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An Adaptive Occlusion Canceller for Hearing Aids

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ABSTRACT

This paper presents an adaptive active noise-control system to reduce the occlusion effect in unvented hearing aids. The proposed system uses a secondary microphone facing inside the ear canal to compare the real acoustic field with the desired one, generating an anti-phase estimate of the occlusion signal. As a result, there is a reduction in the discomfort associated with the occlusion effect. This feature is of particular interest to hearing aid users with severe to profound deafness. Results from objective and subjective experiments by using a real prototype with volunteers are presented, showing the functionality and performance of the proposed architecture.

***Index Terms*—** Occlusion, Hearing Aid, Feedback, Evaluation

1. INTRODUCTION

Lack of hearing is a serious deficiency that impairs social interaction and learning of the individual. Hearing aids are important in reducing these effects (by restoration of the sound quality) but are far from a perfect solution [1]. A major problem associated with users who require high amplification levels is the acoustic feedback [2].

Acoustic feedback results from the acoustic coupling between the microphone and loudspeaker of the hearing aid through the ventilation opening (vent). It is perceived by the user as an unpleasant sound (whistling). The natural solution to this problem would be narrowing the vent (in order to minimize the acoustical transmission from loudspeaker to microphone). However, this leads to another problem known as the occlusion effect.

The occlusion effect, first described by Zwislocki in 1953 [3], is caused by the confinement of acoustical energy (conducted through the skull and jaw) within the ear canal when its cartilaginous portion is completely or significantly occluded by the hearing aid (since placement in the bony portion causes physical discomfort). When the user speaks or chews, vibrations are produced in the cartilaginous portions of the ear canal, which will act as an elastic membrane, leading the user to perceive a muffled version of his voice. This occurs due to a power increase at low frequencies (predominantly in the range of 200 to 500 Hz) [4]. The complete closure of the ear canal may produce an undesired increase of 20 to 30 dB in low-frequency sounds [2]. That is, the narrowing in ventilation openings reduces the acoustic feedback, but favours the onset of the occlusion effect and vice-versa. Although there are many important situations where vent minimization is required, only a reduced number of works on the occlusion effect have been published in the scientific literature.

The first realistic attempt to minimize the occlusion effect through an active noise cancellation system was proposed in [4]. In this work, a fixed feedback controller was embedded in an unvented in-the-ear (ITE) hearing aid by the inclusion of a secondary microphone fitted in the ear mould facing the ear canal. Despite the promising results obtained, the authors reported performance limitations due to the use of a fixed controller. This occurs due to variations of the acoustic system, which are directly related to the characteristics of the speaking process and the maintenance of the conditions found in the fitting process.

This paper presents an adaptive active noise-control system to reduce the occlusion effect. Its adaptive structure is appropriated to deal with dynamic changes of the acoustic channel, avoiding complex fitting processes and preventing performance losses due to ear canal variations or ear mould displacements.

2. MATERIALS AND METHODS

A microphone placed inside the ear canal allows the use of active control strategies for generating a quiet zone (relative to the occluded signal) in the vicinity of the microphone or across the whole ear canal (depending on the acoustic wavelengths involved).

The error signal to be minimized in this approach is the difference between the desired signal (external microphone signal compensated by the hearing aids processing) and the signal that actually reaches the eardrum of the user (measured by the internal microphone). Assuming the sound field inside the ear canal does not significantly vary spatially in relation to the near field of the tympanic membrane (at least for frequencies below 2 kHz) [5] the occlusion effect can be reduced.

2.1. Proposed structure

The block diagram of the proposed structure is shown in Figure 1. Signal $x(t)$ is the speech excitation or sound generated by the hearing aid user with t being the continuous time. F is the transfer function associated to the vocal system, while H represents internal body filtering transformations of $u(t)$ due to the transmission of the user's speech through bones and cartilage to the ear canal. $r(t)$ is the occlusion signal, and $\eta(t)$ the measurement noise. The G system represents the compensation performed by the hearing aid, with output signal $s(n)$, and n being the discrete-time. $d(n)$ is the signal applied to the loudspeaker and $m(n)$ is the signal sampled by the internal microphone located inside the ear canal. The S_1 and S_2 blocks represent the filtering processes related to the digital-to-analog conversion (e.g. reconstruction filter, pre-amplifier, loudspeaker) and the analog-to-digital conversion (e.g. microphone, pre-amplifier, anti-aliasing filter), respectively. The influence of the acoustic path is considered fixed and is embedded in S_1. The block $\hat{S}$ is an estimate of the direct path $S = S_1 * S_2$ obtained during the fitting stage. It provides a filtered version of $s(n)$ and $m(n)$ to compensate the effects of S_1 and S_2 in the adaptation process.

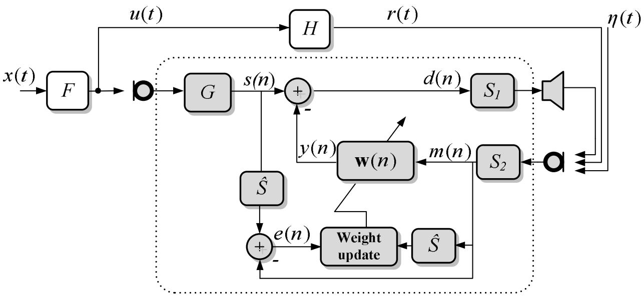

Figure 1. Proposed structure.

The adaptive filter $\mathbf{w}(n) = [\ w_0(n)\ w_1(n)\ \ldots\ w_{N\text{-}1}(n)\ ]^T$ generates an instantaneous estimate of the occluded signal $y(n) = \hat{r}(n)$ by minimization of the instantaneous power of the error signal $e(n)$ given by the difference between the hearing aid processed speech (filtered by $\hat{S}$) and the signal that reaches the hearing aid user's tympanic membrane $m(n)$. Variations of H will be followed by a $\mathbf{w}(n)$ tracking process.

2.2. Direct path

The sound signal changes imposed by the direct path ($S = S_1 * S_2$) can severely affect the performance of the cancellation system. Equalization can be used for minimizing such problem (Figure 2).

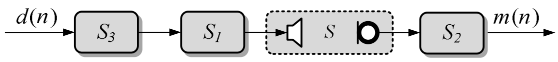

Figure 2. Compensation of the direct path influence.

In this work, as will be justified later, it is assumed that the convolution between the direct path and the equalizer (S_3) can be approximated by a delay of Δ samples, so that $S(\omega) = S_1(\omega)\cdot S_2(\omega)\cdot S_3(\omega) = e^{-j\omega\Delta}$. This will be called as compensated direct path.

2.3. Adaptive Algorithm

Assuming that the direct path can be adequately compensated in order to result in a propagation delay of Δ samples, we obtain:

$$m(n) = d(n-\Delta) + r_f(n) + \eta_f(n) \qquad (1)$$

$$d(n) = s(n) - y(n) \qquad (2)$$

$$y(n) = \sum_{j=0}^{N-1} w_j(n) m(n-j) \qquad (3)$$

where $\eta_f(n) = \sum_{i=0}^{M\text{-}1} S_{2i}\eta(n\text{-}i)$ and $r_f(n) = \sum_{i=0}^{M\text{-}1} S_{2i} r(n\text{-}i)$. By substituting (3) in (2) and the result in (1), one obtains:

$$\begin{aligned} m(n) &= -\sum_{j=0}^{N-1} w_j(n-\Delta) m(n-\Delta-j) \\ &\quad + s(n-\Delta) + r_f(n) + \eta_f(n) \end{aligned} \qquad (4)$$

Assuming that the adaptation process is slow and defining $z(n) = s(n\text{-}\Delta) + r_f(n) + \eta_f(n)$, Eq. (4) may be represented by the block diagram shown in Figure 3, and it can be interpreted as a problem of predicting/estimating $s(n\text{-}\hat{\Delta})$. As a result, the coefficient update problem can be analysed in the scope of the infinite impulse response adaptive filter theory, constrained to a limited set of non-null coefficients. It is important to consider that, as a result of the existing feedback, the performance surface may present multiple minima that are not necessarily global.

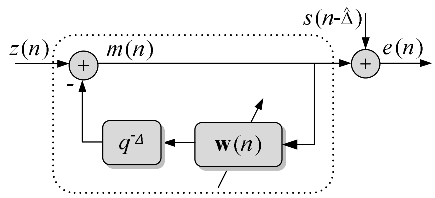

Figure 3. Representation of the proposed structure for slow adaptation [6].

Several update strategies for adaptive filters are available in the literature. Although the algorithms based on stochastic gradient methods, such as the LMS [7] are among the favourites, in feedback structures they lose much of their appeal against the computational cost due to the complexity required to estimate the gradient [8].

Feintuch in 1976 presented a strategy for adapting a recursive filter with very low computational cost [9]. Although a subsequent series of technical notes have disputed the claim that this algorithm would not be able to minimize the mean square error in all situations, the presented results are very promising and cannot be ignored. Its major limitation, however, is the fragility of stability, which has limited its application due to the lack of a deeper theoretical support.

One possibility to alleviate the stability problem, maintaining the low-cost features of [9], is the Simple Hyperstable Adaptive Recursive Filter (SHARF) [10]. The proposed structure can be mapped to the SHARF framework for $S_1*S_2*S_3 = z^{-\Delta}$, small μ, and $e(n)$ substituted by a filtered version of it ($\sum_{i=0}^{P-1} c_i \cdot e(n-i)$, where c_i is the i^{th} coefficient of the SHARF smoothing filter). Due to the recursive nature of the proposed structure (Figure 3) and the speech nonstationarity, there are no strict theoretical guidelines for optimal design of μ and c_i. As a result, these parameters should be chosen parsimoniously. Strategies for polarization of the coefficients (to reduce the open-loop gain of the feedback system) could be used to enhance the stability region. One example is the *Leaky*-LMS algorithm [11], that uses a leakage factor (γ) to control the norm of the coefficient vector, increasing the gain margin at a cost of a tolerable performance loss. The combination of the SHARF and *Leaky* strategies results in the following update strategy

$$\mathbf{w}(n+1) = (1-\mu\gamma)\mathbf{w}(n) + \frac{\mu e(n)\mathbf{m}(n-\Delta)}{\mathbf{m}^T(n-\Delta)\mathbf{m}(n-\Delta)}, \quad (5)$$

where the denominator perform an automatic gain control of the instantaneous power of the signal captured by the internal microphone (it is necessary due to the intrinsic variability of speech).

2.4. Constructive aspects

The proposed controller system was implemented in the development board ADSP-BF537 EZ-KIT *Lite* from Analog Devices. The system was configured to operate at a rate of 16 kHz. This sampling frequency is sufficient to cover the range of hearing loss related to speech intelligibility.

An active ear mould was built from a commercial earphone (assembled with silicone), as shown in Figure 4. Two electret microphones type EM-24046-000 Knowles Electronics and a loudspeaker type ED-27305-000 from the same manufacturer (all specific for using in hearing aids applications) were adapted inside the ear mould. One microphone was placed facing outside, while the other was facing the inside portion of the ear canal together with the loudspeaker. The earphone can be inserted into the ear canal of different users, blocking the ear cavity completely.

Figure 4. Active ear mould.

2.5. Direct path compensation

The impulse response of the equalizer (used for compensation of the direct path) was obtained by an adaptive inverse identification process [12], as shown in Figure 5. The excitation signal $\varepsilon(n)$ was a Gaussian white noise process. The delay of δ samples was necessary to guarantee the causality of $\mathbf{v}(n)$, in case of a non-minimum phase direct path [12].

The ear mould showed in Figure 4, was inserted into the volunteer's ear and the excitation signal was applied by the loudspeaker into the ear canal. The signal captured by the internal microphone was processed according [12] in order to obtain the inverse of the direct path.

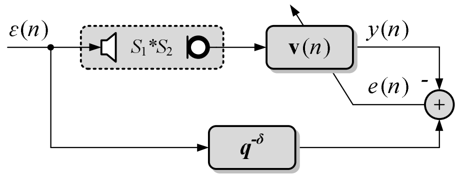

Figure 5. Inverse modelling of the direct path.

The parameters used in the identification process were $N = 100$ coefficients, step-size $\mu = 0.1$ and a delay of $\delta = 50$ samples. In a real situation, this procedure would be performed during the fitting process of the hearing aids. The equalizer consists of a finite impulse response filter whose coefficients correspond to $\mathbf{v}(n)$ after assuring its convergence.

Figure 6 presents the magnitude frequency response of the direct path and its estimated inverse (equalizer) for one volunteer. Their convolution is presented in the same figure. In the time domain, the resulting compensated direct path can be approximated by an impulse shifted by 50 samples.

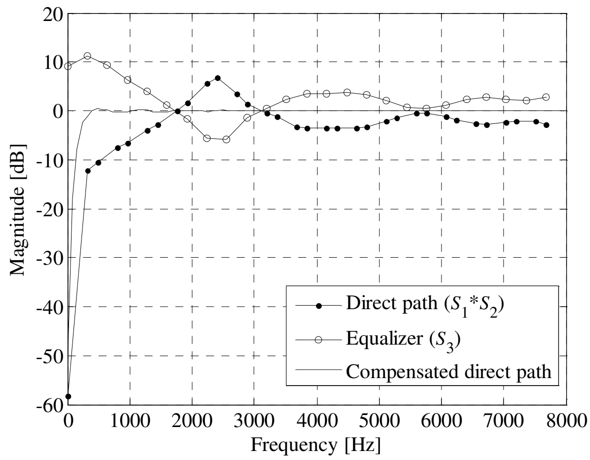

Figure 6. Frequency response of the direct path, equalizer and their convolution.

The analysis of several direct path frequency responses showed a large similarity with the loudspeaker frequency response [13]. This suggests that the loudspeaker is the most important factor in the filtering operations associated with the direct path. As a result, the following approximations can be performed: $\eta_f(n) = \eta(n)$ and $r_f(n) = r(n)$.

2.6. Experiments

The proposed system was evaluated by two different experiments (approved by the Ethics Committee of Universidade Federal de Santa Catarina, processes 2358 and FR476756) with volunteers, using the developed prototype in the platform ADSP-BF537 EZ-KIT *Lite*. Prior to each experiment, the inverse modelling procedure to compensate for the direct path influence was performed. After identification, the ear mould was removed and replaced, in order to simulate daily changes in the acoustic path.

In the first experiment the ability of the proposed system to cancel simple acoustic signals within the ear canal of five volunteers (no complaints regarding hearing problems) was evaluated quantitatively by a procedure similar to the bone conduction tonal audiometry [14]. For this purpose, a bone vibrator was positioned on the mastoid of each volunteer, in order to produce an acoustic field in the ear canal (due to mechanical vibrations in the cartilaginous portions). Four pure tones with frequencies 500 Hz, 1 kHz, 2 kHz and 4 kHz were used as excitation, and individually applied in four different realizations of the same procedure for each volunteer. The used controller parameters were $\hat{\Delta} = 50$ ($\hat{S} = q^{-\hat{\Delta}}$), $\mu = 0.01$, $\gamma = 0.4$ and $N = 200$ coefficients. The power of $m(n)$, from the internal microphone, was measured for the controller turned on and off.

The second experiment, conducted on 15 volunteers, was divided in two parts. In the first part, each volunteer was asked to perform the locution of the phoneme /*u*/ (chosen by having a higher incidence of occlusion, since its formant is close to 300 Hz [15]) with the controller turned on and off. After that, the volunteers were asked to quantify the sound quality and acoustic comfort based on the result obtained when the controller was off. In the second part of the experiment, each volunteer was asked to pronounce different phonetically balanced sentences (between 3 and 6 seconds), previously selected. The controller was switched on and off. At the end of each experiment volunteers were asked to rate the improvement in the subjective quality (speech quality and acoustical comfort) resulting from the use of the proposed controller using a continuous scale from -5 (worse) to 5 (better).

The used parameters for the second experiment were the same as in the first experiment.

3. RESULTS

This section presents results of the two experiments described in Section 2.6 in order to demonstrate the viability and functionality of the proposed structure. Table 1 shows the results related to the first experiment with the bone vibrator (tonal). Results related to the four selected frequencies are presented for each one of the five volunteers named A to E.

Each value in Table 1 represents the measured attenuation of acoustic power taken up by the internal microphone, with the controller turned on, compared to the value obtained for the controller turned off. A total average attenuation of 6.3 dB was observed without any occurrence of power increasing when the controller is turned on, indicating the effectiveness of the proposed system to reduce the sound field in the ear canal for frequencies covering the range associated with speech intelligibility.

Table 1. Results for the bone conduction tonal audiometry experiment

Frequency(Hz)	**Volunteer / Attenuation (dB)**				
	A	***B***	***C***	***D***	***E***
500 Hz	5.6	7.8	6.5	8	6.2
1 kHz	4.6	7.4	6.8	8.9	5.2
2 kHz	3.6	6.3	7	7.2	6.5
4 kHz	4	6.4	6.1	6.1	6.2

The results of the second experiment are shown in the form of box-and-whisker diagrams where each result is defined as presented in Figure 7. In it, the set of sample values comprised between the lower and upper quartiles (denominates q_1 and q_3 respectively) is represented by a rectangle whose median is indicated by a bar. The sample values are considered outliers when greater than $q_3+\varpi\cdot(q_3-q_1)$ or less than $q_3-\varpi\cdot(q_3-q_1)$, whereas q_1 and q_3 are defined as the percentage values of 25% and 75%, respectively. The variable ϖ is defined as the default value of 1.5 [16] and represents the upper and low extremes, which are not considered outliers. The vertical axis indicates subjective satisfaction for the situation in which the controller is turned on, ranging from -5 (worse) to 5 (better), compared to the sound quality when the controller is turned off. The index zero indicates no difference between both situations (controller turned on and off), and negative values indicate amplification of the occlusion effect or speech distortion.

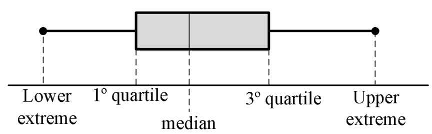

Figure 7. Box and whisker diagram.

Figure 8 shows the results of subjective evaluation for the first part of the second experiment that made use of the locution of the phoneme /*u*/. The sum signal (+) indicates the presence of an outlier. The average of all volunteers resulted in a comparative quality of 2.7.

Figure 9 shows the results obtained in the second part of the second experiment, with full sentences. It shows a significant reduction in the number of outliers. The observed variation among results of different volunteers is likely due to individual subjectivity. The global average resulted in a value of 2.5.

It is important to note that all obtained results were positive, i.e., no volunteer in any experiments reported deterioration in sound quality when the canceller was on.

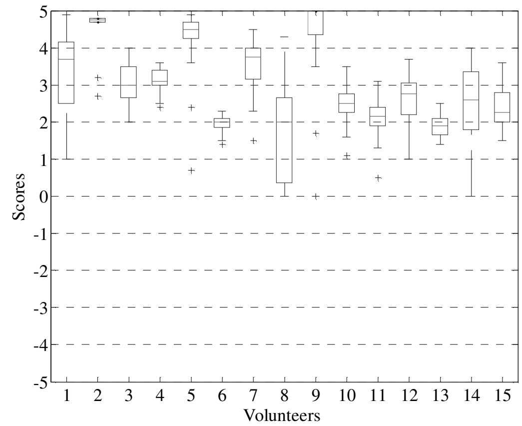

Figure 8. Subjective evaluation for the phonem /u/.

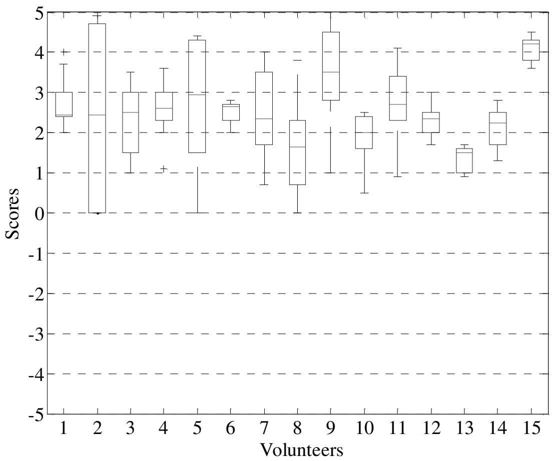

Figure 9. Subjective evaluation for the full sentences.

4. DISCUSSIONS

The fixed controller presented by Mejia (2008) [4] achieved a maximum of 18 dB of power reduction in 300 Hz, but at the cost of amplification in certain ranges (e.g. increase of 3 dB in 1 kHz, and 9 dB in 1.3 kHz). In such architecture it is not possible to guarantee that similar attenuation levels would be obtained after removing and reinserting the ear mould. Additionally, only 10 of 12 subjects reported reduction of the occlusion sensation.

The system proposed in this work showed consistent reduction of the occlusion effect for the four frequencies evaluated with an average reduction of about 6.3 dB of the occlusion signal power. The subjective results indicate that the obtained reduction is sufficient to provide a subjective sensation of "ear opening" (as reported by volunteers), providing adequate acoustic comfort in the limiting case of no ventilation openings.

The variation in the results observed in the diagrams is probably due to the extreme subjectivity involved in the designation of sound quality and with the phonetic content of each sentence, since humans evaluate quality according to their individual perception. The artefacts possibly stems from attention problems and differences in the intonation of the voice of the volunteer during each utterance.

5. CONCLUSION

This paper presented a proposal for an adaptive active noise-control system to reduce the occlusion effect in hearing aids without ventilation opening. The system was implemented in real-time on a digital signal processor.

Objective and subjective experiments were performed showing a significant reduction in the occlusion effect and a consequent improvement in voice quality. These results are of particular interest to hearing aid users with severe to profound deafness.

6. ACKNOWLEDGMENTS

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