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TOWARDS AN ACTIVE HEARING PROTECTION DEVICE FOR MUSICIANS

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TOWARDS AN ACTIVE HEARING PROTECTION DEVICE FOR MUSICIANS

Antoine BERNIER

ABSTRACT

Professional musicians are oftentimes exposed to high levels of sound. Prolonged or severe exposure to high sound levels could lead to permanent hearing loss and compromise their career. The logical solution would be to wear hearing protection devices (HPDs) when appropriate. However, perceptual discomfort associated with wearing HPD can discourage their use by musicians. The perceptual discomfort is caused by two detrimental effects: the occlusion effect and the isolation effect. The occlusion effect is often reported as an augmented, unnatural and annoying perception of one's own voice or instrument mechanically coupled to the head when wearing HPDs. The isolation effect is the unnatural sensation of being isolated from a given sound environment and can be caused by wearing HPDs that do not compensate for psychoacoustical factors and therefore alter the wearer's auditory perception. Both effects are highly unfavorable to the musicians' auditory perception and compromise their capacity to perform to the best of their abilities for their audience. They are among the reasons most often reported by musicians to decide not to wear HPDs.

This master's project presents the concept and first prototype of an active HPD for musicians that aims at solving the detrimental effects while protecting the musician's hearing. A solution for the occlusion effect is presented in the form of an earplug complemented with in-ear active noise control. Practical design issues and required trade-off are analyzed through a literature review and the implementation and characterization of an active occlusion effect reduction system, allowing reduction of the occlusion effect between 8.5 and 12 dB at 250 Hz. A solution for the isolation effect is presented in the form of an earplug complemented with digital signal processing capabilities. Factors that may cause the isolation effect are identified through a literature review and corresponding algorithms that aim at re-establishing the naturalness of the auditory perception while wearing HPDs are presented through the design and implementation of an isolation effect compensation system, allowing up to 15 dB of variable uniform attenuation when used by itself. Both systems working simultaneously in the same device would result in an active HPD for musicians that reduces the occlusion effect and offers uniform variable attenuation up to 25 dB and perceived uniform attenuation up to 25 phons. The aim of this active HPD for musicians is to cause the least perceptual discomfort while protecting a musician's most precious tool: his hearing.

Keywords: Musicians, earplug, hearing protection device, active noise control, in-ear active noise reduction, active occlusion effect reduction, occlusion effect, isolation effect

VERS UN PROTECTEUR AUDITIF ACTIF POUR MUSICIENS

Antoine BERNIER

RÉSUMÉ

Les musiciens professionels sont exposés à de forts niveaux sonores et devraient protéger leur audition afin d'éviter des pertes auditives permanentes qui pourraient potentiellement compromettre leur carrière. Étant donné que de forts niveaux sonores sont souvent inévitables dans le cadre de leur travail, la solution logique serait de porter des protections auditives lorsque nécessaire. Cependant, un inconfort perceptuel est associé au port de protecteurs auditifs et freine leur adoption par les musiciens. Cet inconfort découle de deux effets indésirés : l'effet d'occlusion et l'effet d'isolement. L'effet d'occlusion est souvent décrit comme une perception augmentée, dénaturée et désagréable de sa propre voix ou de son instrument couplé mécaniquement à la tête. L'effet d'isolement est la sensation souvent désagréable d'être isolé de son environnement sonore. L'effet d'occlusion et l'effet d'isolement compromettent la capacité des musiciens d'offrir une performance musicale à la mesure de leurs habiletés losqu'ils portent des protecteurs auditifs. Ils font partie des raisons les plus souvent évoquées par les musiciens pour justifier leur décision de ne pas porter de tels protecteurs.

Ce projet de maîtrise présente le concept et premier prototype d'un bouchon d'oreille actif pour musiciens qui vise à réduire l'inconfort causé par les deux effets néfastes tout en protégeant l'audition du musicien. Une solution pour contrer l'effet d'occlusion est présentée sous la forme d'une protection auditive dotée d'un contrôle actif du bruit intra-auriculaire. Les considérations pratiques et compromis associés à la conception d'un tel contrôle sont analysés à travers une revue de littérature ainsi que par l'implémentation et la caractérisation d'un système actif de réduction de l'effet d'occlusion, permettant d'obtenir une réduction de 8.5 ou 12 dB à 250 Hz. Une solution pour l'effet d'isolement est présentée sous la forme d'une protection auditive dotée d'un traitement numérique de signal. Les principales causes de l'effet d'isolement sont identifiées par le biais d'une revue de littérature et des algorithmes visant à rétablir une perception auditive naturelle lors du port de protections auditives sont présentés à travers la conception et l'implémentation d'un système de compensation de l'effet d'isolement, permettant d'obtenir une attenuation uniforme variable jusqu'à 15 dB. Les deux systèmes combinés forment un protecteur auditif actif pour musiciens permettant de réduire l'effet d'occlusion et de fournir une attenuation uniforme et variable jusqu'à 25 dB, ainsi qu'une attenuation perçue comme uniforme jusqu'à 25 phons. Le but de ce nouveau protecteur auditif est de minimiser l'inconfort perceptuel tout en protégeant l'outil de travail le plus important du musicien : son audition.

Mot-clés : Musiciens, bouchon, protecteur auditif, contrôle actif du bruit, contrôle actif de l'effet d'occlusion, effet d'occlusion, effet d'isolement

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LIST OF ABBREVIATIONS

ADC	Analog to digital converter
ANC	Active noise control
ANR	Active noise reduction
AOER	Active occlusion effect reduction
ARP	Auditory Research Platform
ATF	Acoustic test fixture
BRAMS	International Laboratory for Brain, Music & Sound Research
CB	Critical bands
CIRMMT	Centre for Interdisciplinary Research in Music, Media and Technology
CRITIAS	Sonomax-ÉTS Industrial Research Chair in In-ear Technologies
DAC	Digital to analog converter
DC	Direct current
ECM	Electret condenser microphone
ERB	Equivalent rectangular bandwidth
ETS	École de Technologie Supérieure
FET	Field effect transistor
FIR	Finite impulse response
GM	Gain margin
HPD	Hearing protection device
HTL	Hearing threshold level

IC	Integrated chip
IEM	Inner-ear microphone
IIR	Infinite impulse response
ILS	Internal loudspeaker
ISO	International Standard Organization
MEMS	Micro-electro-mechanical system
MIHL	Music-induced hearing loss
NIHL	Noise-induced hearing loss
NIOSH	National Institute for Occupational Safety and Health
OE	Occlusion effect
OEM	Outer-ear microphone
OER	Occlusion effect reduction
OSHA	Occupational Safety and Health Administration
PEL	Permissible exposure limit
PM	Phase margin
REAT	Real-ear attenuation at threshold
REL	Recommended exposure limit
ROE	Residual occlusion effect
SNR	Signal to noise ratio
SPL	Sound pressure level
WHO	World Health Organization

LIST OF SYMBOLS

dB	deciBel
dB(A)	A-weighted deciBel
dB(HTL)	deciBel of hearing threshold level
dB(SPL)	deciBel of sound pressure level
R	Resistance
L	Inductance
C	Capacitance
L_{eq}	Equivalent sound level
j	Imaginary unit
ω	Frequency in rad/s
ζ	Damping factor
$1/g_c$	The gain that drives a closed loop system into instability
ϕ_c	The phase shift that drives a closed loop system into instability
∞	Infinity
$\mathcal{Z}$	z-transform
$\mathcal{L}^{-1}$	Inverse Laplace transform

INTRODUCTION

Musicians are noise-exposed workers who heavily rely on their auditory sense and should protect it by wearing hearing protectors. However, many musicians do not protect their hearing because they find that the use of hearing protectors is detrimental to their performance, as they cause perceptual discomfort. This master's project presents the concept of a new active hearing protection device (HPD), an earplug for musicians that aims at solving the perceptual discomfort associated with HPDs to enable musicians to protect their hearing. This introduction defines the problem, states the objectives and exposes the methodology of this master's thesis.

Problem statement

Professional musicians are exposed to high levels of sound and should protect their hearing to avoid permanent hearing loss that could compromise their career. Since high sound pressure levels are often required by a musician's work, the logical solution would be to wear HPDs when required. However, perceptual discomfort associated with wearing HPDs can discourage musicians from using them. This perceptual discomfort is assumed to be caused by two detrimental effects: the occlusion effect and the isolation effect.

The occlusion effect is often reported as an augmented, unnatural and annoying perception of one's own voice when wearing HPDs. It affects all musicians whose instrument induces vibrations to the skull, including singers and musicians whose instrument is pressed against any part of the head, such as a trumpet or violin.

The isolation effect is defined, in this work, as the unnatural sensation of being isolated from a given sound environment. It can be caused by wearing HPDs that do not compensate for psychoacoustical factors and therefore modify the wearer's auditory perception. To the author's knowledge, the exact underlying factors that cause one to feel isolated from its sound environment when wearing HPDs have not been exhaustively exposed.

The isolation and occlusion effects are highly unfavorable to the musicians' auditory perception and compromise their capacity to perform to the best of their abilities for their audience. The isolation effect can make it difficult for musicians to judge the sound quality that is being presented to their audience. When, as a consequence of the occlusion effect, an augmented and unnatural perception of one's own voice or instrument is predominantly what is heard, musicians cannot hear the subtle cues that they depend on to adjust their playing. Cues such as knowing how their timbre blends with their colleague's or how loudly their instrument sounds and resonates in a given space can potentially make a big difference in one's performance. These adverse effects are causing some musicians to decide not to wear HPDs.

Solutions to these effects have been proposed in the past. The occlusion effect can be partially addressed by the use of deep insertion earplugs, but this solution causes problems of physical comfort and seems insufficient according to musicians' complaints. The isolation effect can be partially addressed by the use of specialized hearing protection devices, but surveying professional musicians reveals recurring complaints of feeling isolated from their sound environment.

Objectives

The *general objective* of this master's project is to define the characteristics of an earplug that has the least possible negative impact on a wearer's auditory perception.

From the hypothesis that the occlusion effect and the isolation effect cause musicians to decide not to wear HPDs, in regard to the general objective, the following *specific objectives* are derived:

1. Design a method for a system that minimizes the adverse impact of the occlusion effect;
2. Identify the main psychoacoustical factors that can cause the isolation effect when wearing HPDs;
3. Design a method for a system that minimizes the adverse impact of the isolation effect;

4. Investigate the implementation of an electro-acoustic system incorporating the above solutions and allowing further characterization of the occlusion effect and isolation effect experienced by musicians.

Methodology

The methodology of this work starts with a brief review of studies on musicians' hearing loss, exposure to sound, usage rate of currently available hearing protectors and complaints about the perceptual discomfort that they cause. The purpose of this introductory literature review is to factually expose the motivations behind this work and confirm the preponderance of the occlusion and isolation effects.

The occlusion effect is characterized through a literature review that includes a comparison of established solutions as well as a novel approach to solving this effect. This novel approach, in-ear active noise control of the occlusion effect, is reviewed in-depth. The gathered information is used to define the requirements of a system that would minimize the adverse impact of the occlusion effect without compromising physical comfort, like the traditional deep insertion solution. An electro-acoustic system that would solve this effect is designed according to these requirements, fulfilling objective 1.

Regarding the isolation effect, a literature review serves to identify the main psychoacoustical factors that can cause this detrimental effect. The reason for such a review is to fulfill objective 2, an important objective since the causes behind the isolation effect have not been exhaustively defined before, to the author's knowledge. The review on the isolation effect includes the study of established solutions to the isolation effect and their acceptance by musicians. The gathered information is used to define the requirements of a system that would minimize the adverse impact of the isolation effect. An electro-acoustic system that would solve this effect is designed according to these requirements, fulfilling objective 3.

Both systems are implemented separately while ensuring their flexibility and compatibility with each other, and are validated objectively, fulfilling objective 4.

In summary, the methodology outlined below is used, where elements consisting of a literature review have been regrouped:

1. Perform a literature review of topics related to the objectives, specifically:
   a) the perceptual reasons reported by musicians for not wearing HPDs;
   b) the occlusion effect:
      i. the occlusion effect experienced by musicians and solutions that have been devised to reduce the occlusion effect;
      ii. the investigation of in-ear active noise control, a promising solution to the occlusion effect;
   c) the isolation effect:
      i. the psychoacoustical factors that are likely to cause the isolation effect;
      ii. the acceptance of established solutions to the isolation effect;
2. Define the requirements of a system to address and further characterize the occlusion effect and the isolation effect;
3. Design, implement, and validate objectively an electro-acoustic system to reduce the adverse impact of the occlusion effect;
4. Design, implement, and validate objectively an electro-acoustic system to reduce the adverse impact of the isolation effect.

The content of this master's thesis very closely follows the methodology. Chapter 1 consists of a literature review and contains information that was used to identify the elements of the problem and the solution. Based on this information, the requirements and design choices of an active hearing protection device for musicians are presented in chapter 2. The design is presented in two parts: a solution to the occlusion effect, in chapter 3, and a solution to the isolation effect, in chapter 4. Conclusions are then presented and followed by proposed further research.

Contributions

The original contributions of this work include the following:

1. Several prototypes of a hearing protection device for musicians that provide solutions to the occlusion and isolation effects and that have been objectively validated;
2. The hardware, method and design required to answer further research questions, allow for further characterization of the occlusion effect and isolation effect experienced by musicians and improve the prototypes;
3. A significant contribution to the design and implementation of a reconfigurable hardware platform that facilitates in-ear research;
4. One invited presentation, Bernier and Voix (2011), one article in a periodical, Bernier and Voix (2013, article in press), and one article in conference proceedings with an invited presentation, Bernier and Voix (2013). The aforementioned articles are available in appendix VI.

CHAPTER 1

LITERATURE REVIEW

Many different fields are involved in this master's project and are studied in this literature review, divided in 5 sections. Section 1.1 explains the context of the problem, section 1.2 reviews the occlusion effect, section 1.3 reviews active noise control theory and practical concerns, section 1.4 reviews specific practical considerations for active noise reduction in the ear canal and section 1.5 surveys the literature about the isolation effect. Finally, section 1.6 summarizes the literature review.

1.1 Context of the problem

This section covers the context of the problem, reviewing music-induced hearing loss (1.1.1), the legislations and recommendations to prevent hearing loss and how musicians are at risk (1.1.2), and the current efficiency of commercially available HPDs in protecting the hearing of musicians (1.1.3).

1.1.1 Music-induced hearing loss

Noise-induced hearing loss (NIHL) is a condition that affects many workers, where exposure to noise is identified as the cause that led to the hearing loss. Likewise, music-induced hearing loss (MIHL) is a term that applies when the cause of the hearing loss is exposure to music. Chasin (2005, p.15) notes how similar their consequence is:

> *Hearing losses from a wide range of music and noise sources have similar audiometric patterns on a hearing test. The low-frequency sensitivity is either normal or near normal, whereas the hearing sensitivity in the 3000 Hz to 6000 Hz region is reduced. Yet, the hearing sensitivity of an individual to a 8000 Hz sound is much better, and like the lower frequencies can be normal or near normal.*

An audiogram is a measure of one's hearing sensitivity. Octave bands sounds can be used to measure one's thresholds of hearing in deciBel (dB) of hearing threshold level (HTL). A low value in dB(HTL), either negative or null, indicates a superior or perfect hearing in reference to theoretical absolute threshold of audibility for human audition. A value of 30 dB(HTL) at a given frequency indicates that one has lost 30 dB of sensitivity at that frequency. Figure 1.1 shows the typical shape of an audiogram revealing severe noise-induced or music-induced hearing loss.

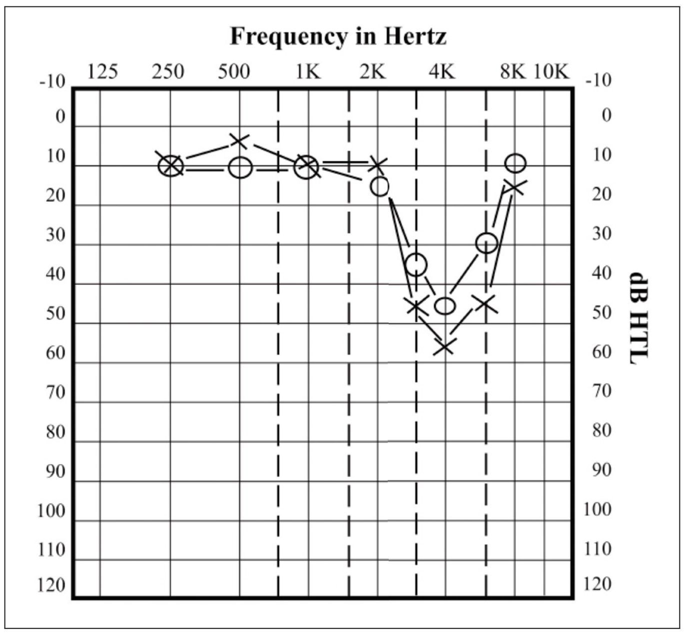

Figure 1.1 Audiogram showing a severe music-induced or noise-induced hearing loss, where a notch in sensitivity is apparent around 4 kHz (O = Left ear, X = Right Ear) Figure courtesy of Dr. Marshall Chasin. Used with permission.

Meanwhile, audiometric examination of professional musicians reveals notches in 25% of cases, according to Eaton and Gillis (2002), 52.5%, according to Royster *et al.* (1991), and up to 70% according to Fabiocchi (2010). Toole (2008, p. 436) claims that *"no amount of experience can compensate for the inability to hear the lowest 20 or 40 dB of musical dynamics, timbral subtleties, distortions, and noises."*

1.1.2 Legislation and recommendations regarding risk of hearing loss

Recommendations and legislation regarding occupational noise exposure have been issued around the world. Although different in their details, the different recommendations and legislation aim at limiting the cumulative sound exposure level that workers are subject to, during a full day of work, to be inferior to a hazardous sound exposure threshold, usually weighted in dB(A).

On one hand, organisms such as the World Health Organization (WHO) or the National Institute for Occupational Safety and Health (NIOSH) define recommended exposure limits (REL) based on scientific studies relating hearing loss to the level and duration of noise exposures, according to Niquette (2012). The recommendations of WHO and NIOSH are different because their goal is different. WHO aims at defining recommendations that would protect all workers, while NIOSH aims at defining recommendations that would protect the majority of workers. Since some humans are more at risk than others because of genetic predisposition as well as other factors, following NIOSH's recommendation does not guarantee that one will not develop NIHL.

On the other hand, organisms such as the Occupational Safety and Health Organization (OSHA) enforce a legislation and define a permissible exposure limit (PEL) resulting from *"debates and compromises that are part of enacting any legislation"*, according to Niquette (2012). Therefore, some legislations may not adequately protect a significant portion of worker.

NIOSH defines the REL as the exposure to an equivalent sound level of 85 dB(A) for a duration of 8 hours, corresponding to a cumulative daily noise dose of 100%. The term "equivalent sound level" (L_{eq}) means that the variations in sound level over time are accounted for, using time averaging, and that the overall exposure is equivalent to the exposure to a steady sound of a given value, 85 dB(A) in this case. As the equivalent sound level increases, the recommended exposure duration decreases. NIOSH defines this time-intensity trade-off, called exchange rate, as 3 dB(A). For example, if the equivalent sound level increases by 3 dB(A), the recommended exposure duration halves. Exposure to an equivalent sound level of 88 dB(A) for 8 hours corresponds to a daily noise dose of 200%, and a daily noise dose of 100% for 4 hours. In most European countries and most of Canada's provinces, the legislation is similar to the NIOSH recommendation.

OSHA defines the PEL, by legislation, as the exposure to an equivalent sound level of 90 dB(A) during an 8-hour day for 5 days a week. An exchange rate of 5 dB is used: as the equivalent sound level increases by 5 dB(A), the allowed exposure duration halves. Exposure to an equivalent sound level of 95 dB(A) for 8 hours corresponds to a daily noise dose of 200%, and a daily noise dose of 100% for 4 hours.

Professional musicians are noise exposed workers and the PEL and REL, although defined in an industrial context, should also apply to musicians. As opposed to typical steady industrial noises, music can vary greatly in intensity, over time.

Many studies have aimed at characterizing the sound levels that musicians are exposed to. The literature review of Patel (2008) compiled the results of different researches and concluded that classical musicians are exposed to sound levels between 80 dB(A) and 110 dB(A), while rock/pop musicians are exposed to higher sound levels, between 88 dB(A) and 117 dB(A). These findings indicate that musicians can be routinely exposed to hazardous levels of sound. When exposed to these levels, the allowed safe exposure duration can very well be below the duration of a normal concert. For example, an equivalent sound level of 100 dB(A) is allowed for 2 hours by the Occupational Safety and Health Association (OSHA) legislation and recommended for less than 15 minutes by the National Institute for Occupational Safety and Health (NIOSH). While concerts are potentially the main source of sound exposure for musicians, hazardous sound exposure is also likely during practice. Hagberg *et al.* (2005) found a correlation between the amount of practice hours and hearing disorder when surveying 407 former music students: *"There was a 1.75 times higher incidence [...] of impaired hearing for musicians with more than 20 h practicing per week before onset of symptoms."* Additionally, O'Brien *et al.* (2013) found that 53% of musicians in their study group would exceed recommended exposure levels in their solitary practices alone, without even considering group rehearsal and performances.

The sound exposure that a marching band composed of music students is subjected to during a performance was measured using dosimeters configured to OSHA's and NIOSH's criteria by Miller *et al.* (2007): *"The daily noise dose values using the OSHA criteria ranged from approximately 200 to 700%, while the values using the NIOSH criteria varied from approximately 1600 to 17000%. The significantly higher noise dose values obtained using the NIOSH criteria were expected due to the different criterion levels and exchange rates used by the two methods."*

The sound pressure level and exposure duration to which professional musicians are exposed as part of their duties indicate that professional musicians are noise exposed workers and should therefore take necessary measures to protect their hearing.

1.1.3 Efficiency of commercially available hearing protection devices

Although specialized HPDs have benefited several musicians, many studies show that professional classical musicians generally do not use hearing protection devices. In France, out of 190 professional classical musicians surveyed by Richoux *et al.* (1998), none used hearing protection devices while 47% reported suffering from symptoms associated to sound overexposure.

Similar populations were surveyed in other European countries, and the use of HPDs was found to be 6% in Finland, 15% in Denmark and about 10% in Germany, as reported by Huttunen *et al.* (2013). The highest usage rate of the studied populations was in the Netherlands, with 52% of the musicians reporting using HPDs in rehearsal, but only 29% using them in concerts.

HPDs seem to be generally more accepted in amplified music, although musicians still find problems with them. Santoni and Fiorini (2010) studied the acceptance of pre-molded musicians earplugs in a Brazilian population of contemporary musicians that were trained with their use, and asked to try them for a period of 3 months. Although general satisfaction towards the earplug was reported by 73.4% of the population, 43.5% reported they disagreed, or strongly disagreed, with using earplugs when performing in shows. Only 4.3% of musicians reported having no negative sensation from wearing the HPD, and "dampened voice" was the most reported negative sensation, with an occurrence of 43%.

Patel (2008), Fabiocchi (2010), Santoni and Fiorini (2010) and Huttunen *et al.* (2013) inquired as to musicians' reasons for not using HPDs, including specialized ones. Here are some of the most prevalent reasons given by musicians as to why they do not wear HPDs:

1. They hinder the musician's performance;
2. They prevent musicians from hearing their colleagues playing;
3. They block out timbre, nuances and dynamics information;
4. They cause communication difficulties;
5. They modify the perception of a musician's own instrument;
6. They protect from loud sounds but that means not being able to hear soft sounds;
7. They are uncomfortable to wear.

All these reasons except comfort indicate a perception shift caused by HPDs. They can be regrouped into two categories. First, modified perception due to the occlusion effect: a musician playing an instrument mechanically coupled to his head while wearing HPDs will hear himself differently. Second, modified perception due to the isolation effect: HPDs reduce the acuteness and reliability of a musician's auditory perception and causes one to feel isolated from his environment. Ultimately, both the occlusion effect and the isolation effect can hinder a musician's performance.

1.2 The occlusion effect

This section presents the causes and characteristics of the occlusion effect (1.2.1), how the occlusion effect is measured (1.2.2), and a review of solutions to occlusion effect (1.2.3).

1.2.1 Characteristics of the occlusion effect

The occlusion effect (OE) is often reported as an unnatural and annoying perception of one's own voice when wearing HPDs. It will affect all musicians whose instrument induces vibrations to the skull, including singers and musicians whose instrument is pressed against any part of the head, such as a trumpet or violin. Although there is a direct solid borne sound path to the cochlea, it is generally accepted that the main objective occlusion effect is due to the existence of another solid borne sound path that ultimately reaches the cochlea by sound generation due to the vibrations of the ear canal walls that cause pressure fluctuations in the air contained in the ear canal. When the ear canal is unoccluded, less energy is transferred to the ear canal by bone conduction as the ear canal has an open-end, hence a lower acoustic impedance, and what is heard is predominantly the sound wave arriving from the air conduction path between the source (e.g. vocal tract) and the ear. However, when the ear canal is occluded, the walls have a strong coupling with the cavity and thus the ear canal sound level is greater and is picked up by the auditory system while the air conduction path is blocked, so what is heard is predominantly the sound wave traveling by bone conduction. Since this effect is more pronounced at low frequencies, below 1000 Hz, the result is an augmented and unnaturally "boomy" perception of one's own voice. Figure 1.2 illustrates how the occlusion effect occurs.

It is apparent from figure 1.3 that the sound pressure level (SPL) increase caused by occlusion effect occurs in the lower frequencies of the speech bandwidth, as measured by Killion (1988). Killion (1988) states that SPL in the occluded ear canal when one is speaking typically amounts to 90 to 100 dB(SPL), and that the occlusion effect results in an amplification of the low frequencies of the talker's own voice by sometimes 20 to 30 dB. Kuk *et al.* (2005) reported a sound pressure increase caused by occlusion effect between 10 and 27 dB for the self-vocalization of /i/, with most of the energy below 800 Hz, and the energy peak around 300 Hz on average.

Unfortunately, to the author's knowledge, no extensive measurement of sound pressure level increase in relation to the occlusion effect has been conducted on musicians.

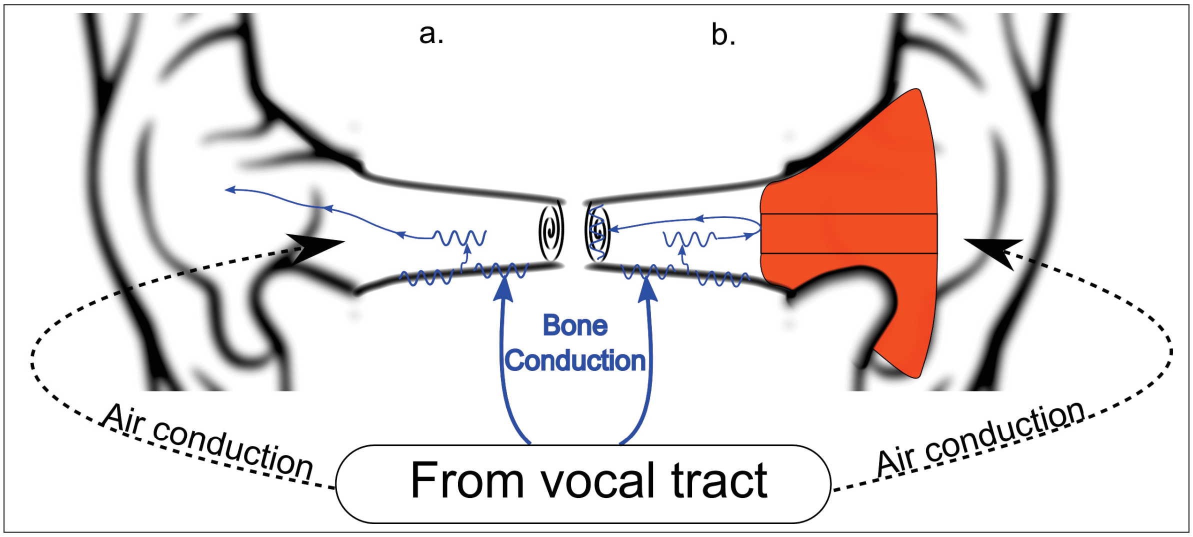

Figure 1.2 Cause of the occlusion effect: a) the air conduction path prevails; b) the bone conduction path prevails, leading to an unnatural and augmented perception of one's own voice

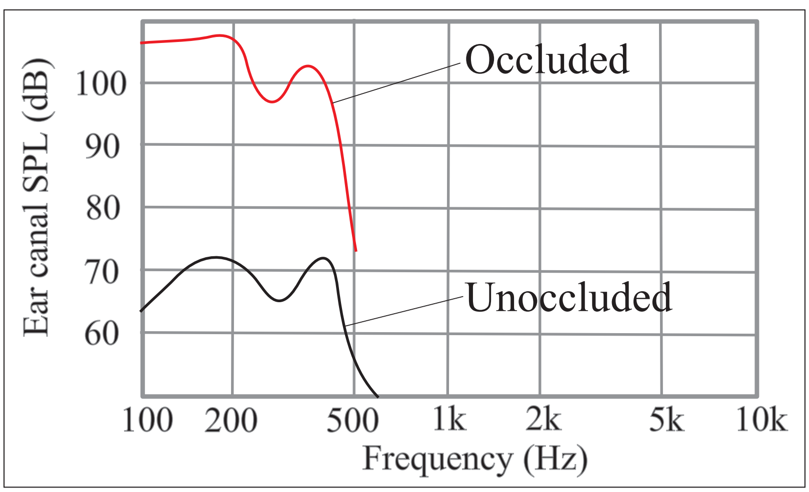

Figure 1.3 Sound pressure level in an unoccluded ear canal and an occluded ear canal when a subject is vocalizing /i/ Adapted with permission from Killion (1988)

1.2.2 Measuring the occlusion effect

In the literature, the occlusion effect is not measured in SPL by audiologists. It is defined by Henry and Letowski (2007, p.70, 72) as an *"increase in the loudness of a bone-conducted sound because of the closing of the ear canal by an earphone, earplug, or other object [...] The physical measure of the occlusion effect does nothing to indicate the degree to which the occlusion can be adversely experienced by the listener."* The objective occlusion effect is thus measured by obtaining the difference between hearing threshold values of bone conducted sound when the ear is occluded and when it is open. Figure 1.4 accurately indicates the different pathways from a given source to the inner ear, showing that body or bone conduction can bypass the ear canal, and how the excitation from a forehead bone vibrator, the vocal cords, turbulences or a mouth speaker travels to the inner ear.

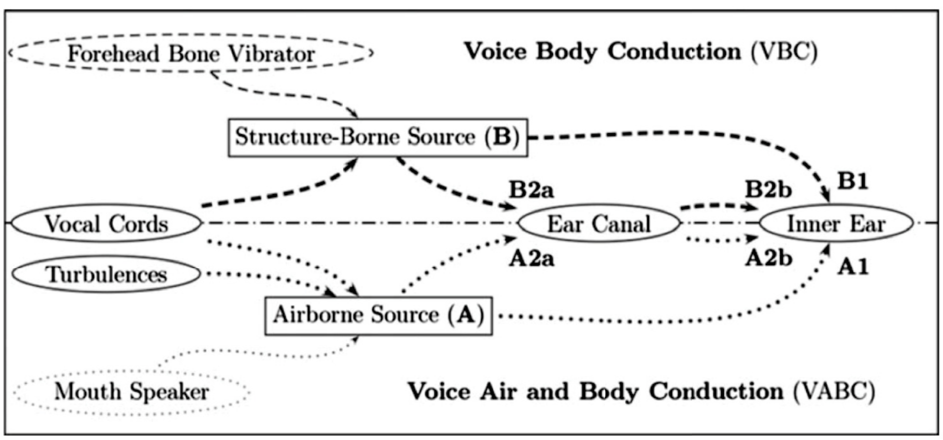

Figure 1.4 Pathways from a source to the inner ear
Reproduced with permission from Le Cocq *et al.* (2010)

Attempts to characterize the relationship between the increase in SPL in the ear canal caused by occlusion effect and the occlusion effect in hearing thresholds difference were notably performed by Goldstein and Hayes (1965), Fagelson and Martin (1998) and Reinfeldt *et al.* (2013). In all studies, a positive correlation is found between the two measures, but the relationship is not unity, and the placement of the bone vibrator is one of the factors that greatly affects the results. Goldstein and Hayes (1965) generally measured greater SPL than hearing threshold

OE below 1 kHz, Fagelson and Martin (1998) measured less SPL than hearing threshold OE at 250 Hz, and Reinfeldt *et al.* (2013) generally measured more SPL than hearing threshold OE at 250 Hz and opposing tendencies at 500 Hz and 1 kHz depending on the bone vibrator placement. The large standard deviations obtained by Reinfeldt *et al.* (2013) indicate that some individuals exhibited an opposing tendency when compared to the average relationship between SPL in the ear canal and hearing threshold occlusion effect. Hence, though the magnitudes of the objective occlusion effect in SPL and the occlusion effect in hearing thresholds difference are strongly correlated, their relationship is not obvious to characterize mathematically.

Typically, the occlusion effect is measured at 250 Hz and 500 Hz, that is, the range where it is perceived the most, and is generally higher at 250 Hz than at 500 Hz. High sound pressure levels below 200 Hz are less perceived because the human ear has a low sensitivity below this point, according to Henry and Letowski (2007). Henry and Letowski (2007, p.72) state that the occlusion effect *"is considered negligible if it is less than 10 dB, mild to moderate when it is between 10 and 20 dB, and severe when it is larger than 20 dB."* The magnitude of the occlusion effect depends on the insertion depth of the occluding device, as discussed in section 1.2.3. Dean and Martin (2000) measured the OE for a shallow insertion and a deep insertion of the same earphone. Their findings are compared in table 1.1.

Table 1.1 Comparison of occlusion effect for shallow and deep insertion as measured by Dean and Martin (2000)

Frequency (Hz)	**Occlusion effect (dB)**	
	Shallow Insertion	**Deep Insertion**
250	17	9
500	14	8
1000	6	-1

1.2.3 Solutions to the occlusion effect

Several types of solutions to the occlusion effect are found in the literature and can yield good results depending on the situation.

First, inserting the occluding device deeply in the ear canal can reduce the occlusion effect, as could be deduced from table 1.1. The magnitude of the increased SPL caused by occlusion effect depends on the remaining volume of the occluded canal, as shown in figure 1.5. Deep insertion tends to reduce the occlusion effect, but is likely to render uncomfortable the actions of speaking, singing, or any action involving mandibular movement since this movement affects the shape of the ear canal. The magnitude of ear canal deformation caused by mandibular movement has been shown to be approximately between 0.2 mm and 2.3 mm, according to Darkner *et al.* (2007), so inserting a device that restricts this deformation can be uncomfortable. On the other hand, earmuffs leaving a large volume in their cup also reduce the occlusion effect, but are bulky and more severely impede localization.

Second, drilling a small hole through the occluding device will significantly decrease the occlusion effect, as studied by Kuk *et al.* (2005). While this method is often applied to hearing aids, hearing protection devices that use such a vent do not offer uniform attenuation, and the method is equivalent to introducing a leak, limiting the achievable attenuation at low frequencies.

Third, the wearer might get accustomed to the occlusion effect, an approach that was listed by Killion (1988) in the context of hearing aids. However, this solution does not seem effective in the context of musician's HPDs. According to Laitinen and Poulsen (2008), the occlusion effect has been found to cause 43% of musicians of a study group to stop wearing hearing protection. The effect has been reported as annoying by 20% of respondents and very annoying by 50% of them. Therefore, it seems that there is a strong tendency to stop using HPDs when the occlusion effect is a problem, rather than becoming accustomed to it.

Fourth, the SPL in the ear canal caused by occlusion effect can be reduced by active noise control. This method seems promising in the case of musicians. The idea is to cancel the sound present in the ear canal by acquiring that sound with an in-ear microphone and processing the sound into an anti-sound to be played by a loudspeaker inside the ear canal, as shown in figure 1.6. The sum of the sound and anti-sound tends to silence. This method has been demonstrated by Mejia *et al.* (2008) to be a promising solution, at least for speech. A prototype was tested on 12 subjects, measuring a peak reduction of the SPL caused by occlusion effect of 15 dB on average, at 300 Hz.

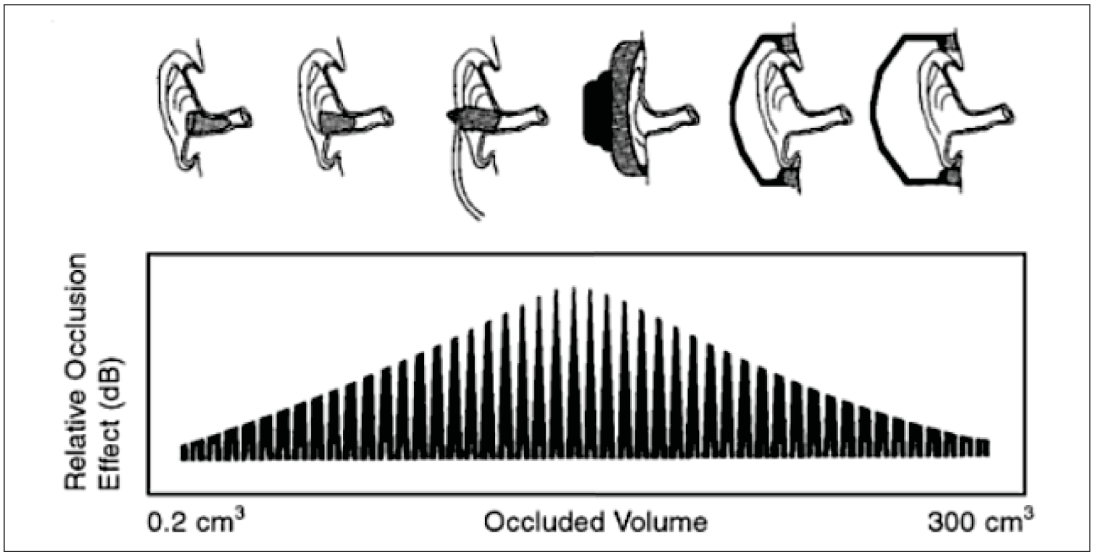

Figure 1.5 Relative occlusion effect per occluded volume based on the type and fit of hearing protectors
Reproduced with permission from Berger (2003)

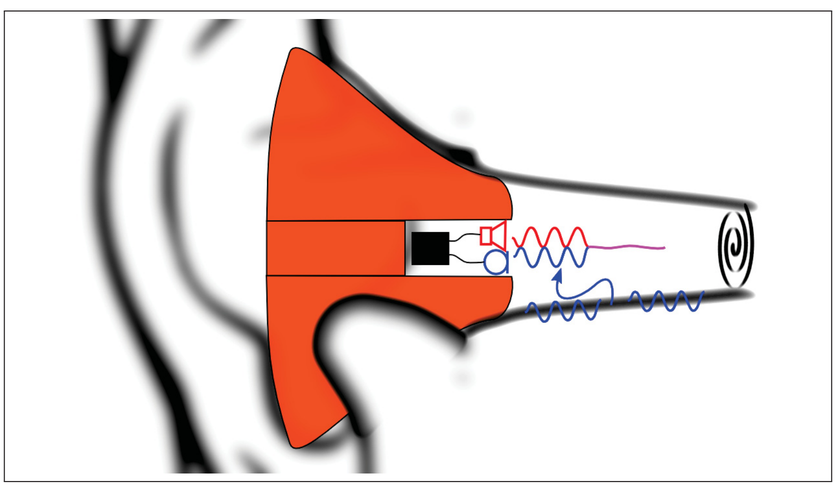

Figure 1.6 Active control of the occlusion effect using an internal microphone, a controller to process the sound into an anti-sound and a speaker to sum the anti-sound to the sound, tending to silence

1.3 Active noise control

Kuo and Morgan (1999) explain the basic principle of active noise control (ANC) as follows:

> *Active noise control involves an electroacoustic or electromechanical system that cancels the primary (unwanted) noise based on the principle of superposition; specifically, an antinoise of equal amplitude and opposite phase is generated and combined with the primary noise, thus resulting in the cancellation of both noises. The ANC system efficiently attenuates low-frequency noise where passive methods are either very expensive or bulky.*

There are two main types of ANC: feedforward and feedback. The feedforward ANC uses an upstream reference microphone to acquire the signal to be canceled, thus gaining advanced information on the noise to be canceled before it reaches the point where its cancellation is desired. Adaptive feedforward ANCs usually use a digital controller to compute the anti-noise, and if the reference microphone is upstream enough, the propagation time to the cancellation point can be used to perform calculations to obtain a suitable anti-noise. Adaptive feedforward controllers use a second microphone at the point where cancellation is desired to verify how well the noise is being canceled. This information serves to adapt calculation on the anti-noise to maximize cancellation. Feedforward controllers are usually efficient and stable, and are used whenever the reference signal is available.

In some cases, however, the reference signal cannot be acquired beforehand, most likely because it is too close to the cancellation point and there is no or too little propagation time. In that case, a feedback ANC can be used. A single microphone is used to acquire the signal to be canceled, and a loudspeaker close to the microphone plays an anti-noise. A short response time is crucial in feedback ANC, and the ideal control should send the anti-noise instantaneously, otherwise its performance is compromised. If delay is present, the anti-noise and noise are not completely synced, and constructive interference can occur and even lead the feedback into instability.

In the case of occlusion effect control, the reference signal is not easy to acquire in practice. A piezoelectric microphone on the throat of a singer or on an instrument could be used to provide advanced information on the signal, but would be in the way of a musician's performance. Therefore, a feedback control strategy is more suitable to this application: a microphone inside the ear canal could pick up the sound resulting from the occlusion effect and a loudspeaker could play a suitable anti-sound. This section reviews the principle of feedback active noise control (1.3.1), its analog implementation (1.3.2) and its digital implementation (1.3.1).

1.3.1 Feedback active noise control

An equivalent block diagram of a feedback ANC is shown in figure 1.7. The loudspeaker and microphone assembly is called the plant, and the transfer function between the loudspeaker's input and the microphone's output, $G(s)$, is the plant transfer function. The disturbance signal $d(t)$ is the signal that must be canceled to obtain a zone of quiet and the error signal $e(t)$ is the error, or residual signal. A controller $H(s)$ acts explicitly as a negative feedback gain, feeding the plant with a command to produce a control signal, or anti-signal.

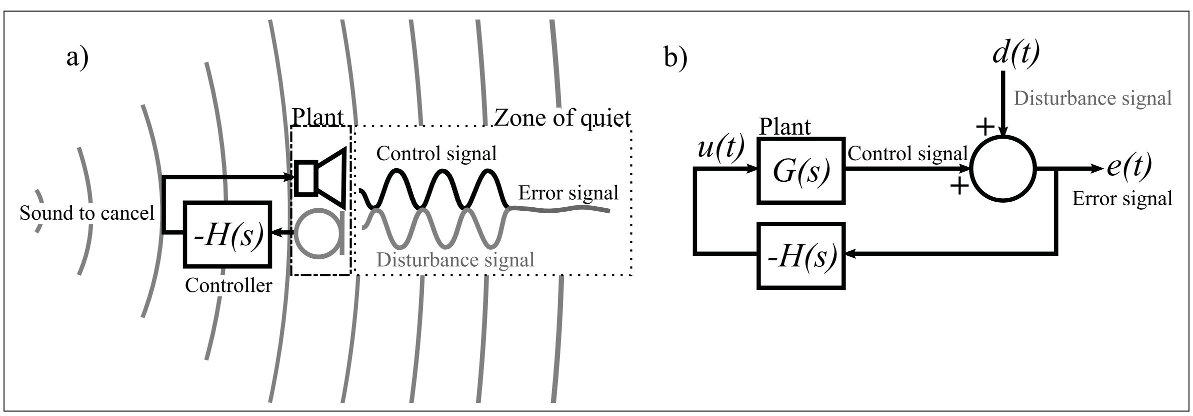

Figure 1.7 Physical block diagram of a feedback control system for the suppression of an acoustic disturbance (a) and its equivalent block diagram (b)
Adapted with permission from Elliot (2001)

In the system from figure 1.7, the Laplace transform of the error signal is given by equation 1.1

$$E(s) = D(s) - H(s)G(s)E(s) \tag{1.1}$$

The transfer function between the disturbance and the error or residual signal can be derived from equation 1.1. This transfer function is referred to as the *sensitivity function*, $S(s)$, expressed in equation 1.2.

$$S(s) = \frac{E(s)}{D(s)} = \frac{1}{1 + H(s)G(s)} \tag{1.2}$$

To ensure that the control provides good rejection of the disturbance, the sensitivity function needs to be small. Inspection of equation 1.2 reveals that if the combined plant and controller frequency response is positive and of high amplitude, the disturbances will be rejected efficiently. On the other hand, if the combined frequency response is negative, the disturbances will be amplified and instability will occur if $H(s)G(s) = -1$. In practice, the plant frequency response will vary greatly across the frequency range. The frequency response of the loudspeaker and the microphone and the acoustic delay due to the distance between them introduces phase shift. As the phase shift approaches 180°, the feedback becomes positive and the system becomes potentially unstable.

The role of the controller $H(s)$, also called compensator, is to compensate the plant's frequency response so that the system is stable and performs well. Typically, it means obtaining high amplitude and a phase around zero in the bandwidth where cancellation is desired and small amplitude where the plant is out of phase, so that the effect of constructive feedback is kept as small as possible.

The term disturbance implies straying from a desired situation, in that case silence or a null signal, is implicit in figure 1.7 to emphasize on the disturbance rejection ability of the system. Figure 1.8 emphasizes on its tracking ability, but it is equivalent to figure 1.7 if $c(t) = 0$ and the plant output, renamed $y(t)$ to be consistent with control literature, contains implicit disturbance. In the system from figure 1.8, the Laplace transform of the plant output $y(t)$ is expressed by equation 1.3.

$$Y(s) = H(s)G(s)[C(s) - Y(s)] \tag{1.3}$$

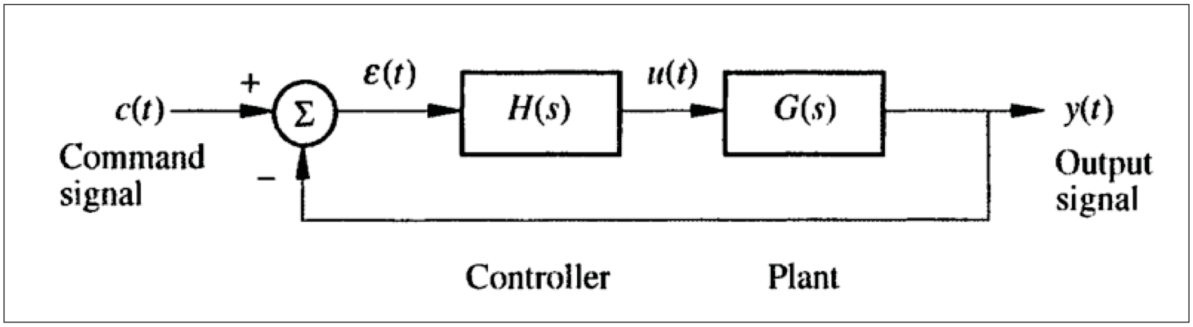

Figure 1.8 Block diagram of a feedback control system having a servo action, in which the output $y(t)$ is arranged to follow the command signal $c(t)$
Reproduced from Elliot (2001)

The ability of the system to track a command signal is characterized by equation 1.4, the transfer function between the plant output and the command signal, often referred to as complementary sensitivity function, $T(s)$.

$$\frac{Y(s)}{C(s)} = \frac{H(s)G(s)}{1 + H(s)G(s)} = T(s) \tag{1.4}$$

If $T(s)$ is large, the system will provide efficient tracking. The sensitivity function and complementary function sum to unity, and the output signal due to both the command signal and the disturbance is shown in equation 1.5.

$$Y(s) = S(s)D(s) + T(s)C(s). \tag{1.5}$$

In general, a small positive sensitivity function is required for good disturbance rejection and provides a large positive complementary sensitivity function that implies good tracking. However, since the plant is an electro-mechanical and acoustical system that has complex dynamics including resonances and phase shifts, the choice of a good compensator, or controller, is critical. In any ANC application, plants have uncertainties and variabilities that can cause the feedback to become unstable, so a trade-off between performance and safety rules the design of a compensator. The next section explains controller design and implementation using analog compensators.

1.3.2 Analog implementation of compensators

Analog compensators have numerous advantages for a practical feedback ANC design. First, they do not introduce a constant group delay like digital implementations. Second, they have proven their practical use in many commercial applications. Third, they are often inexpensive compared to digital compensators.

On the other hand, conventional analog controllers are fixed, so they will not adapt to changes in the plant response. Their implementation can result in characteristics that stray from the design, because of their components' tolerances and their sensitivity to atmospheric conditions. Analog controllers are also limited to implementable topology that can only process the signal in certain ways, so the mathematically optimal controller may not be implementable in practice using analog controllers.

When designing an analog controller for a complex plant such as a loudspeaker/microphone assembly, the most important consideration is to make sure that the system will be stable. A very useful design method is to use a frequency response approach in conjunction with the Nyquist criterion for stability. The Nyquist criterion involves tracing a polar plot of the open-loop frequency response of the compensator/plant subsystem, $H(j\omega)G(j\omega)$. If any point of that subsystem's response is equal to -1, it is apparent from equation 1.2 that the feedback control becomes unbounded. Generally, the Nyquist criterion states that if the polar plot of the open-loop compensator/plant subsystem frequency response encloses the Nyquist point (-1,0) as ω varies from $-\infty$ to ∞, the feedback loop will be unstable. Since the plant has uncertainties that will change its frequency response and can lead an otherwise stable closed-loop system into instability, stability margins must be observed so that the open-loop system never encloses the Nyquist point. Large enough margins will ensure stability for any plant uncertainties, and the system will be said to be *robustly stable*. However, providing large margins is done at the expense of performance. Therefore, stability margins should ideally be just large enough to match the maximum potential change in the frequency response of the plant due to uncertainties. In practice, the achievable performance is limited if the plant has large uncertainties. Two different stability criteria are used in that regard: gain and phase margins.

The gain margin is the increase in overall gain that can be tolerated before the Nyquist point is enclosed. Figure 1.9 shows that if the open-loop frequency response is increased by a gain of $1/g_c$, the system is unstable. The gain margin (GM), in dB, is expressed by equation 1.6. The phase margin (PM) is the phase shift that can be tolerated at the frequency for which the modulus of the open-loop gain is unity before reaching the Nyquist point, as illustrated by ϕ_c on figure 1.9. According to Haugen (2009), reasonable values for gain margin and phase margin are in the range of 6 dB to 12 dB, and 30° to 60°, respectively.

$$\text{GM} = 20\log(1/g_c) \tag{1.6}$$

While the GM and PM are useful in defining the relative stability of a system, they are limited because they do not provide information on the stability when uncertainties in the plant induces both phase and gain changes, as is often the case.

Traditional analog controllers use modular filter elements, such as lead, lag, or lead-lag filters. These filters are better suited for controllers than conventional filters because they introduce less phase shift. Figure 1.10 shows a simple lag filter, built using passive components, and its frequency response.

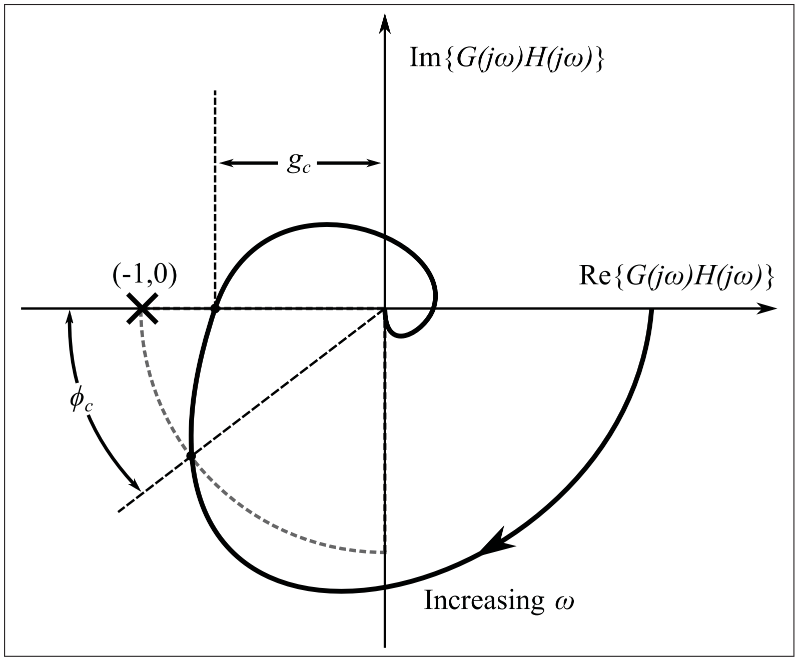

Figure 1.9 Nyquist plot of an open-loop frequency response, showing the gain margin, $20\log(1/g_c)$ and the phase margin, ϕ_c
Reproduced from Elliot (2001)

1.3.3 Digital implementation of compensators

This section examines the consequences of implementing the compensator or controller digitally. Figure 1.11 shows a general block diagram of a continuous plant $G_c(s)$ compensated with a digital controller $H(z)$, along with an anti-aliasing filter $T_A(s)$, to block frequencies superior to half the sample rate, an analog-to-digital (ADC) converter and a digital-to-analog (DAC) converter $T_Z(s)$ to provide an interface between the continuous time-domain and the discrete-time domain, and a reconstruction filter $T_R(s)$ to smooth the response from the DAC. The plant is subject to a disturbance $d_c(t)$.

The z-domain transfer function can be used to characterize the continuous-time system from the perspective of the controller, as it obtains sampled information from the continuous plant's output and sends a sampled signal. Considering all the continuous-time functions to be part of the plant, the z-domain transfer function $G(z)$ of the plant can be obtained by taking the

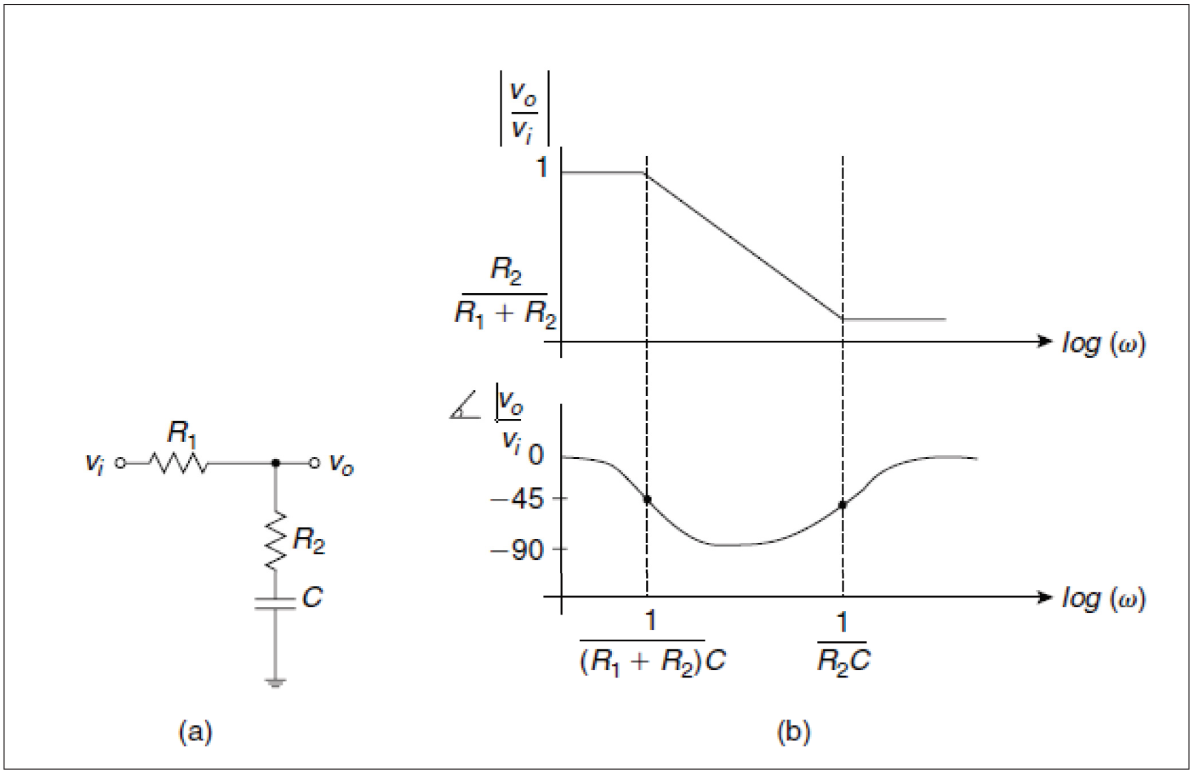

Figure 1.10 A simple lag analog compensator; a) implemented with discrete analog components, b) its frequency response
Reproduced from Pease (2008)

z-transform of the inverse Laplace transform, as shown in equation 1.7.

$$G(z) = \mathcal{Z}\mathcal{L}^{-1}[T_Z(s)T_R(s)G_c(s)T_A(s)] \tag{1.7}$$

The continuous-time disturbance will be discrete from the perspective of the controller and can be expressed by equation 1.8.

$$D(z) = \mathcal{Z}\mathcal{L}^{-1}[T_A(s)D_C(s)] \tag{1.8}$$

Expressing $E(z)$ in a similar fashion to $E(s)$ in equation 1.1 leads to equation 1.9, and the transfer function between the error $E(z)$ and the disturbance $D(z)$ is the discrete sensitivity function $S(z)$, as shown in equation 1.10. It can be seen that the result obtained in equation 1.10 is analogous to the continuous-time sensitivity function from equation 1.2. Assuming

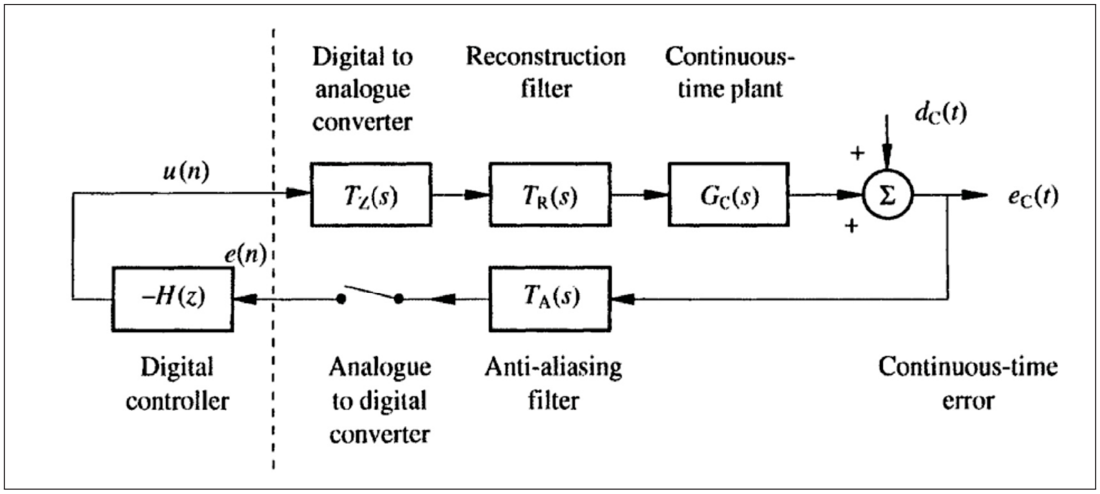

Figure 1.11 Block diagram of a feedback control system with a discrete-time controller, with emphasis on the disturbance rejection ability of the system
Reproduced from Elliot (2001)

that $G(z)$ and $H(z)$ are stable, the closed loop system will be stable if the Nyquist criterion is fulfilled.

$$E(z) = D(z) - H(z)G(z)E(z) \tag{1.9}$$

$$S(z) = \frac{E(z)}{D(z)} = \frac{1}{1 + H(z)G(z)} \tag{1.10}$$

Digital controllers have a great number of advantages over their analog counterparts. First, they offer more flexibility because they can be easily reprogrammed. Second, they are durable, insensitive to temperature fluctuations, and their performance is not affected by the tolerances of components. Third, they allow for performance monitoring and can be adapted or stopped if the plant uncertainties cause degradation of performance or instability.

On the other hand, digital controllers have an inherent delay due to their ADC and DAC, as well as a phase delay of the anti-aliasing filter and reconstruction filter. Digital implementations involve representation of an infinite possibility of values by a finite number of bits, introducing artifacts due to quantification and errors from rounding and saturation that can lead a stable system to instability. Often, their performance is closely linked to performing a large amount of calculus, but the execution time must be inferior to the sampling time, $1/F_s$, which also introduces a delay. In order to reduce delay, a higher F_s will be necessary and therefore a more powerful digital signal processor will be required to permit enough calculus, potentially increasing cost.

As previously stated, delay is critical in feedback active noise control. An increased delay will have a direct negative effect on the bandwidth over which control can be achieved, as well as the maximum attenuation that can be achieved. A general way of calculating the effect of delay Δt on the bandwidth is given by equation 1.11, taken from Elliot (2001). For example, a delay of 1 ms would allow a maximum theoretical control bandwidth of approximately 166 Hz. Figure 1.12 illustrates how the delay affects the attenuation performance of an active headset.

$$\text{Bandwidth(Hz)} < \frac{1}{6\Delta t} \tag{1.11}$$

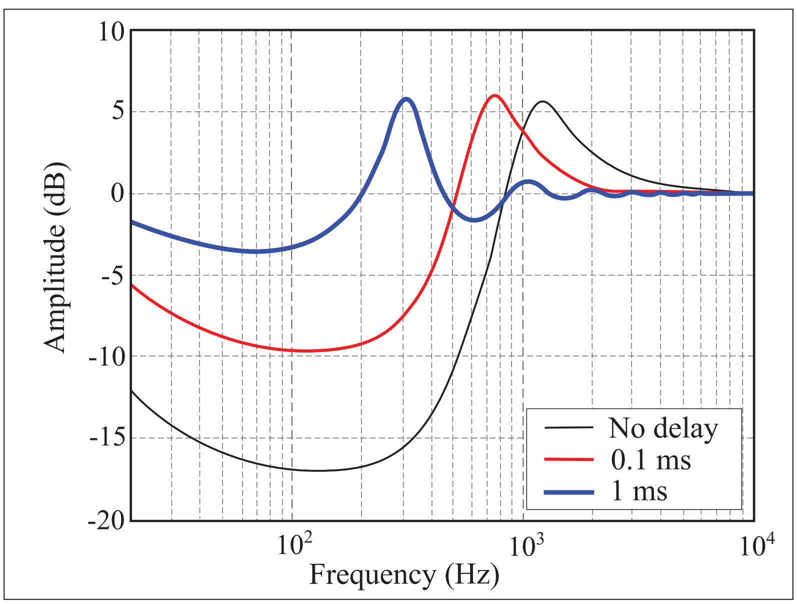

Figure 1.12 Effect of delay on the performance of a feedback ANC
Reproduced with permission from Rafaely (2001)

There are two main ways of implementing a digital controller: Finite-impulse response (FIR) filters, and infinite impulse response (IIR) filters. Both filters can be fixed or adaptive.

As stated by Zimpfer (2000), FIR filters have the advantage of being consistently stable and therefore they are resistant to quantification errors. On the other hand, they have the disadvantage of requiring numerous coefficients, typically several hundred, to properly achieve a given complex frequency response, and this is accentuated if precision is needed in the low frequencies, where active control is often used because it is normally more efficient. A filter with numerous coefficients means an increased number of calculations and therefore more computation time, likely forcing reduction of the sampling frequency, thus increasing delay and requiring the use of an anti-aliasing filter, causing additional phase delay.

IIR filters are the complete opposite. They are not always stable because they include poles, so they rely on the feedback of the output: the future output depends on past and present outputs. Additionally, a stable IIR filter by design can be unstable when it is implemented in finite precision, due to quantification errors. On the other hand, they require considerably less coefficients, often below a hundred, allowing for a high sampling rate that presents two advantages: reduced delay and the possibility of sampling without aliasing using no anti-aliasing filter.

Zimpfer (2000) studied the effect of quantification on digital IIR feedback ANC, showing that the transfer function of the IIR filter was non-linear due to artefacts and depended on the amplitude of the signal. She developed a modified IIR algorithm to decrease the detrimental effect of quantification and achieved digital ANC with performance comparable to continuous-time implementation.

1.4 Active noise reduction in the ear canal

Feedback ANC has often been used to increase the low frequency attenuation of passive earmuffs. As summarized by Herzog (2002):

> *Active noise reduction inside an earmuff benefits from the relatively small volume where the acoustic field should be controlled, but also from the wide availability of transducers suited for this task, leading to affordable commercial products. Moreover, as active noise control (A.N.C.) is more efficient at lower frequencies, its performances may be matched to the passive isolation, resulting in an almost balanced spectrum attenuation.*

Active noise reduction earplug design presents different challenges. Compared to ANR headsets or earmuffs, the volume in which noise must be canceled is much smaller, leading to theoretically better performance. However, a small variation will be much more significant on the small volume enclosed by an earplug than on the larger volume enclosed by an earmuff, leading to large plant uncertainties. Additionally, ANR earplugs require very small transducers to fit in the ear canal, while providing enough sound output power to match the noise to be canceled and exhibiting a frequency response with minimum phase lag.

In summary, two elements have a great impact on the plant response and performance of active control in the occluded ear canal. First, the ear canal is seen by the transducers as an acoustic load that will vary between individuals and will considerably affect the frequency response of the plant. Second, the transducers must be capable of producing high SPL, exhibit minimum phase and be small enough to fit the ear canal. This section consists of a literature review on ear canal variability (1.4.1) and transducer technologies in regard to active noise control in the ear canal (1.4.2).

1.4.1 The effect of ear canal variability on the plant response

Goldstein *et al.* (2005) investigated the effect of the ear canal variation on the plant uncertainties of an universal-fit ANR earplug, through modeling and measurements. Figure1.13 shows the system and model that were used.

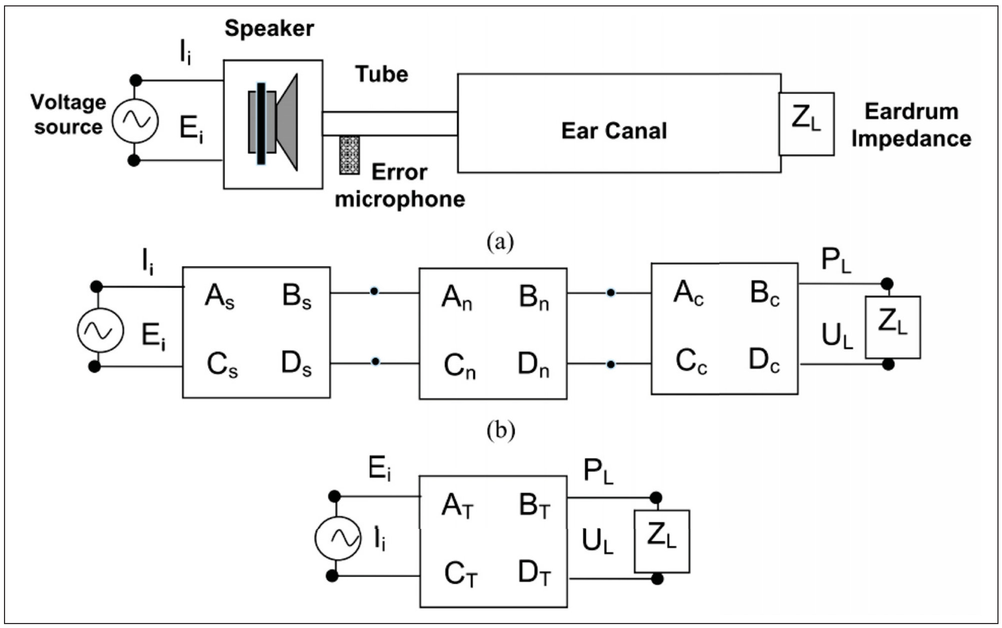

Figure 1.13 Complete dynamic model of deep insertion ANR earplug obtained by cascading four-pole network matrices representing each part of the system Reproduced with permission from Goldstein *et al.* (2005)

The findings of Goldstein *et al.* (2005) are summarized below :

1. *"Plant variation among users can be significant and results from different occluded space dimensions, as well as different eardrum impedances"*;

2. An increase of the occluded ear canal volume will result in a general decrease of the amplitude of the plant response;

3. *"Changes in the ear canal length cause a shift in the frequency of the two high frequency peaks, near 18 kHz and 22 kHz, which originate from acoustic resonances of the occluded space"*;

4. The lower peaks, due to the loudspeaker resonance, *"do not suffer significant shift in frequency due to changes in the input impedance presented by the different ear canal sizes, which indicates that the receiver is close to an ideal volume velocity generator and its response is not much affected by the different acoustic loads"*;

5. *"An increase in the values of the ear drum impedance results in an increase in the magnitude of the pressure frequency response for the frequency range around and below the first response peak at 2500 Hz."* The first response peak corresponds to the first resonance of the loudspeaker;

6. *"The closed loop stability was found to be very sensitive to plant variations resulting from changes in the ear canal size. The changes in length and resulting resonance frequency shifts cause large variations in magnitude and phase in frequency regions with relatively low stability margins, destabilizing the control loop. Changes in radius affect the overall gain of the control loop and can destabilize the system"*;

7. *"It is important to note that when the feedback controller is designed for the small ear canal volume and applied to the large ear canal volume the system in general remained stable, which can be explained by the lower overall gain of the acoustic plant response for the larger occluded space volume."*

The last point is linked to the second one, as explained by Goldstein *et al.* (2005). However, while a system designed for small ear canal volume will be stable in a large one, its performance will decrease due to the lower gain of the plant. Figure 1.14 shows the simulated effect of size of the occluded volume on performance while figure 1.15 shows the effect of variation of the eardrum impedance. To validate the results of the simulation, Goldstein *et al.* (2005) measured the plant response of an ANR earplug in six ears, and compared the results with the predicted plant response for various occluded volumes and ear canal sizes. Figure 1.16 shows the simulated plant responses and figure 1.17 shows the measured plant responses. The measured plant frequency responses vary by 6-8 dB from approximately 1 kHz to more or less 10 kHz and vary by as much as 15 dB at frequencies below 1 kHz.

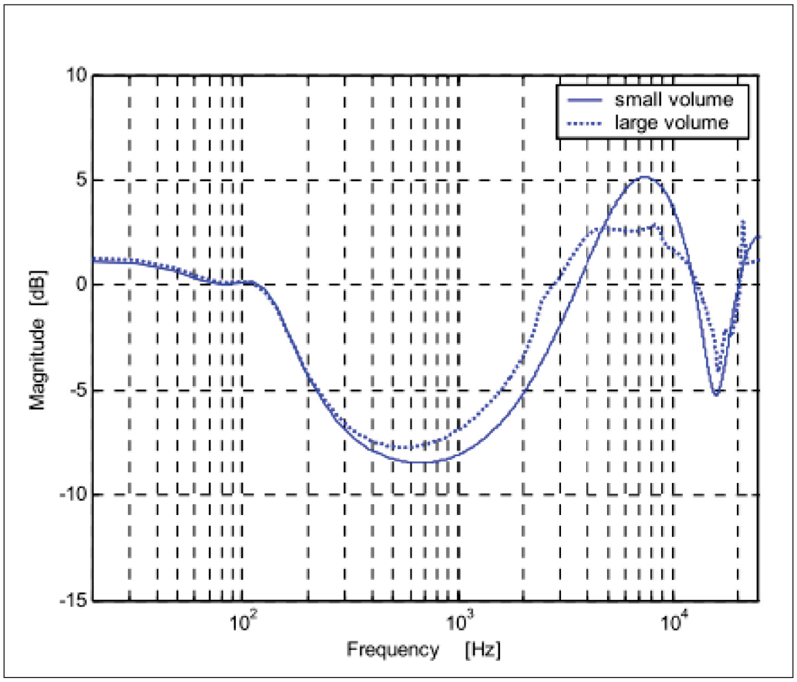

Figure 1.14 Magnitude of the closed loop response for the feedback control system for different occluded space volumes
Reproduced with permission from Goldstein *et al.* (2005)

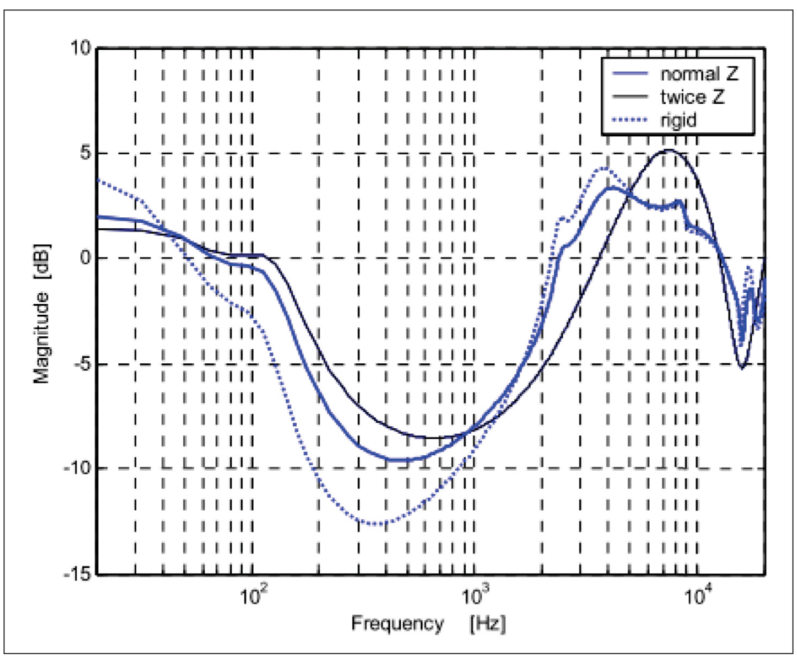

Figure 1.15 Magnitude of the closed loop response for the feedback control system for different occluded eardrum impedances
Reproduced with permission from Goldstein *et al.* (2005)

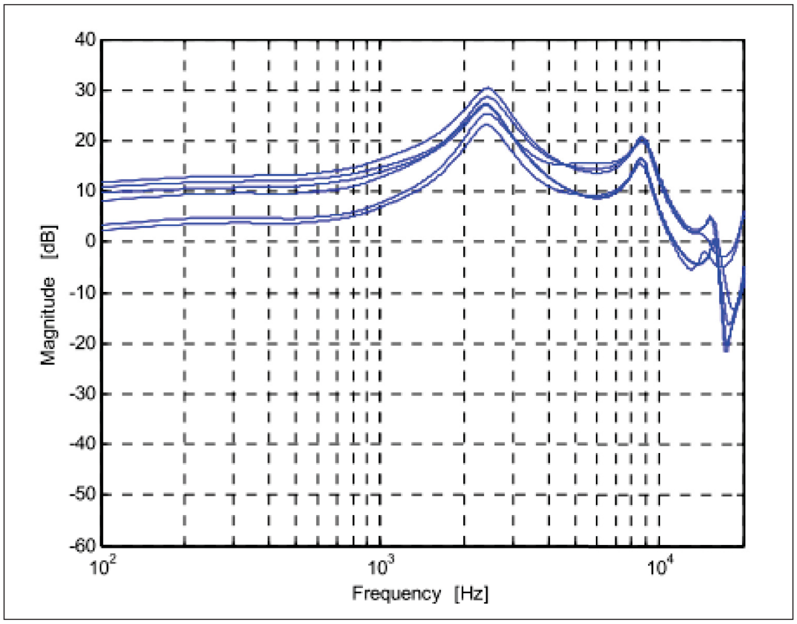

Figure 1.16 Simulated magnitude of the pressure frequency response at the error microphone location for different ears
Reproduced with permission from Goldstein *et al.* (2005)

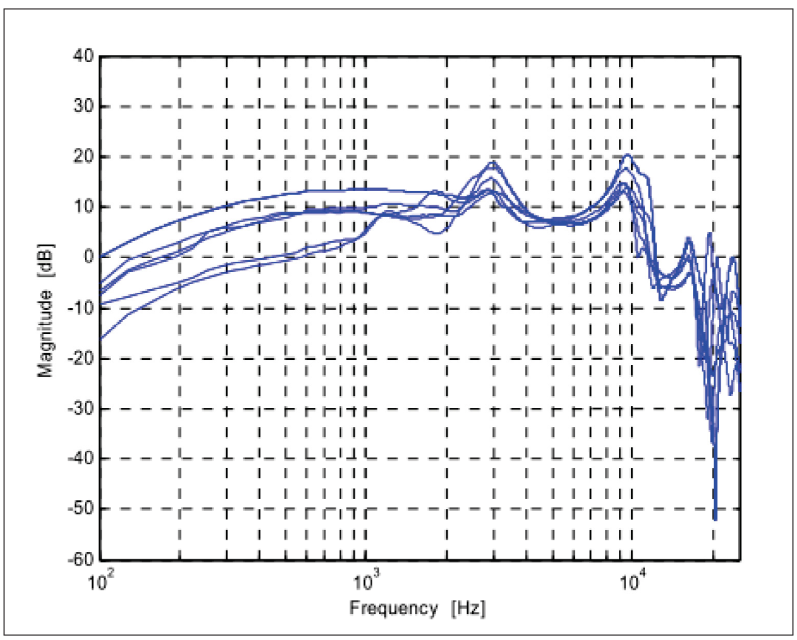

Figure 1.17 Measured magnitude of the pressure frequency response at the error microphone location for different combination of ear canal sizes and eardrum impedance
Reproduced with permission from Goldstein *et al.* (2005)

1.4.2 Transducers technologies in regard to active noise control

When designing an earplug intended for feedback ANC, four major requirements dictate the selection of suitable transducers.

1. The size and shape must be appropriate to fit in the ear canal and allow for optimal placement of the microphone and loudspeaker relative position. Herzog (2002) states that *"a good design should minimize phase delays, by locating the speaker and microphone membrane as close as possible"*;
2. The sound pressure level capabilities of the loudspeaker must match the sound pressure level to cancel;
3. The noise floor of the microphone must be as low as possible, as the inherent noise of the microphone will be played back by the loudspeaker;
4. The plant response, and therefore the frequency response individual transducers, should ideally be uniform and exhibit minimum phase.

Herzog (2002) notes that *"practically, transducers responses are corrected by a control filter, but this can be very difficult if transducers do not have a "smooth" response, or if this response may vary."*

Several transducer technologies exist, each with their advantages and drawbacks. The present section lists and reviews them in regard to active noise control.

1.4.2.1 Moving-coil loudspeakers

Moving-coil loudspeakers are the most commonly available commercially. Figure 1.18 shows a cross section of a typical moving-coil loudspeaker. A coil is mechanically coupled to a diaphragm, also called cone, and rests in a fixed field produced by a magnet. When an electrical current flows through the coil, a corresponding magnetic field is emitted, interacting with the fixed field of the magnet thus applying a force to the coil, pushing it away or towards the magnet. Since it is mechanically coupled to the coil, the cone will push or pull the air it is facing, causing pressure changes and emitting a sound wave. A typical moving coil loudspeaker will be relatively inefficient because the mass of air it can move is small compared to the mass of the diaphragm that needs to be moved. A speaker with a light and large diaphragm will usually be more efficient than a speaker with a small and heavy diaphragm.

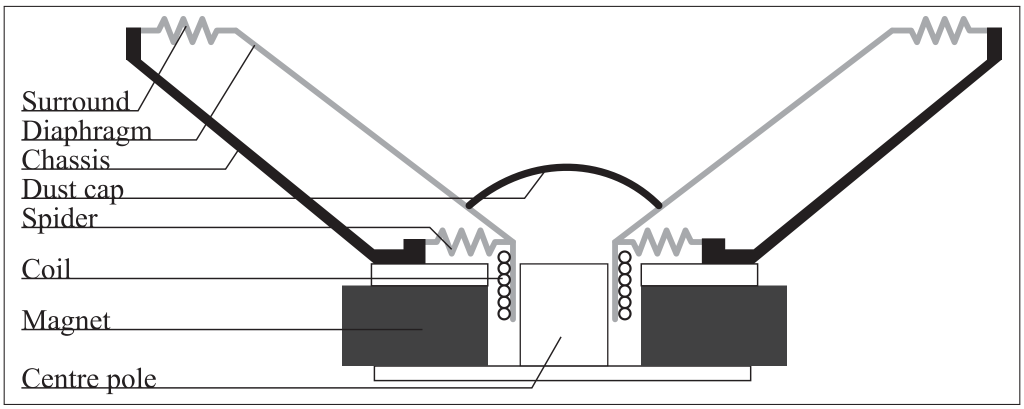

Figure 1.18 Components of a moving-coil loudspeaker

In-ear moving-coil loudspeakers usually provide better bass response than other technologies because they rely on a large diaphragm. In an occluded ear canal, at low frequencies, SPL depends on variation in volume, which in turn depends on the area of the diaphragm and its displacement. The sound pressure level that commercially available loudspeakers of relatively small size can generate is rather impressive. For example, the *FRANKLIN* square dynamic speaker manufactured by *Knowles Electronic* can reproduce a max SPL value of 119.5 dB(SPL) @ 1 kHz for its small size of 15 by 6 by 2 mm.

However, at high displacement corresponding to high SPL, the diaphragm is subject to deformations. Bauer (2000) warns that moving-coil loudspeakers exhibit non-linearities at high levels. Additionally, moving-coil transducers meant to be inserted in the ear canal often exhibit a resonance around 3 kHz, causing a phase shift that is detrimental to an ANR application according to Buck *et al.* (2002), *"the peak at about 3 kHz is meant to reproduce the transfer* *function of the open ear when using closed earphone devices. In feedback systems, this has to be compensated with a filter in order to obtain a stable system with a meaningful level of active attenuation. However, as this compensation cannot be perfect, it usually reduces the bandwidth of the active attenuation."*

1.4.2.2 Balanced armatures loudspeakers

Balanced armature drivers involve a fixed coil around a mobile armature that is mechanically coupled to a diaphragm through a drive rod. Magnets provide a magnetic field that balances the armature in a neutral position. As current flows through the coil, the disturbed magnetic field pushes or pulls the armature, and the diaphragm moves accordingly. Figure 1.19 shows the basic principle of a balanced armature driver. Unlike the moving-coil loudspeakers, the coil of the balanced armature has a fixed position, reducing the mass associated with the diaphragm and allowing for a longer and heavier coil, increasing the sensitivity of the device compared to moving-coil loudspeakers.

Balanced armature drivers, while being very efficient drivers in terms of sound level output, have a number of drawbacks in regard to active noise control, as summarized by Vaudrey *et al.* (2007):

> *Balanced armature manufacturers are currently being motivated by the hearing aid industry and tailor their designs accordingly. Overall phase lag in a design is not important, whereas additional sound power output is important. By adding resonant dynamics, the sound power output of the balanced armature speaker designs are effectively increased at the expense of additional phase lag.*
>
> *There are a variety of dynamic systems in the traditional balanced armature actuator that make it suboptimal for active control: an acoustic system in front of the diaphragm that is separate from the occluded space environment; a port that is used to connect the actuator to a tube in hearing aids that acts as a Helmholtz resonator that also adds additional dynamics in the control band; vibrational modes of the diaphragm and reed itself; a mass-spring-damper system of the mov-*

ing driven diaphragm; the dynamic system of a magnetically driven armature; a compressional mode of the rod connecting the armature and diaphragm.

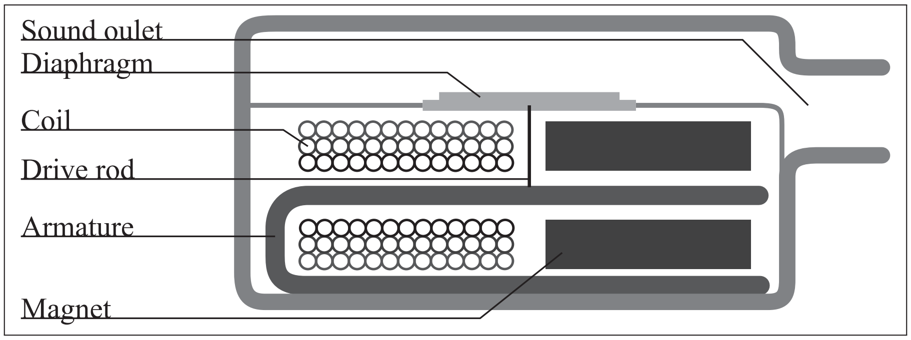

Figure 1.19 Balanced armature driver components

1.4.2.3 Electrostatic loudspeakers

Electrostatic transducers use the force generated by electrical charges. A thin diaphragm with a conductive coating is placed near an electrode or in between two conductive grids that provide a high voltage electric field. Since the diaphragm is very thin and light, and driven over its whole area, as opposed to a moving-coil or balanced armature loudspeaker, they are usually more efficient and more linear. However, their displacement is restrained by the grids that have to be very close for electrical forces to be significant even at high voltage. This leads to limited SPL, in the range of 70 dB(SPL). Moreover, the required high voltage poses practical difficulties and risks.

1.4.2.4 Piezoelectric loudspeakers

Piezoelectric loudspeakers use a crystal material that flexes in response to an electrical field. Low-quality piezoelectric loudspeakers are widely available on the market, but their size and shape are unsuitable for in-ear application. In-ear piezoelectric loudspeakers have been custom-designed for ANR earplug applications in the past, notably by Bauer (2000). Although his custom piezoelectric loudspeaker had quite uniform frequency response characteristics, the sound pressure level capability was only approximately 85 dB(SPL). Piezoelectric loudspeaker also require a large voltage to drive them, posing practical difficulties.

1.4.2.5 Electret condenser microphones

Electret condenser microphones (ECM) are a sub-family of condenser microphones. Condenser microphones generally rely on a metallic diaphragm and a stationary back plate that together form a capacitor. Sound pressure causes the diaphragm to move back and forth, increasing and decreasing the gap between the two plates, causing the capacitance to decrease or increase. This capacitance fluctuation induces changes in potential from an equilibrium point, provided by a DC voltage bias. While conventional condenser microphones require a relatively large voltage bias to function, electret condenser microphone are pre-polarized and require no bias. According to Elko *et al.* (2005), *"Electret microphones dispense with the need for a bias voltage by embedding charge in a polymer that is attached to the backplate. It was the discovery of polymers that could permanently hold an embedded charge that led to the inexpensive electret microphone that is so common today"*. The capacitor formed by the diaphragm and the back plate is of very small value and has a high impedance, so an integrated circuit is used to amplify the signal and lower the output impedance of the electret microphone. Typically, a field effect transistor (FET) is used in low cost design, as its high input impedance is well suited to interface with the small capacitance; the cut-off frequency of the high pass filter formed by the capacitance and the input resistance of the FET falls outside the audible range if the input resistance is high enough. If smaller size or better performances are needed, a specialized integrated chip (IC) is used instead of the FET. The FET or IC requires a bias voltage to function, but this voltage is only about 0.9 V in the case of small microphones designed for hearing aids. Figure 1.20 shows an equivalent circuit schematic for a three lead electret microphone capsule, commonly used in hearing aids.

According to Dillon (2012, p.22), electret microphones have an essentially uniform frequency response. However, when used for hearing aid applications, they often have two deviations from uniform frequency response. First, they are designed to have a low frequencies roll-off: some very low frequency sounds and pressure changes, that may not be audible to humans,

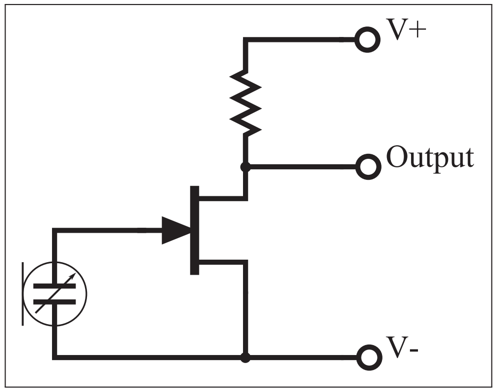

Figure 1.20 Equivalent circuit schematic for a three-lead electret microphone capsule

can be picked up by the microphone and cause it to saturate. To avoid this problem, a back port is sometimes included so that the slow changes in sound pressure occur on both sides of the diaphragm, canceling each other. This causes a high pass effect in the frequency response of the electret microphone. Second, the front port of the microphone, similar to a tube, the compliance of the diaphragm and the volume of air surrounding it exhibit a resonance because they act as a Helmholtz resonator, causing a peak in sensitivity. With proper design, this peak can be tuned to occur at a specific frequency where an increase in sensitivity is desired, or tuned to be out of the audible range.

Electret condenser microphones are traditionally known to offer uniform frequency responses, when designed for this requirement, and high signal-to-noise ratio because of their relatively high sensitivity around 16 mV/Pa and low equivalent input noise lower than 30 dB(A), making them good candidates for active noise control applications. However, for in-ear applications that require smaller sizes than traditional electret microphones, these characteristics come at a relatively high cost.

1.4.2.6 MEMS microphones

Silicon microphones, also called micro-electro-mechanical system (MEMS) microphones, are miniaturized microphones that are constructed using a fabrication process akin to integrated circuits. Traditionally, these microphones offered lower sensitivity, higher noise, and a more uneven frequency response than electret microphones. This trend is changing: in 2012 and 2013, MEMS microphones that match the characteristics of ECMs have been released commercially. These microphones are offered at a lower cost than ECMs and have characteristics that are much more consistent between microphones of the same model because their manufacturing process can be automated.

Most commercially available MEMS microphones use the same principle as condenser microphones, but unlike electret microphones, they are usually not pre-polarized. This allows for the microphones to undergo heat and still function properly; the electret materials lose their charge at high temperatures. However, since the material is not pre-polarized, a large voltage is required to bias them. For this reason, MEMS microphones are often built by including an integrated circuitry in the casing of the microphone to step up the 0.9 V bias available in hearing aids to the higher voltage required by the MEMS. This step-up circuitry is one of the major causes of noise in MEMS.

The sensitivity of capacitive microphone depends on the area and the gap between the plate and diaphragm. Since the diaphragm of MEMS is usually smaller than electret microphones, the gap must be made smaller to compensate. This can easily be done with MEMS technology, but this decreased gap causes more squeeze damping. According to Neumann and Gabriel (2013, p. 198), *"squeeze damping is the damping due to air being squeezed outwards from between two plates moving normal to their surfaces, such as a microphone diaphragm and backplate [...]"*. This phenomenon greatly affects the frequency response of the MEMS microphone and is an internal noise source.

According to Beranek and Mellow (2012, p. 401): *"Because the cost of silicon wafer is fixed, the price of each MEMS microphone depends upon how many can be made from a single wafer. Hence, it is necessary for the diaphragm to be small for economic reasons as well as for miniaturization."* On the other hand, Kaajakari (2004) notes: *"the smaller devices the lower is the signal to noise ratio [...]. Intuitively this may be understood by noting that ratio of mechanical to thermal energy E/kT goes down as the device mass is reduced."* This is because of mechanical-thermal noise, resulting from molecular agitation. According to Gabrielson

(1993), *"any molecular agitation even through solid structures like springs and supports can cause random motion of an object. Often, it is assumed that the preamplifier noise dominates the sensor noise floor [...]; while this is true in some cases, this assumption is particularly dangerous with microminiature sensors for which mechanical-thermal noise can sometimes set a much higher noise floor."*

According to Sessler and Hillenbrand (2009), *"typical silicon condenser microphones have membrane areas of about 1 mm*^{2}*, membrane thicknesses of 0.2 to 0.4* μ*m, air gaps of 1 to 2* μ*m, resonance frequencies in the near ultrasonic range and sensitivities of approximately 10 mV/Pa. They require bias voltages of only a few Volts and their equivalent noise level is 35 to 40 dB(A). They are furthermore shock resistant and insensitive to vibration since their membranes have a relatively small mass per unit area. Such microphones may be operated permanently at temperatures up to 100* $^\circ$*C and up to 260* $^\circ$*C for short periods of time."*

1.5 The isolation effect

Occluding the ear with a HPD has an inherent effect on a wearer's auditory perception on multiple levels. This section reviews phenomena that cause a perception shift and/or a feeling of being isolated from a given sound environment. Three main causes are reviewed: the open and occluded ear responses (1.5.1), the attenuation provided by commercially available HPDs (1.5.2) and the impact of loudness perception on the perceived attenuation (1.5.3).

1.5.1 The open and occluded ear responses

Two components of the external ear affect the transmission of sound pressure to the eardrum: the pinna and the ear canal. The converging geometry of the pinna causes an amplification of sound above 2000 Hz, while the ear canal causes a resonance around 2.7 kHz.

Analogous to a tube, the ear canal has a diameter of about 7 mm and a length of about 25 mm, from the orifice of the canal to the eardrum, according to Henry and Letowski (2007). Just like a tube, it has a characteristic resonance frequency that depends on its length, and the conditions at its boundaries.

A tube that is closed at one end and open at the other is analogous to an unoccluded ear canal: the ear canal entrance is the open end, and the eardrum is the closed end. Such a tube will exhibit a resonance that will amplify a peak frequency and its close surroundings, according to equation 1.12.

$$f(n) = \frac{nc}{4(L + 0.4d)} \tag{1.12}$$

Where $n = 1, 3, 5, ...$, c is the speed of sound, L is the length of the tube and d is the diameter of the tube, resulting in a resonance at the peak frequencies $f(1) \approx 3.08$ kHz , $f(2) \approx 6.17$ kHz, etc.

In reality, the geometry of the pinna and the compliance of the eardrum make the ear canal deviate from the perfect tube analogy, and its resonance has been found to be around 2.7 kHz, as measured by Stenfelt *et al.* (2002). Chasin (2005) quantifies the amplitude of the resonance to be about 15-20 dB. Figure 1.21 shows the high frequency amplification effect of the pinna and the ear resonance.

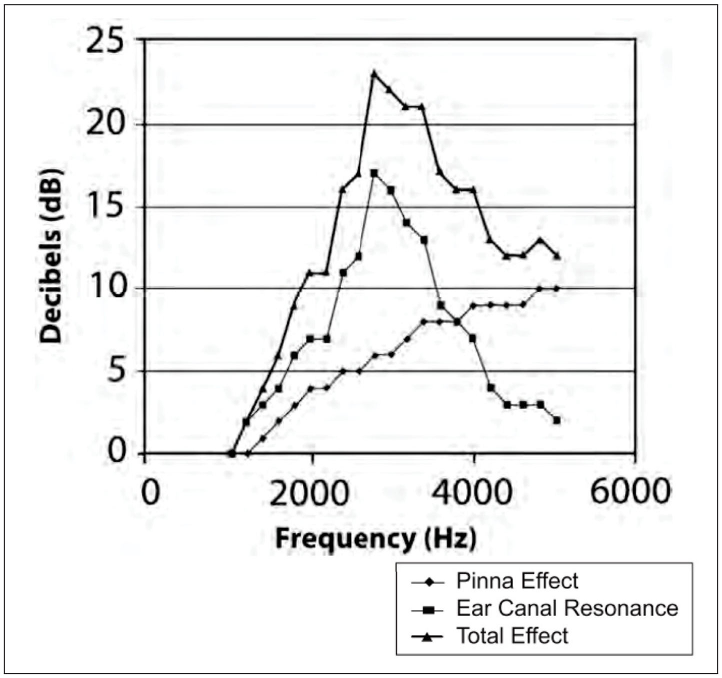

Figure 1.21 The natural amplification of the outer ear, primarily caused by the ear canal resonance and the pinna effect
Figure courtesy of Dr. Marshall Chasin. Used with permission.

A tube that is closed at both ends is analogous to an occluded ear canal: the ear canal entrance is completely obstructed, such as is the case when wearing HPDs. The same tube with different conditions at its boundaries exhibits a different resonance, as equation 1.13 shows.

$$f(n) = \frac{nc}{2(L + 0.3d)} \tag{1.13}$$

Where $n = 1, 2, 3, ...$, c is the speed of sound, L is the length of the tube and d is the diameter of the tube, resulting in resonances at the peak frequencies $f(1) \approx 6.33$ kHz, $f(2) \approx 12.66$ kHz, etc. In reality, the occluded ear resonance has been found to be around 5.5 kHz by Stenfelt *et al.* (2002), but it depends on the insertion depth of the occluding device and the remaining ear canal portion between the tip of the occluding device and the eardrum. Killion *et al.* (1988) found the occluded resonance to be around 8 kHz with musician's custom molded HPDs.

In summary, simply occluding the ear canal will affect the ear response, modifying the ear's natural resonance and resulting in a perception shift.

1.5.2 Attenuation and acceptance of commercially available hearing protection devices

Occluding the ear with a generic hearing protection device, such as those generally used to protect workers in the industry, will not result in uniform attenuation. This is mainly due to a mechanical phenomenon, as qualitatively explained by Berger (2003, p.395):

> *Due to the flexibility of the ear canal flesh, earplugs can vibrate in a piston-like manner, thus limiting their low-frequency attenuation. Earmuffs, too, vibrate as a mass/spring system, the stiffness of the spring depending upon the dynamic characteristics of the earmuff cushion and the circumaural flesh, as well as the volume of the air entrapped inside the earcup. These actions limit their low-frequency attenuation. Representative maximum attenuation values at 125 Hz for earmuffs, pre-molded earplugs, and foam earplugs, are about 20 dB, 30 dB and 40 dB, respectively.*

Therefore, occluding the ear with a HPD typically results in an unbalanced attenuation, more pronounced in the high frequencies than in the low frequencies. This is accentuated by the fact that occluding the ear shifts its natural resonance, as explained in section 1.5.1, increasing the attenuation at and around 2.7 kHz.

Figure 1.22 shows typical shapes of non-uniform attenuation provided by an earplug-type HPD, an earmuff-type HPD, and both devices worn together, as well as the maximum attenuation limit of HPDs. This limit is imposed by the fact that sounds can bypass the HPD by bone and tissue conduction to the inner ear.

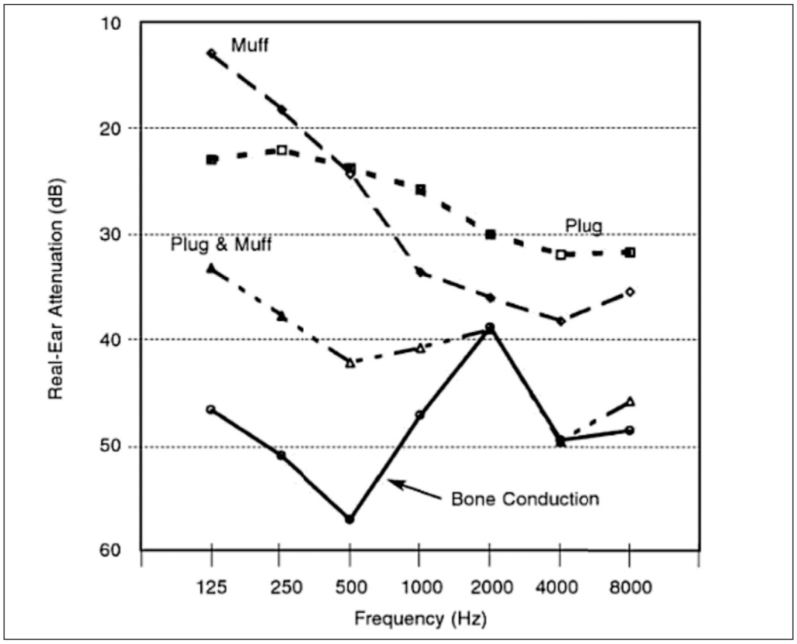

Figure 1.22 Bone-conduction limits to HPD attenuation and an example of the attenuation provided by an earplug, an earmuff, and the two devices worn together Reproduced with permission from Berger (2003)

1.5.2.1 Commercially available earplugs for musicians

In 1988, a solution to the problem of non-uniform attenuation of generic HPDs was presented by Killion *et al.* (1988): *"An Earplug With Uniform 15-dB Attenuation"*. The novel earplug was called ER-15^{TM} and manufactured by Etymotic Research, Inc. Killion *et al.* (1988) stated:

> *Currently available custom-earmold hearing protectors have one defect in common: They muffle the sound. Technically speaking, they give more attenuation at high frequencies than low frequencies. [...] Regardless of their exact construction, a reasonable generalization is that existing custom-earmold hearing protectors produce 10 dB to 20 dB of excessive attenuation. A hearing protector with more uniform response - a high-fidelity earplug, if you will - seems needed.*

The objective that drove the design of the ER-15 was *"[...] to reproduce the shape of the natural frequency response of the normal open ear, but at a reduced level."*, according to Killion *et al.*

(1988). The design is based on passive acoustic elements. Figure 1.23 shows a cross section of the peculiar earplug identifying key acoustic elements of the design and its equivalent electrical circuit.

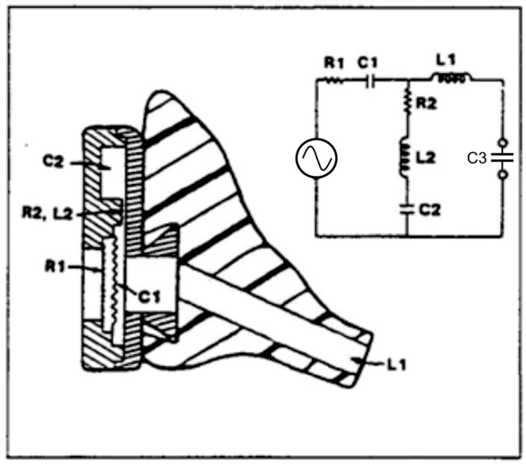

Figure 1.23 Construction of the ER-15 earplug
Reproduced with permission from Killion *et al.* (1988)

The diaphragm C1 and the cavity C2 exhibit compliance, and can be modeled as capacitors. The damping element R1 can be modeled as a resistance, while the sound channel L1 is analogous to an inductance. Three small apertures, though only one can be seen in the cross-sectional view, are represented by the inductance L2, modeling the mass of air in the apertures, and R2, the resistance to flow of that mass of air resulting from friction due to the small size of the apertures. The equivalent circuit of the earplug of figure 1.23 was modified from Killion *et al.* (1988) by the author for analytical purposes by driving it with a voltage source, representing sound pressure level outside the hearing protector, and considering a capacitance C3 at its output, representing the compliance of the occluded ear canal. At low frequencies, below the resonance of the two RLC subcircuits, the capacitors act as a voltage divider, providing attenuation. The R1, L1 and C1 resonant circuit is tuned to enter resonance at 2.7 kHz, offering only a resistance of R1 and therefore allowing more transmission to C3. A scheme is also proposed to nullify the occluded ear canal resonant frequency, in this case around 8 kHz. A second RLC circuit, composed of R2, L2 and C2, is tuned to enter resonance at about 8 kHz, shunting the frequencies in that range to the ground on the equivalent circuit, trapping them acoustically. This creates a dip in the transfer function of the earplug where the peak of the occluded resonance is, canceling the occluded resonance.

The length and diameter of the sound channel L1 therefore have a great impact on the resulting attenuation. If one has a larger and wider ear canal that exhibits a lower resonance than 2.7 kHz, for example 2.5 kHz, the sound channel L1 must be appropriately shorter and wider so that the R1, L1, C1 resonance lowers accordingly. On the contrary, if one has a shorter and narrower ear canal, the sound channel L1 should be longer and narrower. It is therefore critical that the earmold is made properly in order to obtain the desired attenuation. Lack of an acoustic seal between the device and the ear canal would result in the leakage of low frequencies. An undersized sound channel would result in the resonance shifting towards lower frequencies and the earplug over-attenuating the high frequencies as a consequence. The effect of these two situations is illustrated in figure 1.24.

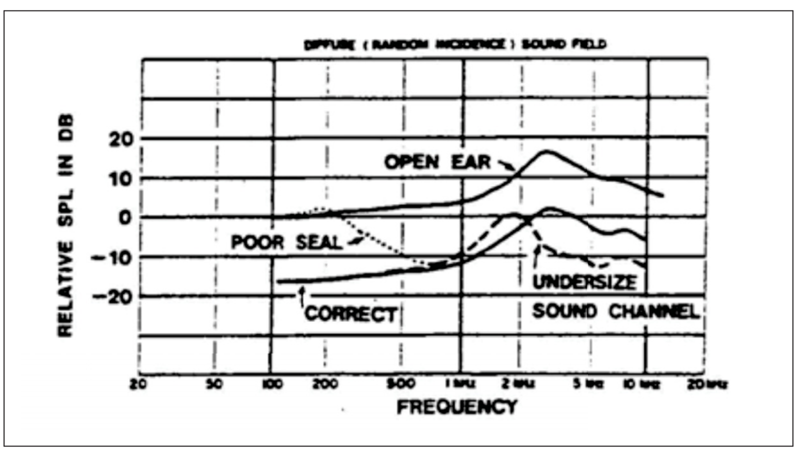

Figure 1.24 Expected eardrum SPL when the ear is open and when it is occluded with the ER-15 for three scenarios
Reproduced with permission from Killion *et al.* (1988)

The inventive earplug was subject to a patent credited to Carlson (1989). It is still available today, 25 years after its release, and comes in three different models providing different attenuation values of 9 dB, 15 dB and 25 dB, obtained by changing the compliance of the diaphragm.

This custom-moulded design was followed by an universal-fit design, patented by Killion *et al.* (1992). In the following excerpt, "Carlson earplug" refers to the ER-15:

> *While giving superb acoustical performance, the Carlson earplug has several limitations:*
>
> 1. *The diaphragm compliance element required by the Carlson earplug is difficult (and thus relatively costly) to manufacture to the tight compliance tolerance required for proper operation.*
> 2. *The Carlson earplug requires a precise and relatively large diameter internal sound channel in order that the Helmholtz resonance between the acoustic mass intrinsic to that internal sound channel and the compliance of air in the earcanal have the proper frequency. A consequent limitation to the Carlson earplug has been the necessity of obtaining a specially manufactured custom earmold for each ear, where it as been found that the earmold manufacturer must individually measure and "tune" the internal sound channel using a special meter manufactured by Etymotic Research in order to provide the correct value of acoustic mass required for proper operation.*

A cross-sectional view of the universal-fit musician earplug, first marketed as ER-20^{TM} and later as ETY-Plugs^{TM} is shown in figure 1.25. Two main parts can be identified: an eartip, comprising a sound channel, and an assembly of an external tube and a cap. The design relies on a large damping element between the two parts, analogous to a resistance, to damp the resonance resulting the residual occluded ear canal volume, analogous to a capacitance, and the air mass in the internal sound channel, analogous to an inductance. The whole system is therefore analogous to a RLC circuit with a large resistance. This makes the length and diameter of the internal sound tube and the residual volume of the occluded ear canal relatively unimportant, two variables that were critical in the design of the ER-15.

The drawback of the use of a large damping element is that it causes undesirable high frequency attenuation, so the purpose of the tube and cap assembly is to acoustically amplify high

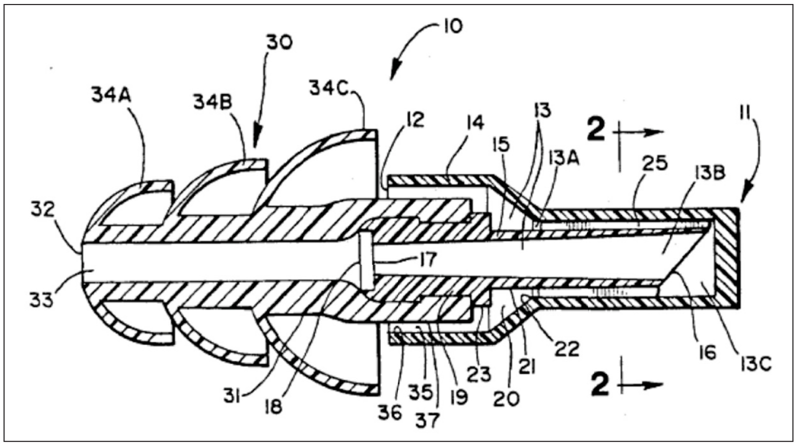

Figure 1.25 Cross-sectional view of a ER-20 earplug
Reproduced with permission from Killion *et al.* (1992)

frequencies in compensation. The external part of the ER-20, the tube and cap assembly, forms a folded horn construction that re-establishes a resonance similar to that of the average open ear, at 2.7 kHz. The opening of the folded horn is located in the pinna, *"where an increased sound pressure level is produced in the 2 to 10 kHz range of frequencies due to the resonances caused by the structure of the conchae and the pinna of the ear"*, according to Killion *et al.* (1992). This results in the attenuation depicted in figure 1.26, comparing the attenuation of the ER-20, or ETY-plugs, to that of a foam E-A-R plugs for shallow and deep insertions.

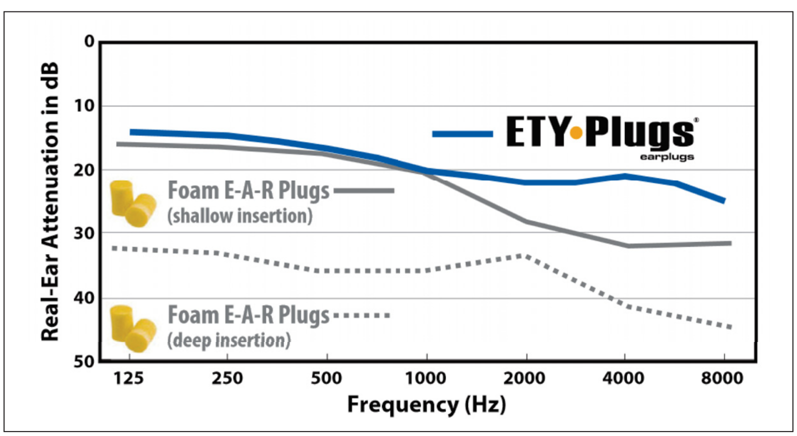

Figure 1.26 Attenuation of the ER-20 compared to foam E-A-R plugs
for shallow and deep insertion
Reproduced with permission from Etymotic Research (2012)

1.5.2.2 Acceptance of commercially available earplugs for musicians

A small number of studies investigating the acceptance of the ER-15 and ER-20 uniform attenuation earplug have been recently published.

Huttunen *et al.* (2013) investigated the usage of ER-15 by 15 symphony orchestra musicians who had them for a median of 15 months, finding that the usage rate was low :

> *Despite elevated hearing thresholds at several frequencies in some subjects, the rate of earplug use was low; only one to three of the 15 musicians reported using their ER-15 earplugs always or nearly always (>95% of the time) either during orchestra rehearsals or concerts, when rehearsing alone or when teaching [...]. One musician used the earplugs often (>80% of the time) during orchestra rehearsals and another used them during rehearsing alone.[...] The subjects were asked with a multiple-choice question whether the use of ER-15 earplugs possibly negatively affected hearing of music and speech. In the questionnaire, hearing of timbre and dynamics was asked in particular. Feelings of distorted or missed timbre/nuances and/or dynamics of music produced by colleagues were reported by 80% of the subjects, and 100% reported that perception of timbre/nuances and/or dynamics of their own playing was affected.*

In an attempt to determine if the attenuation characteristics of their ER-15 was in accordance with the manufacturer's specification, Huttunen *et al.* (2013) measured the field attenuation obtained on musicians, using two different methods. The two methods consisted of standard sound field attenuation using the real-ear attenuation at threshold (REAT), involving octave band noises to measure one's auditory threshold with and without HPDs, and a more precise type of Békésy audiometry, using a continuous sweep tone to track the auditory threshold with and without the HPD. Figure 1.27 shows *"the mean and standard deviation of the attenuation of the ER-15 earplugs obtained in a sound field (REAT, N = 15 subjects) and via headphones (Békésy audiometry, N = 10 subjects), as well as the manufacturer's specification obtained with an artificial head. The error bars and the area between the two gray lines denote the* *inter-subject standard deviations of the attenuation measured in the sound field and with headphones, respectively."*

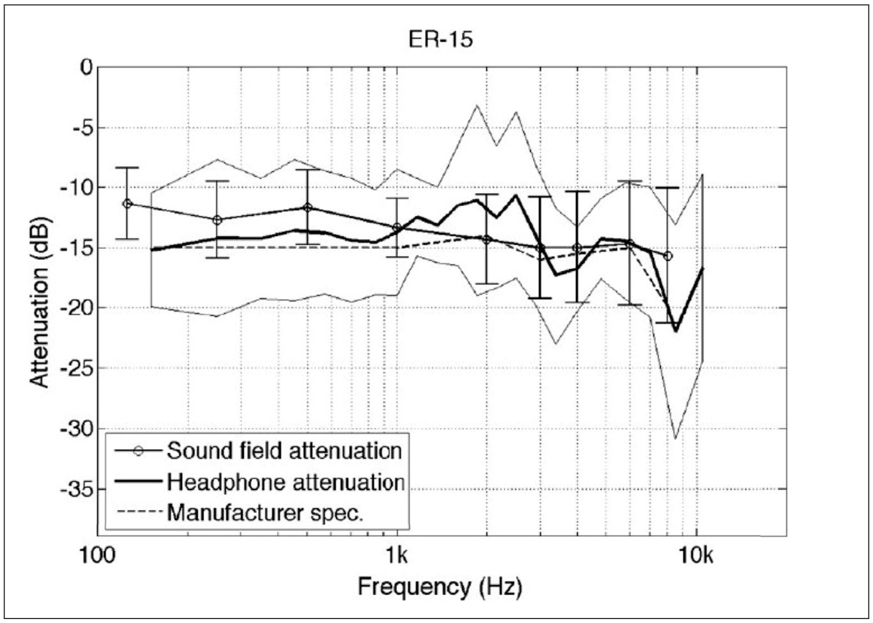

Figure 1.27 Mean and standard deviation of the attenuation of the ER-15 earplugs obtained in a sound field and via headphones as well as the manufacturer's specification obtained with an artificial head
Reproduced with permission from Huttunen *et al.* (2013)

It can be seen that the average of the attenuation is close to the manufacturer's specifications. However, large inter-subject differences were observed. The increase in deviation around 2 kHz is likely attributable to an improper sound channel size causing the resonance action to be unsuitable, as can be deducted by comparing it to the effect of a reduced sound channel size illustrated in figure 1.24.

Huttunen *et al.* (2013) indicate that, in Finland, the process of making ER-15 earplugs does not involve any follow-up to make sure that the earmold is properly made. The large inter-subject variations may be due to this lack of follow-up; the earplug must be carefully tuned to one's ear to provide the right attenuation. However, it is also possible that in some individual cases, the real attenuation, while being according to specifications and quasi-uniform, still alters the musician's auditory perception in unwanted ways and causes him to find that the earplugs distort the sound information.

Regarding the ER-20 universal-fit earplug, Santoni and Fiorini (2010) tested its acceptance with 23 pop/rock musicians that were provided with the hearing protection and were asked to use the HPDs for 3 months. Santoni and Fiorini (2010) came to the following conclusions:

- *"The worst satisfaction evaluation scores about using the HiFi ER 20 hearing protector concerned interference by the protector of high frequency sound perception and full time use of the hearing protector during shows";*
- *"The most common negative sensations while using the hearing protector were dampened voice and pressure in the ears";*
- *"73.9% of musicians scored over 7 to reflect their satisfaction with using the HiFi ER 20 hearing protector, which suggests a favorable tendency towards accepting this device."*

Table 1.2 lists the reported negative sensations and the percentage of occurrence. Only 4.3% of musicians reported perceiving no negative sensations when wearing the HPD.

Table 1.2 Distribution of negative sensations reported by musicians while using the HiFi ER-20 hearing protector (n=23)
Adapted with permission from Santoni and Fiorini (2010)

Negative Sensations	**Number of Subjects**	**Percentage**
Feeling dampened voice	10	43.5%
Pressure in the ear	9	39.1%
Difficulty with music return	7	30.4%
Feeling isolated	7	30.4%
Ear itching	6	26.1%
Interference with music quality	6	26.1%
Communication difficulties	5	21.7%
Feeling of blocked ear	5	21.7%
Ear warmth	2	8.7%
Protector fell from ear	2	8.7%
Mild discomfort	6	8.7%

1.5.3 Loudness and its impact on the perceived attenuation

> *Eventually important is the perception of sound. We do not perceive frequency, we rather perceive pitch; we do not perceive level, but loudness. We do not perceive spectral shape, modulation depth, or frequency of modulation; instead we perceive "sharpness", "fluctuation strength", or "roughness". We also do not perceive time directly; our perception is the subjective duration, often quite different from the physical duration.* Zwicker and Zwicker (1991)

Loudness is defined by Bech and Zacharov (2006) as the perceived magnitude of a sound. It is a psychophysical magnitude strongly correlated to the physical magnitude of sound pressure level: one does not directly feel sound pressure level, one feels a loudness sensation caused by sound pressure level. Since loudness is frequency and SPL dependent, but in a non-linear way, a uniform decrease in SPL at all frequencies composing a sound will not translate to a uniform decrease in loudness at all frequencies of the sound. According to loudness models, if one was to wear perfectly uniform attenuation earplugs and another was not, in the same given sound environment, they could feel different spectral balances: the relative difference in loudness between the frequency components would not be the same. In contrast, if a given earplug was not necessarily uniform in dB of attenuation, but was capable of producing uniformly decreasing loudness over the audio bandwidth, wearing or removing them would not have any effect on the perceived spectral balance. Since spectral balance assessment is used by musicians to blend their instruments together, adjust their playing, and even assess timbre, it is possible that the non-linearity of loudness perception is detrimental to the acceptance of uniform attenuation HPDs. However, loudness is not directly measurable, since it is a perceptual sensation, while physical attenuation is. Therefore, if such uniform loudness attenuation is achievable, a loudness model would be necessary to link the desired uniform perceptual attenuation to the desired physical attenuation. This section attempts to characterize the possible impact of the non-linearity of loudness perception on the auditory perception of a musician wearing HPDs.

1.5.3.1 Loudness of pure tones

Equal loudness contours are a representation of how loud a given sound stimulus will be perceived depending on its frequency and sound pressure level (SPL). Although subjective, the perception of loudness has been found to be generally consistent across human subjects within a given experiment. However, loudness is not trivial to measure, and practical considerations have led to different results among various experiments over the years, causing the exact shape of the curves to considerably vary since they were first studied by Fletcher and Munson (1933) and revised by Robinson and Dadson (1956). Recent convergence in several studies from 1987 to 2002, reviewed by Takeshima and Suzuki (2004), led to the current standard ISO226:2003 of the International Standardization Organization (2004), characterizing equal loudness contours for pure tones. Figure 1.28 shows the revised curves indicating how much SPL is needed at a given frequency for a pure tone to appear as loud as another pure tone at a different frequency, on the same curve. The numbers on the curves indicate the loudness level that a given curve represents; all the pure tones on a given curve have the same loudness level, expressed in phons, and 1 phon is equal to 1 dB(SPL) at 1 kHz only. A sensation that the loudness is doubled corresponds to an increase of 10 phons. The curve labeled MAF, meaning minimum audible field, represents the threshold of hearing.

Careful observation of the curves of figure 1.28 reveals that their shape is different depending on the loudness value that they represent, revealing the non-linearity of loudness perception that has been found in many loudness models, especially in the low frequencies. Similar trends were observed for narrow-band noise by Pollack (1952) and octave band noise by Stevens (1956). A decrease in SPL generally does not correspond to a decrease in loudness of the same magnitude.

The equal loudness contours can vary greatly from one individual to another, especially for frequencies above 1 kHz. Robinson and Dadson (1956) found that hearing loss, closely related to age, is one of the factors that causes variability from the mean curves.

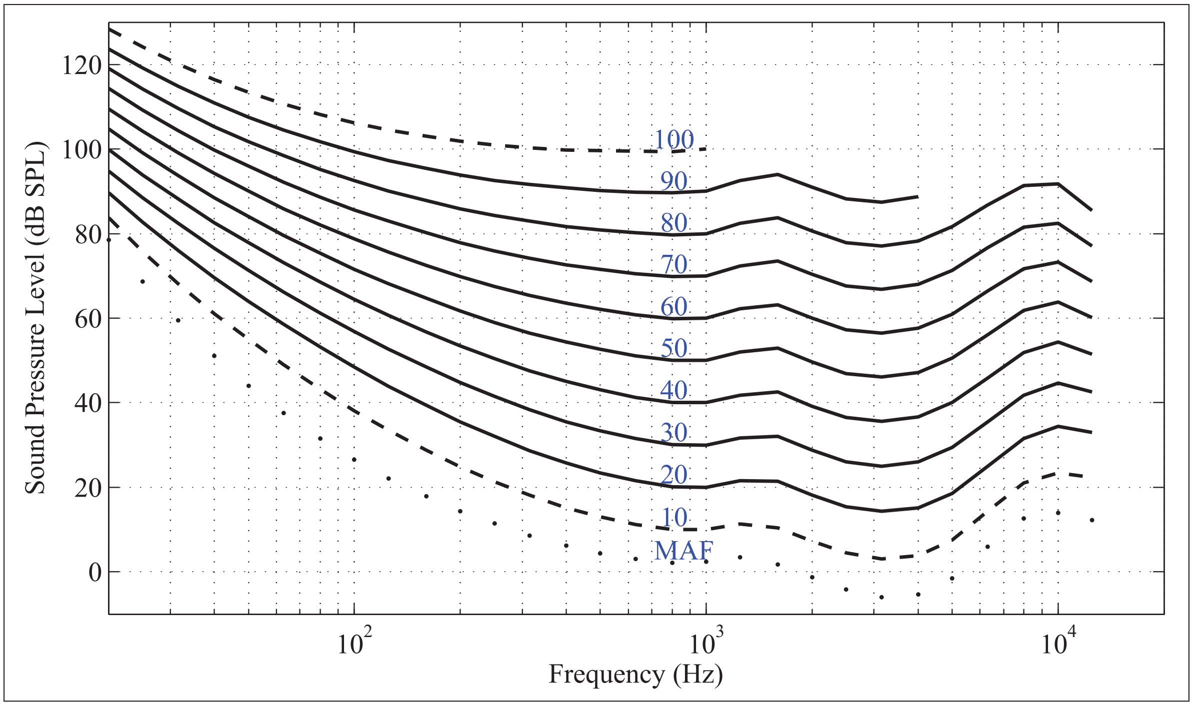

Figure 1.28 Equal loudness contours according to ISO226:2003
Permission to use section 4.1, section 4.2, figure A.1 from ISO 226:2003 was provided by Standards Council of Canada. No further reproduction is permitted without prior written approval from Standards Council of Canada.

The equal loudness curves of the standard ISO226:2003 were obtained under free-field conditions, in an anechoic room. Results reported by Keidser *et al.* (2000) indicate that, as mentioned by Florentine *et al.* (2011), *"low frequency stimuli do not sound as loud when presented in occluded ear canals even though levels have been equated using probe-tube measurements."* This introduces the possibility that loudness perception could be completely different when the ear is occluded, at low frequencies, and that the difference in the curves might not be maintained in occluded conditions. Keidser's finding falls in the category of "the missing 6 dB" phenomenon, a concern raised in 1949 by Leo. L. Beranek. A number of studies have concluded that at low frequencies, 6-10 dB more sound pressure level is required in the ear canal to elicit the same loudness sensation when using headphones than when using a loudspeaker box. Recently however, Völk and Fastl (2011) claimed that this phenomenon is attributable to inter-aural phase differences between the two conditions, loudness not being exclusively dependent on sound level. In the loudspeaker condition, the room, even if anechoic, and the individual listener's morphology have a particular effect on the time-function of the signal arriving in the listener's ear canal, causing particular inter-aural phase differences. In the headphone condition, however, the effect of the room and the morphology of the listener is almost bypassed. Völk and Fastl (2011) showed that, when using headphones and binaural synthesis to simulate a "virtual loudspeaker box", factoring in both the room and the individual morphology, the required pressure loudness was the same in the real loudspeaker conditions and the headphone conditions.

1.5.3.2 Loudness of complex sounds

Sounds encountered in the real world are seldom pure tones, but rather complex sounds composed of multiple components at multiple frequencies. A particular aspect of loudness is that it can stay constant even if there is a variation in the amount of energy in the sound. For example, two close pure tones can elicit the same loudness as one, and a noise of narrow bandwidth can elicit the same loudness as a noise of narrower bandwidth. As the bandwidth of a band-limited noise increases, loudness can stay constant until a certain bandwidth is exceeded, and only then does loudness summation occur. This phenomenon, observed by Zwicker (1961), has led to the concept of auditory filters.

It seems our ear analyzes a complex sound into small frequency intervals, known as auditory filters. Among the representations of the bandwidth of the auditory filters are the critical bands (CB) and the equivalent rectangular bandwidth (ERB). Zwicker and Zwicker (1991) state: "*The critical band concept is based on the well-proven assumption that our auditory system analyzes a broad spectrum in parts that correspond to critical bands.[...] Many experiments dealing with the loudness of sounds of different spectral widths have shown that the instruments our auditory system uses are the critical bands that shape and weigh the many partial loudnesses to be summed up.*" Zwicker's work led to a standardized method for calculating loudness, most often referred to as ISO532B.

Another standardized method, ANSI S3.4:2007, is based on the work of Moore *et al.* (1997). According to Florentine *et al.* (2011, p.127): *currently, the most widely used model of loudness summation is the excitation-pattern model proposed by Moore et al. (1997).*

The Moore *et al.* (1997) model involves the following steps to predict loudness:

1. Filter a given signal with a fixed filter to obtain the SPL at the eardrum from free field, diffuse field or headphones (effect of the external ear);

2. Filter the result with a fixed filter to account for the transformation of the signal by the middle ear;

3. Transform the signal to an excitation pattern, the equivalent of the signal that is exciting the cochlea, through the use of frequency and level dependent filters representing the auditory filters;

4. Transform the excitation to a *specific loudness*, the loudness per ERB, by taking into account factors such as the compression performed by the inner ear and the non-linearity of loudness perception at low frequencies, below 500 Hz;

5. Sum the specific loudness in different ways depending on whether presentation of the signal is monaural, diotic (same signal at both ears), or binaural.

Figure 1.29 shows an example of an auditory filter centered at 1 kHz and how its output level and shape vary with the input level. The auditory filters become increasingly sharp on the high frequency end and not as sharp on the low frequency end as the input level increases. Zwicker and Zwicker (1991) explain: *"we already know that a 1-kHz tone, although it has an infinitely small spectral width, does not lead to an infinitesimally narrow excitation in our auditory system [...] Instead, it results in an excitation over a range increasing with larger SPL values of the 1-kHz tone"*. This is taken into account by Moore and Glasberg (1983): *"The excitation pattern evoked by a given sound is the distribution of internal excitation as a function of some internal variable related to frequency. In terms of the filter bank analogy, the excitation pattern may be conceived as the output of each filter as a function of filter center frequency."*

Figure 1.30 shows an example of calculating the excitation pattern of a 1 kHz tone. Any tone that does not change this excitation pattern significantly (between 0.1 and 1 dB) will be masked and will not be considered to add magnitude to the perceived loudness, as explained in Moore (1996). The non-linearity of loudness perception in the low frequencies is taken into account by the use of variable gains applied to the output of the auditory filters depending on the excitation level, as shown in figure 1.31. The curves relate the gain that must be applied to relate the excitation level with the elicited loudness, in sones. Sones are another loudness unit: 1 sone is equivalent to 40 phons, and each step of 10 phons corresponds to a doubling of the sones, so that 2 sones is equivalent to 50 phons, 4 sones to 60 phons, etc.

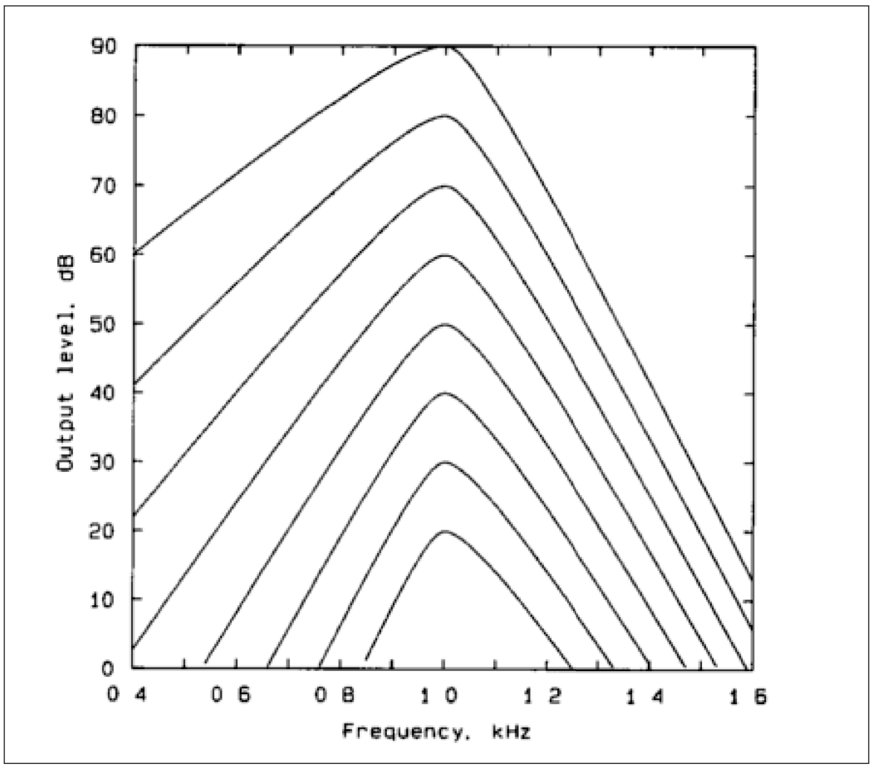

Figure 1.29 Shape of the auditory filter centered at 1 kHz, plotted for input sound levels ranging from 20 to 90 dB(SPL)/ERB
Reproduced with permission from Moore (1996)

As reviewed by Charbonneau (2010), stationary models, such as DIN45631, ISO532B and ANSI S3.4:2007 can be used to predict equal loudness contours following the same general trend as ISO226:2003 reference curves, although notable differences can be observed between models. Out of these models, the ANSI S3.4:2007 model, based on Moore *et al.* (1997), is the one most in agreement with ISO226:2003.

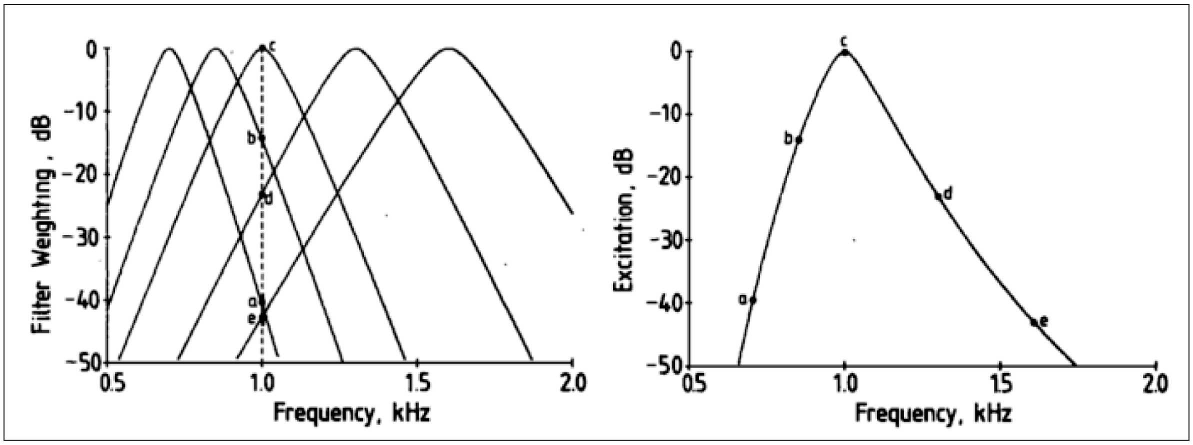

Figure 1.30 Example of deriving an excitation pattern from auditory filter shapes, for a tone at 1 kHz
Reproduced with permission from Moore and Glasberg (1983)

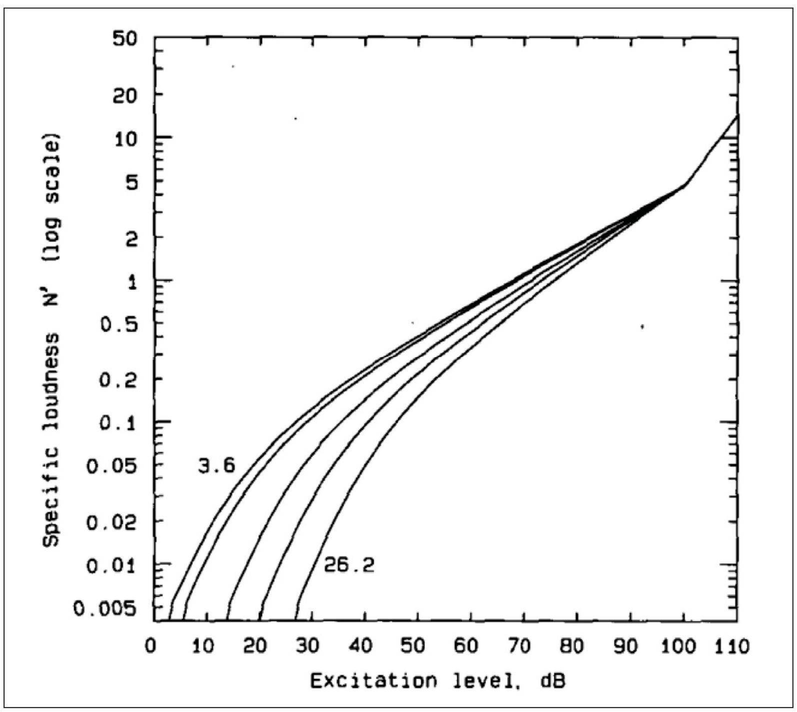

Figure 1.31 Curves relating specific loudness to excitation level where the curve labeled 3.6 applies for all frequencies above 500 Hz; curve 26.2 applies for a frequency of 52 Hz, and other curves correspond to frequencies of 74, 108 and 253 Hz
Reproduced with permission from Moore *et al.* (1997)

A few years following the model applicable to stationary complex sounds, Glasberg and Moore (2002) addressed a limitation of the model: *"Many everyday sounds, such as speech and music, are time-varying, and it would be useful to have a way of calculating loudness of such sounds."* The new model includes calculation of instantaneous loudness, short-term loudness and long-term loudness, by accounting for complex phenomena that are not covered in this work.

1.5.3.3 The impact of loudness and sound pressure level on music perception

Loudness mainly correlates with sound level, but it is also dependent on frequency, bandwidth, incident angle, duration, temporal envelope, spectral complexity, whether a sound is presented binaurally or monaurally and the presence of other sounds, as outlined by Toole (2008) and Florentine *et al.* (2011). Considering loudness in regard to hearing protection could be very similar to considering loudness in sound reproduction: an original sound is reproduced at a different listening level. The effect of the room and the listener's morphology are not bypassed by the HPD, and all factors affecting loudness, but sound pressure level, could be invariant in a hearing protection context.

It is a known phenomena that when uniformly attenuating over the audio bandwidth, low and high frequencies appear to be softer (lose more loudness) than mid frequencies, as reported in Huber and Runstein (2010). The non-linearity of loudness perception has caused sound engineer to apply correcting equalization when recording and mixing. Rumsey and McCormick (2009) explain: *"In practice, if a recording is replayed at a much lower level than that at which it was balanced it will sound lacking in bass and extreme treble – it will sound thin and lack warmth. Conversely, if a signal is replayed at a higher level than that at which it was balanced it will have an increased bass and treble response, sounding boomy and overbright.[...] Rock-and-roll and heavy-metal music often sounds lacking in bass when replayed at moderate sound levels because it is usually balanced at extremely high levels in the studio."* For this reason, it is recommended that mixing engineers work with monitor levels in the 75-90 dB(SPL) range, *"as they more accurately represent the listening levels that are likely to be encountered in the average home (i.e. the Fletcher-Mundson curves will be more closely matched)"*, according to Huber and Runstein (2010).

The same phenomenon has caused some home audio systems manufacturers to include a "loudness" feature in their products: it can be simply a button or a control enabling a boost of low and high frequencies, or only of the low frequencies; there is no standard for a loudness feature. Figure 1.32 shows an example of such a correction, where low and high frequencies are boosted depending on the volume setting. The need for loudness correction in the high frequencies is equivocal. Toole (2008, p.434) claims that *"many loudness controls mistakenly try to follow the shapes of the equal loudness contours rather than the differences in the shapes, and they boost the highs as well."*

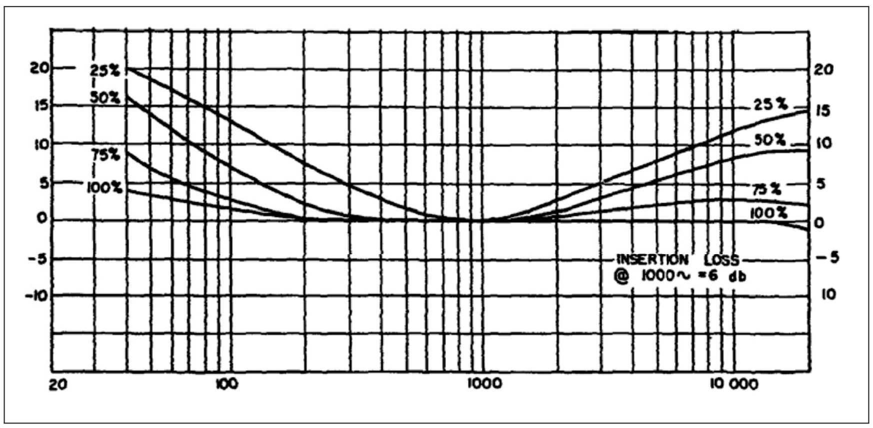

Figure 1.32 Example of frequency equalization performed by a loudness control on a home audio system, depending on the volume setting (%)
Reproduced with permission from Tremaine (1969)

Loudness compensation can also be much more complicated, but would require knowledge of the level at which the original sound was recorded, and the level at which it is being reproduced, to apply a suitable correction filter derived from equal loudness contours, according to Holman and Kampmann (1978). About the precision of such a compensation, Eargle (2003, p. 29-30) states: *"A familiar application of the variation in loudness contours is the loudness control found on most consumer receivers. This control automatically adds bass as you turn the level down. In most receivers the absolute tracking with the phon curves may not be exact, but it is in the right direction and will pretty much adjust the music spectrum so that balances seem natural, whatever the setting of the control."*

Since wearing uniform attenuation earplugs is analogous to turning the volume down, a similar phenomenon can be expected to happen with HPDs. Uniform attenuation of the audio bandwidth would result in a decreased perception of the low and possibly high frequencies compared to mid frequencies. The ideal attenuation, in a musical context, could then be such that it keeps the perceptual balance between the frequency components of the original sound.

1.6 Summary of the literature review

Musicians are noise exposed workers and should protect their hearing by using hearing protection when necessary. Studies on musicians' hearing loss have revealed signs of hearing loss in as much as 70% of cases, likely because of the high SPL that professional musicians are routinely exposed to and the duration of the exposure, but also because of the low usage rate of HPDs, ranging from as low as 6% to 29%. Many of the reasons commonly reported by musicians as to why they don't use HPDs have to do with perceptual discomfort caused by HPDs, impeding with the quality of their work. This perception shift is attributed to two effects: the occlusion effect and the isolation effect. The main purpose of this literature review was to gather information about these problematic effects to properly define the requirements of a solution: a hearing protection device that has minimal impact on the auditory perception.

1.6.1 Regarding the occlusion effect

The occlusion effect mainly manifests itself in the form of increased SPL in the occluded ear canal, mainly at low frequencies, and therefore it is common to measure the objective occlusion effect at 250 Hz and 500 Hz, where it is felt the most. In the literature, the objective occlusion effect is sometimes measured at 1 kHz as well, but is found to have less impact on the magnitude of the subjective occlusion effect. The occlusion effect can be reduced to some extent by the use of a deep insertion, but this solution often lead to physical discomfort problems. Since the SPL caused by occlusion effect is measurable in the ear canal, it can also be canceled using active noise control techniques. It was shown in section 1.3 that a closed-loop system can reject disturbances, in this case the ear canal SPL caused by occlusion effect, but also to track a command signal, meaning that it is possible to decide what command signal the system tracks, and reproduce sound in the musician's ear. When designing such a system, two parameters are of great importance. First, the performance of the system: its ability to reject disturbance and track a command signal. Second, the safety margins of the system: its robustness to uncontrollable changes and variability that can lead a closed-loop system into instability. These two parameters are related: a safe system offers less performance and vice versa.

It has been found in the literature that the performance of active noise control in an occluded ear is subject to great inter-user variability. The shape and size of the ear canal and the acoustic impedance of the eardrum directly affect the variability of the system. Another difficulty of in-ear active noise control is to find transducers small enough to fit in the ear canal while being suitable to active noise control.

1.6.2 Regarding the isolation effect

Three underlying causes have been found as likely contributors to the isolation effect. First, the modification of the ear-canal resonance caused by the occluding device results in a perception shift. Second, the uneven attenuation due to HPDs being generally less efficient at blocking low frequencies results in a perception shift. Third, the non-linearity of loudness perception at low and possibly very high frequencies would theoretically cause the perceived spectral balance of a sound to be changed when it is evenly attenuated, resulting in a perception shift.

The ER-15 and ER-20 are musician's earplugs that attempt to address the first two underlying causes of the isolation effect, summarized above, by using a network of acoustical elements. They aim at reproducing the transfer function of the open-ear, at a lower level, and hence provide a uniform attenuation. Studying the attenuation that they actually provide shows that they work according to specifications, on average, though the standard deviation shows that the attenuation can stray from uniform for a single user. While certainly being a step in the right direction, these earplugs do not seem to resolve the problem for many musicians. Complaints of modified auditory perception are being reported by a great majority of musicians in the studied groups.

CHAPTER 2

DEFINITION OF THE REQUIREMENTS OF THE SYSTEM

Following the literature review, many unknowns were identified in the details of the puzzle of providing satisfactory HPDs for musicians. The information needed to better define the requirements of a system that would negate the detrimental impact of the occlusion and isolation effects are listed below:

1. The SPL in the ear canal, its frequency distribution and the magnitude of the objective and subjective occlusion effect resulting from occluding a musician's ear as he is playing an instrument mechanically coupled to the head;
2. The effect of decreasing the SPL in the musician's ear canal on the magnitude of the subjective occlusion effect, while leaving all other variables unchanged;
3. The exact underlying acoustic and psychoacoustic phenomena that cause some musicians to find that wearing specialized uniform attenuation HPDs results in missing dynamics, timber, and nuances information;
4. The applicability of loudness compensation to hearing protection.

To design a system that would really solve the detrimental impact of the occlusion and isolation effects, these unknowns would have to be characterized to fully define the details of the problem affecting musicians and provide a solution that suits their needs. As a consequence, the work in this master's thesis has to be based on assumptions and hypotheses, derived from what is available in the literature.

The objective of this work is to design and implement an electro-acoustic system that *aims* at negating the detrimental impact of the occlusion and isolation effects, and that *does* provide the basis of a system allowing characterization of the unknowns and testing of compensation strategies, through further research.

This chapter defines the requirements, constraints and design choices of the system to be designed in this project. The envisioned system is composed of an earpiece that interfaces with the ear and a belt pack that contains the electronics and digital signal processors. Two design choices, made *a priori*, are described in section 2.1. The first design choice is that the earpieces are compatible with the SonoFit^{TM} technology. The second design choice is that the belt pack contains the Auditory Research Platform. Sections 2.2 and 2.3 define the respective requirements of the solutions to the occlusion and isolation effects.

2.1 Design Background

This section describes two elements that have a significant impact on the design. First, the SonoFit technology, a process allowing for a quick and personalized physical interface to the ear, is described in section 2.1.1. Second, the Auditory Research Platform, a re-configurable embedded system providing the hardware means to implement prototypes of in-ear technologies and conduct in-ear research is described in section 2.1.2.

2.1.1 The SonoFit technology

The SonoFit technology, developed by Sonomax Technologies Inc., allows for the making of custom molded earpieces within a few minutes and at a lower cost and without involving an audiologist. The principle of the SonoFit technology is illustrated in figure 2.1. The process is analogous to inflating a tiny balloon within one's ear canal to match its shape. An underbody is enveloped by an inflatable skin that is filled with liquid medical grade silicon until it matches the ear canal it is inserted in. The liquid silicon hardens after a few minutes, leaving a permanent custom mold, in which an underbody provides space for the housing of transducers and electronics. A disposable fitting system is provided in the shape of a headband containing pumps that fill the inflating skin with silicon. The headband pushes silicon into the inflating skin until an equilibrium is reached, at which point excess silicon flows back in the pump, preventing over-inflation of the inflating skin. Once the silicon has hardened, the earpiece is detached from the headband and a back cover, called a face plate, is clipped over the silicon intake so that a finished custom-fit earpiece is obtained.

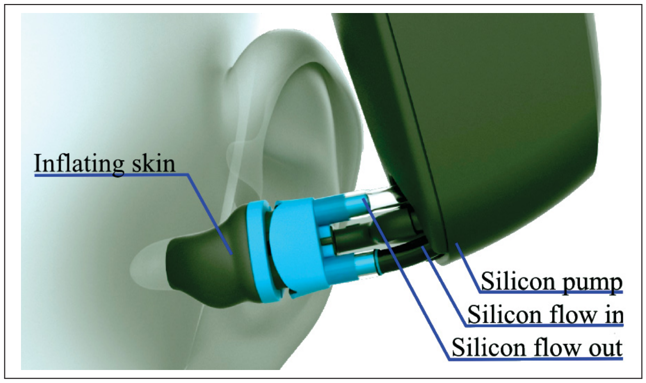

Figure 2.1 The SonoFit technology

Two additional views of an earpiece featuring the SonoFit technology, the Sonomax V4 earpiece, are depicted in figure 2.2. The earpiece a. is attached to the headband and ready to be inflated in the ear canal. The earpiece b. is shown with the faceplate after it is detached. The inflating skin has been removed from earpiece b. to show an important part of the earpiece: the underbody. The available space in the underbody of the Sonomax V4 earpiece imposes constraints on the size and shape of the transducers that can be used for this project.

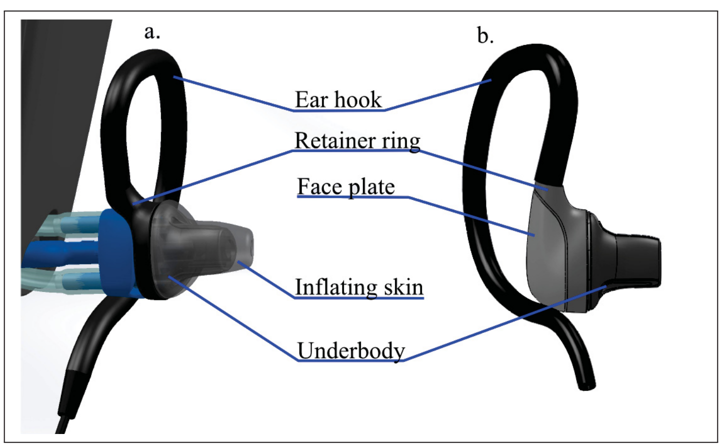

Figure 2.2 Parts composing the Sonomax V4 earpiece: a) ready for custom moulding; b) without inflating skin

2.1.2 The Auditory Research Platform

Through its own research and partnerships with entities such as the International Laboratory for Brain, Music and Sound Research (BRAMS) and the Centre for Interdisciplinary Research in Music, Media and Technology (CIRMMT), CRITIAS pinpointed the need for an Auditory Research Platform (ARP), a platform that would provide the hardware means to conduct research involving in-ear technologies.

General design criteria for the ARP project, initiated in 2011 by the author and colleagues, were portability, flexibility and simplicity of use. At the time of writing, the project is at the final stage of its first implementation, where peripherals and supporting functionalities are being finalized, such as SD^{TM} card interfacing and communication with Android^{TM} devices through Bluetooth^{TM}. A block diagram of the concept of the ARP is depicted in figure 2.3: earpieces are connected to a re-configurable belt pack providing signal processing capabilities. The earpiece can contain multiple transducers depending on the application. The earpieces of the block diagram of figure 2.3 feature an internal loudspeaker (ILS), an in-ear microphone (IEM) and an outer-ear microphone (OEM). Real pictures of the auditory research platform are included in appendix V.

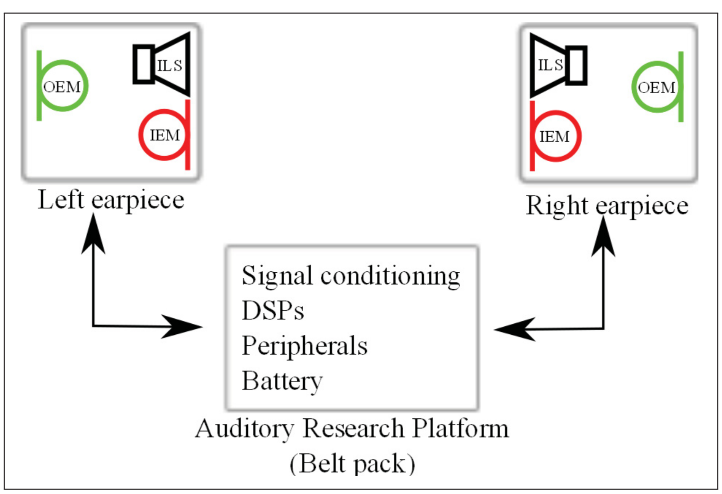

Figure 2.3 Block diagram of the concept of the Auditory Research Platform; a belt pack providing signal processing capabilities is connected to earpieces containing transducers

Because the potential solutions to the occlusion effect and isolation effect developed in this work are likely to be beneficial in applications other than musician's hearing protection, it was decided that the design must use and be complementary to the ARP. This imposes constraints on the design, arising from the following characteristics the ARP:

1. The DSPs of the ARP were partly chosen because they provide an user-friendly block diagram programming interface suitable for non-programmers while permitting programmers to code specialized algorithms using an assembly language. This allows a simplicity of use in research domains where programming is not part of the traditional skill set, like audiology and psychoacoustics. On the other hand, the DSPs have an inherent 1 ms delay in the signal chain from input to output that can be detrimental to some applications, like active noise control;

2. The portability design criterion of the ARP led to a belt pack measuring about 80 mm by 50 mm by 20 mm. The electronics were arranged to leave space of about 50 mm by 30 mm by 10 mm to fit both the battery and the electronics specific to the active HPD for musician design.

The hardware of the ARP is complemented in this project to allow its use for occlusion effect reduction and to implement the solution to the isolation effect.

2.2 Requirements of the solution to occlusion effect

In a musician's hearing protection context, where the ears have to be completely occluded, the most promising solution is active noise control of the sound resulting from occlusion effect. This novel approach needs only shallow insertion of the device and is thus more comfortable for the user while providing occlusion effect reduction. Active control in the ear to reduce the occlusion effect has the side-effect of providing active attenuation that sums up to the passive attenuation of a physical HPD, providing more overall attenuation that is likely to be beneficial to the application. Therefore, an active occlusion effect reduction (AOER) system is designed as part of this master's thesis. This section defines the performance requirements of the AOER

(2.2.1), the controller requirements (2.2.2), and the requirements of the earpiece to be inserted in the ear canal, forming the plant (2.2.3).

2.2.1 Performance requirements

Regarding occlusion effect reduction, the ideal scenario would involve a system that precisely cancels the occlusion effect for each given musician and his instrument. However, the uneven performance of the AOER system reported by Mejia *et al.* (2008) indicates that compromises have to be made between reduction and regeneration, and the solution should be such that it is most efficient where the occlusion effect is most felt in the musicians' case.

General guidelines to define the target performance for the solution can be derived from the literature. In an experiment performed by Kuk *et al.* (2005), involving subjective characterization of the occlusion effect, subjects generally found 5-7 dB of occlusion effect to sound natural. Moreover, Henry and Letowski (2007) state: *"In the evaluation of hearing aids, hearing protectors, and in-the-ear devices, the occlusion effect is typically measured at 250 Hz or 500 Hz and is considered negligible if it is less than 10 dB, mild to moderate when it is between 10 and 20 dB, and severe when it is larger than 20 dB."* This statement provides a first criterion for the target performance of the AOER, which should be such that the remaining OE is below 10 dB. Since the envisioned prototype HPD is meant to be shallowly inserted, to allow a comfortable fit to the ear canal, a good estimation of the initial OE that occluding the ear with the inactive prototype HPD should cause can be obtained by reviewing OE values for similar insertions in the literature. In their study, Dean and Martin (2000) compared different OE for a shallowly inserted earphone and deeply inserted earphone. The OE values reported for a shallow insertion earphone provide a reference for the initial OE that need to be reduced. Since occluding the ear with a deeply inserted device is reported as an acceptable solution to reduce occlusion effect, an acceptable target residual occlusion effect when the prototype HPD is active can be obtained from the occlusion effect caused by deeply inserted devices. The OE values reported for a deep insertion earphone are used as the target residual OE: the solution should be such that it causes no more occlusion effect than a deeply inserted device, which is less than 10 dB in this case. Table 2.1 summarizes, for three important frequencies: The reference OE of a shallowly inserted device, the minimum target residual occlusion effect (ROE) and the consequently required active occlusion effect reduction (AOER).

Table 2.1 Defining the required performances of the occlusion effect reduction system

Frequency (Hz)	**Reference OE (dB)**	**Target ROE (dB)**	**Target AOER (dB)**
250	17	9	8
500	14	8	6
1000	6	-1	7

The OE values reported by Dean and Martin (2000) are measured by obtaining the difference between hearing threshold values of bone conducted sound when the ear is occluded and when it is open, and provide no indication as to the reduction of SPL in the ear canal that should cause the desired OE reduction. To define a SPL reduction target, the author assumes a perfect relationship between the SPL in the ear canal and occlusion effect in hearing threshold difference; for example, it is reported by Fagelson and Martin (1998) that at 250 Hz and 500 Hz, there is less increase in ear canal SPL than there is in OE perception when the ear is occluded, so a sensible approximation could be that decreasing ear canal SPL by 1 dB will result in OER of at least 1 dB or more. On the other hand, at 1 kHz, there is more SPL increase in the ear canal than there is in OE perception, so a sensible approximation could be that an increase in ear canal SPL by 1 dB will result in an increase of OE perception by 1 dB or less. This last approximation would be consistent with the results obtained by Mejia *et al.* (2008), where as much as 9 dB of amplification of the SPL in the ear canal at 1 kHz did not nullify the benefits of the AOER system. Indeed, the occlusion effect reduction system measured by Mejia *et al.* (2008) provided 15 dB of reduction of ear canal SPL at 300 Hz, at the expense of 9 dB regeneration at 1 kHz, and was preferred by 10 subjects out of 12. Both approximations are used to obtain the necessary SPL reduction to achieve a desired OER in a 1:1 ratio.

Results reported by Mejia *et al.* (2008) seem to indicate that reduction at 250 Hz and 500 Hz is more important towards reduction of the perceived OE than reduction at 1 kHz, and that the minimum requirements at 1 kHz in SPL reduction could be decreased to allow for some regeneration.

The final requirements, in SPL occlusion effect reduction, are summarized in table 2.2. The minimum required SPL reduction in the ear canal to cause the minimum required OE reduction is modified to account for the low perceptual effect of 1 kHz regeneration deduced from Mejia *et al.* (2008), and is referred to as minimum target SPL OER. The desired maximum SPL reduction in the ear canal that would cause the maximum desired OE reduction is referred to as maximum target SPL OER. If the device achieves maximum target SPL OER, the SPL in the ear canal when a subject is speaking would ideally be independent of whether he is occluded with the device or not. A median target SPL OER is provided to define a target between minimally acceptable performance and perfect performance.

Table 2.2 Minimum and maximum performance requirements of the OER system to be designed, in target reduction of ear canal SPL caused by OE

Frequency (Hz)	**Min. Target SPL OER (dB)**	**Med. Target SPL OER (dB)**	**Max. Target SPL OER (dB)**
250	8	12.5	17
500	6	10	14
1000	-9	-1.5	6

It is important to note that the information on OE that was used to define guidelines pertains especially to speech stimulation or forehead bone vibration stimulation at specific frequencies, and that some of this information might be inapplicable to musicians in general or to some musicians. For example, the occlusion effect is generally measured from 250 Hz in the literature because high ear canal SPL below 200 Hz has little contribution to the perceived OE, as reported by Henry and Letowski (2007). This assumption might not be applicable if a musician is playing an instrument that produces significantly more energy below 200 Hz than human speech, such as a tuba. Nevertheless, the requirements obtained in this section seem like a reasonable target, given the status of the literature on occlusion effect experienced by musicians.

To meet these performance requirements, the requirements for a suitable controller must be defined.

2.2.2 Controller requirements

Because the signal resulting from occlusion effect can practically only be acquired in the ear canal, where it needs to be canceled, a feedback control is chosen for the application. Because of their audio latency of 1 ms, the DSPs included in the ARP are not suitable for active control. The delay would cause a digital control implemented on these DSPs to behave poorly. Therefore, a dedicated hardware has to be included in the space left in the enclosure of the ARP, as discussed in section 2.1.2. Two options are considered:

1. A feedback analog ANC implementation. The hardware is likely to fit in the allowed space and the cost of prototyping is low. The circuits are relatively simple and the typical performance of feedback analog ANC meets the requirements. Their typical performance at rejecting disturbance is in the scale of 10-20 dB, according to Rudzyn and Fisher (2012);

2. A feedback digital ANC implementation. The hardware is likely to be more difficult to fit the allowed space and the cost of prototyping is higher. Typical performance also meets the requirements, as Zimpfer (2000) demonstrated, and the circuit is easily reconfigurable. While the theory is simple, the practical implementation is more intricate than analog ANC.

The feedback analog ANC option is selected because it theoretically meets the requirements at a lower cost and involves less practical implementation problems, thus providing a good starting point for active occlusion effect control and providing a mean to quantify the effect of SPL reduction in the ear canal on perceived occlusion effect in further research.

2.2.3 Plant's requirements

As previously stated, the plant is a crucial part of an active control system. In the case of active occlusion effect reduction, the plant is composed of two elements that can be controlled:

1. The physical interface to the ear and the body of the earpiece, imposing constraints on the transducers that can be inserted inside the earpiece. This is investigated in section 2.2.3.1;

2. The specific loudspeaker and microphone forming the assembly. This is investigated in section 2.2.3.2.

There is one uncontrollable aspect that affects the plant: inter-user variability of the ear canal, or more specifically its shape, its residual volume when occluded and the impedance of the eardrum. Therefore, safe gain and phase margins requirements need to be defined so that these uncontrollable elements do not render the AOER system unstable. Requirements for gain and phase margins are defined in section 2.2.3.3.

2.2.3.1 Physical interfaces to the ear

Because the device is a hearing protection device, a good acoustic seal is required to protect the user from the high SPL of its environment. There are two main ways of interfacing with the ear when an acoustic seal is required:

1. A custom-fit approach, usually providing physical comfort and a good acoustic seal in one's ear. Since the custom-fit is shaped to the ear canal for a specific insertion depth, a good reproducibility of the acoustic seal and residual occluded volume of the ear canal should be observed. However, a custom-fit approach is usually expensive compared to other solutions, and fits only one ear;

2. A generic-fit approach, traditionally through the use of universal-fit eartips, usually providing less physical comfort than a custom-fit and achieving a good acoustic seal only in certain ears. The reproducibility of the acoustic seal and residual occluded volume is expected to vary. However, a generic-fit approach is usually less expensive and one earpiece built with this approach, along with a wide choice of different universal-fit eartips, could permit a reasonable fit in a great number of ears.

In a hearing protection context, the quality of the acoustic seal is very important if the user is to be protected from ambient sound. Moreover, in an occlusion effect reduction context, the variability of the seal and residual occluded volume has a direct impact on the variability of the plant for a given user. These considerations indicate that a custom-fit approach would be better suited for the application, as preliminary test in appendix I confirms this. However, given that it is envisioned that the prototype developed in this project will be used in further research to collect experimental data and perform perceptual testing on a large number of human subjects, a large number of ears will have to be physically interfaced with. A custom-fit approach would require that a pair of earpieces be built for every subject, considerably increasing the cost of the test, even when considering the SonoFit technology, discussed in section 2.1.1. This practical concern would lead to reject the custom-fit approach and adopt the generic-fit approach for the first stage of the prototype considering that a large number of human subject will need to be tested; In a laboratory setting, the quality of the acoustic seal and residual occluded volume between insertions could always be monitored.

From these contradictory requirements of achieving good reproducibility and testing a large number of subjects at a low cost, it is decided that the first prototype features a generic-fit approach, but that its volume and shape be such that it is compatible with the SonoFit technology, to allow for a quick turnaround in later stages. The prototype earpiece's shape is therefore derived from the Sonomax V4 earpiece, currently having the greatest internal volume available in the underbody among earpieces featuring the SonoFit technology, and thus imposing the least constraint on the size of the transducers that can be used. Two options are then considered for a generic-fit approach:

1. The modification of a Sonomax V4 underbody to use universal-fit eartips;
2. The use of the Sonomax V4 underbody without the inflating skin, by using moldable soft silicon, that does not harden, to take the shape of a given ear canal.

Both options seem sensible, the first option involves only minor modification in the design of the Sonomax V4, and a selection step where a subject would try different universal-fit eartips and select to most suitable one, if any. The second option is certainly the least conventional but the most straightforward, as the existing underbody can be used as is. The quality of the acoustic seal that would be achievable using this method is likely to be good, as the soft silicon approximates a custom-fit. However, since it is not a true custom-fit and that the silicon does not harden, reproducibility of the acoustic seal is expected to be poor; depending on the quantity of soft silicon that is used, the insertion depth and whether the acoustic seal is achieved at the tip or the base of the underbody, changes in occluded volume will occur and introduce variability in the plant. Both options of a generic-fit are retained and therefore two types of earpiece prototypes are built in this project. The first option is referred to as "universal-fit" and the second option as "moldable-fit".

2.2.3.2 The internal loudspeaker and microphone components

Moving-coil loudspeakers, while not being optimal for ANC because they exhibit electro-mechanical resonances, have been used in many ANR headsets and applications. Their large diaphragm permits suitable low frequency reproduction at high sound pressure levels. They are generally inexpensive and widely available commercially. Bauer (2000) concluded that loudspeakers based on piezoelectric technologies were the most suited for active noise reduction. Indeed, publications on ANR using this technology show promising frequency responses but involve the design of a custom-made piezoelectric loudspeaker, as in Lyon (2008); their off-the-shelf availability for in-ear technologies is close to null. The development of such a loudspeaker is beyond the limitations of this work, therefore, piezoelectric transducers are not considered. Considering the motivations in their design outlined in section 1.4.2.2 and the fact that they are generally expensive, balanced-armature loudspeakers are not chosen over moving-coil. Electrostatic speakers simply do not have the required SPL at low frequencies due to their small excursion and are uncommon. These considerations lead to the choice of an inexpensive moving-coil loudspeaker that features a large diaphragm and can provide around 120 dB(SPL).

Regarding the microphone, the application requires a low noise floor, a high SNR, uniform frequency response, and a small package and shape suitable for insertion in the underbody of the earpiece. An electret microphone was found to fulfill these requirements, however at a relatively high price of 10 to 20 times that of the loudspeaker. In 2011, when the microphone was selected, commercially available MEMS microphones could not offer comparable characteristics. In future work, the use of MEMS microphones should be considered.

2.2.3.3 Gain and phase margins requirements

Large uncertainties are expected in the plant, especially if a generic-fit approach is undertaken. The main expected source of uncertainties are the inter-individual difference in the ear canal shape, the residual occluded volume of the ear canal and the eardrum impedance. The measurements of the plant response in various ear canals, reported by Goldstein *et al.* (2005), reveal differences in magnitude as great as 15 dB in the very low frequencies, and 6 to 8 dB in higher frequencies. As the gain margin is usually determined by the magnitude where the plant is out of phase, likely at higher frequencies, the gain margin requirements must account for the 6 to 8 dB of expected variability. This is consistent with the general recommendation of Haugen (2009) to require a gain margin between 6 dB and 12 dB and a phase margin between 30° and 60°. Elliot (2001) reports: *"Wheeler (1986) also conducted a series of experiments in which he measured the response of an active headset on a number of different subjects and found that the variation in the amplitude of the plant response from 1 kHz to 6 kHz was about ±3 dB, and the variation in the phase of the plant response was about ±20°."* While this statement gives a general indication of what the gain and phase margins should be, the plant uncertainties of an ANR earplug are expected to be greater than for an ANR headset, as explained in 1.4. Considering this information, the minimum gain and phase margins requirements for this project are set to 6 dB and 30°. No upper limit is defined.

2.3 Requirements of the solution to isolation effect

Following the literature review, two factors are identified as a cause of isolation effect when wearing hearing protection devices: the modification of the ear canal resonance and a non-uniform attenuation. Solutions for these factors have been presented in the form of commercially available musician's earplugs. However, studies on their field performance and surveying of musicians' opinions tend to indicate that either the attenuation they provide is not uniform enough, possibly because of a lack of follow-up, or that the attenuation is indeed uniform, and that other factors contribute to the isolation effect. The exact underlying acoustic and psychoacoustic phenomena that cause some musicians to find that wearing specialized uniform attenuation HPDs result in missing dynamics, timber, and nuances information are not found in the literature.

If the attenuation provided by musician's earplug is indeed uniform, the non-linearity of loudness perception could explain why some musicians are reluctant to wear specialized uniform HPDs. A literature review on the non-linearity of loudness perception and loudness models indicates that a perfectly uniform attenuation would not be perceived as perfectly uniform.

Since the exact causes of the problem are unknown, attempts at designing an exact solution would be futile. Therefore, the requirements derived in this section are guidelines, so that they cover a wide range of possibilities. In this regard, the versatile ARP is well suited as a platform to house the isolation effect solution. Requirements for three compensations that are likely to diminish the isolation effect are derived in this section; a variable attenuation (2.3.2), a uniform attenuation (2.3.1) and a loudness compensation (2.3.3). Moreover, the solution to the isolation effect should be compatible with the solution to the occlusion effect. The requirements of the first iteration of the complete isolation effect solution are summarized as follow:

1. The solution should be compatible with the solution to the occlusion effect;
2. The transfer function of the HPD must be capable of mimicking the response of an open ear, at a lower level, which is equivalent to a uniform attenuation;
3. The HPD must provide user-selectable, variable attenuation between 0 and 30 dB;
4. The solution must include a loudness correction algorithm that shapes the attenuation of the HPD to consider the non-linearity of loudness perception.

2.3.1 Uniform attenuation

As proposed by Voix and Laville (2005), the system should be able to implement something similar to a "mirror" filter: a filter that would correct the passive attenuation of the HPD so that it is uniform. The "mirror" filter should include a compensation strategy for the open ear and occluded ear resonances that were discussed in section 1.5.1. This involves an external microphone to pick up the sound external to the HPD, a DSP to process the acquired sound so that the attenuation of the HPD is uniform, and an internal loudspeaker to play back the acquired sound, as depicted in figure 2.4. An internal microphone could eventually be used to validate or adapt the transfer function of the HPD.

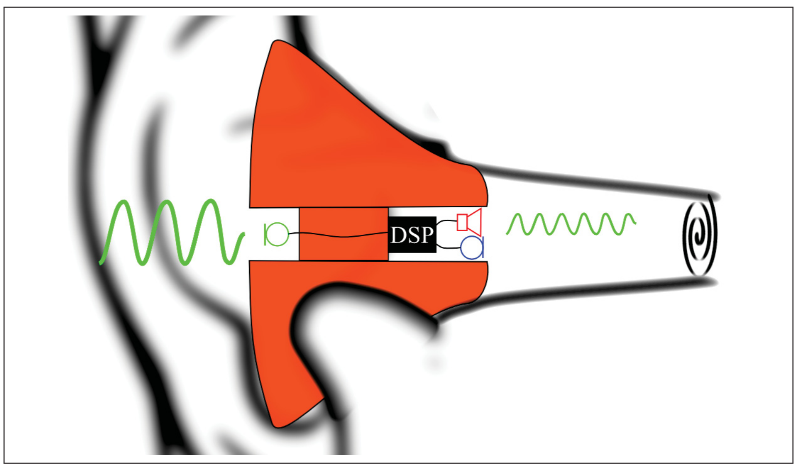

Figure 2.4 Providing uniform attenuation through the use of an active HPD

To achieve uniform attenuation, the transfer function of the HPD should be such that it mimics the response of an open ear, but at a lower level. Since the envisioned HPD only blocks the ear canal, the majority of the effect of the pinna should be maintained when the device is inserted.

The final requirement concerning uniform attenuation, for first iteration of the solution, is that it mimics the transfer function of an open ear canal, at a lower level.

2.3.2 Variable attenuation

According to Patel (2008), classical musicians are exposed to sound pressure levels between 80 dB(A) and 110 dB(A) while rock/pop musicians are exposed to higher sound levels, between 88 dB(A) and 117 dB(A). It is assumed that the lower range of the sound pressure level exposure will occur when a musician is practicing, and the higher range, when a musician is performing. According to the survey of Hagberg *et al.* (2005), musicians may practice for more than 20 hours a week, or 4 hours a day. The length of a performance may be two hours. Levels of 80 dB(A) to 100 dB(A) for 4 hours a day would require an attenuation anywhere between 0 and 12 dB to follow the recommendation of NIOSH (88dB(A) for 4 hours). Levels of 100 to 117 dB(A) for 2 hours would require an attenuation anywhere between 9 dB and 26 dB. The author hence believes that no fixed attenuation could be suitable to protect the hearing of all musicians without eventually being exaggerated in certain situations and insufficient in others. Even one musician may be exposed to a wide variety of sound pressure levels depending on his activities, his musical preferences, his number of practicing hours, the particular instrument that he plays at a given time and the room in which he plays.

Therefore, it is decided that the device should be capable of providing a wide range of attenuation values, ideally from close to zero to approximately 30 dB. Whether it would be best that the device adapts its attenuation automatically to the sound pressure level, promptly or slowly, or that the musician decides what attenuation he needs before using the device is out of the scope of this work. In this regard, the chosen requirement for the exact solution is that it is re-configurable and provides the means to implement almost any derivation of sensible attenuation management strategy. This requirement is fulfilled by the ARP.

The final requirement concerning the magnitude of the attenuation of the first iteration of the solution is that it can provide any user-selectable attenuation between 0 and 30 dB.

2.3.3 Requirements of the loudness compensation solution

In section 1.5.3, the impact of non-uniform loudness perception on musicians wearing hearing protection was explored. While no definitive conclusions are formulated, the fact that it is mentioned the sound engineering literature as a significant factor to be wary of when working with recorded music indicates that its impact could be of importance for musicians.

While many agree that there is a need for a loudness control in the low frequencies, loudness correction in the high frequencies is equivocal. When devising loudness control by using the differences between curves of the standard ISO532B, Holman and Kampmann (1978) found that no high frequency compensation was needed. Moreover, the ANSI S3.4:2007 standard derived from Moore *et al.* (1997) explicitly compensates for the non-linearity of loudness perception below 500 Hz, but not for the high frequencies. This indicates that no loudness control in the high frequencies would be needed. However, Huber and Runstein (2010), Rumsey and McCormick (2009) and Eargle (2003) all mention a decrease in the perception of high frequencies compared to mid frequencies as sound levels go down. For that reason, and the fact that several musicians complain about missing high frequencies when wearing uniform attenuation HPD, it is decided that a loudness compensation for high frequencies would be included in the algorithm: it can always be turned off if psychoacoustic tests reveal that it is unsuitable. Equal loudness curves of ISO226:2003 show explicit non-linearities at low and high frequencies, and equal loudness contours for pure tones are most often referred to by music recording literature and sound reproduction. Therefore, the hypothesis that it is suitable to derive loudness compensation schemes, applicable to music, from equal loudness contours for pure tones is adopted for the first iteration of the device. The proposed strategy for loudness compensation is similar to that of Holman and Kampmann (1978), in that the differences between equal loudness curves are to be compensated.

Compared to the method from Holman and Kampmann (1978), there is a different challenge in the implementation in this work. The envisioned active HPD method derived thus far would theoretically be capable of providing continuous values of uniform attenuation until a maximum is reached. Therefore, the loudness compensation algorithm should be able to adapt to any given value of attenuation, and not just in discrete steps. The Holman and Kampmann (1978) method was separate from volume control and continuous, since it used an analog circuit.

The loudness compensation method derived in this work should be capable of adapting to any selected attenuation while being digital. This poses a technical challenge, as it would be much easier to derive loudness compensation for discrete values of attenuation. Yet, the method must be simple enough not to require too much processing power and update itself quickly.

The final requirement concerning loudness compensation, for first iteration of the solution, is that it provide uniform attenuation in perceived loudness, for any attenuation value that the HPD can provide, using ISO226:2003.

CHAPTER 3

A SOLUTION TO THE OCCLUSION EFFECT

This section presents the design and implementation of a HPD featuring a prototype of an occlusion effect reduction system. It is divided as follow: proposed architecture and design (3.1), implementation experimental validation (3.2) and discussion and conclusions on the solution to occlusion effect proposed in this work (3.3). The requirements for the solution to the occlusion effect, discussed in section 2.2 are summarized below:

1. The use of an analog feedback control;
2. SPL reduction in the ear canal between 8 and 17 dB at 250 Hz, 6 and 14 dB at 500 Hz and -9 and 6 dB at 1 kHz, where -9 dB in reduction corresponds to 9 dB in regeneration;
3. Minimum gain and phase margins of 6 dB and 30° respectively, to ensure closed loop stability in most cases;
4. The use of an inexpensive moving-coil loudspeaker with large SPL capabilities and an electret microphone with a low noise floor, uniform frequency response, and of small size;
5. A physical interface of the earpiece to the ear canal that uses two options for a generic-fit approach: "universal-fit" and "moldable-fit".

3.1 Proposed architecture and design

The proposed architecture of the system for active control of the occlusion effect is presented in figure 3.1. The two main elements are the plant, composed of the internal microphone and loudspeaker assembly inserted in an ear canal, and the controller, composed of the compensator, the microphone pre-amplifier, and the negative feedback. This section presents two prototypes based on two different earpieces, referred to as moldable-fit and universal-fit earpieces or plants.

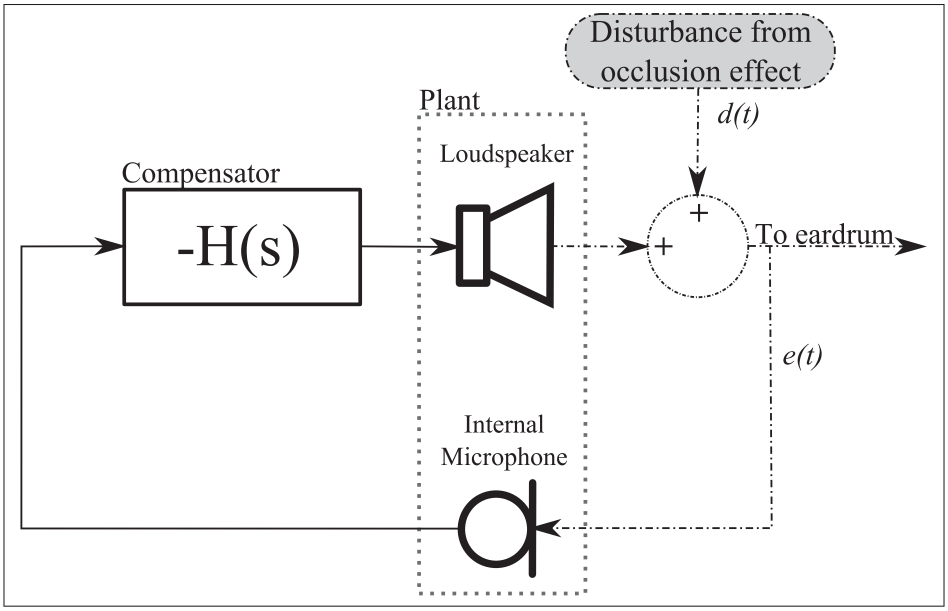

Figure 3.1 Architecture of the proposed system for active occlusion effect reduction

3.1.1 Moldable-fit earpiece

The moldable-fit earpiece uses the underbody of a Sonomax V4 earpiece, without inflating skin. As discussed in section 2.2.3.1, it can be coupled to a variety of ear canal with the use of moldable soft silicon that does not harden. Figure 3.2 shows a cross-section of the moldable-fit earpiece, and illustrates the placement of the internal microphone and loudspeaker assembly. Glue is used as an inert filling to provide an acoustical seal in the underbody and to reduce the volume in front of the loudspeaker before the opening of the underbody for better sound transmission to the ear canal. A picture of the moldable-fit earpiece is available in appendix V.

The moldable-fit plant response, the transfer function between the loudspeaker and microphone when the moldable-fit earpiece is inserted in an ear canal, is modeled by a FIR filter using a system identification procedure, described in appendix II. Figure 3.3 shows the modeled frequency response of the moldable-fit plant. A close look at the plant response reveals a relatively uniform phase up to approximately 2 kHz, although it is not minimum phase. High amplitude is also observed at high frequencies where the plant response becomes out of phase, potentially detrimental to the stability of the ANC.

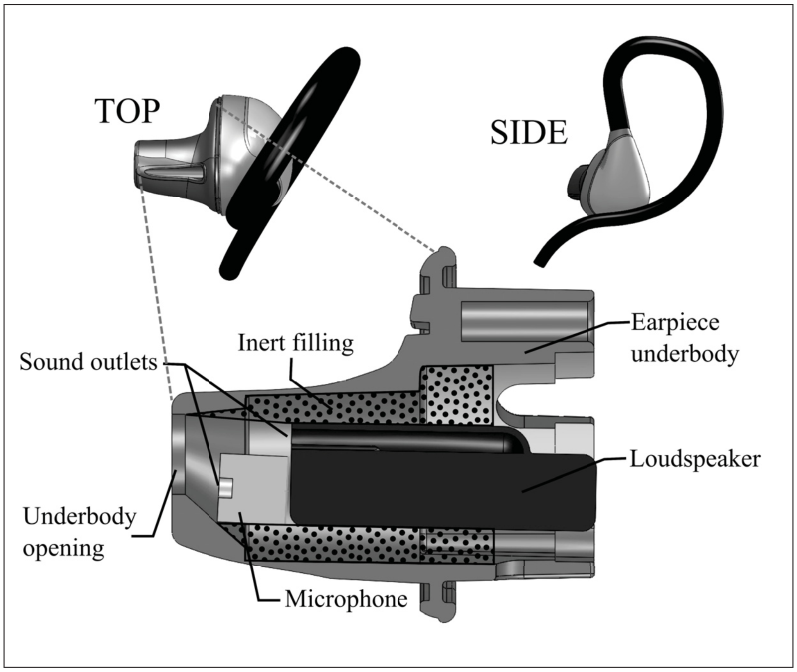

Figure 3.2 Top, side, and cross-sectional view of the moldable-fit earpiece, showing the internal loudspeaker and microphone assembly

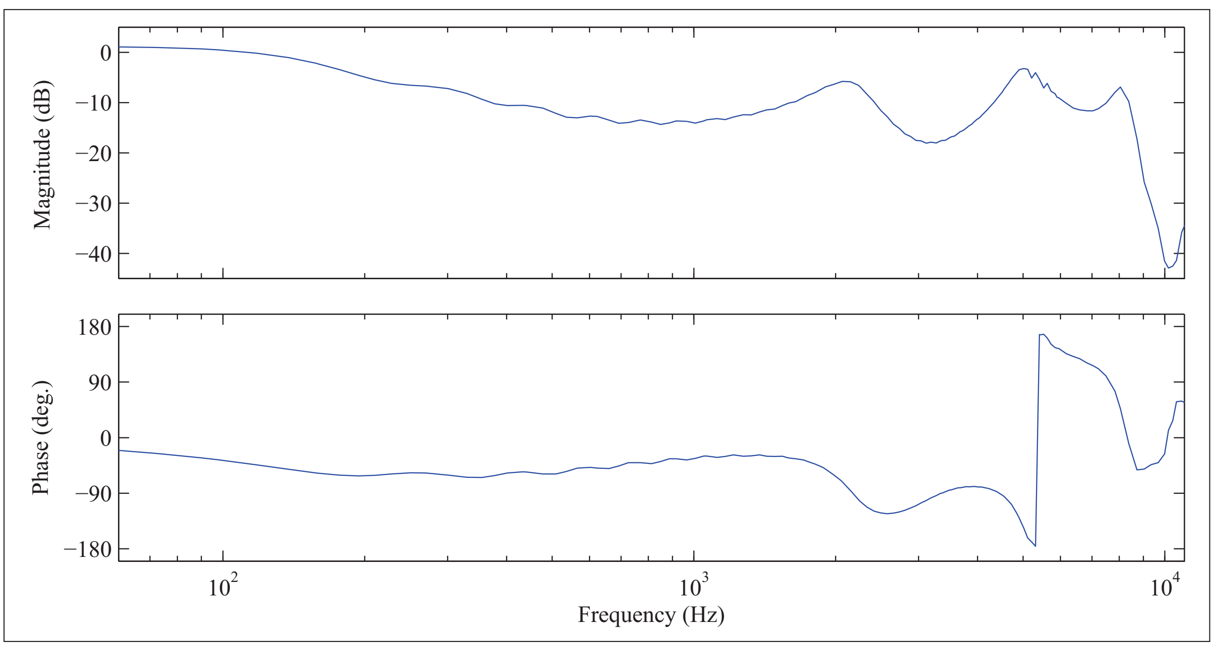

Figure 3.3 Frequency response of the moldable-fit plant in a human ear canal

3.1.2 Universal-fit earpiece

The universal-fit earpiece uses a modified R&D underbody from Sonomax that includes a sound channel allowing the use of universal-fit eartips. Figure 3.4 shows a cross-section of the universal-fit earpiece. A layer of porous material serves to smooth the frequency response of the loudspeaker, and another thin layer protects the microphone from earwax. Both layers dampen the resonance of the sound channel, analogous to a tube. A picture of the universal-fit earpiece is provided in appendix V.

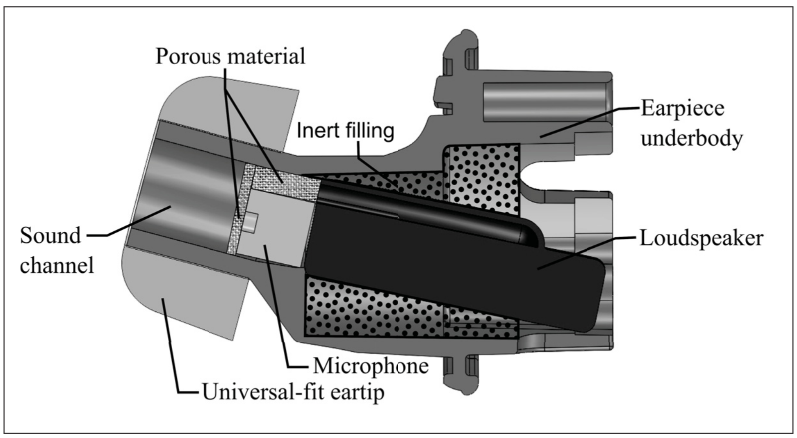

Figure 3.4 Cross-sectional view of the universal-fit earpiece

The universal-fit plant frequency responses, with and without porous material, were modeled using a system identification procedure, and are shown on figure 3.5. The porous material indeed allows damping of the loudspeaker's resonance and the unwanted resonance and anti-resonance caused by the tube. A close look at the plant frequency response reveals a relatively uniform phase up to approximately 3 kHz, although the phase is not minimum. Relatively high amplitude is still observed at high frequencies where the plant frequency response becomes out of phase. Figure 3.6 compares the frequency responses of the moldable-fit and universal-fit plants. Adaptation of the underbody for universal-fit was achieved at the cost of slight amplitude loss in the low frequencies of the universal-fit plant compared to the moldable-fit plant. However, its extended uniform phase in the bandwidth of interest and smoother frequency response make it seem better suited for ANC.

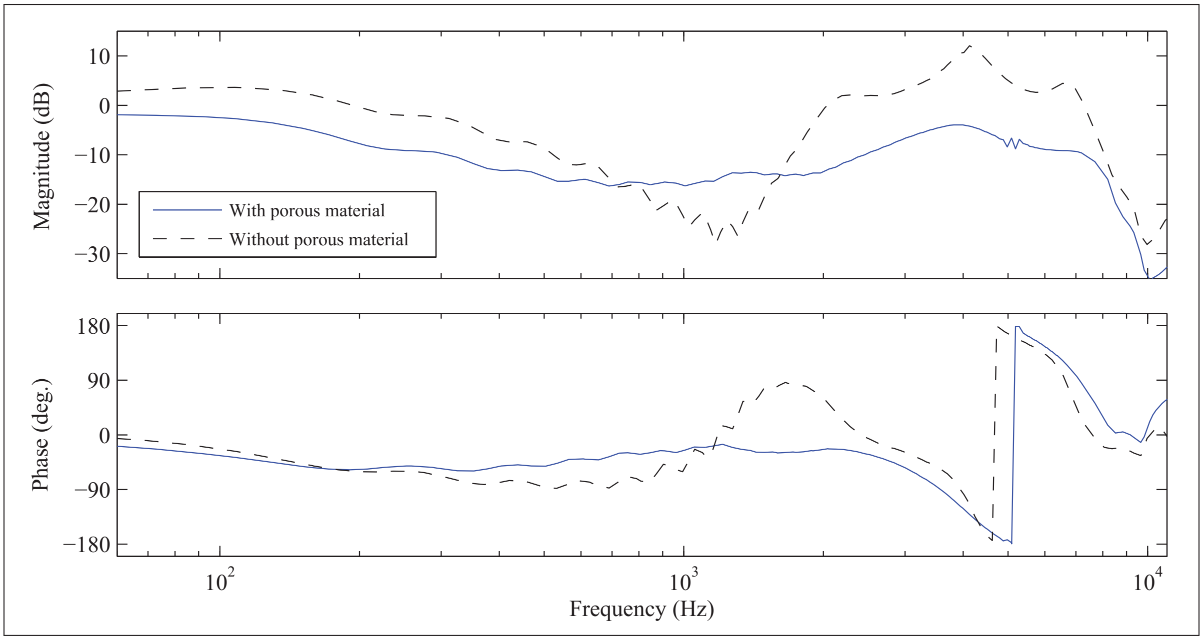

Figure 3.5 Frequency responses of the universal-fit plant in a human ear canal, with and without porous material

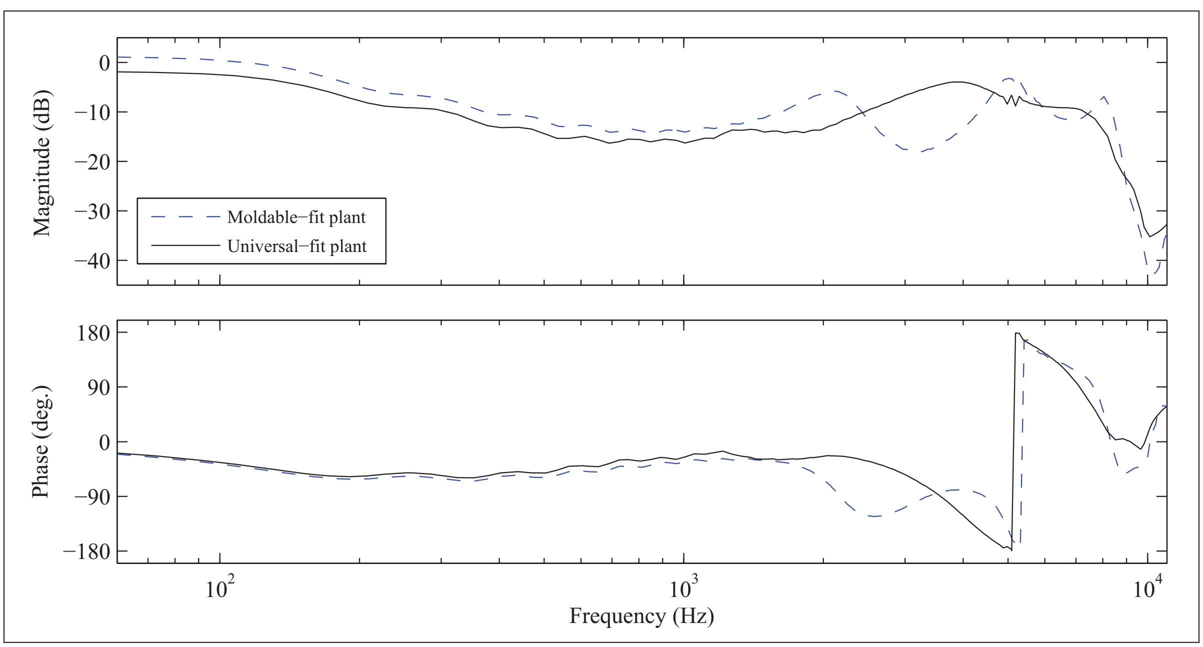

Figure 3.6 Comparison of the frequency responses of the two plants

3.1.3 Design of the controller

This section describes the design of a controller that was originally intended to be used with the moldable-fit plant only, but that proved suitable for the control of the universal-fit plant, with minor modifications. The FIR filters used to model the frequency responses of the plants were taken into MATLAB. Its environment allows one to simulate compensators, cascade them with the plants, and verify the theoretical resulting performance and gain and phase margins. Second order analog filters were simulated through the use of IIR filters.

The controller designed in this work uses two second order filters: one lead compensator and one lag compensator. Figure 3.7 illustrates the frequency response of both lead and lag second order compensators, forming the complete lead-lag fourth order compensator when cascaded together, and the frequency response of the model of the moldable-fit plant. The lag compensator is used to increase the magnitude response in the band of interest, where occlusion effect cancellation should occur, at the cost of phase lag. The lead compensator is used to raise the overall phase while causing decreased magnitude in the very low frequencies. Figure 3.8 shows the frequency response of the uncompensated moldable-fit plant, the lead-lag compensator, and the compensated plant.

The Nyquist plot of the compensated plant of figure 3.9 allows the inspection of gain and phase margins, which are 6.1 dB and 44.9°. These margins are sufficient to meet the minimum requirements of 6 dB and 30°. Figure 3.10 shows the expected occlusion effect reduction from design, predicting a reduction of around 14 dB at 250 Hz, 11 dB at 500 Hz, and an amplification of 0.6 dB at 1000 Hz and 3.3 dB around 1300 Hz, also falling within the boundaries of the requirements. The implementation of the compensator and the resulting performance using the moldable-fit plant and the universal-fit plant are discussed in section 3.2.

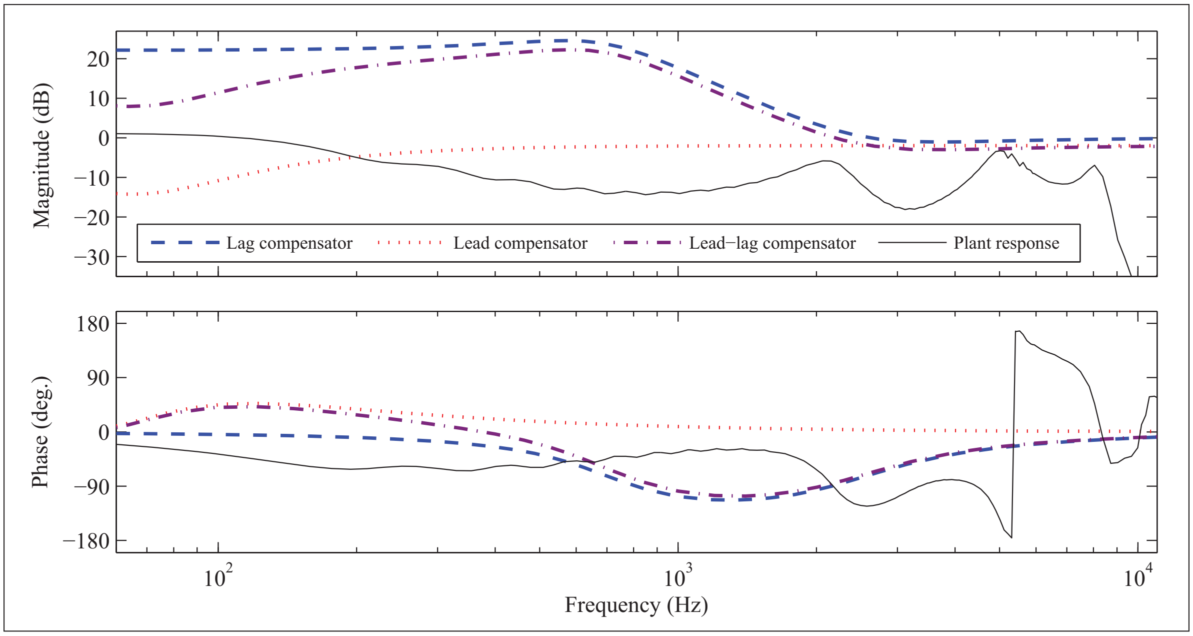

Figure 3.7 Frequency response of the two lead-lag compensators, the complete compensator, and the moldable-fit plant

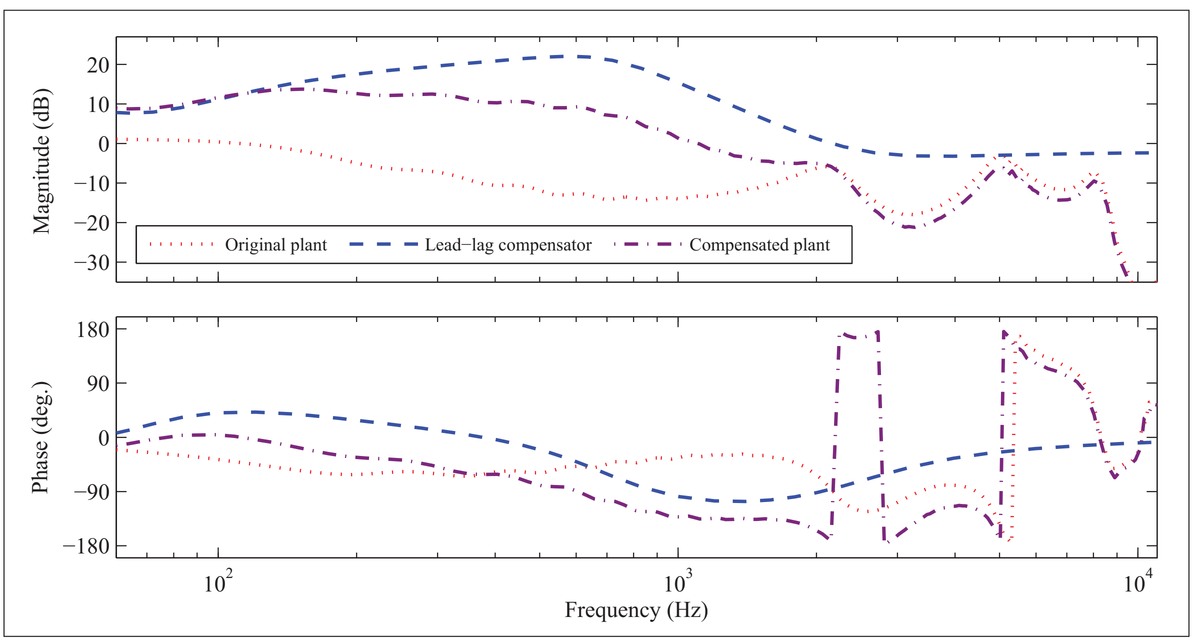

Figure 3.8 Frequency response of the uncompensated moldable-fit plant, the compensator, and the compensated moldable-fit plant

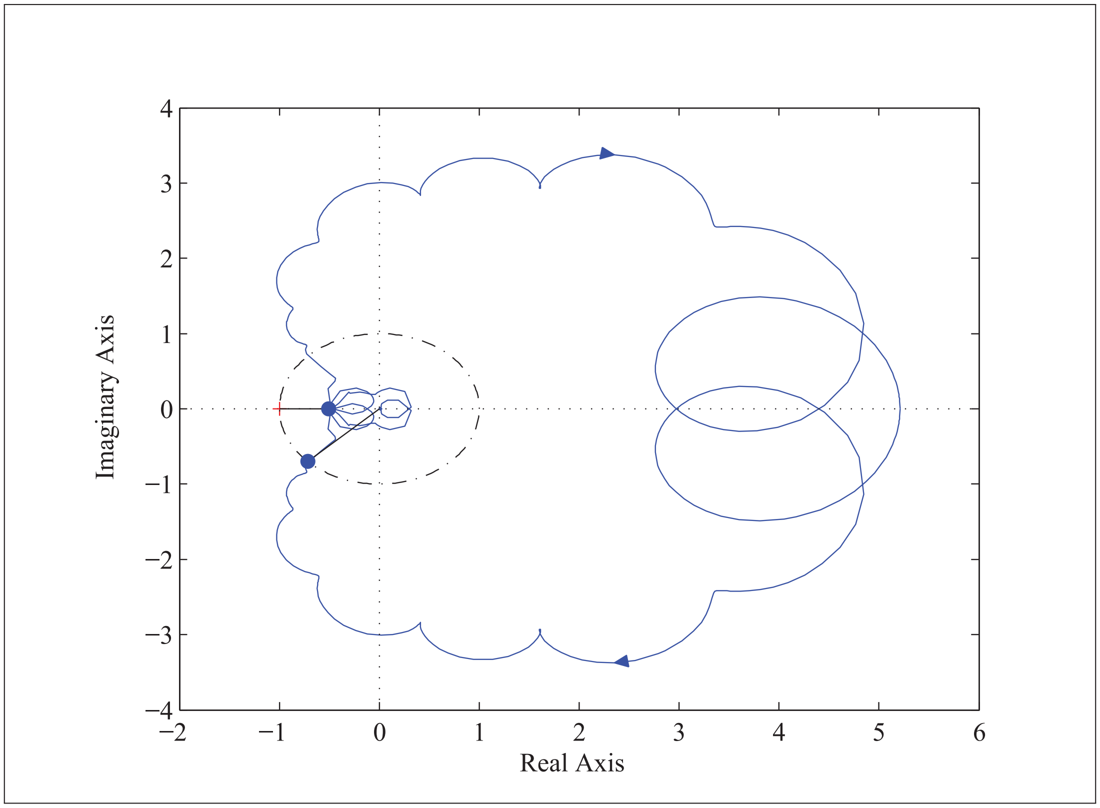

Figure 3.9 Nyquist diagram of the compensated moldable-fit plant by design

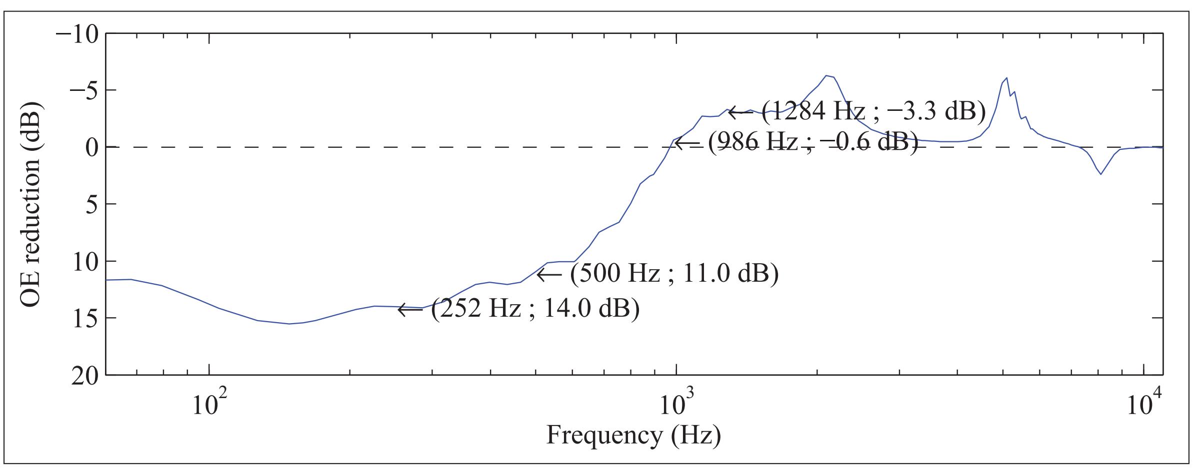

Figure 3.10 Expected active occlusion effect reduction of the moldable-fit prototype by design

3.2 Implementation and experimental validation

This section discusses the implementation of the compensator and the resulting characteristics of the implemented system (3.2.1), the experimental performances of the systems using the moldable-fit earpiece (3.2.2) and the universal-fit earpiece (3.2.3).

3.2.1 Implementation

The compensator was implemented by cascading two second order lead and lag compensators using the tow-thomas circuit topology with 5% tolerance resistors and 10% tolerance capacitors. The pre-amplifier was implemented with a low pass to filter out the DC component, and an inverting amplifier topology that would provide both the feedback loop and the overall gain of the feedback. The complete circuit schematic is available in appendix III. A comparison between the compensated transfer function of the moldable-fit plant from design and implementation is shown in figure 3.11. Small differences can be observed due to the tolerances of the components and the variability of the fit to the ear canal.

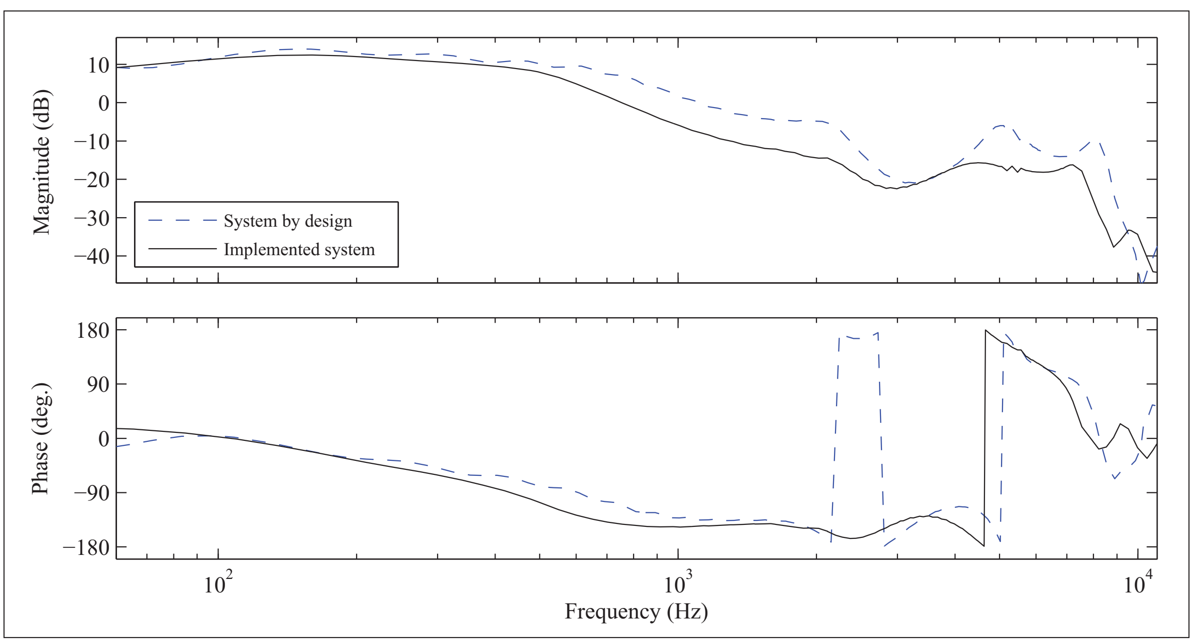

Figure 3.11 Frequency response of the compensated moldable-fit plant from design and implementation

The Nyquist plot of figure 3.12 reveals the gain and phase margins, which are 15.8 dB and 44.3° for the implemented system as opposed to 6.1 dB and 44.9° as predicted by theory. The increase of the gain margin compared to the design can be explained by the reduced magnitude of the implemented system at 2 kHz, previously limiting the gain margin. The decrease in phase margin can be explained by the corresponding decrease in the phase of the system around 700 Hz. The compared theoretical performance of the designed and implemented systems are shown in figure 3.13. Reduced performance of the implemented system in the 800-1000 Hz region can be explained by the reduced amplitude and phase of the implemented system between about 400 Hz and 1000 Hz. The implemented system would theoretically provide occlusion effect reduction that satisfies the requirements.

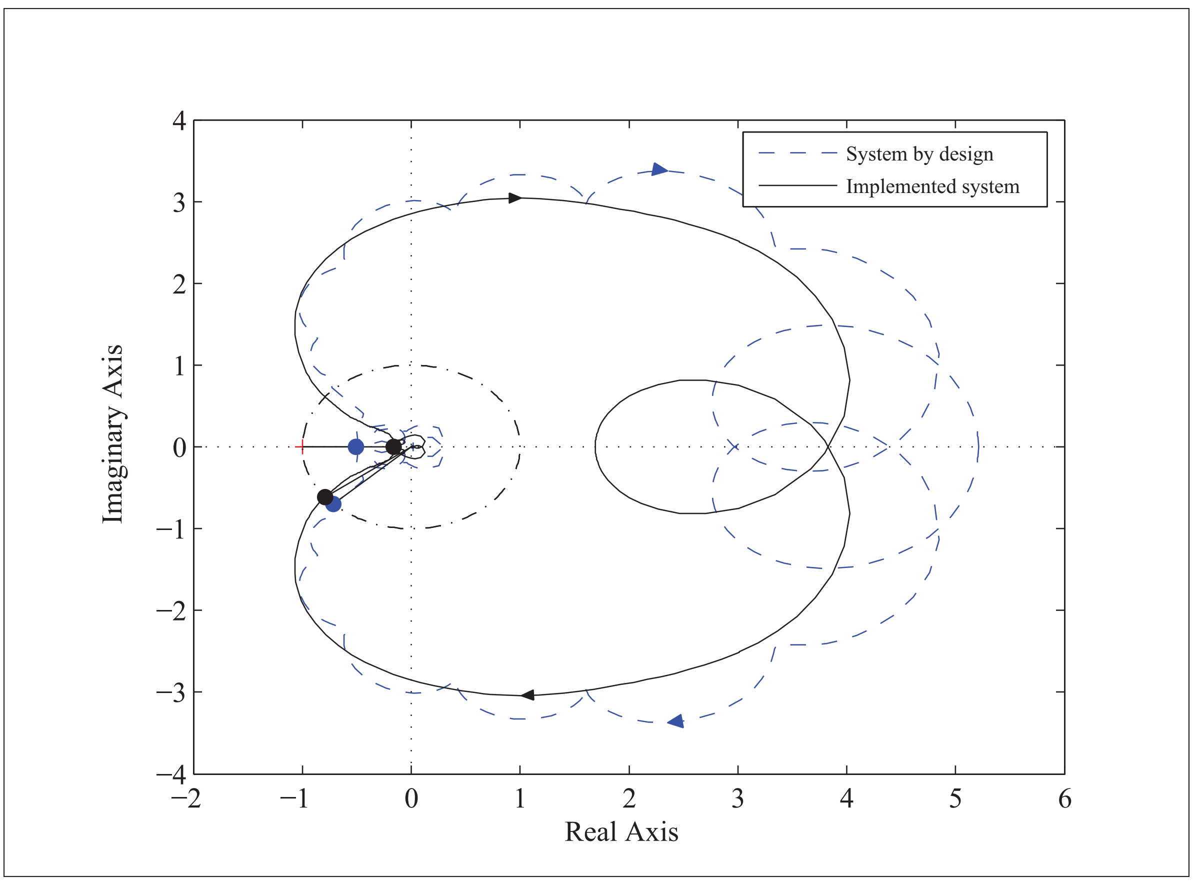

Figure 3.12 Nyquist diagram of the compensated moldable-fit plant from design and implementation

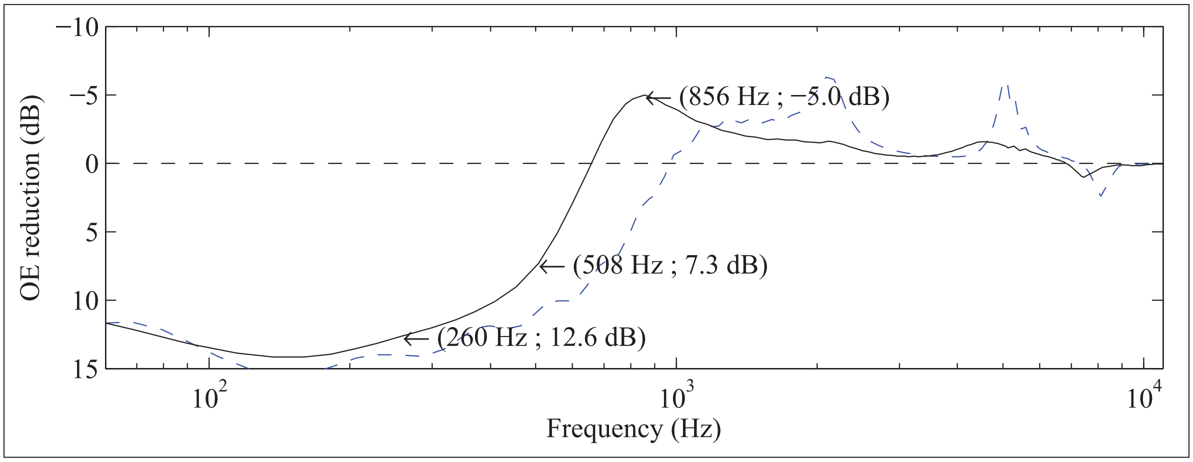

Figure 3.13 Expected active occlusion effect reduction from design and implementation

3.2.2 Experimental validation using the moldable-fit earpiece

To measure the actual SPL reduction in a real ear canal, two moldable-fit earpieces were worn by the author. At first, they were both inactive, meaning that the internal microphones were powered on, but the feedback control was turned off. The sound in both ear canal was then recorded using the internal microphones as the author was humming in a swept-sine-like fashion, to measure any difference in SPL between the two occluded ear canals. Figure 3.14 shows the transfer function between the left and right earpieces. Differences of ± 3 dB are observed in the low frequencies up until 700 Hz, and -5 dB at 1000 Hz. Beyond 1000 Hz, the coherence drops, indicating that the voice of the author did not produce significant energy beyond that frequency. The difference in the acoustic seal and residual occluded volume resulting from variability of the fit between the left and right earpiece might explain these differences. Then, without any modification of the earpieces' fit to the ear canal, the right earpiece feedback was engaged, while the left one remained inactive as the author hummed in the same swept-sine-like fashion. The right earpiece was therefore actively reducing SPL in the ear canal while the left earpiece served as a reference. The transfer function between the left and right internal microphones, corrected for the differences in SPL between the inactive left and right earpiece previously observed, yields the active occlusion effect reduction achieved by the implemented system. A picture of the experimental setup is included in appendix V. Figure 3.15 shows the measured reduction in SPL in the right ear canal with and without the correction, as well as the theoretical occlusion effect reduction expected by design and after implementation.

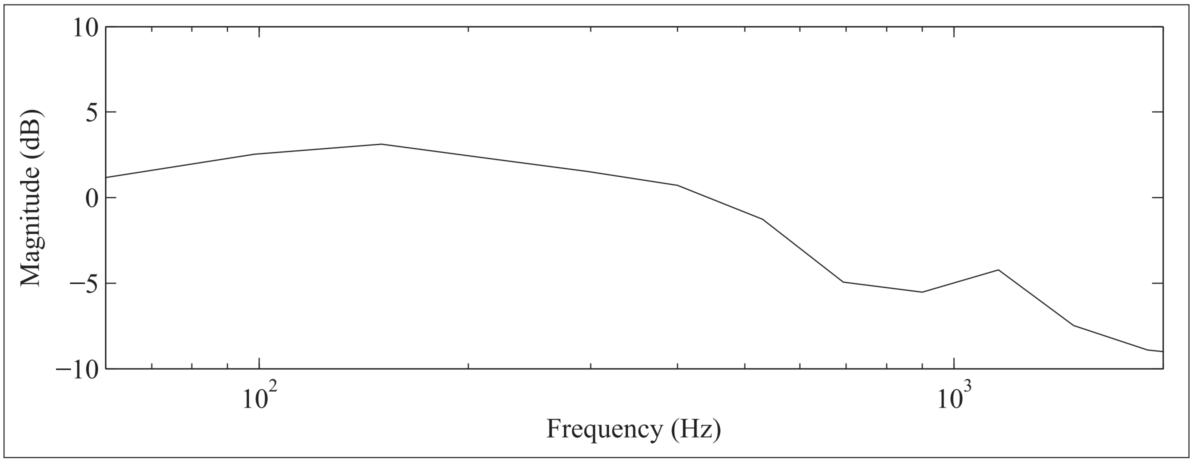

Figure 3.14 Transfer function between the sounds resulting from occlusion effect in the left and right ear canals for inactive earpieces

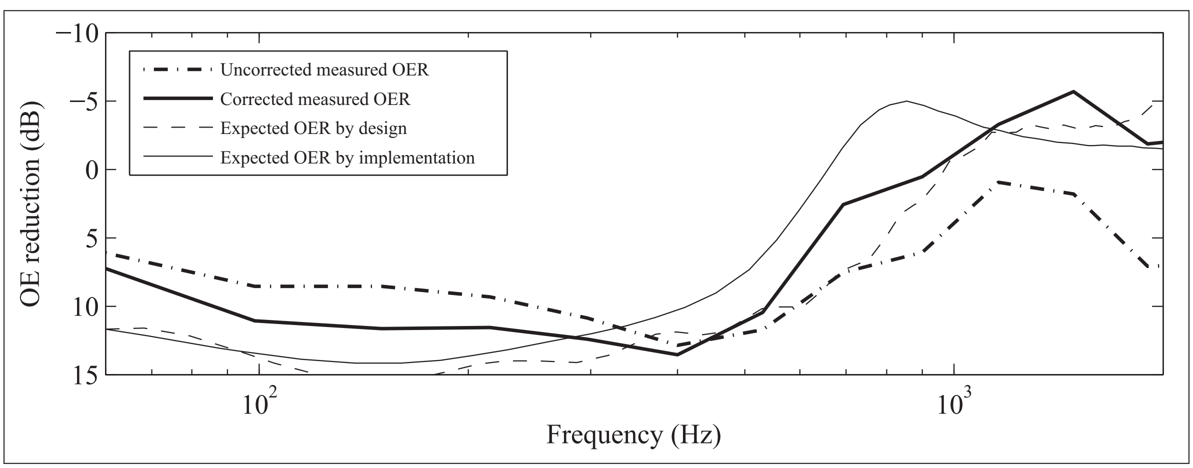

Figure 3.15 Measured occlusion effect reduction; transfer function between the left and right ear canals, with and without correction, compared with the theoretical occlusion effect reduction by design and after implementation

The construction of the moldable-fit plant, although well suited to fit a large number of ear canals while providing a good acoustic seal, encountered a major practical problem. The internal microphone was unprotected and too close to the opening of the earplug, and was eventually damaged.

3.2.3 Experimental validation using the universal-fit earpiece

Since the moldable-fit earpiece and universal-fit earpiece were built with the same components and offer similar frequency responses, the implemented controller intended for the moldable-fit plant was adapted for the universal-fit plant by reducing its gain as much as possible while still achieving the required target AOER. The theoretical performance in occlusion effect reduction are predicted to be 6 and 8.6 dB at 250 Hz and 500 Hz, with minor regeneration of 1.6 dB at 1 kHz, while offering a gain and phase margin of 12.8 dB and 73.6° respectively.

Its performance in reducing SPL in the ear canal was validated with a different method: since the occlusion effect reduction system reduces sound in the ear canal, it also provides active attenuation of external disturbances. Therefore, the performance of the occlusion effect reduction system can be characterized by using a differential audiogram in open and occluded conditions, with the feedback loop disabled and enabled as a variable for the occluded conditions. Two subjects were tested in an audiometric booth with the real ear attenuation at threshold (REAT) method, according to standard ANSI S12.6-1997 and using the REATMaster software, as part of a preliminary test in view of testing a greater number of human subjects for which the author has had the authorization from Comité d'éthique de la recherche (2013). Using octave band noises centered at 63, 125, 250, 500, 1000, 2000, 3150, 4000, 6000 and 8000 Hz, the subjects were tested for their auditory threshold when unoccluded, when occluded with the prototype HPD with the AOER feedback control inactive, and when occluded with the prototype HPD with the AOER feedback control active. The difference in auditory thresholds between the open and the two occluded conditions yields the REAT for the HPD in passive and active mode. The difference in the REAT of the same HPD with inactive then active feedback loop yields the performance of the HPD in active attenuation of external disturbances. Figure 3.16 quantifies the active attenuation provided by the HPD in active and passive mode for two subjects. Figure 3.17 shows the mean active attenuation of external disturbances, and compares it with the expected performance by design.

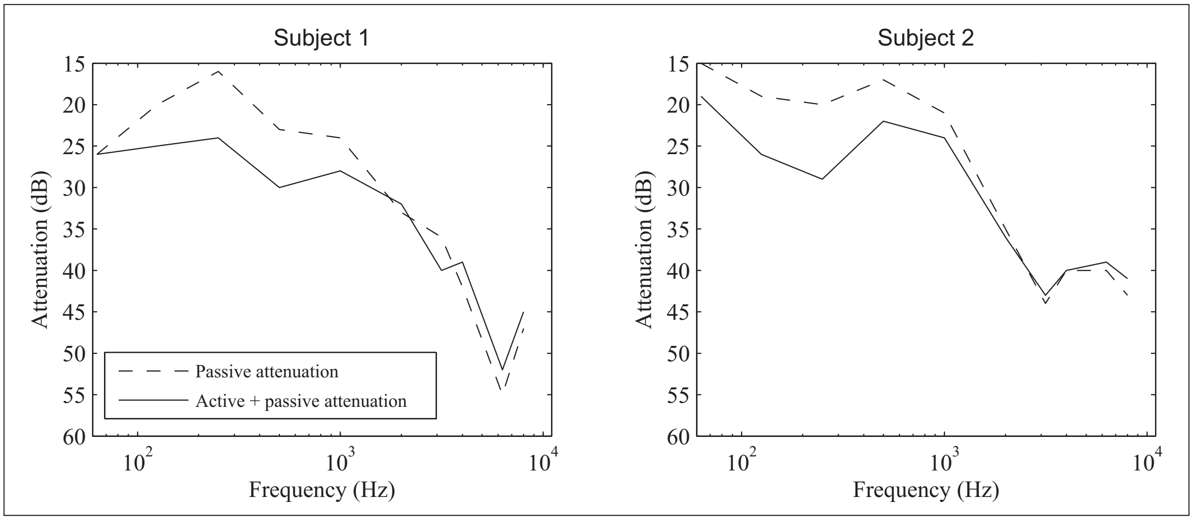

Figure 3.16 Passive attenuation and attenuation in active mode, obtained using narrow band audiometry on two human subjects

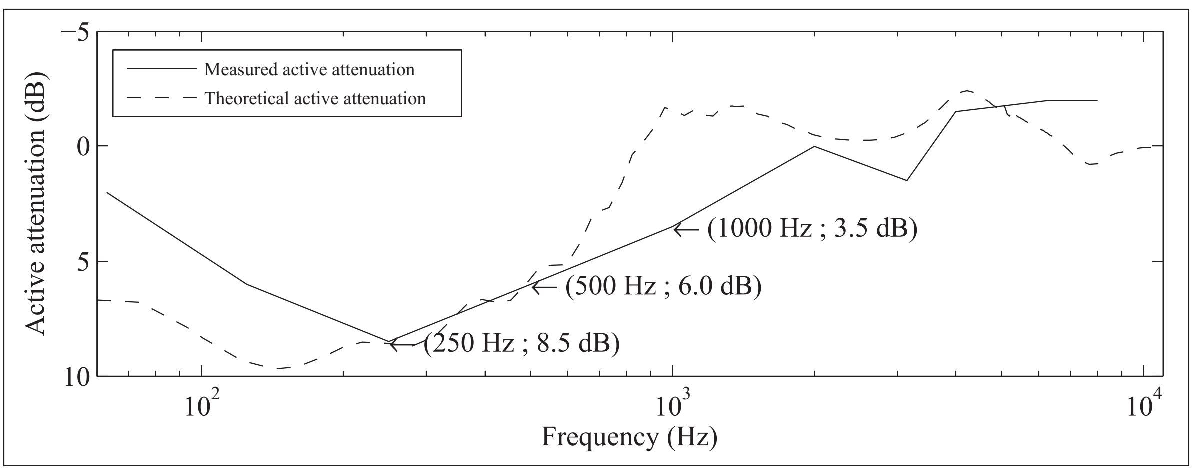

Figure 3.17 Measured active attenuation compared to theoretical performance by design

3.3 Discussion regarding the solution to occlusion effect

In this chapter, a solution to the occlusion effect was presented in the form of an active control of the sound in the ear canal resulting from occlusion effect, or active occlusion effect reduction. Two AOER systems have been implemented in this master's project using a moldable-fit earpiece and an universal-fit earpiece, and experimental performances of both systems meet the requirements defined in section 2.2. The system using moldable-fit earpieces was designed to offer best possible performance while retaining gain and phase margins by design of 6.1 dB and 44.9°, close to the defined minimum requirements. The system based on universal-fit was tuned to offer greater gain and phase margins of 12.8 dB and 73.6° while retaining the required minimum target AOER performance. Table 3.1 summarizes the minimum and maximum target SPL AOER and the performance of the two prototypes.

Table 3.1 Summary of the performances of the two prototypes of active occlusion effect reduction

Frequency (Hz)	Target AOER (dB)		Measured AOER (dB)	
	Min.	Max.	Universal-fit	Moldable-fit
250	8	17	8.5	12
500	6	14	6	10
1000	-9	6	3.5	-3

In general, the experimental performances are close to the theoretical performances for both prototypes. However, the gain margin of the moldable-fit compensated plant was found to vary greatly between design and implementation. This finding is attributed to many factors: the tolerances of the electrical components involved in the implementation and the variability of the fit to the ear canal; the acoustic seal and the residual occluded volume. The moldable-fit physical interface with the ear canal was found to be hard to manage: the soft silicon did not permit to assess that the earplug was centered in the ear canal, and its opening was sometimes partially blocked by the walls of the ear canal, which can be covered with earwax. This led to high variability of the fit between insertions and eventually damaged the internal microphone. Therefore, the moldable-fit option is rejected for further work with a greater number of subjects.

The universal-fit earpiece offered less variability in the ear canal fit between insertions than the moldable-fit earpiece, and its construction ensures that it is more centered in the ear canal and more resistant to earwax. The large gain and phase margins of the prototype using the universal-fit earpiece make it seem well suited for testing on a greater number of human subjects. If better performance is required at the cost of gain and phase margins, a compensator could easily be designed for that purpose using the same technique outlined in section 3.2.1. Table 3.2 shows the estimated residual occlusion effect caused by wearing the universal-fit prototype, estimated by adding the experimental active occlusion effect reduction to the OE values for a shallow insertion device. The ROE is compared with the occlusion effect caused by a shallowly inserted device and a deeply inserted device, according the the measurements of Dean and Martin (2000). The universal-fit prototype causes an occlusion effect similar to that of a deeply inserted device, without the need for deep insertion, likely increasing physical comfort for the same perceived occlusion effect.

Table 3.2 Estimated residual occlusion effect with the universal-fit prototype compared with shallow and deep insertion devices

Frequency (Hz)	**Occlusion effect (dB)**		
	Shallow Insertion	**Deep Insertion**	**Universal-fit**
250	17	9	8.5
500	14	8	8
1000	6	-1	2.5

Anecdotal testing and demonstrating of the universal-fit prototype seems to indicate that moderate performance could make a big difference on the perceived OE, although no definitive conclusions can be made at this stage. This would be consistent with the statement from Henry and Letowski (2007) that occlusion effect *"is considered negligible if it is less than 10 dB"*. The residual occlusion effect caused by the moderate performance of the universal-fit prototype is only about 1 to 1.5 dB away from the 5-7 dB occlusion effect that most participants would find natural, for speech, according to Kuk *et al.* (2005). This indicates that active occlusion effect reduction could indeed be an efficient solution to the occlusion effect experienced by musicians when wearing HPDs.

CHAPTER 4

A SOLUTION TO THE ISOLATION EFFECT

This section presents the design and implementation of a HPD featuring a prototype of an isolation effect compensation system. It is divided as follows: proposed architecture and modelling of the components (4.1), development of isolation effect compensation algorithms and preliminary experimental validation (4.2), the use of the isolation effect compensation system in conjunction with the occlusion effect reduction system (4.3), and discussion regarding the solution to isolation effect proposed in this work (4.4). The requirements for the solution to isolation effect, discussed in section 2.3 are summarized below:

1. Compatibility with the solution to occlusion effect;
2. The use of the Auditory Research Platform to implement the algorithms;
3. A transfer function of the HPD that mimics the response of an open ear, at a lower level, which is equivalent to a uniform attenuation;
4. A user-selectable, variable attenuation between 0 and 30 dB;
5. A loudness correction that shapes the attenuation of the HPD to consider the non-linearity of loudness perception, based on ISO226:2003.

4.1 Proposed architecture and modelling of components

Given the requirements of maintaining compatibility with the active occlusion effect reduction system previously designed, the proposed system architecture is presented in figure 4.1. An external microphone captures the sound in a given environment and is connected to the input of a DSP, provided by the ARP. The output of the DSP is connected to the command signal input of the active occlusion effect reduction system. The internal microphone that is used for occlusion effect control is also connected to the input of the DSP. The external and internal microphones can be calibrated to give an absolute sound pressure level measurement outside and inside the ear canal, while assessing the attenuation of the HPD. The DSP is used to implement the algorithms for isolation effect compensation. An earpiece provides passive attenuation and contains the external microphone and the internal microphone and loudspeaker assembly. The proposed earpiece is a modified version of the universal-fit earpiece, discussed in section 3.1.2, so as to include an external microphone, as shown in figure 4.2.

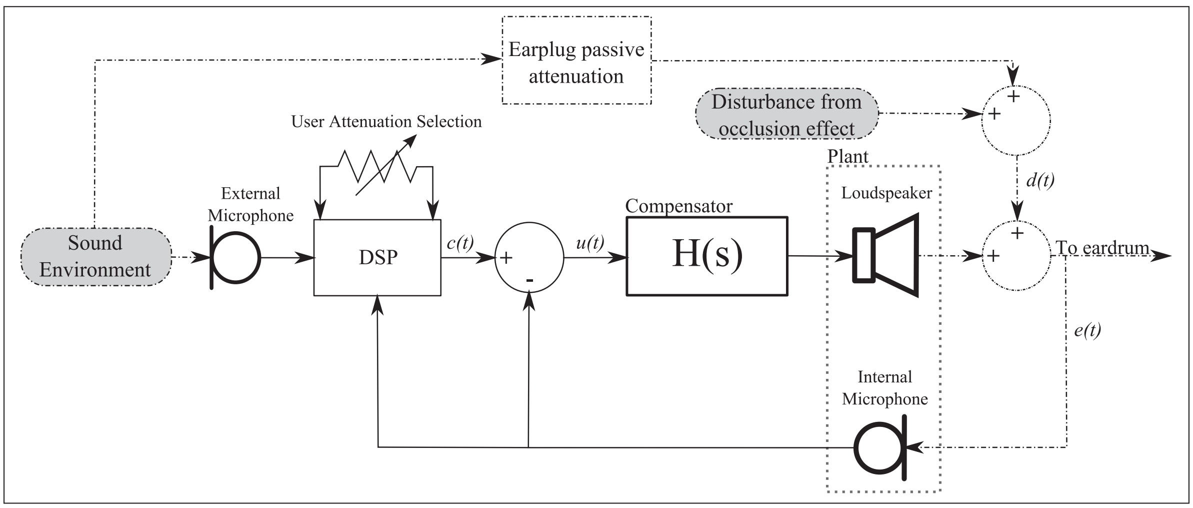

Figure 4.1 Architecture of the proposed system, incorporating both active control of the occlusion effect and the hardware for the isolation effect compensation

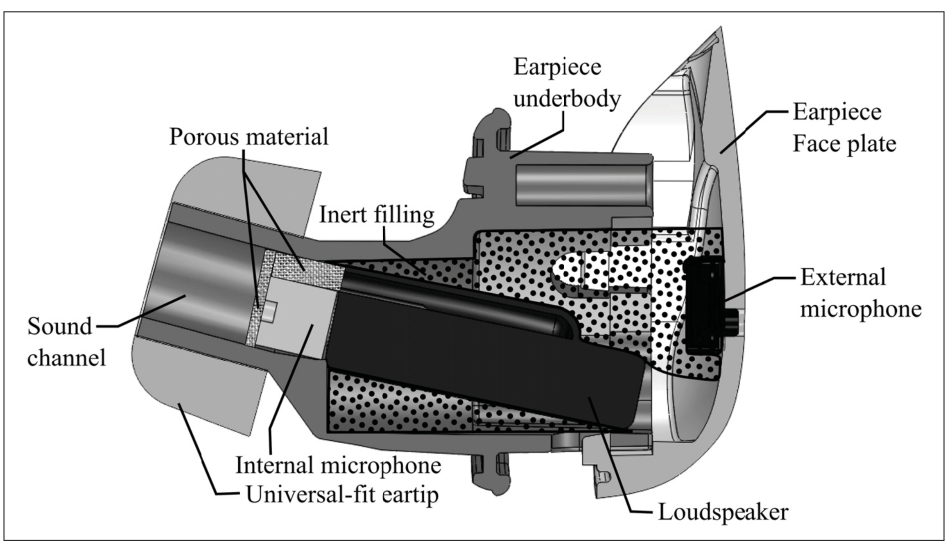

Figure 4.2 Modified universal-fit earpiece including an external microphone

Figure 4.3 illustrates a conceptual view of the system with emphasis on the isolation effect solution from a design perspective. In this conceptual view, there are two paths from the sound environment to the eardrum. The paths and their elements are as follows:

1. The attenuation path, comprised of:
   a) the passive attenuation, provided by the earpiece;
   b) the active attenuation, only if the isolation effect compensation system is used in conjunction with the active occlusion effect reduction system;
2. The electro-acoustic path, comprised of:
   a) the external microphone;
   b) the DSP of the ARP;
   c) the playback mean, exhibiting a different frequency response depending on whether the isolation effect compensation system is used in conjunction with the active occlusion effect reduction system;
   d) the occluded ear canal.

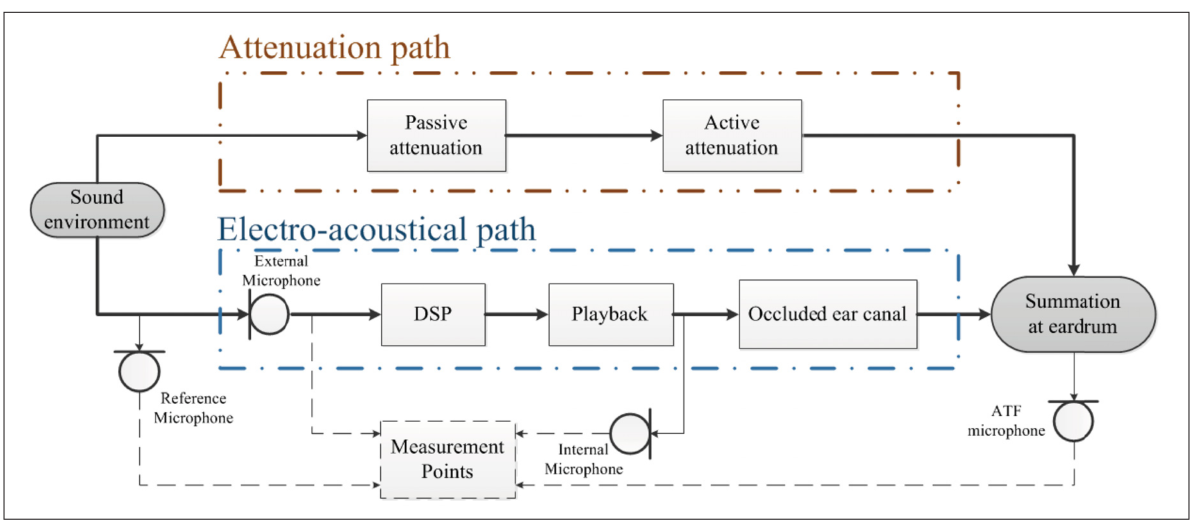

Figure 4.3 Conceptual view of the isolation effect solution from a design perspective

When the two paths meet at the eardrum, destructive and constructive interference is expected to occur because of the phase differences of the signals. This poses a problem if both signals are of the same order of magnitude, but would not have significant impact if the electro-acoustic path prevails.

The following subsections aim at characterizing and modeling each of the elements of the two paths to obtain a model of the whole situation and derive isolation effect algorithms for a realistic scenario. An acoustic test fixture (ATF), in this case a Bruel & Kjær head and torso simulator model 4157 is used to obtain these models. Four important measurement points, that are referred to throughout this chapter, are shown on the conceptual view of figure 4.3. The reference microphone and the ATF microphone are available in a laboratory setting, when the device is worn by an ATF. The external and internal microphones are part of the HPD and are available when the device is worn by a user. A picture of the experimental setup used throughout this chapter is available in appendix V.

4.1.1 The electro-acoustic path

In this section, the electro-acoustic path is modeled with IIR and FIR filters, element by element, either through design of filters or by using the same system identification procedure used in chapter 3 and described in appendix II. The motivation for modeling each element of the paths rather than the just the entire paths or the complete system is to be able to predict how changing a single or many elements will affect the overall performance of the isolation effect compensation system. This is useful to predict how the isolation effect compensation system could behave when used by itself, or in conjunction with the active occlusion effect reduction system, and to identify the elements that could be changed to improve the performance of the isolation effect compensation system.

4.1.1.1 External microphone

The external microphone is the first element of the electro-acoustic path. A miniature electret hearing aid microphone was selected because of its low profile. From the specifications, it has a high pass type second order roll-off at 80 Hz, and a low pass second order roll-off at about

10 kHz with an overshoot of about 3 dB at 7 kHz, so it can be accurately modeled with an IIR filter. Figure 4.4 shows its modeled transfer function obtained by matching the response of IIR filters to the frequency response given in the specifications.

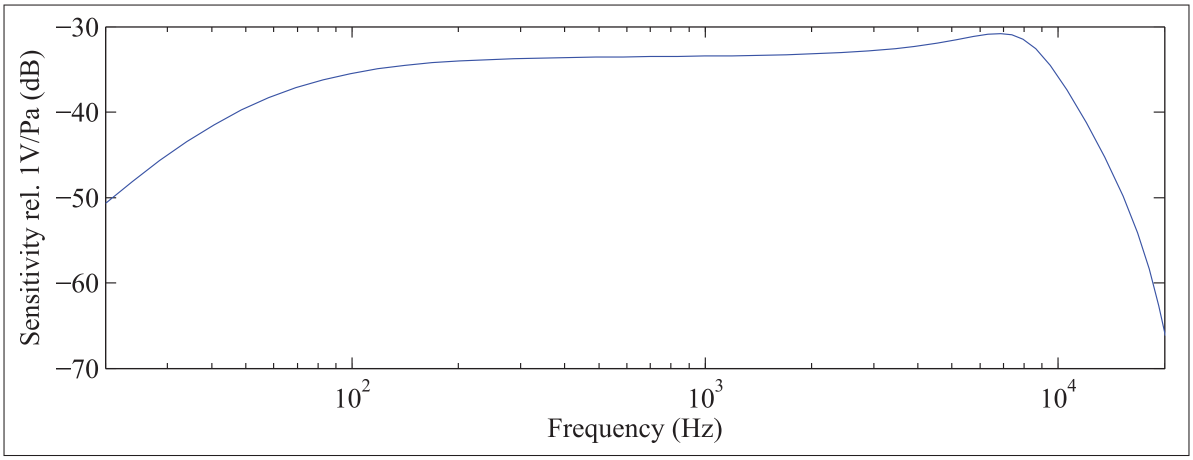

Figure 4.4 External microphone model frequency response

4.1.1.2 Digital signal processor

The DSP provided by the ARP has an inherent input to output delay of 1 ms. Its effect on the electro-acoustic path can be modeled with an all pass FIR filter with a constant group delay of 1 ms. At a sample rate of 48 kHz, 1 ms corresponds to 48 samples. The frequency response of the model of the input to output delay of the DSP is shown in figure 4.5.

4.1.1.3 Playback

Two situations are considered for playback. The first playback situation is when the isolation effect compensation system is used by itself and the output of the DSP is sent to the loudspeaker. In this case, it can be monitored by the internal microphone, at the entrance of the occluded ear canal. This is equivalent to the transfer function of the uncompensated plant. The second playback situation is when the isolation effect is used in conjunction with the AOER system and the output of the DSP is sent to the command signal input of the AOER system.

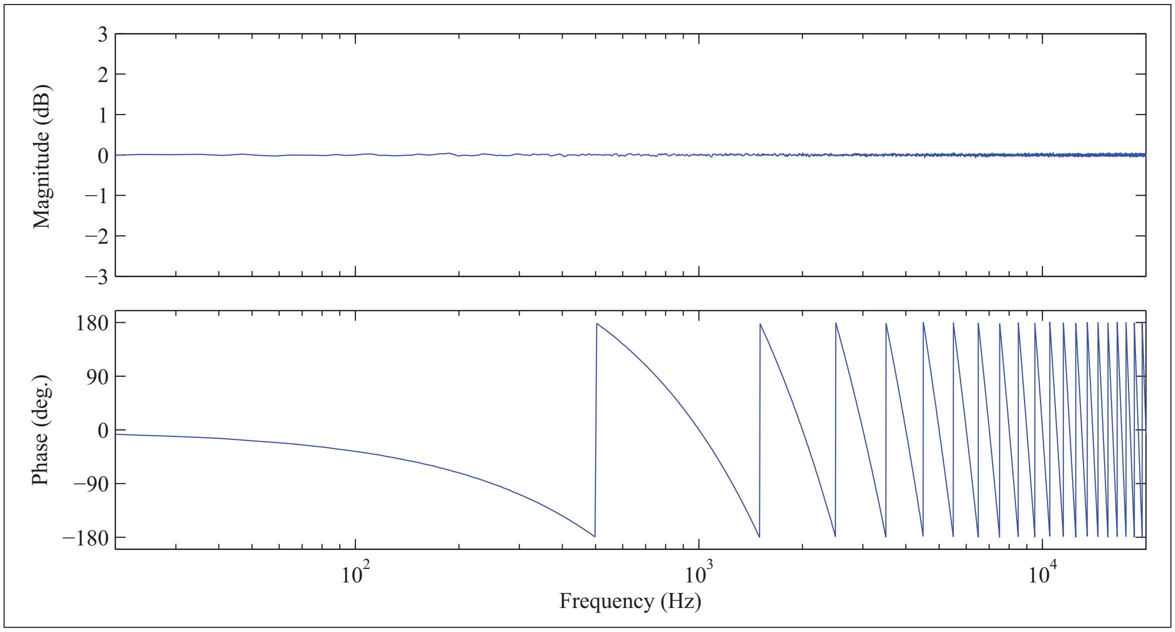

Figure 4.5 Frequency response of the modeled input to output group delay caused by the DSP

This is the transfer function between the internal microphone and the command signal input, the complementary sensitivity function, previously expressed in equation 1.4.

The first playback model, the transfer function of the uncompensated universal-fit plant in the ear canal of the ATF, was modeled with a FIR filter. Figure 4.6 shows a comparison between the model and the measured frequency response of the universal-fit plant on the ATF, as well as the frequency response of same plant in the author's ear canal. The overall rise in magnitude of the frequency response is due to the smaller equivalent volume of the occluded ear canal of the ATF. If the AOER system is used, we can expect the gain margin to be reduced and the relative performance to increase on the ATF, compared to the reported performance of section 3.3.

The second playback situation occurs if the isolation effect compensation system is used in conjunction with the active occlusion effect reduction system. In this case, the playback can be modeled by the sensitivity function of the AOER system, derived from the uncompensated plant frequency response and the compensator frequency response using equation 1.4. The frequency response of the model of the playback provided by the active occlusion effect reduction system is shown in figure 4.7.

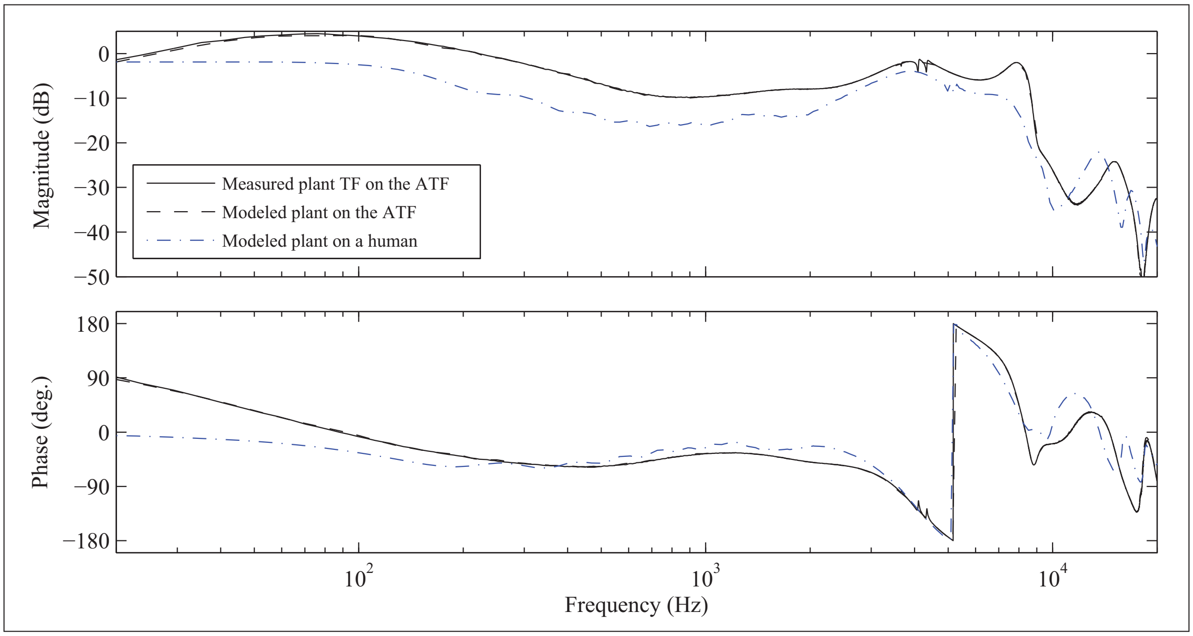

Figure 4.6 Frequency response of the first playback model, the uncompensated plant

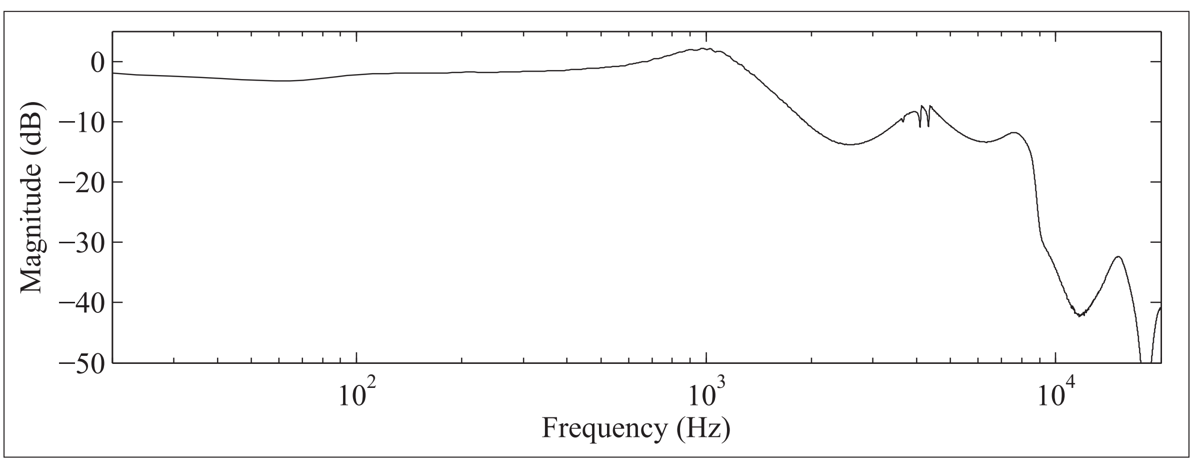

Figure 4.7 Frequency response of the second playback model, the sensitivity function of the occlusion effect reduction system

4.1.1.4 Occluded ear canal

The occluded ear canal of the ATF was modeled by a FIR filter approximating the transfer function from the internal microphone to the "eardrum" microphone of the ATF. Figure 4.8 shows the occluded ear canal measured transfer function from its blocked entrance to the eardrum and the corresponding FIR model frequency response. The obtained FIR filter does not accurately model the occluded ear canal after 9 kHz, but it is sufficient for this analysis, as the loudspeaker does not produce significant SPL past 9 kHz.

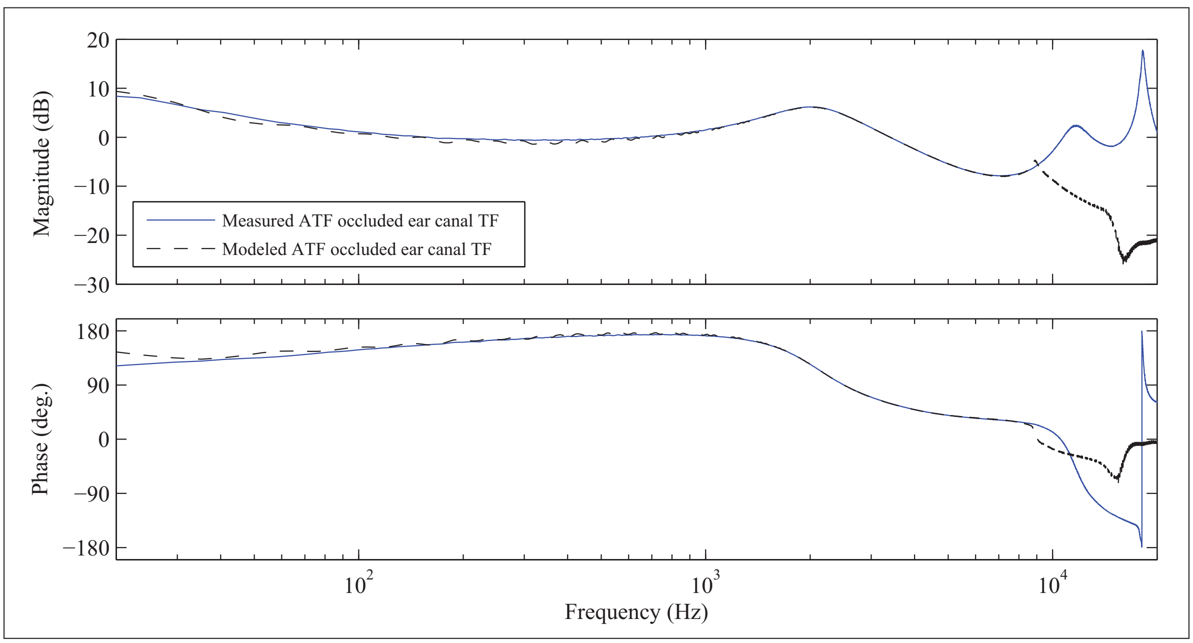

Figure 4.8 Measured transfer function and frequency response of the model of the occluded ear canal

4.1.1.5 Validation of the electro-acoustic path model

Cascading all the separate models of the elements and accounting for the transducers' sensitivity should result in the complete electro-acoustic path model. Measuring its complete transfer function from the sound right before it is picked up by the external microphone to the ATF "eardrum" microphone is difficult because it would require that a reference microphone be positioned exactly where the external microphone is. Since this is physically impossible, the reference microphone is at a distance from the external microphone and this difference in position causes a phase shift, resulting in a difference in phase between the model and the measured transfer function. Additionally, the physical placement of the reference microphone close to the HPD causes the reference microphone to pick up sound reflected by the HPD and the ATF, and therefore amplitude differences between the model and the measured transfer function are observed. Figure 4.9 compares the prediction of the model when a high gain of 20 dB is programmed in the DSP, so that the attenuation path is negligible, with the measured transfer function of the real system using a reference microphone that is placed as close as possible to the outside of the HPD. The case considered here uses the uncompensated plant playback option, without the effect of the active occlusion effect reduction system.

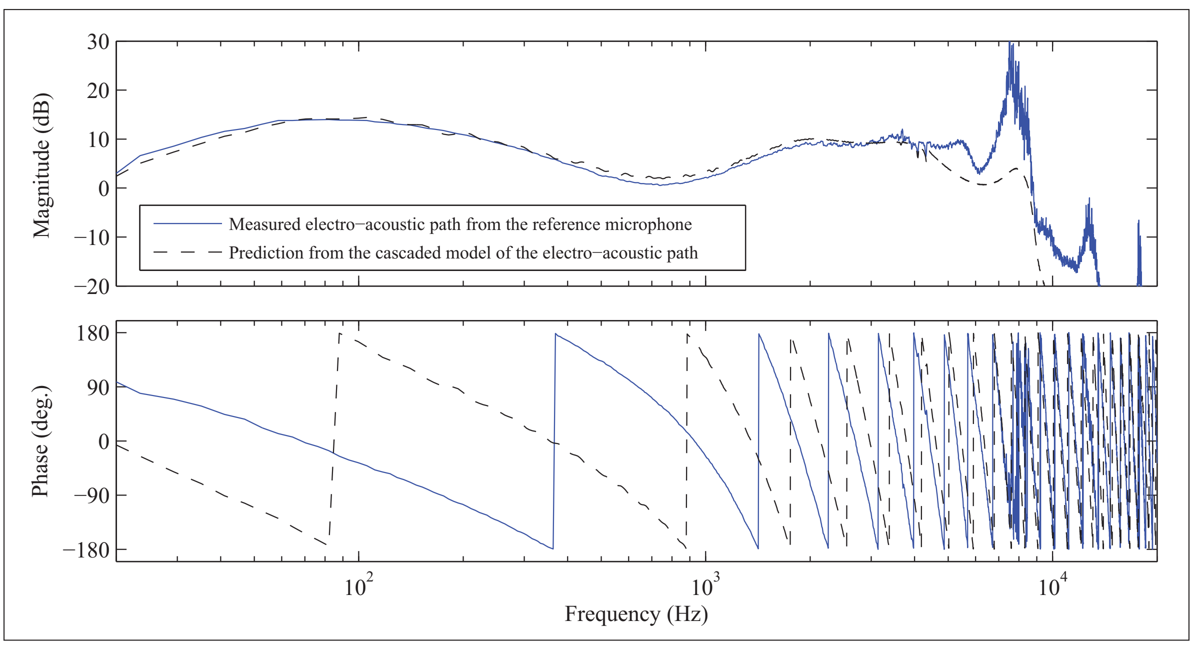

Figure 4.9 Differences between the model and the measure when using a reference microphone

To overcome this limitation, the external microphone is used instead of the reference microphone and corrected for its frequency response using a correction derived from its model. The transfer function from the external microphone output to the "eardrum" microphone output was measured and the effect of the external microphone was factored in, leading to a good agreement between the cascaded model of the electro-acoustic path and its measured transfer function, as shown on figure 4.10. The complete model of the electro-acoustic path can be considered to be a good representation of the real system from about 70 Hz to about 10 kHz, range over which the differences in magnitude are below 3 dB and the differences in phase are below 20 °.

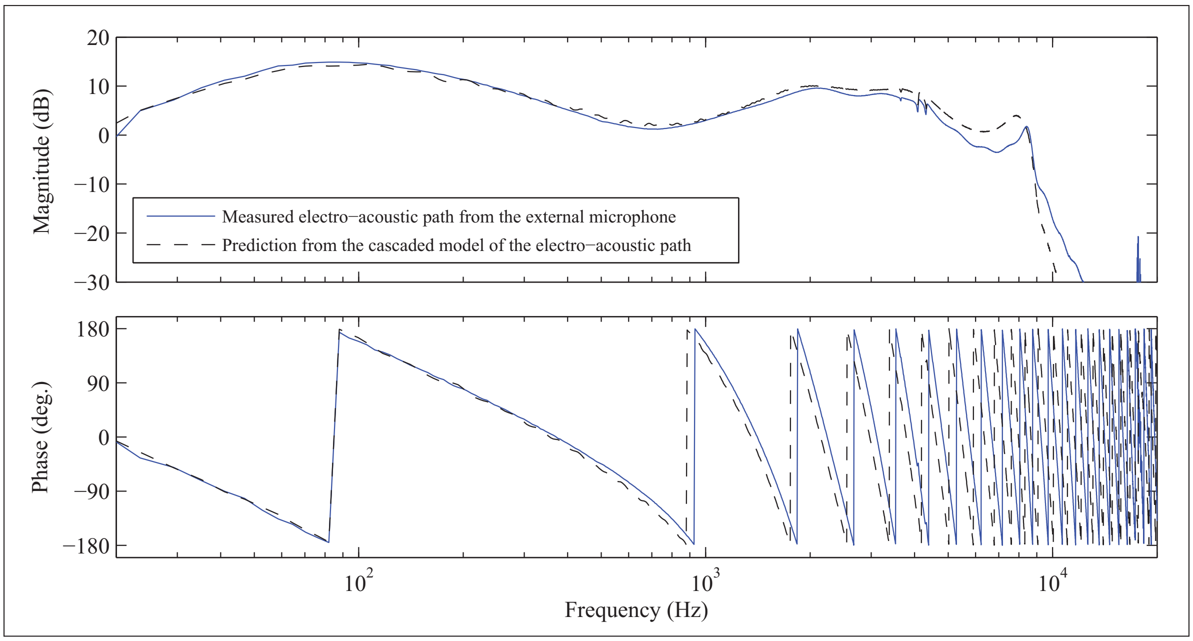

Figure 4.10 Accordance between the model and the measure when correcting for the external microphone frequency response

4.1.2 The attenuation path

The attenuation path is composed of two elements: the passive attenuation and the active attenuation. The active attenuation, provided by the AOER system, adds its effect to the passive attenuation if the isolation effect compensation system is used in conjunction with the active occlusion effect reduction system.

4.1.2.1 Passive attenuation

The passive attenuation refers to the transfer function of the HPD from just outside the HPD to inside the ear canal, with the electro-acoustic path disabled. It is measured with the external microphone and the "eardrum" microphone of the ATF, by applying the same correction for the external microphone as in section 4.1.1.5. The measured transfer function and the frequency response of the model are shown in figure 4.12. The model strays from the measured results in the very low frequencies, in terms of phase, at about 100 Hz.

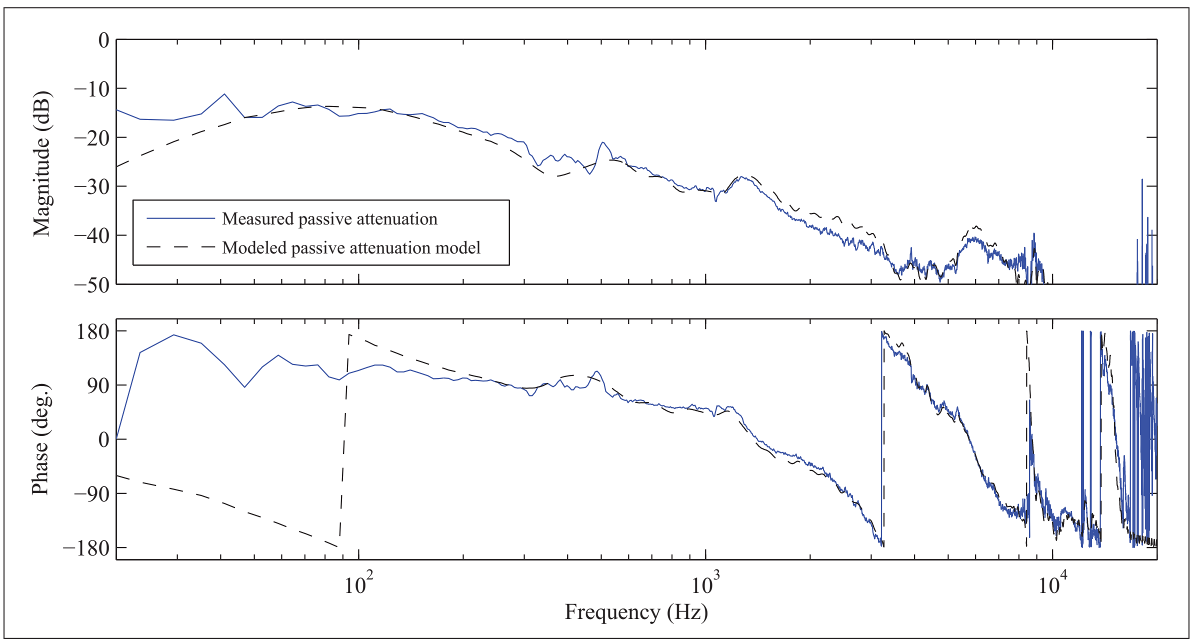

Figure 4.11 Measured and modeled frequency response of the passive attenuation

4.1.2.2 Active attenuation

The active attenuation adds its effect to the passive attenuation's effect if the AOER system is used with the isolation effect compensation system. It can be deduced by obtaining the difference in the measured transfer function of the HPD when the AOER system is inactive and active, as the electro-acoustic path is disabled. Figure 4.12 shows the measured and modeled transfer function of the active attenuation on the ATF. The active attenuation is, as expected, greater than what was measured on two human subjects in section 3.2.3, mainly because of the smaller equivalent occluded volume of the ATF ear canal that causes the plant response to increase in magnitude. However, regeneration of 5-7 dB occurs from 700 Hz to 3 kHz. Figure 4.13 compares the passive attenuation to the combined passive and active attenuation. The effect of the active attenuation benefits the total attenuation below about 700 Hz, the point at which regeneration starts to occur up until about 3 kHz. The models are valid from about 100 Hz to about 10 kHz.

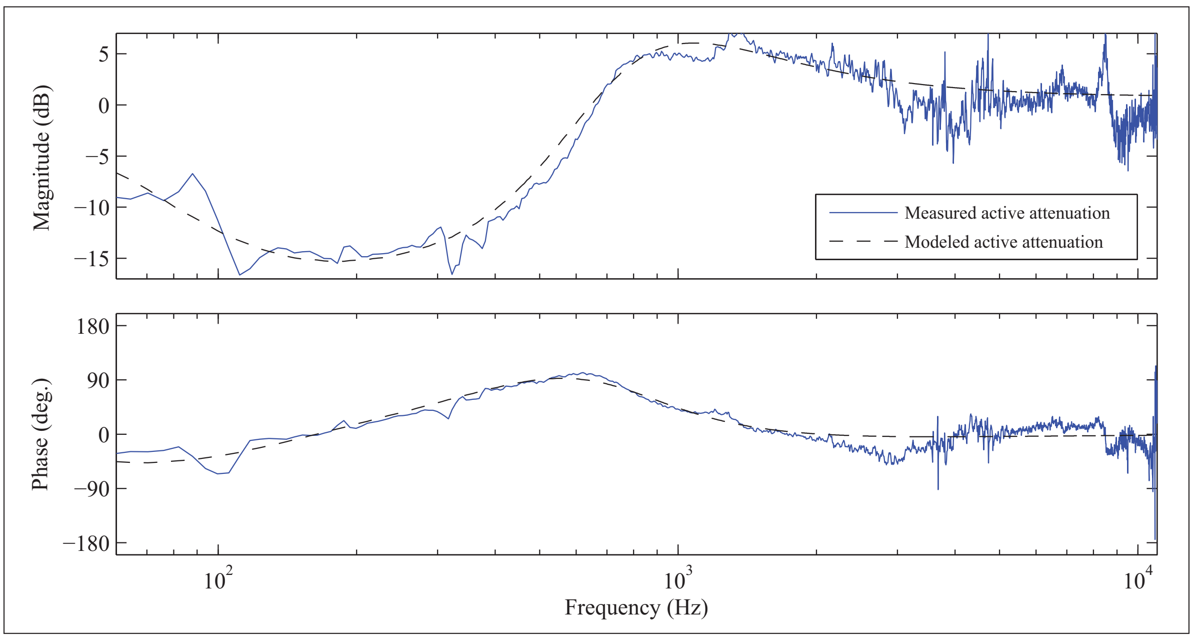

Figure 4.12 Measured and modeled active attenuation

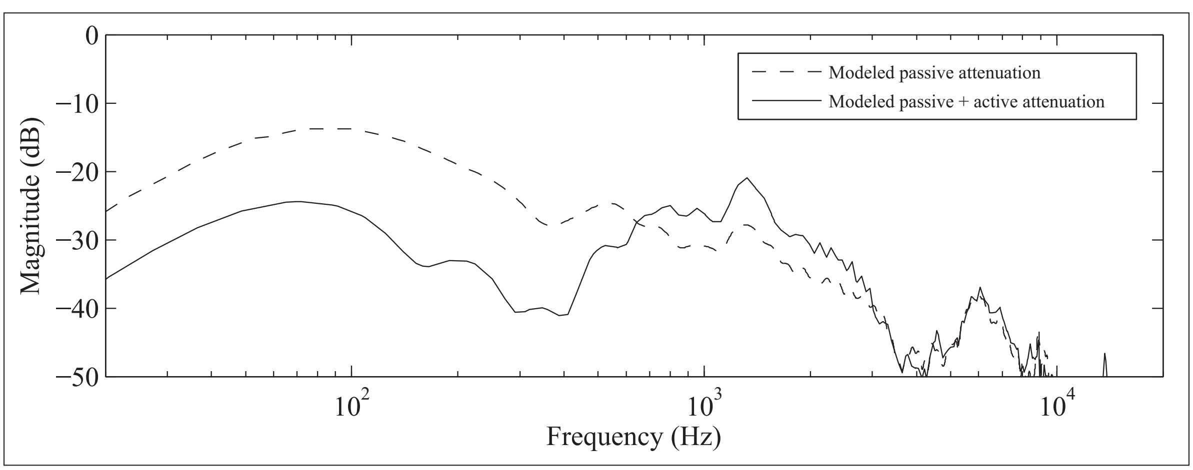

Figure 4.13 Comparison of the attenuation path with and without active attenuation

4.2 Development of the algorithms and preliminary validation

With the electro-acoustic and the attenuation paths of the conceptual view of figure 4.3 modeled and validated experimentally, it is possible to design the algorithms of the isolation effect compensation system. At this point, the distinction between the transfer function of the overall HPD and the attenuation it provides should be clearly established. In a hearing protection context, attenuation is the difference between the occluded ear auditory threshold and the open ear auditory threshold. It can be approximated by the difference between the sound at the eardrum in occluded and open conditions. Since occluding the ear canal modifies its resonance, a uniform magnitude of the transfer function of the HPD is not equivalent to a uniform attenuation.

The HPD has been designed so that the external microphone is as close as possible to the ear canal entrance. This means that some of the high frequency amplification effect of the pinna is picked up by the external microphone, and can thus be played back inside the ear canal by the loudspeaker within the limits imposed by its frequency response. In practice, the HPD sticks out of the ear canal entrance and affects the high frequency amplification of the pinna to some extent, but this is not considered for a first version of the compensation algorithm. The device must therefore only re-establish the open ear canal resonance, so that the open ear response is mimicked at a uniformly reduced and variable level. Therefore, a resonance around 2.7 kHz must be observed in its transfer function. Additionally, if the device is to provide loudness compensation, the target attenuation that includes this compensation must be defined.

Figure 4.14 shows the block diagram of the preliminary isolation effect compensation strategy. The steps to achieve a uniform perceived attenuation are listed below:

1. A compensation filter that corrects the transfer function of the electro-acoustic path so that it is uniform in magnitude;

2. A correction filter so that the transfer function of the electro-acoustic path mimics the transfer function of the open ear;

3. A variable gain, to increase and decrease the magnitude of the transfer function of the electro-acoustic path;

4. A combination correction filter: when the electro-acoustic path and the attenuation path are recombined, destructive and constructive interference will occur if the magnitudes of the paths are of the same order. Since the attenuation path is fixed, the effect of the interference depends on the gain of the electro-acoustic path, and has to be corrected with a combination correction filter so that the overall attenuation of the combined paths is uniform;

5. A loudness compensation to shape the resulting attenuation so that it is not uniform in magnitude but uniform in loudness.

   If these steps are followed and the compensations are suitable, the resulting isolation effect compensation system provides a uniform physical attenuation after step 4. After step 5, it would provide a uniform perceived attenuation.

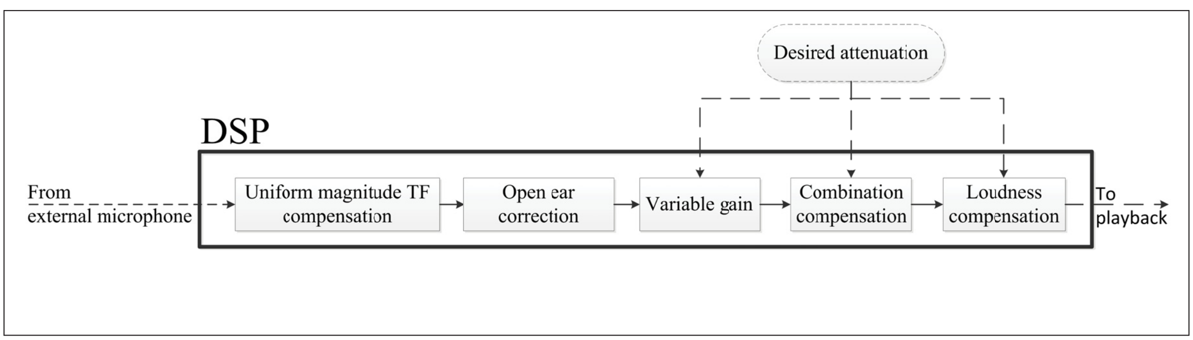

Figure 4.14 Compensation strategy and its explicit steps

Almost all the compensation blocks are to use digital filtering. Since they are all cascaded together, their order is interchangeable because they are linear in this first iteration. The following subsections detail the algorithm of the isolation effect compensation solution.

4.2.1 Uniform magnitude transfer function compensation

The compensation required to obtain a uniform magnitude transfer function from right outside the external microphone to the eardrum depends on the playback strategy that is used. If the device does not include the occlusion effect reduction system, the playback is simply the uncompensated plant. If the device includes the occlusion effect reduction system, the playback is the response of the complementary sensitivity function of equation 1.4.

An example of a uniform magnitude transfer function compensation for the uncompensated plant is presented in figure 4.15, showing the normalized original electro-acoustic path, the correction filter, composed of two second order peaking filters, and the resulting corrected electro-acoustic path. The correction filter is a simplified "mirror" filter of the transfer function of the uncorrected electro-acoustic path. The corrected electro-acoustic path transfer function is uniform in magnitude from about 100 Hz to about 8 kHz within 2 dB at worst.

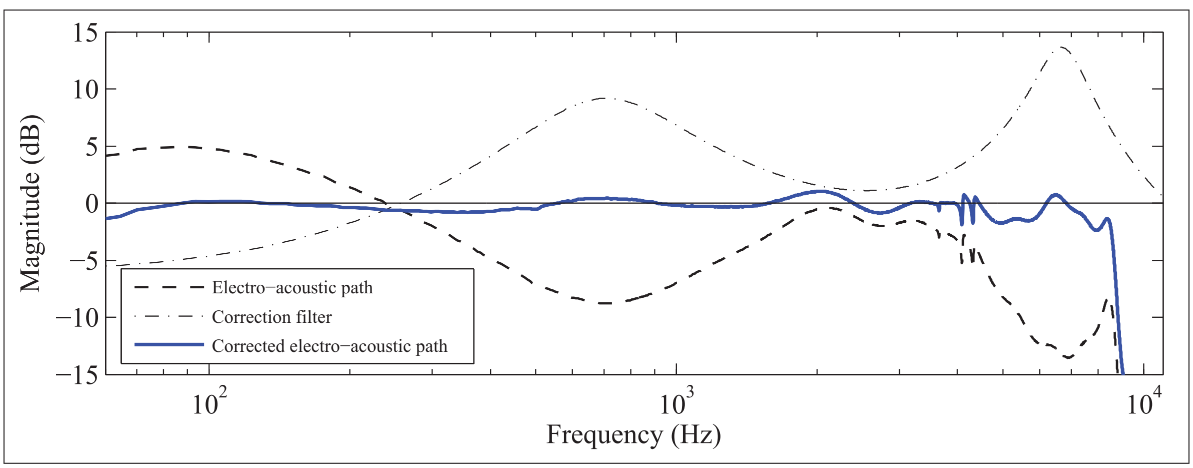

Figure 4.15 Example of uniform magnitude transfer function compensation for electro-acoustic path with playback by the uncompensated plant

4.2.2 Open ear correction

The preliminary chosen open ear correction compensation is simply a wide peak of 15 dB at 2.7 kHz corresponding to the theoretical first resonance of an open ear canal, but could easily be changed if it is not suitable. This strategy is similar to the one used by Killion *et al.* (1988). Figure 4.16 shows the target frequency response of the open ear model and how the corrected electro-acoustic path fares in comparison.

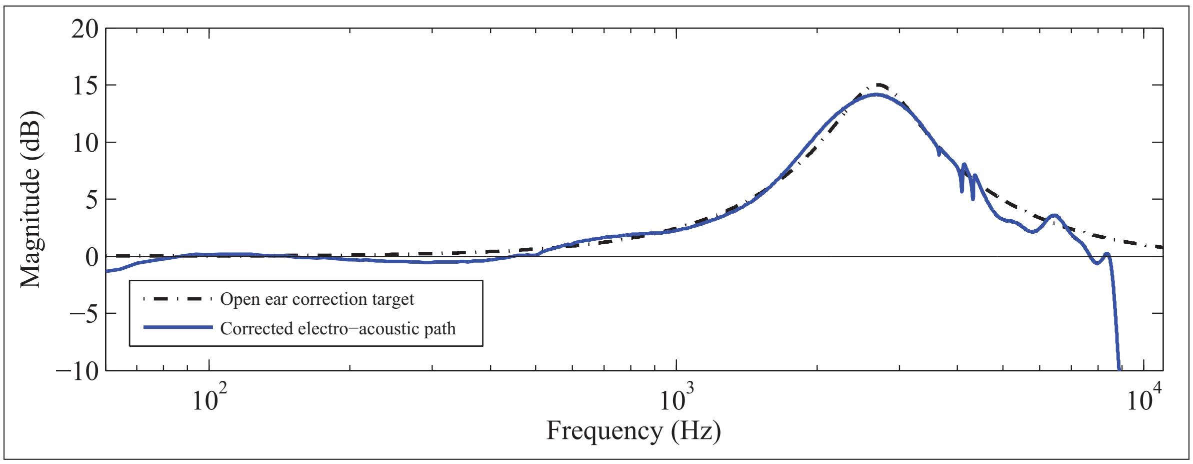

Figure 4.16   Example of an open ear canal correction

4.2.3   Variable gain and combination correction

The variable gain and combination correction are linked and thus presented together. Because of constructive and destructive interference, the acoustic combination of the electro-acoustic and attenuation paths would result in an overall transfer function that would stray from the target. The importance of these constructive and destructive interference depends on the variable gain or the electro-acoustic path: if the gain is such that the electro-acoustic path prevails on the attenuation path, the interference is insignificant. However, as the gain is lowered, the magnitudes of the electro-acoustic path and the attenuation path transfer functions become of the same order and the interference is significant. A pre-compensation filter, referred to as combination compensation, can be applied to the electro-acoustic path so that this interference is controlled. The frequency response of the filter must vary according to the variable gain, which in turn varies according to the desired attenuation, as previously shown in figure 4.14.

Figure 4.17 shows an example of achieving a combined transfer function that would result in a quasi-uniform attenuation of 15 dB. The achievable uniform attenuation is limited by the attenuation path, which provides only about 15 dB of reduction at 100 Hz. According to the models of the paths, their combination primarily causes constructive interference in the low frequencies. To compensate for this, a first order high-pass filter is used. The coefficients of an IIR first order high-pass with variable cut-off frequency, depending on the desired attenuation, can be easily calculated in real time by a DSP. It can be seen on figure 4.17 that the high pass filter permits an overall transfer function of the electro-acoustic combined with the attenuation path close to the target. Figure 4.18 illustrates the effect of the automatic procedure in providing a combined transfer function close to a target, corresponding to a uniform attenuation. As previously discussed, it can be seen that the attenuation path limits the achievable quasi-uniform attenuation to about 15 dB in the low frequencies, and the curves are converging because of this limitation. Nevertheless, the automatic and continuous combination correction would permit to reproduce the open ear response at lower levels, allowing a variable and continuous quasi uniform attenuation between 0 dB to 15 dB.

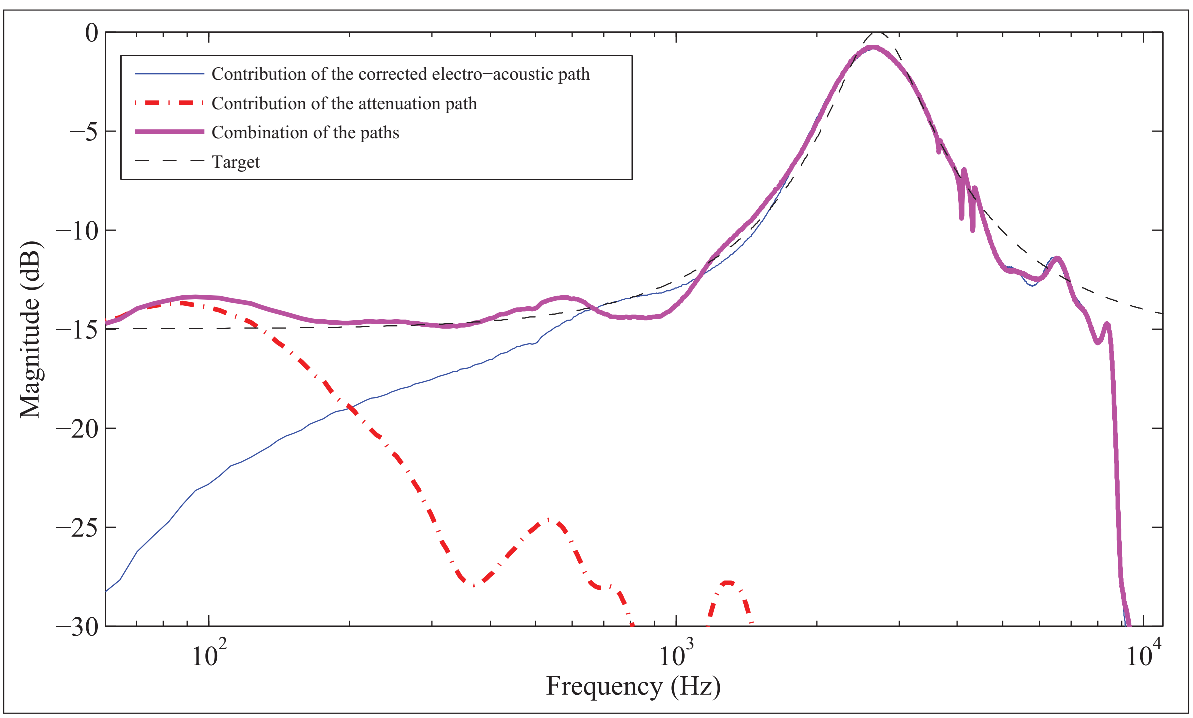

Figure 4.17 Example of obtaining a transfer function corresponding to a uniform attenuation of 15 dB within the constraints of the system

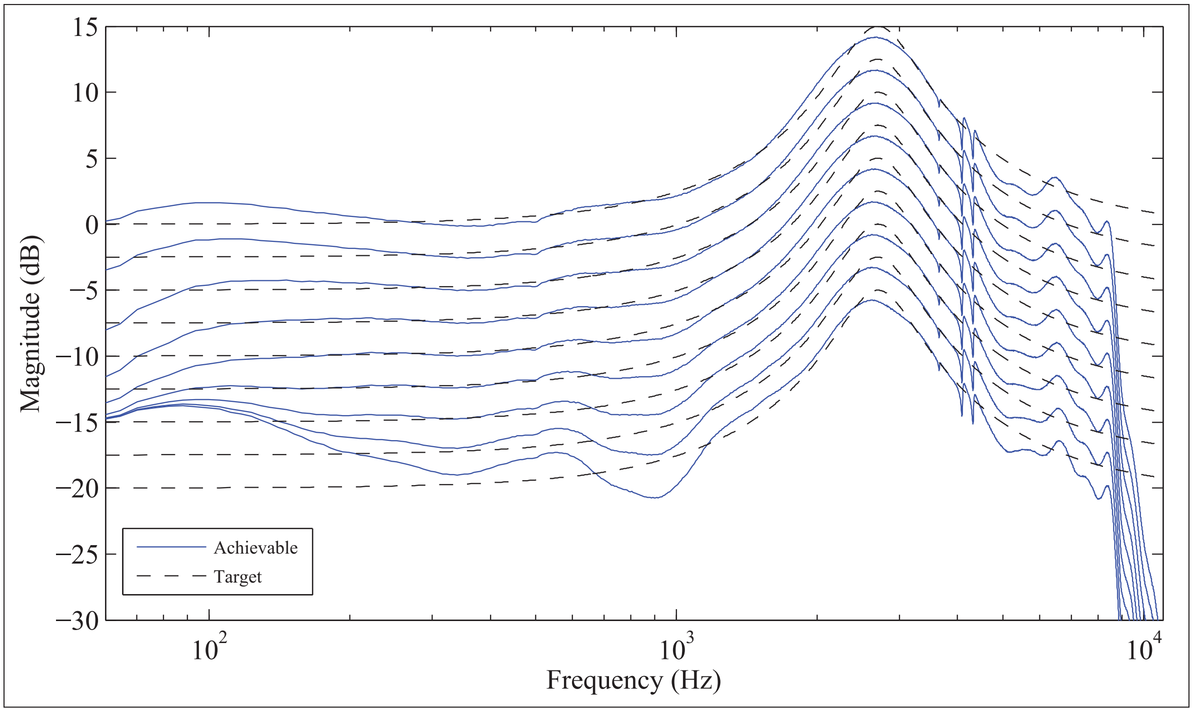

Figure 4.18 Achievable transfer functions using the automatic combination correction compared with target transfer functions corresponding to attenuation values between 0 and 20 dB

4.2.4 Preliminary experimental validation

At this point in the algorithmic steps, the device should be able to provide uniform attenuation up to about 15 dB. To validate the algorithms derived thus far, they were implemented on the ARP and the implemented system was characterized on the ATF. Figure 4.19 compares the target, predicted, and measured overall transfer functions, corresponding to quasi-uniform attenuation of 5, 10, 15 and 20 dB. The prediction is close to the measured results within 2 dB at worst. The measure follows the target curve within 2.5 dB at worst in the case of the quasi-uniform attenuation of 15 dB. As predicted, a uniform attenuation of 20 dB cannot be achieved. The differences between the prediction of the model and the measures can be explained by the slight imperfections in the model and the variability of the fit to the ear canal of the ATF, that changes the transfer function of the attenuation path and the electro-acoustic path. It can be seen that the system is indeed capable of reproducing the open-ear response at a lower level and thus provide a quasi uniform attenuation up to 15 dB, as predicted.

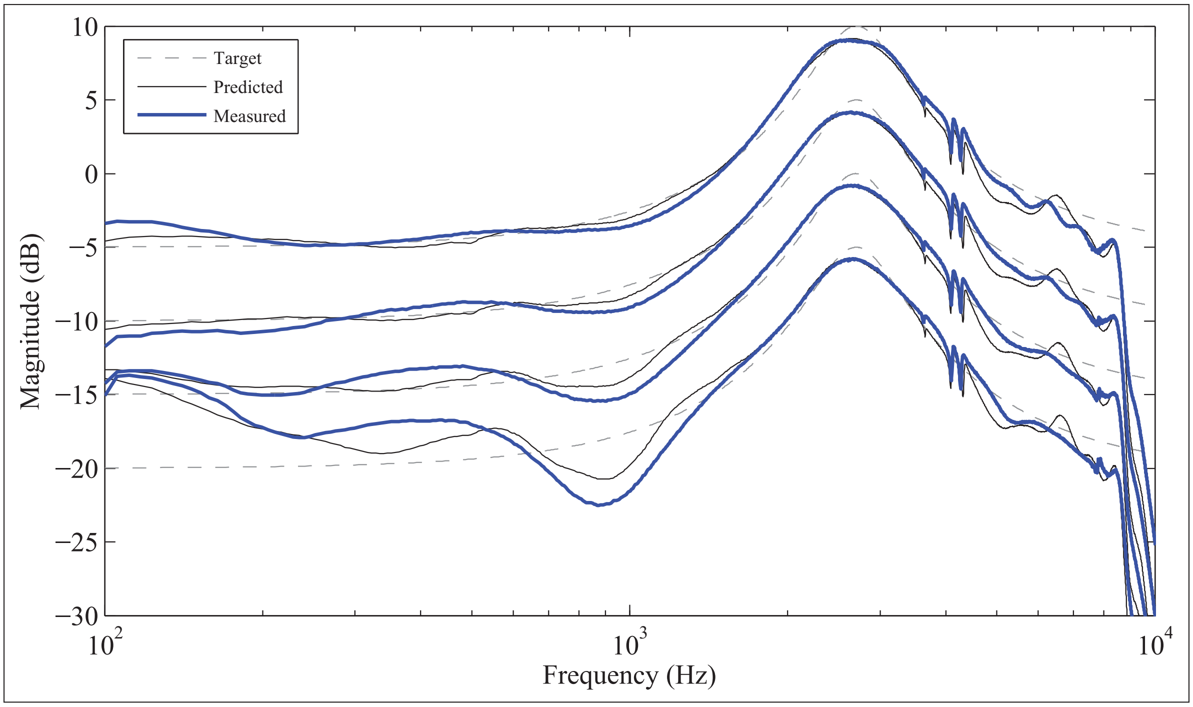

Figure 4.19 Comparison of target, predicted, and measured overall transfer functions, corresponding to quasi-uniform attenuation values of 5, 10, 15 and 20 dB

4.2.5 Loudness compensation

This section derives a method for an attenuation-dependent loudness compensation derived from ISO226:2003. The requirements that drove the method were simplicity and adaptability to any attenuation value. Loudness correction based on equal loudness contours would typically require two parameters to be exact: the sound pressure level of the original signal and the intended sound pressure level of the reproduced signal. In a hearing protection context, this translates to the sound pressure level of the signal that would be present at the eardrum and that same signal uniformly attenuated by a given value. Figure 4.20 shows the effect of reducing a sound from 90 dB(SPL) to 70 dB(SPL) on the perceived loudness as a function of frequency. It results in a corresponding decrease of 20 phons at 1 kHz, but of almost 30 phons at 100 Hz, showing that the original loudness balance is lost.

An investigation of the differences between the equal loudness curves shows that the required SPL decrease to elicit a uniform loudness decrease is fairly constant between the curve, for a given frequency. Figure 4.21 shows the difference between two equal loudness curves that are separated by 20 phons. It can be seen that although the original level is very different (90 dB(SPL) and 60 dB(SPL) at 1kHz), the required SPL difference to elicit the same loudness is very close.

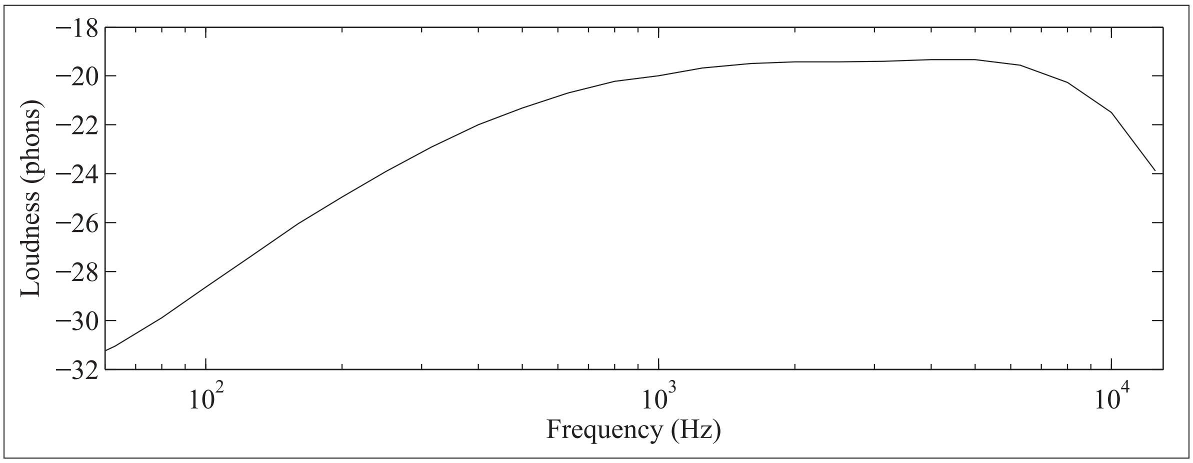

Figure 4.20 Effect of uniformly reducing a sound from 90 dB(SPL) to 70 dB(SPL) on its perceived loudness as a function of frequency

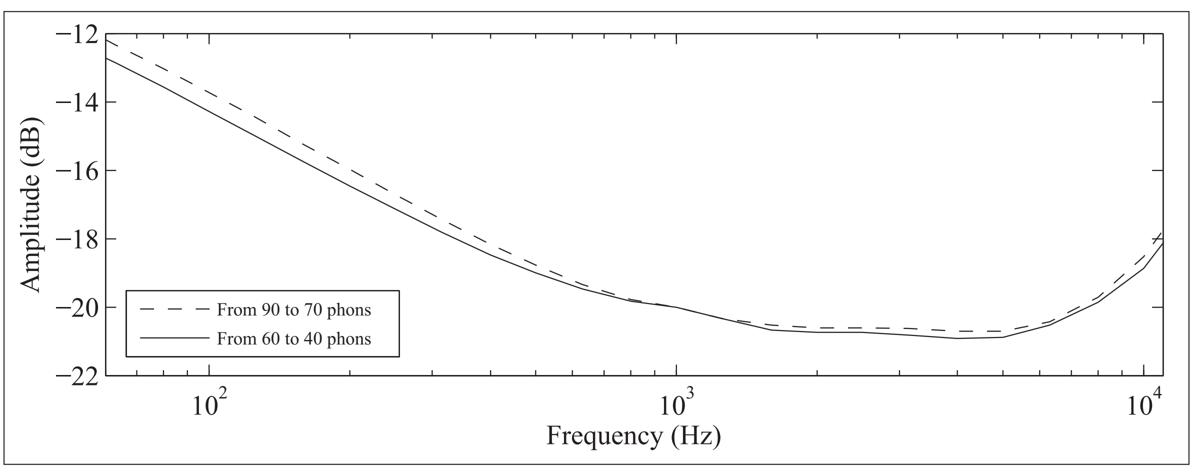

Figure 4.21 Required SPL decrease to ellicit a constant 20 phons loudness decrease between two equal loudness curves of different level

This tendency is found for different uniform decreases in loudness and different original levels, suggesting that a suitable loudness compensation could be derived independently of the original level, considerably simplifying the design. From this observation, target attenuation curves can be derived. Figure 4.22 shows the required attenuation curves to retain loudness spectral balance when decreasing the sound pressure level, derived from ISO226:2003.

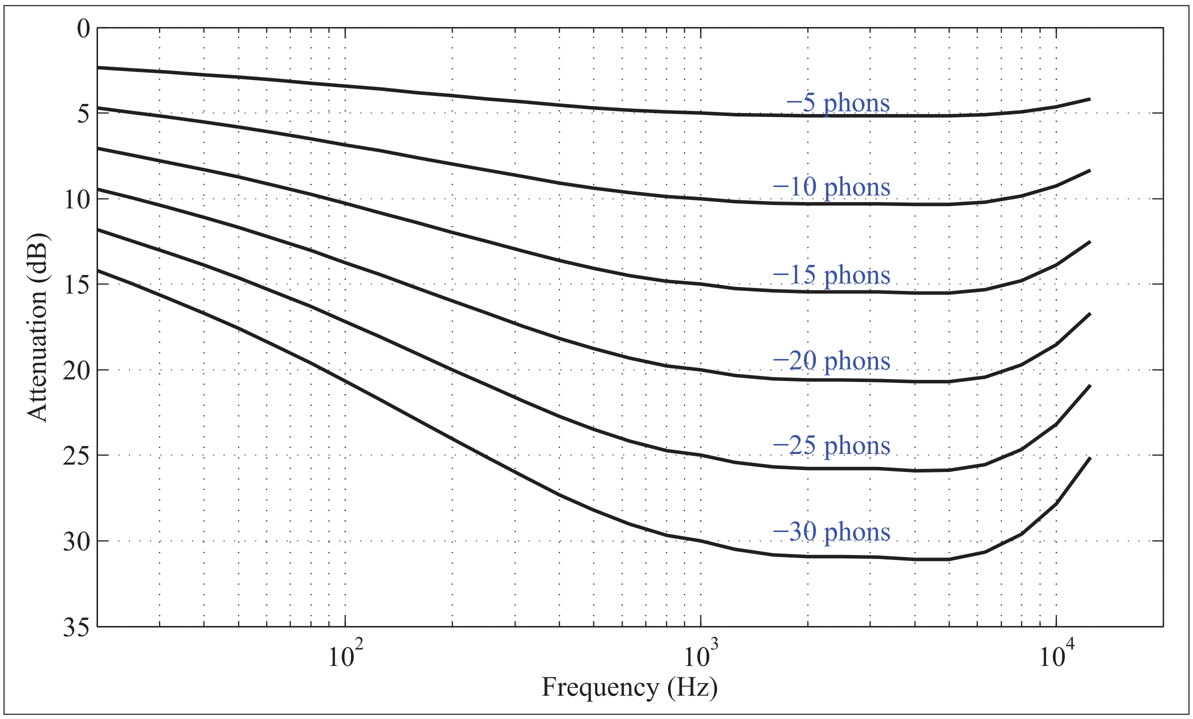

Figure 4.22 Required attenuation curves to achieve a perceived uniform attenuation in loudness derived from ISO226:2003

The required correction could then simply be a filter that matches the curves from figure 4.22, but normalized to 0 dB at 1 kHz. The curve can be closely matched with a first order low shelf filter and a second order high shelf filter, both IIR filters. This technique presents the advantage that the coefficients of these filters can be calculated in real time and therefore adapt the loudness compensation in real time for any attenuation value. A MATLAB script of the algorithm is available in appendix IV. Figure 4.23 shows two examples of fitting the target loudness correction for different uniform attenuation values, revealing that the algorithm performs well from about 100 Hz to 11 kHz.

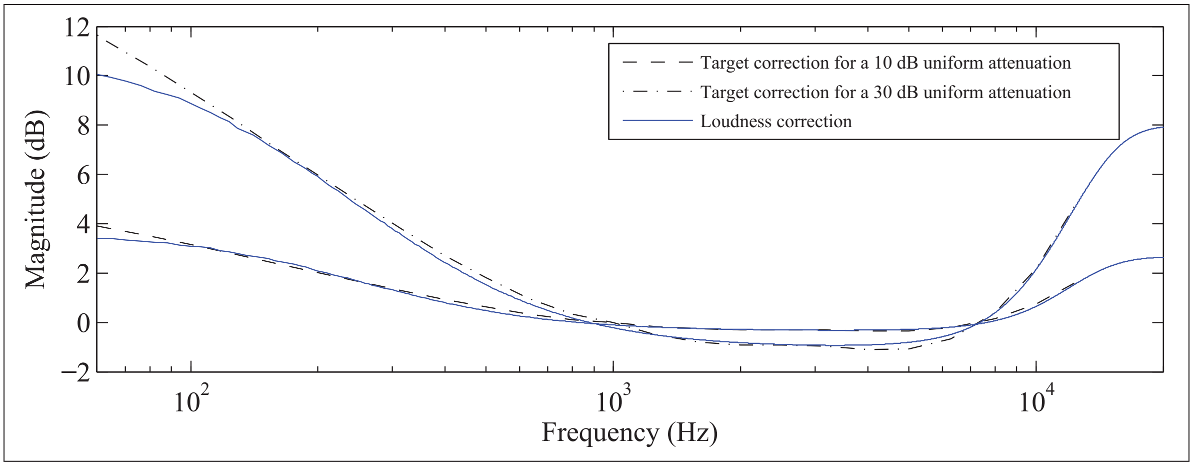

Figure 4.23 Comparison between target and achieved loudness corrections

4.3 Isolation effect compensation system in conjunction with the active occlusion effect reduction system

As previously mentioned, the AOER system has two effects on the isolation effect compensation. First, it changes the characteristics of the playback. The different playback situation requires a different "mirror" correction filter to achieve uniform magnitude of the electro-acoustic path transfer function, as shown in figure 4.24.

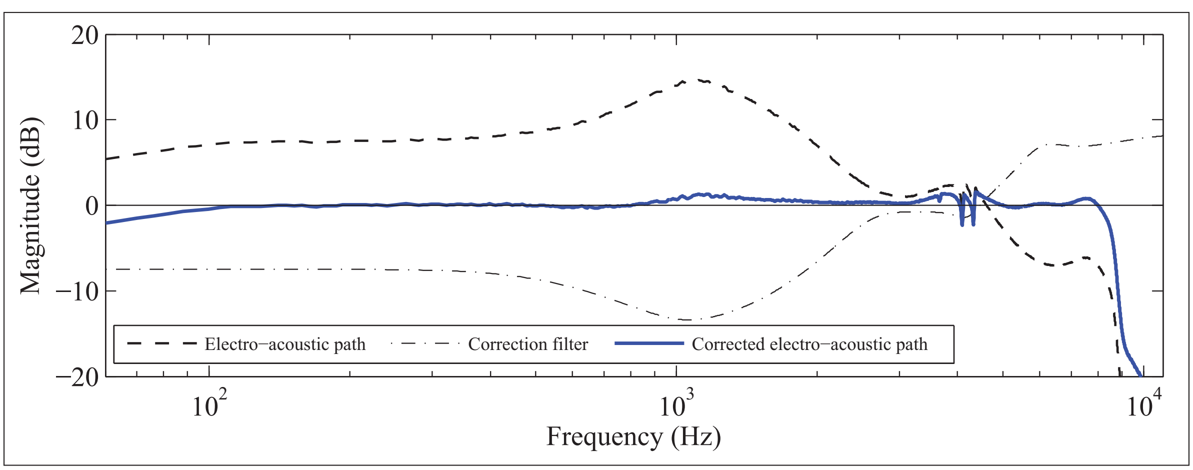

Figure 4.24 Example of uniform magnitude transfer function compensation for the electro-acoustic path with playback by the complementary sensitivity function

Second, the AOER system adds its active attenuation to the passive attenuation, resulting in a different attenuation path. The different attenuation path allows for greater uniform attenuation but requires a different combination compensation algorithm. Figure 4.25 shows an example of achieving an overall transfer function of the HPD corresponding to a uniform attenuation of 20 dB. The small ripples of about 0.5 dB are caused by FIR modeling of the system.

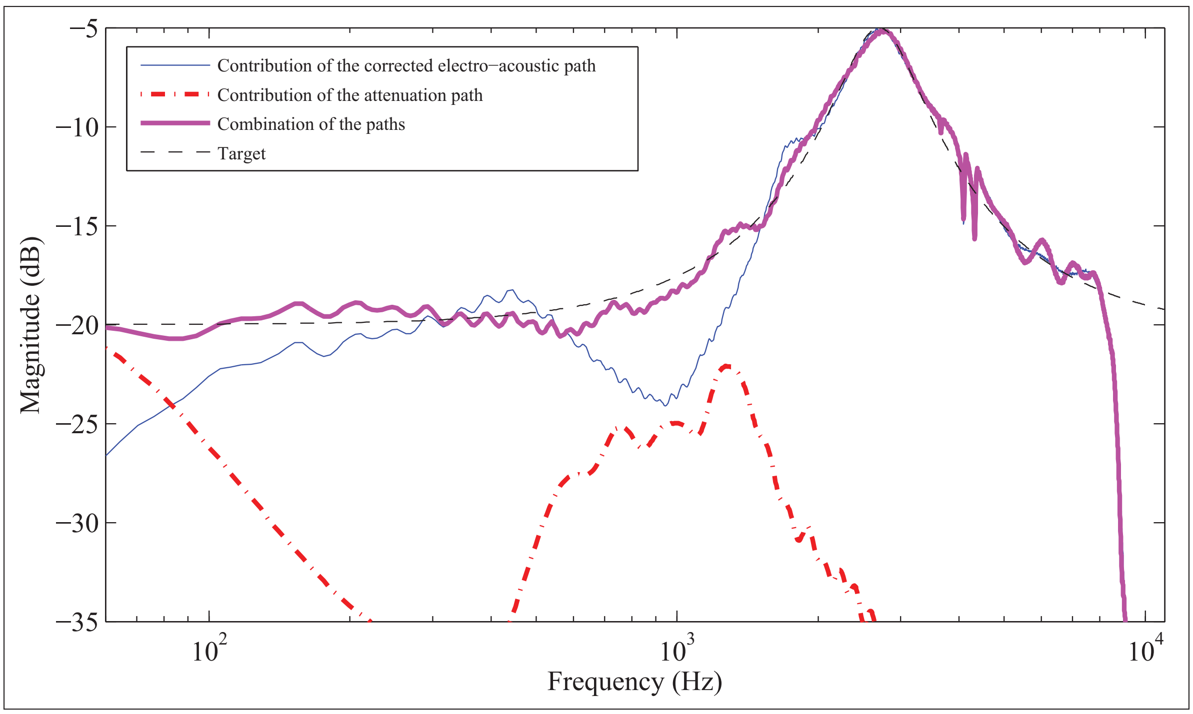

Figure 4.25 Example of obtaining a transfer function corresponding to a uniform attenuation within the new constraints of the system

Because the active attenuation lowers the magnitude of the attenuation path in the low frequencies, the maximum overall achievable uniform attenuation is expected to be greater. However, regeneration caused by the active attenuation in the mid frequencies, as could be observed on figure 4.13, is expected to impose new limits on the maximum overall achievable uniform attenuation. The interference between the two paths is destructive or constructive depending on the frequency. According to the models of the path, the recombination primarily causes constructive interference in the low frequencies. Therefore, the combination compensation filter includes a first order IIR high pass filter with varying cut-off frequency depending on the selected attenuation. Additionally, destructive and constructive interference occurs from 300 Hz to 3 kHz. Therefore, the combination compensation filter is composed of three peaking or notching filters in this band. The center frequencies are constant but the magnitude of the peak or notch depends on the selected attenuation.

Figure 4.26 shows how the automatic combination compensation algorithm enables transfer functions corresponding to quasi-uniform attenuation values from 0 to 25 dB. The achievable transfer functions follow the target transfer functions within about 2 dB for attenuation values of 0 to 20 dB, from 100 Hz to 8 kHz. The transfer function corresponding to a uniform attenuation of 25 dB has a maximum difference with the target curve of about 2.5 dB, occurring at 1300 Hz. The system could therefore provide a quasi uniform variable attenuation up to 25 dB.

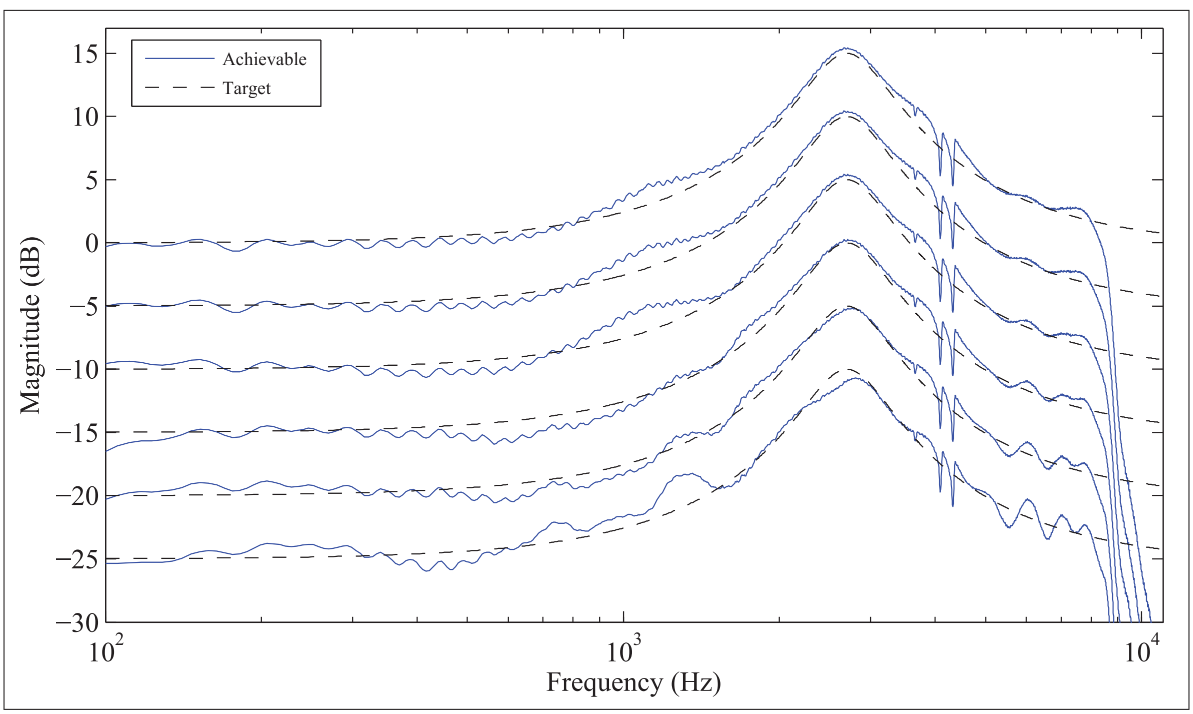

Figure 4.26 Example of obtaining a transfer function corresponding to a uniform attenuation within the new constraints of the system

Figure 4.27 shows the same scenario when including loudness compensation. A quasi-uniform decrease between 0 and 25 phons would be possible within the constraints of the system. The algorithms match the target transfer functions corresponding to a uniform decrease in phons within 3 dB. The details of the parameters of the combination correction have been changed, because the loudness compensation has an impact on the phase of the electro-acoustic path and therefore changes the effect of the interference between the paths. It can be seen that the loudness compensation is of little use at the very high frequencies because the loudspeaker cannot produce significant SPL at those frequencies. Nevertheless, the system could provide a continuous and variable perceived uniform attenuation between 0 dB and 25 dB.

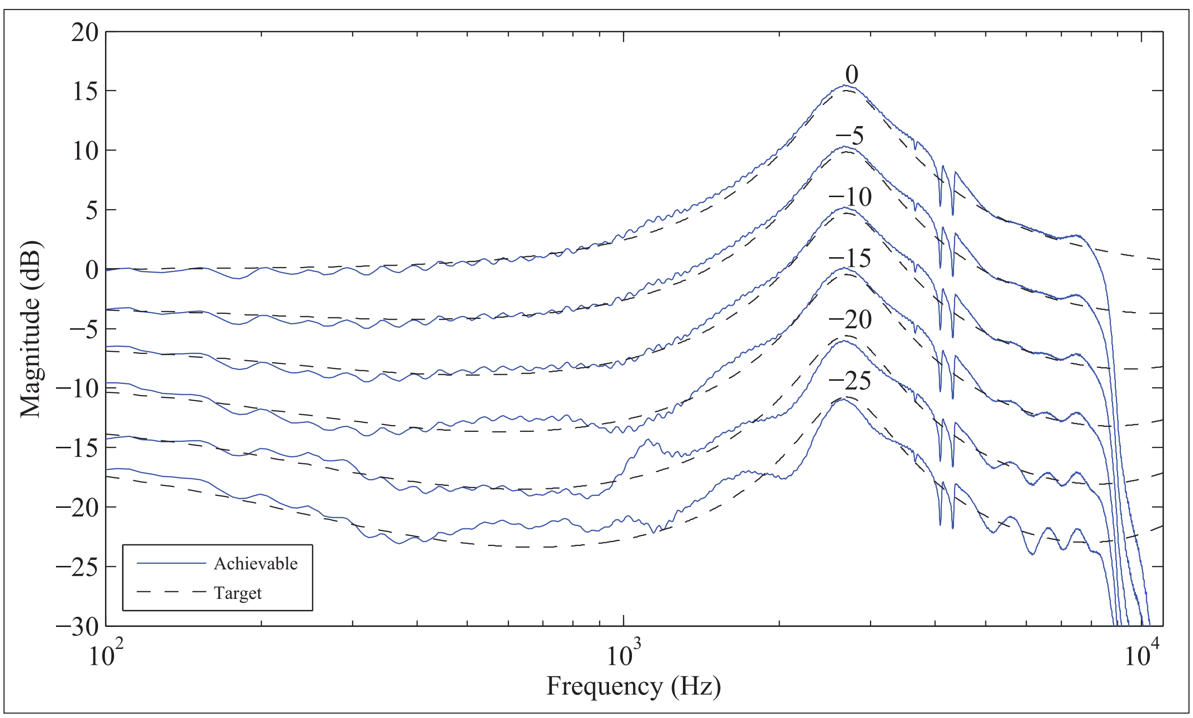

Figure 4.27 Example of obtaining a transfer function corresponding to a perceived uniform attenuation between 0 and 25 phons within the new constraints of the system

4.4 Discussion regarding the solution to isolation effect

In this chapter, a solution to the isolation effect is presented in the form of an isolation effect compensation system. A method was presented to achieve a uniform attenuation with a HPD comprising an external microphone, a DSP, and an internal loudspeaker. A system was implemented and validated experimentally, providing continuous quasi-uniform attenuation between 0 and 15 dB, on an ATF. This validates the method and shows that it is appropriate to design isolation effect compensation algorithms by modeling each element of the system composed of the attenuation path and the electro-acoustic path. Following this validation, the same method was used to predict the interaction between the implemented isolation effect compensation system and the implemented occlusion effect system, when combined. Such a combined system could provide continuous uniform attenuation between 0 and 25 dB, or continuous uniform perceived attenuation between 0 and 25 phons.

The performance of the isolation effect compensation system can therefore increase if used in conjunction with the active occlusion effect reduction system: the active attenuation contributes to make the attenuation path less significant, thus increasing the maximum uniform attenuation that can be achieved. However, the beneficial effect of the AOER system is mitigated by the regeneration that occurs around 1 kHz. Even though the active attenuation lowers the magnitude of the attenuation path by as much as 15 dB in the low frequencies, the magnitude of the attenuation path is increased by 5 to 7 dB in the mid frequencies due to regeneration. As a consequence, the maximum achievable uniform attenuation without and with active attenuation increases from 15 dB to 25 dB; an increase of only 10 dB. Therefore, the maximum uniform attenuation requirement of 30 dB, defined in section 2.3, cannot be met with the system in its current state.

The upper limit on the overall attenuation that can be obtained is defined by the magnitude of the attenuation path. To obtain a greater attenuation, two options could be considered:

1. Increasing the passive attenuation of the device; when the earpiece was constructed, much more design effort was directed towards the electro-acoustic path than the passive attenuation of the device, which received almost no consideration. In the next iteration, requirements for the passive attenuation should be defined and observed;

2. Decreasing the regeneration caused by the active attenuation; though the AOER system provided very little active regeneration when it was tested on humans, the increased magnitude of the plant response in the ATF ear canal compared to the human subjects' ear canals over-estimates both the active attenuation that could be achieved on humans, in the low frequencies, and the active regeneration in the mid frequencies. The requirements regarding the disturbance rejection of the AOER system, affecting both the magnitude of the occlusion effect reduction and the active attenuation, were defined in terms of occlusion effect reduction only. In the next iteration, requirements in terms of active attenuation should be defined and observed.

An efficient and realistic combination of these two options would be an earplug that has a passive attenuation of 20 dB and an active attenuation of 8 dB in the low frequencies, where the passive attenuation is usually less efficient. If the passive attenuation is such that more attenuation than 28 dB is observed in the mid frequencies, where the passive attenuation is usually more efficient than at low frequencies, some regeneration in the mid frequencies could be allowed. A realistic maximum target regeneration would be around 2 dB. If these conditions were met, the requirement of 28 dB of attenuation could be achieved.

The combination of the path was found to be very sensitive to the phase and magnitude of both the electro-acoustic path and the attenuation path. When the overall desired attenuation is set at 20 or 25 dB, near the limit imposed by the attenuation path, inter-user variability of both the electro-acoustic path and the attenuation path is expected to yield results that stray from the desired attenuation. On the contrary, when the system is used for a moderate attenuation of 15 dB and less, the variability of the attenuation path has little effect on the resulting attenuation, so only the variability in the electro-acoustic path would affect the outcome.

Regarding loudness compensation, the isolation effect compensation system is able to accurately meet the required correction in the low frequencies, but not in the high frequencies. This is due to the loudspeaker being unable to produce significant SPL past 9 kHz. To overcome this limitation in further research, a small balanced armature driver, efficient at producing high frequencies, could be included in the design of the earpiece.

Overall, the isolation effect compensation system designed as part of this project is able to meet most of the requirements, indicating that isolation effect compensation using the proposed architecture and method could be a suitable solution to the isolation effect experienced by musicians.

CONCLUSIONS

This thesis investigated the design of an active hearing protection device for musicians. Musicians are noise-exposed workers that heavily rely on their hearing to do their work, but they are reluctant to use hearing protection devices because these modify the acuteness of the musicians' auditory perception. The occlusion effect and the isolation effect were identified as two detrimental factors in the acceptance of hearing protectors by musicians. Both effects were studied through a literature review in order to define the characteristics of a hearing protection device that has a minimal negative impact on a wearer's perception. A solution to the occlusion effect was presented in the form of an active noise control of the sound resulting from occlusion effect in the ear canal: an active occlusion effect reduction system. the requirements of this AOER system have been defined as to reduce the occlusion effect by about 8 to 17 dB at 250 Hz and 6 to 14 dB at 500 Hz. At 1 kHz, the device should ideally provide 6 dB of reduction. However, this proves to be difficult because of the phase shift of transducers around this frequency.

Two prototypes of the AOER system, based on an universal-fit and a moldable-fit earpiece, have been designed, implemented and characterized in this work. The occlusion effect reduction and active attenuation offered by the universal-fit and moldable-fit prototypes was respectively measured to be 8.5 and 12 dB at 250 Hz, 6 and 10 dB at 500 Hz, and 3.5 and -3 dB at 1000 Hz; the minus sign means that regeneration was measured at 1000 Hz for the moldable-fit prototype. The requirements for the solution were met by both prototypes, and it is concluded that active occlusion effect reduction could be an efficient solution to the occlusion effect experienced by musicians.

Further research is needed to validate the solution perceptually, with musicians. Through further research, the occlusion effect reduction that would solve the problem experienced by musicians could be identified and the SPL in the ear canal caused by occlusion effect as a musician is playing, measured. This would help define more precise requirements for a second iteration of an active occlusion effect reduction prototype. Regarding the isolation effect, two factors were identified through the literature review as causes of the isolation effect when wearing earplugs: the modification of the ear canal resonance and a non-uniform attenuation. These factors have been addressed by other solutions in the past, yet the isolation effect remains. A literature review of the non-linearity of loudness perception and loudness models indicates that a perfectly uniform attenuation would not be perceived as uniform. Therefore, a third factor was identified to contribute to the isolation effect, the non-linearity of loudness perception.

A solution to the isolation effect was presented in the form of a HPD complemented with digital signal processing capabilities: an isolation effect compensation system. Such an isolation effect compensation system was required to provide a transfer function that mimics the response of an open ear canal at a lower level to achieve a variable uniform attenuation between 0 and 30 dB, and a variable uniform perceived attenuation, in phons, using a loudness compensation algorithm. A method was proposed to achieve those requirements, and a prototype of isolation compensation system was predicted and validated to be capable of offering a quasi-uniform attenuation between 0 and 15 dB. This validated the method used in this work, but the isolation effect compensation system could not meet the requirement of 30 dB of uniform attenuation, offering only 15 dB.

Using the validated method, it was determined that if the isolation effect compensation system is used in conjunction with the occlusion effect system implemented in this thesis, the maximum achievable uniform attenuation would be around 25 dB, and the maximum achievable perceived uniform attenuation would be around 25 phons. The active attenuation provided by the AOER system would permit this increase in performance. Although it is still insufficient to meet the requirement of 30 dB, an attenuation of 25 dB would be sufficient in most cases. To achieve the requirement of 30 dB, more attention should be given, in the design stage, to the passive attenuation of the device, which was measured to be only 15 dB at 100 Hz, on an ATF.

Nevertheless, it is concluded that isolation effect compensation using the proposed architecture and method could be a suitable solution to the isolation effect experienced by musicians.

Further research is needed to validate the proposed solution to the isolation effect on musicians, rather than on an ATF. Through further research, the shape of the attenuation that musicians would feel comfortable with, if any, could be identified and the hypothesis that the non-linearity of loudness perception contributes to the isolation effect, and to what extent, could be verified and quantified.

The greatest limitation in both solutions presented in this thesis is the variability of the ear canal, causing and comprising inter-user variability of the acoustic seal, the residual occluded volume and the impedance of the eardrum. While an efficient solution could be provided to a single user using the methods described in this work, many users could not be satisfied with the same fixed solution. Therefore, an active hearing protection device for musicians should feature some kind of adaptability to the user. In particular, the plant response, the attenuation path, and the electro-acoustic path need to be known on a given user to apply suitable compensations to the occlusion and isolation effects. There is a need for further research to identify the exact required adaptation and the method to do so.

APPENDIX I

VARIABILITY OF PLANT RESPONSE BETWEEN INSERTIONS

A simple test was performed to assess what interface to the ear would provide the least amount of variability between insertions. The preliminary investigation consisted of measuring the transfer function between the internal loudspeaker and internal microphone for prototypes built with two different approaches: a generic-fit earpiece, using universal-fit eartips, and a custom-fit earpiece. A sequence of the preliminary tests consisted of inserting the earpiece in the ear canal, measuring its transfer function, and removing the earpiece. Three repetitions were performed by the author for the universal-fit and custom-fit earpieces. The universal-fit earpiece was also tested on the ATF for comparison purposes. The least variability between insertions is observed with the custom earpiece on the author, as shown on figure I-1. The most variability is observed with the universal-fit earpiece on the author, as shown on figure I-2. There is less variability between insertions when the universal-fit earpiece is inserted on the ATF, as shown on figure I-3, than on the author. The variability between insertions is about 2 dB for the custom-fit earpiece on the author, about 3 dB for the universal-fit earpiece on the ATF, and about 5 dB for the universal-fit earpiece for the author. More variability could be observed if more than three insertions were compared, but these preliminary results indicate that the custom-fit approach seems to be most suited to limit variability of the plant between insertions, allowing for theoretically better performance on a single user.

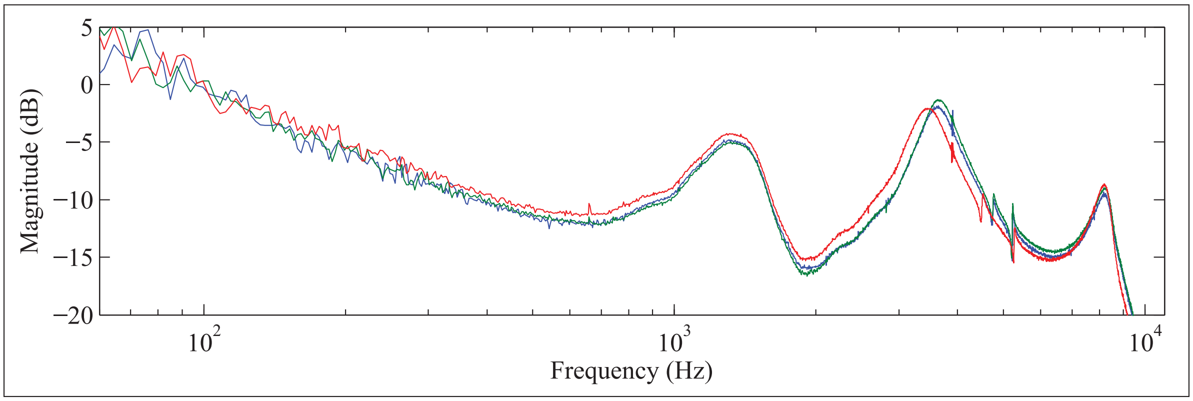

Figure-A I-1 Plant responses for three insertions of a custom-fit earpiece in the author's ear

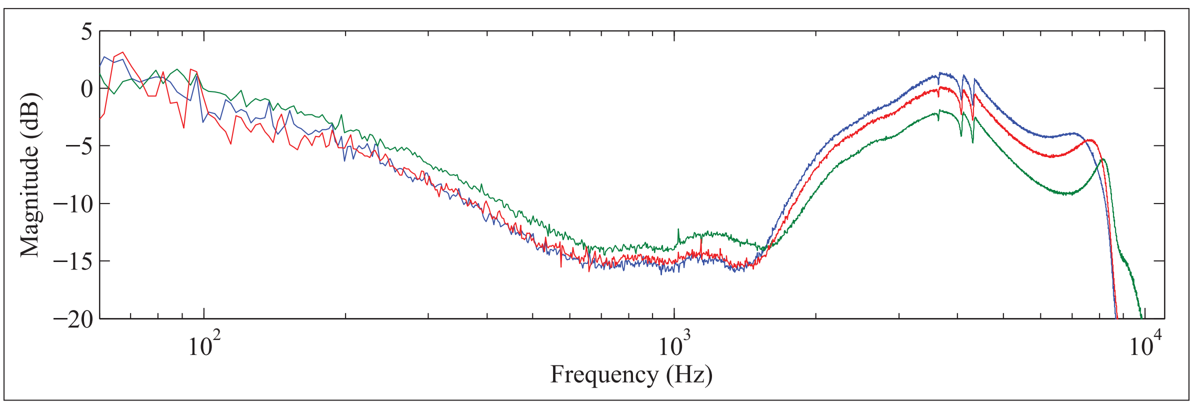

Figure-A I-2 Plant responses for three insertions of an universal-fit earpiece in the author's ear

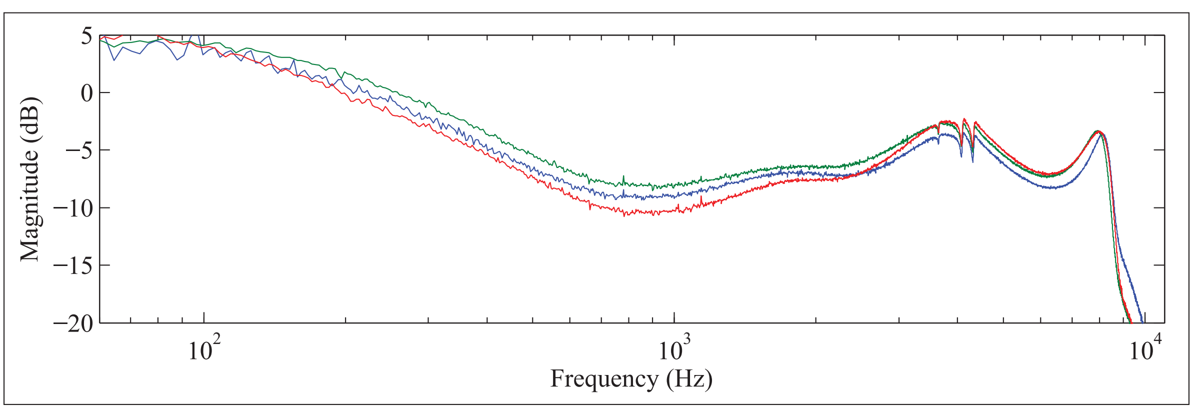

Figure-A I-3 Plant responses for three insertions of an universal-fit earpiece in the ATF's ear

APPENDIX II

SYSTEM IDENTIFICATION PROCEDURE

A system identification procedure involving a FIR filter was used to model the transfer function of different elements throughout this work. Figure II-1 illustrates the idea behind system identification by an adaptive FIR. A wide band signal $x(n)$ is applied to an unknown system and its output $y(n)$ is acquired. The signal $x(n)$ is also sent to the input of an adaptive FIR filter producing an output $y'(n)$, that is subtracted from $y(n)$, yielding an error signal. The error signal is used to adapt the coefficients of the FIR filter, and the process is repeated. As the filter converges to match the unknown system, the error diminishes until a reasonable identification of the unknown system occurs. A LabVIEW^{TM} program was implemented by the author to obtain a quick way of measuring transfer functions and perform system identification. A screen shot of a part of the program showing the system identification graphical user interface is shown in figure II-2. In that figure, the error signal is represented is the top left graph, the bottom left graph shows the filter coefficients, and the two rightmost graphs display the compared amplitude and phase of the measured transfer functions of the unknown system and the FIR filter. The program takes the length of the desired FIR filter as an input parameter as well as a constant that affects the adaptation speed and precision, called step size. Visual inspection of the measured transfer function of the unknown system and the transfer function of the FIR filter allows assessment of the suitability of the identification. The FIR filter coefficients can then be exported in MATLAB^{TM} to be used as a model of the identified system.

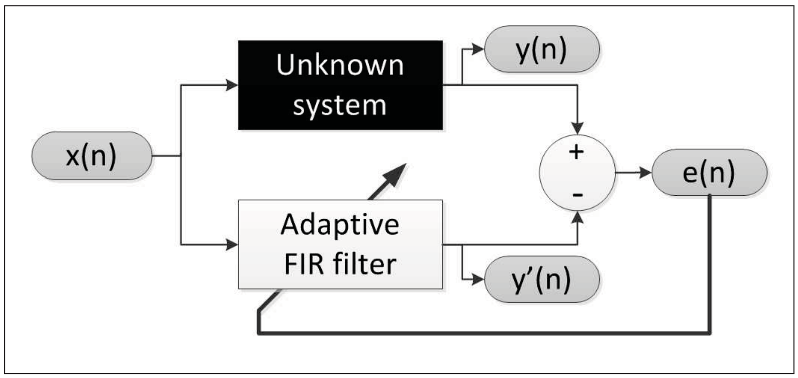

Figure-A II-1   Block diagram of the system identification procedure

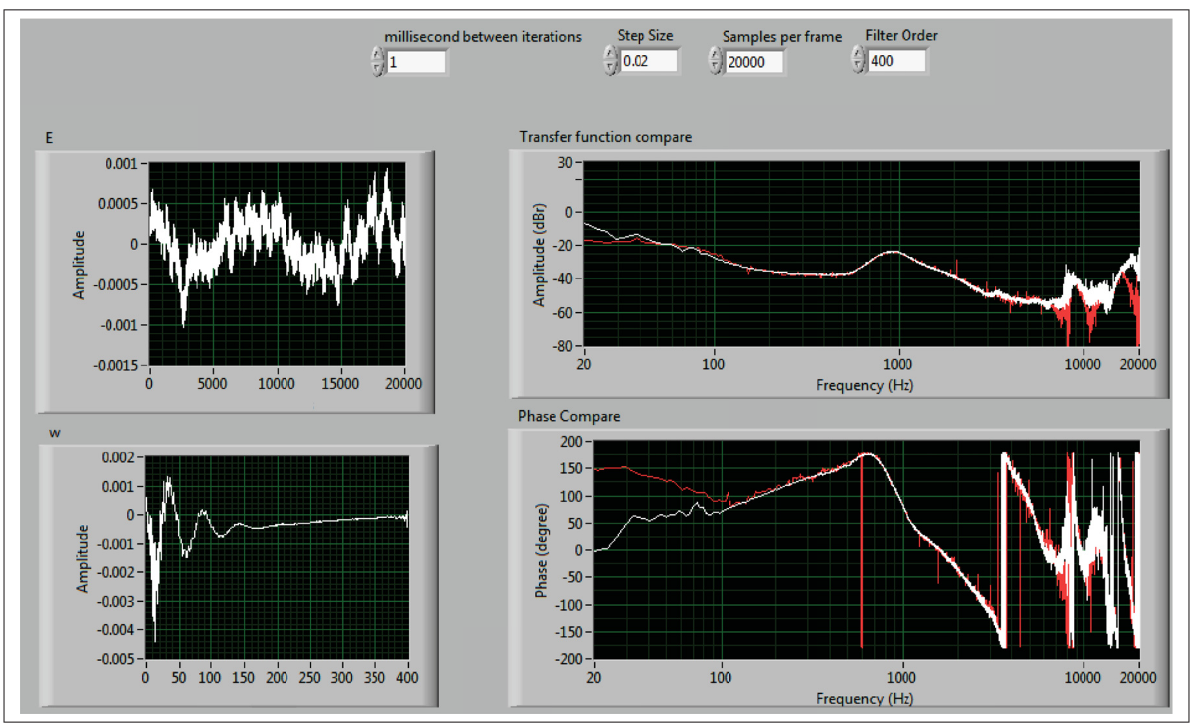

Figure-A II-2   Screen shot of the program used for system identification

APPENDIX III

CONTROLLER CIRCUITS

The controller implemented in this project uses one lead and one lag second order compensator, as well as one difference amplifier. The Tow-Thomas biquad with feedforward topology, depicted in figure III-1, was used to implement the second order lead and lag compensators. The transfer function of a second order controller $H(s)$ is given by equation A III-1. The implementable transfer function using the Tow-Thomas topology is given by equation A III-2, expressed in terms of the components of the topology.

$$H(s) = K\left(\frac{s^2 + 2\zeta_z\omega_z + \omega_z^2}{s^2 + 2\zeta_p\omega_p + \omega_p^2}\right) \tag{A III-1}$$

$$\frac{V_o(s)}{V_i(s)} = \frac{\left(\frac{C_1}{C_2}\right)^2 s^2 + \frac{1}{C}\left(\frac{1}{R_1} + \frac{r}{RR_3}\right)s + \frac{1}{C^2RR_2}}{s^2 + \frac{1}{QCR}s + \frac{C^2}{R^2}} \tag{A III-2}$$

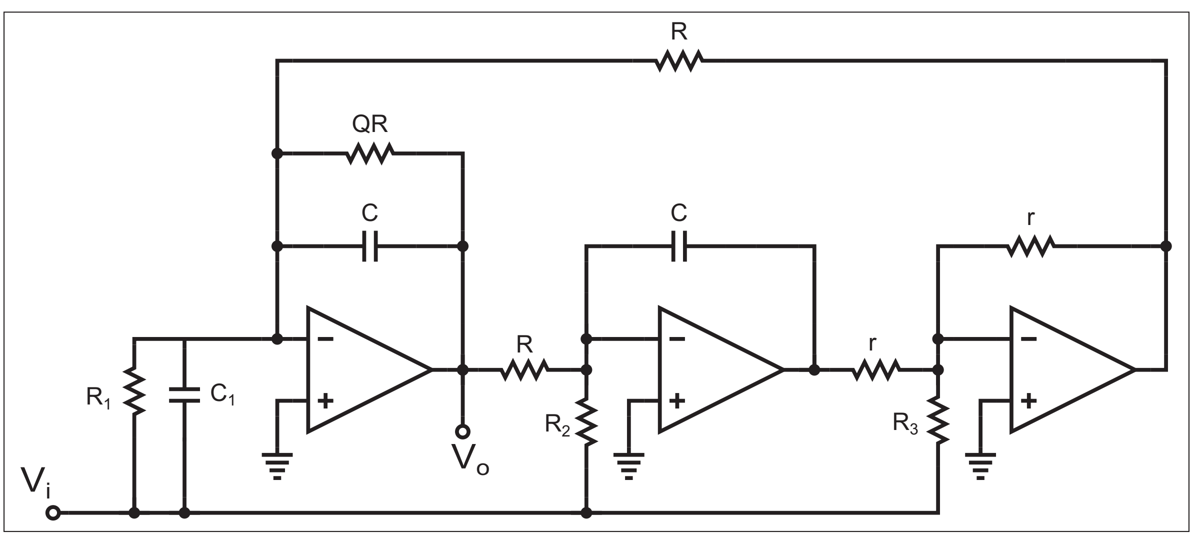

Figure-A III-1 Tow-Thomas biquad with feedforward topology

Posing $H(s) = \frac{V_o(s)}{V_i(s)}$, $K = 1$, $C_1 = C$, $r = 1000\Omega$ and $R_3 = 10000\Omega$ yields equations A III-3, A III-4, A III-5 and A III-6, allowing to determine the required values of the discrete electronic components.

$$R = \frac{1}{\omega_p C} \tag{A III-3}$$

$$Q = \frac{1}{2\zeta_p \omega_p RC} \tag{A III-4}$$

$$R_1 = \frac{1}{2\zeta_z \omega_z C + R/10} \tag{A III-5}$$

$$R_2 = \frac{1}{RC^2\omega_z^2} \tag{A III-6}$$

Table III-1 summarizes the values of the parameters that were chosen for the two second order lead and lag compensators.

Tableau-A III-1 Values of the parameters for one lead and one lag second order compensators

Parameters	**Lag compensator**	**Lead compensator**
ζ_z	0.5	0.4
ω_z	$2\pi2500$	$2\pi70$
ζ_p	0.4	1
ω_p	$2\pi700$	$2\pi105$

A negative feedback loop and a gain K is applied to the cascaded compensators using a difference amplifier, shown in figure III-2, with $R_4 = R_5$ and $R_6 = R_7$ yielding equation A III-7. Pre-amplifiers are used to remove voltage bias from the microphones. The complete circuit of the controller is shown in figure III-3 and the layout of the PCB is shown in figure III-4.

$$V_o = \frac{R_6}{R_4}(V_2 - V_1) = \frac{R_7}{R_5}(V_2 - V_1) = K(V_2 - V_1) \tag{A III-7}$$

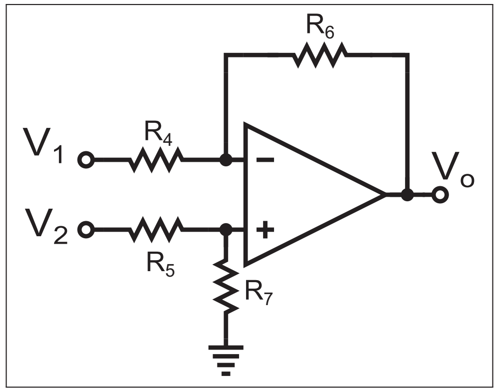

Figure-A III-2   Difference amplifier circuit

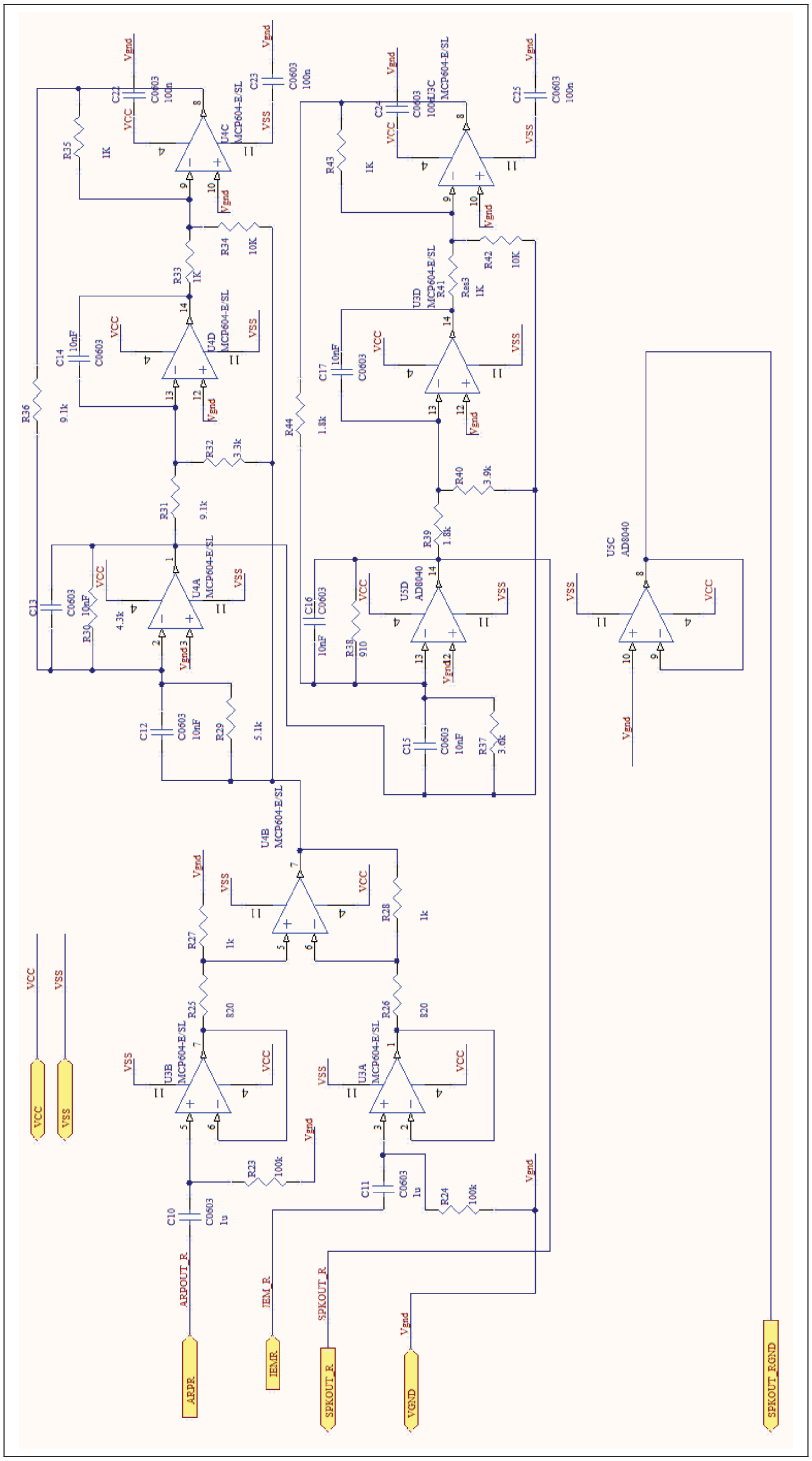

Figure-A III-3 Schematics of the right channel of the AOER controller

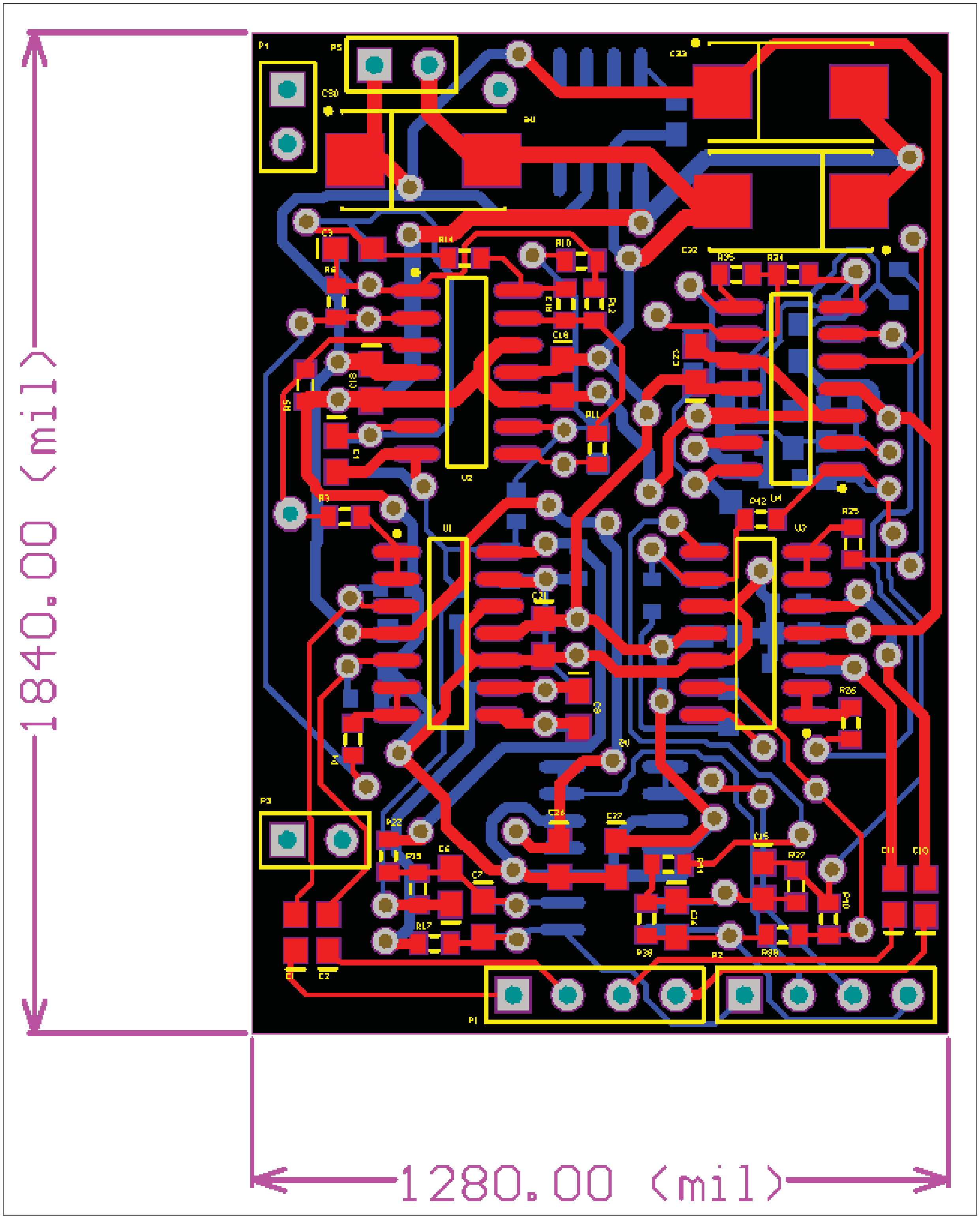

Figure-A III-4   PCB layout of the complete AOER controller

APPENDIX IV

LOUDNESS COMPENSATION SCRIPT

Below is a MATLAB script of the proposed loudness correction algorithm that could calculate coefficients in real time and adapt the correction to a changing attenuation or volume.

```
function [bl,al,bh,ah]  = HPDLC(att,fs)
% This function returns IIR filters coefficients of lowshelf and highshelf
% filters that approximate the required correction that is needed to obtain
% a uniform PERCEIVED attenuation from a uniform attenuation, according
% to ISO226:2003

% Parameters
% Outputs
% bl, al: coefficients of the lowshelf filter;
% bh, ah: coefficients of the highself filter;
% Inputs
% att    : the uniform decrease from the original signal or uniform
           %attenuation value. Designed to work between 0 and 30 dB.
% fs     : The sampling frequency.

%Typical usage
% [x,fs]=wavread('Audiofile.wav');
% y=x;% original signal
% att=20; %attenuation in dB
% yatt=x*10^(-att/20); %attenuated signal
% [bl,al,bh,ah]  = HPDLC(att,fs); %coefficients of filters
% yattcorr=filter(bl,al,yatt); %apply the lowshelf on the att. signal
% yattcorr=filter(bh,ah,yattcorr); %apply the highshelf on the att. signal
%
G=0.4*att;
A=-0.035*att;
fcl=240-4*att;
fch=11400+60*att;
HG=0.3*att;

%lowshelf section - 1st order

V0 = 10^(G/20);
A = 10^(A/20);
H0sur2 = (V0-1)/2;

ab = (tan(pi*fcl/fs)-1)/(tan(pi*fcl/fs)+1);
bl= [H0sur2*(ab+1)+ab H0sur2*(ab+1)+1]*A;
al= [1 ab];

%highshelf section - 2nd order

K = tan((pi * fch)/fs);
V0 = 10^(HG/20);

b0 = (V0+sqrt(2*V0)*K+K^2)/(1+sqrt(2)*K+K^2);
b1 = (2 * (K^2 - V0)) / (1 + sqrt(2)*K + K^2);
b2 = (V0 - (sqrt(2*V0)*K) + K^2) / (1 + sqrt(2)*K + K^2);
a1 = 2*(K^2-1)/(1+sqrt(2)*K+K^2);
a2 =  ((1 - sqrt(2)*K) + K^2) / (1 + (sqrt(2)*K) + K^2);

ah = [  1, a1, a2];
bh = [ b0, b1, b2];
```

APPENDIX V

PICTURES

1 Auditory research platform

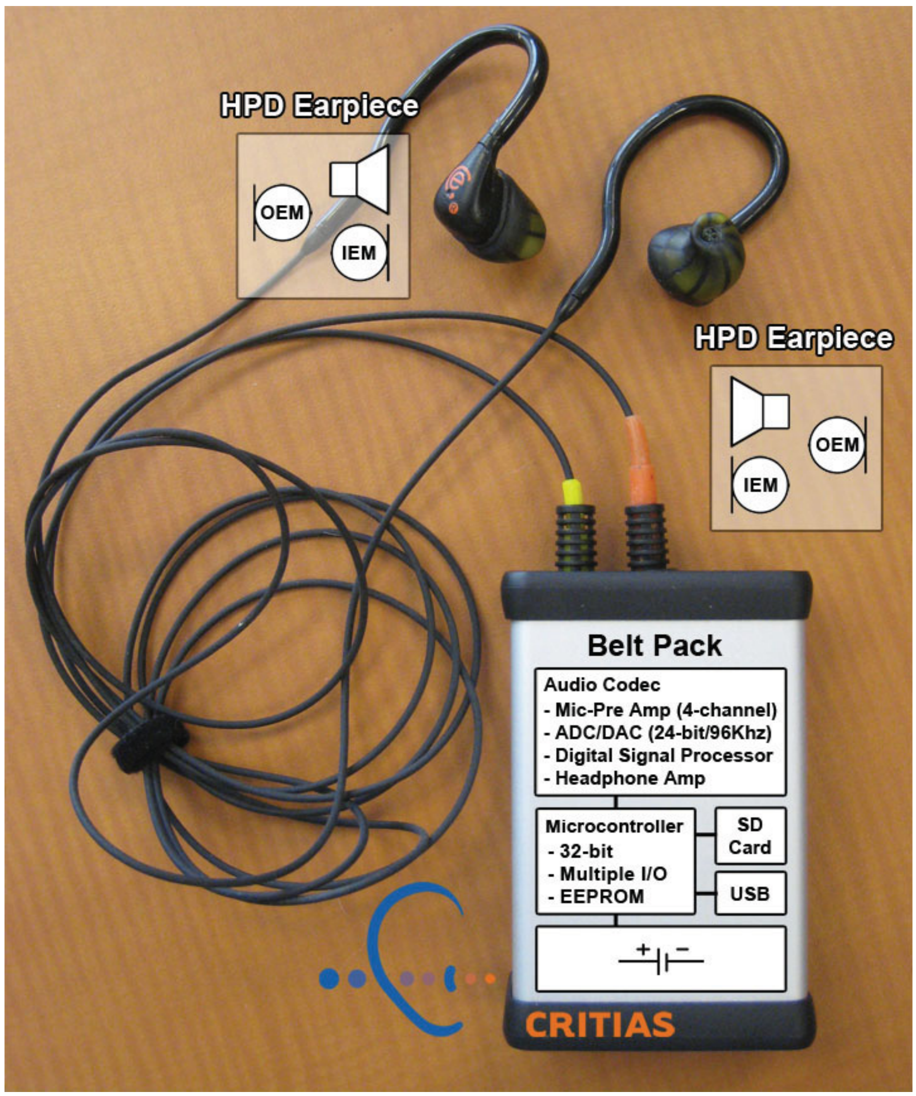

Figure-A V-1 First version of the Auditory Research Platform

2 Earpieces

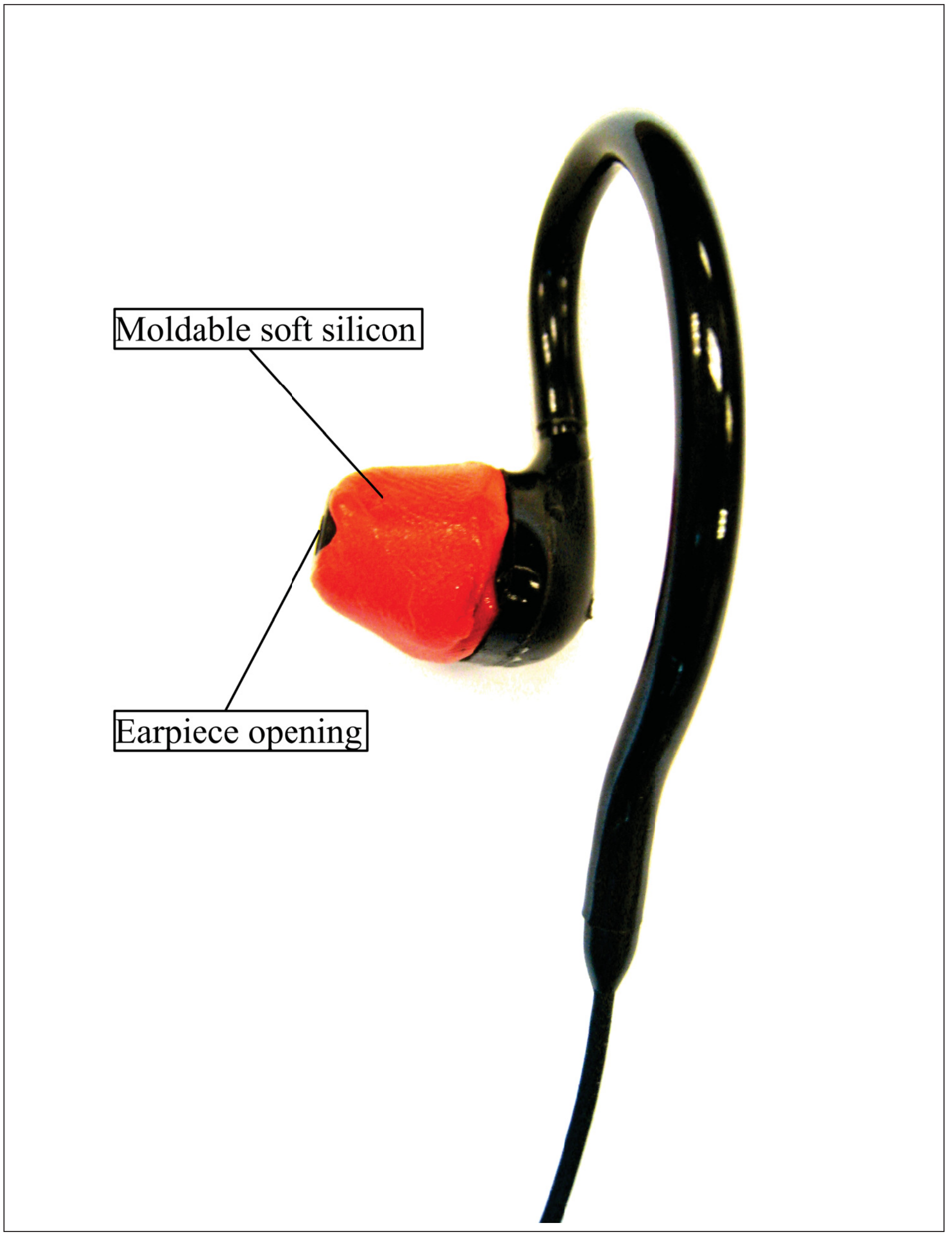

Figure-A V-2 Moldable-fit earpiece

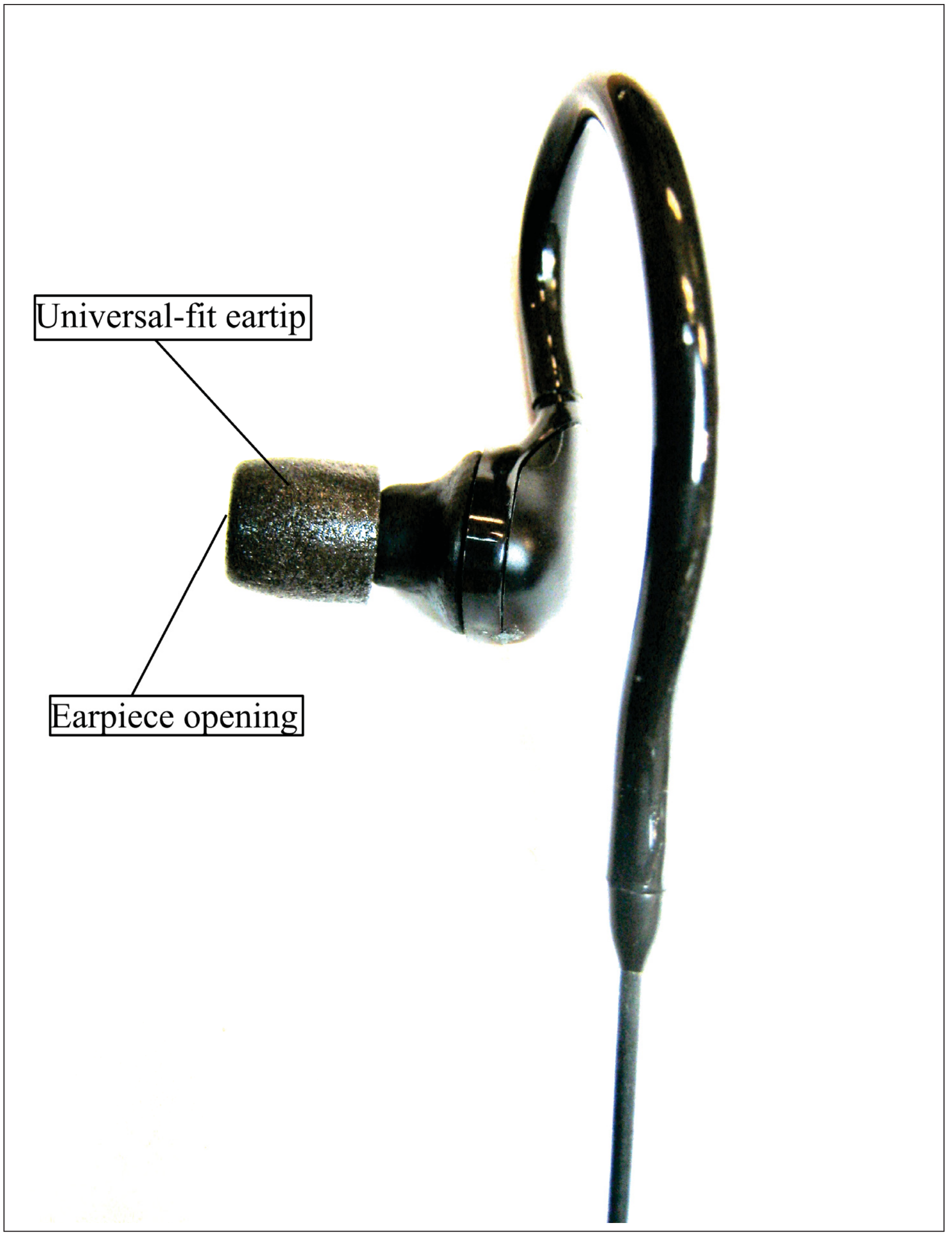

Figure-A V-3 Universal-fit earpiece

3 Active hearing protection device for musicians

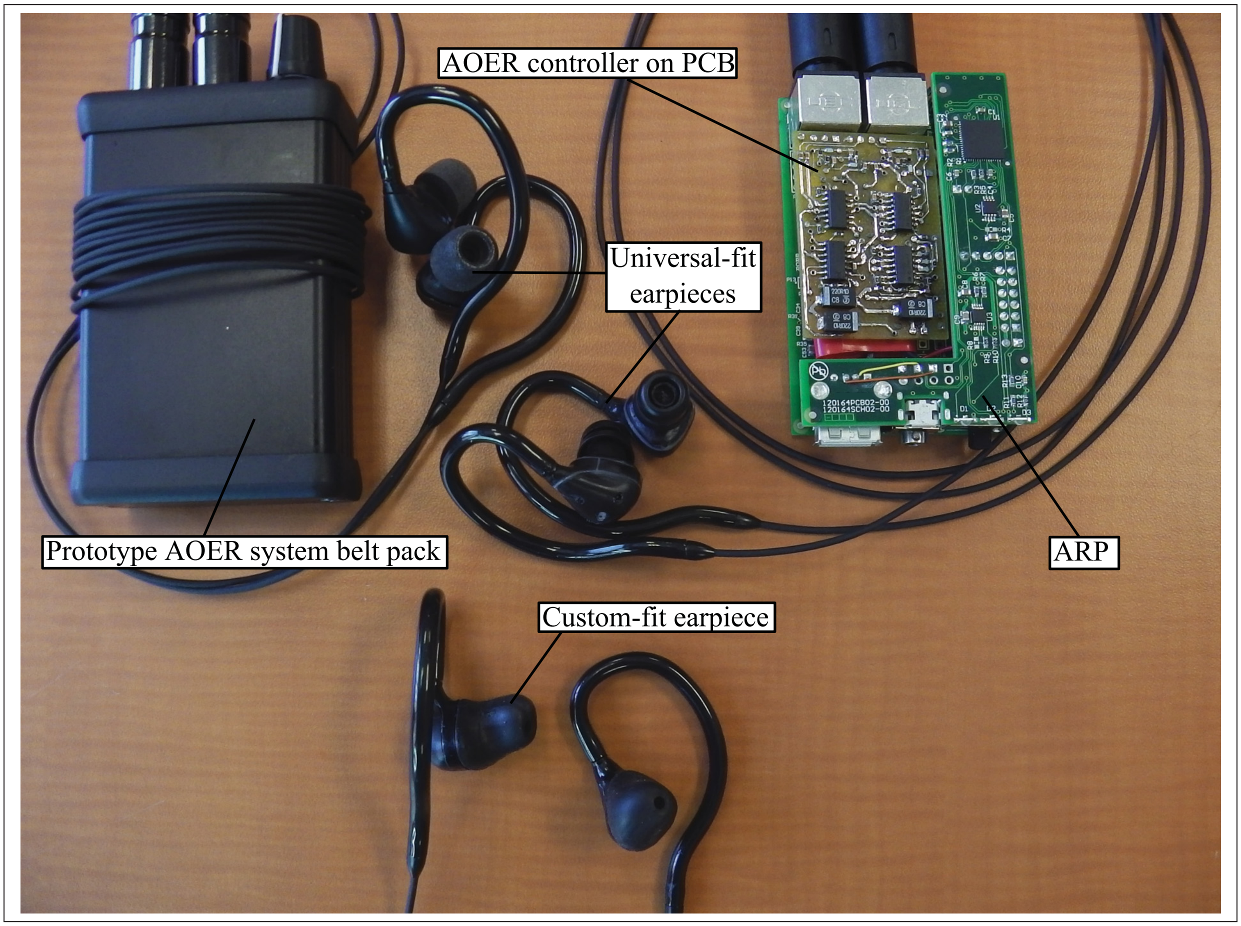

Figure-A V-4 Prototypes of an active HPD for musicians

4 Experimental setup

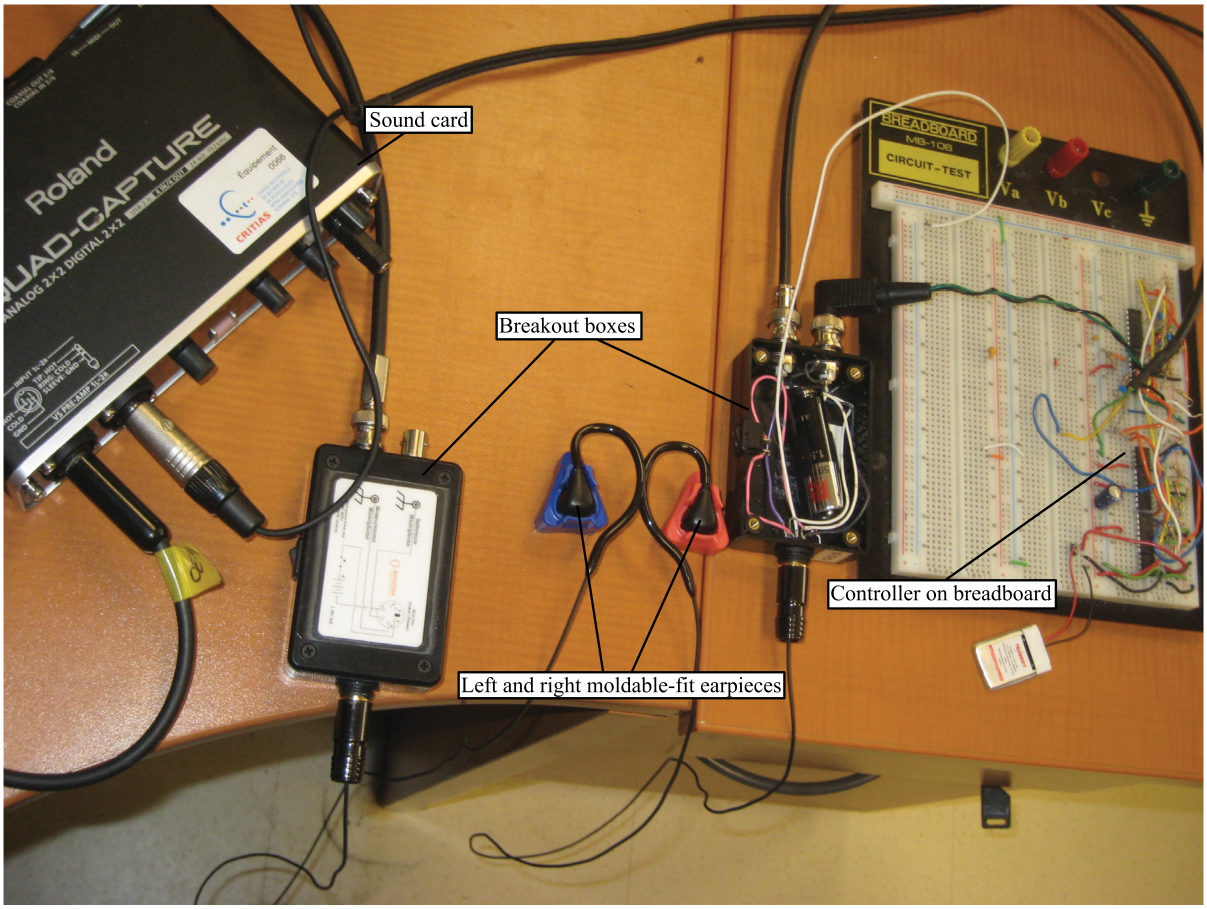

Figure-A V-5 Experimental setup for characterization of the real performance of the moldable-fit AOER system on a human subject

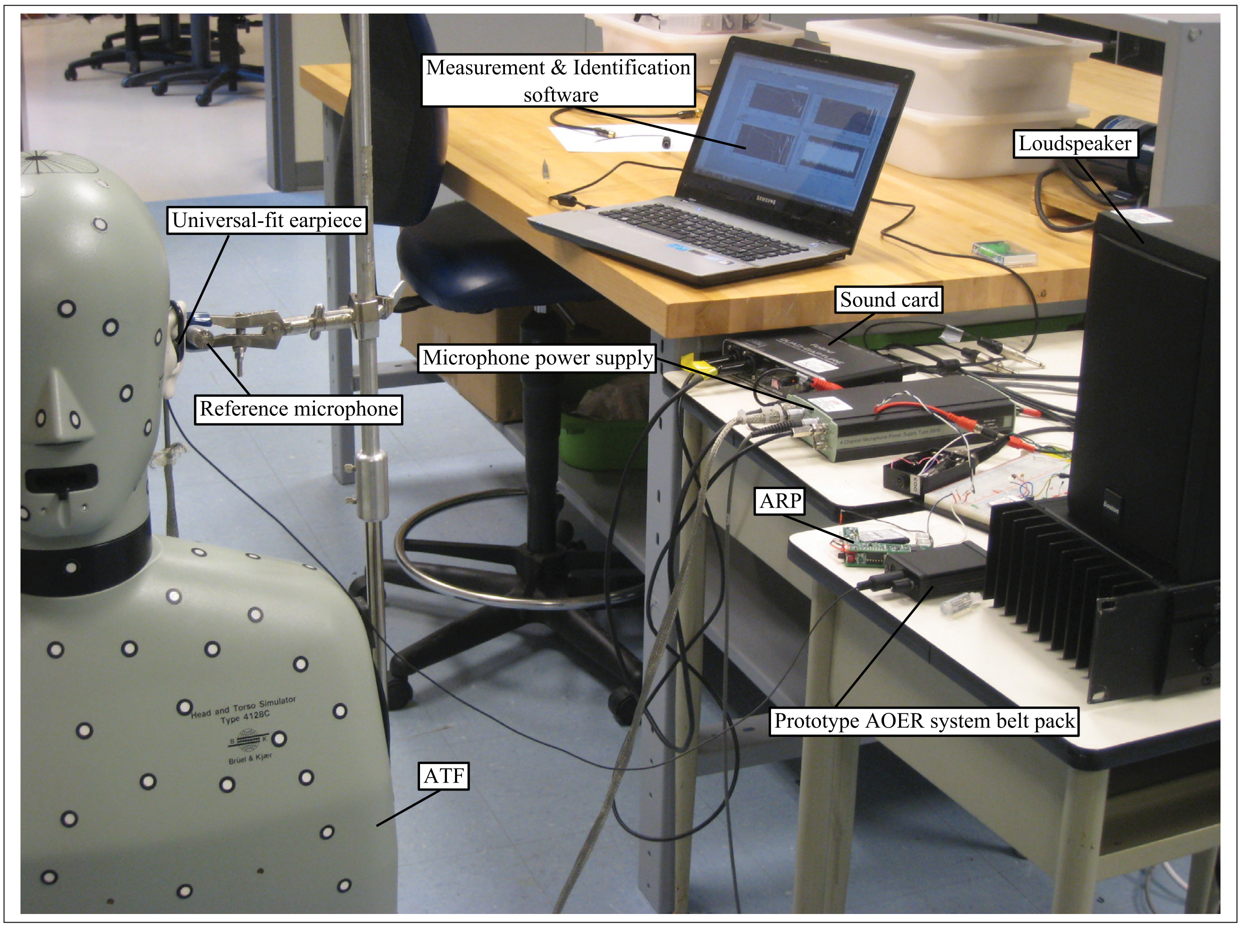

Figure-A V-6 Experimental setup for modeling of the elements of the isolation effect compensation system

APPENDIX VI

PUBLICATIONS

The following article, "An active hearing protection device for musicians" by Antoine Bernier and Jérémie Voix, was published in Proceedings of Meetings on Acoustics, 2013, vol. 19, p. 040015.

Proceedings of Meetings on Acoustics

Volume 19, 2013 http://acousticalsociety.org/

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Montreal, Canada
2 - 7 June 2013

Noise
Session 1pNSa: Advanced Hearing Protection and Methods of Measurement II

1pNSa7. An active hearing protection device for musicians

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Professional musicians have to deal with two main problems when wearing hearing protection devices (HPDs): the occlusion effect and the isolation effect. The occlusion effect is an unnatural and annoying perception of one's own voice when wearing HPDs. It will affect all musicians whose instrument induces vibrations to the skull, including singers and musicians whose instrument is pressed against any part of the head. The isolation effect is the unnatural sensation of being isolated from a given sound environment. It is caused by a non-uniform attenuation of the HPD over the audio spectrum and the absence of compensation for equal loudness contours. These two effects are highly unfavorable to the musicians' auditory perception and compromise their capacity to perform to the best of their abilities for their audience. This paper presents an active HPD for musicians, providing occlusion effect reduction and isolation effect compensation. Preliminary performance of the occlusion effect reduction system is presented and discussed.

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An Active Hearing Protection Device for Musicians

Introduction

Discomfort caused by wearing hearing protection devices (HPDs) can discourage musicians from wearing HPDs to protect their hearing from potentially dangerous noise levels. According to GMMQ (Guilde des musiciens et musiciennes du Québec), representing over 3000 professional musicians, 40% of musicians develop hearing loss during their career [1]. While physical discomfort has already been addressed by solutions such as custom fit hearing protection devices, acoustical and psychoacoustical discomfort remain a problem. Two main causes are responsible for this perceptual discomfort: the occlusion effect and the isolation effect.

The occlusion effect is an unnatural and annoying perception of one's own voice when wearing HPDs. It will affect all musicians whose instrument induces vibrations to the skull, including signers and musicians whose instrument is pressed against any part of the head, such as a trumpet or violin. The vibrations will travel to the ear canal walls by bone conduction, causing those walls to vibrate and causing pressure changes in the air contained in the ear canal, producing an acoustical wave that will be picked up by the auditory system. When the ear canal is unoccluded, most of the energy propagated through the ear canal by bone conduction exits by the ear canal's orifice, and what is heard is predominantly the sound wave arriving from the air conduction path between the source (e.g. vocal tract) and the ear. However, when the ear canal is occluded, the energy wave travelling by bone conduction is trapped within the ear canal and is picked up by the auditory system while the air conduction path is blocked, so what is heard is predominantly the sound wave travelling by bone conduction. Since bone conduction is efficient in conducting low frequencies, the result will be an augmented and unnaturally "boomy" perception of one's own voice. Fig. 1 illustrates how the occlusion effect occurs and indicates pertinent sound pressure level (SPL) measurement points, L_{OP} (SPL in the open ear canal), L_{OC} (SPL in the occluded ear canal) and L_{REF}(SPL from the air conduction path). The transfer function between L_{OC} and L_{REF} shown in Fig. 2 displays an approximation of the SPL increase in the ear canal caused by occlusion effect. Since the air conduction path prevails when the ear canal is unoccluded, L_{REF} can be used to approximate L_{OP}, and is more convenient to measure. L_{OC} and L_{REF} were measured respectively by an internal and external microphone on a prototype earpiece worn by the first author as he was humming. It is apparent from Fig. 2 that the SPL increase caused by occlusion effect occurs in the lower frequencies of the speech bandwidth. This transfer function is only an approximation of the SPL increase caused by occlusion effect, but it is generally consistent with other similar measurements made by [2] or [3].

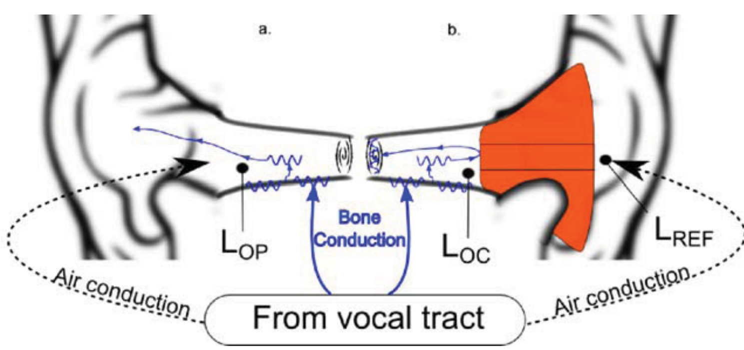

Figure 1: Occlusion effect: a) the sound wave induced by the vibration of the ear canal walls mostly escapes the ear canal and the sound wave travelling along the air conduction path is predominantly heard b) the trapped sound wave propagating from the bone conduction path will cause the eardrum to vibrate, while the air conduction path is blocked, causing an unnatural and augmented perception of one's own voice.

The isolation effect is the unnatural sensation of being isolated from a given sound environment and can be caused by wearing HPDs. Passive HPDs do not necessarily attenuate the entire audio spectrum evenly and do not take equal loudness contours into consideration. Most passive HPDs will attenuate high frequencies much more than low frequencies and, in a musical context, will considerably alter the wearer's perception of timbre. Occluding the ear canal also shifts its main resonance from quarter wavelength to half wavelength [4], further altering the wearer's perception. While passive and active solutions to these problems exist, they involve a fixed attenuation. Although this may be suitable in some situations, in others, a predetermined attenuation will provide either insufficient or excessive protection.

Furthermore, the equal loudness perception curves differ as the stimuli gets louder. Therefore, simply attenuating evenly over the whole audio spectrum does not accurately convey the spectral balance that would be perceived without wearing perfectly uniform-attenuation HPDs. Instead, some high, low midrange and low frequencies will be perceived as more attenuated than the rest of the spectrum, while some mid frequencies may seem less attenuated. Fig. 3 shows a few equal loudness curves at different loudness levels, and the perception shift that would occur if one was to wear uniform-attenuation HPDs.

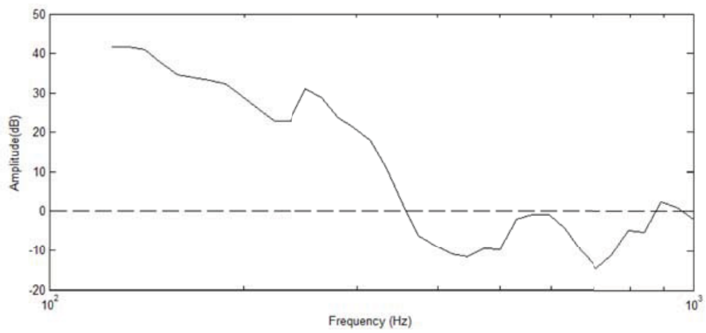

FIGURE 2: Occlusion effect: transfer function between an internal microphone, inside the occluded ear canal, and an external microphone, at the orifice of the ear canal, as the author is humming. The figure shows an estimation of the sound pressure increase caused by the occlusion effect. The sound pressure is seen to increase as high as 40 dB in the lower frequencies.

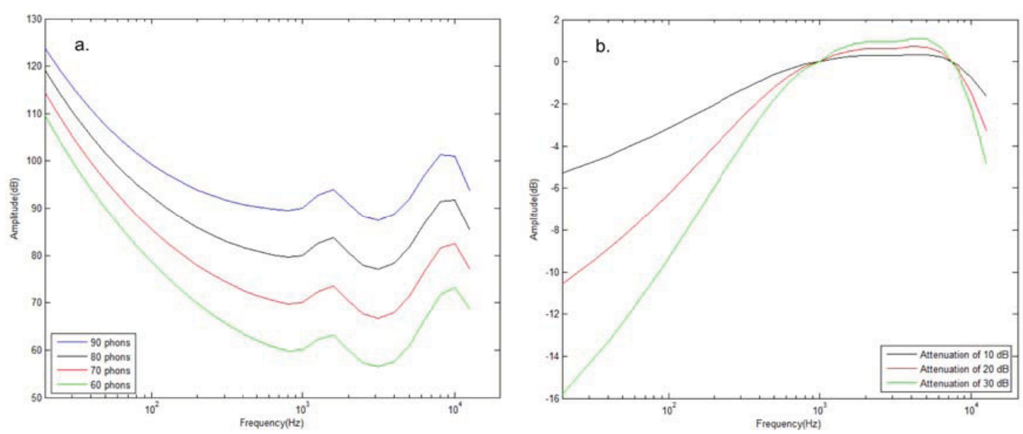

FIGURE 3: Equal loudness contours according to ISO226 [5]: a) The curves represent the required sound pressure level for a given pure tone at one frequency to be perceived as loudly as another pure tone at another frequency on the same curve. Each curve is valid at a specific loudness level, in phons, where 1 phon is set to 1 dB (SPL) at 1 kHz. b) Theoretical resulting perception shift when wearing uniform attenuation hearing protection devices in a 90 dB (SPL) sound environment for different uniform attenuation values.

The isolation and occlusion effects are highly unfavorable to the musicians' auditory perception and compromise their capacity to perform to the best of their abilities for their audience. The isolation effect can make it difficult for musicians to judge the sound quality that is being presented to their audience. When, as a consequence of the occlusion effect, an augmented and unnatural perception of one's own voice or instrument is predominantly what is heard, musicians cannot hear the subtle cues that they depend on to adjust their playing. Cues such as knowing how their timbre blends with their colleague's or how loudly their instruments sounds and resonates in a given space can make a big difference in one's performance. These adverse effects may cause some musicians to decide not to wear HPDs.

Proposed Solution

The proposed solution, addressing previously outlined issues, is a system to provide occlusion effect reduction and isolation effect compensation. The next section presents the occlusion effect reduction system.

Addressing the Occlusion Effect

The occlusion effect reduction system is based on active noise control (ANC) of the low frequency sound wave which becomes predominant in an occluded ear canal. A carefully selected miniature loudspeaker and microphone assembly (referred to as *plant*) is placed in the ear canal, within the HPD. A feedback controller uses the error signal picked up by the internal microphone to generate a corresponding anti-noise with the loudspeaker. The anti-noise adds up to the noise, in the acoustic domain, and reduces the occlusion effect, as shown in Fig. 4. Fig. 5 shows the architecture of the ANC system.

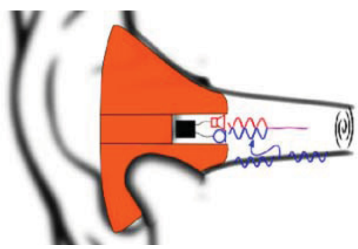

FIGURE 4: Active noise control of occlusion effect: noise in the ear canal is picked up by an internal microphone, and a cancellation signal is generated with the loudspeaker.

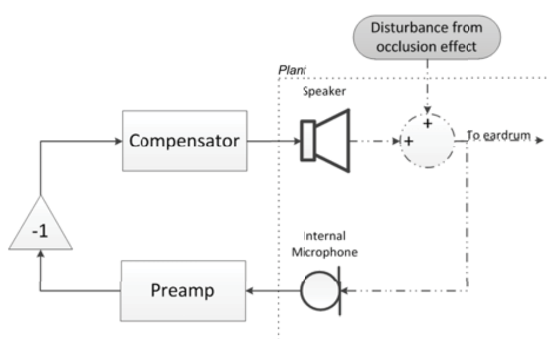

FIGURE 5: Architecture of the occlusion effect ANC system and the *plant* (miniature loudspeaker and microphone assembly in the occluded ear canal).

An analog feedback controller design was selected for its simplicity and stability. The compensated plant response was obtained from the initial plant response, in this case the loudspeaker to microphone transfer function when inserted in the ear canal, as shown in Fig. 6. The theoretical occlusion effect attenuation is shown in Fig. 7, predicting over 10 dB of attenuation from 100 Hz to 400 Hz, where the occlusion effect has been shown to be most significant. Slight regeneration in mid to high frequencies is also observed as a side effect of the feedback control. Regeneration occurs at the frequencies where the plant response's projection on the real axis is negative. Since the acoustical wave resulting from the occlusion effect does not contain much of these frequency components, this regeneration should not prove to be problematic.

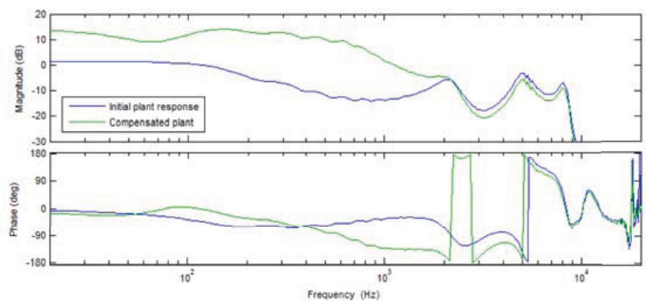

FIGURE 6: Transfer functions of the plant and compensated plant

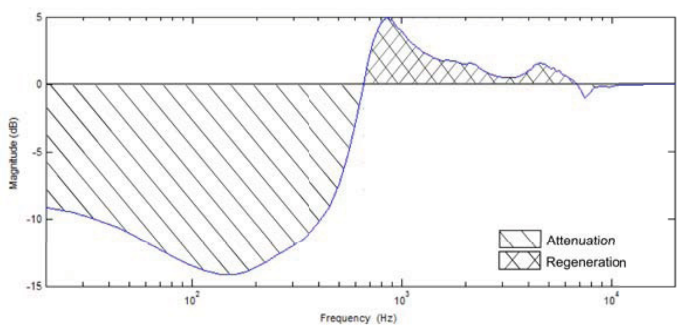

FIGURE 7: Projected occlusion effect attenuation: Attenuation is achieved over the bandwidth of interest, but unavoidable regeneration also occurs.

Addressing the Isolation Effect

This section describes the methodology addressing the isolation effect. An external microphone placed on the outside of the HPD could be used to capture the useful signal, transform and reproduce it at variable volume through the internal miniature loudspeaker. A passive HPD will usually attenuate sound unevenly, letting through more low frequencies than high frequencies. To flatten the attenuation, it is possible to reduce low frequencies by active noise control and amplify high frequencies to match the attenuation level of mid frequencies, as shown in Fig. 8. This procedure is very similar to the one described in [6].

Since ANC of the occlusion effect already cancels some of the low frequencies inside the ear canal, a simple filter can be used to flatten the attenuation and achieve maximum attenuation within the constraints imposed by the performances of the ANC and the passive attenuation of the HPD. This filter includes ear resonance correction for a uniform perceived attenuation. To further tweak the uniform attenuation and compensate for equal loudness contours, a set of filters can be chosen from. Using the outside noise level and the desired attenuation as input parameters, it is possible to account for the shift of perception from one equal loudness curve to another, previously shown in Fig. 3. By doing so, the perceived spectral balance is the same with or without the hearing protection.

The digital signal processor (DSP) housing the filters can measure the sound pressure level outside the HPD. It can then either calculate and apply the required attenuation to follow a certain standard, or apply a user-defined attenuation level. The internal microphone is used to verify that the attenuation is indeed correct. The complete system architecture required to implement both the occlusion effect reduction and isolation effect compensation system is shown in Fig. 9.

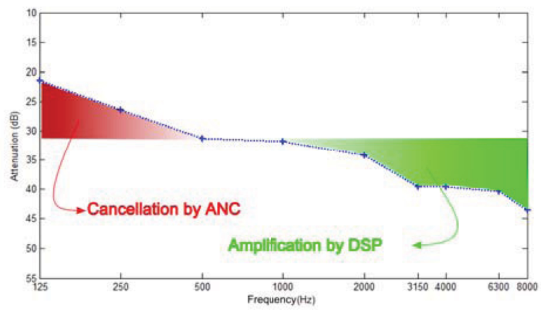

FIGURE 8: Example of achieving uniform attenuation: Since passive attenuation is usually the lowest in the low frequency region, using active noise control in that region will flatten the attenuation. A DSP could then be used to amplify high frequencies and achieve uniform attenuation in that region as well.

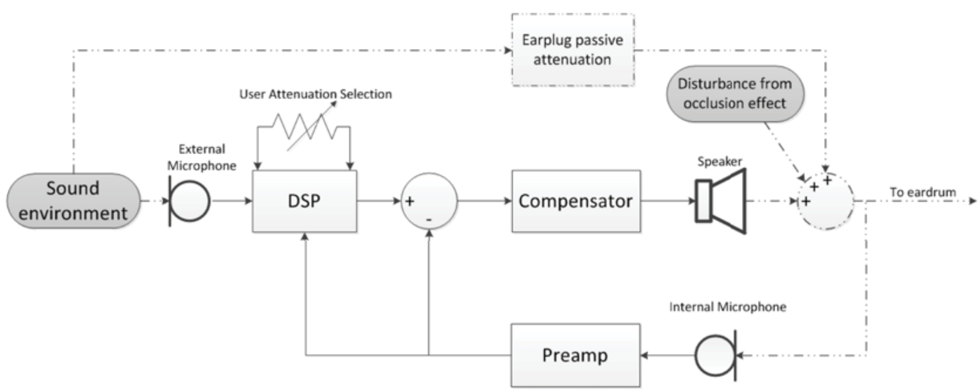

FIGURE 9: Complete system architecture

Preliminary Performance Assessment

At the time of writing, only preliminary performance for occlusion effect reduction had been quantified. First, a correction curve was obtained by characterizing the differences in occlusion effect level when humming, between the first author's left and right ears, occluded by identical earpieces. Then, the occlusion effect ANC system was activated in the first author's right ear, and a transfer function between the reference earpiece (left) and the active earpiece (right) was calculated and corrected using the previously obtained correction curve. Fig. 10 shows the performance of the implemented occlusion effect ANC system in reducing the author's occlusion effect as he hums. A decrease of over 10 dB can be observed from about 100 Hz to about 500 Hz. While this might seem to be a small difference, the perceptual difference is substantial. To better assess the perceptual reduction of the occlusion effect, psychoacoustic tests need to be conducted on a larger number of human subjects.

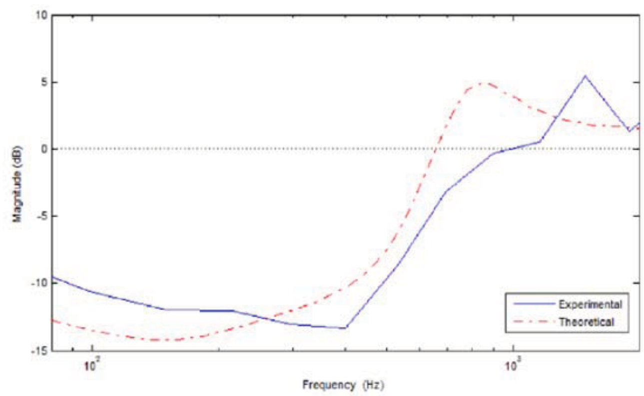

FIGURE 10: Occlusion effect cancellation: the dashed line shows the expected occlusion effect reduction from theory and design. The solid line shows the achieved occlusion effect cancellation for the first author.

A second test was conducted to assess the performance of the ANC system on the attenuation of the plug. First, white noise was played using over-the-ear headphones on a Bruel & Kjær head and torso simulator (HATS) model 4157 as the mannequin was wearing the earplug in passive mode. Second, the procedure was repeated with the earpiece in active mode. The transfer function between the two signals recorded using the microphone located at the eardrum of the HATS is shown in Fig. 11. Although they may seem similar, the measured curve using this method differs from the other experimental attenuation curve from Fig. 10. It is possible that a better fit was achieved on the mannequin than in the author's ears, thus shifting the response of the plant. Soft silicone does not guarantee the same fit with every use, and since it was used to couple the prototype earpiece to the ear canal, the resulting performance can vary. Scheduled test performed on more subjects will provide more information. Nevertheless, the preliminary experimental performance of the active occlusion effect reduction system are comparable to the results presented in [7].

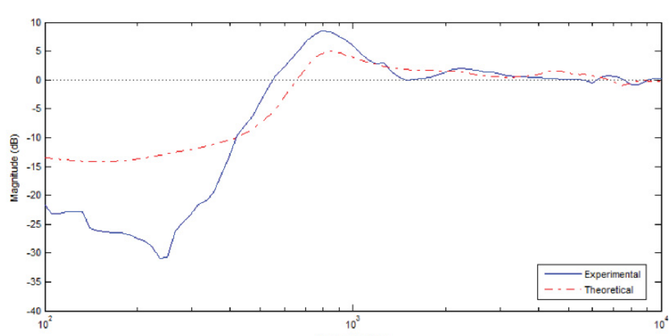

FIGURE 11: Active attenuation: the dashed line shows the expected active attenuation from theory and design. The solid line shows the measured active attenuation. Differences between the curves could be attributable to the variability of the acoustic seal which could in fact influence the *plant* response and therefore the performance of the system.

Having characterized active attenuation, it is possible to present a more realistic overall attenuation scenario. The measured passive attenuation curve of an earplug that would be a good candidate to host the prototype earpiece presented in this paper is shown in Fig. 12 . Projecting the previously presented experimental active attenuation curve on the passive attenuation curve shows the increase in effectiveness of the earplug in the areas where it is the least effective. Given the resulting curve, it is possible to amplify the over-attenuated frequencies to match the attenuation of the least attenuated frequencies, using a DSP. By doing so, higher maximum uniform attenuation can be achieved than what was possible without the active system. Another advantage is the adjustability of the attenuation: using the DSP, the level of uniform attenuation can be adjusted by the user depending on his or her needs. Thus, in the case presented in Fig. 12, uniform attenuation values could range from about 19 dB to any defined lower attenuation bound, such as 6 dB, or even complete bypass of the HPD.

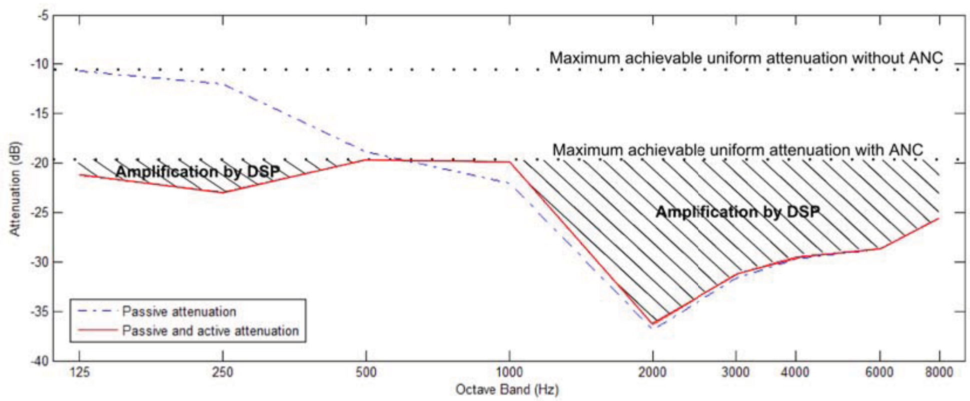

FIGURE 12: Example of achieving uniform attenuation: the dotted line shows the passive attenuation, measured on a human subject, of a potential candidate earplug that could host the prototype. The solid line shows that the experimental active attenuation from Fig. 11 added to the passive attenuation would enable the increase of the maximum achievable uniform attenuation of the hearing protector. Starting from the solid line curve, frequencies lacking intensity could be amplified at the user's ear, using the external microphone, the DSP, and the internal speaker. The maximum uniform attenuation curves achievable by using this method are shown in dotted lines. The figure shows how the active solution would help increase the maximum achievable uniform attenuation.

Conclusions

Musicians still face drawbacks when trying to protect themselves from overexposure to sound. The occlusion effect and isolation effect can discourage musicians from wearing HPDs. Both effects could jeopardize musicians' performance, by altering the auditory perception on which they rely. In this paper, solutions to both effects have been presented, as well as a complete system architecture capable of accounting for these effects. An occlusion effect reduction system was designed, implemented, and preliminary characterization of the performance has been achieved. Preliminary performances are promising, and further validation on a greater number of human subjects is to be done.

ACKNOWLEDGMENTS

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The following article, "Une nouvelle protection auditive adaptée aux musiciens" by Antoine Bernier and Jérémie Voix, is in press and has been accepted by Revue FAMEQ: Musique Pédagogie, 2013.

Une nouvelle protection auditive adaptée aux musiciens

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I. Introduction

L'inconfort perceptuel entraîné par le port d'une protection auditive limite son acceptation par les musiciens et pousse ces derniers à ne pas en faire un usage rigoureux et adéquat. Plusieurs musiciens affirment que les protections auditives entravent leur performance en modifiant leur perception auditive. Le non-port d'une telle protection peut cependant conduire à des pertes auditives qui peuvent être sévères et permanentes, et possiblement compromettre la carrière du musicien. L'inconfort associé aux protections auditives découle principalement de deux effets indésirés : l'effet d'occlusion et l'effet d'isolement.

L'effet d'occlusion est la perception augmentée et dénaturée de sa propre voix lorsque son conduit auditif est occlus, par rapport à la perception naturelle lorsque son conduit auditif est ouvert (**Figure 1**). La raison principale de ce changement de perception est que la vibration des cordes vocales se propage par les os et tissus de la tête et excite les parois du conduit auditif, faisant ultimement vibrer le tympan de façon efficace lorsque le conduit auditif est occlus (**Figure 2**). Le phénomène se manifeste également lorsqu'un instrument à vent est utilisé, puisque celui-ci fait également vibrer les os de la boîte crânienne.

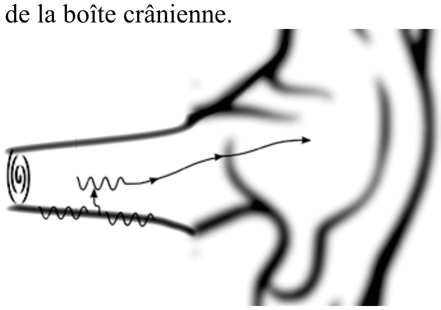

Figure 1: *Les vibrations transmises par les os font vibrer la paroi du canal auditif et l'air contenu dans le canal. La pression sonore créée tend à se dissiper par le chemin le moins résistif en sortant du conduit auditif, puisque celui-ci est ouvert.*

Ce phénomène est indésirable dans plusieurs cas car la perception de sa propre voix à un niveau démesuré peut masquer partiellement ou complètement les sons extérieurs, ce qui est inacceptable pour un musicien qui doit entendre ses collègues lors d'une prestation musicale. De plus, cette perception dénaturée rend le musicien mauvais juge de sa propre prestation. Ultimement, cet effet d'occlusion décourage fortement le port de protection auditive, entraînant des risques pour l'audition à long terme du musicien.

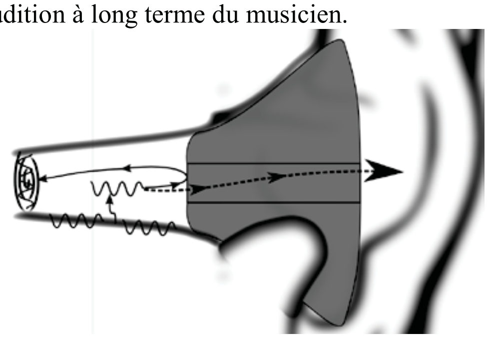

Figure 2: *Le conduit auditif est occlus et le chemin normalement moins résistif est bouché, ce qui fait que la pression sonore régénérée par les vibrations du conduit fait vibrer le tympan, augmentant la perception de sa propre voix. L'effet résultant donne l'impression d'une voix forte dont le timbre est altéré.*

L'effet d'isolement est la sensation souvent désagréable du porteur de bouchons d'être isolé de son environnement sonore et comprend plusieurs facteurs :

Premièrement, l'insertion d'un bouchon annule la résonance naturelle du conduit auditif, qui n'agit plus comme un tuyau résonant ouvert. Notre oreille, habituée à la hausse de niveau créée par cette résonance, ressentira une baisse de niveau aux fréquences concernées.

Deuxièmement, les protecteurs auditifs sont généralement moins efficaces en basses fréquences. Le porteur se retrouve donc affligé d'une différence majeure de perception de l'équilibre spectral (**Figure 3**).

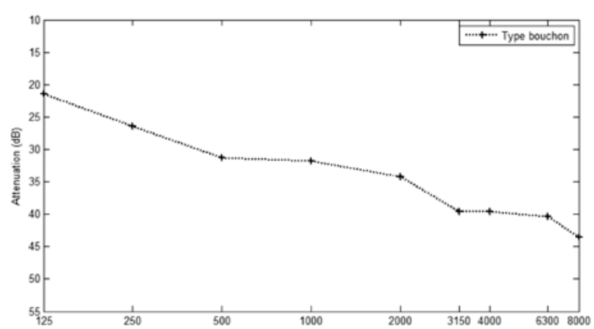

Figure 3: *Courbe d'atténuation typique d'un protecteur de type bouchon, montrant une atténuation moins élevée dans les basses fréquences que dans les hautes fréquences.*

Troisièmement, la sensibilité fréquentielle de l'oreille est non-linéaire: le contenu fréquentiel du même son entendu à différents niveaux sera perçu différemment par notre oreille, comme le témoignent les courbes isosonique (**Figure 4**). Ces courbes impliquent, par exemple, que la basse ressortira plus de l'ensemble si on écoute une chanson à fort niveau qu'à faible niveau. Par conséquent, même un musicien qui porterait des protections auditives compensant pour les facteurs évoqués précédemment n'entendra pas exactement le même équilibre spectral que le spectateur. Cette différence de perception peut être dérangeante lorsque le musicien veut vérifier que sa musique est bien équilibrée aux oreilles du spectateur.

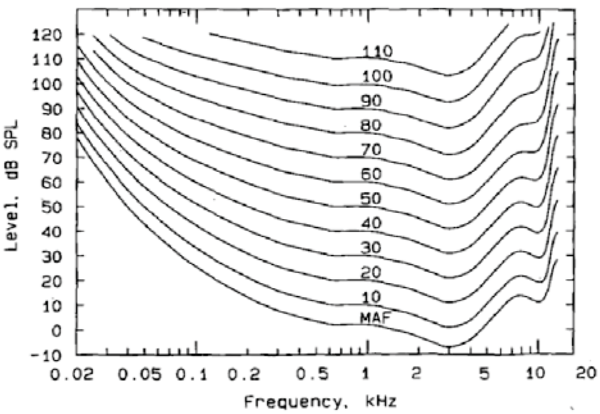

Figure 4 : *Représentation de la non-linéarité de la perception humaine. Les courbes montrent la correspondance entre les niveaux absolus et l'intensité sonore ressentie, selon la fréquence. Les différences de formes entre les courbes témoignent de perceptions différentes dépendamment du niveau sonore. Tiré de [1].*

II. SOLUTION PROPOSÉE

Le projet de maîtrise du premier auteur consiste à développer un prototype de protecteur auditif actif pour musiciens palliant simultanément à l'effet d'occlusion et à l'effet d'isolement. L'effet d'occlusion est annulé par une technique de contrôle actif du bruit, dont le principe est l'annulation d'un son par l'addition d'un son en opposition de phase. Un microphone à l'intérieur du conduit occlus sert de microphone de référence tandis qu'un haut-parleur fournit l'onde sonore d'annulation.

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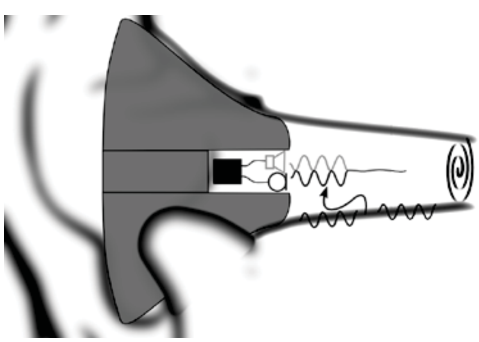

Figure 5 : *Grâce à l'algorithme d'annulation de l'effet d'occlusion, l'onde qui en est responsable (noir) est captée par le micro interne et reproduite à l'inverse (gris pâle) en temps réel par le haut-parleur. La superposition des deux ondes tend vers le silence (gris).*

L'effet d'isolement est pallié par des techniques de traitement du signal transmis sous le protecteur. Un second microphone, à l'extérieur du protecteur auditif, permet de capter l'environnement sonore du porteur. Le haut-parleur interne est utilisé pour reproduire, à niveau moindre, l'environnement sonore dans le canal auditif, en s'assurant que l'équilibre spectral du son reproduit soit le plus près possible de ce qui aurait été entendu sans protecteur auditif (**Figure 6**). Ainsi, si on considère que typiquement, un protecteur auditif atténuera plus les hautes que les basses, le haut-parleur compensera en amplifiant des hautes, et en annulant des basses (**Figure 7**). Le traitement de signal implémenté dans un DSP permet de compenser à la fois pour la disparité fréquentielle de l'atténuation acoustique du bouchon et la résonance perdue de l'oreille en faisant les corrections nécessaires.

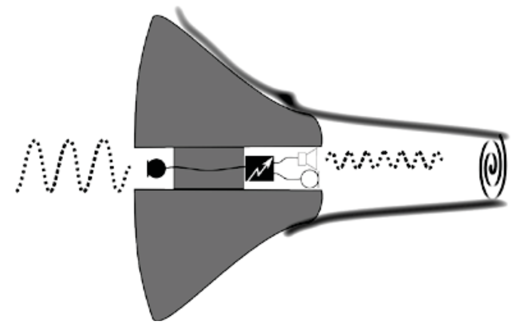

Figure 6 : *Le son extérieur est capté et reproduit par le haut-parleur interne avec les corrections nécessaires, à un niveau variable.*

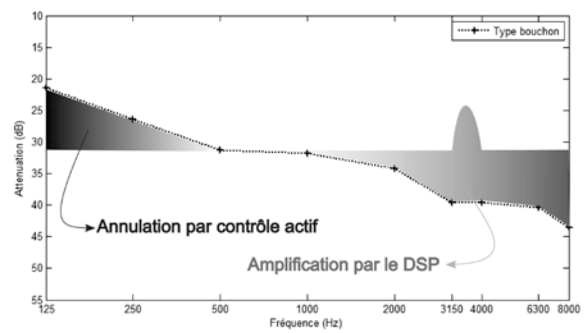

Figure 7 : *Courbe typique d'atténuation fréquentielle d'un protecteur auditif de type bouchon, montrant la correction nécessaire requise pour uniformiser cette atténuation, avec représentation de la compensation pour la résonance perdue de l'oreille.*

Pour pallier à la non-linéarité de l'oreille, qui contribue potentiellement à l'effet d'isolement, une compensation du déséquilibre fréquentiel est implémentée dans le DSP afin que les signaux transmis au tympan, à un niveau moindre, soient perçus avec le même équilibre fréquentiel que s'ils avaient été reçus à des niveaux plus élevés. De cette façon, le musicien entendra le même équilibre sonore que le spectateur et pourra protéger son audition tout en étant bon juge de la qualité sonore de la prestation.

La coexistence des algorithmes est possible car le microphone à l'extérieur du bouchon permet d'identifier le signal utile, tandis que le micro à l'intérieur capte les fuites acoustiques ainsi que l'effet d'occlusion. Les deux microphones sont donc utilisés pour différencier les sons extérieurs, qui doivent être relayés à l'oreille à un niveau sécuritaire, et l'effet d'occlusion et les fuites acoustiques, qui doivent être atténués (**Figure 8**).

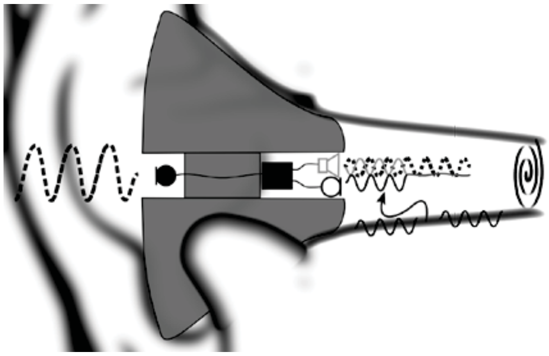

Figure 8 : *Système de bouchon actif pour musicien proposé, montrant l'effet des deux algorithmes superposés fonctionnant en simultané.*

III. CONCLUSION

Le système de bouchon actif pour musicien proposé présente une solution pour les problèmes causés par les protections auditives qui sont les plus dérangeants. Un premier prototype fonctionnel a été réalisé dans le cadre du projet de maîtrise du premier auteur et a permis d'obtenir des résultats prometteurs, quoique limités à un contexte de laboratoire. La prochaine étape est donc la réalisation d'un prototype suffisamment petit et autonome pour être utilisé pendant quelques heures dans une situation réelle. Les défis techniques d'une telle implémentation sont nombreux. Cependant, les bienfaits d'un tel système pourraient potentiellement redéfinir la perception des protections auditives, vues actuellement comme un mal nécessaire.

Finalement, en parallèle avec les efforts qui mèneront à un deuxième prototype, il est nécessaire de concocter un protocole de validation auprès des musiciens. Il est en effet primordial d'obtenir un retour de la part de cette communauté afin de savoir si l'expérience d'un tel prototype comprenant de telles corrections semble bel et bien plus naturelle, par rapport aux bouchons conventionnels ou spécialisés, pour ainsi offrir une solution réelle aux problèmes qu'ils rencontrent.

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