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COMBINED ACTIVE NOISE CONTROL AND NOISE REDUCTION IN HEARING AIDS

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ABSTRACT

This paper presents a combined active noise control and noise reduction scheme for hearing aids to tackle secondary path effects and effects of signal leakage through the fitting. While such leakage contributions and the secondary accoustic path from the reciever to the tympanic membrane are usually not taken into account in standard noise reduction systems, they appear to have a non-negligible impact on the final signal quality. Integrating an active noise control system in the existing noise reduction algorithm helps to compensate for these effects. A Filtered-x Multichannel Wiener Filter is presented and compared experimentally with a classic Multichannel Wiener Filter in a noise reduction framework.

Index Terms— Active noise control, Multichannel Wiener Filter, Noise reduction, Hearing aids.

1. INTRODUCTION

The usage of hearing aids with an open fitting has become more common over the past years mainly owing to the availability of more efficient feedback control schemes and fast signal processing units. Whereas removing the earmold reduces the occlusion effect and improves the physical comfort [1], one major drawback is that the signal leakage through the fitting cannot be neglected anymore. Conventionnal noise reduction systems such as the Generalized Sidelobe Canceller (GSC) [2] or techniques based on Multichannel Wiener Filter (MWF) [3] do not take this contribution into account. Combined with the attenuation in the acoustic path between the loudspeaker and the tympanic membrane (secondary path), the noise leaking through the fitting can override the action of the processing done in the hearing aid.

One efficient way to cancel these undesired signal leakage components is to use Active Noise Control (ANC) [4][5]. Filtered-x algorithms are commonly use to perform ANC[6]. They also allow to include the secondary path in the noise reduction computation. Therefore the error signal that has to be

minimized is the difference between the desired signal and the signal reaching the tympanic membrane rather than the signal fed in the loudspeaker (receiver). A combined use of ANC and noise reduction in a Filtered-x Multichannel Wiener Filter (FxMWF) leads to an almost constant signal-to-noise ratio (SNR) at the tympanic membrane.

This paper will present perfomance comparison between a standard MWF noise reduction scheme and a combined ANC and noise reduction using FxMWF, both applied in hearing aids with an open fitting. The effects of the leakage and the secondary path on the output of MWF are commented in Section 2. Section 3 introduces the combined ANC and noise reduction scheme. Experimental results are presented in Section 4 and, finally Section 5 is a summary of the paper.

2. PROBLEM STATEMENT

Speech enhancement in hearing aids is based on standard noise reduction techniques ignoring the possible effects of signal leakage and secondary path. This section describes an algorithm based on MWF [3] and how the noise leaking through the fitting and the attenuation in the secondary path can affect its performances.

2.1. Signal model

Let N be the filter length and M the number of channels (microphones). The input signal x_m for channel m has a speech part x_m^s and an additive noise part x_m^n. At the time instant k, the input is:

$$x_m[k] = x_m^s[k] + x_m^n[k] \quad m \in \{1 \dots M\} \qquad (1)$$

The column vector $\mathbf{x}_m[k]$ contains the N last samples of channel m. The stack vector gathering all channels is:

$$\mathbf{x}^T[k] = [\mathbf{x}_1^T[k] \dots \mathbf{x}_M^T[k]] \qquad (2)$$

The optimal Wiener filter $\mathbf{w}[k]^T = [\mathbf{w}_1^T[k] \dots \mathbf{w}_M^T[k]]$ is given by:

$$\mathbf{w}[k] = \bar{\mathbf{R}}_{xx}^{-1}[k]\bar{\mathbf{r}}_{xd}[k] \qquad (3)$$

where $\bar{\mathbf{R}}_{xx}[k]$ is the estimate of the correlation matrix of the input $\mathbf{x}$ at time k and $\bar{\mathbf{r}}_{xd}[k]$ is the estimate of the cross-correlation vector between the input $\mathbf{x}$ and the desired signal $d[k]$, which is chosen to be equal to the speech component in the first microphone, up to a delay:

$$
\begin{align}
d[k] &= x_1^s[k - \Delta] = s_{ref}[k] \tag{4} \\
\bar{\mathbf{R}}_{xx}[k] &= E\{\mathbf{x}[k]\mathbf{x}^T[k]\} \tag{5} \\
\bar{\mathbf{r}}_{xd}[k] &= E\{\mathbf{x}[k]d[k]\} \tag{6}
\end{align}
$$

Note that by assuming that the speech and noise components of the input signals are uncorrelated the cross-correlation vector can be easily estimated using:

$$
\begin{align}
\bar{\mathbf{r}}_{xd}[k] &= \bar{\mathbf{r}}_{xx_\Delta}[k] - \bar{\mathbf{r}}_{nn_\Delta}[k] \tag{7} \\
\bar{\mathbf{r}}_{xx_\Delta}[k] &= E\{\mathbf{x}[k]x_1[k - \Delta]\} \tag{8} \\
\bar{\mathbf{r}}_{nn_\Delta}[k] &= E\{\mathbf{x}^n[k]x_1^n[k - \Delta]\} \tag{9}
\end{align}
$$

During noise only periods $\bar{\mathbf{r}}_{nn_\Delta}[k]$ can be estimated, $\bar{\mathbf{r}}_{xx_\Delta}[k]$ can be estimated during noisy speech periods.

2.2. Effect of leakage and the secondary path on MWF performance

Changing from an earmold to an open fitting comes with some advantages such as comfort improvements, reduced occlusion effect and risks of infection and sometimes better sound quality [1]. This change also implies that nothing is left to prevent ambient sound from reaching the tympanic membrane [7]. No direct processing can be done on this signal leaking into the ear canal, therefore its SNR is most probably lower than for the signal processed in the hearing aid, where traditionally a noise reduction algorithm is applied.
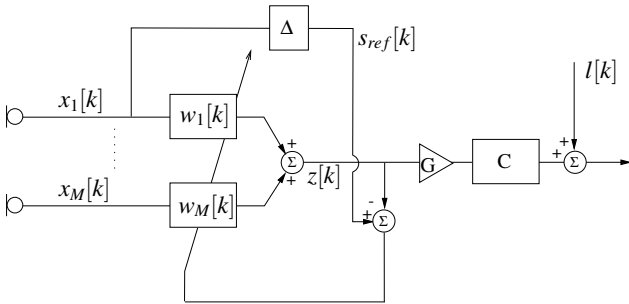

Fig. 1. Multichannel noise reduction system

A standard noise reduction scheme based on MWF is presented in figure 1. The mean square error (MSE) which is minimized corresponds to the error signal:

$$
\begin{align}
e[k] &= d[k] - z[k] \tag{10} \\
z[k] &= \mathbf{w}^T[k]\mathbf{x}[k] \tag{11}
\end{align}
$$

Classic noise reduction schemes ignore the propagation from the loudspeaker to the tympanic membrane (including the receiver response itself). Assuming that the receiver characteristic is approximately linear, the secondary path can be represented by the transfer function C, therefore, including the secondary path, the output is:

$$
out[k] = C * (G \cdot z[k]) \tag{12}
$$

The DC gain of C is lower than 1, so the power of the output is decreased while taking the secondary path into account:

$$
|C * (G \cdot z[k])|^2 \leq |(G \cdot z[k])|^2 \tag{13}
$$

Though the leakage $l[k]$ is not included in the optimization process it is part of the signal reaching the tympanic membrane. Ignoring the secondary path ($C = 1$), the output can be expressed as:

$$
out[k] = G \cdot z[k] + l[k] \tag{14}
$$

It appears clearly that for small amplification gains G the leakage contribution does matter and its SNR may affect the output SNR thus partly cancelling the improvement acheived with the noise reduction in the hearing aid.

From (13) and (14) it comes that taking both leakage signal and secondary path effects into account, the degradation may have even more impact on the output:

$$
out[k] = C * (G \cdot z[k]) + l[k] \tag{15}
$$

In conclusion, whereas secondary path and the leakage are not taken into account in common noise reduction algorithms, they may degrade their perfomances significantly (see also Section 4).

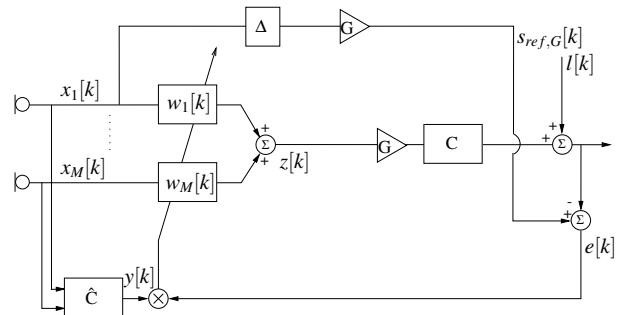## 3. COMBINED ACTIVE NOISE CONTROL AND NOISE REDUCTION

Fig. 2. Multichannel ANC and noise reduction system

The leakage signal is not processed in the hearing aid therefore it is not possible to improve its SNR. It is possible however to attenuate its noise component using ANC. In an

ANC system, the controller output is used to cancel a noise signal and generate a zone of quiet based on destructive interference. In a hearing aid the noise is to be cancelled at the tympanic membrane and so the secondary path plays an important part in the algorithm. Introducing this extra path may lead to instabilities so it is necessary to use Filtered-x algorithms [4][5][6]. Here we present a Filtered-x MWF (FxMWF) based on the estimate of the secondary path:

$$\hat{C}(z) = \sum_{i=0}^{L-1} \hat{c}[i] z^{-i} \qquad (16)$$

The filtered reference signal is:

$$y_m[k] = \sum_{i=0}^{L-1} \hat{c}[i] x_m[k-i] \quad m \in \{1 \dots M\} \quad (17)$$

$$\mathbf{y}_m^T[k] = [y_m[k] \dots y_m[k-N+1]] \qquad (18)$$

$$\mathbf{y}^T[k] = [\mathbf{y}_1^T[k] \dots \mathbf{y}_M^T[k]] \qquad (19)$$

The optimal filter (FxMWF) is:

$$\begin{aligned} \bar{\mathbf{r}}_{yd}[k] &= E\{\mathbf{y}[k]d[k]\} \\ \bar{\mathbf{R}}_{yy}[k] &= E\{\mathbf{y}[k]\mathbf{y}^T[k]\} \\ \mathbf{w}[k] &= \bar{\mathbf{R}}_{yy}^{-1}[k]\bar{\mathbf{r}}_{yd}[k] \end{aligned} \qquad (20)$$

In the case of a simple ANC system with only a noise input ($\mathbf{x} = \mathbf{x}^n$ and $l = l^n$) the desired signal is

$$d[k] = -l^n[k] = -l[k] \qquad (21)$$

The minimized MSE at the time instant k then corresponds to an error signal equal to:

$$e[k] = l[k] + \mathbf{w}^T[k]\mathbf{y}[k] \qquad (22)$$

If the filter $\mathbf{w}$ is adapting slowly:

$$\begin{aligned} e[k] &\approx l[k] + C * [\mathbf{w}^T[k]\mathbf{x}[k]] \\ e[k] &\approx out[k] \end{aligned} \qquad (23)$$

The system is then minizing the sound pressure at the tympanic membrane. For hearing aids systems however, where the aim is to improve the speech-to-noise ratio, the desired signal to be used is a combination of (4) and (21):

$$\begin{aligned} d[k] &= -l^n[k] + G \cdot x_1^s[k - \Delta] \\ d[k] &= -l^n[k] + s_{ref,G}[k] \end{aligned} \qquad (24)$$

The filter described in (20) is then performing a noise reduction combined with ANC, minimizing the noise pressure as well as the difference between the produced signal and the speech reference both at the tympanic membrane. Note once again that by assuming speech and noise components are uncorrelated $\bar{\mathbf{r}}_{yd}[k]$ can be estimated in a similar way as in (7). The system integrating ANC with noise reduction is presented in figure 2.

4. EXPERIMENTAL RESULTS

Both algorithms introduced in Sections 2 and 3 have been tested experimentally and their performance has been compared.

4.1. Experimental setup

The simulations were run on a two-microphones behind-the-ear hearing aid, with a speech source at $0°$ and a multitalker babble noise source at $270°$. The left ear is considered here, facing the noise source. The input SNR is used as a reference measure in standard noise reduction schemes. In our case, as one algorithm also perfoms ANC, the leakage SNR, which can also be considered as the SNR when the hearing aid is turned off, is taken as a reference. The intelligibility-weighted signal-to-noise ratio (SNR) [8] is used here which is defined as

$$\Delta SNR_{intellig} = \sum_i I_i (SNR_{i,out} - SNR_{i,leak}) \qquad (25)$$

where I_i is the band importance function defined in [9] and $SNR_{i,out}$ and $SNR_{i,leak}$ represent the output SNR and the leakage SNR (in dB) of the ith band, respectively.

For a gain G varying from $0dB$ to $20dB$ the inputs are filtered using both of the algorithms previously described. The system is calibrated so that for $G = 0dB$, the leakage and the hearing aid signal have equal power.

4.2. Leakage and secondary path effects, improvements with integrated ANC

To evaluate the effect of the leakage and the secondary path, the input signals are first filtered by a MWF-based noise reduction scheme. Depending on which disturbance is being tested, the signal produced can then be filtered by the secondary path model C and/or the leakage is added, as described in Section 2. The reference SNR (leakage signal), is equal to $-1.3dB$. This value depends on the noise and speech angles as well as input SNR (source signals), which is $5dB$ here.

The degradations induced by the leakage remain small, even for reasonably small gain G (up to $6dB$ (figure 3)) . When introducing both leakage noise and the secondary path the degradation is significant for gains up to $20dB$.

Figure 4 shows that integrating ANC with the noise reduction allows to keep a SNR between $9dB$ and $13dB$ for any gain $G \in [0 \ 20]$ dB. This represents a large improvement compared with a MWF noise reduction algorithm in which leakage and secondary path are not taken into account.
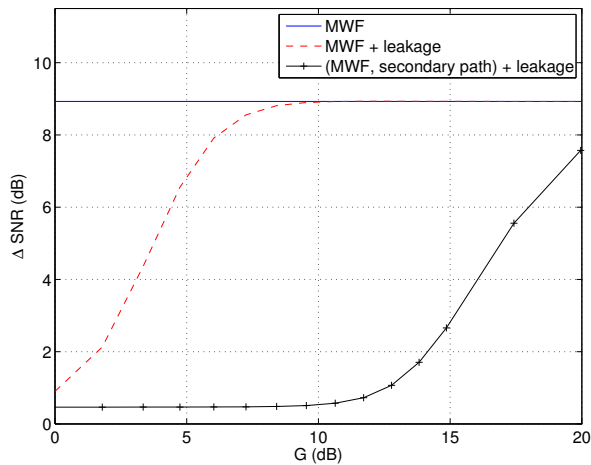
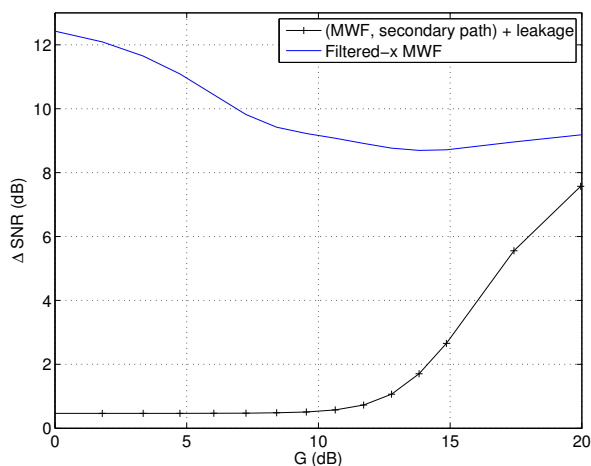**Fig. 3**. Performance comparison for a MWF noise reduction scheme depending on leakage and secondary path

Fig. 4. Performance comparison for noise reduction scheme with or without ANC

5. CONCLUSION

Standard noise reduction techniques used in hearing aids ignore leakage and secondary path effects. With increasing usage of open fitting these aspects cannot be neglected and are in fact seriously degrading noise reduction performances. Integration of ANC with the noise reduction scheme has shown to improve greatly the SNR for low hearing aid gains (between $0 dB$ and $20 dB$), compensating for the noise leaking in the ear canal and taking the secondary path into account in the speech enhancement and amplification process.

6. REFERENCES

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