14th European Signal Processing Conference (EUSIPCO 2006), Florence, Italy, September 4-8, 2006, copyright by EURASIP

ON THE USE OF LINEAR PREDICTION FOR ACOUSTIC FEEDBACK CANCELLATION IN MULTI-BAND HEARING AIDS

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ABSTRACT

An efficient approach to mitigate the howling effect in hearing aids is via the use of Acoustic Feedback Cancellation (AFC). In this paper, the use of a Forward Linear Predictor (FLP) is investigated to improve the performance of an AFC system in multi-band hearing aids. The FLP is used to predict the speech input signal before eliminating it from the error signal of the AFC system. Computer simulations demonstrate that more accurate estimation of the acoustic feedback signal than the conventional AFC system can be obtained. In addition, maximum usable gain of hearing aids required by the users can be achieved when employing the proposed multi-band AFC system.

1. INTRODUCTION

In order to facilitate the hearing impairment problem, hearing aids are normally employed to amplify the sounds of interest. However, hearing-impaired people usually have frequency-dependent characteristics of hearing loss [1]. Thus, conventional hearing aids, which amplify all the sounds at different frequencies with the same amount of amplification, should not be selected. Multi-band hearing aids, where various amplification gains are applied in different frequency bands, are therefore focused in this paper.

For people with sensorineural hearing loss, In-The-Ear (ITE) hearing aids are normally used. The air vent between the hearing aid device and the ear of the user, however, causes the acoustic feedback path. Therefore, the leakage of the amplified sounds from the output of the device is sent back to its microphone. With high values of the amplification gain, the acoustic feedback signal is perceived as whistling or howling by the hearing aid user. As a result, possible maximum gain of the device is limited in order to avoid the howling effect. Acoustic Feedback Cancellation (AFC), which employs an adaptive filter to estimate the acoustic feedback signal, is therefore necessary for hearing aids. Thus, the required amplification gain of the devices can be adjusted according to the hearing loss of the users.

The AFC system is usually categorized into two types; non-continuous and continuous adaptation of the adaptive filters. For the AFC system with non-continuous adaptation [2, 3], the acoustic feedback signal is eliminated solely when the howling effect is detected. Since the training sequence, such as white noise, of the adaptive filter can be perceived by the hearing-aid users, this type of AFC system is only suitable for the people with severe hearing loss.

On the other hand, the AFC system with continuous adaptation [4] – [7], where the adaptive filter continuously identifies the acoustic feedback path, is preferred. However, misconvergence of the adaptive filter occurs. Two main factors of the misconvergence problem are due to the correlation between the input and the output signals of hearing aids and the existence of the speech input energy within the error signal. To sufficiently decorrelate the input and the output signals, a fixed delay of at least 1 *ms* is suggested to be employed in the forward path of the AFC system [6]. To eliminate the speech input signal from the error signal of the AFC system, it is proposed in [8] to use a Forward Linear Predictor (FLP) for speech input signal estimation. The predicted speech signal is then removed from the error signal before sending to the adaptation process. Alternatively, a technique to estimate the inverse of the speech signal model is proposed in [9]. These approaches result in improved performance of the adaptive filter in the AFC system.

When the hearing-aid user wear the device in one ear, the output signal arriving the impaired ear will be delayed, as compared to the signal arriving at the other ear. In order not to destroy the stereo perception of the signals, the inter-aural delay, which is the difference between the group delay of the left-channel and the right-channel signals, should be under the limit of 200 μs [8, 10]. The use of a fixed delay of 1 *ms* to the signal in the forward path will, however, result in fixed inter-aural delay that exceeds the 200 μs limit and thus, will lead to the degradation of the stereo signals. Therefore, in this paper, the use of FLP in [8] without any fixed delay in the forward path is investigated to be employed with the AFC system in multi-band hearing aids. This will be compared to the AFC systems; with and without fixed delay in the forward path, in conventional hearing aids.

This paper is organized as follows. Section II describes the AFC system in hearing aids. In Section III, the use of FLP in the multi-band AFC system are presented, followed by simulation results based on a real speech signal and real database of hearing-impaired people. Finally, the conclusions are given in Section V.

2. THE AFC SYSTEM

A block diagram of the conventional AFC system for hearing aids is illustrated in Fig. 1. The input and the output signals of the system are denoted by $x(n)$ and $s(n)$ respectively. The impulse response of the acoustic feedback path is represented by $\mathbf{f}(n)$. The microphone signal, $d(n)$, contains both the input signal, $x(n)$, and the feedback signal, $y(n)$. The forward path of the hearing aid, $\mathbf{g}(n)$, is represented by the amplification gain, G_0, of the device. Therefore, the feedback signal, $y(n)$, is sent to the ear canal of the HA user via the output signal, $s(n)$. A Finite Impulse Response (FIR) adaptive filter of length L, $\mathbf{w}(n)=[w_0(n), w_1(n), \ldots, w_{L-1}(n)]^T$, is used to estimate the feedback path, $\mathbf{f}(n)$. The error signal, which is the difference between the microphone signal and the estimated feedback signal, $\hat{y}(n)=\mathbf{w}^T(n)\mathbf{s}(n)$, is given by

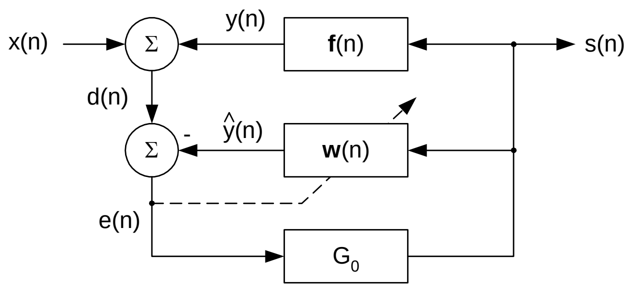


Figure 1: Acoustic feedback cancellation with continuous adaptation.

$$e(n) \;=\; d(n)-\hat{y}(n) \qquad (1)$$

$$\;=\; x(n)+[\mathbf{f}(n)-\mathbf{w}(n)]^T\mathbf{s}(n) \qquad (2)$$

Where the output signal vector of length L is given by $\mathbf{s}(n)=[s(n), s(n-1), \ldots, s(n-L+1)]^T$. Once the adaptive filter can identify the feedback path correctly, it can be seen from Eq.(2) that the error signal, $e(n)$, approaches the speech input signal, $x(n)$. According to linear optimal filtering that chooses to minimize the mean-square value of the error signal [9], the existence of $x(n)$ within $e(n)$, however, makes the adaptive filter not converge to the acoustic feedback path. Hence, the acoustic feedback signal cannot be efficiently eliminated.

3. THE PROPOSED AFC SYSTEM

One of the main reason that the adaptive filter does not converge to the true solution, or diverges from its steady state, is because the error signal, $e(n)$, becomes enormous, i.e. contains the speech input signal, $x(n)$, as shown in the previous subsection. In order to make the error signal, $e(n)$, to be minimum in the mean-square sense, it is suggested in [8] that the speech input signal, $x(n)$, should be estimated by employing the FLP. In this section, Recursive Least Square (RLS)-type FLP in [8] is summarized in brief. Then the hearing loss characteristics of two study cases are given, followed by the calculation of the amplification gain for multi-band hearing aids. Finally, the use of FLP in the proposed multi-band AFC system is presented.

3.1. The forward linear predictor

The error signal, $e(n)$, is used as the input signal to the FLP part, as showin in Fig. 2. With the prediction order of M, the estimation of input signal is given by

$$\hat{e}(n) \;=\; \sum_{j=1}^{M} w_f(j)e(n-j) \qquad (3)$$

where $\mathbf{w}_f^M(n)=[w_{f,1}(n), w_{f,2}(n), \ldots, w_{f,M}(n)]^T$ denotes the prediction coefficient vector of FLP. The forward prediction error is defined as

$$e_1(n) \;=\; e(n)-\hat{e}(n) \qquad (4)$$

The update equation of the prediction coefficients is given by

$$\mathbf{w}_f^M(n+1) \;=\; \mathbf{w}_f^M(n)+\mathbf{k}(n)e_1(n) \qquad (5)$$

The Kalman gain vector is obtained from

$$\mathbf{k}(n) \;=\; \frac{\boldsymbol{\pi}(n)}{\lambda+\mathbf{u}^T(n-1)\boldsymbol{\pi}(n)} \qquad (6)$$

where $\mathbf{u}(n)=[u(n), \ldots, u(n-M+1)]^T$ is the vector of length M of the delayed input signal of FLP and

$$\boldsymbol{\pi}(n) \;=\; \mathbf{P}(n-1)\mathbf{u}(n) \qquad (7)$$

The inverse of the autocorrelation matrix of the signal $u(n)$, $\mathbf{P}(n)$, can be found recursively as

$$\mathbf{P}(n) \;=\; \lambda^{-1}\left(\mathbf{I}_M-\mathbf{k}(n)\mathbf{u}^T(n)\right)\mathbf{P}(n-1) \qquad (8)$$

where $0<<\lambda<1$ is the forgetting factor and $\mathbf{I}_M$ is an $\mathrm{M}\times\mathrm{M}$ identity matrix. When the predicted speech signal, $\hat{e}(n)$, is removed from the error signal, $e(n)$, the new error signal, $e_1(n)$, is subsequently used for the adaptation process of the adaptive filter, $\mathbf{w}(n)$. The desired speech signal, $x(n)$, contained in the error signal, $e(n)$, is sent to the forward path of the hearing aid for amplification, as required.

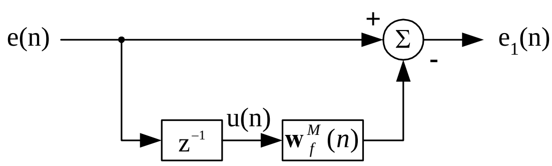


Figure 2: A Forward Linear Predictor.

3.2. Hearing loss characteristics

Two different hearing characteristics, collected from two patients at Chulalongkorn hospital, Bangkok; one with high-frequency hearing loss and the other with moderate-to-severe hearing loss, are shown in the audiogram in Fig. 3. The hearing test used a pure tone that automatically sweeps in frequency over the desired frequency range as measurement signals [11]. The hearing sensitivity of the patients is illustrated in minimum hearing threshold level of pure tone signals perceived for different frequencies, which is called Sound Pressure Level (SPL), measured in dB. The normal hearing threshold, obtained from a person with normal hearing, is also given in Fig. 3 as a reference [12].

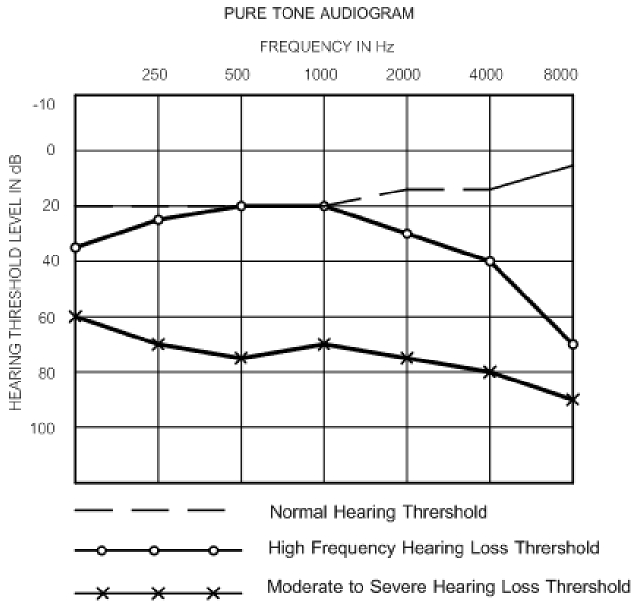


Figure 3: Pure tone audiogram indicating hearing threshold of two patients with different hearing thresholds.

3.3. Amplification gain

Table1: Insertion gain used for two types of hearing loss.

Frequency Range of Input Signal (Hz)	Types of hearing loss	
	High-frequency hearing loss (dB)	Moderate-to-severe hearing loss (dB)
0 - 1k	8	25
1k - 2k	5	28
2k - 4k	10	31
4k - 8k	25	37

To restore normal loudness perception, the gain needed at each frequency for the amplification of the input signal is equal to the threshold loss, which is equal to the difference between the measured SPL characteristics and the normal hearing threshold at any particular frequency. Due to the variation of speech energy across frequencies, the half-gain rule [11] is applied in this paper in order to avoid excessive loudness, i.e. the gain chosen at each frequency is approximately half of the amount of the previously calculated gain. For any selected frequency subband, the average of calculated gains within that band of interest is obtained. Table 1 shows the insertion gains (in dB) according to the audiogram of two patients in Fig. 2. It can be seen that higher levels of amplification gain are required in the high-frequency ranges than in the low-frequency ones.

3.4. The Proposed Multi-band AFC System

According to the hearing loss characteristics, hearing aids should therefore be divided into a number of subbands so that different amplification gains can be applied accordingly.

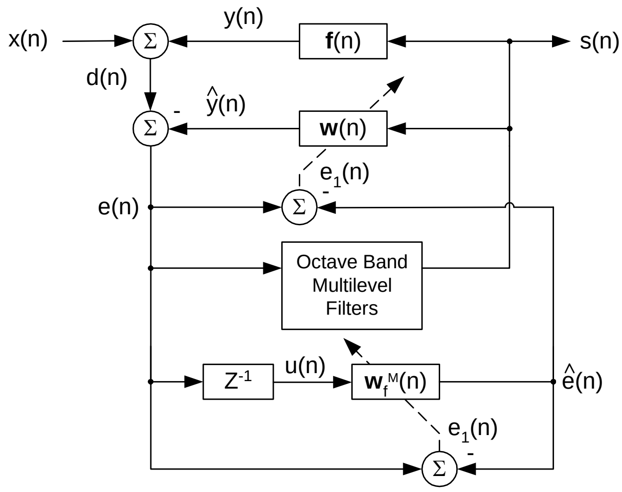


Figure 4: The proposed multi-band AFC system.

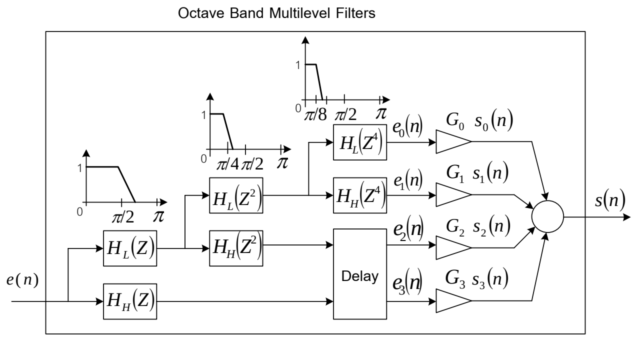


Figure 5: Octave Band Multilevel Filters.

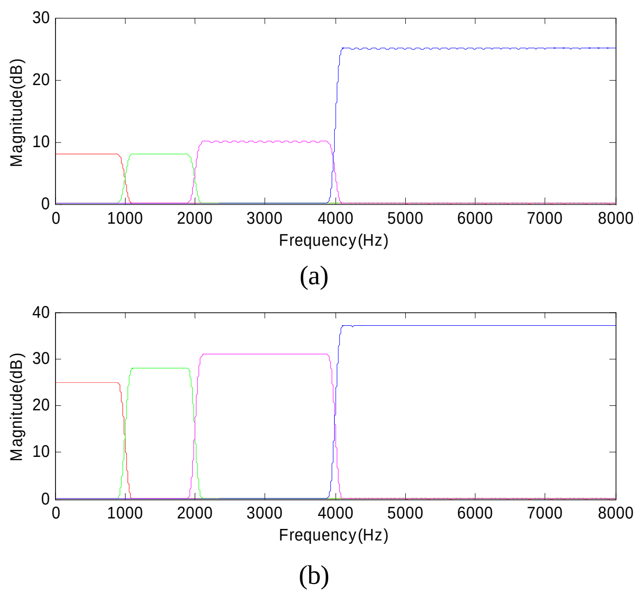


Figure 6: Frequency response of the Octave Band Multilevel Filters for (a) high-frequency hearing loss and (b) moderate-to-severe hearing loss.

A block diagram of the proposed AFC system for multi-band hearing aids, with the use of FLP, is illustrated in Fig. 4. The number of subbands is chosen to be M = 4 due to the frequency range of the hearing test, as given in Table 1.

The Octave Band Multilevel Filters [13] is employed for splitting the error signal, $e(n)$, into M subbands. A single prototype FIR halfband filter is used to generate the filter bank for all subbands via the tree structure, as shown in Fig. 5. With different amplification gains, G_i, according to Table 1, the frequency responses of the Octave Band Multilevel Filters for high-frequency hearing loss and moderate-to-severe hearing loss are depicted in Fig. 6 (a) and Fig. 6 (b), respectively. Subsequently, these subband signals, $s_i(n)$, are summed up to obtain the amplified signal, $s(n)$.

4. SIMULATION RESULTS

A speech signal with sampling rate of 16 *kHz*, normalised to have unity variance, as depicted in Fig. 7, was used as the input signal, $x(n)$, of the hearing aids. In all simulations, the AFC system employed the Normalised Least Mean Square (NLMS) algorithm with step-size $\mu = 0.03$ and $\mu = 0.0055$ for patients with high-frequency hearing loss and moderate-to-severe hearing loss, respectively. The acoustic feedback path was assumed to be time-invariant, **f,** and modelled as shown in Fig. 8. The performance of the AFC system was evaluated via the Weight Error Vector Norm (WEVN), which is given by

$$\mathrm{WEVN}(n) \quad = \quad 10 \times \log_{10} \frac{\left\| \mathbf{f} - \mathbf{w}(n) \right\|^2}{\left\| \mathbf{f} \right\|^2} \tag{9}$$

where $\| . \|$ denotes the Euclidean norm of a vector.

For comparison, the amplification gains in the conventional AFC system were chosen to be the average of those gains in all frequency ranges, i.e. 12 dB for the patients with high-frequency hearing loss and 30 dB for those with moderate-to-severe hearing loss. The conventional AFC systems with and without the use of FLP were compared with the proposed multi-band AFC system. For high-frequency hearing loss, it is shown in Fig. 9 (a) that the WEVN performance when using the proposed technique reaches about –7 dB, which is far better than those of the conventional AFC systems that diverge from their steady state. In the case of moderate-to-severe hearing loss, Fig. 9 (b) illustrates that the proposed AFC system gives much superior WEVN performance than the conventional ones that become divergent either with or without the use of FLP. This can be explained that the feedback signal of the multi-band AFC system is much less than that of the conventional one, as shown in Fig. 10.

For each case, the amplification gain was increased until its maximum value was reached and the output signal, $s(n)$, did not cause any discomfort to the listeners performing listening tests. It can be seen from Fig. 11 that, the multi-band hearing aid scheme enables the peak gain to reach 45 dB for the patient with high-frequency hearing loss and 42 dB for that with moderate-to-severe hearing loss. These are 10 dB higher than the achievable gain of the conventional hearing aid. As a result, this demonstrates that maximum usable gain of hearing aids required by the users can be achieved when employing the proposed multi-band AFC system, whereas lower level of the amplification gain can be applied when employing the conventional AFC system.

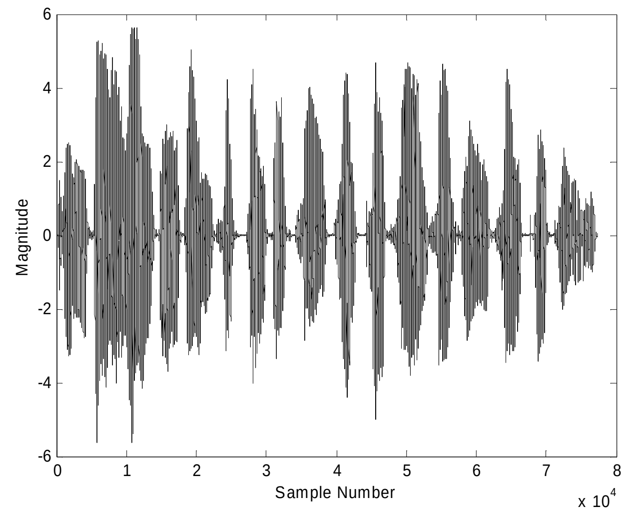


Figure 7: Speech input signal.

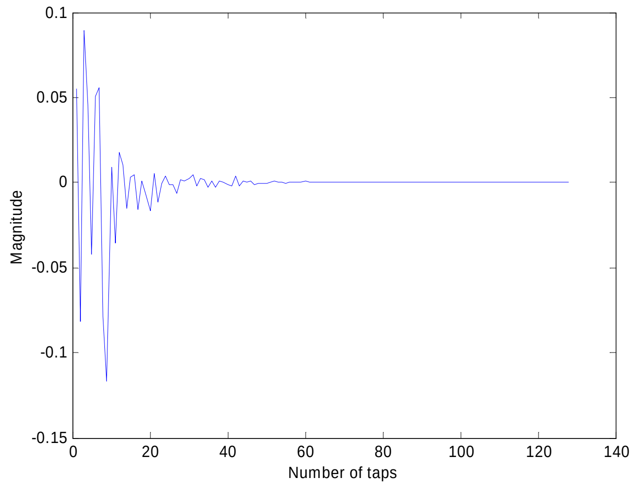


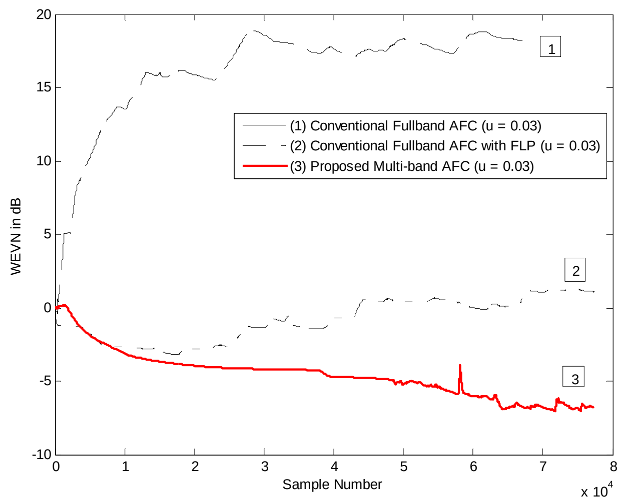
Figure 8: Impulse response of the acoustic feedback path.

5. CONCLUSIONS

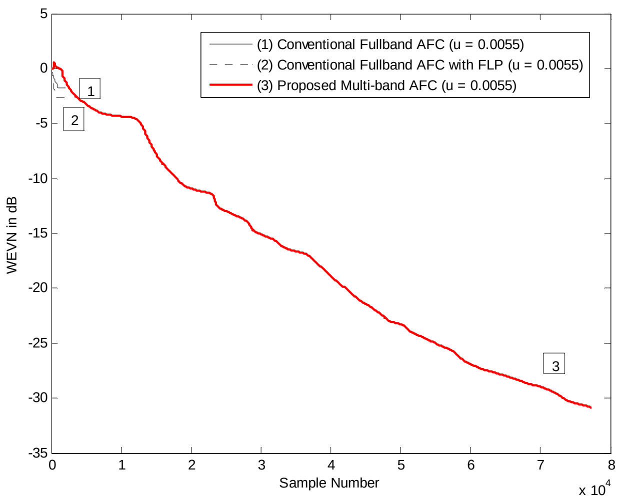
The proposed AFC system for multi-band hearing aids, using FLP has been suggested in this paper. It has been demonstrated that improved performance of the proposed AFC system can be achieved, as compared to the conventional hearing aids, with and without fixed delay in the forward path. By using the proposed multi-band AFC system, with half-gain rule, maximum usable gain of hearing aids to fit the requirements of the patients can be obtained.

6. ACKNOWLEDGMENTS

This work has been supported by Ratchadapiseksompoj Endowment Chulalongkorn University, Center of Excellence in Electrical Power Technology Chulalongkorn University, and the Cooperation Project between the Department of Electrical Engineering and Private Sector for Research and Development, Chulalongkorn University, Thailand. The second and the last authors are indebted to Japan Society for the Promotion of Sciences for their support in the JSPS Invitation Fellowship Program.



(a)



(b)

Figure 9: Comparison of WEVN performance of the AFC systems with (a) high-frequency hearing loss and (b) moderate-to-severe hearing loss.

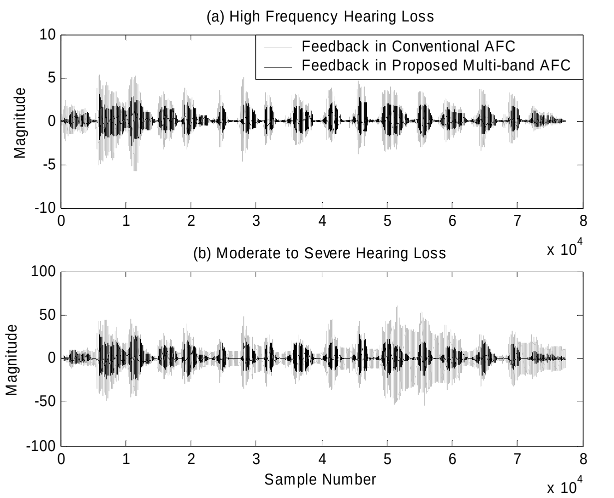


Figure 10: Feedback signal, $y(n)$, of the AFC system for (a) high-frequency hearing loss and (b) moderate-to-severe hearing loss.

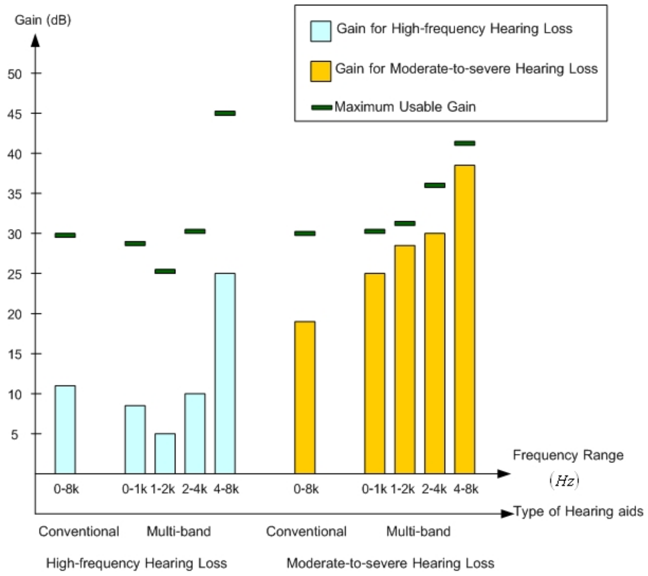


Figure 11: Comparison of maximum amplification gains.

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