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Design of Flexible Signal Processing Algorithm for Reducing Feedback Signal on Digital Hearing Aid System

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Abstract

This paper presents adaptive filtering algorithm for processing noise feedback signal in digital hearing aid system. Conventional hearing aid system suffers from noise feedback signal which degrades the original speech signal and cause signal instability. This can be reduced by utilizing adaptive filtering techniques in the filter bank. However this method is very unstable since their weights are modified frequently. Hence a normalized adaptive filtering technique is utilized in the proposed method. Initial study on signal level on hearing output is carried and proposed algorithm is compared with adaptive beam former which works well in noisy environment. Tradeoff between segmented signal to noise ratio and signal quality is considered. The algorithm achieves a better signal gain quality of 16dB over the limit gain. The algorithm utilizes adaptive feedback noise reduction filter bank having 128 filter coefficients.

Keywords: Hearing Aid, Adaptive beam former, optimal filter, Singular value decomposition.

I. Introduction

In recent years the development of sophisticated digital hearing aid system increases due to the necessity for improving speech intelligibility and better life quality. However due to presence of acoustic feedback from the environment, the performance of the digital hearing aid system degraded as the voice signal travels from speaker to earphone system. This induces instability in the signal at higher frequencies and results in discomfort for hearing impaired patients. In order to minimize this acoustic effect, researchers have developed various adaptive algorithms for reducing feedback signals. One of the most commonly used methods is least mean square algorithm [1] for hearing aid systems. The algorithm suffers from poor stability though the algorithm is simple, linear and offers fewer computations. In order to combat instability problems, normalized least mean square algorithms are utilized [2] whose filter coefficients are normalized and less gain variation over feedback signals. This algorithm gives estimated output of the voice signals. In addition to voice signal estimation, the error signal parameters are also taken in to account in [3] for better control over filter coefficient parameters for adaptation. Some of the literatures discussed on filtered X LMS algorithms [4] and normalized filtered X

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LMS algorithms [5] which comprises of estimation based on filter feedback subsystems in the forward path along with filtered error signal estimate in the feedback path. A study on feedback signal in the ear is carried in [6] based on filtered X LMS algorithms. A Modified adaptive algorithm is embedded along with beamforming techniques for selecting jamming signals [7]. In addition to this techniques, side lobe cancellation methods are developed in [8] for behind the ear hearing aid systems. This system comprises of two stage adaptive beam changing capability with better noise immunity levels. GSVD based optimal filtering algorithms are developed for multiple speech enhancement in [9]. Recently acoustic feedback cancellation in hearing aids using dual adaptive filtering and gain-controlled probe signal is carried in [10]. The algorithm utilizes delay-based normalized least mean square algorithm for reduction of feedback signals. The performance of the signal quality is evaluated based on subjective and objective feedback sound quality evaluation techniques used in [11].

In this paper, singular value decomposition based optimal filtering technique is proposed based for reduction of noise in the feedback signal. The algorithm utilizes weighted filter coefficients whose values are modified adaptively based on SVD. The proposed technique is compared with other conventional algorithms under single noise scenario.

II. Digital Hearing Aid Model

The hearing Aid model used [12] is shown in Fig. 1. The input signal x(n) is obtained from the speaker and the output signal d(n) are utilized derive the estimation of filter coefficient in the feedback path. Let N be the number of the filter coefficient used in the adaptive filter and w(n) be the weights corresponding to the filter. The error signal e(n) is derived from estimated output d(n) and the feedback signal y(n) which is utilized to modifies the weights of the filter. The delay d is used to reduce the instability of the filter coefficient’s used in adaptive filter. When the noise is above the threshold limit due to howling effect, the correlation between error signal e(n) and its past error signal e(n-1) become less correlated and is used to detect presence of noise in the feedback path. The correlation factor is given by

$$\rho = \frac{\left|\sum_{i=0}^{L} e(n-i)(n-D-i)\right|}{\sum_{i=0}^{L} |e(n-i)(n-D-i)|} \tag{1}$$

The adaptive filter estimates the output d(n) by means of convolution between input signal x(n) and weights w(n)

$$\hat{d}(n) = W_n^T(n)x(n) \tag{2}$$

$$W_n^T(n+1) = W_n^T(n) + \Delta W_n^T(n) \tag{3}$$

Where the weights are adjusted by a step size of $\Delta W_n^T(n)$.

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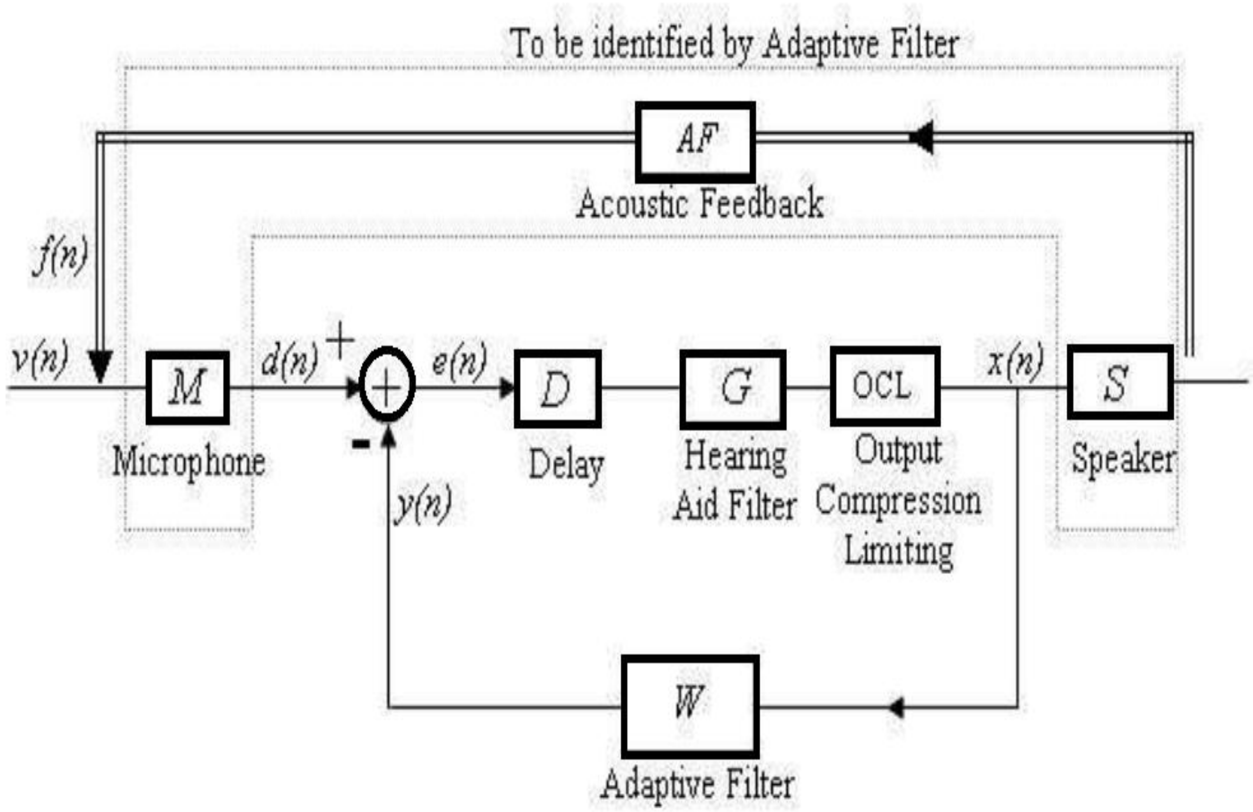

Fig. 1. Simple Digital Hearing Aid model

III. Singular Value Decomposition Based Optimal Filter Technique

In this model the speech signal is constructed from noise input comprising of filtering techniques based on SVD model. Let C(n) is the constructed signal from noise signal N(n)

$$C(n) = d(n) + N(n) \tag{4}$$

The general model of SVD based filter weights are given by

$$W_{SVD} = \epsilon(C(n) * C(n)^T)^{-1} * \{\epsilon(C(n) * C(n)^T)^{-1} - \epsilon(N(n) * N(n)^T)^{-1}\} \tag{5}$$

If an array of microphones are used in the hearing aid system then,

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$$C(n) = \begin{bmatrix} C(n)^T \\ C(n+1)^T \\ C(n+2)^T \\ C(n+3)^T \\ . \\ . \\ C(n+p-1)^T \end{bmatrix}^T \tag{6}$$

$$N(n) = \begin{bmatrix} N(n)^T \\ N(n+1)^T \\ N(n+2)^T \\ N(n+3)^T \\ . \\ . \\ N(n+p-1)^T \end{bmatrix}^T \tag{7}$$

Where p is the total number of array of microphones utilized in the hearing aid systems. The value of input C(n) and noise N(n) signals are obtained from voice feedback systems. The residual noise $\in_n^2$ for a threshold $\propto$ is given by

$$min \in_n^2 < \propto \text{ Where } 0 \leq \propto \leq 1.$$

Therefore equation (5) becomes

$$W_{SVD} = C(n)^{-1} diag \left\{ \frac{q\sigma_i^2 - p\eta_i^2}{q\sigma_i^2} \right\} \begin{bmatrix} C(n)^T \\ C(n+1)^T \\ C(n+2)^T \\ C(n+3)^T \\ . \\ . \\ C(n+p-1)^T \end{bmatrix}^T \tag{8}$$

Where σ_i^2 and η_i^2 are SVD generalized singular values.

IV. Results and Discussions

The input signal C(n) for a period 10 second at a data rate of 64bps is capture using microphone and are shown in Fig.2 below. The signal is sampled and is given to the hearing aid system.

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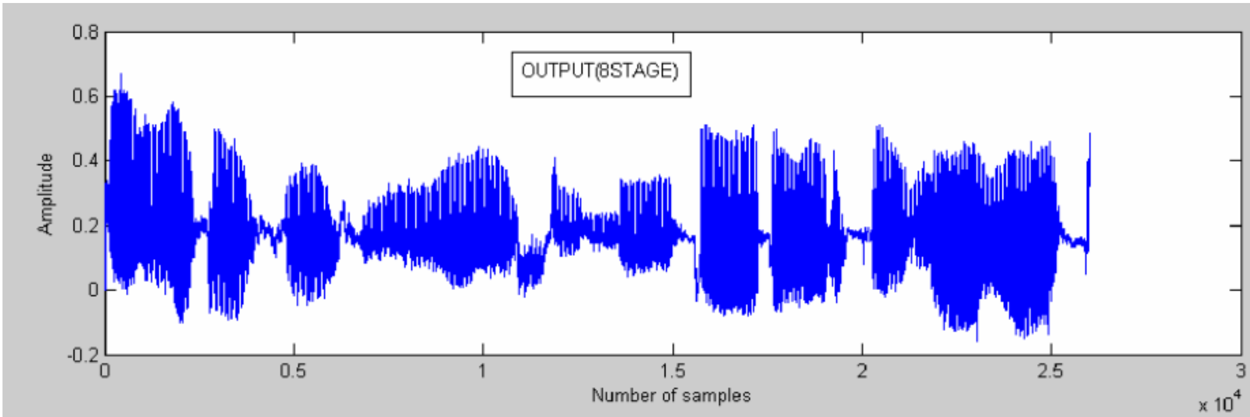

Fig.2 Input voice signal.

The spectral sharpening of the signal after de-correlating with noise feedback signal is shown in Fig. 3.

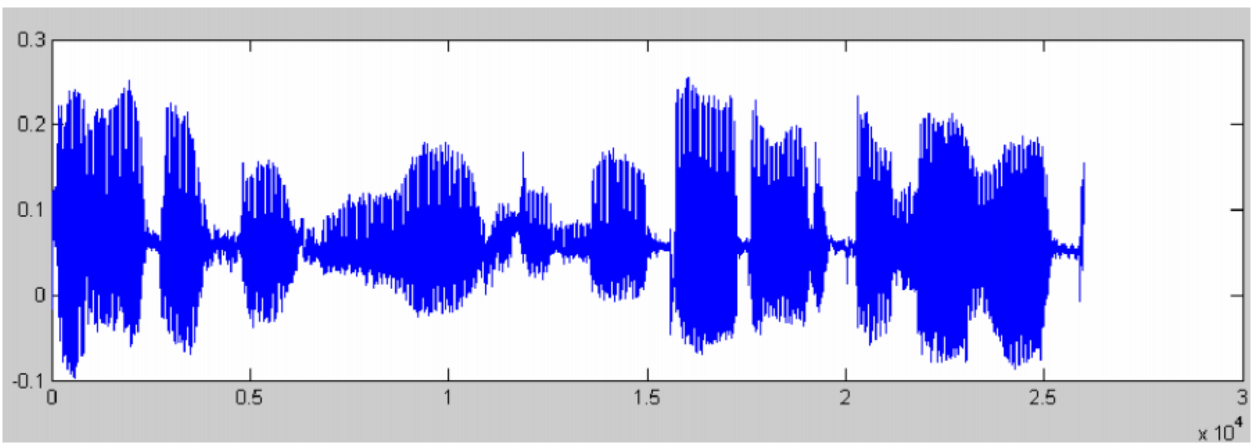

Fig. 3. Waveform of spectral de-correlated noise signal.

The presence of high frequency noise components are removed by means of pre emphasis filter at preprocessing stage. The noise is then estimated from the error signal e(n) and the weights are updated to de-correlate the noise signal form original speech signal. A delay is used for stability of the filter coefficients to reduce howling noise in the aiding system. The frequency response cure for the digital hearing aid system is shown in Fig. 4. It is observed that the hearing aid system gives flat frequency response in the hearing spectrum which makes it ideal for significant gain response characteristics.

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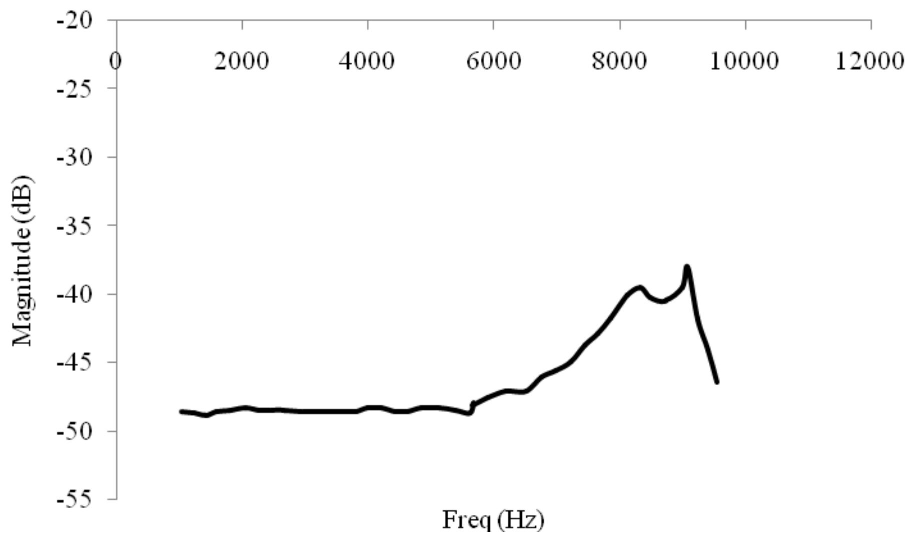
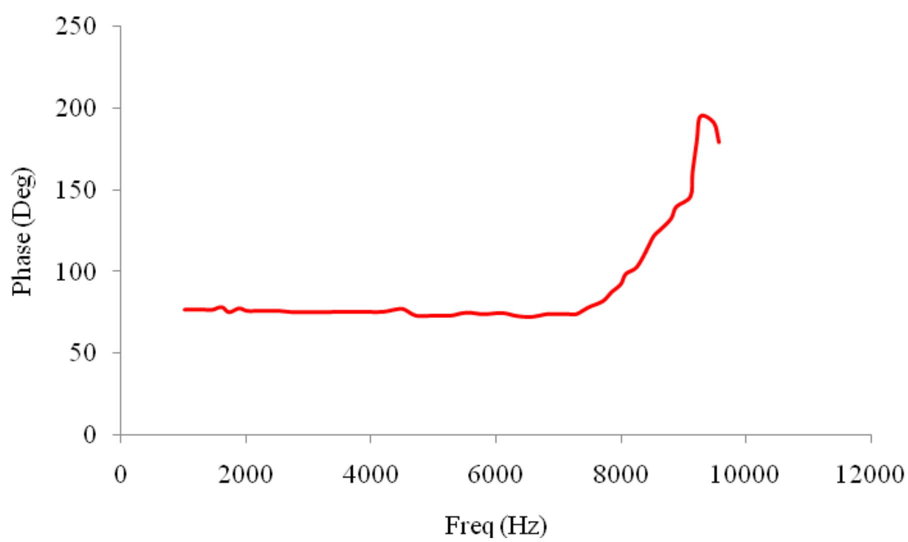


(a)



(b)

Fig. 4. Frequency responseof typical behind the ear hearing aid.

The reference gain curve corresponding to proposed hearing aid system based on SVD optimal filtering technique is shown in Fig. 5. The curve is used to compensate the hearing loos in the hearing frequency spectrum where most of the speech signals are located.

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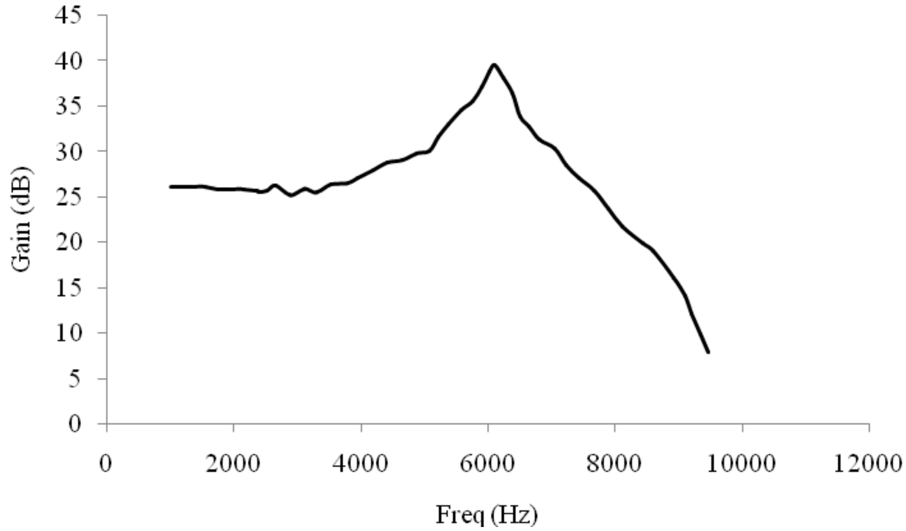

Fig. 5. Reference Gain (dB) curve for compensation of filter response.

Fig. 6 shows signal to noise ratio for proposed digital hearing aid system. It is calculated from average input signal to average noise signal.

$$SNR\ (dB) = \frac{E(C(n))}{E(N(n))} \tag{9}$$

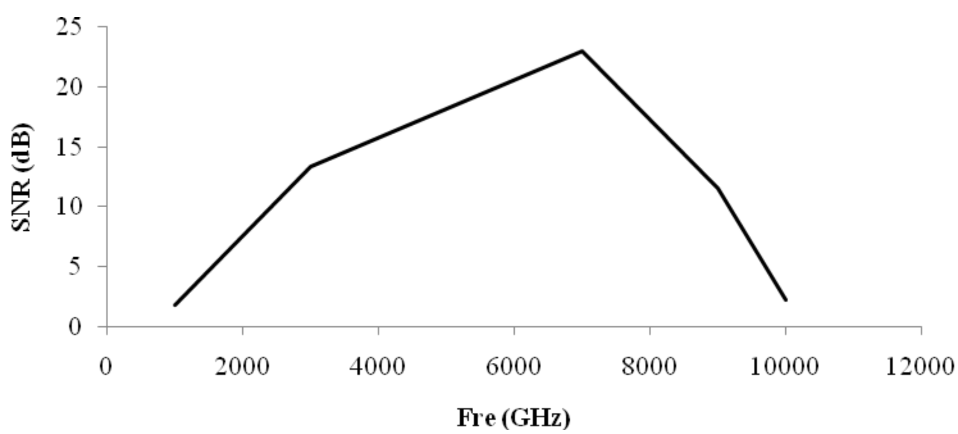

Fig. 6. Signal to noise ratio (dB)

The comparison of performance metrics of the proposed algorithm with other algorithms is given in Table 1. It is inferred from table 1, that the proposed method utilizes minimum number of filter coefficients and attains better signal to noise ratio (SNR) an gain characteristics in the hearing spectrum.

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Table 1. Performance metrics of the proposed algorithm

Parameter	Filter Coefficients	SNR (dB)	Gain (dB)
[10]	256	8.73	6.4
[11]	256	9.8	12.7
[12]	256	17.32	11.6
[13]	256	16.5	10.1
Proposed	128	23.4	16

V. Conclusion

An optimal adaptive filtering model based on singular value decomposition technique is proposed. The model utilizes adaptive weightage whose coefficients are updated based on the weighting function. A delay is used in the filter to increase the stability of filter coefficients. The performance of the proposed model is compared with other conventional model. The algorithm achieves a better signal gain quality of 16dB over the limit gain. The algorithm utilizes adaptive feedback noise reduction filter bank having 128 filter coefficients and attains a significant signal to noise ratio and makes it more appropriate for digital hearing aid systems.

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