CALCULATING THE MAGNITUDE OF TRANSFER FUNCTIONS FOR HEARING AIDS BY INSERTION GAIN MEASUREMENTS

by

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ABSTRACT

CALCULATING THE MAGNITUDE OF TRANSFER FUNCTIONS FOR HEARING AIDS BY INSERTION GAIN MEASUREMENTS

While the hearing loss is one of the most common problem in our lives, it is also the most compensable disability. In general, hearing impairment can be fixed by a hearing aid which amplifies sound levels at different frequencies. A transfer function of a hearing aid determines how much the input signal will be amplified at certain frequencies. However, this function may vary depending on whether the hearing aid is worn or not. Therefore, when a hearing aid is in use, expected results are mostly not achieved if the calibration is done without wearing it. This phenomena is known as a fitting problem in literature and due to the altered transfer function artifacts a small incidence of people having hearing abnormalities prefer using a hearing aid.

In this thesis, to compensate the fitting problem, we designed and developed a hearing aid analyzer which uses a Real Ear Measurement (REM) method. Determining the electroacustic performance of hearing aid in situ is essential for ideal fitting and it varies according to ear shape, ear canal, and hearing sensitivity. In REM, an insertion gain measurement is obtained by inserting a probe microphone into the ear canal and calculating the gain between input and output signals. Fine tuning is performed using the transfer function based on insertion gain measurement. REMs allow determining individual based, actual characteristics of hearing aids. The superiority of our system comes from the fact that it is capable of measuring the transfer function while the hearing aid is in situ. Moreover, our design is battery powered and small in diameter which makes it portable. Finally, our system complies with international standards.

Keywords: Insertion gain measurement, Hearing aid fitting, DSP5509A, AIC33.

ÖZET

İNVİTRO KAZANÇ ÖLÇÜMÜ İLE İŞİTME CİHAZLARI İÇİN TRANSFER FONKSİYONU GENLİKLERİNİN HESAPLANMASI

İşitme kaybı günümüzde en sık görülen problemlerden biri olmasının yanında, en kolay telafi edilebilir bir engeldir. İşitme cihazı, sesi farklı frekanslarda kuvvetlendirir ve işitme kaybı genellikle bu cihaz kullanılarak tedavi edilir. İşitme cihazının transfer fonksiyonu, gelen sesin farklı frekanslarda ne kadar kuvvetlendirileceğini belirler. Bu fonsiyon, cihazın kulağa takılı olup olmamasına göre değişir. Bu yüzden, işitme cihazının ayarı kulakta takılıyken ölçüm yapılıp ayarlanmadıysa, hedeflenen kazançlar tam olarak uygulanamaz. Bu durum literatürde işitme cihazı uyum problemi olarak bilinir ve bu uyum probleminden dolayı ihtiyacı olmasına rağmen bir çok hasta işitme cihazını kullanmamaktadırlar.

Yapılan bu tez çalışmasında, uyum problemini gidermek için, gerçek kulak ölçüm yöntemini (REM) kullanan bir işitme cihazı analizörü tasarlanmıştır. İşitme cihazlarının kulaktayken elektroakustik davranışı işitme cihazı uyumu için oldukça önemlidir ve bu davranış hastanın kendisine has kulak yapısı, kulak kanalı ve işitme hassasiyeti nedeni ile değişiklik göstermektedir. Gerçek kulak ölçüm yönteminde, kulak yoluna yerleştirilen mikrofon ile invitro kazanç ölçümü yapılmakta ve işitme cihazının kazanç değerleri hesaplanmaktadır. Gerçek kulakta ölçümler, işitme cihazının kişiye özgü ve gerçek davranışını bulmayı sağlar. Tasarlanan sistemin asıl üstünlüğü gerçek kulakta işitme cihazının başarımını ölçebiliyor olmasıdır. Ayrıca, tasarım batarya ile beslenmekte ve küçük boyutu ile taşınabilir olmaktadır. Son olarak da, tasarlanan cihaz uluslararası standardlara uygundur.

Anahtar Sözcükler: İnvitro kazanç ölçümü, İşitme cihazı uyumu, DSP5509A, AIC33.

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LIST OF SYMBOLS

f	Fundamental Frequency
F_{ear}	Transfer Function of Ear
F_{ha}	Transfer Function of Hearing Aid
F_{iha}	Transfer Function of Hearing Aid in Situ
F_s	Sampling Frequency
K	Gain
$MCLK$	Master Clock
P	Power
Pr	Pressure
Re	Reference Pressure
RP	Reference Power
SCL	I2C Clock
spp	Samples Per Period
x_i	Received Sound Vector from Internal Microphone
y_i	Received Sound Vector from External Microphone

LIST OF ABBREVIATIONS

ADC	Analog to Digital Interface
AIC	Audio Interface Codec
ANSI	American National Standard Institute
BGA	Ball Grid Array
BTE	Behind-the-Ear
CIC	Completely In-the-Canal
CORFIG	Coupler Response For Flat Insertion Gain
DAC	Digital to Analog Converter
DARAM	Dual Access RAM
DFT	Discrete Fourier Transform
DMA	Direct Memory Access
DSP	Digital Signal Processor
EMC	Electromagnetic Compatibility
EMIF	External Memory Interface
ESD	Electrostatic Discharge
FFT	Fast Fourier Transform
IDE	Integrated Development Environment
IEC	International Electrotechnical Commission
IL	Intensity Level
ISO	International Organization for Standardization
ITE	In-the-Ear
LDL	Loudness Discomfort Levels
McBSP	Multichannel Buffered Serial Port
MCL	Most Comfortable Listening Levels
NAL-NL1	National Acoustic Laboratories Non-Linear
NALR	National Acoustics Laboratory Revised
NAL-R	National Acoustic Laboratory Revised
NAL-RP	National Acoustic Laboratory Revised for severe/Profound

PCB	Printed Circuit Board
PHAA	Portable Hearing Aid Analyzer
PTA	Pure Tone Audiometer
REAR	Real Ear Aided Response
REIG	Real Ear Insertion Gain
REM	Real Ear Measurement
REUR	Real Ear Unaided Response
SPL	Sound Pressure Level
THD	Total Harmonic Distortion

1. INTRODUCTION

1.1 Thesis Motivations

As the population is getting crowded and the average age is increasing, more and more people around the world suffer from hearing losses. According to a recent research the number of people with hearing impairment is nearly 70 million in Europe and 500 million world wide as seen in Figure 1.1 [1]. This rate is 10% of the population and this number is rising dramatically. It is predicted that, by the year 2025, there will be 700 million people with hearing loss all over the world. Hearing loss is a common problem which especially affects elder people, because of aging and deterioration of hearing apparatus. However, recently it is very common to see younger people with hearing abnormalities due to increasing incidence of an intense noise pollution.

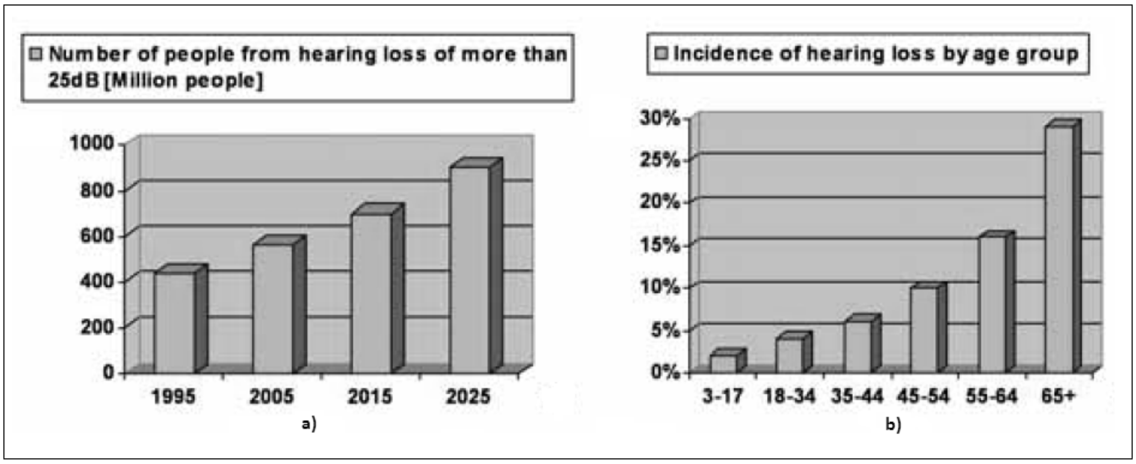

Figure 1.1 Statistics about hearing disorders, and deafness compiled from fact sheets produced by the National Institute on Deafness and Other Communication Disorders (NIDCD): a) forecast of the number of people with hearing loss of more than 25dB; b) incidence of hearing loss by age.

Hearing loss is determined by an audiologist, a trained professional on hearing. He/she tests hearing and performs hearing evaluation tests that indicate the degree of hearing loss and the particular sound frequencies that are impaired. If there is hearing loss, further tests are done to determine how much hearing loss affects the person's ability to understand speech and the type of hearing loss. The first step in hearing testing is audiometry. In this test, the patient wears headphones that play test tones with different frequencies and intensities. The patient informs the tester when a tone is heard by raising his/her hand. After the determination of hearing loss, a suitable medical treatment or one of the appropriate hearing aids and assistive device can be prescribed. Most of the hearing loss (more than 90%) can be compensated by using a hearing aid, but statistics indicate that only 20% of people who need a hearing aid actually use one.

Hearing aid is a device whose function is to amplify the acoustic signal to a level that the individual can hear it properly. In the beginning, hearing aids were mechanical resonating devices with tubes and horns. They were very heavy and not suitable to wear. At the beginning of 1900s, electrical hearing aids were introduced and started to be in use rather than mechanical hearing aids. Since 1950s, hearing aids have become much smaller as seen in Figure 1.2 [2]. The first wearable hearing aid was produced after the mid 1980s. In the 1990s, programmable digital hearing aids were available and they were providing cleaner sound quality and precise fitting [3]. Nowadays, hearing aids are produced so that they provide better sound quality by reducing acoustic feedback and background noise, and increasing audibility of speech sounds. However, according to Kochgin [4], only about 50-60% of the users are satisfied with their hearing aids. This shows us that, although hearing aids were firstly introduced 100 years earlier, hearing technology is not satisfactory yet.

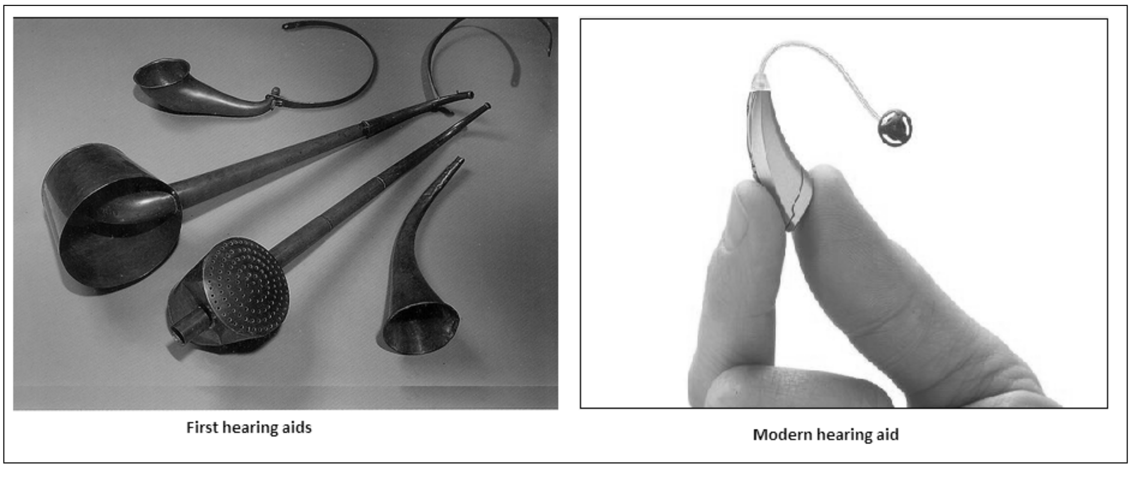

Figure 1.2 Hearing aid history.

In order to increase the usage of hearing aids, they should be fitted properly to the individual's ear. Since the development of electronic hearing aids, several procedures have been introduced for fitting them according to the degree and type of hearing loss of the patients. Basically hearing aid fitting procedure can be classified as comparative and prescriptive procedures. In comparative procedure, the patient tries lots of aids with different adjustments and tests the outcome by means of speech tests. Results can be compared with each other so that the best fitted aid is chosen. Other criteria can also be taken into account like the sound quality. Although, this fitting procedure gives the opportunity to directly test the hearing aid in situ, it requires a certain amount of knowledge and experience from the fitter. Moreover this method is not applicable to children because of being subjective. In contrast prescription procedure is objective and provides controllable result that is less dependent on the quality of the fitter. Also, they can be automated and require relatively little time. In this procedure the desired amplification, which is called "target gain", is calculated by mathematical formulas. But, the relation between calculated target gain and actual performance of hearing aid is not clear. Each individual has different ear characteristics and this can cause a deviation from the target [5].

Measurement of real ear insertion gain (REIG) is a reliable method to determine how closely a hearing aid is programmed to match a prescription target [6]. This measurement gives the amount of gain that is present at the level of the eardrum. REIG is provided by two microphones which are placed near the entrance of the ear canal and eardrum. The difference between these two microphones is defined as real-ear gain. According to Hawkins and Cook [7], prescription methods are not accurate enough. They used 12 different prescription methods, provided by hearing aid manufacturers, and demonstrated that these fitting procedures over-estimated actual real-ear gain, especially at higher frequencies. This situation necessitates using the real-ear measurements to investigate the hearing aid fitting.

Hearing aid analyzer is a platform which tests the hearing aids and performs real-ear measurements. The level of amplification is determined so that the patient receives proper hearing assistance. According to test results, a fine tuning can be done and the target gain can be achieved precisely.

1.2 Thesis Objectives

The main purpose of our study is designing a portable hearing aid analyzer, which we call PHAA throughout in this thesis, for determining the performance of hearing aids while they are worn. Ear behaves like a natural amplifier and its acoustic characteristics are individual based which affect the performance of hearing aids. Therefore, hearing aids' performance is different for each individuals in situ condition. In other words, the transfer functions of hearing aids are different in the ear than outside the ear. In our system, real-ear insertion gain measurement is used to calculate the transfer function. Calculating the transfer function enables us to offer suggestions to the audiologists for better fitting.

Our system is composed of two main parts: presenting the test signals and receiving back the test signals to calculate the transfer function. The main processor is a digital signal processor (DSP) chip which is controlled by a PC software. Also, a voice codec is used for analog to digital and digital to analog conversion. While producing the test signals and measurement setup, we have considered the international standards. After the design some measurements proving the system accuracy, adherence to the distortion obligation and system performance were examined.

We have planned to design PHAA and manufacture it. Yet, there have been no hearing aid analyzer manufactured in Turkey and we want to be the first. PHAA fulfills the national requirements in the standards and it can be marketed world wide. Although, there are big companies in the market, PHAA can compete with their products. PHAA is compact and battery powered which make the usage easier.

In the current design we have proved the concept and in near future we will make it more functional and practical.

1.3 Thesis Organization

The thesis contains different chapters explaining topics related to designing a system for insertion gain measurement.

Chapter 1 gives the general knowledge about determination of hearing loss and hearing aid fitting. Also, the hearing aid and importance of its usage are mentioned briefly. The objectives and motivation of this thesis work are clearly indicated.

In Chapter 2 a background about the thesis is presented. The nature of sound, the anatomy and physiology of hearing, hearing loss and its types, and hearing aids are explained briefly.

Chapter 3 introduces the reader to the theory on which the study is based on. It includes sections about the hearing aid fitting. The steps from hearing loss tests to hearing aid selection and fine tuning are explained.

In Chapter 4 the specifications of a hearing aid analyzer based on the international standards are explained. The requirements for test tones and real ear measurements are highlighted.

Chapter 5 reports the system design and implementation. This part starts with a simulation in MATLAB. Then the block diagram of the system is depicted and explained in detail. In the following section the components that are used in the design are introduced and hardware diagram is explained.The last section describes the software implementation. In this part one can see the main aspects of software used in this design.

Chapter 6 reports experimental results. The evaluation for conformance of the design with the relevant international standards and a comparison between the design and a commercial product is done.

Finally, the conclusions are made in Chapter 7 and an outlook for further research within this area is given. Suggestions, future work, and further topics are also presented in chapter 7.

2. BACKGROUND

2.1 The Nature of Sound

Sound is a mechanical wave that is a production of molecules vibrating in a solid, in a liquid, or in a gas medium. Most of the time, we hear the sounds which travel through the air in the atmosphere. Vibration of an object in the atmosphere causes the movement of particles around it. These moving particles cause the movement of adjacent air particles, carrying the pulse of the vibration through the air. So that, that vibrating object delivers wave of pressure fluctuations through atmosphere to the hearing system. Sound cannot travel through a vacuum.

In humans the audible range of frequencies is usually in the range of 20 Hz to 20 kHz [8]. Frequencies below 20 Hz are generally felt rather than heard and frequencies above 20 kHz can sometimes be sensed by young people. High frequencies are the first to be affected by hearing loss.

While an object is vibrating, the molecules are changing their positions in that object because of the elasticity of the object. In other words, vibration means displacement of a molecule. The total displacement can be defined as one cycle in the movement of the molecule. The total distance determines the magnitude which is called its *intensity*. It determines the loudness of sound. *Loudness* is the way of perception of magnitude. A relational scale is used to measure loudness using the unit of decibel. Firstly, Alexander Graham Bell used this unit which abbreviated as dB [9]. It is a logarithmic unit that indicates the sound intensity to the reference sound pressure level (SPL). In Table 2.1 sound pressure levels of common sounds are tabulated [10].

Sound intensity is the sound power (W) per unit area (m^2). The SI-units for sound intensity are W/m^2. The dynamic range in human hearing and sound intensity is 10^{-12} W/m^2 to 10-100 W/m^2. A more convenient way to express the sound intensity

Table 2.1
Sound Pressure Levels for Common Sounds.

Sound	db SPL	Sound	dB SPL
Rocket Launching pad	180	Jet plane	140
Gunshot blast	130	Car horn	120
Pneumatic drill	110	Power tools	100
Subway	90	Noisy restaurant	80
Busy traffic	75	Conversational speech	66
Pneumatic drill	55	Power tools	40
Soft whisper	30		

is the relative scale with reference human audible sound (10^{-12} W/m^2). Intensity level (IL) means magnitude of sound determining the sound audibility. The level of hearing in humans is hearing level (HL) which is described by the Eq. 2.1

$$\text{dB HL} = 10 \log \left(\frac{P}{RP} \right) \tag{2.1}$$

where P represents the power of a sound wave and RP is the normal hearing threshold which is accepted as the reference power (10^{-12} W/m^2).

The SPL, which is the reference pressure level for sound intensity, is in the range of 0 dB SPL and 140 dB SPL [11]. 140 dB SPL is the threshold of pain that is the point at which the sensation becomes pain. SPL is doubled in every 6 dB increments.

The power of sound waves goes as the square of the pressure. In order to measure the sound pressure, Eq. 2.2 is used;

$$\text{dB SPL} = 10 \log \left(\frac{Pr}{Re} \right)^2 \tag{2.2}$$

which is equivalent to;

$$\text{dB SPL} = 20 \log \left(\frac{Pr}{Re} \right) \tag{2.3}$$

where Pr refers to the sound pressure level (dB) and Re represents the reference sound pressure (Pa) which is equal to 20μPa.

2.2 Hearing Anatomy

Ear is the sensory organ which is responsible for hearing. It is like a biological microphone. A microphone converts vibrations into electrical signals, and the ear shows the similar characteristics with microphone such that vibrations are converted to nervous impulse. The ear is divided into three major areas: external ear, middle ear, and internal ear [12]. These areas can be seen in Figure 2.1 [13].

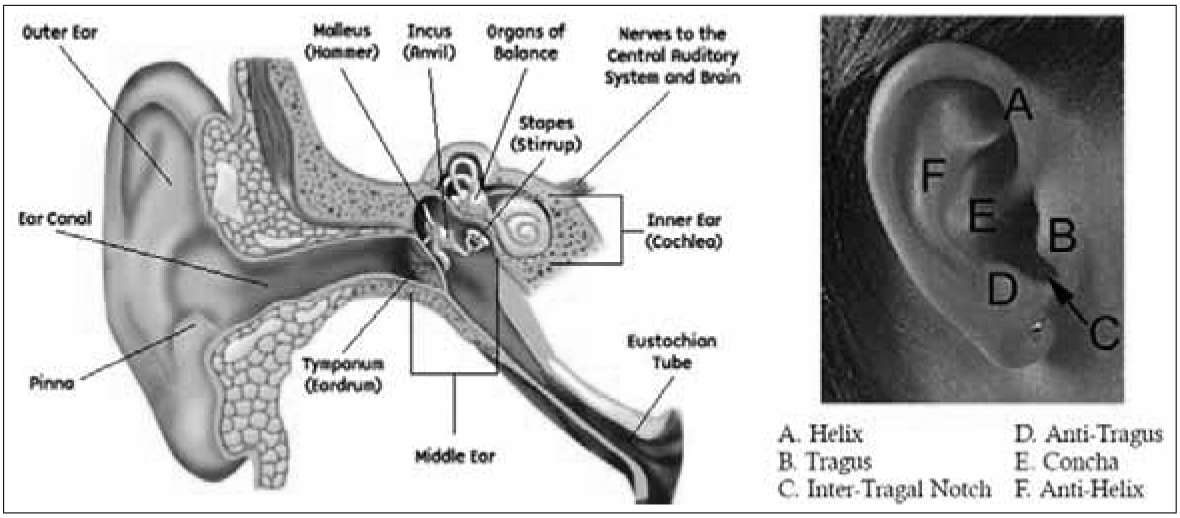

Figure 2.1 Human Ear Anatomy.

2.2.1 Outer Ear

The outer ear is the most external part of the ear. It catches sounds and channels them into the tympanic membrane. It is composed of pinna (auricle), ear canal, and surface of the eardrum. Pinna is the part of the outer ear, which collects the sounds and transmits them to the ear canal. The ear canal is about 4 cm long. It consists of hairy skin with sweat glands and oily sebaceous which together form ear wax. These hairs and wax protect the ear from infections [14]. In the last part of the outer ear, the sound is transmitted to the eardrum.

2.2.2 Middle Ear

The middle ear is an air filled space behind the eardrum (or tympanic membrane), which includes the three ear bones or ossicles: the malleus (or hammer), incus (or anvil), and stapes (or stirrup). It is connected to the back of the nose by a tube called eustachian. The beginning of eustachian tube is also in the middle ear. The outer wall of the middle ear is the eardrum and the end wall is the cochlea. Sound is conducted from the eardrum to the inner ear by three bones, the hammer, incus, and stapes. These bones are ordered so that movement of the tympanic membrane causes movement of the malleus. Movement of malleus causes movement of the incus, which causes movement of the stapes. When the movement of stapes reaches to the oval window, it causes movement of fluid within the inner ear [14].

2.2.3 Inner Ear

It is the end part of ear and also called bony cochlea because of its shape like snail shell as seen in Figure 2.2 [15]. It houses the organ of hearing which is known as membranous labyrinth surrounded by fluid called the perilymph. The beginning of the inner ear is oval window and the end of it is apex. The inner ear is composed of two important parts; organ of hearing (the cochlea) and a vestibular system dedicated for balance. Balance is beyond the scope of this work. Cochlea has a space of 0.2 of a milliliter. There up to 30,000 hair cells which transduce vibration into nervous impulses and nearly 19,000 nerve fibres which transmit signal to the brain [16].

In the middle ear, the energy of pressure waves is transformed into mechanical vibrations. These mechanical signals are propagated as waves in the cochlea and transduced to the nerve impulses to be sent to the brain . The cochlea contains three fluid filled spaces: the scala tympani, the scala vestibuli and the scala media. Scala tympani is connected to the round window and the scala tympani is connected to the oval window. They connect at the apex with the helicotrema which is responsible for pressure equalizing mechanism. The third space, scala media, surrounds the cochlea.

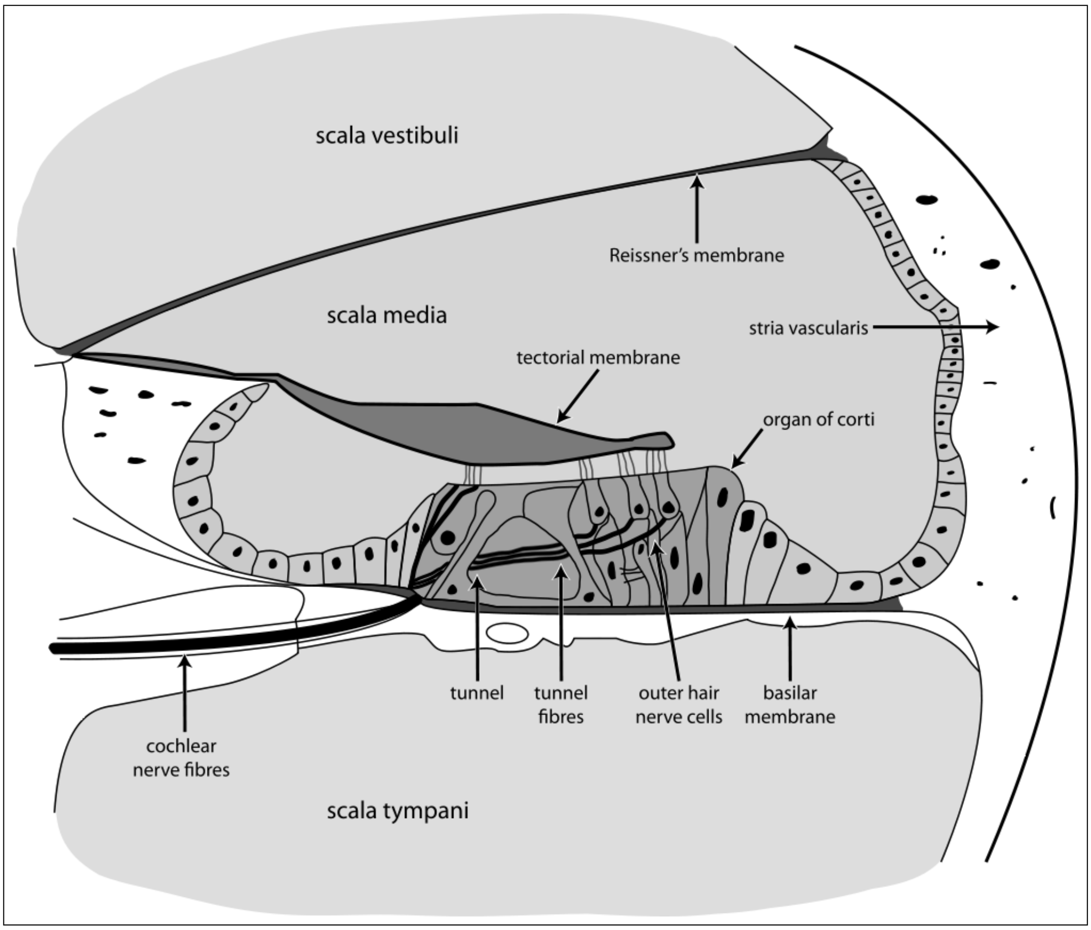

Figure 2.2 Inner Ear.

2.3 Hearing Physiology

Hearing can be summarized by the following sentences: Sounds create vibrations in the air that reach the eardrum. The vibrations in the eardrum cause the movements of tiny bones ordered as a chain. These bones press fluid in the internal ear. This pressure leads the tiny hairy cells to stimulate the neurons that give rise to impulses that travel to the brain. And then, hearing occurs.

The pinna (external part of outer ear) captures the acoustic energy and transfers it into the ear canal. Because of the shape of pinna (natural amplifier), sound is amplified. When these amplified sounds come to the eardrum, they strike the eardrum making it vibrate. Then, ossicles transmit these vibrations to the perilymph of the inner ear.

In the middle ear, acoustic energy is converted into mechanical energy. Here, acoustic energy is amplified by essentially two mechanical phenomenan. First, the area of the tympanic membrane is much greater than the stapes footplate which is connected to the inner ear. This area difference causes the increased pressure on the oval window than the tympanic membrane. Second, because of the nature of the middle ear ossicles acts as a lever and the sound is also amplified here. The total increase in sound energy reaching to the inner ear is about 30 dB.

Cochlea, the main part of the inner ear, contains bony canals filled with fluid. These canals contain sensory epithelium. The energy enters the inner ear from the oval window by the stapes. This energy turns into waves by the help of perilymph. These fluid waves travel through the cochlea and they pass the basilar membrane of the cochlear duct, causing the change of basilar membrane mechanical properties. As the waves travel along the basilar membrane, the amplitude of vibrations changes. The greatest vibration of the basilar membrane occurs at the apex with the low frequency stimuli, and at the base with high frequency stimuli. Perilymph wave displaces basilar membrane, which causes the stereocilia at the apex of both, inner and outer hair cell to be under the influence of a shearing force. This force leads to the change of the resting membrane potential of the hair cell which is subsequently forwarded to its basal end. There a synapse is built up with a dendrite from the auditory nerve. The hair cell membrane potential change is sent via acetylcholine crosswise this synapse which causes depolarization of the nerve fiber. Then, the vibrations are transduced into nervous impulses and sent to the brain [8].

2.4 Hearing Loss

Hearing loss (or deafness) can be defined as the poor ability to receive or process acoustic stimuli. It can occur suddenly or gradually. The level of hearing loss is changing depending on the cause. Also it can be temporary or permanent. There are types of hearing loss such as, congenital hearing loss and gradual gearing loss. Congenital is hearing loss which means born without hearing. Gradual hearing loss is a very common problem which might occur over time and this can affect people of all ages. Hearing loss is a very common problem in societies. 40% of people at the age of 65 and older are suffering of hearing loss [17]. The hearing loss can be classified into five groups: conductive, sensorineural, mixed, central, or functional.

2.4.1 Conductive Hearing Loss

In this type of hearing loss, the sound waves are not conducted efficiently through the outer ear canal to the eardrum and the tiny bones (ossicles) in the middle ear. As a result they are not able to stimulate the sensory cells of the inner ear. The mechanism of creating fluid wave within the cochlea does not work. In other words, the sound waves cannot be transformed into fluid wave within the cochlea. As a result, sensory cells cannot receive any stimuli. The hearing loss is about 60 dB. There are many causes which lead to conductive hearing loss such as impacted wax, external auditory canal atrecia, perforation of the tympanic membrane, ossicular discontinuity, otosclerosis and serous otitis media. Most of the conductive hearing losses are amenable to surgical correction. Performing a myringotomy fluid in the middle ear space, in serous otitis, is an example of these surgical corrections.

2.4.2 Sensorineural Hearing Loss

A sensorineural hearing loss exists when the sensory cells of the cochlea (inner ear) or the auditory nerve fibers are impaired. The transformation from acoustic energy to nervous stimuli does not occur in the cochlea. The causes of this type of hearing loss are; damage to the cochlea, aging (presbycusis), ototoxic medications and tumors such as an acoustic neuroma. Most of the time sensorineural hearing loss cannot be medically or surgically corrected. This is the most common type of permanent hearing loss. The hearing loss can be over the 100 dB.

2.4.3 Mixed Hearing Loss

Mixed hearing loss is a combination of a conductive and sensorineural hearing loss. In mixed hearing loss outer ear or middle ear may be damaged and the inner ear or auditory nerve may be damaged too.

2.4.4 Central Hearing Loss

The hearing loss which occurs in the auditory areas of the brainstem and higher levels (temporal lobe) is called central hearing loss. There is not much knowledge about the causes of central hearing loss. The people who are suffering central hearing loss can hear normally, but cannot process the auditory information (stimulus coming from the ear) efficiently.

2.4.5 Functional Hearing Loss

In functional hearing loss, there is no physiologic basis for this hearing disability. The people with this type of hearing loss are pseudo-patients, called malingerers. This is generally seen in adolescents or persons who apply for pension benefits as a result of hearing loss.

2.5 Hearing Aids

A hearing aid is a device or instrument that can amplify sound waves to help in hearing. There are three major components of a hearing aid: a microphone, an (electronic) amplifier and a receiver (speaker) as seen in Figure 2.3. The microphone picks up the sound waves and transforms them into the electronic signals. These signals are amplified and filtered in the amplifier. The amplification and filtering occurs so that the sound can be heard by the person who is wearing the hearing aid. The third component, the receiver is a kind of speaker. It transforms electrical signals to acoustic signals and delivers these signals to the ear. All hearing aids are designed to amplify the sound that reach to the eardrum [3].

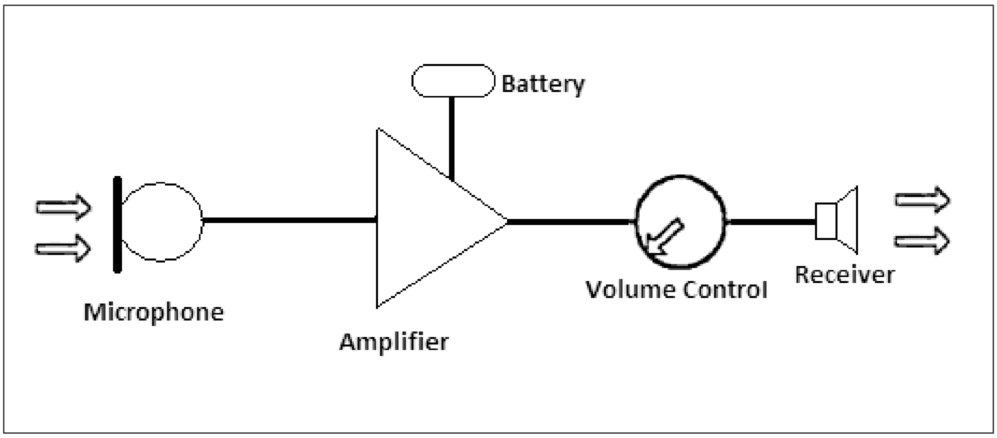

Figure 2.3 Block diagram of a simple hearing aid.

According to different types and degrees of hearing loss, there exist more than 1,000 different models available on the international market today. These can be divided into four common types that differ in size and location on the wearer.

2.5.1 Behind-the-Ear (BTE) Hearing Aids

These have a plastic housing for the microphone, amplifier, battery, and receiver that is worn behind the ear. The plastic house is connected to the earmold by a plastic tube. The earmold extends into the ear canal. Some of them have volume controls. This model is developed decades ago and many users think that they are unattractive and out of date. But, they are more durable than other types and easily interchangeable if they need to be serviced. Also, they can provide better sound quality and tend to be more reliable [18].

2.5.2 In-the-Ear (ITE) Hearing Aids

In-the-ear aids contain custom-made housings that hold all the components. These devices fit into the ear canal without visible wires or tubes. It is not possible to control volume in these devices. Therefore, they are only for people with mild hearing loss. Some users think that it is easier to put on or take off these devices than BTE aids. However, ITE aids are more expensive then BTE, because they are custom made [18].

2.5.3 In-the-Canal (ITC) Hearing Aids

In-the-canal aids fit far into the ear canal, and a smaller portion facing out into the outer ear. ITC aids are very expensive, but they have much superiority such as providing better acoustic and easiness to maintain. Because of its small size it is difficult to insert the battery and adjusting the volume is also harder. Its microphone is positioned for cutting down on wind noise and providing more natural sounds [18].

2.5.4 Completely In-the-Canal (CIC) Hearing Aids

Completely in-the-canal aids fit completely in the ear canal. CIC aids are the smallest size of hearing aids and hardly visible. They are custom designed to fit wearer's ear and provides the most natural hearing process. However the physiology of some wearers (i.e., a very narrow canal) may make it harder for them using CIC aids [18].

3. HEARING AID FITTING

“Hearing aid fitting” is the process of selecting, adjusting and fine-tuning of a suitable hearing aid for the patient [19]. An ideal fitting not only provides a normal hearing level to the patient but also keeps a natural, pleasant sound quality with minimal distortion. Generally, the first step involves determining the hearing loss characteristics by an audiometer measurement device. These characteristics give the amount of hearing loss at varying frequencies from 125 Hz to 8 kHz. In the second step, theoretical prescription method is used to calculate the target insertion gain needed to compensate the hearing loss. In order to decrease rejection rate in hearing aids and increase patient comfort and hearing ability, the hearing aids should be fitted accurately with an accurate prescriptive gain [20, 21]. Therefore, a hearing aid should be programmed according to the most suitable prescription method. After programming the hearing aid it needs to be evaluated for individual usage. The electroacoustic properties of hearing aids, which depend on individuals, are useful but they are not enough to determine how hearing aids are working in a specific real patient’s ear. Besides the prescription methods, extra patient specific methods should be used for the hearing aid fitting. Real ear insertion gain measurement is the best and objective procedure to find the actual insertion gain in situ working condition. For example, a hearing aid should be inserted into the ear and the measurements need to be collected in order to fine tune the amplification of hearing aids and therefore completing a suitable hearing aid fitting.

In this chapter, first we will provide an overview for audiometers and prescription methods. And then real ear insertion gain (REIG) measurement will be described in detail. In this thesis, an REIG measurement device will be designed and developed for calculating the transfer function of the hearing aid under test.

3.1 Audiometers

An audiometer is an instrument to measure the hearing loss. Basically, during this measurement, test sounds at different frequencies are generated and delivered to the patient. According to the patient's response, the level of hearing loss is determined. These generated sounds are very specific which makes the audiometer valuable. A primitive audiometer should perform the basic audiological tests, which involve finding out how much intensity is needed for a patient to hear pure tones at different frequencies. A basic audiometer, which is shown in Figure 3.1 [22], presents sound signals at certain frequencies, controls the intensity of these signals, and delivers them to the patient to get his/her response.

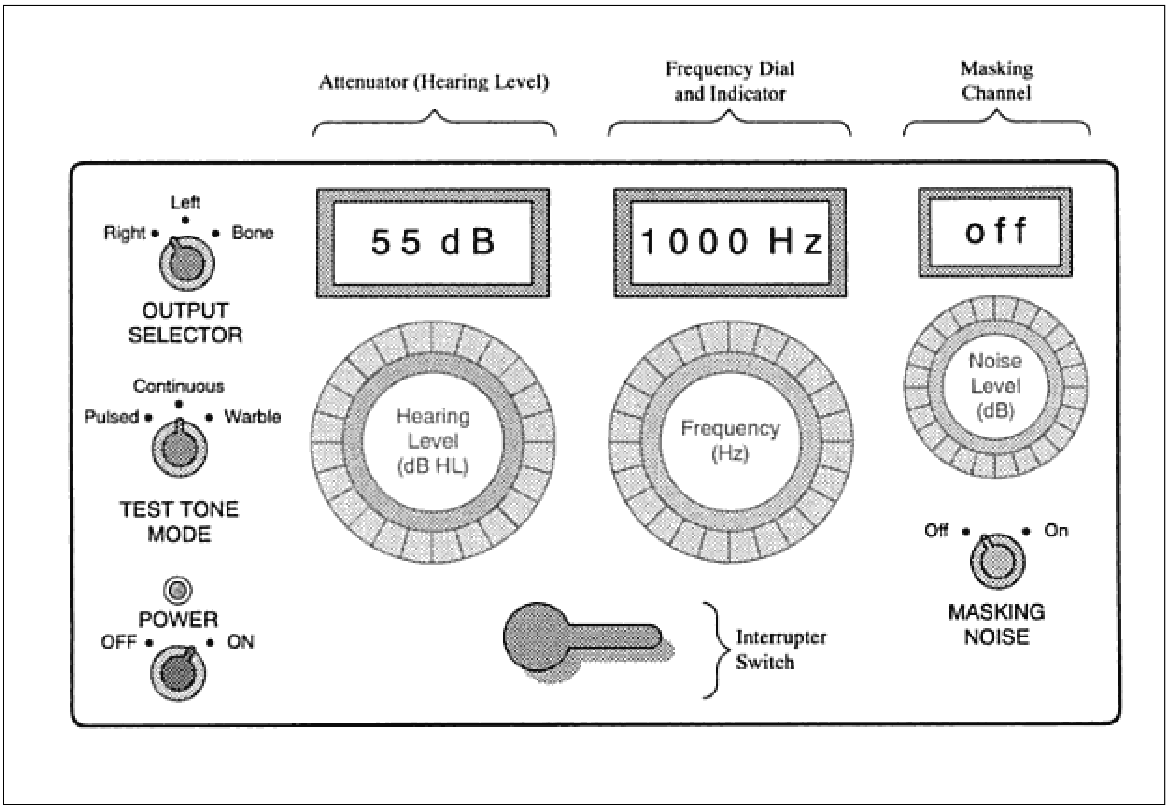

Figure 3.1 Components of Audiometer.

The sound signals generated by the most audiometers include the frequencies of 125, 250, 500, 750, 1000, 1500, 2000, 3000, 4000, 6000, and 8000 Hz. An attenuator or a hearing level controller are used to change the intensity of the test signals. While the basic attenuators can be programmed in 5 dB granularity, more sophisticated models can use 1 dB, 2 dB, or other step sizes. Generally, the range of intensity for hearing loss testing using audiometers is from -10 dB Hearing Level (HL) to 115 dB HL in air-conduction and -10 dB HL to 70 dB HL for bone-conduction. The generated test signals are delivered by special speakers or headphones to the patient [22].

There are various types of audiometers according to the types of signal they produce (pure-tone or speech), frequency range which they operate (limited, normal, or high frequency), the method in which hearing is tested (manual, automatic, or computer controlled), the purpose which they are being used (clinical, diagnostic, industrial, or screening), the number of independent audiometers in one unit (one channel, two channel, or channel and a half), and whether they are portable or not [23]. However, a different classification of audiometers is made by ANSI S3.6-2004 Standard according to the type of signal they produce, mode of operation, and range of auditory function tested on the basis of minimum required features, frequencies, and maximum hearing levels (HLs) they have.

We will give more detail about the most common types of audiometers which are pure-tone, automatic, and speech audiometers.

3.1.1 Pure-tone Audiometers

This type of audiometer is the most basic one which includes pure-tone generator, interrupter switch, amplifier, attenuator, output selector switch, and earphones. The generator produces tones at certain frequencies. The interrupter switch is used for turning on or off the tone which is amplified and routed to the attenuator and attenuated to the desired HL by the HL controller. The output selector switch is used to present the tone to the right or left earphone. Pure-tone audiometers are also classified by ANSI S3.6-2004 according to the required features, frequency and maximum HLs as type 1,2,3, and 4. The first type is the most improved one capable of producing tones at the frequency of 125 Hz to 8000 Hz.

3.1.2 Automatic Audiometers

This kind of audiometer is operated by the patient, enabling him to control the intensity of tone presented to him and thus track his own hearing levels. When the patient hears a tone, he/she provides a positive feedback by depressing the hand switch and the audiometer lowers the level of tone. When the tone is not audible, the listener releases the hand switch, and the audiometer automatically increases the level of tone. Over a period of time, the patient can track his hearing thresholds. ANSI S3.6-2004 types automatic audiometers use the same features as the pure-tone audiometers in terms of frequency, and maximum HLs.

3.1.3 Speech Audiometers

Speech audiometers are mostly the same as the pure-tone audiometers but, they have also a microphone, an external input, and a monitoring meter. The microphone is used for live-voice speech testing, and external input is used for connecting to the playback devices to make the recorded speech test. In current practice, speech audiometer is being used with the pure-tone audiometer in one unit called a clinical or diagnostic audiometer [23]. Speech audiometers are classified as A, B, and C type by ANSI S3.6-2004. Type A has the most required features while the type C has the least.

3.2 Prescription Method

Prescription method is a mathematical formula which takes a hearing threshold as an input and provides a target gain for a specified frequency as an output. In this method, in which amplification is done selectively (providing specific gain as a function of frequency-specific hearing impairment). In 1970s and 1980s, it was firstly used by amplifying the sound linearly. The hearing aids which are programmed according to these linear methods called linear hearing aids. In these devices, a gain is adjusted using only frequency dimension. NAL-RP and POGO are such validated algorithms which use linear amplification. In 1986, Bryne and Dillon [24] developed the most throughly validated method which is entitled the National Acoustics Laboratory Revised (NALR) procedure and this was the origin of many new prescription methods. Then in 1990s, using the improvement of multiband compression technologies, advanced fitting prescription methods, which involves frequency and level dependent gains, and threshold based formulas, are introduced. These are classified as non-linear amplification methods . NAL-NL1 [25], DSL7(i/o) [26] , and Cambridge [27] are examples of such methods used in non-linear hearing aids. These new procedures offer new decisions such as, number of compression channels, compression size, and how to rapidly modify the signal for the best results [28]. Recently it has been a common practice to extend these validated methods to use new manufacturer's proprietary algorithms [29].

Prescription method is a hearing aid fitting method which is not an individual based. Therefore, patient's preference or psychophysical skills are not involved in this fitting process. Prescription formulas are based on the audiological incides, such as thresholds, most comfortable listening levels (MCLs), and loudness discomfort levels (LDLs). Furthermore, it is assumed that a hearing aid wearer has ear canal characteristics that resemble those of an average ear. Unfortunately, neither of these assumptions and estimated MCLs, LDLs, or thresholds can be met with the actual, patient specific, values. For example, there can be variations in the auditory threshold value from one measurement to another or the middle ear resonance and impedance characteristics may be atypical which leads prescribed frequency-gain response inappropriate [30].

In 2005, Mueller carried out a research study by using National Acoustic Laboratories' Revised (NAL-R), Revised for severe/Profound (NAL-RP), and National Acoustic Laboratories-Non-Linear 1 (NAL-NL1) prescription methods as reference and compared eleven different prescription methods. He observed that, there are some differences among these methods and only applying the prescription method is not enough for fitting. He concluded that real ear insertion gain measurement is needed for a better fitting [29]. Elbering [31] claims that, because of big variations of individual ears, ear canals, and external ears, it is impossible to find an optimal common settings for hearing aids.

Clearly, there is a need for new approach for an ideal fitting which should be individual based. Real ear insertion gain measurement is an effective way to figure out the ideal adjustment of hearing aids for a better fitting.

3.3 Real Ear Insertion Gain Measurement

Hearing aid manufacturers propose different types of fitting methods to determine the target frequency-specific gain and there is no consensus about which method is the most applicable [7]. Killion carried out a research study in 2004, and he examined the gain of 16 different hearing aids which are programmed using proprietary algorithm. He concluded that, average programmed gain was much less than normally prescribed by validated procedures [32]. Also, Hawkings and Cook assert that because of the individual coupler response for flat insertion gain (CORFIG) correction variations, the simulated REIG and measured REIG would be different [7]. As a result, the prescriptive gain stated in the manufacturer's software may be different from the gain in the real ear.

The electroacustic performance of hearing aid in situ is very important for fitting and it varies according to patient's different ear shape, ear canal, and hearing sensitivity. In situ measurement enables to figure out the actual, individual based, characteristics of hearing aids. Using an inserted probe microphone in the ear canal for real ear measurement is the most widely used individual based method for verification of fitting. According to the results obtained, fine tuning can be done for ideal fitting. This type of measurement is called real ear insertion gain measurement.

A typical real-ear measurement system is shown in Figure 3.2. A loudspeaker is used to present the test signals which are collected by the special placed microphones. The main processor controls the loudspeaker and analyzes the sound signals received by the microphones. The main components of the measurement system are two mi-

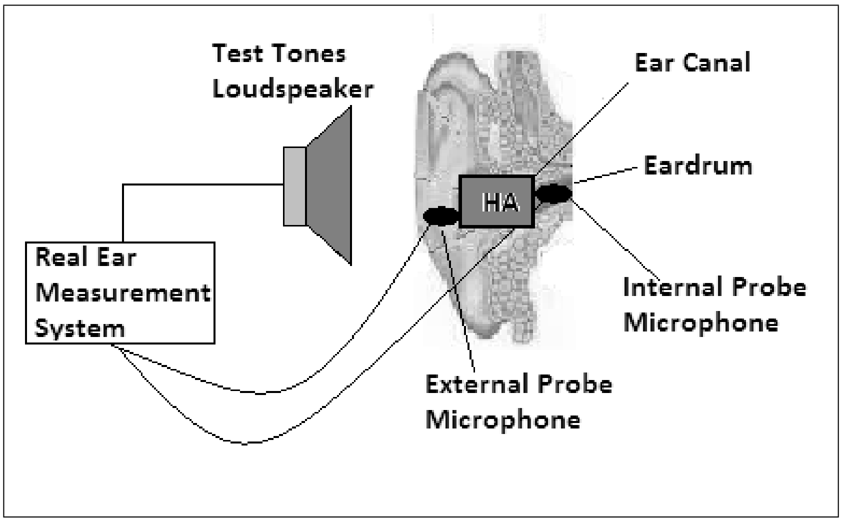

Figure 3.2 A Typical System for Real Ear Measurement.

crophones which are mounted on the head. One of the microphones is called reference microphone and another is called probe microphone. The first one picks up the sounds very close to the hearing aid's microphone just before they enter the hearing aid and the latter one pick ups the sounds very close to the eardrum. Two different measurements are done in this system. One is without hearing aid and the other is with the hearing aid. Without the hearing aid, the signal is generally 12 - 17 dB amplified close to the eardrum because of the ear canal resonance. This measurement is called real-ear unaided response (REUR). In the second measurement, the hearing aid is inserted into the ear and then test signals are presented by the loudspeaker. The reference microphone picks up the sounds from the outside and the probe microphone picks up the sounds which are amplified by the hearing aid. The difference is calculated by the processors and this measurement is called real-ear aided response (REAR). To find the true gain provided by the hearing aid, unaided response is subtracted from the aided response [22]. This method gives the true gain which is called insertion gain or real-ear insertion gain (REIG) as given in Eq. 3.1.

$$REIG = REAR - REUR \tag{3.1}$$

This method wss firstly used in Sweden for fitting hearing aids in children. In 1980 Harford presented a method for measuring the insertion gain. He suggested to use a microphone inserted in the ear canal and connect the microphone to the measuring equipment via a thin wire [33]. This wire passed between earmold and skin. A year later, Laudrisen and Günthersen used a microphone connected to a soft tube inserted in the ear canal to measure the insertion gain [33]. Sweedish Hearing Aid fitting-institutes started to measure the insertion gain by using the pure tone in 1980 [34] and fine tuning was done according to the measured results for an ideal fitting.

In PHAA we determine the hearing aid working condition in terms of frequency and gain while it is inserted into the ear canal and calculate the transfer function of the hearing aid under test. To do this, we measure the REAR values at different frequencies and sound signals intensities. According to the results, we calculate the transfer function and provide suggestions to the audiologists for better fitting.

4. STANDARDS

For hearing aids, standards report a uniform way of testing and describing how hearing aids perform. For example, if all audiometers are calibrated to the same standard, there will be no difference between the measurements taken in different clinics. Compliance with the standards is an obligation in many counties (i.e. Turkey). Therefore, the requirements described in the standards should be satisfied by the manufacturers. In PHAA, our aim is to generate the specific test tones (e.g., similar to pure-tone audiometers) which meet the requirements in IEC 60645 standard. Besides, we setup an environment to make real ear insertion gain measurement by complying the standards IEC 61669 and ISO 12124.

4.1 Standard for Test Tones

As the technological development increases in hearing measurement, the audiometers become more available in a wide range. Many manufacturers are designing new types of audiometers and this makes an obligation to obey some standards in producing different parts of audiometers. Audiometers are composed of different functional units which can be used in different audiometric equipment. By specifying the functional units, it will be possible to specify the performance of these equipments. IEC 60645 standard contains a number of parts and first one of them is about requirements for pure tone audiometers.

Pure-tone audiometers are used for determining the hearing levels, and IEC 60645-1 covers particular requirements for these audiometers. This standard also covers the general requirements for audiometers [35]. The main goal of this International Standard is to be sure that;

- The hearing tests, especially in the frequency range 125 Hz to 8000 Hz, give valid results;

- The measurement results presents a valid comparison between the hearing of the ear tested and the reference threshold of hearing;
- Audiometers are classified according to the frequency range of signals they produce, the method they operate, or the complexity of the range of auditory functions they test.

In the Section 2, normative references are written which constitute provisions of this part of IEC 60645.

This International Standard also contains some useful terms and definitions in Section 3, which are used in measurement process. "Test signal sources", which is our interest, is described in Section 6. The Sub-Section 6.1 describes the "pure tones"; and specifically, we will focus on this Sub-Section 6.1 of the IEC 60645 standard.

4.1.1 Frequency Range and Hearing Level Range

Fixed frequency audiometers shall have test frequencies for which the minimum range of hearing level values is indicated in Table 4.1 [10, 35].

4.1.2 Frequency Accuracy

For fixed audiometers, the frequencies shall be in the tolerance of $\pm 1\%$ for type 1 and 2 audiometers, and $\pm 2\%$ for type 3 and 4 audiometers.

4.1.3 Total Harmonic Distortion (THD)

The maximum total harmonic distortion shall not exceed the tabulated values in Table 4.2 [35].

Table 4.1
Minimum number of frequencies to be provided and the minimum range of values of hearing level for fixed frequency audiometers.

Hearing levels - dB [1]							
	Type 1		Type 2		Type 3		Type 4
Frequency Hz	Air	Bone	Air	Bone	Air	Bone	Air
125	70	-	60	-	-	-	-
250	90	45	80	45	70	35	70
500	120	60	110	60	100	50	70
750	120	60	-	-	-	-	-
1000	120	70	110	70	100	60	70
1500	120	70	110	70	-	-	-
2000	120	70	110	70	100	60	70
3000	120	70	110	70	100	60	70
4000	120	60	110	60	100	50	70
6000	110	50	100	-	90	-	70
8000	100	-	90	-	80	-	-

[1] Maximum hearing level to be at least equal to the tabulated values. Minimum hearing level to be -10 dB for types 1 to 3 and 0 dB for Type 4. For circumaural and insert earphones the maximum hearing levels may be 10 dB lower over the frequency range 500 Hz to 8 kHz.

4.1.4 Rate of Frequency Change

One rate of frequency shall be one octave per minute in the mode of continuous sweep frequency in automatic recording. If the frequency is fixed, a minimum of 30 s shall be allowed at each frequency [35].

4.2 Standards for Real Ear Measurement (REM)

The performance characteristics of hearing aids while they are worn differ from being in simulated ears, due to different individual ears. The measurements which take into account these differences, such as acoustical influence and acoustical coupler, on the performance of hearing aids are important in the fitting of these devices.

Table 4.2
Maximum permissible acoustic THD, expressed in percentage of sound pressure or vibratory force for supra-aural, circumaural, insert earphones and bone vibrators.

	Air conduction			Bone conduction		
Frequency Range (Hz)	125 to 250	315 to 400	500 to 5000	250 to 400	500 to 800	1000 to 4000
Hearing level (dB)	75	90	110	20	50	60
Total Harmonic Distortion (%)	2.5	2.5	2.5	5.5	5.5	5.5
[1] Maximum output level of audiometer. For circumaural insert earphones the hearing level shall be 10 dB less than the levels specified in the table.						

"Real-ear measurement" is a method which is performed for determining performance characteristics of hearing aids in actual use. The accuracy and repeatability of these measurements depend on many factors such as, sound field, the test environment, the nature of the test signal, the hearing aid under evaluation, the method of test signal control, the location of the sound source, the nature of the data acquisition, analysis and presentation as well as the degree of subject movement permitted [36] [37].

The ISO 12124 Standard contains some main definitions, test environment and test methods, and defines the necessary reference points to be used for the measurement of real-ear acoustical characteristics of hearing aids. IEC 61669 Standard specifies the general requirements for test equipment which is designed for measuring the real-ear acoustical characteristics of hearing aids. This international standard also contains some main definitions.

Test signal characteristics, sound measurement requirements, and subject related requirements are the main parts that are observed in PHAA. These are the most important requirements that are needed for an REM. Test of electromagnetic compatibility (EMC), environmental conditions, or safety requirements are not examined because of difficulties in testing and luck of measurement system.

4.2.1 Test Signal Characteristics

Test signals may have narrow-band (pure tone, warble tone, narrow-band noise etc.) or broad-band (composite, speech weighted, etc.) characteristics. In PHAA pure tones are produced for measurements. The level of test signals should be in the range of 50 dB to 90 dB with at least 5 dB steps and the level of signal which is measured at the inlet to the reference microphone shall be within the 3 dB of true value. Also, the test signal should be controlled to within 3 dB of the intended test signal level. The frequency of the test signal should be in range of 200 Hz to 6000 Hz with maximum 2% deviation. The harmonic distortion of the test signal shall not exceed 2% [36, 37]. All these requirements for test tones are accomplished in PHAA.

4.2.2 Sound Measurement Requirements

The probe microphone measurement shall be within 3 dB of the true value over the frequency range 200 Hz to 6000 Hz. Also ambient background noise shall not alter the test result by more than 1 dB at any frequency. Removal of the test signal should cause the level at the field reference point to drop at least 10 dB [36, 37].

4.2.3 Subject Related Requirements

The subject test position is shown in Figure 4.1 [36]. He/she should be comfortably seated upright at the subject test position. It is recommended that the subject be located at a minimum working distance of 0.5 m and at an azimuth angle of sound incidence of 0°and 45°and elevation angle of sound incidence of 0°. The subject shall be instructed to remain silent and avoid unnecessary movements during the measurements. The measurement point in the subject ear shall be chosen such that ±2 mm change in its position shall produce a change of less than 2 dB in the measurement frequency range. For unoccluded measurements, this will generally require a distance of 6 mm from the eardrum. Occluded measurements require that the measurement point be at least 5 mm beyond the hearing aid speaker [36].

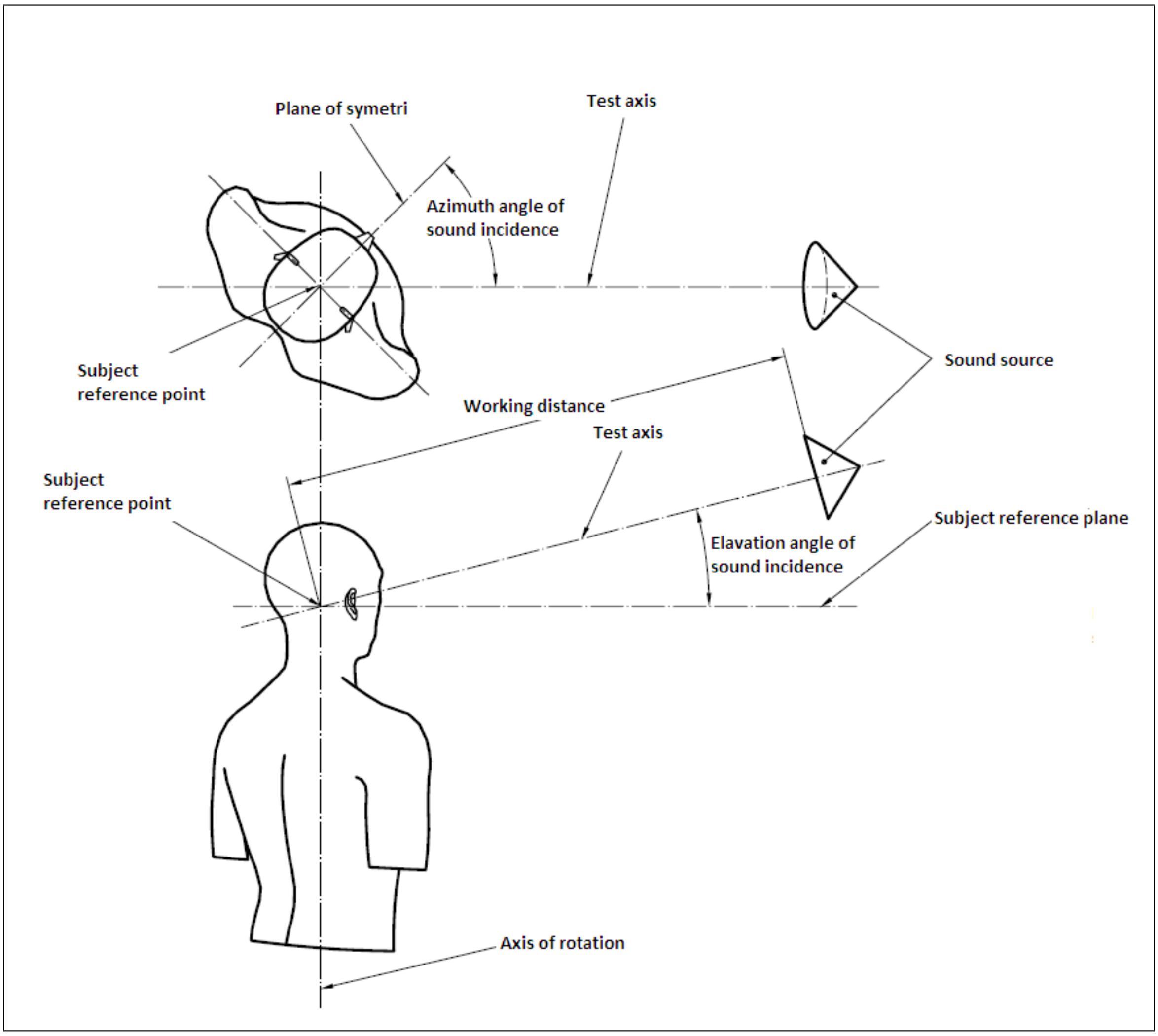

Figure 4.1 REM Test Arrangement.

5. SYSTEM DESIGN AND IMPLEMENTATION

As explained in the 3^{rd} chapter, insertion gain measurement is vital for hearing aid fitting. Hearing aid's acoustical characteristics may change in situ working conditions because of the ear's unique, patient specific characteristics. In other words, the transfer function of a hearing aid is different in the ear than outside the ear. Assume that the transfer function of a hearing aid is F_{ha} without the ear effect, and F_{iha} when it is inserted into the ear canal. As mentioned previously, because of the natural ability, the ear behaves like an amplifier. To consider this fact, we need to take into account the ear's transfer function, F_{ear} (See Figures 5.1 and 5.2). Therefore, the actual transfer function of a hearing aid in situ is F_{iha}, which is the multiplication of F_{ear} and F_{ha}:

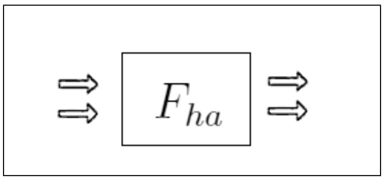

Figure 5.1 Hearing Aid Transfer Function.

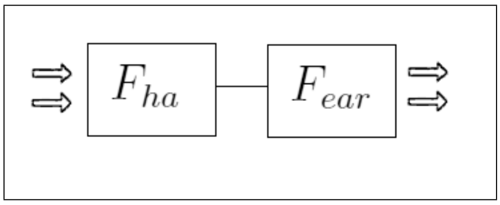

Figure 5.2 Actual Transfer Function of Hearing Aid.

$$F_{iha} = F_{ear} \cdot F_{ha} \tag{5.1}$$

We propose a system that calculates the transfer function (F_{iha}) to inform the audiologist the deviation from the prescribed hearing aid function for a better fitting. Two main parts of PHAA to calculate the actual gain includes generation of pure tones and receiving the tones back from the microphones.

The proposed system, shown in Figure 5.3, composed of two main parts that are Sinusoid Generation and Insertion Gain Computation. In the first part, we aim to generate the test tones in an accurate way to present to the patient's ear. And, in the second part, insertion gain measurement is done by receiving back the previously sent test tones. The microphones, named as MIC1 and MIC2 in the figure, are intended to place before and after the hearing aid while doing the insertion gain measurements.

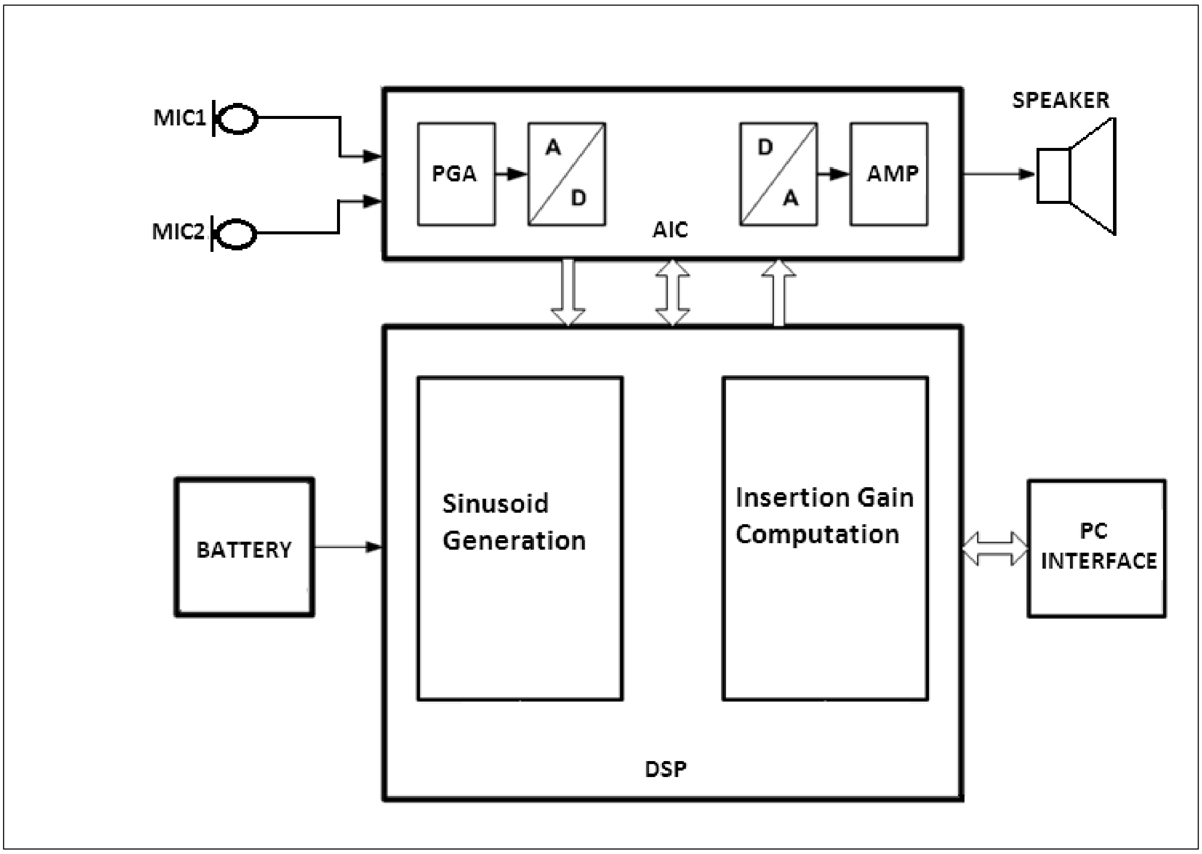

Figure 5.3 Block Diagram of the System.

To meet the requirements specified in the standards for pure tones generation (IEC 60645-1) and insertion gain measurements (ISO 12124 and IEC 61669) makes PHAA challenging. The dynamic range for pure tones is from 40 dB to 90 dB. Also it is an obligation to attenuate the intensity by the 5 dB step size. To meet this condition, a very precise controllable attenuator is needed. In PHAA, an audio interface codec AIC33 attenuates the signal in 5 dB steps. Keeping the source test signals within the limits of THD is the most challenging requirement to meet. AIC33's improved digital analog converters (DACs) adresses this challenge. They have digital interpolation filter and delta-sigma modulator with analog reconstruction filter. They provide us to generate test signals with very low THD. Moreover, a special design is needed to meet the frequency accuracy requirements in the standard. For fixed audiometers, the frequencies shall be in the tolerance of ± 1 for type 1 and 2 audiometers, and ± 2 for type 3 and 4 audiometers. To obtain desired frequencies from 125 Hz to 16000 Hz, in the limits of frequency deviations, DSP processors superiority is useful.

The digital pure tones are generated according to IEC 60645-1 standard. The generated digital pure-tones in DSP are sent to the AIC to transform into analog signals with a desired gain. Then they released to the test environment by a speaker which is powered by batteries. The two microphones are used to gather the sounds before and after the hearing aid, which will be used to compute the transfer function of a hearing aid by DSP.

Before the system setup, a simulation is done in MATLAB tool to make sure that the desired signal values (meet the standards' requirements) are possible to obtain. In this Chapter, first the simulation will be explained in detail and then the DSP processor and AIC codec which are used in our system will be described.

5.1 MATLAB Work

In pure tone audiometers, the source signals are special sine waves in certain frequencies. These signals should meet the requirements of standard IEC 60645-1. THD and frequency accuracy are the main two requirements for these signals.

Firstly, in order to produce these source signals and check whether they satisfy those two requirements in simulation, a suitable tool should be chosen. MATLAB program is suitable extensively used for mathematical operations and signal generations. It is capable of handling advanced logarithmic functions and Fast Fourier Transform (FFT) of a signal, which are the most needed operations in this study. Therefore, MATLAB is chosen for production and test of the signals.

One of the main requirements for the source signals is frequency accuracy. In the standard IEC 60645-1, it is stated that the test signals should be at the frequencies of 125, 250, 500, 750, 1000, 1500, 2000, 3000, 4000, 6000, and 8000 Hz with the accuracy of mostly 2.5% as depicted in Table 4.2. In order to work with a wide range frequency of 16000 Hz is also included in PHAA. According to the Nyquist Sampling Criterion, frequency sampling must be at least twice faster than the highest signal frequency, as given in Eq. 5.2.

$$F_s \geq 2f_{max} \tag{5.2}$$

Although a 32 kHz sampling frequency is enough to provide the criterion (maximum fundamental frequency is 16 kHz), 48 kHz sampling frequency is selected to obtain higher signal accuracy. This sampling frequency is also compatible with the processors used in PHAA. The signals are generated at the required frequencies with the sampling rate at 48 kHz by $sin()$ function of MATLAB. This makes it possible to generate a pure sine wave in an accurate way. One solution to get this accuracy is computing the FFT of the signal and checking whether the signal is on the desired frequency. In Figure 5.4 one can see that there is no any harmonics that is worth.

In order to analyze the sampling rate effect on the signal, comparison of two sine waves with two different frequencies both sampled with the same sampling rate are shown in Figures 5.5 and 5.6. Frequency of the sine wave in Figure 5.5 is 1 kHz and frequency of the second sine wave in Figure 5.6 is 16 kHz, where both having the same sampling rate (f_s) at 48 kHz. Samples per one period (spp) can be found as:

$$spp = f_s/f_0 \tag{5.3}$$

Therefore, one period is represented by 48 samples for 1 kHz sine wave, however, 3 samples are used to represent 1 period of 16 kHz signal.

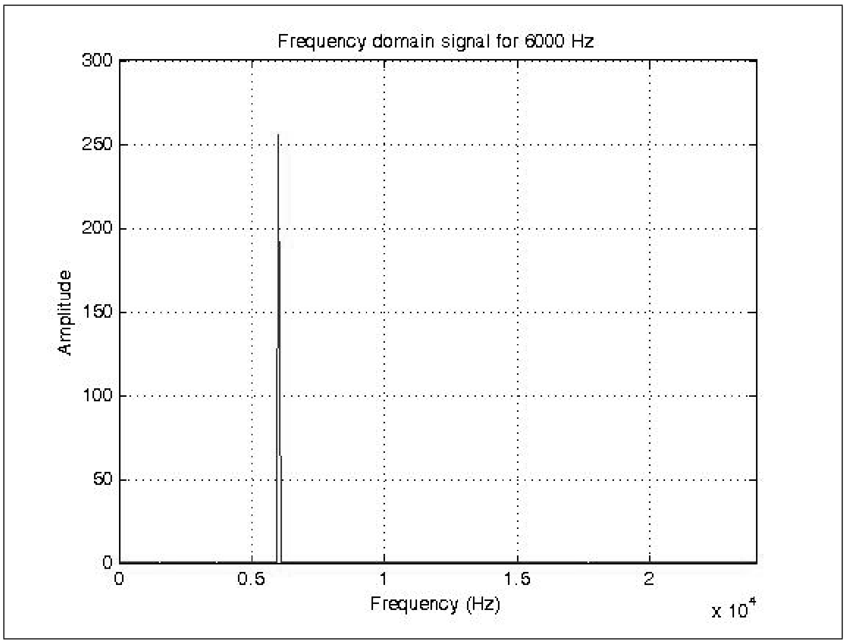

Figure 5.4 FFT of 6000 Hz sine signal in MATLAB.

Another major requirement for the source signals is to meet the total harmonic distortion requirement, which is at most 2.5% of the signal as shown in Table 4.2. THD can be expressed as the given Eq. 5.4 that is the ratio of the sum of the powers of all higher harmonic frequencies to the power at the first harmonic, or fundamental frequency.

$$THD = \frac{P_2 + P_3 + P_4 + ... + P_\infty}{P_1} = \frac{\sum_{n=2}^{\infty} P_n}{P_1} \tag{5.4}$$

This can be equivalently written as

$$THD = \frac{P_{total} - P_1}{P_1} \tag{5.5}$$

It is hard to take some measurements in the time domain, while it is easier in frequency domain, such as harmonic distortion. Examining the signal in frequency domain makes it possible to see the harmonics and amplitudes. FFT can be used for analyzing the signals in the frequency domain. Typical syntax in MATLAB for FFT is $fft(x, N)$ where x is the signal $x[n]$ which is going to be transformed and N is the

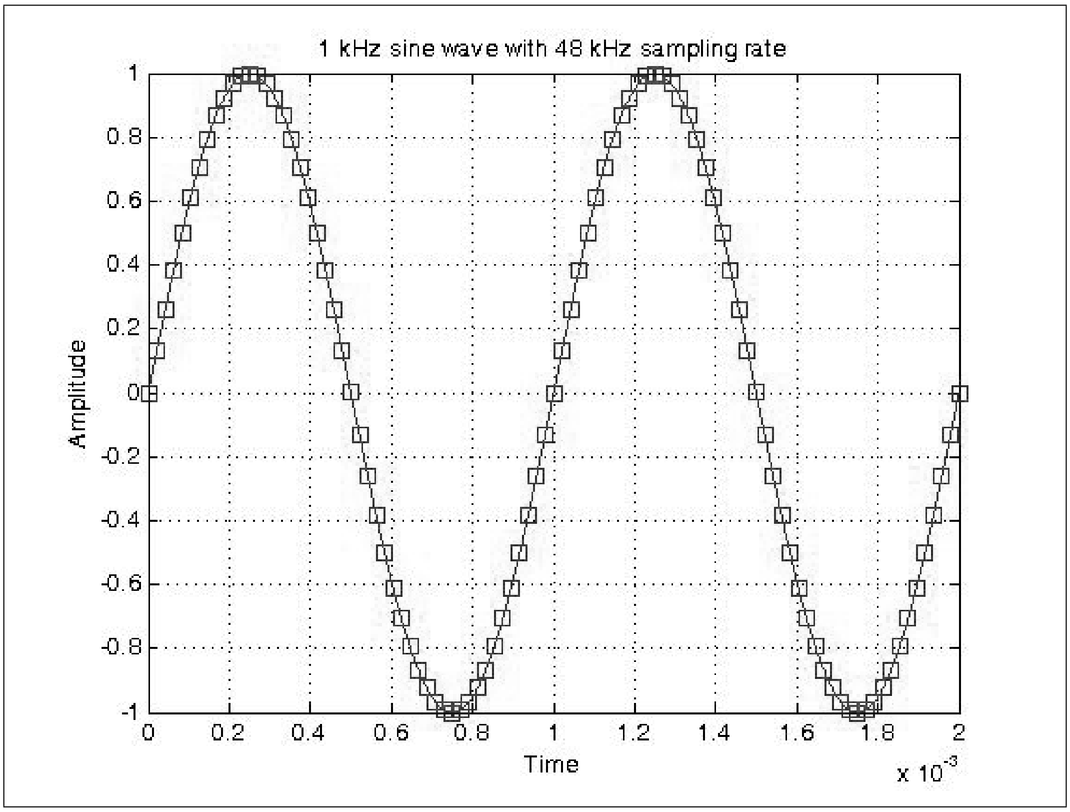

Figure 5.5 1kHz sine wave with 48 kHz sampling rate.

number of points in FFT. N is chosen to be the same length with the analyzed signal in our simulations. The THD results of the signals at test frequencies converges to 0 in MATLAB simulations, where the THD calculation is done by using the Eq.5.4. However, these simulated results would not represent the results in real world. The main reason for this difference is the effect of the limitations of analog reconstruction filters, which are used in the output of the ADCs. The process of reconstruction, also known as interpolation, produces a continuous smooth analogue signal from a digital input. In this process a reconstruction filter that is actually a low-pass filter is used to prevent aliasing (means distortion). While this filter diverges from being ideal, distortion will be higher. For a more realistic approach, generated test signals in simulation are played in a computer to measure the THD and frequency accuracy for each desired frequency. The produced digital signals are constructed into analog signals in the PC sound card and presented to the environment from the sound card output (headphone jack). Then these signals, pure sine sound waves, are measured by using an analyzer called $MinilyzerML1$. Measurement system design is shown in

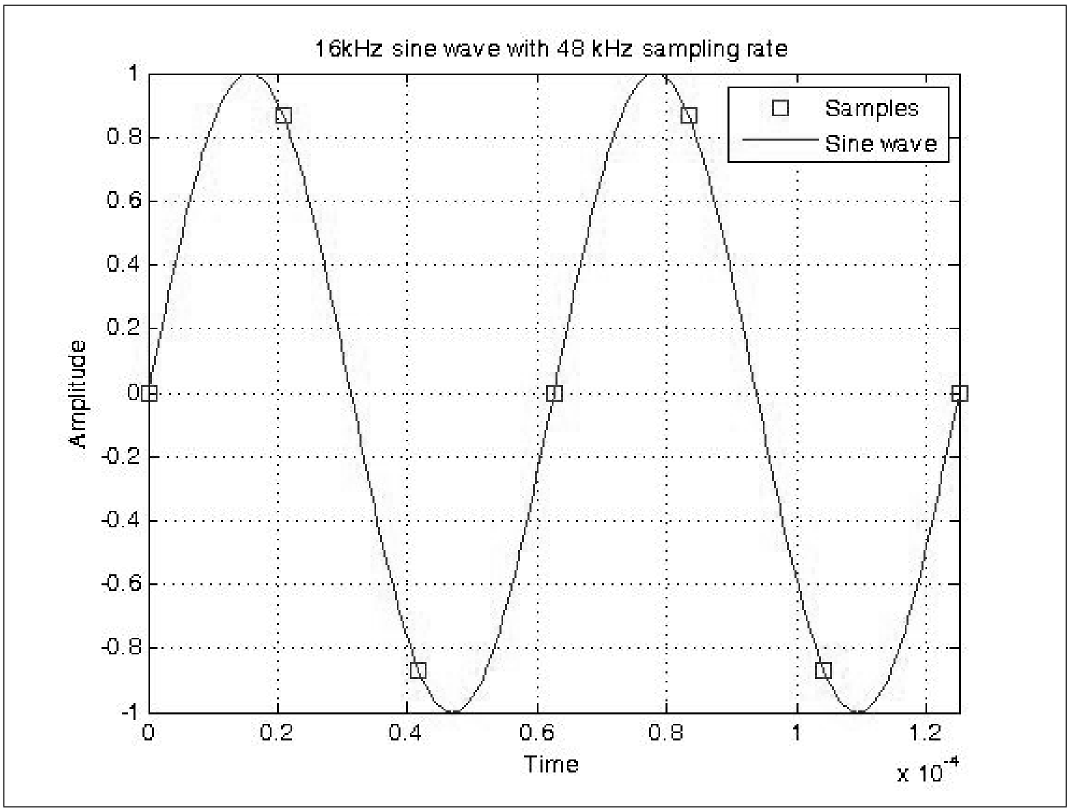

Figure 5.6 16kHz sine wave with 48 kHz sampling rate.

Figure 5.7 and the results are tabulated in Table 5.1.

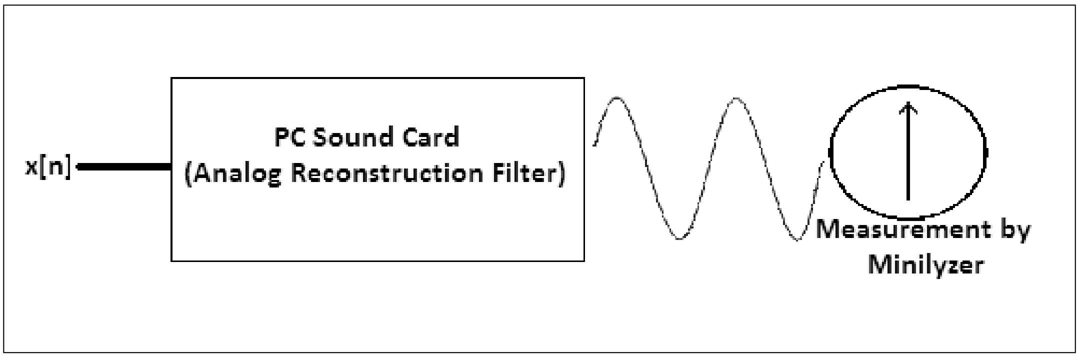

Figure 5.7 THD Measurement by Minilyzer.

The results obtained in MATLAB and measured by analyzer show that, produced source digital signals in simulation fulfill the requirements of the standard IEC 60645-1 in means of THD and frequency accuracy when we reconstruct the signal by using the PC sound card. While the THD converges to 0 in MATLAB, here it goes

Table 5.1
Measured THD and Frequency.

$TargetFreq.$	$MeasuredFreq.$	THD	$TargetFreq.$	$MeasuredFreq.$	THD
125 Hz	125 Hz	1.35	2 kHz	2 kHz	0.20
250 Hz	250 Hz	0.82	3 kHz	3 kHz	0.19
500 Hz	500 Hz	0.64	4 kHz	4 kHz	0.13
750 Hz	750 Hz	0.49	6 kHz	6 kHz	0.07
1 kHz	1 kHz	0.39	8 kHz	8 kHz	0.12
1.5 kHz	1.5 kHz	0.29	16 kHz	16 kHz	0.13

up to 1.35%, because the sound card of the PC does not have an ideal reconstruction process. Keeping all these analysis and results in mind, in order to fulfill the requirements of THD, the design of the PHAA should have a reliable reconstruction process and this is achived by using AIC.

5.2 Block Diagram of the System

As seen in Figure 5.3, the system consists of two important parts, digital signal processor (DSP) and the analog interface circuit (AIC). AIC is vital for PHAA because of its ability to reconstruct the signals with very low THD. A detailed description of how AIC utilizes this process is widely explained in Section 5.3.2. AIC is interfaced with a low-power C55x fixed-point DSP for signal acquisition. To achieve noise immunity, two microphones, located in the ear before and after the hearing-aid device, are differentially connected to the AIC. There is a highly flexible programmable microphone gain amplifier inside the AIC integrated circuit that allows the gain to vary from 0 dB to +59.5 dB. The A/D converter in the AIC has a 16-bit resolution and it supports the sampling rate of 96 kHz with a very low total harmonic distortion (-93 dB at 1 kHz) [38]. The sinusoidal signal generation and insertion gain computation take place in the DSP. The DSP is a 200 MHz fixed point DSP commonly used in low power communication and audio signal processing applications. It can be interfaced with the PC using the XDS510 DSP emulator, which allows real-time signal monitoring both in time and frequency. At the signal playback end of the system, there is a D/A converter with similar specifications (16-bit resolution, differential outputs) of the A/D converter. It can be programmed independently up to 96 kHz. The AIC front end has an internal programmable gain amplifier that can be programmed from 0 dB to 9 dB. There is a programmable attenuation stage in the AIC also that could attenuate the signal from 0 dB to -78 dB. This attenuator is used to mimic the operation of a Pure Tone Audiometer (PTA). The attenuation is programmed to increase or decrease in 5 dB steps.

The AIC is programmed its I2C digital interface and data is provided in both directions via serial port interface (SPI), whereas the multiple channel buffered serial port of DSP is used to communicate with the AIC and I2C channel to program the AIC. The aforementioned AIC is manufactured with TLV320AIC33 product number by Texas Instruments.

5.3 Hardware Diagram

In PHAA, a pre-produced board is used. However, this board was not appropriate for the purpose of PHAA. In order to achieve the desired objectives of the system the following changes are done:

- two microphone inputs are added,
- a high power output is added,
- unnecessary components are removed in order to eliminate the noise on the card,
- extra power conversion is done for AIC's high power output,
- power distribution is re-managed for efficient power consumption.

Altium Designer is used to draw the schematics and printed circuit board (PCB) of the circuit presented in Appendix A. There are four pages of the schematic which are: device power and DSP, flash, DSP power, and audio. The power conversion for the system requirements and DSP5509A connection with the AIC33 and other components are drown in the first page. Flash memory connection with the DSP5509A is shown in the second page. In the third page, DSP5509A power connection is drown. In audio, the last page, the AIC33 codec is drown and the connection of microphones and speaker are shown. DSP5509A is the main processor of the system with a 179-pin ball grid array (BGA). The main advantage of using BGAs is producing a miniature package for an integrated circuit with many hundreds of pins. BGAs are being produced with more pins and with decreasing spacing between the pins. But this was causing difficulties for the soldering process. As package pins got closer together, the danger of accidentally bridging adjacent pins with solder grew. In order to check the soldering faults we used the x-ray machine which is a special microscope. AIC33 is also a BGA typed with 80 pins. It is connected to the DSP5509A by McBSP0 channel for data transfer and I2C for comment transfer.

In PHAA, a 4.2V battery is used to supply the system voltage. DSP5509A needs 1.6V core voltage at 200Mhz and 1.8V for I/O pins. AIC33 needs 3.3V analog voltage, 1.8V digital core voltage, and 1.8V for I/O pins. All these voltages are distributed by using voltage translators. TPS7301, which is shown in Figure 5.8, is used for converting the battery voltage (4.2V) to 3.3V. TPS62000 and TPS76318 are used for converting 3.3V to 1.6V and 1.8V respectively.

The system clock is generated by a 12 MHz oscillator and delivered to both DSP5509A and AIC33. DSP's and AIC's master clocks (MCLK) are sourced by this oscillator. DSP is configured to work at 48 kHz for the serial data communication by dividing the MCLK. AIC is at slave mode for serial data communication and it is clocked by DSP. The I2C standard clock (SCL) is 100 kHz and generated by dividing the system clock again. DSP is the master and AIC is clocked by DSP in I2C communication.

The microphone connection to the AIC is provided by using BGF100 shown in Figure 5.9 It is a microphone filter with low pass characteristics. It also enables an

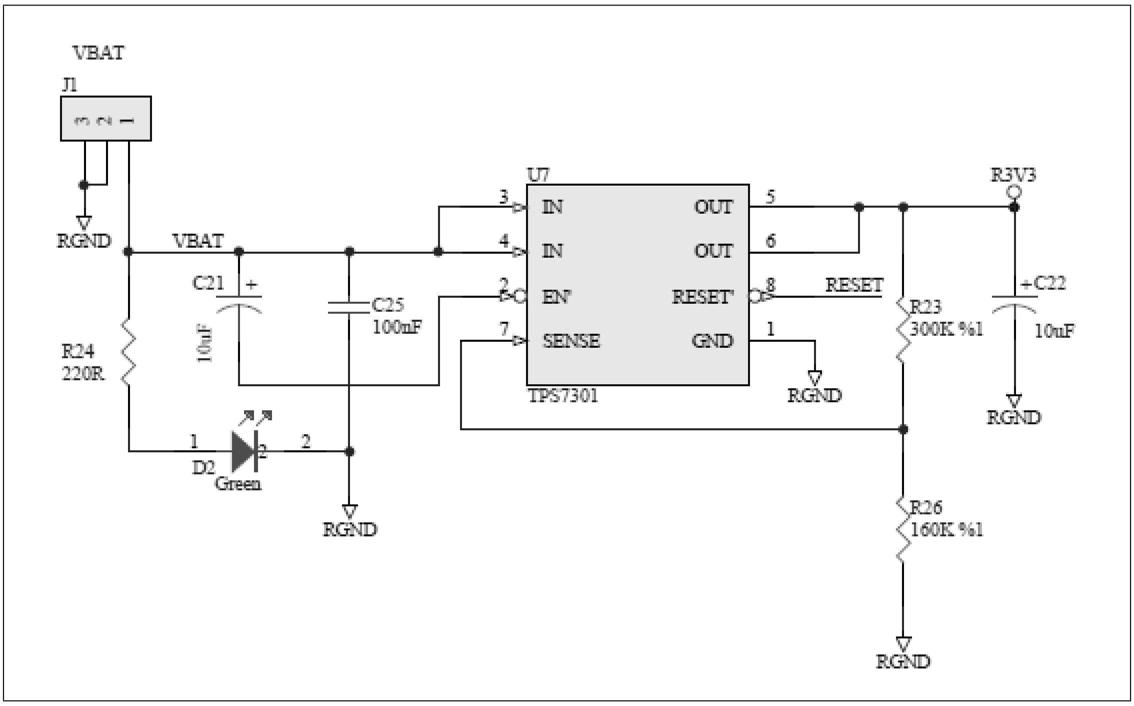

Figure 5.8 TPS7301.

electrostatic discharge (ESD) protection at the input pins up to 15 kV discharge [39].

5.3.1 DSP

The C55x^{TM}DSP architecture is very successful in terms of high performance and low power. The TMS320C55x^{TM} family is the industry's best combination of power saving and performance. This leads us to choose one of the member of this family in PHAA. The TMS320VC5509A fixed-point digital signal processor (DSP), shown in Figure 5.10, is a member of this family which has a CPU supporting an internal bus structure that is consist of one program bus, three data read buses, two data write buses, and additional buses dedicated to peripheral and DMA activity. In a single cycle three data reads and two data writes can be achieved by these buses. Besides the transfers in CPU, DMA (direct memory access) controller can perform two data transfers independently in a cycle [40].

The DSP architecture enables to make complex mathematical operations that

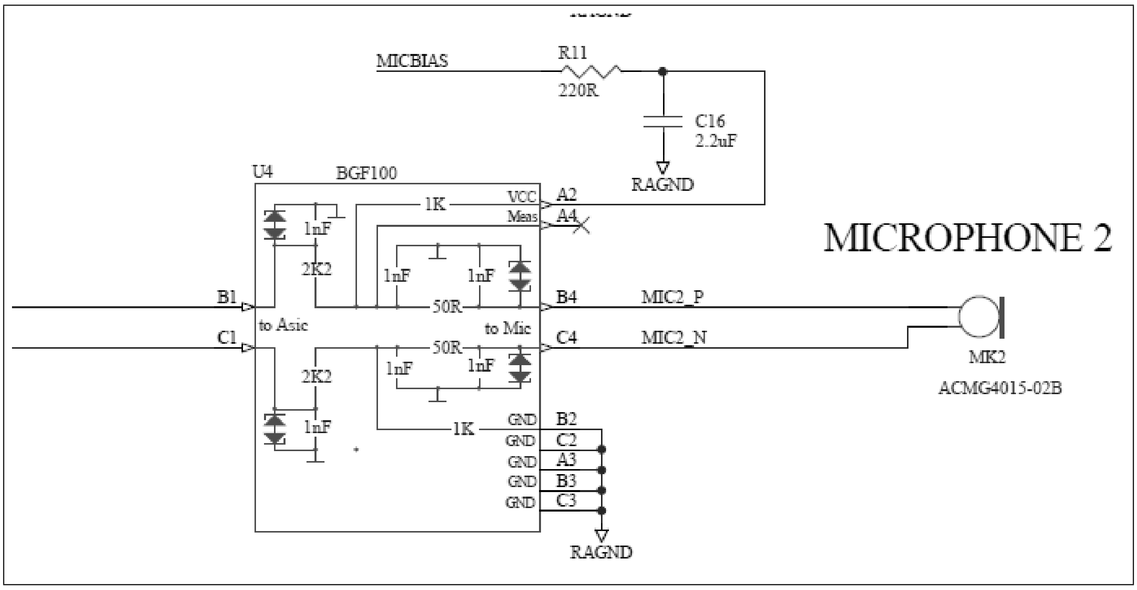

Figure 5.9 BGF100.

involve a significant amount of multiplication and addition. In DSP, these calculations are done in a single cycle (compared to multiple cycles for RISC processors) with the help of the multiply/accumulate (MAC) hardware inside the arithmetic logic unit (ALU). In TMS320VC5509A there are two MACs each capable of 17-bit x 17-bit multiplication in a single cycle. In addition, the Harvard architecture of the DSP (multiple busses) allows instruction and operand fetches in the same cycle for increased speed of operation. This high speed calculating ability makes finding the transfer function easier in PHAA.

The core supports 200 MHz clock rate at 1.6 V, 144 MHz at 1.35 V, and 108 MHz at 1.2 V. In PHAA the DSP is powered at 1.6 V by a battery, therefore the clock rate can go up to 200 MHz. A crystal oscillator is used for generation of the system master clock which is 192MHz.

There is a 256 KB On-Chip memory composed of 64 KB Dual Access Ram (DARAM) and 192 KB Single Access RAM (SARAM). The 5509A supports 8 blocks of 8K bytes of DARAM and 24 blocks of 8K byte of SARAM. DSP5509A also has 64K Bytes On-Chip ROM that provides non-volatile storage for program or data. In PHAA, a 512K Bytes external flash memory is interfaced to DSP. The code size in PHAA is

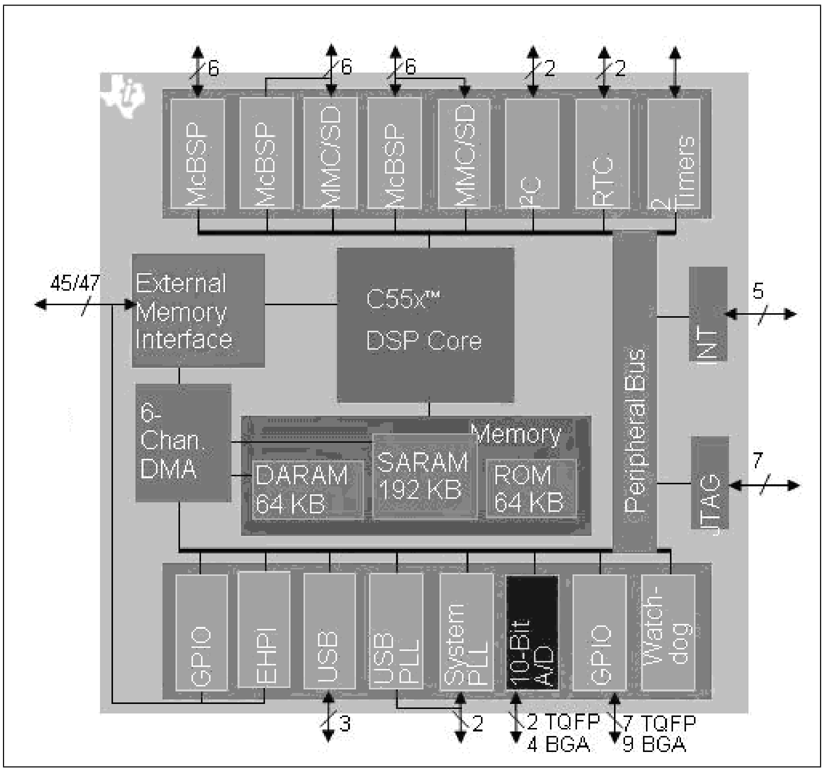

Figure 5.10 TMS320VC5509A DSP Block Diagram.

nearly 51 kilobytes which is stored at this external flash memory. The storage area can be changed by modifying the memory map file of the program.

DSP5509A supports peripherals such as external memory interface (EMIF), two 20-bit timers, watchdog timers, USB interface, or three serial ports which are a combination of up to three multichannel buffered serial ports (McBSPs) and up to two MultiMedia/Secure Digital Card interfaces. DSP5509A has three full-duplex McBSPs that allow direct interface to other C55x DSPs, codec, and other devices in a system. In PHAA McBSP_0 channel is used to interface to the AIC processor(See in Figure 5.11). Full-duplex properties are used in this connection which means the data transfer occurs in both directions.

I2C multi-master and slave interface in standard and fast mode, is an additional peripherals supported by DSP5509A. The comment communication between the DSP and AIC is established via I2C channel as seen in Figure 5.11. I2C communication between AIC and DSP is standard mode in which the clock rate is 100 kHz. DSP is

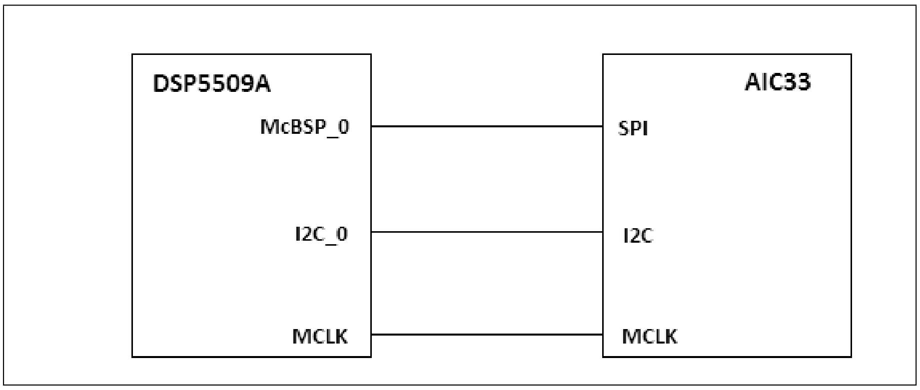

Figure 5.11 DSP to AIC Connection.

the master and AIC is the slave with address of 0x1B. DSP sends the configuration comments to AIC by the *i2c_write*() function.

The 5509A is supported by the TI' s Code Composer Studio Integrated Development Environment (IDE), DSP/BIOS, Texas Instrument's algorithm standard, and the industry's largest third-party network. PHAA work with TI's Code Composer Studio (CCS) development environment. Code Composer communicates with the design through an external emulator (XDS510). It provides real-time analysis tools for monitoring without halting the processor. The 5509A is also supported by the C55x DSP Library which features more than 50 foundational software kernels (FIR filters, IIR filters, FFTs, and various math functions) as well as chip and board support libraries [40].

5.3.2 AIC33

TLV320AIC33 is a low power stereo audio codec with variable sampling rates from 8 kHz to 96 kHz for A/D and D/A conversion with a very low, 100 dB, signal to noise ratio (SNR). There are multiple inputs and outputs programmable in single-ended or fully differential configuration. Register based power control leads DAC playback at 48 kHz, 14 mW from a 3.3 V analog supply, making it ideal for PHAA. Because, it can be powered up with a battery which eliminates the DC power noise effect [38].

The codec has four high-power output drivers as well as three differential output drivers. The high power outputs can drive four separate channels of single-ended 16 ohm headphones using ac-coupling capacitors, or stereo 16 ohm headphones without capacitors. Also these outputs are capable of driving 8 ohm speakers in pairs, connected in bridge-terminated load. Furthermore, it is possible to use these outputs in four different single-ended output configured mode. The two fully differential stereo outputs can drive the 10 ohm load and the fully differential mono output can drive a 10 ohm too. In PHAA, high power outputs are used to present the analog signals to the speaker in fully differential configuration as seen in Figure 5.12.

There are ten audio input pins in the codec. Inputs can be programmed in single-ended or fully differential configuration. To eliminate the noise effect, the microphones are connected to the input in fully differential mode in the PHAA. LINE1L is used for reference microphone and LINE1R is used for inserted microphone. The connection of microphones to the AIC is seen in Figure 5.12.

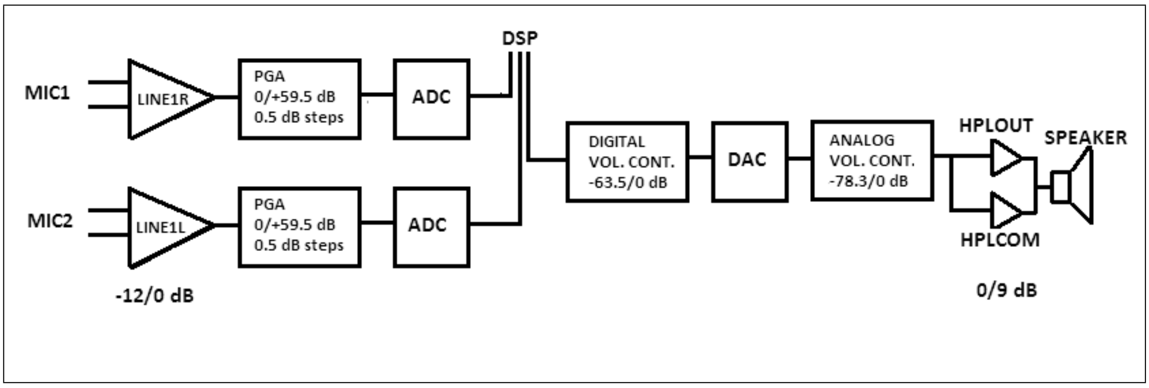

Figure 5.12 Simplified Block Diagram of AIC and Peripherals Connection.

The codec input path contains integrated microphone bias, digitally controlled stereo microphone amplifier, and automatic gain control (AGC). The input signal can be amplified up to 59,5 dB before the ADCs. In PHAA this feature makes it possible to calibrate the microphones. This path also has the ability to mix the input signal by mix/mux capability among the multiple input signals. The output path includes mix/mux capability from the stereo DAC and selected inputs, through programmable volume controls, to the various outputs. In the output path, there are three volume controls; before DACs, after DACs, and before outputs. The first one, digital volume control in the range of -63,5 to 0 dB which makes it possible to bring the signal to a desired level in PHAA. The second one is analog volume control ranging from -78,3 dB to 0 dB. It is used as attenuator in PHAA with the step size of 5 dB. Lastly, a level control with the range of 0 dB to 9 dB is used for fine tuning of the test sounds in PHAA [38].

A general digital to analog converter (DAC) needs a complex hardware design to filter high frequency components and reconstruct the signals smoothly [41]. AIC's oversampling ability and having a good reconstruction block enable us to convert digital signals to analog signals with a very low THD without any further analog filter. AIC's DAC blocks contain digital audio processing block, digital interpolation filter, multi-bit digital delta-sigma modulator, and analog reconstruction filter. The digital interpolation filter upsamples the digital signals before they are sent to the delta-sigma modulator and reconstruction filter stage. It also has enhanced image filtering and reduces signal images caused by the upsampling process which are below 20 kHz. For example, upsamling a 8 kHz signals generates signal images at multiples of 8 kHz such as 16 kHz or 24 kHz. These images are filtered by digital interpolation filter. There is also a third order multi-bit delta-sigma modulator with an analog reconstruction filter. This provides high resolution, low noise performance by using oversamling. In this block, the upsampled rate is kept constant at 128xFsref. In PHAA the sampling frequency is at 48 kHz, therefore the digital delta-sigma-modulator operates at a rate of 6,144 MHz. This oversampling ratio is high enough to ensure that the quantization noise occurred after delta-sigma modulation stays low within the frequency band below 20 kHz at all sample rates [38].

The serial control bus supports SPI (programmable for I2S, DSP, left/right justified, or TDM modes), or I2C protocols. The data connection between the DSP and AIC is done via SPI programmed to DSP mode and the comment connection is done via I2C in standard mode. AIC33 is always a slave in the I2C transmission with slave address of $0x1B$ [38].

5.4 Software Implementation

After exploring the standards (IEC 60645, ISO 12124, and IEC 61669) in detail, we first determined the design needs and the requirements for our software implementation. The software implementation is composed of two main parts. The first part includes generating and presenting the test tones by satisfying the frequency accuracy and low THD. The second part includes receiving the sounds from the test environment and calculating the transfer function of the hearing aid which is being tested.

5.4.1 Generating/Presenting the Test Tones

As mentioned before in Sections 4. and 5.1, presenting the test tones with a very low THD and frequency accuracy are the two main requirements in the Standard 60645-1. According the simulation in MATLAB environment, look-up tables are prepared for the generation of desired sinusoidal signals (at the frequencies of 125, 250, 500, 1000, 1500, 2000, 3000, 4000, 6000, 8000, and 16000 Hz) in DSP. In order to make sure that these generated signals are similar to generated signals in MATLAB simulation, the FFT of these signals are compared. $fft()$ function in DSP library is used to compute the Discrete Fourier Transform (DFT) of the generated signals. In Figure 5.4 and 5.13 the FFT of a 6000 Hz sine wave is depicted. From the figures, it is seen that the signals are completely the same.

Using an interrupt routine, these generated tables are sent to the AIC sample by sample with the 48 kHz serial bus clock rate. Theoretically, again in section 5.1 it is proved that 48 kHz sampling rate is very suitable for the generation of test tones. The D/A converters in the AIC33 are also programmed to 48 kHz clock rate. The data is transmitted by the McBSP_0 channel to the AIC's SPI. AIC33 is configured by DSP5509A via the I2C channel to support a fully differential output for speaker. After the selection of test tones in means of frequency, it is sent to the AIC33. The signal is attenuated to a desired level after conversion into the analog form in AIC33 and then sent to the high power output which is connected to a speaker. The attenuation is

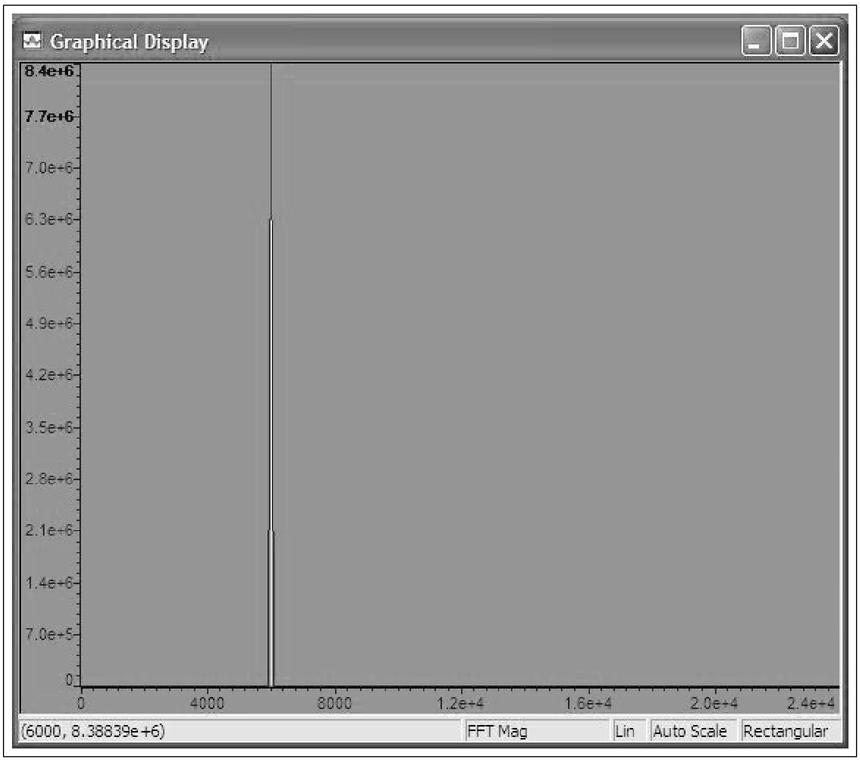

Figure 5.13 FFT of 6000 Hz sine signal in DSP.

done again by configuring the AIC33 via I2C channel.

5.4.2 Receiving the Sounds and Calculating the Transfer Function

The same with the presenting the test tones, AIC33 is programmed to 48 kHz sampling to receive signals from both reference and inserted microphones. These microphones are connected to the AIC33 fully differentially to eliminate the noise effect. From the SPI of AIC33, received data are sent to McBSP_0 channel of the DSP5509A sample by sample. In DSP, the received data are buffered in order to calculate the transfer function. An important trick in the software is addressing this buffer to a temporary buffer. While the interrupt routine is running the buffer will be written by the new data values. In order to calculate the gain at a certain time this changing buffer should not be used. The addressed buffer is used for the calculation. The frame size (buffer size) is chosen 384, because it is the minimum common number for reconstructing the test tones. The least frequency of the test signals is 125 Hz and the sampling rate is chosen 48 kHz. The minimum number of sample to produce a periodical sinusoid at these values is 48000 / 125 = 384 sample. These 384 values are used to calculate the power of each microphone channel for each test frequency ranging from 125 Hz to 16 kHz and power ratio between the two microphone is found by the equation 5.6 where K is the gain, x_i, and y_i is the sound vector received from internal microphone and external microphone respectively.

$$K = 10 \log \frac{\sum_{i=0}^{n} x_i}{\sum_{i=0}^{n} y_i} \tag{5.6}$$

6. EXPERIMENTAL RESULTS

A hearing aid analyzer should present the test tones and calculate the actual gain to evaluate the hearing aid in situ performance. We have proposed to design an analyzer that verifies the specifications in the standards of IEC 606045-1, ISO-12124, and IEC 61669. In order to take the measurements in a accurate way, the calibration of the system and test setup should be carefully done with the defined procedures.

The calibration of the microphones are done with a calibrator which produces 1 kHz audio sine wave at two different amplitudes (94 and 114 dB SPL).

The environmental noise is measured with a sound level meter which can measure the sound intensities between 60 to 130 dB. It is ensured that the environmental noise is less enough not to change the test results by more than 1 dB.

THD and frequency accuracy in pure sine waves with different intensities and insertion gain of a hearing aid are calculated in this chapter.

6.1 Verification of Test Tones

The THD, dB SPL, and frequency accuracy of the presented test tones are measured by the Minilyzer. By considering the measurement results given in Table 6.1 and 6.2, one can see that the test tones are generated successfully in the limits of frequency accuracy and THD (THD is less than 2% and frequency accuracy ± 1 in type 1 audiometer).

Table 6.1
Measured Frequency Accuracy (Measured by Miliyzer).

Frequency (Hz)	Measured Freq.(Hz)	Frequency (Hz)	Measured Freq.(Hz)
125	125	2000	2000
250	250	3000	3000
500	500	4000	4000
750	750	6000	6000
1000	1000	8000	8000
1500	1500	16000	16000

6.2 Verification of Hearing Aid

We have realized the tests with the hearing aid produced by Oticon with the GO 05-151984 model number. This hearing aid is programmed with the NOAH programmer according the settings depicted in Figure 6.1

The target gain is calculated by the prescription method in the programmer and it is seen in Figure 6.2.

The target gain versus frequency is tabulated in Table 6.3 when the input intensity is 50 dB.

Then, in order to check whether the hearing aid is programmed according to the desired values, a hearing aid analyzer from Affinity is used. Affinity is widely used by audiologists and it is clinically accepted device using for hearing aid fitting. This can be a reference test device for evaluating PHAA. We applied the Full-on Gain test which draws a graph of frequency versus gain with a 50 dB input as seen in Figure 6.3.

According to the Affinity hearing aid analyzer, the applied gains for the hearing aid is shown in the Table 6.3 in case of the input intensity is 50 dB and for a frequency range of 250 to 6000 Hz.

Table 6.2
THD (%) of all the frequencies versus given intensity (Measured by Miliyzer).

Hz/dB	40	45	50	55	60	65	70	75	80	85	90
125	0.70	0.38	0.25	0.14	0.09	0.05	0.03	0.02	0.02	0.02	0.7
250	0.70	0.38	0.24	0.14	0.09	0.05	0.03	0.02	0.02	0.02	0.7
500	0.70	0.40	0.30	0.20	0.13	0.15	0.20	0.04	0.05	0.05	0.73
750	0.80	0.40	0.25	0.20	0.13	0.15	0.07	0.05	0.08	0.08	0.72
1000	0.70	0.30	0.30	0.28	0.12	0.12	0.06	0.04	0.07	0.08	0.72
1500	0.80	0.40	0.28	0.25	0.13	0.11	0.07	0.06	0.06	0.08	0.73
2000	0.70	0.20	0.27	0.24	0.14	0.13	0.08	0.07	0.06	0.06	0.75
3000	0.70	0.25	0.25	0.22	0.10	0.11	0.06	0.04	0.06	0.08	0.74
4000	0.77	0.24	0.26	0.25	0.11	0.12	0.09	0.07	0.08	0.09	0.65
6000	0.76	0.30	0.25	0.24	0.10	0.11	0.11	0.09	0.09	0.10	0.48
8000	0.76	0.32	0.31	0.25	0.16	0.10	0.17	0.16	0.17	0.02	0.34
16000	0.90	0.30	0.30	0.28	0.26	0.26	0.23	0.20	0.20	0.20	0.55

While examining the Table 6.3 one can see that the target gains for each frequency are applied with the deviation of 2 dB except the gains at 4 kHz and 6 kHz. This might be caused by poor ability of hearing aid in amplifying the higher frequencies.

Within the following procedure, we test the program in the hearing aid with PHAA. Firstly, we simulate the test environment by using a plastic tube and eliminating the background noise. Next, we present the test tones at the frequencies of 250, 375, 500, 750, 1000, 1500, 2000, 3000, 4000, and 6000 Hz and calculate the respective gains. The gain versus frequency graphic is shown in Figure 6.4. Notice that the graph in the figure seems very similar to the graph obtained by using Affinity, but the gain values are have a difference around 2 dB.

The obtained results with PHAA are very close to the results obtained with the Affinity hearing aid analyzer. The deviations at each frequencies are around 2 dB as we see in Figure 6.4. One possible reason of such an unexpected difference might be environmental noise or measurement errors, because it is hard to simulate the same

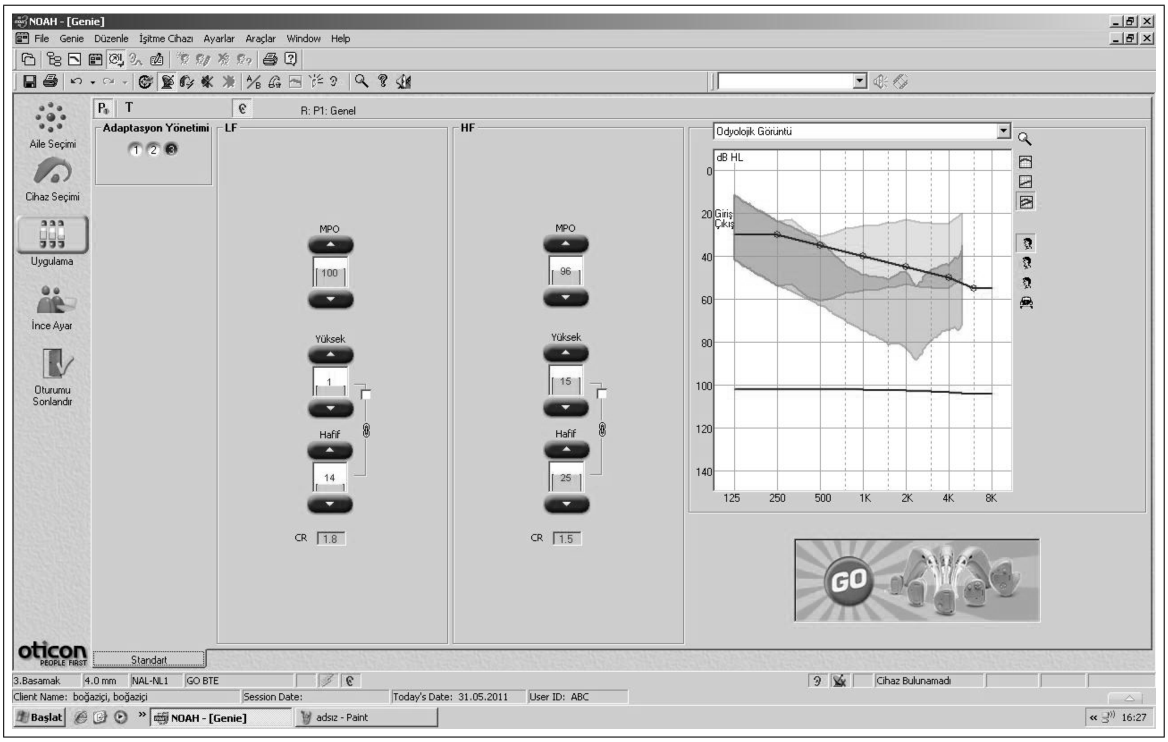

Figure 6.1 Initial Settings of Hearing Aid by NOAH.

test environment with the Affinity hearing aid analyzer. We can say that PHAA is quite successful in measuring the applied gain to the target hearing aid.

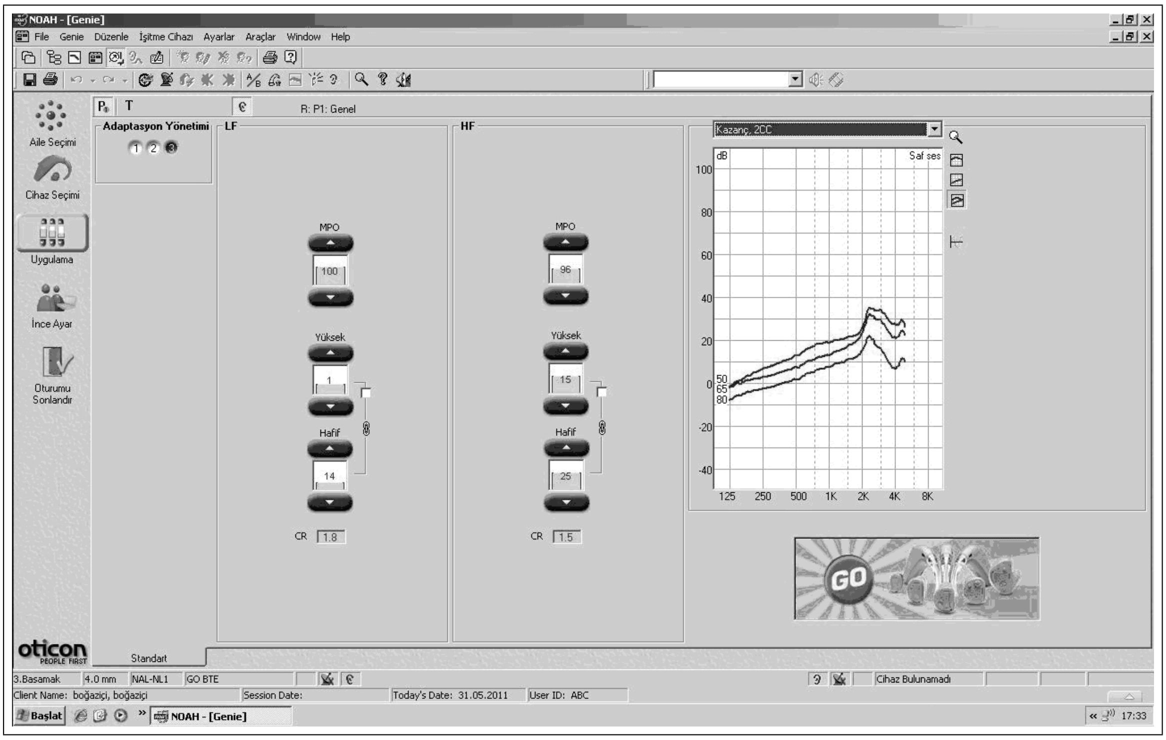

Figure 6.2 Programmed Target Gain with NOAH.

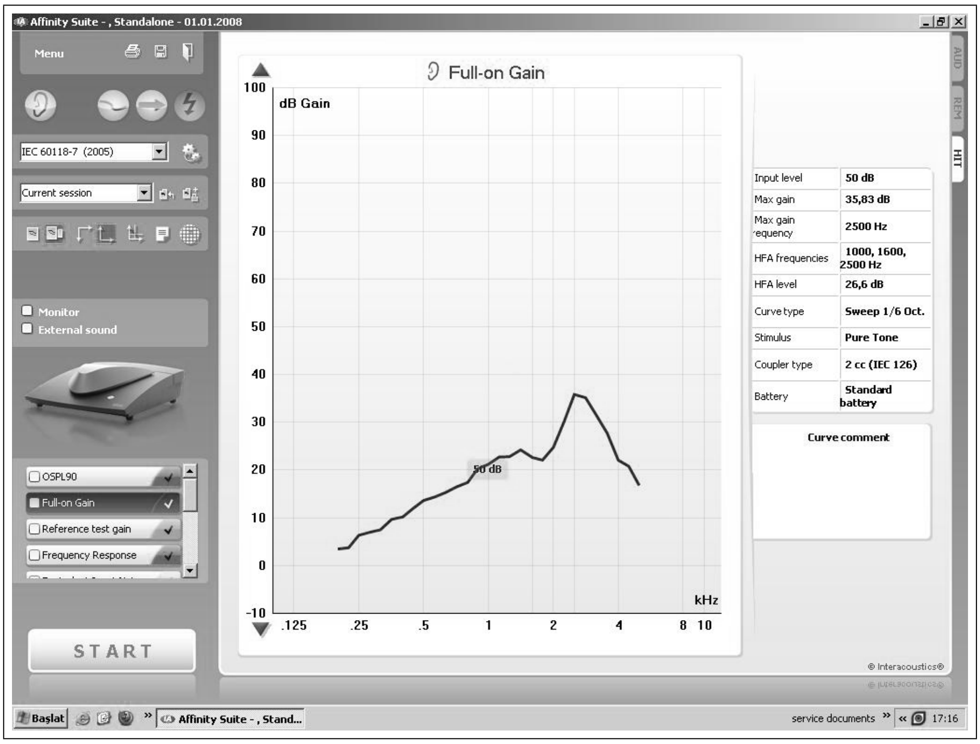

Figure 6.3 Measured Target Gain with Affinity Hearing Aid Analyzer.

Table 6.3
Gains by NOAH, Affinity, and PHAA.

Frequency (Hz)	NOAH	Affinity	PHAA
250	8	6	8
375	10	8	10
500	13	14	16
750	18	16	17
1000	19	21	19
1500	22	23	23
2000	26	24	23
3000	32	32	30
4000	28	22	21
6000	26	15	16

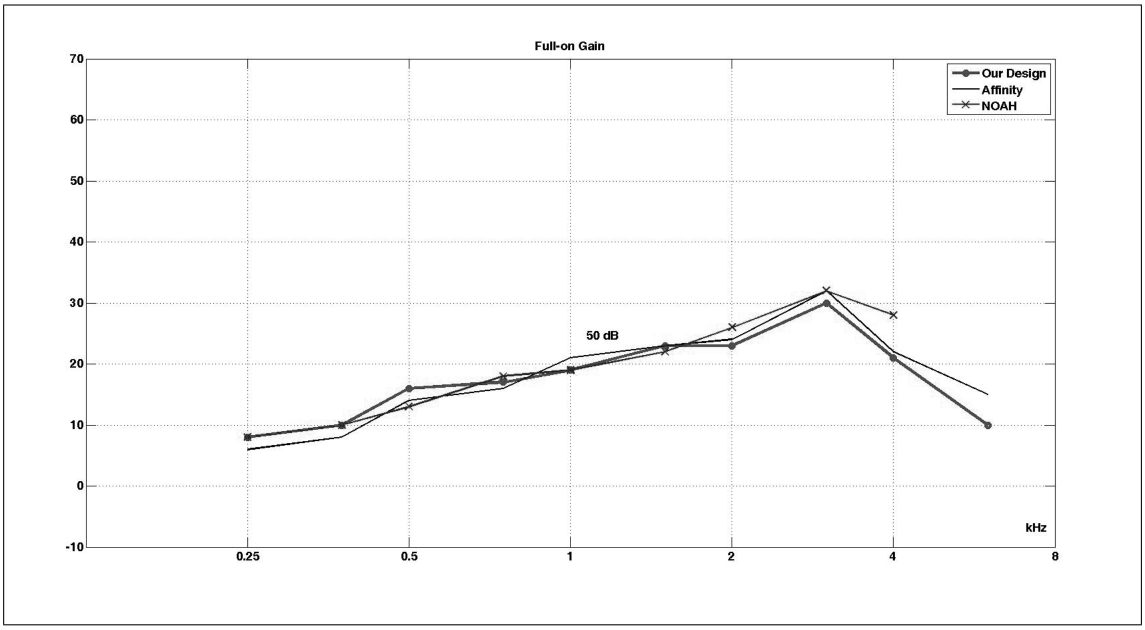

Figure 6.4 Results of Derived by NOAH, Affinity, and PHAA.

7. CONCLUSION AND FUTURE WORK

Hearing loss is a common problem in our lives and it can be compensated by using a hearing aid. Unfortunately, it is well known that only about 20% of people who face hearing loss use a hearing aid. The major reason of such a situation is hearing aid fitting problem. This thesis have mainly focused on designing a portable hearing aid analyzer to determine hearing aids performance while it is in use. To fulfill this purpose, we have designed a hearing aid analyzer and calculated the transfer function of hearing aids by insertion gain measurements. Thanks to the results obtained from the measurements, some useful feedback can be provided to the audiologists for a fine tuning.

In order to provide a better understanding of hearing, Chapter 2 has reviewed the topics of sound nature, hearing anatomy and physiology, hearing loss, and hearing aids and also has given a brief information about some definitions of hearing.

In Chapter 3, we have introduced the hearing aid fitting problem and explained the procedures for hearing aid fitting. Furthermore, types of audiometers have been given. In this chapter, it has been asserted that the prescription method, generally the only applied method for fitting procedure is not enough for an ideal fitting, and also claimed that REM as an individual based test should be done for a better patient satisfaction. Additionally, the setup for a REM has been defined in this Chapter.

The requirements in the international standards for test tones and REM have been investigated in Chapter 4. IEC 60645-1 Standard (for test tones) and IEC 61669 and ISO 12124 Standards (for REM) have been examined. THD, frequency accuracy, and REM setup requirements are the main subjects in these standards that make the design challenging.

PHAA can be divided into two main parts. The first part includes presenting the test tones in a accurate way to the patient's ear. The second part includes receiving back the tones and calculating the transfer function of the hearing aid while it is inserted to the ear. In Chapter 5, a simulation in MATLAB environment has been explained and a method has been improved for generating the test tones. This work constitutes the basis of the first part of PHAA. Then, the block diagram of the PHAA has been depicted and the flowchart has been explained in detail. Also, we have described the main processors and detailed the software implementation.

In Chapter 6, we have applied some tests to evaluate conformance of PHAA with the relevant international standards. Therefore we have made some measurements to check whether PHAA:

- meets the THD requirements for pure tones,
- can present the test tones at the desired frequencies,
- can calculate the transfer function of a hearing aid with insertion gain measurement.

In the light of these measurements, we have concluded that:

- the pure tones are produced within the limits of THD determined in the standards,
- the test tones are at the desired frequencies without any deviation,
- the proposed PHAA can calculate the transfer function of a hearing aid in a similar manner of a commercial hearing aid analyzer.

In addition to these features, PHAA has some advantages such as it is battery powered and portable due to its small size. Indeed, these properties make our design superior to its relevant predecessors.

This thesis can be considered as a basis for future studies in this field. Future works are planned as

- manufacturing PHAA and make it commercial,
- making some innovations in REM such as placing the probe microphone into the earmold,
- designing a compact device including audiometer and analyzer, so that the hearing tests can be done with only one device.

APPENDIX A. SCHEMATICS

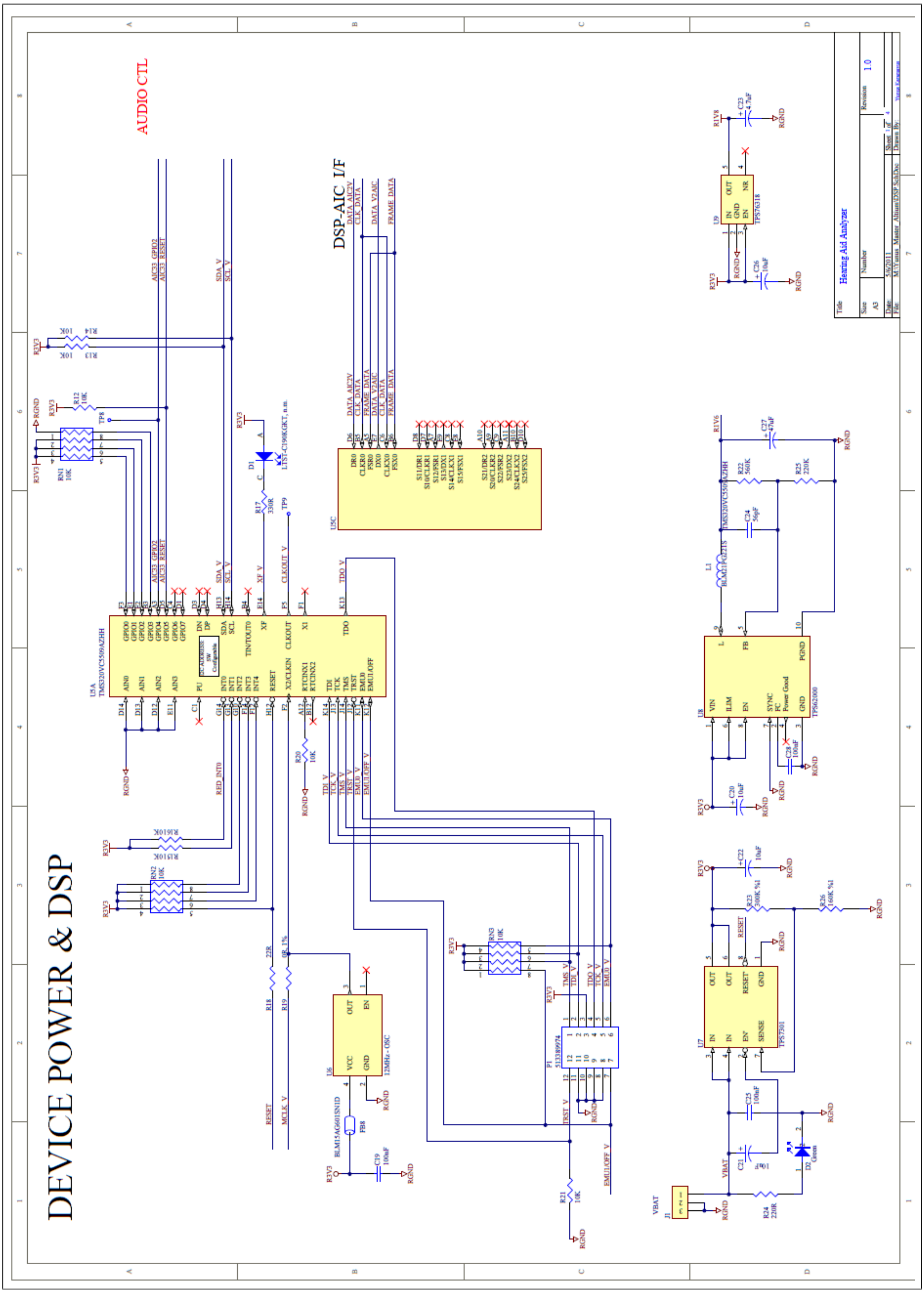

Figure A.1 Schematic of Device Power and DSP.

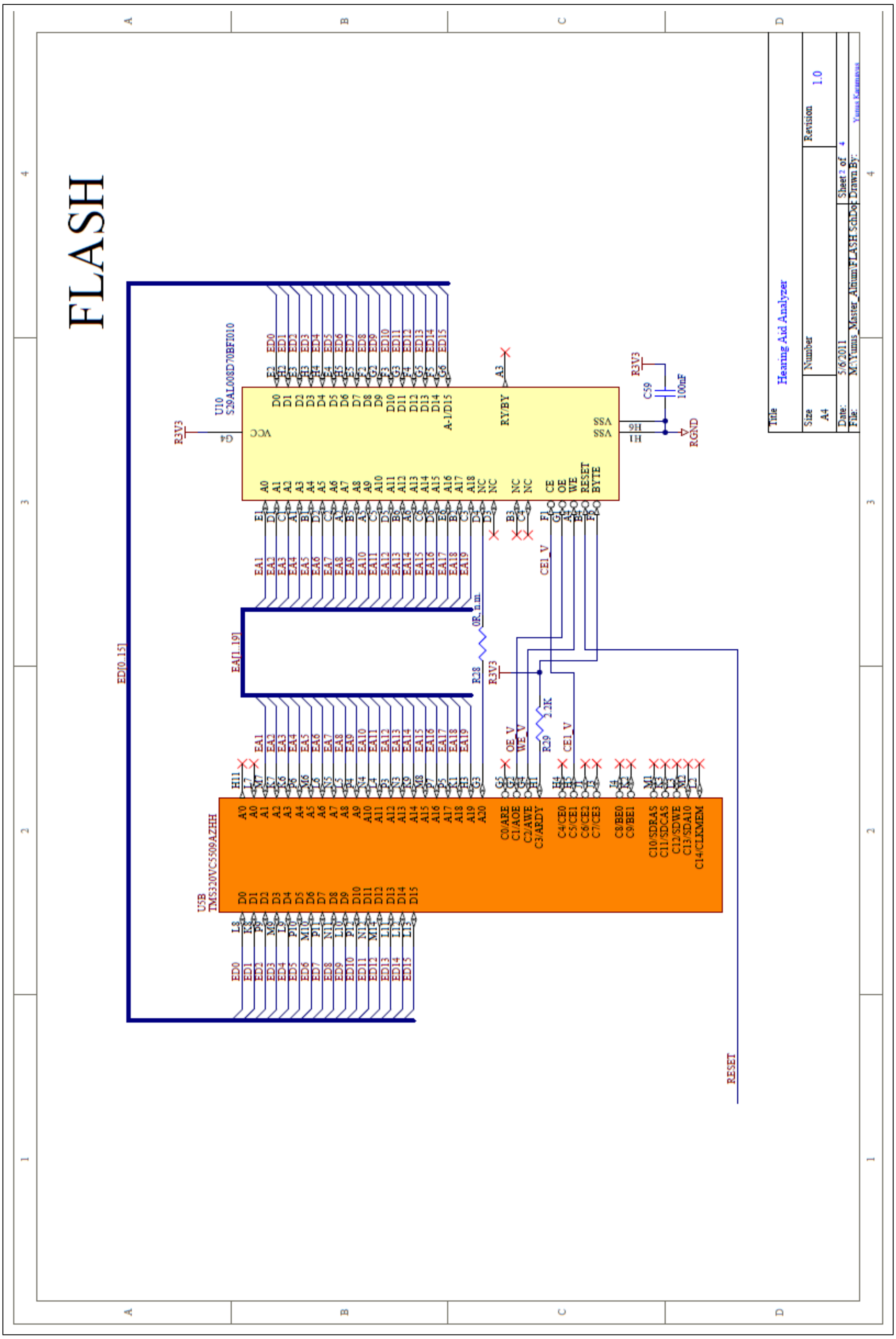

Figure A.2 Schematic of Flash.

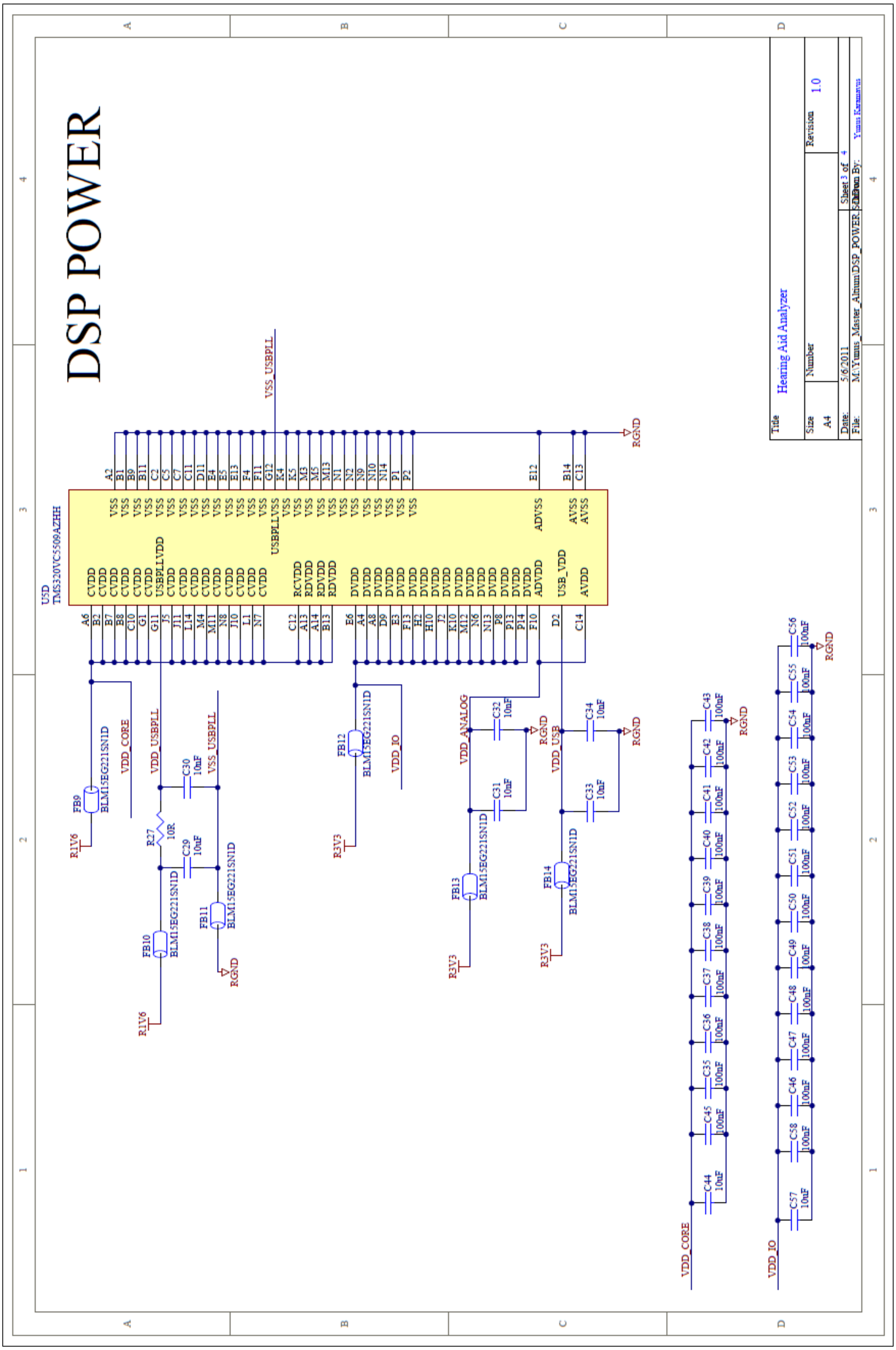

Figure A.3 Schematic of DSP Power.

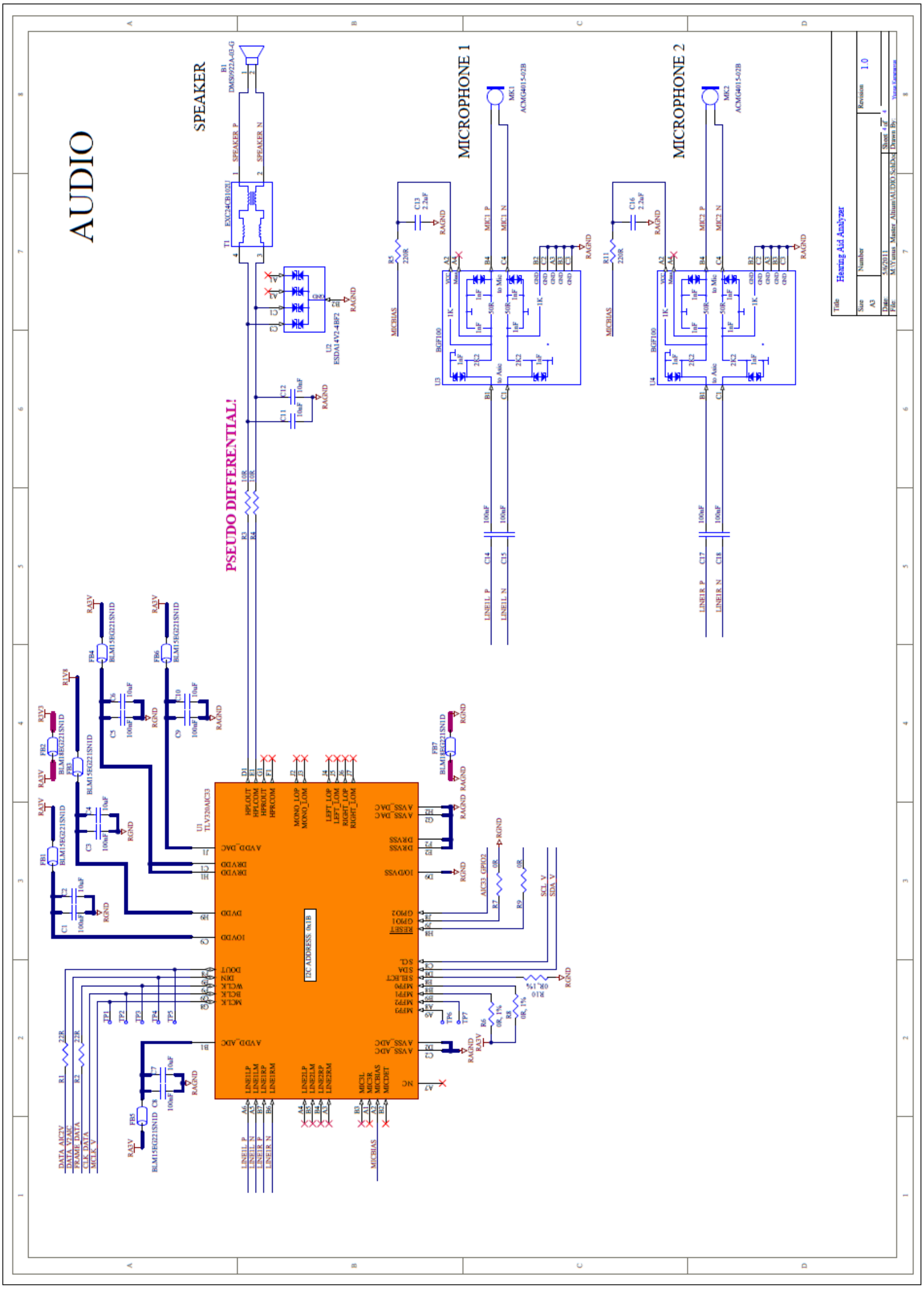

Figure A.4 Schematic of Audio.

APPENDIX B. BILL OF MATERIALS

Hearing Aid Analyzer BOM

Comment	Description	Designator	Footprint	LibRef	Quantity
DMS0922A-03-G	Dynamic Miniature Speaker (Buzzer) Pads	B1	DMS0922A-03-G	PAD_SPEAKER	1
100nF	Capacitor, Ceramic MLCC Capacitor	C1, C3, C5, C8, C9, C14, C15, C17, C18, C19, C35, C36, C37, C38	0402	CAP, CAP CER MLCC	33
10uF	Capacitor, Ceramic MLCC Capacitor	C2, C4, C6, C7, C10, C26, C29, C44, C57	0603	CAPACITOR POL, CAP CER MLCC	9
10nF	Ceramic MLCC Capacitor	C11, C12, C30, C31, C32, C33, C34	0402	CAP CER MLCC	7
2.2uF	Ceramic MLCC Capacitor	C13, C16	0402	CAP CER MLCC	2
10uF	Capacitor	C20, C21, C22	TANTAL-B	CAPACITOR POL	3
4.7uF	Capacitor	C23	0603	CAPACITOR POL	1
56pF	Capacitor	C24	0603	CAP	1
100nF	Capacitor	C25, C28	0603	CAP	2
47uF	Capacitor	C27	TANTAL-B	CAPACITOR POL	1
LTST-C190KGKT, n.m.	Led Green Clear 0603	D1	LED 0603	LED_GREEN	1
Green		D2	SMD_LED	LED	1
BLM15EG221SN1D	Ferrite Chip Bead 220R 700mA 0R28 0402	FB1, FB3, FB4, FB5, FB6, FB9, FB10, FB11, FB12, FB13, FB14	0402	BLM15EG221SN1D	11
BLM18EG221SN1D	Ferrite Chip Bead 220R 2A 0R05 0603	FB2, FB7	0603	BLM18EG221SN1D	2
BLM15AG601SN1D	Ferrite Chip Bead 220R 700mA 0R28 0402	FB8	0402	BLM15EG221SN1D	1
	Connector	J1	PWR2.5	CON3	1
BLM21PG221S		L1	1206	INDUCTOR	1
ACMG4015-02B	Microphone Pads	MK1, MK2	ACMG4015-02B	PAD_MICROPHONE	2
513389974	Board to Board Connector Receptacle 12 pin 0.4mm	P1	MOLEX-51338-9974	CONN_MOLEX_REC_12PIN	1
22R	Fixed Thick Film Chip Resistor	R1, R2, R18	0402	RES	3
10R	Fixed Thick Film Chip Resistor	R3, R4, R27	0402	RES	3
220R	Fixed Thick Film Chip Resistor	R5, R11	0402	RES	2
0R, 1%	Fixed Thick Film Chip Resistor	R6, R8, R10, R19	0402	RES	4
0R	Fixed Thick Film Chip Resistor	R7, R9	0402	RES	2
10K	Fixed Thick Film Chip Resistor	R12, R13, R14, R15, R16, R20, R21	0402	RES	7
330R	Fixed Thick Film Chip Resistor	R17	0603	RES	1
560K		R22	0603	RES1	1
300K %1		R23	0603	RES1	1
220R		R24	0603	RES1	1
220K		R25	0603	RES1	1
160K %1		R26	0603	RES1	1
0R, n.m.	Fixed Thick Film Chip Resistor	R28	0402	RES	1
2.2K	Fixed Thick Film Chip Resistor	R29	0402	RES	1
10K	Chip Resistor Array	RN1, RN2, RN3	EXB28V	RESISTOR_PACK4	3
EXC24CB102U	Ferrite Noise Filter 2 Mode	T1	EXC24CB/CP	EXC24CB102U	1
TEST POINT	Test Point	TP1, TP2, TP3, TP4, TP5, TP6, TP7, TP8, TP9	PIN1	TEST POINT	9
TLV320AIC33	Stereo Audio Codec for Portable Audio/Telephony (80-Ball)	U1	S-PBGA-N80-0.5MM-AN	TLV320AIC33	1
ESDA14V2-4BF2	Quad Bidirectional Transil Array for ESD Protection	U2	ESDA14V2-4BF2	ESDA14V2-4BF2	1
BGF100	Microphone Filter and ESD Protection	U3, U4	BGF100	BGF100	2
TMS320VC5509AZHH	TMS320VC5509A Fixed-Point Digital Signal Processor	U5	S-PBGA-N179	TMS320VC5509A	1
12MHz - OSC		U6	JO22	CRYSTAL12.288M	1
TPS7301		U7	R-PDSO-G-8	TPS7301	1
TPS62000		U8	S-PDSO-G10	TPS62000	1
TPS76318	Low Vin LDO with Reverse Current Protection	U9	IC_TPS73216_SOT23_5	IC_TPS73216_SOT23_5	1
S29AL008D70BFI010	Spansion 8 Megabit Boot Sector Flash Memory	U10	FBGA-48	S29AL008D	1

Figure B.1 List of Materials Used in System.

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