Estimation of Stapedius-Muscle Activation using Ear Canal Absorbance Measurements
(An Application of Signal Processing in Physiological Acoustics)

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Abstract

The stapedius muscle, which is located in the middle ear, goes into contraction when the ear is exposed to high sound intensities. This muscle activation is called ‘the acoustic reflex’. Measurement of the acoustic reflex is clinically of importance since it can reveal diagnostic information about the middle ear’s pathologies. Moreover, this muscle-activation alters the acoustic characteristics of the middle ear (i.e. the acoustic impedance and the power reflectance), which in turn, can significantly manipulate one’s perception of sounds. In the present study, these acoustic characteristics are measured in the ear canal by means of absorbance measures using equivalent Thevenin circuit theory. The quantities are then compared to form the shift responses between the baseline (before the activation) and the post-activator effect. This project investigates the shifts in power reflectance and admittance of the middle ear caused by the stapedius-muscle contraction. The wideband characterization (0.1- 8 kHz) of these acoustic reflex-induced shifts is achieved using chirp signals as a probe and through ipsilateral broadband noise activator. The data acquisition and signal processing of the project are carried out using MATLAB software. The hardware consists of National Instruments USB-6212 data acquisition interface and low noise microphone system Etymotic Research ER-10B+. A group of 10 adults including 5 males and 5 females are recruited as the participants for the project. The measurements of the reflectance shifts indicate that the most robust frequency region affected by the acoustic reflex is up to 4 kHz whereas for the admittance shifts, this region is up to 2 kHz. In addition, it is shown that the stapedius-muscle contraction leads to the attenuation of the low-frequency transmission into the middle ear (less than 1 kHz) consistent with a stiffness-controlled system in this range of frequencies. In contrast, the results imply that the activation of the stapedius muscle leads to a slight enhancement of the frequency transmission in the range of 1-4 kHz (corresponding to the speech frequency band). These findings suggest a beneficial role for the stapedius-muscle contraction in the perception of speech during vocalization. Furthermore, the implemented methods in this project can be useful in better understanding the effect of the stapedius-muscle contraction on the speech perception both in normal hearing and hearing impaired persons.

KEY WORDS: Signal Processing, Physiological Acoustics, Middle Ear, Stapedius Muscle, Acoustic Reflex, Thevenin Circuit, Power Reflectance, Absorbance Measurements.
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Chapter 1

Introduction

In this chapter first, a brief overview of the anatomy and physiology of the middle ear as well as the acoustic reflex is described. Then, acoustic characteristics of the middle ear, which the purpose of the current project is to investigate their changes due to the acoustic reflex, are introduced.

1.1. Middle Ear

The middle ear is a part of the ear starts from eardrum (tympanic membrane) and ends at the oval window of the cochlea. This portion includes ossicular chain that is malleus, incus and stapes, suspended by supporting ligaments which transfers the vibration of eardrum to the inner ear’s membrane and as a result to waves in fluid. The malleus is composed of a manubrium which is attached to the tympanic membrane and a head which is articulated with the incus in its posterior part and following that the incus has a joint with the head of the stapes. Also, there is the Eustachian tube in the tympanic cavity which joins there to the nasal cavity in order to make the pressure equalized between these areas [1, 2, 3]. Different parts of the ear are illustrated in figure 1 [4].

![Fig. 1. The human ear consists of outer, middle and inner part [4].](image)

There are two tiny muscles namely tensor tympani muscle and stapedius muscle located in the middle ear which are the smallest muscles in human body. The tensor tympani muscle
attaches the manubrium of the malleus to the wall of the middle ear cavity. A branch of the trigeminal nerve innervates the tensor tympani. The activation of this muscle which is due to stimulation of the skin of the human ear and face area pulls the manubrium inwards resulting in a greater stiffness of the eardrum. In other words, the mechanics of the middle ear are altered that leads to make less low frequency transmission through the middle ear [1, 2, 3].

The stapedius muscle attaches the head of the stapes to the area between the oval and round windows. In spite of its very small size, a branch of the facial nerve richly innervates the stapedius muscle. The activation of this muscle can be due to either excessive sound stimulation or vocalization. Its contraction pulls the stapes into the direction perpendicular to its piston-like motion that makes the stapes being tilted. As a consequence, the annular ligament attaching the stapes footplate to the oval window is being stretched and become stiffer. Again, this alterations decrease the transmission of low frequencies through the middle ear [1, 2, 3]. This attenuation in transmission of low frequencies will be investigated through the current project.

The middle ear muscles are more specifically shown in figure2 [5].

![Fig. 2. Middle ear muscles including Tensor tympani muscle and Stapedius muscle [5].](image)

The physiological role of the middle ear muscles is not completely distinguished. It has been confirmed that their contractions increases the stiffness of the ossicular chain which contributes to decrease the low-frequency sound energy transmitted to the cochlea. It has also been reported that vocalization seems to stimulate both muscles. In general, their activity might contribute to protect the inner ear against extreme stimulation, a decrease of low-frequency masking as well as to improve the signal to noise ratio for hearing the speech-
like sounds. On the other hand, it has been recently declared that there is a hypothesis which says the middle ear muscle-contraction might lead to the long-term regulation of inner ear fluid pressure by forcing endolymph to flow to or from the endolymphatic sac [1].

1.2. Acoustic Reflex

Since the activation of the stapedius muscle can be of the acoustic origin, its contraction is termed as the acoustic reflex which is three- or four-neuron arc in the low brainstem. Different information can be interpreted from the occurrence of the acoustic reflex. For instance, information about the afferent auditory system, the function of auditory brainstem, the integrity of the facial nerve which innervates the stapedius muscle, and the functional status of the middle ear can be explored. Also, assessing the acoustic reflex has been successfully applied for infants such as neonates with auditory dysynchrony. It is reportedly useful for some of them who suffer from poor functional hearing while their screening of cochlear function using evoked otoacoustic emissions does show no sign [6]. Furthermore, acoustic reflex reveals information about the function of auditory system during vocalization. Hence, the acoustic reflex which is going to be investigated in our project is a significant physiological test of auditory function used in the diagnosis of conductive, cochlear and retrocochlear disorders.

1.3. Acoustic Characteristics of the Middle Ear

When the acoustic reflex occurs, contraction of the stapedius muscle makes the ossicular chain be stiffened and as a consequence the acoustic impedance measured in the ear canal at the tympanic membrane is increased. Therefore, an indirect, noninvasive way to assess the acoustic reflex is to study the related change in acoustic impedance or admittance of the middle ear which is the method this paper is based on.

1.4. Definition of Acoustic Impedance and Admittance [7]

To figure out the average motion of the tympanic membrane in response to the external sound needs to define acoustic impedance or in other words acoustic admittance. Considering sound pressure stimuli as a function of time \( (t) \) and sinusoids of frequency \( (f) \), \( p(t) = P \cos(2\pi ft + \theta_p) \), then the rate of alternating volume displacement (volume velocity) at the tympanic membrane would be also a sinusoid, \( u(t) = U(f) \cos(2\pi ft + \theta_u(f)) \), with a frequency-dependent magnitude and phase. It should be taken into account that in order to have this relationship, the sound intensity must be within the range where the middle ear is linear; particularly, less than 120 dB SPL. Hence, the complex quantity of the frequency-dependent acoustic impedance is defined as the ratio of the sound pressure
to the volume velocity that gives its magnitude equal to \(|Z(f)| = P/U(f)| \) and its phase angle equal to \(\angle Z(f) = \theta_p - \theta_U(f)\). In this example of stimuli in contrast to sound pressure and volume velocity which vary with time, the magnitude and the phase angle of acoustic impedance are time-independent.

In other words, the Cartesian complex form of the impedance \(Z = R + j(\omega L - \frac{1}{\omega C})\) explicitly represents the resistance \(R\) and the reactance (including inductance \(L\) and capacitance \(C\)) of its electrical equivalent in the real and imaginary parts respectively. In fact, the electrical analogy for sound pressure and volume velocity are voltage and current in that order. Figure 3 shows the electrical equivalent of the ear canal that is a series combination of resistance, capacitance and inductance which imply damper, spring (stiffness) and mass function of its mechanical equivalent respectively.

![Ear Canal](image)

**Fig. 3.** Electrical equivalent of the ear canal impedance: \(Z = R + j\left(\omega L - \frac{1}{\omega C}\right)\).

Alternatively, the admittance can be used to describe acoustic characteristic of the middle ear which is defined as the reverse of the acoustic impedance. Considering complex numbers, the simple relationship between these two quantities is \(Y(f) = 1/Z(f)\). This is the relation between their units as well. According to international system of units (SI), acoustic siemens and acoustic ohm are the units used for acoustic admittance and impedance respectively. 1 acoustic ohm = 1 Pa·s·m⁻³. Also, the term immittance is used to describe the combination of these two quantities.

### 1.5. Definition of Acoustic Reflectance and Absorbance [7]

Studying acoustic waves in the tubes makes it easier to explain the definition of reflectance and absorbance. Therefore, assuming a long narrow rigid-walled tube of constant cross-section as a one-dimensional acoustic device there is a sound source on one side of the tube's long axis while the other side is terminated by an object with unknown immittance.
The forward (incident) wave is generated by the sound source and the backward wave is generated by the reflection of forward wave at the tube’s terminal. The assumption of rigid-walled tube with uniform cross-section which is reasonable to take into account in many circumstances makes the effects of air viscosity small enough to be negligible. Pressure reflectance is defined as the ratio of the complex sound pressure (magnitude and phase) of the incident wave to the reflected wave. Equation 1a and 1b represent the mathematics of the sinusoidal incident and reflected waves as \( p^+ \) and \( p^- \) respectively, which are the functions of time \( t \) and distance \( x \) from the source of the reflection, where the radian frequency \( \omega = 2\pi f \) and considering the speed of sound \( c \), \( k = \omega / c \).

\[
p^+(t,x) = P^+ \cos(\omega t - kx + \theta_p^+), \quad \text{(Eqn.1a)}
\]
\[
p^-(t,x) = P^- \cos(\omega t + kx + \theta_p^-), \quad \text{(Eqn.1b)}
\]

Therefore, according to this definition, the magnitude and the phase of the pressure reflectance is achieved by equation 2a and 2b respectively.

\[
|R(f,x)| = \frac{p^-}{p^+}, \quad \text{(Eqn.2a)}
\]
\[
\angle R(f,x) = (kx + \theta_p^-) - (-kx + \theta_p^+) = 2kx + (\theta_p^- - \theta_p^+), \quad \text{(Eqn.2b)}
\]

As the equation 2b shows, the phase of the pressure reflectance is dependent on the measurement location that is the distance from the source of the reflection.

Also, the pressure reflectance at any distance \( x \) away from the tube termination can be measured in terms of the acoustic impedance at that distance and the characteristic impedance of the air-filled tube. Equation 3a demonstrates the characteristic impedance \( Z_c \) which depends on the density of air \( \rho_0 \), the speed of sound in air \( C \) and the cross-sectional area of the tube \( S \).

\[
Z_c = \frac{\rho_0 C}{S}, \quad \text{(Eqn.3a)}
\]
\[
R(f,x) = \frac{Z(f,x)-Z_c}{Z(f,x)+Z_c}, \quad \text{(Eqn.3b)}
\]

The relationship between the acoustic pressure reflectance and the acoustic impedance is expressed by the equation 3b.

In addition to pressure reflectance, the power reflectance (sometimes called energy reflectance) is also used in the acoustic terminology. The power or energy reflectance is defined as the ratio of the power in the reflected wave to the incident wave. More specifically, it equals the square of the magnitude of the pressure reflectance which is demonstrated by the equation 4.
\[ R = |R(f)|^2 \quad \text{(Eqn.4)} \]

Hence, the power reflectance is a real number and while it includes no phase information, is constant along the tube’s length. This fact makes it to be independent of the measurement position.

On the other hand, absorbance describes the sound power that is absorbed by the tube’s termination. Therefore, it is achieved by subtracting the power reflectance from the value of one which is stated in equation 5a. Absorbance can also be quantified in dB levels and be called transmittance (equation 5b).

\[ A = 1 - R \quad \text{(Eqn.5a)} \]
\[ T = 10 \log_{10} A \quad \text{(Eqn.5b)} \]
Chapter 2

Method and Materials

In this chapter first, different techniques for measuring the acoustic immittance in the ear canal are introduced. The basic principle for the acoustic reflex detection is described. Subsequently, different techniques for detecting the acoustic reflex, which have been implemented in other research, are stated. Stimulus, data acquisition system and measurement assembly implemented in the current project are then explained. Besides, it is described how acoustic immittance, reflectance and their alterations caused by the acoustic reflex are measured.

2.1. Different Techniques for Measuring Immittance in the Ear Canal

There are four different techniques for measuring the acoustic immittance in the ear canal. Two are direct and two are indirect measurements [7].

The direct methods use the sound pressure produced by a volume velocity source. In the first method the equivalent sound source is calibrated and its characteristics are determined. Then it estimates the unknown immittance from sound pressure measurements in the ear canal. The second method (acoustic bridges) uses the comparison of sound pressures produced in loads of known and unknown immittance.

The indirect methods use the measurements of acoustic reflectance to determine the acoustic immittance. In the first method a long tube is coupled to the ear canal and sound pressure is measured along the length of the tube. The second wave-tube method measures the sound pressure at different locations within the ear canal.

The technique of measuring acoustic impedance which is going to be used by the current project is categorized in the first direct method.

2.2. Basic Principle for the Acoustic Reflex Detection

As it was briefly mentioned in the introduction of this paper, detecting the acoustic reflex is based on the idea of occurring alterations in the acoustic impedance and consequently reflectance measured in the middle ear which is due to the stapedius-muscle contraction. Accordingly, detecting the acoustic reflex is obtained in three steps which are first, measuring the power reflectance in the middle ear in normal quiet circumstances (baseline), second, stimulating the ear to elicit the acoustic reflex and measuring the power reflectance
in this condition (post activator effect) and finally comparing these two power reflectance to get the reflectance shifts caused by the acoustic reflex [6,8].

2.3. Different Techniques for Measuring the Acoustic Reflex

In terms of which kind of probe signal is used in order to measure its reflectance at the tympanic membrane, which kind of activator is used in order to elicit the acoustic reflex and either ipsilateral or contralateral ear are investigated, there are different techniques to detect the acoustic reflex.

The clinical middle-ear muscle reflex tests usually utilize the 226 Hz probe tone to get the response from the ear canal and measure its reflectance. The reflex elicitor can be presented either to the same ear in which the probe reflectance is monitoring (ipsilateral) or to the opposite ear (contralateral). The common stimuli applied as the reflex elicitor are 0.5, 1, 2 and 4 kHz tones as well as the broadband noise (BBN) [8].

Another approach is to use wideband probe stimulus instead of 226 Hz tone, for measuring its reflectance in the ear canal. This technique which was employed by Feeney and Keefe at 1999 for the first time applied chirps as the probe signal [9]. Feeney et al. did investigation on a group of infants and adults using a train of 8 chirps, each with duration equal to 40 msec, as a probe signal and a contralateral stimulus consist of wideband noise between 2.5-11 kHz [6]. Also, another study has been carried out by Keefe et al. in which a probe signal is made of 1 sec of 22 clicks while the ipsilateral stimulus is 500 msec of BBN [8].

The advantages and disadvantages of different techniques will be discussed later in chapter4.

2.4. Stimulus, Data Acquisition and Measurement Assembly

In the present study, the probe signal consists of 10 chirps (sweep frequency in the range of 100-8000 Hz), each with 20 msec duration and 30 msec silence between each two. The ipsilateral reflex elicitor is composed of 1 sec broadband noise (BBN) with cutoff frequencies equal to 100 Hz and 8 kHz. Totally, the whole stimulus contains first, a chirp train with approximate duration of 500 msec and then iteratively one BBN and one chirp train repeated for five times.

The responses to the 10 chirps in the first train which are recorded in absence of the reflex elicitor are averaged and it contributes to the baseline reflectance. The responses to the first two chirps after the offset of each five BBN activator are averaged and it contributes to the post activator reflectance. The reason for choosing the response from the first two chirps is
the fact that the effect of stapedius contraction, in particular, the acoustic reflex remains around 100 msec after the offset of each activator [8].

Since chirps are short time signals and involve wide band of frequencies, they are widely implemented as excitation or stimulus signals for investigation the parameters and their changes in dynamic objects in various areas of engineering, for instance, in biomedical investigation including bioimpedance measurement and impedance spectroscopy [11].

In fact, a chirp is a signal of many cycles with sinusoidal oscillation, a short time upsweep or downsweep frequency signal. In other words, the frequency increases or decreases with the time respectively. There are different types of chirp such as linear, quadratic, and logarithmic depending on how the frequency changes over the time. The chirp used in this project is logarithmic upsweep signal in which the instantaneous frequency is given by equation 6a and 6b.

\[ f(t) = f_0 \cdot \beta^t \quad \text{,} \quad \text{(Eqn6.a)} \]
\[ \beta = \left( \frac{f_1}{f_0} \right)^{\frac{1}{t_1}} \quad \text{, where } f_1 > f_0 \quad \text{,} \quad \text{(Eqn6.b)} \]

An example of the logarithmic chirp signal and the spectrogram of it are illustrated in figure 4a and figure 4b respectively [12,13].

The stimulus is digitally generated by the use of programming in MATLAB and data acquisition toolbox software installed in a computer which is connected to the data acquisition hardware. National Instrument, NI USB-6212, is the interface implemented for this project. The data acquisition hardware plays the role of both analog to digital and digital
to analog converter. The generated data from the computer, the stimulus, is converted into an analog signal and output to the earphone. The response signal from the examined object is input to the microphone, amplified through low noise Etymotic research system and converted into bits which can be read by the computer and analyzed to extract meaningful information.

The components of the data acquisition system used in this project as well as their relationships are illustrated in figure 5.

![Data Acquisition System Diagram]

**Fig. 5.** Data acquisition system’s components and their relationships.

The measurement assembly is comprised of an Etymotic Research, model ER-10B+ low noise microphone and an ER-2 earphone together with a standard compressible foam earplug which can be inserted into the ear canal with full insertion depth (14 mm if possible, depending on the shape and size of the ear canal [6]) as well as into the calibration tube. According to the Keefe et al. [10], the design applied in this equipment let the microphone probe tip extends approximately 3 mm beyond the foam surface where the earphone probe tip is and this helps to reduce the contribution of the evanescent mode coupling between the source and the receiver. In order to have better seal in the ear canal, it has been suggested allowing the earplug to be kept in the ear canal for two minutes prior to reflex measurement which is used in this project as well [6].
2.5. Electrical Equivalent of the Measurement Device and the Ear Canal

The measurement device is modeled as a voltage source and internal impedance which is connected to the ear canal in a series circuit. The ear canal acts as a load for this electrical circuit. In terms of acoustic, sound pressure difference plays the role of voltage difference. Therefore, the voltage source is replaced by the sound pressure generator in the circuit shown in figure 6. The goal is to measure the impedance of the ear canal. Then it is possible to determine the reflectance based on the known theory which will be explained in the following parts of this paper.

The pressure at the entrance of the ear canal is recorded by the means of microphone. Thus, if the Thevenin (Norton) parameters of the source, more specifically, the measurement device parameters are found, using pressure divider equation gives the impedance of the ear canal.

2.6. Calibration: Determination of the Source Characteristics

In order to determine the Thevenin pressure \( P_0 \) and the Thevenin impedance \( Z_0 \) of the sound delivery system, four cavities method developed by Keefe et al. [10] is used in this project. We use four tubes of optimized lengths \( L_1 = 1.1, L_2 = 1.7, L_3 = 2, L_4 = 3 \) cm, each with diameter of 8 mm, as a load at a time for the Thevenin circuit (figure 7). The tube diameter is selected to be similar as much as possible to the average human ear-canal diameter. The optimized lengths are chosen based on a study by Voss et al. [14].
Fig. 7. The Thevenin circuit consists of the measurement device and the calibration tube.

A train of 10 chirps with the same features as previously explained as the stimulus is transmitted to each calibration tube and simultaneously recorded by the microphone at the tube’s entrance. The pressure at each tube’s entrance is computed then through two following steps; first, correlation and averaging in time domain, second, calculating one-third octave intervals in frequency domain.

2.6.1. Cross-Correlation and Averaging

The first point where the maximum match between the transmitted chirp signal and the reflected signal occurs is detected by cross-correlating two signals in time domain using MATLAB software. Then, according to the predefined time distances between transmitted chirp signals in a train, their echoes are detected and averaged to form the pressure response. Averaging helps to enhance the signal to noise ratio of the signal recorded by the microphone.

2.6.2. Calculation of the Pressure at One-third Octave Intervals by means of Moving Average Technique

The average of detected sequences in the signal back to the microphone which have maximum correlation with the chirp signal is transformed into frequency domain using Fast Fourier Transform. Since analyzing the pressure response for all frequencies of its spectrum is impractical and time consuming, the whole spectrum is divided into one-third octave bands with central frequencies equal to 125, 250, 500, 1000, 2000, 4000 and 8000 Hz which are of importance for audio analysis. The one-third octave interval is defined as its upper frequency limit is given by its lower frequency limit times the cube root of two. Specifically, considering $f_c$ as the center frequency, the lower band-edge frequency is given by the formula $f_{\text{low}} = f_c / 2^{\frac{1}{6}}$ and the upper band-edge frequency is given by the formula
f_{upp} = f_c \cdot 2^{\frac{1}{6}} \text{ which contribute to the percent fractional bandwidth per each one-third octave interval being constant as } BW = 100 \left[ \left( f_{upp} - f_{low} \right) / f_c \right] \approx 23.2\% .

The sampling frequency is set to 20 kHz assuring to avoid aliasing. The moving average technique is used to taking average of spectrum in each one-third octave interval and assigning it to the corresponding central frequency. This can also regarded as a low pass filtering smoothen the spectrum. As a result, the pressure response recorded by microphone is now obtained at central frequencies.

According to pressure divider equation in the circuit described in figure 7, the relation between the source pressure \((P_0)\) and the pressure recorded by microphone at the entrance of the tube \((P_L)\) is expressed by equation 7.

\[
P_0 = P_L \cdot \left( \frac{Z_0}{Z_L} + 1 \right) , \quad \text{(Eqn.7)}
\]

The tubes are assumed to be viscothermal lossless, thus, the impedance at the entrance of the closed ending tube is estimated by equation 8 [7,10]. It indicates that the impedance is frequency dependent and complex quantity.

\[
Z_L = -j \frac{\rho_0 c}{S} \cot\left( \frac{2\pi f}{c} L \right) , \quad \text{(Eqn.8)}
\]

Where, \(\rho_0\) is the density of the air, \(c\) is the speed of sound, \(S\) is the cross-sectional area of the tube, \(f\) is the frequency, and \(L\) is the tube’s length.

All calculations are done in frequency domain. Accordingly, the pressure spectrum at the tube entrance and the frequency-dependent impedance of the tube (calculated at one-third octave central frequencies) are known. Considering two unknown parameters of \(P_0\) and \(Z_0\), solving equation 7 for two different length of tubes is sufficient to find out the Thevenin parameters. Increasing the number of tubes leads to an over-determined equation system which results in more accurate estimates of the Thevenin parameters [10]. When 4 numbers of tubes are used then the equation 7 can be written in the matrix form which gives the equation 9.

\[
\begin{pmatrix}
P_{L_1} Z_{L_1} \\
P_{L_2} Z_{L_2} \\
P_{L_3} Z_{L_3} \\
P_{L_4} Z_{L_4}
\end{pmatrix} = \begin{pmatrix}
Z_{L_1} & -P_{L_1} \\
Z_{L_2} & -P_{L_2} \\
Z_{L_3} & -P_{L_3} \\
Z_{L_4} & -P_{L_4}
\end{pmatrix} \begin{pmatrix}
P_0 \\
Z_0
\end{pmatrix} , \quad \text{(Eqn.9)}
\]

If we name the column array containing 4 rows of the product of the measured pressure in each tube and the tube impedance as \(A\) and the two column array of 4 rows of the tube impedance and -1 times the measured pressure as \(B\) then the estimated least-square values for the Thevenin parameters are calculated according to equation 10 [7].
\[
\begin{bmatrix}
P_0 \\ Z_0
\end{bmatrix}
= (B^T B)^{-1} B^T A ,
\]  
(Eqn.10)

However, according to Keefe et al. several factors such as chirp stimulus waveform, the sensitivity and the frequency response of the transducers and connected signal conditioners as well as the properties of the calibration tubes have an effect on the Thevenin parameters [10].

2.7. Measurements in the Ear Canal

After determination of the source characteristics, the described stimulus is transmitted to the ear canal in order to process chirp’s reflection both in quiet condition (baseline) and in activator condition (post activator effect).

As it was stated in the stimulus and data acquisition part, a series of chirps and BBN are transmitted to the ear canal. The responses to the chirps are averaged to form the ear canal pressure response with high signal to noise ratio. Therefore, it is of importance to determine where each chirp echo exactly starts and where it ends in the signal back to the microphone. Correlation technique which previously explained for calibration tubes is again used for this purpose.

2.8. Calculating Acoustic Impedance and Reflectance Shifts in the Ear Canal

Considering the electrical equivalent of the measurement device in series with the ear canal as a load of the circuit illustrated in figure6, again using pressure divider equation results in solving for the ear canal impedance. As it is stated by equation11, the only unknown quantity is the ear canal impedance \( Z_{EC} \). The Thevenin parameters of the source \( (P_0 \text{ and } Z_0) \) were determined by calibration method and the pressure at the entrance of the ear canal \( (P_{EC}) \) was determined by averaging the correlated parts of the reflected signal recorded by microphone.

\[
Z_{EC} = Z_0 \left( \frac{P_{EC}}{P_0 - P_{EC}} \right) ,
\]  
(Eqn.11)

It should be taken into account that all quantities here are frequency-dependent which have complex values, including real part and imaginary part. Also, all calculations are done at central frequencies of one-third octave bands.

There is a theory based on that, the reflectance then will be computed from the impedance of the ear canal [10,14]. The relationship between the normalized impedance \( (Z_n) \) and the reflectance \( (R) \) is expressed by equation12.
According to equation 13, the ear canal impedance \((Z_{EC})\) has been normalized by the characteristic impedance of the ear canal \((Z_c)\).

\[
Z_n = \frac{Z_{EC}}{Z_c} , \quad \text{(Eqn.13a)}
\]

\[
Z_c = \frac{\rho_0 c}{A} , \quad \text{(Eqn.13b)}
\]

Where \(\rho_0\) is the density of air, \(c\) is the speed of sound and \(A\) is the cross-sectional area of the ear canal.

The purpose of normalization is to reduce between-subject and between-test variability. In fact, there are variations in physical size of the ear between different participants (an important consideration for comparing male, female and juvenile data) as well as in ambient temperature and atmospheric pressure at the time of testing. Both \(\rho_0\) and \(c\) are functions of temperature and pressure. Hence, normalization by the characteristic impedance reduces this variability problem [15].

The constants at 20°C temperature used for all calculations are shown in table 1.

Finally, the power reflectance that is the square of the reflectance magnitude is computed separately for the baseline and for the post-activator effect. Their corresponding power reflectance are then compared to detect the shifts occurred between them which are induced by the acoustic reflex. In fact, the reflectance shift which is defined by equation 14, is the post-activator power reflectance \((R_p)\) minus the baseline power reflectance \((R_b)\) [6, 9, 17].

Furthermore, admittance shift is another parameter of investigation which is defined by equation 15. This is the ratio of admittance magnitude in post-activator condition \((|Y_p|)\) minus admittance magnitude in baseline condition \((|Y_b|)\) relative to the baseline [6, 8, 9, 17].

\[
\Delta R = R_p - R_b , \quad \text{(Eqn.14)}
\]

\[
\Delta Y = \frac{|Y_p| - |Y_b|}{|Y_b|} , \quad \text{(Eqn.15)}
\]
Table 1. Table of constants used for calculations.

<table>
<thead>
<tr>
<th>Name</th>
<th>Constant</th>
<th>Value</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>Speed of sound</td>
<td>$C$</td>
<td>34321</td>
<td>cm/s</td>
</tr>
<tr>
<td>Density of air</td>
<td>$\rho_0$</td>
<td>0.001204</td>
<td>g/cm$^3$</td>
</tr>
<tr>
<td>Ear canal diameter</td>
<td>$d$</td>
<td>0.8</td>
<td>cm</td>
</tr>
<tr>
<td>Tube diameter</td>
<td>$d$</td>
<td>0.8</td>
<td>cm</td>
</tr>
<tr>
<td>Ear canal area</td>
<td>$A$</td>
<td>0.5027</td>
<td>cm$^2$</td>
</tr>
<tr>
<td>Tube area</td>
<td>$S$</td>
<td>0.5027</td>
<td>cm$^2$</td>
</tr>
</tbody>
</table>

There are several factors that affect the measurement of immittance and consequently reflectance in the ear canal [7,16]. These factors including measurement location within the ear canal (distance from tympanic membrane), estimates of ear canal cross-sectional area, variations across canal cross-section, variation in middle ear cavity volume, variation in the rigidity of the ear canal walls and viscosity losses will be discussed later in this study.

For instance, the assumed value for the ear canal diameter is chosen very close to the average diameter of the human ear canal which is equal to 0.74 cm. However, it has been confirmed that an error in the assumed canal diameter affects the magnitude of the reflectance but the general shape remains the same. In a study by Voss et al. they have shown this fact by investigating calculations with an area both 20 percent greater and 20 percent less than the average diameter [14]. Their result correspond to two different subject are illustrated in figure 8. It also shows that the reflectance magnitude in different subjects is not exactly same as each other.

Although the shape of power reflectance in different subjects, with different assumptions as well as different measurement techniques is not similar but they approximately conform the same pattern. The result of two other studies of power reflectance are illustrated in figure 9a [18] which correspond to two adult subjects and figure 9b [19] which represents the reflectance in left and right ears, collapsed across age group.
Fig. 8. Reflectance in two subjects, solid lines correspond to the assumed canal area with an average diameter, dotted lines correspond to the assumed canal area with diameter twenty percent smaller and dashed line correspond to twenty percent greater than the average [14].

Fig. 9. Power reflectance curve resulted from two different studies, (a) and (b).

Also, the result of different measurement techniques for detection of the admittance shifts and the reflectance shifts more or less conform the same general pattern which can be seen in different studies by Feeney et al. for instance, figure 2 and 3 of a research article [6], figure 2 and 4 of another [9] and figure 1 and 2 of a further [17].

Figure 10 shows one example of admittance shifts and reflectance shifts detected by wideband ipsilateral measurement at different levels of activator including a maximum of 90 dB SPL in 4 dB descending steps which are represented by lighter and thinner lines [8].
Fig. 10. Normalized admittance shifts (top) and normalized reflectance shifts (bottom), obtained by wideband (1 sec of 22 clicks as the probe signal) ipsilateral measurement at different levels of activator (500 msec BBN) including a maximum of 90 dB SPL in 4 dB descending steps, represented by lighter and thinner lines [8].
Chapter 3

Results and Analysis

In this chapter first, the results of the calibration procedure and the determination of Thevenin parameters are demonstrated. Second, the accuracy and sensitivity of the implemented method in the current project, for measuring the power reflectance in the ear canal, are investigated and its results are presented. Finally, the measured reflectance shifts and admittance shifts induced by stapedius-muscle contraction are illustrated and analyzed.

3.1. Calibration and Thevenin Parameters

The stimulus contains a train of 10 chirps with characteristics described in chapter 2 was transmitted to the calibration tubes. The root-mean-square level of the chirps recorded by microphone was set to 65 dB SPL. Figure 11 illustrates a part of the train which has been zoomed to be distinguishable.

![Graph](image)

**Fig. 11.** Part of the chirp train as the probe signal.

We tried transmitting chirps at 80, 75 and 70 dB SPL as well. Although higher levels of chirps resulted in better correlation achieved between transmitted and reflected signals as well as with higher signal to noise ratio and without any need of amplifying signal in the receiver,
but those higher levels may elicit the acoustic reflex. It may be the reason that we could not detect reflectance shifts in the ear canal using those levels of chirps. For instance, figure 12 shows high correlation between a 80 dB SPL chirp signal of a chirp train presented to the 1.1cm calibration tube (bottom row) and the reflected signal of the last chirp of the train (middle row) as well as the averaged reflected signal of 10 chirps (top row). Apparently, averaging reduced the noise and yielded to the smoothed signal. Also, considering 20msec length of each chirp of the train, we used 1msec prior and 1msec following each reflected chirp for averaging. In particular, totally ten 22msec reflected signals were averaged.

![Fig. 12. An example of 80 dB SPL chirp signal of a chirp train presented to the 1.1cm calibration tube (bottom row), the reflected signal of the last chirp of the train (middle row), the averaged reflected signal of 10 chirps recorded by microphone (top row).](image)

The results from calibration tubes of 4 different lengths ($L_1 = 1.1$, $L_2 = 1.7$, $L_3 = 2$, $L_4 = 3$ cm) are illustrated through figures 13 to 24. For each tube, the probe stimulus and the reflected signal recorded by microphone are represented both in time domain and in frequency domain. Besides, spectrogram of the probe and the microphone are demonstrated (figures 13, 16, 19, 22).

Results of correlation and averaging according to the process stated above are also shown for each tube (figures 14, 17, 20, 23). These results achieved using an option of 40 dB amplifying in receiver which has been provided by Etymotic Research ER-10B+ low noise microphone. Indeed, without using this modification, reflected signals of chirps at 65 dB SPL involved high amount of noise and could not be correlated. Although with using +40 mode of microphone system, reflected signals still involved more noise compared to figure 12 where chirps are presented at 80 dB SPL, but this amount of noise were acceptable for obtaining reflectance shifts in the ear canals.
Subsequently, the averaged reflected signal recorded by microphone that forms the sound pressure at the entrance of the tube was transformed to the frequency domain by Fast Fourier Transform. Then pressures for one-third octave bands with central frequencies equal to 125, 250, 500, 1000, 2000, 4000 and 8000 Hz were calculated for each tube (figures 15, 18, 21, 24).

Calibration process based on Thevenin equivalent circuit, calculated impedance and pressure at the entrance of each tube resulted in Thevenin source parameters, $P_0$ and $Z_0$, which are complex numbers over central frequencies of one-third octave intervals.

$L_1 = 1.1$ cm:

**Fig. 13.** Top row, from left to right: probe stimulus contains a train of 10 chirps in time domain, spectrogram of the stimulus, probe stimulus in frequency domain. Bottom row, from left to right: reflected signal recorded by microphone, spectrogram of the recorded signal, amplitude of reflected signal in frequency domain. All measured in the calibration tube with the length of 1.1cm.
Fig. 14. A 65 dB SPL chirp signal of a chirp train presented to the 1.1cm calibration tube (bottom row), the reflected signal of the last chirp of the train (middle row), the averaged reflected signal of 10 chirps recorded by microphone (top row).

Fig. 15. The amplitude of averaged reflected chirp signal recorded by microphone in frequency domain (top row), its representation in one-third octave intervals (bottom row) correspond to 1.1cm calibration tube.
$L_2 = 1.7\text{ cm}$:

**Fig. 16.** Top row, from left to right: probe stimulus contains a train of 10 chirps in time domain, spectrogram of the stimulus, probe stimulus in frequency domain. Bottom row, from left to right: reflected signal recorded by microphone, spectrogram of the recorded signal, amplitude of reflected signal in frequency domain. All measured in the calibration tube with the length of 1.7cm.

**Fig. 17.** A 65 dB SPL chirp signal of a chirp train presented to the 1.7cm calibration tube (bottom row), the reflected signal of the last chirp of the train (middle row), the averaged reflected signal of 10 chirps recorded by microphone (top row).
Fig. 18. The amplitude of averaged reflected chirp signal recorded by microphone in frequency domain (top row), its representation in one-third octave intervals (bottom row) correspond to 1.7cm calibration tube.

$L_3 = 2 \text{ cm}$:

Fig. 19. Top row, from left to right: probe stimulus contains a train of 10 chirps in time domain, spectrogram of the stimulus, probe stimulus in frequency domain. Bottom row, from left to right: reflected signal recorded by microphone, spectrogram of the recorded signal, amplitude of reflected signal in frequency domain. All measured in the calibration tube with the length of 2cm.
Fig. 20. A 65 dB SPL chirp signal of a chirp train presented to the 2cm calibration tube (bottom row), the reflected signal of the last chirp of the train (middle row), the averaged reflected signal of 10 chirps recorded by microphone (top row).

Fig. 21. The amplitude of averaged reflected chirp signal recorded by microphone in frequency domain (top row), its representation in one-third octave intervals (bottom row) correspond to 2cm calibration tube.
$L_4 = 3 \text{ cm}$:

**Fig. 22.** Top row, from left to right: probe stimulus contains a train of 10 chirps in time domain, spectrogram of the stimulus, probe stimulus in frequency domain. Bottom row, from left to right: reflected signal recorded by microphone, spectrogram of the recorded signal, amplitude of reflected signal in frequency domain. All measured in the calibration tube with the length of 3 cm.

**Fig. 23.** A 65 dB SPL chirp signal of a chirp train presented to the 3 cm calibration tube (bottom row), the reflected signal of the last chirp of the train (middle row), the averaged reflected signal of 10 chirps recorded by microphone (top row).
3.2. Accuracy and Sensitivity of the Implemented Method for Measuring Power Reflectance

In order to investigate the accuracy of the measurement system, first, the power reflectance in normal circumstances with the stimulus probe containing a train of 10 chirps was examined in a subject’s ear canal. As figure 25 shows the pattern of the resulted power reflectance was in good agreement with the general pattern illustrated in figure 9 [18,19] of chapter 2. Power reflectance was scaled between values 0 and 1 and was examined over one-third octave frequencies from 125 to 8000 Hz. According to Keefe et al., in frequencies above 10.7 kHz the effect of evanescent mode becomes considerably involved in the measurements of the adult ear canals [18]. For this reason, the frequencies above 8 kHz were excluded in our investigation. Reflectance value equal to zero represents no reflectance occurs and almost all sound power are absorbed (refer to equation 5) and transmitted into the middle ear. On the other hand, reflectance value equal to 1 represents no transmission of sound energy into the middle ear and all reflected back.

Accordingly, power reflectance curve in figure 25 indicates that, at low frequencies there is high energy reflectance while it is decreasing with increasing frequency up to 2 kHz which seems to be the absorption resonant frequency depends on resonant structures of the subject’s middle ear. Also, the result confirms that minimum reflectance occurs in the range of 1-4 kHz and sound power mostly transmitted into the middle ear in this range of frequency which is most important for speech perception. At higher frequencies, power reflectance begins to increase.
In other words, since the power reflectance is derived from the acoustic impedance (refer to the equation 3 and 4) and the major frequency-dependent part of the impedance is the reactance (refer to the definition of impedance: \( Z = R + j \left( \omega L - \frac{1}{\omega C} \right) \)), the decrease of power reflectance in low frequencies is consistent with a stiffness-controlled system in which \( C \) is dominant and its increase at frequencies above the resonance frequency is consistent with a mass-controlled system in which \( L \) is dominant.

Similarly, the resulted transmittance in dB levels is illustrated in figure 26.

![1/3 octave ear canal power Reflectance](image1)

**Fig. 25.** One-third octave power reflectance versus center frequency (Hz) measured in the ear canal.

![1/3 octave ear canal transmittance](image2)

**Fig. 26.** One-third octave transmittance (dB) versus center frequency (Hz) measured in the ear canal.
Second, in order to investigate the sensitivity of the measurement system, the same subject was asked to hold breath, close her nose and force pressure into her ears. The effect of more pressure in the middle ear cavity was examined on power reflectance measured in the ear canal. The resulted power reflectance and transmittance are illustrated in figure 27 and 28 respectively. It indicates that there are some changes in power reflectance in this condition compare to the normal condition in her ear canal as it was expected in general.

For instance, a valley followed by a peak was observed at lower frequencies of power reflectance in second condition. The power reflectance at 500 Hz increased from value close to 0.56 in first condition to a value close to 0.92 in second condition while at 1000 Hz it decreased from approximately 0.44 to 0.16.

In fact, these changes are not similar for different subjects as well as for same subject in different experiments which is due to several factors affected the measurement. We will discuss about sources of variability and complications later in chapter4. However, the resulted changes in power reflectance due to more pressure in the middle ear cavity proved the sensitivity of our measurement system in general.

![1/3 octave ear canal power Reflectance](image)

*Fig. 27. One-third octave power reflectance versus center frequency (Hz) measured in the ear canal while subject holds breath and forces pressure into her ears.*
Fig. 28. One-third octave transmittance (dB) versus center frequency (Hz) measured in the ear canal while subject holds breath and forces pressure into her ears.

3.3. Reflectance Shifts and Admittance Shifts Induced by Stapedius-Muscle Contraction

The acoustic reflex detection was performed on 10 adult participants including 5 male subjects and 5 female subjects. The whole stimulus duration was around 8 sec begins with a train of 10 chirps with the total duration around 0.5 sec, then 1 sec BBN and repeated for 5 times as it is illustrated in figure 29. Considering 2 minutes prior to measure the acoustic reflex, letting the earplug to expand in the ear canal, the whole experiment duration for each ear was around 2-3 minutes if the collected data was acceptable. For instance, if the recorded signal was too noisy and unable to be correlated the program returned error and stop to work. Also, in this project we did not consider the point of choosing left ear or right ear. For each subject an ear which resulted better shifts was taken into account. The overall root-mean-square level of BBN recorded by microphone was set to 90 dB SPL and for chirps it was set to 65 dB SPL. Because of small intensity of chirps compare to BBNs they are not visible in the scale of figure 29. Figure 30 illustrates a part of the stimulus which has been zoomed to show the chirp signals. Besides, spectrogram of the probe stimulus and the signal recorded by microphone are illustrated in figure 31.
Fig. 29. The whole stimulus presented to the ipsilateral ear canal, contains chirps and BBN signals.

Fig. 30. A part of stimulus presented to the ipsilateral ear canal, contains chirp signals.
The resulted reflex-induced shifts in power reflectance separately averaged in male and female subjects as well as averaged in total 10 participants are plotted in figures 32 and 33 respectively. The shifts in admittance magnitude were also investigated in the same way and the results are illustrated in figures 34 and 35. Blue color represents shifts correspond to male subjects and green color represents shifts correspond to female subjects while the error bars stand for the variances between participants.

The mean shifts averaged in 10 participants indicated that the most changes in power reflectance as well as in admittance between the baseline condition and the post-activator condition have been occurred at frequencies lower than 1000 Hz. This was consistent with changes in stiffness at low frequencies induced by stapedius-muscle contraction.

The mean curve of reflectance shifts averaged in 10 participants began to increase from a positive value close to zero at 125 Hz and had a peak by 250 Hz, then decreased again to a value close to zero at 1kHz and approximately being around zero for higher frequencies. More specifically, there were positive shifts at frequencies below 1000 Hz while the mean shifts between 1 and 4 kHz were negative and of reduced magnitude compared to lower frequencies. Again, at 8 kHz the mean shift was positive but relatively small in magnitude. Therefore, the most effective frequencies to show the reflex induced shifts were up to 4 kHz.
The positive shifts implied that the power reflectance has been increased in post-activator condition compared to baseline condition. On the other hand, transmittance has been decreased where the shifts are positive. As a result, more transmittance occurred in the range of 1 kHz to 4 kHz in post-activator condition relative to baseline. This can be of importance for speech perception.

The mean curve of admittance shifts averaged in 10 participants began from a negative value by 125 Hz, had a valley at 250 Hz and increased to the positive value at 1000 Hz, then the mean approached to zero by 2 kHz and higher frequencies.

However, the resulted variance in the admittance shifts was greater compared to the reflectance shifts. The variance among female subjects was larger than male subjects as well.
Fig. 32. One-third octave shifts in power reflectance \( (\delta r = R_p - R_b) \) versus center frequency, blue color represents averaged in males and green color represents averaged in females. The error bars represent calculated variances.

Fig. 33. One-third octave shifts in power reflectance \( (\delta r = R_p - R_b) \) versus center frequency, averaged in 10 participants. The error bars represent calculated variances.
Fig. 34. One-third octave normalized shifts in admittance magnitude ($\Delta Y = \frac{|Y_p| - |Y_b|}{|Y_b|}$) versus center frequency, blue color represents averaged in males and green color represents averaged in females. The error bars represent calculated variances.

Fig. 35. One-third octave normalized shifts in admittance magnitude ($\Delta Y = \frac{|Y_p| - |Y_b|}{|Y_b|}$) versus center frequency, averaged in 10 participants. The error bars represent calculated variances.
Chapter 4

Discussion

In this chapter, first, possible sources of variations in the measurements are discussed. Some of these variations originate from changes across canal cross-section, whereas others originate from variations in the rigidity of the ear-canal walls. Furthermore, the effect of viscosity losses are explained separately and our assumptions in the current project are discussed. Second, the advantages of wideband measurement using chirps as a probe instead of pure tone are reported and our experience of using different stimulus lengths is presented. Third, it is discussed why ipsilateral broadband reflex elicitor has been used in the presented project rather than contralateral pure tone. Finally, some crucial considerations and criteria in our experiment in order to obtain the reflectance shifts are stated.

4.1. Sources of Variability in Immittance/Reflectance Measurements on Human Ear Canal

There are some variability that complicates the measurement of immittance and consequently reflectance inside the human ear canal. These complications are consequences of variations across canal cross section, variations in the rigidity of the ear canal walls and, last but not least, variations in the viscous losses.

There are some other sources of variability that can possible affect the measurements. A research study by Voss et al. on normal cadaver ears demonstrates that the measurement location within the ear canal, or in other words the distance from tympanic membrane, have small and rather negligible effects on reflectance measurement [16]. However, their result indicates that variation in middle ear cavity volume has relatively large effect on resonances of the middle ear cavity and reflectance measurement. These variations are not currently considered in our study.

4.1.1. Variations across Canal Cross Section

Despite we assumed the ear canal to have a uniform cross section during our broad band measurements the ear canal cross section is not totally uniform along its entire length. Indeed, the canal geometry comprises local constrictions and expansions as well as bends and curves. Also, there is considerable tapering of the canal near the tympanic membrane. These deviations from uniformity are of more importance with increasing the stimulus frequency where the deviations become a significant fraction of a sound wavelength. In particular, it likely happens when the deviations are larger than a tenth of the wavelength of the sound [7]. Hence, this fact limits the use of high frequencies in wideband measurement.
of immittance and reflectance. In the current project, acoustic characteristics of the middle ear were investigated in the range of 100-8000 Hz.

Besides, it has been confirmed that the sudden variation in the ear canal area is a source of non-uniformity in sound pressure across a canal cross section. This can affect measurements at a single point in adult human ear canal and cause uncertainties in the immittance and reflectance at frequencies above 5-6 kHz [7].

One suggested technique which is less affected by these variations is based on finely spaced measurements of sound pressure within the ear canal. It enables investigations at very high frequencies, even as high as 20 kHz [7].

4.1.2. Variation in the Rigidity of the Ear Canal Walls

while we assumed the perfect rigid ear canal, there are variations in its rigidity. The outer half of the ear canal is composed of cartilage and other soft tissues. This can cause variations in the rigidity of the ear canal along its entire length.

In adults’ ear canals, the deviations from rigidity are small enough (compared to the compliance of the air in the canal) and can be ignored [7]. However, in case of infants and newborns, which are not subjects of the current project, this assumption is of less validity. It has been suggested that using stimulus with frequency range of 1000 Hz and higher is less affected by the variation in ear canal walls rigidity for infants and newborns [7].

4.1.3. The Effect of Viscous Losses

In the current project we ignored the viscous losses both in the ear canal and in the calibration tubes. In fact, part of the sound power is absorbed as sound travels between the measurement location and the reflector. This is because of the viscous interactions between the air and the walls in the ear canal and in the calibration tubes. Although these viscous losses play no substantial role over dimensions related to reflectance measurements in the human ear canal, their effect on calibration tubes are not negligible. Studies prove that viscous losses are most significant at the resonant and anti-resonant frequencies of the tube impedances and thermal losses are of importance at frequencies below 100 Hz [7]. Hence, considering viscothermal losses in calculations for calibration tubes results in more accurate fitting procedure used to characterize the output pressure and impedance of the measurement device.
4.2. Chirps, Wide Band versus Pure-tone Probe

In the current project we use chirps as a probe instead of pure-tone signals. Chirps have benefits from two points of view.

First, they are very short time signals thus can perfectly detect changes in dynamic objects which need quick measurements. In the case of reflectance measurement on human ear canal, many different parameters such as swallowing, breathing, even smallest body movement, pulsation of blood and any other biological noise can affect the measurement. Therefore, it is practically necessary to carry out the measurements in shortest possible time. Different duration of chirps and the whole stimulus were examined in current study to choose the one which gives better result. For instance, once, a train of 10 chirps, each with duration equal to 40 msec and 60 msec silence between each two were investigated as a probe signal. Another experiment was done with a stimulus consisting of around 1 sec chirp train with duration of 20 msec and 1 sec BBN each, repeated for 10 times (around 21 sec stimulus in total). Finally, it was decided to use a train of 10 chirps, with duration of 20 msec each one and subsequently, 1 sec BBN which are repeated for 5 times (around 8sec in total) for this study. This stimulus gives relatively desired result for the detection of acoustic reflex.

Second, the chirps involve a very wide range of frequencies. Whereas using pure tones (e.g. 226 Hz), as it is currently used in clinical measurement of acoustic reflex, provides information of the middle ear’s acoustic characteristics only at a specific frequency and provide no information at other frequencies. In other words, the use of wide band probe makes it possible to obtain reflectance changes over a wide bandwidth of chirps.

From this point of view, it has been confirmed that pure-tone measurements are insufficient for analyzing the reflectance shifts since pathologies may not be evenly apparent at the same frequency in every ear. Besides, the effective frequency ranges differ across different ages [8]. The wideband technique has been specially suggested to track the best frequency for measurements in infants and postnatal middle ear development [8]. Furthermore, it is reported that using wideband rather than a single frequency probe assures measurement of acoustic reflex at lower activator levels [8].

Hence, considering that the frequency range and the time duration of chirps can be designed independently from each other, chirps provide wideband high-speed measurement together with lower required activator level.

4.3. Ipsilateral versus Contra lateral and Broadband versus Pure-tone Activator

In the current project we used ipsilateral measurement with BBN activator. This method was chosen because many studies in acoustic reflex measurement agree that generally the
activator threshold for ipsilateral measurement is less than contralateral one as well as for broadband noise rather than pure-tone activator (e.g. 0.5, 1, 2 and 4 kHz) [8].

In fact, acoustic reflex threshold depends upon the technique of measurement, probe and activator type. For instance, reflex threshold using clinical method and contralateral broadband noise is reported in the range of 75 to 100 dB SPL in a research study [20]. Some cases of hearing loss and Tinnitus have been also reported after measurement of acoustic reflex using 120 dB HL activator [8]. Thus, there are safety-related concerns with this kind of measurements which emphasizes the use of lower activator levels. The current project presented broadband activator at level of 90 dB SPL in ipsilateral ear to avoid risk of using high intensity.

Furthermore, another advantage of ipsilateral measurement is that it can be implemented for young infants more simply in comparison with the contralateral one [8]. In other words, it paves the way for further use of the presented method for studying the acoustic reflex in the infants’ ears.

4.4. Critical Considerations and Criteria

According to our experiment a significant condition required in order to obtain the acoustic reflex was the considerable difference between sound intensity levels of the chirps and the BBN activator. As stated above, the overall BBN level was set at 90 dB SPL. We tried different intensity levels of chirps such as 80, 75, 70, and 65 dB SPL but the best shifts were captured at 65 dB SPL. Higher levels of chirps might act as acoustic reflex elicitors that were not of interest. The problem with lower levels of chirps was that the reflected signal involved high amount of noise which disturbed the main signal. Etymotic Research microphone utilized in this project, ER-10B+, provides two modes of amplifiers with noise control solutions to allow 20 dB and 40 dB higher output levels from the receiver. Consequently, using +40 mode was crucial to succeed in obtaining acoustic reflex shift with high signal to noise ratio.

Another important consideration was that the probe needs to be completely sealed in the ear canal. That was achieved by allowing the earplug to stay in the canal around 2 minutes prior to the measurement. Its effect on the amount of signal to noise ratio and thus on detection of valid shifts was much more than one would expect.

Also, the magnitude of the resulted reflectance shift at 250 Hz (approximately 0.45 in average) plotted in figure 33 was relatively higher than what we expected (less than 0.2) based on other studies of detecting acoustic reflex. That was also observed in admittance shifts. It may be caused by the less accurate baseline condition at 250 Hz. Further investigation is required to be able to evaluate the accuracy of the baseline and post-activator conditions at this frequency. For instance, one criterion suggested by Keefe et al. accepted the recorded data as satisfactory if the input resistance was positive [18].
Accordingly, assessing the real part of the ear-canal impedance may help achieve more precise analysis. Another criterion used by Feeney et al. suspected the leaky probe where the power reflectance in adults was lower than 0.8 at 250 Hz [6].

It is strongly suggested to take these criteria into account for further investigation. Moreover, increasing the number of transmitted chirps from 10 to 20 in baseline is another possibility that may lead to more accurate baseline reflectance.

Regarding the larger variance observed in admittance shifts compared to the reflectance shifts, one possible reason can be the fact that the admittance magnitude varies depending on the measurement point in the ear canal while the power reflectance is independent of the probe location provided that slow variations in cross-sectional area and low viscosity losses are assumed [7,14]. In fact, the distance between the probe microphone and the tympanic membrane affects the acoustic impedance of the ear canal. Therefore, variations of the earplug insertion depth among different subjects make this distance vary from measurement to measurement which may affect the measured admittance in the ear canal.

Finally, a number of other considerations can be recommended: Assessing audiogram test of participants prior to acoustic reflex measurement, calibrating the source parameters for each experiment, analyzing the reflectance at more central frequencies and performing the measurements in double-walled sound booth may contribute to more accurate results.
Chapter 5

Conclusion and Future Work

This project provided a wideband investigation of acoustic characteristics of the middle ear and their changes due to stapedius-muscle contraction, so-called acoustic reflex. Relatively complete characterization of the acoustic reflex using ipsilateral broad band noise at lower intensities was obtained here. The mean positive reflectance shifts and negative admittance shifts in low frequencies (below 1 kHz) confirmed that stapedius-muscle contraction increases the power reflectance as well as impedance of the ear canal. Our findings were consistent with a stiffness-controlled system for the low frequencies in which contraction of the stapedius muscle made the ossicular chain become stiffer which in turn, according to the electrical equivalent, increased the impedance of the ear canal. In other words, contraction of the stapedius muscle leads to increase of the ear canal’s impedance and thereby attenuation of the lower frequency components. This low-frequency attenuation can also be regarded as a protective function of the middle ear against high intensity sounds of low frequency, for example, during chewing.

Besides, our measurements of the reflectance shifts indicated that the most robust frequency region affected by the acoustic reflex was up to 4 kHz. Whereas, for the admittance shifts, this region was up to 2 kHz. Furthermore, small negative reflectance-shifts in the range of 1-4 kHz were observed. This suggests that at this frequency range (which corresponds to speech frequency band) the stapedius-muscle contraction was beneficial in the sense that the sound-transmittance was increased. This might result in better speech perception immediately after middle-ear muscle contraction.

It has been confirmed that in addition to presenting high sound intensity to the ear which makes the middle-ear muscle go to contraction, it can also occur shortly before vocalization [21]. Therefore, our result may help explain how this muscle contraction affects the hearing of one’s own voice as well as hearing the others voice during vocalization. This fact opens new possibilities which need further investigation. Besides, other potential uses of the presented project for assessing acoustic reflex on specific groups of subjects such as infants as well as hearing-impaired persons can be studied in the future.
References


