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An improved block adaptive system for effective feedback cancellation in hearing aids

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ABSTRACT

The hearing aid being a battery operated, portable device requires short processing delay, low computational complexity, with appreciable acoustic feedback cancellation effect. The prediction error method (PEM) and PEM with shadow filter (PEM-SH) based adaptive feedback canceller (AFC) referred as PEMAFC and PEMAFC-SH respectively reduces the amount of bias present in the estimate of feedback path. The available partitioned block frequency domain adaptive filter (PBFAF) based implementation of PEMAFC (PBFAF-P) and PEMAFC-SH (PBFAF-PS), offers a potential option for modelling an adaptive filter with many taps along with short block processing delay. However, the PBFAF suffers from large computational load because of the involvement of computationally expensive gradient constraints in each partition. Though removing or alternately applying the gradient constraint saves some computations but it results in significant performance degradation. With an objective of substantially reducing the computational burden and simultaneously retaining the performance, this paper develops an improved partitioned block Hartley domain adaptive filter (IPBHAF) and then employs it for effective feedback cancellation in hearing aids. Further, the IPBHAF with modified step size (IPBHAF-M) is proposed to achieve both fast convergence and better steady state performance. The simulation based experiments demonstrate the superior performance of IPBHAF-M based implementations of PEMAFC (IPBHAF-MP) and PEMAFC-SH (IPBHAF-MPS) over the PBFAF-P and PBFAF-PS in terms of both computational complexity and feedback cancellation performance.

1. Introduction

Hearing aids assist in listening to hearing impaired people. The basic hearing aid consists of a microphone, an amplifier and a receiver. Different types of hearing aids are available in the market to suit different users depending on the type of hearing loss and the age of the user. The major problem faced by most of the hearing aid users is the degradation of sound quality due to the effect of acoustic feedback which is perceived as howling or whistling by the users. This phenomenon deteriorates the quality of speech, limits the maximum stable gain and the amplification provided by the hearing aid [1,2]. The close proximity of the receiver and microphone, leakage from the receiver, vent and structural transmission from the shell of the hearing aid are the prime factors responsible for the acoustic feedback in hearing aids. The measured acoustic feedback path comprises of the characteristics of the microphone, D/A converter, A/D converter and the speaker along with the physical path between the receiver and the microphone. The transfer function of the acoustic feedback path is subjected to variations due to the changes in the nearby acoustics of the user [3].

There are several methods reported in the literature using adaptive feedback cancellers (AFC) to suppress the effect of acoustic feedback, broadly divided into continuous and non-continuous mode of adaptation. In non-continuous mode of adaptation [4,5], the basic operation of the hearing aid is interrupted and the adaptive filter coefficients are adapted by inserting a probe noise or training signal at the hearing aid output. On the other hand, continuous adaptive feedback canceller (CAF) does not require an identification signal and continuously adapts to the coefficients of the adaptive filter [5]. The CAF produces a biased estimate of the feedback path because of the presence of forward path which makes the whole system closed loop. The forward path introduces correlation between the incoming desired signal and the input to the loudspeaker when the desired signal is spectrally colored. This biased estimate of the feedback path causes cancellation of desired signal along with the feedback signal, thus leads to signal distortion.

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Many solutions have been reported in the literature to tackle the problem of biased feedback path estimate like using delays or nonlinearities in the forward path or cancellation path [6], employing constrained [7] and band limited adaptation [8], Filtered-XLMS (FXLMS) [9], PEM based AFC [10], two stage methods for unbiased feedback cancellation [11], PEM based adaptive filtering with row operations (PEM-AFROW) [12], transient mean square analysis of PEMAFCs [13], clipping used in the feedback path [14], physical performance based evaluation of feedback cancellation methods [15], using subjective measures to assess the sound quality [16], AFC based on weighted adaptive projection sub gradient [17], and a low delay method is proposed in [18].

Further, other methods like using all pass filters having time varying poles [19], a band limited linear predictive coding based approach [20], PEM based AFC with two signal model [21], insertion of specifically designed probe noise signals [22,23], improved prediction error filter [24], pitch and formant estimation based method [25], using two microphone based approach [26] have been developed for reducing the bias while estimating the feedback path. Usually, all the above methods work towards reducing the bias in the estimate of the feedback path. However, for practical application like the hearing aid which is a battery operated device, having limited power supply, another important issue which needs to be addressed is the computational complexity associated with the adaptive feedback canceller. Therefore, designing an efficient adaptive feedback canceller involves a trade off between achieving a better steady state performance by obtaining an unbiased estimate of the feedback path, convergence rate of the feedback cancellation algorithm and the computational complexity associated with it.

The least mean square (LMS) algorithm exhibits poor convergence for colored input signal and also has high computational complexity per input sample when used for modelling an adaptive filter with many taps [27]. The convergence rate is enhanced by using block frequency domain adaptive filters (BFAF) for modelling the feedback path [28]. However, the BFAF suffers from large block processing delay when modelling the impulse response with large number of coefficients. Further, with the motive to reduce the block processing delay (latency), the PBFAF has been used for realizing the PEMAFC and PEMAFC-SH by Spriet et al. [29]. The PBFAF uses block length of smaller size and replaces a large point convolution task by summing up the outputs of smaller convolutions. However, the PBFAF is computationally expensive, since it involves two FFT operations per partition for computing the gradient constraint. The omission of the gradient constraint [30,31] though saves computations but introduces significant amount of performance degradation, which may not be acceptable for hearing aid users. It may even sometime cause divergence of the weights of the adaptive filter as they get polluted by wrap-around artifacts [27].

In order to achieve substantial reduction in the computational complexity without compromising the steady state performance, this paper proposes an improved partitioned block Hartley domain adaptive filter (IPBHAF) for feedback cancellation in hearing aids. The Hartley transform involves less memory and computations as it is a real valued transform and has self inverting property which considerably reduces the cost of hardware implementation of the transform domain algorithm [32]. The IPBHAF utilizes the concept of sinusoidal constraints to mitigate the effect of wrap around artifacts introduced as a result of the omission of gradient constraints.

A proper choice of step size is required to maintain balance between the convergence rate and the steady state performance [33]. A variable step-size is used with LMS algorithm [34] and NLMS [35] algorithm for feedback cancellation in hearing aids. Therefore, a bin wise varying modified step size [36] is used along with IPBHAF termed as IPBHAF with modified step size (IPBHAF-M) to gain two fold advantage of better feedback cancellation accuracy and convergence rate. In this paper the PEM and PEM-SH based feedback canceller is realized by IPBHAF-M as IPBHAF-MP and IPBHAF-MPS respectively.

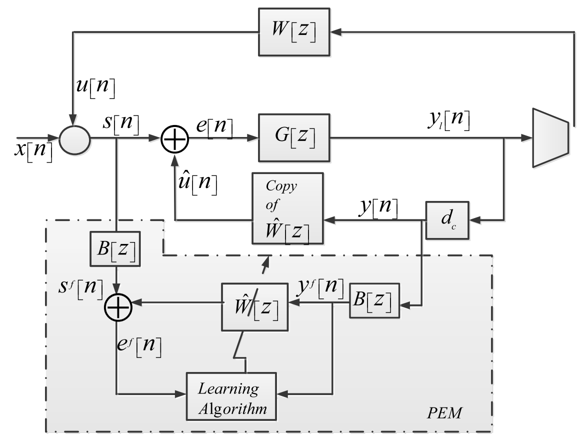

Fig. 1. PEM based feedback canceller.

The organization of the paper proceeds as follows: Section 2 outlines the PBHAF based implementation of PEMAFC (PBHAF-P) and PEMAFC-SH (PBHAF-PS). The proposed IPBHAF-M based feedback cancellation scheme is described in Section 3. Simulation based experiments are carried out to assess the performance of the proposed method in Section 4. Finally, Section 5 provides the conclusion of the paper.

2. PBHAF based feedback cancellation scheme for hearing aids

This section describes the PBHAF based feedback cancellation scheme for hearing aids. The PEMAFC and PEMAFC-SH are implemented in Hartley domain by employing PBHAF. The partitioned block adaptive filtering scheme results in faster convergence, low block processing delay and reduced computational complexity. The impulse response of the feedback path is partitioned into partitions of smaller lengths. PBHAF implements the whole system in Hartley domain, which further saves the memory and arithmetic computations.

In the following section, the time domain vectors are represented by lower-case bold italics, whereas the time domain matrix is represented by upper-case bold italics. The vectors in Hartley domain are denoted by upper-case italics whereas matrix in Hartley domain is written as underlined upper-case italics.

2.1. PBHAF based implementation of PEMAFC (PBHAF-P)

The CAF technique results in biased estimation of the feedback path when the input signal is spectrally colored because of the presence of forward path. The PEM based approach is used to obtain an adaptive model of the input signal and then the microphone and the loudspeaker signals are filtered with the signal model to pre-whiten the desired signal component present in them [10]. Fig. 1 shows the PEM based approach for feedback cancellation. $x[n]$ represents the desired signal which is amplified and delivered to the tympanic membrane of the hearing aid user as $y_1[n]$. An estimate of the feedback signal $\hat{u}[n]$ is used to cancel the effect of acoustic feedback from the microphone signal $s[n]$.

The desired input signal is represented as

$$x[n] = \boldsymbol{a}^T[n]\boldsymbol{r}[n] \tag{1}$$

where $\boldsymbol{a}[n]$ represents the coefficient vector of the adaptive signal model $A[z]$, $\boldsymbol{r}[n] = [r[n], r[n-1] \cdots r[n-N_f+1]]^T$ is a zero mean