ACOUSTIC FEEDBACK CANCELLATION FOR HEARING-AIDS, USING MULTI-DELAY FILTER

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ABSTRACT

Acoustic feedback cancellation in hearing-aid needs robust and efficient adaptive filter techniques. In this paper, we propose a method based on the Generalized-Multi-Delay Filter algorithm. This algorithm is particularly attractive in the hearing-aid context because it requires only small size transforms. We derive a new method for the step-size control which improves the behavior of the algorithm and enables to reach the stability conditions in hearing-aid.

1. INTRODUCTION

In a hearing-aid device, the acoustic feedback path between the receiver and the microphone is the source of instability for the whole closed-loop system. When the system is unstable the device sound quality is deteriorated since distortion and high-level self-sustaining oscillations can set-up. These phenomena are very unpleasant for the hearing-impaired and limit the potential gain of the device.

This study proposes to introduce in the hearing aids context an improved transform domain echo cancellation algorithm, called Generalized Multi-Delay Filter (GMDF) in a specific implementation using the Fast Hartley Transform (FHT). This implementation was initially introduced in [6] and is derived from time-domain Block Least Mean Square (BLMS) adaptive filtering.

This technique was originally developed for hand-free-telephones and modern teleconferencing systems. Its basic idea consists in a segmentation of the feedback path impulse response estimation which is updated directly in the Hartley domain. It also takes advantage of the FHT as a real transform having the self-inverting property.

In spite of GMDF efficiency, it is not sufficient to reach the stability conditions in the particular context of hearing-aid without a better control of the step-size of the adaptation process. To achieve this goal, the present paper proposes a new transform domain normalization strategy at marginal computational cost.

2. FEEDBACK IN HEARING AID

In a hearing aid system (described in Figure 1), the acoustic source signal \( s[n] \), where \( n \) is the time index, is corrupted by the additive feedback signal \( u[n] \) originating from the output signal \( y[n] \) leaking back to the input. In classical acoustic echo cancellation system, these signals are usually named:

- \( s[n] \) : the near-end talker signal,
- \( u[n] \) : the echo signal,
- \( y[n] \) : the far-end talker signal.

Hence, without feedback reduction, the closed-loop system transfer function is:

\[
T(z) = \frac{z^{-D}.G(z)}{1 - z^{-D}.G(z).H(z)}
\]

where the intended hearing loss compensation and the feedback path transfer functions are approximated respectively by \( G(z) \) and \( H(z) \). \( D \) is the time delay introduced by the algorithm in samples.

Figure 1: Hearing-aid device with its intended transfer function \( G(z) \) and an adaptive filter \( H(z) \) estimating the feedback path \( H(z) \)

Assuming that \( G(z) \) and \( H(z) \) are two linear time-invariant transfer functions, self-oscillation phenomenon can occur and causes unpleasant whistling sound if it
exists \( z \) on the unit circle for which:

\[
\begin{align*}
|G(z).H(z)| &> 1, \text{ and} \\
\arg(G(z).H(z)) &> 2m\pi \text{ rad, } m \in \mathbb{N}.
\end{align*}
\] (2)

According to condition (2), it is obvious that the gain provided by \( G(z) \) has to be limited in order to keep the system in stable conditions. However limiting the gain prevents from efficiently compensating the hearing loss.

In the recent years, many different approaches have been studied to reduce acoustic feedback [2, 3, 4]. Two different approaches are generally considered:

- non-continuous adaptation: the adaptive feedback cancellation filter is update only during silence or when oscillation is detected,
- continuous adaptation generally based on LMS adaptive filtering.

In many non-continuous schemes, a burst of pseudo-random noise sequence is generated at the output of the hearing-aid in order to identify the feedback path.

This article focuses on LMS-based continuous adaptation scheme according to the following reasons:

- feedback path can vary between two updates
- regularly switching off the system and generating noise can be unpleasant for the hearing impaired.

LMS-based feedback cancellation systems adapt the coefficients of the filter \( \tilde{H}(z) \) that estimates the feedback signal through \( \hat{u}[n] \) by minimizing, in the Least Mean Square (LMS) sense, the error signal:

\( x[n] = v[n] - \hat{u}[n] \).

The closed-loop system transfer function with adaptive feedback cancellation is given by (see Fig. 1):

\[
T(z) = \frac{z^{-D}.G(z)}{1 - z^{-D}.G(z).\left(H(z) - \bar{H}(z)\right)}
\] (3)

According to Equation (3), the stability condition becomes:

\[
|G(z).\left(H(z) - \bar{H}(z)\right)| < 1
\] (4)

LMS-based algorithms have to deal with drastic conditions of use compared to classical acoustic echo cancellation:

- the high gain in the device negatively balances the stability condition.

According to this particular context, the present paper proposes to use an improved algorithm from modern acoustic echo cancellation system.

### 3. GENERALIZED MULTI-DELAY FILTER IN HARTLEY DOMAIN

Ferrara introduced in [1] an exact and fast implementation of Block-LMS in the frequency domain which takes advantage of the Discrete Fourier Transform (DFT) circular convolution property. This method is known as Fast LMS (FLMS).

The second advantage of the FLMS is that the frequency transforms coefficients are less correlated than in the time domain. This enables a faster and nearly uniform convergence rate of all the modes of the adaptive filter.

In fact, FLMS can be transposed with any transforms having the circular convolution property. In [6], FLMS was exactly transposed in the Discrete Hartley Transform (DHT) domain.

The DHT of a M-dimensional real vector \( x = [x_0, \ldots, x_{M-1}] \) is closely related to the DFT and defined by:

\[
X_k^H = \frac{1}{\sqrt{M}} \sum_{m=0}^{M-1} x_m \left(\cos\left(\frac{2\pi mk}{M}\right) + \sin\left(\frac{2\pi mk}{M}\right)\right)
\] (5)

Using DHT in the context of hearing aid seems very attractive for the following reasons:

- it is a real transform,
- it has the self-inverting property \((DHT = DHT^{-1})\) and thus reduces the implementing cost of programming a transform function in the hearing device DSP chip.

A modification of FLMS was initially proposed in [7]. This implementation, known as Multi-Delay Filter (MDF), consists in sectioning the impulse response of length \( N \) in \( K \) short successive segments of length \( M \). Its initial aim was to use \( K \) M-dimensional transforms instead of one larger N-dimensional in order to deal with large impulse responses in room acoustical environment. The use of short transform is very interesting according to the low computational power and fixed-pointed arithmetic of hearing aid dedicated DSP chips. This algorithm also enables to reduce the delay to a correct value (10-20ms).

In [5] a Weighted OverLap-Add (WOLA) version of MDF was introduced and is known as Generalized MDF\( \alpha \) (GMDF\( \alpha \)) where \( \alpha \) stands for the overlap factor \( \alpha = \frac{M}{\pi \ell} \) where \( M \) is the size of the transform and
As for the FLMS, GMDFα can be transposed to DHT domain (GMDFα-DHT).

Consequently, the present paper suggests the use of GMDFα-DHT for acoustic feedback cancellation in hearing aids. A complete description of the algorithm is given in [6].

In GMDFα-DHT the time-domain tap weight vector is divided into K segments and the algorithm directly handles with the M-dimensional DHT of each segment : $H_s^k$, $k \in [0; K - 1]$.

At each block index s, the update can be written:

$$H_s^{k+1} = H_s^k + C \mu_s \odot (Y_s^k * E_s)$$

where

- $C$ is a constraint matrix,
- $\mu_s$ is the step-size parameter which controls the convergence of the algorithm (see definition in section 4)
- $Y_s^k$ is the DHT of the hearing aid output data also sectioned in K blocks.
- $E_s$ is the DHT of the error signal which is the input data for the hearing aid.
- $\odot$ is the Schür component by component product
- $\star$ denotes cross-correlation equivalent operation in the Hartley domain.

GMDFα-DHT is also a very flexible solution for feedback cancellation in hearing-aid. Indeed, different strategy could be chosen on each block and in each frequency bin.

4. NEW ADAPTATION STEP-SIZE DEFINITION

To ensure the convergence of the adaptive filter at marginal computational cost, simple update rules have to be investigated. Defining a new control of the step-size in equation (6) seems to be a good choice. The step-size $\mu_s$ is given by:

$$\mu_s = \mu_0 T_s$$

where

- $\mu_0$ is constant and given by : $\mu_0 = \frac{1}{K+1}$ (see [5])
- $T_s$ is the transform domain normalization vector.

Indeed, $T_s$ is an important parameter which controls the convergence of the adaptation process. It is given by:

$$T_s = \left[ \frac{1}{P_{s,1}}, \ldots, \frac{1}{P_{s,M}} \right]^T$$

where the $i^{th}$ spectral component $P_{s,i}$ is obtain by first order low-pass filtering of the $i^{th}$ power spectrum component of $Y_s$:

$$P_{s+i,i} = \gamma P_{s,i} + (1 - \gamma) |Y_s,i|^2, 0 < \gamma < 1$$

A new definition of $T_s$ is proposed here. According to the particular conditions in this closed-loop system, the idea behind this new normalisation strategy is to adopt a more careful adaptation rule. This may slow down the convergence but will prevent from some bias.

4.1. Minimum threshold of the Power Spectrum in Sub-band

If $P_{s,i}$ has a very small value, the signal is probably carrying no valuable information in the $i^{th}$ frequency bin and is not properly exciting the closed-loop system. However, according to equation (8) small values of $P_{s,i}$ induce a fast adaptation of the gradient. In order to reach a satisfactory trade-off, a new definition of $T_s$ is proposed.

Power spectral coefficients are grouped in $K_B$ sub-bands $\{B_k\}_{k \in [1; K_B]}$ and the mean spectral power of each subband is evaluated.

$$P_{s,k}^m = mean \{P_{s,i}\}_{i \in B_k}$$

Then, in order to prevent from normalizing the gradient by small values, $P_{s,k}^m$ is used as a minimum threshold for the corresponding subband.

$$T_{s,i} = 1 / max (P_{s,i}, P_{s,k}^m)$$

4.2. Balancing with the Error Power Spectrum

In modern acoustic echo cancellation systems, double-speech detectors are commonly used in order to freeze the adaptation process when the far-end talker signal is too weak or in case of double-talk situation.

In the hearing-aid context, the algorithmic complexity of double-speech detectors seems to be prohibitive according to the dedicated DSP chips capabilities. Nevertheless as the echo and source signals are highly correlated, a strategy has to be proposed to slow down the adaptation process when the error signal power is too important.

Recently, the introduction of a delay at the output of the device has been proposed to reduce correlation between the echo and source signals [8]. This approach has been integrated to the present study since it enables to reduce the bias in the identification of

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1 Due to the close relation between DFT and DHT, the bins of the transform domain are indicated as frequency bins.
the echo path. It should also be noticed that a delay is induced by block processing in transform domain algorithms.

Let us consider that the power spectrum of the error signal reaches a high level. Two situations are then probable:

(A) the source signal \( s[n] \) is present and the algorithm would consider it as an error. Thus, adaptation should be slowed down.

(B) the source signal is not present. Thus, the convergence is not achieved and the adaptation should be sped up.

As it is impossible to determine in which situation the system is, a careful strategy needs to be chosen. The proposed strategy is to balanced the normalization vector \( T_s \) with the spectrum power of both the output signal \( Y_s \) and the error signal \( E_s \).

In situation (A), this strategy would tend to slow down the adaptation. In situation (B), it seems coherent to normalize the error signal by its own power spectrum.

As for \( P_s \), let us define the error signal power spectrum by:

\[
\tilde{P}_{s+1,i} = \gamma \tilde{P}_{s,i} + (1 - \gamma) |E_{s,i}|^2 \tag{12}
\]

\[
\tilde{P}_{s,i}^m = \text{mean} \left\{ \tilde{P}_{s,i} \right\}_{(i \in B_k)} \tag{13}
\]

The new definition of the normalization vector thus becomes:

\[
T_{s,i} = 1/ \left[ \max (P_{s,i}, P_{s,i}^m) + \rho \cdot \max (\tilde{P}_{s,i}, \tilde{P}_{s,i}^m) \right] \tag{14}
\]

Where \( \rho \) is a parameter which balances the relative importance of \( P_s \) and \( \tilde{P}_s \).

5. SIMULATION RESULTS

Computer simulations were conducted to evaluate the efficiency of the \textit{GMDF\textalpha-DHT} algorithm in reducing acoustic feedback.

\textit{GMDF\textalpha-DHT} was also compared to the proposed implementation with a novel stepsize parameter control according to the previous section.

Two different impulse responses measurements of In-The-Ear hearing aid devices were used (see Figure 2). These impulse responses have been normalized to exactly reach the critical stability value.

Figures (3) and (4) present the results through spectrogram comparisons. A 5s speech sentence at the sampling rate of 16kHz was used as the source signal and the impulse response models were switched at the middle of the duration in order to simulate a fast change in the feedback path. The simulated

![Figure 2](image-url)

\textit{Figure 2:} Impulses responses and their respective power spectrum

hearing-aid gain was static and set respectively to 5dB and 15dB in experiments illustrated in Figures (3) and (4).

With the gain set to 5dB (see Figure 3), the original \textit{GMDF\textalpha-DHT} implementation prevents efficiently from self-oscillations until the switch between the impulse responses (top of the figure). This switch causes instability and self-oscillations for about 2s. After this relatively long time of convergence, the adaptive filter comes back to a stable solution without any self-oscillation. In same conditions, with the novel stepsize definition, no self-oscillation occurs even at the switch time (see bottom of the figure).

In the second test, with the gain set to 15dB, a complete instability of the system can be observed for the original \textit{GMDF\textalpha-DHT} implementation (see top of Figure 4). In the same test condition, the novel strategy prevents from system instability. Sound quality is quite the same than in the previous test with the same algorithm.

According to these experiments, the novel strategy appears to efficiently improve the behavior of the \textit{GMDF\textalpha-DHT}.

Similar results have been observed with many different set-up. Notably, high gain values can be provided with a compression stage in the hearing-aid. Indeed non-linearities in the \textit{hearing-loss compensation stage} of the device (see Fig. 1) improve the decorrelation of the source and echo signals and thus improve the feedback path estimation.

Independently of sound quality and distortion, no self-oscillation phenomenon occurs in experiments conducted with higher gain values when the new step-size definition is used. However, further studies are needed in order to defined properly the right gain margin for hearing-aid device provided by the proposed feedback cancellation algorithm.
6. CONCLUSION

In this paper, a modern acoustic echo cancellation algorithm was introduced to the hearing-aid particular context. A novel strategy for the transform-domain normalization was also proposed in order to have a better controlled on the adaptation process of the adaptive filter. Simulation results presented in section 5 show that high-gain value are reachable thanks to this new normalization.

7. REFERENCES