mathbf{X}^T(n))^{-1}\mathbf{X}(n)\mathbf{d}(n)\\ &= (\mathbf{X}^T(n))^{-1}\mathbf{X}^{-1}(n)\mathbf{X}(n)\mathbf{d}(n)\\ &= (\mathbf{X}^T(n))^{-1}\mathbf{d}(n) \qquad (10)\end{aligned}$$

Since $\mathbf{X}^T(n)$ is a Toeplitz matrix, (10) can be solved in $2N^2 + 8Nlog_2(N) - N$ arithmetic operations as shown in the Appendix [6]. These many operations are expensive to implement in one iteration. Fortunately, we can solve a large portion of the algorithm in $N-1$ iterations as we get successive samples with a maximum $4N$ operations during any iteration as shown in step 2 of the Appendix. After N iterations, $8Nlog_2N + N$ operations are left to complete the computations in (10). We complete these operations in the next d iterations where d is a number of the order of $2log_2N$ so that approximately $4N$ operations are completed during each iteration. Therefore, we can obtain the initial estimate of the altered feedback path with the least-squares method in $2N+d-1$ iterations with linear complexity.

3.2. Gain processing

The gain function of the hearing is reduced by a constant β for the first N samples after howling is detected to reduce the effect of the dominant spectral components created in the signal by the howling action. In all the simulations presented in the next section we used $\beta = 0.01$. The gain is increased and kept at the prescribed level for the patient in the next $N + d$ samples. Maintaining a high gain at these samples helps to obtain a better estimate of the altered feedback path with the least-squares method [1, 4]. The gain is reduced for the next D samples because these samples were generated during the transition period with no feedback cancellation and a high gain and therefore may contain many unwanted spectral components of the feedback. Alternating low and high gain values for short periods of times allows stable operation of the system. Furthermore, since gain values are altered for short periods, the distortions may not be perceivable because of the temporal masking effect of the human auditory system [7].

4. RESULTS AND DISCUSSION

Simulations were conducted in MATLAB with the feedback path obtained from an in-the-ear hearing aid. The impulse response of the feedback path was modeled using a FIR filter with 256 coefficients. The critical gain of the feedback path was 37 dB. An FIR adaptive filter with 256 taps was used to estimate the feedback path. The gain and delay used for simulations were set to 50 dB and 128 samples, respectively in all simulations presented here. The howling detector declared the onset of howling whenever the correlation factor in (6) exceeded 0.9.

In the simulations, we introduced a sudden change in the feedback path by negating all coefficients of the feedback path sometime after the adaptive filter has reached the steady state. The howling detector sensed howling in about 652 samples (approximately 40 ms) after this change. The classical method reduces the gain as soon as howling is detected by 40 dB and increases it slowly back to the prescribed gain over the subsequent 50000 iterations. These parameters were selected through experimentation such that the hearing aid is stable and produces the least distorted output signal possible for the conventional scheme. In simulations involving the method of this paper, the gain is reduced by 40 dB intermittently during the transition period as explained in Section 3.2. The parameter d was chosen to be 16.

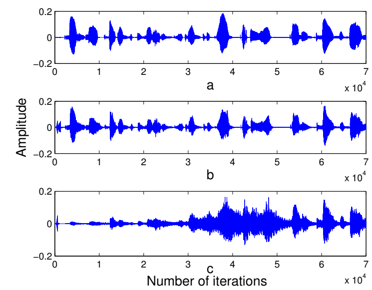

Fig. 2. Scaled output signals after howling detection: (a) the desired output (b) with the new scheme (c) with slowly increasing gain

Figure 2 shows the output waveforms (normalized to have similar amplitude ranges for the three sub-plots) after the onset of howling for both schemes. It is clear that the classical scheme does not provide sufficient amplification. The method of this paper appears to reproduce the input signal reasonably faithfully at the output. We can see that the effect of the delay in howling detection in both Figures (2b) and (2c) for a very short duration at the beginning of the plots. There are slight differences between the signals in Figures (2a) and (2b). This is due to build up of uncancelled feedback during the transition period. However, these differences are not perceptually bothersome because they occur over very short durations.

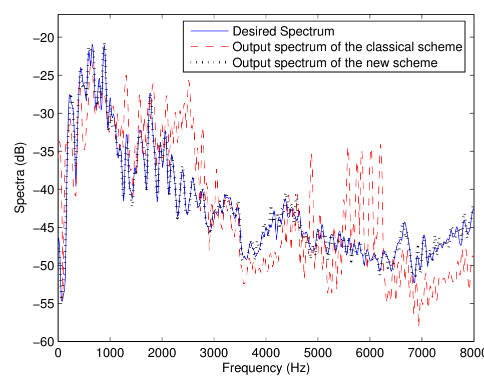

Fig. 3. Comparision of the output spectra of the two schemes

As can be expected, the classical scheme produced relatively faint output during the transition period leading to some lack of intelligibility. The classical method attained the maximum gain in about 50000 iterations (3.125 sec) whereas the transient period for the new scheme was less than 657 iterations (0.0375 sec) for $N = 256$.

The outputs of the two schemes are compared in the frequency domain in Figure 3. The spectrum was calculated for the signals shown in Figure 2. The new scheme matches the desired response closely whereas the conventional scheme produces a significant amount of distortions in the process of ramping up the gain. These distortions occur because of the slow convergence of the classical scheme. On the other hand the new scheme adapts to the altered feedback path quickly and therefore cancels out most of the undesired spectral components of the feedback. Figure 4 displays the misalignment between the true feedback path and the estimated feedback path just prior and just after the onset of howling. The misalignment was calculated as

$$\text{misalignment(dB)} = 20\log\frac{\|\mathbf{w}(n) - \mathbf{h}(n)\|}{\|\mathbf{h}(n)\|} \tag{11}$$

At the beginning of the plot the system was adapted to the feedback path and the misalignment was low. The misalignment increased suddenly when the feedback path changed. With the first method, convergence was quite slow as can be seen from the dashed line curve in Figure 4. With the new scheme, the misalignment did not change during the transition period because the adaptive filter was not updated. The new scheme accurately estimated the altered feedback path during the transition period and updated the adaptive filter at the end of the transition period thus suddenly reducing the misalignment in a very short time (approximately 32 ms).

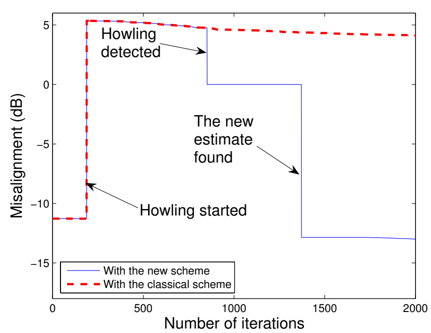

Fig. 4. Misalignment around the onset of howling for the two schemes compared in the example

5. CONCLUSION

A novel howling suppression scheme based on the least-squares method is presented in this paper. The proposed scheme has faster convergence than conventional howling suppression methods. The least-squares estimate used in the proposed howling suppression method is implemented with a linear complexity. The system also employs a perceptually motivated gain control algorithm that allows stable operation. The distortions occurring during the transition period are kept below the perceptual threshold of the listener because they are masked by the temporal masking phenomenon in the human auditory system. This scheme can also be applied to suppressing the feedback in other systems such as acoustic/network echo cancellers.

6. APPENDIX

In the following presentation "$\sim$" on a column vector represents reverse order of entries, "$\star$" denotes element by element multiplication of vectors and $\mathbf{0}_{N\times 1}$ is a column vector with N zeros.

Step 1. **Initialization:**

$\lambda_0 = x(n-N+1)$

$a_0 = b_0 = r_0 = s_0 =$ null vector

Step 2. **Recursions:** for $k = 0$ to $N-2$

$$\alpha_k = \frac{-1}{\lambda_k}(x(n-N+k+2) + \mathbf{a}_k^T\tilde{\mathbf{r}}_k) \tag{12}$$

$$\beta_k = \frac{-1}{\lambda_k}(x(n-N-k) + \mathbf{s}_k^T\tilde{\mathbf{b}}_k) \tag{13}$$

$$\mathbf{a}_{k+1} = \begin{bmatrix} \mathbf{a}_k + \alpha_k\tilde{\mathbf{b}}_k \\ \alpha_k \end{bmatrix} \quad \mathbf{b}_{k+1} = \begin{bmatrix} \beta_k \\ \tilde{\mathbf{b}}_k + \beta_k\mathbf{a}_k \end{bmatrix} \tag{14}$$

$$\lambda_{k+1} = \lambda_k(1 - \alpha_k\beta_k) \tag{15}$$

$$\mathbf{r}_k = \begin{bmatrix} \mathbf{r}_k \\ x(n-N+2) \end{bmatrix} \quad \mathbf{s}_k = \begin{bmatrix} \mathbf{s}_k \\ x(n-N-k) \end{bmatrix} \tag{16}$$

Step 3. **Compute Output:**

$$\mathbf{ae} = \begin{bmatrix} 1 \\ \mathbf{a}_{N-1} \\ \mathbf{0}_{N\times 1} \end{bmatrix} \quad \mathbf{be} = \begin{bmatrix} 1 \\ \mathbf{0}_{N\times 1} \\ \tilde{\mathbf{b}}_{N-1} \end{bmatrix} \quad \mathbf{de} = \begin{bmatrix} \mathbf{d}(n) \\ \mathbf{0}_{N\times 1} \end{bmatrix}$$

$\mathbf{aef} = \text{FFT}(\mathbf{ae}), \mathbf{bef} = \text{FFT}(\mathbf{be}), \mathbf{def} = \text{FFT}(\mathbf{de})$

$\mathbf{uf} = \mathbf{aef} \star \mathbf{def}, \quad \mathbf{vf} = \mathbf{bef} \star \mathbf{def}$

$\mathbf{u} = \text{IFFT}(\mathbf{uf}), \quad \mathbf{v} = \text{IFFT}(\mathbf{vf})$

$\mathbf{p}(m) = \mathbf{v}(m), \mathbf{q}(m) = \mathbf{u}(m+N)$ for $1 \le m \le N$

$\mathbf{p}(m) = \mathbf{q}(m) = 0$ for $N+1 \le m \le 2N$

$$\mathbf{x}f = \mathbf{aef}(m)\mathbf{pf}(m) + (-1)^m\mathbf{bef}(m)\mathbf{qf}(m) \quad \text{for } 1 \le m \le 2N \tag{17}$$

Compute $\mathbf{x} = \text{IFFT}(\mathbf{xf})$

$$\hat{\mathbf{w}}(n) = \frac{\mathbf{x}(1:N)}{\lambda_{N-1}} \tag{18}$$

7. REFERENCES

[1] A. Kaelin, A. Lindgren, and Wyrsch S., "A digital frequency domain implementation of a very high gain hearing aid with compensation for recruitment of loudness and acoustic echo cancellation," *Signal Processing*, vol. 64, pp. 71–85, 1998.

[2] J. Hellgren, "Analysis of feedback cancellation in hearing aids with filtered-X LMS and the direct method of closed loop identification," *IEEE Trans. Speech Audio Process.*, vol. 10, no. 2, pp. 119–131, February 2002.

[3] Hsiang-Feng Chi, Shawn X. Gao, Sigfrid D. Soli, and Alwan Abeer, "Band-limited feedback cancellation with a modified filtered-X LMS algorithm for hearing aids," *Speech Communication*, vol. 39, pp. 147–161, 2003.

[4] B. Farhang-Boroujeny, *Adaptive Filters: Theory and Applications*, John Wiley and Sons, New York, NY, 1998.

[5] J. Cioffi and Kailath T., "Fast recursive-least-squares transversal filters for adaptive filtering," *IEEE Trans. Acoust. Speech Signal Process.*, vol. 32, no. 1, pp. 304–337, April 1984.

[6] J. R. Jain, "An efficient algorithm for a large Toeplitz set of linear equations," *IEEE Trans. Acoust. Speech Signal Process.*, vol. 27, no. 6, pp. 612–615, December 1979.

[7] W. A. Yost and D. W. Nielsen, *Fundamental of hearing*, Hold Rinehart and Winston, Austin, TX, second edition, 1968.