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Integrated active noise control and noise reduction in hearing aids

Romain Serizel

Dissertation presented in partial
fulfillment of the requirements for
the degree of Doctor
in Engineering Sciences

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Preface

Five years (almost), and now it is my turn to write a preface to my PhD. The good news is this means that I have finally completed my doctoral research. The bad news is that writing a preface appears to be as difficult as many persons before me claimed it was. So let me keep it simple and just take some time to express my gratitude to those who helped me during my PhD and who made this possible.

Firstly, I would like to thank Prof. Marc Moonen for giving me the opportunity to join his group, for his continuous encouragement, guidance, and most importantly for believing in me. I have learned a lot during the five years I have spent here.

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From 2006 to 2009 I was a Marie Curie Fellow involved in the EST-SIGNAL project (Early Stage Training in Signal processing). I would like to thank the Marie Curie Actions for funding the first years of my doctoral research.

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Jeanne and Steven. Thank you for coming here to my defence.

Last but not least, I would like to thank my dear girlfriend Pernelle for her patience and constant support, especially so during the past few (intense) months.

Abstract

In every day life conversations and listening scenarios the desired speech signal is rarely delivered alone. The listener most commonly faces a scenario where he has to understand speech in a noisy environment. Hearing impairments, and more particularly sensorineural losses, can cause a reduction of speech understanding in noise. Therefore, in a hearing aid compensating for such kind of losses it is not sufficient to just amplify the incoming sound. Hearing aids also need to integrate algorithms that allow to discriminate between speech and noise in order to extract a desired speech from a noisy environment. A standard noise reduction scheme in general aims at maximising the signal-to-noise ratio of the signal to be fed in the hearing aid loudspeaker. This signal, however, does not reach the eardrum directly. It first has to propagate through an acoustic path and encounter some perturbations which are, by design, neglected in standard noise reduction schemes.

In an open fitting setup, there is no earmold to prevent ambient sound from reaching the eardrum or to prevent the sound amplified by the hearing aid from leaving the ear canal. This signal leakage through the open fitting combined with the attenuation in the acoustic path between the hearing aid loudspeaker and the eardrum, *i.e.*, the so-called secondary path, can then override the action of the noise reduction in the hearing aid. Active noise control can be used to compensate for the effects of this signal leakage. The principle of active noise control is to generate a zone of quiet based on destructive interference, in this case at the eardrum. In the hearing aids framework, however, active noise control alone is not sufficient. It has to be performed together with the noise reduction algorithm.

This thesis first presents an integrated active noise control and noise reduction scheme for hearing aids to tackle secondary path effects and effects of noise leakage through an open fitting. Integrating active noise control and noise reduction in a single set of filters allows to compensate for the signal leakage and the secondary path effects. The implementation of the integrated active noise control and noise reduction scheme in hearing aids, however, comes with a number of problems primarily due to the dimensions of the devices.

Firstly, the integrated active noise control and noise reduction scheme does not allow to balance between the noise reduction and the active noise control. In some circumstances it would be useful to emphasise one of the functional blocks. Secondly, the integrated active noise control and noise reduction scheme relies on an ear canal microphone which should ideally be located at the eardrum. In practice, however, the ear canal microphone is distant from the eardrum and the sound reaching the eardrum is basically unknown and uncontrolled. Finally, the use of active noise control in hearing aids, is limited by the size of the devices and the number of microphones available on each device. The number of noise sources that can be compensated for by the active noise control is limited by the number of microphones available. Also the small separation between the microphones and the loudspeaker results in a short acoustic propagation time and hence small causality margins.

In order to solve these problems, variations on the integrated active noise control and noise reduction scheme are also presented in this thesis. Firstly, changing the original problem to a constrained problem leads to weighted integrated active noise control and noise reduction schemes. A first weighted integrated active noise control and noise reduction scheme is derived that allows to emphasise either the active noise control (providing an improved signal-to-noise ratio) or the noise reduction (providing a lower speech distortion). A speech intelligibility weighted integrated active noise control and noise reduction scheme is then derived that allows to focus on reducing speech distortion at the eardrum or on minimising the residual noise at the eardrum. Secondly, an integrated approach to active noise control and noise reduction that is based on an optimisation over a zone of quiet generated by the active noise control is then proposed. This approach allows to overcome the ear canal microphone location problem. Finally, a binaural approach is introduced that allows to access extra microphones from the contralateral hearing aid and to design a scheme with increased causality margin.

Korte Inhoud

In dagelijkse conversaties zijn de luisteromstandigheden zelden dusdanig dat alleen het gewenste spraaksignaal aangeleverd wordt. De luisteraar is meestal geconfronteerd met omstandigheden waarin hij spraak moet verstaan in een omgeving met achtergrondruis. Gehoorbeperkingen, en in het bijzonder sensorineuraal gehoorverlies, kunnen zorgen voor een vermindering van de verstaanbaarheid van spraak in ruis. Om in een hoorapparaat dit soort gehoorverlies te compenseren, is het bijgevolg niet voldoende om het inkomende geluid enkel te versterken. Hoorapparaten dienen ook algoritmen te bevatten die toelaten om een onderscheid te maken tussen spraak en ruis, zodanig dat het gewenste spraaksignaal uit de omgeving met achtergrondruis geëxtraheerd kan worden. In het algemeen heeft een standaard ruisonderdrukkingsschema als doel om de signaal-ruisverhouding te maximaliseren van het signaal dat naar de luidspreker van het hoorapparaat gestuurd wordt. Dit signaal bereikt het trommelvlies echter niet rechtstreeks. Eerst propageert het signaal door een akoestisch pad en ondergaat het enkele storingen die in standaard ruisonderdrukkingsschema's bij ontwerp worden verwaarloosd.

In een hoorapparaat met open aanpassing is er geen oorstukje dat voorkomt dat omgevingsgeluid het trommelvlies bereikt. Deze signaallek door de open aanpassing, gecombineerd met de verzwakking in het akoestisch pad tussen de luidspreker van het hoorapparaat en het trommelvlies, het zgn. secundaire pad, kan dan de werking van de in het hoorapparaat uitgevoerde ruisonderdrukking ongedaan maken. “Active Noise Control” (ANC) kan aangewend worden om de effecten van deze signaallek te compenseren. Het principe van ANC is om een stiltezone te genereren die gebaseerd is op destructieve interferentie, in dit geval aan het trommelvlies. In de context van hoorapparaten is ANC alleen echter niet voldoende. ANC dient uitgevoerd worden samen met het ruisonderdrukkingsalgoritme.

Dit proefschrift stelt in een eerste fase een geïntegreerd ANC- en ruisonderdrukkingsschema voor, dat effecten van het secundaire pad en effecten van signaallek in hoorapparaten met open aanpassing aanpakt. Het integreren van

ruisonderdrukking en ANC in één enkele set filters maakt het mogelijk om signaallek en effecten van het secundaire pad te compenseren. De implementatie van het geïntegreerd ANC- en ruisonderdrukkingsschema brengt evenwel talrijke problemen met zich mee, voornamelijk wegens de dimensies van de apparaten.

Het geïntegreerd ANC- en ruisonderdrukkingsschema laat niet toe om af te wegen tussen ruisonderdrukking en ANC. In sommige omstandigheden zou het nuttig zijn om de nadruk te leggen op één van de functionele blokken. Daarnaast vertrouwt het geïntegreerd ANC- en ruisonderdrukkingsschema op de aanwezigheid van een microfoon in de gehoorgang, idealiter gepositioneerd ter hoogte van het trommelvlies. In de praktijk bevindt de microfoon in de gehoorgang zich echter op een zekere afstand van het trommelvlies, en is het geluid dat het trommelvlies bereikt ongekend en ongecontroleerd. Ten slotte is het gebruik van ANC in hoorapparaten beperkt door de afmetingen van de apparaten en door het aantal microfoons dat beschikbaar is op elk apparaat. Het aantal ruisbronnen dat kan gecompenseerd worden via ANC is immers beperkt door het aantal beschikbare microfoons. Tevens resulteert de korte afstand die de microfoons van de luidspreker scheidt in een korte akoestische propagatietijd en dus in kleine causaliteitsmarges.

Om deze problemen op te lossen, stelt dit proefschrift ook variaties voor op het geïntegreerd ANC- en ruisonderdrukkingsschema. Het veranderen van het originele probleem naar een probleem met beperkingen leidt tot een gewogen schema dat toelaat om de nadruk te leggen op ofwel ANC (wat zorgt voor een verbeterde signaal-ruisverhouding) ofwel ruisonderdrukking (wat zorgt voor een lagere spraakdistortie). Daarna wordt een geïntegreerde aanpak voor ANC en ruisonderdrukking voorgesteld, die gebaseerd is op een optimalisatie over een stiltezone gegenereerd door ANC. Deze aanpak laat toe om het positioneringsprobleem van de microfoon in de gehoorgang te overwinnen. Ten slotte wordt een binaurale aanpak voorgesteld die toelaat om extra microfoons van het contralaterale hoorapparaat te gebruiken, en om een schema met een hogere causaliteitsmarge te ontwerpen.

Glossary

Mathematical functions

a	scalar a
$\mathbf{a}$	vector $\mathbf{a}$
$\mathbf{A}$	matrix $\mathbf{A}$
a^*	complex conjugate of a
$\mathbf{A}^T$	transpose of the matrix $\mathbf{A}$
$\mathbf{A}^H$	Hermitian transpose of the matrix $\mathbf{A}$
e^a	exponential of a
$\mathbf{e}_i$	N-dimensional column vector with the ith element equal to 1 and all the other elements equal to 0
$\mathbf{e}_{i,\Delta}$	N-dimensional column vector with the ith element equal to $\mathrm{e}^{-j\omega\Delta}$ and all the other elements equal to 0
$\mathbb{E}\{.\}$	expectation operator
J	MSE criterion
$aJ(S)$	average MSE criterion over zone S
I_P	identity filter of length P
$\mathbf{r}_{xy}$	cross-correlation vector between vector $\mathbf{x}$ and vector $\mathbf{y}$
$\mathbf{R}_x$	autocorellation matrix of vector $\mathbf{x}$
$\tilde{\mathbf{R}}_x$	empirical autocorellation matrix of vector $\mathbf{x}$
$\Re\{.\}$	real part
$\forall$	for all
$\partial/\partial x(.)$	partial first derivative with respect to x
$a \approx b$	a is approximately equal to b
$a \triangleq b$	a is defined as b
$a \in [b,\ c]$	a is in the interval $[b,\ c]$
$a \notin [b,\ c]$	a is not in the interval $[b,\ c]$

Signals

$\mathbf{A}$	M-dimensional steering vector, which contains the acoustic transfer functions from the speech source position to the hearing aid microphones
$C(z)$	secondary path transfer function
$\hat{C}(z)$	estimate of the secondary path
$d_{NR}[k]$	time-domain desired signal for the NR at time k
$\delta 1(\mathbf{r})$	plane wave model for a propagation from the ear canal microphone to a point $\mathbf{r}$
$\delta 2(\mathbf{r})$	monopole-point source model for a propagation from the ear canal microphone to a point $\mathbf{r}$
$e[k]$	time-domain error signal at time k
$e_{\mathrm{NR}}[k]$	time-domain error signal for the NR at time k
$e_{\mathrm{ANC}}[k]$	time-domain error signal for the ANC at time k
$e_{\mathrm{Int}}[k]$	time-domain error signal for the integrated ANC and NR at time k
$E(\omega)$	frequency-domain error signal
$E_{\mathrm{NR}}(\omega)$	frequency-domain error signal for the NR
$E_{\mathrm{ANC}}(\omega)$	frequency-domain error signal for the ANC
$E_{\mathrm{Int}}(\omega)$	frequency-domain error signal for the integrated ANC and NR
$l[k]$	time-domain leakage signal at time k
$L(\omega)$	frequency-domain leakage signal
$\mathrm{Pow}^{n}_{\mathrm{leak}}$	power of the noise component of the leakage signal at the eardrum (in dB)
$\mathrm{Pow}^{n}_{\mathrm{out}}$	power of the residual noise and the power at the eardrum (in dB)
$a\mathrm{Pow}_{\mathrm{out}}$	average power of the output signal out over the zone S (in dB)
$\Delta\mathrm{Pow}$	residual noise power improvement (in dB)
$\mathrm{SNR}_{\mathrm{out}}$	SNR of the output signal (in dB)
$a\mathrm{SNR}_{\mathrm{out}}$	average SNR of the output signal out over the zone S (in dB)
$t_m[k]$	time-domain signal in the BTE microphone m at time k filtered by $\mathbf{w}_m[k]$
$\mathbf{t}_m[k]$	P-dimensional data vector of the signal $t_m[k]$
$\mathbf{t}[k]$	MP-dimensional stacked signal vector
$\mathbf{u}[k]$	time-domain NR part of the integrated ANC and NR filter
$\mathbf{U}$	frequency-domain NR part of the integrated ANC and NR filter
$\mathbf{v}[k]$	time-domain ANC adaptive filter
$\mathbf{V}$	frequency-domain ANC adaptive filter
$\mathbf{w}[k]$	time-domain MWF
$\mathbf{W}(\omega)$	frequency-domain MWF

$x_m[k]$	time-domain signal in the BTE microphone m at time k
$x_m^n[k]$	noise component of the signal $x_m[k]$
$x_m^s[k]$	speech component of the signal $x_m[k]$
$x_{L,m}[k]$	time-domain signal in the left hearing aids mth BTE microphone at time k
$x_{R,m}[k]$	time-domain signal in the right hearing aids mth BTE microphone at time k
$\mathbf{x}_m[k]$	N-dimensional data vector of the signal $x_m[k]$
$\mathbf{x}[k]$	MN-dimensional stacked signal vector
$X_m(\omega)$	frequency-domain signal in the BTE microphone m
$X_m^n(\omega)$	noise component of the signal $X_m(\omega)$
$X_m^s(\omega)$	speech component of the signal $X_m(\omega)$
$\mathbf{X}(\omega)$	M-dimensional stacked signal vector
$y_m[k]$	time-domain filtered mth BTE microphone signal at time k
$y_{\text{Cas}}[k]$	time-domain filtered reference signal in the cascaded ANC and NR.
$y_{Cas,m}[k]$	time-domain mth filtered reference signal in the multichannel cascaded ANC and NR.
$\mathbf{y}_m[k]$	N-dimensional data vector of the signal $y_m[k]$
$\mathbf{y}_{\text{Cas}}[k]$	N_{ANC}-dimensional data vector of the signal $y_{\text{Cas}}[k]$
$\mathbf{y}_{\text{Cas},m}[k]$	N_{ANC}-dimensional data vector of the signal $y_{\text{Cas},m}[k]$
$\mathbf{y}[k]$	MN-dimensional stacked signal vector
$\mathbf{y}_{\text{Cas,Multi}}[k]$	MN_{ANC}-dimensional stacked signal vector
$Y_m(\omega)$	frequency-domain filtered mth BTE microphone signal
$\mathbf{Y}(\omega)$	M-dimensional stacked signal vector
$z[k]$	output signal of the filter $\mathbf{w}[k]$
$\mathbf{z}[k]$	P-dimensionnal stacked vector of the filter output $z[k]$
$\tilde{z}[k]$	time-domain signal in the ear canal microphone at time k
$Z(\omega)$	output signal of the filter $\mathbf{W}(\omega)$
$\tilde{Z}(\omega)$	frequency-domain signal in the ear canal microphone
$\tilde{Z}_{\text{HA}}(\omega)$	hearing aid contribution to the frequency-domain signal in the ear canal microphone
$\xi_2(S)$	average propagation coefficient over zone S for one monopole-point source
$\xi_{2,2}(S)$	average propagation coefficient over zone S for two monopole-point sources
$\xi_{2,1}(S)$	average propagation coefficient over zone S for one monopole-point source and one plane wave source

Parameters

c_0	speed of sound in air
δ	degree of causality
δ	NR delay
Δ_{alg}	processing structural delay
Δ_{HA}	hearing aids delays (D/A, A/D converters...)
Δ_{pri}	propagation delay from the sound source to the eardrum (primary path)
Δ_{ref}	propagation delay from the sound source to the BTE microphones
Δ_{sec}	propagation delay from the receiver to the eardrum (secondary path)
f	frequency-domain variable (Hz)
f_s	sampling frequency (Hz)
G	amplification in the hearing aid
k	discrete time index
λ_f	exponential forgetting factor for frequency-domain correlation matrices
λ_t	exponential forgetting factor for time-domain correlation matrices
m	microphone index
M	number of microphones in an hearing aid
μ	trade-off parameter between ANC and NR
ν	auxiliary parameter between ANC and NR ($\nu = \frac{\mu+1}{\mu}$)
η	auxiliary paramter between ANC and NR ($\eta = \frac{1}{\mu+1}$)
N	filter length of filter $\mathbf{W}$
N_{ANC}	filter length of the ANC filter $\mathbf{v}$
N_{DFT}	DFT window length
P	length of estimate of the secondary path $C(z)$
Q	number of noise sources (speech sources plus noise sources)
$\mathbf{r}$	spatial variable
$\mathbf{r}_0$	spatial location of the hearing aid loudspeaker
r	radial coordinate of point $\mathbf{r}$
ρ	output SNR of the MWF-based NR
S	desired zone of quiet
$\bar{S}$	area of zone S
θ	angular coordinate of point $\mathbf{r}$
ω	frequency-domain variable ($\text{rad} \cdot s^{-1}$)

Acronyms and Abreviations

aMSE	average mean squared error
aSNR	average signal-to-noise ratio
A/D	analog-to-digital
AFC	acoustic feedback cancellation
ANC	active noise control
ANR	active noise reduction
BTE	behind-the-ear
CIC	completly-in-the canal
D/A	digital-to-analog
dB	decibel
DFT	discrete Fourier transform
DRC	dynamic range compression
DSP	digital signal processing
FFT	fast Fourier transform
DTFT	discrete-time Fourier transform
FxLMS	filtered-x least mean square
FxMWF	filtered-x multichannel Wiener filter
GSC	generalised sidelobe canceller
HINT	hearing in noise test
IDFT	inverse discrete Fourier transform
IDTFT	inverse discrete-time Fourier transform
IFFT	inverse fast Fourier transform
ITC	in-the-canal
ITE	in-the-ear
LCMV	linearly constrained minimum variance
LMS	least mean squares
MSE	mean squared error
MWF	multichannel Wiener filter
NLMS	normalized least mean squares
NR	noise reduction
RLS	recursive least squares
RP	remote point
SD	speech distortion
SDW-MWF	speech distortion weighted multichannel Wiener filter
SNR	signal-to-noise ratio
SRT	speech reception threshold
STFT	short time Fourier transform
VAD	voice activity detector
WASN	wireless acoustic sensor network
ZQ	zone of quiet

Contents

Chapter 1

Introduction

The work presented in this thesis is motivated by the growing use of hearing aids with an open fitting. Whereas removing the earmold reduces the occlusion effect and improves the physical comfort, one major drawback is that nothing is left to prevent the ambient sound to reach directly the eardrum and the sound produced by the hearing aid loudspeaker to reach out of the ear canal.

One consequence of the ageing of the population as well as of precocious exposure to sounds at a damaging level, is that the number of hearing impaired persons has been steadily growing during the past decades. People with hearing impairment will most commonly suffer from sensorineural losses which alter the hearing in several ways: some sounds become inaudible because the absolute audibility thresholds are increased, other sounds are perceived incorrectly due to spectral deformations. As a consequence from these degradations, a person suffering from sensorineural losses usually has a decreased ability to understand speech.

In a common listening scenario the desired speech is usually present together with some disturbances (*e.g.*, other talkers, background noise...) and can easily be unintelligible for a person suffering from sensorineural losses. Therefore, in a hearing aid compensating for such a kind of losses it is not sufficient to just amplify the incoming sound. Hearing aids also need to integrate algorithms which allow to discriminate between speech and noise in order to extract desired speech from a noisy environment.

With the recent advances in electronics and digital signal processing (DSP), hearing aids have evolved from a simple amplification/playback system to a device that can incorporate different DSP algorithms. In addition to a standard noise reduction (NR) algorithm used to compensate for speech intelligibility deficit caused by sensorineural losses, a hearing aid can perform a large variety of DSP algorithms, *e.g.*, acoustic feedback cancellation (AFC), dynamic range compression (DRC)...

Moreover, integration of performant AFC algorithms in hearing aids have allowed not only usage of higher amplifications but also broader access to hearing aids with an open fitting. In an open fitting setup, there is no earmold to prevent ambient sound from reaching the eardrum. This signal leakage through the open fitting combined with the attenuation in the acoustic path between the hearing aid loudspeaker and the eardrum (the so-called secondary path) can override the action of the NR processing done in the hearing aid. Active noise control (ANC) can be used to compensate for the effects of this signal leakage. The principle of ANC is to generate a zone of quiet based on destructive interference, in this case at the eardrum.

In this thesis an *integrated active noise control and noise reduction* scheme is first presented to tackle the signal leakage problem. Chapter 2 and Chapter 3 describe the problem statement and the different ways to combine and finally integrate ANC and NR. Chapter 4 to Chapter 6 present extensions of the integrated ANC and NR scheme.

1.1 Preliminaries

This section briefly describes the auditory system, the effects of hearing impairment on sound perception, the different hearing aid devices commercially available at the moment of writing and the acoustic signals commonly involved in an every day life listening scenario.

1.1.1 The auditory system

The auditory system, presented in Figure 1.1, is the part of the human body responsible for hearing. It can be decomposed in three different parts: the outer-ear, the middle-ear and the inner-ear [30, 107].

The outer-ear consists of the pinna and the ear canal. Its goal is to gather the sound energy and to focus it toward the eardrum.

The middle-ear is the part of the ear between the eardrum and the oval window of the cochlea. It contains three ossicles which convert the sound vibration transmitted from the eardrum to waves transmitted to the fluid in the cochlea.

The inner-ear contains the cochlea and the auditory nerve. The cochlea is divided in its length by the basilar membrane. When a sound wave is present in the cochlea, the basilar membrane moves and the hair cells that are attached to the membrane convert the mechanical waves to a neural signal. The auditory nerve then transmits these neural pulses to the brain where they are interpreted as different sounds.

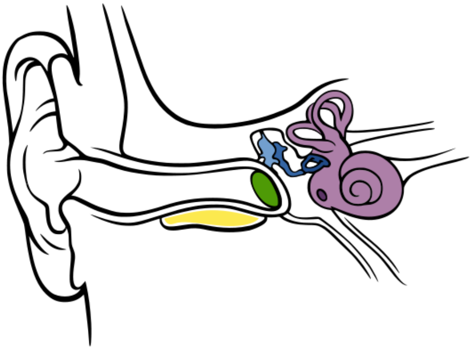


Figure 1.1: Auditory system

1.1.2 Hearing impairment

Hearing impairments can be classified in different categories depending on the part of the auditory system which is affected. Problems in the outer-ear and the middle ear, affecting the transmission of the sound to the inner-ear are called conductive hearing losses. They can usually be corrected by a medical intervention or amplification of the sound. The losses caused by problems in the inner-ear are called sensorineural losses. They compose a large majority of the hearing loss cases (about 90%) and can cause the impairment of various hearing abilities [17, 71].

Reduced audibility

Hearing impaired persons may require a sound to have a higher level to be audible (the so-called hearing threshold) as compared to persons with a normal hearing [32, 115]. Therefore, some sounds might appear to be softer than they really are and some other sounds might not be heard at all. Whereas people with a severe hearing loss may not hear the speech at all, people with a moderate or a mild hearing loss may perceive the speech but not understand it properly. Persons suffering from sensorineural hearing losses might just hear someone who is actually talking clearly but it will appear to them as if it was just mumbling [43, 54, 73]. Indeed, due to the increased hearing threshold some of the phonemes may not be audible and one sound may be confused with another sound [45, 61, 127]. To overcome this problem, the sound can be amplified for the particular frequencies where losses are localised.

Reduced dynamic range

Organic sounds are not usually emitted at a constant level. When a sound is too soft, it can be amplified but if a sound is amplified too much it can cause pain to the listener, when the sound level reaches the so-called threshold of loudness discomfort [115]. The difference between the hearing threshold and the threshold of loudness discomfort is the range within which a listener can hear a sound without any pain. It is commonly called the dynamic range of hearing.

With sensorineural impairments, the hearing threshold will usually increase more than the threshold of loudness discomfort. Therefore the dynamic range of hearing is reduced and loud sounds should not be amplified by the same amount as soft sounds. To overcome this problem the sound can be amplified by a non-linear gain in order to decrease its dynamic range (*i.e.*,the difference between the loudest sound and the softest sound). This is the so-called DRC [105, 106].

Reduced frequency/temporal resolution

The frequency selectivity between sounds takes place in the cochlea (which acts as a filterbank): hair cells present at the base of the cochlea are sensitive to high frequencies, while hair cells located at the apex react to low frequencies. When a frequency component is present in a sound, the corresponding area in the cochlea is excited. When a sound is composed of two frequency tones which are closely spaced, the cochlea is excited on a single broad region and the brain is unable to discriminate between the tones. This is the so-called frequency masking.

With a damaged cochlea, one specific frequency can be exciting a broader area of the basilar membrane than with a normal cochlea. Hence the impact of the frequency masking becomes more important and the frequency resolution (*i.e.*, the smallest frequency difference needed to be able to discriminate between two frequency tones) is reduced [33, 38, 104]. This can cause confusion between phonemes and therefore a decreased ability to understand speech.

In a similar way, when two sounds follow each other immediately, the louder sound can mask the softer one. The time separation needed to discriminate between two consecutive sounds can be increased with sensorineural losses. Therefore, the temporal (feedforward or feedback) masking has a great impact on the way hearing impaired persons perceive sounds and on their ability to understand speech [13, 22, 128].

1.1.3 Hearing aids

From the acoustic era in the late 17th century [5, 46] through the carbon era and the transistor era during the 20th century up to the present digital era, hearing aids have been used to compensate for hearing impairments. Because of the ageing of the population and the earlier exposure to sounds at a damaging level (*e.g.*, in the Netherlands 10% of the population in the age range $12 - 15$ already suffer permanent losses [62]) the number of hearing impaired persons has increased steadily during the past decades. This, combined with an easier access to new techniques and to products which are designed to be more appealing (or at least less repelling), the hearing aids market has become a fastly expending market.

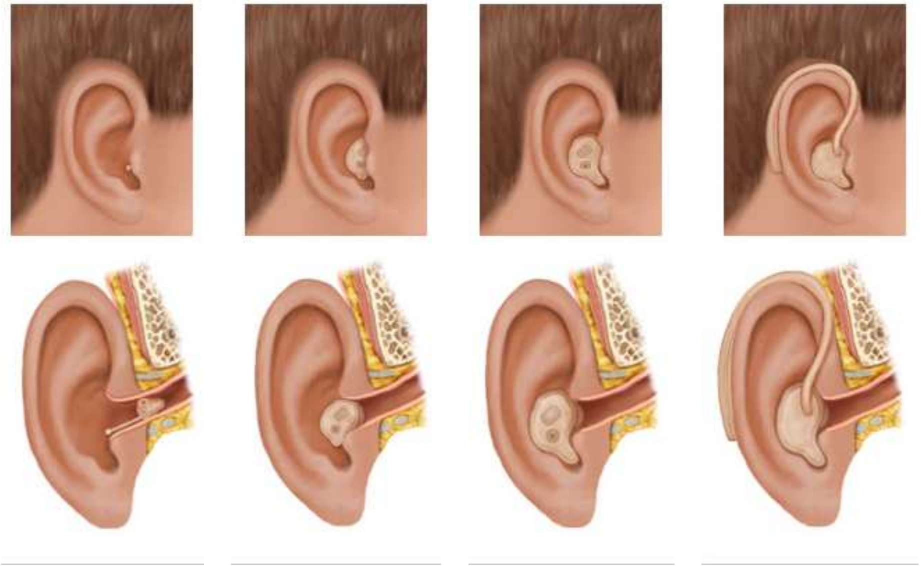
As an impact from the broader range of hearing aid users, the constant demand for improved comfort and better speech intelligibility requires that hearing aids provide powerful DSP algorithms. The miniaturisation constraints and the need for DSP have established hearing aids as an area offering very challenging problems in terms of DSP. The kind of DSP applied in the hearing aid depends on the type of hearing losses to be compensated for, as well as on the type of hearing aid used. At the moment of writing, commercially available hearing aids can be categorised in four different types [51].

Canal aids

Canal aids are designed to fit entirely in the ear-canal and can be of two types: completely-in-the-canal (CIC) or in-the-canal (ITC). CIC hearing aids are seated completely in the ear canal and are therefore almost invisible. CIC are however difficult to adjust or remove. ITC hearing aids are slightly larger than CIC and they protrude a bit from the ear canal. They are therefore slightly easier to manipulate. Canal aids are limited in gain and signal processing power due to their small size. They are therefore more appropriate for mild to moderate hearing losses.

In-the-ear hearing aids

In-the-ear (ITE) hearing aids are designed to fit entirely in the outer-ear (*i.e.*, ear canal and pinna). ITE hearing aids are bigger than CIC and ITC hearing aids and therefore leave more room for a larger receiver (*i.e.*, the hearing aid loudspeaker) and integrated circuits and batteries (*i.e.*, more signal processing power).



(a) CIC (b) ITC (c) ITE (d) BTE

Figure 1.2: Different types of hearing aids [42]

Behind-the-ear hearing aids

In a behind-the-ear (BTE) hearing aid the electronics are in a small case hooked behind the ear. The receiver is in the BTE case and the sound is conducted to the ear canal through a plastic tube which terminates by an earmold. The earmold is custom-made to fit the ear canal of the user and prevent ambient sound to reach directly the eardrum. The earmold also prevents the sound from the receiver to reach the BTE microphones, hence reducing the feedback effects and allowing for high amplifications. BTE hearing aids are larger than CIC, ITC and ITE hearing aids and leave more room for a bigger receiver and more DSP power. BTE hearing aids are therefore suited for mild to profound hearing losses. For conciseness, a BTE hearing aid will often be referred to as a BTE, from now on.

Open fitting BTE hearing aids

Open fitting BTE hearing aids are a kind of BTE hearing aids which has been used more commonly during the past years. The main difference with a classic BTE hearing aid is that the earmold is removed (and sometimes replaced by a small piece of silicon). The absence of the earmold reduces the occlusion effect, the risk of infections in the ear canal and it improves the physical comfort [52, 53] but it also leaves nothing to prevent the ambient sound to reach directly the eardrum. The amplification in open fitting hearing aids is also limited due to the higher risk of feedback caused by the absence of the earmold. Hence, the use of open fitting BTE is particularly dependent on performant AFC algorithms [7, 44, 79, 92, 113, 120]. An evaluation of state-of-the-art AFC algorithms compared with AFC algorithms implemented in commercial hearing aids can be found in [112]. This type of hearing aids is particularly suited for mild to moderate hearing losses.

1.1.4 Acoustic signals

Common listening scenarios (Figure 1.3) involve at least one speech source and sometimes noise sources or concurrent speech sources. The characteristics of speech signals and noise signal have an influence on the way they are dealt with. The speech signals and noise signals that can be encountered in a hearing aid framework are briefly described here. More detailed descriptions of speech signal characteristics and speech processing in general can be found in [16, 85, 88]

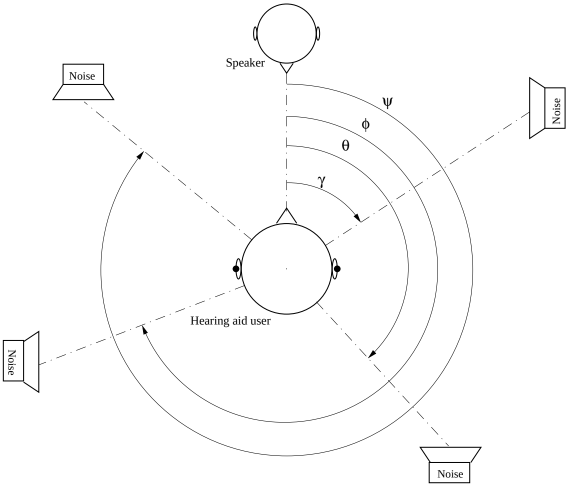


Figure 1.3: Common listening scenario

Speech signals

According to the speech production model, the speech signal can be categorised in two different kinds of speech: voiced speech and unvoiced speech.

- Voiced speech signal is band-limited: the main portion of its energy concentrates below 4kHz and its mean frequency envelope decays at 6dB/Octave.
- Unvoiced speech is broadband and has a much flatter spectrum

In both voiced speech and unvoiced speech most of the speech signal content of interest for speech understanding mainly lies between 300Hz and 3.4kHz. Hence, a sampling rate at 8kHz is usually sufficient to achieve a good speech intelligibility. In this thesis, however, the sampling rate is fixed to 16kHz in order to comply with most of the current standards for wideband speech systems, *e.g.*, hearing aids.

Speech is a non-stationary signal but it can be assumed to be short-time stationary in the order of 20ms to 30ms. Speech is also an intermittent signal, which means that there is silence between spoken words. In a typical conversation, the silence rate can be considered to be about 50%.

Noise signals

The knowledge on the noise signals is usually less extensive than the knowledge on the speech signal. Indeed, a large variety of noise signals can be present in common listening scenarios. The noise can be broadband or band-limited, intermittent or persistent, stationary or non-stationary. In the case of concurrent speakers, the noise signal can even have the same characteristics as the desired speech signal. Finally, the noise can be diffuse (*i.e.*, coming from every direction, such as wind, road noise in a car. . .) or localised (*i.e.*, one or several localised noise sources such as car engine, broadcasting system. . .). Furthermore, when the noise is localised, the position of the noise sources is usually unknown, whereas the speech source is most commonly assumed to be facing the listener.

1.2 Noise reduction in hearing aids

Every day life listening scenarios usually include more than one sound source. It can be a scenario with multiple speech sources (*e.g.*, cocktail party problem) or a scenario with one speech source and background noise sources (*e.g.*, road traffic, fan, music from a broadcasting system. . .). The desired speech signal is rarely delivered alone and the listener most commonly faces a scenario where he has to understand speech in a noisy environment [23, 43].

Hearing impairments, and more particularly sensorineural losses, can cause a reduction of speech intelligibility. Therefore, a person suffering from sensorineural losses may have a great difficulty to understand speech in a noisy environment. A person affected by mild to severe hearing losses may need a signal-to-noise ratio (SNR) up to 10dB to understand speech, when normal hearing persons are able to understand speech with a SNR down to -5dB [23, 86]. The speech reception threshold (SRT) is a model introduced by Plomp [86], which is defined as the SNR at which 50% of the speech can be correctly understood by the listener. It has been shown in [87] that increasing the SNR by 1dB around the SRT can sometimes be sufficient to generate a significant improvement in everyday communication. Hence, there is obviously a need for algorithms that enhance the speech and get rid of the unwanted noise (*i.e.*, to increase the SNR), the so-called NR algorithms [16, 63].

NR algorithms can be categorised in two types: single channel NR algorithms and multichannel NR algorithms. In the specific case of hearing aids, a third category can be considered that extends the multichannel NR algorithms: the binaural NR algorithms.

1.2.1 Single channel noise reduction

In single channel NR algorithms, only one microphone is available. Therefore, single channel NR algorithms cannot exploit spatial separation between the sound sources and have to rely on differences in signal characteristics (*e.g.*, spectral differences, temporal differences…). Single channel NR algorithms allow to ease listening effort [64] and to increase the SNR. An increase of SNR, however, does not necessarily mean an improved speech intelligibility and single channel NR algorithms are generally reported to provide no improvement in term of speech intelligibility [4, 72, 121]. Also, as speech and noise usually share common spectral components, the SNR improvement owing to a single channel NR algorithm comes at the price of a speech distortion (SD) degradation.

1.2.2 Multichannel noise reduction

Recent commercial BTE and ITE hearing aids can integrate multiple microphones, *i.e.*, at least 2 microphones and even up to 3 microphones in some cases (*e.g.*, Siemens Triano 3). Multichannel NR algorithms can rely on spatial separation between the sound sources as well as on spectral and temporal differences. In the case of hearing aids, the speech sources and the noise sources are usually located at different spatial positions. Therefore, multichannel NR algorithms are preferred to single channel NR schemes as they can take advantage of this spatial separation.

Fixed beamformers

Fixed beamformers compose a first simple class of multichannel NR algorithms which can separate signals coming from different directions [119]. The fixed beamformer tries to steer toward the direction from where the desired speech signal comes (in hearing aids this is typically the forward field of view) and to reject signals coming from other directions. Fixed beamformers are therefore data-independent. They only depend of the direction of arrival of the desired speech signal.

The main categories of fixed beamformers include: delay-and-sum beamformers, filter-and-sum beamformers [48] or superdirective microphone arrays [12, 50].

Adaptive beamformers

Adaptive beamformers try to steer toward the direction of the desired speech signal and to adaptively minimise the contributions from the noise sources coming from other directions. This typically yields a constrained optimisation problem. In a linearly constrained minimum variance beamformer (LCMV) the power of the output signal is minimised while the response for a signal coming from the desired speech direction (most commonly the forward looking direction) is constrained [34]. The LCMV beamformer (also known as Frost beamformer) is then preserving the desired signal while minimising the contribution due to noise signals arriving from directions other than the direction of the desired speech signal.

The generalised sidelobe canceller (GSC) ,*i.e.*, the Griffiths-Jim beamformer, is an alternative approach to the LCMV where the optimisation problem is reformulated as an unconstrained problem [37]. The GSC can be decomposed as a fixed beamformer steering toward the desired speech source, a blocking matrix and a multichannel adaptive filter [41, 69].

- The fixed beamformer produces the so-called speech reference.
- The blocking matrix produces the so-called noise references.
- The multichannel adaptive filter cancels, in the speech reference, the noise components which are correlated to the noise references.

The multichannel adaptive filter needs to be adapted during speech only periods in order to preserve the speech. The multichannel adaptive filter hence exploits the intermittent characteristic of the speech signal and relies on a voice activity detector (VAD) [67]. Since the position of the noise sources and the noise signals characteristics are unknown and can vary over the time, these beamformers are signal dependent and can indeed be referred to as adaptive beamformers. This category of beamformers applied to a two microphone setup has proved its efficiency for hearing aid applications and is still widely used in commercial hearing aids [66, 114].

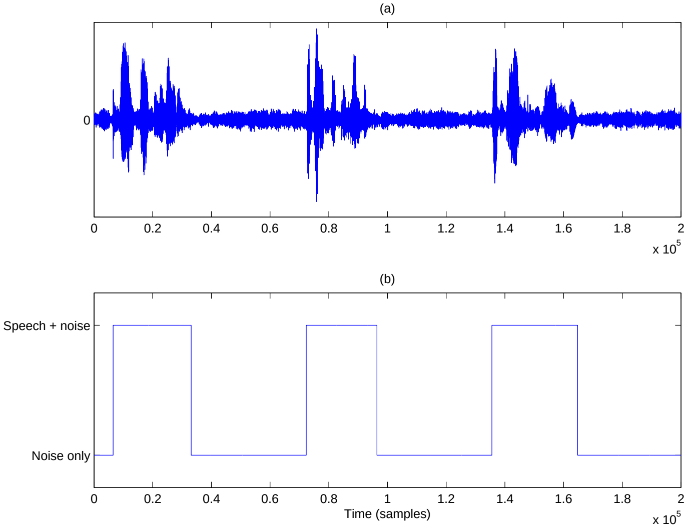


Figure 1.4: (a) Speech and noise signal, (b) Ideal voice activity detector

Multichannel Wiener filters

Multichannel Wiener filters (MWF) represent another class of multichannel NR algorithms which are defined by an unconstrained optimisation problem [10, 18, 20, 81, 93, 107]. The MWF-based NR algorithms minimise a mean squared error criterion (MSE) to compute the optimal estimate of the desired signal. The desired signal is the (unknown) speech component of the signal present in one arbitrarily chosen microphone.

MWF-based NR algorithms take advantage of both spectral differences and spatial separation between the sound sources and can be shown to outperform the GSC under some circumstances [20, 111]. The speech signal and the noise signals, however, often share some spectral components. MWF-based NR algorithms therefore inevitably introduce SD degradation on the output signal. These degradations can be attenuated by designing a filter based on a trade-off between a NR objective and a SD objective. This is the so-called speech distortion weighted multichannel Wiener filter (SDW-MWF)[20, 21, 78].

1.2.3 Binaural noise reduction

It has been known for years that binaural hearing offers advantages over monaural hearing such as: better speech intelligibility, enhanced localisation, improved quality of listening [56, 65, 70]... If such information is really helpful for normal hearing persons, it may even be more important for persons with a hearing impairment.

In the hearing aids framework, the NR techniques mentioned above are usually applied on a monaural setup or on a bilateral setup (when the listener wears two hearing aids working independently) and the SNR improvement can come with a degradation of binaural cues. These binaural cues are helpful for sound source localisation and their disappearance can put the hearing aid user at disadvantage. In a binaural setup, the hearing aid user is wearing two hearing aids which can communicate, *e.g.*, via a wireless link. The NR schemes applied in hearing aids can then be adapted to take advantage of this setup to deliver improved SNR [11, 19, 122] and to preserve binaural localisation cues [55, 118].

1.3 Signal leakage and secondary path

In a hearing aid context, standard NR algorithms (as the algorithms presented above) in general aim at maximising the SNR of the signal to be fed in the hearing aid loudspeaker. The NR algorithms are designed based on the signals recorded by the hearing aid microphones and sometimes based on information on the sound sources localisation. In a hearing aid framework, however, there are some other perturbations to be taken into account (Figure 1.5):

- the signal fed into the hearing aid loudspeaker does not immediately reach the eardrum, it has first to propagate through an acoustic path (the so-called secondary path)
- a portion of the ambient sound can reach directly the eardrum (the so-called signal leakage)

These perturbations are, by design, neglected by standard NR algorithms. They can however have an impact on the performance of a NR scheme applied in a hearing aid framework, especially so with an open fitting BTE.

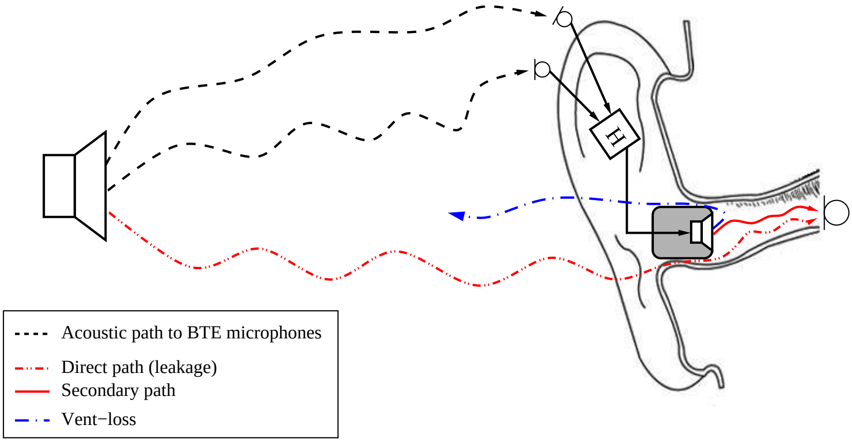


Figure 1.5: Hearing aid with an open fitting

1.3.1 Secondary path

In the hearing aid framework, the so-called secondary path represents the acoustic propagation from the input of the hearing aid loudspeaker to the eardrum. This secondary path includes the acoustic propagation from the loudspeaker to the eardrum (including the loudspeaker response itself) and the effects of the sound transferred from the ear canal to the open field through the fitting (also known as "vent-loss").

In the context of hearing aids with an open fitting there is no earmold left to prevent the sound from leaving the ear canal. Therefore the "vent-loss" is rather important. The secondary path then usually acts as an attenuation and the signal actually reaching the eardrum has lower energy than the signal fed in the hearing aid loudspeaker.

1.3.2 Signal leakage

A hearing aid with an open fitting has no earmold (or an earmold with a large vent path). Therefore, there is nothing to prevent ambient sound from leaking into the ear canal, which results in an additional leakage signal reaching the eardrum. In literature this leakage signal can also be referred to as "vent-through" or "direct sound" [17, 51].

The leakage signal is not processed in the hearing aid, therefore its SNR is generally lower than for the signal provided by the hearing aid, *i.e.*, the output signal of the NR algorithm. Combined with the attenuation caused by the secondary path, this low SNR leakage signal can then override the effects of the NR algorithm in the hearing aid.

In the case of a BTE, the signal leakage is not recorded by the BTE microphones. Therefore no processing can be done in the hearing aid directly on this leakage signal in order to attenuate its noise component.

1.4 Active noise control

Noise barriers are commonly used in soundproofing where they are placed between a noise source and a recipient to damp the noise and prevent it to reach the recipient. One major problem with such passive noise cancellation systems is that their size (*e.g.*, the thickness of the damping material) depends on the frequency components of the noise to be cancelled. The size of a homogeneous material needed to muffle a sound is inversely proportional to the frequency of the sound to be muffled. Therefore, cancelling low frequency noises can require large noise barriers. Perforated panels and Helmholtz resonators can improve sound absorption [39] but they are still rather heavy solutions and can by no means be applied to mobile systems. In 1953, Olson *et al.* [82] were the first to describe an electronic sound absorber prefiguring what should be known later as ANC [25, 26, 59] (which can be also referred to as active noise reduction (ANR)).

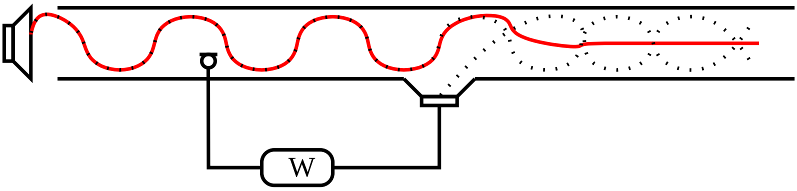


Figure 1.6: Active noise control

ANC relies on destructive interferences and on the superposition principle. Instead of trying to muffle the disturbance coming from the primary source as in a passive system, a secondary source generates a so-called anti-noise (*i.e.*, a signal which is in phase opposition with the signal to be cancelled). The anti-noise is added to the disturbance resulting in the cancellation of both signals (Figure 1.6). In practice it is rarely possible to achieve global control and the noise can only be controlled over a zone surrounding the error microphone. This is the so-called zone of quiet. Outside of the zone of quiet, the interferences are no longer destructive and might even become constructive.

First attempts to actively cancel noise focused on one-dimensional problems, such as noise control in a duct [47, 116], before being extended to three-dimensional control problems in the late 1970's [2, 68]. All these early systems, however, were fixed controllers working as an open loop. Only Onoda and Kido [83] had been working on a closed-loop adaptive system cancelling a pure tone until Burgess' work on an adaptive broadband ANC scheme in 1981 [8]. Since then, with the development of DSP chips, ANC has become a fertile research area.

During the past years, with miniaturisation of electronic components, ANC has appeared in a large variety of commercial applications: fan or engine noise cancellation in factories [57, 117], road noise and car engine noise cancellation in automotive vehicles [14, 15, 27, 58, 84], ear protections, headphones [35, 91]... Depending on the noise scenario (*i.e.*, number of noise sources, noise signal characteristics, number of microphones available...) a different control strategy can be more appropriate: feedback ANC or feedforward ANC.

1.4.1 Feedback active noise control

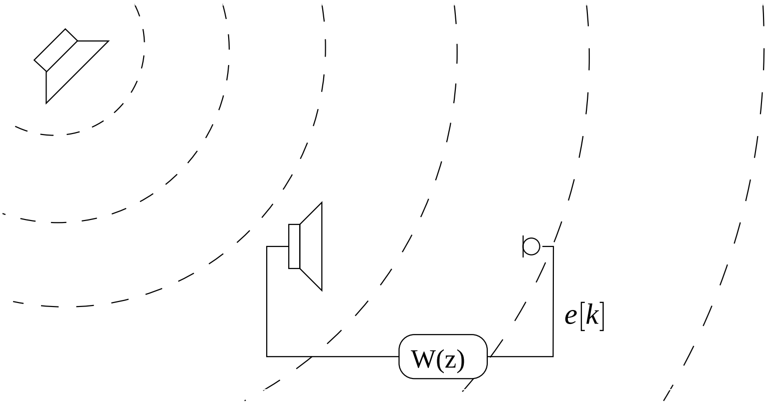
In feedback ANC (Figure 1.7(a)) the signal recorded by the microphone corresponds to a combination of the original noise (coming from the primary source) and of the feedback loop signal (coming from the secondary source). If the secondary path (the path from the secondary source to the microphone) frequency response is flat with no phase shift, the transfer function in the feedback path $W(z)$ is just a gain.

Unfortunately, in real systems, the frequency response of the secondary path is never flat and phase shifts may lead to stability issues (especially at high frequencies) and degrade the ANC performance. A simple feedback ANC would therefore be effective only on a very limited bandwidth. It is possible, however, to implement compensating filters in the feedback loop to overcome the phase shift problem and extend the bandwidth over which it is possible to achieve feedback ANC [9, 123].

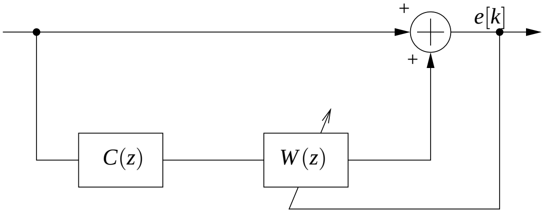
In [31] Forsythe *et al.* have suggested that the feedback ANC can be implemented as a feedforward controller acting as a linear predictor (Figure 1.7(b)). The performance of feedback ANC is therefore highly dependent on the signal predictability [24] and feedback ANC is usually applied to cancel periodic noises such as fan noise, turbine noise, propeller noise in an aircraft...

1.4.2 Feedforward active noise control

In feedforward ANC (Figure 1.8(a)) a reference signal is available from the reference microphones (on the left). If this reference signal is correlated to the



(a)

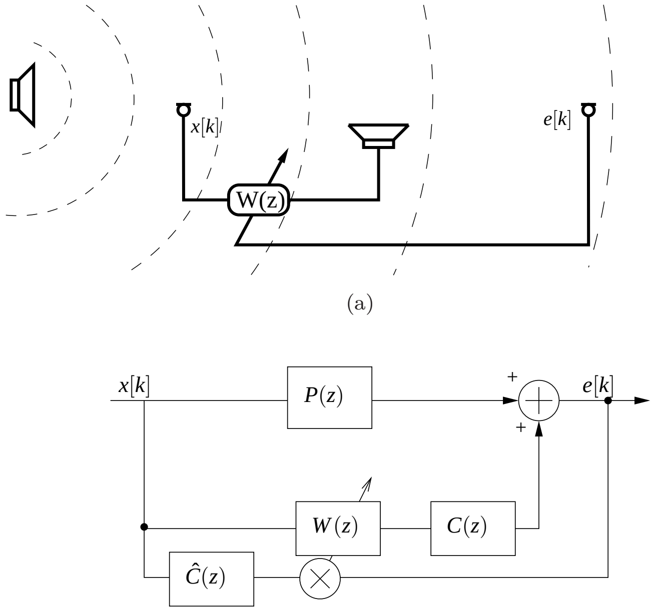


(b)

Figure 1.7: (a) Feedback active noise control. (b) Equivalent feedforward representation

disturbance, it is possible to design a control strategy to cancel the noise at the secondary source. The reference signal is filtered by the controller $W(z)$ to produce the anti-noise which is fed in the secondary source. The error microphone (on the right) is used to monitor the performance of the ANC. Without this microphone the system works in open loop and any change on the secondary path could lead to decreased performance and even to instability.

In practice, the feedforward ANC cannot be implemented based on standard adaptive filter algorithms such as recursive least squares filter (RLS) or least mean squares filter (LMS) [125]. The signal from the controller is subject to phase shifts in the secondary path and using LMS or RLS would lead to an unstable system. A solution is to introduce the same phase shifts in the reference signal [8, 74, 124]. This is achieved by filtering the reference signal by an estimate of the secondary path (Figure 1.8(b)), hence the name Filtered-x LMS (FxLMS).



(a)

(b)

Figure 1.8: (a) Feedforward active noise control. (b) Equivalent Filtered-x structure

Another problem comes from the fact that the secondary source does not emit only in the direction of the error microphone, where the noise is to be cancelled. The secondary source usually also emits sounds toward the reference microphone. The reference signal is therefore corrupted by the signal produced by the secondary source and feedback effects might happen, leading to instabilities. The simplest way to solve this problem is to filter a version of the signal fed in the secondary source by a model of the feedback path and to subtract this filtered signal from the reference signal. This technique is similar to some approaches used in AFC [120].

1.4.3 Active noise control in hearing aids

In hearing aids and especially in open fitting BTE's, the signal leakage which is neglected in standard NR algorithms can have an impact on the NR performance, *i.e.*, on the actual SNR at the eardrum. In a BTE setup the leakage signal is not recorded by the BTE microphones. It is therefore not possible to improve its SNR using standard NR algorithms. The signal in the BTE microphones is, however, highly correlated with the leakage signal. It is therefore possible to attenuate the leakage signal's noise component using feedforward ANC.

Feedforward ANC relies on the presence of an error microphone. In this thesis, 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 (*i.e.*, at the end of the tube which is used to conduct the sound from the BTE to the ear canal). It is then possible to implement a feedforward ANC scheme in the hearing aid, using the BTE microphones as reference microphones and the ear canal microphone as an error microphone.

In the case of hearing aids, performing ANC alone is not sufficient and ANC would have to be performed together with the NR. ANC and NR basically have opposite objectives: the NR aims at producing an output signal with high SNR while the ANC produces an output signal which is a phase opposite version of the noise signal (*i.e.* a signal with low SNR). Therefore, combining ANC and NR in hearing aids is not trivial.

To be effective, a scheme combining ANC and NR also has to satisfy some constraints set by the hearing aid framework itself. Constraints on the number of microphones available and on the separation between the microphones will, *e.g.*, have an effect on the number of noise sources that the scheme can compensate for and its robustness to non-causality.

1.4.4 Causality

The performance of feedforward ANC schemes is highly dependent on the causality of the system [25]. The distance between the reference microphones and the secondary source must be sufficient to allow causal design. In the case of hearing aids with an open fitting, the causality criterion (Figure 1.9) can be defined as follows:

> The acoustic delay from the noise source to the ear canal microphone Δ_{pri} has to be larger than the sum of the acoustic 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 acoustic delay is the direct propagation time between a sound source and a microphone (BTE microphone or ear canal microphone). If the source is about 1 meter away from the listener, Δ_{pri} and Δ_{ref} are about 50 taps (*i.e.*, sampling periods) at $16kHz$ (with $\Delta_{\text{pri}} \geq \Delta_{\text{ref}}$). In hearing aids, Δ_{sec} is just a few taps at $16kHz$ (see also Figure 1.10).

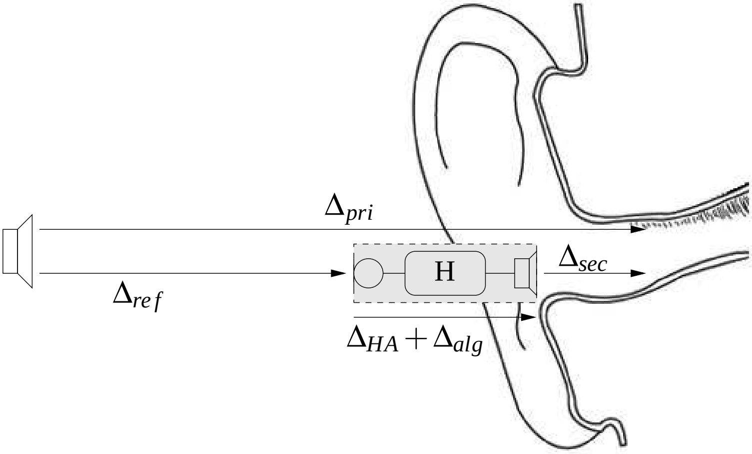


Figure 1.9: Delays in hearing aid system

$$\Delta_{\text{ref}} + \Delta_{\text{HA}} + \Delta_{\text{alg}} + \Delta_{\text{sec}} \leq \Delta_{\text{pri}} \tag{1.1}$$

The causality margin δ is then defined as:

$$\delta = \Delta_{\text{pri}} - (\Delta_{\text{ref}} + \Delta_{\text{HA}} + \Delta_{\text{sec}}) \tag{1.2}$$

, *i.e.*, the delay (number of taps) that can be introduced by the DSP algorithms in the hearing aids such that the system still satisfies (1.1):

$$\Delta_{\text{alg}} \leq \delta \tag{1.3}$$

When $\delta \geq 0$ the DSP algorithms can introduce a delay $\Delta_{\text{alg}} \geq 0$. It is then possible to design a causal ANC. When $\delta < 0$ the ANC has to be designed as a non-causal filter.

In practice, this criterion does not define a hard limit but it gives an indication on the performance to be expected from an ANC scheme. The bandwidth on which it is possible to achieve good ANC performance reduces with the causality margin δ specified in (1.2). When (1.3) is not satisfied, the ANC efficiency vanishes quickly [57]. Delay is thus a critical problem in ANC and many approaches have been developed to try to deal with it [75, 91]. Note that in this thesis the causality problem is investigated from an experimental point of view only and no attempt is made to find a theoretical formulation.

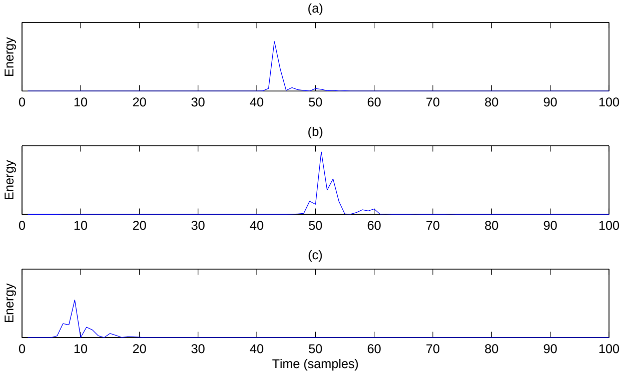


Figure 1.10: (a) Impulse response from speech source to a BTE microphone, (b) Impulse response for the leakage path, (c) Impulse response for the secondary path

In this thesis the hearing aid processing delay (*i.e.*, Analog-to-Digital (A/D) converter delays, Digital-to-Analog (D/A) converter delays. . .) has been neglected ($\Delta_{\mathrm{HA}} = 0$) so as to focus on the performance improvement owing to the integrated approach and all its variations. This is not a realistic assumption but it is a necessary step in the development of these DSP algorithms. The causality criterion (1.4) and the causality margin (1.2) can then be rewritten as follows:

$$\Delta_{\mathrm{ref}} + \Delta_{\mathrm{alg}} + \Delta_{\mathrm{sec}} \leq \Delta_{\mathrm{pri}} \tag{1.4}$$

$$\delta = \Delta_{\mathrm{pri}} - (\Delta_{\mathrm{ref}} + \Delta_{\mathrm{sec}}) \tag{1.5}$$

1.5 Outline of the thesis

This section presents the main research objectives of this thesis. A chapter by chapter overview of the thesis is then given, with reference to the published contributions related to each chapter.

1.5.1 Main research objectives

This thesis introduces an integrated ANC and NR scheme for speech enhancement in hearing aids, to tackle the signal leakage and secondary path problem. This thesis focuses on open fitting BTE's which are the type of hearing aids where the signal leakage impact is the most problematic. This thesis only aims at providing a first proof-of-concept and it is not claimed that the schemes presented here can at this early stage be readily implemented in real-time and studied during listening tests.

In a common listening scenario, sound sources can be moving, especially the noise sources. The algorithms considered here are therefore adaptive algorithms. The small spatial separation between the BTE microphones induces highly correlated BTE microphone signals. Under such constraint a priori information about the desired signal and the microphone characteristics and is hence appealing for hearing aid applications. It has therefore been chosen to focus on MWF-based filters only.

ANC can be used to compensate for the noise component of the leakage signal but in the case of hearing aids, the ANC has to be applied together with the NR. The main objective of this thesis is to introduce a way to combine ANC and NR. First a scheme is described that integrates ANC and NR in a single set of filters. This scheme can work on a very simple scenario. In practice however, some constraints limit the implementation of such a scheme. Variations of the original scheme are then introduced to overcome some of these limitations. Two weighted approaches to integrated ANC and NR are introduced. The first scheme allows to emphasise either the ANC or the NR and the second scheme allows to focus on reducing the SD or to minimise the residual noise at the eardrum. An integrated approach to ANC and NR that is based on an optimisation over a desired zone of quiet is then proposed. This approach allows to overcome the ear canal microphone location problem. Finally, a binaural approach that allows to access extra microphones from the contra-lateral hearing aid is introduced as a solution to the limited number of microphones available on one hearing aid and, to some extent, to the causality problem.

1.5.2 Chapter by chapter overview

Chapter 2 presents an MWF-based NR scheme and the impact of secondary path effects and signal leakage on its SNR performance. The experimental setup and the performance measures are introduced. The performance is analysed theoretically when only one speaker of interest is present (the so-called single speech source scenario). A multichannel ANC based on Filtered-x MWF (FxMWF) is then presented and its performance is analysed. The results presented in this chapter can be found in [99, 101].

In **Chapter 3**, the different ways to combine ANC and NR are compared. The integrated ANC and NR scheme is introduced and its performance is derived in the case of a single speech source. The theoretical performance is compared to the simulation performance. The results presented in this chapter have been published in [101, 98, 99, 95].

Chapter 4 presents weighted approaches to integrated ANC and NR. A first weighted integrated ANC and NR scheme allows to emphasise ANC when no speech is present or to emphasise NR when the ANC is ineffective on the type of noise that is corrupting the speech signal. In a similar way as with SDW-MWF a second weighted approach, the speech distortion weighted integrated ANC and NR scheme (SDW-ANC/NR), allows to focus on ANC or on reducing the SD depending on the weight applied. In a single speech source scenario and when the number of sources (speech source plus noise sources) is less than or equal to the number of microphones, the SDW-ANC/NR can be shown to deliver a constant SNR at the eardrum. The results presented in this chapter can also be found in [96, 102].

In **Chapter 5** the problem of the ear canal microphone location is addressed. To be effective the integrated ANC and NR scheme needs a microphone in the ear canal. The noise is controlled at the ear canal microphone which should ideally be located at the eardrum. In practice, however, the microphone can only be located a few millimetres away from the eardrum and so the sound that reaches the eardrum is basically unknown and uncontrolled. A so-called Zone-of-Quiet approach is presented in this chapter that allows to design a filter to control the sound over a zone away from the ear canal microphone. The results presented in this chapter can be found in [97, 103].

Chapter 6 introduces a binaural extension to the integrated ANC and NR scheme. The monaural integrated ANC and NR scheme performance is limited by the number of microphones on one BTE and by the spatial separation between the BTE microphones and the ear canal microphone. A binaural approach to integrated ANC and NR (based on previous work on binaural MWF) allows to overcome partly these limitations. The results presented in this chapter can also be found in [100]

Finally, **Chapter 7** presents overall conclusions of the thesis and provides suggestions for future research.

Chapter 2

Noise Reduction and Active Noise Control in a Hearing Aid Framework

This chapter presents an MWF-based NR scheme [21] and the impact of secondary path effects and signal leakage on its SNR performance. While such leakage contributions and the secondary acoustic path from the loudspeaker to the eardrum are usually not taken into account in standard NR systems, they appear to have a non-negligible impact on the final SNR. The signal model, the experimental setup and the performance measures are first introduced. A theoretical performance analysis is then provided in the case of a single speech source. A multichannel ANC based on FxMWF is then presented to compensate for the noise component of the leakage signal and its performance is analysed.

Section 2.1 introduces the signal model and the performance measures that are used in this thesis. The MWF-based NR scheme is described in Section 2.2. The effects of the noise leakage and the secondary path on the output of MWF-based NR are commented on in Section 2.3. In Section 2.4 a multichannel ANC is introduced as a solution to tackle the signal leakage problem. Experimental results are presented in Section 2.5 and finally Section 2.6 presents a summary of the chapter.

2.1 Signal Model

This section introduces the signal models (time-domain signal model and frequency-domain signal model) that are used in this thesis. The experimental setup and the measures used to compare the performance of the different schemes along this thesis are then presented.

2.1.1 Time-domain model

Let M be the number of microphones (channels) in one hearing aid and N the filter length. The number of sound sources (speech plus noise sources) will be denoted by Q. The time-domain signal x_m for microphone m has a desired speech part x_m^s and an additive noise part x_m^n, *i.e.*,:

$$x_m[k] = x_m^s[k] + x_m^n[k] \quad m \in \{1 \dots M\} \tag{2.1}$$

where k is the discrete time index. The discrete time index is related to the actual time by:

$$t = \frac{k}{f_s} \tag{2.2}$$

with f_s the sampling frequency.

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 (2.1) 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.

In practice, in order to distinguish “speech plus noise periods” from “noise only periods” it is necessary to use a VAD. The performance of the VAD can affect the performance of the MWF filters [110]. In this thesis, however, as research on VAD is a vast subject of itself, it has been chosen to assume that a perfect VAD is available and to focus on the performance improvement owing to the introduction of ANC in hearing aids and to the different variations on the integrated ANC and NR scheme.

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

$$\mathbf{x}_m[k] = [x_m[k] \dots x_m[k-N+1]]^T \quad m \in \{1 \dots M\} \tag{2.3}$$

where ^{T} denotes the transpose operator. The compound vector gathering all channels is:

$$\mathbf{x}[k] = [\mathbf{x}_1^T[k] \dots \mathbf{x}_M^T[k]]^T \tag{2.4}$$

An NM-dimensional MWF $\mathbf{w}[k] = [\mathbf{w}_1^T[k] \ldots \mathbf{w}_M^T[k]]^T$ will be designed and applied to these signals, which minimises an MSE criterion:

$$J_{\mathrm{MSE}}[k] = \mathbb{E}\{|e[k]|^2\} \tag{2.5}$$

Where $e[k]$ is an error signal to be defined next, depending on the scheme applied, and $\mathbb{E}\{.\}$ is the expectation operator.

The filter output signal $z[k]$ (*i.e.*, the signal to be fed into the hearing aid loudspeaker) is defined as:

$$z[k] = \mathbf{w}[k]^T \mathbf{x}[k] \tag{2.6}$$

The autocorrelation matrices of the speech component and the noise component of the microphone signals are given by:

$$\mathbf{R}_{x^s}[k] = \mathbb{E}\{\mathbf{x}^s[k]\mathbf{x}^s[k]^T\} \tag{2.7}$$

$$\mathbf{R}_{x^n}[k] = \mathbb{E}\{\mathbf{x}^n[k]\mathbf{x}^n[k]^T\} \tag{2.8}$$

In a stationary scenario, and if the speech signal and the noise signal are uncorrelated, $\mathbf{R}_{x^n}$ can be estimated during ”noise only periods“ and $\mathbf{R}_{x^s}$ can be estimated during ”speech plus noise periods“ using:

$$\mathbf{R}_x[k] = \mathbb{E}\{\mathbf{x}[k]\mathbf{x}[k]^T\} \tag{2.9}$$

$$\mathbf{R}_{x^s}[k] = \mathbf{R}_x[k] - \mathbf{R}_{x^n}[k] \tag{2.10}$$

In practice, the correlation matrices are estimated recursively. The estimate of the autocorrelation matrix of the microphone signals is updated during “speech plus noise periods”, using:

$$\tilde{\mathbf{R}}_x[k] = \lambda_t \tilde{\mathbf{R}}_x[k-1] + (1-\lambda_t)\mathbf{x}[k]\mathbf{x}[k]^T \tag{2.11}$$

where $\lambda_t \in [0\,1]$ is an exponential forgetting factor that depends on the number of past samples to be taken into account. Here $\lambda_t = 1 - \frac{1}{N_{\mathrm{samples}}}$ with $N_{\mathrm{samples}} = 20000$ samples. This clearly exceeds the stationarity of speech signals but it also allows to have a stable estimation for the correlation matrices, which is the main concern at this stage. In “noise only periods” $\tilde{\mathbf{R}}_x$ is not updated, i.e., $\tilde{\mathbf{R}}_x[k] = \tilde{\mathbf{R}}_x[k-1]$.

The estimate of the autocorrelation matrix of the noise component of the microphone signals is updated similarly during “noise only periods”, using:

$$\tilde{\mathbf{R}}_{x^n}[k] = \lambda_t \tilde{\mathbf{R}}_{x^n}[k-1] + (1-\lambda_t)\mathbf{x}[k]\mathbf{x}[k]^T \tag{2.12}$$

$$= \lambda_t \tilde{\mathbf{R}}_{x^n}[k-1] + (1-\lambda_t)\mathbf{x}^n[k]\mathbf{x}^n[k]^T \tag{2.13}$$

In “speech plus noise periods” $\tilde{\mathbf{R}}_{x^n}$ is not updated, i.e., $\tilde{\mathbf{R}}_{x^n}[k] = \tilde{\mathbf{R}}_{x^n}[k-1]$.

The estimate of the autocorrelation matrix of the speech component of the microphone signal is then given by:

$$\tilde{\mathbf{R}}_{x^s}[k] = \tilde{\mathbf{R}}_{x}[k] - \tilde{\mathbf{R}}_{x^n}[k] \tag{2.14}$$

2.1.2 Frequency-domain model

The frequency-domain representation of a signal $x[k]$ is obtained with the discrete-time Fourier transform (DTFT):

$$X(\omega) = \mathcal{F}\{x[k]\} = \sum_{k=-\infty}^{\infty} x[k]\mathrm{e}^{-j\omega k} \tag{2.15}$$

where $\omega = 2\pi f$ is the frequency-domain variable.

The spectrum $X(\omega)$ is 2π-periodic in ω and the frequency $\omega = \pi$ represents the Nyquist-frequency. The spectrum $X(\omega)$ can be transformed back to time-domain with the inverse discrete time Fourier transform (IDTFT):

$$x[k] = \mathcal{F}^{-1}\{X(\omega)\} = \frac{1}{2\pi}\int_{-\pi}^{\pi} X(\omega)\mathrm{e}^{j\omega k}\mathrm{d}\omega \tag{2.16}$$

In practice, the spectrum $X(\omega)$ is approximated by applying the discrete Fourier transform (DFT) on a windowed frame of the length N_{DFT} of the signal $x[k]$. This is referred to as the short-time Fourier transform (STFT). The lth component of the DFT of the pth frame of $x[k]$ is computed as follows:

$$X(l,p) = \sum_{k=0}^{N_{\mathrm{DFT}}-1} h[k]x[pN_{\mathrm{DFT}}+k]\mathrm{e}^{\frac{-j2\pi kl}{N_{\mathrm{DFT}}}} \quad l \in \{0 \ldots N_{\mathrm{DFT}}-1\} \tag{2.17}$$

where $h[k]$ is a window function (*e.g.*, rectangular, Hann, Blackman,…) used to attenuate some of the effects of the DFT approximation.

The pth frame of time-domain signal $x[k]$ can by obtained from the discrete spectrum $X(l,p)$ by applying the inverse discrete Fourier transform (IDFT):

$$x[pN_{\mathrm{DFT}}+k] = \frac{1}{N_{\mathrm{DFT}}} \sum_{l=0}^{N_{\mathrm{DFT}}-1} X(l,p)\mathrm{e}^{\frac{j2\pi kl}{N_{\mathrm{DFT}}}} \; ; \; k \in \{0 \ldots N_{\mathrm{DFT}}-1\} \tag{2.18}$$

Both the DFT and the IDFT can be computed efficiently using the fast Fourier transform (FFT) and the inverse fast Fourier transform (IFFT), respectively. In the remainder of this thesis, only the actual Fourier transform $X(\omega)$ is used in the derivations for conciseness. During the simulations, the length of the DFT is set to $N_{\mathrm{DFT}} = 2N$ so that time-domain filters and frequency-domain filters have the same length. A $2N$-coefficient frequency-domain filter indeed corresponds to a time-domain filter with a N-coefficient causal part and a N-coefficient anticausal part. The anticausal part is then set to zero (truncated) to obtain a N-coefficient causal time-domain filter. Any frequency-domain analysis therefore corresponds to a (causal) time-domain analysis to the extent that the effect of this truncation is limited.

The frequency-domain signal $X_m(\omega)$ for microphone m has a desired speech part $X_m^s(\omega)$ and an additive noise part $X_m^n(\omega)$, i.e.:

$$X_m(\omega) = X_m^s(\omega) + X_m^n(\omega) \quad m \in \{1 \dots M\} \tag{2.19}$$

The compound vector gathering all microphone signals is:

$$\mathbf{X}(\omega) = [X_1(\omega) \dots X_M(\omega)]^T \tag{2.20}$$

An MWF $\mathbf{W}(\omega) = [W_1(\omega) \dots W_M(\omega)]^T$ will be designed and applied to these signals, which minimises an MSE criterion:

$$J_{\mathrm{MSE}}(\omega) = \mathbb{E}\{|E(\omega)|^2\} \tag{2.21}$$

Where $E(\omega)$ is an error signal to be defined next, depending on the scheme applied.

The filter output signal $Z(\omega)$ is defined as:

$$Z(\omega) = \mathbf{W}(\omega)^H \mathbf{X}(\omega) \tag{2.22}$$

where ^{H} denotes the Hermitian transpose.

The autocorrelation matrices of the speech component and the noise component of the microphone signals are respectively given by:

$$\mathbf{R}_{X^s}(\omega) = \mathbb{E}\{\mathbf{X}^s(\omega)\mathbf{X}^s(\omega)^H\} \tag{2.23}$$

$$\mathbf{R}_{X^n}(\omega) = \mathbb{E}\{\mathbf{X}^n(\omega)\mathbf{X}^n(\omega)^H\} \tag{2.24}$$

In a stationary scenario, and if the speech signal and the noise signal are uncorrelated, $\mathbf{R}_{X^n}(\omega)$ can be estimated during ”noise only periods“ and $\mathbf{R}_{X^s}(\omega)$ can be estimated during ”speech plus noise periods“ using:

$$\mathbf{R}_X(\omega) = \mathbb{E}\{\mathbf{X}(\omega)\mathbf{X}(\omega)^H\} \tag{2.25}$$

$$\mathbf{R}_{X^s}(\omega) = \mathbf{R}_X(\omega) - \mathbf{R}_{X^n}(\omega) \tag{2.26}$$

In practice, the correlation matrices are estimated recursively. The estimate of the autocorrelation matrix of the microphone signals is updated during “speech plus noise periods”, using:

$$\tilde{\mathbf{R}}_X(\omega) = \lambda_f \tilde{\mathbf{R}}_X(\omega) + (1 - \lambda_f)\mathbf{X}(\omega)\mathbf{X}(\omega)^H \tag{2.27}$$

where $\lambda_f \in [0\,1]$ is an exponential forgetting factor that depends on the number of past samples to be taken into account. Here $\lambda_f = 1 - \frac{1}{N_{\text{samples}}}$ with $N_{\text{samples}} =$ 20000 samples. In “noise only periods” $\tilde{\mathbf{R}}_x$ is not updated.

The estimate of the autocorrelation matrix of the noise component of the microphone signals is updated similarly during “noise only periods”, using:

$$\begin{aligned} \tilde{\mathbf{R}}_{X^n}(\omega) &= \lambda_f \tilde{\mathbf{R}}_{X^n}(\omega) + (1 - \lambda_f)\mathbf{X}(\omega)\mathbf{X}(\omega)^H && (2.28) \\ &= \lambda_f \tilde{\mathbf{R}}_{X^n}(\omega) + (1 - \lambda_f)\mathbf{X}(\omega)^n \mathbf{X}^n(\omega)^H && (2.29) \end{aligned}$$

In “speech plus noise periods” $\tilde{\mathbf{R}}_{x^n}(\omega)$ is not updated.

The estimate of the autocorrelation matrix of the speech component of the microphone signal is then given by:

$$\tilde{\mathbf{R}}_{X^s}(\omega) = \tilde{\mathbf{R}}_X(\omega) - \tilde{\mathbf{R}}_{X^n}(\omega) \tag{2.30}$$

2.1.3 Experimental setup and performance measures

Experimental setup

All the simulations presented in this thesis were run on acoustic path measurements obtained with a CORTEX MK2 manikin head and torso equipped with artificial ears and a two-microphone BTE hearing aid. The sound sources (FOSTEX 6301B loudspeakers) were positioned at 1 meter from the centre of the head. The speech source was located at 0° and the noise sources at 30°, 90°, 270° and 330°. The eardrum at which the noise is to be controlled here is the left eardrum, facing the noise source at 270°. 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 seconds long signals. The speech was composed of three sentences from the HINT database [80] concatenated with silence periods. The noise was the multitalker babble from Auditec [3]. All the signals were sampled at 16kHz.

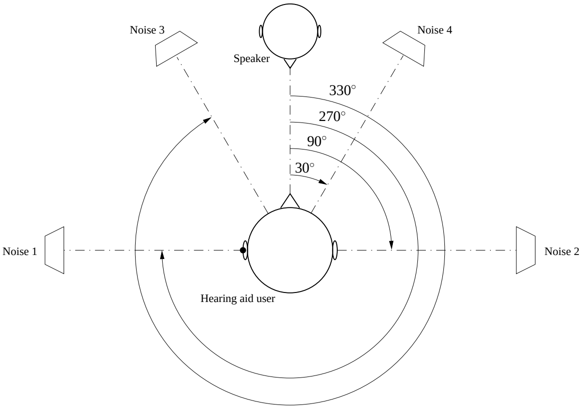


Figure 2.1: Experimental setup

Residual noise power

A first performance measure is the reduction of the residual noise power at the eardrum. The reference power is the power of the leakage signal's noise component, which can also be considered as the residual noise power at the eardrum when the hearing aid is turned off. The performance measure is then written as:

$$\Delta\mathrm{Pow}^n = \mathrm{Pow}^n_{\mathrm{eardrum}} - \mathrm{Pow}^n_{\mathrm{leak}} \tag{2.31}$$

where $\mathrm{Pow}^n_{\mathrm{eardrum}}$ and $\mathrm{Pow}^n_{\mathrm{leak}}$ are the power (in dB) of the residual noise at the eardrum for a particular processing scheme and the power of the noise component of the leakage signal at the eardrum. In general, the power of a signal x is given by:

$$\mathrm{Pow}_x[k] = 10\log_{10}(\mathbb{E}\{|x[k]|^2\}) \tag{2.32}$$

$$\mathrm{Pow}_X(\omega) = 10\log_{10}(\mathbb{E}\{|X(\omega)|^2\}) \tag{2.33}$$

Signal-to-noise ratio

A second performance measure (*e.g.*, used for NR and integrated ANC and NR schemes) is the intelligibility-weighted SNR [36]. Evaluation of NR in hearing aids can rely on a variety of performance measures and most commonly perceptual measures such as the SRT, listening effort scaling or preference rating. Evaluating perceptual measures requires listening tests which were not carried out during this thesis. The intelligibility-weighted SNR, however, has been shown to be correlated with the SRT and is therefore used here as a performance measure.

The leakage signal SNR, which can also be considered as the SNR when the hearing aid is turned off, is taken as a reference measure. The intelligibility-weighted SNR improvement can then be defined as:

$$\Delta\mathrm{SNR}_{\mathrm{intell}} = \sum_i I_i(\mathrm{SNR}_{\mathrm{i,eardrum}} - \mathrm{SNR}_{\mathrm{i,leak}}) \tag{2.34}$$

where I_i is the band importance function defined in [1] and $\mathrm{SNR}_{\mathrm{i,eardrum}}$ and $\mathrm{SNR}_{\mathrm{i,leak}}$ represent the output SNR at the eardrum for a particular processing scheme and the leakage SNR (in dB), respectively, of the ith one third octave band. In general the SNR of a signal $x[k]$ is given by:

$$\mathrm{SNR}_x[k] = 10\log_{10}\frac{\mathbb{E}\{|x^s[k]|^2\}}{\mathbb{E}\{|x^n[k]|^2\}} \tag{2.35}$$

$$\mathrm{SNR}_X(\omega) = 10\log_{10}\frac{\mathbb{E}\{|X^s(\omega)|^2\}}{\mathbb{E}\{|X^n(\omega)|^2\}} \tag{2.36}$$

Spectral distortion

All the schemes described in this thesis rely on MWF and therefore have an effect on speech distortion. An intelligibility weighted spectral distortion measure is used to analyse the effect of the different schemes on the signal in terms of speech distortion. The spectral distortion measure is defined as follows:

$$\mathrm{SPD}_{\mathrm{intell}} = \sum_i I_i \mathrm{SPD}_i \tag{2.37}$$

where I_i is the band importance function defined in [1] and SPD_i the average spectral distortion (in dB) in the i-th one third octave band,

$$\mathrm{SPD}_i = \frac{1}{(2^{1/6} - 2^{-1/6})f_i^c}\int_{2^{-1/6}f_i^c}^{2^{1/6}f_i^c} |10\log_{10} G^s(f)|df \tag{2.38}$$

with centre frequencies f_i^c and $G^s(f)$ the squared magnitude of the transfer function for the speech component from the input to the output of the processing scheme.

2.2 Multichannel Wiener filter-based noise reduction

2.2.1 Time-domain formulation

The MWF is an unconstrained multichannel optimal (in the MSE sense) filtering technique. The aim of the MWF (Figure 2.2) is to estimate an unknown desired speech signal based on the microphone signals $\mathbf{x}[k]$. The desired signal $d_{NR}[k]$ is arbitrarily chosen to be equal to the (unknown) speech component in the first microphone, up to a delay:

$$d_{NR}[k] = G \cdot x_1^s[k-\Delta] \tag{2.39}$$

where the gain G is the amplification that compensates for the hearing loss and Δ is the NR delay.

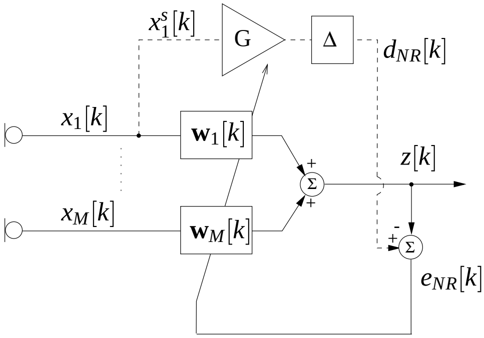


Figure 2.2: Multichannel Wiener filter-based noise reduction

The MWF-based NR is designed to minimise the squared distance between the filtered microphone signal $z[k]$ and the desired speech signal $d_{NR}[k]$. Therefore, the MSE criterion to be minimised is:

$$J_{\mathrm{NR}}[k] = \mathbb{E}\{|e_{\mathrm{NR}}[k]|^2\} \tag{2.40}$$

$$e_{\mathrm{NR}}[k] = z[k] - d_{NR}[k] \tag{2.41}$$

$$= \mathbf{w}[k]^T \mathbf{x}[k] - d_{NR}[k] \tag{2.42}$$

The corresponding Wiener filter is:

$$\boxed{\mathbf{w}_{\mathrm{NR}}[k] = \mathbf{R}_x[k]^{-1}\mathbf{r}_{xd_{NR}}[k]} \tag{2.43}$$

Here $\mathbf{R}_x[k]$ is the correlation matrix of the input $\mathbf{x}[k]$ (as defined in 2.9) and $\mathbf{r}_{xd_{NR}}[k]$ is the cross-correlation vector between the input $\mathbf{x}[k]$ and the desired signal $d_{NR}[k]$:

$$\mathbf{r}_{xd_{NR}}[k] = E\{\mathbf{x}[k]d_{NR}[k]\} \tag{2.44}$$

In a stationary scenario, and if the speech signal and the noise signal are uncorrelated the cross-correlation vector can be estimated using:

$$\mathbf{r}_{xd_{NR}}[k] = G\cdot(\mathbf{r}_{xx_{1,\Delta}}[k] - \mathbf{r}_{x^n x^n_{1,\Delta}}[k]) \tag{2.45}$$

$$\mathbf{r}_{xx_{1,\Delta}}[k] = \mathbb{E}\{\mathbf{x}[k]x_1[k-\Delta]\} \tag{2.46}$$

$$\mathbf{r}_{x^n x^n_{1,\Delta}}[k] = \mathbb{E}\{\mathbf{x}^n[k]x^n_1[k-\Delta]\} \tag{2.47}$$

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

In practice the NR delay Δ is an integer, smaller than the filter length N (the NR delay is usually set to $\Delta = \frac{N}{2}$). The cross-correlation vectors $\mathbf{r}_{x_{1,\Delta}}[k]$ and $\mathbf{r}_{x^n x^n_{1,\Delta}}[k]$ can then be expressed through the autocorrelation matrix of the microphone signals $\mathbf{R}_x[k]$ and the autocorrelation matrix of the noise component of the microphone signals $\mathbf{R}_{x^n}[k]$, respectively:

$$\mathbf{r}_{xx_{1,\Delta}}[k] = \mathbf{R}_x[k]\mathbf{e}_\Delta \tag{2.48}$$

$$\mathbf{r}_{x^n x^n_{1,\Delta}}[k] = \mathbf{R}_{x^n}[k]\mathbf{e}_\Delta \tag{2.49}$$

where $\mathbf{e}_\Delta$ is the N-dimensional column vector with the Δth element equal to 1 and all the other elements equal to 0.

The MWF can finally be expressed as follows:

$$\boxed{\mathbf{w}_{\mathrm{NR}}[k] = G\cdot\mathbf{R}_x[k]^{-1}(\mathbf{R}_x[k] - \mathbf{R}_{x^n}[k])\mathbf{e}_\Delta} \tag{2.50}$$

2.2.2 Frequency-domain formulation

In the single speech source case it is possible to derive simpler frequency-domain formulae for the MWF-based filters and for their SNR performance. Besides, in the context of hearing aids, it can be desired to apply a frequency-dependent gain $G(\omega)$. Therefore, it is needed to derive a representation of the MWF in the frequency-domain. The desired speech signal is still arbitrarily chosen to be the (unknown) speech component of the first microphone signal ($m = 1$), up to a delay

Δ. This can now be written as:

$$D_{\mathrm{NR}}(\omega) = \mathbf{G}_{1,\Delta}(\omega)^H \mathbf{X}^s(\omega) \tag{2.51}$$

$$\mathbf{G}_{1,\Delta}(\omega) = [G(\omega)\mathrm{e}^{j\omega\Delta}|0\dots0]^T \tag{2.52}$$

$$= G(\omega)\cdot \mathbf{e}_{1,\Delta}(\omega) \tag{2.53}$$

where $\mathbf{e}_{1,\Delta}(\omega)$ is the N-dimensional column vector with the first element equal to $\mathrm{e}^{j\omega\Delta}$.

The MWF aims to minimise the squared distance between the filtered microphone signal $Z(\omega)$ and the desired speech signal $D_{\mathrm{NR}}(\omega)$. The corresponding MSE criterion is:

$$J_{\mathrm{MSE}}(\omega) = \mathbb{E}\{|E_{\mathrm{NR}}(\omega)|^2\} \tag{2.54}$$

$$E_{\mathrm{NR}}(\omega) = \mathbf{W}(\omega)^H \mathbf{X}(\omega) - \mathbf{G}_{1,\Delta}(\omega)^H \mathbf{X}^s(\omega) \tag{2.55}$$

The MWF solution is given as:

$$\boxed{\mathbf{W}_{\mathrm{NR}}(\omega) = \mathbf{R}_X(\omega)^{-1}(\mathbf{R}_X(\omega) - \mathbf{R}_{X^n}(\omega))\mathbf{G}_{1,\Delta}(\omega)} \tag{2.56}$$

2.2.3 Single speech source case

In the single speech source case, the autocorrelation matrix of the speech component of the microphone signal $\mathbf{R}_{X^s}(\omega)$ is rank-1 and can be rewritten as:

$$\mathbf{R}_{X^s}(\omega) = P^s(\omega)\mathbf{A}(\omega)\mathbf{A}(\omega)^H \tag{2.57}$$

where $P^s(\omega)$ is the power of the speech signal and $\mathbf{A}(\omega)$ is the M-dimensional steering vector, which contains the acoustic transfer functions from the speech source position to the hearing aid microphones (including room acoustics, microphone characteristics, and head shadow effect).

The Woodbury identity can be applied to compute the inverse of the matrix pencil $(\mathbf{R}_X(\omega) = \mathbf{R}_{X^s}(\omega) + \mathbf{R}_{X^n}(\omega))$ [11]. The filter (2.56) can then be rewritten as follows (see Appendix A.1):

$$\boxed{\mathbf{W}_{\mathrm{NR}}(\omega) = \frac{\mathbf{R}_{X^n}(\omega)^{-1}\mathbf{R}_{X^s}(\omega)}{\rho(\omega)+1}\mathbf{G}_{1,\Delta}(\omega)} \tag{2.58}$$

with

$$\rho(\omega) = P^s(\omega)\mathbf{A}(\omega)^H \mathbf{R}_{X^n}(\omega)^{-1}\mathbf{A}(\omega) \tag{2.59}$$

In the absence of signal leakage and secondary path effects the output SNR of the system is given by:

$$\mathrm{SNR}_{\mathrm{NR(noLeakage)}}(\omega) = \frac{\mathbf{W}_{\mathrm{NR}}(\omega)^H \mathbf{R}_{X^s}(\omega) \mathbf{W}_{\mathrm{NR}}(\omega)}{\mathbf{W}_{\mathrm{NR}}(\omega)^H \mathbf{R}_{X^n}(\omega) \mathbf{W}_{\mathrm{NR}}(\omega)} \tag{2.60}$$

This expression reduces to (see Appendix A.2):

$$\boxed{\mathrm{SNR}_{\mathrm{NR(noLeakage)}}(\omega) = \rho(\omega)} \tag{2.61}$$

This result is similar to the SNR formula in [11] and does not depend on the gain $G(\omega)$. This means that, under a single speech source scenario, the MWF delivers a constant output SNR for any gain. This result, however, does not include the effects of the signal leakage and the secondary path effects, as discussed in the next section.

2.3 Secondary path and leakage

A standard NR scheme in general aims at maximising the SNR of the signal $z[k]$ to be fed in the hearing aid loudspeaker. The signal $z[k]$, however, does not reach the eardrum directly. It first has to propagate through an acoustic path and encounter some perturbations which are, by design, neglected in standard NR schemes. The NR scheme based on MWF, as applied in the hearing aid context, is presented in Figure. 2.3.

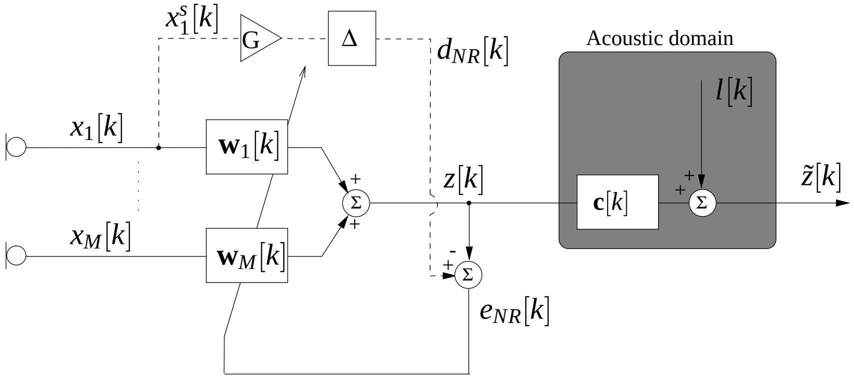


Figure 2.3: Multichannel noise reduction system in the hearing aids context

2.3.1 Time-domain formulation

Secondary path

The secondary path includes the acoustic propagation from the loudspeaker to the eardrum (including the loudspeaker response itself) and the effects of the sound transferred from the ear canal to the open field through the fitting (also know as "vent-loss"). Assuming that the loudspeaker characteristic is approximately linear, the secondary path can be represented by a filter coefficient vector $\mathbf{c}[k]$ of length P.

$$\mathbf{c}[k] = [c_0[k] \dots c_{P-1}[k]]^T \tag{2.62}$$

Corresponding to a transfer function:

$$\hat{C}(z) = \sum_{i=0}^{L-1} \hat{c}_i[k] z^{-i} \tag{2.63}$$

The secondary path can be estimated off-line using classic identification methods based for example on LMS algorithms, or on-line by adding random noise to the signal exciting the secondary path, as introduced by Eriksson et al. in [29] and later refined by Kuo et al. [60] and Zhang et al. [126].

In this thesis the secondary path $\mathbf{c}[k]$ is estimated off-line though a practical implementation probably requires an on-line estimation. The secondary path estimate corresponds to a filter coefficient vector of length $\hat{P}$:

$$\hat{\mathbf{c}}[k] = [\hat{c}_0[k] \dots \hat{c}_{\hat{P}-1}[k]]^T \tag{2.64}$$

In this thesis $\hat{P} = 32$.

In the context of hearing aids with an open fitting, the "vent-loss" is significant and the secondary path usually acts as an attenuation. The power of the signal actually reaching the eardrum is lower than the power of the output signal $z[k]$ when taking the secondary path into account.

Signal leakage

The hearing aid with an open fitting has no earmold to prevent ambient sound from leaking into the ear canal, which results in an additional leakage signal $l[k]$ reaching the eardrum [17]. In the literature this leakage signal can also be referred to as "vent-through" or "direct sound" [17, 51].

The leakage signal is not processed in the hearing aid, therefore its SNR is generally lower than for the signal provided by the hearing aid, *i.e.*, the output signal of the NR algorithm. In the case of a BTE, the signal leakage is not recorded by the BTE microphones. Therefore no processing can be done in the hearing aid directly on this leakage signal in order to attenuate its noise component.

Signal at the eardrum

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

$$\tilde{z}[k] = \mathbf{c}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-P+1] \end{bmatrix} + l[k] \tag{2.65}$$

It clearly appears that for small gains G the attenuation caused by the secondary path and the additive leakage signal contribution do matter. The leakage signal SNR may affect the output signal SNR thus partly cancelling the improvement achieved with the NR scheme in the hearing aid.

In conclusion, whereas the secondary path and the leakage are not taken into account in conventional NR algorithms, they may degrade their performance significantly (see also Figure 2.7(a)). These degradations appear for every type of hearing aid but they are the most significant in BTE hearing aids with an open fitting where the earmold is removed (see also Figure 2.7(b)).

2.3.2 Frequency-domain formulation and single speech source case

In a single speech source scenario, the MWF-based NR filter reduces to (2.58) and it is possible to derive a simple formula for its output SNR. In this section, the frequency-domain representation of the problem is used and the frequency-domain leakage signal L will be used instead of the time-domain leakage signal $l[k]$. The signal reaching the eardrum can then be expressed as follows:

$$\tilde{Z}(\omega) = C(\omega)Z(\omega) + L(\omega) \tag{2.66}$$

Signal leakage

Taking only the signal leakage (and no secondary path) effects into account, the signal reaching the eardrum can be expressed as follows:

$$\tilde{Z}(\omega) = Z(\omega) + L(\omega) = \mathbf{W}(\omega)^H \mathbf{X}(\omega) + L(\omega) \tag{2.67}$$

The output SNR of the standard NR scheme can then be rewritten (see Appendix A.3):

$$\mathrm{SNR}_{\mathrm{NR(Leakage)}}(\omega) = \frac{\mathbb{E}\{|\mathbf{W}(\omega)^H \mathbf{X}^s(\omega) + L^s(\omega)|^2\}}{\mathbb{E}\{|\mathbf{W}(\omega)^H \mathbf{X}^n(\omega) + L^n(\omega)|^2\}} \tag{2.68}$$

$$\boxed{\mathrm{SNR}_{\mathrm{NR(Leakage)}}(\omega) = \frac{\frac{\rho(\omega)^2}{(\rho(\omega)+1)^2} P_{D_{\mathrm{NR}}}(\omega) + \frac{\rho(\omega)}{\rho(\omega)+1}\alpha(\omega) + P_{L^s}(\omega)}{\frac{\rho(\omega)}{(\rho(\omega)+1)^2} P_{D_{\mathrm{NR}}}(\omega) + \frac{1}{\rho(\omega)+1}\alpha(\omega) + P_{L^n}(\omega)}} \tag{2.69}$$

with

$$\begin{aligned} P_{D_{\mathrm{NR}}}(\omega) &= \mathbf{G}_{1,\Delta}(\omega)^H \mathbf{R}_{X^s}(\omega) \mathbf{G}_{1,\Delta}(\omega) && (2.70) \\ &= G(\omega)^2 \mathbb{E}\{X_1^s(\omega) X_1^s(\omega)^H\} && (2.71) \\ P_{L^s}(\omega) &= \mathbb{E}\{|L^s(\omega)|^2\} && (2.72) \\ P_{L^n}(\omega) &= \mathbb{E}\{|L^n(\omega)|^2\} && (2.73) \end{aligned}$$

where $P_{D_{\mathrm{NR}}(\omega)}$ is the power of the desired speech signal and $P_{L^s}(\omega)$ and $P_{L^n}(\omega)$ are the power of the speech component and the noise component of the leakage signal, respectively.

$$\begin{aligned} \alpha(\omega) &= \mathbf{G}_{1,\Delta}(\omega)^H \mathbf{r}_{sl}(\omega) + \mathbf{r}_{sl}(\omega)^H \mathbf{G}_{1,\Delta}(\omega) && (2.74) \\ &= G(\omega)(\mathbf{e}_{1,\Delta}(\omega)^H \mathbf{r}_{sl}(\omega) + \mathbf{r}_{sl}(\omega)^H \mathbf{e}_{1,\Delta}(\omega)) && (2.75) \end{aligned}$$

The cross-correlation vector $\mathbf{r}_{sl}(\omega)$ between the speech component of the microphone signals and the speech component of the leakage signal is given by:

$$\mathbf{r}_{sl}(\omega) = \mathbb{E}\{\mathbf{X}^s(\omega) L^s(\omega)^*\} \tag{2.76}$$

where * denotes the conjugate of a complex number.

If the speech component and the noise component of the microphone signals and the leakage signal are uncorrelated, $\mathbf{r}_{sl}(\omega)$ can be estimated using:

$$\mathbf{r}_{sl}(\omega) = \mathbb{E}\{\mathbf{X}(\omega) L(\omega)^*\} - \mathbb{E}\{\mathbf{X}^n(\omega) L^n(\omega)^*\} \tag{2.77}$$

where $\mathbb{E}\{\mathbf{X}(\omega)L(\omega)^*\}$ is estimated during “speech plus noise periods” and $\mathbb{E}\{\mathbf{X}^n(\omega)L^n(\omega)^*\}$ and is estimated during “noise only periods”. The leakage signal can be estimated from the ear canal microphone signal $\tilde{Z}(\omega)$:

$$L(\omega) = \tilde{Z}(\omega) - Z(\omega) \tag{2.78}$$

From equations (2.71), (2.75) and (2.69) one can observe that the output SNR is a rational function of the gain $G(\omega)$. The denominator and the numerator of this function are both of degree 2. Therefore, the two extreme cases for the output SNR of the filter (2.58), when the signal leakage is taken into account, are given by:

$$\lim_{G(\omega)\to 0} \mathbf{W}_{\mathrm{NR(Leakage)}}(\omega) = \frac{P_{L^s}(\omega)}{P_{L^n}(\omega)} = \mathrm{SNR}_{\mathrm{leakage}}(\omega) \tag{2.79}$$

$$\lim_{G(\omega)\to \infty} \mathbf{W}_{\mathrm{NR(Leakage)}}(\omega) = \rho(\omega) = \mathrm{SNR}_{\mathrm{NR(noLeakage)}}(\omega) \tag{2.80}$$

When the gain $G(\omega)$ is small, the output SNR is equivalent to the leakage SNR, *i.e.*, the NR has no effect on the signal delivered at the eardrum. When the gain G is large, the output SNR is equivalent to the output SNR of the NR scheme without leakage, *i.e.*, the signal leakage has no effect on the signal delivered at the eardrum (see Figure 2.7(a)).

Signal leakage and secondary path

Taking both the signal leakage effect and the effect of the secondary path into account, the signal reaching the eardrum is expressed as in (2.66). The output SNR of the MWF-based NR scheme can then be rewritten:

$$\mathrm{SNR}_{\mathrm{NR(Leak+Sec)}}(\omega) = \frac{\mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^s(\omega) + L^s(\omega)|^2\}}{\mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^n(\omega) + L^n(\omega)|^2\}} \tag{2.81}$$

$$\boxed{= \frac{|C(\omega)|^2 \frac{\rho(\omega)^2}{(\rho(\omega)+1)^2} P_{D_{\mathrm{NR}}}(\omega) + (C(\omega) + C(\omega)^*) \frac{\rho(\omega)}{\rho(\omega)+1}\alpha(\omega) + P_{L^s}(\omega)}{|C(\omega)|^2 \frac{\rho(\omega)}{(\rho(\omega)+1)^2} P_{D_{\mathrm{NR}}}(\omega) + (C(\omega) + C(\omega)^*) \frac{1}{\rho(\omega)+1}\alpha(\omega) + P_{L^n}(\omega)}}$$

This expression is very similar to (2.69) except that due to the attenuation caused by the secondary path $C(\omega)$, the system needs a larger gain $G(\omega)$ to restore the performance back to the performance of the NR scheme in the absence of signal leakage and secondary path effects (Figure 2.7(a)).

The gain $G(\omega)$ can be designed to compensate for the effects of the secondary path. This solution, however, provides a fixed compensation and can become partly ineffective if the secondary path is changing over time (*e.g.*, when the BTE case or the ear-canal tube is moving. . .).

In the single speech source scenario, the theory confirms the observations made from simulations (see also Figure 2.7(a)). The secondary path and the signal leakage degrade the NR performance when the gain $G(\omega)$ is small. One solution to compensate for these perturbations is to use ANC.

2.4 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, or some of its components, using ANC. The signal in the BTE microphones is highly correlated with the leakage signal. It is therefore possible to apply a feedforward ANC strategy. Feedforward ANC relies on the presence of an ear canal microphone. In this thesis, 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 at the end of the tube which is used to conduct the sound from the BTE to the ear canal. It is then possible to implement a feedforward ANC scheme, using the BTE microphones as reference microphones and the ear canal microphone as an ear canal microphone.

This section presents a time-domain multichannel feedforward ANC (Figure 2.4) and its frequency-domain representation (Figure 2.5), both used to attenuate the leakage signal $l[k]$.

2.4.1 Time-domain formulation

In feedforward ANC systems, 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 [6, 8, 25, 125] based on an estimate of the secondary path and on the filtered microphone signal defined as:

$$y_m[k] = \hat{\mathbf{c}}[k]^T \begin{bmatrix} x_m[k] \\ \vdots \\ x_m[k-\hat{P}+1] \end{bmatrix} \quad m \in \{1 \dots M\} \tag{2.82}$$

$$\mathbf{y}_m[k] = [y_m[k] \dots y_m[k-N+1]]^T \tag{2.83}$$

$$\mathbf{y}[k] = \begin{bmatrix} \mathbf{y}_1[k] \\ \vdots \\ \mathbf{y}_M[k] \end{bmatrix} \tag{2.84}$$

The aim of the ANC (Figure 2.4) is to minimise the power of the residual signal at the eardrum. The residual signal at the eardrum $e_{\text{ANC}}[k]$ is equal to the ear

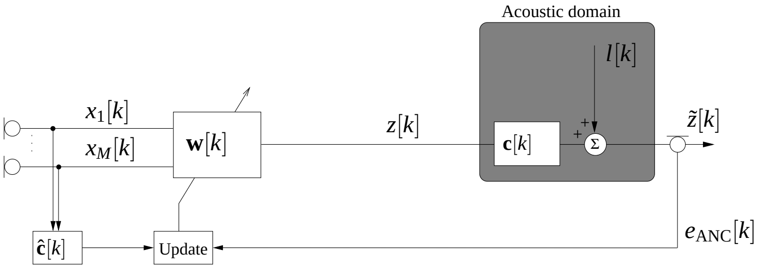


Figure 2.4: Multichannel active noise control

canal microphone signal $\tilde{z}[k]$:

$$e_{\mathrm{ANC}}[k] = \tilde{z}[k] \tag{2.85}$$

$$= \mathbf{c}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-P+1] \end{bmatrix} + l[k] \tag{2.86}$$

$$= \mathbf{c}[k]^T \mathbf{z}[k] + l[k] \tag{2.87}$$

where $\mathbf{z}[k]$ is the P-dimensional column vector containing the P last samples of the filter output $z[k]$:

$$\mathbf{z}[k] = [z[k] \dots z[k-P+1]]^T \tag{2.88}$$

$$z[k] = \mathbf{w}[k]^T \mathbf{x}[k] \tag{2.89}$$

The ANC MSE design criterion is then:

$$J_{\mathrm{ANC}}[k] = \mathbb{E}\{|e_{\mathrm{ANC}}[k]|^2\} \tag{2.90}$$

Assuming that the secondary path identification error is small ($\hat{\mathbf{c}}[k] \approx \mathbf{c}[k]$) and that the filter $\mathbf{w}_{\mathrm{ANC}}[k]$ is adapting slowly, the MSE criterion (2.90) can be written as follows:

$$J_{\mathrm{ANC}}[k] \approx \mathbb{E}\{|\mathbf{w}[k]^T \mathbf{y}[k] + l[k]|^2\} \tag{2.91}$$

The Filtered-x MWF (FxMWF) minimising (2.90) is then:

$$\boxed{\mathbf{w}_{\mathrm{ANC}}[k] = -\mathbf{R}_y[k]^{-1} \mathbf{r}_{yl}[k]} \tag{2.92}$$

Here $\mathbf{R}_y[k]$ is the correlation matrix of the filtered microphone signals $\mathbf{y}[k]$ and $\mathbf{r}_{yl}[k]$ is the cross-correlation vector between the filtered microphone signals $\mathbf{y}[k]$ and the leakage signal $l[k]$:

$$\mathbf{R}_y[k] = \mathbb{E}\{\mathbf{y}[k]\mathbf{y}[k]^T\} \tag{2.93}$$

$$\mathbf{r}_{yl}[k] = \mathbb{E}\{\mathbf{y}[k]l[k]\} \tag{2.94}$$

In practice, the leakage signal $l[k]$ can be estimated from the ear canal microphone signal $\tilde{z}[k]$:

$$l[k] = \tilde{z}[k] - \mathbf{c}[k]^T\mathbf{z}[k] \tag{2.95}$$

$$\approx \tilde{z}[k] - \hat{\mathbf{c}}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-\hat{P}+1] \end{bmatrix} \tag{2.96}$$

2.4.2 Frequency-domain formulation

The previous multichannel ANC scheme can be represented in the frequency-domain (2.5). Note that in practice, to ensure the causality of the scheme, the filter coefficients are computed in the frequency-domain while the actual filtering operation is performed in the time-domain, in a similar way as presented in [75].

As it was also stated in Section 2.1.2 the time-domain filter obtained by taking the $2N$-IDFT of the frequency-domain vector coefficient contains an N-dimensional causal part and a N-dimensional anticausal part. The time-domain filter effectively applied to the microphone signals is truncated to the N-dimensional causal part.

Note that, due to the inverse DFT and the truncation, the effect of causality on the frequency-domain version of the ANC schemes is unclear and difficult to analyse. Therefore, in this thesis the frequency-domain ANC schemes are always operating on a system with a sufficiently positive causality margin ($\delta \geq 0$), so that the effect of the truncation is indeed limited.

The filtered microphone signals are now expressed as follows:

$$\begin{aligned} Y_m(\omega) &= \hat{C}(\omega)X_m(\omega) \quad m \in \{1 \dots M\} \\ \mathbf{Y}(\omega)^T &= [Y_1(\omega) \dots Y_M(\omega)] \end{aligned} \tag{2.97}$$

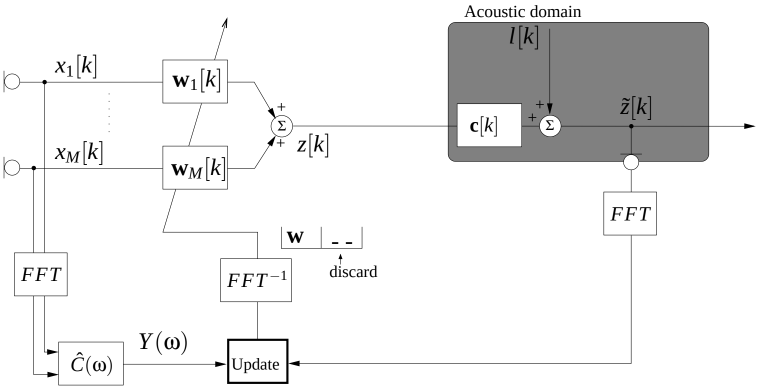


Figure 2.5: Frequency-domain multichannel active noise control

The ANC scheme is then designed to minimise the following MSE:

$$J_{\text{ANC}}(\omega) = \mathbb{E}\{|E_{\text{ANC}}(\omega)|^2\} \tag{2.98}$$

$$E_{\text{ANC}}(\omega) = C(\omega)\mathbf{W}(\omega)^H\mathbf{X}(\omega) + L(\omega) \tag{2.99}$$

Assuming that the secondary path identification error is small ($\hat{C}(\omega) \approx C(\omega)$) and that the filter $\mathbf{W}(\omega)$ is adapting slowly, the MSE criterion (2.98) can be written as follows:

$$J_{\text{ANC}}(\omega) \approx \mathbb{E}\{|\mathbf{W}(\omega)^H\mathbf{Y}(\omega) + L(\omega)|^2\} \tag{2.100}$$

The Filtered-x MWF (FxMWF) minimising (2.98) is then:

$$\boxed{\mathbf{W}_{\text{ANC}}(\omega) = -\mathbf{R}_Y(\omega)^{-1}\mathbf{r}_{YL}(\omega)} \tag{2.101}$$

Here $\mathbf{R}_Y(\omega)$ is the correlation matrix of the filtered microphone signals $\mathbf{Y}(\omega)$ and $\mathbf{r}_{YL}(\omega)$ is the cross-correlation vector between the filtered microphone signals $\mathbf{Y}(\omega)$ and the leakage signal $L(\omega)$.

$$\mathbf{R}_Y(\omega) = \mathbb{E}\{\mathbf{Y}(\omega)\mathbf{Y}(\omega)^H\} \tag{2.102}$$

$$\mathbf{r}_{YL}(\omega) = \mathbb{E}\{\mathbf{Y}(\omega)L(\omega)^*\} \tag{2.103}$$

The leakage signal $L(\omega)$ can be estimated from the ear canal microphone signal $\tilde{Z}(\omega)$ as follows:

$$L(\omega) = \tilde{Z}(\omega) - C(\omega)Z(\omega) \tag{2.104}$$

$$\approx \tilde{Z}(\omega) - \hat{C}(\omega)Z(\omega) \tag{2.105}$$

2.5 Experimental results

Recent commercial hearing aids usually integrate 2 microphones (at most 3 microphones), therefore, the filters analysed here are 2-channel filters. This section briefly analyses the SNR performance of a 2-channel MWF-based NR scheme and the impact of signal leakage and of secondary path effects. In a single speech source scenario, the simulated SNR performance is then compared with the theoretical SNR performance. A 2-channel ANC scheme is finally tested experimentally and its performance is analysed.

MWF-based NR

In the first experiment, the effect of signal leakage and the secondary path effects are neglected. The $\Delta\mathrm{SNR}_{\mathrm{intell}}$ of MWF-based NR filters of different lengths N is evaluated for noise angles varying from 270° to 90°. The $\Delta\mathrm{SNR}_{\mathrm{intell}}$ is then evaluated for a number of noise sources increasing from 1 to 4.

MWF-based NR, signal leakage and secondary path

The second experiment shows the effect of leakage and the secondary path effects on the MWF-based NR performance. For a gain G varying from 0dB to 20dB the inputs are filtered using the NR scheme and the SNR improvement (2.34) taking into account secondary path effects and/or signal leakage is evaluated.

The system is calibrated so that for $G = 0\mathrm{dB}$, for a source at 0°, the leakage and the signal fed in the loudspeaker have equal power at the eardrum.

ANC

The third experiment presents the performance of a 2-channel ANC for a single noise source, depending on the noise source location and on the causality margin of the system. The performance is then analysed for a number of noise sources varying from 1 to 4.

2.5.1 Multichannel Wiener filter-based noise reduction

The output SNR performance of an MWF-based 2-channel NR scheme is first briefly analysed when the signal leakage and the secondary path effects are neglected. The input SNR is 0dB.

Figure 2.6(a) shows the SNR improvement of a 2-channel MWF-based NR scheme for different filter lengths and for a noise angle varying from 270° to 90°. The speech source is facing the hearing aid user (at 0°). This figure shows the impact of the spatial separation of the sources on the output SNR performance. When the noise source is "well separated" from the speech source (*i.e.*, when the noise source is at 270° or 90°) the SNR improvement can reach up to $12 \sim 16$dB. When the spatial separation decreases so does the SNR performance and the MWF-based NR schemes merely deliver a 2dB SNR improvement when the speech source and the noise source are located at the same position.

Figure 2.6(b) presents the influence of the number of noise sources on the output SNR performance of the MWF-based NR scheme. To evaluate the effects of the number of sound sources (speech plus noise sources) on the performance of the MWF-based NR scheme, multiple noise sources are used to compose the input signals. When only one noise source is active, the noise source at 270° (facing the hearing aid) is used. Then the noise source at 90° is added for the 2 noise sources scenario, the noise source at 330° is added for the 3 noise sources scenario and finally the noise source at 30° is added for the 4 noise sources scenario. The noise sources are calibrated to have equal power at the hearing aid microphones and the input SNR is kept to 0dB. For any filter length N the SNR improvement performance of the MWF-based NR schemes quickly decreases when the number of noise sources is growing.

Note also that when the filter lent N is increased, so does the delay Δ introduced by the MWF-based NR scheme. In hearing aids, this delay is critical and should not exceed 10ms (*i.e.*, the limit above which the sound of a person speaking and the image of his lips moving appear un-synchronised). It is therefore chosen that, in the remainder of this thesis, the filter length N will not exceed 128 (this corresponds to a 4ms delay at 16kHz).

Secondary path and leakage

To evaluate the effect of the leakage and the secondary path, the input signals are first filtered by an MWF-based NR scheme (fig. 2.3). Depending on which disturbance is being tested, the signal produced can then be filtered by the secondary path model $\mathbf{c}[\mathbf{k}]$ and/or the leakage can be added. One noise source is present at 270° (facing the hearing aid). The leakage signal SNR is used as a reference here and is equal to -1.3dB. This value depends on the noise and speech angles as well as the input SNR (source signals), which is 5dB here.

Figure 2.7(a) shows the impact of signal leakage and secondary path effects of the output SNR of an MWF-based NR scheme applied in the context of hearing aids with an open fitting. The SNR improvement of the MWF-based NR scheme (with and without perturbation) is evaluated for a gain G varying from 0dB to 20dB. When the leakage is the only disturbance considered, the degradations induced by the leakage remain small even for reasonably low gain G (down to 10dB. However, introducing both the leakage and the secondary path the degradations are significant for gains up to at least 20dB.

These degradations stay significant up to 15dB even when replacing the open fitting by a vented earmold (Figure 2.7(b)). In this thesis, however, it is chosen to focus on hearing aids with an open fitting as they are the type of hearing aids for which leakage signal and secondary effects induce the most significant degradations.

Frequency-domain version

The same experiment is now run using a frequency-domain version of the 2-channel MWF-based NR scheme with $N = 128$. The gain G is flat, *i.e.*, it takes the same value for all frequencies. The system is calibrated so that for $G = 0$dB, for a source at 0°, the leakage signal and the signal fed in the loudspeaker have equal full-band power.

Figure 2.8(a) shows the impact of signal leakage and the secondary path effects on the output SNR of a frequency-domain MWF-based NR scheme applied in the context of hearing aids with an open fitting. The results are comparable to the result presented for the time-domain version of the NR scheme.

Figure 2.8(b) presents the theoretical output SNR improvement of the MWF-based NR with no perturbations, with signal leakage and with signal leakage and secondary path effects, in a single speech source scenario. These theoretical performances are based on the formulae derived in this chapter. The theoretical SNR improvement is presented for a gain G varying from 0dB to 20dB.

When the signal leakage is the only disturbance considered, the degradations induced by the signal leakage remain small even for reasonably small gain G (down to 8dB (Figure 2.8(b)). However, introducing both the signal leakage and the secondary path the degradations are significant for gains up to at least 20dB. This confirms the results presented in Figure 2.8(a).

2.5.2 Active noise control

In this part the performance of a 2-channel ANC scheme is presented. The first experiment is set with only one noise source. The noise source can be located at 0° (facing the hearing aid user), 90° (facing the right ear) or at 270° (facing the left ear). The BTE is still worn on the left ear and the noise sources are calibrated such that the residual noise power at the eardrum without ANC is equal for any source position. In each case, ANC schemes are run for different causality margins δ (1.5) and the residual noise power performances are compared. In practice, a variable artificial delay Δ_{HA} is added to the microphone signals in order to have a system with a variable causality margin.

Figure 2.9(a) shows the residual noise power improvement at the left eardrum for the ANC scheme depending on the causality margin δ. Wherever the noise source is located, the ANC is able to reduce significantly the noise for a causality margin down to (at best) $\delta \geq -3$. When the noise source is facing the hearing aid the system needs to be strictly causal ($\delta > 0$). When the causality criterion is satisfied ($\Delta_{\mathrm{alg}} \leq \delta$), ANC allows to reduce the residual noise power at the eardrum by 9dB to 18dB depending on the noise source location. When the causality criterion is not satisfied, the ANC is not able to deliver any improvement of the residual noise power at the eardrum.

Figure 2.9(b) presents the influence of the number of noise sources on the residual noise power performance of the ANC scheme when the causality margin is sufficiently high ($\delta = 12$). To evaluate the effect of the number of noise sources, multiple noise sources are used to compose the noise signal to be cancelled. When only one noise source is active, the noise source at 270° (facing the hearing aid) is used. Then the noise source at 90° is added for the 2 noise sources scenario, the noise source at 330° is added for the 3 noise sources scenario and finally the noise source at 30° is added for the 4 noise sources scenario. The noise sources are calibrated to have equal power at the hearing aid microphone. The performance of the 2-channel ANC quickly decreases with the number of noise sources.

2.6 Conclusion

State-of-the-art hearing aids include NR algorithms that enhance the speech and get reduce the unwanted noise (*i.e.*, increase the SNR). MWF-based NR algorithms represent one class of these NR algorithms which can take advantage of the spatial separation between sources as well as characteristic differences between the speech signal and the noise signals.

Standard NR techniques, including MWF techniques, ignore signal leakage and secondary path effects. These aspects can seriously degrade the NR performance and should not be neglected. Degradations induced by the signal leakage and the secondary path appear for any kind of BTE but they are most significant in BTE's with an open fitting. Therefore, this thesis focuses on BTE's with an open fitting. The signal leakage and the secondary path effects remain, however, a problem in BTE's with a vented earmold.

It is possible to implement a feedforward ANC scheme in hearing aids, using the BTE microphones as reference microphones and the ear canal microphone as an ear canal microphone. The ANC allows to reduce the residual noise power caused by the leakage at the eardrum. The ANC, however, is highly sensitive to delays and strict causality constraints in hearings aids may limit its benefits.

Finally, performing ANC alone is not sufficient, *i.e.*, ANC would have to be performed together with the NR.

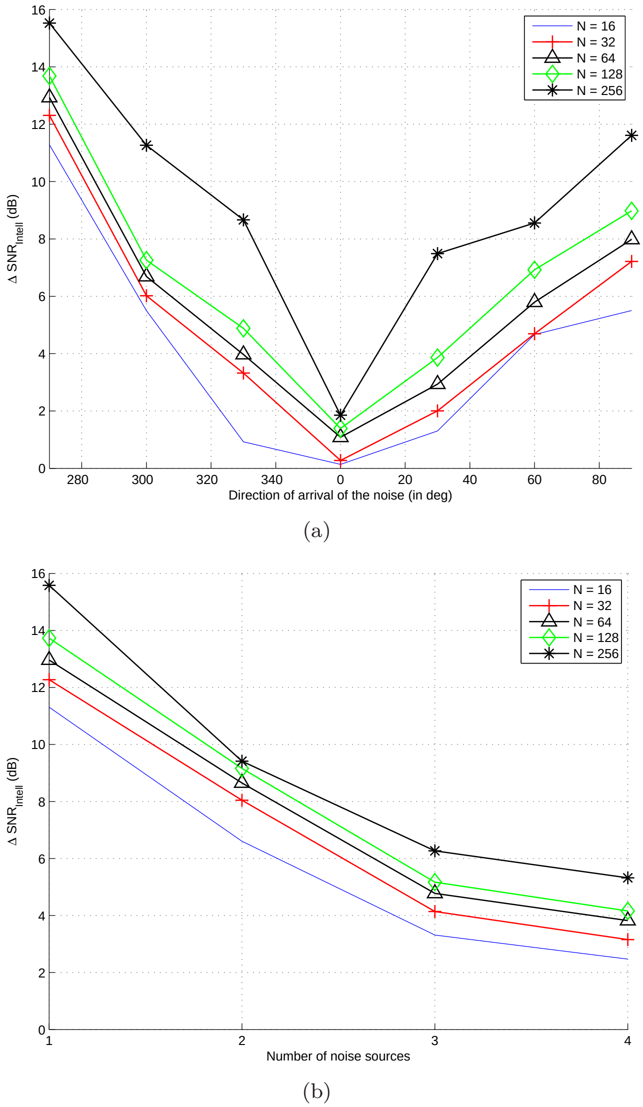
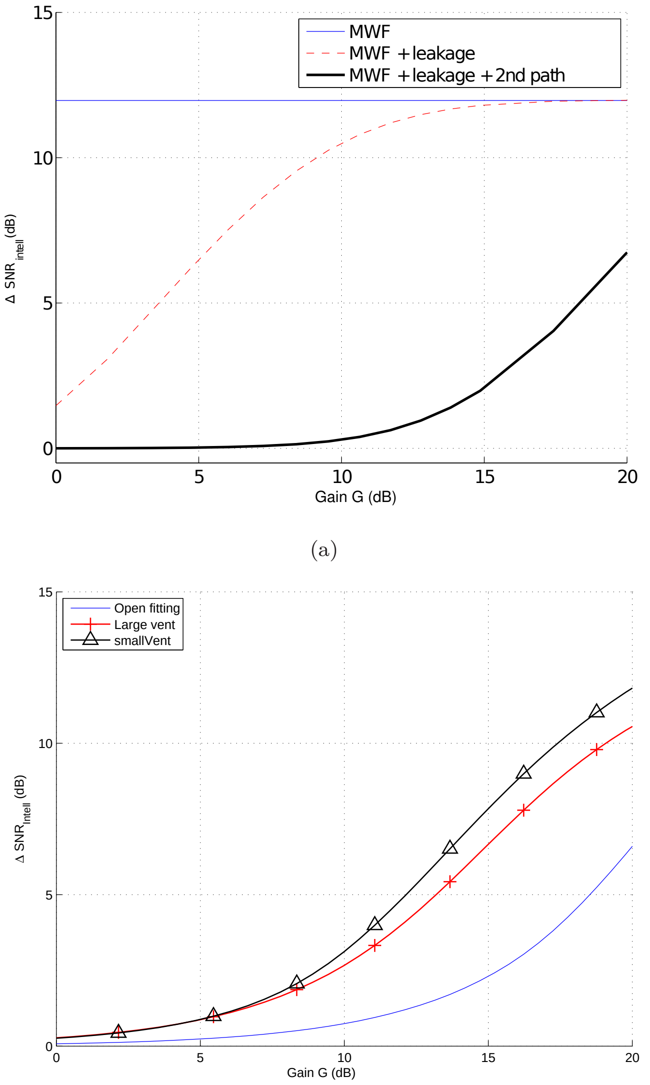


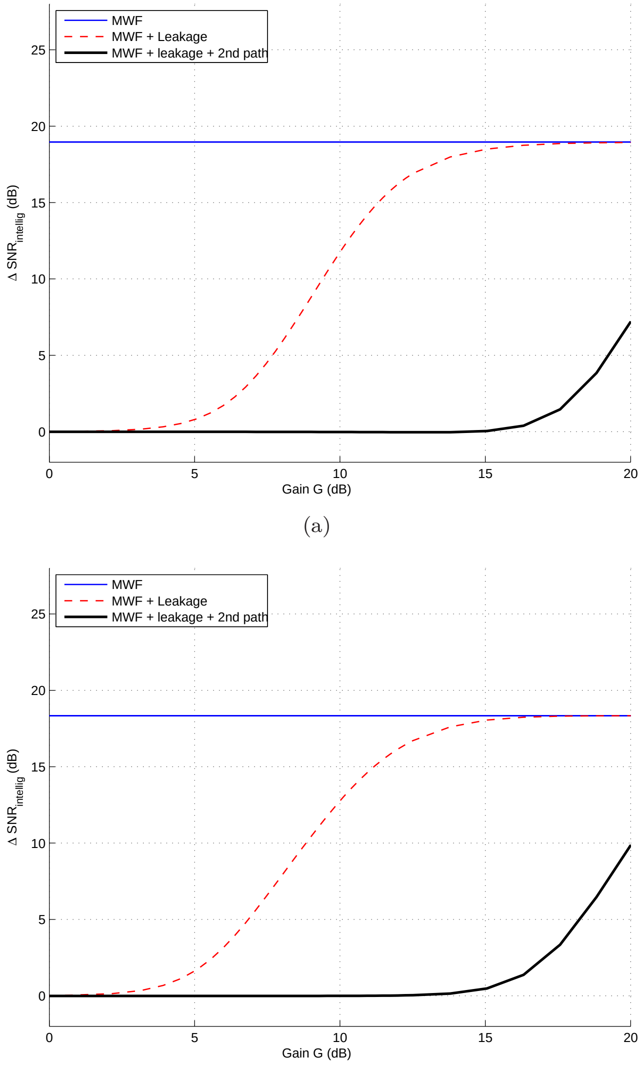
Figure 2.6: (a) SNR performance for a multichannel Wiener filter noise reduction scheme depending on the filter length for different noise angles. (b) SNR performance for a multichannel Wiener filter noise reduction scheme depending on the number of noise sources



(a)

(b)

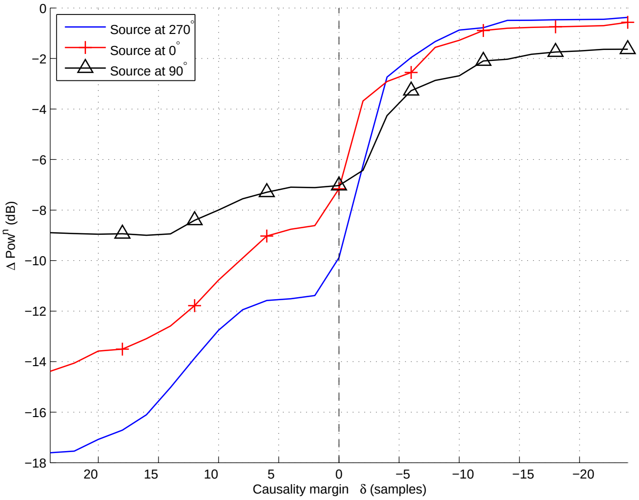
Figure 2.7: (a) Performance comparison for a multichannel Wiener filter noise reduction scheme depending on leakage and secondary path for an open fitting. (b) Performance comparison for a multichannel Wiener filter noise reduction scheme with leakage and secondary path, depending on the fitting



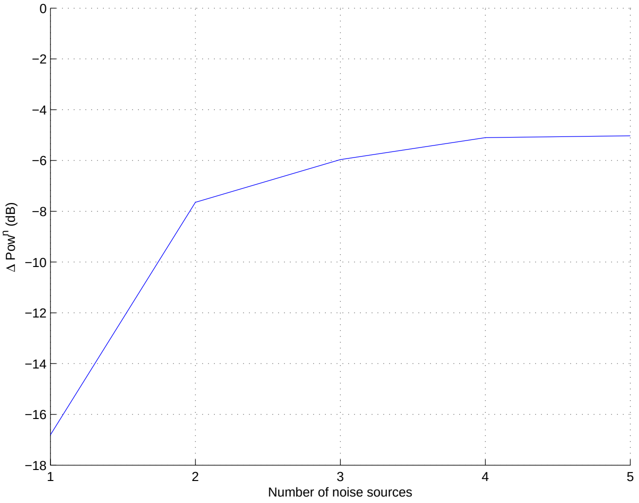
(a)

(b)

Figure 2.8: (a) Simulated SNR for the frequency-domain MWF-based NR scheme under a single speech source depending on signal leakage and secondary path. (b) Theoretical SNR for the frequency-domain MWF-based NR scheme under a single speech source depending on signal leakage and secondary path.



(a)



(b)

Figure 2.9: (a) Residual noise power performance for a multichannel ANC with a single noise source, depending on the causality margin. (b) Residual noise power performance for a multichannel ANC depending on the number of noise sources.

Chapter 3

Combined and Integrated Active Noise Control and Noise Reduction

This chapter presents combined ANC and NR schemes for hearing aids to tackle secondary path effects and effects of noise leakage through an open fitting. The different ways to combine ANC and NR are first compared. Using a NR algorithm and an ANC 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 NR and ANC. The theoretical SNR performance of the integrated ANC and NR is finally derived in the case of a single speech source.

3.1 Introduction

It has been shown in the previous chapter that ANC is an efficient way to tackle the signal leakage problem in hearing aids with an open fitting. In the hearing aids framework, however, ANC alone is not sufficient. It has to be performed together with a NR algorithm. There are different ways to combine ANC and NR. Here, the cascading of the two 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) [95, 98].

In a cascaded implementation of an MWF-based 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 thus 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 improves the performance. However, the delay needed to achieve a high NR performance is still added to the system latency. 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 SNR at the eardrum up to the causality limit ($\delta = 0$).

This chapter 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.

Section 3.2 introduces different schemes combining ANC and NR. The integrated ANC and NR scheme is presented in Section 3.3. The causality problem is described in Section 3.4. Formulae for the output SNR of the integrated ANC and NR scheme are derived in the single speech source case in Section 3.5. The experimental results are presented in Section 3.6, and finally Section 3.7 presents the conclusions of this chapter.

3.2 Combined active noise control and noise reduction

3.2.1 Noise reduction and single channel active noise control in cascade

A straightforward way to combine ANC and NR is to cascade the two functional blocks, using the output of the noise canceller as the input to the ANC system (Figure 3.1).

The filtered reference signal is then:

$$y_{\mathrm{Cas}}[k] = \hat{\mathbf{c}}[k]^T \mathbf{z}[k] \tag{3.1}$$

$$\mathbf{y}_{\mathrm{Cas}}[k] = [y_{\mathrm{Cas}}[k] \dots y_{\mathrm{Cas}}[k - N_{\mathrm{ANC}} + 1]]^T \tag{3.2}$$

The ANC output signal is:

$$\mathbf{v}[k]^T\mathbf{z}[k] \tag{3.3}$$

where the filter $\mathbf{v}[k]$ of length N_{ANC} is designed to minimise the MSE:

$$J_{\text{ANC}}[k] = \mathbb{E}\{|e_{\text{ANC}}[k]|^2\} \tag{3.4}$$

Here, $e_{\text{ANC}}[k]$ is an error signal, constructed from the ear canal microphone signal $\tilde{z}[k]$, as will be described below.

The lower branch with $\hat{\mathbf{c}}[k]$ in Figure 3.1 (and also in subsequent figures), represents the adaptation (gradient estimation) of $\mathbf{v}[k]$ with a standard Filtered-x adaptive filter algorithm.

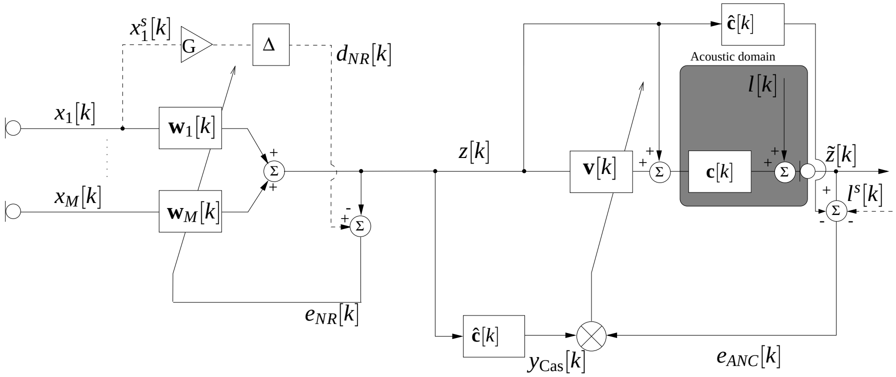


Figure 3.1: Multichannel NR and single channel ANC system in cascade

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 localise the speaker. Therefore, it is 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 cancelled.

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 $e_{\text{ANC}}[k]$ as indicated in Figure 3.1. Note however that, as $l^s[k]$ is unknown, this subtraction is not done explicitly, but implicitly by adapting the filter $\mathbf{v}[k]$ only in "noise only periods" where $l^s[k] = 0$. During "speech plus noise periods", the remaining leakage signal could then be considered as an estimate of the speech component of the leakage:

$$l^s[k] = l[k] - l^n[k] \tag{3.5}$$

The ANC itself would also tend to remove the desired speech component (NR output signal $z[k]$) so this signal has to be added back to the ANC output signal (loudspeaker input signal) and then the same signal filtered by the estimated secondary path $\hat{\mathbf{c}}[k]$ (corresponding to an estimate of the NR output as delivered at the eardrum) is subtracted from the ANC error signal $e_{\text{ANC}}[k]$, as explained in [35] and [58], leading to:

$$e_{\text{ANC}}[k] = \mathbf{c}[k]^T \begin{bmatrix} \bar{z}[k] \\ \vdots \\ \bar{z}[k-P+1] \end{bmatrix} + l[k] - (l^s[k] + \hat{\mathbf{c}}[k]^T \mathbf{z}[k]) \tag{3.6}$$

where

$$\bar{z}[k] = \mathbf{v}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-N_{\text{ANC}}+1] \end{bmatrix} + z[k] \tag{3.7}$$

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

$$e_{\text{ANC}}[k] \approx \mathbf{v}[k]^T \mathbf{y}_{\text{Cas}}[k] + l^n[k] \tag{3.8}$$

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

$$\begin{aligned} \tilde{z}[k] &= e_{\text{ANC}}[k] + \hat{\mathbf{c}}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-\hat{P}+1] \end{bmatrix} + l^s[k] && (3.9) \\ &\approx e_{\text{ANC}}[k] + \mathbf{c}[k]^T \mathbf{z}[k] + l^s[k] && (3.10) \end{aligned}$$

This is the sum of the minimised error signal, the enhanced speech signal as delivered at the eardrum and the speech component of the leakage signal.

In noise only periods, the MSE criterion (3.4) can be rewritten as follows:

$$J^n_{\text{ANC}}[k] = \mathbb{E}\{|e^n_{\text{ANC}}[k]|^2\} \approx \mathbb{E}\{|\mathbf{v}[k]^T \mathbf{y}^n_{\text{Cas}}[k] + l^n[k]|^2\} \tag{3.11}$$

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 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 signal of the ANC, is non-zero and can be used to produce the anti-noise. Still, this noise component may be small and then the ANC may exhibit a poor performance.

3.2.2 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 Figure 3.2, which will exhibit improved performance.

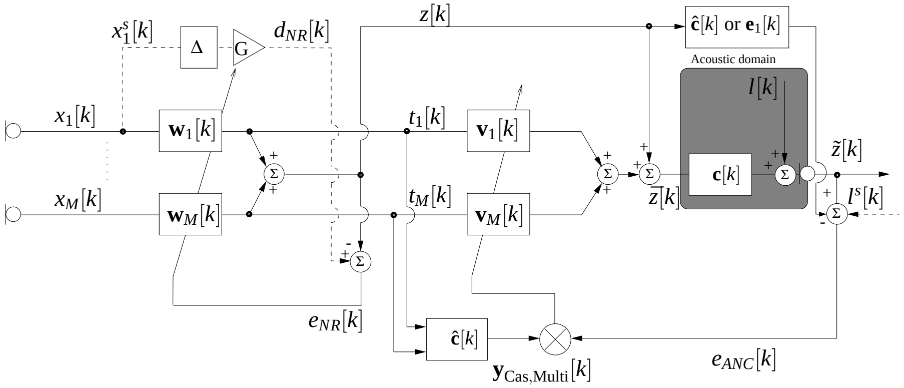


Figure 3.2: Multichannel NR and ANC systems in cascade

The NR algorithm is the same as described in Chapter 2. 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:

$$t_m[k] = \mathbf{w}_m[k]^T \mathbf{x}_m[k] \quad m \in \{1 \dots M\} \tag{3.12}$$

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

$$\begin{aligned} \mathbf{t}_m[k] &= [t_m[k] \dots t_m[k - N_{\text{ANC}} + 1]]^T \quad m \in \{1 \dots M\} \\ \mathbf{t}[k]^T &= [\mathbf{t}_1[k]^T \dots \mathbf{t}_M[k]^T] \end{aligned} \tag{3.13}$$

The filtered reference signals in the ANC are defined as:

$$y_{Cas,m}[k] = \hat{\mathbf{c}}[k]^T \begin{bmatrix} t_m[k] \\ \vdots \\ t_m[k-\hat{P}+1] \end{bmatrix} \quad m \in \{1 \dots M\} \tag{3.14}$$

Let $\mathbf{y}_{\text{Cas},m}[k]$ denote the column vector containing the last N_{ANC} samples of $y_{\text{Cas},m}[k]$ and let $\mathbf{y}_{\text{Cas,Multi}}[k]$ denote the compound vector gathering all the channels.

$$\begin{aligned} \mathbf{y}_{\text{Cas},m}[k] &= [y_{\text{Cas},m}[k] \dots y_{\text{Cas},m}[k-N_{\text{ANC}}+1]]^T \quad m \in \{1 \dots M\} \\ \mathbf{y}_{\text{Cas,Multi}}[k]^T &= [\mathbf{y}_{\text{Cas},1}[k]^T \dots \mathbf{y}_{\text{Cas},M}[k]^T] \end{aligned} \tag{3.15}$$

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

$$J_{\text{ANC}}[k] = \mathbb{E}\{|c[k]^T \begin{bmatrix} \bar{z}[k] \\ \vdots \\ \bar{z}[k-P+1] \end{bmatrix} + l^n[k] - \hat{\mathbf{c}}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-\hat{P}+1] \end{bmatrix} |^2\} \tag{3.16}$$

where

$$\bar{z}[k] = \mathbf{v}[k]^T \mathbf{t}[k] + z[k] \tag{3.17}$$

Assuming that the secondary path identification error is small ($\hat{\mathbf{c}}[k] \approx \mathbf{c}[k]$) and that the filter $\mathbf{v}[k]$ is adapting slowly, the error signal reduces to:

$$e_{\text{ANC}}[k] \approx \mathbf{v}[k]^T \mathbf{y}_{\text{Cas,Multi}}[k] + l^n[k] \tag{3.18}$$

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

$$\tilde{z}[k] = e_{\text{ANC}}[k] + \hat{\mathbf{c}}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-\hat{P}+1] \end{bmatrix} + l^s[k] \tag{3.19}$$

$$\approx e_{\text{ANC}}[k] + \mathbf{c}[k]^T \mathbf{z}[k] + l^s[k] \tag{3.20}$$

This is the sum of the minimised error signal, the enhanced speech signal as delivered at the eardrum 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 that reaches the eardrum, while reducing the impact of noise leakage on the final output.

In noise only periods, the MSE criterion (3.16) can be rewritten as follows:

$$J_{\text{ANC}}[k] = \mathbb{E}\{|e^n_{\text{ANC}}[k]|^2\} \approx \mathbb{E}\{|\mathbf{v}[k]^T \mathbf{y}^n_{\text{Cas,Multi}}[k] + l^n[k]|^2\} \tag{3.21}$$

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 $\mathbf{v}[k]$ on the speech component of the signal is unknown and cannot be controlled. Also, as long as $\hat{\mathbf{c}}[k] \approx \mathbf{c}[\mathbf{k}]$ in the upper branch (Figure 3.2), the filter $\mathbf{v}[k]$ does not compensate for the secondary path effects.

An attempt to compensate for this extra path can be to replace $\hat{\mathbf{c}}[k]$ in the upper branch (Figure 3.2) by $\mathbf{e}_1[k]$ as defined in Section 2.3.2, which corresponds to the identity filter of length P such that, for any signal $x[k]$:

$$\mathbf{e}_1[k]^T \mathbf{x}[k] = x[k] \tag{3.22}$$

The MSE criterion (3.16) is then modified into:

$$J_{\text{ANC}}[k] = \mathbb{E}\{|c[k]^T \begin{bmatrix} \bar{z}[k] \\ \vdots \\ \bar{z}[k-P+1] \end{bmatrix} + l^n[k] - z[k]|^2\} \tag{3.23}$$

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

$$\begin{aligned} \mathbf{v'}_m[k] &= \mathbf{v}_m[k] + \mathbf{e}_1 \quad m \in \{1 \dots M\} && (3.24) \\ \mathbf{v'}[k] &= [\mathbf{v'}_1[k]^T \dots \mathbf{v'}_M[k]^T]^T && (3.25) \end{aligned}$$

the ANC error signal can be written as follows:

$$e_{\text{ANC}}[k] = \mathbf{c}[k]^T \begin{bmatrix} \bar{z'}[k] \\ \vdots \\ \bar{z'}[k-P+1] \end{bmatrix} + l^n[k] - z[k] \tag{3.26}$$

where

$$\bar{z'}[k] = \mathbf{v'}[k]^T \mathbf{t}[k] \tag{3.27}$$

and $\mathbf{v'}[k]$ is the optimisation vector.

Assuming that the secondary path identification error is small ($\hat{\mathbf{c}}[k] \approx \mathbf{c}[k]$) and that the filter $\mathbf{v'}$ is adapting slowly, the error reduces to:

$$e_{\text{ANC}} \approx \mathbf{v'}[k]^T \mathbf{y}_{\text{Cas,Multi}}[k] + l^n[k] - z[k] \tag{3.28}$$

Assuming that the speech and noise components are uncorrelated, the MSE criterion (3.23) can be written as

$$\begin{aligned} J_{MSE}[k] \approx\ & \mathbb{E}\{|\mathbf{v}'[k]^T \mathbf{y}^n_{\text{Cas,Multi}}[k] - z^n[k] + l^n[k]|^2\} \\ & + \mathbb{E}\{|\mathbf{v}'[k]^T \mathbf{y}^s_{\text{Cas,Multi}}[k] - z^s[k]|^2\} \end{aligned} \tag{3.29}$$

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 rest of this thesis, 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 Chapter 2.

From (3.29) it can be seen that the desired signal to be used here is:

$$\begin{aligned} d_{\text{ANC}}[k] &= z[k] - l^n[k] \\ &= z^n[k] + z^s[k] - l^n[k] \end{aligned} \tag{3.30}$$

The optimal filter $\mathbf{v}'[k]$ is then given by:

$$\boxed{\mathbf{v}'[k] = \mathbf{R}_y[k]^{-1}\mathbf{r}_{yd_{\text{ANC}}}[k]} \tag{3.31}$$

Here $\mathbf{R}_y[k]$ is the correlation matrix of the filtered reference signal $\mathbf{y}_{\text{Cas,Multi}}[k]$ (as defined earlier) and $\mathbf{r}_{yd_{\text{ANC}}}[k]$ is the cross-correlation vector between the filtered reference signal $\mathbf{y}_{\text{Cas,Multi}}[k]$ and the target signal $d_{\text{ANC}}[k]$:

$$\mathbf{r}_{yd_{\text{ANC}}}[k] = \mathbb{E}\{\mathbf{y}_{\text{Cas,Multi}}[k] d_{\text{ANC}}[k]\} \tag{3.32}$$

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

$$\mathbf{r}_{yd_{\text{ANC}}}[k] = \mathbf{r}_{y^s d^s_{\text{ANC}}}[k] + \mathbf{r}_{y^n d^n_{\text{ANC}}}[k] \tag{3.33}$$

$$= \mathbf{r}_{y^s z^s}[k] + \mathbf{r}_{y^n d^n_{\text{ANC}}} \tag{3.34}$$

$$= \mathbf{r}_{yz}[k] - \mathbf{r}_{y^n z^n}[k] + \mathbf{r}_{y^n d^n_{\text{ANC}}} \tag{3.35}$$

with

$$\mathbf{r}_{yz}[k] = \mathbb{E}\{\mathbf{y}_{\text{Cas,Multi}}[k] z[k]\} \tag{3.36}$$

$$\mathbf{r}_{y^n z^n}[k] = \mathbb{E}\{\mathbf{y}^n_{\text{Cas,Multi}}[k] z^n[k]\} \tag{3.37}$$

$$\mathbf{r}_{y^n d^n_{\text{ANC}}}[k] = \mathbb{E}\{\mathbf{y}^n_{\text{Cas,Multi}}[k] d^n_{\text{ANC}}[k]\} \tag{3.38}$$

While $\mathbf{R}_y[k]$ and $\mathbf{r}_{yz}[k]$ are estimated during speech plus noise periods, $\mathbf{r}_{y^n z^n}[k]$ and $\mathbf{r}_{y^n d^n_{\text{ANC}}}[k]$ can be estimated during noise only periods, based on (3.28):

$$d^n_{\text{ANC}}[k] = z^n[k] - l^n[k] \tag{3.39}$$

$$\approx \mathbf{v'}[k]^T \mathbf{y}^n_{\text{Cas,Multi}}[k] - e^n_{\text{ANC}}[k] \tag{3.40}$$

The first term on the right hand side in (3.29) is the difference between the noise component in $z[k]$ and the noise component of the signal (loudspeaker + leakage) reaching the eardrum. The term corresponds to the difference between the amplified desired speech signal and the speech component of the signal reaching the eardrum, where the secondary path effect has been cancelled effectively. Therefore, minimising (3.29) corresponds to compensating for the secondary path effect on the reference signal while cancelling the effect of the noise leakage at the eardrum. The output signal (at the eardrum) is:

$$\tilde{z}[k] = e_{\text{ANC}}[k] + z[k] + l^s[k] \tag{3.41}$$

which is the sum of the minimised error, the enhanced speech signal and the speech component of the leakage signal. Unlike in (3.20), the desired speech signal is not filtered by the secondary path. The scheme therefore minimise the noise power at the eardrum and compensate for the effects of the secondary path on the desired speech.

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 Figure 3.2). 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 ANC and NR does not seem to be the most efficient approach (see also Section 3.6 for an evaluation of the system in Figure 3.2).

3.2.3 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 (Figure 3.3). 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.

The desired signal for the ANC filter is still given by (3.30):

$$d_{\text{ANC}} = z[k] - l^n[k] \tag{3.42}$$

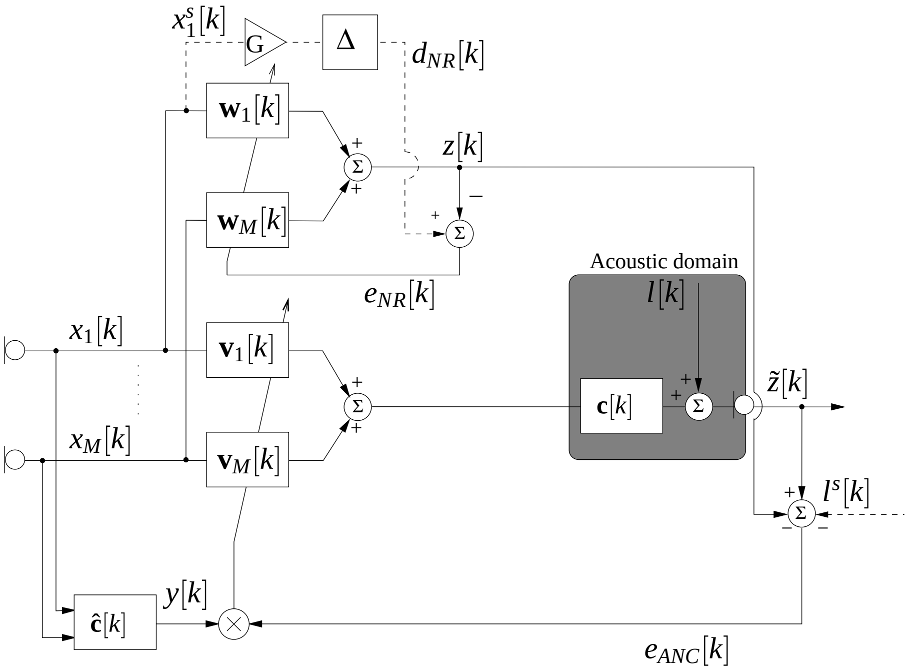


Figure 3.3: Active noise control and noise reduction system in parallel

The MSE to be minimised is then:

$$J_{\text{ANC}}[k] = \mathbb{E}\{|e_{\text{ANC}}[k]|^2\} \tag{3.43}$$

$$e_{\text{ANC}}[k] = \mathbf{c}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-P+1] \end{bmatrix} + l[k] - (z[k] + l^s[k]) \tag{3.44}$$

where

$$z[k] = \mathbf{v}[k]^T \mathbf{x}[k] \tag{3.45}$$

Assuming that the secondary path estimation error is small ($\hat{\mathbf{c}}[k] \approx \mathbf{c}[k]$) and that the filter $\mathbf{v}[k]$ is varying slowly, the MSE criterion is rewritten as follows:

$$J_{\text{ANC}}[k] \approx \mathbb{E}\{|\mathbf{v}[k]^T \mathbf{y}[k] - z[k] + l^n[k]|^2\} \tag{3.46}$$

and the optimal filter is:

$$\boxed{\mathbf{v}[k] = \mathbf{R}_y[k]^{-1} \mathbf{r}_{yd_{\text{ANC}}}[k]} \tag{3.47}$$

This expression is similar to (3.31) except that the reference signals ($\mathbf{y}[k]$) are different to those used to compute the cascaded filter ($\mathbf{y}_{\mathrm{Cas}}[k]$).

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

$$\begin{aligned} J_{\mathrm{ANC}}[k] &= \mathbb{E}\{|\mathbf{c}[k]^T\mathbf{z}^n[k] - z^n[k] + l^n[k]|^2\} \\ &\quad + \mathbb{E}\{|\mathbf{c}[k]^T\mathbf{z}^s[k] - z^s[k]|^2\} \end{aligned} \tag{3.48}$$

The filter $\mathbf{v}[k]$ thus minimises the noise sound pressure at the eardrum as well as the influence of the secondary path on the desired signal that reaches the eardrum. The output of the system is the sum of this minimised error, the enhanced speech signal and the speech component of the leakage signal:

$$\tilde{z}[k] = e_{\mathrm{ANC}}[k] + z[k] + l^s[k] \tag{3.49}$$

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 explicit estimate of the desired speech signal ($z[k] = \mathbf{w}[k]^T\mathbf{x}[k]$). To further simplify the system, this signal can implicitly be replaced by the desired signal of the NR ($d_{\mathrm{NR}}[k]$). This is pursued in the next section.

3.3 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 (Figure 3.4).

The algorithm relies on a Filtered-x version of the MWF (FxMWF) based on an estimate of the secondary path $\hat{\mathbf{c}}[k]$ and on the filtered reference signals $\mathbf{y}[k]$.

The aim of the integrated ANC and NR scheme is to improve the SNR at the eardrum, also compensating for the degradations caused by signal leakage. The filter structure is derived from the parallel ANC and NR scheme where the estimated desired signal $z[k]$ is replaced by the (unknown) NR desired signal to be estimated $d_{\mathrm{NR}}[k]$. Therefore, the desired signal (at the eardrum) to be used is:

$$\begin{aligned} d_{\mathrm{Int}}[k] &= -l^n[k] + G\cdot x_1^s[k-\Delta] \\ &= -l^n[k] + d_{\mathrm{NR}}[k] \end{aligned} \tag{3.50}$$

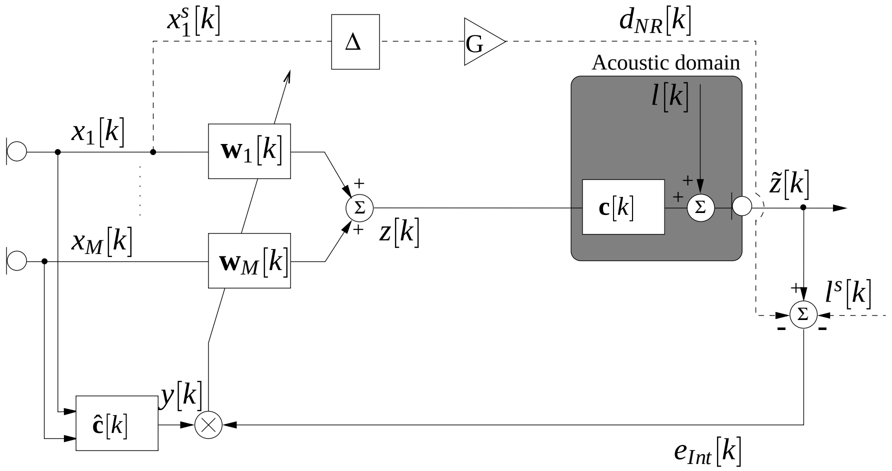


Figure 3.4: Integrated multichannel active noise control and noise reduction system

The MSE criterion to be minimised is then :

$$J_{\text{Int}}[k] = \mathbb{E}\{|e_{\text{Int}}[k]|^2\} \tag{3.51}$$

$$= \mathbb{E}\{|\mathbf{c}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-P+1] \end{bmatrix} + l^n[k] - d_{\text{NR}}[k]|^2\}$$

where

$$z[k] = \mathbf{w}[k]^T \mathbf{x}[k] \tag{3.52}$$

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

$$e_{\text{Int}}[k] \approx \mathbf{w}[k]^T \mathbf{y}[k] + l^n[k] - d_{\text{NR}}[k] \tag{3.53}$$

Assuming that, the noise and speech components in (3.53) are uncorrelated, the criterion (3.51) can be rewritten as follows:

$$J_{\text{Int}}[k] = \mathbb{E}\{|\mathbf{w}[k]^T \mathbf{y}^s[k] - d_{\text{NR}}[k]|^2\}$$

$$+ \mathbb{E}\{|\mathbf{w}[k]^T \mathbf{y}^n[k] + l^n[k]|^2\} \tag{3.54}$$

The first term of the right hand side corresponds to the estimation of the desired speech signal $d_{\text{NR}}[k]$ including secondary path compensation while the second term specifies the noise sound pressure at the eardrum and thus corresponds to the ANC. The filter described in (3.55) is therefore performing a FxMWF-based NR, which takes the secondary path into account, combined with an ANC.

The optimal filter (FxMWF) minimising (3.54) is:

$$\boxed{\mathbf{w}[k] = \mathbf{R}_y[k]^{-1}\mathbf{r}_{yd_{\mathrm{Int}}}[k]} \tag{3.55}$$

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

$$\mathbf{r}_{yd_{\mathrm{Int}}}[k] = \mathbb{E}\{\mathbf{y}[k]d_{\mathrm{Int}}[k]\} \tag{3.56}$$

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

$$\mathbf{r}_{yd_{\mathrm{Int}}}[k] = \mathbf{r}_{y^s d_{\mathrm{NR}}}[k] - \mathbf{r}_{y^n l^n}[k] \tag{3.57}$$

$$= G \cdot (\mathbf{r}_{yx_{1,\Delta}}[k] - \mathbf{r}_{y^n x^n_{1,\Delta}}[k]) - \mathbf{r}_{y^n l^n}[k] \tag{3.58}$$

with

$$\mathbf{r}_{yx_{1,\Delta}}[k] = \mathbb{E}\{\mathbf{y}[k]x_1[k-\Delta]\} \tag{3.59}$$

$$\mathbf{r}_{y^n x^n_{1,\Delta}}[k] = \mathbb{E}\{\mathbf{y}^n[k]x^n_1[k-\Delta]\} \tag{3.60}$$

$$\mathbf{r}_{y^n l^n}[k] = \mathbb{E}\{\mathbf{y}^n[k]l^n[k]\} \tag{3.61}$$

While $\mathbf{r}_{yx_{1,\Delta}}[k]$ is estimated during speech plus noise periods, $\mathbf{r}^n_{y^n x_{1,\Delta}}[k]$ and $\mathbf{r}_{y^n l^n}[k]$ can be estimated during noise only periods with:

$$l^n[k] \approx e^n_{\mathrm{Int}}[k] - \mathbf{w}[k]^T\mathbf{y}^n[k] \tag{3.62}$$

3.4 Robustness to causality

As explained in [25], the main condition to achieve a good performance with the (feedforward) ANC system is that a causality criterion is satisfied (see also Figure 1.9). The bandwidth on which it is possible to achieve good ANC performance reduces with the causality margin δ defined in (1.5). When (1.3) is not satisfied, the ANC efficiency vanishes quickly [57]. Delay is thus a critical problem in ANC and many approaches have been developed to try to deal with it [75, 91].

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 centimetres. This corresponds to a few tens of microseconds, *i.e.*, only a few taps (*i.e.*, sampling periods) for standard sampling frequencies.

3.4.1 Noise reduction and active noise control in cascade

In the cascaded schemes presented in Sections 3.2.1 and 3.2.2, the input of the ANC is the output of a standard multichannel NR, which itself introduces a delay (Figure. 2.3):

$$\Delta_{alg} = \Delta + \Delta_{\mathrm{ANC}} \tag{3.63}$$

Where Δ and Δ_{ANC} are the delay introduced by the NR and the ANC, respectively.

The causality criterion (1.4) can then be rewritten as follows:

$$\Delta_{\mathrm{ref}} + \Delta + \Delta_{\mathrm{ANC}} + \Delta_{\mathrm{sec}} \leq \Delta_{\mathrm{pri}} \tag{3.64}$$

$$\Delta_{\mathrm{ANC}} + \Delta \leq \Delta_{\mathrm{pri}} - (\Delta_{\mathrm{ref}} + \Delta_{\mathrm{sec}}) = \delta \tag{3.65}$$

$$\Delta_{\mathrm{ANC}} \leq \delta - \Delta \tag{3.66}$$

Usually, the NR delay Δ is set to half of the NR filter length (N) and will already exceed the few taps available for processing (δ). Therefore, the causality criterion specified in (3.66) cannot be fulfilled and the ANC may not be able to yield good performance. Reducing the NR delay Δ increases the effective causality margin for the ANC 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 its causality margin decreases, 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.

3.4.2 Integrated active noise control and noise reduction

If the speech signal and the noise signal are uncorrelated, the integrated ANC and NR scheme (Section 3.3), minimising the MSE (3.54) can be split into a sum of two filters:

$$\mathbf{w}[k] = \mathbf{u}[k] + \mathbf{v}[k] \tag{3.67}$$

where:

$$\mathbf{u}[k] = \mathbf{R}_y[k]^{-1}\mathbf{r}_{y^s d_{\mathrm{NR}}}[k] \tag{3.68}$$

$$\mathbf{v}[k] = -\mathbf{R}_y[k]^{-1}\mathbf{r}_{y^n l^n}[k] \tag{3.69}$$

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 cancelling the noise leakage. The filters integrating NR and ANC can thus be seen as the sum of two sets of filters, one for the NR and the other one for the ANC. Therefore, the input signals of the ANC do not depend on the delay introduced in the NR part (*i.e.*, $\Delta_{alg} = \Delta_{\mathrm{ANC}}$) and it is possible to design a causal active noise controller to be integrated with the NR as long as:

$$\Delta_{\mathrm{ref}} + \Delta_{\mathrm{ANC}} + \Delta_{\mathrm{sec}} \leq \Delta_{\mathrm{pri}} \tag{3.70}$$

$$\Delta_{\mathrm{ANC}} \leq \Delta_{\mathrm{pri}} - (\Delta_{\mathrm{ref}} + \Delta_{\mathrm{sec}}) = \delta \tag{3.71}$$

$$\Delta_{\mathrm{ANC}} \leq \delta \tag{3.72}$$

That is, there is no performance tradeoff to be done between the NR and the ANC. The whole system merely requires a positive causality margin in order to design a causal ANC ($\delta \geq 0$).

3.5 Signal-to-noise ratio performance for a single speech source scenario

This section first presents a frequency-domain description of the integrated ANC and NR scheme introduced previously and then analyses its output SNR performance in the single speech source case. Note again that in practice, as explained in Section 2.4.2 to ensure causality, the filter coefficients are computed in the frequency-domain while the filtering operation itself is performed in the time-domain (Figure 2.4), in a similar way as presented in [75]. The time-domain filter obtained by taking the $2N$-IDFT of the frequency-domain coefficient vector contains an N-dimensional causal part and an N-dimensional anticausal part. The time-domain filter applied to the microphone signals is truncated to the N-dimensional causal filter. Therefore, in this thesis the frequency-domain integrated ANC and NR schemes are always operating on a system with a positive causality margin ($\delta \geq 0$) and the SNR formulae derived are also valid only as long as the causality margin δ is sufficiently positive so that the effect of the truncation is limited.

3.5.1 Integrated active noise control and noise reduction

The frequency-domain representation of the scheme that integrates NR and ANC in a single set of adaptive filters is given below.

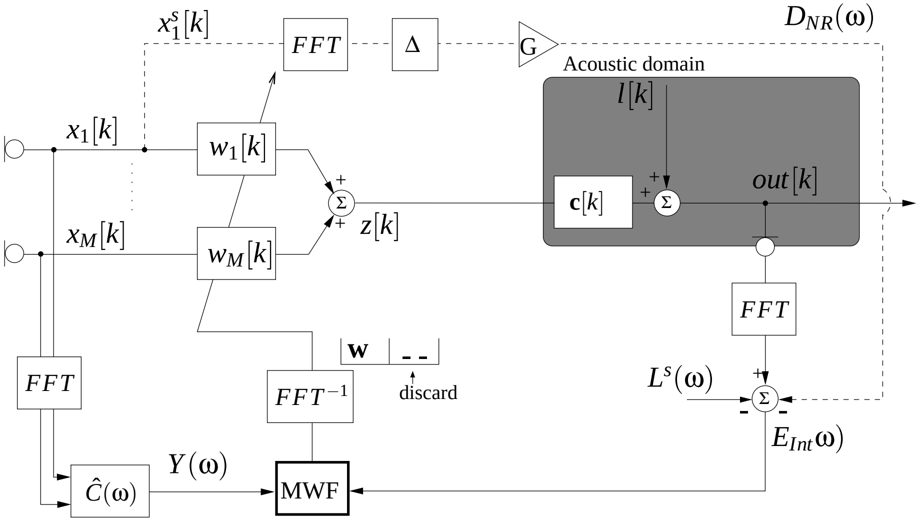


Figure 3.5: Integrated active noise control and noise reduction

The aim of the integrated ANC and NR scheme is to minimise the squared distance between a desired speech signal $D_{\mathrm{NR}}(\omega)$ and the speech signal delivered at the eardrum $C(\omega)Z^s(\omega)$ and at the same time to minimise the residual noise at the eardrum $\tilde{Z}^n(\omega)$. Therefore the desired signal to be used here is:

$$D_{\mathrm{Int}}(\omega) = \mathbf{G}_{1,\Delta}(\omega)^H \mathbf{X}^s(\omega) - L^n(\omega) = D_{\mathrm{NR}}(\omega) - L^n(\omega) \tag{3.73}$$

and the MSE criterion to be minimised is:

$$J_{\mathrm{Int}}(\omega) = \mathbb{E}\{|E_{\mathrm{Int}}(\omega)|^2\} \tag{3.74}$$

$$\begin{aligned} E_{\mathrm{Int}}(\omega) &= C(\omega)Z(\omega) - D_{\mathrm{Int}}(\omega) \qquad (3.75) \\ &= C(\omega)\mathbf{W}(\omega)^H \mathbf{X}(\omega) - D_{\mathrm{NR}}(\omega) + L^n(\omega) \end{aligned}$$

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

$$E_{\mathrm{Int}}(\omega) = \mathbf{W}(\omega)^H \mathbf{Y}(\omega) - D_{\mathrm{NR}}(\omega) + L^n(\omega) \tag{3.76}$$

The optimal filter (FxMWF) minimising (3.74) is then:

$$\boxed{\mathbf{W}_{\text{Int}}(\omega) = \mathbf{R}_Y(\omega)^{-1}(\mathbf{r}_{YD_{\text{NR}}}(\omega) - \mathbf{r}_{YL^n}(\omega))} \tag{3.77}$$

Where $\mathbf{r}_{YD_{\text{NR}}}(\omega)$ and $\mathbf{r}_{YL^n}(\omega)$ are the cross-correlation vector between the filtered microphone signal $\mathbf{Y}(\omega)$ and the desired speech signal $D_{\text{NR}}(\omega)$ and the noise component of the leakage signal $L^n(\omega)$, respectively:

$$\mathbf{r}_{YL^n}(\omega) = \mathbb{E}\{\mathbf{Y}(\omega)L^n(\omega)^*\} \tag{3.78}$$

$$\mathbf{r}_{YD_{\text{NR}}(\omega)} = \mathbb{E}\{\mathbf{Y}(\omega)D_{\text{NR}}(\omega)^*\} \tag{3.79}$$

When speech and noise are uncorrelated these cross-correlation vectors reduce to:

$$\mathbf{r}_{YL^n}(\omega) = \mathbb{E}\{\mathbf{Y}^n(\omega)L^n(\omega)^*\} \tag{3.80}$$

$$\mathbf{r}_{YD_{\text{NR}}}(\omega) = \mathbb{E}\{\mathbf{Y}^s(\omega)D_{\text{NR}}(\omega)^*\} \tag{3.81}$$

Alternatively, the autocorrelation matrix $\mathbf{R}_Y(\omega)$ and the cross-correlation vectors $\mathbf{r}_{YD_{\text{NR}}}(\omega)$ and $\mathbf{r}_{YL^n}(\omega)$ can be expressed as follows:

$$\mathbf{R}_Y(\omega) = \mathbb{E}\{\hat{C}(\omega)\mathbf{X}(\omega)\mathbf{X}(\omega)^H\hat{C}(\omega)^H\} \tag{3.82}$$

$$= |\hat{C}(\omega)|^2\mathbf{R}_X(\omega) \tag{3.83}$$

$$\mathbf{r}_{YL^n}(\omega) = \mathbb{E}\{\hat{C}(\omega)\mathbf{X}^n(\omega)L^n(\omega)^*\} \tag{3.84}$$

$$= \hat{C}(\omega)\mathbf{r}_{X^nL^n}(\omega) \tag{3.85}$$

$$\mathbf{r}_{YD_{\text{NR}}}(\omega) = \mathbb{E}\{\mathbf{Y}^n(\omega)D_{\text{NR}}(\omega)^*\} \tag{3.86}$$

$$= \hat{C}(\omega)\mathbf{r}_{XD_{\text{NR}}}(\omega) \tag{3.87}$$

$$= \hat{C}(\omega)\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) \tag{3.88}$$

The filter (3.77) can then be rewritten as follows:

$$\boxed{\mathbf{W}_{\text{Int}}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}(\mathbf{R}_{X^s}(\omega) + \mathbf{R}_{X^n})(\omega)^{-1}(\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) - \mathbf{r}_{X^nL^n}(\omega))} \tag{3.89}$$

3.5.2 Output signal-to-noise ratio when the number of sources is less than or equal to the number of microphones

In [99], the output SNR of the integrated ANC and NR scheme has been derived when the number of sources (speech plus noise sources) is less than or equal to the number of microphones ($Q \leq M$). Under this assumption it can be shown that the leakage signal can be rewritten as a linear combination of the microphone signals:

$$L(\omega) = \mathbf{P}(\omega)^H \mathbf{X}(\omega) \tag{3.90}$$

$$\mathbf{P}(\omega)^T = [P_1(\omega) \dots P_M(\omega)] \tag{3.91}$$

The filter (3.89) then becomes:

$$\mathbf{W}_{\mathrm{Int}(Q \leq M)}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2} (\mathbf{R}_{X^s}(\omega) + \mathbf{R}_{X^n}(\omega))^{-1} (\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) - \mathbf{R}_{X^n}(\omega)\mathbf{P}(\omega)) \tag{3.92}$$

Single speech source case

In the single speech source scenario, the Woodbury identity can be applied to invert the pencil matrix $(\mathbf{R}_s(\omega) + \mathbf{R}_n(\omega))$, leading to (see also Appendix B.1):

$$\boxed{\mathbf{W}_{\mathrm{Int}(Q \leq M)}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2} [\frac{\mathbf{R}_n(\omega)^{-1}\mathbf{R}_s(\omega)}{1 + \rho(\omega)} (\mathbf{G}_{1,\Delta}(\omega) + \mathbf{P}(\omega)) - \mathbf{P}(\omega)]} \tag{3.93}$$

Note that this filter $W_{\mathrm{Int}(Q \leq M)}(\omega)$ can be separated into two filters:

$$\mathbf{U}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2} \frac{\mathbf{R}_n(\omega)^{-1}\mathbf{R}_s(\omega)}{1 + \rho(\omega)} \mathbf{G}_{1,\Delta}(\omega) \tag{3.94}$$

$$\mathbf{V}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2} [\frac{\mathbf{R}_n(\omega)^{-1}\mathbf{R}_s(\omega)}{1 + \rho(\omega)} \mathbf{P}(\omega) - \mathbf{P}(\omega)] \tag{3.95}$$

The first filter $\mathbf{U}(\omega)$ is an FxMWF-based NR filter that also compensates for the effects of the secondary path. Expression (3.94) is indeed very similar to (2.58).

The second filter $\mathbf{V}(\omega)$ is an ANC filter that estimates the noise component of the leakage signal. The output signal of $\mathbf{V}(\omega)$ after filtering by the secondary path, is given by:

$$\begin{aligned}
C(\omega)\mathbf{V}(\omega)^H\mathbf{X}(\omega) &= \frac{C(\omega)\hat{C}(\omega)^*}{|\hat{C}(\omega)|^2}\mathbf{P}(\omega)^H\frac{\mathbf{R}_n(\omega)^{-1}\mathbf{R}_s(\omega)}{1+\rho(\omega)}\mathbf{X}(\omega) - \mathbf{P}(\omega)^H\mathbf{X}(\omega) \\
&\approx \frac{C(\omega)\hat{C}(\omega)^*}{|\hat{C}(\omega)|^2}(L^s(\omega) - L(\omega)) \qquad (3.96) \\
&\approx -\frac{C(\omega)\hat{C}(\omega)^*}{|\hat{C}(\omega)|^2}(L^n(\omega)) \qquad (3.97) \\
&\approx -L^n(\omega) \qquad (3.98)
\end{aligned}$$

assuming that the secondary path identification error is small ($\hat{C}(\omega) \approx C(\omega)$). The output SNR of the system, taking both the signal leakage and the secondary path effects into account, is given by (see also Appendix B.2):

$$SNR_{\mathrm{Int}(Q\leq M)}(\omega) = \frac{\mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^s(\omega) + L^s(\omega)|^2\}}{\mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^n(\omega) + L^n(\omega)|^2\}} \qquad (3.99)$$

$$\boxed{SNR_{\mathrm{Int}(Q\leq M)}(\omega) = \rho(\omega) = SNR_{NR(noLeakage)}(\omega)} \qquad (3.100)$$

The secondary path and the noise component of the leakage signal are included in the cost function of the integrated ANC and NR scheme (3.74). Therefore, the signal leakage has no effect on the performance of the system and the scheme delivers a constant output SNR for any gain $G(\omega)$. The SNR is then equal to the SNR of the MWF-based NR scheme in the absence of signal leakage and the secondary path effects (see also Section 3.6).

3.5.3 Output signal-to-noise ratio when the number of sources is larger than the numbers of microphones

When the number of sound sources is larger than the number of microphones ($Q > M$), the leakage signal can only be approximated by a linear combination of the input signals. The leakage signal can then be rewritten as:

$$\begin{aligned}
L(\omega) &= \tilde{\mathbf{P}}(\omega)^H\mathbf{X}(\omega) + e_L(\omega) \qquad (3.101) \\
\tilde{\mathbf{P}}(\omega)^T &= [\tilde{P}_1(\omega)\ldots\tilde{P}_M(\omega)] \qquad (3.102)
\end{aligned}$$

where $e_L(\omega)$ is the estimation error. Here the filter $\tilde{\mathbf{P}}(\omega)$ is designed to minimise the mean-square value of $e_L(\omega)$:

$$\mathbb{E}\{|e_L(\omega)|^2\} = \mathbb{E}\{|L(\omega) - \tilde{\mathbf{P}}(\omega)^H \mathbf{X}(\omega)|^2\} \tag{3.103}$$

Based on this expression for the leakage signal, the MSE criterion (3.74) used to design the integrated filter $\mathbf{W}_{\text{Int}}(\omega)$ can be rewritten as:

$$J_{\text{Int}}(\omega) = \mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H \mathbf{X}(\omega) + \tilde{\mathbf{P}}(\omega)^H \mathbf{X}^n(\omega) + e_L^n(\omega) - \mathbf{G}_{1,\Delta}(\omega)^H \mathbf{X}^s(\omega)|^2\} \tag{3.104}$$

The estimation error $e_L(\omega)$ is orthogonal to the microphone signals and to the microphone signals filtered by $\tilde{\mathbf{P}}(\omega)$ [40]:

$$\mathbb{E}\{\mathbf{X}(\omega) e_L(\omega)^*\} = 0 \tag{3.105}$$

$$\mathbb{E}\{\tilde{\mathbf{P}}(\omega)^H \mathbf{X} e_L(\omega)^*\} = 0 \tag{3.106}$$

If the filter $\mathbf{W}(\omega)$ varies slowly, it can be shown that $e_L(\omega)$ is also orthogonal to the microphone signals filtered by $\mathbf{W}(\omega)$:

$$\mathbb{E}\{\mathbf{W}(\omega)^H \mathbf{X}(\omega) e_L(\omega)^*\} = 0 \tag{3.107}$$

The integrated filter $\mathbf{W}_{\text{Int}(Q>M)}(\omega)$ can then be rewritten as in (3.92):

$$\mathbf{W}_{\text{Int}(Q>M)}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2} (\mathbf{R}_{X^s}(\omega) + \mathbf{R}_n(\omega))^{-1} (\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) - \mathbf{R}_{X^n}(\omega)\tilde{\mathbf{P}}(\omega)) \tag{3.108}$$

which is (3.92) with $\mathbf{P}(\omega)$ replaced by $\tilde{\mathbf{P}}(\omega)$.

The absence of the estimation error $e_L(\omega)$ in formula (3.108) implies that the integrated ANC and NR scheme can compensate only for the estimated part of the leakage signal. The scheme can fully compensate for signal leakage when the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones and the estimation error $e_L(\omega)$ is zero. When the number of sources is larger than the number of microphones, the integrated ANC and NR scheme will only compensate for the estimated leakage signal ($\tilde{\mathbf{P}}(\omega)^H \mathbf{X}(\omega)$).

Single speech source case

In the single speech source scenario, the Woodbury identity can be applied to invert the pencil matrix $(\mathbf{R}_s(\omega) + \mathbf{R}_n(\omega))$ in (3.108), leading to an expression similar to (3.100):

$$\boxed{\mathbf{W}_{\mathrm{Int}(Q>M)}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\frac{\mathbf{R}_{X^n}(\omega)^{-1}\mathbf{R}_{X^s}(\omega)}{1+\rho(\omega)}(\mathbf{G}_{1,\Delta}(\omega) + \tilde{\mathbf{P}}(\omega)) - \tilde{\mathbf{P}}(\omega)]} \tag{3.109}$$

Assuming that the secondary path identification error is small ($\hat{C}(\omega) \approx C(\omega)$), the output SNR of the system is given by (see also Appendix B.3):

$$SNR_{\mathrm{Int}(Q>M)}(\omega) = \frac{\mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^s(\omega) + L^s(\omega)|^2\}}{\mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^n(\omega) + L(\omega)n|^2\}}$$

$$\boxed{= \frac{\frac{\rho(\omega)^2}{(\rho(\omega)+1)^2}(P_{D_{\mathrm{NR}}}(\omega) + \beta(\omega) + \tilde{\mathbf{P}}(\omega)^H R_s(\omega)\tilde{\mathbf{P}}(\omega)) + E_{e_L^s}(\omega)}{\frac{\rho(\omega)}{(\rho(\omega)+1)^2}(P_{D_{\mathrm{NR}}}(\omega) + \beta(\omega) + \tilde{\mathbf{P}}(\omega)^H R_s(\omega)\tilde{\mathbf{P}}(\omega)) + E_{e_L^n}(\omega)}} \tag{3.110}$$

where $E_{e_L^s}(\omega)$ and $E_{e_L^n}(\omega)$ are the energy of the speech component and the noise component in the estimation error e_L, respectively:

$$E_{e_L^s}(\omega) \quad = \quad \mathbb{E}\{|e_L^s(\omega)|^2\} \tag{3.111}$$

$$E_{e_L^n}(\omega) \quad = \quad \mathbb{E}\{|e_L^n(\omega)|^2\} \tag{3.112}$$

and $\beta(\omega)$ is defined as follows:

$$\beta(\omega) \triangleq \tilde{\mathbf{P}}(\omega)^H\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) + \mathbf{G}_{1,\Delta}(\omega)^H\mathbf{R}_{X^s}(\omega)\tilde{\mathbf{P}}(\omega) \tag{3.113}$$

When the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones, the estimation error $e_L(\omega)$ is zero. Formula (3.110) then reduces to (3.93). In this case, the integrated ANC and NR scheme delivers a constant output SNR equal to the output SNR of the MWF-based NR in the absence of signal leakage and secondary path effects. This confirms what was shown in the previous section.

When the number of sources is larger than the number of microphones, the estimation error $e_L(\omega)$ is not zero in general. One can then again identify the two extreme cases for the output SNR of the integrated ANC and NR scheme:

$$\begin{aligned}
\lim_{G(\omega)\to 0} \mathbf{W}_{\mathrm{Int}(Q>M)}(\omega) &= \frac{\frac{\rho(\omega)^2}{(\rho(\omega)+1)^2}\tilde{\mathbf{P}}(\omega)^H R_s(\omega)\tilde{\mathbf{P}}(\omega) + E_{e_L^s}(\omega)}{\frac{\rho(\omega)}{(\rho(\omega)+1)^2}\tilde{\mathbf{P}}(\omega)^H R_s(\omega)\tilde{\mathbf{P}}(\omega) + E_{e_L^n}(\omega)} \\
\lim_{G(\omega)\to \infty} \mathbf{W}_{\mathrm{Int}(Q>M)}(\omega) &= \rho(\omega) = SNR_{NR(noLeakage)}(\omega) \qquad (3.114)
\end{aligned}$$

As the number of sources grows, the energy of the estimation error $e_(\omega)L$ is expected to grow also. The extreme output SNR case for low gain then becomes:

$$\lim_{(G(\omega),Q)\to(0,\infty)} \mathbf{W}_{\mathrm{Int}(Q>M)}(\omega) = \frac{E_{e_L^s}(\omega)}{E_{e_L^n}(\omega)} = SNR_{e_L}(\omega) \qquad (3.115)$$

The integrated ANC and NR scheme then tends to behave more like an FxMWF-based NR under signal leakage effects as it only compensates for the secondary path effects (see also Section 3.6).

3.6 Experimental results

The algorithms introduced in this chapter have been tested experimentally and their performance has been compared to the performance of the MWF-based NR presented in Chapter 2. The performance of the integrated ANC and NR scheme is then analysed in a single speech source scenario.

3.6.1 Experimental setup

The time-domain filter lengths are set to $N = 64$ and $N_{\mathrm{ANC}} = 64$, and the NR delay is set to half of the NR filter length ($\Delta = 32$). The secondary path $\mathbf{c}[k]$ is estimated offline using an identification technique based on the Normalised Least Mean Square (NLMS) algorithm. The length of the estimated path $\hat{\mathbf{c}}[k]$ is set to $\hat{P} = 32$.

Signal leakage and secondary path effects

The first experiment shows the effect of leakage on the NR performance and the improvement achieved by ANC. For an gain G varying from 0dB to 20dB the inputs are filtered using the three algorithms previously described (the standard MWF-based NR scheme, the cascaded NR and multichannel ANC scheme and the integrated ANC and NR scheme) and the SNR improvement is evaluated. The system is calibrated so that for $G = 0\text{dB}$, for a source at 0°, the leakage and the signal fed in the loudspeaker have equal power.

Causality

The second test aims to demonstrate the impact of delay on the ANC performance with the different algorithms. The gain is fixed to $G = 10\text{dB}$. For a varying causality margin δ (1.5), the SNR performances of the MWF-based NR, the cascaded NR and multichannel ANC scheme and the integrated ANC and NR scheme are compared. In practice, a variable artificial delay Δ_{HA} is added to the microphone signals in order to have a system with a variable causality margin.

Frequency-domain version

Two experiments are then presented in frequency-domain in the single speech source case. The frequency-domain filter length is set to $N = 128$, and the NR delay is set to half of the NR filter length ($\Delta = 64$).

The first experiment shows the output SNR of the MWF-based NR and the integrated ANC and NR scheme when only a noise source at 270° is active. The second experiment aims at showing the effect of the number of sources (speech plus noise sources) on the output SNR of the integrated ANC and NR scheme.

3.6.2 Combined active noise control and noise reduction

Figure 3.6(a) shows the performance of the combinations of NR and ANC (cascaded NR and multichannel ANC (Figure 3.2) and integrated ANC and NR (Figure 2.4)) compared to a standard MWF-based NR, for gain $G \in [0\ 20]$ dB. The causality margin is artificially kept sufficiently high (by adding a delay to Δ_{Pri}) so that no performance trade-off has to be made between the NR and the ANC ($\delta = 48$, *i.e.*, for $\Delta_{alg} \leq 48$ the criterion (1.4) is fulfilled).

For gains up to 15dB, cascading NR and ANC allows to maintain a constant SNR improvement of around 4dB, which is already significantly better than the performance of the MWF-based NR alone. When the gain is increasing above 15dB, the performance converges to the performance of the MWF-based NR alone. The integrated approach delivers an almost constant SNR improvement around 12dB for all values of gain. This is better than the MWF-based NR alone or the cascaded NR and ANC scheme.

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.

3.6.3 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 2 samples (rather than the 48 samples used previously).

To see how the delays in the system can affect performance, for each algorithm the gain is set to $G = 5\text{dB}$, and a variable delay (Δ_{HA}) is added to the microphone signals (BTE microphones), to allow the causality margin to vary between -16 and 48. Figure 3.6(b) shows the SNR improvement for each algorithm as a function of the causality margin δ.

When the causality margin is high, the performance for the cascaded NR and multichannel ANC scheme is close to what was obtained in Figure 3.6(a). When the causality margin decreases and becomes lower than the NR delay ($\delta \leq \Delta = 32$), the SNR improvement starts to decrease and eventually converges to the improvement obtained with the classic MWF-based NR. This is due to the fact that for a causality margin lower than 32, the ANC effectively has to be designed as a non-causal system ($\Delta_{\text{ANC}} \leq 0$). Therefore, its performance is reduced. The integrated ANC and NR scheme delivers an almost constant SNR improvement as long as the overall system is causal ($\delta \geq 0$).

Figure 3.7(a) shows the SNR improvement given by the three algorithms for a realistic causality margin ($\delta = 2$), which corresponds to what has been measured on the transfer functions used for the simulations. For lower gains, $G \leq 15\text{dB}$, the cascaded ANC and NR scheme delivers only minor improvement (around 1dB) compared to the standard MWF-based NR scheme. The performance of these two schemes then tend to converge as the gain is increased. The integrated ANC and NR scheme on the other hand maintains an SNR improvement of more than 10dB. Therefore, the integrated ANC and NR scheme seems to offer a practical way to introduce ANC in hearing aids.

Finally, Figure 3.7(b) shows the SNR improvement for $\delta = -16$ for the integrated ANC and NR scheme and the MWF-based NR scheme. Here, the performance improvement achieved with the integrated ANC and NR scheme is mainly due to the secondary path compensation. Fig. 3.7(b) also shows the SNR improvement obtained with a FxMWF-based NR scheme that compensates for the effect of the secondary path but does not include the ANC. This latter scheme is indeed found to achieve a similar performance improvement as the integrated ANC and NR scheme in a non-causal ($\delta \leq 0$) scenario.

3.6.4 Frequency-domain version, single speech source scenario

A frequency-domain version of the integrated ANC and NR scheme has been implemented in order to compare the simulated SNR performance with the theoretical formulae derived in this chapter in the single speech source case. Formulae derived for frequency-domain version of the integrated ANC and NR are valid only for a causal ($\delta \geq 0$) scenario. Therefore, during these experiment the causality margin is (artificially) set to $\delta = 12$.

Output signal-to-noise ratio in a single noise source scenario

Figure 3.8(a) presents the theoretical output SNR of the MWF-based NR with signal leakage, with signal leakage and secondary path effects and the theoretical output SNR of the integrated ANC and NR scheme, based on the formulae derived in Sections 2.2 and 3.3. The theoretical SNR is presented for a gain G varying from 0dB to 20dB. The gain G is flat, *i.e.*, it takes the same value for all frequency bands. The system is calibrated so that for $G = 0$dB, for a source at 0°, the leakage signal and the signal fed in the loudspeaker have equal full-band power.

When the signal leakage is the only disturbance considered, the degradations induced by the signal leakage remain small even for reasonably small gain G (down to 8dB). When both the signal leakage and the secondary path are introduced, however, the degradations are significant for gains up to at least 20dB. The integrated ANC and NR scheme delivers a constant output SNR for any gain G. This confirms the results observed from Section 3.6.2.

Figure 3.8(b) presents the simulated output SNR for the MWF-based NR (with signal leakage only and with signal leakage and secondary path effects) and the simulated output SNR for the integrated ANC and NR scheme. The gain G varies from 0dB to 20dB. The performance of the MWF-based NR under the different disturbances is similar to the theoretical performance. The integrated ANC and NR scheme delivers an almost constant output SNR which also confirms the theoretical results, *i.e.*, that the integrated ANC and NR scheme allows to restore the NR performance and deliver a constant output SNR for any gain G in the scenario where the number of sources is less than or equal to the number of microphones.

Output signal-to-noise ratio in a multiple noise sources scenario

To evaluate the effects of the number of sound sources (speech plus noise sources) on the performance of the integrated ANC and NR scheme, multiple noise sources are used to compose the input signals. When only one noise source is active, the noise source at 270° (facing the hearing aid) is used. Then the noise source at 90°

is added for the 2 noise sources scenario, the noise source at 330° is added for the 3 noise sources scenario and finally the noise source at 30° is added for the 4 noise sources scenario.

As explained in Section 3.5.3 the number of sound sources (speech plus noise sources) has an impact on the performance of the integrated ANC and NR scheme. The number of noise sources similarly has an impact on the performance of MWF-based NR schemes, *i.e.*, ρ depends on the number of noise sources (see also Figure 2.6(b)). In order to observe the effects of the number of sound sources (speech plus noise sources) on the ANC part of the integrated ANC and NR schemes it is then more convenient to look at the normalised output SNR ($\frac{SNR}{\rho}$).

Figure 3.9(a) presents the simulated output SNR performance of the integrated ANC and NR scheme for a gain of 0dB (where the integrated ANC and NR scheme acts as an ANC filter that estimates the noise component of the leakage signal) and a gain of 20dB (where the integrated ANC and NR scheme acts as an MWF-based NR when secondary path effects and signal leakage are neglected) and for a number of noise sources varying from 1 to 4. For a gain of 20dB the output SNR decreases when the number of noise sources increases. This shows the effect of the number of noise sources on the NR part of the integrated ANC and NR scheme. For a gain of 0dB the degradation of the output SNR introduced by the number of noise sources is related to the fact that the integrated ANC and NR scheme can compensate only for an estimate of the leakage signal when the number of sources becomes larger than the number of microphones (as explained in Section 3.5.3).

Figure 3.9(b) presents the normalised simulated output SNR of the integrated ANC and NR scheme, for a gain varying from 0dB to 20dB and for scenarios with 1 to 4 noise sources. For large gains, the integrated ANC and NR scheme delivers the same output SNR as the MWF-based NR in absence of signal leakage and secondary path effects for any number of noise sources. For small gains, when the number of noise sources increases, the output SNR decreases. The ANC can compensate only for the estimated part of the leakage signal and the integrated ANC and NR scheme tends to behave as a FxMWF-based NR compensating for the secondary path. This can be seen in Figure (3.10) where the simulated output SNR performance of the integrated ANC and NR scheme and of the MWF-based NR are plotted for the 4 noise sources scenario. The integrated ANC and NR scheme then behaves as the MWF-based NR under signal leakage effects (*i.e.*, when the effects of the secondary path are compensated for).

3.7 Conclusion

Standard NR techniques used in hearing aids ignore leakage and secondary path effects. When open fitting BTE's are used these effects 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 was shown to provide SNR improvements between 4dB and 12dB depending on the approach used.

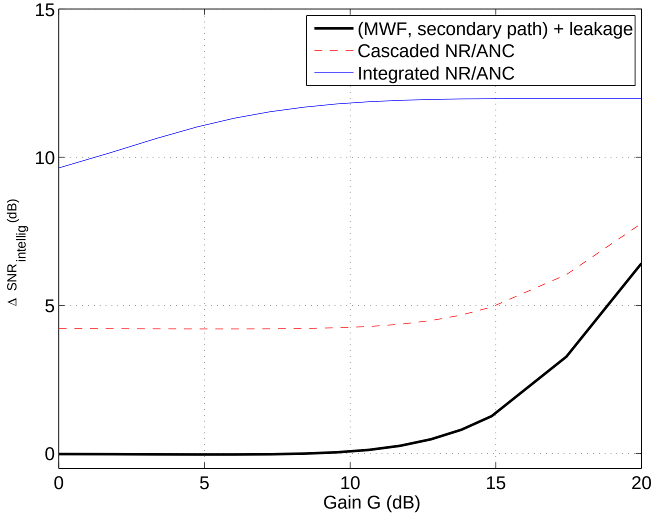
The ANC performance is conditioned by the system causality margin which differs for the two algorithms evaluated here. A cascaded ANC and NR scheme can deliver good performance (SNR improvement around 4dB) as long as the system causality margin is high enough to support the NR latency ($\delta \geq N/2$).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

An alternative integrated approach of ANC with NR has shown to improve the SNR by about 12dB for low hearing aid gains (between 0dB and 20dB), as long as the causality margin of the system is positive ($\delta \geq 0$). When the causality margin of system becomes negative, the integrated ANC and NR scheme can still outperform standard MWF-based NR schemes 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.

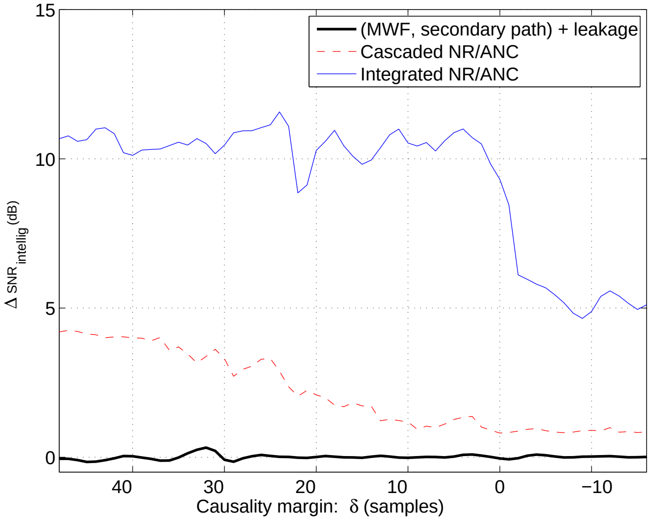
In the single speech source case, it has been shown theoretically that the signal leakage and the secondary path effects degrade the performance of a standard NR based on MWF, especially so when a small gain G is applied in the hearing aid. This confirms previous observations and simulation results as well as the need, with increased usage of hearing aids with open fitting, for a scheme that compensates for signal leakage.

It has been shown that integrating NR and ANC in a single set of filters allows to compensate for the signal leakage and the secondary path effects. When the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones the leakage signal can be written as a linear combination of the microphone signals. It is then possible to derive a simple formula for the output SNR of the integrated ANC and NR scheme. The integrated ANC and NR scheme then allows to restore the NR performance and delivers a constant output SNR for any gain G as in absence of signal leakage and secondary path effects.

When the number of sound sources is larger than the number of microphones, it is possible to rewrite the leakage signal as the sum of a linear combination of the microphone signals and an estimation error. The integrated ANC and NR scheme can then compensate only for the estimated part of the leakage signal. When the number of sound sources grows so does the estimation error, and the integrated ANC and NR scheme then tends to behave like an FxMWF-based NR, which also compensates for the secondary path effect. Therefore, it still improves the output SNR performance compared to a standard MWF-based NR.

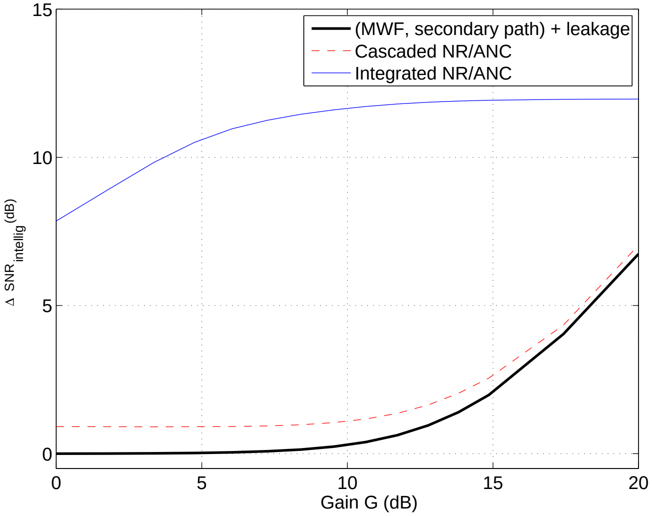


(a)

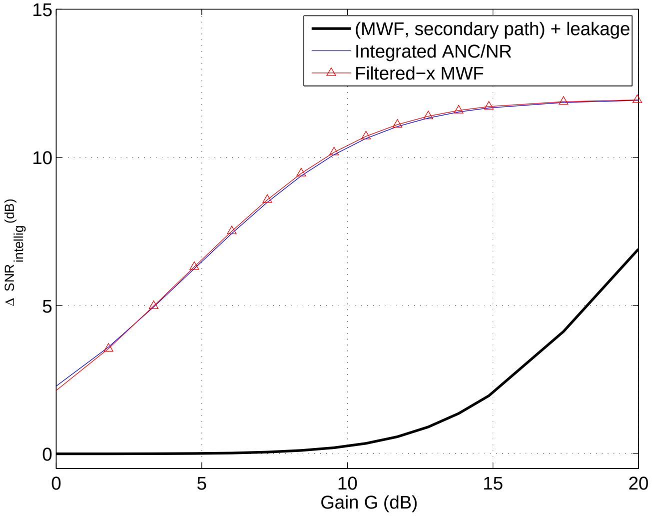


(b)

Figure 3.6: Performance comparison for NR scheme with or without ANC: (a) when the causality margin is sufficiently high for a cascaded scheme ($\delta = 48$) (b) Performance depending on the causality margin

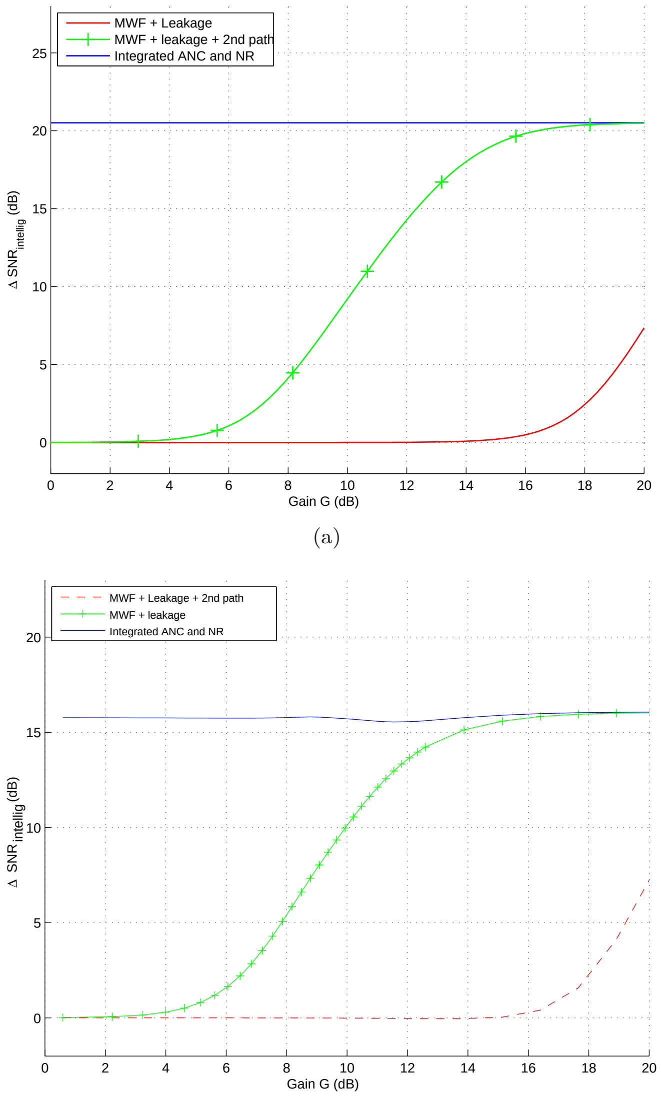


(a)



(b)

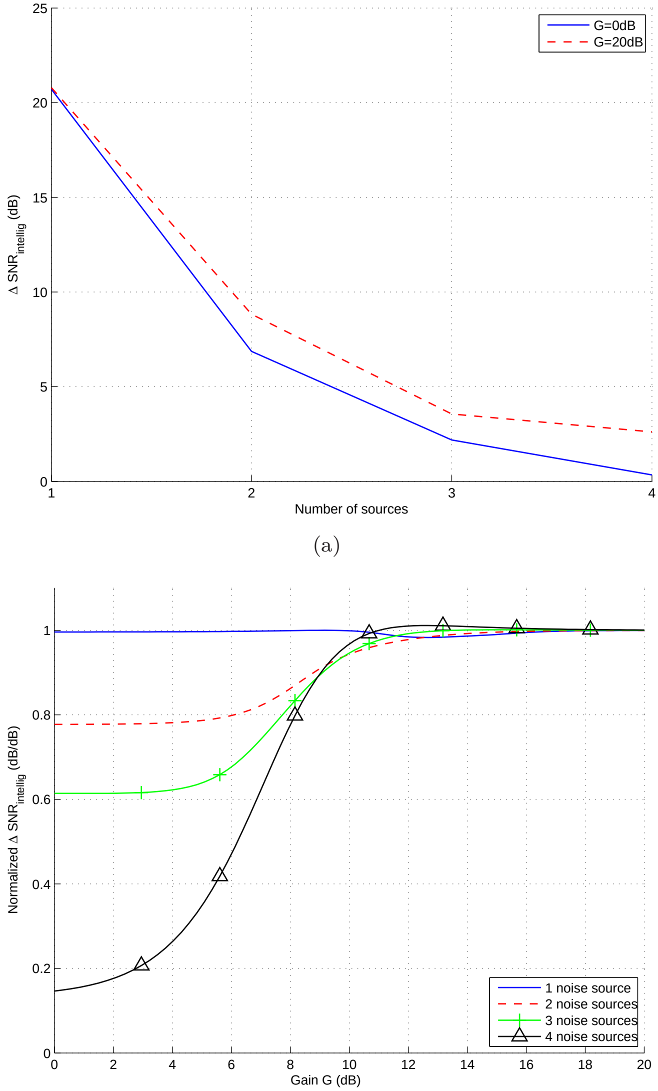
Figure 3.7: (a) Performance comparison for NR scheme with or without NR for a system with low causality margin, $\delta = 2$. (b) Performance for a non-causal system, $\delta = -16$



(a)

(b)

Figure 3.8: SNR output for the MWF-based NR and the integrated ANC and NR scheme in a single speech source scenario: (a) Theoretical output SNR. (b) Simulated output SNR



(a)

(b)

Figure 3.9: (a) Simulated output SNR for the integrated ANC and NR scheme depending on the number of noise sources ($G = 0$dB and $G = 20$dB)Theoretical output SNR. (b) Normalised simulated output SNR for the integrated ANC and NR scheme (noise source at 270°, 90°, 330° and 90°).

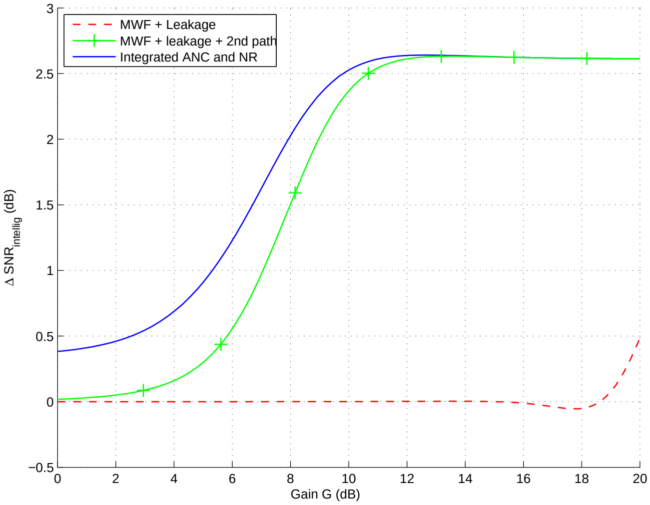


Figure 3.10: Simulated output SNR for MWF-based NR scheme and for the integrated ANC and NR scheme (noise source at 30°, 90°, 270° and 330°)

Chapter 4

Weighted Integrated Active Noise Control and Noise Reduction

The integrated ANC and NR scheme introduced in Chapter 3 does not allow to balance between the NR and the ANC and may introduce higher speech distortion (SD) than a standard MWF-based NR. In some circumstances, however, it would be useful to emphasise one of the functional blocks.

Changing the original optimisation problem to a constrained optimisation problem leads to a scheme based on a weighted mean squared error criterion which allows to focus on the ANC or on the NR. It is similarly possible to derive a scheme that allows to focus on reducing the SD or on reducing the residual noise at the eardrum (*i.e.*, ANC). In a single speech source scenario and when the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones, it is possible to derive a simple formula for the output SNR of the latter scheme. It can then be shown that the scheme delivers a constant SNR at the eardrum for any weighting factor.

4.1 Introduction

The objectives of the integrated ANC and NR scheme presented in Chapter 3 are to attenuate the noise component of the leakage (*i.e.*, ANC) and to minimise the difference between an unknown desired speech signal and the signal delivered at the eardrum (*i.e.*, NR), the balance between these two objectives being fixed. In some cases however, it will be useful to adjust the trade-off between ANC and NR, *e.g.*, when the input signal does not contain any desired speech component or when the ANC is found to be inefficient.

The concept of weighted NR has been introduced in [28] and later applied in the MWF framework to derive the so-called SDW-MWF [20][21][77]. A similar approach is used here to derive a weighted version of the integrated ANC and NR scheme based on the FxMWF. The weighted scheme then allows to emphasise either the ANC or the NR ultimately providing a lower residual noise power at the eardrum or a lower SD at the eardrum depending on the weight applied.

This weighted ANC and NR scheme, however, does not reduce to the original unweighted integrated ANC and NR scheme for any weighting factor. Besides, focusing on NR allows to reduce SD compared to unweighted ANC and NR but NR itself still introduces SD. Similarly to SDW-MWF, a speech distortion weighted integrated ANC and NR scheme (SDW-ANC/NR) is then derived that truly allows to balance between reducing the SD and reducing the residual noise at the eardrum. In the single speech source scenario and when the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones, it is possible to derive theoretically the output SNR of the SDW-ANC/NR scheme at the eardrum. The SDW-ANC/NR scheme is then shown to deliver a constant SNR at the eardrum for any weighting factor.

This chapter presents weighted approaches to integrated ANC and NR. Section 4.2 introduces a weighted integrated ANC and NR scheme that allows to balance between ANC and NR. An SDW-ANC/NR is then presented and the theoretical output SNR is derived in a single speech source scenario in Section 4.3. The performance of the original unweighted integrated ANC and NR scheme, the weighted integrated ANC and NR scheme and the SDW-ANC/NR scheme formulated here, all of them based on FxMWF and applied in hearing aids with an open fitting, are compared in Section 4.4. Finally, Section 4.5 presents a summary of this chapter.

4.2 Weighted integrated active noise control and noise reduction

The integrated ANC and NR scheme introduced in Chapter 3 imposes a fixed trade-off between the NR and the ANC. It is possible, however, to modify the optimisation problem in order to derive a filter with a variable ANC versus NR trade-off.

4.2.1 Fixed trade-off between noise reduction and active noise control

The unweighted integrated ANC and NR scheme minimises an MSE criterion (3.54) which can be viewed as the sum of an ANC term and a SD term as follows:

$$\begin{aligned}\mathbb{E}\{|e_{\text{Int}}|^2\} =&\mathbb{E}\{|\mathbf{c}[k]^T\mathbf{z}^s[k]-d_{\text{NR}}[k]|^2\}\\ &+\mathbb{E}\{|\mathbf{c}[k]^T\mathbf{z}^n[k]+l^n[k]|^2\}\end{aligned} \tag{4.1}$$

Therefore, the unweighted integrated ANC and NR scheme may exhibit lower noise attenuation performance at the eardrum than an ANC filter alone, minimising the MSE criterion:

$$\mathbb{E}\{|e_{\text{ANC}}|^2\} =\mathbb{E}\{|\mathbf{c}[k]^T\mathbf{z}^n[k]+l^n[k]|^2\} \tag{4.2}$$

On the other hand, when the leakage signal is dominant (*e.g.*, for low gains G), the ANC term dominates the SD term in the optimisation, eventually leading to a large SD. Therefore, the unweighted integrated ANC and NR scheme is found to introduce more SD than a standard NR scheme minimising the MSE criterion:

$$\mathbb{E}\{|e_{\text{NR}}|^2\} =\mathbb{E}\{|\mathbf{c}[k]^T\mathbf{z}[k]-d_{\text{NR}}[k]|^2\} \tag{4.3}$$

When the input signal does not contain any speech, the NR is not needed and the system should just focus on minimizing the residual noise at the eardrum. In this case an ANC scheme alone can perform better than the unweighted integrated ANC and NR scheme and deliver lower residual noise at the eardrum.

On the other hand, if the background noise is, *e.g.*, high-frequency noise when typically the ANC is found to be inefficient, using a NR scheme alone may reduce the SD introduced by the unweighted integrated ANC and NR scheme (see also Section 4.4).

The MSE criterion (4.1) which is minimised by the unweighted integrated ANC and NR scheme, cannot be decomposed in the sum of an ANC (4.2) term and an NR term (4.3). It is therefore not possible to derive straightforwardly a scheme that balances between the ANC and the NR from the unweighted integrated ANC and NR scheme. Hence, it is necessary to first reformulate the optimisation problem.

4.2.2 Constrained problem formulation

The algorithm described in this section applies a different weight to the ANC objective (4.2) and to the NR objective (4.3) of the integrated ANC and NR scheme. Therefore, unlike for the unweighted integrated ANC and NR scheme described in Chapter 3, the two MSE criteria have to be computed explicitly. A weight is then applied to one of the criteria, allowing to emphasise the effect of one of the two functional blocks.

The problem can be seen as trying to minimise the residual noise at the eardrum *(i.e.*, ANC) under the constraint that the difference between the desired signal and the filtered signal, as delivered to the eardrum (*i.e.*, NR), is kept below a given threshold:

$$\min_{\mathbf{w}} \mathbb{E}\{|e_{\mathrm{ANC}}|^2\}, \text{ subject to } \mathbb{E}\{|e_{\mathrm{NR}}|^2\} \leq T \tag{4.4}$$

Introducing the Lagrange-multiplier $\mu > 0$, the MSE criterion to be minimised is then :

$$J_{\mu,\mathrm{Int}}[k] = \mu\mathbb{E}\{|e_{\mathrm{NR}}|^2\} + \mathbb{E}\{|e_{\mathrm{ANC}}|^2\} \tag{4.5}$$

The Lagrange-multiplier $\mu \in]0, \infty[$ then acts as a trades-off parameter between the NR and the ANC:

- When $\mu \rightarrow 0$, the MSE criterion in (4.5) reduces to (4.2). The system then behaves as a standard ANC algorithm. The algorithm then achieves a high noise attenuation performance but it may also introduce significant SD, as the speech component is not taken into account in the optimisation process.
- When $\mu \rightarrow \infty$, the MSE criterion in (4.5) reduces to (4.3). The system then behaves as an FxMWF-based NR algorithm. The algorithm introduces less SD but the noise attenuation performance at the eardrum is decreased. The effects of the leakage signal are not compensated for any more.

In practice, to compute the MSE in (4.5) the two error signals ($e_{\mathrm{ANC}}[k]$ and $e_{\mathrm{NR}}[k]$) have to be computed explicitly. Assuming that the secondary path identification error is small ($\hat{\mathbf{c}}[k] \approx \mathbf{c}[k]$), these error signals can be written as follows:

$$\begin{aligned} e_{\mathrm{ANC}}[k] &\approx \mathbf{w}[k]^T\mathbf{y}^n[k] + l^n[k] & (4.6)\\ e_{\mathrm{NR}}[k] &\approx \mathbf{w}[k]^T\mathbf{y}[k] - d_{\mathrm{NR}}[k] & (4.7) \end{aligned}$$

If the noise and the speech components in (4.7) are uncorrelated, the MSE criteria (4.2) and (4.3) can be rewritten as follows:

$$\begin{aligned} \mathbb{E}\{|e_{\mathrm{ANC}}|^2\} &\approx \mathbb{E}\{|\mathbf{w}[k]^T\mathbf{y}^n[k] + l^n[k]|^2\} & (4.8)\\ \mathbb{E}\{|e_{\mathrm{NR}}|^2\} &\approx \mathbb{E}\{|\mathbf{w}[k]^T\mathbf{y}^s[k] - d_{\mathrm{NR}}[k]|^2\} & \\ &\quad +\mathbb{E}\{|\mathbf{w}[k]^T\mathbf{y}^n[k]|^2\} & (4.9) \end{aligned}$$

The optimal filter (FxMWF), which minimises the MSE criterion in (4.5), is then:

$$\mathbf{w}_\mu[k] = \mathbf{R}_\mu[k]^{-1}\mathbf{r}_\mu[k] \tag{4.10}$$

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

$$\mathbf{R}_\mu[k] \quad = \quad \mu\mathbf{R}_{y^s}[k] + (1+\mu)\mathbf{R}_{y^n}[k] \tag{4.11}$$

$$\mathbf{r}_\mu[k] \quad = \quad \mu\mathbf{r}_{y^s d_{\mathrm{NR}}}[k] - \mathbf{r}_{y^n l^n}[k] \tag{4.12}$$

Note that if the speech and the noise components of the input signals are again uncorrelated the correlation matrix and the cross-correlation vector can be estimated using:

$$\mathbf{R}_\mu[k] \quad = \quad \mu\mathbf{R}_{y}[k] + \mathbf{R}_{y^n}[k] \tag{4.13}$$

$$\mathbf{r}_{y^s d_{\mathrm{NR}}}[k] \quad = \quad G \cdot (\mathbf{r}_{y x_{1,\Delta}}[k] - \mathbf{r}_{y^n x^n_{1,\Delta}}[k]) \tag{4.14}$$

The expression of the optimal filter can then be simplified by substituting (4.12) and (4.13) in (4.10):

$$\boxed{\mathbf{w}_\mu[k] = (\mu\mathbf{R}_{yy}[k] + \mathbf{R}_{y^n y^n}[k])^{-1}(\mu\mathbf{r}_{y^s d_{\mathrm{NR}}}[k] - \mathbf{r}_{y^n l^n}[k])} \tag{4.15}$$

From (4.15) it appears clearly that the two extreme cases for the filter $\mathbf{w}_\mu[k]$ are given by:

$$\lim_{\mu\to 0} \mathbf{w}_\mu[k] \quad = \quad -\mathbf{R}_{y^n y^n}[k]^{-1}\mathbf{r}_{y^n l^n}[k] \tag{4.16}$$

$$\lim_{\mu\to \infty} \mathbf{w}_\mu[k] \quad = \quad \mathbf{R}_{yy}[k]^{-1}\mathbf{r}_{y^s d_{\mathrm{NR}}}[k] \tag{4.17}$$

Here (4.16) is the expression of an ANC filter that minimises the noise pressure at the eardrum and (4.17) is the expression of a FxMWF-based NR filter that also compensates for the secondary path. The filter described in (4.15) therefore integrates the two functional blocks with the coefficient μ used as a trade-off parameter between the NR and the ANC.

4.2.3 Single speech source case

In the single speech source case it is possible to derive simpler frequency-domain formulae for the MWF-based filters.

The frequency-domain representation of the weighted integrated ANC and NR scheme (4.15) is given by:

$$\mathbf{W}_\mu(\omega) = (\mu \mathbf{R}_Y(\omega) + \mathbf{R}_{Y^n}(\omega))^{-1}(\mu \mathbf{r}_{Y^s D_{\mathrm{NR}}}(\omega) - \mathbf{r}_{Y^n L^n}(\omega)) \tag{4.18}$$

The leakage signal can be approximated (estimated) by a linear combination of the input signals:

$$L(\omega) = \tilde{\mathbf{P}}(\omega)^H \mathbf{X}(\omega) + e_L(\omega) \tag{4.19}$$

where $e_L(\omega)$ is the estimation error.

The weighted MSE criterion 4.5 can then be rewritten as follows:

$$\begin{aligned} J_{\mu,\mathrm{Int}}(\omega) =& \mu \mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H \mathbf{X}(\omega) - \mathbf{G}_{1,\Delta}(\omega)^H \mathbf{X}^s(\omega)|^2\} \\ &+ \mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H \mathbf{X}^n(\omega) + \tilde{\mathbf{P}}(\omega)^H \mathbf{X}^n(\omega) + e_L^n(\omega)|^2\} \end{aligned} \tag{4.20}$$

The estimation error $e_L(\omega)$ is orthogonal to the microphone signals (3.105) and to the microphone signals filtered by $\tilde{\mathbf{P}}(\omega)$ (3.106). If the filter $\mathbf{W}(\omega)$ varies slowly $e_L(\omega)$ is also orthogonal to the microphone signals filtered by $\mathbf{W}(\omega)$ (3.107).

The weighted integrated ANC and NR filter (4.18) can then be rewritten as follows:

$$\begin{aligned} \mathbf{W}_\mu(\omega) =& \mu(\omega)\frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\mu(\omega)\mathbf{R}_{X^s}(\omega) + (\mu(\omega)+1)\mathbf{R}_{X^n}(\omega)]^{-1}\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) \\ &- \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\mu(\omega)\mathbf{R}_{X^s}(\omega) + (\mu(\omega)+1)\mathbf{R}_{X^n}(\omega)]^{-1}\mathbf{R}_{X^n}(\omega)\tilde{\mathbf{P}}(\omega) \\ \mathbf{W}_\mu(\omega) =& \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\mathbf{R}_{X^s}(\omega) + \nu(\omega)\mathbf{R}_{X^n}(\omega)]^{-1}\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) \\ &- \nu(\omega)\eta(\omega)\frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\mathbf{R}_{X^s}(\omega) + \nu(\omega)\mathbf{R}_{X^n}(\omega)]^{-1}\mathbf{R}_{X^n}(\omega)\tilde{\mathbf{P}}(\omega) \end{aligned} \tag{4.21}$$

with

$$\nu(\omega) = \frac{\mu(\omega)+1}{\mu(\omega)} \qquad \nu(\omega) \in \,]1,\ \infty[\tag{4.22}$$

$$\eta(\omega) = \frac{1}{\mu(\omega)+1} \qquad \eta(\omega) \in \,]0,\ 1[\tag{4.23}$$

In a single speech source scenario, the auto-correlation of the speech signal $\mathbf{R}_{X^s}(\omega)$ is rank-1. The pencil matrix $(\mathbf{R}_{X^s}(\omega) + \nu(\omega)\mathbf{R}_{X^n}(\omega))$ can then be inverted by applying the Woodbury identity and the filter (4.18) can be expressed as follows (see also Appendix C.1):

$$\boxed{\mathbf{W}_\mu(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\frac{\mathbf{R}_{X^n}(\omega)^{-1}\mathbf{R}_{X^s}(\omega)}{\nu(\omega)+\rho(\omega)}(\mathbf{G}_{1,\Delta}(\omega) + \eta(\omega)\tilde{\mathbf{P}}(\omega)) - \eta(\omega)\tilde{\mathbf{P}}(\omega)]} \tag{4.24}$$

The expression is very similar to the expression for the so-called MWF-η in [11]. The weighted integrated ANC and NR scheme can then be seen as an SDW-MWF with partial production of anti-noise. When $\eta(\omega) \to 0$ no anti-noise is produced and the weighted integrated ANC and NR scheme behaves as an FxMWF-based NR. Increasing $\eta(\omega)$ will introduce the anti-noise and when $\eta(\omega) \to 1$ the weighted integrated ANC and NR scheme tends to produce only anti-noise, *i.e.*, the filter acts as an ANC scheme.

4.3 Speech distortion weighted integrated active noise control and noise reduction

The MSE criterion minimised by the weighted integrated ANC and NR scheme introduced in the previous section does not relate directly to the MSE criterion minimised by the unweighted integrated ANC and NR introduced in Chapter 3. Therefore, the weighted ANC and NR scheme does not reduce to the original unweighted integrated ANC and NR scheme for any weighting factor. Besides, focusing on NR allows to reduce SD compared to unweighted integrated ANC and NR but NR itself still introduces SD. In this section a speech distortion weighted integrated ANC and NR (SDW-ANC/NR) scheme is derived that allows to balance between reducing the SD and reducing the residual noise at the eardrum.

4.3.1 Constrained problem formulation

Similarly to SDW-MWF in [20, 21, 77], it is possible to derive an integrated ANC and NR scheme that applies a different weight to the ANC objective and to the SD objective. The problem can be seen as trying to minimise the residual noise at the eardrum *(i.e.*, ANC) under the constraint that the difference between the desired speech signal and the speech component of the filtered signal, as delivered to the eardrum (*i.e.*, SD), is kept below a given threshold:

$$\min_{\mathbf{w}} \mathbb{E}\{|e_{\mathrm{ANC}}|^2\}, \text{ subject to } \mathbb{E}\{|e_{\mathrm{SD}}|^2\} \leq T \tag{4.25}$$

where e_{ANC} and e_{SD} are defined as:

$$e_{\mathrm{ANC}}[k] \approx \mathbf{c}[k]^T \mathbf{z}^n[k] + l^n[k] \tag{4.26}$$

$$e_{\mathrm{SD}}[k] \approx \mathbf{c}[k]^T \mathbf{z}^s[k] - d_{\mathrm{NR}}[k] \tag{4.27}$$

Introducing the Lagrange-multiplier $\mu > 0$, the MSE criterion to be minimised is then :

$$J_{\mu,\mathrm{Int}}[k] = \mu \mathbb{E}\{|e_{\mathrm{SD}}|^2\} + \mathbb{E}\{|e_{\mathrm{ANC}}|^2\} \tag{4.28}$$

The Lagrange-multiplier $\mu \in]0, \infty[$ acts as a trades-off parameter between the SD and the ANC:

- When $\mu \to 0$, the MSE criterion in (4.28) reduces to (4.2). The system then behaves as a standard ANC algorithm. The algorithm then achieves a high noise attenuation performance but it may also introduce significant SD, as the speech component is not taken into account in the optimisation process.
- When $\mu \to \infty$, the MSE criterion in (4.28) reduces to (4.32). The system then minimizes the SD but the noise attenuation performance is decreased. The effects of the leakage signal are not compensated for any more.

In practice, to compute the MSE in (4.28) the two error signals ($e_{\mathrm{ANC}}[k]$ and $e_{\mathrm{SD}}[k]$) have to be computed explicitly. Assuming that the secondary path identification error is small ($\hat{\mathbf{c}}[k] \approx \mathbf{c}[k]$), these error signals can be written as follows:

$$e_{\mathrm{ANC}}[k] \approx \mathbf{w}[k]^T \mathbf{y}^n[k] + l^n[k] \tag{4.29}$$

$$e_{\mathrm{SD}}[k] \approx \mathbf{w}[k]^T \mathbf{y}^s[k] - d_{\mathrm{NR}}[k] \tag{4.30}$$

The MSE criteria (4.2) and (4.32) can be rewritten as follows:

$$\mathbb{E}\{|e_{\mathrm{ANC}}|^2\} \approx \mathbb{E}\{|\mathbf{w}[k]^T \mathbf{y}^n[k] + l^n[k]|^2\} \tag{4.31}$$

$$\mathbb{E}\{|e_{\mathrm{SD}}|^2\} \approx \mathbb{E}\{|\mathbf{w}[k]^T \mathbf{y}^s[k] - d_{\mathrm{NR}}[k]|^2\} \tag{4.32}$$

The optimal filter (FxMWF), which minimises the MSE criterion in (4.28), is then:

$$\mathbf{w}_\mu[k] = \mathbf{R}_\mu[k]^{-1}\mathbf{r}_\mu[k] \tag{4.33}$$

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

$$\mathbf{R}_\mu[k] = \mu\mathbf{R}_{y^s}[k] + \mathbf{R}_{y^n}[k] \tag{4.34}$$

$$\mathbf{r}_\mu[k] = \mu\mathbf{r}_{y^s d_{\text{NR}}}[k] - \mathbf{r}_{y^n l^n}[k] \tag{4.35}$$

The optimal filter can then be rewritten as follows by substituting (4.35) and (4.34) in (4.33):

$$\boxed{\mathbf{w}_\mu[k] = (\mu\mathbf{R}_{y^s y^s}[k] + \mathbf{R}_{y^n y^n}[k])^{-1}(\mu\mathbf{r}_{y^s d_{\text{NR}}}[k] - \mathbf{r}_{y^n l^n}[k])} \tag{4.36}$$

This solution will be referred to as speech distortion weighted integrated ANC and NR (SDW-ANC/NR) scheme.

From (4.36) it appears clearly that the two extreme cases for the filter $\mathbf{w}_\mu[k]$ are given by:

$$\lim_{\mu\to 0}\mathbf{w}_\mu[k] = -\mathbf{R}_{y^n y^n}[k]^{-1}\mathbf{r}_{y^n l^n}[k] \tag{4.37}$$

$$\lim_{\mu\to\infty}\mathbf{w}_\mu[k] = \mathbf{R}_{y^s y^s}[k]^{-1}\mathbf{r}_{y^s d_{\text{NR}}}[k] \tag{4.38}$$

Here (4.37) is the expression of an ANC filter, which minimises the residual noise at the eardrum, and (4.38) is the expression of a FxMWF-based filter that minimizes the SD at the eardrum. The filter described in (4.36) therefore integrates the two functional blocks with the coefficient μ used as a trade-off parameter between the SD and the ANC.

4.3.2 Single speech source case

In the single speech source case it is possible to derive simpler frequency-domain formulae for the MWF-based filters and for their SNR performance.

The frequency-domain representation of the SDW-ANC/NR scheme (4.36) is given by:

$$\mathbf{W}_\mu(\omega) = (\mu(\omega)\mathbf{R}_{Y^s}(\omega) + \mathbf{R}_{Y^n}(\omega))^{-1}(\mu(\omega)\mathbf{r}_{Y^s D_{\text{NR}}}(\omega) - \mathbf{r}_{Y^n L^n}(\omega)) \tag{4.39}$$

The leakage signal can be approximated (estimated) by a linear combination of the input signals (4.19). The weighted MSE criterion 4.28 can then be rewritten as follows:

$$\begin{aligned} J_{\mu,\mathrm{Int}}(\omega) =& \mu(\omega)\mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^s(\omega) - \mathbf{G}_{1,\Delta}(\omega)^H\mathbf{X}^s(\omega)|^2\} \\ &+ \mathbb{E}\{|C(\omega)\mathbf{W}(\omega)^H\mathbf{X}^n(\omega) + \tilde{\mathbf{P}}(\omega)^H\mathbf{X}^n(\omega) + e_L^n(\omega)|^2\} \end{aligned} \tag{4.40}$$

The estimation error $e_L(\omega)$ is orthogonal to the microphone signals (3.105) and to the microphone signals filtered by $\tilde{\mathbf{P}}(\omega)$ (3.106). If the filter $\mathbf{W}(\omega)$ varies slowly $e_L(\omega)$ is also orthogonal to the microphone signals filtered by $\mathbf{W}(\omega)$ (3.107).

The SDW-ANC/NR filter (4.39) can then be rewritten as follows:

$$\begin{aligned} \mathbf{W}_\mu(\omega) =& \mu(\omega)\frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\mu(\omega)\mathbf{R}_{X^s}(\omega) + \mathbf{R}_{X^n}(\omega)]^{-1}\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) \\ &- \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\mu(\omega)\mathbf{R}_{X^s}(\omega) + \mathbf{R}_{X^n}(\omega)]^{-1}\mathbf{R}_{X^n}(\omega)\tilde{\mathbf{P}}(\omega) \end{aligned} \tag{4.41}$$

$$\tag{4.42}$$

In a single speech source scenario, the pencil matrix $(\mu(\omega)\mathbf{R}_{X^s}(\omega) + \mathbf{R}_{X^n}(\omega))$ can then be inverted by applying the Woodbury identity and the filter (4.39) can be expressed as follows (see also Appendix C.2):

$$\boxed{\mathbf{W}_\mu(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\frac{\mathbf{R}_{X^n}(\omega)^{-1}\mathbf{R}_{X^s}(\omega)}{\frac{1}{\mu(\omega)} + \rho(\omega)}(\mathbf{G}_{1,\Delta}(\omega) + \tilde{\mathbf{P}}(\omega)) - \tilde{\mathbf{P}}(\omega)]} \tag{4.43}$$

The expression is very similar to the expression for the integrated ANC and NR (3.109) with a scaling factor in the numerator.

4.3.3 Output signal-to-noise ratio when the number of sources is less than or equal to the number of microphones

When the number of sources (speech plus noise sources) is less than or equal to the number of microphones ($Q \leq M$) the leakage signal can be rewritten as a linear combination of the microphone signals.

$$L(\omega) = \mathbf{P}(\omega)^H\mathbf{X}(\omega) \tag{4.44}$$

The filter (4.43) then becomes:

$$\mathbf{W}_{\mu}(\omega) = \frac{\hat{C}(\omega)}{|\hat{C}(\omega)|^2}[\frac{\mathbf{R}_{X^n}(\omega)^{-1}\mathbf{R}_{X^s}(\omega)}{\nu(\omega)+\rho(\omega)}(\mathbf{G}_{1,\Delta}(\omega)+\mathbf{P}(\omega))-\mathbf{P}(\omega)] \tag{4.45}$$

The output SNR of the SDW-ANC/NR scheme at the eardrum can then be expressed as follows (see also Appendix C.3):

$$\boxed{\mathrm{SNR}_{\mu,\mathrm{Int(Q\leq M)}}(\omega) = \frac{\rho(\omega)^2(P_{D_{\mathrm{NR}}}(\omega)+\alpha(\omega)+P_{L^s}(\omega))}{\rho(\omega)(P_{D_{\mathrm{NR}}}(\omega)+\alpha(\omega)+P_{L^s}(\omega))} = \rho(\omega)} \tag{4.46}$$

It is shown in [108, 109], that in a single speech source scenario the weighting factor $\mu(\omega)$ of an SDW-MWF scheme acts as a scaling factor between the different SDW-MWF filters and that the frequency-domain output SNR is therefore independent of this weighting factor $\mu(\omega)$. In the case of the SDW-ANC/NR the weighting factor μ does not merely act as a scaling factor between the different realisations of the SDW-ANC/NR filter (see (4.45)). In the single speech source scenario and when the number of sources (speech source plus noise sources) is less than or equal to the number of microphone, however, the weighting factor $\mu(\omega)$ acts as a scaling factor on the power of the speech signal at the eardrum and the power of the noise signal at the eardrum. Therefore, the SNR at the eardrum is again independent of the weighting factor $\mu(\omega)$ (4.46).

The weighting factor $\mu(\omega)$, however, has an effect on the SD and the residual noise power at the eardrum which can be expressed as follows (see also Appendix C.4):

$$\mathrm{SD}_{\mu,\mathrm{Int(Q\leq M)}}(\omega) = \frac{1}{(1+\rho(\omega)\mu(\omega))^2}(P_{D_{\mathrm{NR}}}(\omega)+\alpha(\omega)+P_{L^s}(\omega)) \tag{4.47}$$

$$P^n_{\mu,\mathrm{Int(Q\leq M)}}(\omega) = \frac{\mu(\omega)^2\rho(\omega)}{(1+\rho(\omega)\mu(\omega))^2}(P_{D_{\mathrm{NR}}}(\omega)+\alpha(\omega)+P_{L^s}(\omega)) \tag{4.48}$$

It appears from the previous equations that when $\mu(\omega) \to 0$ the SDW-ANC/NR scheme behaves as an ANC scheme and the residual noise power at the eardrum tends to 0. When $\mu(\omega) \to \infty$, the SDW-ANC/NR scheme minimizes the SD and so the SD at the eardrum tends to 0.

4.4 Experimental results

The weighted integrated ANC and NR scheme and the SDW-ANC/NR scheme introduced in this chapter have been tested experimentally and their performances have been compared with the performance of the unweighted integrated ANC and NR scheme described in Chapter 3.

The weighted integrated ANC and NR scheme is first considered and then the SDW-ANC/NR scheme is analysed. For both the weighted integrated ANC and NR scheme and the SDW-ANC/NR scheme, the influence of the weighting factor μ on the power of the residual noise at the eardrum and on the SD of the desired signal is first examined. Note that is the case of frequency-domain scheme the weighting factor is chosen to be constant for all frequencies. The impact of μ on the output SNR at the eardrum is then considered.

4.4.1 Experimental setup

The filter lengths are set to $N = 64$, and the NR delay is set to half of the NR filter length ($\Delta = 32$). The secondary path $\mathbf{c}[k]$ is estimated offline using an identification technique based on the NLMS algorithm. The length of the estimated path $\hat{\mathbf{c}}[k]$ is set to $\hat{P} = 32$. The causality margin of the system is set to a positive causality margin ($\delta = 12$).

One noise source is present at 270° (facing the hearing aid). The position of the sources and the SNR for the source signals leads to a so-called leakage SNR (which corresponds to the SNR when the hearing aid is turned off) equal to $-1.3dB$. 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 at the eardrum.

Noise power, speech distortion and signal-to-noise ratio

The weighted integrated ANC and NR scheme and the SDW-ANC/NR scheme have been applied for different gains G and for a trade-off parameter μ varying from 0 to 3. The SD of the desired signal and the residual noise power at the eardrum have been computed to analyse the impact of μ on the performance of each functional block. The output SNR performance is then computed and presented as a function of μ

Performance measures

In order to compare the weighted integrated ANC and NR scheme or the SDW-ANC/NR scheme, with the unweighted integrated ANC and NR scheme, the performance measures defined in Chapter 2 are redefined as follows.

The *noise attenuation improvement* is redefined as

$$\Delta POW = POW_{\text{weight}} - POW_{\text{unweight}} \tag{4.49}$$

where POW_{weight} and POW_{unweight} are the broadband power (in dB) of the noise signal at the eardrum obtained with the weighted integrated ANC and NR scheme or the SDW-ANC/NR scheme, and with the unweighted integrated ANC and NR scheme, respectively.

The *speech distortion improvement* is redefined as

$$\Delta \text{SPD}(\omega) = \text{SPD}_{\text{intellig,weight}}(\omega) - \text{SPD}_{\text{intellig,unweight}}(\omega) \tag{4.50}$$

where $\text{SPD}_{\text{intellig,weight}}$ and $\text{SPD}_{\text{intellig,unweight}}$ represent the output SD (in dB) for the weighted integrated ANC and NR scheme or the SDW-ANC/NR scheme, and for the unweighted integrated ANC and NR scheme, respectively.

The *intelligibility-weighted SNR improvement* [36] is redefined as

$$\Delta SNR_{intellig} = \sum_i I_i (SNR_{i,\text{weight}} - SNR_{i,\text{unweight}}) \tag{4.51}$$

where $\text{SNR}_{\text{i,weight}}$ and $\text{SNR}_{\text{i,unweight}}$ represent the output SNR of the weighted integrated ANC and NR scheme or the SDW-ANC/NR scheme, and the unweighted integrated ANC and NR scheme of the ith band, respectively.

4.4.2 Weighted integrated active noise control and noise reduction

In this subsection, the performance of the weighted integrated ANC and NR scheme introduced in Section 4.2 is analysed and compared to the performance of the unweighted integrated ANC and NR scheme presented in Chapter 3.

Noise power and speech distortion performance

To derive the filter introduced in Section 4.2 the MSE criterion has been explicitly decomposed as the sum of two criteria, and a weight is applied to emphasise one of the functional blocks (ANC or NR). In order to analyse the impact of the weighting factor μ on the NR criteria and on the ANC criterion, the SD of desired signal at the eardrum and the residual noise power at the eardrum have to be computed.

Figures 4.1 and 4.2 present the noise attenuation improvement (4.49) and the SD improvement (4.50), for the weighted integrated ANC and NR scheme compared against the unweighted integrated ANC and NR scheme as a function of μ and for different values of the gain G.

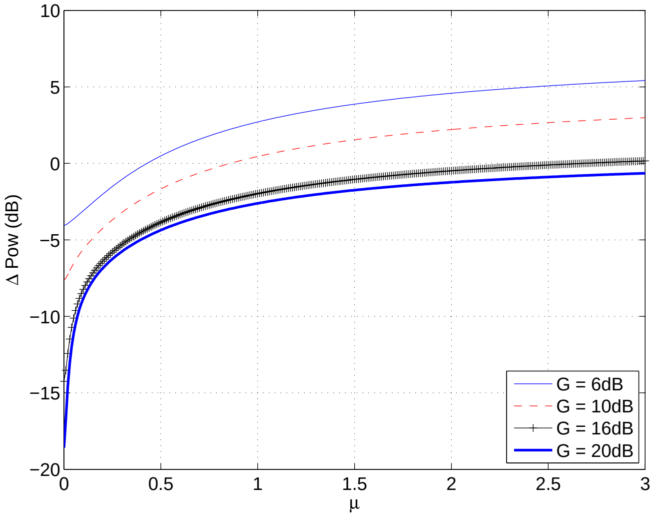


Figure 4.1: Noise attenuation improvement of the weighted integrated ANC and NR scheme compared to the integrated active noise control and noise reduction scheme

When $\mu \rightarrow 0$, the weighted integrated ANC and NR scheme is attenuating the noise at the eardrum more efficiently than the unweighted integrated ANC and NR scheme (Figure 4.1), *i.e.*, it behaves as an ANC algorithm. This also means that the algorithm introduces a lot of SD (Figure 4.2).

When μ increases, the noise attenuation improvement vanishes (Figure 4.1) while the SD decreases (Figure 4.2). When $\mu \rightarrow \infty$, the weighted integrated ANC and NR scheme behaves as a standard NR and some improvement can be done on the SD performance compared against the unweighted integrated ANC and NR scheme.

Signal-to-noise ratio performance

For all values of the gain G, the unweighted integrated ANC and NR scheme provides an SNR improvement of about $10dB$. Figure 4.3 presents the SNR improvement (4.51) of the weighted integrated ANC and NR scheme compared against the SNR performance of the unweighted integrated ANC and NR scheme as a function of μ and for different values of the gain G.

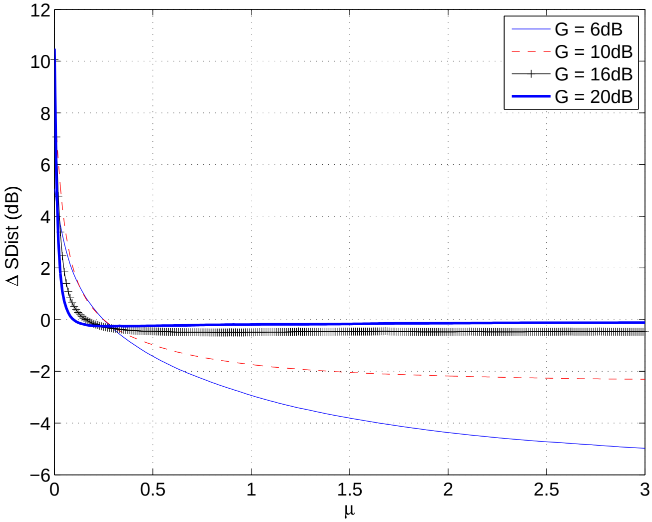


Figure 4.2: Speech distortion improvement of the weighted integrated ANC and NR scheme compared to the integrated active noise control and noise reduction scheme

For low values of μ (up to around 0.5), the weighted integrated ANC and NR scheme provides an SNR improvement that can be $4dB$ higher than the SNR improvement obtained with the unweighted integrated ANC and NR scheme. When μ increases, the weighted scheme behaves more like a standard NR scheme, and then the unweighted integrated ANC and NR scheme exhibits a better SNR performance for gains G up to 20dB.

In this particular scenario, the output SNR exhibits a local maximum for $\mu_0 \approx 0.1$ and the maximum output SNR is higher than the output SNR of the unweighted integrated ANC and NR scheme. In general SNR and SD are the performance measures of interest in the case of hearing aids. Hence, when $0 \leq \mu \leq \mu_0$, decreasing the weighting factor is degrading the SNR performance and the SD performance. Therefore, the weighting factor should then be set to at least μ_0.

Figure 4.4 presents the output SNR performance of the weighted integrated ANC and NR schemes depending on the weighting factor μ and for input SNR varying from 0dB to 15dB. For every input SNR, the output SNR appears to have a maximum value. The optimum weighting factor μ_0 for which this maximum output SNR value is achieved is increasing when the input SNR is decreasing.

- When the microphone signals SNR is low, the output SNR of the MWF-based NR scheme ρ is low and the optimal weighting factor μ_0 is high. The

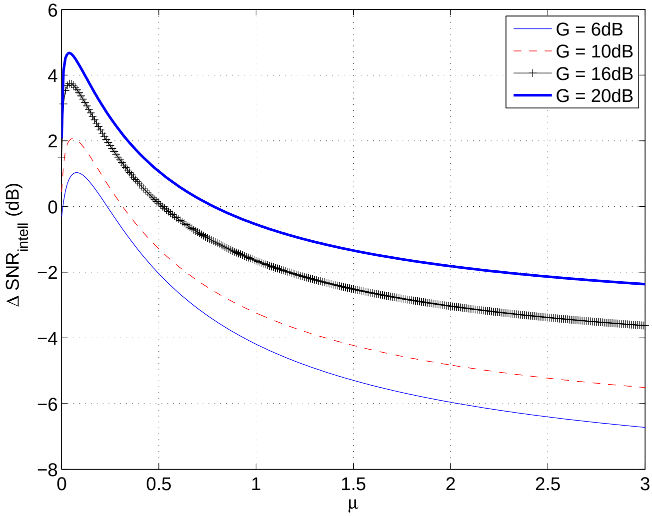


Figure 4.3: Signal-to-noise ratio improvement of the weighted integrated ANC and NR scheme compared to the integrated active noise control and noise reduction scheme

weighted integrated ANC and NR with an optimal weight then behaves as a NR to increase the SNR of the signal to be fed in the hearing aid loudspeaker.

- When the microphones signals SNR is high, there is no need for NR (besides, a MWF-based NR would not improve the SNR much in this case). The optimal weighting factor μ_0 appears then to be close to 0 and the weighted integrated ANC and NR behaves as an ANC, which prevents the ambient noise from reaching the eardrum.

Figure 4.5 presents the output SNR improvement performance for an MWF-based NR, the unweighted integrated ANC and NR scheme and the weighted integrated ANC and NR scheme with two different values for the weighting factor ($\mu = \mu_0$ and $\mu = 3$). The input SNR is set to 0dB.

When $\mu = \mu_0$ the weighted integrated ANC and NR scheme achieves a higher output SNR performance than the unweighted integrated ANC and NR for any gain G. When the weighting factor is increasing, the weighted integrated ANC and NR tends to behave as an FxMWF-based NR scheme that compensates for the secondary path attenuation. The output SNR improvement performance of the weighted integrated ANC and NR scheme for a weighting factor $\mu = 3$ exhibits indeed the same behaviour as an FxMWF-based NR that also compensates for the

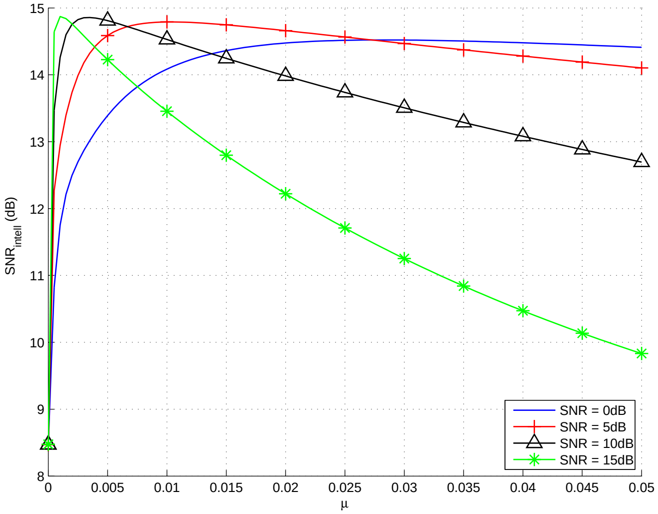


Figure 4.4: SNR performance of the weighted integrated ANC and NR depending on the weighting factor and the input SNR

secondary path (see also Chapter 3) and achieves higher output SNR performance than the standard MWF-based NR.

4.4.3 Speech distortion weighted integrated active noise control and noise reduction

In this subsection, the performance of the SDW-ANC/NR scheme introduced in Section 4.3 is analysed and compared to the performance of the unweighted integrated ANC and NR scheme presented in Chapter 3.

Noise power and speech distortion performance

Figures 4.6 and 4.7 present the noise attenuation improvement (4.49) and the SD improvement (4.50), for the SDW-ANC/NR scheme compared against the unweighted integrated ANC and NR scheme as a function of μ and for different values of the gain G.

When $\mu \rightarrow 0$, the SDW-ANC/NR scheme is attenuating the noise at the eardrum more efficiently than the unweighted integrated ANC and NR scheme (Figure 4.6), *i.e.*, it behaves as an ANC algorithm. This also means that the algorithm introduces a lot of SD (Figure 4.7).

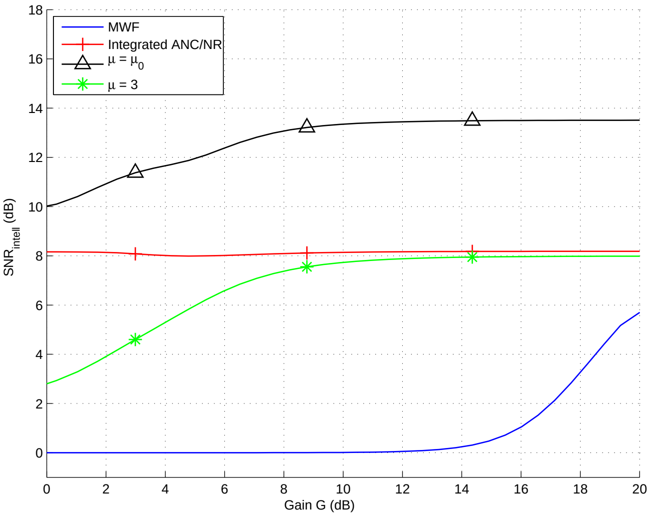


Figure 4.5: Performance comparison between the MWF-based NR, the integrated ANC and NR and the weighted integrated ANC and NR ($\mu = \mu_0$ and $\mu = 3$)

When μ increases, the noise attenuation improvement vanishes (Figure 4.6) while the SD decreases (Figure 4.7). When $\mu \to \infty$, the SDW-ANC/NR scheme minimizes the SD at the eardrum.

Signal-to-noise ratio performance

For all values of the gain G, the unweighted integrated ANC and NR scheme provides an SNR improvement of about $10dB$. Figure 4.8 presents the SNR improvement (4.51) of the SDW-ANC/NR scheme compared against the SNR performance of the unweighted integrated ANC and NR scheme as a function of μ and for different values of the gain G.

For all values of the weighting factor, the SDW-ANC/NR scheme delivers an SNR improvement that is almost constant and equal to the SNR improvement obtained with the unweighted integrated ANC and NR scheme. In terms of the SNR at the eardrum, the SDW-ANC/NR scheme therefore maintains the performance of the unweighted integrated ANC and NR scheme (*i.e.*, the performance of a MWF-based NR when signal leakage and the secondary path are not taken into account) for any weighting factor μ.

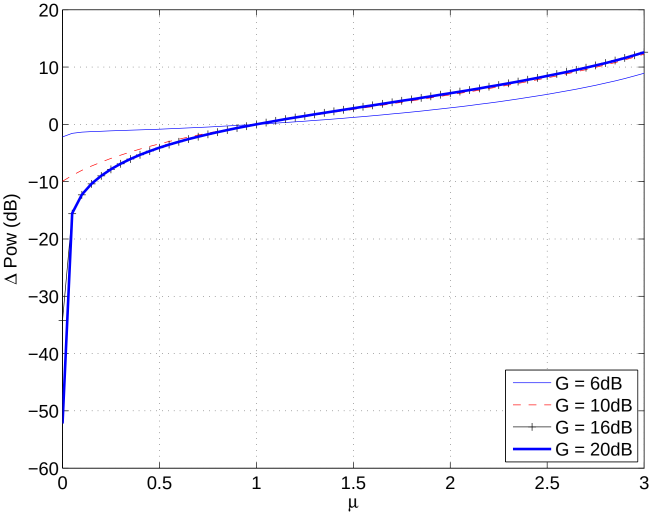


Figure 4.6: Noise attenuation improvement of the SDW-ANC/NR scheme compared to the integrated ANC and NR scheme

4.5 Conclusion

An integrated ANC and NR scheme has been introduced in the previous chapter to tackle the secondary path effects and the effects of signal leakage in the framework of hearing aids with an open fitting. The objectives of the integrated ANC and NR scheme are to attenuate the noise component of the leakage (*i.e.*, ANC) and to minimise the difference between the desired speech signal and the signal delivered at the eardrum (*i.e.*, NR), the balance between these two objectives being fixed.

The concept of weighted NR applied in the MWF framework to derive the SDW-MWF has been extended here to derive weighted versions of the integrated ANC and NR scheme based on FxMWF.

The first weighted integrated ANC and NR scheme introduced in this chapter allows to emphasise either the ANC or the NR. When the signal does not contain any speech, the weighted integrated ANC and NR scheme allows to focus on ANC and minimise the power of the residual noise signal at the eardrum. On the other hand, if the ANC is found to be inefficient for the considered background noise scenario the emphasis can be put on the NR. The weighted integrated ANC and NR scheme then exhibits improved SD performance compared to the unweighted integrated ANC and NR scheme.

This weighted ANC and NR scheme, however, does not reduce to the original

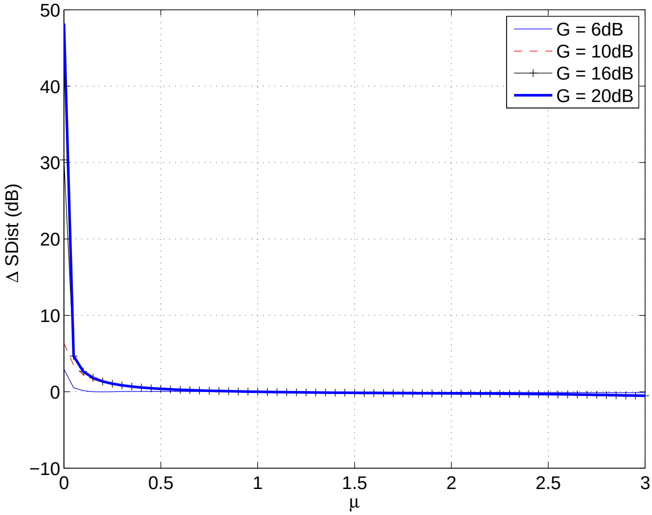


Figure 4.7: Speech distortion improvement of the SDW-ANC/NR scheme compared to the integrated ANC and NR scheme

unweighted integrated ANC and NR scheme for any weighting factor. Besides, focusing on NR allows to reduce the SD compared to unweighted integrated ANC and NR but the NR itself is still introducing SD. A SDW-ANC/NR scheme has been derived that allows to balance between reducing the SD at the eardrum and minimising the residual noise at the eardrum (*i.e.*, ANC). In the single speech source scenario and when the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones, a formula for the output SNR of the SDW-ANC/NR scheme at the eardrum has been derived. The SDW-ANC/NR scheme has then been shown to deliver a constant SNR at the eardrum for any weighting factor.

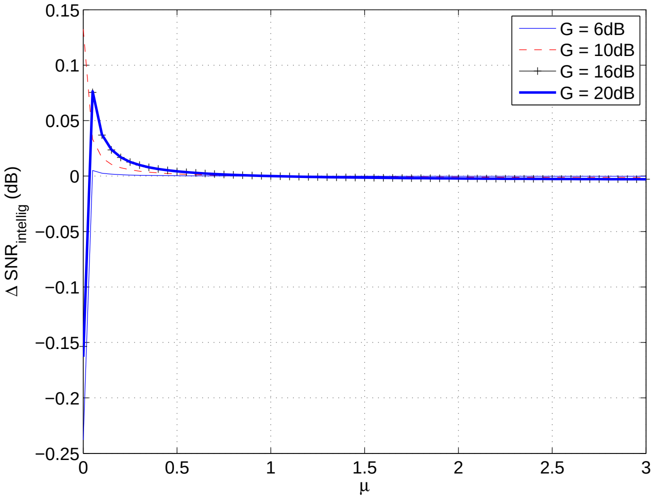


Figure 4.8: Signal-to-noise ratio improvement of the SDW-ANC/NR scheme compared to the integrated ANC and NR scheme

Chapter 5

Zone Of Quiet Based Integrated Active Noise Control and Noise Reduction

This chapter focuses on an integrated approach to active noise control and noise reduction which is based on an optimisation over a zone of quiet generated by the active noise control. A basic integrated active noise control and noise reduction scheme has been introduced in Chapter 3 to tackle secondary path effects and effects of noise leakage through an open fitting. This scheme, however, only takes the sound pressure at the ear canal microphone into account. For an integrated active noise control and noise reduction scheme to be efficient, it is desired to achieve active noise control at the eardrum, which in practice is away from the ear canal microphone. In some cases it can also be desired to achieve noise control over a zone not limited to a single point.

5.1 Introduction

An integrated ANC and NR scheme has been introduced in Chapter 3 to tackle the effects of the secondary path and signal leakage effects. The objectives of this scheme are to attenuate the noise component of the leakage (*i.e.*, ANC) and to minimise the difference between an unknown desired speech signal and the signal delivered at the eardrum (*i.e.*, NR). In practice however, the so-called ear canal microphone, used to construct the error signal in the scheme, cannot be located exactly at the eardrum. Besides, the spatial distribution of sound sources and the geometry of the ear canal do not allow to achieve control over the complete ear canal. Therefore, the actual SNR at the eardrum is basically unknown and uncontrolled.

ANC allows to generate zones of quiet based on destructive interference. The size and shape of these zones of quiet depend on the type of the sound sources and the frequency components of the signal to be cancelled [76][49][90]. It is therefore possible to determine the performance of the scheme described in Chapter 3 at any point of the ear canal, based on the spatial distribution of the sound sources and on the position of the ear canal microphone.

Based on this idea, it is possible to define a particular design criterion, *e.g.*, an MSE criterion, to be minimised at any chosen point in the ear canal and to derive a filter designed to control the noise at this remote point.

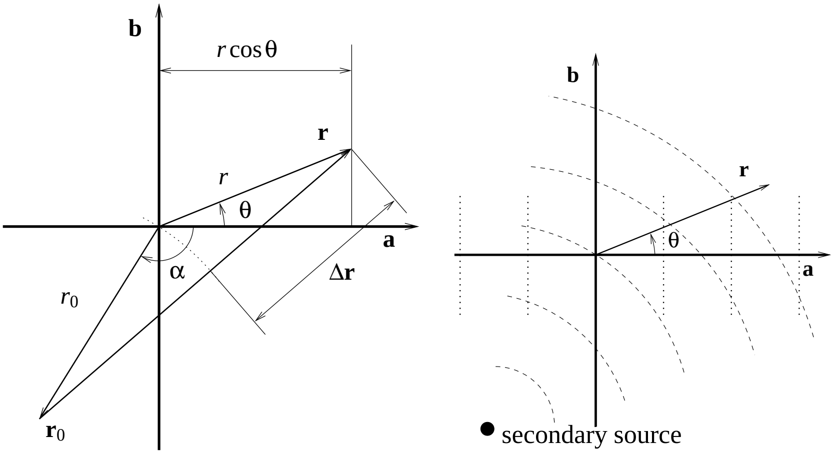
In a similar way, it is also possible to compute an average of the MSE criterion over a desired zone of quiet. In the approach presented here a filter is then derived from the minimisation of this averaged-MSE (aMSE) criterion in order to achieve a control that is more robust on the desired zone of quiet than with a scheme minimising a standard MSE criterion as in Chapter 3.

This chapter also presents a performance comparison between the standard schemes (an ANC scheme and an integrated ANC and NR scheme) and the alternative schemes introduced here. All the schemes presented here are based on FxMWF and applied in hearing aids with an open fitting.

Section 5.2 introduces the approach to derive an ANC scheme and an integrated ANC and NR scheme designed to control the noise at a remote point. The theoretical output SNR of the remote-point integrated ANC and NR is then derived for a single speech source scenario when the number of sound sources is less than or equal to the number of microphones. A zone of quiet based approach to ANC and integrated ANC and NR, to control the noise over a desired zone, is presented in Section 5.3. Experimental results are presented in Section 5.4 and finally Section 5.5 presents a summary of the chapter.

5.2 Integrated active noise control and noise reduction at a remote point

The scheme described in Chapter 3 achieves noise control at the ear canal microphone. Ideally, it should be located at the eardrum. In practice, however, this is not possible, *i.e.*, the controlled point can be located a few tens of millimetres away from the eardrum. The ANC can then only generate a so-called zone of



(a) Secondary source (b) Sound field

Figure 5.1: Sources distribution (a) and sound field (b). The dashed circles represent the sound from the secondary source, the dotted lines represent the sound from the leakage.

quiet around the ear canal microphone and the signal which is actually reaching the eardrum is unknown and uncontrolled.

5.2.1 Remote-point model

To study the performance of the schemes over the zone of quiet, the ear canal microphone is chosen to be the origin of the 2-dimensional coordinate system used to determine each point of the space (presented in Figure 5.1(a)).

It is possible then to estimate the sound pressure at a particular point ($\mathbf{r} = (r\sin\theta\ ,\ r\cos\theta)$) of the ear canal, when the sound pressure at the ear canal microphone and the type of the sound sources generating the sound field in the ear canal are known [49, 76, 90].

External sound sources

The external sound sources (the speech source and the noise sources) are assumed to be far enough from the hearing aid user so that the leakage signal can be viewed as a far field signal and a plane wave model can be used. Assuming also that the ear canal acts as a waveguide and that the plane waves from different directions are uncorrelated [89], the sound field resulting from the leakage can be modelled as one plane wave propagating along the ear canal axis (chosen to be the axis $\mathbf{a}$) as presented in Figure 5.1(b). The sound pressure at point $\mathbf{r}$, resulting from the leakage signal can then be expressed as follows:

$$L(\omega,\mathbf{r}) = L(\omega,\mathbf{0})\ \delta_1(\omega,\mathbf{r}) = L(\omega)\ \delta_1(\omega,\mathbf{r}) \tag{5.1}$$

$$\delta_1(\omega,\mathbf{r}) \triangleq \mathrm{e}^{-j\omega\frac{r\cos\theta}{c_0}} \tag{5.2}$$

where c_0 is the speed of sound in air and $L(\omega,\mathbf{0})$ is the sound pressure resulting from the leakage signal at the ear canal microphone.

Hearing aid loudspeaker

The hearing aid loudspeaker (secondary source), is located at:

$$\mathbf{r}_0 = (r_0\sin\alpha\ ,\ r_0\cos\alpha) \tag{5.3}$$

The ear canal is assumed to be in the near field of the hearing aid loudspeaker. A simple model to study the sound field generated by the hearing aid loudspeaker is then the monopole-point-secondary-source approach [49]. The use of this model can be justified by the small dimension of the hearing aid loudspeaker. The sound field generated by the hearing aid loudspeaker is then assumed to be circular with exponentially decaying amplitude:

$$Z_{\mathrm{HA}}(\omega,\mathbf{r}) = Z_{\mathrm{HA}}(\omega,\mathbf{0})\delta_2(\omega,\mathbf{r}) \tag{5.4}$$

$$\delta_2(\omega,\mathbf{r}) \triangleq \frac{r_0}{|\mathbf{r}-\mathbf{r}_0|}\mathrm{e}^{-j\omega\frac{\Delta\mathbf{r}}{c_0}} \tag{5.5}$$

where $Z_{\mathrm{HA}}(\omega,\mathbf{0})$ is the sound pressure resulting from the hearing aid signal $Z(\omega)$ at the ear canal microphone.

$$Z_{\mathrm{HA}}(\omega,\mathbf{0}) = C(\omega)Z(\omega) = Z_{\mathrm{HA}}(\omega) \tag{5.6}$$

Note that the subsequent schemes can be adapted if these acoustic models are replaced by other models. The methodology applied stays valid as long as it is possible to find a mathematical model to describe the acoustic propagation from the sources (speech source, noise sources and secondary source) to the eardrum

Performance measures at a remote point

Based on the previous models, it is possible to determine the sound field resulting from the external sources and the hearing aid loudspeaker at any point of the ear canal. Using Parseval theorem, the power of the total sound pressure signal and the SNR at the point $\mathbf{r}$ can be expressed as follows:

$$\mathrm{Pow}(\omega, \mathbf{r}) = \frac{1}{2\pi}\int_{-\infty}^{\infty} |\delta_1(\omega, \mathbf{r})L(\omega) + \delta_2(\omega, \mathbf{r})Z_{\mathrm{HA}}(\omega)|^2 \mathrm{d}\omega \tag{5.7}$$

$$\mathrm{SNR}(\omega, \mathbf{r}) = \frac{\mathrm{Pow}^s(\omega, \mathbf{r})}{\mathrm{Pow}^n(\omega, \mathbf{r})} \tag{5.8}$$

It appears that interferences that are destructive at the ear canal microphone (thus allowing ANC) might become constructive at a distant point. The noise component of the leakage would then be amplified at the eardrum instead of being attenuated.

5.2.2 Active noise control

When the noise is to be controlled at a point $\mathbf{r}$ distant from the ear canal microphone and when the acoustic propagation from the sources to the distant point can be modelled, the MSE criterion to be minimised (2.90) can be rewritten at point $\mathbf{r}$ based on equations (5.4) and (5.1):

$$J_{\mathrm{ANC}}(\omega, \mathbf{r}) = \mathbb{E}\{|E_{\mathrm{ANC}}(\omega, \mathbf{r})|^2\} \tag{5.9}$$

$$E_{\mathrm{ANC}}(\omega, \mathbf{r}) = \delta_2(\omega, \mathbf{r})C(\omega)\mathbf{W}(\omega)^H\mathbf{X}(\omega) + \delta_1(\omega, \mathbf{r})L(\omega) \tag{5.10}$$

Assuming that the secondary path identification error is small ($\hat{C}(\omega) \approx C(\omega)$) and that the filter $\mathbf{W}$ is adapting slowly, the MSE criterion (5.9) can be written as follows:

$$J_{\mathrm{ANC}}(\omega, \mathbf{r}) \approx \mathbb{E}\{|\delta_2(\omega, \mathbf{r})\mathbf{W}(\omega)^H\mathbf{Y}(\omega) + \delta_1(\omega, \mathbf{r})L(\omega)|^2\} \tag{5.11}$$

The optimal filter (FxMWF) minimising (5.9) is then:

$$\mathbf{W}_{\mathrm{ANC}}(\omega, \mathbf{r}) = -\frac{\delta_2(\omega, \mathbf{r})\delta_1(\omega, \mathbf{r})^*}{|\delta_2(\omega, \mathbf{r})|^2}\mathbf{R}_Y(\omega)^{-1}\mathbf{r}_{YL}(\omega) \tag{5.12}$$

Here $\mathbf{R}_Y(\omega)$ is the correlation matrix of the filtered microphone signals $\mathbf{Y}(\omega)$, as defined in (2.102), while $\mathbf{r}_{YL}(\omega)$ is the cross-correlation vectors between the filtered microphone signals $\mathbf{Y}(\omega)$ and the leakage signal $L(\omega)$, as defined in (2.103).

Assuming once again that the filter $\mathbf{W}(\omega)$ is adapting slowly, the MSE criterion (5.11) can be written as follows:

$$J_{\mathrm{ANC}}(\omega, \mathbf{r}) \approx \mathbb{E}\{|\mathbf{W}(\omega)^H\mathbf{Y}(\omega, \mathbf{r}) + L(\omega, \mathbf{r})|^2\} \tag{5.13}$$

The filter (5.12) then reduces to:

$$\boxed{\mathbf{W}_{\mathrm{ANC}}(\omega,\mathbf{r}) = -\mathbf{R}_Y(\omega,\mathbf{r})^{-1}\mathbf{r}_{YL}(\omega,\mathbf{r})} \tag{5.14}$$

Here $\mathbf{R}_Y(\omega,\mathbf{r})$ is the correlation matrix of $\mathbf{Y}(\omega,\mathbf{r})$, *i.e.*, the filtered microphone signals $\mathbf{Y}(\omega)$ estimated at the remote point (RP) $\mathbf{r}$ and $\mathbf{r}_{YL}(\omega,\mathbf{r})$ is the cross-correlation vector between $\mathbf{Y}(\omega,\mathbf{r})$ and $L(\omega,\mathbf{r})$, *i.e.*, the leakage signal $L(\omega)$ at the RP $\mathbf{r}$.

$$\begin{aligned} \mathbf{R}_Y(\omega,\mathbf{r}) &= \mathbb{E}\{\mathbf{Y}(\omega,\mathbf{r})\mathbf{Y}(\omega,\mathbf{r})^H\} && (5.15) \\ \mathbf{r}_{YL}(\omega,\mathbf{r}) &= \mathbb{E}\{\mathbf{Y}(\omega,\mathbf{r})L(\omega,\mathbf{r})^*\} && (5.16) \end{aligned}$$

Filter (5.14) is then an FxMWF-based ANC filter applied to the filtered microphone signals Y to attenuate the leakage signal L both, filtered by the corresponding acoustic models $\delta_2(\omega,\mathbf{r})$ and $\delta_1(\omega,\mathbf{r})$, respectively. It can then be used to compensate for the signal leakage at a RP $\mathbf{r}$.

5.2.3 Integrated active noise control and noise reduction

In a similar way, a filter $\mathbf{W}_{\mathrm{Int}}(\omega,\mathbf{r})$ may be defined corresponding to an RP $\mathbf{r}$. This filter then minimises the MSE criterion (3.74) rewritten at point $\mathbf{r}$, based on (5.1) and (5.4):

$$\begin{aligned} J_{\mathrm{Int}}(\omega,\mathbf{r}) &= \mathbb{E}\{|E_{\mathrm{Int}}(\omega,\mathbf{r})|^2\} && (5.17) \\ E_{\mathrm{Int}}(\omega) &= \delta_2(\omega,\mathbf{r})C(\omega)\mathbf{W}(\omega)^H\mathbf{X}(\omega) + \delta_1(\omega,\mathbf{r})L^n(\omega) - D_{NR}(\omega) && (5.18) \end{aligned}$$

Assuming that the secondary path identification error is small ($\hat{C}(\omega) \approx C(\omega)$) and that the filter $\mathbf{W}(\omega)$ is adapting slowly, the MSE criterion (5.17) can be written as follows:

$$J_{\mathrm{Int}}(\omega,\mathbf{r}) \approx \mathbb{E}\{|\delta_2(\omega,\mathbf{r})\mathbf{W}(\omega)^H\mathbf{Y}(\omega) + \delta_1(\omega,\mathbf{r})L^n(\omega) - D_{NR}(\omega)|^2\} \tag{5.19}$$

If the speech and noise signals are uncorrelated, the optimal filter (FxMWF) minimising (5.19) is then:

$$\boxed{\begin{aligned} \mathbf{W}_{\mathrm{Int}}(\omega,\mathbf{r}) &= \frac{\mathbf{R}_Y(\omega)^{-1}(\delta_2(\omega,\mathbf{r})\mathbf{r}_{YD_{\mathrm{NR}}}(\omega) - \delta_2(\omega,\mathbf{r})\delta_1(\omega,\mathbf{r})^*\mathbf{r}_{YL^n}(\omega))}{|\delta_2(\omega,\mathbf{r})|^2} \\ \mathbf{W}_{\mathrm{Int}}(\omega,\mathbf{r}) &= \frac{\mathbf{R}_Y(\omega)^{-1}(\delta_2(\omega,\mathbf{r})\mathbf{r}_{Y^sD_{\mathrm{NR}}}(\omega) - \delta_2(\omega,\mathbf{r})\delta_1(\omega,\mathbf{r})^*\mathbf{r}_{Y^nL^n}(\omega))}{|\delta_2(\omega,\mathbf{r})|^2} \end{aligned}} \tag{5.20}$$

Filter (5.20) then allows to deliver a desired speech signal and compensate for the noise component of the leakage signal at a point $\mathbf{r}$ which is away from the ear canal microphone.

Alternatively, (5.19) can be written as:

$$J_{\text{Int}}(\omega, \mathbf{r}) \approx \mathbb{E}\{|\mathbf{W}(\omega)^H \mathbf{Y}(\omega, \mathbf{r}) + L^n(\omega, \mathbf{r}) - D_{NR}(\omega)|^2\} \tag{5.21}$$

and then filter (5.12) reduces to:

$$\mathbf{W}_{\text{Int}}(\omega, \mathbf{r}) = \mathbf{R}_Y(\omega, \mathbf{r})^{-1} \mathbf{r}_{YD_{\text{Int}}}(\omega, \mathbf{r}) \tag{5.22}$$

Here $\mathbf{r}_{YD_{\text{Int}}}(\omega, \mathbf{r})$ is the cross-correlation vector between the filtered microphone signals $\mathbf{Y}(\omega)$ and the desired signal $D_{\text{Int}}(\omega)$ at the RP $\mathbf{r}$.

$$\mathbf{r}_{YD_{\text{Int}}}(\omega, \mathbf{r}) = \mathbb{E}\{\mathbf{Y}(\omega, \mathbf{r}) D_{\text{Int}}(\omega, \mathbf{r})^*\} \tag{5.23}$$

If speech and noise are uncorrelated $\mathbf{r}_{YD_{\text{Int}}}(\omega, \mathbf{r})$ can be expressed as follows:

$$\begin{aligned} \mathbf{r}_{YD_{\text{Int}}}(\omega, \mathbf{r}) &= \mathbf{r}_{Y^s D_{NR}}(\omega, \mathbf{r}) - \mathbf{r}_{Y^n L^n}(\omega, \mathbf{r}) \\ &= \mathbb{E}\{\mathbf{Y}^s(\omega, \mathbf{r}) D_{NR}(\omega, \mathbf{r})^H\} - \mathbb{E}\{\mathbf{Y}^n(\omega, \mathbf{r}) L^n(\omega, \mathbf{r})^*\} \end{aligned} \tag{5.24}$$

The filter (5.22) can then be rewritten as:

$$\boxed{\mathbf{W}_{\text{Int}}(\omega, \mathbf{r}) = \mathbf{R}_Y(\omega, \mathbf{r})^{-1} (\mathbf{r}_{Y^s D_{NR}}(\omega, \mathbf{r}) - \mathbf{r}_{Y^n L^n}(\omega, \mathbf{r}))} \tag{5.25}$$

Filter (5.25) provides an integrated ANC and NR applied to the filtered microphone signals $\mathbf{Y}(\omega)$ and the leakage signal $L(\omega)$ both estimated at the RP $\mathbf{r}$.

Single speech source scenario

It has been explained in Chapter 3 that the leakage signal can be approximated by a linear combination of the input signals

$$L(\omega) = \tilde{\mathbf{P}}(\omega)^H \mathbf{X}(\omega) + e_L(\omega) \tag{5.26}$$

The leakage signal at point $\mathbf{r}$ can similarly be approximated by a linear combination of the inputs:

$$\begin{aligned} L(\omega, \mathbf{r}) =& \delta_1(\omega, \mathbf{r}) L(\omega) \\ =& \tilde{\mathbf{P}}^H(\omega, \mathbf{r}) \mathbf{X}(\omega) + e_L(\omega, \mathbf{r}) \end{aligned} \tag{5.27}$$

where

$$\tilde{\mathbf{P}}(\omega,\mathbf{r}) = \delta_1(\omega,\mathbf{r})\tilde{\mathbf{P}}(\omega) \tag{5.28}$$

$$e_L(\omega,\mathbf{r}) = \delta_1(\omega,\mathbf{r})e_L(\omega) \tag{5.29}$$

The estimation error $e_L(\omega,\mathbf{r})$ is then orthogonal to the microphone signals (3.105) and to the microphone signals filtered by $\tilde{\mathbf{P}}(\omega,\mathbf{r})$ (3.106). If the filter $\mathbf{W}(\omega,\mathbf{r})$ varies slowly $e_L(\omega,\mathbf{r})$ is also orthogonal to the microphone signals filtered by $\mathbf{W}(\omega,\mathbf{r})$ (3.107).

The integrated ANC and NR scheme at RP $\mathbf{r}$ (5.22) can then be rewritten as:

$$\begin{aligned}\mathbf{W}_{\mathrm{Int}}(\omega,\mathbf{r}) =& \frac{\hat{C}(\omega)}{|\delta_2(\omega,\mathbf{r})\hat{C}(\omega)|^2}\mathbf{R}_X(\omega)^{-1}\delta_2(\omega,\mathbf{r})\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega)\\ &- \frac{\hat{C}(\omega)}{|\delta_2(\omega,\mathbf{r})\hat{C}(\omega)|^2}\mathbf{R}_X(\omega)^{-1}\delta_2(\omega,\mathbf{r})\mathbf{R}_{x^n}(\omega)\tilde{\mathbf{P}}(\omega,\mathbf{r})\end{aligned} \tag{5.30}$$

Let the secondary path to the RP and its estimate be defined as follows:

$$C(\omega,\mathbf{r}) \triangleq C(\omega)\delta_2(\omega,\mathbf{r}) \tag{5.31}$$

$$\hat{C}(\omega,\mathbf{r}) \triangleq \hat{C}(\omega)\delta_2(\omega,\mathbf{r}) \tag{5.32}$$

The filter (5.30) then reduces to:

$$\mathbf{W}_{\mathrm{Int}}(\omega,\mathbf{r}) = \frac{\hat{C}(\omega,\mathbf{r})}{|\hat{C}(\omega,\mathbf{r})|^2}\mathbf{R}_X(\omega)^{-1}(\mathbf{R}_{X^s}(\omega)\mathbf{G}_{1,\Delta}(\omega) - \mathbf{R}_{x^n}(\omega)\tilde{\mathbf{P}}(\omega,\mathbf{r})^H) \tag{5.33}$$

In the single speech source case, the autocorrelation matrix of the microphone signals is rank-1 and applying the Woodbury identity to invert $\mathbf{R}_X(\omega)$ leads to:

$$\boxed{\mathbf{W}_{\mathrm{Int}}(\omega,\mathbf{r}) = \frac{\hat{C}(\omega,\mathbf{r})}{|\hat{C}(\omega,\mathbf{r})|^2}[\frac{\mathbf{R}_X(\omega)^{-1}\mathbf{R}_{X^s}(\omega)}{1+\rho(\omega)}(\mathbf{G}_{1,\Delta}(\omega) + \tilde{\mathbf{P}}(\omega,\mathbf{r})) - \tilde{\mathbf{P}}(\omega,\mathbf{r})]} \tag{5.34}$$

The output SNR of the RP-based integrated ANC and NR scheme estimated at point $(\omega, \mathbf{r})$ is given by

$$SNR(\omega, \mathbf{r}) = \frac{\mathbb{E}\{|C(\omega, \mathbf{r})\mathbf{W}(\omega)^H\mathbf{X}^s(\omega) + \mathbf{X}^s(\omega)\tilde{\mathbf{P}}(\omega, \mathbf{r})^H|^2\}}{\mathbb{E}\{|C(\omega, \mathbf{r})\mathbf{W}(\omega)^H\mathbf{X}^n(\omega) + \mathbf{X}^n(\omega)\tilde{\mathbf{P}}(\omega, \mathbf{r})^H|^2|^2\}} \tag{5.35}$$

Depending on the number of sound sources (speech source plus noise sources) the output SNR is then:

$$\boxed{\begin{aligned} SNR_{\text{Int}(Q\leq M)} &= \rho(\omega) = SNR_{NR(noLeakage)}(\omega) \\ SNR_{\text{Int}(Q>M)} &= \frac{\frac{\rho(\omega)^2}{(\rho(\omega)+1)^2}(P_{D_{NR}}(\omega) + \beta(\omega) + \tilde{\mathbf{P}}(\omega)^H R_s(\omega)\tilde{\mathbf{P}}(\omega)) + E_{e^s_L}(\omega)}{\frac{\rho(\omega)}{(\rho(\omega)+1)^2}(P_{D_{NR}}(\omega) + \beta(\omega) + \tilde{\mathbf{P}}(\omega)^H R_s(\omega)\tilde{\mathbf{P}}(\omega)) + E_{e^n_L}(\omega)} \end{aligned}} \tag{5.36}$$

The RP-based integrated ANC and NR then allows to restore the performance of the integrated ANC and NR at point $\mathbf{r}$.

5.3 Integrated active noise control and noise reduction over a zone of quiet

The ultimate goal is to control the signal at the eardrum. Therefore, it is more relevant to design a filter adapted to a desired zone of quiet rather than a single RP. This section introduces the ZQ model and the ANC scheme and the integrated ANC and NR scheme derived from this model.

5.3.1 Zone of quiet model

When the SNR is to be controlled over a desired zone of quiet, the SNR computed at one spatial point does not appear to be a satisfying measure. The average signal power and the average-SNR (aSNR) over a desired ZQ S are defined as:

$$\text{aPow}(\omega, S) \triangleq \frac{1}{\bar{S}}\int_S \text{Pow}(\omega, \mathbf{r})\mathrm{d}S \tag{5.37}$$

$$\text{aSNR}(\omega, S) \triangleq \frac{1}{\bar{S}}\int_S \text{SNR}(\omega, \mathbf{r})\mathrm{d}S \tag{5.38}$$

where $\bar{S}$ is the area of S.

The filters introduced in the previous sections minimise MSE criteria at a particular point (the ear canal microphone in Chapter 3 or an RP in Section 5.2) and may exhibit degraded performance in terms of average signal power or average SNR over a desired ZQ (see also Section 5.4).

In order to derive new filters which are efficient on a desired ZQ, the aMSE criterion over a desired ZQ has to be defined:

$$aJ_{\text{MSE}}(\omega, S) = \frac{1}{\bar{S}} \int_S J_{\text{MSE}}(\omega, \mathbf{r}) \mathrm{d}S \tag{5.39}$$

5.3.2 Active noise control

When the noise is to be controlled over a desired ZQ S away from the ear canal microphone and when the acoustic propagation from the sources to an RP can be modelled, the aMSE criterion to be minimised can be written as:

$$\begin{aligned} aJ_{\text{ANC}}(\omega, S) &= \frac{1}{\bar{S}} \int_S \mathbb{E}\{|E_{\text{ANC}}(\omega, \mathbf{r})|^2\} \mathrm{d}S & (5.40) \\ E_{\text{ANC}}(\omega, \mathbf{r}) &= C(\omega)\mathbf{W}(\omega)^H \mathbf{X}(\omega, \mathbf{r}) + L(\omega, \mathbf{r}) & (5.41) \end{aligned}$$

Assuming that the secondary path identification error is small ($\hat{C}(\omega) \approx C(\omega)$) and that the filter $\mathbf{W}(\omega)$ is adapting slowly, the aMSE criterion (5.40) can be written as follows:

$$aJ_{\text{ANC}}(\omega, S) \approx \frac{1}{\bar{S}} \int_S \mathbb{E}\{|\delta_2(\omega, \mathbf{r})\mathbf{W}(\omega)^H \mathbf{Y}(\omega) + \delta_1(\omega, \mathbf{r})L|^2\} \mathrm{d}S \tag{5.42}$$

Integration being a linear operator, the aMSE criterion can be rewritten as:

$$\begin{aligned} aJ_{\text{ANC}}(\omega, S) \approx \; & \frac{\mathbf{W}(\omega)^H \mathbf{R}_Y(\omega) \mathbf{W}(\omega)}{\bar{S}} \int_S |\delta_2(\omega, \mathbf{r})|^2 \mathrm{d}S \\ & + 2\Re[\frac{\mathbf{W}(\omega)^H \mathbf{r}_{YL}(\omega)}{\bar{S}} \int_S \delta_2(\omega, \mathbf{r}) \delta_1(\omega, \mathbf{r})^* \mathrm{d}S] \\ & + \frac{\mathbf{R}_L(\omega)}{\bar{S}} \int_S |\delta_1(\omega, \mathbf{r})|^2 \mathrm{d}S \end{aligned} \tag{5.43}$$

where $\mathbf{R}_L(\omega)$ is the auto-correlation of the leakage signal $L(\omega)$:

$$\mathbf{R}_L(\omega) = \mathbb{E}\{|L(\omega)|^2\} \tag{5.44}$$

The FxMWF minimising (5.40) is then:

$$\boxed{\mathbf{W}_{\mathrm{ANC}}(\omega, S) = -\frac{\xi_{2,1}(\omega, S)}{\xi_{2,2}(\omega, S)}\mathbf{R}_{Y}(\omega)^{-1}\mathbf{r}_{YL}(\omega)} \tag{5.45}$$

with

$$\xi_{2,2}(\omega, S) \triangleq \frac{1}{S}\int_{S} |\delta_2(\omega, \mathbf{r})|^2 \mathrm{d}S \tag{5.46}$$

$$\xi_{2,1}(\omega, S) \triangleq \frac{1}{S}\int_{S} \delta_2(\omega, \mathbf{r})\delta_1(\omega, \mathbf{r})^* \mathrm{d}S \tag{5.47}$$

5.3.3 Integrated active noise control and noise reduction

In a similar way, a filter $\mathbf{W}_{\mathrm{Int}}(\omega, S)$ may be defined corresponding to a desired ZQ S. This filter then minimises the aMSE criterion:

$$aJ_{\mathrm{Int}}(\omega, S) = \frac{1}{\bar{S}}\int_{S} \mathbb{E}\{|E_{\mathrm{Int}}(\omega, \mathbf{r})|^2\}\mathrm{d}S \tag{5.48}$$

$$E_{\mathrm{Int}}(\omega, \mathbf{r}) = C(\omega)\mathbf{W}(\omega)^H\mathbf{X}(\omega, \mathbf{r}) + L^n(\omega, \mathbf{r}) - D_{\mathrm{NR}}(\omega) \tag{5.49}$$

Assuming that the secondary path identification error is small ($\hat{C}(\omega) \approx C(\omega)$), that the filter $\mathbf{W}(\omega)$ is adapting slowly and that the noise and speech signals are uncorrelated, the aMSE criterion (5.48) can be written as:

$$aJ_{\mathrm{Int}}(\omega, S) \approx \frac{1}{S}\int_{S} \mathbb{E}\{|\delta_2(\omega, \mathbf{r})\mathbf{W}(\omega)^H\mathbf{Y}^s(\omega) - D_{\mathrm{NR}}(\omega)|^2\}\mathrm{d}S$$

$$+\frac{1}{S}\int_{S} \mathbb{E}\{|\delta_2(\omega, \mathbf{r})\mathbf{W}(\omega)^H\mathbf{Y}^n(\omega) + \delta_1(\omega, \mathbf{r})L^n(\omega)|^2\}\mathrm{d}S \tag{5.50}$$

Integration being a linear operator, the aMSE can be rewritten as:

$$\begin{aligned} aJ_{\text{Int}}(\omega,S) \quad &\approx \quad \frac{\mathbf{W}(\omega)^H\mathbf{R}_Y(\omega)\mathbf{W}(\omega)}{\bar{S}}\int_S|\delta_2(\omega,\mathbf{r})|^2\mathrm{d}S \\ &+ 2\Re[\frac{\mathbf{W}(\omega)^H\mathbf{r}_{YL^n}(\omega)}{\bar{S}}\int_S\delta_2(\omega,\mathbf{r})\delta_1(\omega,\mathbf{r})^*\mathrm{d}S] \\ &- 2\Re[\frac{\mathbf{W}(\omega)^H\mathbf{r}_{YD_{\text{NR}}}(\omega)}{\bar{S}}\int_S\delta_2(\omega,\mathbf{r})\mathrm{d}S] \\ &+ \frac{\mathbf{R}_{L^n}(\omega)}{\bar{S}}\int_S|\delta_1(\omega,\mathbf{r})|^2\mathrm{d}S + |G(\omega)|^2\mathbf{R}_Y(\omega) \end{aligned} \tag{5.51}$$

Here $\mathbf{R}_{L^n}(\omega)$ is the auto-correlation of the noise component of the leakage signal $L^n(\omega)$:

$$\mathbf{R}_{L^n}(\omega) = \mathbb{E}\{|L^n(\omega)|^2\} \tag{5.52}$$

The FxMWF minimising (5.39) is then:

$$\boxed{\mathbf{W}_{\text{Int}}(\omega,S) = \mathbf{R}_{ZQ}(\omega,S)^{-1}\mathbf{r}_{ZQ}(\omega,S)} \tag{5.53}$$

where

$$\mathbf{R}_{ZQ}(\omega,S) \quad = \quad \xi_{2,2}(\omega,S)\mathbf{R}_Y(\omega) \tag{5.54}$$

$$\mathbf{r}_{ZQ}(\omega,S) \quad = \quad \xi_2(\omega,S)\mathbf{r}_{Y^sD_{\text{NR}}}(\omega) - \xi_{2,1}(\omega,S)\mathbf{r}_{Y^nL^n}(\omega) \tag{5.55}$$

where $\xi_{2,2}(\omega,S)$ and $\xi_{2,1}(\omega,S)$ are defined in (5.46) and (5.47), respectively and $\xi_2(\omega,S)$ is defined as:

$$\xi_2(\omega,S) \triangleq \frac{1}{S}\int_S\delta_2(\omega,\mathbf{r})\mathrm{d}S \tag{5.56}$$

5.4 Experimental results

The schemes introduced in Section 5.2 and Section 5.3 have been tested experimentally for both ANC and integrated ANC and NR and their performance have been compared with the performance of a standard ANC scheme and the original integrated ANC and NR scheme described in Chapter 3, respectively.

5.4.1 Experimental setup

In these experiments, the ear canal microphone was considered to be located 15mm away from the eardrum. The hearing aid loudspeaker is located 20mm away from ear canal microphone and the angle α is set to $\frac{5\pi}{6}$ (as presented on Figure 5.2).

The filter length is set to $N = 128$, and, in the case of the integrated ANC and NR schemes, the NR delay is set to half of the filter length ($\Delta = 64$). The secondary path $C(z)$ is estimated off-line using an identification technique based on the NLMS algorithm. The length of the estimated path $\hat{C}(z)$ is set to $L = 32$. Frequency-domain filters are considered in the chapter therefore the causality margin of the system is (artificially) set to a positive value ($\delta = 12$). In the case of integrated ANC and NR schemes, the gain is set to $G = 5$dB.

The RP-based schemes (ANC and integrated ANC and NR) are designed to optimise the MSE criterion at the central point of the eardrum (point $\mathbf{r}_{TM}$ in Figure 5.2).

In the case of ZQ-based schemes (ANC and integrated ANC and NR schemes), two different zones are considered. The first zone (zone 1 in Figure 5.2) is the portion of the ear canal between the ear canal microphone and the eardrum (considered to be a 15mm by 10mm rectangular zone). The second zone (zone 2 in Figure 5.2) is the close neighbourhood of the eardrum (considered to be a 5mm by 10mm rectangular zone).

Active noise control

The performance measure used for the ANC schemes is the residual noise power at the eardrum. The residual noise power at the eardrum is estimated from time-domain signals that reach the eardrum. The reference power is the power of the noise component of the leakage signal, which can also be considered as the residual noise power when the hearing aid is turned off. Note that it can be shown from (5.1) that, as far as the plane wave model is concerned, the power of the noise component of the leakage signal is stationary in space, *i.e.*, in the ear canal. The performance measure is then written as:

$$\Delta\mathrm{Pow}^{n}(\mathbf{r}) = \mathrm{Pow}^{n}_{\mathrm{eardrum}}(\mathbf{r}) - \mathrm{Pow}^{n}_{\mathrm{leak}} \tag{5.57}$$

for the RP schemes, and:

$$a\Delta\mathrm{Pow}^{n}(S) = \frac{1}{S}\int_{S} \Delta\mathrm{Pow}^{n}(\mathbf{r})\mathrm{d}S \tag{5.58}$$

for the ZQ-based schemes.

Here $\mathrm{Pow}^{n}_{\mathrm{eardrum}}(\mathbf{r})$ and $\mathrm{Pow}^{n}_{\mathrm{leak}}$ are the power (in dB) of the residual noise at point $\mathbf{r}$ for a particular processing scheme and the power of the noise component of the leakage signal, respectively. They are computed from full-band time-domain signals.

$$\mathrm{Pow}^{n}_{\mathrm{res}}(\mathbf{r}) = \mathbb{E}\{|z_{\mathrm{HA}}[k,\mathbf{r}] + l^{n}[k,\mathbf{r}]|^{2}\} \tag{5.59}$$

$$\mathrm{Pow}^{n}_{\mathrm{leak}} = \mathbb{E}\{|l^{n}[k]|^{2}\} \tag{5.60}$$

Integrated ANC and NR scheme

The performance measure used for integrated ANC and NR schemes is the intelligibility-weighted SNR. The intelligibility-weighted SNR is estimated from time-domain signals filtered by the schemes described in this chapter. The leakage SNR, which can also be considered as the SNR when the hearing aid is turned off, is used as a reference measure. Note that it can be shown from (5.1) that the SNR of the leakage signal is stationary in space, *i.e.*, in the ear canal.

The intelligibility-weighted SNR improvement [36] is defined as:

$$\Delta\mathrm{SNR}_{\mathrm{intellig}}(\mathbf{r}) = \sum_{i} I_i(\mathrm{SNR}_{i,\mathrm{out}}(\mathbf{r}) - \mathrm{SNR}_{i,\mathrm{leak}}) \tag{5.61}$$

for the RP schemes, and:

$$a\Delta\mathrm{SNR}_{\mathrm{intellig}}(S) = \frac{1}{S}\int_{S}\Delta\mathrm{SNR}_{\mathrm{intellig}}(\mathbf{r})\mathrm{d}S \tag{5.62}$$

for the ZQ-based schemes.

Here I_i is the band importance function and $\mathrm{SNR}_{i,\mathrm{out}}(\omega,\mathbf{r})$ and $\mathrm{SNR}_{i,\mathrm{leak}}$ represent the output SNR at point $\mathbf{r}$ for a particular processing scheme and the leakage SNR (in dB), respectively, of the ith one third octave band.

5.4.2 Active noise control

The residual noise power has been computed in the ear canal and the performance is presented for a standard ANC scheme (Figure 5.3(a)), an RP-based ANC scheme designed to minimise the noise at point $\mathbf{r}_{TM}$ (Figure 5.3(b)), a ZQ-based ANC scheme designed to minimise the noise power over zone 1 (Figure 5.3(c)) and a ZQ-based ANC scheme defined to minimise the noise power over zone 2 (Figure 5.3(d)). All these schemes are based on an FxMWF.

The standard ANC scheme allows to reduce the noise power by about 10dB at the ear canal microphone but its performance is degraded to a reduction of 4dB at the eardrum ($\mathbf{r}_{TM}$). The RP-based ANC scheme allows to reduce the noise

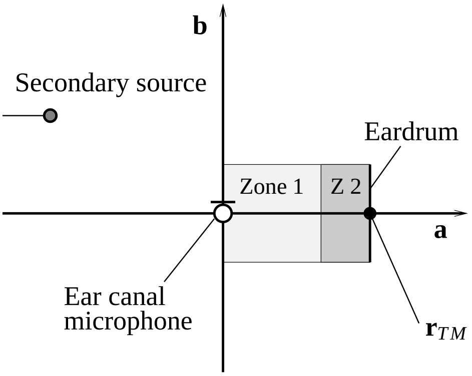


Figure 5.2: Simulation setup

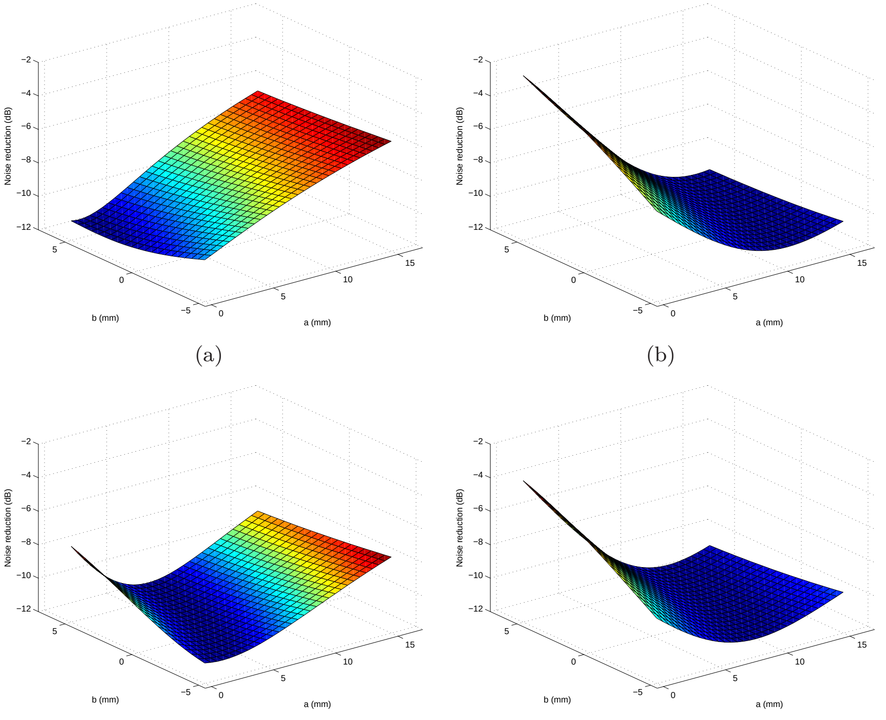
	Pow(0)	Pow($\mathbf{r}_{TM}$)
Standard ANC	-10.7	-5.8
RP-based ANC	-4	-10.6

Table 5.1: Noise power at the ear canal microphone and at the eardrum (in dB).

power by about 10dB at $\mathbf{r}_{TM}$ and therefore restores the ANC performance at the eardrum (Table 5.1). When the noise is to be controlled over a desired zone of quiet, however, the RP-based ANC scheme does not provide a satisfying approach as its residual noise power quickly increase when moving away from $\mathbf{r}_{TM}$.

The noise reduction performance of the ZQ-based ANC designed to minimise the noise over the close neighbourhood of the eardrum (zone 2) is shown in Figure 5.3(d). When the standard ANC scheme achieves an average noise reduction of 6dB over the considered zone, the ZQ-based ANC scheme achieves an average performance of about 10dB. Note also that the maximum noise power over the considered zone drops from -5.4dB with the standard ANC scheme to -9.6dB with the ZQ-based ANC scheme (Table 5.2). The risk of constructive interferences is then highly reduced.

When the noise is controlled over the full ear canal (zone 1) the average noise reduction delivered by the standard ANC scheme is 8dB. The ZQ-based ANC scheme set to minimise the noise over zone 2 (Figure 5.3(c)) achieves an average noise reduction of about 10dB, improving the performance of the standard scheme by almost 2dB. The maximum noise power over the ear canal also drops from -5.4dB with the standard ANC scheme to -7.4dB with the ZQ-based ANC scheme (Table 5.3).



(a)

(b)

(c)

(d)

Figure 5.3: Noise reduction for multichannel ANC: (a) Standard Scheme, (b) RP-based, (c) ZQ-based ANC (Zone 1), (d) ZQ-based ANC (Zone 2)

	min(Pow)	max(Pow)	aPow
Standard ANC	-7.5	-5.4	-6.4
ZQ-based ANC (Zone 2)	-10.9	-9.6	-10.6

Table 5.2: Minimum, maximum and average noise power over zone 2 (in dB).

It appears that for the standard ANC scheme the noise reduction performance quickly decreases when the ear canal microphone is away from the eardrum. The RP-based approach allows to restore the ANC performance at a single point but its performance also decreases when moving away from the optimised point. When the noise is to be considered over a ZQ, the ZQ-based approach has shown to provide significant performance improvement. It also allows to keep the maximum noise power over the ZQ at a lower level than with the standard ANC scheme, thus preventing for possible constructive interferences.

	min(Pow)	max(Pow)	aPow
Standard ANC	-10.9	-5.4	-8
ZQ-based ANC (Zone 1)	-10.9	-7.4	-9.7

Table 5.3: Minimum, maximum and average noise power over zone 1 (in dB).

	Pow(0)	Pow($\mathbf{r}_{TM}$)
Original integrated ANC and NR	9.1	4.4
RP-based integrated ANC and NR	6.7	9.7

Table 5.4: Signal-to-noise ratio performance at the ear canal microphone and at the eardrum (in dB).

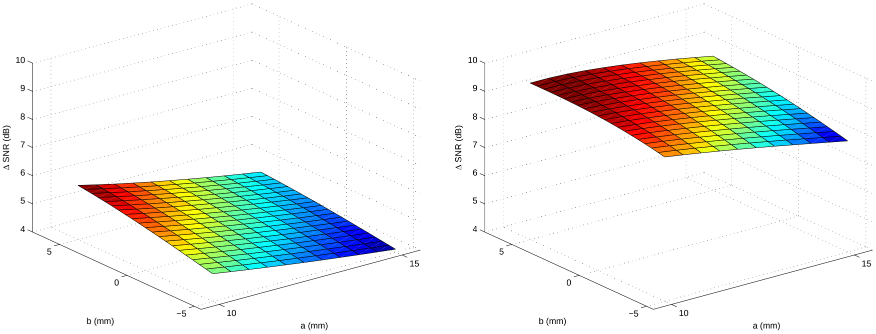
5.4.3 Integrated active noise control and noise reduction

The SNR has been computed over the ear canal and the aSNR has been computed over the two zones of quiet for the output of the original integrated ANC and NR scheme, the RP-based integrated ANC and NR scheme designed to minimise the MSE criterion (5.17) at point $\mathbf{r}_{TM}$, a ZQ-based integrated ANC and NR scheme designed to minimise the aMSE criterion (5.48) over the ear canal (zone 1) and a ZQ-based integrated ANC and NR scheme designed to minimise the aMSE criterion over the neighbourhood of the eardrum (zone 2).

The original integrated ANC and NR scheme delivers a SNR of about 9dB at the ear canal microphone but its performance is fluctuating over the ear canal (Figure 5.5(a)) and its output SNR eventually decreases to 4.4dB at the eardrum ($\mathbf{r}_{TM}$).

The RP-based integrated ANC and NR scheme delivers a 9.7dB SNR at $\mathbf{r}_{TM}$ and therefore restores at the eardrum the original integrated ANC and NR scheme performance at the canal microphone (Table 5.4). When the noise is to be controlled over a desired zone of quiet, however, the RP-based integrated ANC and NR scheme does not provide a satisfying approach as its SNR quickly decreases when moving away from $\mathbf{r}_{TM}$.

The SNR performance of the ZQ-based integrated ANC and NR scheme designed to minimise the noise over the close neighbourhood of the eardrum (zone 2) is shown in Figure 5.4(b) and compared to the performance of the original integrated ANC and NR scheme over the same zone (Figure 5.4(a)). When the original integrated ANC and NR scheme achieves an aSNR of 4.9dB over the considered zone, the ZQ-based integrated ANC and NR scheme achieves an aSNR of about 9dB. Note also that the minimum SNR over the zone increases from 4.1dB with



(a) (b)

Figure 5.4: SNR performance of integrated ANC and NR schemes over zone 2: (a) Original scheme, (b) ZQ-based scheme (zone 2).

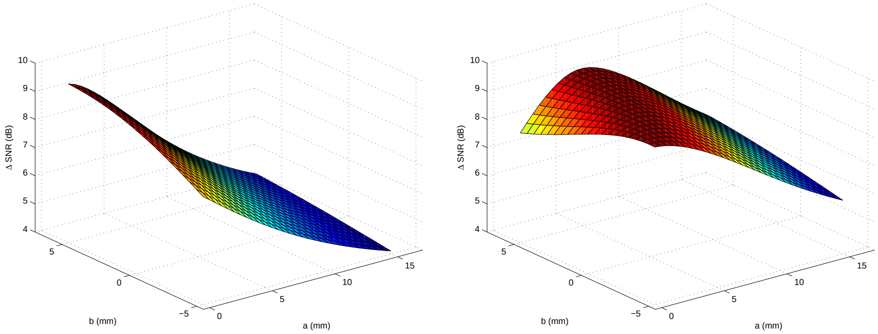
	min(SNR)	max(SNR)	aSNR
Original integrated ANC an NR	4.1	5.9	4.9
ZQ-based integrated ANC and NR (Zone 2)	7.9	10	9

Table 5.5: Minimum and average signal-to-noise ratio, over zone 2 (in dB).

the original integrated ANC and NR scheme to about 8dB with the ZQ-based integrated ANC and NR scheme (Table 5.2).

When the zone of quiet to be considered is the full ear canal (zone 1) the aSNR (Figure 5.5(a)) delivered by the original integrated ANC and NR scheme is 6.4dB. The ZQ-based integrated ANC and NR scheme designed to minimise the aMSE over zone 1 (Figure 5.5(b)) delivers an aSNR of 8.3dB. Note also that the minimum SNR over the ear canal increases from 4.1dB with the original ANC and NR scheme to about 5.9dB with the ZQ-based integrated ANC and NR scheme (Table 5.2).

It appears that the performance of the original integrated ANC and NR scheme quickly decreases when the ear canal microphone is away from the eardrum. The RP-based approach allows to restore performance of the integrated ANC and NR scheme at a single point but its performance also decreases when moving away from the optimised point. The ZQ-based approach, adjusted to the zone on which the noise is to be cancelled, exhibits improved performance in aSNR compared to the original integrated ANC and NR scheme. It also allows to improve the minimum SNR, therefore reducing the impact of the performance degradation where it is most significant.



(a) (b)

Figure 5.5: SNR performance of integrated ANC and NR schemes over zone 1: (a) Standard scheme, (b) ZQ-based scheme (zone 1)

	min(SNR)	max(SNR)	aSNR
Original integrated ANC and NR	4.1	9.6	6.4
ZQ-based integrated ANC and NR (Zone 1)	5.9	9.6	8.3

Table 5.6: Minimum and average signal-to-noise ratio, over zone 1 (in dB).

5.5 Conclusion

It has been shown in previous chapters that ANC is an efficient solution to the signal leakage problem in hearing aids with an open fitting. A regular ANC scheme and the integrated ANC and NR scheme allow to control the noise at the ear canal microphone. Ideally, this microphone should be located at the eardrum. In practice however, the ear canal microphone is away from the eardrum and hence the sound reaching the eardrum is basically unknown and uncontrolled.

The RP-based approach, based on an MSE criterion expressed at an RP away from the ear canal microphone, applied to the ANC scheme and the integrated ANC and NR scheme, has been shown to provide a performance improvement compared to the original schemes. Under a single speech source scenario and when the number of sound sources (speech source plus noise sources) is less than or equal to the number of hearing aid microphones, it is possible to derive a formula for the RP-based integrated ANC and NR SNR performance. This confirms that the RP-based approach allows to restore at an RP the integrated ANC and NR scheme SNR performance at the ear canal microphone. The RP-based approach, however, cannot be a satisfying approach when the noise is to be controlled over a ZQ rather than at a single point.

The so-called ZQ-based approach allows to optimise the average of a particular design criterion, here the MSE criterion, over a ZQ generated by the ANC. This approach, applied to the ANC scheme and the integrated ANC and NR scheme, exhibits improved performance, compared to the original schemes, especially when the noise is to be cancelled on a zone that is away from the ear canal microphone. The ZQ-based approach also allows to reduce the impact of constructive interference over the desired ZQ.

The ZQ based approach thus provides a more robust and more realistic way to apply ANC and solve the signal leakage problem in the framework of hearing aids with an open fitting.

Chapter 6

Binaural Integrated Active Noise Control and Noise Reduction

This chapter presents a binaural approach to integrated active noise control and noise reduction. The monaural integrated active noise control and noise reduction scheme introduced in Chapter 3 only has access the microphones of one hearing aid and so is subject to constraints on the number of microphones and the spatial separation between the microphones. The binaural approach allows to have access to extra microphones from the contra-lateral hearing aid. Moreover, the spatial separation between the microphones on two hearing aids allows to design a scheme with increased causality margin.

6.1 Introduction

It has been known for years that binaural hearing offers advantages over monaural hearing such as a: better speech intelligibility, enhanced localisation, improved quality of listening. . . [56, 65, 70] If binaural information is really helpful for normal hearing persons, it may become tremendously important for persons with an hearing impairment.

State-of-the-art hearing aids perform NR in order to improve their output SNR and hence to allow for a better speech understanding in background noise and to ease listening effort [63]. Conventional NR systems such as the GSC [37] or techniques based on the MWF [10, 21, 94] are commonly used.

When these processing schemes are applied in a monaural setup or a bilateral setup (*i.e.*, when the listener wears two hearing aids working independently), the SNR improvement can come with a degradation of binaural localisation cues which can put the hearing aid user at disadvantage. In a binaural setup, the hearing aid user wears two hearing aids which can communicate, *e.g.*, via a wireless link. The NR schemes applied in hearing aids can be adapted to take advantage of this setup to deliver improved SNR [19, 122] and to preserve binaural localisation cues [55, 118].

Conventional NR techniques, be they monaural, bilateral or binaural, do not take leakage effects into account. Combined with the attenuation in the secondary path, the noise leaking through the open fitting directly to the eardrum can then override the action of the NR. One efficient way to tackle this problem is to use ANC [25, 59] and integrated this with the NR as introduced in Chapter 3.

It has been shown in Chapter 3 that, to be effective, the integrated ANC and NR scheme needs to be designed as a causal system. In the case of hearing aids, the causality margins are rather small and so this causality may quickly become a limitation. It has also been shown in Chapter 3 that in the single speech source case, the integrated ANC and NR scheme can compensate for noise sources only as long as the number of sources (speech source and noise sources) is less than or equal to than the number of microphones available (which is maximally three in the case of commercial hearing aids). A binaural integrated ANC and NR scheme can benefit from the increased number of available microphones as well as from the causality margin increase owing to the (outpost) location of the contra-lateral microphones.

The signal model for binaural processing, the binaural MWF-based NR and the secondary path effects and the effects of the signal leakage on the NR performance are described in Section 6.2. A binaural multichannel ANC scheme are presented in Section 6.3. A binaural integrated ANC and NR scheme are presented in Section 6.4. Experimental results are presented in Section 6.5 and finally conclusions are presented in Section 6.6.

6.2 Background and problem statement

This section introduces the signal model and notation for a binaural setup. The binaural MWF and the effects of secondary path and signal leakage effects on the performance of a MWF-based binaural NR are then presented.

6.2.1 Signal model

Figure 6.1 presents a typical listening scenario where the hearing aid user is wearing two hearing aids. In a binaural setup, signals from both hearing aids are used to compute the hearing aid loudspeaker signals. The section therefore redefines the signal model for a binaural setup.

Let M be the number of microphones (channels) on each hearing aid and N the filter length. The time-domain signal $x_{L,m}$ for microphone m in the left hearing aid has a desired speech part $x^s_{L,m}$ and an additive noise part $x^n_{L,m}$, *i.e.*,:

$$x_{L,m}[k] = x^s_{L,m}[k] + x^n_{L,m}[k] \quad m \in \{1 \dots M\} \tag{6.1}$$

The column vector $\mathbf{x}_{L,m}[k]$ contains the N last samples of the microphone signal m in the left hearing aid:

$$\mathbf{x}_{L,m}[k] = [x_{L,m}[k] \dots x_{L,m}[k-N+1]]^T \quad m \in \{1 \dots M\} \tag{6.2}$$

Similarly, the time-domain signal $x_{R,m}$ for microphone m in the right hearing aid can be specified as:

$$x_{R,m}[k] = x^s_{R,m}[k] + x^n_{R,m}[k] \tag{6.3}$$

$$\mathbf{x}_{R,m}[k] = [x_{R,m}[k] \dots x_{R,m}[k-N+1]]^T \quad m \in \{1 \dots M\}$$

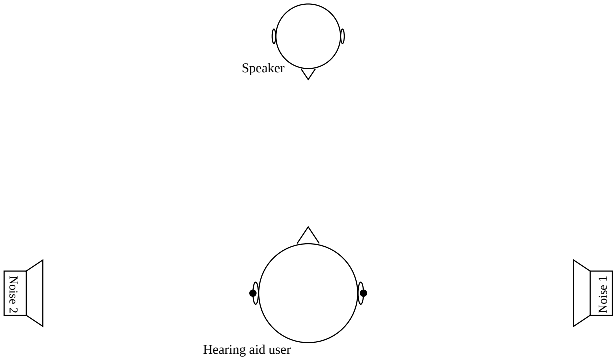


Figure 6.1: Typical listening scenario

The MN-dimensional compound vectors $\mathbf{x}_L$ and $\mathbf{x}_R$ gathering all microphone signals from the left and the right hearing aid respectively and the $2MN$-

dimensional compound signal vector $\mathbf{x}$ are defined as follows:

$$\mathbf{x}_L[k] = \begin{bmatrix} \mathbf{x}_{L,1}[k] \\ \vdots \\ \mathbf{x}_{L,M}[k] \end{bmatrix} \tag{6.4}$$

$$\mathbf{x}_R[k] = \begin{bmatrix} \mathbf{x}_{R,1}[k] \\ \vdots \\ \mathbf{x}_{R,M}[k] \end{bmatrix} \tag{6.5}$$

$$\mathbf{x}[k] = \begin{bmatrix} \mathbf{x}_L[k] \\ \mathbf{x}_R[k] \end{bmatrix} \tag{6.6}$$

6.2.2 Binaural multichannel Wiener filter

In a binaural setup, the hearing aid user is wearing two hearing aids which can communicate, *e.g.*, via a wireless link. It is assumed in the remainder of this chapter that signals can be transmitted from one hearing aid to the other without any degradation or transmission delay. In practice, the propagation delay itself is negligible compared to the acoustic propagation delay from one ear to the other ear. The processing delay (conversion, detection, forward error correction…) on the other hand depends on the parameters of the transmission (robustness, sampling frequency…).

The binaural MWF [11, 55] is an extension of the MWF [10, 21, 94] presented in Chapter 2. The goal is to estimate a desired speech signal pair based on the microphone signals $\mathbf{x}_L$ and $\mathbf{x}_R$.

The desired speech signal pair is arbitrarily chosen to be the (unknown) speech component of the first microphone signal ($m = 1$) from the left hearing aid and the right hearing aid, up to a delay Δ:

$$d_{\mathrm{NR},L}[k] = G_L \cdot x^s_{L,1}[k-\Delta] \tag{6.7}$$

$$d_{\mathrm{NR},R}[k] = G_R \cdot x^s_{R,1}[k-\Delta] \tag{6.8}$$

where G_L and G_R are the hearing aid gains of the left and right hearing aid respectively.

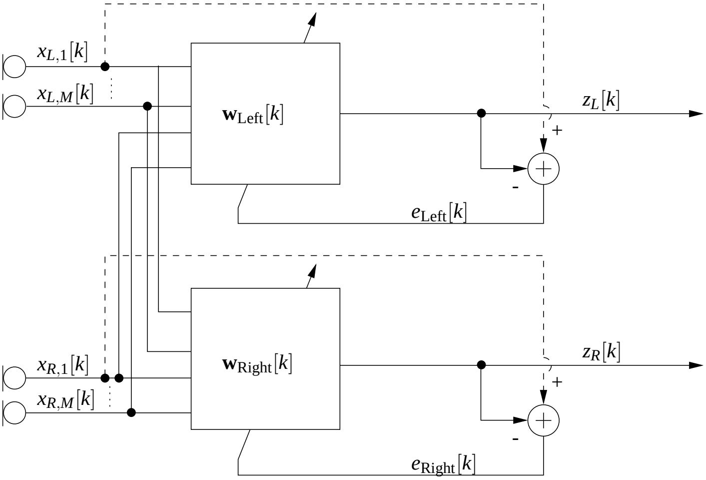


Figure 6.2: Binaural multichannel Wiener filter

In each hearing aid, a $2MN$-dimensional MWF is designed to estimate the corresponding desired speech signal:

$$\mathbf{w}_{\text{Left}}[k] = \begin{bmatrix} \mathbf{w}_{\text{Left},L}[k] \\ \mathbf{w}_{\text{Left},R}[k] \end{bmatrix} = \begin{bmatrix} \mathbf{w}_{\text{Left},L,1}[k] \\ \vdots \\ \mathbf{w}_{\text{Left},L,M}[k] \\ \mathbf{w}_{\text{Left},R,1}[k] \\ \vdots \\ \mathbf{w}_{\text{Left},R,M}[k] \end{bmatrix} \tag{6.9}$$

$$\mathbf{w}_{\text{Right}}[k] = \begin{bmatrix} \mathbf{w}_{\text{Right},L}[k] \\ \mathbf{w}_{\text{Right},R}[k] \end{bmatrix} = \begin{bmatrix} \mathbf{w}_{\text{Right},L,1}[k] \\ \vdots \\ \mathbf{w}_{\text{Right},L,M}[k] \\ \mathbf{w}_{\text{Right},R,1}[k] \\ \vdots \\ \mathbf{w}_{\text{Right},R,M}[k] \end{bmatrix} \tag{6.10}$$

The filters $\mathbf{w}_{\text{Left}}[k]$ and $\mathbf{w}_{\text{Right}}[k]$ minimise the errors $e_{\text{Left}}[k]$ and $e_{\text{Right}}[k]$ respectively:

$$e_{Left}[k] = d_{\text{NR},L}[k] - (\mathbf{w}_{\text{Left},L}[k]^T\mathbf{x}_L[k] + \mathbf{w}_{\text{Left},R}[k]^T\mathbf{x}_R[k]) \tag{6.11}$$

$$e_{Left}[k] = d_{\text{NR},R}[k] - (\mathbf{w}_{\text{Right},L}[k]^T\mathbf{x}_L[k] + \mathbf{w}_{\text{Right},R}[k]^T\mathbf{x}_R[k]) \tag{6.12}$$

In the MWF-based NR framework, the binaural filtering process (illustrated on Figure 6.2) minimises the following MSE criteria:

$$\begin{aligned} J_{\text{NR,Left}}[k] &= \mathbb{E}\{|e_{\text{NR,Left}}[k]|^2\} \\ &= \mathbb{E}\{|d_{\text{NR},L}[k] - \mathbf{w}^T_{\text{Left}}[k]\mathbf{x}[k]|^2\} \end{aligned} \tag{6.13}$$

$$\begin{aligned} J_{\text{NR,Right}}[k] &= \mathbb{E}\{|e_{\text{NR,Right}}[k]|^2\} \\ &= \mathbb{E}\{|d_{\text{NR},R}[k] - \mathbf{w}^T_{\text{Right}}[k]\mathbf{x}[k]|^2\} \end{aligned} \tag{6.14}$$

The MWF-based NR filters are then given by:

$$\mathbf{w}_{\text{NR,Left}}[k] = \mathbf{R}_x^{-1}[k]\mathbf{r}_{xd_{\text{NR},L}}[k] \tag{6.15}$$

$$\mathbf{w}_{\text{NR,Right}}[k] = \mathbf{R}_x^{-1}[k]\mathbf{r}_{xd_{\text{NR},R}}[k] \tag{6.16}$$

Here $\mathbf{R}_x[k]$ is the correlation matrix of the input $\mathbf{x}[k]$

$$\mathbf{R}_x[k] = \mathbb{E}\{\mathbf{x}[k]\mathbf{x}[k]^T\} \tag{6.17}$$

and $\mathbf{r}_{xd_{L,\text{NR}}}[k]$ and $\mathbf{r}_{xd_{R,\text{NR}}}[k]$ are the cross-correlation vectors between the input $\mathbf{x}[k]$ and the desired signals for the left hearing aid $d_{\text{NR},L}[k]$ and the right hearing aid $d_{\text{NR},R}[k]$, respectively:

$$\mathbf{r}_{xd_{\text{NR},L}}[k] = G_L\cdot\mathbb{E}\{\mathbf{x}[k]x^s_{L,1}[k-\Delta]\} \tag{6.18}$$

$$\mathbf{r}_{xd_{\text{NR},R}}[k] = G_R\cdot\mathbb{E}\{\mathbf{x}[k]x^s_{R,1}[k-\Delta]\} \tag{6.19}$$

The filter output signals $z_L[k]$ and $z_R[k]$ (*i.e.*, the signal to be fed in the hearing aid loudspeakers) are defined as:

$$z_L[k] = \mathbf{w}_{\text{NR,Left}}[k]^T \mathbf{x}[k] \tag{6.20}$$

$$z_R[k] = \mathbf{w}_{\text{NR,Right}}[k]^T \mathbf{x}[k] \tag{6.21}$$

6.2.3 Multichannel Wiener filter-based noise reduction, secondary path and signal leakage

In the subsequent sections, only the filter designed for the left hearing aid ($\mathbf{w}_{\text{Left}}[k]$) will be considered. All the reasoning, however, can be transposed to the filter in the right hearing aid ($\mathbf{w}_{\text{Right}}[k]$) as well.

The filter (6.15) is designed without taking the effects of the signal leakage and the secondary path effects into account. Figure 6.3 presents a binaural MWF-based NR where the secondary path and the signal leakage are added.

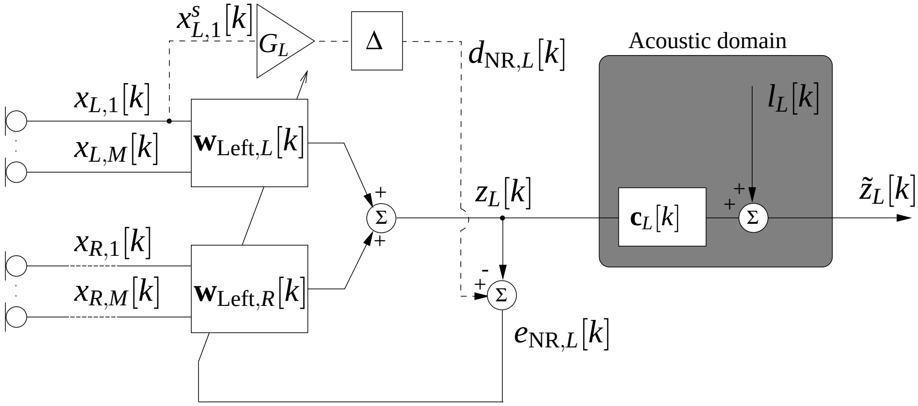


Figure 6.3: Binaural MWF-based NR in the left hearing aid

The secondary path which represents the propagation from the loudspeaker to the eardrum has been introduced in Chapter 2. Assuming that the loudspeaker characteristic is approximately linear, the secondary path can be represented by a filter coefficient vector $\mathbf{c}[k]$ of length P.

A 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 $l_L[k]$ reaching the eardrum (see also Chapter 2).

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

$$\tilde{z}_L[k] = \mathbf{c}_L[k]^T \begin{bmatrix} z_L[k] \\ \vdots \\ z_L[k-P+1] \end{bmatrix} + l_L[k] \tag{6.22}$$

It clearly appears that for small gains G_L the leakage SNR may affect the output SNR thus partly cancelling the improvement achieved with the NR. In conclusion, whereas the secondary path and the leakage are not taken into account in conventional NR algorithms, they may degrade their performance significantly.

For conciseness, the filter $\mathbf{w}_{\text{Left}}[k]$ and the corresponding error $e_{\text{Left}}[k]$ will be denoted $\mathbf{w}[k]$ and $e[k]$; G_L, $z_L[k]$, $\mathbf{c}_L[k]$ and $l_L[k]$ will be denoted G, $z[k]$, $\mathbf{c}[k]$ and $l[k]$ respectively in the remainder of the chapter.

6.3 Binaural active noise control

The goal of the binaural ANC (Figure 6.4) is to extend the monaural multichannel ANC presented in Section 2.4 based on existing work on binaural MWF [55, 11]. The binaural ANC scheme can then benefit from the location of hearing aid microphones on both sides of the head to access microphone signals with longer propagation time (*e.g.*, acoustic propagation from the right hearing aid microphones to the left ear canal microphone (Figure 6.5)). In some scenarios, such signals will allow to design system with higher causality margin and therefore to overcome partly the latency problem.

The binaural ANC scheme again relies on a Filtered-x structure. The filtered reference signal is now defined as:

$$\mathbf{y}[k] = \begin{bmatrix} \mathbf{y}_L[k] \\ \mathbf{y}_R[k] \end{bmatrix} \tag{6.23}$$

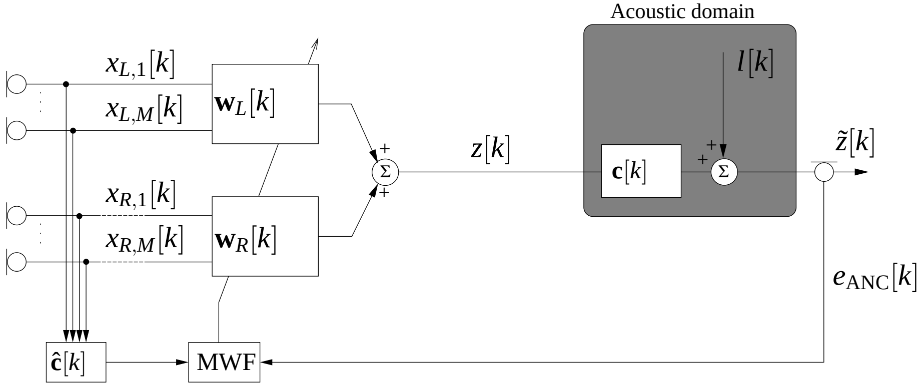


Figure 6.4: Binaural active noise control in the left hearing aid

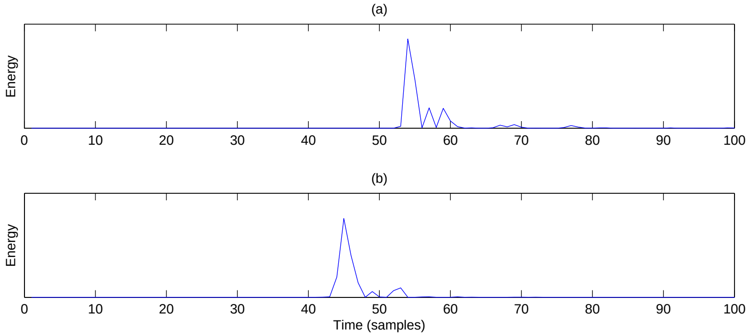


Figure 6.5: (a) Impulse response from a noise source at 90° to a microphone of the left BTE, (b) Impulse response from a noise source at 90° to a microphone of the right BTE

where $\mathbf{y}_L[k]$ is the filtered reference signal derived from the microphone signals in the left hearing aid:

$$y_{L,m}[k] = \hat{\mathbf{c}}[k]^T \begin{bmatrix} x_{L,m}[k] \\ \vdots \\ x_{L,m}[k-\hat{P}+1] \end{bmatrix} \quad m \in \{1 \dots M\} \tag{6.24}$$

$$\mathbf{y}_{L,m}[k] = [y_{L,m}[k] \dots y_{L,m}[k-N+1]]^T \tag{6.25}$$

$$\mathbf{y}_L[k] = \begin{bmatrix} \mathbf{y}_{L,1}[k] \\ \vdots \\ \mathbf{y}_{L,M}[k] \end{bmatrix} \tag{6.26}$$

and where $\mathbf{y}_R[k]$ is the filtered reference signal derived from the microphone signals in the right hearing aid, is defined as follows:

$$y_{R,m}[k] = \hat{\mathbf{c}}[k]^T \begin{bmatrix} x_{R,m}[k] \\ \vdots \\ x_{R,m}[k-\hat{P}+1] \end{bmatrix} \quad m \in \{1 \ldots M\} \tag{6.27}$$

$$\mathbf{y}_{R,m}[k] = [y_{R,m}[k] \ldots y_{R,m}[k-N+1]]^T \tag{6.28}$$

$$\mathbf{y}_R[k] = \begin{bmatrix} \mathbf{y}_{R,1}[k] \\ \vdots \\ \mathbf{y}_{R,M}[k] \end{bmatrix} \tag{6.29}$$

The ANC output signal is now:

$$z[k] = \mathbf{w}^T[k]\mathbf{x}[k] \tag{6.30}$$

where the filter $\mathbf{w}[k]$ is designed to minimise the MSE:

$$J_{\text{ANC}}[k] = \mathbb{E}\{|e_{\text{ANC}}[k]|^2\} \tag{6.31}$$

Here, $e_{\text{ANC}}[k]$ is an error signal, constructed from the ear canal microphone signal $\tilde{z}[k]$:

$$e_{\text{ANC}}[k] = \tilde{z}[k] \tag{6.32}$$

$$= \hat{\mathbf{c}}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-\hat{P}+1] \end{bmatrix} + l[k] \tag{6.33}$$

Assuming that the secondary path identification error is small ($\hat{C} \approx C$) and that the filter $\mathbf{w}[k]$ is adapting slowly, the MSE criterion (6.31) can be written as follows:

$$J_{\text{ANC}}[k] \approx \mathbb{E}\{|\mathbf{w}^T[k]\mathbf{y}[k] + l[k]|^2\} \tag{6.34}$$

The Filtered-x MWF (FxMWF) minimising (6.34) is then:

$$\boxed{\mathbf{w}_{\text{ANC}}[k] = -\mathbf{R}_y[k]^{-1}\mathbf{r}_{yl}[k]} \tag{6.35}$$

Here $\mathbf{R}_y[k]$ is the correlation matrix of the filtered microphone signals $\mathbf{y}[k]$ and $\mathbf{r}_{yl}[k]$ is the cross-correlation vector between the filtered microphone signals $\mathbf{y}[k]$ and the leakage signal $l[k]$:

$$\mathbf{R}_y[k] = \mathbb{E}\{\mathbf{y}[k]\mathbf{y}[k]^T\} \tag{6.36}$$

$$\mathbf{r}_{yl}[k] = \mathbb{E}\{\mathbf{y}[k]l[k]\} \tag{6.37}$$

6.4 Binaural integrated active noise control and noise reduction

The goal of the binaural integrated ANC and NR scheme (Figure 6.6) is to extend the monaural integrated ANC and NR scheme presented in the Chapter 3. The binaural integrated ANC and NR can then benefit from the extra contra-lateral hearing aid microphones in order to compensate for the effect of more noise sources. The ANC part of the integrated ANC and NR scheme can also benefit from the causality margin improvement owing to the contra-lateral hearing aid microphones location.

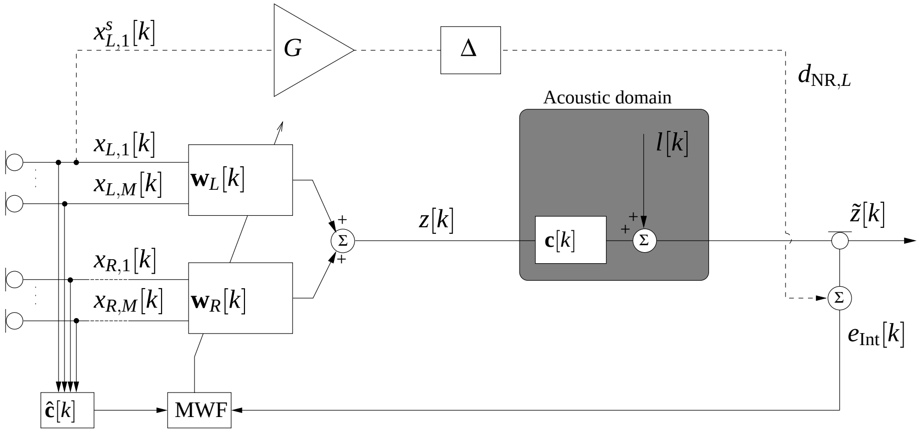


Figure 6.6: Binaural integrated active noise control and noise reduction

The overall desired signal (at the eardrum) to be used is:

$$d_{\mathrm{Int},L}[k] = d_{\mathrm{NR},L}[k] + l^s[k] \tag{6.38}$$

Hence the MSE criterion to be minimised is:

$$J_{\mathrm{Int}}[k] = \mathbb{E}\{|E_{\mathrm{Int}}[k]|^2\} \tag{6.39}$$

$$\begin{aligned} E_{\mathrm{Int}}[k] &= \mathbf{c}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-P+1] \end{bmatrix} + l[k] - d_{\mathrm{Int},L}[k] \\ &= \mathbf{c}[k]^T \begin{bmatrix} z[k] \\ \vdots \\ z[k-P+1] \end{bmatrix} + l^n[k] - d_{\mathrm{NR},L}[k] \end{aligned} \tag{6.40}$$

where

$$z[k] = \mathbf{w}^T[k]\mathbf{x}[k] \tag{6.41}$$

Assuming that the secondary path identification error is small ($\hat{\mathbf{c}}[k] \approx \mathbf{c}[k]$) and that the filter $\mathbf{w}[k]$ is adapting slowly, the MSE criterion (6.39) can be written as follows:

$$J_{\mathrm{Int}}[k] \approx \mathbb{E}\{|\mathbf{w}^T[k]\mathbf{y}[k] + l^n[k] - d_{\mathrm{NR},L}[k]|^2\} \tag{6.42}$$

The FxMWF minimising (6.39) is then:

$$\mathbf{w}_{\mathrm{Int}}[k] = \mathbf{R}_y^{-1}[k](\mathbf{r}_{yd_{\mathrm{NR},L}}[k] - \mathbf{r}_{yl^n}[k]) \tag{6.43}$$

Here $\mathbf{R}_y[k]$ is the correlation matrix of the filtered microphone signals $\mathbf{y}[k]$ and $\mathbf{r}_{y^s d_{\mathrm{NR},L}}[k]$ and $\mathbf{r}_{y^n l^n}[k]$ are the cross-correlation vectors between the filtered microphone signals $\mathbf{y}[k]$ and the desired signal $d_{\mathrm{NR},L}[k]$ and the noise component of the leakage signal $l^n[k]$, respectively.

Assuming that the speech and noise signals are uncorrelated, the filter (6.43) can be rewritten as follows:

$$\boxed{\mathbf{w}_{\mathrm{Int}}[k] = \mathbf{R}_y^{-1}[k](\mathbf{r}_{y^s d_{\mathrm{NR},L}}[k] - \mathbf{r}_{y^n l^n}[k])} \tag{6.44}$$

6.5 Experimental results

The binaural versions of the multichannel ANC scheme presented in Section 6.3 and of the integrated ANC and NR scheme presented in Section 6.4, have been tested experimentally and their performance has been compared to the performance of the monaural ANC scheme introduced in Section 2.4 and the monaural integrated ANC and NR scheme introduced in Chapter 3.

6.5.1 Experimental setup

The filter length is set to $N = 128$, and the NR delay is set to half of the NR filter length ($\Delta = 64$). The secondary path $\mathbf{c}[k]$ is estimated off-line using an identification technique based on the NLMS algorithm. The length of the estimated path $\hat{\mathbf{c}}[k]$ is set to $L = 32$.

Active noise control

The performance measure used for the ANC schemes is the residual noise power at the eardrum. The reference power is the power of the leakage signal's noise component, which can also be considered as the residual noise power when the hearing aid is turned off. The formula for the residual noise power improvement at the eardrum is given in (2.34).

Integrated active noise control and noise reduction

The performance measure used for the integrated ANC and NR schemes is the intelligibility-weighted SNR [36]. The leakage signal SNR, which can also be considered as the SNR when the hearing aid is turned off, is taken as a reference. The formula for the speech intelligibility-weighted SNR is given in (2.34).

6.5.2 Active noise control

In this part, three different 2-channel ANC schemes are compared. Two monaural schemes are considered (one using the microphones signals from the left hearing aid, the other one using the microphone signals from the right hearing aid) and compared to a binaural scheme using one microphone signal from each hearing aid. The binaural scheme can also run with four microphone signals but it has been decided to use only two microphone signals in order to have a fair comparison with the monaural schemes.

Single noise source

The first experiment is set up with only one noise source. The noise source can be located at 0° (facing the hearing aids user), 90° (facing the right ear) or at 270° (facing the left ear). In each case the three different schemes are run for different degrees of causality δ (1.5) and the residual noise power performance is evaluated.

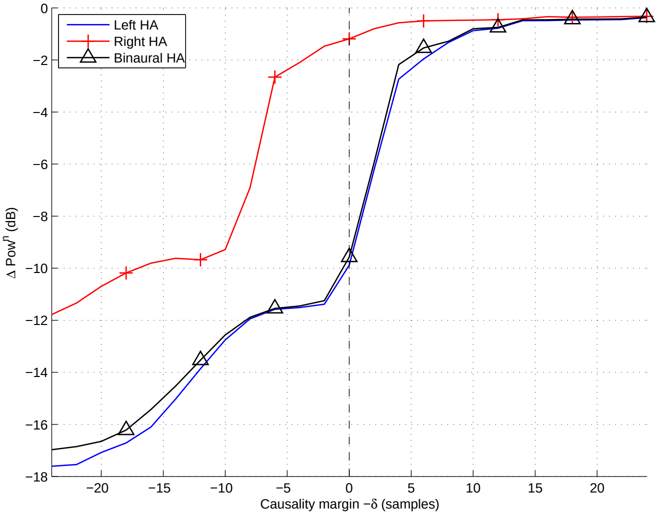


Figure 6.7: Noise reduction for multichannel (monaural, binaural) active noise control with a single noise source on the left of the listener (270°)

Figure 6.7 presents the residual noise power improvement at the left eardrum for the three schemes when the source is facing the left hearing aid (270°). The noise signal then reaches the microphones of the left hearing aid before reaching the left eardrum. It is therefore possible to design a causal system based on the microphone signals from the left hearing aid, even if the causality margin is rather small. On the other hand, the noise signal reaches the microphones of the right hearing aid after reaching the left eardrum. The ANC scheme to design based on the right hearing aid microphone signals is then non causal. The binaural ANC scheme is based on a microphone signal from each hearing aid. It can be seen on Figure 6.7 that the binaural ANC scheme matches the residual noise power performance of the monaural scheme applied on the microphone signals from the left hearing aid.

The residual noise power improvement at the left eardrum when the noise source is facing the listener (0°) is shown on Figure 6.8. The noise signal reaches the microphones of the left hearing aid at the same time as it reaches the microphones of the right hearing aid. The causality margins are then the same if the system is based on the microphone signals from the left hearing aid or from the right hearing aid. The binaural ANC scheme is based on one microphone signal from each hearing aid and its performance is, in this scenario, similar to the performance of the monaural schemes.

Figure 6.9 presents the noise power improvement of the residual noise at the left eardrum for the three schemes when the noise source is facing the right hearing

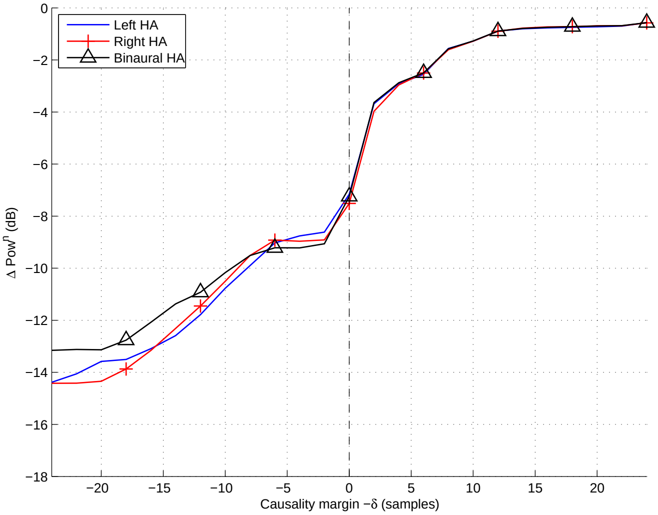


Figure 6.8: Noise reduction for multichannel (monaural, binaural) active noise control with a single noise source facing the listener (0°)

aid (90°). The noise signal then reaches the left eardrum shortly after it reaches the microphones of the left hearing aid. The monaural ANC scheme based on the microphone signals from the left hearing aid then has to be designed with low causality margin. In this scenario however, the noise signal reaches the microphones of the right hearing aid before reaching the left eardrum. Therefore, the ANC scheme based on the microphone signals from the right hearing aid can be designed with a larger causality margin.

Once again in this scenario, as the binaural ANC scheme is based on a microphone signal from each hearing aid, it can be seen on Figure 6.9 that the binaural ANC scheme matches the residual noise power performance of the best of the two monaural schemes (in this case, the ANC scheme based on the microphones signals from the right hearing aid).

Multiple noise sources

The second experiment is set up with two noise source: one on each side of the listener, *e.g.*, one noise source at 270° and the other noise source at 90°. The residual noise power improvement at the left eardrum is presented in Figure 6.10. Each of the monaural ANC schemes is well suited to attenuate the noise signal from one of the sources (namely the source which is facing the hearing aid of which the microphone signals are used in the ANC) but the attenuation of the

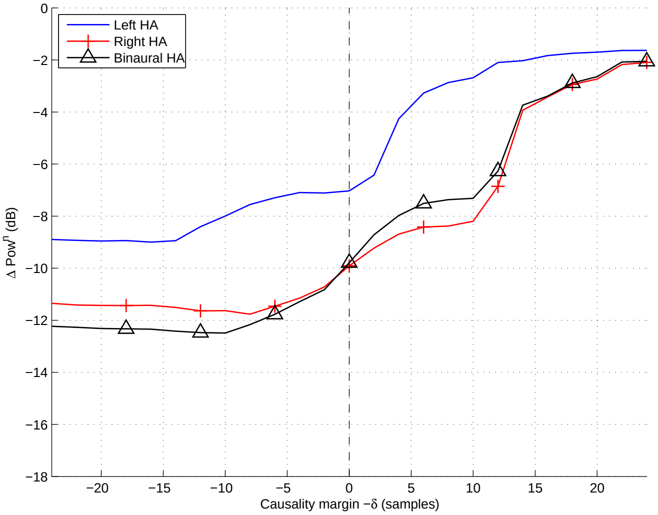


Figure 6.9: Noise reduction for multichannel (monaural, binaural) active noise control with a single noise source on the right of the listener (90°)

other noise source can be problematic (see also above). The binaural ANC scheme on the other hand delivers a better performance than any of the monaural scheme in this particular case.

6.5.3 Integrated active noise control and noise reduction

In this section, the performance of the integrated ANC and NR schemes are compared. The first experiment aims at showing the effect of causality on different integrated ANC and NR schemes while the second experiment focuses on the impact of the number of sources on the integrated ANC and NR schemes.

Single noise source

In the first experiment there is only one noise source which can be located at 90° or at 270°. The gain G is set to 10dB. For each scenario the three different schemes are run for different degrees of causality δ (1.5) and their speech-intelligibility weighted SNR improvement is evaluated.

Three different 2-channel integrated ANC and NR schemes are compared here. As in the previous part, two monaural schemes are considered (one using the microphones signals from the left hearing aid, the other one using the microphone

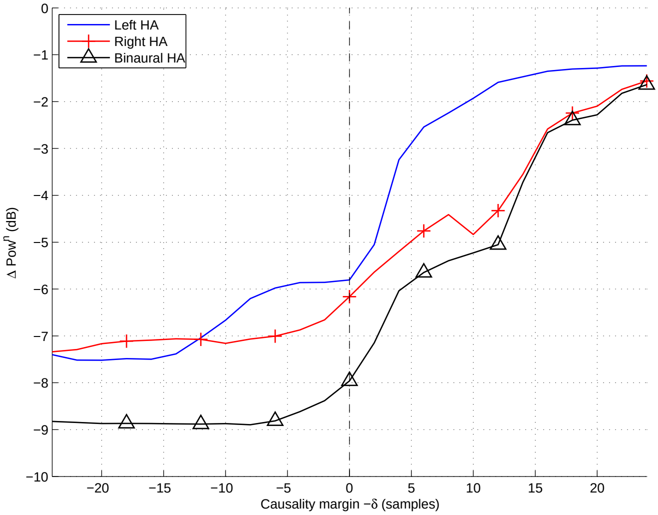


Figure 6.10: Noise reduction for multichannel (monaural, binaural) active noise control with two noise sources (270° and 90°)

signals from the right hearing aid) and compared to a binaural scheme using one microphone signal from each hearing aid. Once again, it has been chosen to limit the number of channels of the binaural scheme to two in order to have a fair comparison with the monaural schemes.

Figure 6.11 presents the SNR improvement at the left eardrum for the three schemes when the source is facing the left hearing aid (270°). In this scenario the integrated ANC and NR scheme based on the microphone signals from the left hearing aid is causal. It can therefore achieve an SNR improvement up to the causality bound. The integrated ANC and NR scheme based on the microphone signals from the right hearing aid on the other hand is non-causal and requires a causality margin $\delta \geq 8$. Below this value, the ANC is ineffective and the only SNR improvement is due to the compensation of the effects of the secondary path. The binaural scheme achieve a performance similar to the scheme based on the microphone signals from the left hearing aid and only require a positive causality margin.

The SNR power improvement at the left eardrum when the noise source is facing the right ear (90°) is shown in Figure 6.12. In this scenario the integrated ANC and NR scheme based on the microphone signals from the left hearing aid is effective for a positive causality margin whereas the schemes based on the microphone signals from the right hearing aid and the binaural scheme delivers SNR improvement for

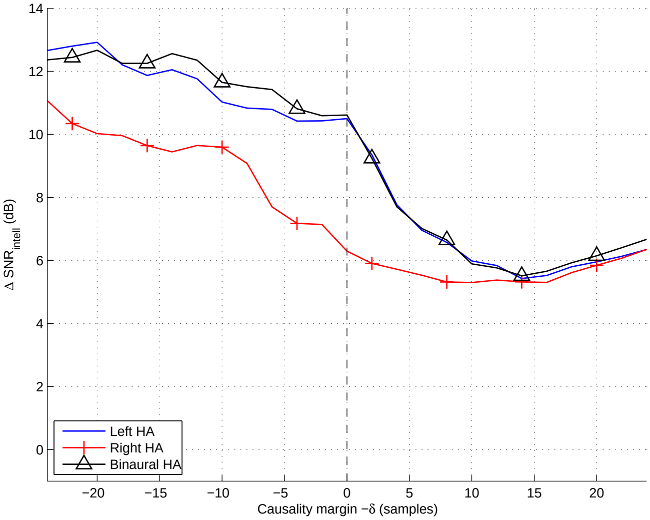


Figure 6.11: Integrated active noise control and noise reduction for multichannel (monaural, binaural) active noise control with a single noise source (270°)

a causality margin down to $\delta \geq -8$. Once again, the binaural scheme matches the SNR improvement performance of the best of the two monaural schemes (in this case, the integrated ANC and NR scheme based on the microphone signals from the right hearing aid).

Multiple noise sources

The aim of the second experiment is to evaluate the effects of the number of sound sources (speech source plus noise sources) on the performance of the integrated ANC and NR scheme. The the causality margin of the system is (artificially) set to a positive value ($\delta = 12$). When only one noise source is active, the noise source at 270° is used. Then the noise source at 90° is added for the 2 sources scenario, the noise source at 330° is added for the 3 sources scenario and finally the noise source at 30° is added for the 4 sources scenario.

It is shown in Figure 6.13 that the number of noise sources has an impact on the performance of MWF-based NR schemes when signal leakage effects and the effect of the secondary path are neglected, *i.e.*, the performance of the integrated ANC and NR scheme when the gain G is high. In order to observe the effects of the number of sound sources (speech source plus noise sources) on the ANC part of the integrated ANC and NR schemes it is more convenient to look at the normalised

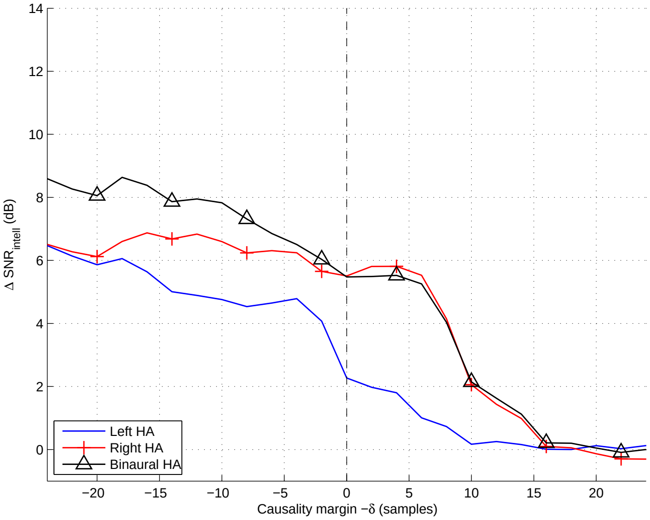


Figure 6.12: Multichannel (monaural, binaural) integrated active noise control and noise reduction with a single noise source (90°)

output SNR improvement:

$$\overline{\Delta\mathrm{SNR}_{\mathrm{intellig}}} = \frac{\Delta\mathrm{SNR}_{\mathrm{intellig}}}{\Delta\rho_{\mathrm{intellig}}} \tag{6.45}$$

Where $\Delta\rho_{\mathrm{intellig}}$ is the intelligibility-weighted SNR improvement for the MWF-based NR scheme when no perturbation (signal leakage and secondary path) is taken into account. It is defined as follows:

$$\Delta\rho_{\mathrm{intellig}} = \sum_i I_i(\rho_i - \mathrm{SNR}_{\mathrm{i,leak}}) \tag{6.46}$$

Where ρ_i represents the output SNR (in dB) of the ith one third band for the MWF-based NR scheme with no perturbation.

Figure 6.14–6.16 presents the normalised output SNR improvement ($\overline{\Delta\mathrm{SNR}_{\mathrm{intellig}}}$) of the three integrated ANC and NR schemes (2-channel monaural based on the microphone signals from the left hearing aid, 3-channel binaural (two microphone signals from the left hearing aid and one microphone signal from the right hearing aid) and 4-channel binaural), for an gain varying from 0dB to 25dB and for scenarios with 1 to 4 noise sources. For large gains, the integrated ANC and

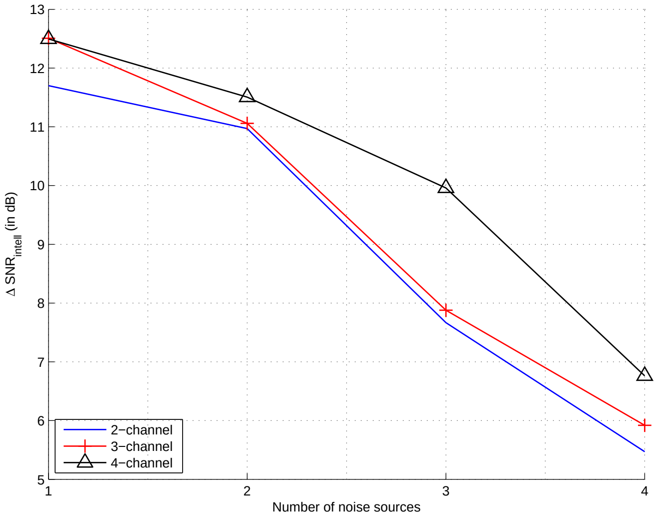


Figure 6.13: Multichannel (2-channel monaural, 3-channel binaural and 4-channel binaural) integrated active noise control and noise reduction with $G = 25\text{dB}$

NR schemes deliver a normalised output SNR ($\overline{\Delta\text{SNR}_{\text{intell}}}$) of about 1, *i.e.*, the integrated scheme delivers the same output SNR as the MWF-based NR in absence of signal leakage and secondary path effects for any number of noise sources.

The 2-channel integrated ANC and NR is able to deliver an almost constant SNR improvement for any gain when only one noise source is present. When the number of noise sources is larger, the normalised SNR performance at low gains is decreased to about 50% for any number of noise sources (Figure 6.14).

The 3-channel integrated ANC and NR scheme delivers a normalised SNR of at least 80% as long as the number of noises sources does not exceed two (Figure 6.15). Then the normalized SNR performance of the 3-channel integrated ANC and NR scheme is reduced to about 65%.

The 4-channel integrated ANC and NR scheme allows to maintain a normalised SNR performance of 80% as long as the number of sources is lower than three, then the performance is decreased to about 65% (Figure 6.16).

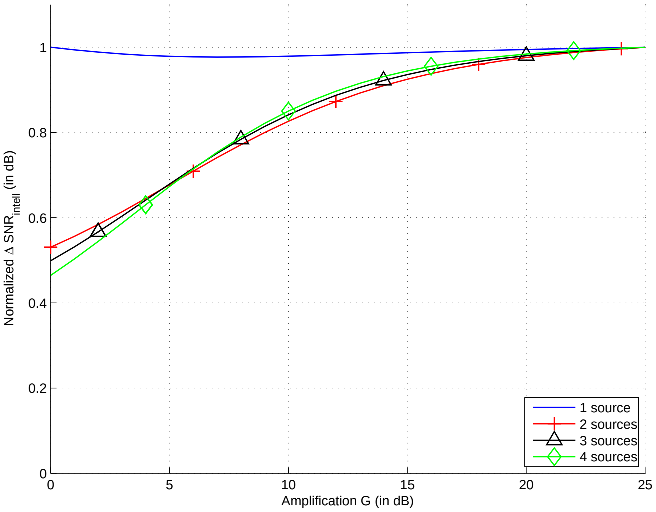


Figure 6.14: Normalised signal-to-noise ratio performance of a 2-channel monaural scheme integrated active noise control and noise reduction schemes

6.6 Conclusion

It has been shown in previous chapters that ANC provides an efficient solution to the signal leakage problem in hearing aids with an open fitting. The hearing aids, however, have small dimensions. Therefore, the integrated ANC and NR scheme is subject to strong constraints on causality and on the number of noise sources that can be compensated for.

The binaural approach to integrated ANC and NR allows to access the microphone signals from the contra-lateral ear. These microphones are distant from the ear canal microphone where the noise is to be cancelled. The propagation time from the contra-lateral microphones to the ear canal microphone is therefore larger and allows, in some scenarios, to design a scheme with a higher causality margin.

Moreover, the binaural integrated ANC and NR scheme is based on a larger number of microphone signals. Therefore, this approach allows to attenuate the noise from a larger number of sources than the monaural integrated ANC and NR schemes.

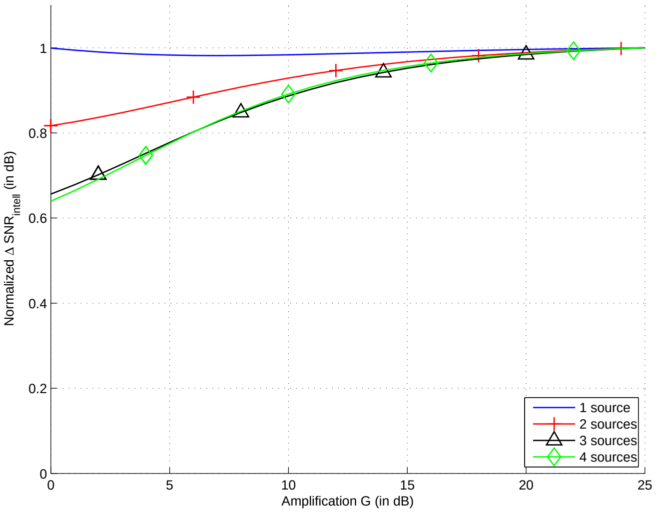


Figure 6.15: Normalised signal-to-noise ratio performance of a 3-channel binaural scheme integrated active noise control and noise reduction schemes

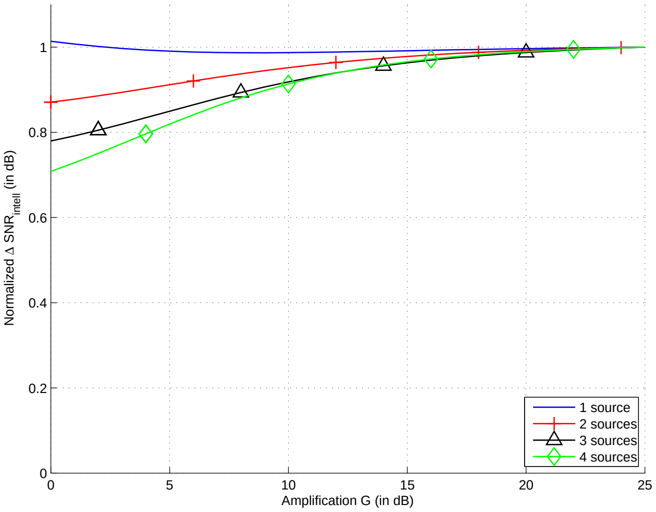


Figure 6.16: Normalised signal-to-noise ratio performance of a 4-channel binaural scheme integrated active noise control and noise reduction schemes

Chapter 7

Conclusions and Further Research

This chapter summarises conclusions of the thesis and provides some suggestions for further research.

7.1 Conclusion

A person suffering from sensorineural losses might have a great difficulty to understand speech in a noisy environment. Therefore, state-of-the-art hearing aids include NR algorithms that enhance the speech and get rid of the unwanted noise (*i.e.*, increase the SNR).

Over the past few years, the usage of hearing aids with an open fitting has become more common. Whereas removing the earmold reduces the occlusion effect and improves the physical comfort, one major drawback is that there is no earmold left to prevent the ambient sound to reach directly the eardrum. Standard NR techniques by design ignore this signal leakage and the secondary path effects. These aspects, however, can seriously degrade the NR performance and should not be neglected.

In this thesis, we have introduced ANC as a solution to the signal leakage and secondary path problem. The different ways to combine ANC and NR have been explored and an integrated ANC and NR scheme has been presented to tackle the signal leakage effects and the effects of secondary path. The original integrated ANC and NR scheme has later been refined to overcome partly some of the limitations introduced by the hearing aids framework.

In **Chapter 2** a MWF-based NR algorithm has been described. MWF-based NR algorithms represent one class of the multichannel NR algorithms which can take advantage of the spatial separation between sources as well as characteristic differences between the speech signal and the noise signals. For these reasons MWF-based NR algorithms are well suited to hearing aids framework where the noise sources are generally spatially separated from the speaker (who produces the signal of interest).

Standard NR techniques, including MWF techniques, ignore leakage and secondary path effects. In a single speech source scenario, it is possible to derive the output SNR performance of a MWF-based NR. These results have been extended to the case where signal leakage and secondary path are taken into account. The theoretical formulae allowed to verify the observations made from simulations. The degradations induced by the effects of signal leakage and the secondary path effects can seriously degrade the NR performance. Therefore, these degradations cannot be neglected, especially so when dealing hearing aids with open fittings where the signal leakage and the secondary path effects are the most significant.

It is possible to implement a feedforward ANC scheme in hearing aids, using the BTE microphones as reference microphones and an ear canal microphone as an error microphone. A multichannel ANC has been presented and has shown to allow to reduce the residual noise power caused by the leakage signal at the eardrum. The ANC, however, is highly sensitive to delays and strict causality constraints in hearings aids may limit its benefits. Finally, in the case of hearing aids performing ANC alone is not sufficient and ANC would have to be performed together with an NR.

In **Chapter 3** different ways to combine ANC and NR have been presented. The SNR performance of two algorithms (a cascaded ANC and NR scheme and an integrated ANC and NR scheme) have been analysed and compared to the performance of a MWF-based NR. 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 4dB) 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.

The integrated ANC and NR scheme has shown to improve the SNR by about 12dB for low hearing aid gains (between 0dB and 20dB), as long as the system is causal. When the system becomes non-causal, the integrated ANC and NR scheme can still outperform standard NR algorithms by taking the secondary path into account in the speech enhancement process, thereby reducing the impact of the leakage on the output signal.

Theoretical output SNR of the integrated ANC and NR has been derived for a single speech source scenario. When the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones the leakage signal can be written as a linear combination of the microphone signals. It has then been shown that the integrated ANC and NR scheme allows to restore the NR performance and delivers a constant output SNR for any amplification gain.

When the number of sound sources is larger than the number of microphones, it is possible to rewrite the leakage signal as the sum of a linear combination of the microphone signals and an estimation error. The integrated ANC and NR scheme can then compensate only for the estimated part of the leakage signal. When the number of sound sources grows so does the estimation error. The integrated ANC and NR scheme then tends to behave like an FxMWF-based NR that compensates for the secondary path effect. Therefore, it still improves the output SNR performance compared to a standard MWF-based NR.

The use of ANC in hearing aids, however, is limited by the size of the devices and the number of microphones available on each device. The number of noise sources that can be compensated for by the ANC is limited by the number of microphones available. Also the small separation between the microphones and the loudspeaker results in low acoustic propagation time. Therefore, the causality margin, which is crucial for the ANC to be effective, is very low.

Besides, the integrated ANC and NR relies on an ear canal microphone which should ideally be located at the eardrum. In practice however, the ear canal microphone is distant from the eardrum and the sound reaching the eardrum is basically unknown and uncontrolled.

Finally, the objectives of the integrated ANC and NR scheme are to attenuate the noise component of the leakage (*i.e.*, ANC) and to minimise the difference between an unknown desired speech signal and the signal delivered at the eardrum (*i.e.*, NR), but the balance between these two objectives is fixed. In some cases however, it would be useful to adjust the trade-off between ANC and NR, *e.g.*, when the input signal does not contain any speech or when the ANC is found to be inefficient.

In **Chapter 4**, the concept of weighted NR applied in the MWF framework to derive the SDW-MWF has been extended to derive two weighted versions of the integrated ANC and NR scheme based on FxMWF. The weighted integrated ANC and NR scheme then allows to emphasise either the ANC or the NR whereas the SDW-ANC/NR scheme allows to focus on reducing the SD at the eardrum or on minimising the residual noise at the eardrum.

When the signal does not contain any speech, the weighted integrated ANC and NR scheme allows to focus on ANC and minimise the power of the residual noise signal at the eardrum. On the other hand, if the ANC is found to be inefficient for the considered background noise scenario the emphasis can be put on the NR. The weighted integrated ANC and NR scheme then exhibits improved SD performance compared to an un-weighted integrated ANC and NR scheme.

This weighted ANC and NR scheme, however, does not reduce to the original un-weighted integrated ANC and NR scheme for any weighting factor. Besides, focusing on NR allows to reduce SD compared to integrating ANC and NR but NR itself is still introducing SD. A SDW-ANC/NR scheme has been derived that allows to balance between reducing the SD at the eardrum and minimising the residual noise at the eardrum (*i.e.*, ANC). In the single speech source scenario and when the number of sound sources (speech plus noise sources) is less than or equal to the number of microphones, a theoretical formula for the output SNR of the SDW-ANC/NR scheme at the eardrum has been derived. The SDW-ANC/NR scheme has then been shown to deliver an almost constant SNR at the eardrum for any weighting factor.

Chapter 5 introduced a remote-point approach and a zone-of-quiet approach both applied to the integrated ANC and NR schemes, to tackle the ear canal location problem.

The remote-point approach, based on a MSE criterion expressed at a distant point, applied to the ANC scheme and the integrated ANC and NR scheme, has been shown to allow performance improvement compared to the standard schemes. In a single speech source scenario, the remote-point integrated ANC and NR scheme has been shown to restore, at a remote point, the performance delivered by the integrated ANC and NR scheme at the ear canal microphone. It has proved however be an unsatisfying approach when the noise is to be controlled over a desired zone of quiet and not at a single point.

The zone-of-quiet approach allows to optimise the average of a particular design criterion, here the MSE criterion, over a zone of quiet generated by the ANC and thus to derive filters which are designed to control the noise over the desired zone of quiet. This approach, applied to the ANC scheme and the integrated ANC and NR scheme, has exhibited improved performance, compared against the original schemes, especially so when the noise is to be cancelled on a zone which is distant from the ear canal microphone. The zone-of-quiet based approach also allowed to reduce the impact of constructive interference over the desired zone of quiet.

The zone-of-quiet based approach has been shown to provide a more robust and more realistic way to apply ANC and solve the signal leakage problem in the framework of hearing aids with an open fitting.

Finally, **Chapter 6** presented a binaural approach to integrated ANC and NR that allows to access the microphone signals from the contra-lateral ear.

The microphones from the contra-lateral hearing aid are distant from the ear canal microphone where the noise is to be cancelled. The propagation time from these microphones to the ear canal microphone is then larger than in a monaural scheme.

Therefore, the binaural approach has been shown to allow, in some scenarios, to design a scheme with an increased causality margin.

The binaural integrated ANC and NR scheme is based on the microphone signals from both ear. Therefore, this approach also allowed to attenuate the noise from a larger number of sources than the monaural integrated ANC and NR schemes.

7.2 Suggestions for further research

The implementation of the integrated ANC and NR schemes in hearing aids comes with numerous problems principally due to the dimensions of the devices. In the case of hearing aids with an open fitting the separation between the loudspeaker and the microphone is small and there is no earmold to prevent the sound from the hearing aid loudspeaker to reach the BTE microphones. Therefore, the acoustic coupling between the loudspeaker and the BTE microphones is high and the feedback can cause important degradations to the signal. The effects of the feedback coupling on the integrated ANC and NR scheme have been neglected so far and still have to be investigated. The integrated ANC and NR could then be combined with AFC to achieve more robust processing in more realistic scenarios.

It is explained in this thesis that the ANC performance is highly constrained by the dimensions of the hearing aid devices and the small amount of microphones available. In a binaural setup, the hearing aid user is wearing two hearing aids which can communicate through a wireless link. The binaural integrated ANC and NR scheme can benefit from the extra amount of hearing aid microphones available in order to compensate for the effect of more noise sources. The ANC part of the integrated ANC and NR scheme can also benefit from the causality margin improvement owing to the location of the contra-lateral hearing aid microphones. The binaural setup could be considered as a very simple wireless acoustic sensor network (WASN) and one could imagine a scenario where the hearing aids of several users communicate together to provide better, *e.g.*, speech reference, noise references... The binaural approach to integrated ANC and NR could be extended to larger WASN. A WASN approach to the integrated ANC and NR scheme could then benefit from a larger number of microphone spread over the auditory scene to increase causality margins as well as the number of noise sources that can be compensated for.

In this thesis, the binaural approach to integrated ANC and NR has been introduced partly to overcome to some extent the causality limitations induced by the small dimensions of hearing aids. The improvement owing to the binaural integrated ANC and NR scheme cannot, however, be considered as fully satisfactory. Indeed, in this thesis the hearing aid processing delays Δ_{HA} (*i.e.*, A/D converter delays, D/A converter delays...) have been neglected so as to focus on the performance improvement owing to the integrated ANC and NR approach and all its variations. Should these functional delays be taken into account, the integrated ANC and NR scheme would need to have a causality margin by far higher than the causality margins achieved with schemes presented here. Therefore causality is still a very challenging problem in the implementation of ANC in hearing aids. In order to try to solve this causality problem, the effect of causality on the integrated ANC and NR scheme performance should be first formulated and analysed theoretically.

All the schemes introduced in this thesis, 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. The consequences of this new disturbance on the performance of the integrated ANC and NR scheme and on its stability still have to be investigated.

Finally, all the results presented in this thesis are coming from simulations run on acoustic path measurements. The integrated ANC and NR may, however, not behave exactly similarly in the “real-world”. Besides, a hardware implementation of the integrated ANC and NR schemes presented in this thesis may rise technical problems which have not been thought about yet (quantisation, fixed-point conversion. . .).

Appendix A

Appendix to Chapter 2

A.1 Single speech source MWF (2.58)

In a single speech source scenario, if the speech signal and the noise signals are uncorrelated, the correlation matrix $\mathbf{R}_X$ of the microphone signals X can be rewritten as follows:

$$\mathbf{R}_X = \mathbf{R}_{X^n} + P^s \mathbf{A}\mathbf{A}^H \tag{A.1}$$

Applying the Woodbury identity, the inverse of the matrix $\mathbf{R}_X$ is:

$$\begin{aligned} \mathbf{R}_X^{-1} &= \mathbf{R}_{X^n}^{-1} - \mathbf{R}_{X^n}^{-1}\mathbf{A}(P^{s-1} + \mathbf{A}^H \mathbf{R}_{X^n}^{-1}\mathbf{A})^{-1}\mathbf{A}^H \mathbf{R}_{X^n}^{-1} \\ &= \mathbf{R}_{X^n}^{-1} - \frac{\mathbf{R}_{X^n}^{-1} P^s \mathbf{A}\mathbf{A}^H \mathbf{R}_{X^n}^{-1}}{(1+\rho)} \end{aligned} \tag{A.2}$$

The MWF-based NR filter (2.56) can then be rewritten as follows:

$$\begin{aligned} \mathbf{W}_{\mathrm{NR}} &= \mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} - \frac{\mathbf{R}_{X^n}^{-1}\mathbf{A}P^s\mathbf{A}^H\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\rho+1}\mathbf{G}_{1,\Delta} \qquad \text{(A.3)} \\ &= \mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} - \frac{\mathbf{R}_{X^n}^{-1}P^s\mathbf{A}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}P^s\mathbf{A}\mathbf{A}^H}{\rho+1}\mathbf{G}_{1,\Delta} \\ &= \mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} - \frac{\rho\mathbf{R}_{X^n}^{-1}P^s\mathbf{A}\mathbf{A}^H}{\rho+1}\mathbf{G}_{1,\Delta} \end{aligned}$$

$$\mathbf{W}_{\mathrm{NR}} = \mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} - \frac{\rho\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\rho+1}\mathbf{G}_{1,\Delta}$$

$$\mathbf{W}_{\mathrm{NR}} = \frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\rho+1}\mathbf{G}_{1,\Delta} \tag{A.4}$$

■

A.2 MWF output SNR, single speech source case (2.61)

In the single speech source case, the output SNR of the MWF-based NR (2.56) can be expressed as follows:

$$\begin{aligned}\mathrm{SNR}_{\mathrm{NR(noLeakage)}} &= \frac{\mathbf{W}_{\mathrm{NR}}^H\mathbf{R}_{X^s}\mathbf{W}_{\mathrm{NR}}}{\mathbf{W}_{\mathrm{NR}}^H\mathbf{R}_{X^n}\mathbf{W}_{\mathrm{NR}}} \\ &= \frac{\frac{1}{(\rho+1)^2}\mathbf{G}_{1\Delta}^H\mathbf{R}_{X^s}^H\mathbf{R}_{X^n}^{-1}{}^H\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta}}{\frac{1}{(\rho+1)^2}\mathbf{G}_{1\Delta}^H\mathbf{R}_{X^s}^H\mathbf{R}_{X^n}^{-1}{}^H\mathbf{R}_{X^n}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta}}\end{aligned} \tag{A.5}$$

The autocorrelation matrices are Hermintian matrices, *i.e.*,:

$$\mathbf{R}_{X^s}^H = \mathbf{R}_{X^s}$$

$$\mathbf{R}_{X^n}^H = \mathbf{R}_{X^n}$$

The inverse of a Hermitian matrix being a Hermitian matrix:

$$\mathbf{R}_{X^n}^{-1}{}^H = \mathbf{R}_{X^n}^{-1}$$

The output SNR of the MWF-based NR filter can then be rewritten as follows:

$$\begin{aligned}\mathrm{SNR}_{\mathrm{NR(noLeakage)}} &= \frac{\mathbf{G}_{1\Delta}^H\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta}}{\mathbf{G}_{1\Delta}^H\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta}} \\ &= \frac{P^{s3}\mathbf{G}_{1\Delta}^H\mathbf{A}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}\mathbf{A}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}\mathbf{A}\mathbf{A}^H\mathbf{G}_{1,\Delta}}{P^{s2}\mathbf{G}_{1\Delta}^H\mathbf{A}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}\mathbf{A}\mathbf{A}^H\mathbf{G}_{1,\Delta}} \\ &= \frac{\rho^2P^s\mathbf{G}_{1\Delta}^H\mathbf{A}\mathbf{A}^H\mathbf{G}_{1,\Delta}}{\rho P^s\mathbf{G}_{1\Delta}^H\mathbf{A}\mathbf{A}^H\mathbf{G}_{1,\Delta}}\end{aligned} \tag{A.6}$$

$$\mathrm{SNR}_{\mathrm{NR(noLeakage)}} = \rho \tag{A.7}$$

■

A.3 MWF output SNR, single speech source case with leakage (2.69)

The output SNR for an MWF-based NR when signal leakage is present is given by the following formula:

$$\mathrm{SNR}_{\mathrm{NR(Leakage)}} = \frac{\mathbb{E}\{|\mathbf{W}_{\mathrm{NR}}^H \mathbf{X}^s + L^s|^2\}}{\mathbb{E}\{|\mathbf{W}_{\mathrm{NR}}^H \mathbf{X}^n + L^n|^2\}} \tag{A.8}$$

The power of the speech component at the eardrum is then given by:

$$\begin{aligned} P_{\tilde{Z}^s} =& \mathbb{E}\{|\mathbf{W}_{\mathrm{NR}}^H \mathbf{X}^s + L^s|^2\} \\ =& \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^s} \mathbf{W}_{\mathrm{NR}} + \mathbb{E}\{|\mathbf{W}_{\mathrm{NR}}^H \mathbf{X}^s L^{s*}|\} \\ & + \mathbb{E}\{|L^s \mathbf{X}^{sH} \mathbf{W}_{\mathrm{NR}}|^2\} + \mathbb{E}\{|L^s|^2\} \\ =& \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^s} \mathbf{W}_{\mathrm{NR}} + \mathbf{W}_{\mathrm{NR}}^H \mathbf{r}_{sl} + \mathbf{r}_{sl}^H \mathbf{W}_{\mathrm{NR}} + P_{L^s} \end{aligned} \tag{A.9}$$

Similarly, the power of the speech component at the eardrum is given by:

$$\begin{aligned} P_{\tilde{Z}^n} =& \mathbb{E}\{|\mathbf{W}_{\mathrm{NR}}^H \mathbf{X}^n + L^n|^2\} \\ =& \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^n} \mathbf{W}_{\mathrm{NR}} + \mathbb{E}\{|\mathbf{W}_{\mathrm{NR}}^H \mathbf{X}^n L^{n*}|\} \\ & + \mathbb{E}\{|L^n \mathbf{X}^{nH} \mathbf{W}_{\mathrm{NR}}|^2\} + \mathbb{E}\{|L^n|^2\} \\ =& \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^s} \mathbf{W}_{\mathrm{NR}} + \mathbf{W}_{\mathrm{NR}}^H \mathbf{r}_{nl} + \mathbf{r}_{nl}^H \mathbf{W}_{\mathrm{NR}} + P_{L^n} \end{aligned} \tag{A.10}$$

where $\mathbf{r}_{sl}$ and $\mathbf{r}_{nl}$ are the cross-correlation vectors between the speech component and the noise component, respectively of the microphone signals and the speech component of the leakage signal, and where P_{L^s} and P_{L^s} are the powers of the speech component and the noise component of the leakage signal, respectively.

$$\begin{aligned} \mathbf{r}_{sl} &= \mathbb{E}\{\mathbf{X}^s L^{s*}\} \\ \mathbf{r}_{nl} &= \mathbb{E}\{\mathbf{X}^n L^{n*}\} \\ P_{L^s} &= \mathbb{E}\{|L^s|^2\} \\ P_{L^n} &= \mathbb{E}\{|L^n|^2\} \end{aligned}$$

The leakage signal can be approximated by a linear combination of the input signals and can then be rewritten as:

$$\begin{aligned} L &= \tilde{\mathbf{P}}^H \mathbf{X} + e_L \\ \tilde{\mathbf{P}}^T &= [\tilde{P}_1 \dots \tilde{P}_M] \end{aligned} \tag{A.11}$$

where e_L is the estimation error. Here the filter $\tilde{\mathbf{P}}$ is designed to minimize the mean-square value of e_L:

$$\mathbb{E}\{|e_L|^2\} = \mathbb{E}\{|L - \tilde{\mathbf{P}}^H \mathbf{X}|^2\}$$

The estimation error e_L is then orthogonal to the microphone signals and to the microphone signals filtered by $\tilde{\mathbf{P}}$ [40]:

$$\begin{aligned} \mathbb{E}\{X e_L^H\} &= 0 \\ \mathbb{E}\{\tilde{\mathbf{P}}^H \mathbf{X} e_L^H\} &= 0 \end{aligned}$$

The cross-correlation vectors $\mathbf{r}_{sl}$ and $\mathbf{r}_{nl}$ can the be rewritten as follows:

$$\begin{aligned} \mathbf{r}_{sl} &= \mathbb{E}\{\mathbf{X}^s \mathbf{X}^{sH} \tilde{\mathbf{P}}\} + \mathbb{E}\{\mathbf{X}^s e_L^H\} \\ &= \mathbf{R}_{X^s} \tilde{\mathbf{P}} \end{aligned} \tag{A.12}$$

$$\begin{aligned} \mathbf{r}_{nl} &= \mathbb{E}\{\mathbf{X}^n \mathbf{X}^{nH} \tilde{\mathbf{P}}\} + \mathbb{E}\{\mathbf{X}^n e_L^H\} \\ &= \mathbf{R}_{X^n} \tilde{\mathbf{P}} \end{aligned} \tag{A.13}$$

The output SNR can then be rewritten as follows:

$$\mathrm{SNR}_{\mathrm{NR(Leakage)}} = \frac{\mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^s} \mathbf{W}_{\mathrm{NR}} + \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^s} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^s}{}^H \mathbf{W}_{\mathrm{NR}} + P_{L^s}}{\mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^s} \mathbf{W}_{\mathrm{NR}} + \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^n} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^n}{}^H \mathbf{W}_{\mathrm{NR}} + P_{L^n}}$$

In a similar way as in Appendix A.2, the SNR can be rewritten:

$$\mathrm{SNR}_{\mathrm{NR(Leakage)}} = \frac{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^s} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^s}{}^H \mathbf{W}_{\mathrm{NR}} + P_{L^s}}{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \mathbf{W}_{\mathrm{NR}}^H \mathbf{R}_{X^n} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^n}{}^H \mathbf{W}_{\mathrm{NR}} + P_{L^n}}$$

where $P_{D_{\mathrm{NR}}}$ is the power of the desired speech signal:

$$P_{D_{\mathrm{NR}}} = \mathbf{G}_{1,\Delta}^H \mathbf{R}_{X^s} \mathbf{G}_{1,\Delta} \tag{A.14}$$

In a single speech source scenario, the SNR expresses as follows:

$$\mathrm{SNR}_{\mathrm{NR(Leakage)}} = \frac{\frac{\rho^2 P_{D_{\mathrm{NR}}}}{(\rho+1)^2} + \frac{\mathbf{G}_{1,\Delta}{}^H \mathbf{R}_{X^s}{}^H \mathbf{R}_{X^n}^{-1}{}^H \mathbf{R}_{X^s} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^s}{}^H \mathbf{R}_{X^n}^{-1} \mathbf{R}_{X^s} \mathbf{G}_{1,\Delta}}{\rho+1} + P_{L^s}}{\frac{\rho^2 P_{D_{\mathrm{NR}}}}{(\rho+1)^2} + \frac{\mathbf{G}_{1,\Delta}{}^H \mathbf{R}_{X^s}{}^H \mathbf{R}_{X^n}^{-1}{}^H \mathbf{R}_{X^n} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^n}{}^H \mathbf{R}_{X^n}^{-1} \mathbf{R}_{X^s} \mathbf{G}_{1,\Delta}}{\rho+1} + P_{L^n}}$$

The autocorrelation matrices are Hermitian:

$$\begin{aligned}
\mathrm{SNR}_{\mathrm{NR(Leakage)}} =& \frac{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \mathbf{G}_{1,\Delta}{}^H \frac{\mathbf{R}_{X^s} \mathbf{R}_{X^n}^{-1}}{\rho+1} \mathbf{R}_{X^s} \tilde{\mathbf{P}}}{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \mathbf{G}_{1,\Delta}{}^H \frac{\mathbf{R}_{X^s}}{\rho+1} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \frac{\mathbf{R}_{X^s}}{\rho+1} \mathbf{G}_{1,\Delta} + P_{L^n}} \\
&+ \frac{\tilde{\mathbf{P}}^H \mathbf{R}_{X^s} \frac{\mathbf{R}_{X^n}^{-1} \mathbf{R}_{X^s}}{\rho+1} \mathbf{G}_{1,\Delta} + P_{L^s}}{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \mathbf{G}_{1,\Delta}{}^H \frac{\mathbf{R}_{X^s}}{\rho+1} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \frac{\mathbf{R}_{X^s}}{\rho+1} \mathbf{G}_{1,\Delta} + P_{L^n}} \\
=& \frac{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \frac{\rho}{\rho+1} (\mathbf{G}_{1,\Delta}{}^H \mathbf{R}_{X^s} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^s} \mathbf{G}_{1,\Delta}) + P_{L^s}}{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \frac{1}{\rho+1} (\mathbf{G}_{1,\Delta}{}^H \mathbf{R}_{X^s} \tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H \mathbf{R}_{X^s} \mathbf{G}_{1,\Delta}) + P_{L^n}}
\end{aligned}$$

Let α be defined as follows:

$$\alpha = \mathbf{G}_{1,\Delta}^H \mathbf{r}_{sl} + \mathbf{r}_{sl}^H \mathbf{G}_{1,\Delta} \tag{A.15}$$

The output SNR of the MWF-based NR when leakage signal is present is then given by:

$$\mathrm{SNR}_{\mathrm{NR(Leakage)}} = \frac{\frac{\rho^2}{\rho+1} P_{D_{\mathrm{NR}}} + \frac{\rho}{\rho+1} \alpha + P_{L^s}}{\frac{\rho}{\rho+1} P_{D_{\mathrm{NR}}} + \frac{1}{\rho+1} \alpha + P_{L^n}} \qquad \blacksquare \tag{A.16}$$

Appendix B

Appendix to Chapter 3

B.1 Integrated ANC and NR scheme, single speech source case ($Q \leq M$) (3.93)

When the number of sources (speech plus noise sources) is less than or equal to the number of microphones ($Q \leq M$) the leakage signal can be rewritten as a linear combination of the microphone signals:

$$\begin{aligned} L &= \mathbf{P}^H \mathbf{X} && \text{(B.1)} \\ \mathbf{P}^T &= [P_1 \dots P_M] && \text{(B.2)} \end{aligned}$$

The integrated ANC and NR filter can then be separated into the two following filters:

$$\begin{aligned} \mathbf{W}_{\mathrm{Int(Q\leq M)}} =& \frac{\hat{C}}{|\hat{C}|^2}(\mathbf{R}_{X^s} + \mathbf{R}_{X^n})^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} \\ & - \frac{\hat{C}}{|\hat{C}|^2}(\mathbf{R}_{X^s} + \mathbf{R}_{X^n})^{-1}\mathbf{R}_{X^n}\mathbf{P} \qquad \text{(B.3)} \\ =& \mathbf{U}_{\mathrm{NR}} + \mathbf{V}_{\mathrm{ANC}} \end{aligned}$$

The NR filter can be rewritten as follows (see Appendix A.1):

$$\mathbf{U}_{\mathrm{NR}} = \frac{\hat{C}}{|\hat{C}|^2}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\rho + 1}\mathbf{G}_{1,\Delta} \qquad \text{(B.4)}$$

Applying the Woodbury identity, the inverse of the matrix $\mathbf{R}_X$ is:

$$\mathbf{R}_X^{-1} = \mathbf{R}_{X^n}^{-1} - \frac{\mathbf{R}_{X^n}^{-1} P^s \mathbf{A}\mathbf{A}^H \mathbf{R}_{X^n}^{-1}}{(1+\rho)} \tag{B.5}$$

The ANC filter then reduces to:

$$\begin{aligned}\mathbf{V}_{\text{ANC}} &= -(\mathbf{R}_{X^n}^{-1} - \frac{\mathbf{R}_{X^n}^{-1} P^s \mathbf{A}\mathbf{A}^H \mathbf{R}_{X^n}^{-1}}{(1+\rho)})\mathbf{R}_{X^n}\mathbf{P} \\ &= \frac{\hat{C}}{|\hat{C}|^2}[\frac{\mathbf{R}_{X^n}^{-1} P^s \mathbf{A}\mathbf{A}^H}{1+\rho}\mathbf{P} - \mathbf{P}] \\ &= \frac{\hat{C}}{|\hat{C}|^2}[\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{P} - \mathbf{P}] \end{aligned} \tag{B.6}$$

Finally, under a single speech source and when the number of sound sources (speech source plus noise sources) is less or equal than the number of microphones ($Q \leq M$), the integrated ANC and NR filter is given as follows:

$$\mathbf{W}_{\text{Int(Q}\leq\text{M)}} = \frac{\hat{C}}{|\hat{C}|^2}[\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}(\mathbf{G}_{1,\Delta} + \mathbf{P}) - \mathbf{P}] \qquad \blacksquare \tag{B.7}$$

B.2 Output SNR for the integrated ANC and NR scheme, single speech source case ($Q \leq M$) (3.100)

The output SNR for the integrated ANC and NR scheme is given by the following formula:

$$\text{SNR}_{\text{Int(Q}\leq\text{M)}} = \frac{\mathbb{E}\{|C\mathbf{W}_{\text{Int(Q}\leq\text{M)}}^H \mathbf{X}^s + L^s|^2\}}{\mathbb{E}\{|C\mathbf{W}_{\text{Int(Q}\leq\text{M)}}^H \mathbf{X}^n + L^n|^2\}} \tag{B.8}$$

$$\tag{B.9}$$

In the single speech source case, when the number of sound sources (speech source plus noise sources) is less than or equal to the number of microphones, the integrated ANC and NR filter can be written as follows:

$$\mathbf{W}_{\text{Int(Q}\leq\text{M)}} = \frac{\hat{C}}{|\hat{C}|^2}(\mathbf{W}_1 - \mathbf{P}) \tag{B.10}$$

The power of the speech component at the eardrum is then given by:

$$\begin{aligned}
P_{\tilde{Z}^s} =&\mathbb{E}\{|C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^{H}\mathbf{X}^s + L^s|^2\}\\
=&C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^{H}\mathbf{R}_{X^s}\mathbf{W}_{\mathrm{Int(Q\leq M)}}C\\
&+ C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^{H}\mathbf{R}_{X^s}\mathbf{P} + \mathbf{P}^H\mathbf{R}_{X^s}{}^H\mathbf{W}_{\mathrm{Int(Q\leq M)}}C^*\\
&+ P_{L^s}\\
=&\mathbf{P}^H\mathbf{R}_{X^s}\mathbf{P} - \mathbf{P}^H\mathbf{R}_{X^s}\mathbf{W}_1 - \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{P} + \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{W}_1\\
&+ \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{P} - \mathbf{P}^H\mathbf{R}_{X^s}\mathbf{P} + \mathbf{P}^H\mathbf{R}_{X^s}\mathbf{W}_1 - \mathbf{P}^H\mathbf{R}_{X^s}\mathbf{P}\\
&+ P_{L^s} \qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad \text{(B.11)}\\
=&\mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{W}_1\\
=&\mathbf{P}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{P} + \mathbf{G}_{1,\Delta}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{G}_{1,\Delta}\\
&+ \mathbf{P}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{G}_{1,\Delta}\\
&+ \mathbf{G}_{1,\Delta}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{P}\\
P_{\tilde{Z}^s} =&\frac{\rho^2}{(\rho+1)^2}\left(P_{L^s} + \mathbf{P}^H\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{G}_{1,\Delta}^H\mathbf{R}_{X^s}\mathbf{P} + P_{D_{\mathrm{NR}}}\right) \qquad\qquad \text{(B.12)}
\end{aligned}$$

Similarly, the power of the noise component at the eardrum is given by:

$$\begin{aligned}
P_{\tilde{Z}^n} =&\mathbb{E}\{|C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^{H}\mathbf{X}^n + L^n|^2\}\\
=&C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^{H}\mathbf{R}_{X^s}\mathbf{W}_{\mathrm{Int(Q\leq M)}}C^*\\
&+ C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^{H}\mathbf{R}_{X^n}\mathbf{P} + \mathbf{P}^H\mathbf{R}_{X^n}{}^H\mathbf{W}_{\mathrm{Int(Q\leq M)}}C^*\\
&+ P_{L^n}\\
=&\mathbf{P}^H\mathbf{R}_{X^n}\mathbf{P} - \mathbf{P}^H\mathbf{R}_{X^n}\mathbf{W}_1 - \mathbf{W}_1^H\mathbf{R}_{X^n}\mathbf{P} + \mathbf{W}_1^H\mathbf{R}_{X^n}\mathbf{W}_1\\
&+ \mathbf{W}_1^H\mathbf{R}_{X^n}\mathbf{P} - \mathbf{P}^H\mathbf{R}_{X^n}\mathbf{P} + \mathbf{P}^H\mathbf{R}_{X^n}\mathbf{W}_1 - \mathbf{P}^H\mathbf{R}_{X^n}\mathbf{P}\\
&+ P_{L^n} \qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad\qquad \text{(B.13)}
\end{aligned}$$

$$
\begin{aligned}
P_{\tilde{Z}^n} =& \mathbf{W}_1^H \mathbf{R}_{X^n} \mathbf{W}_1 \\
=& \mathbf{P}^H \frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho} \mathbf{R}_{X^n} \frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho} \mathbf{P} + \mathbf{G}_{1,\Delta}^H \frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho} \mathbf{R}_{X^n} \frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho} \mathbf{G}_{1,\Delta} \\
& + \mathbf{P}^H \frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho} \mathbf{R}_{X^n} \frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho} \mathbf{G}_{1,\Delta} \\
& + \mathbf{G}_{1,\Delta}^H \frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho} \mathbf{R}_{X^n} \frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho} \mathbf{P} \\
=& \frac{1}{(\rho+1)^2} (\mathbf{P}^H \mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{P} + \mathbf{G}_{1,\Delta}^H \mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} \\
& + \mathbf{P}^H \mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{G}_{1,\Delta}^H \mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{P}) \\
P_{\tilde{Z}^n} =& \frac{\rho}{(\rho+1)^2} (P_{L^s} + \mathbf{P}^H \mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{G}_{1,\Delta}^H \mathbf{R}_{X^s}\mathbf{P} + P_{D_{\mathrm{NR}}}) \qquad \text{(B.14)}
\end{aligned}
$$

Finally, combining (B.12) and (B.14), the output SNR of the integrated ANC and NR scheme in a single speech source scenario and when the number of sound sources (speech source plus noise sources) is less than or equal to the number of microphones is given by:

$$
\begin{aligned}
\mathrm{SNR}_{\mathrm{Int(Q\leq M)}} &= \frac{\frac{\rho^2}{(\rho+1)^2}(P_{L^s} + \alpha + P_{D_{\mathrm{NR}}})}{\frac{\rho}{(\rho+1)^2}(P_{L^s} + \alpha + P_{D_{\mathrm{NR}}})} \\
\mathrm{SNR}_{\mathrm{Int(Q\leq M)}} &= \frac{\frac{\rho^2}{(\rho+1)^2}}{\frac{\rho}{(\rho+1)^2}} \\
\mathrm{SNR}_{\mathrm{Int(Q\leq M)}} &= \rho \qquad \blacksquare \qquad \text{(B.15)}
\end{aligned}
$$

B.3 Output SNR for the integrated ANC and NR scheme, single speech source case ($Q > M$) (3.110)

The output SNR for the integrated ANC and NR scheme is given by the following formula:

$$
\mathrm{SNR}_{\mathrm{Int(Q>M)}} = \frac{\mathbb{E}\{|C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^H \mathbf{X}^s + L^s|^2\}}{\mathbb{E}\{|C\mathbf{W}_{\mathrm{Int(Q\leq M)}}^H \mathbf{X}^n + L^n|^2\}} \qquad \text{(B.16)}
$$

When the number of sound sources (speech source plus noise sources) is larger than the number of microphones, the leakage signal can be approximated by a linear combination of the input signals and can then be rewritten as:

$$\begin{aligned} L &= \tilde{\mathbf{P}}^H \mathbf{X} + e_L \\ L^s &= \tilde{\mathbf{P}}^H \mathbf{X}^s + e_{L^s} \\ L^n &= \tilde{\mathbf{P}}^H \mathbf{X}^n + e_{L^n} \end{aligned}$$

where e_L is the estimation error and e_{L^s} and e_{L^n} are the approximation error speech component and noise component respectively.

The estimation error e_L, its speech component e_{L^s} and its noise component e_{L^s} are orthogonal to the microphone signals and to the microphone signals filtered by $\tilde{\mathbf{P}}$ [40]:

$$\begin{aligned} \mathbb{E}\{X e_L^H\} &= 0 \\ \mathbb{E}\{X e_{L^s}^H\} &= 0 \\ \mathbb{E}\{X e_{L^n}^H\} &= 0 \\ \mathbb{E}\{\tilde{\mathbf{P}}^H \mathbf{X} e_L^H\} &= 0 \\ \mathbb{E}\{\tilde{\mathbf{P}}^H \mathbf{X} e_{L^s}^H\} &= 0 \\ \mathbb{E}\{\tilde{\mathbf{P}}^H \mathbf{X} e_{L^n}^H\} &= 0 \end{aligned}$$

If the filter $\mathbf{W}$ varies slowly, e_L, e_{L^s} and e_{L^n} are also orthogonal to the microphone signals filtered by $\mathbf{W}$:

$$\mathbb{E}\{\mathbf{W}^H \mathbf{X} e_L^H\} = 0$$

$$\mathbb{E}\{\mathbf{W}^H \mathbf{X} e_{L^s}^H\} = 0$$

$$\mathbb{E}\{\mathbf{W}^H \mathbf{X} e_{L^n}^H\} = 0$$

The output SNR of the integrated ANC and NR scheme can then be rewritten as follows:

$$\mathrm{SNR}_{\mathrm{Int(Q>M)}} = \frac{\mathbb{E}\{|C\mathbf{W}^H_{\mathrm{Int(Q\,M)}}\mathbf{X}^s + \tilde{\mathbf{P}}^H \mathbf{X}^s + e_{L^s}|^2\}}{\mathbb{E}\{|C\mathbf{W}^H_{\mathrm{Int(Q\leq M)}}\mathbf{X}^n + \tilde{\mathbf{P}}^H \mathbf{X}^n + e_{L^n}|^2\}} \tag{B.17}$$

The power of the speech component at the eardrum is then given by:

$$
\begin{aligned}
P_{\tilde{Z}^s} =& \mathbb{E}\{|C\mathbf{W}^H_{\text{Int(Q>M)}}\mathbf{X}^s + L^s|^2\} \\
=& C\mathbf{W}^H_{\text{Int(Q>M)}}\mathbf{R}_{X^s}\mathbf{W}_{\text{Int(Q>M)}}C^* \\
& + C\mathbf{W}^H_{\text{Int(Q>M)}}\mathbf{R}_{X^s}\tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}{}^H\mathbf{W}_{\text{Int(Q>M)}}C^* + \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + E_{e_L^s} \\
& \left.\begin{array}{l}
+\mathbb{E}\{C\mathbf{W}^H_{\text{Int(Q>M)}}\mathbf{X}^s e_L^{s\,H}\} + \mathbb{E}\{e_L^s\mathbf{X}^{sH}\mathbf{W}_{\text{Int(Q>M)}}C^*\} \\
+\mathbb{E}\{\tilde{\mathbf{P}}^H\mathbf{X}^s e_L^{s\,H}\} + \mathbb{E}\{e_L^s\mathbf{X}^{sH}\tilde{\mathbf{P}}\}\}
\end{array}\right\} = 0
\end{aligned}
$$

where $E_{e_L^s}$ is the energy of the speech component in the estimation error e_L:

$$
E_{e_L^s} \quad = \quad \mathbb{E}\{|e_L^s|^2\}
$$

In the single speech source case, when the number of sound sources (speech source plus noise sources) is less than or equal to the number of microphones, the integrated ANC and NR filter can be written as follows:

$$
\mathbf{W}_{\text{Int(Q>M)}} = \frac{\hat{C}}{|\hat{C}|^2}(\mathbf{W}_2 - \tilde{\mathbf{P}}) \tag{B.18}
$$

The power of the speech component at the eardrum can then be rewritten as follows:

$$
\begin{aligned}
P_{\tilde{Z}^s} =& \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} - \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\mathbf{W}_2 - \mathbf{W}_2^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + \mathbf{W}_2^H\mathbf{R}_{X^s}\mathbf{W}_2 \\
& + \mathbf{W}_2^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} - \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\mathbf{W}_2 - \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} \\
& + \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + E_{e_L^s} \qquad\qquad (\text{B.19}) \\
=& \mathbf{W}_2^H\mathbf{R}_{X^s}\mathbf{W}_2 + E_{e_L^s} \\
=& \tilde{\mathbf{P}}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\tilde{\mathbf{P}} + \mathbf{G}_{1,\Delta}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{G}_{1,\Delta} \\
& + \tilde{\mathbf{P}}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{G}_{1,\Delta} \\
& + \mathbf{G}_{1,\Delta}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^s}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\tilde{\mathbf{P}} \\
& + E_{e_L^s}
\end{aligned}
$$

$$P_{\tilde{Z}^s} = \frac{\rho^2}{(\rho+1)^2}(\tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{G}_{1,\Delta}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + P_{D_{\mathrm{NR}}}) + E_{e_L^s} \tag{B.20}$$

Similarly, the power of the noise component at the eardrum is given by:

$$\begin{aligned} P_{\tilde{Z}^n} =& \tilde{\mathbf{P}}^H\mathbf{R}_{X^n}\tilde{\mathbf{P}} - \tilde{\mathbf{P}}^H\mathbf{R}_{X^n}\mathbf{W}_2 - \mathbf{W}_2^H\mathbf{R}_{X^n}\tilde{\mathbf{P}} + \mathbf{W}_2^H\mathbf{R}_{X^n}\mathbf{W}_2 \\ &+ \mathbf{W}_2^H\mathbf{R}_{X^n}\tilde{\mathbf{P}} - \tilde{\mathbf{P}}^H\mathbf{R}_{X^n}\tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H\mathbf{R}_{X^n}\mathbf{W}_2 - \tilde{\mathbf{P}}^H\mathbf{R}_{X^n}\tilde{\mathbf{P}} \\ &+ \tilde{\mathbf{P}}^H\mathbf{R}_{X^n}\tilde{\mathbf{P}} + E_{e_L^n} \qquad\qquad (\text{B.21}) \\ =& \mathbf{W}_2^H\mathbf{R}_{X^n}\mathbf{W}_2 + E_{e_L^n} \end{aligned}$$

$$\begin{aligned} P_{\tilde{Z}^n} =& \tilde{\mathbf{P}}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^n}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\tilde{\mathbf{P}} + \mathbf{G}_{1,\Delta}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^n}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{G}_{1,\Delta} \\ &+ \tilde{\mathbf{P}}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^n}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{G}_{1,\Delta} \\ &+ \mathbf{G}_{1,\Delta}^H\frac{\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}}{1+\rho}\mathbf{R}_{X^n}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\tilde{\mathbf{P}} \\ &+ E_{e_L^n} \\ =& \frac{1}{(\rho+1)^2}(\tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\tilde{\mathbf{P}} + \mathbf{G}_{1,\Delta}^H\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} \\ &+ \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{G}_{1,\Delta}^H\mathbf{R}_{X^s}\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}\tilde{\mathbf{P}}) + E_{e_L^n} \end{aligned}$$

$$P_{\tilde{Z}^n} = \frac{\rho^2}{(\rho+1)^2}(\tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{G}_{1,\Delta}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} + P_{D_{\mathrm{NR}}}) + E_{e_L^s} \tag{B.22}$$

Let β be defined as follows:

$$\beta \triangleq \tilde{\mathbf{P}}^H\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{G}_{1,\Delta}^H\mathbf{R}_{X^s}\tilde{\mathbf{P}} \tag{B.23}$$

Combining (B.20) and (B.22), the output SNR of the integrated ANC and NR scheme in the single speech source case and when the number of sound sources (speech source plus noise sources) is less than or equal to the number of microphones is given by:

$$\mathrm{SNR}_{\mathrm{Int(Q\leq M)}} = \frac{\frac{\rho^2}{(\rho+1)^2}(P_{D_{NR}} + \beta + \tilde{\mathbf{P}}^H R_s \tilde{\mathbf{P}}) + E_{e^s_L}}{\frac{\rho}{(\rho+1)^2}(P_{D_{NR}} + \beta + \tilde{\mathbf{P}}^H R_s \tilde{\mathbf{P}}) + E_{e^n_L}} \qquad \blacksquare \tag{B.24}$$

Appendix C

Appendix to Chapter 4

C.1 Weighted integrated ANC and NR scheme, single speech source case (4.24)

The weighted integrated ANC and NR filter can be separated into the two following filters:

$$\begin{aligned}\mathbf{W}_\mu =&\frac{\hat{C}}{|\hat{C}|^2}(\mathbf{R}_{X^s}+\nu\mathbf{R}_{X^n})^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta}\\ &-\eta\nu\frac{\hat{C}}{|\hat{C}|^2}(\mathbf{R}_{X^s}+\nu\mathbf{R}_{X^n})^{-1}\mathbf{R}_{X^n}\mathbf{P} \qquad \text{(C.1)}\\ =&\mathbf{U}_{\mu,\text{NR}}+\mathbf{V}_{\mu,\text{ANC}}\end{aligned}$$

Applying the Woodbury identity, the inverse of the pencil matrix $\mathbf{R}_{X^s}+\nu\mathbf{R}_{X^n}$ leads to:

$$\begin{aligned}(\mathbf{R}_{X^s}+\nu(\mathbf{R}_{X^n})^{-1} &= \frac{1}{\nu}(\mathbf{R}_{X^n}^{-1}-\frac{1}{\nu}\mathbf{R}_{X^n}^{-1}\mathbf{A}(P^{s\,-1}+\frac{1}{\nu}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}\mathbf{A})^{-1}\mathbf{A}^H\mathbf{R}_{X^n}^{-1})\\ &= \frac{1}{\nu}(\mathbf{R}_{X^n}^{-1}-\frac{\mathbf{R}_{X^n}^{-1}P^s\mathbf{A}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}}{(\nu+\rho)}) \qquad \text{(C.2)}\end{aligned}$$

The NR filter can be rewritten as follows (see Appendix A.1):

$$\mathbf{U}_{\mu,\text{NR}} = \frac{\hat{C}}{|\hat{C}|^2}\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\rho+\nu}\mathbf{G}_{1,\Delta} \qquad \text{(C.3)}$$

The ANC filter then reduces to:

$$\begin{aligned}\mathbf{V}_{\mu,\mathrm{ANC}} &= -\frac{1}{\nu}(\mathbf{R}_{X^n}^{-1} - \frac{\mathbf{R}_{X^n}^{-1} P^s \mathbf{A}\mathbf{A}^H \mathbf{R}_{X^n}^{-1}}{(\nu+\rho)})\nu\eta\mathbf{R}_{X^n}\mathbf{P} \\ &= \frac{\hat{C}}{|\hat{C}|^2}[\eta\frac{\mathbf{R}_{X^n}^{-1} P^s \mathbf{A}\mathbf{A}^H}{1+\rho}\mathbf{P} - \eta\mathbf{P}] \\ &= \frac{\hat{C}}{|\hat{C}|^2}[\eta\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{1+\rho}\mathbf{P} - \eta\mathbf{P}] \qquad \text{(C.4)}\end{aligned}$$

Finally, based on the previous expressions for the NR filter (C.2) and the ANC filter (C.4), in the single speech source case, the weighted integrated ANC and NR filter is given as follow:

$$\mathbf{W}_{\mu} = \frac{\hat{C}}{|\hat{C}|^2}[\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\nu+\rho}(\mathbf{G}_{1,\Delta} + \eta\tilde{\mathbf{P}}) - \eta\tilde{\mathbf{P}}] \qquad \blacksquare \qquad \text{(C.5)}$$

C.2 SDW-ANC/NR scheme, single speech source case (4.43)

The weighted integrated ANC and NR filter can be separated into the two following filters:

$$\begin{aligned}\mathbf{W}_{\mu} =& \frac{\hat{C}}{|\hat{C}|^2}(\mathbf{R}_{X^s} + \nu_1\mathbf{R}_{X^n})^{-1}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} \\ & - \nu_1\frac{\hat{C}}{|\hat{C}|^2}(\mathbf{R}_{X^s} + \nu_1\mathbf{R}_{X^n})^{-1}\mathbf{R}_{X^n}\mathbf{P} \qquad \text{(C.6)}\end{aligned}$$

where

$$\nu_1 = \frac{1}{\mu} \qquad \text{(C.7)}$$

Applying the Woodbury identity, the inverse of the pencil matrix $\mathbf{R}_{X^s} + \nu\mathbf{R}_{X^n}$ leads to:

$$\begin{aligned}(\mathbf{R}_{X^s} + \nu(\mathbf{R}_{X^n})^{-1} &= \frac{1}{\nu_1}(\mathbf{R}_{X^n}^{-1} - \frac{1}{\nu_1}\mathbf{R}_{X^n}^{-1}\mathbf{A}(P^{s\,-1} + \frac{1}{\nu_1}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}\mathbf{A})^{-1}\mathbf{A}^H\mathbf{R}_{X^n}^{-1}) \\ &= \frac{1}{\nu_1}(\mathbf{R}_{X^n}^{-1} - \frac{\mathbf{R}_{X^n}^{-1} P^s \mathbf{A}\mathbf{A}^H \mathbf{R}_{X^n}^{-1}}{(\nu_1+\rho)}) \qquad \text{(C.8)}\end{aligned}$$

In the single speech source case, the SDW-ANC/NR filter expresses as follows:

$$\mathbf{W}_{\mu} = \frac{\hat{C}}{|\hat{C}|^2}[\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\nu_1 + \rho}(\mathbf{G}_{1,\Delta} + \tilde{\mathbf{P}}) - \tilde{\mathbf{P}}] \tag{C.9}$$

$$\mathbf{W}_{\mu} = \frac{\hat{C}}{|\hat{C}|^2}[\frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\frac{1}{\mu} + \rho}(\mathbf{G}_{1,\Delta} + \tilde{\mathbf{P}}) - \tilde{\mathbf{P}}] \tag{C.10}$$

$$\blacksquare \tag{C.11}$$

C.3 Output SNR for the SDW-ANC/NR scheme, single speech source case ($Q \leq M$) (4.46)

The output SNR for the SDW-ANC/NR scheme is given by the following formula:

$$\mathrm{SNR}_{\mathrm{Int(Q\leq M)}}(\mu) = \frac{\mathbb{E}\{|C\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}^H\mathbf{X}^s + L^s|^2\}}{\mathbb{E}\{|C\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}^H\mathbf{X}^n + L^n|^2\}} \tag{C.12}$$

$$\tag{C.13}$$

In a single speech source scenario, when the number of sound sources (speech source plus noise sources) is less than or equal to than the number of microphones, the SDW-ANC/NR filter can be written as follows:

$$\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}} = \frac{\hat{C}}{|\hat{C}|^2}(\mathbf{W}_{\mu,1} - \mathbf{P})\mathbf{W}_{\mu,1} = \frac{\mathbf{R}_{X^n}^{-1}\mathbf{R}_{X^s}}{\nu_1 + \rho}(\mathbf{G}_{1,\Delta} + \tilde{\mathbf{P}}) \tag{C.14}$$

The power of the speech component at the eardrum is then given by:

$$\begin{aligned} P_{\tilde{Z}^s} =& \mathbb{E}\{|C\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}^H\mathbf{X}^s + L^s|^2\} \\ =& C\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}^H\mathbf{R}_{X^s}\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}C^* \\ &+ C\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}^H\mathbf{R}_{X^s}\mathbf{P} + \mathbf{P}^H\mathbf{R}_{X^s}{}^H\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}C^* \\ &+ P_{L^s} \end{aligned}$$

$$\begin{aligned}
P_{\tilde{Z}^s} =& \mathbf{P}^H \mathbf{R}_{X^s} \mathbf{P} - [\mathbf{P}^H \mathbf{R}_{X^s} \mathbf{W}_1 + \mathbf{W}_1^H \mathbf{R}_{X^s} \mathbf{P}] + \mathbf{W}_1^H \mathbf{R}_{X^s} \mathbf{W}_1 \\
&+ \mathbf{W}_1^H \mathbf{R}_{X^s} \mathbf{P} - \mathbf{P}^H \mathbf{R}_{X^s} \mathbf{P} + \mathbf{P}^H \mathbf{R}_{X^s} \mathbf{W}_1 - \mathbf{P}^H \mathbf{R}_{X^s} \mathbf{P} \\
&+ P_{L^s} \qquad \text{(C.15)} \\
=& \mathbf{W}_1^H \mathbf{R}_{X^s} \mathbf{W}_1 \qquad \text{(C.16)} \\
=& \frac{\rho^2}{(\nu + \rho)^2} (P_{D_{\mathrm{NR}}} + \alpha + P_{L^s}) \qquad \text{(C.17)}
\end{aligned}$$

Similarly, the power of the noise component at the eardrum is given by:

$$\begin{aligned}
P_{\tilde{Z}^n} =& \mathbb{E}\{|C \mathbf{W}_{\mu,\mathrm{Int(Q \leq M)}}^H \mathbf{X}^n + L^n|^2\} \\
=& C \mathbf{W}_{\mu,\mathrm{Int(Q \leq M)}}^H \mathbf{R}_{X^n} \mathbf{W}_{\mu,\mathrm{Int(Q \leq M)}} C^* \\
&+ C \mathbf{W}_{\mu,\mathrm{Int(Q \leq M)}}^H \mathbf{R}_{X^n} \mathbf{P} + \mathbf{P}^H \mathbf{R}_{X^n}{}^H \mathbf{W}_{\mu,\mathrm{Int(Q \leq M)}} C^* \\
&+ P_{L^n} \\
=& \mathbf{P}^H \mathbf{R}_{X^n} \mathbf{P} - [\mathbf{P}^H \mathbf{R}_{X^n} \mathbf{W}_1 + \mathbf{W}_1^H \mathbf{R}_{X^n} \mathbf{P}] + \mathbf{W}_1^H \mathbf{R}_{X^n} \mathbf{W}_1 \\
&+ \mathbf{W}_1^H \mathbf{R}_{X^n} \mathbf{P} - \mathbf{P}^H \mathbf{R}_{X^n} \mathbf{P} + \mathbf{P}^H \mathbf{R}_{X^n} \mathbf{W}_1 - \mathbf{P}^H \mathbf{R}_{X^n} \mathbf{P} \\
&+ P_{L^n} \qquad \text{(C.18)} \\
=& \mathbf{W}_1^H \mathbf{R}_{X^n} \mathbf{W}_1 \qquad \text{(C.19)} \\
=& \frac{\rho}{(\nu + \rho)^2} (P_{D_{\mathrm{NR}}} + \alpha + P_{L^s}) \qquad \text{(C.20)}
\end{aligned}$$

Combining (C.17) and (C.20), the output SNR of the SDW-ANC/NR scheme in the single speech source case and when the number of sound sources (speech source plus noise sources) is less than or equal to the number of microphones is given by:

$$\mathrm{SNR}_{\mu,\mathrm{Int(Q \leq M)}} = \frac{\rho^2 (P_{D_{\mathrm{NR}}} + \alpha + P_{L^s})}{\rho (P_{D_{\mathrm{NR}}} + \alpha + P_{L^s})} = \rho$$

$$\blacksquare \qquad \text{(C.21)}$$

C.4 Speech distortion for the SDW-ANC/NR scheme, single speech source case ($Q \leq M$) (4.47)

The SD for the SDW-ANC/NR scheme is given by the following formula:

$$\mathrm{SNR}_{\mathrm{Int(Q\leq M)}}(\mu) = \mathbb{E}\{|C\mathbf{W}^H_{\mu,\mathrm{Int(Q\leq M)}}\mathbf{X}^s - D_{\mathrm{NR}}|^2\} \tag{C.22}$$

In the single speech source case, when the number of sound sources (speech source plus noise sources) is less than or equal to the number of microphones, the SDW-ANC/NR filter can be written as in (4.43). The SD at the eardrum is then given by:

$$\begin{aligned}
P_{\tilde{Z}^s} =& \mathbb{E}\{|C\mathbf{W}^H_{\mu,\mathrm{Int(Q\leq M)}}\mathbf{X}^s + D_{\mathrm{NR}}|^2\} \\
=& C\mathbf{W}^H_{\mu,\mathrm{Int(Q\leq M)}}\mathbf{R}_{X^s}\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}C^* \\
& - C\mathbf{W}^H_{\mu,\mathrm{Int(Q\leq M)}}\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} - \mathbf{G}^H\mathbf{R}_{X^s}{}^H\mathbf{W}_{\mu,\mathrm{Int(Q\leq M)}}C^* \\
& + P_{D_{\mathrm{NR}}} \\
=& \mathbf{P}^H\mathbf{R}_{X^s}\mathbf{P} - [\mathbf{P}^H\mathbf{R}_{X^s}\mathbf{W}_1 + \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{P}] + \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{W}_1 \\
& - \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} + \mathbf{P}^H\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} - \mathbf{G}_{1,\Delta}{}^H\mathbf{R}_{X^s}\mathbf{W}_1 + \mathbf{G}_{1,\Delta}{}^H\mathbf{R}_{X^s}\mathbf{P} \\
& + P_{D_{\mathrm{NR}}}
\end{aligned} \tag{C.23}$$

$$\begin{aligned}
=& P_{L_s} - [\mathbf{P}^H\mathbf{R}_{X^s}\mathbf{W}_1 + \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{P}] + \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{W}_1 \\
& - \mathbf{W}_1^H\mathbf{R}_{X^s}\mathbf{G}_{1,\Delta} - \mathbf{G}_{1,\Delta}{}^H\mathbf{R}_{X^s}\mathbf{W}_1 + \alpha + P_{D_{\mathrm{NR}}}
\end{aligned} \tag{C.24}$$

$$\begin{aligned}
=& \frac{\rho^2}{(\rho+\nu_1)^2}(P_{D_{\mathrm{NR}}} + \alpha + P_{L_s}) \\
& - \frac{\rho}{(\rho+\nu_1)}(2P_{D_{\mathrm{NR}}} + 2\alpha + 2P_{L_s}) + P_{D_{\mathrm{NR}}} + \alpha + P_{L_s}
\end{aligned} \tag{C.25}$$

$$= (P_{D_{\mathrm{NR}}} + \alpha + P_{L_s})\frac{\rho^2 - \rho(\rho+\nu_1) + (\rho+\nu_1)^2}{(\rho+\nu_1)^2} \tag{C.26}$$

$$= (P_{D_{\mathrm{NR}}} + \alpha + P_{L_s})\frac{\nu_1^2}{(\rho+\nu_1)^2} \tag{C.27}$$

$$= (P_{D_{\mathrm{NR}}} + \alpha + P_{L_s})\frac{1}{(\rho\mu+1)^2} \qquad \blacksquare \tag{C.28}$$

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Publication List

International Journal Papers

1. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., "Integrated active noise control and noise reduction in hearing aids.",*IEEE Transactions on Speech Audio and Language*, 18(6):1137–1146, August 2010.

2. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., "Output snr analysis of integrated active noise control and noise reduction in hearing aids under a single speech source.", Accepted for publication in*EURASIP Signal Processing*, February 2011.

3. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., "A zone of quiet based approach to integrated active noise control and noise reduction for speech enhancement in hearing aids.", Submitted to *IEEE Transactions on Speech Audio and Language*, February 2011.
4. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., "Binaural integrated active noise control and noise reduction in hearing aids.", to be Submitted, KULeuven, 2011.
5. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., "Speech Distortion Weighted Integrated Active Noise Control and Noise Reduction in Hearing Aids.", to be Submitted, KULeuven, 2011.

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1. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., "Combined active noise control and noise reduction in hearing aids.", In *11th International Workshop on Acoustic Echo and Noise Control (IWAENC)*, Seattle (WA),

United States, September 2008.

2. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., “A weighted approach integrated active noise control and noise reduction in hearing aids.”, In *The 17th European Signal Processing Conference (EUSIPCO-2009)*, Glasgow, Scotland, August 2009.

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4. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., “Output SNR analysis of integrated active noise control and noise reduction in hearing aids for a single speech source scenario.”, In *The 18th European Signal Processing Conference (EUSIPCO-2010)*, Aalborg, Denmark, August 2010.

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1. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., “Signal processing challenges in hearing aids with open fittings.”, In *The IEEE Benelux Signal Processing Symposium (BSPS-2010*), Delft, The Netherlands, April 2010.

2. R. Serizel, M. Moonen, J. Wouters, and S. H. Jensen., “Active Noise Control in Hearing Aids.”, In IAP VI/4-DYSCO Study Day, Gent, Belgium, May 2010, p. 14.

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Curriculum vitae

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