Feedback cancellation in Hearing Aids

A Solution with Adaptive Filtering

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Abstract

The application of adaptive filters in feedback cancellation in vented and unvented hearing aids (HA), together with their feedback path properties when worn by a head and torso simulator, have been analyzed and studied in this project.
A vented and an unvented *in the ear* (ITE) HA were mounted in a head and torso simulator (HATS) and the feedback paths were measured with different objects placed near the HAs.
Adaptive feedback cancellation (AFC) algorithms based on the least mean square(LMS) algorithm have been simulated with white noise and speech input signals. Due to the short distance between the microphone and receiver in HAs, the feedback and input signal can be correlated in the case of a highly autocorrelated input signal such as speech. The simulations performed show that this correlation does not make the estimation of the feedback path accurate enough, and hence the feedback cancellation not effective. Decorrelation methods were studied and the delay in the forward path option was simulated. The simulation results show that this is an effective solution to decorrelate the input and feedback signals and hence achieve a better estimation. When a white noise input signal was used, an added stable gain(ASG) of 20 dB was obtained in the simulations.
The LMS approach was implemented in a digital signal processor as an adaptive feedback canceller. For a white noise input signal the ASG of the AFC was found to be 12dB.

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Synopsis

Der er i dette projekt opstillet et adaptivt feedback cancellation (AFC) system, der adaptivt tilpasser sig den akustiske tilbagekobling for et høreapparat.
Anvendte applikationer, såvel som den akustiske tilbagekobling er blevet analyseret.
I forbindelse med målinger af den akustiske tilbagekobling er et høreapparat monteret i en head and torso simulator (HATS), hvor der endvidere, på skift, er placeret en række forskellige objekter i nærheden af høreapparatet, for således at måle ændringerne i den akustiske tilbagekobling.
Et adaptiv feedback cancellation algoritme, baseret på Least Mean Square (LMS), er designet og simuleret i Matlab.
På baggrund af den korte afstand mellem mikrofon og højtaler, i høreapparatet, vil der være korrelation mellem det akustiske tilbagekoblet signal fra højtaleren og inputtet til mikrofonen.
Metoder til at udføre dekorrelation er analyseret og en simpel metode i simuleringerne er valgt. Ved at introducere en tidsforsinkelse i forward stien for HA, opnås en bedre estimering af den akustiske tilbagekobling.
Der er i simuleringerne opnået en forbedring i forstærkning på 20 dB, for hvid støj input.
En AFC, baseret på LMS, er implementeret på DSP.
Der er opnået en 12 dB forbedring i forstærkning.
En tidsforsinkelse af signalet er ikke implementeret.

Preface

This report has been made by group 862 at the Department of Acoustics. The report concerns feedback cancellation in vented hearing aids - a solution with adaptive filtering. The target audience is students at the Department of Acoustics and others with interest in feedback cancellation, hearing aids and adaptive filters.

Simulations have been performed in Matlab and the solution is implemented in assembly language on a TMS320C50 digital signal processor.

Figures and illustrations are referred to with a number. (e.g. figure 3.1, meaning the first figure in chapter 3)

References to literature is marked with a number (e.g. [10]) Footnotes are marked with a raised *number*[1].

Special thanks to our contact at Oticon, Jesper Krogh Christensen, for making measurement on a real hearing aid a reality by supplying two hearing aids.

Aalborg University, 31. May 2006:

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Contents

Chapter 1

Introduction

For people with hearing impairment q hearing aids (HA) are an option to recover some of their hearing capability. A typical hearing aid consists basically on a microphone that collects the incoming sound, an amplifying module and a loudspeaker (usually called receiver). The purpose is to reproduce the original sound, reinforced it the gaining the signal. This is the basic scheme in the two common kinds of hearing aids , the *in the ear* (ITE) and *behind the ear* (BTE) devices (see appendix C). The specific hearing aid appropriate for one subject or the gain needed can be individual regarding to the disorder, but one common problem every current hearing aid user may experience is the matter concerning feedback.

Feedback finds expression in a howling tone, which is annoying to the users of the hearing aids. Feedback is due to the short distance between the microphone and the receiver in the hearing aid device that allows the already highly amplified sound from receiver to reach the microphone again, establishing a closed-loop (see appendix B).

To find the origin of feedback in real current hearing aids it is necessary to explain first another well known effect caused by the use of hearing aids: *the occlusion effect*. An occlusion effect occurs when some object completely fills the outer portion of the ear canal (e.g., try to block your ears with the tip of your fingers). What this does is trap the bone-conducted sound vibrations of a person's own voice in the space between the tip of the object and the eardrum. Normally, when people talk (or chew) these vibrations escape through the open ear canal and the person is unaware of their existence. But when the ear canal is blocked, the vibrations are reflected back toward the eardrum and increase the loudness perception of their own voice. It is an annoying effect that may leads to the discarding of the hearing aids.

Some ITE hearing aids are trying to avoid this occlusion effect by introducing a small vent. However, the vent encourages the feedback to appear due to the new low impedance path from the receiver to the microphone and will be the dominant component of the acoustic feedback path (see figure 1.1). An amplification of the sound escaping through the vent is produced and this amplified sound will be sent back to the receiver again. If the amplified sound arrives in phase with the incoming sound, and if the gain of the loop is greater than one, feedback will occur.

This is one of the main challenges for hearing aids designers: how to avoid feedback in vented hearing aids in an efficient way.

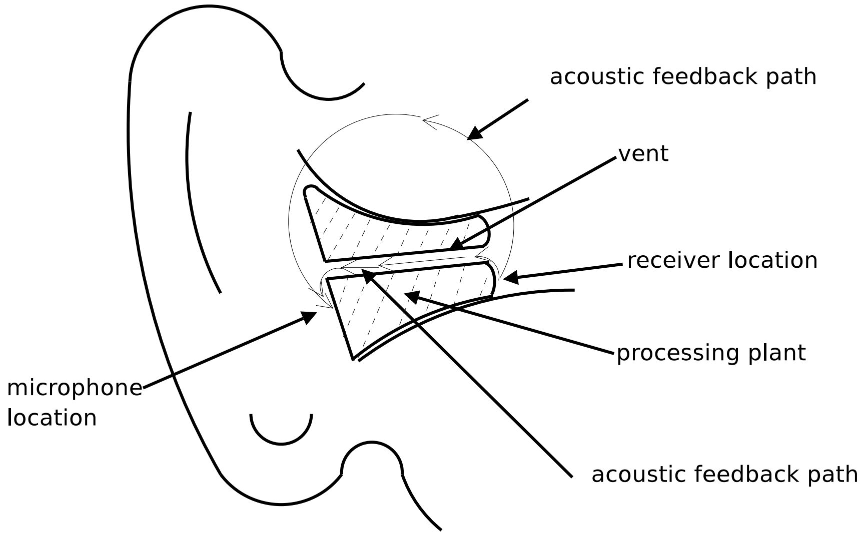

Figure 1.1: Schematic diagram of a ITE hearing aid in the ear canal. The feedback-path is due to leakages and the vent is included to prevent the occlusion effect.

1.1 Scope of the project

The problem explained has been already studied by many different researchers, whom have proposed many solutions. Basically there are two kinds of approaches: the feedback reduction by properly designing the shell, microphone and receiver; and feedback reduction by digital signal processing. One of the feedback reduction technique is the feedback-path cancellation method[13]. Common feedback reduction methods are explained in section 2.1.2, but this project work has made use of the feedback path cancellation approach, specifically, using adaptive filters. The reason for that selection is that feedback characteristics change along variations of the feedback path, and the feedback path is influenced by the changes in acoustical environment where the hearing aid works. Then, it is preferable to have a system that can cancel the feedback dynamically and track the variations of the feedback path.

Therefore, an adaptive system to cancel out the feedback will be suitable. The system tries to adapt to the acoustic feedback path and therefore eliminating the sound fed back through leakages. However, this solution has its own set of problems due to the particularity of the signals present in the hearing aid. Consequently different techniques have been studied, but the project problem statement can generally be expressed as:

How to develop a feedback cancellation system in vented hearing aids, cancelling the acoustics feedback path, using adaptive filtering.

1.2 Limitations of the project

There are two limitations in this project:

- First, it was not possible to get real commercial vented hearing aids. Instead, two ITE unvented hearing aid shells were kindly provided by Oticon. The hearing aids shells were not complete commercial hearing aids. They were consisted of a small amplifier circuit and a shell with receiver and microphone. As the shells were unvented, it was decided to drill two holes in one of them in order to encourage feedback. This perforated hearing aid will be referred as a *vented* hearing aid.

- Second, the measurements were performed in a manikin and a manikin represents an average user, real users are different from one to another. This variation causes performance differences of the hearing aid, especially in terms of feedback, when the hearing aid is worn by different users.

1.3 Report structure

The report contains the necessary information about the system developed in both theoretical and practical matters. This system is a hearing aid feedback canceller based on adaptive filter. The implementation follows the continuous adaptation model. To explain all the steps involved in the analysis, design and implementation of the system, the report is presented in the following structure:

- Chapter 1, *Introduction*: Introduction to the topic of feedback in hearing aids and its possible solutions as well as the problem statement are presented. At the end a short description of the report content and organization is provided.

- Chapter 2, *Analysis*: This chapter is divided in two main parts. The first part describes of the phenomenon of feedback in hearing aids, in which an explanation about its origins and techniques for its prevention is included. The second part provides the adaptive filter theory necessary to support the processing involved in this project, from the different possibilities available, to the one that was finally implemented.

- Chapter 3, *Design*: In order to make a good design of the adaptive feedback canceller (AFC), software simulations using MATLAB were performed. To improve the accuracy of the simulations some empirical data regarding the feedback path of a real system was needed. So first, a working system was implemented based on a hearing aid shell. Then, measurements of the hearing aid feedback path were made. Finally, the simulations of the AFC based on the feedback path measured were carried out.

- Chapter 4, *Implementation*: Because of the short latency required for the system, a digital signal processor(DSP) was chosen to implement the AFC. In this chapter considerations made during the implementation of the algorithms in the DSP are explained.

- Chapter 5, *Results*: Three sets of measurements were made. The first set was done to determine the added stable gain when the AFC was running in the DSP. The second set was done to obtain the output sound pressure levels in an ear simulator mounted on a HATS with a 90dB input level ($OSPL_{90}$). The third set was for the gain of the hearing aid system with and without the AFC running when the input signal is white noise.

- Chapter 6, *Discussion and conclusions*: Different considerations regarding feedback cancellation in hearing aids are discussed here. The methods of solution analyzed throughout the report are brought into context in terms of hearing aid users, manufacturers and the actual work of the group as the project was being carried out. Limitations and validity of the presented work are discussed as well, together with possible further work in terms of system development and additional measurements required to describe the performance of the system in a proper manner. Some conclusions regarding the feedback path measurements and the simulations performed are given as well.

- *Bibliography*: The reader can find here all the references made in the report. This is a list of all the monographes, papers and any kind of sources that support this work.

- Appendix:

  - Appendix A, *Human audition*: A brief description of the anatomy of the human auditory system as well as an overview of human loudness perception can be found here.

  - Appendix B, *Feedback and Nyquist Theorem*: An introduction to feedback and stability of systems with feedback through the Nyquist Theorem.

  - Appendix C, *Types of hearing aids*: An overview on the most common hearing aid devices available nowadays.

  - Appendix D, *MATLAB code*: Code of MATLAB functions developed during the simulations of the feedback canceller are listed here.

  - Appendix E, *Standards for hearing aids measurements*: Explanation about hearing aids standards and summary of the one followed in this project, the IEC 60118-8 -*Hearing aids: Methods of measurement of performance characteristics of hearing aids under simulated in situ working conditions*.

  - Appendix F, *Specification sheets*: Specifications of the microphone and receiver mounted in the hearing aid shell.

Chapter 2

Analysis

This chapter gives a deeper explanation of the problem this project is dealing with: feedback in hearing aids; and the different techniques to avoid it. The first part explains the cause of feedback in hearing aids, describing its origins and variable character. It also explains the main approaches to solve the problem, from general to specific solutions proposed in this work, and why such solution is chosen. The second part is dedicated to the theoretical basis of the adaptive filters technique, the proposed solution for feedback in hearing aids in this project. First, an explanation of the advantages of adaptive filtering are given, then the different algorithms and their respective problems are studied, and finally, solutions to these problems when dealing with hearing aid devices are proposed.

2.1 Feedback in hearing aids

One of the necessary conditions for feedback to occur is the existence of a closed loop from the receiver, the output of the hearing aid, to the microphone, the input of the hearing aid. But the existence of such closed loop does not imply the occurrence of feedback. There are two conditions for feedback to occur:

1. The closed loop gain of the electro-acoustical circuit should be greater than one.
2. The input and the feedback signals should be in phase.

Then, the determining factors for feedback to occur are related to the gain of a system and the distance between microphone and receiver, comprising the input and output of a system. In hearing aids, the distances between microphone and receiver are very short compared to, for instance, a PA-system [1]. The distance is approximately 25 mm and along with a high gain, which is required for certain areas of the hearing spectra, feedback is likely to happen.

The whole process can be explained as follows: as sound wave from the receiver propagates in the ear canal, it will travel towards the eardrum and some of the wave get reflected back to the hearing aid itself. As mechanical vibrations can occur in the hearing aid, such vibration could

[1] A PA-system (e.g. concerts) comprise of several microphones and speakers, but compared to the hearing aid, the distance between the microphone and speaker is somewhat different

transfer sound waves to the microphone through the hearing aid mechanical structure. To avoid the undesired *occlusion effect* a vent is introduced to hearing aids. However, the vent immediately becomes the most influential path of the feedback. Sound waves can travel directly or indirectly from the receiver to the microphone through the vent (see appendix A). In the next section, the main components of the feedback path are systematically expressed.

2.1.1 Origins of feedback in hearing aids

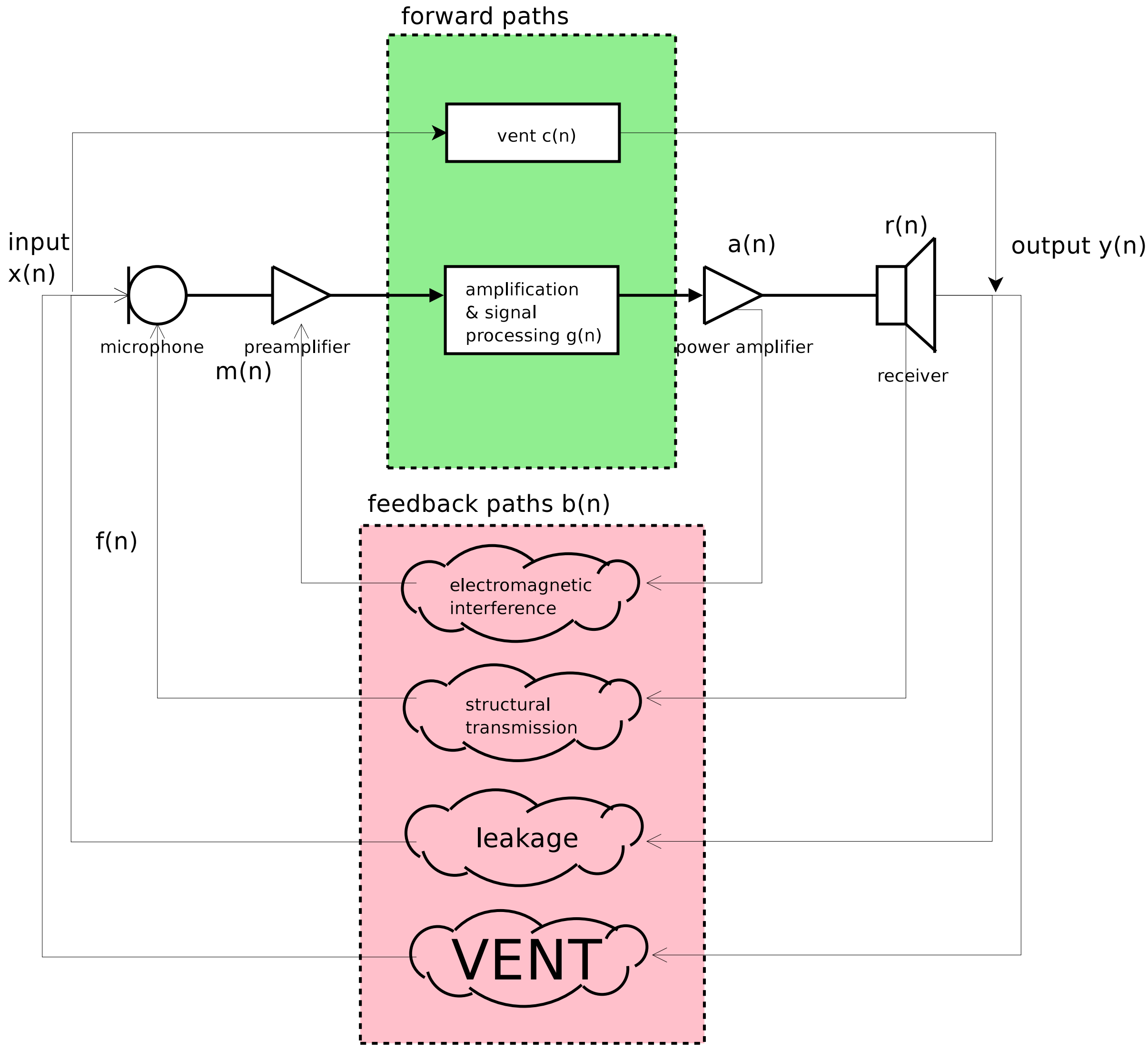

Figure 2.1: Scheme of the forward and feedback paths of a vented ITE hearing aid.

Figure 2.1 depicts the hearing aid-ear system composition regarding to the forward paths and the undesired feedback paths. The lower-case letters represent impulse responses of the system components. The input and output signals are represented as x(n) and y(n) respectively. The microphone and microphone preamplifier signals are represented by m(n). The receiver power amplifier and the receiver as a(n) and r(n), respectively. The feedback paths are represented as b(n), the vent forward transmission path as c(n), and the feedback signal as f(n). The hearing aid forward path processing gain is represented as g(n).

The main components of the feedback path are:

1. Sound waves transmitted from the receiver to the microphone through
   (a) the vent;
   (b) the leakage between the hearing aid and the ear canal;
   (c) the solid structure of the hearing aid itself.
2. The electromagnetic signals generated by the output of the receiver power amplifier picked up by the microphone preamplifier circuitry.

The feedback path in point *2* is included here for completeness. From now, the feedback path due to point *2* will be ignored. It will be included in the feedback path measured empirically because it is difficult (and not necessary) to isolate its effect.

One important point to note is that the feedback paths in *1.a* and *1.b* are not time constant. For instance, although the physical attributes of the vent are not time varying, the feedback path may vary significantly due to external changes outside the vent, such as the case when the hearing aid user covers the hearing aid with his hand. Also when the user is laughing or chewing, the fitting of the hearing aid changes, therefore the feedback path is also changed.

2.1.1.1 Effects of the vent in hearing aids

The main path through which the sound from the receiver can be fed back to the microphone of hearing aids is the vent. A vent is defined as *an opening for the escape of a gas or liquid or for the relief of pressure*. And there are advantages and disadvantages regarding the introduction of a vent in a hearing aid.

Advantages:

- It can reduce the occlusion effect in a hearing aid.
- The perception of internal hearing aid noise will be reduced, especially in the low frequencies.
- The feeling of pressure in the ear is relieved.
- It reduces moisture build-up in the ear canal.

All these advantages can be further increased by making the vent as large as possible.

Disadvantages:

- The volume of the vent combined with the ear canal cavity can produce a Helmholtz resonator, sometimes resulting in an echo or barrel effect when the hearing aid user speaks.
- It encourages feedback. As vent size increases, the amount of acoustic leakage increases, and therefore the probability of feedback increases.

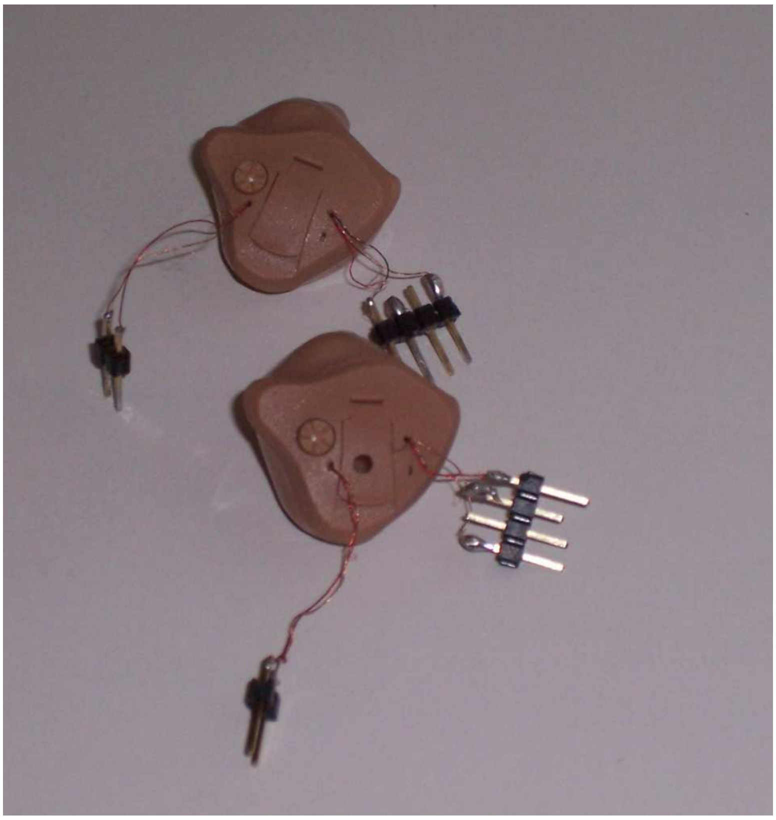

Figure 2.2: Figure of the two hearing aids used during the experiments: one was unvented, the other was drilled at both sides, producing a vent through the shell.

For the experimental study performed during this work, a standard commercial hearing aid vent was not used. In fact, commercial vents are not always a tube with constant diameter [13] as one may think. Therefore, as only two unvented hearing aid shells were provided by Oticon, two holes were drilled on one of the shells. One hole with diameter 1mm was drilled next to the receiver outlet and the other hole with diameter 2mm was drilled next to the microphone to emulate the effect of real hearing aids vents. From now on, it should be considered that although every time in this report such configuration is referred as *vented hearing aid*, it is not a standard commercial hearing aid vent. However it is called like that for practical reasons. The results shown later on the report, show that it can be a good approximation to standard vents, given the similarity observed with the feedback path responses obtained by other researchers. Both the unvented and *vented* hearing aids are shown in figure 2.2

2.1.1.2 Feedback study for the hearing aid system

Referring to figure 2.1, if no type of feedback cancellation is assumed, the output signal frequency response Y can be represented as [18]:

$$Y = \frac{X[C + G(MAR)]}{1 - BC - G(MARB)} \tag{2.1}$$

The upper case letters represent frequency responses of the system components described previously. For example MARB is the product of the Fourier transforms of the signals m(n), a(n), r(n) and b(n) (see figure 2.1). Given that C has a low-pass nature and B is high-pass with reduced gain

[18], their product can be safely neglected. Then, the approximate solution for the output signal frequency response will become:

$$Y = \frac{X[C + G(MAR)]}{1 - G(MARB)} \tag{2.2}$$

From the denominator of equation 2.2, it can be noticed that the system can become unstable if G(MARB) approaches 1. The Nyquist criterion will guarantee stability if:

$$|G(MARB)| < 1 \tag{2.3}$$

System stability will be guaranteed only if the hearing aid gain G is maintained low.

2.1.2 Feedback reduction in hearing aids

In this section, several methods commonly used in hearing aid industry for solving the feedback problem are presented. The most important methods described in this section can help to reduce feedback oscillations electronically. All these techniques can be useful, but none of them will cancel feedback completely [13]. It has to be noted that to obtain a better overall feedback reduction, combination of good shell designs together with electronic feedback reduction techniques have to be considered. First, two techniques that not necessarily require digital processing are explained, they can be called *physical* techniques. The complicated task regarding the mold or shell design, is to make it open enough to avoid occlusion and yet closed enough to prevent feedback to be highly encouraged. Also in this sense the directivity of the microphone and receiver have been considered. Next, the many digital/electronic approaches are investigated. The final method presented in this section is the chosen one, and the supporting reasons for that selection will be presented as the advantages and disadvantages of every method are briefly presented.

2.1.2.1 Feedback reduction by the use of directional microphone and receiver

The possibility of choosing a desired microphone directivity for preventing the encouragement of feedback can be considered. Unfortunately, as sound can come from any direction, at some direction the microphone sensitivity can be too low so that the incoming sound will be severely damped. For the same reason the receiver can also be designed with such a directivity pattern that the reinforced sound points mostly to the eardrum.

2.1.2.2 Feedback reduction by a proper design of the acoustical feedback path

A proper design of the acoustical feedback path may provide an improvement on the final feedback perceived by the hearing aid user. It means that by changing the size and shape of the vent or introducing specific acoustic treatment to it (such as changing its radius at some section, or introducing small resonators like the muffler in a car) could change the acoustic response of the vent and, consequently, the feedback path characteristics. Therefore, the frequency response of the feedback path could be controlled in some extent this way.

2.1.2.3 Feedback reduction by gain frequency response

One of the conditions for feedback is that the closed loop gain (total gain traveling forward through the hearing aid amplifier and transducers and then returning through the feedback path) is greater than 0 dB at some frequency. At this same frequency, the phase shift around the entire loop must also be close to an integral number of periods. One way to avoid feedback is decreasing the gain at those frequencies where this conditions are met. Feedback is likely to occur near the peaks of the gain frequency response curve of the hearing aid. The simplest way to avoid feedback is to turn the overall gain down (to a point that can be below the user's requirement). This is obviously unsatisfactory since all frequencies will be attenuated and it would give the user an inadequate aid. Another approach is to decrease only the amplitude of a peak. This can be done with multichannel hearing aids (many subbands), but if there are only few channels the gain reduction would affect an unnecessary wide frequency range. The use of digital filters can provide a finer control of the gain-frequency response shape. Unfortunately, the frequencies at which feedback occurs do not remain fixed over time. If the hearing aid is slightly moved in the ear, the person moves their jaw, or places a hand close to the ear, the characteristics of the leakage path and hence the feedback frequency will change [9].

2.1.2.4 Feedback reduction by phase control

Another way of preventing feedback is to change the phase at those frequencies that meet the conditions for feedback. This can be done using all-pass filters. These filters have a flat frequency response while the phase can be manipulated to change at some or all frequencies. The problem is that it can be useful at some particular frequency, but as the gain may be increased, another frequency will become a problem. Moreover, if the characteristics of the feedback path change, the required phase manipulations will vary. A rudimentary control of phase is to reverse the connections on the receiver. This will help 50% of the time, for some settings of the volume control [9] and only for low frequencies.

2.1.2.5 Feedback reduction by frequency shifting

If the signal leaking back to the microphone has a different frequency from the original input signal, the two sounds would not be in phase and the amplitude summation would not be as effective. This is the basic principle behind the frequency shifting method. Unfortunately there are disadvantages. A large frequency shift is needed to achieve a significant increase in gain without oscillation [9]. As a result, the quality or pitch of the sound will be changed. Sophisticated methods for altering the frequencies of speech signals without affecting voice quality are needed.

2.1.2.6 Feedback reduction by feedback path cancellation

This method has been implemented in this project. By estimating the feedback path signal and subtracting it from the input, the method acts as a gain and phase control feedback reduction at the same time. Besides, the adaptive nature of the estimation makes it track the changes in the acoustical environment as well. Like the other methods it has limitations. The estimation has to be accurate and fast in order not to cancel the input signal. Sinusoids and impulse train inputs will tend to be cancelled. The method may need a probe signal such as a white noise to determine the feedback path more accurately. This noise signal may be heard by the user and can be annoying. However, it can be avoided with signal processing techniques, so that the estimation is still accurate enough, without injecting an external probe signal. A 10 dB of added gain without oscillation is considered satisfactory by applying this method [9].

2.1.3 Feedback study considering an adaptive feedback canceller (AFC)

Figure 2.3 shows a similar but modified scenario from figure 2.1. An AFC has been put in the forward processing path of the hearing aid.

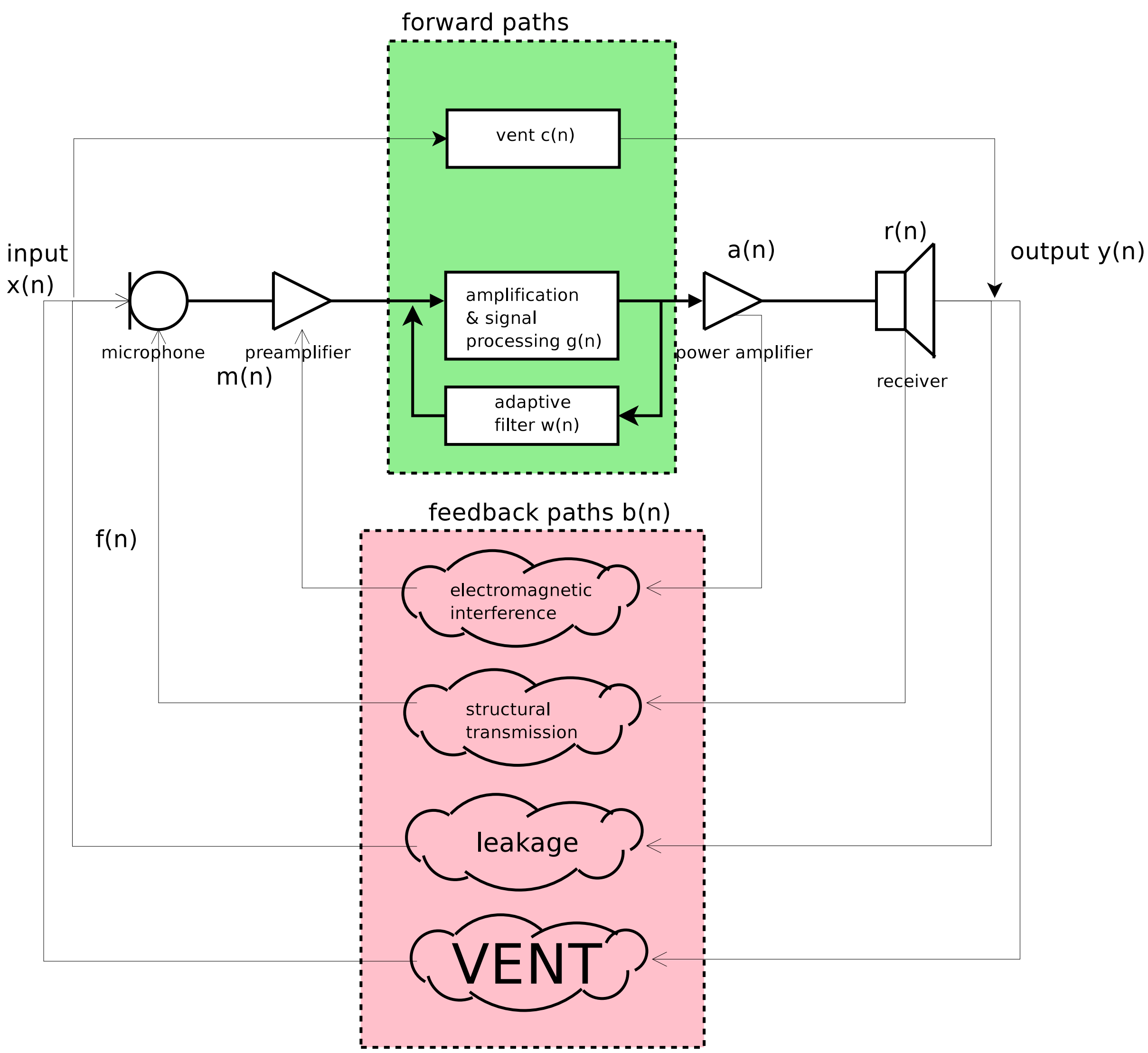

Figure 2.3: Hearing aid system with adaptive filter cancellation filter in the forward path

A similar analysis of the output signal frequency response can be made, this time attention is paid on the influence of the AFC on the system stability conditions. Assuming again $|BC| \ll 1$, the output Y will be given by [18]:

$$Y = \frac{X[C+G(WC+MAR)]}{1-G(MARB-W)}, \tag{2.4}$$

where W is the feedback cancellation filter frequency response. The denominator of equation 2.4 shows that the system will be stable if either the hearing aid processing gain G is low, or if the cancellation filter W comes close to cancel the feedback path MARB. In this case, the Nyquist stability criterion will be:

$$|G(MARB-W)| < 1 \tag{2.5}$$

Feedback cancellation occurs when $W \equiv MARB$. For the system to be stable, a close match between the cancellation filter and the feedback path is required at high processing gains, and the degree of that matching can be lower at lower gains.

The error signal to control the adaptive cancellation filter is given by (assuming again $|BC| \ll 1$) [18]:

$$E \equiv \frac{XM}{1-G(MARB-W)} \tag{2.6}$$

In closed loop operation adaptation can be difficult because the error can be minimized either by reducing the magnitude of the numerator or increasing the magnitude of the denominator. A disadvantage is that for sinusoidal inputs, the denominator of equation 2.6 can become large at the frequency of the of the sinusoid, converting equation 2.6 in a notch filter. This mode of operation can lead to cancellation of sinusoidal input signals.

2.2 Adaptive filters

2.2.1 Basic adaptive filters theory

Any filtering operation involves the extraction of information about a quantity of interest at time t using data measured up to and including t.

Filters can be classified as linear or nonlinear. Linear filters produce an output that is a linear function of the filter input. Otherwise the filter is nonlinear. Different approaches to solve a linear filtering problem can be found.

In statistical approaches like the Wiener filtering, the mean and autocorrelation of the input signal have to be known and unwanted additive noise is assumed. The aim is to design a linear filter such that with the noisy data as input the effects of noise at the filter output are minimized[1]. Normally a *cost function* is defined, which combines important characteristics of the system into a single real-valued number. It acts as an index of the performance of the filter. A useful approach is taking the cost function as the mean square value of the *error signal* defined as the difference between some desired response and the filter output. The resulting solution of this filter optimization problem is known as the Wiener filter solution. This solution is said to be *optimum in the mean square error sense*. However, it not only requires a priori information about the mean and autocorrelation of the input data, but also assumes stationarity of the input signals.

However when the statistical characteristics of the input signal are not known in advance, it is not possible to design the Wiener filter or else the design may no longer be optimal [1]. To overcome this limitation, adaptive filters may be used. This kind of systems are self-designing, which means they rely on a recursive algorithm, making it possible to perform satisfactorily in an environment where complete knowledge of statistical properties of the input signal is not available. The algorithm starts from a set of predetermined initial conditions, and it may converge to the optimum Wiener solution in some statistical sense, after successive iterations in a stationary environment. In non-stationary environments, it offers *tracking* capabilities, in that it can track time variations in the statistics of the input data, provided that variations are sufficiently slow.

In more precise terms, it can be said that typical digital filters have their coefficients fixed throughout the time of execution. Therefore, time stationary filters are useful in situations when the target environment is time stationary and the properties of the environment are known in advance. On the other hand, adaptive filters have their coefficients adapt to certain changes. The aim is to make a filter that tracks the reference input as close as possible by minimizing a *cost function*.

Adaptive filters are considered in this project as the feedback path of the hearing aid is neither time stationary nor its characteristics are completely known in advance.

In adaptive filter theory a reference signal is fed into the adaptive filter, then the filter adjusts its coefficients by minimizing the cost function. The cost function can be the mean square values of the error signals or other criteria. A variety of recursive algorithms have been developed for the operation of adaptive filters. Among those, the most widely used are LMS (Least Mean Squared), NLMS (Normalized Least Mean Squared) and RLS (Recursive Least Squares). This methods will be discussed with some detail in section 2.2.2. The choice of one algorithm over the other can be determined by some of the following factors [1]:

- *Rate of Convergence* It is defined as the number of iterations the algorithm requires to converge close enough to the Wiener solution under stationary inputs. A fast rate of convergence allows the filter to adapt quickly to an environment of unknown statistics.

- *Misalignment* It is a quantitative measure of how the final value of the mean squared error deviates from the mean squared error produced by the Wiener filter.

- *Tracking* In a non time stationary environment, the algorithm is required to track statistical variations of the input signal.

- *Robustness* An adaptive filter is said to be robust if small disturbances produce only small estimation errors.

- *Computational Requirements* Among the issues to be considered here, there is

  - the number of arithmetic operations (multiplications, additions, etc) required to complete one iteration of the algorithm,
  - the size of memory locations required to store the data, and
  - the effort required to program the algorithm on a computer.

- *Structure* The way in which the algorithm is implemented in hardware form will be determined by the structure of the information flow in the algorithm.

- *Numerical Properties* Inaccuracies are produced by quantization errors due to A/D conversion and the digital representation of internal calculations. This last form of quantization errors involves more design challenges. If the adaptive filter is insensitive to variations in the word length used in its digital implementation it is said to be *numerically robust*.

2.2.2 Adaptive filters algorithms

2.2.2.1 Least mean square (LMS) filter

One kind of adaptive filters is called the Least Mean Square (LMS) filter. The LMS filter belongs to the class of linear Finite-duration Impulse Response (FIR) adaptive filters. Its structure is based on a tapped delay line.

The LMS algorithm filters the signals, generates an estimation error by comparing its output to a desired response, and adjusts the filter coefficients according to the estimation error automatically. It is a simple algorithm that requires no pertinent correlation functions and no matrix inversions [1]. It adapts its filter coefficients such that the square of the error signals, $e(n)^2$ is minimized.

Figure 2.4 shows a feedback cancellation diagram that uses an adaptive filter *W(z)*, and can be seen as a basic hearing aid system that uses LMS to cancel out the feedback signal *y(n)*. If the cancellation is correct, the adaptive filter *W(z)* will correctly estimate the feedback path *F(z)*. The input signal *x(n)* can be considered as speech, the desired signal *d(n)* is the sum of *x(n)* and *y(n)*. The output signal to the receiver *s(n)* is the input to the adaptive filter with coefficient taps *w(n)*. The LMS based adaptive filter updates its coefficients based on *d(n)* and the scalar filter output *y_(n)*, which is given by:

$$y\_(n) = w'(n)s(n),$$

where $w'(n)$ is the transposed coefficient vector. The estimation error is given by

$$e(n) = d(n) - y\_(n)$$

The essence of the LMS filter can be described as the formula [1]

$$\begin{pmatrix} updated \\ value\ of \\ tap\ weight \\ vector \end{pmatrix} = \begin{pmatrix} old\ value\ of \\ tap\ weight \\ vector \end{pmatrix} + \begin{pmatrix} learning \\ rate \\ parameter \end{pmatrix} \begin{pmatrix} tap\ input \\ vector \end{pmatrix} \begin{pmatrix} error \\ signal \end{pmatrix}$$

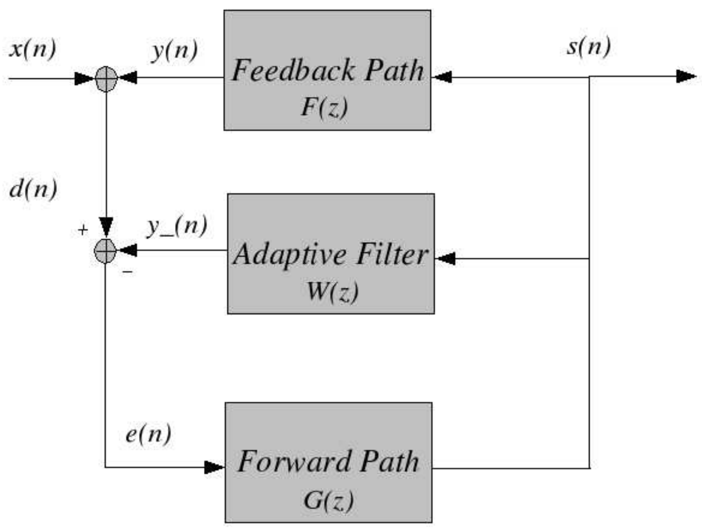

Figure 2.4: Basic feedback cancellation diagram

where the error signal is the difference between the reference input to the filter and the actual response of the FIR filter due to that reference input.

The equation can be mathematically written as:

$$\mathbf{w}(n+1) = \mathbf{w}(n) + \mu \mathbf{s}(n) e^{*}(n)$$

The mean square error is a second order function of tap weights.

Under the context of hearing aids, the feedback path is not time constant. The resulting estimation system is non time stationary.

In ideal cases, *w(n)* will converge to *f(n)*, the feedback path filter coefficients. In this case the feedback gets cancelled and the $e(n)$ will be the desired signal $x(n)$.

2.2.2.2 Normalized least mean square(NLMS) filter

A more refined approach of basic LMS filter is using a non constant step size. Such technique in adjusting the step size is known as the Normalized Least Mean Square (NLMS) filter. In the basic

LMS filter the correction term is given by the multiplication of the step-size, the tap input vector and the error signal. So when the input vector or the reference signal is large, the correction term will become very large and cause a gradient noise amplification problem.

To resolve the issue, the normalized LMS (NLMS) filter can be used. As the name implies, the correction term of a NLMS filter is normalized with respect to the squared Euclidean norm of the reference signal at iteration n.

$$\mathbf{w}(n+1) = \mathbf{w}(n) + \frac{\mu}{\|\mathbf{s}(n)\|^2}\mathbf{s}(n)e^*(n)$$

The most important point of the NLMS filter is that the convergence time is potentially faster than that of LMS filter. In addition, to solve the potential problem of divide by zero in the above equation, a slightly modified version of the same equation is shown as:

$$\mathbf{w}(n+1) = \mathbf{w}(n) + \frac{\mu}{a+\|\mathbf{s}(n)\|^2}\mathbf{s}(n)e^*(n)$$

where a is a small constant and $a > 0$.

2.2.2.3 Recursive least square(RLS) filter

Recursive least squares algorithm is used in adaptive filters to find the filter coefficients that recursively produce the least squares of the error signal. The idea behind RLS filters is to minimize a weighted least squares error function. So, the RLS algorithm achieves recursively an exact least squares solution. In the RLS algorithm, the computation of the correction is applied by updating the old estimate of the coefficient vector utilizing all the past available information. The correction applied to the previous estimate consists of the product of two factors: the true estimation error $\eta(n)$ and the gain vector $k(n)$.

The gain vector itself consists of $\Phi^{-1}(n)$, the inverse of the deterministic correlation matrix, multiplied by the tap-input vector *s(n)*. The major difference between the LMS and RLS algorithms is therefore the presence of $\Phi^{-1}(n)$ in the correction term of the RLS algorithm. The term $\Phi^{-1}(n)$ has the effect of decorrelating the successive tap inputs, thereby making the RLS algorithm self-orthogonalizing. The RLS algorithm converges in the mean square sense within less than $2M$ iterations, where M is the length of the adaptive filter. The rate of convergence of the RLS algorithm is therefore, in general, faster than that of the LMS algorithm by an order of magnitude. But it is not generally used since it is unstable when its used with a forgetting factor λ. Even though, the RLS algorithm, in theory, exhibits zero misalignment [1].

On the other hand, the LMS algorithm always exhibits a nonzero misalignment; however, this misalignment may be made arbitrarily small by using a sufficiently small step-size parameter μ.

2.2.2.4 Summary of the RLS algorithm

Given the set of input samples $s(1), s(2), s(3), ..., s(N)$ and the set of desired response $d(1), d(2), d(3), ..., d(N)$.

1. Initialization of RLS algorithm:

   In RLS algorithm there are two variables involved in the recursion for which initial values in order to start the recursions must be provided:

   (a) $w(n+1)$

   (b) P_{n-1}

   Then, the value of $w(0)$ and $P(0)$ must be stated:

   - $w(0)$: If some priori information about the parameters w was known this information will be used to initialize the algorithm. Otherwise, the typical initialization is w(0) = 0.

   - $P(0)$: P(n) is defined as

$$P(n) = \Phi^{-1}(n) = \left[\sum_{i=i_1}^{n} \lambda^{n-1} \underline{s}(i)\underline{s}^T\right]^{-1} \tag{2.7}$$

     However, the approximate initialization is commonly used. It doesn't require matrix inversion:

$$P(0) = \delta \cdot I \tag{2.8}$$

     δ should be a large or small positive constant for low and high SNR respectively.

2. For each time instant, n = 1, . . . , N, Compute:

   (a)

$$\pi = s(n)P(n-1) \tag{2.9}$$

   (b)

$$k(n) = \frac{\pi(n)}{\lambda + s(n-1)'s(n)} \tag{2.10}$$

   (c)

$$\varepsilon(n) = d(n) - w'(n-1)s(n) \tag{2.11}$$

   (d)

$$w(n) = w(n-1) + k(n)\varepsilon^*(n) \tag{2.12}$$

   (e)

$$P(n) = \lambda^{-1}P(n-1)(1 - k(n)s'(n)) \tag{2.13}$$

2.2.2.5 Subband adaptive filter

By separating the input signal into several subbands, it is possible to run adaptive filters in each of the subbands with a fraction of the complexity compared to the wide-band approach [8]. Besides, the overall spectral shape of the signal in each subband is "flattened", leading to a decrease in the eigenvalue spread. Figure 2.5 shows a two band example of the structure of the adaptive filtering [8].

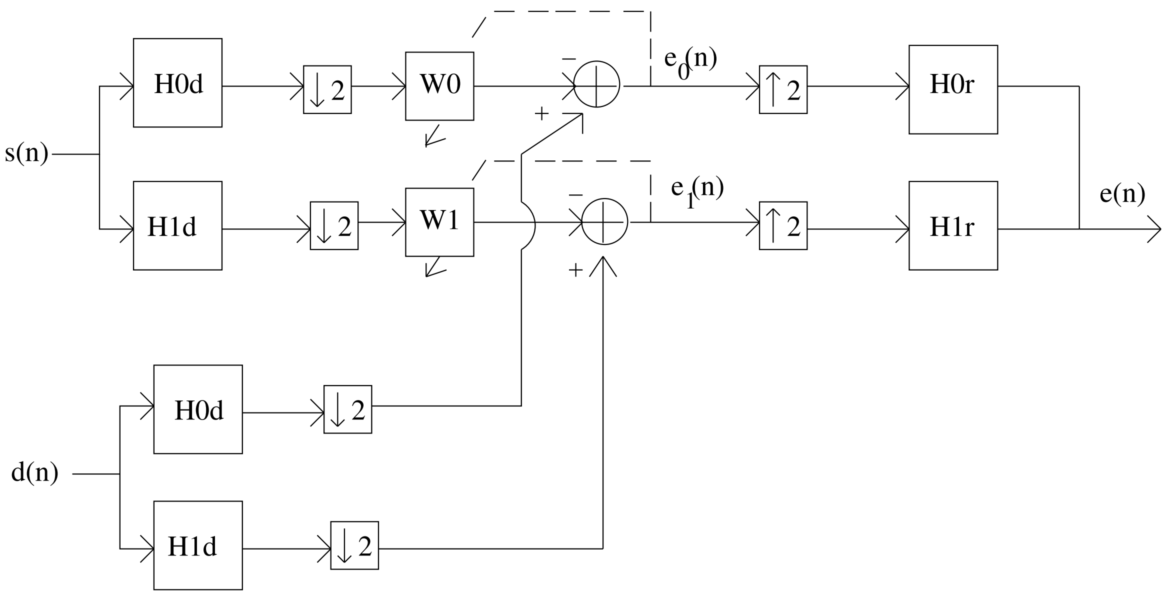

Figure 2.5: Two band structure for subband adaptive filtering

The filters at the beginning and the end of the diagram are the analysis and synthesis quadrature mirror filters (QMF) respectively. The analysis filters are required to separate the signal into *M* subbands and the synthesis filters ensure proper reconstruction (*M*=2 in this case). These filters can be taken as recommended in [20].

Mathematically,

$$H_0r(z) = M(-H_1d(-z)) \tag{2.14}$$

$$H_1r(z) = M(H_0d(-z)) \tag{2.15}$$

and $H_1d(z) = H_0d(-z)$ can be chosen, so that they are half-band QMF. The factor of *M* comes from the fact that when a signal is up-sampled by *M* and then filtered, the signal is scaled by a factor of *M*.

As after the filtering the information in the signal is band-limited, decimation of the signal by a factor of *M* can be done. Thus, the complexity of each adaptive filter is *1/M* of that of the wideband approach.

The subband algorithm uses a version of the NLMS algorithm, which is described by the following equations [8]:

$$P_s(n) = \lambda P_s(n-1) + (1-\lambda)s^2(n-1) \tag{2.16}$$

$$P_e(n) = \lambda P_e(n-1) + (1-\lambda)e^2(n-1) \tag{2.17}$$

$$e(n) = d(n) - w^T(n-1)s(n-1) \tag{2.18}$$

$$w(n) = w(n-1) + \frac{2\alpha e(n)s(n-1)}{N(P_s(n)+P_e(n))} \tag{2.19}$$

where

- $s(n)$ is the input signal vector to the adaptive filter defined as $[s(n), s(n-1), ..., s(n-N+1)]$ where N is the size of the adaptive filter (see basic diagram on figure 2.4 of section 2.2.2.
- $e(n)$ is the error signal used by the adaptive algorithm to update the filter coefficients.
- $P_s(n), P_e(n)$ are the estimates of the power of $s(n)$ and $e(n)$ respectively.
- λ is the forgetting factor.
- $\frac{\alpha}{N(P_s(n)+P_e(n))}$ is the NLMS adaptation step size after [6].

Simulations were performed (see section 3.3) in order to see the behavior of the subband adaptive filter cancellation algorithm when applied to a system such as the one shown in figure 2.4.

2.3 Adaptive filters applied to feedback cancellation in hearing aids

Feedback in hearing aids limits the maximum usable gain of the device and degrades the overall system response [2]. Feedback cancellation in hearing aids using adaptive filtering involves the estimation of the feedback signal and the subtraction of that feedback signal from the microphone input signal. The final objective is to increase the maximum usable gain of the instrument while preserving the intelligibility of speech.

An implementation of adaptive filtering in feedback cancellation should work with any hearing aid configuration and under real acoustic situations. It also should be computationally efficient so as to offer a practical system. A real system has to work in a changing acoustical environment.

The acoustic environment for a hearing aid will change, for example,

- when a telephone receiver is moved close to the aided ear,
- when a hand is moved close to the device to adjust the volume control of the hearing aid, and
- when the position of the hearing aid is shifted due to eating or talking [2].

Therefore, a fixed coefficient filter cannot be used as such filter cannot adjust to changes in the acoustical environment. An adaptive filter, on the other hand, will update the estimated feedback path whenever changes in acoustical environment occur. Based on the form of the adaptation procedure, non-continuous and continuous adaptation systems have been proposed for hearing aids feedback cancellers.

2.3.1 Non continuous adaptation

Non-continuous adaptation systems usually use white noise as a probe signal under certain occasions to estimate the acoustic feedback path[3]. The normal hearing aid processing is interrupted only when:

1. the input of the hearing aid is silent,
2. feedback is detected in the hearing aid.

The adaptation starts with disengaging the microphone to the amplifier of the hearing aid temporarily. Then, the training white noise sequence is injected into the input of the amplifiers. Next, the white noise is amplified and then fed into the receiver. The signal that the microphone picked up is transmitted via the feedback path of the hearing aid. Hence, a reliable reference signal is obtained. The adaptive filter then adjust its coefficients according to the reference signal. Once the adaptation is complete the microphone is re-engaged to the amplifier and the hearing aid is switched back into working mode [2].

When using a white noise training signal, a relatively accurate feedback path estimation can be achieved. However, adaptation is limited to periods when silence or howling is detected. In addition, the interference nature of the training sequence can lower the signal-to-noise ratio (SNR), and it can be effective only in limited cases[3], [2].

Regarding the processing requirements, the time constraint for the filter to adapt is very tight. The adaptive filter should be able to finish its adaptation within a few milliseconds.

2.3.2 Continuous adaptation

Continuous adaptation systems constantly adapt the filter coefficients based on the input signal, not requiring the use of any training signal. Figure 2.4 shows the basic diagram for a continuous adaptation filter.

The adaptive filter *W(z)* continuously tries to adapt to the feedback path transfer function *F(z)* [4]. The principal disadvantage of continuous adaptation is that the desired signal *d(n)* is the sum of the input signal *x(n)* and the feedback signal *y(n)*. However the input signal *x(n)* and the hearing aid output signal *s(n)* are correlated, causing the feedback signal *y(n)* to be correlated with the input signal *x(n)* [4]. Due to this fact, the adaptive filter cannot estimate the feedback path properly as it will be demonstrated in the next section. However, different methods have been proposed to decorrelate the input and output signals. They will also be discussed in the next section.

2.3.3 Correlation problem in continuous adaptation systems

Continuous adaptation systems present many advantages compared to non continuous counterparts. However, due to their closed loop nature, continuous adaptation systems have performance problems on the adaptive filter involved. The problem comes from the correlation between the input signal applied to the hearing aid and the final output. In this case the LMS algorithm does no converge, and therefore, it introduces a bias in the output. To solve the bias problem hearing aid designers use additional techniques, such as delaying modules to decorrelate the mentioned signals. A convergence analysis of a continuous adaptive system based on the LMS algorithm can be done to show this problem.

In continuous adaptation systems the adaptive algorithm updates the filter coefficients based on the input signal, as shown in figure 2.4. Based on the hearing aid output *s(n)* and the desired signal *d(n)*, the adaptive filter *W(z)* tries to estimate the feedback path transfer function *F(z)*. Because *d(n)* is the sum of the input signal *x(n)* and the feedback signal *y(n)*, the estimation will only be correct under the assumption that *x(n)* and *y(n)* are uncorrelated. Unfortunately this condition is not satisfied in most practical cases[4].

A requirement for the convergence of the LMS algorithm is that the adaptive filter tap weights difference vector must converge to zero in the steady state. However if the basic system in figure 2.4 was used it is not possible for the system to converge. The adaptive filter tap weights difference vector is defined as:

$$\varepsilon(n) = \hat{w}(n) - f \tag{2.20}$$

where $f^T = [f(0)f(1)f(2)...f(N)]$, $f(n) = Z^{-1}[F(z)]$, $\hat{w}$ is the adaptive filter tap weights vector and N is the order of the filter.

To check the convergence of ε is equivalent to evaluate its expected value when $n \to \infty$.

In order to find that expression the state equation of the update step of the algorithm was considered. The state equation is:

$$\hat{w}(n+1) = \hat{w}(n) + \mu s(n)e(n) \tag{2.21}$$

In this equation, the adaptive filter input signal vector is represented by $s(n)$ and

$s^T = [s(n)s(n-1)s(n-2)...s(n-N)]$, $\hat{w}^T(n)$
can be expressed $\hat{w}^T(n) = [\hat{w}_0(n)\hat{w}_1(n)\hat{w}_2(n)...\hat{w}_N(n)]$ and, finally:

$$e(n) = d(n) - s^T(n)\hat{w}(n). \tag{2.22}$$

Combining equations 2.20, 2.21 and 2.22, an iterative expression for the adaptive filter tap weights difference vector can be found:

$$\varepsilon(n+1) = \varepsilon(n) - \mu s(n)s^T(n)\varepsilon(n) + \mu x(n)s(n) \tag{2.23}$$

Taking the expected value of this expression and assuming small values for the convergence factor μ [1] lead us to the equation:

$$E[\varepsilon(n+1)] = [I - \mu E[s(n)s^T(n)]E[\varepsilon(n)] + \mu E[x(n)s(n)] \tag{2.24}$$

$E[\varepsilon(n)]$ will not converge to zero because of the term $\mu E[x(n)s(n)]$. In steady state, when $n \to \infty$:

$$E[\varepsilon] = E^{-1}[s(n)s^T(n)]E[x(n)s(n)] \tag{2.25}$$

In the system modeled if it cannot be assumed that s(n) and x(n) are uncorrelated, then their cross-correlation vector is non-zero and the mean of the adaptive filter taps will not converge to zero. Therefore, the use of an LMS system (the demonstration can be extended to NLMS [11]) will lead to a biased solution in hearing aids feedback cancellation if further processing was added.

There are different possible techniques to avoid this problem. One of the techniques consists of adding some other processing in the basic block diagram as depicted in the figure 2.4 that decorrelates the input and output of the hearing aid. Some of these possibilities are suggested in [12] and they are:

- The introduction of delay either on the forward path or in the feedback path. This technique will be further explored in chapter 3. It is based on the statistical properties of speech signals.
- The addition of non-linear processing to one of those signals. A nonlinear forward path can reduce the correlation between the input and feedback signals [17]. This technique was not further explored because of the lack of substantial reference material applying it and because of time issues.
- The addition of artificial noise. This approach is also used in non continuous adaptation systems. A probe signal *r(n)* independent of the input signal *x(n)* can be injected into the system in order to get a better estimation of the feedback path. Generally *r(n)* is a noise signal.

2.3.4 Performance descriptors

In order to test the performance of the AFC algorithms, it is necessary to define some parameters used when referring to AFC in hearing aids [16], [17]:

- *Weighted Error Vector Norm (WEVN)*

  It is a normalized version of the difference tap weight vector (defined in section 2.3.3), and it is defined as:

$$WEVN(n) = 10log_{10}\frac{\|w(n)-f\|^2}{\|f\|^2} \tag{2.26}$$

  A high negative dB value in the WEVN means the estimation is accurate. On the other hand, a positive dB value indicates a high bias in the estimation.

- *Maximum Stable Gain (MSG)*

  It is defined as the maximum allowable gain without oscillation, assuming a flat forward path frequency response gain $G(e^{j\omega})$:

$$MSG = 20log_{10}min\frac{1}{|F(e^{j\omega})-W(e^{j\omega})|}, \tag{2.27}$$

  where $F(e^{j\omega})$ is the real feedback path frequency response, $W(e^{j\omega})$ is the estimated feedback path frequency response, and *min* is the minimum operator.

  For any forward path frequency response $G(e^{j\omega})$, the maximum stable gain generally becomes:

$$MSG = 20log_{10}min\frac{1}{|G(e^{j\omega})(F(e^{j\omega})-W(e^{j\omega}))|} \tag{2.28}$$

- *Added Stable Gain (ASG)* It is defined as the additional gain that is possible by using the feedback canceller $W(e^{j\omega})$, assuming a flat forward path frequency response:

$$ASG = MSG - 20log_{10}min\frac{1}{|F(e^{j\omega})|}, \tag{2.29}$$

  For the case of any forward path frequency response
  $G(e^{j\omega})$, the ASG becomes:

$$ASG = 20log_{10}min\frac{1}{|G(e^{j\omega})(F(e^{j\omega})-W(e^{j\omega}))|} - 20log_{10}min\frac{1}{|G(e^{j\omega})F(e^{j\omega})|} \tag{2.30}$$

$$ASG = 20log_{10}min\frac{|F(e^{j\omega}|}{|F(e^{j\omega})-W(e^{j\omega})|} \tag{2.31}$$

Equation 2.31 shows that the ASG does not depend directly on the forward path characteristics but on how good the estimation of the AFC is. Therefore, it can be obtained by directly applying equation 2.29.

These descriptors of the performance of the AFC will be used in the simulations, where real feedback path responses will try to be cancelled (see section 3.3). They will also be used in the final measurement of the implemented system (see section 5.1), where the ASG will be obtained when the forward path is handled by the DSP.

Chapter 3

System design

This chapter describes how the adaptive algorithm was designed and simulated in MATLAB. It also gives an overview of the hardware elements used in the system. A description of the feedback path study required to input data to the simulations performed in MATLAB is given as well.

The goal was to design a feedback canceller which can be included in a general system such as the one depicted in fig. 3.1. First, the rest of the hearing aid system had to be designed and physically implemented before the testing of the AFC algorithm. When the rest of the system was implemented, some data related to the hearing aid could be extracted. These data were basically the impulse response of the feedback path of the hearing aid when the hearing aid was worn by a head and torso simulator. The real system responses were used to run the simulation algorithms in MATLAB in order to obtain reliable results.

This chapter is divided into three parts:

- First, a summary of the specifications of the desired system is given. This is followed by a section with a short description of the design of the hardware involved.
- Second, a report of the measurements of the feedback path is presented. The measurements were based on the system explained before.
- At last, the design of the feedback canceller is explained. The explanation is based on the feedback path measurements. Different simulations that made use of variations of adaptive filter techniques were run in MATLAB.

3.1 System Specifications

Feedback cancellation for an "in the ear" (ITE) vented device is being considered because it is the one in which feedback effect is stronger. ITE hearing aids usually provide less amplification than the bigger-sized "behind the ear" (BTE) hearing aids due to a weaker amplifier and smaller battery. ITE hearing aids are recommended for people who have mild to moderately severe hearing losses. This is the device around which the AFC system was developed. The system is based on the commercial shell *Oticon GO ITE Power*, and a receiver and a microphone with external connections, all of them provided by Oticon[1]. The relative positions of the receiver, microphone and the drilled

[1]More information in *www.oticon.com*

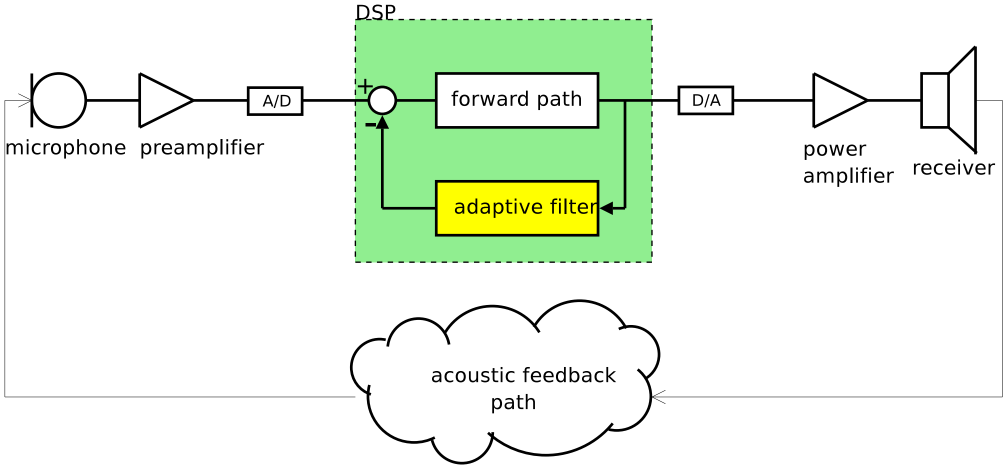

Figure 3.1: Block diagram of the total system

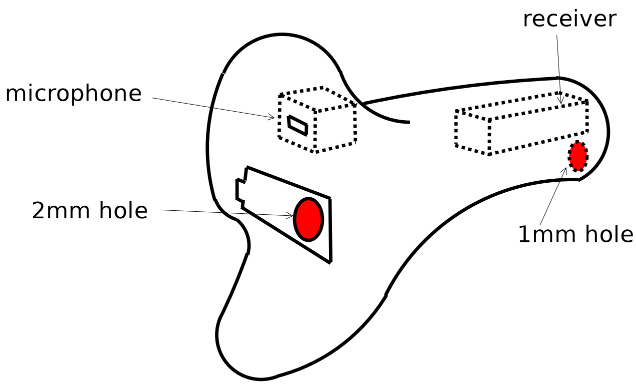

Figure 3.2: Components inside the hearing aid shell

holes within the shell are shown in figure 3.2. The specification sheets of microphone and receiver are attached in appendix F.

3.1.1 Adaptive filter canceller requirements

The objective was to implement an adaptive feedback canceller which performance was:

- Frequency range: 200 - 8 kHz to cover the whole speech frequency range. The sampling rate of the analog to digital conversion is 16 kHz the cut off frequency of 8 kHz.
- Adaptive filter length: At the mentioned sample rate, the feedback paths to be estimated were found to be approximately 100 taps long (see section 3.2).
- Processing power: The AFC chosen is based on NLMS algorithm. This algorithm is going to be executed in a DSP. The number of operations required to execute this algorithm is around 4M (additions and sums) [1], where M=100 taps. The time required for processing the 4M operations at 16 kHz is $(1/16000)/400 = 1.5626 * 10^{-7}$ seconds. That means the DSP processing power should be above 6.4 million instruction per second (MIPS). Because of the availability and to provide enough programing flexibility a DSP of 20 MIPS was chosen.

- Maximum latency: Because of the method chosen, a delay of approximately 1 ms is required in the forward path. To do this accurately, the combined latency of the A/D D/A and DSP processing should be much lower. At 16 kHz sample rate the latency of the A/D D/A will be approximately 0.125 ms and within the same time the 20 MIPS DSP can execute up to 2500 instructions. So, the maximum latency present in the system is only a fraction of a millisecond.
- Added stable gain of 8 dB: This is considered as a fairly good performance of cancellers based on continuous adaptive filters [9], [17].

3.1.2 DSP system

- DSP speed: 20 MIPS

- DSP model: TMS320C50

- Analog-to-digital and digital-to-analog converters: Crystal CS4218 AD/DA converter.
  - Resolution: 16 bit
  - Sampling rate: 16 kHz
  - Signal-to-noise ratio: 96 dB, it is the quantization noise when using 16 bit resolution.

- Debugger/monitor tool: D/M320C/2x/2xx/5x

3.1.3 Hardware

The components involved in the whole system are:

- Microphone
- Microphone preamplifier
- Analog-to-digital (A/D) converter
- DSP system
- Digital-to-analog (D/A) converter
- Power amplifier
- Receiver

The microphone and receiver were provided with the shell by Oticon. The DSP board, A/D and D/A converters were commercial devices described in the previous section. The power amplifier was a Fostex headphone amplifier.

The microphone preamplifier supplied was in the form of a circuit board which measured 6mm x 3mm. However the schematic diagram bundled with the circuit board showed the circuit was not completed and the function of the pins were not known. Special considerations and design work were needed. With the help of Mr. Jesper Krogh Christensen of Oticon the correct way to make the connections between the circuit board and other components like the microphone, receiver, power supply, extra capacitors and resistors was known.

Due to the fact that the output voltage of a microphone is low, a preamplifier in close proximity is needed to boost the signal voltage for two reasons: to prevent 50 Hz hum from power outlets from contaminating the signal; to minimize the resistive loss due to transmitting through long cables.

The first two operational amplifiers in the microphone preamplifier circuit are configured as seen in figure 3.3. The power supply voltage of the circuit is 1.3V as specified by Oticon.

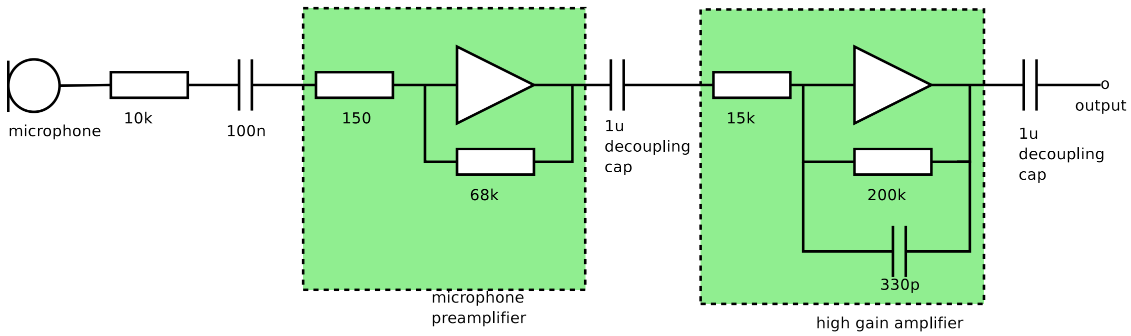

Figure 3.3: Schematic of the microphone preamplifier circuit

The signal from the microphone is first fed through the first amplifier a gain of 6.6, see equation 3.1.

$$G \approx \frac{68k\Omega}{10150\Omega} = 6.6 \tag{3.1}$$

A capacitor of $1\mu F$ is connected between the two amplifiers to avoid DC from the output of the first part. The size of the decoupling capacitor is chosen such that it has no effect on the frequency response of the amplifiers.

The second amplifier in the Oticon circuit is also used as the microphone preamplifier to provide an additional gain of approximately 13, see equation 3.2.

$$G \approx \frac{200k\Omega}{15k\Omega} = 13.3 \tag{3.2}$$

For the same reason as mentioned before, a capacitor of $1\mu F$ is connected on the output side of the second part of the microphone amplifier. To verify the gain of the amplifier circuit, an measurement was carried out. The setup of the experiment is shown in figure 3.4

The Oticon amplifier circuit has an output stage for driving the receiver. However, after measurements and casual listening test a significant amount distortion was found. It was decided that the output stage on the Oticon circuit board should be replaced by a Fostex headphone amplifier. So the forward path amplifier is consisted of the following amplifiers: the microphone preamplifier, the high gain amplifier from the Oticon amplifier circuit and a Fostex headphone amplifier. From

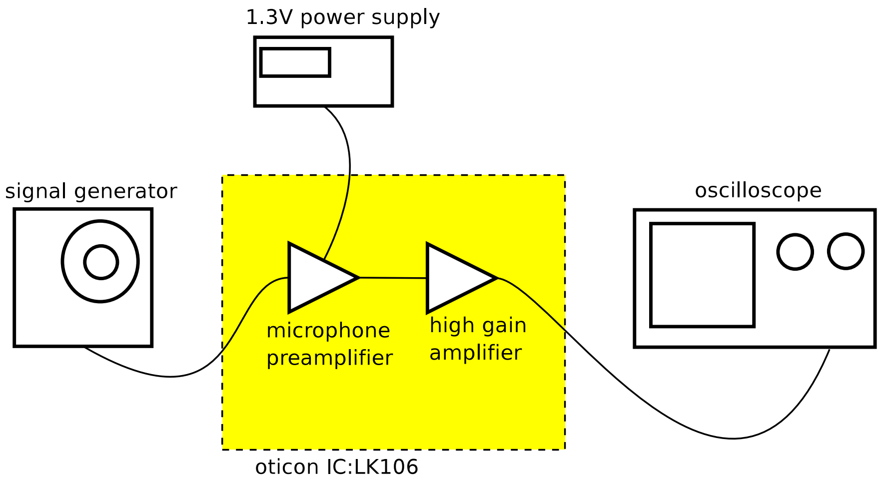

Figure 3.4: Experiment setup for gain measurement of the Oticon microphone preamplifier

now on when the microphone preamplifier is referred, it means the combined microphone preamplifier and high gain amplifier.

To verify overall gain of forward path, an measurement was carried out. The setup of the experiment is shown in figure 3.5

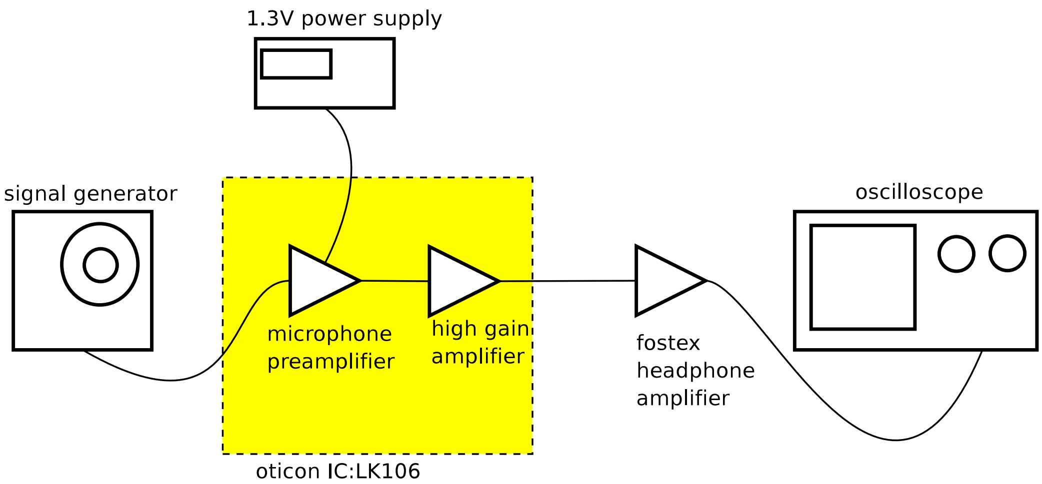

Figure 3.5: Experiment setup for gain measurement of the Oticon microphone preamplifier and Fostex headphone amplifier

In both experiments the input voltage of the microphone preamplifier was fixed at 2mV peak to peak. This value was chosen as it is similar to the typical output voltage of the microphone. The measurement results of the two experiments are shown in the table 3.1, with all voltages expressed as peak to peak values.

From the results, the gain of the microphone amplifier was found to be 55. The calculated gain was 88. The measured gain did not closely match with the calculated gain. One of the possible

Table 3.1: Measured output voltage of microphone preamplifier and headphone amplifier under different input frequency

Frequency	Voltage of microphone preamp	Voltage of Fostex amp	Overall gain
250 Hz	110 mV	1.25 V	56 dB
500 Hz	110 mV	1.25 V	56 dB
1 kHz	110 mV	1.25 V	56 dB
2 kHz	97 mV	1.10 V	55 dB
4 kHz	65 mV	0.75 V	51 dB
8 kHz	40 mV	0.45 V	47 dB

reasons was the input resistance of the microphone may have a effect on the overall gain.

The gain starts to rolloff at around 2 kHz. The results agree with the circuit diagram as shown in figure 3.3. In the amplifier on the right hand side there is a shunt capacitor to the 200 k resistor, at high frequency the capacitor decreases the impedance of that resistor. As a result the gain of the high gain amplifier decreases as frequency increases.

3.2 Experimental study of feedback path on a real ITE hearing aid

The theoretical analysis of the feedback path, of course, is not enough to describe the real behaviour of hearing aids in terms of the feedback problem that hearing aid users may suffer from. For experimental analysis of feedback in hearing aids an ITE hearing aid device provided by the danish company Oticon was used. As it has been mentioned in previous sections, a vent usually included to prevent the occlusion effect introduces an additional acoustical feedback path that encourages feedback in a big extent.

The goal of the first experiment was to measure the feedback path of the provided hearing aid when the hearing aid was mounted on the B&K Head And Torso Simulator (HATS) 4128C. Experiments were carried out in both an unvented and the *vented* (see section 2.1.1.1) hearing aid. First the effect of introducing a vent in the hearing aid was studied. Then for the vented hearing aid different acoustic situations such as vary the volume levels and introducing objects near the hearing aid were studied. With these results, it was possible to run simulations of the feedback canceller algorithm.

The purpose of the measurement is to obtain the impulse response of the electro acoustic feedback path of the vented and unvented hearing aid under different acoustic environments. The experiment set up and procedure are explained in the following section.

3.2.1 Test set-up

Figure 3.6 shows the test set-up for the measurement of the acoustic feedback path. The setup is valid for both the original and the vented hearing aid.

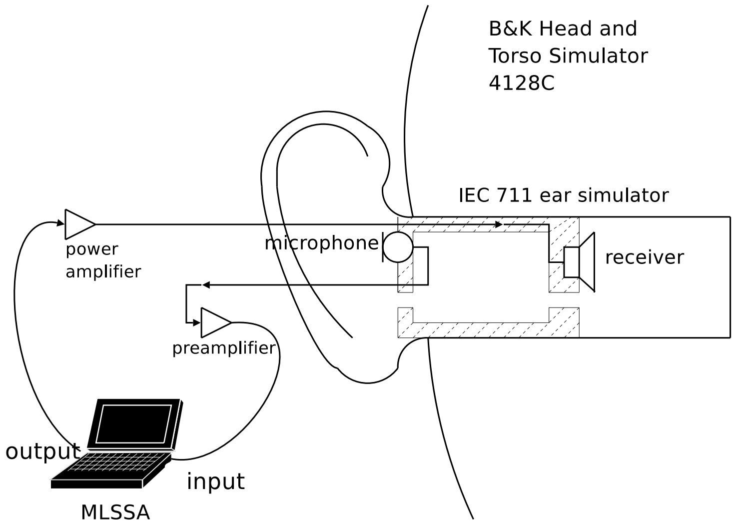

Figure 3.6: Test set-up, measurement of the impulse response of the hearing aid.

The reasons to use the HATS are to emulate the real situation of the acoustic system hearing aid-user (including the pinna effects) and to follow the standard IEC-60118 (the standard for measuring the characteristics of hearing aids). An almost identical experiment setup will be used when the AFC is implemented on a DSP.

The experiments were carried out in the VR-Laboratory of the Acoustics Department, Aalborg University. The laboratory has good sound isolation and dry acoustic environment. Measurements can be carried out in an non anechoic room because the sound level radiated by the hearing aid receiver towards the room is not large enough to excite the room, the microphone will not pick up the influence of the room.

3.2.2 List of equipment

Instrument	Model
Measurement system	MLSSA
Microphone	Knowles Electronics FG-23653-C36 (Oticon)
Receiver	Knowles Electronics EC-26675-A33 (Oticon)
Receiver amplifier	Fostex PH-5 headphone amplifier
Microphone pre amplifier	based on circuit IC:LK106 (Oticon)
Head and Torso Simulator	B&K 4128 C
Experiments room	Virtual Reality Lab (B3-103)

Table 3.2: List of equipment used in the test set-up for the measure of the impulse response of the acoustic feedback path.

3.2.3 Applied signal

Using the *MLSSA* system, a pseudo-random noise signal is sent to the input of the receiver's amplifier of the hearing aid. The acoustic signal produced by the receiver was picked up by the

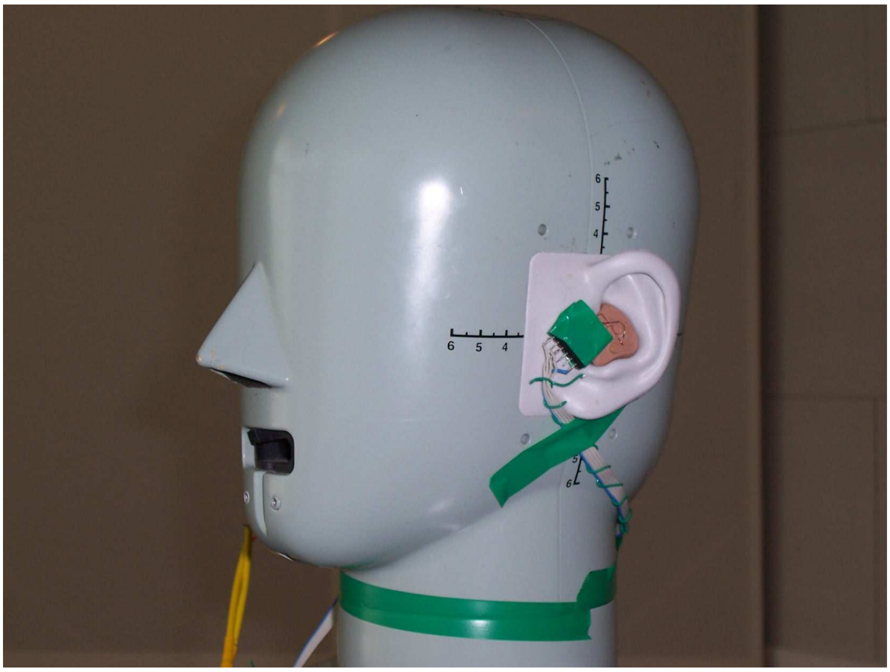

Figure 3.7: Mounting of the hearing aid on the *B&K* head and torso simulator

microphone of the hearing aid. The output level of the receiver was varied by changing the output voltage of the MLSSA system. The details of the MLSSA setup are shown in the table 3.3.

Parameter	**value**
length of the MLSSA sequence	16383
bust amplitude	0.08203 V
acquisition mode	cross correlation
acquisition length	65536 samples
sampling rate	32 kHz
pre-average cycles	12

Table 3.3: MLSSA setup details

3.2.4 Procedure

The hearing aid is mounted in the ear canal of the HATS as shown in figure 3.8. The receiver and microphone of the hearing aid are respectively connected to the output of the receiver amplifier and input of the microphone preamplifier provided by Oticon. The MLSSA input and output are connected to the output of the microphone preamplifier and input of the receiver amplifier. During the measurements, the microphone picked up the signal applied to the receiver transmitted through the feedback path. In this way the impulse response of the feedback path is measured.

To measure the effect of objects in close proximity of the hearing aid on the feedback path, following objects were placed close to the hearing aid one by one:

- a mobile phone (Nokia 2650)

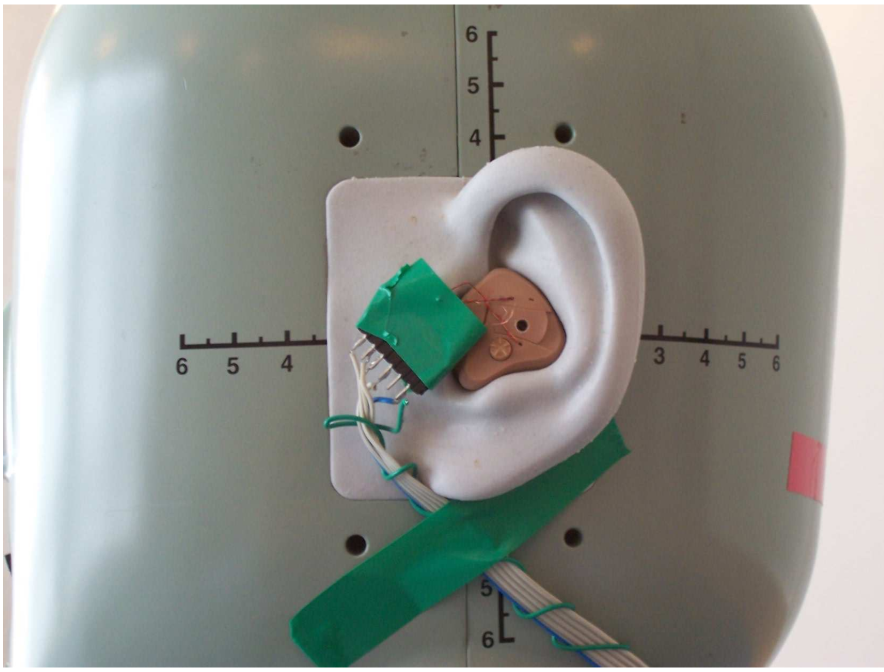

Figure 3.8: Mounting of the hearing aid on the *B&K* head and torso simulator

- a baffle (a wooden table oriented such that the table top is normal to the hearing aid)
- a human hand
- a sofa

In this way, it was possible to simulate real situations that hearing aids users may experience.

The mobile phone was placed close to the "ear" of the HATS. The phone was placed at various distances from the ear. To ensure the feedback path was constant during the measurement process, the mobile phone was held by a microphone stand. The purpose is to examine whether the impulse response of the acoustic feedback path would change by varying the distance between the ear and object. The baffle is placed to simulate the situation when the user stands close to a relatively reflecting surface. The measurement of the effect of the hand was tried to simulate the effect when a user cover his ear by hand. A sofa was also used to check the change of the feedback path in cases like when the user is very close to sound absorbing surfaces like a bed or a sofa.

3.2.5 Results

All the experiments were carried out using the *Oticon*'s shell for ITE devices. The shells supplied were originally unvented. Two holes were drilled on one of shells to emulate the effect of standard vents. One point to note is the acoustical path added is formed by a cavity with two holes. As explained in section 2.1.1.1 this is not a real vent but for practical reasons (and because it is a good approximation) it is referred as a vent in this report.

In this section the main results of the measurements of the feedback path obtained are presented. They are organised in two parts, first the results obtained under the analysis of a unvented hearing aid; second the effect of external acoustic changes on the feedback path.

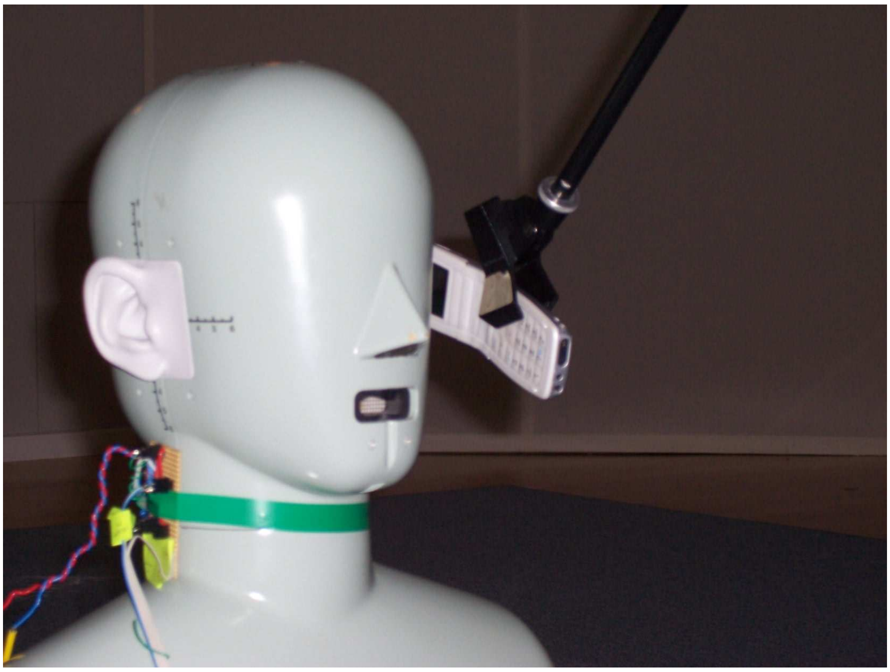

Figure 3.9: Set up for the feedback path measurement when the mobile phone is placed close to an unvented hearing aid.

3.2.5.1 Unvented hearing aid

In this section the feedback measurements obtained with the unmodified hearing aid are presented. The feedback in the unvented hearing aid was due to the leakage of imperfect fitting and structural transmission through the hearing aid only. The impulse response of the feedback path was measured and the corresponding frequency response was obtained. In the frequency response, if the level is above 0 dB, it means that the input voltage level to the MLSSA is greater output voltage level. The MLSSA input and output are amplified by the microphone preamplifier and receiver amplifier respectively. In the frequency range where the response has level greater than 0 dB, feedback is likely to occur. But of course, input and output signals have to be in the phase before feedback can occur.

1. Effect of mobile phone The effect of the presence of the mobile phone next to the hearing aid was investigated. The measurement setup is depicted in figure 3.9.

   Figure 3.10 shows the response of an unvented hearing aid in three different environments. The first curve is the result without any object close to the HATS' pinna. The second curve is the result with the mobile phone placed at a distance of 10 mm from the pinna. And the third curve is the result with the same mobile phone placed in contact with the pinna.

   The results show a clear change of the feedback, mostly in terms of amplitude. The amplitude increases as an object is placed closer to the ear. The shape of the frequency response, however, does not vary dramatically.

2. Effect of the hand

   The next plot in the figure 3.11 shows the responses when no obstacle is present and when the user's hand covering the ear. Again feedback is encouraged when a hand is covering the ear.

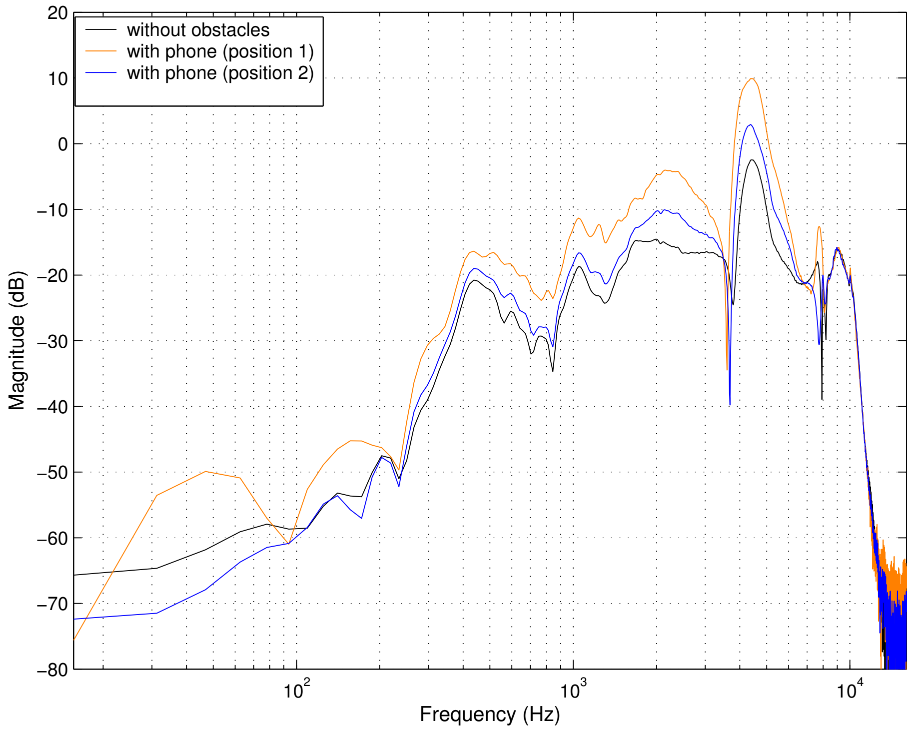

Figure 3.10: Feedback path measurement when the mobile phone is placed close to an unvented hearing aid. Feedback path changes along the distance between device and phone.

Although it is not observable in the figure, the experiment was repeated with small variations in the exact positioning of the hand. The results showed that the frequency response varied with those small changes of the position. The result is useful in simulating the case when the user is adjusting the hearing aid gain or fitting.

3.2.5.2 Vented hearing aid

As explained before, two holes were drilled in one of the available hearing aids in order to emulate the effect of the vent on the real devices. The first hole with diameter of 1mm was drilled next to the receiver. The second hole with diameter of 2 mm was drilled next to the microphone. The impulse response of the feedback path of the vented hearing aid, depicted in figure 3.12, was chosen as the reference feedback path for the simulations performed in MATLAB.

1. Effect of different kind of venting.

   The measurement tried to show the difference in feedback path frequency response for the same hearing aid with no vent, with one hole next to the receiver and with two holes emulating a vent. The results are presented in figure 3.13.

   Definitely feedback is encouraged and its frequency response shape changes significantly. In the case of two holes feedback might occur between the frequencies of 600 and 3500 Hz. The shape demonstrates a good approximation with real vents, as other researchers' measurements present similar shapes [14].

2. Effect of mobile phone

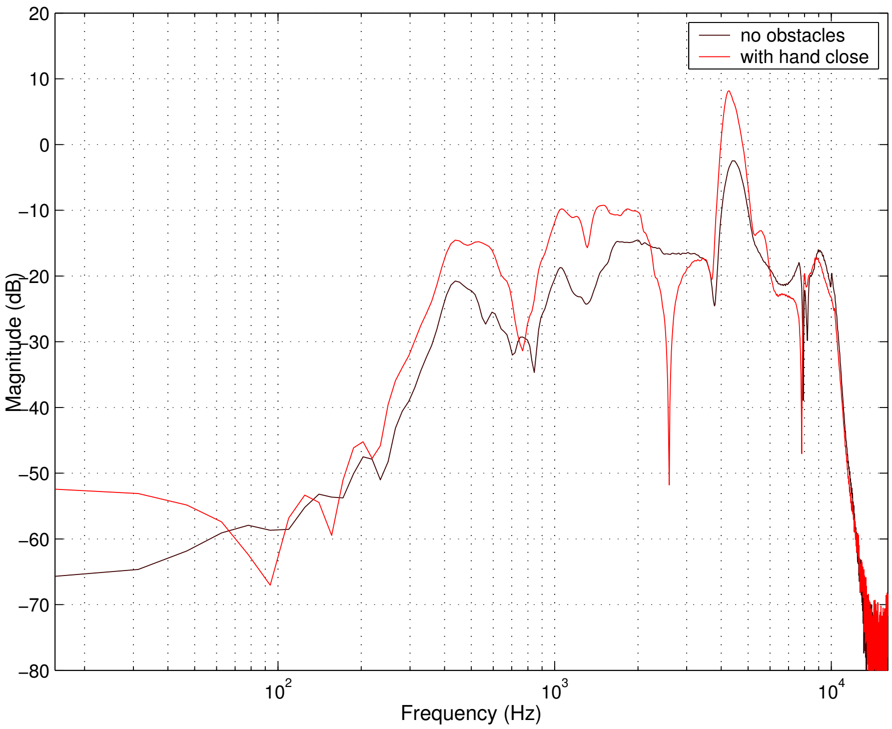

Figure 3.11: Frequency response of the feedback path when the HATS ear is covered by a hand. Feedback is highly encouraged.

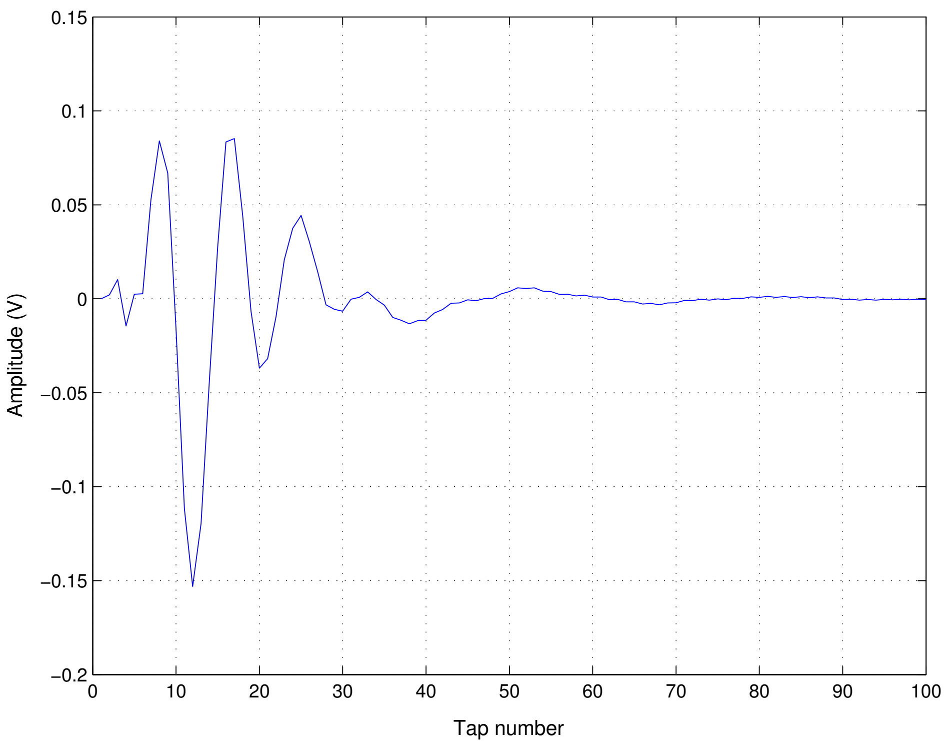

Figure 3.12: Impulse response of the feedback path with a *vented* hearing aid in HATS. This is the impulse response taken as reference for the system simulations.

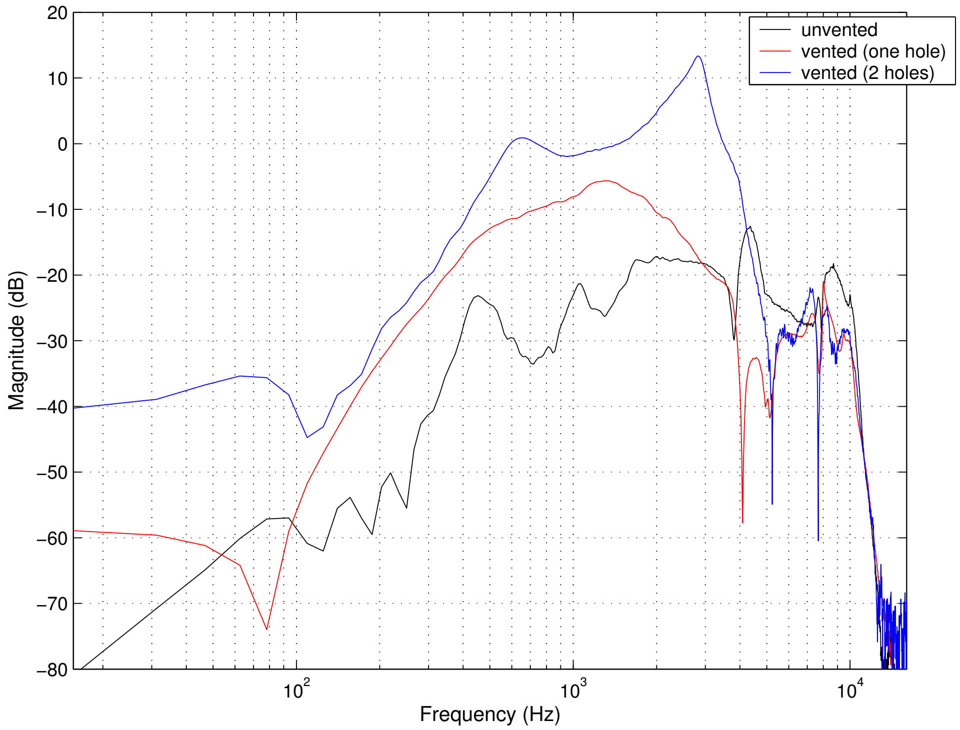

Figure 3.13: Comparison of the feedback path frequency response between an unvented HA, a hearing aid with one hole next to the receiver and one with two holes, the second next to the microphone.

The measurement was similar to the one performed with the unvented hearing aid except in this case a vented hearing aid was used. The frequency response of the feedback path shifted up by placing the same mobile phone close to the ear, as clearly shown in figure 3.14. The changes in frequency response were small when compared to the results in the unvented hearing aid case.

3. Effect of hand, sofa or sound reflective surfaces

   Real situations such as when the user stands next to a wall (sound reflective surface), sits on a sofa, lies on a bed (non-reflective surfaces), or tries to adjust the hearing aid gain by placing his hand next to the ear were also simulated.

   The results showed that if the objects were placed further than around 10 cm from the pinna, their influences were insignificant, even when the surface is relatively sound reflective. It was shown that users will not suffer from the proximity effect of a wall if they don't put the ear deliberately close to the wall, as usually the shoulders will not allow the wall to be closer than 15 cm.

   As it can be seen in figure 3.15, when the HATS is lying on an sound absorbing surface such as a bed or sofa, the effect is still very small even though the head is in contact with that surface. The feedback is only slightly encouraged below a certain frequency due to the acoustic nature of textile materials, the materials absorb a lot at high frequencies. In the same plot the response when the subject puts his hand next to the ear was shown. Feedback is not only highly encouraged in certain frequencies, also its frequency response's shape was changed. The shape will be different for different positions of the hand.

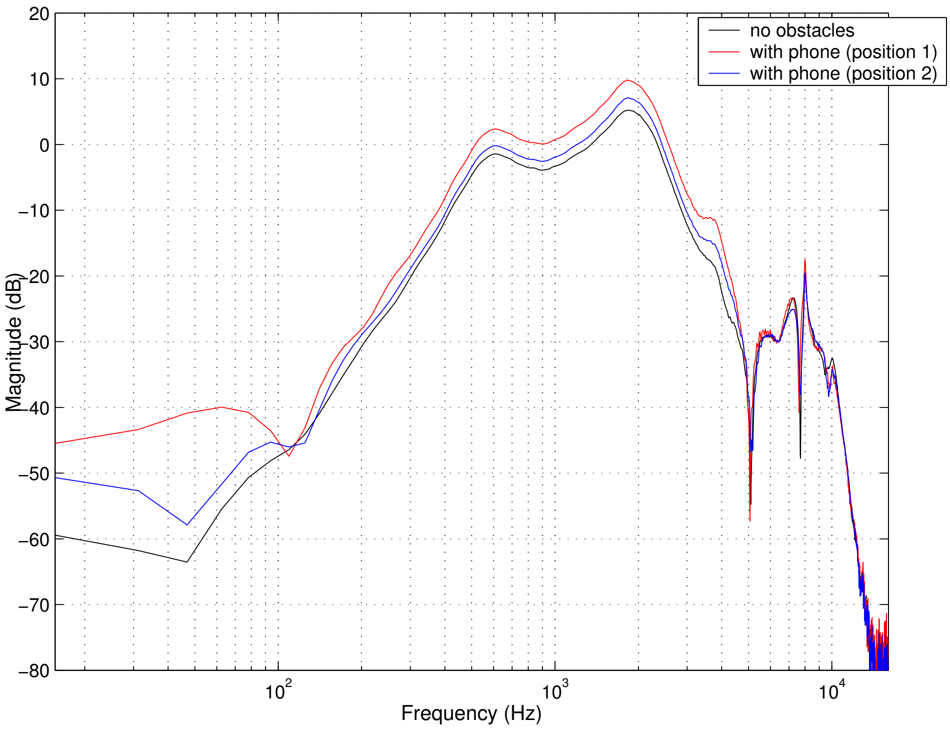

Figure 3.14: Feedback path measurement when the mobile phone is placed close to a vented hearing aid. Feedback path changes along the distance between device and phone.

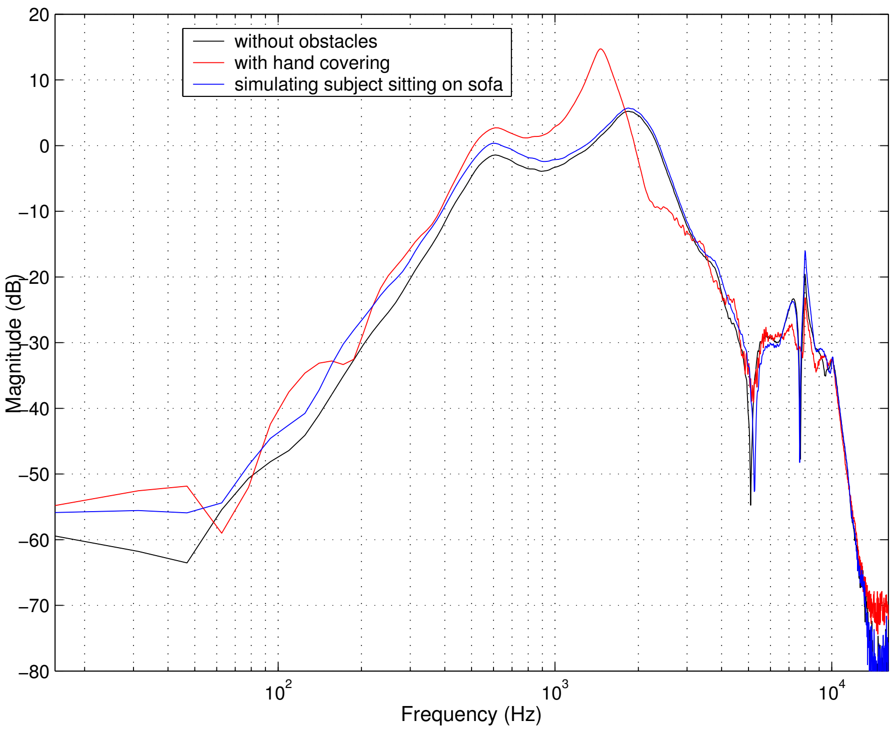

Figure 3.15: Feedback path of a vented hearing aid simulating when a subject is sitting on a sofa and when the user covers his ear with a hand.

3.2.6 Conclusions

Based on the results presented in previous section important conclusions could be drawn:

- Every time when the hearing aid (HA) was refitted to perform new measurements the feedback path response showed small changes even though there was no variation in the external acoustical environment. In the set up used (HATS), the fitting was very good as the physical dimensions of the ear canal in the head and torso simulator allow a very tight fitting, but little variations could still be found. So it is expected that in real users for which the fitting is usually less perfect, the changes could be more noticeable.
- When there is no vent feedback might appear in ITE hearing aids. The most vulnerable feedback frequency range is between 2000 and 4000 Hz for the hearing aid under analysis.
- The vent does encourage feedback in all frequencies up to 4000 Hz. The frequency range where the feedback could occur is lowered down a bit but widened in a noticeable extent.
- The vent smooths the frequency response in its fine structure.
- Little changes in the acoustic environment close to the ear will result in significant changes of the feedback path. This result shows it is necessary to design a cancellation algorithm with fast tracking capability. On the other hand, acoustical environment changes not in close proximity to the hearing aid have a less critical effect on the feedback path.

3.3 Design simulation

Several custom made Matlab functions (see Appendix D) were developed in order to test the performance of the algorithms to be implemented. The feedback cancellation approach is based on the feedback path estimation method (see section 2.1.2). Adaptive feedback cancellation (AFC) algorithms were designed based on the LMS algorithm. LMS based algorithms were preferred because of its simplicity and reduced number of computations required compared to algorithms such as RLS. The normalized version, NLMS, was used in order to avoid gradient noise amplification (see section 2.2.2).

3.3.1 General algorithm description

Two algorithms were developed in order to simulate a hearing aid system. The first algorithm simulates the basic diagram as shown in figure 3.16. This is the basic version without any decorrelation method. As it was demonstrated in section 2.3.3, it would not be possible for this first algorithm to converge with a speech input signal. The next section shows the method chosen to solve the correlation problem.

3.3.1.1 Solving the correlation problem

Delay based methods have been proposed to decorrelate the input and feedback signals in hearing aids [4], [5], [6], as mentioned also in section 2.3.3. For speech input signals, a suitable delay value

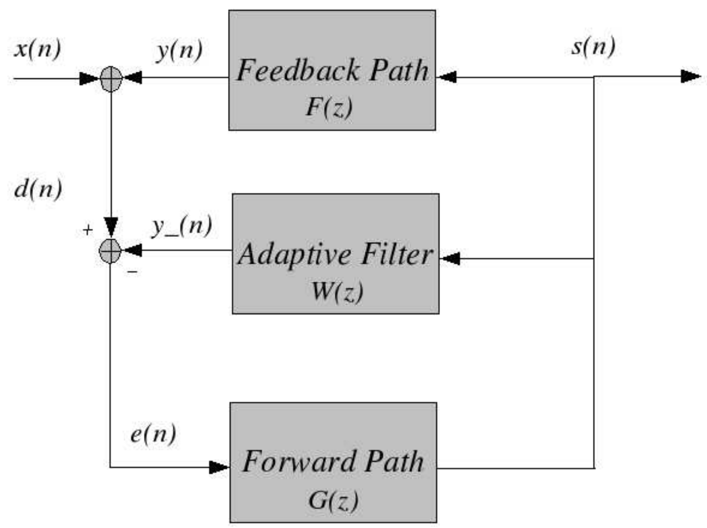

Figure 3.16: Basic feedback cancellation diagram

should be determined so as to minimize the correlation between the input signal and feedback signal. To determine that value it is necessary to examine the autocorrelation of a speech signal. Figure 3.17 shows a typical autocorrelation sequence of speech. A female voice was recorded at sampling rate of 16 kHz for 20 seconds and the normalized autocorrelation sequence of this signal was plotted.

It can be seen that the autocorrelation shows a rapid decay after one ms, and no significant reduction is obtained with a higher time lag. Therefore, it can be said that by adding a suitable delay between the input and output can decorrelate the input and feedback signals to some extent[3].

Whether the delay is placed in the forward path or the cancellation path has been a subject of study [6], [7]. These previous works have shown that introducing delays in both the forward and cancellation paths can partially decorrelate the signals, but introducing delays in the forward path is more beneficial. This is due to the fact that the bias in the estimation of the feedback path will depend on the forward path characteristics if the cancellation path is delayed [4]. Figure 3.18 shows a block diagram of the feedback cancellation system depicted in 3.16, but with a delay in the forward path.

In figures 3.16 and 3.18, the signal *d(n)* is the signal from the microphone. This signal is the sum of the input signal *x(n)* (e.g., speech) and the feedback signal *y(n)*. The output signal *s(n)* is the signal fed to the receiver, and it is equal to the filtered error signals through the forward path frequency response *G(z)*. In the simulations, the forward path was modeled as a all pass zero phase system with an arbitrary gain. In both figures 3.16 and 3.18, *F(z)* represents the feedback path frequency

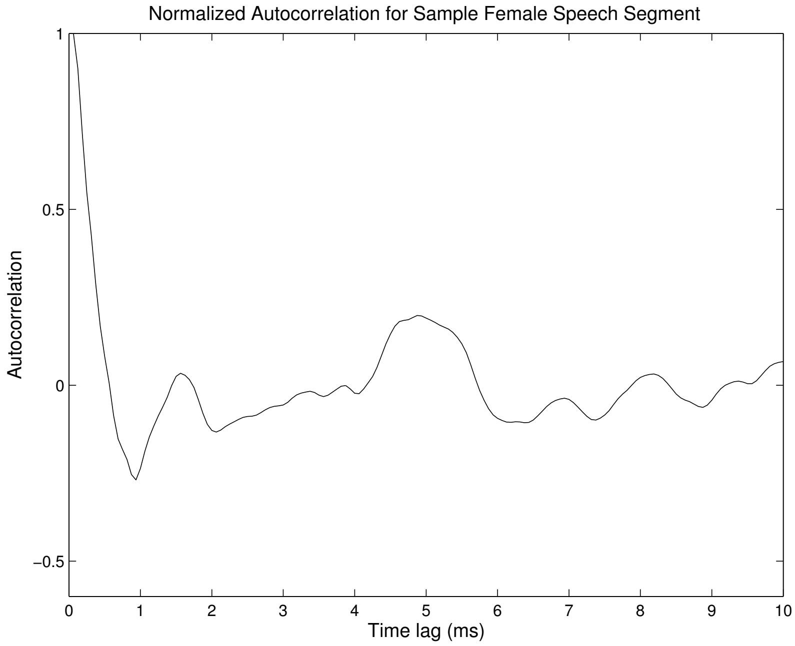

Figure 3.17: Normalized Autocorrelation sequence for a 20 s speech signal

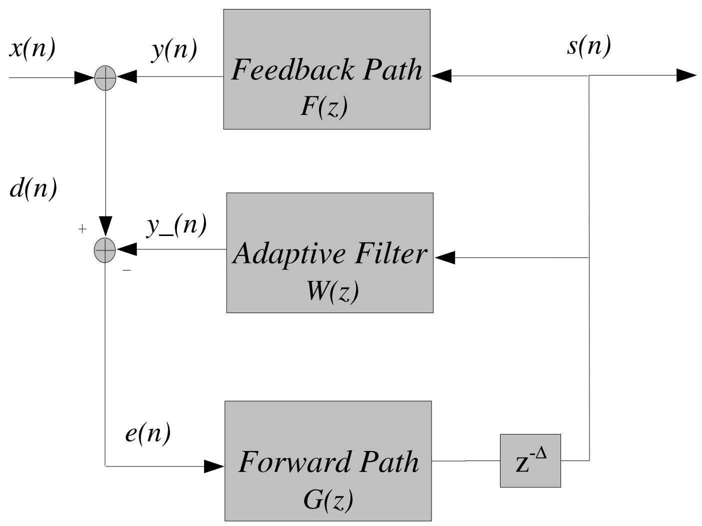

Figure 3.18: Feedback cancellation diagram with decorrelation delay

response. The frequency response covers the whole electroacoustical transfer function including receiver, microphone and acoustical feedback path. This impulse response was obtained through measurements (see section 3.2.5) and was used in the simulations. The measured data shows that a 100 tap impulse response was long enough to describe the feedback paths. For both algorithms, the adaptive filter estimates the feedback signal *y(n)*, then this estimated feedback signal was subtracted from the input. The flow chart for the algorithm is shown in figure 3.19.

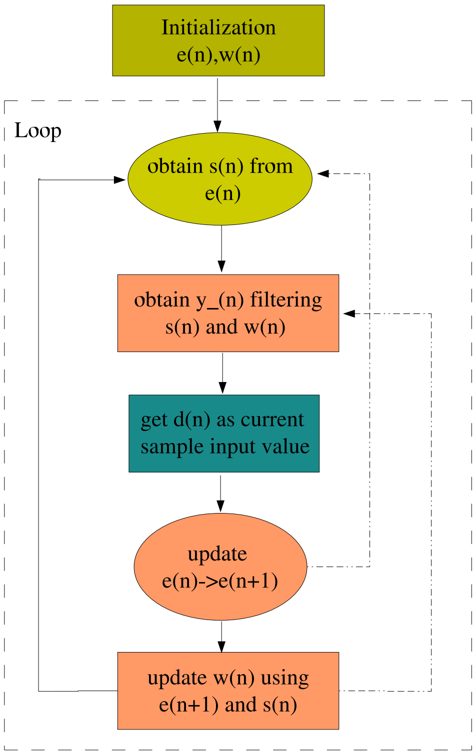

Figure 3.19: Flow chart of the AFC algorithms with and without delay

In this flow chart, during initialization, the first estimated coefficients *w(n)* are set to zero and the first M (M=100) error signals are set to the first M input signal samples. Then, an infinite loop runs performing the NLMS algorithm based on the input signal *s(n)* , and the current input *d(n)*. The input signal *s(n)* is formed from e(n) and this e(n) has been updated by the new error signals *e(n+1)*. The difference between the two algorithms is that in the first algorithm the input signal to the receiver *s(n)* is obtained from *e(n)* directly, while in the second version the input signal to the receiver *s(n)* is obtained from a delayed version of *e(n)*.

3.3.2 Simulation of the basic hearing aid system

The first algorithm that simulates the basic system as shown in figure 3.16 was tested with two different input signals, a white noise signal and a speech signal. The speech signal was a 3 second long female voice with autocorrelation function as shown in figure 3.20.

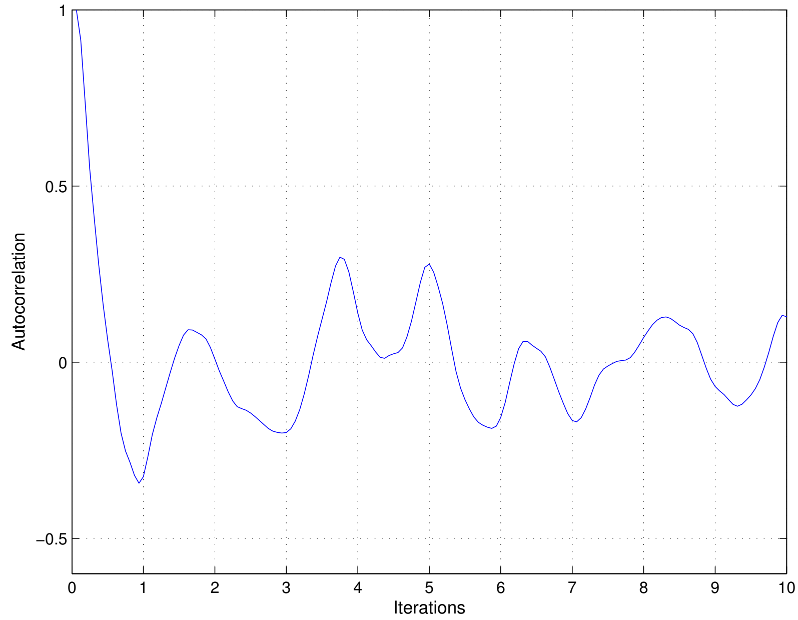

Figure 3.20: Normalized autocorrelation for the speech signal used in the simulations

The feedback path to be estimated is a 100 tap impulse response obtained experimentally for the ITE hearing aid with a *vent*. This impulse response is shown in figure 3.21.

In this simulation the feedback path was stationary. The forward path gain was set to four, simulating some reinforcing gain put to the incoming sound. The results are given in terms of the *Weighted Error Vector Norm(WEVN)*, which is also known as misalignment, as defined in section 2.3.4. A high negative dB value in the WEVN means the estimation is accurate. On the other hand, a positive dB value indicates a high bias or a high inaccuracy in the estimation. Figure 3.22 shows the results.

As expected, only for the case with the white noise input the estimation is accurate enough to make the WEVN converge to a negative dB value. For a speech signal, the correlation between input and feedback signals produces a high bias in the estimation. The number of iterations performed can be seen as a measure of the speed of the cancellation, provided that there is one iteration for every input sample. Setting a very small value for the stepsize μ, gives a more accurate estimation but slower convergence. Higher values for the stepsize produce faster convergence, but the estimation is not as accurate. The stepsize was set to 0.01 for this simulation. This stepsize value was chosen as a suitable tradeoff between convergence speed and accuracy.

3.3.3 Simulation of the hearing aid system with a decorrelation delay

The algorithm that uses a decorrelation delay in the forward path was tested for a speech input signal and compared to the one without decorrelation method. The forward path was delayed by 1.3 milliseconds, which corresponds to 20 samples at a sample rate of 16 kHz. As in the previous case, the feedback path was fixed and obtained from the same measurement. The forward path

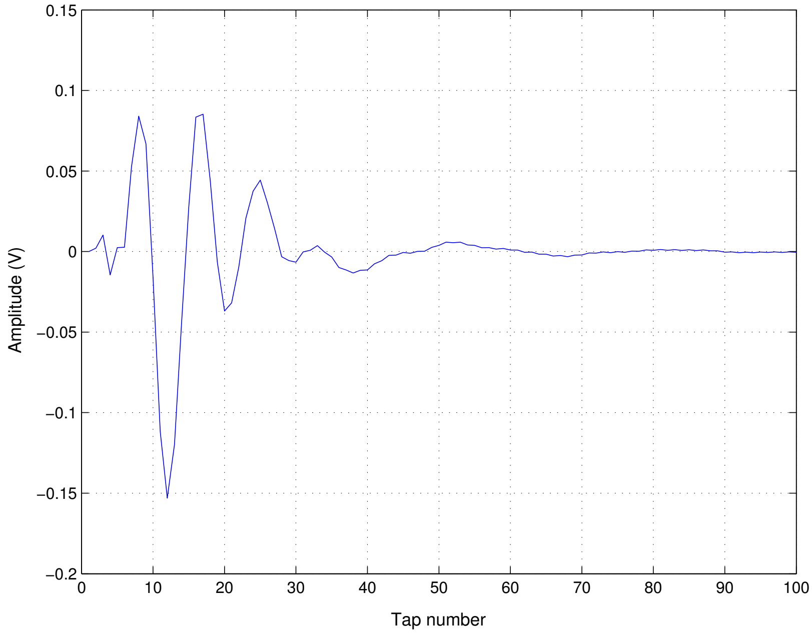

Figure 3.21: Feedback path impulse response to be estimated

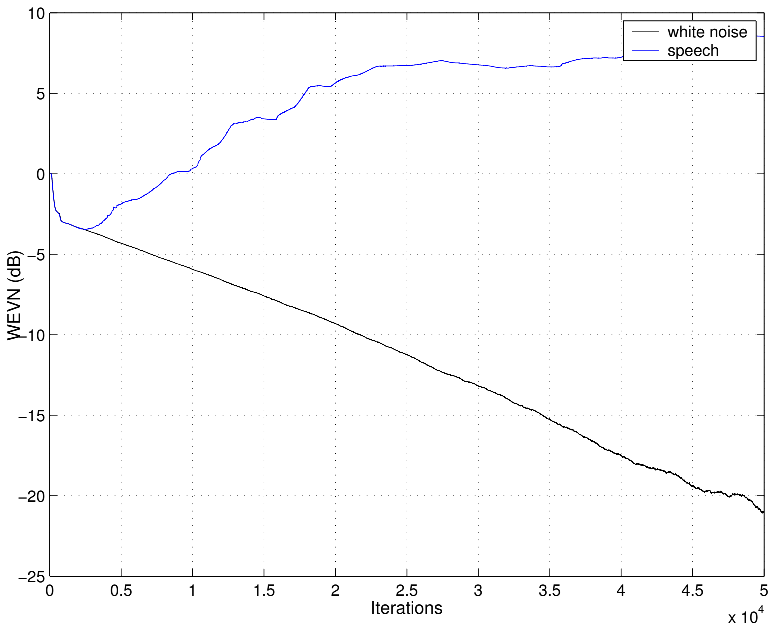

Figure 3.22: Misalignment of the basic AFC for a white noise and a speech signal input

gain and the stepsize were unchanged. From now on the gain and the stepsize will be kept fixed for the next simulations. The results are shown in figure 3.23.

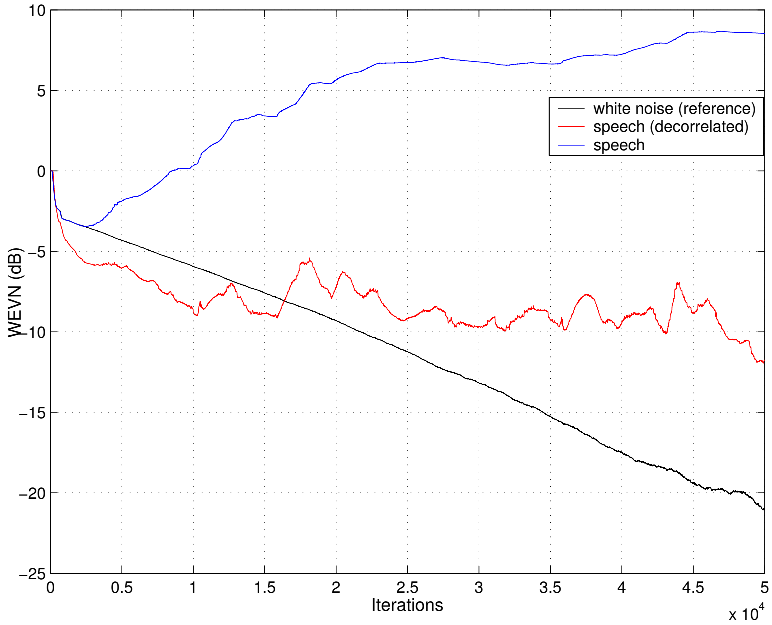

Figure 3.23: Misalignment of the AFC with a speech input signal with and without decorrelation delay

For the ease of comparison, the WEVN curve for a white noise input is plotted as well. It can be seen that there is a significant improvement in the estimation when the forward path is delayed. The input and feedback signals are successfully decorrelated with a minimum extra complexity added to the algorithm.

Figure 3.24 shows the WEVN for different gains in the forward path. At higher gains a better estimation can be achieved. These results are in concordance with the ones shown in IEEE Transactions on Speech and Audio Processing[4].

3.3.4 Simulation of the hearing aid system with a decorrelation delay and a variable feedback path

To further evaluate the behavior of the system in more realistic situations a variable feedback path was simulated. The variation in feedback path was made by updating the feedback path impulse response after a certain number of iterations in the algorithm. It has to be noted that in the simulation the changes in feedback path are abrupt while in real situations the changes are continuous. However, the sudden changes in the acoustical environment make it even more difficult for the algorithm to adapt. The simulation tries to mimic the situation when a user places a mobile phone at different distances from his ear. The feedback path impulse responses used in the simulation are taken from the measurements performed described in section 3.2.5. The impulse responses used and their effective iteration range are shown in the table 3.4

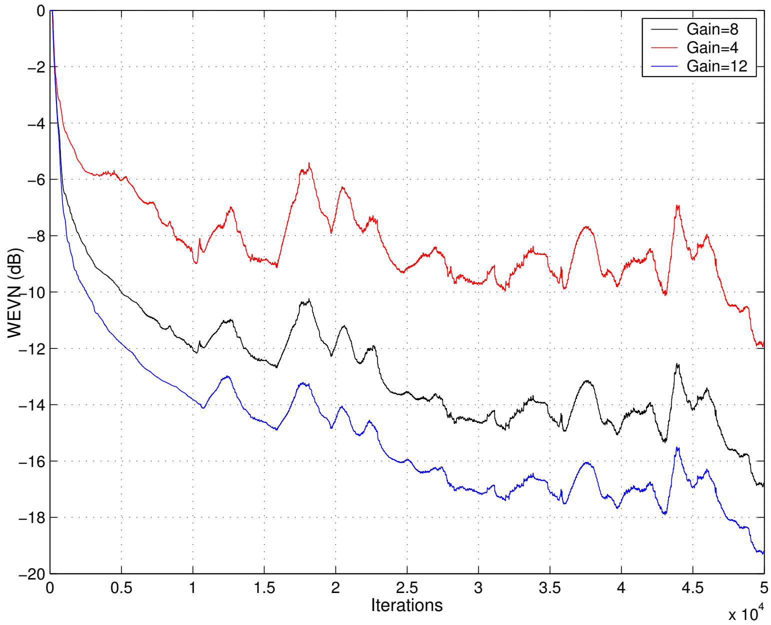

Figure 3.24: Misalignment of AFC for a decorrelated speech input signal at different gains

Situation	**Effective iteration range**
no obstacle near the ear	0 to 20000
mobile phone in contact with the ear	20001 to 40000
mobile phone 10mm from the ear	40001 to 60000

Table 3.4: Variation of impulse responses with respect to iterations range

The feedback paths iteration domains are labeled from FP1 to FP3. It can be seen that the algorithm adapts to the changes in the feedback path in a good fashion. The misalignment is only slightly increased after the feedback path is updated and it remains at a sufficiently low level.

3.3.5 Estimation of added stable gain for the hearing aid with the proposed AFC

In this section, an example of how the proposed AFC can provide the hearing aid with more gain without oscillation is made. The added stable gain (ASG), as defined in section 2.3.4, was obtained in a simulation using a given feedback path measurement. This gain will depend not only on the feedback path frequency response, but also on how accurate the estimation of the AFC is. The simulation result is considered as an example, because in the final test of the system implemented, another measurement of the feedback path would be required. This feedback path may differ in some extent to the one used in this simulation. The electroacoustical feedback path of a vented hearing aid was used and its frequency response is plotted in figure 3.26.

For every iteration, the coefficients of the adaptive filter and hence the filter's frequency response was updated. By using the equation 2.29 and the frequency response of each iteration, the ASG

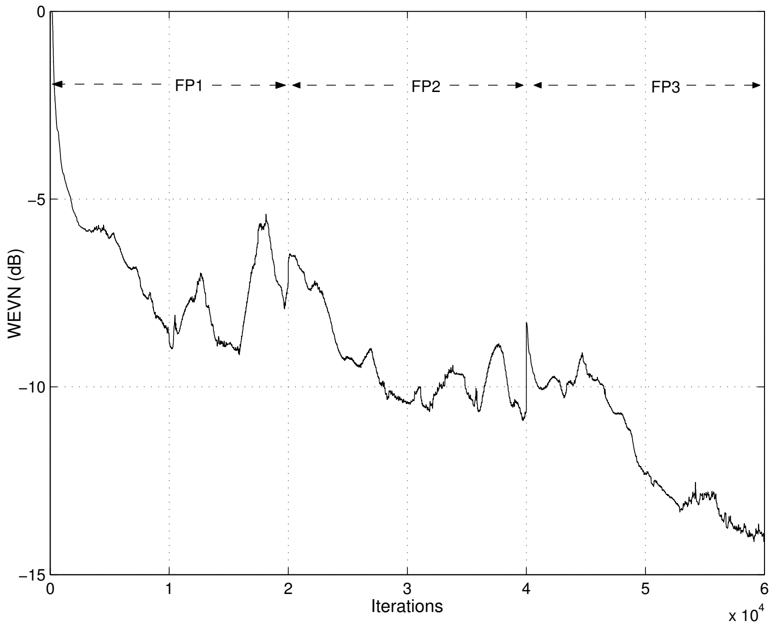

Figure 3.25: Misalignment of the AFC for a decorrelated speech signal and variable feedback path

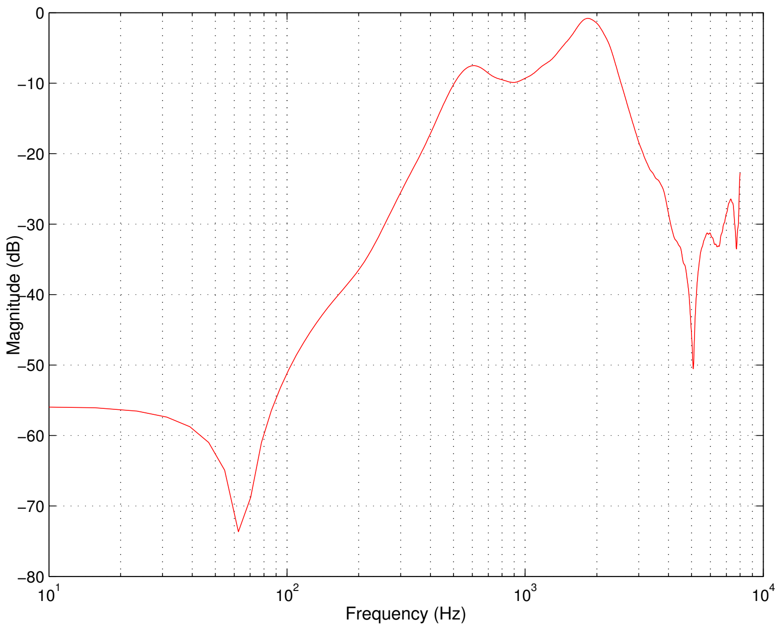

Figure 3.26: Frequency response of the feedback path used for calculating the ASG

curve was obtained and the result is shown in figure 3.27.

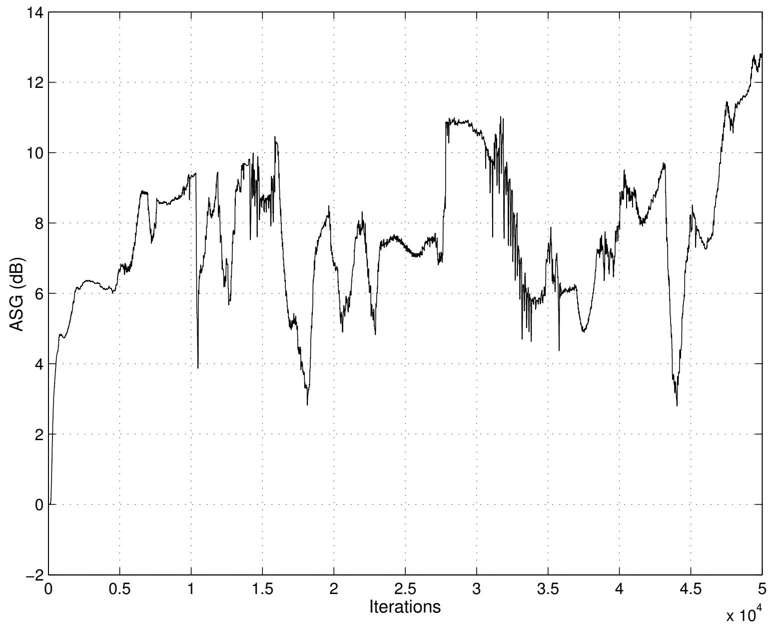

Figure 3.27: Added stable gain for speech input signal

The ASG is positive throughout the iterations, which means that an additional gain without oscillation would be obtained with the AFC. Because of the time varying characteristics of a speech signal, the ASG changes approximately between 3 and 12 dB for the speech segment considered. In order to observe how the ASG varies in a longer segment, with the same settings, a 20 second speech input signal was used and the corresponding ASG obtained is shown in figure 3.28.

A variation in ASG between approximately 2 and 16 dB is observed in general. Because of the changes in ASG values with speech input signals, an additional simulation of the ASG under white noise input is needed as well.

Figure 3.29 shows the steady state ASG for a white noise input signal. An ASG of around 20 dB was achieved by the AFC for a white noise input signal. Although the value seems to be relatively high, laboratory measures have indicated that 20 dB of added gain without oscillation can be achieved [19].

3.3.6 Simulation of AFC with two subbands

This section describes the simulation results of the subband AFC study. This approach was not implemented, but simulations were carried out to observe how a subband AFC behaves. The subband AFC approach explored in the analysis section was simulated in order to see how the performance of the AFC is changed when the input signal is filtered and separated into several subbands. The separation of the input into subbands allows the AFC to act on each of the subbands with lower complexity. For simplicity, a subband structure of two bands was used (see figure

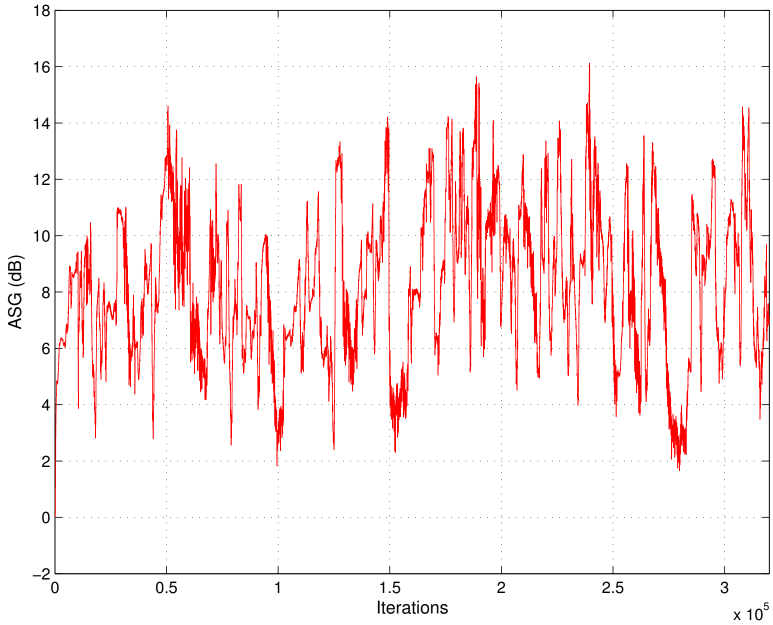

Figure 3.28: Added stable gain for 20 second speech input signal

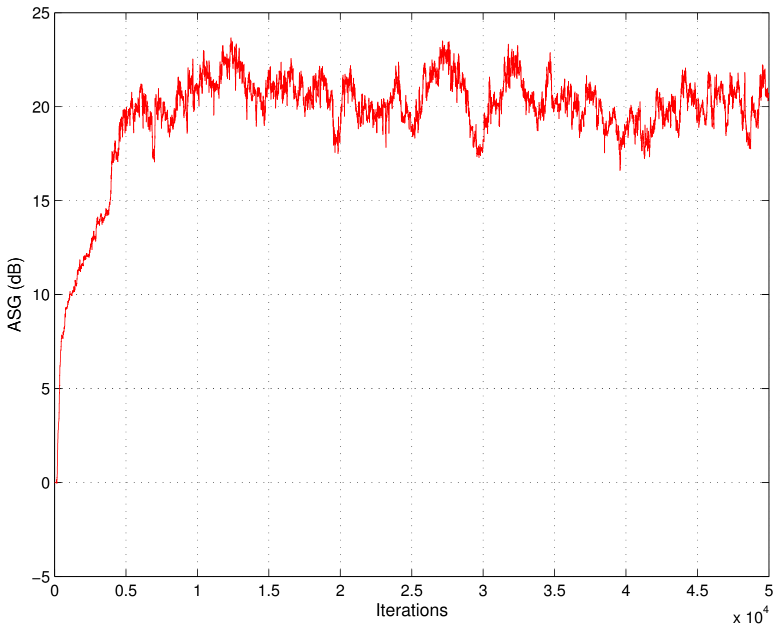

Figure 3.29: Added stable gain for white noise input signal

2.5). A MATLAB function was developed to simulated this (see Appendix D). The steps in the MATLAB function are:

- The signals d(n) and s(n) are filtered by two analysis quadrature mirror filter (QMF).
- The signals are downsampled by a factor of two.
- Two adaptive filters run on each of the bands.
- The signals are upsampled by a factor of two.
- The signals are filtered by the two synthesis QMF.

Suitable values for the constants α and forgetting factor λ (see equations 2.16 to 2.19) were chosen experimentally after considering recommendations given in the reference [8].

The analysis and synthesis QMF used are halfband FIR filters of order 32. The relationship between them is described by equations 2.14 and 2.15. The forward path gain was set to 4, to simulate some reinforcement to the incoming sound and keep it the same as in the wideband approach.

At first, the algorithm was tested for a white noise input of 50000 samples and was compared to the wideband algorithm. The results are given in terms of the misalignment (WEVN). Figure 3.30 shows the results.

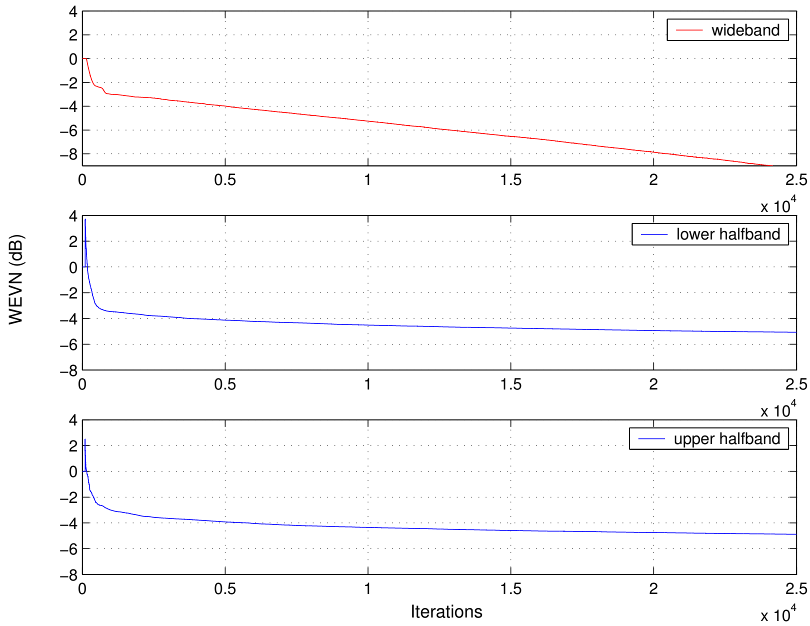

Figure 3.30: Misalignment of the AFC for 2 subbands

For the ease of comparison the wideband AFC is shown up to iteration 25000 in the first plot in figure 3.30. It has to be noted that the wideband AFC actually needs 50000 iterations to process the 50000 samples set in the simulation. However, as shown in the second and third plot, each AFC running in parallel needs only 25000 iterations to process the whole input signal. From this simulation, the convergence of the subband AFC can be considered to be faster than that for the wideband case. However, the wideband algorithm gives more accurate results. The WEVN was found to give much lower dB values than the subband AFC when all the 50000 iterations were observed. This results are in agreement with the observations done in M.G. Siqueira et al. paper[8].

The algorithm was also tested with a speech input signal. The results are shown in figure 3.31.

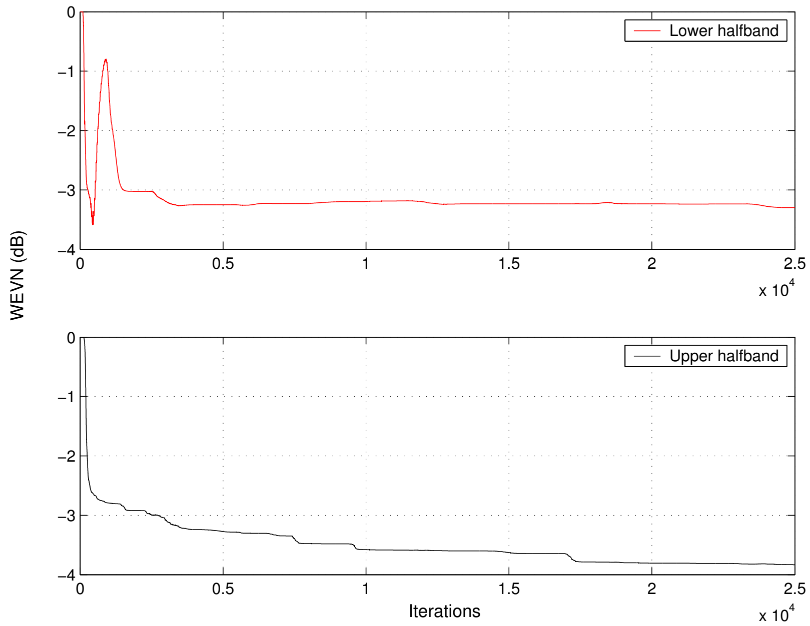

Figure 3.31: Misalignment of the AFC for two subbands for a speech input signal

It is interesting to note that for the subband AFC there is no high bias in the misalignment for a speech input signal as it was observed in the wideband case with no decorrelation method applied. The possible reason for this is that the combined group delay of the analysis and synthesis filters already decorrelate the input signal from the feedback signal to some extent. It was seen that with greater order of the QMF chosen, the misalignment is improved to some extent.

Chapter 4

Implementation

Because of the short input output latency requirement of the adaptive feedback canceller system, the option for implementing the system in the *SIMULINK* environment had to be discarded, as in this environment the latency between the input and output was greater than 100 ms. Therefore, it would not be possible to reproduce the real life situation where input and output are correlated due to the short delay.

Therefore, a digital signal processor (DSP TMS320C50) is considered as the best platform to implement the adaptive filter algorithm. In fact, commercial digital hearing aids are using special digital signal processing hardware to handle their hearing aid functions, such as feedback cancellation adaptive filters and intelligent noise reduction.

4.1 General implementation considerations

Before implementing the adaptive feedback canceller, general implementations considerations are presented, thoughts that will lead to a better understanding of the underlying choices in the algorithm implemented.

4.1.1 Overflow and scaling

One of the potential problems of using a fixed point DSP is overflow. To understand the overflow problem, it is necessary to understand the internal representations of positive and negative numbers in the DSP. Most DSP's use two's complement as the internal representation of numbers. Under two's complement representations with 16-bit register, the maximum allowable value is 32767 and the minimum allowable value is -32768. The two's complement binary values are shown in the table 4.1.

In multiply and add operation the value can become so large that it exceeds the maximum allowable range. As seen from table 4.1, if a number is added to the positive maximum, the value will become a negative number. This situation is known as overflow. For instance, if one was added to the 32767, the result would become -32768.

To prevent overflow, care must be taken. One method is scaling down the input by a certain factor,

Decimal number	**Two's complement representation**
32767	0111 1111 1111 1111
-32768	1000 0000 0000 0000

Table 4.1: Decimal representation of the maximum and minimum number in two's complement with a 16 bit representation.

and scale up by the same factor at the final output. A proper scaling factor can minimize the loss in the signal to noise ratio without causing the DSP overflowing. If the input is scaled down too much, the signal to noise ratio will decrease such that most of the input precision is lost. Furthermore, if the input is not scaled overflow may occur.

Representing the numbers in such a way that calculations are only performed on numbers less than one, meaning working on fractional numbers, can prevent overflow. The multiplication of two numbers less than one will always be less than one. For addition of two numbers with their magnitude less than one, it is not sure whether the result would be greater than one or not. One method to tackle the problem is the use of conditional shifting. The DSP has a branch on overflow assembly language statement and could therefore branch to a state in the algorithm performing shifting of the numbers.

4.1.2 Representing floating point numbers in fixed point

To store the required range of numbers, including fractional parts of a number on a fixed point DSP, the designer can introduce a decimal point, such that a number of bits are representing the fraction of the number.

A format named the $Qformat$ is used to describe this representation.

For instance $Qm.n$ means a representation where m is the number of bits to represent the integer part and n is the number of bits to represent the fractional part. In case of a 16 bit word storage the sum $m+n$ is 15, with the extra one bit for determining the sign of the number.

Some trade-off has to be made between the number of bits in representing the integer part and that of the fractional part. On one hand the designer may want to allocate as many bits to the integer part as possible to accommodate the largest magnitude number, and on the other hand the designer will also want to allocate as many bits in the fractional part as possible to preserve the precision. However, the sum of integer bits and fractional bits is only 15 (the total number of bits for representing af number is $16-1$, since one bit is used as the sign bit in two's complement).

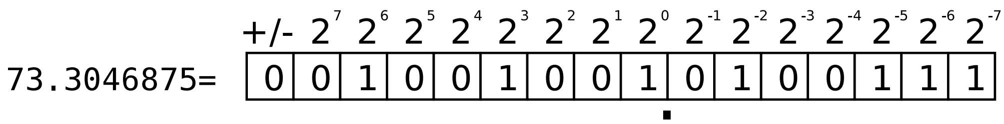

Figure 4.1: Floating point number represented in fixed point environment

Figure 4.1 shows the floating point number 73.3046875 stored as the fixed point number 9383 with a $Q9.7$ representations, meaning 7 bits are representing the fraction. Figure 4.2 shows three example Q formats and their respective positive and negative values.

format	minimum value	maximum value
Q15.0	-32768	32767
Q1.14	-2	1.999939
Q0.15	-1	0.999969

Table 4.2: Maximum and minimum values under different Q formats

4.1.2.1 Arithmetics under Q representation

Regarding arithmetic operations on Q represented numbers, the designer has to consider the following. Addition and subtraction can be performed directly as long as the two numbers are in the same Q format. One point to note is, that if two $Qm.n$ numbers are added or subtracted, the sum can end up to the $Qm+1.n$ format.

Multiplications of two Q represented numbers ($Qm_1.n_1$, $Qm_2.n_2$) can be performed regardless of the Q format. The effect of multiplication regarding to the decimal point is seen in the final result, meaning that the Q representation will become $Qm_1+m_2.n_1+n_2$.

So finally, considering a scenario comprising multiplications and addition/subtraction, the designer properly has to introduce shifting of a number to fulfill the criteria stated above.

4.2 Adaptive feedback canceller implementation

An assembly language program was developed for the TMS320C50 DSP. All the details mentioned in previous sections have been considered. Based on the design, the implementation is an adaptive feedback cancellation (AFC) filter using LMS calculation. The AFC does not comprise any decorrelation between the input and output, hence no delays in the forward path. The flowchart of the algorithm is depicted in figure 4.2.

The algorithm is divided into following routines.

`INITBUF` Setting up the two circular buffers.

`INITVAR` Setting up important variables, such as filter length and step-size, before running the main routine.

`FILLUP` Preparing the buffers by filling up the error vector buffer with pure samples from the A/D converter, before running the main adaptive process.

`MAIN` Main program, running the LMS routine by performing filtering, error calculation, update of error vector and update of coefficients. Furthermore the main routine is fetching new samples from the converter and sending the output from the adaptive process to the converter.

`FINIT` Preparatory steps before filtering, such as setting up the filtering order.

`FIR` The filtering process, multiplying the filter coefficients with the input samples.

`ERROR` Calculating the difference between the estimate $y\_(n)$ and the desired response $d(n)$.

`UPDATE` Updating the coefficients based on the step-size, current error and error vector product.

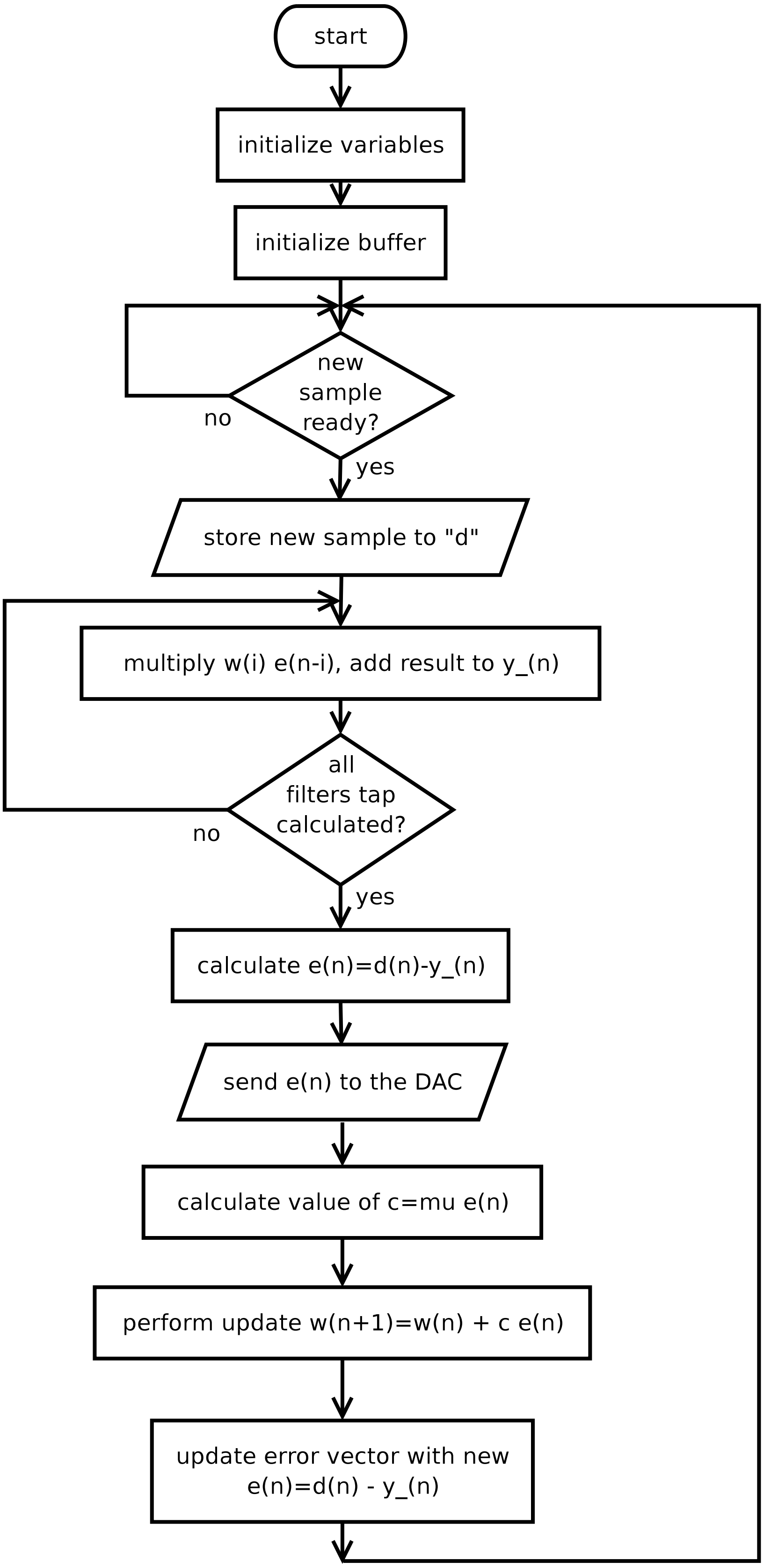

Figure 4.2: Flowchart showing the the algorithm performing the LMS calculations

Before doing any adaptation, the system is first initialized by setting up the circular buffers and declaring important variables like filtering order and the analog-digital converter interrupt routine.

Two circular buffers are used. The DSP offers logic managing the circular buffers after setup. To visualize the situation, the reader can think of a moving window applied on a input sequence as shown in figure 4.3a). The window is shifted to the right to calculate a new output value. The sector is shifted clockwise for calculating successive outputs. The circular buffers are set up by the code 4.1

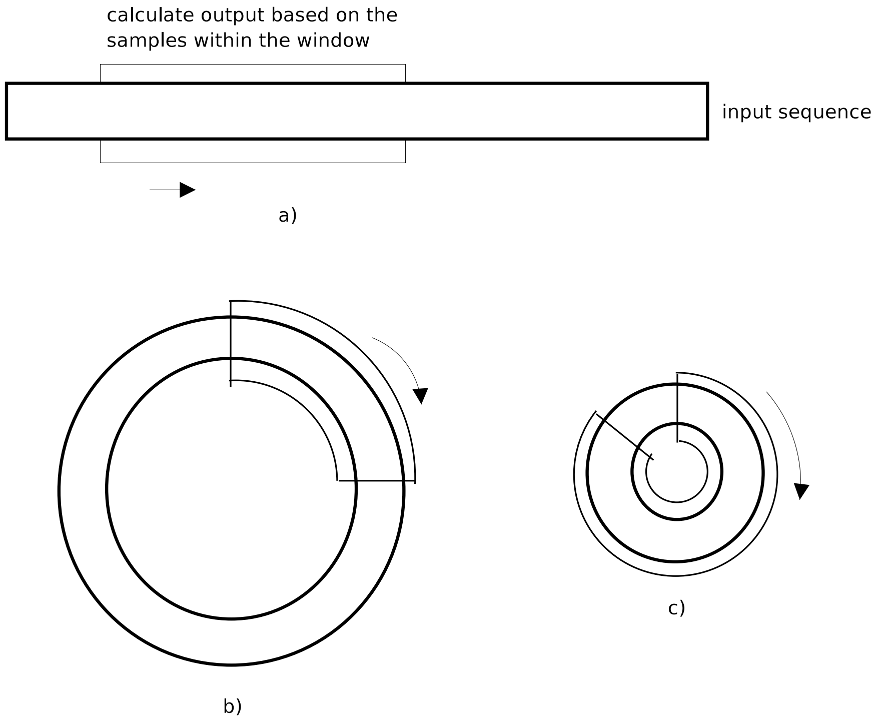

Figure 4.3: Linear buffer to circular buffer

Circular buffer 1 is used for the error vector and circular buffer 2 for the coefficient vector. Circular buffer 1 is set up in a decrementing fashion compared to the circular buffer 2 which is in an incrementing fashion. The difference in direction is due to the calculation of the convolution sum in the filtering.

The adaptive process is based on the interrupt from the A/D converter. Therefore, whenever a new sample is ready for processing, the adaptive algorithm is executed, performing one iteration of the adaptation.

The adaptation comprises three steps

- Filtering (`FINIT` and `FIR`)

- Error calculation (`ERROR`)

- Coefficient update (`UPDATE`)

Code 4.1 Circular buffer set up - two buffers used

```
INITBUF
        splk    #0feh,CBCR      ; Set up AR6 and AR7 as pointers
                                ; for the circular buffers
; * Circular buffer #1 *
        LACC    #09FFh          ; Load 9FFh into ACC
        SAMM    CBSR1           ; Set CB1 start at AFFh
                                ; by storing value to the CBSR1 register
        SAMM    AR6             ; SET AR6 register with 9FFh
        SUB     #63             ; Subtract 63 from ACC
                                ; Filterlength is 64
        SAMM    CBER1           ; Store CB1 register with 9C0h


; * Circular buffer #2 *
        LACC    #0A80h          ; Same procedure here
        SAMM    CBSR2           ;
        SAMM    AR7
        ADD     #63
        SAMM    CBER2
```

4.2.1 Filtering

The filter is implemented as a FIR filter. A typical FIR filter equation is shown in equation 4.1.

$$y(n) = \sum_{i=0}^{N-1} w_i(n)e(n-i) \tag{4.1}$$

Where n is at the current time instance and N is the filter order.

Code 4.2 present the implementation of the filtering.

To keep full precision (32 bit) in each multiplication and addition, during the filtering, the accumulator B in the DSP is used. The final sum is then stored in 16 bit. The reason for keeping 32 bit precision is to avoid truncation after each multiplication.

4.2.2 Error calculation

The error is calculated on behalf of the difference between the desired response and the estimation y_(n) from the filtering process. The piece of code performing the operation is shown in code 4.3. The reason for the negating of the y_(n) is only due to the previous result from the filtering. The convolution sum y_(n) is still in the accumulator and therefore simply negated and followed by an addition of the value d(n).

4.2.3 Update of the coefficients

Finally the update of coefficients is performed. The consideration here is to calculate the product of the step-size and the newest error ($e(n) = d(n) - y\_(n)$) first and secondly do the multiplication

Code 4.2

```
FIR
; * Coefficients *
        LDP     #0                          ; Load page 0
        MAR     *,AR7                       ; Ind. addressing, using AR7
        LT      *+                          ; Load coefficient w(i) to TREG
                                            ; Increment pointer
; * Error vector *
        LDP     #0                          ; Load page 0
        MAR     *,AR6
        MPY     *-                          ; Product in PREG

        PAC                                 ; Load ACC with PREG
        ADDB                                ; Add ACCB to ACC
        SACB                                ; Store ACC in ACCB
        .
        .
 * Condition check   *
 * Filtering done?   *

        .
        SACH    y                           ; Store final sum
                                            ; ACC high to y
```

Code 4.3

```
ERROR   NEG                                 ; -y(n)
        ADDH    d                           ; Add d(n)
        SACH    err                         ; Store to "err"
```

by each element in the error vector ($\hat{e}(n)$) and updating each element in the coefficient vector.

Code 4.4

```
UPDATE  LDP     #20             ; Select data page 20
        LACL    order           ; Load floops into ACC
        SACL    q               ; Store that to variable q
        LT      err             ; Load TREG with "err"
        MPY     u               ; Multiply by "u"
        PAC                     ; Store PREG to ACC
        ADD     one,15          ; Round the result
        SACH    errf            ; Store high ACC to "errf"
        LT      errf            ; Load TREG with "errf"
        LDP     #0              ; Select data page 0
        LARK    AR1,64-1        ; Use AR1 as counter
        MAR     *,AR6           ; Indirect addressing using AR6
        MPY     *-,AR7          ; Multiply "errf" by ê(n)
                                ; Decrement pointer
                                ; NEXT AR7

ADAPT   ZALR    *,AR6           ; Zero ACC low,load high,round
                                ; NEXT AR6
        MPYA    *-,AR7          ; Multiply and accumulate
                                ; previous product
                                ; Decrement pointer
                                ; NEXT AR7
        SACH    *+,0,AR1        ; Update coefficients
                                ; Increment pointer
                                ; Shift by 0 (no shifting)
                                ; NEXT AR1
        BANZ    ADAPT,*-,AR7    ; Branch to ADAPT when AR1>0
```

Chapter 5

Results

The results of the measurements are presented here together with the necessary setups to perform them. The main measurements are the added stable gain (ASG) measurement for the adaptive feedback cancellation filter and the output sound pressure level with an input of 90dB ($OSPL_{90}$) measurement.

5.1 Added stable gain

The measurement of the added stable gain (ASG) is based on the difference between the maximum gain achieved in the hearing aid system without causing feedback for two settings: with and without the adaptive feedback cancellation algorithm running on the DSP.

Preliminary tests showed that the added stable gain of the adaptive feedback cancellation filter was rather high, so that under the original forward path gain, feedback could not be achieved. One can either introduce more gain in the forward path by scaling the output from the DSP (in the digital domain) or in the analog domain by adding an extra amplifier. The approach of scaling in the digital domain was rejected as it might clip the output of the signal or cause overflow to occur if not implemented properly. To add extra gain in the forward path, an additional Denon preamplifier DAP 2500A was added to the signal path to increase the overall gain of the hearing aid system. The connection of the equipment is shown in figure 5.1.

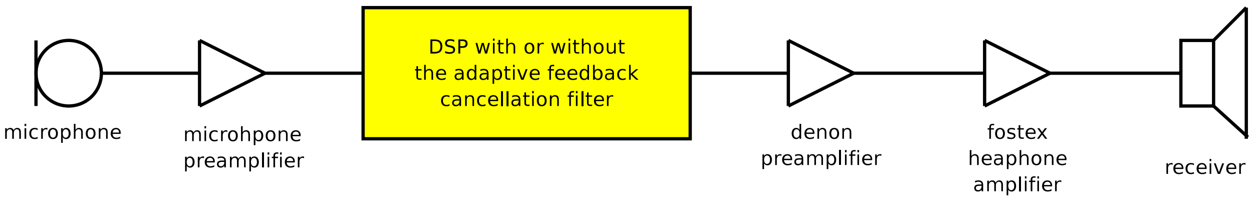

Figure 5.1: Block diagram for equipment connection in the ASG measurement

In the simulations performed in chapter 3.3, white noise sequences were applied to the input. Similarly, a white noise acoustic signal was applied to the microphone of the hearing aid. The reason why speech input signals were not used is that the algorithm with decorrelation delays was not implemented on the DSP. However, no significant detrimental effects were noticed when speech was an input to the system. The hearing aid used was the one with a vent. Two different sound pressure levels (SPLs) of the white noise were applied onto the microphone. The SPLs measured with the presence of the HATS were 66dB and 70dB.

For consistency, the volume at the headphone amplifier was set to maximum and the overall gain of the forward path was controlled by the volume knob of the Denon preamplifier. By connecting a pair of headphones to the Fostex headphone amplifier, one can monitor the signals applied to the receiver of the hearing aid.

With the presence of the white noise acoustic signals, the forward path gain of the hearing aid was slowly increased until feedback was just about to occur. The judgment of whether there was annoying feedback was by determining if a feedback howling tone was present by listening to the signal through the headphones. To avoid changes in the hearing aid fitting in the measurements, these volume settings at the preamplifier were noted until all the measurements involving the HATS were done.

When all measurement requiring the HATS were done, the vented hearing aid was disconnected and the volume knob of the Denon preamplifier was restored to the settings previously noted. To extract information about the added stable gain only the positions of the knobs in the Denon preamplifier were needed, given that they will give the difference in dB for the settings noted. The MLSSA was used to obtain the required gain data. The block diagram of the gain measurement is shown in the figure 5.2. The results of the measurements are shown in the following table:

SPL of white noise	**Added stable gain**
66dB	11.0dB
70dB	12.7dB

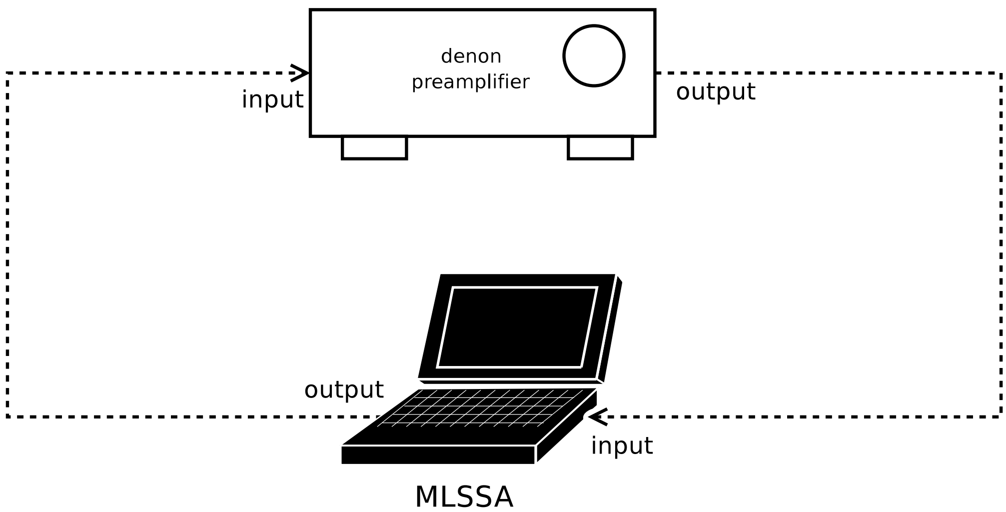

Figure 5.2: Block diagram for the measurement of forward path gain using MLSSA

5.2 $OSPL_{90}$ measurement

The aim of the $OSPL_{90}$ measurement is to obtain the *in situ* output sound pressure level from a hearing aid mounted in a HATS with a reference input sound pressure level of 90dB. According to the IEC 60118-8 standard (see Appendix E), the input sound source should produce frequencies over the range 200 to 8000 Hz keeping the reference input SPL constant. The frequencies considered in the measurement were 250 Hz, 500 Hz, 1000 Hz, 2000 Hz, 4000 Hz and 8000 Hz. In order to keep the reference input at the same level, the output voltage levels sent to the sound source were recorded for each octave band.

The measurement was made only for the hearing aid system without the AFC running. Because of the nature of the AFC algorithm implemented, possible interference or cancellation of sinusoidal inputs was expected and it was also observed. This is why the $OSPL_{90}$ measurement was discarded for the AFC implementation. To measure the performance of the hearing aid system with the AFC, and to check that the system was not severely attenuating the input for input signals other than pure tones, a test of the gain of the hearing aid system with AFC was done for a white noise input. The result was compared to the gain of the hearing aid system without AFC for the same white noise input. The results of this measurement are presented in the next section.

The standard specifies that the right ear on a HATS should be used. However, as the hearing aid shells supplied by the Oticon are both for the left ear, only the left ear on the HATS was used. A B&K type 4128 HATS was used for the measurements.

The measurements were performed in the VR-Lab. The room was considered acoustically dry enough to provide essentially free field conditions for the measurements under consideration. In the calibration phase, a B&K measurement microphone type 4133 was placed 1 meter from the sound source, a coaxial loudspeaker. According to the IEC 118 standard, the test point chosen to perform the measurements has a 270° azimuth angle of sound incidence and the elevation angle was set to 0°. This corresponds to the sound source pointing directly to the left ear and at the same height of the hearing aid microphone. The height of the test point was set to 73 cm. This would allow a lower frequency limit of 118 Hz for the measurement (assuming a sufficiently acoustically dry surface of the floor). Figure 5.3 shows this situation when the OPSL90 measurement was actually measured during this work.

In order to obtain the required reference input levels, amplified signals of sine waves from B&K signal generator type 1049 were applied to the loudspeaker. The voltage level fed to the sound source was then adjusted until at every octave band a 90dB SPL was recorded at the measurement microphone. This was done without the HATS positioned in the test point.

Next, the measurement microphone was substituted by the HATS wearing the hearing aid system. The microphone in the HATS' left ear was also previously calibrated with a B&K type 4220 calibrator. The HATS was put at the test point position. The same tone stimulus signals were applied and the SPL measured by the HATS microphone were recorded.

The result of the measurement is shown in the figure 5.4. The figure shows that there is a variation between 83 and 123 dB in the output SPL measured in the ear simulator. The middle frequencies

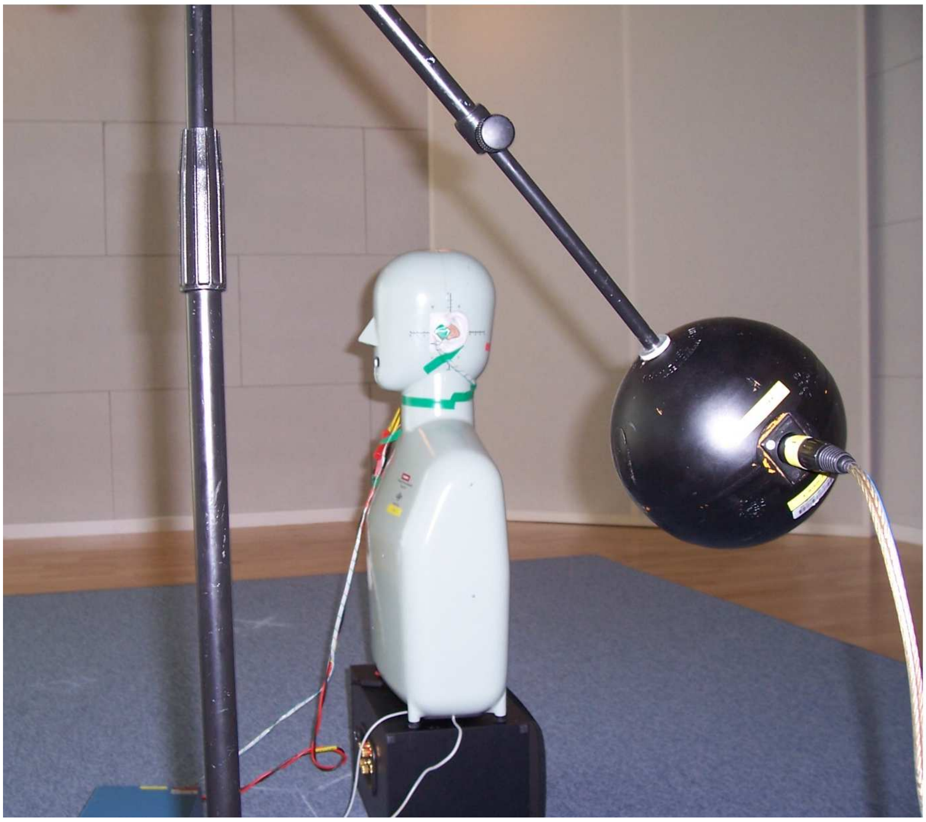

Figure 5.3: Manikin and sound source positioned in the VR-Lab for measurement of the $OSPL_{90}$

present gains that are over 25 dB. The negative gain value found in 8 kHz can be due to the fact that the antialiasing filter in the A/D converter severely attenuates frequencies close to 8 kHz.

In commercial hearing aids the output SPL is usually limited to 100 or 110 dB as the whole purpose of an hearing aid is to help the hearing impaired by amplifying sound in an intelligent way. Most hearing aids will limit the output by operating in compression mode for a input SPL of 90dB.

5.3 Hearing aid system gain under white noise input

In order to compare the performance of the hearing aid system with and without AFC for a non pure tone input, a test of the hearing aid gain was done for a white noise input signal. The reference microphone was placed near the hearing aid microphone mounted on the HATS, in order to obtain a reference level at the input of the hearing aid. The levels in the ear simulator were measured for the cases with and without AFC. For this setup, the influence of the manikin is not being considered. The reference levels at the hearing aid position and the levels in the ear simulator were input to the B&K analyzer type 2133. A third octave band analysis was done and the gain levels obtained for the cases with and without AFC are plotted in figure 5.5.

In general the system without AFC presents a slightly higher gain. The differences are in general less than 5 dB and rather small compared to the gain of the system which mainly remains over 20 dB throughout the frequency range considered.

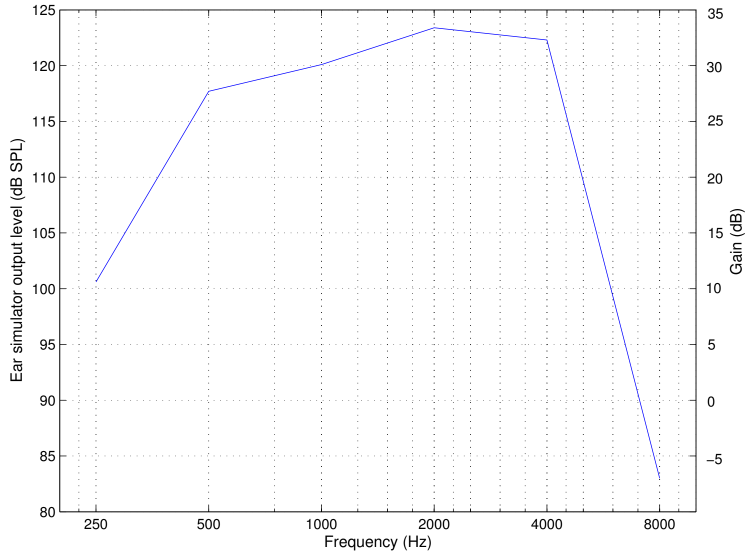

Figure 5.4: $OSPL_{90}$ measurement results within the frequency range 200 Hz to 8 kHz

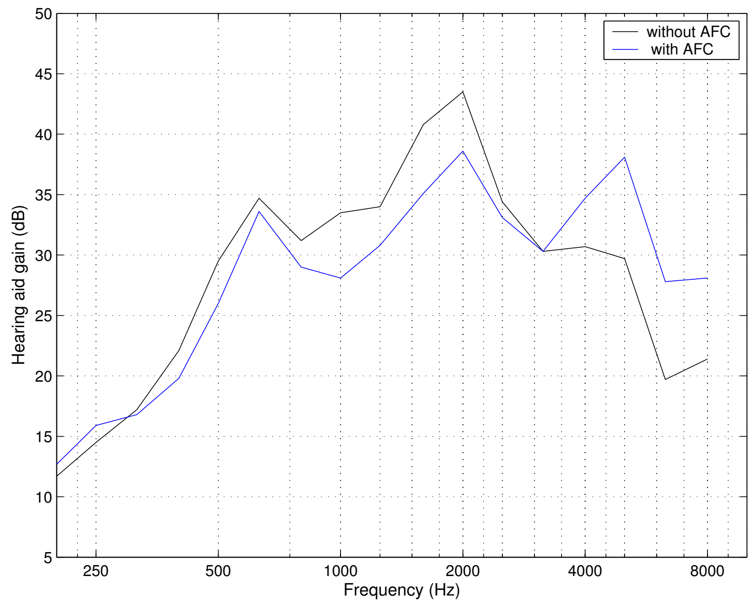

Figure 5.5: Hearing aid gain for white noise input signal

Chapter 6

Discussion and conclusions

6.1 Discussion

Several methods have been proposed to avoid the effects of feedback in hearing aids ([13], [2], [3], etc) and each method has its own trade-off. It is not trivial to determine which method to select. From previous works, it can be said that in general, some approaches are better than others. But even so, hearing aid manufacturers may implement methods in their commercial hearing aids feedback cancellers that in principle have some disadvantages.

In commercial hearing aids, the principal factors determining which approaches to be chosen are always quality and economics. The optimal solution could be just a trade off between the two factors. But there are other related factors that have to be considered in finding a particular adequate solution. A hearing aid is not a device to create feedback, it is a device designed to solve, at least partially, hearing impediments. Then, the solution for a feedback canceller should always consider for which kind of hearing aid it is designed. For example, if a person suffers from a mild hearing disability that requires not much gain and only in certain frequencies, it is possible that in this situation feedback simply will not occur. This means that the quality factor should have the user as reference. In that sense, the final evaluation of the hearing aid feedback canceller, should require listening tests.

The economic factor comprises some other issues. The main one is the complexity of the implementation of the device. Normally the more complex, the more expensive, especially considering the limitation of physical space in hearing aids. The user should also be taken as reference, because if the product is too expensive no user could afford it, and the problem would remain where it was. This should not limit researchers, but should be taken into account.

Coming back to the users' necessities, it is very important to remark that not every objective improvement in the feedback aid canceller will be perceived so for them. For example, it will not be worth to put efforts in reducing system noise, if the user threshold of hearing is over that level of noise.

Although, as mentioned, the feedback cancellation system selection should consider the type of hearing impediment for which the system was designed. However, this project is a research work focused on the problems of feedback cancellation. In this sense, one have to try to obtain the best objective results, with the solution that does not exceed a reasonable limit of complexity. After the analysis work, it was clear that the best method to overcome the feedback problem should be an adaptive feedback cancellation filter. Also, in that section, it has been argued why the continuous adaptation was considered as the starting point. The basis for the AFC algorithms was the LMS adaptive filter. Variations of that algorithm were simulated to compare results.

Among the variations of the LMS algorithm, it was decided to simulate two variations that have the greatest potential to improve the performance of the AFC: the normalized version of the LMS algorithm (NLMS) and the addition of subband filters [20]. The splitting of operating frequency range of the hearing aid to two or more subbands allowed the AFC in each subband to achieve faster adaptation, however accuracy of the feedback path estimation was lowered. Its efficiency in computation has to be balanced with a relatively higher complexity in the implementation. If the DSP capacity or battery consumption is a constrain, then the subband approach could be a good alternative. On the other side, the NMLS had also theoretically better performance than the LMS in terms of adaptation time and other advantages like the independence of the correction term from the level of the current sample input vector $s(n)$.

An important key point was that the LMS algorithm should not work for highly self correlated input signals, such speech [11]. The simulation results confirmed this phenomena. The system worked properly for white noise as input signal but not for speech. Therefore, one of the possibilities to decorrelate the input signal and output signal of an hearing aid was chosen: to apply a delay in the forward path. As expected, this variation really helped to solve the problem. The NLMS algorithm was investigated in depth using simulations. The aim of the investigation was to try to get an objective value of the improvement provided by the canceller. The added stable gain shown in section *Design* suggested that a good performance of the algorithm could be expected.

The implementation in principle showed something different. While the LMS algorithm was working properly, that means it reduced feedback significantly and was adapting to changes in the feedback path, the NLMS algorithm it was impossible to make work in the way it should. Due to limited time, it was impossible to check why it did not work as expected and the rest of the performance tests were made on the LMS based system. It seemed that the performance of such canceller was better than expected for speech as input signal, as it was actually reducing feedback and the speech signal was still intelligible. The question that emerged then was to what extent did it work properly in fact. The answer should require listening tests, which is beyond the limits of the present work. Then, loss in intelligibility, or other effects could be noticed. On the other hand, it could also be possible that the input signal and the signal fed back to the microphone were somehow decorrelated by the system implementation, even though it was not explicit in the algorithm.

From that time on, the experiments were carried out with a white noise signal as input, leaving the correlation problem and intelligibility discussion for further works. The performance of the canceller can be described in terms of the ASG obtained, 12 dB. Another result was to observe the feedback tracking of the algorithm. A measurement of this was not documented in a proper objective way in the report due to its implicit difficulty and time issues. Acoustical environment changes next to the hearing aid induced a momentous howling tone that could be ceased quickly. This demonstrated the good adaptation of the algorithm.

However, the 12 dB of ASG obtained cannot be considered as an excellent result, as other researches already referred in this work [19] could report up to 20 dB in ASG. Besides, there were also unwanted effects on the input signal. For example, as the gain was increased, the system noise heard was also increased, sometimes in non acceptable extent. Most of that noise, however, cannot be directly associated to the canceller algorithm, as it was present also when the DSP system was not acting as canceller. However, a proper measurement of the signal to noise ratio (SNR) was not done for the implemented system.

6.2 Conclusions

From the feedback path measurement results some general characteristics of the feedback path behavior can be inferred:

- Introducing a vent causes not only a significant amount of leakage that highly encourages feedback, but it also changes dramatically the shape of the the feedback path frequency response.
- With a fixed configuration, the introduction of a reflective object near the hearing aid at different distances can cause more a change in the amplitude of the feedback path frequency response than a change in the frequency response overall shape.

From the simulations performed some other conclusions can be made:

- It can be concluded that in theory an adaptive LMS based AFC algorithm will not estimate the feedback path properly if the input signal is a speech.
- The addition of a small delay in the forward path will effectively decorrelate the speech signal and hence allow the AFC to obtain a better estimation of the feedback path.
- The subband LMS approach can reduce the number of operations required for each adaptive filter to perform, but the estimation tends to be not as accurate as in the wideband case. It also seems that because of the group delay involved in filtering the signals, the method requires no extra decorrelation delay to estimate the feedback path when the input is speech.
- With higher gains in the forward path, better feedback path estimations can be achieved, which is beneficial because at higher gains feedback will be more likely to occur.

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Appendix

Appendix A

Human audition

A.1 Anatomy of the auditory system

Since this project covers hearing aids performance, it is important to have an overview on the human auditory system and human perception. In order to give a more practical explanation about them the natural direction of hearing process, from when a sound wave arrives to the outer part of the ear to when it evokes an auditory event was followed. But to analyze it systematically a initial division has to be made. There are two main parts in the auditory path of the human being in which they can be found many organs involved. These main parts are [10]:

1. The peripheral auditory pathway.
2. The central auditory pathway.

Even if both parts are very important for the understanding of human hearing, since the typical hearing aids considered in this work are placed in the peripheral auditory pathway, the central auditory pathway will not be discussed in depth, but it is necessary to give a short description of the peripheral.

The main purpose of peripheral auditory pathway is to detect and interpret the sound. Three subparts can be considered within the peripheral auditory pathway. The three parts of the ear are shown below in figure A.1 and are called *outer ear*, *middle ear* and *inner ear*.

- Outer ear:

  In the peripheral auditory pathway, the outer ear receives the sound and sends it to the *ear drum* or *tympanic membrane* through the ear canal. This is also called the *Acoustic Path*. Its most external element is the *pinna*, in charge of the sound collection. And the pinna is connected by the *concha* to the *auditory canal*. Is precisely in this part of the peripheral auditory pathway where the ITE hearing aids are located, obstructing the ear canal.

- Middle ear:

  Also called *tympanic cavity*, it transforms the energy of a sound wave into the internal mechanical vibrations of the bone structure formed by the *hammer*, the *anvil* and the *stirrup*.

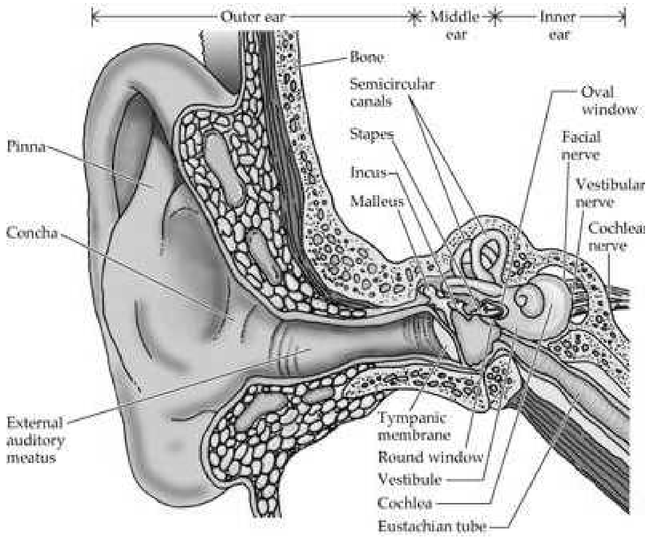

Figure A.1: The outer, middle and inner ear and their different organs.

Then, these vibrations will be converted into a compressional waves in a fluid at the inner ear. So it can be considered as the bridge between the vibration of the ear drum at the end of the outer ear and the vibration of the oval window at the inner ear.

- Inner ear:

  It is made up of a series of intercommunicating cavities that together form the bony labyrinth: the cochlea, the vestibule and the semicircular canals. Its main function is to transform the energy of the compressional wave within the inner ear fluid into nerve impulses which can be transmitted to the brain. In the inner ear the most important structure is the cochlea which is the organ responsible the mentioned transformation of middle ear fluid vibrations into neural firings.

  The cochlea is playing the most important role in the analysis of sound made by means of a frequency-space analysis and also acting as an intelligent amplifier that provides more gain to lower intensity sounds. From the cochlea, the sensory information is conveyed to the central auditory pathway.

A.2 Human auditory perception

This section does not intend to be a profound description of the human ability of hearing, but only a brief introduction to the basic principles of sound cognition. Then, below there is a basic explanation about how human beings to hear in terms of perception and not of the physical signal only. The human perception of sounds is a big subject and it can be investigated in huge depth, It has been briefly outlined here because it is related to the requirements of a hearing aid system in the sense of that when factors like selecting the frequency range of operation (better efficiency where the ear is more precise) or maximum noise level admitted (a small amount of noise can be admissible when it is masked by the input signal) are considered.

The first thing to analyze is what human beings can hear and what human beings can not. What human beings perceive are changes in pressure. So first question is what kind of changes human perceive as sound. This leads to the consideration of two different domains, the frequency of the sound wave and the level of it. Because not every frequency of sound is audible for human auditory system. And even if the frequency is in audible range for humans not every sound wave can be perceived.

The standard frequency range considered as audible range is the range between 20 Hz and 20 kHz. However this range may change much along different subjects together with aging.

On the other side the loudness perception is influenced also by other effects, as the kind of the sonorous stimulus (e.g. pure tones or complex sounds and its bandwidth), the frequency of this stimulus and the possible simultaneous presence of other sound waves.

The first consideration about sound loudness perception is the pseudo-logarithmic response of human auditory system to sound pressure changes. Because of that, sound loudness is often represented in decibel scales. But this non-linear response is not constant along the frequency. To represent that response, measurements were made on a very large population of young subjects[21]. The results are known as the curves of human audibility and equal loudness (A.2).

In these curves it is observed that the ear identifies middle frequency sounds as louder when equal sound pressure is applied. So it is in this range where the human ear is sharpener or more efficient. To perceive a similar sensation of loudness much higher pressure is required in low and high frequencies. Furthermore, this changes in loudness perception along frequency are different along wave pressure itself: when the level of the sound is higher the curves become flatter.

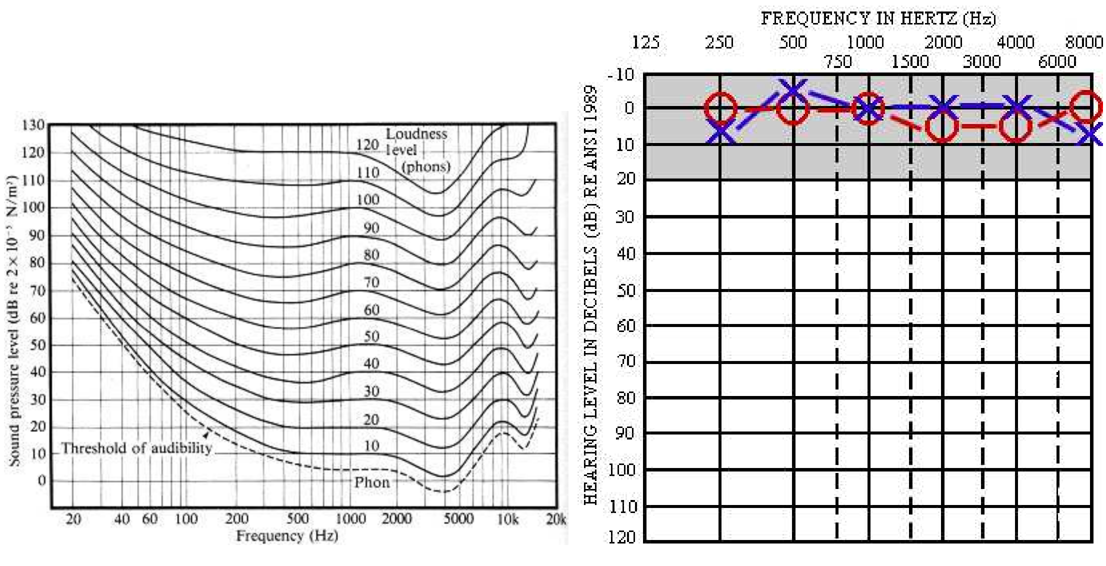

Figure A.2: Human threshold of audition and equal loudness curves (left) and example of a typical audiogram (right). Audiograms are normalized to the threshold of audition.

When testing the hearing abilities of subjects, the results are presented normalized on the basis of the threshold of hearing (the lower curve in left figure in fig. A.2). In that figure it is also shown an example of a typical audiogram.

Another important effect on loudness perception is that not every sound that is above the threshold of hearing can be perceive. If a sound is given to a subject, while a much louder is also present, then it is possible that the subject does not perceive the lower one. This is known as masking and hearing designers may take advantage of it as some amount of noise can be admissible in the system since the input sound will be louder.

Appendix B

Fundamentals of feedback

Feedback can refer to feedback in cybernetics, control theory and related disciplines where it is used in various areas dealing with complex systems, such as engineering, architecture, economics, and biology. It can also allude to a type of message that a receiver transmits back to a source in response to having received a message. As well as it occurs as audio feedback, which is the "howl" heard in microphone or guitar amplification systems.

But in every case feedback is considered as a process whereby a function of the output signal of a system is passed (fed back) to the input. In a system with a feedback path lines are usually drawn starting from the input through the system and to the output. The feedback path is shown by another arrowed line, directed from output outside the system to an input, resulting in a loop on the diagram. That is called feedback loop.

B.1 Nyquist's stability theorem

The Nyquist plot allows us to predict the stability and performance of a closed-loop system by observing its open-loop behavior. Stability of the closed-loop control system may be appointed directly by computing the poles of the closed-loop transfer function. In contrast, the Nyquist stability criterion allows stability to be determined without computing the closed-loop poles.

A Nyquist plot is used in automatic control and signal processing for assessing the stability of a system with feedback. It is represented by a graph in polar coordinates in which the gain and phase of a frequency response are plotted. The plot of these phasor quantities shows the phase as the angle and the magnitude as the distance from the origin. The frequency response of a system can be viewed via the Nyquist diagram.

The most important criteria of the Nyquist's Stability Theorem can be summarized as (see figure B.2):

1. If the open-loop transfer function F(s) is stable, then the closed-loop system is unstable for any encirclement of the point -1.
2. If the open-loop transfer function F(s) is unstable, then there must be one counter clock-wise encirclement of -1 for each pole of F(s) in the right-half of the complex plane.

3. The number of surplus encirclements is exactly the number of unstable poles of the closed-loop system.

The Nyquist plot is very useful in looking at the stability of an open feedback system. If the magnitude function of a frequency that is phase-shifted 180° is greater than or equal to unity then the closed system will be unstable as it can be seen in figure B.1.

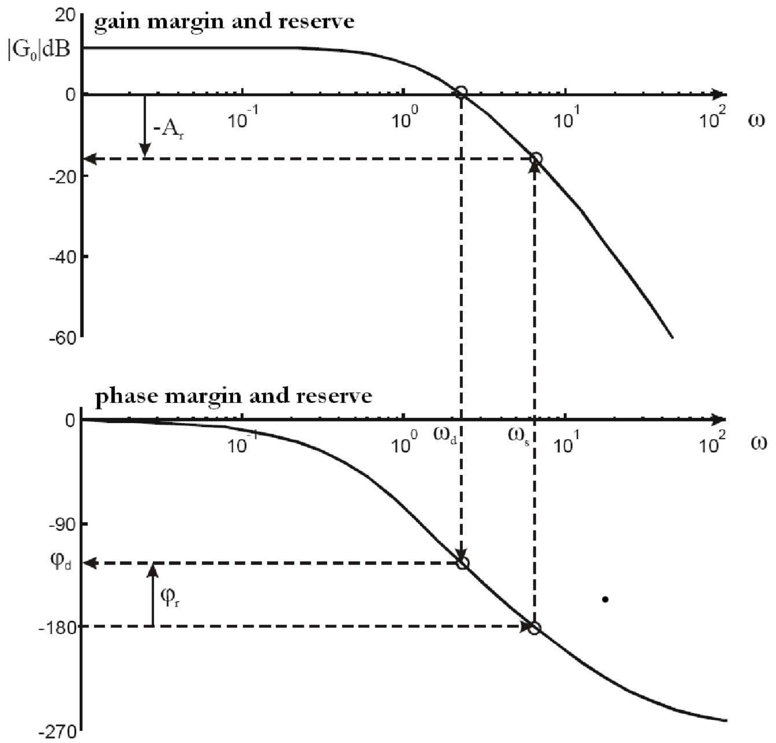

Figure B.1: Bode plot

The Nyquist criterion can be used for design purposes regardless of open-loop stability (the Bode design methods assume that the system is stable in open loop)[1]. Therefore, this criterion is used to determine closed-loop stability when the Bode plots display confusing information.

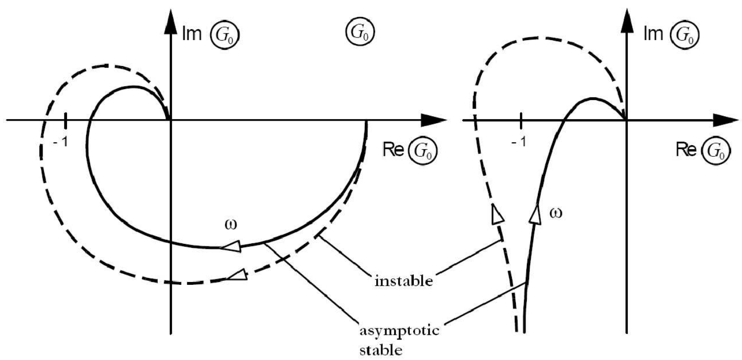

Figure B.2: Instable and stable Nyquist plot depending on the position of the system curve in relation to -1.

[1] Bode plot shows the frequency response of a system in terms of amplitude and phase.

Appendix C

Overview of different types of hearing aid

There are many different kinds of hearing aids. They differ from size, form and mode of transmission of acoustics signals. The behind the ear (BTE) type, in the ear (ITE) type and the in the canal (ITC) type hearing aid amplify sound and the resulting sound is acoustically transmitted to the ear drum through air.

They consist basically on a microphone to pick up the signal that arrives the ear, and amplification module that reinforces that signal, and a receiver that reproduces the amplified sound and leads it to the eardrum. Currently they include a digital signal processing module where different improvements are implemented, such as limiters, compressors, noise reduction, or feedback cancellation systems. They are relatively small devices that the user wears in or next to the ear but that can be taken off easily.

The following table summarizes the main features of different types of hearing aids which are depicted below in figure C.1.

Figure C.1: Example of real *behind the ear*, *in the ear* and *in the canal* hearing aids.

Type	Classification and Summary
Behind The Ear (BTE)	Ear worn hearing aid. One or more microphone, a central processing unit and a receiver are housed on the main body behind the ear; the receiver is coupled to the ear canal via a tube. They usually provide more gain as larger battery and more powerful amplifier can be housed within the hearing aid casing.
In The Ear (ITE)	Ear worn hearing aid. ITE hearing aids are smaller than the BTE; one or more microphones, a central processing unit and a loudspeaker are housed in the same structure; the whole unit is partially inserted in the ear canal and is fitted on the concha of the ear. They usually provide less gain than the BTE counterpart due to smaller size.
In The Canal (ITC)	Ear worn hearing aid. ITC is similar but smaller than the ITE type; the whole hearing aid unit can be placed in the ear canal.
Body Aids	Body worn hearing aid. Are used by people with profound hearing loss. The aid is attached to a belt or a pocket and connected to the ear by a wire. Because of its large size, it is able to incorporate many signal processing options, but it is usually used only when other types of hearing aids cannot be used.

Appendix D

Matlab codes

The algorithm that simulates the proposed AFC system is presented at first here. The second program shown is the subband AFC simulation. The *.m* files can be found in the CD attached to the report.

```
////////////////////////////////////////////////
////////////// Proposed AFC simulation /////////
////////////////////////////////////////////////

clear all;
N=50000; \%number of samples (ir iterations) of the input signal
%input=randn(N,1); Used when testing with a white noise input signal

[x,FS,NBITS]=WAVREAD('Katspeech.wav');   %Speech segment used in the simulations
x=x(1:N);
input=x;                                 %input signal to the system

M=100;                                   %length of the adaptive filter
FP1;                                     %load measured data of feedback path
FP=FP1(1:M);                             %take the first M samples
f=FP';
w=zeros(M,1);                            %initial coefficient values set to 0
mu=0.01;                                 %numerator of the time varying step size
                                         %    parameter

G=4;                                     %gain of the forward path

SD=20;                                   %sample delay in the case of speech
                                         %    input signal
e=x(1:M)';                               %initialize the error vector with the
                                         %    first sampled input values


%%%%%%%%%%%%%%  NLMS  %%%%%%%%%%%%%%%%

for n=M+SD:N-SD

    s=e(n-SD:-1:n-M-SD+1)*G;               %output signal to the receiver, delayed by
                                           %     SD samples
    y_(n)=w'*s';                           %filtered signal through the estimated
                                           %     feedback path
    y(n)=f*s';                             %simulated feedback signal through the
                                           %     measure feedback path
    d(n)=x(n)+y(n);                        %simulated desired signal that is the sum
                                           %     of the input plus feedback signals
    e(n+1)=d(n)-y_(n);                     %error signal update
    w=w+(mu/(0.0001+s*s'))*s'*e(n+1);      %NLMS coefficient update
    diff=abs(f'-w);
    WEVN(n)=10*log10((diff'*diff)/(f*f')); %misalignment calculation

end;
```

```
%%%%%%%%%%%   Calculation of performance parameters  %%%%%%%%%%%%%%%%%%%

MSG_without=20*log10(min(1./(G*abs(fft(f))))); %Maximum Stable Gain w/o feedback
                                               %     cancellation
MSG=20*log10(min(1./(G*abs(fft(f)-fft(w'))))); %Maximim Stable Gain with feedback
                                               %     cancellation
ASG=MSG-MSG_without;                           %Added stable gain because of AFC

plot(WEVN,'k');                                %display results in terms of the
                                               %     misalignment

///////////////////////////////////////////////
///////////// subband AFC simulation//////////
///////////////////////////////////////////////


clear all; M=100; N=50000;              %input=randn(N,1); white noise input signal
[x,FS,NBITS]=WAVREAD('Katspeech.wav');  %Speech segment used in the simulations
x=x(1:N);
input=x;                                %input signal to the system

G=4;                                    %forward path gain
D=2;                 %number of subbands
w0=zeros(M/D,1);     %initialization of each adaptive filter, complexity is reduced by 1/D;
w1=zeros(M/D,1);     %=================================================================;
FP1;                 % load feedback path data stored
f=FP1(1:100)';       %use the first 100 taps

ff=0.999999;         %forgeting factor
alpha=0.0001;        %mu for subband nlms

LO_D=fir1(32,0.5);   %H0(z)
HI_D=qmf(LO_D,1);    %H1(z)=H(0)(-z)
LO_R=-LO_D;          %G0(z)=-H1(-z)=-H0(z)
HI_R=HI_D;           %G1(z)=H0(-z)=H1(z)

S=M/D;               %size of each of the subband adaptive filters (50)

%%%feedback path to be estimated is downsampled

for i=1:S fd(i)=f(D*i); end;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

% simulate input signal splited into 2 bands by the analysis qmf filters %

inputlow=filter(LO_D,1,input);                %Band 0 id the output from the lowpass
inputhigh=filter(HI_D,1,input);               %Band 1 id the output from the highpass

%%%%%%%%%

e0=inputlow(1:M)';  %initialize each eror vector with first input values for each band
e1=inputhigh(1:M)'; %=================================================================%

%initialize input and error power estimate for each band
Ps0=0; Pe0=0;

Ps1=0; Pe1=0;
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
```

```
for n=M:N/D

    %%%%%%%%%%%%%%%%% adaptive filter 0 %%%%%%%%%%%%%%%%%%%%%%%%%%

    s0fd=e0(n:-D:n-M+1)*G;                              %output signal from band 0
    Ps0=ff*Ps0+(1-ff)*s0fd*s0fd';                       %power estimation of Ps0
    y0_(n)=w0'*s0fd';                                   %filtered signal through the
                                                        %    estimated feedback path
    y0(n)=fd*s0fd';                                     %simulated feedback signal band 0
    d0(n)=inputlow(D*n)+y0(n);                          %desired signal is downsampled by 2
    e0(n+1)=d0(n)-y0_(n);                               %error signal for band 0
    Pe0=ff*Pe0+(1-ff)*e0(n+1)^2;                        %power estimation  of error
                                                        %    signal for band 0
    w0=w0+(2*alpha/(S*(0.0001+Ps0+Pe0)))*s0fd'*e0(n+1); % subband nlms adaptation
    diff0=abs(fd'-w0);
    WEVN0(n)=10*log10((diff0'*diff0)/(fd*fd'));         %misalignament calculation for AF0

    %%%%%%%%%%%%%%% adaptive filter 1 %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

    s1fd=e1(n:-D:n-M+1)*G;                             %output signal from band 1
    Ps1=ff*Ps1+(1-ff)*s1fd*s1fd';                      %power estimation of Ps1
    y1_(n)=w1'*s1fd';                                  %filtered signal through the
                                                       %    estimated feedback path
    y1(n)=fd*s1fd';                                    %simulated feedback signal band 1
    d1(n)=inputhigh(D*n)+y1(n);                        %desired signal is downsampled by 2
    e1(n+1)=d1(n)-y1_(n);                              %error signal for band 1
    Pe1=ff*Pe1+(1-ff)*e1(n+1)^2;                       %power estimation of error signal
                                                       %    for band 1
    w1=w1+(2*alpha/(S*(0.0001+Ps1+Pe1)))*s1fd'*e1(n+1);%subband nlms adaptation
    diff1=abs(fd'-w1);
    WEVN1(n)=10*log10((diff1'*diff1)/(fd*fd'));        %misalignament calculation for AF1

end;


%%% upsample stage of the output error signals %%%%

for i=1:length(e0)*D if rem(i-1,D)==0; e1up(i)=e1(ceil(i/4));
e0up(i)=e0(ceil(i/4)); else
e1up(i)=0;
e0up(i)=0;
end; end;

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%


%filter each upsampled error signal with the reconstruction filters

e0f=filter(D*LO_R,1,e0up);
e1f=filter(D*HI_R,1,e1up);

e=e0f+e1f;                            %total error signal for the output


%% display results for each band in terms of the misalignment %%
subplot(2,1,1); plot(WEVN0); subplot(2,1,2); plot(WEVN1);
```

Appendix E

Standards for hearing aids measurements

To grant comparable results when testing and presenting the features of different hearing aids, measurements should be performed following a standard. For hearing aids there are two main sets of standards. One has been developed by the *American Nationals Standards Institute (ANSI)* and it is the ANSI S3.22 and the other was made by the *International Electro-acoustical Commission (IEC)* and is referred as IEC 118 or 60118. They present many similarities and only some differences, but those differences can make the final values obtained vary significantly when the test has been performed either with one or the other. Therefore, when presenting the characteristics of any hearing aid, it is necessary to state which standard has been followed.

The main difference between the two standards mentioned is that ANSI specifies that hearing aids should be measured in a 2-cc coupler, whereas when using IEC it can be substituted by an ear simulator. During this work it was decided to follow the standard IEC 60118-8 titled *Hearing aids-Part 8: Methods of measurement of performance characteristics of hearing aids under simulated in situ working conditions*, which explains how to measure a hearing aid mounted on a manikin.

The reason for choosing this standard is because for the measurements required, it was important to emulate real conditions, as the output SPL in the ear or feedback will depend on the acoustical environment near the hearing aid. As it is mentioned in the standard objective *the purpose of this report is to describe test methods which simulate the acoustical effects of a median adult wearer on the performance of a hearing aid*. This project is inscribed in this frame. Parameters as the fitting or the pinna will influence the SPL in the ear (and the feedback), therefore to have a good approximation of the hearing aid performance it was recommendable to make the measurements on a *head and torso simulator* (HATS).

This standard will provide a reliable and repeatable way to do the measurements, but it should always be considered that the results achieved on an individual person may differ substantially from the ones got on a HATS. The measurement of the $OSPL_{90}$ intends to characterize the frequency response of the system when mounted on an average person.

A *head and torso simulator* or *manikin* is a human body approximation extending from the top of the head to the waist. It is specifically designed to simulate the acoustic diffraction produced by a median adult human head and torso. The head includes two pinna simulators, and contains at least one ear simulator. In turn, an ear simulator consists of a principal cavity, acoustic load networks and a calibrated microphone located so that the sound pressure at the microphone corresponds approximately to the sound existing at the human eardrum.

E.1 Standard measurement of the simulated *in situ* $OSPL_{90}$ frequency response

The $OSPL_{90}$ is the output sound pressure level in the ear simulator produced by an input signal of 90 dB when the hearing aid gain control is set at full on gain. Consequently, the *in situ* $OSPL_{90}$ frequency response is the $OSPL_{90}$ measured in the ear simulator as a function of frequency. The structure of the standard regarding this measurement is briefly explained below:

E.1.1 Test equipment

Regarding the test equipment the standard specifies:

- The acoustical requirements for the test space: such as essentially free field conditions or distance from the manikin to the closest surface of wavelength / 4 ($\lambda/4$) and to the sound source of $1m$.
- The sound source: such shape, dimensions and accurate flat frequency response over the range 200 Hz, 8000 Hz, provided with an uniform wavefront at the manikin position area and a with small harmonic distortion.
- The requirements of a manikin: shape, dimensions, etc.
- The ear simulator: in accordance with IEC Publication 711.
- The equipment for the different measurements: it is specified in terms of the calibration precision, maximum deviation allowed in sound level reception, harmonic distortion and internal noise of the devices used, tolerances of indicators, and accuracy in frequency responses.

E.1.2 Test conditions

The test conditions are also precisely explained, regarding:

- The choice of the test point.
- The ambient conditions (temperature, humidity and atmospheric pressure).
- The no dressing of the manikin.
- The location of the hearing aid. It is explained that the right ear should be used, unless otherwise stated. During this work this could not be complied as only left ear hearing aids were available.

- The normal operating conditions for the hearing aid, like the power supply and the gain control. Regarding controls it says the position of them shall be stated in the report of the measurement performed.
- Any other particular accessories used should be stated.

E.1.3 Measurements

Different measurements are described in the standard IEC 118-8. As general instructions it is said that the data should only be quoted for that part of the frequency range between 200 Hz and 8000 Hz which the output from the hearing aid falls by at least 10 dB when the signal source is switched off. It also specifies that when different methods are allowed, it should be stated which is the one used. The measurements described are the following:

- First, the procedure for the adjustment of the reference input sound pressure level is explained. The reference input corresponds to the level at each frequency in the absence of the manikin. This level must be kept constant at 90 dB. This will be a previous measurement for all the detailed after.
- Full-on simulated insertion gain measured by two different methods
  - The constant reference input SPL method.
  - The constant ear simulator SPL method.
- Directional characteristics. Three different measurements are explained:
  - Manikin directional response (MDR).
  - Simulated *in situ* directional response (SISDR).
  - Simulated insertion directional response (SIDR).

However, the only measurement described in this standard used during this project was the **Simulated *in situ* $OSPL_{90}$ measurement**. With this measurement it was possible to obtain a response of the hearing aid when mounted on a HATS and at maximum gain. The test procedure is as follows:

1. Place the manikin at the reference point (in essentially free field conditions and at a distance to the closest surface bigger than $\lambda/4$ and to the sound source of $1m$.). The hearing aid should be mounted in the way corresponding to the actual use.
2. The gain control is turned full-on and other controls are set to their required positions. The maximum gain was considered the gain at which the output began to be distorted.
3. At a suitable frequency, the reference input should be set at 90 dB.
4. The frequency of the sound source is varied over the range 200-8000 Hz keeping the reference input SPL constant at 90 dB. The ear simulator SPL is recorded as a function of frequency.

E.1.4 Frequency response recording charts

In accordance with the IEC Publication 263: *Scales and Sizes for Plotting Frequency Characteristics and Polar Diagrams*, all frequency functions should be plotted on a grid having a linear decibel ordinate and a logarithmic frequency abscissa scale.

Appendix F

Specifications sheets