Integrated Active Noise Control and Noise Reduction in Hearing Aids[1]

Romain Serizel[2], Marc Moonen[2],
Jan Wouters[3] and Søren Holdt Jensen[4]

August 2010
Published in the IEEE Transactions on Audio, Speech and Language Processing
Vol. 18, No. 6, August 2010

[1]This report is available by anonymous ftp from *ftp.esat.kuleuven.be* in the directory *pub/sista/resrizel/reports/09-69.pdf*

[2]K.U.Leuven, Dept. of Electrical Engineering (ESAT), Research group SCD (SISTA) Kasteelpark Arenberg 10, 3001 Leuven, Belgium, Tel. +32 16 32 9607, Fax +32 16 321970, E-mail: *romain.serizel@esat.kuleuven.be*. This research work was carried out at the ESAT laboratory of the Katholieke Universiteit Leuven, in the frame of the Marie-Curie Fellowship EST-SIGNAL program (http://est-signal.i3s.unice.fr) under contract No. MEST-CT-2005-021175, and the Concerted Research Action GOA-AMBioRICS. The scientific responsibility is assumed by its authors.

[3]Katholieke Universiteit Leuven, Department of Neurosciences, ExpORL, O. & N2, Herestraat 49/721, 3000 Leuven, Belgium, E-mail: Jan.Wouters@med.kuleuven.be

[4]Aalborg University, Department of Electronic Systems, MISP, Niels Jernes Vej 12 A6-3, 9220 Aalborg, Denmark, E-mail: shj@es.aau.dk

Abstract

This paper presents combined active noise control and noise reduction schemes for hearing aids to tackle secondary path effects and effects of noise leakage through an open fitting. While such leakage contributions and the secondary acoustic path from the loudspeaker 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-to-noise ratio.

Using a noise reduction algorithm and an active noise control system in cascade may be efficient as long as the causality margin of the system is large enough. Putting the two functional blocks in parallel and then integrating them is found to lead to a more robust algorithm. A Filtered-x Multichannel Wiener Filter is presented and applied to integrate noise reduction and active noise control. The cascaded scheme and the integrated scheme are compared experimentally with a Multichannel Wiener Filter in a classic noise reduction framework without active noise control, where the integrated scheme is found to provide the best performance.

Integrated Active Noise Control and Noise Reduction in Hearing Aids

Romain Serizel, Marc Moonen, *Fellow, IEEE*, Jan Wouters, and Søren Holdt Jensen, *Senior Member, IEEE*

***Abstract*—This paper presents combined active noise control and noise reduction schemes for hearing aids to tackle secondary path effects and effects of noise leakage through an open fitting. While such leakage contributions and the secondary acoustic path from the loudspeaker 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-to-noise ratio. Using a noise-reduction algorithm and an active noise control system in cascade may be efficient as long as the causality margin of the system is large enough. Putting the two functional blocks in parallel and then integrating them is found to lead to a more robust algorithm. A Filtered-x Multichannel Wiener Filter is presented and applied to integrate noise reduction and active noise control. The cascaded scheme and the integrated scheme are compared experimentally with a Multichannel Wiener Filter in a classic noise reduction framework without active noise control, where the integrated scheme is found to provide the best performance.**

***Index Terms*—Active noise control (ANC), hearing aids, multichannel Wiener filter, noise reduction (NR).**

I. 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 noise leakage through the fitting cannot be neglected anymore. Conventional noise reduction (NR) systems such as the Generalized Sidelobe Canceller (GSC) [2] or techniques based on the Multichannel Wiener Filter (MWF) [3] do not take this contribution into account. Combined with the attenuation in the acoustic path between the sound source (hearing aid loudspeaker) and the tympanic membrane (the so-called 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 this undesired noise leakage is to use active noise control (ANC) [4], [5]. The principle of ANC is to generate a zone of quiet, in this case at the tympanic membrane, canceling the effect of noise leakage. To acheive feedforward ANC at the tympanic membrane, it is assumed that, in all the subsequent systems, a microphone is present in the ear canal to provide an error signal. In the hearing aids framework, ANC then has to be performed together with a NR algorithm. There are different ways of combining ANC and NR. Here, the cascading of both functional blocks will be considered first and then the integration of ANC and NR into one filter set will be described, based on an initial parallel combination of the functional blocks and a Filtered-x version of the MWF algorithm (FxMWF) [6].

In a cascaded implementation of the standard NR scheme with a single-channel ANC algorithm, the output of the NR, which is supposed to have a low power noise component, is used as the input of the ANC to produce the so-called anti-noise. The two functional blocks have opposite targets. Therefore cascading NR and single-channel ANC is found to be inefficient.

Using a multichannel ANC instead of a single-channel ANC allows to have input signals with higher power noise components which improve the performance. However, the delay needed to achieve a high NR performance is still added to the system latency. In ANC algorithms, this delay is a critical parameter and can reduce drastically the noise cancellation capabilities [4]. Therefore, ANC used in cascade with a standard NR scheme can only compensate for noise leakage as long as causality margins are sufficient to include the NR delay and still allow the ANC processing to be causal. When the NR delay grows, the ANC benefits decrease and vanish quickly.

In an integrated use of ANC and NR based on FxMWF, the delay from the NR algorithm does not interfere with the leakage cancellation part. The system is more robust to latency and can almost provide a constant signal-to-noise ratio (SNR) at the tympanic membrane up to the causality bound. Also, the use of a filtered-x algorithm [7]–[9] in the integrated approach allows to include the secondary path effect in the NR computation. 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 into the loudspeaker. Therefore, even with higher system latencies, integrating ANC and NR can lead to performance improvements compared to a classic NR scheme where the noise leakage and the secondary path effect are not taken into account.

Manuscript received March 30, 2009; revised July 29, 2009. First published September 25, 2009; current version published July 14, 2010. This work was carried out at the ESAT Laboratory of Katholieke Universiteit Leuven, in the frame of the Marie-Curie Fellowship EST-SIGNAL Program (http://est-signal.i3s.unice.fr) under Contract MEST-CT-2005-021175, and the Concerted Research Action GOA-AMBioRICS. The scientific responsibility is assumed by the authors. The associate editor coordinating the review of this manuscript and approving it for publication was Prof. Michael L. Seltzer.

R. Serizel and M. Moonen are with the Department of Electrical Engineering, Katholieke Universiteit Leuven, ESAT-SCD, B-3001 Leuven, Belgium.

J. Wouters is with the Division of Experimental Otorhinolaryngology, Katholieke Universiteit Leuven, ExpORL, B-3000 Leuven, Belgium.

S. H. Jensen is with the Department of Electronic Systems, Aalborg University, DK-9220 Aalborg, Denmark.

Color versions of one or more of the figures in this paper are available online at http://ieeexplore.ieee.org.

Digital Object Identifier 10.1109/TASL.2009.2030948

1558-7916/$26.00 © 2010 IEEE

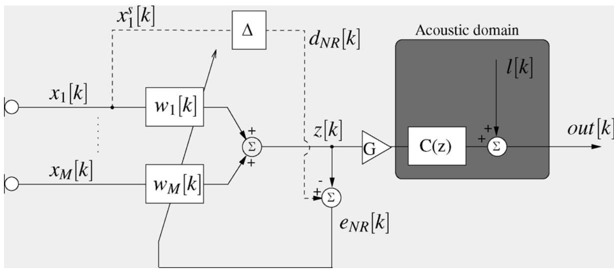

Fig. 1. Multichannel noise reduction system in the hearing aids context.

This paper will present a performance comparison between a standard MWF-based NR without ANC, a cascaded version of NR and ANC, and an integrated ANC and NR using FxMWF; all applied in hearing aids with an open fitting. The effects of the noise leakage and the secondary path on the output of MWF-based NR are commented on in Section II. Section III introduces different schemes combining ANC and NR. The causality problem is described in Section IV. The experimental results are presented in Section V, and finally Section VI presents the conclusions of this paper.

II. Problem Statement

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

A. Signal Model and Multichannel Wiener Filter Basics

Let N be the filter length and M the number of microphones (channels). The signal x_m for microphone m has a desired speech part x_m^s and an additive noise part x_m^n, i.e.,

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

where k is the time index.

In the sequel, superscripts s and n will also be used for other signals and vectors, to denote their speech and noise component, respectively. Signal model (1) holds for so-called "speech plus noise periods." There are also "noise-only periods" (i.e., speech pauses), during which only a noise component is observed.

The column vector $\mathbf{x}_m[k]$ contains the N last samples of the channel m

$$\mathbf{x}_m[k] = [x_m[k] \dots x_m[k-N+1]]^T. \tag{2}$$

The compound vector gathering all channels is

$$\mathbf{x}^T[k] = [\mathbf{x}_1^T[k] \dots \mathbf{x}_M^T[k]]. \tag{3}$$

In the NR context (Fig. 1), the output of the system is

$$z[k] = \mathbf{w}^T[k]\mathbf{x}[k] \tag{4}$$

with $\mathbf{w}^T[k] = [\mathbf{w}_1^T[k] \dots \mathbf{w}_M^T[k]]$ the optimal Wiener filter which minimizes the mean squared error (MSE)

$$J_{\mathrm{MSE}}[k] = E\{|d_{NR}[k] - \mathbf{w}^T[k]\mathbf{x}[k]|^2\} \tag{5}$$

and which is given by

$$\mathbf{w}[k] = \mathbf{R}_{xx}^{-1}[k]\mathbf{r}_{xd_{NR}}[k]. \tag{6}$$

Here, $\mathbf{R}_{xx}[k]$ is the correlation matrix of the input $\mathbf{x}[k]$ and $\mathbf{r}_{xd_{NR}}[k]$ is the cross-correlation vector between the input $\mathbf{x}[k]$ and the desired signal $d_{NR}[k]$, which is chosen to be equal to the (unknown) speech component in the first microphone, up to a delay

$$\mathbf{R}_{xx}[k] = E\{\mathbf{x}[k]\mathbf{x}^T[k]\} \tag{7}$$

$$\mathbf{r}_{xd_{NR}}[k] = E\{\mathbf{x}[k]d_{NR}[k]\} \tag{8}$$

$$d_{NR}[k] = x_1^s[k-\Delta]. \tag{9}$$

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

$$\mathbf{r}_{xd_{NR}}[k] = \mathbf{r}_{xx_{1,\Delta}}[k] - \mathbf{r}_{x^n x^n_{1,\Delta}}[k] \tag{10}$$

$$\mathbf{r}_{xx_{1,\Delta}}[k] = E\{\mathbf{x}[k]x_1[k-\Delta]\} \tag{11}$$

$$\mathbf{r}_{x^n x^n_{1,\Delta}}[k] = E\{\mathbf{x}^n[k]x_1^n[k-\Delta]\}. \tag{12}$$

While $\mathbf{r}_{xx_{1,\Delta}}[k]$ and $\mathbf{R}_{xx}[k]$ are estimated during the speech-plus-noise periods, $\mathbf{r}_{x^n x^n_{1,\Delta}}[k]$ can be estimated during the noise-only periods.

B. Multichannel Wiener Filter-Based Noise Reduction With Leakage and Secondary Path Effects

The NR scheme based on MWF, as applied in the hearing aids context, is presented in Fig. 1. The gain G is the amplification that compensates for the hearing losses. It is considered here to be a broadband gain.

Classic NR schemes ignore the so-called secondary path, i.e., the propagation from the loudspeaker to the tympanic membrane (including the loudspeaker response itself). Assuming that the loudspeaker characteristic is approximately linear, the secondary path can be represented by the transfer function $C(z)$. As explained in [6], the dc gain of $C(z)$ is lower than 1, so the power of the output is decreased when taking the secondary path into account.

The hearing aid with an open fitting has no earmold to prevent ambient sound from leaking into the ear canal, which results in additional leakage signal $l[k]$ reaching the tympanic membrane [10]. No direct processing can be done on this signal; therefore, its SNR is generally lower than for the signal provided by the hearing aid.

Taking both, the leakage signal and the secondary path effect into account, leads to the following output signal model

$$out[k] = C * (G \cdot z[k]) + l[k]. \tag{13}$$

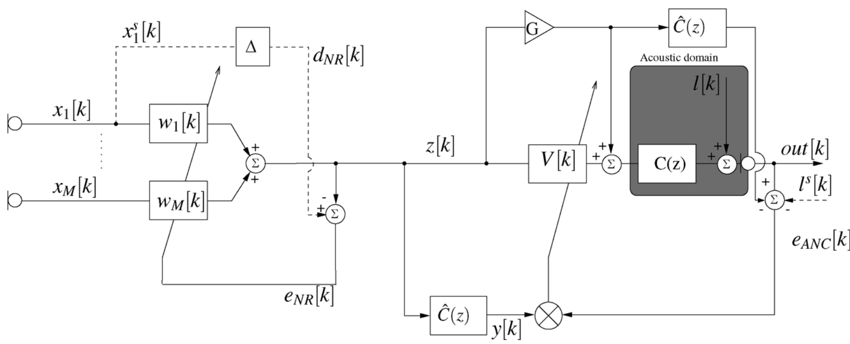

Fig. 2. Multichannel noise reduction and single-channel active noise control systems in cascade.

It clearly appears that, for small amplification gains G, the attenuation caused by the secondary path and the additive leakage contribution do matter. The leakage SNR may affect the output SNR thus partly canceling the improvement achieved with the NR in the hearing aid. In conclusion, whereas the secondary path and the leakage are not taken into account in convetional NR algorithms, they may degrade their performance significantly (see also Section V for an evaluation of the system shown on Fig. 1).

III. Combined Noise Reduction and Active Noise Control

The leakage signal is not processed in the hearing aid therefore it is not possible to improve its SNR using standard NR algorithms. It is possible however to attenuate the leakage signal's noise component using ANC. In all the subsequent systems, it is assumed that a microphone is present in the ear canal to provide an error signal. Commercial hearing aids currently do not have an ear canal microphone, but it is technically possible to include a microphone on the eartip. This section describes different approaches to combine the NR scheme (as presented in Section II) and the ANC algorithm. Two of these schemes (shaded Figs. 3 and 5) will then be used for further evaluation in the next sections (and compared against the scheme shown Fig. 1).

A. Noise Reduction and Single-Channel Active Noise Control in Cascade

A straightforward way to combine ANC and NR is to cascade both functional blocks, using the output of the noise canceller as the input to the ANC system (Fig. 2).

In an ANC system, the controller output is designed to cancel a noise signal and generate a zone of quiet based on destructive interference. In a hearing aid, the noise is to be canceled at the tympanic membrane and, as in any ANC system, the secondary path plays an important part in the algorithm. Introducing this extra path may lead to instabilities. Therefore, it is necessary to use so-called filtered-x algorithms [4], [7]–[9] based on an estimate of the secondary path:

$$\hat{C}(z) = \sum_{i=0}^{L-1} \hat{c}_i z^{-i} \tag{14}$$

$$\hat{\mathbf{c}} = [\hat{c}_0 \ldots \hat{c}_{L-1}]^T \tag{15}$$

and the filtered reference signal

$$y[k] = \hat{\mathbf{c}}^T \mathbf{z}[k] \tag{16}$$

where $\mathbf{z}[k] = [z[k] \ldots z[k-N+1]]^T$.

The secondary path can be estimated offline using classic identification methods based for example on Least Mean Square (LMS) algorithms, or online by adding random noise to the signal exciting the secondary path, as introduced by Eriksson *et al.* in [11] and later refined by Kuo *et al.* [12] and Zhang *et al.* [13].

The ANC output signal is $\mathbf{v}^T[k]\mathbf{z}[k]$, where the filter $\mathbf{v}[k]$ is designed to minimize the MSE

$$J_{\mathrm{MSE}}[k] = E\{|e_{\mathrm{ANC}}[k]|^2\}. \tag{17}$$

Here, $e_{\mathrm{ANC}}[k]$ is an error signal, constructed from the ear canal microphone signal $out[k]$, as will be described next.

The lower branch with $\hat{C}(z)$ in Fig. 2 (and also in subsequent figures), represents the adaptation (gradient estimation) of $\mathbf{v}[k]$ a standard filtered-x adaptive filter algorithm.

The goal here is to cancel the noise component of the leakage signal while preserving the speech signal estimate provided by the NR. In the hearing aids context, the speech component of the leakage signal can provide cues which, e.g., are helpful to localize the speaker. Therefore, it was chosen here to cancel only the noise component of the leakage signal and preserve the speech component. All the schemes presented here, however, can straightforwardly be modified for the case where the full leakage signal is to be canceled. Cancelling only the noise component of the leakage signal effectively corresponds to removing the unknown speech component of the leakage $l^s[k]$ from the ANC error signal $e_{\mathrm{ANC}}[k]$ as indicated in Fig. 2. Note however that, as $l^s[k]$ is unknown, this subtraction is not done explicitly. The noise component of the leakage $l^n[k]$ can be canceled by adapting the filter $\mathbf{v}[k]$ in noise only periods. During speech plus noise periods, the remaining leakage signal could then be considered as an estimate of the speech component of the leakage:

$$l^s[k] = l[k] - l^n[k]. \tag{18}$$

The ANC itself would also tend to remove the desired speech component (NR output signal $z[k]$) so this signal, multiplied by the amplification gain G, has to be added back to the ANC output signal (loudspeaker input signal) and then the same

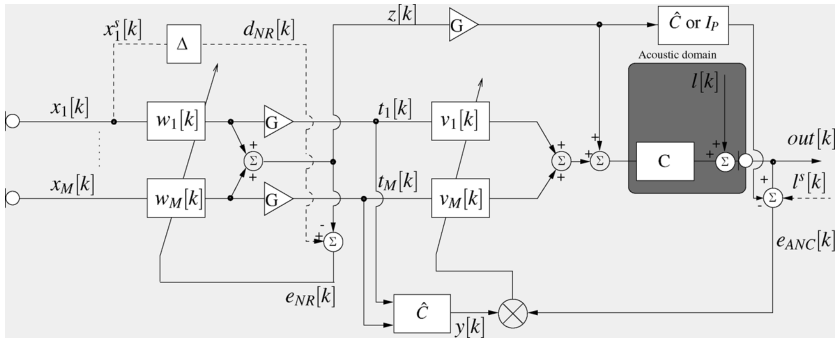

Fig. 3. Multichannel noise reduction and active noise control systems in cascade.

signal filtered by the estimated secondary path $\hat{C}(z)$ (corresponding to an estimate of the NR output as delivered at the tympanic membrane) is subtracted from the ANC error signal $e_{\mathrm{ANC}}[k]$, as explained in [14] and [15], leading to:

$$e_{\mathrm{ANC}}[k] = C*\mathbf{v}^T[k]\mathbf{z}[k]+C*G\cdot z[k]+l[k]-l^s[k]-G\cdot\hat{\mathbf{c}}^T\mathbf{z}[k]. \quad (19)$$

Assuming that the secondary path identification error is small ($\hat{C}(z)\approx C$), the upper branches with G and $\hat{C}(z)$ do not contribute to $e_{\mathrm{ANC}}[k]$. Furthermore, by also assuming that the filter $\mathbf{v}$ is adapting slowly, the error reduces to:

$$e_{\mathrm{ANC}}[k]\approx\mathbf{v}^T[k]\mathbf{y}[k]+l^n[k] \quad (20)$$

where $\mathbf{y}[k]=[y[k]\dots y[k-N+1]]^T$.

The output of the combined system (at the tympanic membrane) is given by

$$out[k]=e_{\mathrm{ANC}}[k]+G\cdot\hat{\mathbf{c}}^T\mathbf{z}[k]+l^s[k] \quad (21)$$
$$out[k]\approx e_{\mathrm{ANC}}[k]+C*(G\cdot z[k])+l^s[k]. \quad (22)$$

This is the sum of the minimized error signal, the enhanced speech signal as delivered at the tympanic membrane and the speech component of the leakage signal.

Assuming that the speech and the noise components in (20) are uncorrelated, the MSE criterion (17) can be rewritten as follows:

$$J_{\mathrm{MSE}}[k]\approx E\{|\mathbf{v}^T[k]\mathbf{y}^s[k]|^2\}+E\{|\mathbf{v}^T[k]\mathbf{y}^n[k]+l^n[k]|^2\}. \quad (23)$$

A simple ANC scheme is then obtained when the MSE criterion is modified to:

$$J^n_{\mathrm{MSE}}[k]=E\{|e^n_{\mathrm{ANC}}[k]|^2\}\approx E\{|\mathbf{v}^T[k]\mathbf{y}^n[k]+l^n[k]|^2\} \quad (24)$$

resulting in a standard filtered-x adaptive filtering operated in noise only periods.

In the case of a perfect NR the noise component of the input signal of the ANC is zero. This is problematic as the feedforward ANC is designed to produce so-called anti-noise based on the noise component of its input. In practice, the NR is never perfect and so the noise component of its output, i.e., the input of the ANC, is nonzero and can be used in the ANC to produce the anti-noise. Still, this noise component may be small and then the ANC may exhibit a poor performance.

B. Noise Reduction and Multichannel Active Noise Control in Cascade

The NR output signal $z[k]$ is the sum of filtered microphone signals. These filtered microphone signals themselves generally have a significant noise component, typically larger than the noise component in the sum signal. These signals therefore provide more suitable input signals for the ANC than the NR output $z[k]$ itself. A cascaded scheme with a multichannel ANC can then be derived from the initial cascade of MWF-based NR and single channel ANC as indicated in Fig. 3, which will exhibit improved performance.

The NR algorithm is the same as described in Section II. The multichannel input of the ANC is the output of the NR before the summation. Let $t_m[k]$ denote the mth filtered microphone signal

$$t_m[k]=\mathbf{w}_m^T[k]\mathbf{x}_m[k]\quad m\in\{1\dots M\}. \quad (25)$$

Let $\mathbf{t}_m[k]$ denote the column vector containing the last P samples of $t_m[k]$ (P being the length of the ANC filter) and let $\mathbf{t}[k]$ denote the compound vector gathering all the channels

$$\mathbf{t}_m[k]=[t_m[k]\dots t_m[k-P+1]]^T\quad m\in\{1\dots M\}$$
$$\mathbf{t}^T[k]=[\mathbf{t}_1^T[k]\dots\mathbf{t}_M^T[k]]. \quad (26)$$

The filtered reference signals in the ANC are defined as

$$y_m[k]=\hat{\mathbf{c}}^T\mathbf{t}_m[k]\quad m\in\{1\dots M\}. \quad (27)$$

Let $\mathbf{y}_m[k]$ denote the column vector containing the last P samples of $y_m[k]$ and let $\mathbf{y}[k]$ denote the compound vector gathering all the channels:

$$\mathbf{y}_m[k]=[y_m[k]\dots y_m[k-P+1]]^T\quad m\in\{1\dots M\}$$
$$\mathbf{y}^T[k]=[\mathbf{y}_1^T[k]\dots\mathbf{y}_M^T[k]]. \quad (28)$$

The ANC output signal is equal to $\mathbf{v}^T[k]\mathbf{t}[k]$, where $\mathbf{v}[k]^T=[\mathbf{v}_1^T[k]\dots\mathbf{v}_M^T[k]]$ is a multichannel adaptive filter of length P, which minimizes the MSE

$$J_{\mathrm{MSE}}[k] = E\{|C*\mathbf{v}^T[k]\mathbf{t}[k]+C*G\cdot z[k]+l^n[k]-G\cdot\hat{\mathbf{c}}^T\mathbf{z}[k]|^2\}. \quad (29)$$

Assuming that the secondary path identification error is small ($\hat{C}(z) \approx C$) and that the filter $\mathbf{v}$ is adapting slowly, the error signal reduces to

$$e_{\mathrm{ANC}}[k] \approx \mathbf{v}^T[k]\mathbf{y}[k] + l^n[k]. \tag{30}$$

The output of the combined system (at the tympanic membrane) is given by

$$\mathrm{out}[k] = e_{\mathrm{ANC}}[k] + G \cdot \hat{\mathbf{c}}^T \mathbf{z}[k] + l^s[k] \tag{31}$$
$$\mathrm{out}[k] \approx e_{\mathrm{ANC}}[k] + C * (G \cdot z[k]) + l^s[k]. \tag{32}$$

This is the sum of the minimized error signal, the enhanced speech signal as delivered at the tympanic membrane and the speech component of the leakage signal. Cascading a standard NR scheme and the ANC algorithm thus allows to improve the SNR of the signal, while reducing the impact of noise leakage on the final output.

Assuming that the speech and the noise components in (30) are uncorrelated, the MSE criterion (29) can be rewritten as follows:

$$J_{\mathrm{MSE}}[k] \approx E\{|\mathbf{v}^T[k]\mathbf{y}^s[k]|^2\} + E\{|\mathbf{v}^T[k]\mathbf{y}^n[k] + l^n[k]|^2\}. \tag{33}$$

A simple ANC scheme is then again obtained when the MSE criterion is modified to

$$J^n_{\mathrm{MSE}}[k] = E\{|e^n_{\mathrm{ANC}}[k]|^2\} \approx E\{|\mathbf{v}^T[k]\mathbf{y}^n[k] + l^n[k]|^2\} \tag{34}$$

resulting in a standard filtered-x adaptive filtering operated in noise only periods. One consequence of such adaptation during noise only periods is that the effect of the filter $\mathbf{v}$ on the speech component of the signal is unknown and cannot be controlled. Also, as long as $\hat{C}(z) \approx C$ in the upper branch (Fig. 3), the filter $\mathbf{v}$ does not compensate for the secondary path effects.

An attempt to compensate for this extra path can be to replace $\hat{C}(z)$ in the upper branch (Fig. 3) by I_P, which is the identity filter of length P such that, for any signal $x[k]$, $I_P * x[k] = x[k]$. The MSE criterion (29) is then modified into

$$J_{\mathrm{MSE}}[k] = E\{|C * \mathbf{v}^T[k]\mathbf{t}[k] + C * G \cdot z[k] + l^n[k] - G \cdot z[k]|^2\}. \tag{35}$$

Introducing $\mathbf{v'}[k]$ as the compound vector gathering all the $\mathbf{v'}_m[k]$ such that

$$\mathbf{v'}_m[k] = \mathbf{v}_m[k] + G \cdot I_P \quad m \in \{1 \ldots M\} \tag{36}$$
$$\mathbf{v'}^T[k] = [\mathbf{v'}^T_1[k] \ldots \mathbf{v'}^T_M[k]] \tag{37}$$

the ANC error signal can be written as follows:

$$e_{\mathrm{ANC}}[k] = C * \mathbf{v'}^T[k]\mathbf{t}[k] + l^n[k] - G \cdot z[k] \tag{38}$$

where $\mathbf{v'}[k]$ is the optimization vector.

Assuming that the secondary path identification error is small ($\hat{C}(z) \approx C$) and that the filter $\mathbf{v'}$ is adapting slowly, the error reduces to

$$e_{\mathrm{ANC}} \approx \mathbf{v'}^T[k]\mathbf{y}[k] + l^n[k] - G \cdot z[k]. \tag{39}$$

Assuming that the speech and noise components in (35) are uncorrelated, the MSE criterion (35) can be written as

$$J_{\mathrm{MSE}}[k] \approx E\{|\mathbf{v'}^T[k]\mathbf{y}^n[k] - G \cdot z^n[k] + l^n[k]|^2\} + E\{|\mathbf{v'}^T[k]\mathbf{y}^s[k] - G \cdot z^s[k]|^2\}. \tag{40}$$

Here, the coefficients of the filter have to be updated during both noise only periods and speech plus noise periods. This makes the use of standard ANC algorithms based on gradient estimation inconvenient. Therefore, here and in the subsequent schemes, the adaptive filters are computed based on the estimation of second order statistics of the speech signals and the noise signals, as in the MWF approach of Section II.

From (40), it can be seen that the desired signal to be used here is

$$d_{\mathrm{ANC}}[k] = -l^n[k] + G \cdot z[k]$$
$$= -l^n[k] + G \cdot z^n[k] + G \cdot z^s[k]. \tag{41}$$

The optimal filter $\mathbf{v'}$ is then given by

$$\mathbf{v'}[k] = \mathbf{R}^{-1}_{yy}[k]\mathbf{r}_{yd_{\mathrm{ANC}}}[k]. \tag{42}$$

Here, $\mathbf{R}_{yy}[k]$ is the correlation matrix of the filtered reference signal $\mathbf{y}[k]$ and $\mathbf{r}_{yd_{\mathrm{ANC}}}[k]$ is the cross-correlation vector between the filtered reference signal $\mathbf{y}[k]$ and the target signal $d_{\mathrm{ANC}}[k]$

$$\mathbf{R}_y[k] = E\{\mathbf{y}[k]\mathbf{y}^T[k]\} \tag{43}$$
$$\mathbf{r}_{yd_{\mathrm{ANC}}}[k] = E\{\mathbf{y}[k]d_{\mathrm{ANC}}[k]\}. \tag{44}$$

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

$$\mathbf{r}_{yd_{\mathrm{ANC}}}[k] = \mathbf{r}_{y^s d^s_{\mathrm{ANC}}}[k] + \mathbf{r}_{y^n d^n_{\mathrm{ANC}}}[k] \tag{45}$$
$$= G \cdot \mathbf{r}_{y^s z^s}[k] + \mathbf{r}_{y^n d^n_{\mathrm{ANC}}} \tag{46}$$
$$= G \cdot [\mathbf{r}_{yz}[k] - \mathbf{r}_{y^n z^n}[k]] + \mathbf{r}_{y^n d^n_{\mathrm{ANC}}} \tag{47}$$

with

$$\mathbf{r}_{yz}[k] = E\{\mathbf{y}[k]z[k]\} \tag{48}$$
$$\mathbf{r}_{y^n z^n}[k] = E\{\mathbf{y}^n[k]z^n[k]\} \tag{49}$$
$$\mathbf{r}_{y^n d^n_{\mathrm{ANC}}}[k] = E\{\mathbf{y}^n[k]d^n_{\mathrm{ANC}}[k]\}. \tag{50}$$

While $\mathbf{R}_{yy}[k]$ and $\mathbf{r}_{yz}[k]$ are estimated during speech plus noise periods, $\mathbf{r}_{y^n z^n}[k]$ and $\mathbf{r}_{y^n d^n_{\mathrm{ANC}}}[k]$ can be estimated during noise only periods, where based on (39)

$$d^n_{\mathrm{ANC}}[k] = G \cdot z^n[k] - l^n[k] \tag{51}$$
$$\approx \mathbf{v'}^T[k]\mathbf{y}^n[k] - e^n_{\mathrm{ANC}}[k]. \tag{52}$$

The first term on the right-hand side in (40) corresponds to the difference between the amplified desired speech signal and the speech component of the signal reaching the tympanic membrane, where the secondary path effect has been canceled effectively. The second term is the difference between the noise component in the amplified $z[k]$ and the noise component of

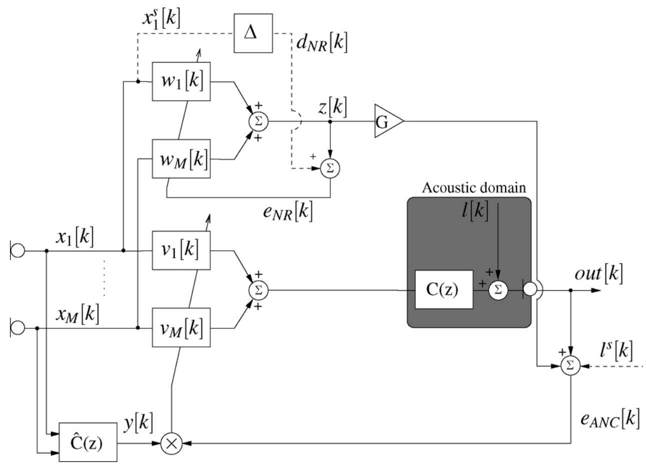

Fig. 4. Active noise control and noise reduction system in parallel.

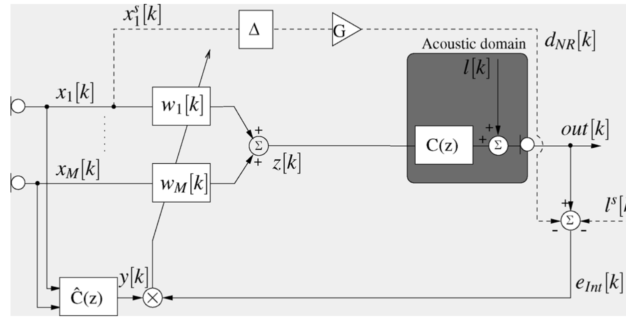

Fig. 5. Integrated multichannel active noise control and noise reduction system.

the signal (loudspeaker + leakage) reaching the tympanic membrane. Therefore, minimizing (40) corresponds to compensating for the secondary path effect on the reference signal while canceling the effect of the noise leakage. The output signal (at the tympanic membrane) is

$$out[k] = e_{\mathrm{ANC}}[k] + G \cdot z[k] + l^s[k] \tag{53}$$

which is the sum of the minimized error, the enhanced speech signal as delivered at the tympanic membrane and the speech component of the leakage signal.

In the cascaded implementation, the input of the ANC is related to the output of the NR which is then also the reference signal for the secondary path cancellation (upper branch on Fig. 3). This is problematic, the ANC needs an input with a strong noise component, while the secondary path cancellation ideally has to be applied to the desired speech signal. To design a performant algorithm for ANC and secondary path cancellation, the input signal of the ANC and the secondary path cancellation reference signal have to be different signals. Therefore, cascading does not seem to be the most efficient approach (see also Section V for an evaluation of the system in Fig. 3).

C. Noise Reduction and Multichannel Active Noise Control in Parallel

As an alternative to cascading the NR with the ANC, the two functional blocks can also be put in parallel (Fig. 4). Here, $\mathbf{w}[k]$ is an MWF applied in the context of NR and $\mathbf{v}[k]$ is a multichannel ANC filter, also compensating for the secondary path effects.

Assuming that the speech and noise components are uncorrelated, the MSE to be minimized by $\mathbf{v}[k]$ is

$$\begin{aligned} J_{\mathrm{MSE}}[k] = E\{|C * \mathbf{v}^T[k]\mathbf{x}^n[k] - G \cdot z^n[k] + l^n[k]|^2\} \\ + E\{|C * \mathbf{v}^T[k]\mathbf{x}^s[k] - G \cdot z^s[k]|^2\}. \end{aligned} \tag{54}$$

The filter $\mathbf{v}[k]$ thus minimizes the noise sound pressure at the tympanic membrane as well as the influence of the secondary path on the output signal. The output of the system is the sum of this minimized error, the amplified enhanced speech signal and the speech component of the leakage signal:

$$out[k] = e_{\mathrm{ANC}}[k] + G \cdot z[k] + l^s[k]. \tag{55}$$

The parallel system is thus combining the NR, the ANC and the compensation of the secondary path effect. In this approach the reference signal for the secondary path compensation is an estimate of the speech signal ($z[k] = \mathbf{w}[k]^T\mathbf{x}[k]$) amplified by G. To further simplify the system, this signal can implicitly be replaced by the desired signal of the NR ($d_{NR}[k]$). This is pursued in the next section.

D. Integrated Active Noise Control and Noise Reduction

This section introduces an algorithm integrating both the NR and the ANC in a single set of adaptive filters (Fig. 5).

The algorithm relies on a filtered-x version of the MWF (FxMWF) based on an estimate of the secondary path $\hat{C}(z)$. The filtered reference signals are now

$$y_m[k] = \hat{\mathbf{c}}^T\mathbf{x}_m[k] \quad m \in \{1 \ldots M\} \tag{56}$$
$$\mathbf{y}_m[k] = [y_m[k] \ldots y_m[k-N+1]]^T \tag{57}$$
$$\mathbf{y}^T[k] = [\mathbf{y}_1^T[k] \ldots \mathbf{y}_M^T[k]]. \tag{58}$$

The aim of the integrated scheme is to improve the speech-to-noise ratio, and so the desired signal (at the tympanic membrane) to be used is

$$\begin{aligned} d_{Int}[k] &= -l^n[k] + G \cdot x_1^s[k-\Delta] \\ &= -l^n[k] + d_{NR}[k]. \end{aligned} \tag{59}$$

The MSE criterion to be minimized is then

$$\begin{aligned} J_{\mathrm{MSE}}[k] &= E\{|e_{Int}[k]|^2\} \\ &= E\{|C * \mathbf{w}^T[k]\mathbf{x}[k] + l^n[k] - d_{NR}[k]|^2\}. \end{aligned} \tag{60}$$

Assuming that the secondary path identification error is small ($\hat{C}(z) \approx C$) and that the filter $\mathbf{w}$ is adapting slowly, the error signal can be rewritten as follows:

$$e_{Int}[k] \approx \mathbf{w}^T[k]\mathbf{y}[k] + l^n[k] - d_{NR}[k]. \tag{61}$$

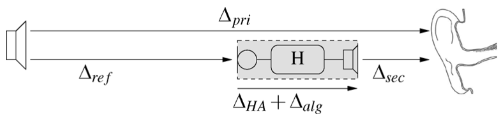

Fig. 6. Delays in hearing aid system environment.

Assuming that, the noise and speech components in (61) are uncorrelated, the criterion (60) can be rewritten as follows:

$$J_{\mathrm{MSE}}[k] = E\{|\mathbf{w}^T[k]\mathbf{y}^s[k] - d_{NR}[k]|^2\} + E\{|\mathbf{w}^T[k]\mathbf{y}^n[k] + l^n[k]|^2\}. \quad (62)$$

The first term of the right-hand side corresponds to the secondary path compensation on the speech component of the output signal while the second term specifies the noise sound pressure at the tympanic membrane and thus corresponds to the ANC. The filter described in (63) is thus performing a NR, which takes the secondary path into account, combined with a ANC.

The optimal filter (FxMWF) minimizing (62) is

$$\mathbf{w}[k] = \mathbf{R}_{yy}^{-1}[k]\mathbf{r}_{yd_{Int}}[k]. \quad (63)$$

Here, $\mathbf{R}_{yy}[k]$ is the correlation matrix of the filtered reference signal $\mathbf{y}[k]$ and $\mathbf{r}_{yd_{Int}}[k]$ is the cross-correlation vector between the filtered reference signal $\mathbf{y}[k]$ and the desired signal $d_{Int}[k]$

$$\mathbf{R}_{yy}[k] = E\{\mathbf{y}[k]\mathbf{y}^T[k]\} \quad (64)$$
$$\mathbf{r}_{yd_{Int}}[k] = E\{\mathbf{y}[k]d_{Int}[k]\}. \quad (65)$$

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

$$\mathbf{r}_{yd_{Int}}[k] = \mathbf{r}_{y^s d_{NR}}[k] - \mathbf{r}_{y^n l^n}[k] \quad (66)$$
$$= \mathbf{r}_{yx_{1,\Delta}}[k] - G \cdot \mathbf{r}_{y^n x^n_{1,\Delta}}[k] - \mathbf{r}_{y^n l^n}[k] \quad (67)$$

with

$$\mathbf{r}_{yx_{1,\Delta}}[k] = E\{\mathbf{y}[k]x_1[k-\Delta]\} \quad (68)$$
$$\mathbf{r}_{y^n x^n_{1,\Delta}}[k] = E\{\mathbf{y}^n[k]x^n_1[k-\Delta]\} \quad (69)$$
$$\mathbf{r}_{y^n l^n}[k] = E\{\mathbf{y}^n[k]l^n[k]\}. \quad (70)$$

While $\mathbf{R}_{yy}[k]$ and $\mathbf{r}_{yx_{1,\Delta}}[k]$ are estimated during speech plus noise periods, $\mathbf{r}_{y^n x^n_1[k-\Delta]}$ and $\mathbf{r}_{y^n l^n}[k]$ can be estimated during noise only periods with

$$l^n[k] \approx e^n_{Int}[k] - \mathbf{w}^T[k]\mathbf{y}^n[k]. \quad (71)$$

IV. Robustness to Causality

As explained in [4], the main condition to get the (feedforward) ANC system working is that a causality criterion is satisfied. That is (Fig. 6) the acoustic delay from the noise source to the ear canal microphone Δ_{pri} is longer than: the sum of the delay from the source to one of the reference microphones Δ_{ref}, the delay associated with the processing within the hearing aid Δ_{HA}, the algorithmic delay Δ_{alg} and the acoustic delay of the secondary path Δ_{sec}.

The bandwidth on which it is possible to achieve good ANC performance reduces with the "degree of causality" (i.e., the delay margin specified in (73)). When (72) is not satisfied, the ANC efficiency vanishes quickly [16]. Delay is thus a critical problem in ANC and many approaches have been developed to try to deal with it [17], [18]

$$\Delta_{\mathrm{ref}}+\Delta_{\mathrm{HA}}+\Delta_{\mathrm{alg}}+\Delta_{\mathrm{sec}} \le \Delta_{\mathrm{pri}} \quad (72)$$
$$\delta = \Delta_{\mathrm{pri}} - (\Delta_{\mathrm{ref}}+\Delta_{\mathrm{HA}}+\Delta_{\mathrm{sec}}). \quad (73)$$

In case of hearing aids, the delay available for processing is linked to the distance between the microphones and the loudspeaker which is not more than a few centimeters. This corresponds to a few tens of microseconds, i.e., only a few samples for standard sampling frequencies.

A. Noise Reduction and Active Noise Control in Cascade

In the cascaded schemes presented in Sections III-A and III-B, the input of the ANC is the output of a standard multichannel NR, which itself introduces a delay $\Delta_{\mathrm{alg}} = \Delta$ (Fig. 1). Usually, this delay is set to half of the NR filter length (N) and will already exceed the few taps available for processing. Therefore, the causality criterion specified in (72) cannot be fullfilled and so the ANC may not be able to yield good performance. Reducing the NR delay increases the causality of the system, but this also has an impact on the NR performance.

Cascading NR and ANC therefore requires a tradeoff between the performance of the two functional blocks. Besides, knowing that the ANC's performance quickly decreases as the non-causality increases, the NR delay Δ has to be decreased drastically in order to improve the ANC efficiency. In realistic scenarios, finding a satisfying tradeoff may be impossible, which may make the ANC useless.

B. Integrated Noise Reduction and Active Noise Control

In the integrated approach (Section III-D), the filter minimizing the MSE (62) can be split into a sum of two filters

$$\mathbf{w}[k] = \mathbf{u}[k] + \mathbf{v}[k] \quad (74)$$

where

$$\mathbf{u}[k] = \mathbf{R}_{yy}^{-1}[k]E\{(\mathbf{y}^s)^T[k]d_{NR}[k]\} \quad (75)$$
$$\mathbf{v}[k] = -\mathbf{R}_{yy}^{-1}[k]E\{(\mathbf{y}^n)^T[k]l^n[k]\}. \quad (76)$$

The filter $\mathbf{u}[k]$ describes a NR which also compensates for the secondary path effects while the filter $\mathbf{v}[k]$ is an ANC system canceling the noise leakage. So, under the assumption that speech and noise components are uncorrelated, the set of filters integrating NR and ANC can be seen as the sum of two sets of filters, one for NR and the other for the ANC. Therefore, the output of the ANC does not depend on the delay introduced in the NR part (i.e., $\Delta_{\mathrm{alg}} = 0$) and so it is possible to design a causal active noise controller to be integrated with the NR as long as

$$\Delta_{\mathrm{ref}} + \Delta_{\mathrm{HA}} + \Delta_{\mathrm{sec}} \le \Delta_{\mathrm{pri}}. \quad (77)$$

That is, there is no performance tradeoff to be done between the NR and the ANC.

V. Experimental Results

The algorithms introduced in Section II (Fig. 1) and Section III-B (Fig. 3) and Section III-D (Fig. 5) have been tested experimentally and their performance has been compared.

A. Experimental Setup

The simulations were run on acoustic path measurements with a manikin head and torso equipped with artificial ears and a two-microphone behind-the-ear (BTE) hearing aid. The speech source was located at 0° and a noise source at 270°. The BTE was worn on left ear, facing the noise source. Commercial hearing aids currently do not have an ear canal microphone, therefore the artificial ear eardrum microphone is used here to generate the error signal. The tests were run on 22-s-long signals. The speech was composed of three sentences from the HINT database [19] concatenated with silence periods. The noise was the multitalker babble from Auditec [20]. All the signals were sampled at 16 kHz.

The (BTE microphone) input SNR is often used as a reference measure in standard NR schemes. In our case, as two algorithms also perfom 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) improvement [21] is used here as a performance measure, which is defined as

$$\Delta\mathrm{SNR}_{\mathrm{intellig}} = \sum_i I_i(\mathrm{SNR}_{i,\mathrm{out}} - \mathrm{SNR}_{i,\mathrm{leak}}) \qquad (78)$$

where I_i is the band importance function defined in [22] and $\mathrm{SNR}_{i,\mathrm{out}}$ and $\mathrm{SNR}_{i,\mathrm{leak}}$ represent the output SNR and the leakage SNR (in dB) of the ith band, respectively.

The filter lengths are set to $N = 64$ and $P = 64$, and the NR delay is set to half of the NR filter length ($\Delta = 32$). The secondary path $C(z)$ is estimated offline using an identification technique based on the Normalized Least Mean Square (NLMS) algorithm. The length of the estimated path $\hat{C}(z)$ is set to $L = 32$.

The first experiment shows the effect of leakage on the NR performance and the improvement achieved by ANC. For an amplification gain G varying from 0 dB to 20 dB the inputs are filtered using the three algorithms previously described and the $\Delta\mathrm{SNR}_{\mathrm{intellig}}$ is evaluated. The system is calibrated so that for $G = 0$ dB, for a source at 0°, the leakage and the signal fed in the loudspeaker have equal power.

The second test aimed to demonstrate the impact of delay on the ANC performance with the different algorithms. With a fixed amplification gain G, for a varying degree of causality δ(73), the system performances are compared.

B. Leakage and Secondary Path Effects, Improvements With Active Noise Control

To evaluate the effect of the leakage and the secondary path, the input signals are first filtered by an MWF-based NR scheme (Fig. 1). Depending on which disturbance is being tested, the signal produced can then be filtered by the secondary path model $C(z)$ and/or the leakage can be added, as described in Section II.

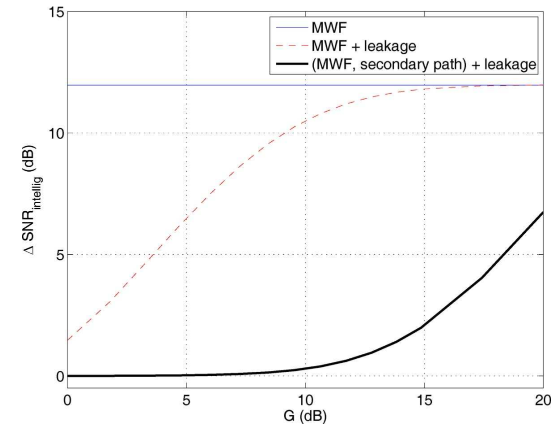

Fig. 7. Performance comparison for a Multichannel Wiener Filer noise reduction scheme depending on leakage and secondary path.

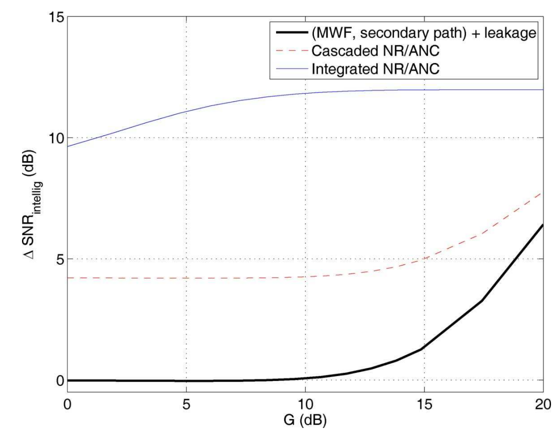

Fig. 8. Performance comparison for noise reduction scheme with or without active noise control, $\delta = 48$.

The reference SNR (leakage signal), is equal to -1.3 dB. This value depends on the noise and speech angles as well as the input SNR (source signals), which is 5 dB here.

When the leakage is the only disturbance considered, the degradations induced by the leakage remain small even for reasonably low gain G [down to 10 dB (Fig. 7)]. However, introducing both the leakage and the secondary path the degradations are significant for gains up to at least 20 dB. This shows that for small amplification gains, as usually used with open fittings, there is a need for leakage cancellation.

Fig. 8 shows the performance of the combinations of NR and ANC [cascaded (Fig. 3) and integrated (Fig. 5)] compared to a classic NR, for gain $G \in [0\ 20]$ dB. The degree of causality is here kept high enough ($\delta = 48$, i.e., for $\Delta_{\mathrm{alg}} \leq 48$ the criterion (72) is fullfilled) so that no performance tradeoff has to be made between the NR and the ANC. For gains up to 15 dB, cascading NR and ANC allows to maintain a constant SNR improvement of around 4 dB, which is already significantly better then the performance of the NR alone. When the gain is increasing above 15 dB, the performance converges to the performance of the NR alone. The integrated approach gives an almost constant SNR improvement around 12 dB for all values of the gain. This is better than NR alone or the cascaded NR and ANC.

All these results are given for a system where the degree of causality is sufficient for any processing, which is not the case in a realistic system.

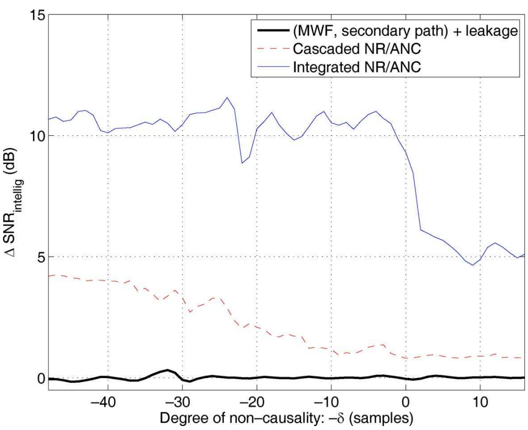

Fig. 9. Performance depending on the degree of non-causality.

C. Causality Study

In hearing aids, the causality margins are much smaller than what was used for the previous simulations. Based on the transfer functions which are used here, the degree of causality for a signal coming from an angle of 270° (the noise direction of arrival) is seen to be about two samples (rather than the 48 samples used previously).

To see how the delays in the system can affect performance, for each algorithm the amplification is set to $G = 10$ dB, and a delay is added to the input signals (BTE microphones), to allow the degree of causality to vary between -16 and 48. Fig. 9 shows the SNR improvement for each algorithm as a function of the degree of causality δ.

When the degree of causality is high, the obtained performance for the cascade is close to what was obtained in Fig. 8. When the degree of causality decreases and becomes lower than the NR delay ($\Delta = 32$), the SNR improvement starts to decrease and eventually converges to the improvement obtained with the classic NR. This is due to the fact that for a degree of causality lower than 32, the ANC has to be designed as a non-causal system, therefore its performance is reduced. The integrated approach gives an almost constant SNR improvement as long as the overall system is causal.

Fig. 10 shows the SNR improvement given by the three algorithms for a realistic degree of causality equal to 2, which corresponds to what has been measured on the transfer functions used for the simulations. For lower gains, $G \leq 15$ dB, the cascade approach gives only minor improvement (around 1 dB) compared to the standard scheme, then their performances tend to converge as amplification is increased. The integrated approach on the other hand maintains an SNR improvement of more than 10 dB. Therefore, this integrated approach seems to offer a practical way to introduce ANC in hearing aids.

Finally, Fig. 11 shows the SNR improvement for $\delta = 16$ for the integrated approach and the NR only approach. Here, the performance improvement achieved with the integrated approach is mainly due to the secondary path compensation. Fig. 11 also shows the SNR improvement obtained with a filtered-x MWF algorithm that does not include the ANC, which is indeed found to achieve a similar performance improvement.

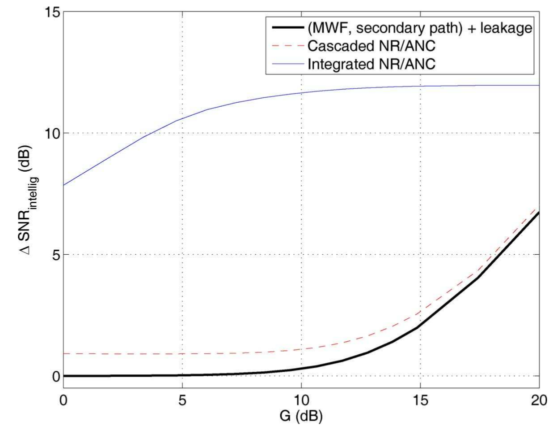

Fig. 10. Performance comparison for noise reduction scheme with or without active noise control, $\delta = 2$.

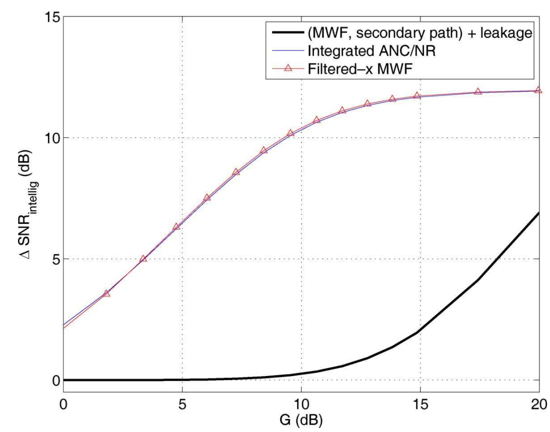

Fig. 11. Performance for a non-causal system, $\delta = -16$.

VI. Conclusion

Standard NR techniques used in hearing aids ignore leakage and secondary path effects. When open fittings are used these aspects cannot be neglected and are in fact found to seriously degrade the NR performance. ANC can then be used to reduce the impact of the leakage, it has shown to provide SNR improvements between 4 dB and 12 dB depending on the approach used.

The ANC performance is conditioned by the system causality which differs for the two algorithms evaluated here. A cascaded approach can give good results (SNR improvement around 4 dB) as long as the system causality is high enough to support the NR latency ($\delta > N/2$). This is not a realistic assumption for hearing aids where the latency margin is in fact close to zero.

An alternative integrated approach of ANC with NR has shown to improve the SNR by about 12 dB for low hearing aid gains (between 0 dB and 20 dB), as long as the system is causal. When the system becomes non-causal, the integrated approach can still outperform standard NR algorithms by taking the secondary path into account in the speech enhancement and amplification process, thereby reducing the impact of the leakage on the output signal. All the previous schemes, however, rely on the presence of an ear canal microphone. Commercial hearing aids currently do not have an ear canal microphone. Adding this extra microphone might induce some problems such as bone conduction to the ear canal microphone when the user talks. These problems would have to be investigated before practical tests.

REFERENCES

[1] J. Kiessling, "Sounds towards the tympanic membrane," in *Proc. 8th EFAS Congr.*, Heidelberg, Germany, Jun. 2007.

[2] L. J. Griffiths and C. W. Jim, "An alternative approach to linearly constrained adaptive beamforming," *IEEE Trans. Antennas Propagat.*, vol. AP-30, no. 1, pp. 27–34, Jan. 1982.

[3] S. Doclo, A. Spriet, J. Wouters, and M. Moonen, "Frequency-domain criterion for the speech distortion weighted multichannel wiener filter for robust noise reduction," *Speech Commun.*, vol. 49, no. 7-8, pp. 636–656, 2007.

[4] S. J. Elliott and P. A. Nelson, *Active Control of Sound*. Cambridge, MA: Academic, 1993.

[5] S. M. Kuo and D. R. Morgan, "Active noise control: a tutorial review," *Proc. IEEE*, vol. 87, no. 6, pp. 943–973, Jun. 1999.

[6] R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen, "Combined active noise control and noise reduction in hearing aids," in *Proc. 11th Int. Workshop Acoust. Echo Noise Control (IWAENC)*, Sep. 2008.

[7] J. C. Burgess, "Active adaptive sound control in a duct: a computer simulation," *J. Acoust. Soc. Amer.*, vol. 70, no. 3, pp. 715–726, Sep. 1981.

[8] B. Widrow and S. D. Stearns, *Adaptive Signal Processing*. Englewood Cliffs, NJ: Prentice-Hall, 1985.

[9] E. Bjarnason, "Analysis of the filtered-x LMS algorithm," *IEEE Trans. Speech Audio Process.*, vol. 3, no. 6, pp. 504–514, Nov. 1995.

[10] H. Dillon, *Hearing Aids*. New York: Thieme, 2001.

[11] L. J. Eriksson and M. C. Allie, "Use of random noise for on-line transducer modeling in an adaptive active attenuation system," *J. Acoust. Soc. Amer.*, vol. 85, no. 2, pp. 797–802, Feb. 1989.

[12] S. M. Kuo and D. Vijayan, "A secondary path modeling technique for active noise controlsystems," *IEEE Trans. Speech Audio Process.*, vol. 5, no. 4, pp. 374–377, Jul. 1997.

[13] M. Zhang, H. Lan, and W. Ser, "Cross-updated active noise control system with online secondary path modeling," *IEEE Trans. Speech Audio Process.*, vol. 9, no. 5, pp. 598–602, Jul. 2001.

[14] W. S. Gan and S. M. Kuo, "An integrated audio and active noise control headset," *IEEE Trans. Consumer Electron.*, vol. 48, no. 2, pp. 242–247, May 2002.

[15] S. M. Kuo and B. M. Finn, "An integrated audio and active noise control system," in *Proc. IEEE Int. Symp. Circuits Syst. ISCAS'93*, 1993, pp. 2529–2532.

[16] X. Kong and S. M. Kuo, "Study of causality constraint on feedforward active noise control systems," *IEEE Trans. Circuits Systems II: Analog Digital Signal Process.*, vol. 46, no. 2, pp. 183–186, Feb. 1999.

[17] D. R. Morgan and J. C. Thi, "A delayless subband adaptive filter architecture," *IEEE Trans. Signal Process.*, vol. 43, no. 8, pp. 1819–1830, Aug. 1995.

[18] B. Rafaely and M. Furst, "Audiometric ear canal probe with active ambient noise control," *IEEE Trans. Speech Audio Process.*, vol. 4, no. 3, pp. 224–230, May 1996.

[19] M. Nilsson, S. D. Soli, and A. Sullivan, "Development of the Hearing in Noise Test for the measurement of speech reception thresholds in quiet and in noise," *J. Acoust. Soc. Amer.*, vol. 95, no. 2, pp. 1085–1099, Feb. 1994.

[20] "Auditory Tests (Revised), Compact Disc, Auditec,". St. Louis, MO, Auditec, 1997.

[21] J. E. Greenberg, P. M. Peterson, and P. M. Zurek, "Intelligibility-weighted measures of speech-to-interference ratio and speech system performance," *J. Acoust. Soc. Amer.*, vol. 94, no. 5, pp. 3009–3010, Nov. 1993.

[22] *American National Standard Methods for Calculation of the Speech Intelligibility Index*, ANSI S3.5–1997, Acoust. Soc. Amer., 1997.

Romain Serizel received the M.Eng. degree in automatic system engineering from ENSEM, Nancy, France, in 2005 and the M.Sc. degree in signal processing from Université Rennes 1, Rennes, France, in 2006. He is currently pursuing the Ph.D. degree under the supervision of Prof. M. Moonen in the Electrical Engineering Department (ESAT-SCD), Katholieke Universiteit Leuven, Leuven, Belgium.

His research interests include hearing aids systems and digital signal processing for audio.

Marc Moonen (M'94–SM'06–F'07) received the electrical engineering degree and the Ph.D. degree in applied sciences from Katholieke Universiteit Leuven, Leuven, Belgium, in 1986 and 1990, respectively.

Since 2004, he has been a Full Professor in the Electrical Engineering Department, Katholieke Universiteit Leuven, where he is heading a research team working in the area of numerical algorithms and signal processing for digital communications, wireless communications, DSL, and audio signal processing.

Prof. Moonen received the 1994 K.U.Leuven Research Council Award, the 1997 Alcatel Bell (Belgium) Award (with P. Vandaele), the 2004 Alcatel Bell (Belgium) Award (with R. Cendrillon), and was a 1997 "Laureate of the Belgium Royal Academy of Science." He received a journal best paper award from the IEEE TRANSACTIONS ON SIGNAL PROCESSING (with G. Leus) and from Elsevier Signal Processing (with S. Doclo). He was chairman of the IEEE Benelux Signal Processing Chapter (1998–2002), and is currently Past-President of EURASIP (European Association for Signal Processing) and a member of the IEEE Signal Processing Society Technical Committee on Signal Processing for Communications. He has served as Editor-in-Chief for the *EURASIP Journal on Applied Signal Processing* (2003–2005), and has been a member of the editorial board of IEEE TRANSACTIONS ON CIRCUITS AND SYSTEMS II (2002–2003), and *IEEE Signal Processing Magazine* (2003–2005) and *Integration, the VLSI Journal*. He is currently a member of the editorial board of *EURASIP Journal on Applied Signal Processing*, *EURASIP Journal on Wireless Communications and Networking, and Signal Processing*.

Jan Wouters was born in Leuven, Belgium, in 1960. He received the physics degree and the Ph.D. degree in sciences/physics from the Katholieke Universiteit Leuven, Leuven, Belgium, in 1982 and 1989, respectively.

From 1989 until 1992, he was a Research Fellow with the Belgian National Fund for Scientific Research (FWO) at the Institute of Nuclear Physics (UCL Louvain-la-Neuve and K.U. Leuven) and at NASA Goddard Space Flight Center. Since 1993, he has been a Professor at the Neurosciences Department of the K.U. Leuven (Full Professor since 2001). His research activities center around audiology and the auditory system, signal processing for cochlear implants, and hearing aids. He is author of about 145 articles in international peer-reviewed journals and is a reviewer for several international journals.

Dr. Wouters received an Award of the Flemish Ministry in 1989, a Fullbright Award and a NATO Research Fellowship in 1992, and the Flemish VVL Speech Therapy–Audiology Award in 1996. He is member of the International Collegium for ORL (CORLAS), a Board Member of the International Collegium for Rehabilitative Audiology (ICRA), and is responsible for the Laboratory for Experimental ORL and the audiology program at K.U. Leuven.

Søren Holdt Jensen (S'87–M'88–SM'00) received the M.Sc. degree in electrical engineering from Aalborg University, Aalborg, Denmark, in 1988, and the Ph.D. degree in signal processing from the Technical University of Denmark, Lyngby, in 1995.

He is a Full Professor at Aalborg University and is currently heading a research team working in the area of numerical algorithms and signal processing for speech and audio processing, image and video processing, multimedia technologies, and digital communications. Before joining the Department of Electronic Systems at Aalborg University, he was with the Telecommunications Laboratory of Telecom Denmark, Ltd., Copenhagen, Denmark, the Electronics Institute of the Technical University of Denmark, the Scientific Computing Group of Danish Computing Center for Research and Education (UNI•C), Lyngby, Denmark, the Electrical Engineering Department of Katholieke Universiteit Leuven, Leuven, Belgium, and the Center for PersonKommunikation (CPK) of Aalborg University.

Prof. Jensen was an Associate Editor for the IEEE TRANSACTIONS ON SIGNAL PROCESSING, and is currently Member of the Editorial Board of *Elsevier Signal Processing* and the *EURASIP Journal on Advances in Signal Processing*. He is a recipient of an European Community Marie Curie Fellowship, former Chairman of the IEEE Denmark Section, and Founder and Chairman of the IEEE Denmark Section's Signal Processing Chapter.