MATURATION OF SKULL PROPERTIES WITH IMPLICATIONS FOR THE FITTING AND VERIFICATION OF THE SOFT BAND BONE-ANCHORED HEARING SYSTEM FOR INFANTS AND YOUNG CHILDREN

by

ALLISON RUTH MACKEY

B.Sc., Queen's University, 2007

A THESIS SUBMITTED IN PARTIAL FULFILLMENT OF THE REQUIREMENTS FOR THE DEGREE OF

MASTER OF SCIENCE

in

The Faculty of Graduate Studies
(Audiology and Speech Sciences)

THE UNIVERSITY OF BRITISH COLUMBIA
(Vancouver)

March 2013

ABSTRACT

Soft band bone-anchored hearing systems (BAHS) are optimal for individuals with conductive or mixed hearing losses. Although it is understood that bone-conduction hearing is different between infants and adults, few studies have attempted to explain why these differences exist or how they affect the fitting of a soft band BAHS. The main objectives in this study were: (i) to better understand how properties of the developing skull contribute to the maturation of bone-conduction attenuation and sensitivity, and (ii) to determine how future BAHS fitting and verification protocols should be adjusted for infants and young children.

The transcranial attenuation of pure-tone bone-conduction stimuli was measured on infants and young children (age 1 month to 7 years) and adults using sound pressure in the ear canal when the transducer was placed on different positions across the skull. In addition, the mechanical impedance magnitude for the forehead and temporal bone was collected for contact forces of 2, 4, and 5.4 N using an impedance head, a BAHS transducer, and a specially-designed holding device. This study was the first to measure mechanical impedance of the skull, which is an essential component to bone-conduction hearing, on young children and infants.

Transcranial attenuation was greatest for young infants, and decreased throughout maturation. Attenuation was also greater from the forehead compared to the contralateral temporal bone for infants and children over 10 months of age. In addition, mechanical impedance was lowest for the youngest infants and increased throughout maturation for low frequencies, but for high frequencies, these infants had the highest impedance on the temporal bone only. The effect of contact force was significant for low frequencies, and the effect of placement was significant for high frequencies. These results suggest that the properties of the developing skull relate to infant-adult differences in transcranial attenuation, and the mechanical impedance of the skin and subcutaneous tissue may explain the infant-adult differences in bone-conduction sensitivity. The results also provide important implications for fitting and verifying output from the BAHS for infants and young children.

PREFACE

This study was reviewed and approved by the Clinical Research Ethics Board of the University of British Columbia (Certificate # H12-00398). This study is in collaboration with researchers at the Institute of Reconstructive Sciences in Medicine in Edmonton, Alberta.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

LIST OF ABBREVIATIONS

Abbreviation	Definition
A	Acceleration
ABR	Auditory brainstem response
ANOVA	Analysis of variance
ANSI	American National Standards Institute
ASSR	Auditory steady-state response
BAHA	Bone-anchored hearing aid
BAHS	Bone-anchored hearing system
BEST	Balanced electromagnetic separation transducer
C	Compliance
CROS	Contralateral routing of signal
CSF	Cerebral spinal fluid
dB	Decibels
dB HL	Decibels hearing level
dB SPL	Decibels sound pressure level
DPOAE	Distortion-product otoacoustic emission
DSL [i/o]	Desired sensation level input/output
F	Force
Hz	Hertz
IEC	International Electrotechnical Commission
LTASS	Long-term average speech spectrum
M	Mass
MANOVA	Multivariate analysis of variance
MRL	Minimum response level
N	Newton
P	Probability
R	Resistance
RECD	Real-ear-to-coupler difference
RETFL	Reference equivalent threshold force level

RF	Resonant frequency
SSD	Single-sided deafness
TM	Tympanic membrane
μN/V	Micro-Newton per Volt
VRA	Visual reinforcement audiometry
Z	Mechanical impedance

ACKNOWLEGEMENTS

First, I would like to acknowledge my supervisor, Dr. Susan Small for her support and advice throughout my thesis and for her time spent guiding me through the process. In addition, I would like to thank my committee members, Dr. Navid Shahnaz and Dr. Anna van Maanen for their input and feedback on my project. I am very grateful for the support and collaborative efforts of Dr. Bill Hodgetts and members of the Bone-Conduction Amplification Laboratory in Edmonton, Alberta. To my colleagues and members of the Pediatric Audiology Lab (Moko Chen, Ricky Lau, Myron Huen, Regina Wenk, Foong Yen Choong, Katie MacDonald, and Heather Holliday), thank you so much for your assistance and flexibility throughout the data collection phase of the project and for your continuous encouragement. Finally, I would like to acknowledge my family who have provided me with tremendous support throughout my graduate degree.

CHAPTER 1: Literature Review: The Transmission of Bone-Conducted Sound and the Bone-Anchored Hearing System for Infants and Young Children

1.1 Introduction

With the onset of newborn hearing screening programs, the number of newborns identified with hearing loss has increased. For example, Yoshinaga-Itano (2003) reported that between 1986-1992, an average of 1.5 newborns per year were identified with hearing loss in Colorado, and 86 newborns were identified in 1999 after the onset of newborn hearing screening. In order for children to receive the best possible opportunity for language development, optimal amplification fitting protocols for young children with all types of hearing loss is necessary. One amplification device available to infants and young children is the bone-anchored hearing system (BAHS). This device is specifically designed for individuals with a conductive or mixed hearing loss who cannot use air-conduction hearing aids. The BAHS is positioned on the temporal bone behind the ear and transmits sound to the cochlea via bone conduction. The BAHS was originally developed to fasten to an implant in the temporal bone; however, the BAHS can also be attached to a soft band that wraps around the head, which is typically used for children under age 5 years. Although the BAHS is clinically available and widely used, the fitting and verification protocols for the device are not well developed compared to the protocols we use to fit air conduction hearing aids, particularly for infants and young children.

One factor that requires consideration when fitting a BAHS is the infant-adult difference in bone-conduction thresholds. For both behavioural and physiological measures of threshold, infants have shown better sensitivity to bone-conduction stimuli, particularly for low frequencies (Hulecki & Small, 2011; Small & Stapells, 2008a). Small and Stapells (2008a) reported that auditory steady state response (ASSR) thresholds for 500 and 1000 Hz worsen, 2000 Hz improves, and 4000 Hz remains similar until at least 2 years of age. Few studies have investigated the infant-adult differences in bone-conduction hearing, and even fewer studies have investigated an explanation for why these differences exist. Although the

BAHS is clinically available for young children and infants, research is required to understand the mechanisms of bone conduction that contribute to infant-adult sensitivity differences in order to provide optimal fitting of these devices.

In addition, there is no standard clinical BAHS fitting and verification method designed specifically for infants and young children. The recommended fitting procedure for adults begins with determining if the individual is a candidate for a BAHS and selecting the appropriate device. Then, during the fitting session with the BAHS, bone-conduction thresholds are acquired with the device *in situ* through the manufacturer software. Specific recommendations for positioning the soft band BAHS on the head are very limited and often not evidence based. For example, manufacturers state that the BAHS can be positioned on a convenient position on the child's skull (e.g., Cochlear, 2011a), even though studies have shown that bone-conduction sensitivity is poorer when the transducer is placed on the forehead (Small, Hatton, & Stapells, 2007; Stuart, Yang, & Stenstrom, 1990) In addition, manufacturers suggest tightening the soft band so that it remains snug but comfortable on the child's head. Hodgetts, Scollie and Swain (2006) found that tightening the band from 2 to 5 N increased the output force only slightly when the BAHS was placed on an artificial mastoid, and therefore the authors suggested that soft band tightness is not as important as other factors, such as the volume control settings. Prescription algorithms for air-conduction hearing aids include independent formulae, such as the desired sensation level input/output model (Scollie et al., 2005). For calculating the prescribed output force for the BAHS, the only available algorithm was developed for adults by the manufacturer. To verify that the device is providing enough gain, it is recommended that aided thresholds through the soundfield are collected (e.g., National Deaf Children's Society [NDCS], 2010). Limitations for measuring aided soundfield thresholds include interaction with the noise floor and nonlinear processing in the BAHS (Nicholson, Christensen, Dornhoffer, Martin, & Smith-Olinde, 2011). Additionally, measures of aided threshold do not convey how sound is being processed by the device at a normal conversational level.

The current fitting and verification protocol requires a child to perform behavioural threshold responses numerous times, which is difficult for a young child performing visual reinforcement audiometry (VRA) and impossible for an infant under 6 months of age.

Verification procedures are particularly important for young children and infants who cannot provide subjective feedback of their listening experience. New procedures are in development for verifying the BAHS output through an artificial mastoid or skull simulator. Measures of force output are acquired from the device, which is attached to a skull simulator and placed in a hearing aid test box (Håkansson & Hodgetts, 2009). Test box measures are better suited for a young child who is unable to sit still and silent through *in situ* measurements. However, the artificial mastoid and skull simulator were developed with adult skull properties in mind (Håkansson & Carlsson, 1989). The mechanical impedance of the skull is one feature that was accounted for when developing the verification tools (Håkansson, Carlsson, & Tjellstrom, 1986). Flottorp and Solberg (1976) found that mechanical impedance on the skin-covered mastoid was slightly lower for children age 9-10 years compared to adults, but no study has investigated the skull properties for children younger than 9 years of age. This information is required to fill in the gap in the literature regarding mechanical impedance in the early years of life to provide a better framework for studies on the maturation of bone conduction hearing and to determine whether tools such as the artificial mastoid are appropriate to verify aided output from the soft band BAHS for infants and young children.

The current study investigates (1) the transmission of bone-conducted sound across the skull and (2) the mechanical impedance magnitude of the skin-covered skull for infants and young children and adults. For each participant, sound pressure was measured when a bone-conducted stimulus was presented at different positions on the skull via a probe-tube microphone positioned in the ear canal. Mechanical impedance was also measured using a BAHS transducer, impedance head and calibrated coupling device. The specific objectives of this study were as follows: (i) to compare attenuation of bone-conducted sound across 5 age groups (0-7 years and adults), frequency, and for different positions on the skull, (ii) to compare mechanical impedance across infants and young children of different ages and adults, (v) to compare mechanical impedance between skull positions that correspond to the placement of the BAHS, and (vi) to compare mechanical impedance between different contact forces that correspond to the soft band tightness. The following sections discuss what we know about the mechanisms of bone-conduction hearing, how maturation affects bone-conduction sensitivity, the history and design of the bone-anchored hearing system, the factors that affect bone-conduction transmission, the procedures for fitting hearing devices, and the mechanical impedance characteristics of the skull.

1.2 Mechanisms of Bone Conduction

Audiologists make use of one's ability to hear through bone conduction for both diagnostic and amplification purposes; however, our understanding of bone-conduction hearing is relatively limited, in part due to its complexity. Early studies hypothesized that bone-conduction stimulation may travel through a different auditory pathway from the cochlea than air-condition stimulation, or that a different organ is stimulated altogether. Researchers have since established that bone-conduction hearing is generated from the same cochlear stimulation as air-conduction hearing (Khanna, Tonndorf, & Queller, 1976; Stenfelt & Håkansson, 2002; Stenfelt, Puria, Hato, & Goode, 2003).

Transmission of bone-conducted sound begins with the vibration of the skull and results in pressure waves through the cochlea. Many theories have been proposed regarding the mechanism by which this process occurs. Because each theorized mechanisms originates from a specific part of the ear, each theory is explained in relation to this location in separate sections. Two theories, the compression theory and the theory of cochlear fluid inertia, involve mechanisms in the inner ear. The theory of middle ear ossicle inertia involves a mechanism in the middle ear. The osseotympanic theory involves a mechanism in the outer ear, and in addition, a phenomenon called the occlusion effect is described which corroborates the osseotympanic theory. Finally, a theory of soft-tissue conduction is discussed, which involves transmission of sound through the soft-tissue in the head. Figure 1.1 portrays comprehensive diagram of the mechanisms of bone conduction from Stenfelt and Goode (2005).

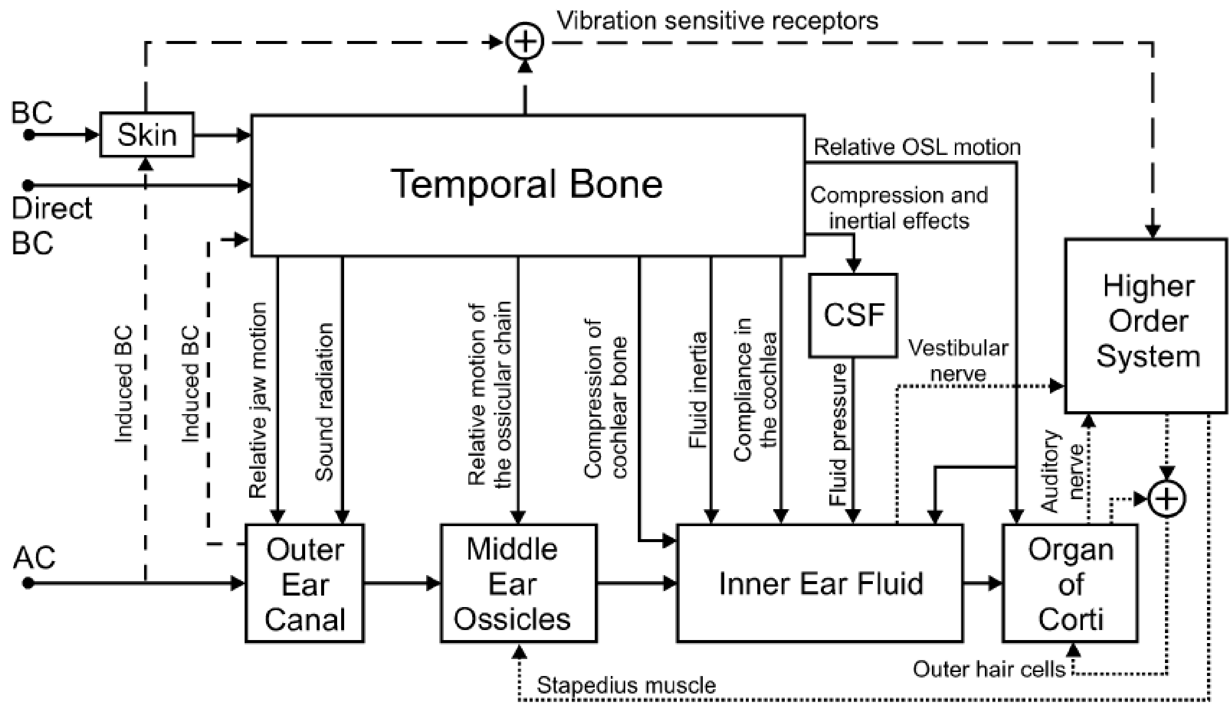

Figure 1.1 Schematic representation of the mechanisms of bone-conduction hearing. Presented with permission from Stenfelt and Goode (2005).

1.2.1 Inner ear

1.2.1.1 Compression theory

Tonndorf (1962) described the compression theory originally developed by Herzog and Krainz. The compression theory was based on the principle that a pressure difference exists between the two scalae within the cochlear shell. The theory assumed that pressure is created from a difference in stiffness between the round and oval window and a difference in volume between the scalae. Vibratory energy arising from a bone-conducted stimulus would cause the cochlear shell to compress and expand, changing the spaces containing the perilymphatic fluid. The scalae volume difference would be maintained while the fluid itself cannot be compressed. The compression of the cochlea would cause the round and oval window to flex with different volumes resulting in a pressure build-up across the cochlear partition (Tonndorf, 1962). Stenfelt, Hato, and Goode (2004) measured window volumes and found 5-15 dB more fluid was displaced at the round window than oval window for low frequencies, but for high frequencies (7 kHz and above), 5-15 dB more fluid was displaced at the oval window than the round window. The compression theory predicted that the pressure build-up in the cochlea would result in pressure waves across the cochlear partition (Tonndorf, 1962).

Subsequent research demonstrated that the compression theory was insufficient to fully explain bone-conduction hearing. First, when the stiffness of the round window was increased to match that of the oval window, eliminating the stiffness differential between the two windows, Kirikae (1959d) found no change in cochlear microphonics when a decrease was expected. Second, Carhart (1950) found that bone-conduction hearing was worse at certain frequencies for patients with otosclerosis. Otosclerosis involves the fixation of the stapes and stiffening of the oval window, which increases the stiffness differential between the two windows. Carhart's findings were the opposite of what would be predicted by the compression theory.

Currently, researchers believe that the compression theory plays only a minor role in contribution to bone-conduction hearing. Stenfelt, Hato and Goode (2004) measured volume displacement of the windows using laser Doppler vibrometer and confirmed that for high frequencies, little change in fluid flow at the round window occurred when the stapes was glued in place. Little change was also observed when a hole was drilled in to the footplate of the stapes. Based on these findings, Stenfelt and Goode (2005a) claimed that the compression theory has little effect on bone-conduction hearing, and any contribution would likely be for frequencies below 4000 Hz. For low frequencies, Stenfelt, Hato and Goode (2004) found fluid flow increased at the round window with alteration to the stapes; however, other studies have found that the cochlea actually does not physically distort at frequencies below 800-1000 Hz, but instead moves as one rigid mass (Håkansson, Brandt, Carlsson, & Tjellström, 1994), negating this theory for low frequencies.

1.2.1.2 Theory of cochlear fluid inertia

A second theory that involves fluids of the inner ear is based on inertia and vibration of the cochlear fluids. Kirikae (1959a) described a model by Ranke who represented the inertia of the inner ear as a U-shaped tube. If one side of the tube has more liquid, the difference in mass between the two columns of liquid would result in inertia, and the fluid would begin to move. In the cochlea, the perilymphatic fluid in the scala vestibuli has a larger mass than in the scala tympani. Based on this theory, bone-conducted vibration causes acceleration of the cochlear bone influencing fluid inertia. Other factors involved in the inertial mechanism of inner ear fluid may include: the compliance of the round and oval windows, the compliance of the bone and fluid adjacent to the round and oval windows, the viscosity of the fluid, the resistance of the fluid flow ducts, and the compliance of other bone and structures, such as blood vessels (Stenfelt, Hato, & Goode, 2004).

Ranke referred to the cumulative “third window” as all compliant structures in the cochlea other than the round or oval window (Tonndorf, 1962). Stenfelt and Goode (2005a) provided support for the presence of a third window. They reported that when an intense vibratory force was delivered to the cochlea, less than 1,000,000th of the total cochlear fluid volume was displaced at the round window as it flexes. Therefore, they predicted that if one of the windows is obstructed, another compliant structure in the cochlea would likely account for this small displacement. Tonndorf (1962) found a difference in cochlear microphonics after the cochlear aqueduct was closed, but only a slight difference was observed when the cochlear windows themselves were closed. Additionally, Stenfelt, Hato and Goode (2004) compared volumes of fluid displacement to air- and bone-conduction stimuli and found a frequency-dependent difference in volume between windows for bone-conduction stimuli only. They concluded that the difference in window displacement across frequency was evidence for the existence of a third window (Stenfelt & Goode, 2005a). In conclusion, Stenfelt and Goode (2005a) described inertial fluid movement as a significant contributor to bone-conduction hearing, particularly for frequencies below 1000 Hz.

1.2.2 Middle ear

1.2.2.1 Theory of middle ear ossicle inertia

The theory of middle ear ossicle inertia was first investigated and described by Barany (1938b). The middle ear ossicles are connected to the temporal bone through multiple loose ligament connections. Other connections from the ossicles include the tympanic membrane (TM), muscle tendons, and the oval window. Each ossicular bone also connects to another in the chain. The inertia of the ossicular chain can be described in two ways: through relative motions between the temporal bone and the ossicular chain as a whole, and through relative motions between the individual ossicles. Barany (1938b) described the theory of ossicular inertia as the chain's movement in relation to the bony walls of the middle ear. For a low frequency stimulus, when the mode of vibration is strictly translational (i.e., the movement of the bony wall is perpendicular to rotational axis of the ossicles), the ossicles should act as a pendulum and move in phase with the vibration of the temporal bone. In a more recent study, Stenfelt, Hato and Goode (2002) controlled the direction of oscillation and found little difference when the mode of vibration was altered, thereby contradicting Barany's theory that the line of stimulation must be perpendicular to the axis of rotation.

Research has also investigated frequency specific responses for middle ear ossicle inertia. Kirikae (1959c) conducted an experiment measuring the cochlear microphonics in a cat. He found that when mass-loading the TM, an increase in sensitivity to bone-conducted stimuli was revealed for low frequencies. He attributed this finding to the increased inertial response between the temporal bone and ossicular chain for low frequencies only. However, based on various manipulations of the membranes, windows, and ossicles of the middle ear in cats, Brinkman (1965) claimed that middle ear ossicular inertia is an important contributions to bone-conduction hearing for the whole range of frequencies from 250 – 8000 Hz. Huizing (1960b) found an increase in bone-conduction responses in frequencies up to 2500 Hz while mass-loading the TM. He predicted that for low frequencies, the skull moves as a whole unit in a translational motion; however, for high frequencies, the system is much more complex, and the vibratory motion of the skull is divided into individual segments.

A more recent study investigated the motion of the ossicles. Using a technique called laser Doppler vibrometry, Stenfelt, Hato and Goode (2002) were able to make fine-detailed measurements in temporal bone specimens. They looked at the differential reference velocity of the umbo of the malleus and the footplate of the stapes in reference to the promontory of the cochlea, and could determine the relative motion of these structures to controlled bone-conducted stimuli. The ossicles moved at a relative velocity and phase similar to the temporal bone for low frequencies up to 1000 Hz. Between 1000 and 2000 Hz, the umbo and footplate reached their resonant frequency. The resonant frequency of the umbo of the malleus was at 1800 Hz, while the resonant frequency of the footplate was at 1500 Hz. Alterations were made to the specimens, such as mass-loading the TM, gluing the malleus or the stapes, and breaking the joint between the incus and the stapes. Mass-loading the TM resulted in a lower resonant frequency of the entire ossicular chain. The differential relative velocity increased in the low frequencies for both ossicles. This result is similar to other studies with mass-loaded TMs. No differences were observed in relative motion above 2000 Hz. After gluing the malleus or stapes to the temporal bone, the resonant frequency for the non-glued structure increased to a higher frequency; however, the relative velocity was attenuated at the new resonant frequency. Finally, no differences were recorded when the stapes was dislocated from the incus. Based on these more recent findings, the middle ear ossicular inertia appeared not to influence bone-conduction hearing for frequencies below 1000 Hz (Stenfelt, Hato, & Goode, 2002).

Stenfelt and Goode (2005a) concluded that ossicular inertia contributes to some degree for bone-conduction hearing in the low and middle frequencies; however, it is not likely a significant contributing factor.

1.2.3 Outer ear

1.2.3.1 Osseotympanic theory

Tonndorf, Greenfield and Kaufman (1966) claimed that the osseotympanic theory of bone-conduction was originally developed by Bezold in 1885. The basis of the theory is that sound is radiated outward from the walls of the external auditory canal and transmitted into the system via the air conduction.

Studying the sound pressure in the ear canal is a technique often used to investigate mechanisms of bone conduction. For instance, Berthold attempted to theoretically explain the outcomes of the tuning fork tests by measuring the sound pressure in the ear canal and in the tympanic cavity (Bárány, 1938a). In addition, Mach hypothesized that as vibrating matter approached a steady state, the energy transfer into a system has an opposite energy transfer and bone-conduction hearing increased when outward vibration was hindered through some disease or lesion. Therefore, conductive losses could be measured though the sound pressure in the ear canal during bone-conduction stimulation (Bárány, 1938a). This speculation was disproved in the 1950’s; however, researchers continued to theorize on the origin of sound pressure in the ear canal during bone-conduction stimulation (Huizing, 1960a).

Bekesy was the first to fully describe the osseotympanic theory of bone conduction as sound radiating from the walls of the ear canal. He also believed that the motion of the jaw relative to the skull could carry vibrations through the temporal-mandibular joint to the external ear canal (Puria & Rosowski, 2012). Tonndorf, Greenfield and Kaufman (1966), through their work with cochlear microphonics in cats, supported Bekesy's hypothesis of sound radiation from the ear canal. Stenfelt, Wild, Hato and Goode (2003) found that removal of the jaw did not significantly change the sound pressure measured in the ear canal, contradicting Bekesy’s early theory that jaw movement transmitted vibrations.

Huizing (1960a) conducted a study on the sound pressure in the ear canal with alterations of the TM. His first conclusion was that the sound pressure measured in the ear canal did not always correspond to bone-conduction sensitivity. For example, when the tympanic cavity or tympanic membrane was increased in either stiffness or mass, no change was recorded in sound pressure, yet an increase in threshold was measured for air-conducted sounds. Bone-conduction thresholds increased slightly for an increase in stiffness, but decreased with an increase in mass. Other studies using a computer simulation (Egolf, Feth, Cooper, & Franks, 1985) and an ear simulator (Gilman & Dirks, 1986), found that when increasing the impedance of the tympanic membrane, the sound pressure in the ear canal increased. Additionally, Martin, Westwood and Bamford (1996) found higher mean real-ear-to-coupler difference (RECD) values for children who had otitis media with effusion compared to children with no known middle ear pathology, particularly for low and mid frequencies. Tsai, Ostroff, Korman and Chen (2005) investigated the sound pressure in the occluded and unoccluded ear canal to bone-conducted stimuli for individuals with and without otosclerosis. They found that when the ear canal was occluded, the sound pressure increased more for those with otosclerosis. An increase in impedance caused by the stiffening of the stapes may have forced the bone-conduced sound back to the ear canal. These studies indicate that sound pressure in the ear canal is directly affected by the status of the middle ear, and that measures of sound pressure in the ear cannot predict the amount of bone-conducted sound reaching the cochlea.

Huizing (1960a) investigated the frequencies where sound pressure could be measured in an unoccluded ear. For frequencies below 1200 Hz, the sound pressure in the ear canal was measurable at 30-40 dB for an 80 dB SPL stimulus. However, above 1200 Hz, the sound pressure quickly decreased until 2500 Hz after which it could not be recorded. Khanna, Tonndorf and Queller (1976), through subjective cancellation experiments, reported that for an unoccluded ear, combined inner- and middle-ear mechanisms dominated bone-conduction hearing for high frequencies (above 900 Hz). For low frequencies (700 Hz or less), the outer-ear component dominated bone-conduction hearing. At the point of subjective cancellation of air- and bone-conducted sounds, they found similar changes in sound pressure measured across frequency for the bone-conducted stimuli alone and for both the air- and bone-conducted stimuli combined. They proposed that sound pressure in the ear canal can estimate the input of the bone-conducted stimulus into the skull, even if it is not a measure of hearing threshold.

A more recent experiment by Stenfelt, Wild, Hato and Goode (2003) investigated the osseotympanic mechanism of bone-conduction. They measured the sound pressure in the ear canal as well as the motion velocity of the umbo of the malleus using laser Doppler vibrometry. When they removed the cartilaginous portion of the canal wall, they found that sound pressure measured in the ear canal decreased by approximately 5-10 dB for all frequencies below 1000 Hz and 10-15 dB for frequencies between 1000 and 4000 Hz.

1.2.3.2 Occlusion effect

Bekesy theorized that when the ear canal was occluded, sound pressure trapped in the ear canal would lead to an increase in bone-conduction hearing. Bekesy also demonstrated that when occluding the ear by placing an ear plug past the cartilaginous portion of the ear canal, the increase in bone-conduction hearing was diminished (Tonndorf, Greenfield, & Kaufman, 1966). Tonndorf (1966) further described the occlusion effect. He outlined how the ear canal is analogous to a high-pass filter. When the ear is occluded, the filter is disrupted, and low frequencies become trapped in the external ear space.

In a more recent study, Stenfelt, Wild, Hato and Goode (2003) measured sound pressure in the ear canal and velocity of the umbo of the malleus, and found that occluding an intact ear canal resulted in a 10-20 dB increase in sound pressure for frequencies under 1000

Hz. After removing the soft-tissue of the ear canal, the occlusion effect was reduced to 5-10 dB. With the removal of the soft-tissue and TM, the occlusion effect diminished. Based on their results, the authors concluded that a significant portion of sound radiation into the ear canal originates from the cartilaginous portion of the ear canal. Finally, by comparing the sound pressure in the ear canal to the velocity measured at the umbo of the malleus, the authors determined that the contribution of the sound pressure in the unoccluded ear canal is 10 dB less than the contribution of the middle ear ossicle inertia. However, when the ear canal is occluded the sound pressure in the ear canal plays a significant part in hearing through bone conduction for low frequencies (between 400 and 1200 Hz), even though the 10-15 dB increase in sound pressure measured in the ear canal due to occlusion was higher than the 5-10 dB increase in umbo velocity.

1.2.4 Soft-tissue conduction

Research has shown that changing the pressure of cerebral spinal fluid (CSF) can change the pressure of the inner ear (e.g., Beentjes, 1972). Yoshida and Uemura (1991) investigated the effects of a CSF pressure change on basilar membrane activation in guinea pigs by measuring the cochlear microphonics. They found that pressure changes in the endolymph and perilymph of the inner ear correlated to the pressure change of the CSF in a linear manner with minimal time lag between the increase of CSF pressure and increase of inner ear pressure. Moreover, cochlear microphonics from air-conducted stimuli were suppressed after the increase of CSF pressure. The authors hypothesized a possible change in hearing threshold with CSF movement; however, they predicted that CSF contribution to threshold was only 10 dB or less.

Subsequent studies further investigated this mechanism and supported that the fluid transmission of CSF to the inner ear accounts for a substantial portion of bone-conduction hearing (Freeman, Sichel, & Sohmer, 2000; Sohmer, Freeman, Geal-Dor, Adelman, & Savion, 2000). Results of a series of research studies on animals and humans found no significant threshold differences with the bone-conduction transducer positioned directly on the exposed brain of a craniotomy compared to the transducer on the mastoid bone. Similarly, no differences were found between standard bone-conduction thresholds and thresholds when the transducer was positioned on the eye of the adult subjects or on the fontanelle of neonates (Freeman, Sichel, & Sohmer, 2000; Sohmer, Freeman, Geal-Dor, Adelman, & Savion, 2000). Bone-conduction thresholds were also better when the transducer was held to the thinnest part of temporal bone compared to the mastoid. When the transducer was placed on the fontanelle, the vibration measured on the temporal bone was 14 dB smaller compared to when the transducer was on the bone at a comparable distance. Therefore, when the fluid is set into motion, little to no vibration is transmitted back from fluid to bone. These results supported the theory that fluid vibratory transmission is a key component to bone-conduction hearing (Sohmer, Freeman, Geal-Dor, Adelman, & Savion, 2000).

A follow-up study measured the ABR to a bone-conducted stimulus in rodents when a saline tube was connected between a craniotomy of two animals. The response was measured in one animal while bone-conducted stimuli were delivered to the other. The discovery that bone-conduction responses could be measured in another animal through the transfer of fluid also supported the theory of a non-osseous bone-conduction pathway. Because the title "non-osseous bone-conduction" is an oxymoron, it was changed to "soft-tissue conduction" (Perez, Adelman, & Sohmer, 2011).

Stenfelt, Hato and Goode (2004) believed that fluid flow into the cochlea is frequency specific, and at lower frequencies, the fluid flow is likely higher due to the lower impedance of the cochlear and vestibular aqueducts, while at higher frequencies, the impedance of the aqueducts is high. It is likely that fluid transmission does play some role in bone-conduction hearing, in combination with the osseous mechanisms; however, the specific contributions of each mechanism are not understood. Researchers today agree that bone-conduction hearing cannot be explained simply by one mechanism, but instead, is a complex frequency-dependant combination of mechanisms (Stenfelt & Goode, 2005a; Tonndorf, 1968).

1.3 Maturation of the Auditory System in Relation to Bone-Conduction Hearing

Candidacy criteria for the BAHS now include infants and young children; however, only a small number of studies have investigated the maturation of bone-conduction hearing sensitivity, and even fewer have examined an explanation for the infant-adult differences in bone-conduction sensitivity. Studies have shown that bone-conduction sensitivity is both frequency and age dependent and has been measured across ages using both behavioural and physiological measures (e.g., Hulecki & Small, 2011; Small & Stapells, 2008a). As mentioned, once the sound activates the basilar membrane, the auditory nerve is fired in the same manner for both air- and bone-conduction. Therefore, when considering sensitivity differences for bone-conduction hearing, maturation of air-conduction hearing must also be considered. In addition, studies have outlined some structural and functional changes that occur throughout the development of the auditory system. The following sections will first discuss the maturation of bone-conduction sensitivity, and then examine studies on the maturation of the auditory system with an attempt to explain the infant-adult differences in bone-conduction hearing.

1.3.1 Maturation of bone-conduction hearing

Visual reinforcement audiometry (VRA) is a technique that uses operant conditioning to measure the minimum response level (MRL) to auditory stimuli for infants as young as 6 months of age. The clinician conditions the infant to produce a head turn toward a visual reward after the perception of a sound. Only a few studies have measured bone-conduction MRLs in infants. Gravel (1989) was the first to assemble bone-conduction data for infants. She reported no findings for differences in MRL across frequency for normal hearing infants. Recently Hulecki and Small (2011) measured bone-conduction MRLs for young (7-15 month) and old (18-30 month) infants. MRLs were not significantly different between older or younger infants; however, frequency specific effects were found for infants in both age groups. Specifically, MRLs at 500 and 1000 Hz were better than MRLs at 2000 and 4000 Hz. Means for 500 and 1000 Hz were 9.0 and 9.8 dB HL for the young infants and 10.4 and 8.6 dB HL for the older infants, while means for 2000 and 4000 Hz were 14.3 and 14.0 dB HL for the young infants and 13.0 and 16.5 dB HL for the old infants. Casey (2012) identified similar behavioural trends across frequency. A significant effect of frequency was found for bone-conduction thresholds, while no significant effect was found for air-conduction. Results from the small collection of behavioural studies correspond well with trends across frequencies found in studies measuring physiological thresholds to BC stimuli.

Auditory steady-state responses (ASSRs) and ABRs are two physiological measures that are commonly used for threshold estimation in both research and clinical settings. Studies using frequency-specific physiological measures have found frequency-dependent trends between infants and adults (Casey, 2012; Foxe & Stapells, 1993; Nousak & Stapells, 1992; Small & Stapells, 2008a; Small & Stapells, 2006; Stapells & Ruben, 1989; D. W. Swanepoel, Ebrahim, Friedland, Swanepoel, & Pottas, 2008; Vander Werff, Prieve, & Georgantas, 2009). For example, Foxe and Stapells (1993) investigated infant and adult ABRs elicited to 500 and 2000 Hz tone-bursts and found that the mean wave V latency elicited to 500 Hz stimuli was significantly shorter for infants (2 weeks to 13 months, with the majority under 6 months) than for adults. Therefore, the authors suggested that the 500-Hz stimulus is more effective in infants than adults. Although group mean differences were not significant, trends indicated that infant thresholds were similar to those of adults at 500 Hz and thresholds were 5.5 dB worse compared to adults at 2000 Hz. They also found that bone-conduction thresholds for infants were better at 500 than 2000 Hz by 10.5 dB (Foxe & Stapells, 1993).

More recently, Small and Stapells (2008a; 2006) looked at bone-conduction thresholds using frequency-specific ASSRs. These studies were the first to identify maturational changes in bone-conduction responses for ASSRs across a range of frequencies. They found similar results to Foxe and Stapells (1993) for trends across frequency for young infants (0-11 months). For older infants (12-24 months), there were smaller differences in threshold across frequency; however, thresholds were still better at 1000 compared to 2000 Hz. Throughout maturation, the general trend is that bone-conduction thresholds tend to worsen in the low frequencies (500 and 1000 Hz), improve at 2000 Hz, and remain similar at 4000 Hz (Small & Stapells, 2008a). Small and Stapells (2008a) predicted that bone-conduction sensitivity changes with maturation until at least 2 years of age. One important consideration is that infant-adult differences in bone-conduction sensitivity are more prominent for physiological compared to behavioural measures (Hulecki & Small, 2011; Small & Stapells, 2008a; Small & Stapells, 2006).

In contrast to research using bone-conduction stimuli, studies comparing infant and adult threshold differences to air-conduction stimuli have shown that infants are less sensitive than adults using both behavioural (e.g., Parry, Hacking, Bamford, & Day, 2003; Vander Werff, Prieve, & Georgantas, 2009) and physiological techniques for threshold estimation (e.g., Rance & Tomlin, 2006; Sininger & Abdala, 1996; Van Maanen & Stapells, 2009). Sininger, Abdala and Cone-Wesson (1997) measured threshold using click and tone-burst ABR and calibrated the stimuli in the ear canal to control for ear canal size and properties. The authors still found significantly higher thresholds for newborn infants compared to adults. Furthermore, Casey (2012) investigated the maturational air-bone gap for infants and found a significant air-bone gap at 500 and 1000 Hz using ASSR, but not for 2000 or 4000 Hz. No significant air-bone gap was found for adults. Explanations are needed to account for the frequency-specific sensitivity differences between infants and adults for bone-conduction measurements.

1.3.2 Maturation of the auditory system

Physiological measures of hearing provide evidence that the auditory system undergoes significant postnatal development. In order to attempt to explain maturation of the ABR and ASSR for bone-conduction stimuli, an understanding of the maturation of the auditory system is required. The following sections will outline the changes in each structure of the auditory system: the brainstem, cochlea, middle ear, and outer ear. In addition, the maturation of the skull, skin and subcutaneous tissue will also be discussed due to their relevance to the transmission of bone-conducted sounds.

1.3.2.1 Brainstem

Studies using physiological measures have postulated that maturation of the brainstem contributes substantially to infant-adult differences in the ABR or ASSR. Changes occur throughout development in the central nervous system, including the auditory pathway. For example, adult neurons responsible for auditory input in the brainstem are 50-60% larger in size than newborn auditory neurons (Moore & Linthicum, 2007) Neuronal connections are responsible for signal transmission. By interpreting responses of the ABR and modeling the auditory pathway, researchers were able to predict components responsible for the maturation of the ABR. Multiple studies originally postulated a correlation between an increase in myelin density and decrease in ABR wave V latency throughout first postnatal year (Jiang, Zheng, Sun, & Liu, 1991; Rotteveel, de Graaf, Colon, Stegeman, & Visco, 1987; Salamy, 1984). However, by interpreting ABR peaks individually, Moore, Ponton, Eggermont, Wu and Huang (1996) found that conduction time across individual axons is mature at birth. The authors predicted that as the length of the auditory pathway increases, a similar increase in conduction velocity is responsible for a stable conduction time throughout postnatal maturation. Ponton, Moore and Eggermont (1996) investigated the conduction time for the ABR sequence for infants and adults when a synapse was involved. The authors found that when the ABR peak latency differences contain a synaptic junction, the latency remains immature until 1-2 years of age, while the asynaptic intervals are mature at birth.

Brainstem maturation has been reported to mature substantially within the first year and continue throughout the first few years of life; however, higher-order cortical pathways in the auditory system continue to mature well into adolescence (Moore & Linthicum, 2007).

1.3.2.2 Cochlea

It is generally understood that the cochlea is structurally mature at the time of birth. Specifically, the newborn cochlea is adult size and shape, and hair cell innervations are complete (Lavigne-Rebillard & Pujol, 1988). There are some indications that the ossification of the otic capsule bone continues after birth. More specifically, an onset of ossification occurs around the time of birth in the endochondral layer of the bone. Up until about age 3 years, the three layers of otic capsule (endosteal, periosteal, and endochondral) are continuously fusing to create one indistinct layer of bone (Eby & Nadol, 1986).

Studies have used distortion-product otoacoutic emissions (DPOAEs) to investigate the functional maturation of the cochlea more extensively (Abdala & Sininger, 1996; A. M. Brown, Sheppard, & Russell, 1994; Eggermont, Brown, Ponton, & Kimberley, 1996). Using this technique, researchers have been able to quantify the presence and functionality of the cochlear amplifier, independent of the subsequent neuronal connections up the auditory nerve (Abdala & Sininger, 1996; Abdala, Sininger, Ekelid, & Zeng, 1996; Eggermont, Brown, Ponton, & Kimberley, 1996). The overall conclusion drawn from these studies is that the cochlea is adult-like at birth, including features such as frequency resolution and the cochlear amplifier (Abdala & Sininger, 1996; Abdala, Sininger, Ekelid, & Zeng, 1996; Abdala, 2001;

Eggermont, Brown, Ponton, & Kimberley, 1996). Therefore, immaturities in the cochlea are unlikely to contribute to infant-adult differences in bone-conduction sensitivity.

1.3.2.3 Middle ear

Similar to the cochlea, maturation of the middle ear occurs primarily during prenatal development. The ossicles in the middle ear have reached adult size and shape at birth and are fully ossified. Changes to the ossicles that may occur after birth include pneumatization and remodeling of the bone (Eby & Nadol, 1986), internal bone erosion of the stapes footplate (Saunders, Kaltenbach, & Relkin, 1983), and release of residual mesenchyme clinging to the ossicles after birth (Saunders, Kaltenbach, & Relkin, 1983; Takahara, Sando, Hashida, & Shibahara, 1986; Takahara & Sando, 1987). Mesenchyme is tissue from the mesoderm that differentiates to form structures during development. Mesenchyme, or a derivative of mesenchyme, may be present in the middle ear cavity until 5 months of age before it clears out (Eby & Nadol, 1986). Studies using multifrequency tympanometry have investigated changes across age with high-frequency stimuli, representing changes in mass systems in the middle ear (Calandruccio, Fitzgerald, & Prieve, 2006; Holte, Margolish, & Cavanaugh, 1991; Keefe, Bulen, Arehart, & Burns, 1993; Meyer, Jardine, & Deverson, 1997; Sanford & Feeney, 2008). The increase in compliance at high frequencies from birth to adulthood may be attributed to slight ossicular restructuring after birth. Although the ossicles are physically adult-like at birth, subtle changes may occur in and around the bones during postnatal development.

A salient change that occurs after birth is the growth and adjustment of the tympanic cavity. The middle-ear cavity has been shown to enlarge as a whole, in addition to specific growth of the antral and mastoid sinuses (Anson & Donaldson, 1981; Saunders, Kaltenbach, & Relkin, 1983). Mastoid growth and pneumatization will be discussed further in the section on temporal bone development. Ikui, Sando, Haginomori and Sudo (2000) investigated the growth of the tympanic cavity using a three-dimensional computer reconstruction modeling. They found the overall volume enlarges to 1.5 times the size of a newborn tympanic cavity, and that increase in height of the tympanic cavity contributed the most to this increase in volume. The tympanic cavity is divided into 3 parts: the epitympanum, which connects directly to the mastoid, the mesotympanum, defined by the boundaries of the tympanic membrane, and the hypotympanum, the most inferior portion of the cavity adjacent to the inferior temporal bone, mandible and neck muscles. Results of their study showed the greatest growth was in the epitympanum and hypotympanum. It was not surprising that they found that the mesotympanum showed little signs of growth postnatally, given that the tympanic membrane is adult size at birth (Saunders, Kaltenbach, & Relkin, 1983).

The tympanic ring, also called the tympanic annulus or ectotympanic ring, is a ring of bone and fibre that partially encircles the tympanic membrane. Although the tympanic membrane does not grow postnatally, the tympanic ring changes significantly postnatally. After birth, the bony portion of the ring continues to fuse, and extends laterally. The tympanic ring develops into the bony portion of the external auditory meatus. The bony ring also continues to increase in thickness until adolescence (Saunders, Kaltenbach, & Relkin, 1983). Ikui, Sando, Sudo and Fujita (1997) mapped the angle of the tympanic membrane and annulus in comparison to the oval window and internal auditory meatus. They found a significant change in the plane of the tympanic annulus from birth to adulthood. More specifically, newborns exhibited tympanic annuli in a more horizontal plane in comparison with the vertical plane exhibited in adult temporal bone specimens. This was considered to be the most significant postnatal change in the middle ear anatomy, and is projected to settle into adult form by as late as 4-5 years of age (Eby & Nadol, 1986). An adjustment in tympanic annulus orientation produces a change in the orientation of the developed ossicles, a change in shape of the developed tympanic membrane, and one potential explanation for the increase in compliance with age for high frequencies (Calandruccio, Fitzgerald, & Prieve, 2006; Holte, Margolish, & Cavanaugh, 1991; Keefe, Bulen, Arehart, & Burns, 1993; Sanford & Feeney, 2008; Saunders, Kaltenbach, & Relkin, 1983).

Sanford and Feeney (2008) also found that energy reflectance of the middle ear for infants was more sensitive to changes in static pressure at high frequencies, so that the push and pull of the tympanic membrane on the ossicles due to positive or negative pressure change creates a disarticulation or a hyperarticulation of the ossicles, creating more or less energy reflectance, respectively. The authors attributed this finding to the prediction that infant ossicles are not as firmly joined, and therefore are more susceptible to pressure change in both directions.

Nousak and Stapells (1992) commented that changes in ossicular structure, growth of the tympanic cavity, development of the tympanic ring, and presence of mesenchyme in the middle ear cavity could all contribute to the infant-adult differences in bone-conduction sensitivity. Stenfelt, Hato, and Goode (2002) claimed that middle ear ossicular inertia likely contributes to mid-frequency bone-conduction hearing, and therefore, these significant changes to the middle ear within the first few years of life could provide some explanation to infant-adult bone-conduction sensitivity differences.

Significant change in the middle ear structures occurs throughout postnatal development, and studies on maturation of tympanometric functions can provide valuable insight to how these changes affect frequency-specific sound transmission through the middle ear to bone-conducted stimuli. However, Stapells and Ruben (1989) reported that physiological bone-conduction thresholds for infants who present with middle ear fluid had similar thresholds to those without middle ear fluid and consequently, middle-ear factors cannot completely explain the maturation of bone-conduction hearing.

1.3.2.4 Outer ear

The structure and function of the external auditory meatus, or the ear canal, develops up to about age 12 years (Abdala & Keefe, 2012). Keefe, Bulen, Campbell, and Burns (1994) estimated ear canal lengths to be 14.0, 16.5, 17.5, 20.0, and 21.0 mm for 1, 3, 6, 12 and 24 months, respectively. Bagatto, Seewald, Scollie and Tharpe (2006) estimated ear canal lengths of 16.5 and 17.5 mm for 3 and 6 month old infants, respectively. In addition to smaller ear canal length, the diameter of the ear canal was also reported to be narrower in infants than adults. Infant ear canals also have a more oval-shaped cross section (Saunders, Kaltenbach, & Relkin, 1983). Keefe, Bulen, Campbell and Burns (1994) described ear canal diameter as increasing from 4.4 mm to 7.7 mm for infants 1 and 24 months, respectively.

In addition, ear canal walls undergo restructuring throughout early development. Newborn ear canal walls are primarily made up of cartilage. As described previously, the tympanic ring extends laterally during postnatal development up to at least 1 year of age (Anson & Donaldson, 1981), and the lateral extensions of the tympanic ring may continue to grow until adolescence (Saunders, Kaltenbach, & Relkin, 1983). These lateral extensions of the tympanic ring eventually become the bony medial walls of the external auditory meatus (Anson & Donaldson, 1981; Saunders, Kaltenbach, & Relkin, 1983).

Studies investigating properties of the ear canal have primarily focused on establishing protocols for probe-tube measurements for fitting hearing aids (Bagatto, Seewald, Scollie, & Tharpe, 2006; Feigin, Kopun, Stelmachowicz, & Gorga, 1989), investigating the resonant properties of ear canals (Bentler, 1989; Kruger, 1987; Kruger & Ruben, 1987), and developing an understanding of tympanometric results for low-frequency stimuli (Holte, Margolish, & Cavanaugh, 1991; Keefe, Bulen, Arehart, & Burns, 1993; Keefe, Bulen, Campbell, & Burns, 1994; Sanford & Feeney, 2008). Studies investigating resonant frequencies agree that ear-canal resonance decreases with increasing age. Newborn ear-canal resonance values are 5000 or 6000 Hz and decrease to the adult value of approximately 2700 Hz by about 6-12 months of age (Kruger, 1987; Kruger & Ruben, 1987). The absorptive nature of the cartilaginous canal walls of the infant decrease the acoustic energy transferred into the middle ear. As a consequence, ear canals walls are set into vibration and the energy is dissipated, leading to a higher absolute impedance and lower admittance magnitudes for infants compared to adults (Holte, Margolish, & Cavanaugh, 1991; Keefe, Bulen, Arehart, & Burns, 1993; Sanford & Feeney, 2008). Sanford and Feeney (2008) measured a 30% change in mean admittance and energy reflectance for frequencies between 750 and 2000 Hz between 4 and 24 weeks of age. Keefe, Bulen, Arehart, and Burns (1993) attributed this difference in admittance magnitude primarily to changes in external ear canal properties.

Insights from these projects provide valuable information for the maturation of bone-conduction hearing. For instance, recent research has looked at the occlusion effect for young (0-7 months) and older infants (10-22 months) and adults, and found that sound pressure measured at the eardrum to bone-conduction stimuli was larger for infants compared to adults for both unoccluded and occluded ears. They also found a greater increase in sound pressure at the eardrum for low frequencies when occluding the ear, likely due to infants having smaller ear canals. A higher proportion of cartilage in the ear canal also may contribute to the increase in low frequency sound pressure after occlusion (Stenfelt, Wild, Hato, & Goode, 2003). However, there were no differences in young infant ASSR thresholds when the ear was occluded compared to when unoccluded (Small, Hatton, & Stapells, 2007; Small & Hu, 2011). These differences were reported to emerge in older infants (Small & Hu, 2011). Therefore, although the sound pressure difference is observed for low-frequency bone-conduction tones, it does not likely contribute to the better hearing sensitivity for these sounds. These results, in conjunction with wideband tympanometry studies (e.g., Keefe, Bulen, Arehart, & Burns, 1993), suggested that the cartilage in the ear canal may absorb vibratory energy from the acoustic stimuli. Even though the sound pressure in the ear canal was higher for young infants, it cannot be assumed that this energy is transferred through the auditory pathway (e.g., Keefe, Bulen, Arehart, & Burns, 1993). Even in adults, the increase in sound pressure measured at the ear drum due to occlusion did not correspond directly to a decrease in thresholds for low-frequency stimuli (Stenfelt, Wild, Hato, & Goode, 2003; Stenfelt & Reinfeldt, 2007).

1.3.2.5 Skull

The most likely contributor to better low-frequency bone-conduction hearing sensitivity for infants is the difference in size and structure of the skull. The skull continues to grow until the second or third decade of life (Steele & Bramblett, 1988). Some researchers postulate that one contributing factor to infant-adult differences in bone-conduction sensitivity is the overall size of the cochlea compared to the overall size of the head. The infant cochlea is adult-size, but the infant skull and temporal bone is substantially smaller and more delicate than that of an adult. Therefore, energy may take a more direct path to the cochlea for infants, while energy is more easily dissipated in adult skulls (Foxe & Stapells, 1993; Nousak & Stapells, 1992).

Notably, the skull of infants is comprised of individual bones that have not fully fused together. Throughout development, sutures are created when bones begin to join together. Sutures develop at different rates up to young adulthood when the skull is considered mature (Steele & Bramblett, 1988). Fontenelles are the soft-tissue gaps between portions of unfused bone. The presence of fontenelles at birth allow for the cranium of the newborn infant to compress during birth. In addition, fontenelles and immature sutures allow for the rapid growth of the infant brain. The average head circumference of the newborn infant is 33-35 cm, and the average head circumference for a 1 year old infant is 46 cm (G. H. Sperber, Guttman, & Sperber, 2001; World Health Organization, 2007). Sutures contribute most to the growth of the skull until age 4 years (G. H. Sperber, Guttman, & Sperber, 2001). During this time, sutures are not ossified, yet allow for new bone growth to occur at the edge of each bone segment. The brain-growth process stimulates ossification, thus expanding the cranial bone (Opperman, 2000). After age 4 years, a process called surface apposition and resorption contributes to growth. This includes resorption of the bone and flattening of the curvature of some bones to allow for the brain to grow further (G. H. Sperber, Guttman, & Sperber, 2001). The fontanelles include the anterior fontanelle at the crown of the head (commonly known as the “soft-spot”), the posterior fontanelle, two posterolateral fontanelles and two anterolateral fontanelles.

Some studies have hypothesized that immature sutures may enclose vibratory energy within the temporal bone, thereby increasing the effective stimulation to the ipsilateral cochlea (Foxe & Stapells, 1993; Small & Stapells, 2008a; Small & Stapells, 2008b; Stapells & Ruben, 1989; Stuart, Yang, & Stenstrom, 1990; Stuart, Yang, & Botea, 1996). A study by Sohmer Freeman, Geal-Dor, Adelman and Savion (2000) investigated the acceleration across the fontanelle of an infant’s skull for a click stimulus. They found that 14 dB of vibration was lost from the fontanelle to the temporal bone, suggesting that vibratory energy may be dissipated when it travels across soft-tissue. The loss of energy across the fontanelle may be one contributing factor to the higher transcranial attenuation reported for infants compared to adults.

The measurement of transcranial attenuation to bone-conduction vibration is an important consideration for teasing out mechanisms contributing to infant-adult differences in bone-conduction sensitivity. The transcranial attenuation for adults has substantial intersubject variability, but is conservatively considered to be 0 dB for low frequencies and 15 dB for 4000 Hz (Studebaker, 1967). Similar frequency-specific trends were found for children ages 7-16 years with severe-profound unilateral sensorineural hearing loss using unmasked thresholds with contralateral mastoid bone-conduction stimulation. Attenuation for 500, 1000, 2000, and 4000 Hz was 6.2, 7.2, 11.0, and 16.9 dB, respectively (Vanniasegaram, Bradley, & Bellman, 1994).

Transcranial attenuation for infants was indirectly estimated by comparing ipsilateral and contralateral channels during physiological measures of hearing. Measurements for ABR studies included wave V amplitude and latency. Studies found that infants have longer wave V latency and lower amplitude measures for recordings made in the contralateral channel compared to the ipsilateral channel (Foxe & Stapells, 1993; Picton, Durieux-Smith, & Moran, 1994; Stapells & Ruben, 1989; Stuart, Yang, & Botea, 1996; Yang, Rupert, & Moushegian, 1987). Small and Stapells (2008b) compared ipsilateral and contralateral channels using measurements of ASSR amplitude, phase delay, and ASSR threshold. By combining amplitude and phase delay for a measure of asymmetry, they found a 78% occurrence of asymmetry for infants, while adults only had a 44% occurrence of asymmetry using the same criterion and stimulus intensity. Infants also had significantly poorer thresholds for ASSRs recorded from the contralateral channel, while adults had no significant difference between channels. These physiological studies have estimated that transcranial attenuation of bone-conduction stimuli is higher for infants than for adults (Foxe & Stapells, 1993; Small & Stapells, 2008b; Stapells & Ruben, 1989; Yang, Rupert, & Moushegian, 1987). Based on their findings, Small and Stapells (2008b) predicted infants have at least 10-30 dB of transcranial attenuation to bone-conduction stimuli, and attenuation is much more variable for infants compared to adults.

Foxe and Stapells (1993) found frequency-specific differences when comparing ipsilateral and contralateral recording channels to bone-conduction stimuli for infants. For a 2000-Hz signal, no differences in wave V amplitude were present between ipsilateral and contralateral channels; however, for a 500-Hz stimulus, substantial reduction of wave V amplitude was measured across channels. This trend is consistent with reported frequency-dependent differences in sensitivity to bone-conducted stimuli (Small & Stapells, 2008a; Small & Stapells, 2006). However, Small and Stapells (2008b) did not find any frequency-specific differences to ASSR thresholds measured across channels.

Not only is the shape and structure anatomically different between infants and adults, but the actual make-up of the skull bone itself shows notable differences. For one, infants have thinner skull bones, and thickness increases with increasing skull circumference (Breisch, Haas, Masoumi, Chadwick, & Krous, 2010). In addition, neonate skull bones are unilaminar (composed of a single layer), in contrast to adult skull bones consisting of multiple layers, called tables. Tables are made up of a think outer layer and a thin but dense inner layer. Between the two layers lies diploë, a spongy bone tissue. The thick yet lightweight, tablular structure of adult bones creates a high stiffness/mass ratio (G. H. Sperber, Guttman, & Sperber, 2001) with large mechanical energy absorbing capabilities (Margulies & Thibault, 2000). The literature provides mixed results regarding when this adult-like structure begins to take form, possibly as early as 6 months of age (Margulies & Thibault, 2000) to as late as 4 years of age (G. H. Sperber, Guttman, & Sperber, 2001). Studies report that infant unilaminar skulls have a smaller elastic modulus compared to adult skulls (Kriewall, 1982; Margulies & Thibault, 2000). Generally, substances with a small elastic modulus are less stiff compared to substances with a high elastic modulus (Margulies & Thibault, 2000). O'Brian and Liu (2005) measured the speed of sound through bone matter and found that the compact tables of the skull bone transmitted sound at a rate of 2600-3100 m/s, while the spongy diploë matter present in adult skulls transmitted sound at a slower rate of 2200-2500 m/s.

In addition, Margulies and Thibault (2000) investigated mechanical properties of infant sutures with regards to pediatric head trauma. Results showed that immature sutures are made up of a membrane that joins plates of bone, and when loaded with a force, the membrane easily separates from the adjacent bone. These results are in contrast to the stiffer sutures between cranial bones in the adult skull connected by fibres containing mechanical energy-absorbing collagen. The periosteal and endosteal membranes in infant sutures demonstrate a poorer ability to absorb an applied physical force in comparison to adult sutures.

A closer examination of the frontal and temporal skull bones and their surrounding sutures and fontanelles is provided in the following sections. The temporal bone and the frontal bone are the most common positions for stimulating the skull via bone conduction.

1.3.2.5.1 Temporal bone

Growth and development of the temporal bone has been studied to understand the surgical implications to cochlear implantation and BAHS titanium fixture implantation in the developing skull (Dahm, Shepherd, & Clark, 1993; Eby & Nadol, 1986; Granström, Bergström, Odersjö, & Tjellström, 2001; Simms & Neely, 1989). Overall, the temporal bone grows in length on average 9.5% per year under 30 months of age, 6.5% per year under 4 years of age, and 1.4% over 4 years of age (Simms & Neely, 1989). The thickness of the temporal bone for children under age 16 years averages to 2.5 mm (Granström, Bergström, Odersjö, & Tjellström, 2001), while the adult skull thickness is about 6 mm on average (Federspil et al., 2010). The temporal bone is made up of four parts: the squamous, petrous, tympanic and styloid bone, in addition to the mastoid, which originates from the squamous and petrous bones (Dahm, Shepherd, & Clark, 1993). The development of the mastoid process occurs postnatally and becomes prominent after the first year of life due to ongoing pneumatization and the strengthening of the sternocleidomastoid muscle. Pneumatization is considered one of the most significant changes of temporal bone development postnatally. The age at which pneumatization is complete varies across studies and is likely different between genders. One study found pneumatization finishes as early as age 6 years (Schillinger, 1939); however, it is more likely that the development of the mastoid process continues until adolescence, with the majority of change occurring within the first year of life (Dahm, Shepherd, & Clark, 1993; Eby & Nadol, 1986). The growth of the mastoid extends in three directions: lateral, anteroinferior, and superoinferior, with the most growth in the superoinferior direction (Dahm, Shepherd, & Clark, 1993; Eby & Nadol, 1986).

In addition to temporal bone growth, two fontanelles are present between the temporal bone and adjacent cranial bones. The anterolateral fontanelle lies between the squamous portion of the temporal bone and the frontal bone, and closes at approximately 3 months of age. The poterolateral fontanelle lies between the petrous component of the temporal bone and the occipital bone, and closes around 2-3 years of age (G. H. Sperber, Guttman, & Sperber, 2001).

1.3.2.5.2 Frontal bone

Although the anatomy of the temporal bone is typically the focus when explaining bone-conduction hearing, some clinical protocols allow for placing the bone-conduction transducer on the forehead of the patient. This practice has been demonstrated both for diagnostic and clinical purposes (e.g., Cochlear, 2011a; Small, Hatton, & Stapells, 2007; Whittle, 1965). Therefore, an explanation of frontal bone development is warranted. Similar to the temporal bone, early studies have found pneumatization and growth of the frontal bone continues into early adulthood to allow for continued growth of facial bones (Ford, 1958). Young (1957) found that the height of the frontal bone grows rapidly within the first 2-3 years after birth, then decelerates in growth rate. The frontal bone was reported to increase in an arch-like configuration until age 3 years, then progressively flatten until 16 years of age. In addition, frontal bones continue to increase in thickness throughout early development (Young, 1957).

The frontal bone is divided into two halves by the frontal suture running from the bregma (or the anterior fontanelle) down to the nasion. The frontal suture is typically mature and fused by age 6-8 years. Chondroid tissue, an immature form of cartilage, is present at the frontal suture and is eventually replaced by bone. The anterior fontanelle, located at the crown of the head, between the frontal and parietal bones, usually does not close until 2-3 years of age (G. H. Sperber, Guttman, & Sperber, 2001).

1.3.2.6 Skin and subcutaneous tissue

Early studies purported that the skin of a newborn infant is fully mature (W. L. Weston, Lane, & Morelli, 1996); however, newer techniques to investigate the physiology of infant skin have revealed differences in the skin structure of neonates (Stamatas, Nikolovski, Mack, & Kollias, 2011). Because bone-conduction stimuli are presented transcutaneously for diagnostic purposes across ages, as well as for amplification purposes for children under age 5 years, an understanding of the infant-adult differences in the properties of the skin and subcutaneous tissue is important when investigating the maturation of bone-conduction hearing.

Two layers make up the structure of the skin: the epidermis and the dermis. Overall, skin thickness increases as an individual grows. Skin thickness at the forehead for children age 2-3 years is 1.18 compared to 1.99 mm for adults, primarily attributed to the overall difference in thickness of the dermal layer (Seidenari, Giusti, Bertoni, Magnoni, & Pellacani, 2000). Differences in collagen and elastin fibre may contribute to the thinner dermis in infants (Stamatas, Nikolovski, Mack, & Kollias, 2011). Collagen is less dense in infants compared to adults (Vitellaro-Zuccarello, Cappelletti, Rossi, & Sari-Gorla, 2005), and continues to increase in density until age 3-5 years (Widdowson, 1968). Elastin fibres are also different in structure between infants and adults. Notably, they are smaller in size and contain less elastin matrix, a substance which defines their structural maturity. The skin of infants and young children contains the ability to bind to and retain water, which allows for an increased compressibility of the skin and increased hydration (Nikolovski, Stamatas, Kollias, & Wiegand, 2008; Seidenari, Giusti, Bertoni, Magnoni, & Pellacani, 2000; Stamatas, Nikolovski, Mack, & Kollias, 2011).

The infant epidermis and stratum corneum, the outermost of the five layers of the epidermis, are also much thinner compared to adults. For infants 6-24 months of age, the stratum corneum is 30% thinner than for adults, and the epidermis as a whole is 20% thinner for infants. In the epidermis, skin cells are much smaller in size but are much more densely packed compared to adult skin surface (Stamatas, Nikolovski, Mack, & Kollias, 2011).

The differences in infant and adult subcutaneous tissue have not been as fully researched, particularly for the head; however, research on cochlear implant surgery has investigated some trends. Lupin and Gardiner (2001) found that the thickness of the scalp (i.e., skin and underlying soft-tissue) at the temporal bone increased with increasing age. They also found that scalp thickness varied across different parts of the temporal bone, with thicker scalp on the mastoid process in comparison to the temporal bone directly anterior to the mastoid. Raine, Lee, Strachan, Totten and Khan (2007) found the median scalp thickness in children ages 2-15 years and adults was 3 and 5 mm, respectively.

To fully understand the mechanisms and maturation of bone-conduction hearing, it is necessary to have an understanding of the maturation of the auditory system, including structures of the head involved in bone-conduction. Developmental changes in anatomy and functionality of these structures can help explain the developmental changes observed in bone-conduction hearing. Additional research is required to provide a more complete explanation as to why infants show differences in their response to bone-conducted stimuli compared to adults, with respect to both transcranial attenuation and frequency-specific trends in bone-conduction sensitivity. One means for studying the underlying mechanisms for these infant-adult differences in bone-conduction hearing is to measure how bone-conduction stimuli is transmitted across the skull for adults and children of different ages.

1.4 Bone-Anchored Hearing Systems

Air- and bone-conduction thresholds are used for classifying types of hearing losses into conductive, sensorineural, or mixed. Amplification devices also use air- or bone-conduction stimulation to manage hearing loss. Most hearing aids amplify air-conducted sound. Sound is picked up from the microphone, amplified in the sound processor, and presented into the ear canal of an individual. Bone-anchored hearing systems have been designed for individuals with conductive or mixed hearing losses with a mild sensorineural component who have problems wearing air-conduction hearing aids. Chronically draining ears or craniofacial anomalies such as atresia are two examples of such problems. More discussion on candidacy is included in a subsequent section. A bone-anchored hearing system is enclosed in a rectangular or oval-shaped plastic casing. Sound is amplified from the microphone then converted into vibration, which is presented at the skull of the individual.

The nomenclature for bone-anchored hearing systems has changed over time, as bone-conduction products have evolved. The original devices that used bone-conduction stimulation for amplification were not anchored to the bone, but used a steel headband that wrapped around the top of the head to hold the device in place. These were first referred to as bone-conduction hearing aids; however, for the purpose of this thesis, they will be referred to as conventional bone-conduction hearing systems. Later, when the titanium abutment was developed, the device that was actually anchored to the skull was referred to as a bone-anchored hearing aid (BAHA). The soft band with the attached device was called BAHA Softband, even though it was not physically anchored to the bone. Recently, Cochlear trademarked the title Baha in reference to their own bone-anchored hearing systems (including the soft band version). Therefore, all devices, with the exception of the conventional bone-conduction hearing system, will be referred to as a bone-anchored hearing systems (BAHS) or devices throughout this discussion.

The second nomenclature issue to address is the differentiation between the terms "steel headband," "testband," "headband," and "soft band." These terms are often used interchangeably throughout the literature as technology changes. The "steel headband" was used originally for conventional bone-conduction hearing systems and is now used as a "testband" for short-term trials with the BAHS. A newer "headband" is also used for patient test trials before undergoing surgery for a BAHS implant and for adult long-term users who benefit from a BAHS but cannot undergo surgery for an abutment. Finally, the “soft band” refers to the fabric band that wraps around the head and is used with the BAHS, typically for the pediatric population.

1.4.1 History of the bone-anchored hearing system

1.4.1.1 Conventional bone-conduction hearing system

Bone-anchored hearing systems (BAHS) have become significantly more sophisticated over time. Early versions were essentially bone oscillators with a microphone attached to a steel headband or incorporated into eyeglasses. Analog circuitry and only limited adjustments to the sound were possible. The steel headbands were uncomfortable and because of the awkward fit on the head, the device was frequently poorly positioned on the mastoid. In addition, many patients had complaints about the cosmetically unappealing appearance of the steel headband (Håkansson, Tjellström, Rosenhall, & Carlsson, 1985).

1.4.1.2 Osseointegration

In the 1960's, Professor Per-Ingvar Brånemark, began using titanium implants for conditions of the jaw. These dental implants proved to be successful in the short and long term and effectively osseointegrated into the bone of the jaw without reaction (Adell, Lekholm, Rockler, & Brånemark, 1981). In 1977, Anders Tjellström surgically placed the first titanium implant in the temporal bone of a patient who used a conventional bone-conduction hearing system and osseointegration was achieved. Over the course of five years, no reaction was observed and the osseointergrated implant remained stable (Tjellström et al., 1983). Because of the successful outcome, the team continued to offer implantation to interested candidates.

1.4.1.3 Early sound processors

In 1982, the first BAHS (HC-200) was provided to candidates with the implant (Håkansson, Tjellström, Rosenhall, & Carlsson, 1985; Håkansson, Carlsson, Tjellström, & Liden, 1994; Snik et al., 2005). The HC-200 attached to the implant through a titanium abutment, which was screwed into the implant. The HC-200 transmitted sound percutaneously (through punctured skin), as opposed to the conventional bone-conduction systems, which transmitted sound transcutaneously (across intact skin). This original device was only slightly modified when it was released as the Classic, which was used throughout the 1990's. An initial study compared the BAHS to the conventional bone-conduction hearing system and found that patients reportedly preferred the BAHS after only two weeks of use, but they found no significant difference in speech discrimination testing (Håkansson, Tjellström, Rosenhall, & Carlsson, 1985). Subsequent studies, however, found lower aided sound-field thresholds (H. R. Cooper, Burrell, Powell, Proops, & Bickerton, 1996), better speech reception thresholds in quiet (Snik, Dreschler, Tange, & Cremers, 1998), better word discrimination (Håkansson, Carlsson, Tjellström, & Liden, 1994), and better clinical impact scores in the Hearing Handicap and Disability Index questionnaire (Hol et al., 2004) with the BAHS compared to the conventional bone-conduction hearing system.

1.4.1.4 Power sound processors

As mentioned earlier, the BAHS was developed for managing conductive hearing losses and mixed hearing losses with mild sensorineural components. Individuals with a significant sensorineural component to their hearing loss do not benefit as well with a BAHS (Snik, Mylanus, & Cremers, 1995). Functional gain is defined as the difference between unaided and aided thresholds. The important consideration for functional gain for a BAHS is the amount of gain that is available to the patient in excess of what is sufficient to overcome the gap between air- and bone-conduction thresholds (air-bone gap). The additional gain of the Classic was limited to approximately 5-10 dB. Snik, Jorritsma, Cremers, Beynon and Van den Berge (1992) introduced the super-bass BAHS as an alternative for individuals with bone-conduction thresholds of 45-65 dB HL. The super-bass device (HC220, or later, Cordelle) was worn on the body, with a cable connecting the processor to the transducer on the abutment. The additional gain for the Cordelle was 10-15 dB and 5-10 dB for the low and high frequencies, respectively (Carlsson & Håkansson, 1997).

1.4.1.5 Recent modifications

In 1999, the Food and Drug Administration in the U.S. approved the implant for use in adults and children over the age of 5 years. Soon after, the BAHS soft band was developed for young children and infants under the age of 5 years (Christensen, Smith-Olinde, Kimberlain, Richter, & Dornhoffer, 2010). The soft band is a fabric band that wraps around the child's head. A plastic disk on the soft band is positioned against the child's skull and is equipped with a coupling piece that fits into the snap on the transducer.

Ongoing product development has focused on the improvement of the cosmetic appeal of the processor and the increase in the amount of power available in a head-worn device, and the advancement of the programming capabilities of the digital devices. Additional improvements have also been made on the abutment shape and material to help limit the chance of infection around the abutment site.

1.4.2 Candidacy

The candidacy criteria for fitting a BAHS have changed with advances in research and technology. Originally, the BAHS was designed for individuals with a conductive or mixed hearing loss, with at least one of the following conditions: (1) atresia where surgery could not be performed, (2) continuously draining ear from operated atresia or chronic otitis media, or (3) severe inflammation of the ear canal where an ear mold could not be used. If the patient did not have any of these conditions, but experienced a significant occlusion effect to the point that they could not tolerate an ear mold, a BAHS was also potentially warranted.

Yoshinaga-Itano, Sedey, Coulter and Mehl (1998) introduced the implications for early intervention. In their study, they found significantly lower language scores in children where amplification was provided after 6 months of age compared to those where amplification was provided before 6 months of age. Therefore, the Joint Committee on Infant Hearing recommended that infants be fit with amplification devices before the age of 6 months (Joint Committee on Infant Hearing, 2007). Young infants who are candidates for a BAHS should be managed by a team consisting of at least an otolaryngologist, speech language pathologist, and an audiologist (National Deaf Children's Society [NDCS], 2010; Snik et al., 2005; Snik, Leijendeckers, Hol, Mylanus, & Cremers, 2008).

When determining candidacy for the BAHS, the criteria is identical for individuals across all ages. As mentioned, the original use of the BAHS was for those with bilateral conductive or mixed losses. Individuals with unilateral loss may also be candidates for a BAHS, including those with unilateral conductive loss and unilateral profound sensorineural loss, also known as single-sided deafness (SSD).

1.4.2.1 Conductive hearing loss

Mylanus, van der Pouw, Snik and Cremers (1998) found that individuals with an air-bone gap of 30 dB or more had greater benefit with the BAHS than air-conduction hearing aids, in comparison to individuals with a smaller conductive component to their hearing loss. A large gain requirement in air-conduction hearing aids causes distortion, which can lead to poorer sound quality and higher chance of feedback. Therefore, for individuals to see significant advantages with the BAHS over the air-conduction hearing aid, the BAHS is recommended to those with a conductive hearing loss containing an air-bone gap of 30 dB or greater (Mylanus, van der Pouw, Snik, & Cremers, 1998).

Some discussion in the literature involves the benefit of bilateral BAHS (reviewed by Janssen, Hong, & Chadha, 2012). Although Priwin, Stenfelt, Granström, Tjellström and Håkansson (2004) found that adults showed better speech perception and localization of sounds with bilateral BAHS, the issue is debated for adults because some authors believe that the bone-conducted signals may cross to the contralateral cochlea and result in interference (Snik et al., 2005; Snik, Leijendeckers, Hol, Mylanus, & Cremers, 2008; Stenfelt, 2005). The situation for infants may be different. As previously described, the transcranial attenuation of a bone-conducted sound is higher for infants compared to adults. Priwin, Jönsson, Hultcrantz and Granström (2007) found improved speech recognition in noise and sound localization with bilateral BAHS for children age 6 years and up. Therefore, the contralateral bone-conducted signal may not cause significant interference, and bilateral BAHS may be useful, particularly for children with bilateral conductive losses (Snik, Leijendeckers, Hol, Mylanus, & Cremers, 2008). Only one study to date has found improved performance with a bilateral soft band BAHS with a single subject 2 years of age (Hol, Cremers, Coppens-Schellekens, & Snik, 2005).

As mentioned earlier, individuals who are fitted with a BAHS commonly have conductive hearing losses due to chronic otitis media with ear drainage or congenital malformations of the ear (e.g., atresia). These conditions will be described in more detail in the following sections. People with otosclerosis and tympanosclerosis may also be candidates for a BAHS. Although not listed in the original criteria, individuals with otosclerosis and tympanosclerosis who could not undergo surgery or where surgical attempts at correcting the loss has failed may find more benefit with a BAHS compared to an air-conduction hearing aid if their air-bone gap is 30 dB or greater (Burrell, Cooper, & Proops, 1996; Snik et al., 2005).

1.4.2.1.1 Chronic otitis media

Individuals with chronic ear drainage make up the greatest number of adult BAHS users (Tjellström, Håkansson, & Granström, 2001). Placing an air-conduction hearing aid ear mold in a draining ear canal blocks the infection in the ear and prevents the fluid from drying. Studies have shown that using a BAHS reduces the frequency of ear infections compared to air-conduction hearing aids (Macnamara, Phillips, & Proops, 1996; McDermott, Sunil, Reid, & Proops, 2002; Mylanus, van der Pouw, Snik, & Cremers, 1998). For infants and young children with chronic otitis media where pressure equalization tubes cannot be inserted, the BAHS soft band may be another option. Ramakrishnan, Davison and Johnson (2006) found improved parent and teacher scores on a Modified Listening Situation Questionnaire after children over age 6 years with otitis media wore the BAHS soft band at school. However, unaided thresholds of the children were 20-30 dB HL; which is often better than aided thresholds measured through the soundfield (Christensen, Smith-Olinde, Kimberlain, Richter, & Dornhoffer, 2010). No studies have investigated the audiological benefit or long-term outcomes for children with otitis media using a BAHS soft band.

1.4.2.1.2 Congenital malformation of the ear

Among children, the majority of BAHS users are those with conductive hearing loss due to congenital malformations of the ear, such as atresia (Tjellström, Håkansson, & Granström, 2001). Infants born with a conductive loss due to atresia of the ear canal may eventually undergo corrective surgery; however, Declau, Cremers and Heyning (1999) recommended that reconstructive surgery should not take place until the child is 5-6 years of age. Therefore, a BAHS could be used as a temporary intervention strategy. They also suggested that a BAHS could potentially provide the child with better hearing thresholds than reconstructive surgery and both options should be considered. Infants with congenital conductive losses that cannot be surgically repaired may be candidates for the BAHS on a permanent basis (Snik, Leijendeckers, Hol, Mylanus, & Cremers, 2008; Tjellström, Håkansson, & Granström, 2001).

1.4.2.2 Mild-moderate sensorineural hearing loss

The sensorineural component of an individual's hearing loss must be considered when deciding candidacy for the BAHS. The Classic BAHS specifications recommend that the patient have bone-conduction thresholds less than 45 dB HL. Håkansson et al. (1990) revealed that 80% of individuals with bone-thresholds of 40 dB HL or less were satisfied with their device. The more powerful body-worn models allow for bone-thresholds between 30-60 dB HL (Snik, Jorritsma, Cremers, Beynon, & Van den Berge, 1992; Snik et al., 2005; Tjellström, Håkansson, & Granström, 2001). Head-worn power devices were recently developed and are now available to individuals with bone-conduction thresholds up to 55 dB HL (Cochlear, 2011b).

1.4.2.3 Unilateral conductive hearing loss

For adults, a conservative estimate of transcranial attenuation is 0 dB, and therefore, bone-conducted sound is assumed to transmit to both cochleae almost equally; therefore, for individuals with unilateral conductive hearing loss, sound is reaching the normal-hearing ear both through natural hearing and through the bone-conducted transmission from the contralateral mastoid, potentially causing interference. As described in an earlier section, infants have higher transcranial attenuation, and therefore the sound reaching the contralateral cochlea may not cause as much interference; however, this prediction has not been investigated.

Studies measuring subjective benefit have revealed improved patient outcome measures and good compliance with the device worn on the side with poorer hearing compared to when unaided (Hol, Snik, Mylanus, & Cremers, 2005; Kunst et al., 2008; Wazen, Spitzer, Ghossaini, Kacker, & Zschommler, 2001). However, studies measuring audiological benefit in adults have shown inconsistent findings. Hol, Snik, Mylanus and

Cremers (2005) found significant improvement in speech recognition threshold and localization tests when subjects wore the BAHS compared to when unaided. Wazen, Spitzer, Ghossaini, Kacker and Zschommler (2001) did not find improved word recognition scores; however, they did show pure-tone and speech reception threshold improvement in the aided condition through the soundfield.

Two studies have investigated benefit with the BAHS for unilateral conductive hearing loss in children over age 6 years compared to when unaided. Kunst et al. (2008) did not find significant improvement for localization tests, but did find improvement in speech recognition in noise in five of the eight children tested. Priwin, Jönsson, Hultcrantz and Granström (2007) found a decrease in localization scores and no improvement in speech recognition in noise; however, a ceiling in performance was reached as most children performed well in the unaided condition. The benefit of the BAHS for children age 5 or younger who have unilateral conductive losses has not been investigated.

1.4.2.4 Single-sided deafness

Individuals with single-sided deafness (SSD) often complain of poor localization and trouble hearing speech in noise, especially when the talker is situated on the side of the deaf ear (Bosman, Hol, Snik, Mylanus, & Cremers, 2003). The use of the BAHS for SSD is for contralateral routing of the signal to reduce the head-shadow effect. This technique is similar to that used in the contralateral routing of signal (CROS) air-conduction hearing aid, where the signal is captured by the microphone on the patient's deaf ear, then routed to the receiver built into an ear mold in the better hearing ear (Dillon, 2001). An ear mold may partially occlude the normal hearing ear and prevent natural sounds from entering the ear canal, and hearing aid receivers can produce poor sound quality. These disadvantages to using air-conduction CROS aids can be avoided with the BAHS. The BAHS is also recommended for young children with SSD. However, as described, transcranial attenuation of bone-conducted sound is higher for infants than adults, and therefore, research is needed to understand the transcranial properties of bone-conducted sound for infants to predict the outcomes of children using the BAHS for trancranial routing of the signal.

Investigation of the benefits of the BAHS to individuals with SSD has yielded mixed results. Some studies showed that participants subjectively reported better outcomes with the BAHS compared to the air-conduction CROS (Bosman, Hol, Snik, Mylanus, & Cremers, 2003; Wazen et al., 2003). In contrast, Snik et al. (2005) reported that 25% of patients in the Nijmegen clinic fitted with a BAHS steel headband on a trial basis did not find benefit and stopped wearing the device. Improvement of speech recognition scores also varied across studies. Most studies found that either the BAHS or air-conduction CROS system resulted in improved speech recognition scores compared to the unaided condition (Bosman, Hol, Snik, Mylanus, & Cremers, 2003; Christensen & Dornhoffer, 2008; Hol, Bosman, Snik, Mylanus, & Cremers, 2005; Niparko, Cox, & Lustig, 2003); however, only one study found improved speech recognition scores with the BAHS compared to the air-conduction CROS (Niparko, Cox, & Lustig, 2003). Studies have found that sound localization does not improve with the use of a BAHS or air-conduction CROS (Bosman, Hol, Snik, Mylanus, & Cremers, 2003; Hol, Bosman, Snik, Mylanus, & Cremers, 2005; Niparko, Cox, & Lustig, 2003). Studies on the benefits of the BAHS for those with SSD have used only adults or teenagers as their research subjects, and no study has looked at the effects of a BAHS for infants or young children with SSD.

1.4.3 Implant and abutment or soft band for pediatric BAHS users

The Food and Drug Administration in the United States of America approved BAHS implants for children age 5 years or older who fit the candidacy criteria. Children younger than 5 years of age are provided with a BAHS soft band. In other countries, the implant is used for children as young as 18 months (Tjellström, Håkansson, & Granström, 2001); however, most guidelines recommend that children are not implanted until at least 3-4 years of age (National Deaf Children's Society [NDCS], 2010; Snik et al., 2005). Two applications of the BAHS in children are described in the following section, the implant and abutment, and the soft band.

1.4.3.1 Implant and abutment

As discussed in a previous section, the skulls of young children are different than adults in their structure, mineral composition, and thickness; therefore osseointegration has been carefully studied for infants and young children. Granström, Bergström, Odersjö and

Tjellström (2001) recommended that the skull be at least 2.5 mm thick for a 3 mm implant. By 5-7 years of age, the skull should be 2.5 mm thick; however, the authors acknowledged that the surgery can be performed successfully on children as young as 2-4 years of age if the surgeon is aware of precautions and uses special techniques to stimulate bone growth (Davids, Gordon, Clutton, & Papsin, 2007). Granström, Bergström, Odersjö and Tjellström (2001) found that osseointegration was successful and had a low percentage of implant failures from their sample (5.8%). Some reports show similar failure rates of 3.4% (Jacobsson, Tjellstrom, Fine, & Andersson, 1992), 4.2% (Marsella, Scorpecci, Pacifico, Presuttari, & Bottero, 2012) and 5% (Tietze & Papsin, 2001), while others have a higher failure rate of 15% (Papsin, Sirimanna, Albert, & Bailey, 1997) and 16.3% (de Wolf, Hol, Huygen, Mylanus, & Cremers, 2008). Generally, failure rates among children are either lower than adults (Snik et al., 2005; Tjellström, Håkansson, & Granström, 2001) or very similar (Zeitoun, De, Thompson, & Proops, 2002). However, special surgical techniques are required to succesfully implant in children with thin bones (Snik et al., 2005; Snik, Leijendeckers, Hol, Mylanus, & Cremers, 2008). Zeitoun, De, Thompson and Proops (2002) also noted that individuals with craniofacial anomalies may have different bone thickness, which needs to be considered. Some reports claimed these failure rates are largely due to trauma (Seemann, Liu, & Di Toppa, 2004; Zeitoun, De, Thompson, & Proops, 2002). Even without failures, children with the implant will likely undergo revision surgery due to the growth and development of subcutaneous tissue and bone (Granström, Bergström, Odersjö, & Tjellström, 2001; Hol, Snik, Mylanus, & Cremers, 2005; Tjellström, Håkansson, & Granström, 2001).

One study investigating outcome measures in children with the percutaneous BAHS has found significant improvement in aided threshold and positive caregiver feedback with regards to listening situations compared to when unaided (Seemann, Liu, & Di Toppa, 2004).

1.4.3.2 Soft band

The BAHS soft band is an elastic band that wraps around the child’s head. A plastic disk with an abutment snap on the outside is incorporated into the band. The BAHS snaps onto the abutment on the plastic disk instead of the abutment connected to the implant in the skull. Contrary to the percutaneous implant system, the soft band works through transcutaneous stimulation.

Limited research is available on outcomes with the soft band BAHS. Hol, Cremers, Coppens-Schellekens and Snik (2005) and Verhagen, Hol, Coppens-Schellekens, Snik and Cremers (2008) investigated speech and language development and aided thresholds for a small sample of children wearing the soft band BAHS. Initially, they found that children had a mean speech and language delay of one month with the soft band BAHS (Verhagen, Hol, Coppens-Schellekens, Snik, & Cremers, 2008); however, a long-term follow-up of two children in the sample using the soft band BAHS showed that speech and language scores fell into a range typically for their age-matched peers with normal hearing (Hol, Cremers, Coppens-Schellekens, & Snik, 2005).

Snik, Leijendeckers, Hol, Mylanus and Cremers (2008) described a single-case of a child fitted with a BAHS soft band at age 4 months and bilateral implants at 44 months. Measures of receptive and expressive language showed that initial scores at age 24 months were above average but then dropped to below average over the following two years. After implantation, scores returned to above average. This case study demonstrated that soft band use of the BAHS was sufficient for basic language development; however, better and more stable hearing was required for complex language development. Based on the results from these small studies, it is recommended that children using the soft band are fitted with a more powerful device to achieve the necessary gain (Hol, Cremers, Coppens-Schellekens, & Snik, 2005; Snik et al., 2005), and that the transition to the implant is made soon after the child is of eligible age, in order to provide the best outcomes for speech and language development (Verhagen, Hol, Coppens-Schellekens, Snik, & Cremers, 2008). Percutaneous and transcutaneous stimulus delivery is discussed in greater detail in a later section.

1.5 Factors that Affect Bone-Conduction Transmission

When an auditory signal is transmitted via bone conduction, there are many factors which contribute to the final intensity of the stimulus at the cochlea. These factors include the contact area on the skull, and the force applied to the area, the position on the head where the stimulus is delivered, and whether the transducer is attached percutaneously or transcutaneously. These factors will be discussed separately in the following sections.

1.5.1 Contact force and area

The quantity of force used to couple the bone-conduction transducer to the skull, whether for diagnostic or intervention purposes, is referred to as contact force. Studies have shown that thresholds generally decrease as the contact force increases for low-frequency bone-conduction stimuli (Lau, 1986; Watson, 1938); however, for forces greater than approximately 500 grams, the effect of contact force diminishes (Lau, 1986; Von Békésy, 1960). In addition, contact force affects bone-conducted signals more for low frequencies than for high frequencies (Lau, 1986; Watson, 1938). Whittle (1965) reported that subjects often complained of pain with coupling forces of 7.5 N, and recommended a standardized coupling force of 4.5 N. The ANSI standard for the recommended contact force is 5.4 N, corresponding to 550 grams +/- 50 grams (American National Standards Institute [ANSI], 1996).

Researchers have further investigated the effect of contact force on physiological thresholds in infants. Yang, Stuart, Stenstrom and Hollett (1991) measured click-ABR wave V latencies on newborn infants, using a variety of contact forces (225, 325, 425, and 525 grams). They found that latencies decreased as the contact force increased. The authors also commented that increasing the contact force to 525 grams caused the transducer to slip easily outside of the elastic band used to couple the transducer to the head. Therefore, they recommended a contact force of 400-450 grams for physiological measures on infants. Based on their recommendations, subsequent studies measuring infant physiological thresholds have used 425 grams as their calibrated contact force when coupling the bone-conduction transducer to the infant's skull (e.g., Small, Hatton, & Stapells, 2007).

In a clinical setting, parents are instructed to tighten the soft band to a level that is snug but comfortable. It is typically suggested to parents that they should be able to slide no more than one or two fingers under the band. To investigate the optimal force to apply to the soft band, Hodgetts, Scollie and Swain (2006) measured the output vibratory force of the BAHS soft band when applied to an artificial mastoid with contact forces of 2, 3, 4, and 5 N. They defined a loose soft band fitting to be approximately 2 N. They found only slight differences between the 2 and 5 N conditions, and only for high frequencies and none of the differences in output reached significance. Based on these results, it is plausible that a tight fitting soft band may not be necessary for delivering enough output to the child. However, these measurements were recorded on an artificial mastoid with adult skull properties and not directly on a pediatric sample. The properties of the skull for children under age 7 have not yet been investigated, and therefore it is unknown whether the artificial mastoid is an appropriate measurement tool to estimate the skull of this population. In addition, although no significant differences were observed with a loose fitting contact force, the artificial mastoid is a stationary object, and the loose fitting BAHS may not be optimal if you factor in movement of the child.

Verstraetan, Zorowski, Somers, Riff and Offeciers (2009) also compared contact force through investigation of the steel headband and a newer headband that put less pressure on the head to improve user comfort. They found that audiometric and speech reception thresholds were comparable between the two types of band. Therefore, because hearing is not compromised, when applying a signal transcutaneously through a BAHS, a force of 5.4 N was not necessary for achieving good results with a BAHS. More research is needed to directly compare the effects of contact force on hearing function for infants and children using a soft band device.

Contact area is another factor involved in bone-conduction sensitivity. Watson (1938) and Khanna, Tonndorf and Queller (1976) found that thresholds were lower for high frequencies when a larger contact disk was used, whereas there was little change seen for low frequencies. Khanna, Tonndorf and Queller (1976) also found significant variation in acceleration levels measured at threshold with a variety of contact disk sizes (diameter of 1.6 to 3.0 cm). Generally, a larger disk resulted in lower acceleration, particularly for frequencies between 1000 and 4000 Hz. Contact area was not investigated in the present study.

1.5.2 Skull location

Many studies have provided evidence that the mastoid is generally more sensitive to bone-conduction sensitivity than the forehead (McBride, Letowski, & Tran, 2008; Small, Hatton, & Stapells, 2007; Studebaker, 1967; Watson, 1938; P. B. Weston, Gengel, & Hirsh, 1967). Weston, Gengel and Hirsh (1967) found that average thresholds at the forehead for 500 Hz were 15-20 dB higher than thresholds at the mastoid. For frequencies 1000 Hz and higher, forehead thresholds were only slightly higher. In addition, threshold measures were more variable for the mastoid than the forehead placement. Subsequently, McBride, Letowski and Tran (2008) measured bone-conduction thresholds at 11 different skull positions. The mean thresholds for 500, 1000, 2000, and 4000 Hz on the mastoid were 4, 0, 0 and 6 dB HL while the mean thresholds for the forehead were 15, 7, 4, and 15 dB HL. In addition to thresholds, Huizing (1960b) noted that measurements of sound pressure depended on the position of the transducer (ipsilateral mastoid, contralateral mastoid, forehead, crown, and occipital bone). For a 500 Hz stimulus, attenuation was greatest when the transducer was placed on the frontal and occipital bone, and least when recorded on ipsi- and contralateral mastoids.

Bekesy believed that the variability of skin thickness at the mastoid made it a suboptimal position for the transducer, and that the stable bone and skin thickness across the forehead make a better surface for delivering consistent stimuli (Whittle, 1965). Although test-retest reliability may be lower for the mastoid than the forehead (P. B. Weston, Gengel, & Hirsh, 1967), agreement that adult thresholds are better at the mastoid have warranted standards committees to assign the mastoid as the recommended transducer position for diagnostic bone-conduction audiometry (American National Standards Institute [ANSI], 1996).

The effects of transducer position on adult bone-conduction hearing are well established, however, less research is available for infants. Yang, Rupert and Moushegian (1987) investigated ABR wave V latencies for infants with the transducer on both the mastoid and forehead. They found that latencies were shorter with the transducer on the mastoid compared to the forehead for neonates and 1-year old infants. The same pattern was found for adults; however, the latency differences between the two transducer positions were not significant and substantially smaller compared to the infant groups. Stuart, Yang and Stenstrom (1990) also recorded ABR wave V latency to bone-conduction clicks in newborn infants. They found that the latency changed when the transducer was moved across three different positions on the temporal bone: the mastoid, superior to the pinna, and supero-posterior to the pinna. The shortest latencies were with the transducer on the mastoid; however, effective coupling of the transducer to the mastoid was difficult and responses were variable. As a result, the researchers recommended positioning the transducer on the temporal bone, supero-posterior to the pinna.

More recently, Small, Hatton and Stapells (2007) investigated ASSR threshold differences for three skull locations (mastoid, high temporal bone, and forehead) for newborn infants with normal hearing. Thresholds were similar when the transducer was placed on the temporal bone and the mastoid. However, thresholds were significantly poorer with the transducer on the forehead. Mean thresholds for the mastoid, temporal bone, and forehead can be observed in Table 1.1. Based on these findings, the authors recommended that the bone-conduction transducer should be placed on either the mastoid or temporal bone but not the forehead when performing physiological bone-conduction measures on infants.

Table 1.1. Mean ASSR thresholds for transducer positions for neonatal infants with normal hearing in dB HL (Small, Hatton & Stapells, 2007).

	Frequency (Hz)			
Transducer Position	**500**	**1000**	**2000**	**4000**
Mastoid	17.3	14	32.3	26
Temporal bone	16	16.7	34.6	33.3
Forehead	30.7	26.7	51.1	44

1.5.3 Percutaneous or transcutaneous transmission

With the invention of the BAHS, a series of investigations were completed to determine the benefit of a percutaneous over a transcutaneous system. Håkansson, Tjellström and Rosenhall (1984) found significantly lower thresholds across frequencies 500 to 6000 Hz with the transducer attached percutaneously compared to transcutaneously. The largest mean differences (20 dB) were around the resonant frequency of the transducer (750 to 850 Hz). To follow up on their findings, Håkansson, Tjellström and Rosenhall (1985) measured the acceleration level at threshold for frequencies 250 to 6000 Hz with percutaneous and transcutaneous stimulation. They aimed to create a correction factor to predict benefit from a percutaneous system for individuals wearing a transcutaneous BAHS. Large variation was observed between subjects and no consistent patterns were found across frequency. The average decrease in threshold with percutaneous stimulation was 21 dB (range 16-28 dB). Similar threshold changes were observed in subsequent studies (Mylanus, Snik, & Cremers, 1994; Stenfelt & Håkansson, 1999). Verstraeten, Zarowski, Somers, Riff and Offeciers (2009) measured aided thresholds through the soundfield and found a similar pattern of results.

Håkansson, Tjellström and Rosenhall (1984; 1985) predicted that the difference between transcutaneous and percutaneous thresholds was due to the compliance and resistance of the skin and subcutaneous tissue that attenuates and directs the acceleration through the skin. In response to their prediction, studies have attempted to find methods to predict threshold variation for individuals before obtaining the BAHS implant and abutment (Mylanus, Snik, & Cremers, 1994; Stenfelt & Håkansson, 1999). Mylanus, Snik and Cremers (1994) investigated thresholds transcutaneously and percutaneously, but also measured the thickness of the skin and subcutaneous tissue to explore whether skin thickness was correlated to threshold difference. No significant correlations were found between skin and subcutaneous tissue thickness and the threshold differences between stimulation modes. Therefore, success with the percutaneous device cannot be predicted from either a physical measure of skin and subcutaneous tissue or a correction factor.

In summary, there is no known procedure that can predict the benefit a particular individual will receive with the percutaneous BAHS over a transcutaneous system; however, it has been shown that thresholds are lower and therefore the benefit is greater with the percutaneous system over the transcutaneous system. Snik et al. (2005) recommended that candidacy evaluations for a percutaneous BAHS should include a trial with a transcutaneous BAHS, either on a headband or soft band. Those who find benefit with a transcutaneous system will likely find equal or greater benefit with a percutaneous system. However, for those who do not find benefit with a transcutaneous system, it is unknown whether a percutaneous system will be helpful. More discussion on the fitting protocol for BAHS is provided in the following section.

1.6 Fitting Hearing Devices

The process of fitting a hearing device begins with conversion of the individual's hearing thresholds into graphs that match the output scale of the hearing device and account for ear canal properties of the individual. Second, the audiologist must calculate the amount of gain required to acquire suitable audibility for understanding speech. This step is referred to as prescribing aided output targets for speech. Based on these values, an appropriate hearing device can be selected, and gain can be adjusted. Finally, the audiologist should perform an objective test to determine if the output of the device is reaching the previously prescribed targets. This step is referred to as verification of the hearing device.

Dickinson (2010) described the difference in fitting and verification protocols for air-conduction and BAHS. Air-conduction hearing aid fittings are performed using a prescriptive approach, based on formulas and algorithms from the severity and configuration of hearing impairment. BAHS fittings use an evaluative approach, in which the audiologist adjusts the hearing device based on a clinical observation and the individual's experience and preferences.

The fitting procedure for air-conduction hearing aids has been well documented (e.g., Valente et al., 2006); however, the fitting procedure currently used for BAHS contains significant limitations. Before outlining the BAHS fitting procedure, a clear understanding of the fitting procedure for air-conduction hearing aids is helpful.

1.6.1 Current practice for fitting air-conduction hearing aids

The protocol for fitting air-conduction hearing aids has been extensively researched and documented by professional organizations for both adults (e.g., Valente et al., 2006) and children (American Academy of Audiology [AAA], 2004; The Pediatric Working Group, 1996). The following sections provide a brief overview of this protocol.

1.6.1.1 Converting thresholds

Real-ear-to-coupler difference (RECD) is a measure to account for individual ear canal size and acoustics. To select an appropriate hearing aid for an individual, air-conduction thresholds must first be converted from dB HL to real-ear dB SPL. Because ear canal acoustics for infants and young children change rapidly throughout early development, thresholds should be converted to real-ear dB SPL to accurately compare any change in hearing thresholds over time (Tharpe, Sladen, Huta, & McKinley Rothpletz, 2001). RECD is measured by placing a plastic probe-tube microphone in the individual's ear canal along with either a foam insert earphone or the hearing aid ear mold which is connected to the hearing aid analyzer (Bagatto, 2001; Tharpe, Sladen, Huta, & McKinley Rothpletz, 2001). Children will not always tolerate this set up; therefore, the average RECD values for the particular age group of the child may be used (Bagatto, Scollie, Seewald, Moodie, & Hoover, 2002). However, Bagatto and Scollie (2010) predicted up to +/- 15 dB of error when using average RECD values, and therefore measured RECD values are always preferred.

1.6.1.2 Prescribing output targets for aided speech

Selection of an appropriate hearing aid depends on the calculated target output levels for an individual based on their hearing thresholds. Many algorithms are available for calculating these targets; however, two are most commonly used: the Desired Sensation Level multistage input/output algorithm (DSL[i/o]) version 5 (Scollie et al., 2005) and the National Acoustics Laboratories nonlinear algorithm (Byrne, Dillon, Ching, Katsch, & Keidser, 2001).

1.6.1.3 Verification of aided output

As defined earlier, verification of the hearing aid is performed to confirm that hearing aid output provides enough gain for speech to be audible to the individual. Verification can either be completed through real-ear measures with the probe-tube microphone at the level of the eardrum, or it can be completed in the hearing aid analyzer text box, using the RECD values to predict the output in the ear canal. The output of the hearing aid is matched to the targets prescribed through the fitting algorithm. An SPLogram is a plot consisting of the person's real-ear SPL thresholds, prescribed targets, predicted loudness-discomfort-levels, and hearing aid output for a speech signal (Dillon, 2001).

For adults, real-ear aided output can be measured directly in the person's ear; however, the procedure requires the individual to sit still and quiet for the duration of the measurement, with their head in an upright and stable position. This is often not possible with young children. By using the RECD to convert the signal online, the real-ear aided output can be approximated when measured in a 2-cc coupler (Bagatto et al., 2005; Bagatto & Scollie, 2010).

1.6.2 Current practice for fitting bone-anchored hearing systems

Fitting a BAHS begins with measuring both air- and bone-conduction thresholds. Candidacy and device selection is determined based on the bone-conduction thresholds and size of the air-bone gap. As described earlier, children under age 5 years are fitted with the BAHS soft band; however, different manufacturers and models are available from which the audiologist can select the most appropriate device.

Although some census documents are available regarding fitting guidelines for BAHS, research in this area is limited, particularly in comparison to air-conduction hearing aid fitting protocols. The National Deaf Children’s Society (NDSC) in the UK has outlined pediatrics fitting guidelines for the BAHS (National Deaf Children's Society [NDCS], 2010). In addition, Snik et al. (2005) reported on decisions from a round-table discussion with experts in the field on currently available procedures for fitting a BAHS for both adults and children. In North America, it has been reported that the American Academy of Audiology will also develop consensus documents for both adults and children (Sockalingam, 2012). Researchers agree that current fitting protocols for the BAHS have significant limitations, and an updated fitting procedure is necessary.

The current fitting procedure is discussed in the following sections. Verification of the BAHS is directly related to this research, and therefore, currently-used verification procedures are discussed, followed by a synopsis of recent investigations on potential future approaches to verification of the BAHS output.

1.6.2.1 Converting thresholds

Recently, BAHS manufacturers have made it possible to estimate bone-conduction thresholds in units of force through the device itself on the head of the individual. For an air-conduction hearing aid fitting, the RECD converts air-conduction thresholds in dB HL to real-ear dB SPL. Measuring thresholds *in situ* through the BAHS itself eliminates this step.This procedure is particularly important for individuals with the implant and abutment because research has shown significant thresholds differences between transcutaneous and percutaneous stimulus delivery (Håkansson, Tjellström, & Rosenhall, 1984; Håkansson, Tjellström, & Rosenhall, 1985; Mylanus, Snik, & Cremers, 1994; Verstraeten, Zarowski, Somers, Riff, & Offeciers, 2009). In addition, Christensen, Smith-Olinde, Kimberlain, Richter and Dornhoffer (2010) showed a significant decrease in aided soundfield thresholds when the soft band BAHS was compared to the conventional bone-conduction hearing device, which is similar to a diagnostic bone oscillator. In this case, the stimulus was presented transcutaneously for both devices, yet threshold differences were found. Mechanical differences between the two instruments were likely contributing factors. The difference between the diagnostic transducer and BAHS revealed that *in situ* threshold measurements may also be useful for soft band users. Although it is recommended for thresholds to be measured through the BAHS, retesting thresholds may not always be feasible for the pediatric population, due to attention issues that limit testing time.

1.6.2.2 Prescribing output targets for aided speech

Currently, the only widely used formulae for prescribing targets are manufacturer-developed algorithms built into the software (e.g., Oticon Medical, 2011). For his own lab, Hodgetts (2010b) modified a version of the DSL [i/o] to prescribe appropriate targets for aided speech within the individual’s dynamic range. In a study by Hodgetts, Hagler, Håkansson and Soli (2011), subjects had better outcome measures for a variety of speech tests when a prescriptive (audibility-derived) approach was used for their BAHS fitting, compared to when an evaluative (patient-derived) fitting approach was used. Aided output measurements revealed significantly more gain in the high frequencies for the prescriptive approach, but subjective reports from patients did not show significant differences. Therefore, a prescriptive approach is a more appropriate strategy for BAHS fitting, and more research is required to develop prescriptive fitting formulae for BAHS, for both adults and children. This method shows a lot of promise, but this approach is not yet clinically available.

1.6.2.3 Verification of aided output

Current verification techniques for the BAHS have significant limitations. Available methods for verification include measurements of functional gain and performance on speech tests. Aided soundfield thresholds with warble tone stimuli have been recommended as the best available verification method to test BAHS performance (National Deaf Children's Society [NDCS], 2010; Nicholson, Christensen, Dornhoffer, Martin, & Smith-Olinde, 2011).

Functional gain is calculated from the unaided and aided thresholds through the soundfield. Adjustments are made to the BAHS as needed. If possible, speech testing may also be completed while both unaided and aided to show benefit with speech stimuli. One limitation is that soundfield testing has poor test-retest reliability and frequency specificity. Other limitations include interactions with the noise floor and nonlinear processing of the device (Nicholson, Christensen, Dornhoffer, Martin, & Smith-Olinde, 2011). For a pediatric BAHS user, if all fitting procedures are carried out in accordance with recommendations (National Deaf Children's Society [NDCS], 2010; Snik et al., 2005), the child would potentially have thresholds measured through (1) the air- and bone-conduction transducers, (2) the BAHS manufacturer software *in situ*, (3) the soundfield while unaided, and (4) the soundfield while aided. This fitting protocol is time consuming for the clinic, and unrealistic for young children tested using VRA. It is also not possible for children under 6 months of age, who are diagnosed through physiological testing. However, there is no better option currently available for verification of the BAHS (Nicholson, Christensen, Dornhoffer, Martin, & Smith-Olinde, 2011).

1.6.3 Verification methods for bone-anchored hearing systems under investigation

The necessity for a more effective and efficient verification protocol of the BAHS is apparent. Only a few studies have looked at possible future methods for verification, and more research is required to develop an appropriate yet inexpensive tool for BAHS verification.

1.6.3.1 Sound pressure in the ear canal

One proposed technique is to measure the sound pressure in the ear canal during *in situ* stimulation from the BAHS (Dickinson, 2010; Hodgetts, Håkansson, Hagler, & Soli, 2010). Hodgetts, Håkansson, Hagler and Soli (2010) measured threshold the device *in situ* with a probe tube microphone in the ear canal. The sound pressure radiating in the ear canal from bone-conducted stimulation was recorded. The aided long-term average speech spectrum (LTASS) at different input levels was delivered directly to the BAHS, and sound pressure in the ear canal was recorded. In the clinic, gain would then be adjusted to obtain optimal audibility across a range of frequencies. Dickenson (2010) also suggested this technique for verifying the BAHS for SSD patients by measuring the sound radiating in the contralateral ear canal.

The advantages to using real-ear measures for verification are its clinically feasibility and low cost; however, there are also limitations to using this method. First, for many BAHS users, bone-conduction thresholds are near normal. Therefore, the noise floor will likely interfere with the measure of sound pressure in the ear canal both for threshold and for low-input LTASS measurements (Hodgetts, Håkansson, Hagler, & Soli, 2010). For children, this approach is even less feasible. Children who are not able to sit through real-ear measurements with air-conduction hearing aids will not likely have the patience for this similar type of probe tube measurement. In addition, congenital conductive hearing loss due to craniofacial anomalies make up the largest number of pediatric BAHS users (Tjellström, Håkansson, & Granström, 2001). Atresia, microtia, or chronic middle-ear drainage will make it difficult or impossible to perform probe tube measures (Hodgetts, Håkansson, Hagler, & Soli, 2010; Nicholson, Christensen, Dornhoffer, Martin, & Smith-Olinde, 2011).

1.6.3.2 Acceleration and force measurements

1.6.3.2.1 In situ

Hodgetts, Håkansson, Hagler and Soli (2010b) proposed a second procedure where acceleration is directly measured from the BAHS as it is stimulated on the abutment of the individual. To make this measurement in their study, special equipment that is not clinically available was required. The measurement assembly included a balanced electromagnetic separation transducer (BEST) and an accelerometer. The BEST is unique in that it vibrates equally throughout its core, and therefore measurements could be accurately taken from the backside of the transducer (Håkansson, 2003). During *in situ* bone-conduction stimulation, a measure of acceleration from the device was made using an accelerometer, which was mounted on the backside of the BEST transducer attached to the abutment on the skull.

This measurement procedure paralleled the technique measuring sound pressure in the ear canal. Thresholds were acquired in acceleration level. The input LTASS was then directed to the BAHS and the output spectrum was mapped in acceleration level. Clinically, the gain would be adjusted to optimize audibility within the measured dynamic range.

In contrast to probe tube techniques, the noise floor is not a factor when measuring threshold through acceleration *in situ*, and a BAHS for any individual, regardless of his or her medical or anatomical situation, can be verified using this method. However, the BEST is not clinically available and the cost to manufacture this type of transducer is high (Hodgetts, Håkansson, Hagler, & Soli, 2010). Similarity, accelerometers are also very expensive. If the equipment were to become readily available, new techniques and technical details would require audiologists to undergo extensive training on using the equipment (Hodgetts, 2011). Finally, similar to previously described methods, young children may not be able to sit long enough for to complete *in situ* acceleration measurements. If an *in situ* measurement is developed, alternate methods must become available for children who are unable to perform real-time measurements of BAHS performance.

1.6.3.2.2 Hearing aid analyzer test box

From these limitations, some potential solutions using test box measures have been proposed. First, Dickinson (2010) suggested coupling a test rod that comes with the device to the measurement microphone of the hearing aid analyzer. The test rod is a device used to test the BAHS. An individual can hold the test rod coupled to the BAHS to his or her mastoid to listen through the BAHS. In the hearing aid analyzer test box, the reference microphone should be positioned next to the microphone of the BAHS. However, this method cannot be used to map output to an audible level within the dynamic range, but can only be used as a relative measure to track changes in gain at follow-up appointments.

A more comprehensive method of verification uses the test box in conjunction with a skull simulator or artificial skull. The skull simulator is effectively a 2-cc coupler for the mastoid (Hodgetts, 2010a). The TU-1000 is a skull simulator designed for making electroacoustic analyses of the BAHS (Håkansson & Carlsson, 1989). The skull simulator is not considered an artificial skull or mastoid because it does not perfectly match the properties of an adult skull. Artificial mastoids used for calibrating transcutaneous bone-conduction transducers are bulky and expensive and not practical for measuring BAHS output in a hearing aid analyzer test box. It is also important to note that while artificial mastoids contain the properties of the skin-covered skull, the skull simulator is intended for percutaneous BAHS use.

By using a skull simulator in the test box, the force can be measured and recorded through the hearing aid analyzer. The skull simulator itself would need to be coupled to the hearing aid analyzer (Håkansson & Carlsson, 1989). *In situ* thresholds in units of force would be entered into the hearing aid analyzer. The LTASS would be projected from the speakers in the test box, and the aided LTASS output measured from the skull simulator would be displayed with reference to the inputted thresholds (Dickinson, 2010; Håkansson & Hodgetts, 2009; Hodgetts, 2010a). It is important to note that the aided output is in reference to the skull simulator, while the bone-conduction thresholds are in reference to the individual's skull; however, because the skull simulator is an adequate simulation of the adult skull (discussed in more detail in section 2.7.4), a real-mastoid to skull-simulator difference is not necessary. A conversion calculation would be necessary to convert the measured force values from the BAHS to the values in the SPLogram on the hearing aid analyzer monitor (Håkansson & Hodgetts, 2009; Hodgetts, 2010a). Currently, the cost of a skull simulator is expensive (Hodgetts, 2010a); however, this method of verification is the most practical to date and a promising option for the future of BAHS fitting (Hodgetts, 2010b). An image of the skull simulator in the test box from Dickinson (2010) is presented in Figure 1.2.

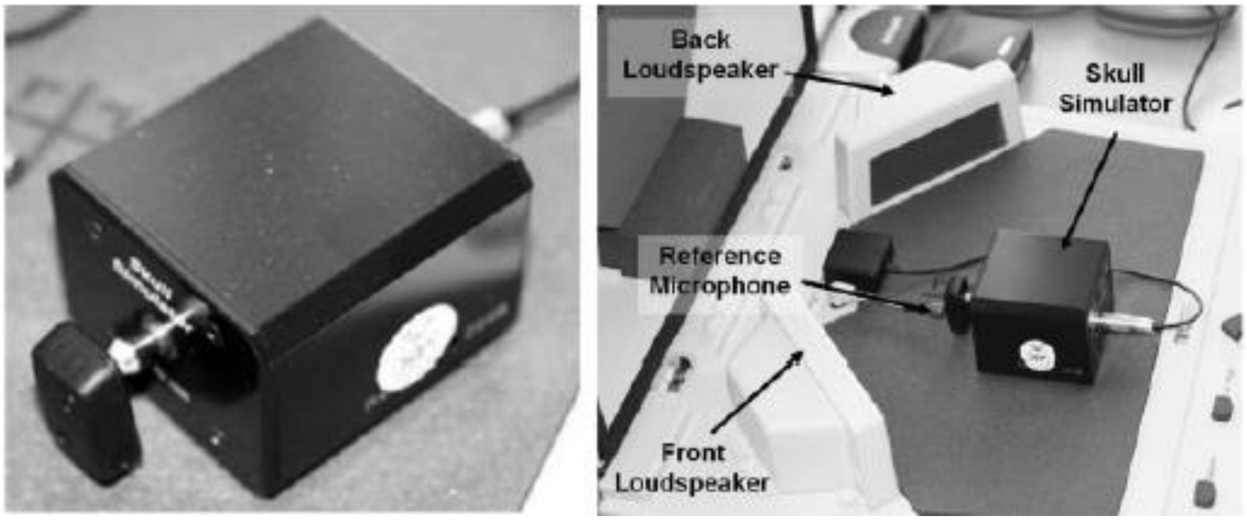

Figure 1.2 Image of the skull simulator with an attached BAHS and the skull simulator set-up in a hearing aid analyzer test box for verification of the BAHS. Presented with permission from Dickenson (2010)**.**

1.7 Mechanical Impedance

As stated earlier, the skull simulator does not perfectly match the properties of an average adult skull. Artificial mastoids were developed to contain properties of the adult human skin-covered skull, such as the standard for mechanical impedance, a mass that corresponds to the total mass of the adult skull, and a rubber pad that corresponds to the skin and subcutaneous tissue above the mastoid bone (American National Standards Institute [ANSI], 1987; Stenfelt & Håkansson, 1998). Artificial mastoids, such as the Brüel and Kjær Artificial Mastoid Type 4930, are commonly used for calibrating bone-conduction transducers; however, they are bulky and expensive and not practical for measuring BAHS output in a hearing aid analyzer test box. Stefelt and Håkansson (1998) developed a miniature artificial mastoid with properties matching those of an average adult head to be used in conjunction with the TU-1000 to measure the output of a BAHS in the test box. In addition to understanding the physical properties of the skull for calibration purposes, it is also important to consider differences in anatomical and impedance characteristics across groups of individuals of different ages to learn more about the maturation of bone-conduction hearing and its underlying mechanisms.

Mechanical impedance is the amount that a system opposes an applied force (Moser, 2009). Mechanical impedance in audiology is typically described in relation to the outer and middle ear systems. In bone-conduction audiometry and BAHS fitting, mechanical impedance of the head is an important factor when considering the amount of opposition from the head in response to a vibratory force. Mechanical impedance can be written in terms of a complex number that contains both a magnitude and phase component. The magnitude of mechanical impedance includes both real and imaginary parts. It can be calculated from the resistance of a system, which is related to its friction, and the reactance of a system, which is related to its stiffness and mass. As a force is applied, the stiffness and mass components of the system store energy while the resistance of the system dissipates energy. The magnitude of mechanical impedance can therefore be described as $Z = R + j(m\omega - k/\omega)$, where R is the resistance, $m\omega$ is the positive reactance defined by the mass, k/ω is the negative reactance defined by the stiffness, and *j* is a complex constant (Moser, 2009). Mechanical impedance (Z) can also be calculated through the force and acceleration outputs with the equation $F = |Z| \times A/\omega$, where F is the force, and A is the acceleration (Håkansson, Tjellström, & Rosenhall, 1985).

Mechanical impedance changes as a function of frequency. As frequency increases, a positive slope corresponds to a positive phase angle and a system controlled through mass. A negative slope corresponds to a negative phase angle and a system controlled through stiffness. The resonant frequency of the system is the point at which the impedance is at its minimum value, where the stiffness and mass component are equal, and the impedance is defined by the resistance (Henry & Letowski, 2007).

In the following sections, measurements of mechanical impedance and resonant frequency for adult heads will be discussed. Because the mechanical impedance is valuable for both transcutaneous transmission of bone-conduction stimulation and for percutaneous transmission with the BAHS, studies have investigated mechanical impedance for both the skin-covered skull and the temporal bone using the BAHS implant and abutment. For the skin-covered skull impedance, differences between groups of individuals will be discussed, in addition to the contact force, contact area, and placement on the skin-covered skull. For the impedance of the temporal bone, differences between groups of individuals will be discussed.

1.7.1 Skin-covered skull

Mechanical impedance of the skin-covered mastoid for adults has been extensively researched. Values for the adult mastoid are published in IEC 373 (1990) and ANSI S3.13 (1987) and are used for designing and calibrating artificial mastoids, which are used to calibrate transcutaneous bone-conduction transducers.

Dadson (1954) and Corliss and Koidan (1955) were the first to measure impedance values for the intention of designing an artificial mastoid. They found that stiffness dominated the impedance measurement until the high-frequencies, after which mass dominated the system. They also found that some damping, or energy dissipation, was present (Corliss & Koidan, 1955; Dadson, Robinson, & Greig, 1954). A damping mechanism refers to the resistance of the system, where friction will eventually cause the energy to dissipate and the system will come to rest. A system without damping will store energy in the reactance indefinitely (Moser, 2009). The trend in impedance across frequency has been replicated by all studies on mechanical impedance of the skin-covered skull (Cortes, 2002; Flottorp & Solberg, 1976; Håkansson, Carlsson, & Tjellstrom, 1986; Smith & Suggs, 1976). The results from the two original studies were combined to create the first standardized measurement of mechanical impedance (Flottorp & Solberg, 1976).

More recently, measures of mechanical impedance have been made with updated methods and instrumentation for the purpose of contributing to the design of an artificial mastoid (Flottorp & Solberg, 1976), and for understanding the dynamic properties of the skull (Smith & Suggs, 1976). Flottorp & Solberg (1976) collected and analyzed mechanical impedance at 10 frequencies between 125 and 6300 Hz. Results of their study indicated that impedance values originally collected for ISO/IEC standards (Corliss & Koidan, 1955; Dadson, Robinson, & Greig, 1954) were not within their confidence intervals for impedance magnitudes, and they recommended that standards needed to be updated (Flottorp & Solberg, 1976).

Håkansson, Carlsson and Tjellström (1986) found a similar pattern of impedance magnitude and phase. They created a three-parameter model of impedance, which included the mass of the outer layer of soft-tissue (M), the compliance of the skin and subcutaneous tissue (C) and the damping effect or resistance (R). The values they calculated were consistent with the findings from Flottorp and Solberg (1976). The average resonant frequency was 3000 Hz with a standard deviation of 590 Hz. Variability in the measurement between studies was likely due differences in methodology, including contact force and area (Håkansson, Carlsson, & Tjellstrom, 1986). Cortes (2002) compared mechanical impedance of the skin-covered mastoid to the Brüel & Kjaer artificial mastoid and reported mean differences of approximately 7 dB between the two measurements, with the artificial mastoid containing higher impedance values throughout the frequency range.

1.7.1.1 Group differences

Smith and Suggs (1976) reported on potential changes in mechanical impedance across age due to the ossification of skull sutures; however, no evidence of age-dependent differences were found in his small sample. The only study to look at different groups of individuals was Flottorp and Solberg (1976) who examined the impedance measurements at the mastoid and forehead for different age groups (young: 9-10 years; adult: 18-38 years; older adult: 48-71 years). When comparing the young group and the adult group, only slight differences were found. For the forehead, the young group had slightly lower impedance values than the adult group for frequencies below 2000 Hz (mean of 2 dB). No differences were observed for the mastoid between the young and adult groups. The mean impedance for the older group was 4 dB lower than for the adult group for the forehead. For the mastoid, mean impedance values for the older group were 6 dB lower than the adult group (Flottorp & Solberg, 1976). This is the only study to date that measured impedance magnitude for children; however, based on the literature from skull development, most of the development in the skull, skin, and subcutaneous tissue have reached adult levels by the age of 9 years (e.g., G. H. Sperber, Guttman, & Sperber, 2001).

1.7.1.2 Contact force and area

An early investigation reported no changes in stiffness or mass values when the coupling force was increased from 500 to 1000 grams (Corliss & Koidan, 1955). Although threshold variation with contact force has been extensively researched (Lau, 1986; Von Békésy, 1960; Watson, 1938; Whittle, 1965; Yang, Stuart, Stenstrom, & Hollett, 1991), Cortes (2002) was the first to measure mechanical impedance with multiple contact forces. She found that the mechanical impedance of the mastoid increased with increasing contact force until 6 N, after which the impedance plateaus.

Although no study has measured the effect of contact area directly on measures of mechanical impedance, Håkansson, Carlsson and Tjellström (1986) used a smaller disk area compared to previous studies (e.g., Corliss & Koidan, 1955; Flottorp & Solberg, 1976) and found some discrepancies in their measurement, which they attributed to the differences in applied contact force. They found a lower mass value with a smaller disk. The mass is attributed to the amount of skin and subcutaneous tissue directly under the plate. Therefore, a lower mass would be expected. A smaller volume of contact skin should amount to a larger compliance value, which was also observed. These results correspond well to differences in threshold for high frequencies when only contact force was varied (Khanna, Tonndorf, & Queller, 1976). Håkansson, Carlsson and Tjellström (1986) also found lower resistance magnitude. They predict that with fewer "particles" undergoing a velocity change due to vibration, smaller values of resistance should be recorded. The difference in their results compared to other studies are in line with the use of a smaller contact area.

1.7.1.3 Skull location

Corliss and Koidan (1955) measured the impedance at the mastoid and forehead and calculated similar values of stiffness, mass and resistance for each location. Smith and Suggs (1976) found negligible variations between frontal and occipital locations. Flottorp and Solberg (1976) also found little variation between the mastoid and forehead for low frequencies. The forehead impedance was slightly lower than the mastoid impedance across frequency, but significant differences were only found between locations at 1500 and 2000 Hz. These authors noted that the placement of the disk on the forehead was completed without any contact issues, while placement on the mastoid contained more difficulties with achieving contact over the entirety of the disk; therefore, interpretations of differences between locations should be made with caution.

1.7.2 Skull

As described in a previous section, sensitivity differences are observed between percutaneous and transcutaneous transmission of bone-conduction stimuli. Before the development of the BAHS, Franke (1956) explored mechanical impedance with the purpose of understanding mechanical vibrations of the skull. He measured impedance of the skin-covered skull on live subjects and impedance of the skull on cadavers in order to obtain better coupling. He found that the reactance for the cadaver skull was substantially higher than for the skin-covered skull due to its stiffness component.

Later, Tjellström et al. (1980) developed a model for skin and skull impedance, and took preliminary measurements of the impedance of the skull through the abutment. By comparing their result to those of Flottorp and Solberg (1976), they discovered that the mechanical impedance of the skull was 10-25 dB greater than the impedance of the skin-covered skull measured by Flottorp and Solberg (1976). Håkansson, Carlsson and Tjellström (1986) followed-up by more thoroughly measuring both skin-covered skull and skull impedance for a range of frequencies on the same subjects. The shape of the skull impedance function was different from the impedance of the skin-covered skull. The impedance generally increased at low frequencies until reaching one or two antiresonances between 100 and 350 Hz. Then, the impedance decreased through to 10,000 Hz, indicating a stiffness-dominated system. Comparing skin-covered skull and skull impedances, the authors concluded that the skull impedance is much higher than the skin impedance (10-30 dB). Significantly lower velocity of the transducer was needed to drive the signal to a certain acceleration when attached to the abutment. Measurement values collected through Håkansson Carlsson and Tjellström (1986) have been replicated with good agreement using updated methodology and equipment (Woelflin, 2011).

1.7.2.1 Group differences

A thesis completed by Woeflin (2011) investigated group differences in percutaneous mechanical impedance with the abutment. Specifically, he looked at differences in impedance across age, and among individuals with a history of major ear surgery (e.g., mastoidectomy) or craniofacial abnormalities. He found that the skull impedance for older individuals (50-80 years) is significantly greater than for younger individuals (18-49 years) between 200-600 Hz. This is likely due to the stiffening and calcification of the skull as individuals age. Subjects who had undergone major ear surgery had significantly lower skull impedance for frequencies between 220 and 1100 Hz. One explanation is that the surgical procedure altered the properties of the adjacent bone, causing a decrease in stiffness. Finally, no significant differences were found between those with and without craniofacial abnormalities. This is the only report to date on group differences for the impedance magnitude of the skull.

1.7.3 Comparison of skin-covered skull and skull impedance

Using a combination of impedance and sensitivity data, researchers have made some predictions on the contributing factors of the skin, subcutaneous tissue, and skull to bone-conduction stimuli. Modeling these factors can help researchers better understand bone-conduction transmission of sound.

First, because the mechanical impedance of the skull is substantially higher than the mechanical impedance of the skin, it is likely that the properties of the skin and subcutaneous tissue determine the impedance measurement of the skin-covered mastoid (Håkansson, Carlsson, & Tjellstrom, 1986; Tjellström et al., 1980). Tjellström et al. (1980) predicted that variations in measurements by age observed by Flottorp and Solberg (1976) are likely due to the development of the skin and subcutaneous tissue.

The high frequencies for skin-covered skull mechanical impedance are dominated by mass. The difference curves calculated from acceleration threshold for transcutaneous and percutaneous signal delivery indicate an increasing difference between the two measurements for frequencies 4000 Hz and higher. Therefore, Håkansson, Tjellström and Rosenhall (1985) predicted that the outer skin mass is contributing to the attenuation of the transcutaneous signal corresponding to the increase in impedance magnitude after this point.

Researchers have suggested a cascade model to describe skin-covered skull and skull mechanical impedance. In their model, acceleration delivered to the skin-covered skull interacts with the compliance and resistance of the skin and subcutaneous tissue. The resulting force is then delivered to the skull and is representative of hearing sensitivity (Håkansson, Tjellström, & Rosenhall, 1984; Håkansson, Tjellström, & Rosenhall, 1985; Håkansson, Carlsson, & Tjellstrom, 1986). Using magnitude values for both skull and skin impedance and an analytic network program (ANP-3), researchers have been able to model skin and skull impedance independently and find good correspondence between the two models (Håkansson, Tjellström, & Rosenhall, 1985; Håkansson, Carlsson, & Tjellstrom, 1986).

1.7.4 Development of the skull simulator

The artificial mastoid represents the mechanical impedance of the skin-covered mastoid and does not represent the mechanical impedance of the skull, which is several magnitudes higher than the skin-covered mastoid. Therefore, the artificial mastoid is not useful for electrovibrational verification of the BAHS when applied to the abutment. In contrast, the skull simulator is useful for measuring the output from the BAHS when it will be coupled to the abutment. Håkansson and Carlsson (1989) reported that the impedance calculated from the transducer of the BAHS is significantly smaller than the impedance of the skull. Therefore, the impedance value of the skull is irrelevant as long as it is substantially larger than the impedance of the transducer. This is denoted by the criterion Z_L (load impedance) >> Z_M (transducer output impedance), where $Z_L = Z_T$ (skull impedance). In order to fulfill this criterion, the skull simulator has a rigid mass of 50 g insulated by springs. The mass is critical in that a higher mass would be impractical in terms of portability, but any mass lower than 50 g would conflict with the impedance criterion.

As mentioned, the criterion for the skull simulator is based on findings that the mechanical impedance of the skull is substantially higher than the impedance of the transducer; however, this is not the case for the impedance of the skin-covered skull. Only a couple studies have investigated the effect of the mechanical impedance of the skin-covered skull on the force output from the artificial mastoid. Flottorp and Solberg (1976) mentioned that the variation in mechanical impedance among their adult subjects produced a maximum deviation of 5-6 dB of force output at the resonant frequencies of the transducer, specifically 500 and 1000 Hz, in their case. Similarly, Lundgren (2010) measured the impedance of the B-71 bone-conduction transducer and measured variability in force and acceleration output based on the standard deviation in mechanical impedance for adult subjects. He found a standard deviation in force and acceleration output of 0-5 dB depending on the frequency, which is based on variability in mechanical impedance and resonant properties of the bone-conduction transducer. Stenfelt and Håkansson (1998) created a miniature artificial mastoid that can be used in conjunction with the skull simulator to accurately measure the force output for a transcutaneous BAHS. The mechanical impedance magnitude was in good accordance with the IEC (1990) standards of mechanical impedance of the skin-covered mastoid; however, phase measurements were outside of the standard tolerances. Output force measurements were consistent with the Brüel & Kjaer type 4930 artificial mastoid, and therefore their device could be a means for verifying the BAHS in the hearing aid analyzer testbox for soft band devices. An important consideration is that all values used to create tools for BAHS verification, such as the artificial mastoid and skull simulator, have used average adult measurements.

1.7.5 Force and acceleration

As discussed, measurements of the BAHS *in situ* use an accelerometer to measure acceleration (Hodgetts, Håkansson, Hagler, & Soli, 2010); however, all measurements through the skull simulator measure force output. The distinction between these two measurements is important for understanding the measurement of mechanical impedance. Håkansson and Carlsson (1989) noted that because of the difference between the magnitudes of the impedance for the skull and the skull simulator, only force output can be computed from the BAHS through the skull simulator. Unlike force, acceleration is highly dependent on the skin and subcutaneous tissue and the status of the coupling condition. This explains how acceleration is directly affected by change in contact area of the bone-conduction transducer, while force is not affected (Håkansson, Tjellström, & Rosenhall, 1985). Velocity is incorporated into the measure of acceleration, and therefore, low measures of acceleration equate to low measures of velocity, gain, distortion, and power consumption (Håkansson, Tjellström, & Rosenhall, 1984; Håkansson, Tjellström, & Rosenhall, 1985). Acceleration is also sensitive to individual differences in mechanical impedance.

Force measurements are only slightly affected by the skin and subcutaneous tissue. The difference in threshold between percutaneous and transcutaneous stimulus methods was found to be only approximately 10 dB when measured in force (Carlsson, Håkansson, & Ringdahl, 1995). However, force is more sensitive to movement artefacts from the patient than acceleration, and the mass of the load must be compensated in the force measurement. If the mechanical impedance (Z) and acceleration (A) are known, force (F) can be calculated with the equation $F = |Z| \times A/\omega$. Mechanical impedance can also be calculated from measures of force and acceleration using this equation (Håkansson, Tjellström, & Rosenhall, 1985).

The importance of collecting mechanical impedance values is twofold. First, they provide valuable insights to the transmission of a bone-conducted signal through the skin to the underlying bone. Second, verification protocols for BAHS will soon require the use of a skull simulator or artificial mastoid. The initial step to determining the appropriateness of a verification tool for infants and young children is to quantify the mechanical impedance values of the immature skull.

1.8 Rationale for Thesis

The purpose of this thesis is to investigate the properties of the skull for infants and young children. The study aims to investigate the transmission of sound across the maturing skull through a physical measurement of sound pressure in the ear canal, to isolate factors such as the brainstem, cochlea, and middle and outer ears from a measure of transcranial attenuation. In addition, results of this thesis also aim to fill in a substantial gap in the literature on the maturation of bone-conduction hearing by collecting mechanical impedance values for infants and young children, a population that was neglected from studies during early investigations of mechanical impedance of the skin-covered skull (Flottorp & Solberg, 1976). These findings will help researchers better understand the maturational differences in sensitivity to bone-conduction stimuli and better explain the mechanisms responsible for bone-conduction hearing. Additionally, studying the properties of the maturing skull will help contribute to better fitting and verification protocols for BAHS soft bands for infants and young children. Specifically, it is important to understand how verification tools, such as the artificial mastoid and skull simulator, which were developed with adult skull properties in mind, should be used to verify the BAHS soft band for infants and young children. Finally, results of the attenuation of bone-conducted sound across the skull will allow researchers and clinicians to better understand how infants and young children with unilateral conductive hearing loss and single-sided deafness will benefit from a BAHS, and how infants and young children will benefit from a bilateral BAHS fitting.

First, properties of transcranial attenuation of bone-conducted sounds were analyzed using a measure of sound pressure in the ear canal. Measurements were collected when the bone-conduction transducer was placed on the temporal bone ipsilateral to the test ear, contralateral to the test ear, and while on the forehead of each individual. Four audiometric frequencies were analyzed (500, 1000, 2000, and 4000 Hz). Secondly, the mechanical impedance of the skin-covered skull was investigated. Magnitude and phase values were collected for frequencies 100-10 000 Hz, for measurement made at the mastoid and forehead. Three different contact force levels were investigated (2, 4, and 5.4 N). Participants ranged in age from 1 month to 7 years. An adult group was included to investigate infant-adult differences for all measurements made.

As described, previous studies have shown significant differences between infants and adults for hearing sensitivity to bone-conducted sounds as well as the structure of the skull and anatomy and physiology of the auditory system. However, explanations regarding the underlying mechanisms for these differences in bone-conduction hearing are limited. In addition, the BAHS is available for young children and infants in the form of a transcutaneous soft band device; however, the current fitting and verification protocol for the BAHS is not precise and should be improved to offer the same standard that is used to fit air-conduction hearing aids. New verification protocols require the use of tools with the properties of an average adult skull in mind, and it is undetermined whether these devices are useful for infants and young children who have significant differences in their skull and skin anatomy. This study is an important first step to solving these quandaries through an investigation of the transmission of bone-conducted sound across the developing skull and the measurement of the mechanical impedance of the skin-covered skull for infants and young children.

CHAPTER 2: Maturation of Skull Properties with Implications for the Fitting and Verification of the Soft Band Bone-Anchored Hearing System for Infants and Young Children

2.1 General Methods

Two separate experiments were conducted within this study. In Experiment 1, the transcranial attenuation of bone-conducted sound was compared for infants, young children, and adults using measures of sound pressure in the ear canal. In Experiment 2, the maturation of the mechanical impedance of the skin-covered skull was investigated. The same participants were tested in both experiments and will be described in detail in the General Methods section. Procedures common to both experiments will also be discussed in the General methods section. A description of the methodology and results for each individual experiment will be provided in sections 2.2 and 2.3.

2.1.1 Participants

Eighty-two individuals participated in the study, including 65 infants and children (mean age: 30.8 months; range: 1-88 months; 36 female) and 17 adults (mean age: 25.5 years; range: 20-32 years; 14 female). All participants involved in the study participated in both experiments; however, not all participants completed all conditions. The details regarding participant exclusion are described in detail in Appendix A. All participants or participants' parent/guardians reported no history of major ear surgery or craniofacial abnormalities, with the exception of one child who had cochlear implantation surgery in both ears. Participants were classified into the following five groups according to age ranging from young infants to school-aged children: Group A: 1-10 months (n=18), Group B: 11 months-2 years (n=13), Group C: 2-4 years (n=15), Group D: 4-7 years (n=19), and an adult comparison group (n=17). Because of the rapid development in middle ear status and higher prevalence of otitis media with effusion within the first year of life (Paradise et al., 1997), it is important to note that among the 18 infants in Group A, 10 were younger than 6 months of age. Participants were recruited from the community using posters and an e-mail message. Otoscopy was performed on the test ear for each participant. If cerumen was determined to interfere with the procedure for the test ear, the other ear was examined. No participant was excluded due to cerumen impaction. A GSI-38 automatic tympanometer was used for screening middle-ear status. Sixty participants had static admittance and middle ear pressure within a normal range. Two children did not complete the screening. One child had significant negative pressure in the test ear due to recovery of a known ear infection. Two children presented with flat tympanograms for their test ear.

2.1.2 Procedure

Experiment order was counterbalanced and pseudo-randomized. In instances where children were not comfortable completing one experiment, the other was attempted. Additionally, for children who participated with siblings, alterations to the randomly assigned experiment order were often made to keep children interested in the study. The study procedures were approved by UBC Clinical Research Ethics Board. Participants or their parent or legal guardian signed a consent form outlining the procedure for both Experiment 1 and 2 before testing commenced. Participants were requested to dedicate approximately 30 minutes of their time to participation in both parts of this study. When testing was complete, participants or their parent/guardians were provided an honorarium. Testing was completed in the Pediatric Audiology Lab at the University of British Columbia in the School of Audiology and Speech Sciences.

2.2 Experiment 1: Transcranial Attenuation of Bone-Conducted Sound

2.2.1 Methods

2.2.1.1 Materials and calibration

A Fonix 6500-CX real-ear analysis system, probe-tube assembly, and foam earplug were used to take measurements of sound pressure in the ear canal. The bone-conducted stimulus presented at the skull was generated using a GSI-16 audiometer with Radioear B-71 bone oscillator.

Stimulus calibration was conducted using a B&K Mastoid 4930 artificial mastoid and a Larson and Davis system 824 sound level meter. The transducer was coupled to the artificial mastoid with 5.4 N of force, and 0 dB HL was calibrated to the Reference Equivalent Threshold Force Levels (RETFLs) in dB re: 1 μN. Stimuli were presented with frequencies of 500, 1000, 2000, and 4000 Hz at intensities of 50, 50, 50 and 60 dB HL respectively. A stimulus intensity of 60 dB HL was selected due to an interaction with the noise floor observed during the pilot testing with adults.

2.2.1.2 Procedure

Testing was completed in a double-walled sound attenuated booth. Average ambient noise levels measured in the booth with a Larsen Davis System 824 for one-octave wide bands centered at 500, 1000, 2000, and 4000 Hz were 1, 3, 6, and 7 dB SPL, respectively. Infants and children were seated in the lap of their parent/guardian. Silent toys or movies on a video monitor were used to maintain attention. A probe-tube was inserted into one of the participant’s ears. The test ear was typically randomly assigned; however, in some cases, the opposite ear was selected due to the presence of excessive cerumen, a myringotomy tube, or intolerance to insertion of the probe tube in that ear. The probe tube was inserted following recommendations from Bagatto (2001) and Bagatto, Seewald, Scollie, and Tharpe (2006). Both sources recommended leaving 5 mm between the end of the probe tube and the ear drum. Bagatto (2001) recommended an insertion depth of 28 mm for adult females, 31 mm for adult males, and 15-25 mm for children. Bagatto, Seewald, Scollie, and Tharpe (2006) recommended inserting the probe tube approximately 11 mm into the ear canal for infants. Once the probe tube was inserted, the canal was left unoccluded or occluded depending on the test frequency. First, a measure of sound pressure (in dB SPL) was recorded when no stimulus was presented to determine ambient noise in the ear canal. For each participant, the transducer was positioned on three locations on the skull. First, it was placed on the high temporal bone ipsilateral to the test ear, posterior and slightly anterior to the top of the pinna; second, on the middle of the forehead; and third, on the contralateral temporal bone, similar position to the ipsilateral temporal bone). The bone-conduction transducer was held in place by hand at a contact force of 400-450 g. Researchers were trained to hold the oscillator pressing down with their fingertips on the top of the transducer at this contact force by selfmonitoring using a compressive spring scale (Small, Hatton, & Stapells, 2007). A measurement of sound pressure in the ear canal was captured for stimuli at each frequency and skull position.

The experimental conditions were the same for all stimulus frequencies; however, for 500 and 1000 Hz, the ear canal was unoccluded, whereas the ear canal was occluded for 2000 and 4000 Hz. Studies have shown that bone-conduction hearing at 4000 Hz is confounded to some degree due to the airborne radiation from the bone-conduction transducer (Lightfoot, 1979). Small and Hu (2011) also found that sound pressure in the ear canal at 2000 Hz is significantly lower in an occluded ear than an unoccluded ear with the bone-conduction transducer positioned on the ipsilateral mastoid for adults, which is consistent with air-conduction radiated sound contributing to an unoccluded measure of sound pressure in the ear canal. It was therefore deemed prudent that for 2000 and 4000 Hz, measurements should be made with an occluded ear to gain a more accurate measure of bone-conduction sound generated in the ear canal originating from vibration of the skull.

2.2.1.3 Analysis

To achieve a measure of transcranial attenuation of bone-conducted sound, attenuation was calculated by subtracting the sound pressure measured at the forehead and contralateral temporal bone from the sound pressure measured at the ipsilateral temporal bone. Data from the children with abnormal or incomplete tympanograms (n=5) were included in the analysis because their individual data were consistent with the pattern of results observed through box plots. Ambient sound pressure in the ear canal was evaluated for each age group and condition. One subject was excluded from analyses due to high ambient noise, as noted in Appendix A. A mixed-model analysis of variance (ANOVA) was performed to compare one between-subject variable (age group) and a multivariate analysis of variance (MANOVA) was performed to compare two within-subject variables (transducer position and frequency). A MANOVA was chosen over a mixed-model ANOVA for the within-subject variables because the assumption of sphericity was violated, and a MANOVA is recommended over non-parametric tests when this assumption is violated (Hill & Lewicki, 2006). The transducer position variable contained two factors: (i) ipsilateral temporal bone–forehead and (ii) ipsilateral–contralateral temporal bone. The frequency variable contains four factors: (i) 500 Hz, (ii) 1000 Hz, (iii) 2000 Hz, and (iv) 4000 Hz. As mentioned previously, five age groups were included in the analysis. An alpha of 0.05 was used as the criterion for statistical significance. A conservative Bonferonni approach was used to perform multiple comparison post hoc analyses in order to compare contrasts for significant main effects and interactions.

2.2.2 Results

The mean transcranial attenuation for each experimental condition is displayed in Figure 2.1. Notably, a maturational effect of transcranial attenuation was shown through decreasing attenuation with an increase in age category (i.e., Group A through Adult). Results of a mixed-model ANOVA revealed a significant main effect of age group [$F(4,71) = 4.25$, $p = 0.003$]. *Post hoc* comparisons and their significance levels for all significant main and interaction effects, including the main effect of age, are presented in Table 2.1. These results indicated significantly less attenuation for adults than for Group A (1-10 months) and Group B (11 months-2 years). Interaction effects were also analyzed through the Hotelling's T MANOVA. Results indicated a significant interaction between transducer position and age group [$F(4,71) = 2.99$, $p = 0.024$]. As shown in Table 2.1, for the ipsilateral–forehead condition, attenuation was significantly lower for adults compared to Group A and Group B, and for ipsilateral–contralateral attenuation, the comparison between adults and Group A approached significance, but no other ipsilateral-contralateral age group comparison was significant.

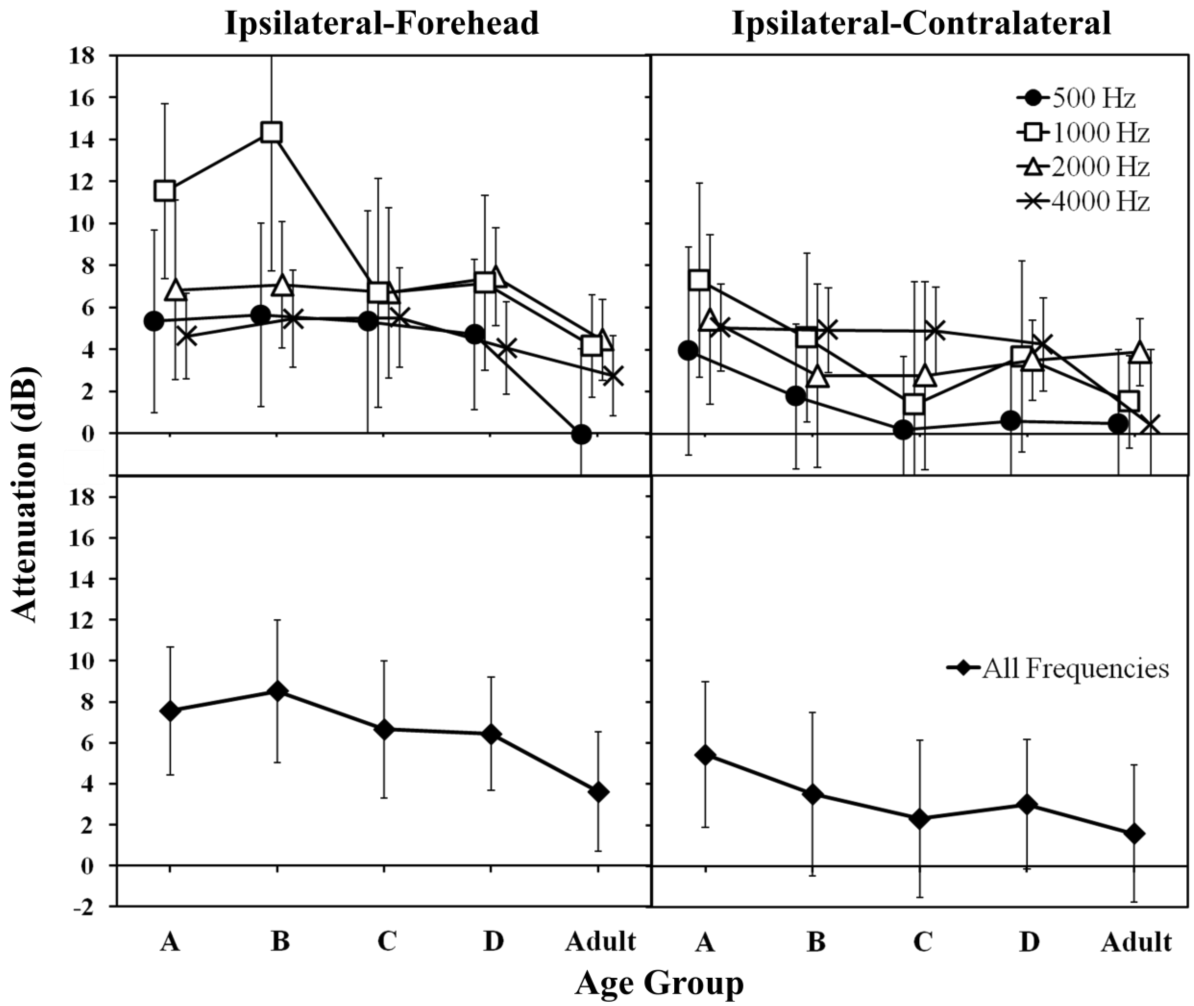

Figure 2.1 Attenuation of bone-conducted sound across the skull detailed by comparisons of sound pressure in the ear canal when bone-conduction stimuli are presented at different positions on the skull for each age group. Error bars represent 95% confidence intervals around the mean.

Table 2.1 Attenuation of bone-conducted sound *post hoc* analyses *p*-values for the main effects of age group and frequency, and the interactions of age group-by-position and frequency-by-position.

		Main Effect	Position: Ipsilateral-Forehead (I-F)	Ipsilateral-Contralateral (I-C)	I-F vs. I-C
Age Group					
Group A					0.23
	vs. B	1.0	1.0	1.0	
	vs. C	0.72	1.0	0.57	
	vs. D	0.84	1.0	1.0	
	vs. Adult	0.003*	0.04*	0.05†	
Group B					<0.001*
	vs. C	1.0	1.0	1.0	
	vs. D	1.0	1.0	1.0	
	vs. Adult	0.03*	0.005*	1.0	
Group C					<0.001*
	vs. D	1.0	1.0	1.0	
	vs. Adult	0.84	0.58	1.0	
Group D					<0.001*
	vs. Adult	0.32	0.47	1.0	
Adult					0.19
Frequency					
500 Hz					<0.001*
	vs. 1000 Hz	0.002*	<0.001*	0.03*	
	vs. 2000 Hz	0.3	1.0	0.03*	
	vs. 4000 Hz	1.0	1.0	0.02*	
1000 Hz					<0.001*
	vs. 2000 Hz	0.63	0.002*	1.0	
	vs. 4000 Hz	0.03*	<0.001*	1.0	
2000 Hz					0.004*
	vs. 4000 Hz	1.0	0.08†	1.0	
4000 Hz					1.0

A "*" indicates contrast is significant at $p<0.05$. A "†" indicates contrast is marginally significant at $p<0.1$.

Figure 2.1 also shows that the attenuation from the forehead position was greater than the attenuation from the contralateral temporal bone, as confirmed by the Hotelling's T MANOVA results [$F(1,71) = 140.94$, $p < 0.001$]. This effect is particularly evident in the graphs collapsed across all frequencies. Based on the interaction between age group and transducer position, *post hoc* analyses revealed that the ipsilateral–forehead condition had significantly more attenuation than the ipsilateral–contralateral condition for Groups B, C, and D, but not for Group A and adults. Finally, a significant interaction was revealed between transducer position and stimulus frequency [$F(3,69) = 8.88$, $p < 0.001$]. Interestingly, when collapsed across age groups, the ipsilateral-forehead condition showed significantly more attenuation than the ipsilateral-contralateral condition for 500, 1000 and 2000 Hz, but not for 4000 Hz (Table 2.1).

In addition, Hotelling's T results revealed a significant main effect of frequency [$F(3,69) = 4.74$, $p = 0.004$]. The *post hoc* results, as described in Table 2.1, indicated that 1000 Hz had significantly more overall attenuation than 500 Hz and 4000 Hz. Based on the interaction between transducer position and frequency, for the ipsilateral–forehead condition, the attenuation for 1000 Hz was significantly greater than for 500, 2000 and 4000 Hz, and the attenuation for 2000 Hz was greater than for 4000 Hz at a level that approached significance. For the ipsilateral–contralateral condition, 500 Hz had significantly less attenuation than 1000, 2000 and 4000 Hz. The interaction between frequency and age group was not significant [$F(12,203) = 0.60$, $p = 0.84$], nor was the 3-way interaction of frequency, position and age group [$F(12,203) = 1.29$, $p = 0.23$].

An additional comparison was made to account for any airborne radiation of sound from the bone-conduction transducer. A small sample of children (n = 4) and adults (n = 5) completed additional measures with the bone-conduction transducer held off the temporal bone on the ipsilateral side to determine if airborne sound may contribute to the measures of sound pressure in the ear canal through a head shadow effect. As mentioned, the ear canal was unoccluded for 500 and 1000 Hz and occluded for 2000 and 4000 Hz stimuli. Mean differences between ambient sound (i.e., no stimulus presented at the skull) and airborne radiation (i.e., stimulus is on but transducer is not in contact with skull) for 500, 1000, 2000, and 4000 Hz were 7.73, 19.76, 5.32, and 8.49 dB, respectively. Therefore, interpreting results from 1000 Hz should completed with some caution; however, any effects where attenuation for infants was greater than adults should not be attributed to a head shadow effect because an opposite trend would be expected due to adults having a larger head circumference.

2.3 Experiment 2: Mechanical Impedance of the Skin-Covered Skull

2.3.1 Methods

2.3.1.1 Materials and calibration

2.3.1.1.1 Equipment

Figure 2.2 illustrates the configuration of the equipment used to measure mechanical impedance. As shown in this figure, a BAHS transducer (Oticon Ponto) was wired to a 3.5 mm stereo plug, which, along with a B&K 8001 impedance head, was attached to a plastic holding device created at the iRSM institute for the purpose of this study. The device was constructed so that springs could be calibrated to a desired contact force and the examiner could monitor the amount of contact force applied to the skull. Springs were wrapped around three posts on the holding device that slide into the handle portion of the device. A contact plate with an area of 2.0 cm^2 was screwed into the top of the impedance head. The bottom of the impedance head and transducer were connected together with a 10-32 screw attached to an abutment. The BAHS transducer was snapped onto the abutment, which was screwed into the top of the impedance head. Two B&K 2647 A charge-to-deltatron amplifiers linked the impedance head to a NI-cDAQ analogue input module via two B&K mini coaxial cables. One cable carried acceleration data from the impedance head, while the other carried force data. A 3.5 mm stereo jack was connected to an output module with input cables connected to the BAHS transducer. The NI-cDAQ sent data to a laptop computer via a USB-A to USB-B cord to be analyzed by the BCAD Software Suite written in LabView by the iRSM group for the purpose of calculating acceleration, force, and mechanical impedance magnitude and phase values online.

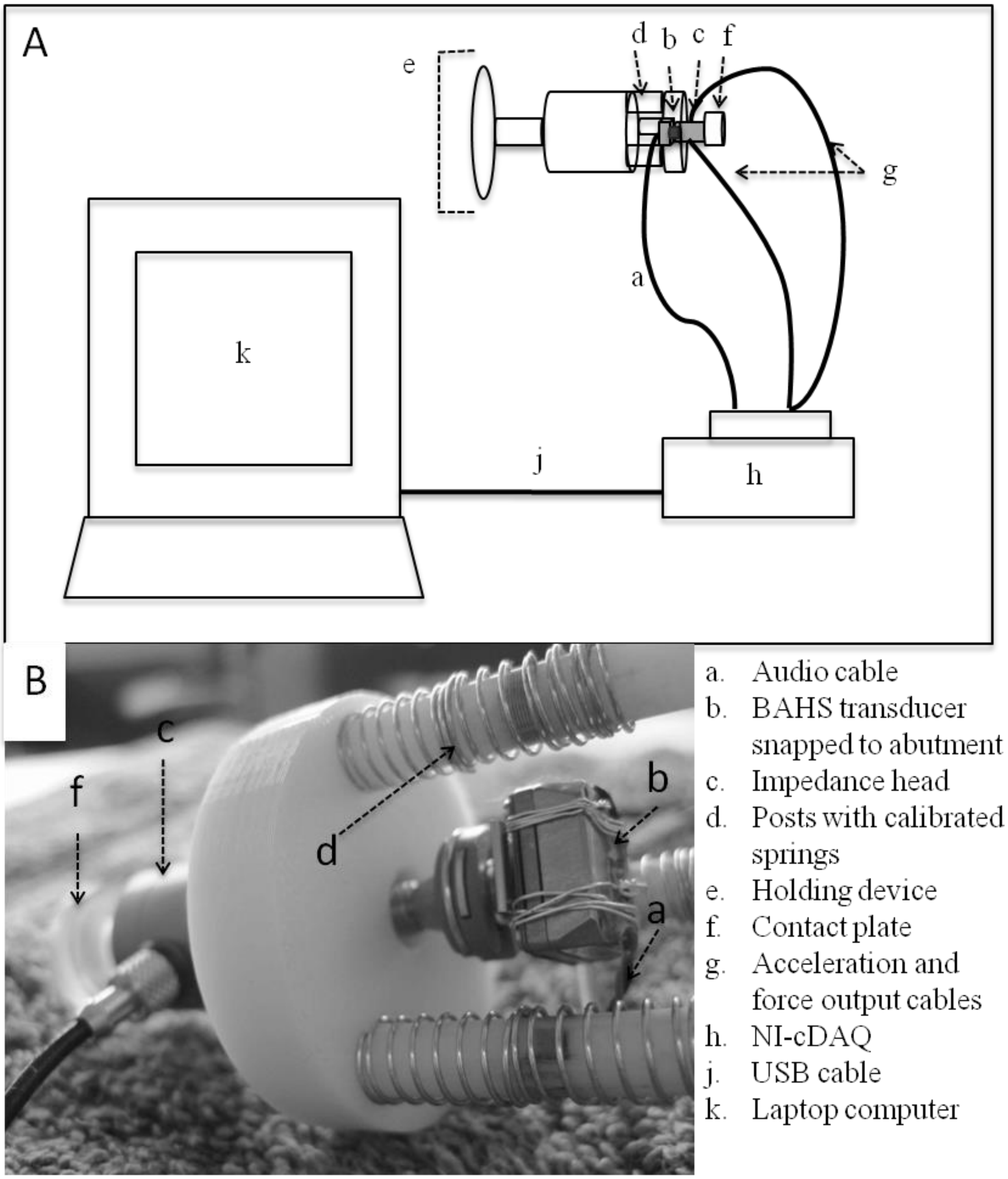

Figure 2.2 A. Diagram of the setup for measuring mechanical impedance. **B**. Picture of BAHS transducer with abutment, impedance head, and contact plate attached to the bottom portion of the holding device with the calibrated springs.

2.3.1.1.2 Calibration of contact force levels

Contact force magnitude was calibrated on the holding device by placing different springs along the posts of the holding device along with a small amount of oil for lubrication. A weight corresponding to the desired force was positioned between two tables. The holding device was then held underneath the weight and pressed upward. Springs were adjusted so that they compressed about half way down the holding device posts. The amount of force necessary to lift the weight off the table was noted by a marker drawn on the posts of the holding device. This procedure was repeated for each desired force, using different marker colours and spring combinations. A researcher was trained to hold the device against each participant's head so that the springs lined up to the appropriate marking, thus applying the necessary contact force. Only one researcher (A. Mackey) completed all measurements for all participants.

Three forces were calibrated. First, 5.4 N was selected because it is the ANSI (1996) coupling force standard for calibrating bone-conduction transducers, and also corresponds to the force used by adult studies of bone-conduction hearing and mechanical impedance. Second, 4 N was selected as the clinically recommended tightness for soft band users. Audiologists at the iRSM have recommended that parents tighten the soft band so that two fingers stacked one on top of the other can fit under the band. To determine the amount of contact force applied using this recommendation, the soft band force was measured using different sizes of Styrofoam balls to represented average head circumference for children of different ages. A soft band (Oticon) was wrapped around the largest Styrofoam ball with a circumference of 65.3 cm and tightened to fit the recommendations. A spring scale was used to measure the force of the band at this setting. Measurements were repeated on two smaller Styrofoam balls (circumferences 39.9 and 47.7 cm). These dimensions correspond roughly to the average head size of a 2 month old and 18 month old, respectively (Kuczmarski et al., 2002). Finally, the same measurement was completed on five adult heads, all of whom were members of the Pediatric Audiology Lab. For all measurements completed, 4 N matched audiologist soft band tightness recommendations. Additionally, 4 N falls into the 400-425 g coupling force range used for measuring physiological responses from bone-conduction stimuli with infants (e.g., Small, Hatton, & Stapells, 2007; Yang, Stuart, Stenstrom, & Hollett, 1991). Finally, a contact force value less than the clinically recommended soft band tightness was selected to predict the validity of BAHS verification measurements when the soft band is not tightened to recommendations. Measured with a spring scale, 2 N was the force provided by the soft band when the band was loosened considerably but did not fall off the head. Hodgetts, Scollie and Swain (2006) also use 2 N as their lowest contact force level on an artificial mastoid, equating it to a reasonably loose soft band. Calibration was performed before any testing commenced and mid-way through the testing phase.

2.3.1.2 Procedure

Participants were positioned either sitting in a chair, sitting in the parent/guardian's lap, or while being held over the guardian's shoulder. Woelflin (2011) described that vibratory energy is not carried below the level of the neck, and therefore, the recording can be taken in any comfortable position, as long as the participant's head is not positioned against another object. The holder was pressed either against the participant's flattest part of the high temporal bone, anterior and posterior to the top of the pinna, or on the forehead at a force level of 2, 4, or 5.4 N. Starting force and position were pseudorandomized and counterbalanced. In cases where the child was predicted to not sit still through the entire test session, 5.4 N on the temporal bone condition was completed first. In cases where the child did not tolerate the device placed against their forehead, the temporal bone was attempted. The researcher or parent/guardian's fingertips were used on the opposite side of the skull to reduce movement artefact. This does not interfere with the impedance measurements due to the decoupling of the skin of the skull to the tips of the fingers (Håkansson, Carlsson, & Tjellstrom, 1986). A sine-sweep of 201 logarithmically-spaced frequencies (100-10,000 Hz) was delivered via the BAHS transducer with 50 mV amplitude, corresponding to a vibratory force of 0.02 N. Each sweep took approximately 20 seconds to complete. For each frequency, the impedance head provided a measure of force and acceleration that was sent to the BCAD Software Suite for analysis. Mechanical impedance and phase were calculated online. The output text file from BCAD Software Suite included acceleration data (m/s^2), force data (N), and both phase (degree) and magnitude (Ns/m) components of impedance for each frequency measured.

The holding device was then moved to the second position and the process was repeated for the initial force. The procedure was repeated for the other two contact forces. In most cases, force level conditions increased step-wise and then decreased (e.g., if 2 N was completed first, 4 N was performed next, or if 5.4 N was completed first, 4 N was next). However, in cases when the researcher felt the participant might not complete all of the conditions, the force 5.4 N on the high temporal bone was completed immediately. In cases where large movement artefacts were observed, this condition was repeated if possible after all other conditions were completed. Nine adult subjects completed all conditions twice. Based on visual inspection from the graph of impedance magnitude-by-frequency, the two runs of the same condition appeared to overlap consistently for each participant, showing good test-retest reliability.

2.3.1.3 Analysis

Impedance values were logarithmically scaled using the formula 20×log(X/1) and represented in units dB re 1 Ns/m. Each sweep was inspected for movement artefacts. Large artefacts were removed and a linear formula was used to interpolate the missing data. Data from one adult was removed altogether due to excess noise from a loose connection during recording. Other data points that were excluded due to instances of noise where linear interpolation could not be completed are summarized in Appendix A. The number of frequencies replaced due to movement for all participants was noted. For the temporal bone, the number of data points removed for 5.4, 4, and 2 N were 33, 142 and 85 points, respectively. For the forehead, the number of data points removed for 5.4, 4, and 2 N were 113, 148 and 220, respectively. Resonant frequency was determined manually for each sweep. Resonant frequency was defined as the lowest impedance magnitude before values began to increase steadily as frequency increased. As described in Chapter 1, measures of mechanical impedance are driven by different physical properties across the frequency sweep. In the case of the skin-covered skull, frequencies below the resonant frequency are driven by the stiffness of the system and frequencies above the resonant frequency are driven by the mass of the system. Therefore, it was not reasonable to include the whole sweep in a single analysis. Each frequency sweep of impedance magnitude data was divided into low and high frequency sets based on the mean resonant frequency. Low frequency sets included data from 100 to 1000 Hz. High frequency sets included data from 1995 to 10 000 Hz.

Impedance magnitude data for high and low frequency sets were analyzed using mixed-model analysis of variance (ANOVAs) and multivariate analysis of variances (MANOVAs). Due to the correlated nature of the data, the assumption of sphericity was violated for repeated-measures in both high- and low-frequency data sets. Therefore, results of the MANOVAs were used to compare within-subject factors. Within-subject variables included contact force, position, and frequency. One requirement for running a MANOVA is that the number of variables is less than the sample size, which is not held true for this analysis if considering each frequency measured as a separate variable. Therefore, frequencies closest to audiometric inter-octave frequencies were selected from each set. Five frequencies were analyzed for low frequency data (125, 251, 501, 758, and 1000 Hz). Six frequencies were analyzed for high frequency data (1995, 3019, 3981, 6025, 7943 and 10 000 Hz). The contact force variable contained three factors: (i) 5.4 N, (ii) 4 N and (iii) 2 N. The position variable contained two factors: (i) temporal bone and (ii) forehead. The between-subject variable was age group with five factors: (i) Group A (0-10 months), (ii) Group B (11 months-2 years), (iii) Group C (2-4 years), (iv) Group D (4-7 years), and (v) adults. An alpha criterion of 0.05 was used for statistical significance. Conservative Bonferonni multiple comparison *post hoc* analyses were completed on significant main effects and interactions. Resonant frequency and the impedance magnitude at the resonant frequency were analyzed with the same independent variables. Mixed-model ANOVAs were calculated for resonant frequency and impedance magnitude at resonant frequency, and multiple comparison Bonferonni *post hoc* analyses were completed on significant main effects and interactions.

2.3.2 Results

Figures 2.3, 2.4 and 2.5 display the mean impedance magnitude across the full frequency sweep for each variable of interest (age group, skull position and contact force). Each main effect is displayed in the graph collapsed across the other two variables. All three graphs display the same trend across frequency. Impedance magnitude decreases as frequency increases until the resonant frequency is reached. The resonant frequencies is the lowest impedance magnitude value on the graph. After this point, impedance magnitude increases as frequency increases. Based on these graphs, different effects between and among conditions were observed for low and high frequencies (below and above the resonant frequency); therefore, the data were partitioned into low- and high-frequency segments, which was necessary to fit a general linear model. The effect of frequency within each partitioned data set was not a variable of interest in the analyses; any trends across frequency can be seen in Figures 2.3, 2.4 and 2.5. The resonant frequency properties, including the resonant frequency itself and the impedance magnitude at resonant frequency, were analyzed separately for the variables age group, skull position and contact force.

Figure 2.6 provides a representation of the mean impedance magnitude for age group, skull position and contact force combined across the five low frequencies and six high frequencies selected for the analyses. Table 2.2 summarizes the statistical findings for both high- and low-frequency data sets. Specifically, for the low frequencies, all main effects and 2-way interactions for the selected variables were significant. For the high frequencies, the main effect of skull position and contact force was significant, as were the interactions between age and skull position and between contact force and skull position. The main effect of age group approached significant. Results from the analyses of mechanical impedance are described in detail in the following sections corresponding to low- and high-frequency data sets, followed by the results from the resonant frequency analyses.

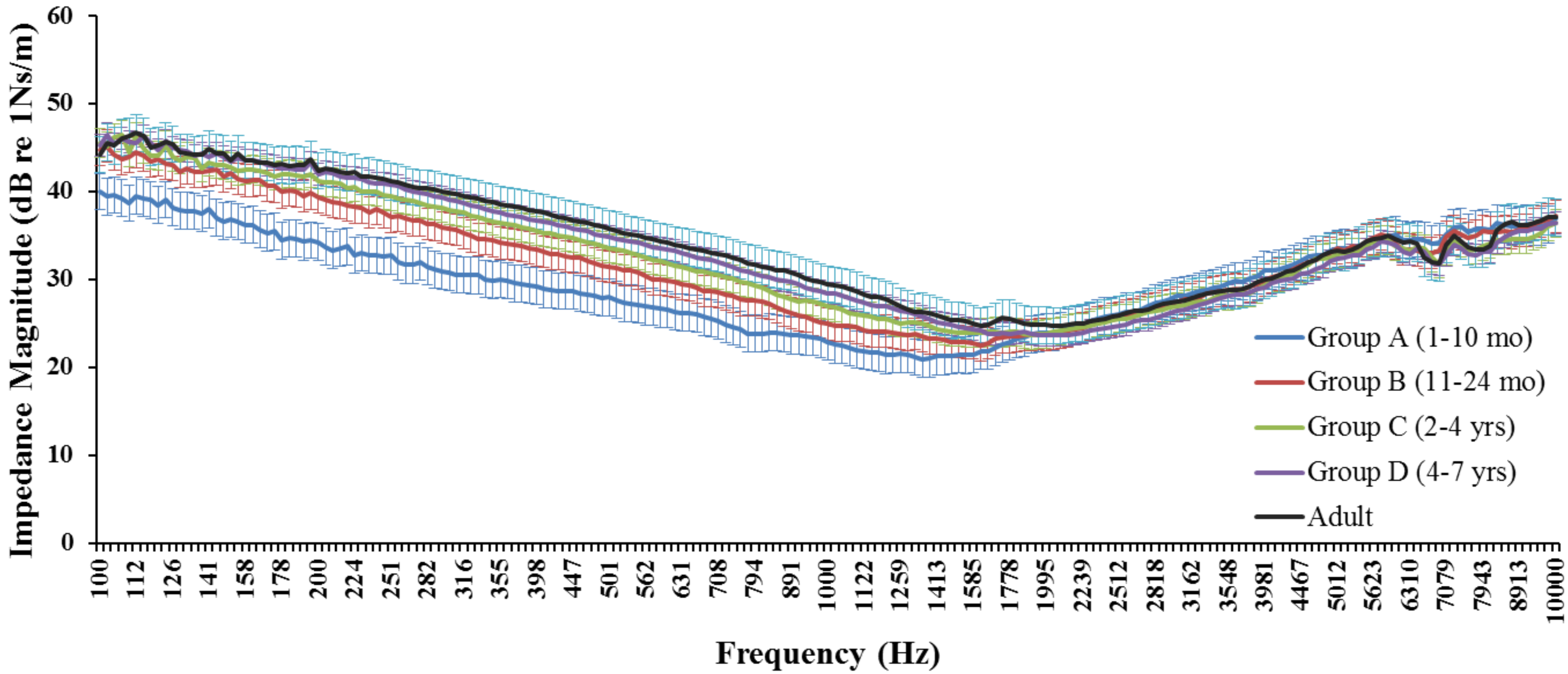

Figure 2.3 The mechanical impedance magnitudes for each age group across a 201 frequency sweep. Error bars represent 95% confidence intervals around the mean.

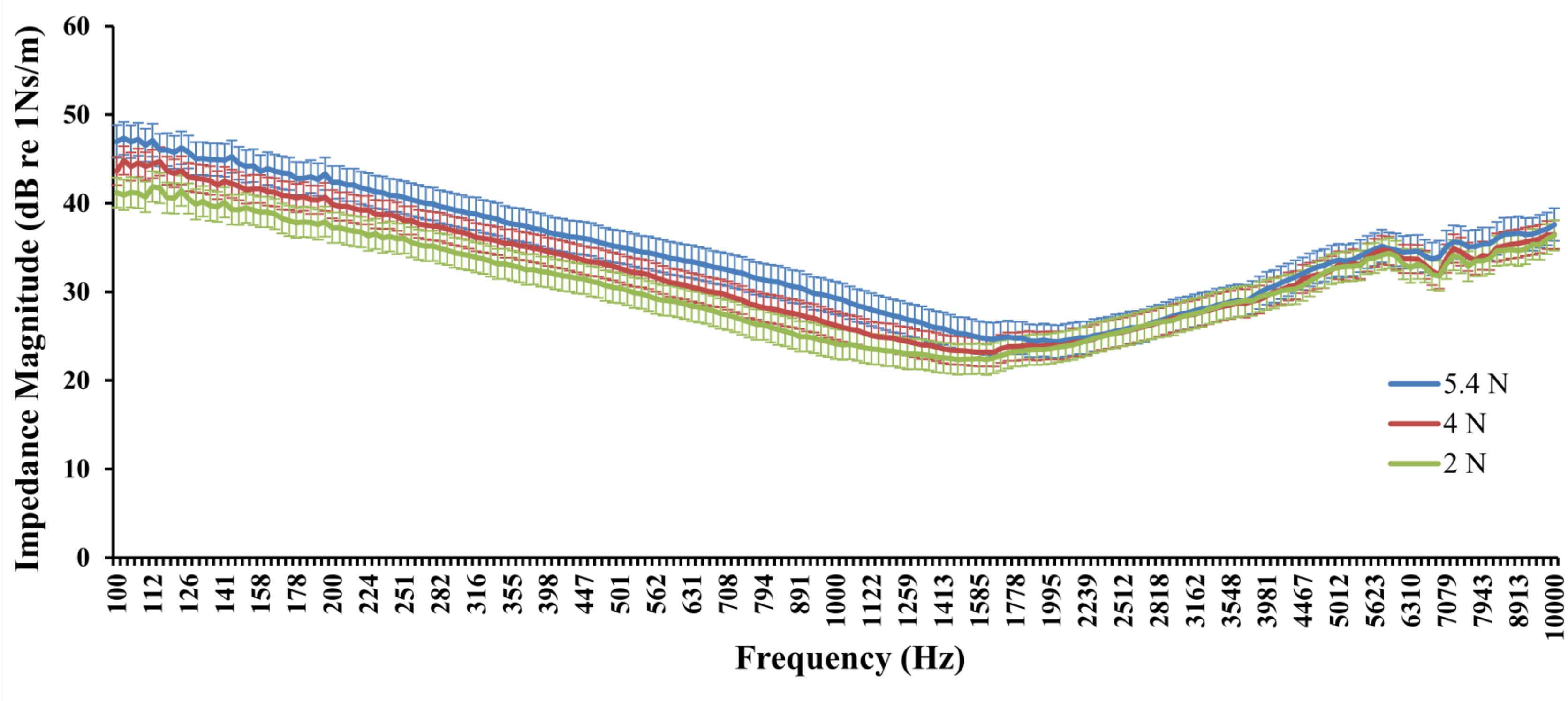

Figure 2.4 The mechanical impedance magnitudes for each contact force across a 201 frequency sweep. Error bars represent 95% confidence intervals around the mean.

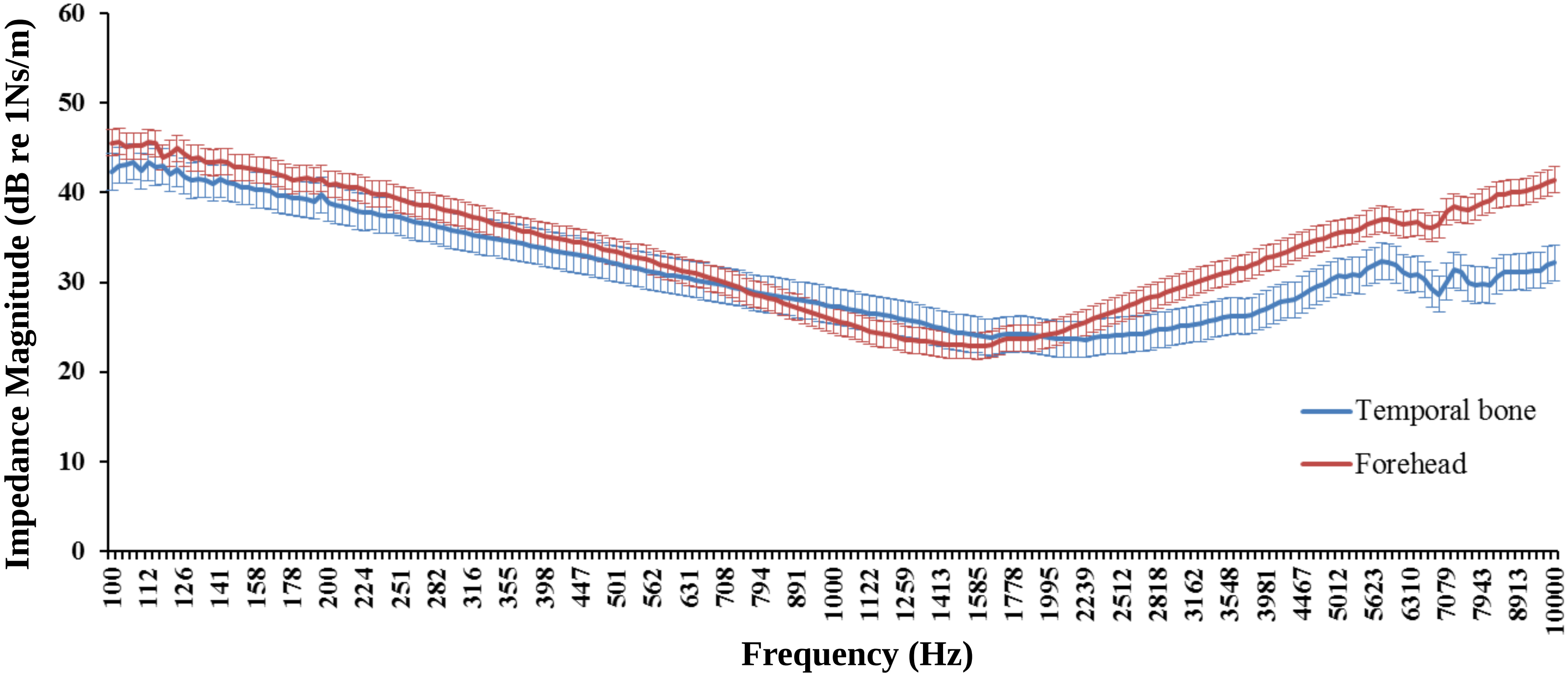

Figure 2.5 The mechanical impedance magnitudes for each skull position across a 201 frequency sweep. Error bars represent 95% confidence intervals around the mean.

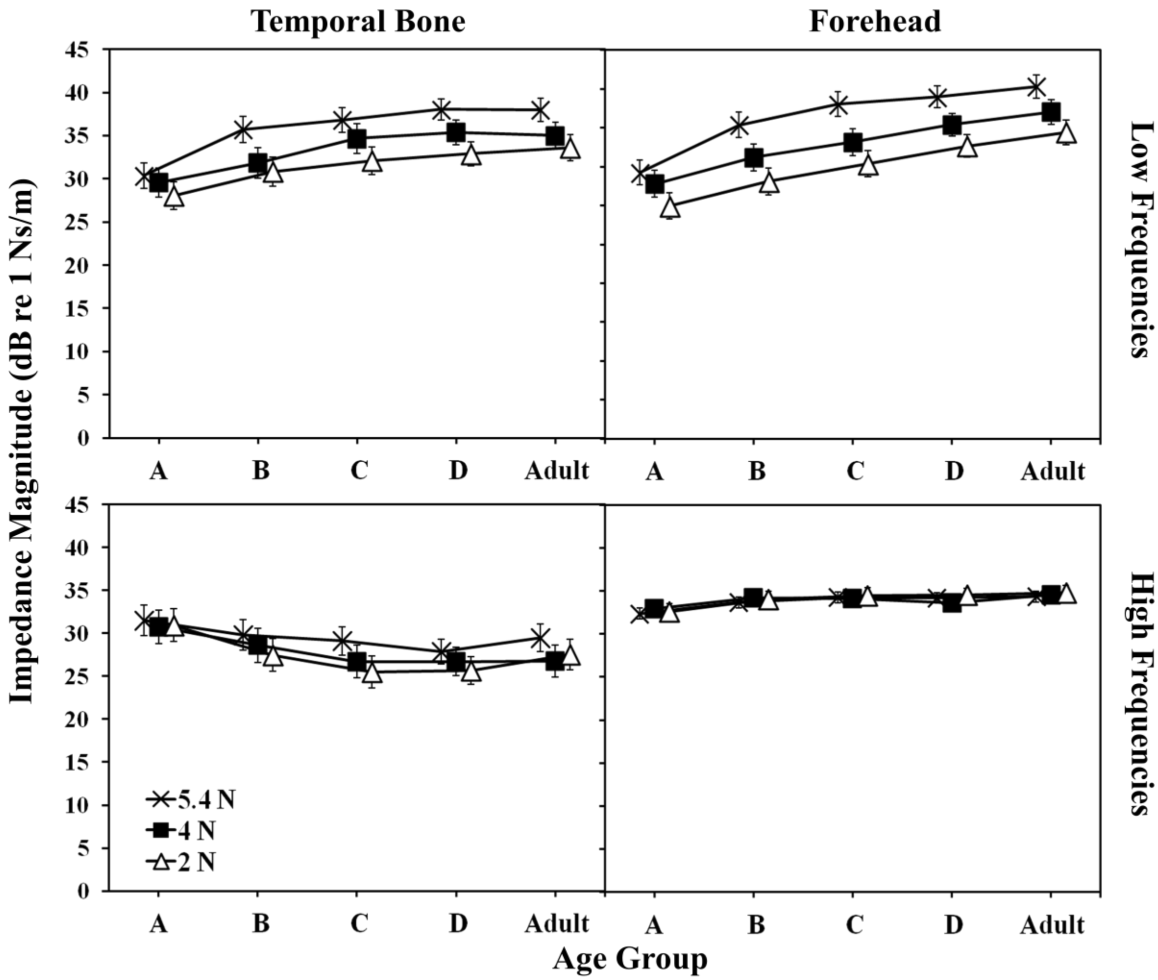

Figure 2.6 Mean impedance magnitude values measured at the temporal bone and forehead for each age group and contact force, collapsed across selected frequencies for low- and high-frequency data sets. Error bars represent 95% confidence intervals around the mean.

Table 2.2 Results from statistical analyses for the selected variables from mechanical impedance magnitude for both high- and low-frequency data sets.

	Low Frequency			High Frequency		
	F	**df**	**p**	**F**	**df**	**p**
Age Group	33.6	4, 71	<0.001*	2.1	4, 69	0.09†
Contact Force	237.5	2, 70	<0.001*	9.0	2, 68	<0.001*
Skull Position	7.0	1, 71	0.01*	326.0	1, 69	<0.001*
Age X Force	2.9	8, 138	0.005*	0.9	8, 134	0.53
Age X Position	2.6	4, 71	0.04*	11.3	4, 69	<0.001*
Position X Force	3.8	2, 70	0.03*	14.1	2, 68	<0.001*
Age X Position X Force	0.6	8, 138	0.75	0.8	8, 134	0.58

Based on results from mixed-model and multivariate ANOVAs.
A "*" indicates result is significant at $p<0.05$. A "†" indicates result is marginally significant at $p<0.1$.

2.3.2.1 Low frequency

Means and standard deviations for each measured condition for the five low frequencies selected for analysis are presented in Table 2.3, which will be mentioned briefly in the current section and referenced further in chapter 3 during a discussion related to mechanical impedance means between age groups. In the following sections, trends will be discussed as they relate to the selected variables of age group, contact force, and skull position.

2.3.2.1.1 Age group

From Figure 2.6, a clear trend of increasing impedance with increasing age category was observed for low frequencies overall, with the youngest age group (Group A) showing particularly low impedance values when compared to the other groups. This trend can also be observed in the full frequency sweep in Figure 2.3. Results of a mixed-model ANOVA revealed a significant main effect of age group, as presented in Table 2.2. Significance levels for *post hoc* comparisons for each significant main effect and interaction are presented in Table 2.4. The *post hoc* analysis for age group confirmed that Group A had significantly lower impedance when compared to all other age groups. Group B also had significantly lower impedance compared to Group D and adults. Additionally, Group C had lower impedance than adults at a level that approached significance. The other contrasts between age categories were not significant.

As displayed in Table 2.4, *post hoc* comparisons for a significant two-way interaction effect between skull position and age group revealed that, Group A had significantly lower impedance magnitude compared to all other age groups for both temporal bone and forehead positions, which is in line with the main effect observed for age. However, Group B had significantly lower impedance magnitude compared to Group D and adults for the forehead position, but not for the temporal bone position. *Post hoc* analyses for the age group and contact force interaction indicated that the difference in impedance between Group A and Group B was driven by the 5.4 N contact force only. Impedance was not significantly different between the two groups for the other two forces, as shown in the low-frequency graphs from Figure 2.6. It is also important to note that the standard deviations across low frequencies for 4 and 2 N for Group B were typically larger than the standard deviations for 5.4 N (Table 2.3). Therefore, the larger variability for 4 and 2 N conditions for Group B may have contributed to this interaction. Group A had significantly lower impedance magnitude compared to all other age groups for all measured contact forces, consistent with the main effect. Group B had significantly lower impedance magnitude compared to adults for each contact force, but comparisons to Group D were significant for 4 and 2 N but not for 5.4 N.

Table 2.3 Means and standard deviations for impedance magnitude for each age group, skull position, and contact force measured across selected low frequencies.

			Frequency (Hz)				
			125	**251**	**501**	**758**	**1000**
Group A	**Temporal Bone**	**5.4 N**	36.6 (7.4)	32.6 (4.3)	28.7 (3.1)	27.2 (2.5)	26.8 (2.4)
		4 N	38.1 (5.8)	31.7 (3.9)	28.3 (3.0)	25.4 (3.4)	24.4 (3.4)
		2 N	36.2 (4.3)	29.4 (4.5)	26.2 (2.8)	24.2 (3.3)	24.4 (2.9)
	Forehead	**5.4 N**	40.7 (5.7)	35.1 (3.1)	29.8 (3.4)	25.4 (3.0)	23.2 (3.4)
		4 N	40.6 (3.9)	34.7 (2.6)	28.2 (2.7)	23.4 (3.6)	20.7 (2.9)
		2 N	36.9 (6.2)	33.0 (2.8)	26.4 (2.6)	20.3 (4.6)	18.0 (2.6)
Group B	**Temporal Bone**	**5.4 N**	45.3 (3.7)	39.1 (2.5)	33.7 (2.6)	31.6 (2.8)	28.9 (3.3)
		4 N	40.2 (8.6)	35.3 (6.0)	30.7 (4.0)	27.3 (4.4)	25.8 (4.3)
		2 N	40.0 (5.7)	35.6 (5.6)	29.4 (3.2)	25.3 (3.7)	23.9 (5.0)
	Forehead	**5.4 N**	47.9 (4.0)	40.7 (1.9)	34.2 (3.0)	31.3 (2.2)	27.6 (3.2)
		4 N	44.4 (6.0)	37.5 (3.1)	31.8 (2.7)	26.6 (3.2)	22.6 (4.7)
		2 N	40.4 (6.8)	34.8 (4.4)	29.0 (3.3)	24.1 (4.9)	20.7 (3.4)
Group C	**Temporal Bone**	**5.4 N**	42.9 (8.0)	41.5 (1.7)	35.9 (2.1)	32.6 (1.9)	31.0 (1.9)
		4 N	43.5 (3.3)	38.6 (2.2)	33.0 (2.6)	30.0 (3.7)	28.2 (2.8)
		2 N	40.0 (4.6)	37.0 (2.8)	30.7 (2.3)	27.5 (3.1)	25.2 (3.3)
	Forehead	**5.4 N**	51.0 (5.6)	43.1 (2.5)	37.1 (2.4)	33.0 (3.1)	29.4 (3.4)
		4 N	43.9 (6.2)	40.2 (3.3)	33.6 (3.7)	28.6 (3.9)	25.3 (3.8)
		2 N	42.3 (4.2)	36.1 (3.6)	31.2 (2.6)	26.6 (3.8)	22.9 (3.4)
Group D	**Temporal Bone**	**5.4 N**	47.7 (7.9)	42.5 (2.8)	36.0 (3.3)	33.3 (3.2)	30.7 (3.7)
		4 N	45.1 (5.6)	39.6 (2.6)	33.7 (3.0)	30.4 (3.4)	28.1 (3.1)
		2 N	39.2 (8.3)	37.8 (2.4)	31.7 (3.0)	29.4 (3.2)	26.6 (3.8)
	Forehead	**5.4 N**	48.1 (7.5)	44.6 (2.5)	38.6 (2.5)	34.7 (3.1)	31.9 (2.8)
		4 N	45.7 (6.5)	41.5 (2.7)	36.1 (2.5)	30.7 (3.9)	28.0 (4.0)
		2 N	43.9 (4.2)	38.9 (3.1)	32.9 (3.2)	28.3 (3.4)	25.4 (3.4)
Adult	**Temporal Bone**	**5.4 N**	47.4 (4.7)	41.8 (2.4)	36.9 (2.0)	33.1 (2.7)	30.7 (2.5)
		4 N	43.4 (4.8)	39.0 (3.5)	33.7 (3.6)	30.5 (4.3)	28.4 (3.7)
		2 N	42.2 (5.6)	37.5 (4.1)	32.3 (4.4)	29.1 (4.6)	26.9 (4.3)
	Forehead	**5.4 N**	50.3 (6.1)	45.4 (1.9)	39.7 (2.2)	35.8 (2.3)	32.7 (3.0)
		4 N	44.9 (6.4)	42.6 (3.0)	37.5 (2.8)	33.7 (2.3)	30.5 (2.9)
		2 N	44.2 (5.9)	40.4 (2.0)	34.4 (2.3)	30.9 (2.6)	27.5 (2.0)

Table 2.4 Low-frequency *post hoc* analyses *p*-values for the main effects of age group and contact force, and the interactions of age group-by-position, age group-by-force and force-by-position.

		Main Effect	Position			Force					
			Temporal Bone (TB)	Forehead (F)	TB vs. F	5.4N	4N	2N	5.4 vs. 4 N	5.4 vs. 2N	4 vs. 2N
Age Group											
Group A					1.0				1.0	<0.001*	0.02*
	vs. B	<0.001*	0.01*	0.008*		<0.001*	0.36	0.22			
	vs. C	<0.001*	<0.001*	<0.001*		<0.001*	<0.001*	<0.001*			
	vs. D	<0.001*	<0.001*	<0.001*		<0.001*	<0.001*	<0.001*			
	vs. Adult	<0.001*	<0.001*	<0.001*		<0.001*	<0.001*	<0.001*			
Group B					1.0				<0.001*	<0.001*	0.07†
	vs. C	0.17	1.0	1.0		1.0	1.0	1.0			
	vs. D	<0.001*	0.12	0.002*		0.11	0.003*	0.03*			
	vs. Adult	<0.001*	0.13	<0.001*		0.02*	<0.001*	<0.001*			
Group C					1.0				<0.001*	<0.001*	<0.001*
	vs. D	0.74	1.0	1.0		1.0	1.0	1.0			
	vs. Adult	0.07†	1.0	0.03*		1.0	1.0	0.31			
Group D					1.0				<0.001*	<0.001*	<0.001*
	vs. Adult	1.0	1.0	1.0		1.0	1.0	1.0			
Adult					0.01*				<0.001*	<0.001*	0.02*
Force											
5.4 N					<0.001*						
	vs. 4 N	<0.001*	<0.001*	<0.001*							
	vs. 2 N	<0.001*	<0.001*	<0.001*							
4 N					0.09†						
	vs. 2 N	<0.001*	<0.001*	<0.001*							
2 N					1.0						

A "*" indicates contrast is significant at $p<0.05$. A "†" indicates contrast is marginally significant at $p<0.1$.

2.3.2.1.2 Contact force

For the impedance magnitude for contact force, Figures 2.4 and 2.6 portray a trend of increasing impedance with increasing contact force. Hotelling's T MANOVA results confirmed a significant main effect of contact force, and *post hoc* analyses revealed that all comparisons between contact forces were significant, as displayed in the main effect column of Table 2.4. *Post hoc* comparisons revealed that all comparisons between contact forces within each age group were significant with the exception of 5.4 to 4 N for Group A. The comparison between 4 and 2 N for Group B was marginally significant. However, as mentioned previously, the larger variation observed for 4 and 2 N conditions from Group B (Table 2.3) may partly explain this marginal effect. All comparisons between contact forces for each skull position were also significant.

2.3.2.1.3 Skull position

The impedance magnitude across the low frequencies in Figure 2.5 for skull positions revealed that the trend lines with increasing frequency may intersect when frequencies approach resonance. The interaction observed is likely due to differences in resonant frequency properties, which will be discussed in a later section. A main effect of skull position revealed that the impedance of the forehead was significantly greater than the impedance of the temporal bone, although Figure 2.5 depicts a greater difference between skull positions for high frequencies, discussed in the next section.

When comparing between skull positions for each age group individually, the impedance for the forehead was only significantly higher than the temporal bone for adults. No other comparison between positions was significant. Similarly, impedance of the forehead was significantly larger than the impedance of the temporal bone for a contact force of 5.4 N and only marginally significant for 4 N. No significant difference was found between positions for 2 N. Therefore, the main effect of skull position is driven primarily by adults and by the 5.4 N contact force conditions. The 3-way interaction between age group, contact force and skull position was not significant.

2.3.2.2 High frequency

Means and standard deviations for each measured condition for high frequencies are presented in Table 2.5. Table 2.5 will not be directly discussed in the results but will be referenced later in the discussion of the results. Similar to the low frequencies, results for the high frequencies will be discussed in the following sections based the selected variables of age group, contact force and skull position.

2.3.2.2.1 Age group

Represented in both Figure 2.3 and Figure 2.6, the large effect for age seen for the low frequencies was not apparent across the high frequencies. Although the main effect of age did approach significance based on a mixed-model ANOVA for high frequencies, *post hoc* analyses revealed that no age group was significantly different than any other age group, as shown in Table 2.6. *Post hoc* analyses for the interaction between age group and skull position indicated that Group A had significantly higher impedance magnitude than Group C, D and adults for the temporal bone position; however, impedance between groups was not significantly different for the forehead position. In contrast with low frequencies, an interaction between age group and contact force was not significant for high frequencies.

Table 2.5 Means and standard deviations for impedance magnitude for each age group, skull position, and contact force measured across selected high frequencies

			Frequency (Hz)					
			1995	**3019**	**3981**	**6025**	**7943**	**10000**
Group A	**Temporal Bone**	**5.4 N**	24.9 (2.2)	26.8 (2.7)	29.6 (3.7)	34.7 (4.2)	36.4 (3.9)	36.7 (7.2)
		4 N	24.4 (2.2)	27.2 (3.3)	30.0 (3.5)	34.2 (4.4)	33.9 (8.8)	34.6 (7.9)
		2 N	23.8 (2.8)	27.4 (3.3)	30.0 (4.1)	35.0 (3.5)	34.5 (5.5)	34.9 (5.8)
	Forehead	**5.4 N**	23.5 (1.8)	29.3 (1.0)	32.3 (0.9)	35.9 (1.6)	35.8 (5.7)	37.4 (5.7)
		4 N	23.3 (2.6)	28.8 (1.7)	31.9 (0.9)	35.8 (1.6)	37.7 (2.5)	39.9 (3.3)
		2 N	24.3 (2.6)	28.9 (1.9)	32.6 (1.2)	34.4 (2.5)	36.7 (2.7)	38.8 (6.6)
Group B	**Temporal Bone**	**5.4 N**	25.2 (1.4)	25.7 (3.5)	28.3 (4.9)	31.9 (5.5)	33.0 (4.9)	34.7 (6.1)
		4 N	23.3 (2.2)	25.3 (4.1)	26.7 (4.0)	32.0 (4.7)	32.1 (6.8)	32.2 (4.3)
		2 N	21.8 (3.3)	23.8 (4.5)	25.3 (5.9)	31.4 (5.6)	30.5 (5.5)	32.0 (4.5)
	Forehead	**5.4 N**	23.2 (1.8)	29.6 (1.5)	33.1 (1.5)	36.5 (1.3)	39.5 (1.1)	40.1 (4.1)
		4 N	24.5 (1.6)	30.2 (0.7)	33.0 (2.0)	36.2 (3.0)	39.4 (1.1)	42.3 (0.9)
		2 N	24.6 (2.4)	29.7 (1.5)	33.0 (0.9)	36.7 (1.7)	38.6 (2.3)	41.7 (2.6)
Group C	**Temporal Bone**	**5.4 N**	24.6 (3.1)	25.4 (3.4)	27.5 (3.2)	33.0 (2.8)	31.2 (5.1)	32.9 (3.7)
		4 N	22.7 (2.7)	23.3 (4.0)	24.7 (4.8)	30.5 (6.5)	28.6 (4.2)	30.4 (3.4)
		2 N	21.6 (3.7)	23.1 (4.4)	25.0 (4.8)	28.9 (5.3)	25.1 (5.7)	29.0 (3.6)
	Forehead	**5.4 N**	24.3 (2.4)	29.3 (1.4)	33.1 (1.9)	36.8 (2.1)	39.7 (0.9)	42.3 (1.4)
		4 N	24.0 (1.8)	29.9 (1.4)	32.8 (1.5)	37.5 (1.2)	39.1 (2.3)	41.5 (2.4)
		2 N	25.6 (1.7)	30.5 (1.5)	33.3 (1.2)	36.1 (1.9)	38.9 (3.1)	42.2 (2.4)
Group D	**Temporal Bone**	**5.4 N**	24.6 (3.6)	24.8 (3.3)	26.3 (4.3)	30.9 (5.1)	28.4 (6.2)	32.5 (3.1)
		4 N	23.9 (2.2)	23.8 (5.3)	24.9 (6.0)	30.7 (4.7)	26.7 (6.6)	30.1 (4.4)
		2 N	21.8 (3.1)	22.6 (5.1)	24.7 (5.0)	30.1 (4.8)	25.4 (6.5)	29.2 (3.7)
	Forehead	**5.4 N**	23.6 (1.1)	28.8 (1.0)	32.7 (1.0)	37.4 (0.8)	39.9 (0.9)	43.1 (0.8)
		4 N	23.8 (1.5)	28.8 (2.1)	33.58613	36.9 (1.7)	38.8 (2.6)	41.1 (3.1)
		2 N	24.1 (2.3)	30.1 (2.0)	33.6 (1.4)	37.4 (2.0)	39.2 (3.7)	42.7 (2.2)
Adult	**Temporal Bone**	**5.4 N**	25.0 (1.6)	26.4 (2.7)	29.3 (2.6)	32.7 (2.6)	30.2 (4.9)	33.3 (3.8)
		4 N	25.1 (4.0)	25.2 (4.1)	26.3 (4.1)	30.4 (3.5)	24.3 (6.1)	29.2 (3.5)
		2 N	23.9 (3.4)	25.7 (3.9)	26.5 (3.7)	32.0 (3.3)	26.1 (6.7)	30.8 (3.9)
	Forehead	**5.4 N**	25.4 (1.8)	29.0 (1.5)	31.8 (3.4)	37.4 (1.3)	39.7 (2.6)	42.6 (2.0)
		4 N	24.6 (1.1)	29.2 (1.0)	32.8 (1.0)	37.2 (1.2)	40.2 (1.7)	43.1 (2.0)
		2 N	24.6 (1.6)	29.7 (1.7)	32.9 (1.6)	38.0 (1.5)	40.2 (1.4)	43.2 (2.1)

Table 2.6 High-frequency *post hoc* analyses *p*-values for the main effects of age group and contact force, and the interactions of age group-by-position and force-by-position.

		Main Effect	Position: Temporal Bone (TB)	Position: Forehead (F)	Position: TB vs. F
Age Group					
Group A					1.0
	vs. B	1.0	0.09†	1.0	
	vs. C	0.45	<0.001*	1.0	
	vs. D	0.1	<0.001*	1.0	
	vs. Adult	1.0	0.002*	0.52	
Group B					<0.001*
	vs. C	1.0	1.0	1.0	
	vs. D	1.0	0.43	1.0	
	vs. Adult	1.0	1.0	1.0	
Group C					<0.001*
	vs. D	1.0	1.0	1.0	
	vs. Adult	1.0	1.0	1.0	
Group D					<0.001*
	vs. Adult	1.0	1.0	1.0	
Adult					<0.001*
Force					
5.4 N					<0.001*
	vs. 4 N	0.003*	<0.001*	1.0	
	vs. 2 N	<0.001*	<0.001*	1.0	
4 N					<0.001*
	vs. 2 N	1.0	1.0	1.0	
2 N					<0.001*

A "*" indicates contrast is significant at $p<0.05$. A "†" indicates contrast is marginally significant at $p<0.1$.

2.3.2.2.2 Contact force

Based on Figure 2.4, a more prominent effect of increasing contact force was apparent across the low frequencies compared to high frequencies; however, a main effect of contact force was also significant for high frequencies. *Post hoc* analyses revealed that impedance magnitude for 5.4 N was significantly larger than for both 4 and 2 N; however, impedance magnitude for 4 N was not significantly different than 2 N. Similarly, the interaction effect between position and contact force also revealed that for the temporal bone position, impedance magnitude for 5.4 N was significantly greater than for 4 and 2 N; however, 4 N was not significantly different from 2 N. For the forehead position, no comparisons between contact forces were significant.

2.3.2.2.3 Skull position

In contrast to contact force, where a more prominent effect was observed for low frequencies, Figure 2.5 displays the difference in skull positions across the full frequency sweep, where it is apparent that the effect of skull position was more prominent for the high frequencies compared to low frequencies. Impedance for the forehead was significantly higher than impedance measured at the temporal bone (Table 2.2).

When comparing skull positions for each age group individually, impedance magnitude was significantly higher for the forehead position compared to the temporal bone for each age group, with the exception of Group A. A difference in skull position was not significant for that age group only. Additionally, impedance magnitude was significantly higher for the forehead position than for the temporal bone position for each contact force. It is interesting to note that these results are in contrast to the results for low frequencies, where significant differences between skull positions were driven by the adult group and by 5.4 N contact force factors only.

2.3.2.3 Resonant frequency properties

Figure 2.7 summarizes the resonant frequency properties for each variable, including the resonant frequency itself, and the impedance magnitude at each resonant frequency, which corresponds to the magnitude of resistance, or friction, of the system. Notably, an increase in resonant frequency can be observed in addition to an impedance magnitude as groups increase in age. Interestingly, the trend across age is more apparent for the forehead position compared to the temporal bone. In addition, a similar increase in both resonant frequency and impedance is present for increasing contact force.

To support these observations, resonant frequency was analyzed with a mixed-model ANOVA. Results revealed a significant main effect of age group [$F(4,69)=7.84$), $p<0.001$].

Bonferroni multiple comparison *post h*oc analyses were completed on each significant main effect. For age group, resonant frequency was found to be significantly lower for Group A than for Group C (p=0.004), Group D (p<0.001) and adults (p<0.001). No other significant contrasts were found between age groups. In addition, mixed-model ANOVA results indicated a significant effect of contact force [$F(2,138)$ = 27.63, p<0.001] and position [$F(1,69)$ = 90.34, p<0.001]. *Post hoc* analyses on contact force indicated that resonant frequency for 5.4 N was significantly greater than the resonant frequency for 4 N (p<0.001) and 2 N (p<0.001). Similarly, the resonant frequency for 4 N was significantly greater than for 2 N (p=0.005). No two- or three-way interaction effects were significant for this mixed-model ANOVA.

Results from the mixed-model ANOVA for impedance magnitude at resonant frequency revealed a significant main effect of age group [$F(4,69)$ = 24.6, p<0.001]. Significance values for *post hoc* comparisons are located in Table 2.7. *Post hoc* testing indicated that adults had significantly higher impedance values at resonant frequency than all other groups (A, B, C, and D). Additionally, Groups A and B also had significantly lower impedance values than Groups C and D. In addition, the interaction between age group and skull position was significant [$F(4,69)$ = 9.3, p<0.001]. For the forehead position, impedance for adults was significantly higher than Groups A and B and marginally significantly higher than Group D, but for the temporal bone position, impedance for adults was only significantly higher than Group B and marginally higher than Group A. Additionally, Groups A and B were significantly different from Groups C and D for the forehead conditions, but not for the temporal bone conditions. Therefore, the main effect of group is largely based on the measurements from the forehead.

The mixed-model ANOVA also revealed a significant main effects of contact force [$F(2,138)$=47.9, p<0.001], where *post hoc* analyses indicated that the mechanical impedance measured at resonant frequency for 5.4 N was significantly greater than for 4 and 2 N, and impedance at 4 N was significantly greater than 2 N. Finally, a significant main effect of position was revealed [$F(1,69)$ = 4.9, p=0.03]; however, the interaction between age group and position indicated that impedance for the temporal bone was only significantly greater than the forehead for Group A. No other contrasts between positions were significant for any other age group.

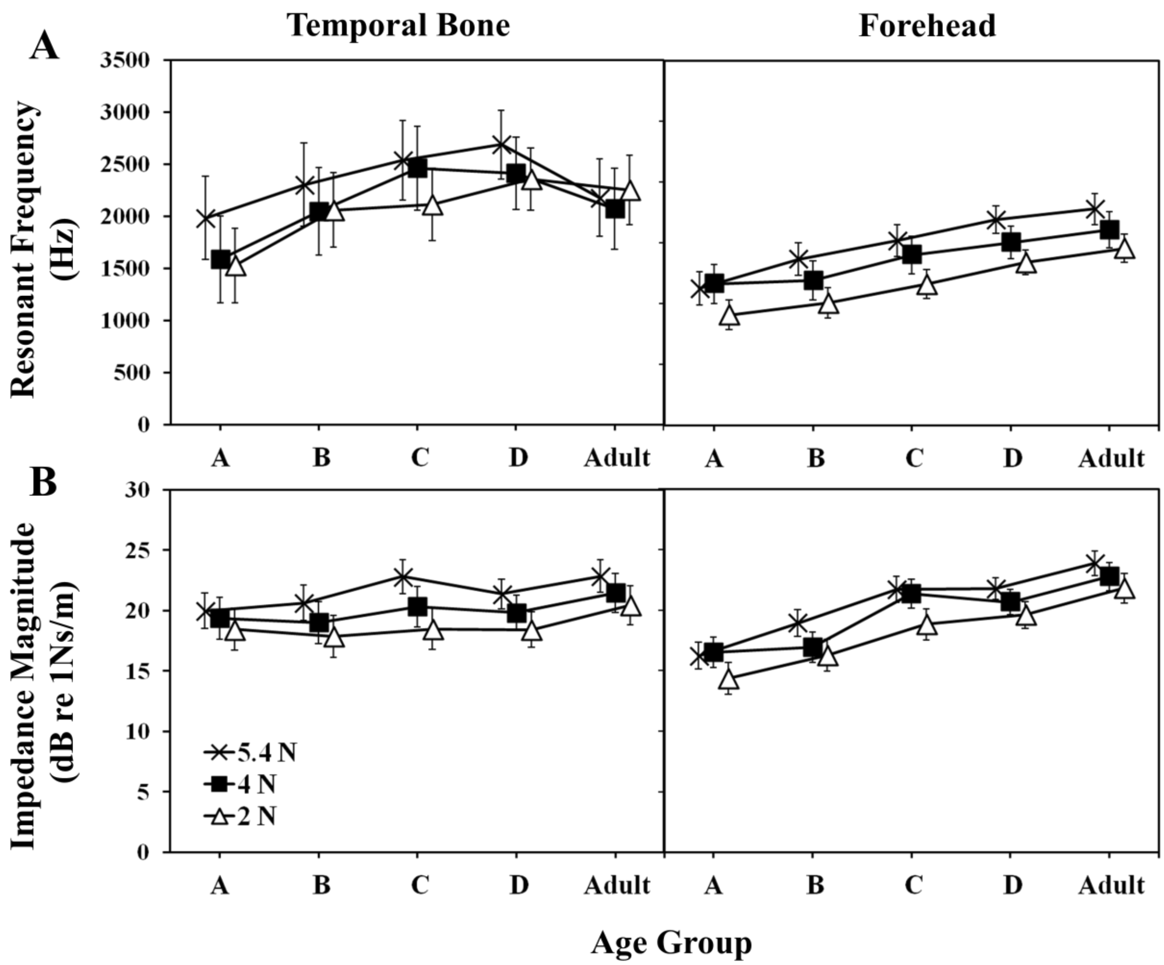

Figure 2.7 A. Mean resonant frequency at the temporal bone and forehead for each age group and contact force. **B.** Mean impedance magnitude values at resonant frequency at the temporal bone and forehead for each age group and contact force. Error bars represent 95% confidence intervals around the mean.

Table 2.7 Impedance magnitude at resonant frequency *post hoc* analyses *p*-values for the main effects of age group and contact force, and the interaction of age group-by-position.

		Main Effect	Position		
			Temporal Bone (TB)	Forehead (F)	TB vs. F
Age Group					
Group A					<0.001*
	vs. B	1.0	1.0	1.0	
	vs. C	<0.001*	1.0	<0.001	
	vs. D	<0.001*	1.0	<0.001	
	vs. Adult	<0.001*	0.06†	<0.001	
Group B					0.5
	vs. C	<0.001*	1.0	<0.001	
	vs. D	0.002*	1.0	<0.001	
	vs. Adult	<0.001*	0.04*	<0.001	
Group C					1.0
	vs. D	1.0	1.0	1.0	
	vs. Adult	0.03*	1.0	0.1	
Group D					1.0
	vs. Adult	0.002*	0.43	0.06†	
Adult					1.0
Force					
5.4 N					
	vs. 4 N	<0.001*			
	vs. 2 N	<0.001*			
4 N					
	vs. 2 N	<0.001*			
2 N					

A "*" indicates contrast is significant at $p<0.05$. A "†" indicates contrast is marginally significant at $p<0.1$.

CHAPTER 3: Discussion and Conclusions

Mechanical impedance of the skin-covered skull is an important factor upon which the study of bone-conduction hearing is founded. As mentioned earlier, to date, studies have only investigated the mechanical impedance for individuals between ages 9 and 71 years (Flottorp & Solberg, 1976). This study fills in a significant gap in the literature by collecting the mechanical impedance for infants and young childen 7 years of age and younger. By investigating the mechanical properties of the skull, this study contributes to an explanation of the maturational differences in bone-conduction sensitivity and relates the maturation of skull properties to the fitting and verification of the soft band BAHS. To achieve the goals of this study, a special tool was developed and tested to collect mechanical impedance in infants and young children with ease, something that would have been difficult or impossible using previously published methods. Specific contributions to bone-conduction research methodology will be discussed first, followed by a detailed discussion of the results and implications regarding the maturation of bone-conduction hearing and fitting and verification of the BAHS.

3.1 Mechanical Impedance Methodology

Previous methods for measuring impedance in older children and adults used bulky stationary equipment that consisted of a series of weights that applied a calibrated contact force to the head (Corliss & Koidan, 1955; Cortes, 2002; Flottorp & Solberg, 1976; Håkansson, Carlsson, & Tjellström, 1986). Researchers in the Bone-Conduction Amplification Laboratory at the iRSM developed a specialized portable hand-held device for measuring impedance that possesses numerous advantages compared to previous methods. The iRSM researchers converted calibrated weights into an innovative compressive spring gauge that can be monitored by the experimenter. The simplified appliance has a short set-up time that accommodates the more limited attention span of infants and young children. In addition, the BCAD Software created by researchers at iRSM allows for quick automated online calculations of mechanical impedance. Experimenters can observe the impedance calculation while recording and can determine immediately if any troubleshooting is necessary. This user-friendly application and portable device was very effective in the present study and will allow for single- or multi-site collection of impedance data in future research.

In order to determine the validity of the portable measurement tool, the adult data from the current study were compared to previously published impedance data. Table 3.1 provides a comparison of average mechanical impedance values from the adult temporal bone condition (5.4 N) in the current study compared to normative impedance values published in IEC 373 (1990), which are used to calibrate the artificial mastoid. Impedance values from the current study were slightly lower than the reference impedance values across all frequencies defined. Cortes (2002) also found that the impedance collected from 30 adult subjects were lower than impedance values from the artificial mastoid across all frequencies. The slight differences in frequency between the two sets of data in Table 3.1 were due to the logarithmic spacing of the frequencies collected in the current study, and did not account for the differences observed in impedance values. The largest differences between the two data sets were for the frequencies between 1250-1600 Hz. One factor that contributed to differences in the mid-frequencies was the resonant frequency. The average resonant frequency for the current study was 2174 Hz, compared to 3000 Hz for IEC 373 (1990).

A second method for comparing the impedance data from the current study with previously collected data is to use the calculated compliance (inverse of stiffness), mass and resistance parameter values based on the cascade model described by Håkansson, Tjellström and Rosenhall (1985) and Håkansson, Carlsson and Tjellström (1986). Compliance of the skin-covered skull (C_S) was determined from the impedance magnitude at 125 Hz, and mass of the skin-covered skull (M_S) was calculated from the equation $f_r = 1/[2\pi(MsCs)^{0.5}]$, based on the resonant frequency (RF). Finally, resistance (R_S) was calculated as the average impedance value at the resonant frequency (Håkansson, Carlsson, & Tjellstrom, 1986). Measurement conditions and parameter values calculated from the impedance measured by Flottorp and Solberg (1976), Håkansson, Carlsson and Tjellström (1986) and the present study are presented in Table 3.2. Compliance calculated from the present study is smaller and mass is larger compared to previous studies. The current resistance value lies in between values calculated from previous studies, and the resonant frequency is lower than previous studies. It is important to note that the present study measured the mechanical impedance of the temporal bone at the location where a BAHS would be placed, and previous studies measured the impedance at the flattest portion of the mastoid process. In addition, the current study used a contact force of 5.4 N and a contact area of 2.0 cm^2.The larger mass and smaller compliance value, in comparison to previous studies, may be attributed to the difference in temporal bone location and the larger contact disk area used (Håkansson, Carlsson and Tjellström, 1986), even though contact area for all studies falls within the ANSI (1996) tolerances (1.75±0.25 cm^2). Håkansson, Carlsson and Tjellström (1986) attributed their smaller mass and resistance and slightly larger compliance values than those reported by Flottorp and Solberg (1976) to a smaller contact disk area and higher contact force. Contact area has previously been shown to be a sensitive parameter for bone-conduction threshold measurement, particularly for high frequencies (Khanna, Tonndorf, & Queller, 1976; Watson, 1938). Despite small differences in impedance values across studies, the new methodology used in the present study allows for relatively rapid, accurate, and repeatable measurements of mechanical impedance for individuals as young as 6 weeks of age.

Table 3.1 Mechanical impedance magnitude for IEC 373 (1990) and for the current study.

IEC 373		Current study	
Frequency (Hz)	**Impedance (dB re 1 Ns/m)**	**Impedance (dB re 1 Ns/m)**	**Frequency (Hz)**
250	44.3	41.8	**251**
315	42.9	40.3	**316**
400	41.3	38.9	**398**
500	39.9	36.9	**501**
630	38.5	34.8	**630**
750	37.4	33.1	**758**
800	37.0	32.6	**794**
1000	35.5	30.7	**1000**
1250	34.0	28.0	**1258**
1500	32.4	26.2	**1513**
1600	31.9	25.2	**1621**
2000	29.8	25.0	**1995**
2500	27.8	25.3	**2511**
3000	27.2	26.4	**3019**
3150	27.3	26.7	**3162**
4000	29.5	29.3	**3981**
5000	32.6	32.0	**5011**
6000	34.4	32.7	**6025**
6300	34.6	32.6	**6309**
8000	35.1	30.2	**7943**

Table 3.2 Parameter values compliance (C_S), mass (M_S), resistance (R_S) and resonant frequency (RF) and the conditions used for collecting impedance of the skin-covered skull from Flottorp and Solberg (1976), Håkansson, Carlsson and Tjellström (1986) and the current study.

Parameter	Flottorp and Solberg (1976)	Håkansson, Carlsson and Tjellström (1986)	Current Study
C_S (m/N)	4.4×10^{-6}	5.3×10^{-6}	2.6×10^{-6}
M_S (Ns^2/m)	0.6×10^{-3}	0.53×10^{-3}	2.1×10^{-3}
R_S (Ns/m)	20.0	9.0	14.5
RF (Hz)	3000	3000	2174
Condition			
Contact Force (N)	5.4	10	5.4
Contact Area (cm^2)	1.75	1.50	2.00

3.2 Infant-Adult Differences in Bone-Conduction Hearing

The following two sections examine the findings from each experiment in the current study and discuss how they might explain infant-adult differences in bone-conduction sensitivity.

3.2.1 Transcranial attenuation of bone-conducted sound

The findings from Experiment 1, which used measures of sound pressure in the ear canal to predict trancranial attenuation of bone-conducted sound, are consistent with previous studies that showed that transcranial attenuation of vibratory sound is greater in infants compared to adults. By measuring sound pressure in the ear canal, it is possible to isolate some of the factors that contribute to immaturities in physiological measures of bone-conduction hearing sensitivity, such as infant-adult differences in brainstem maturation. A measure of sound pressure in the ear canal represents only a physical measure of attenuation of bone-conducted sound across the skull.

Most studies using physiological measures of transcranial attenuation have not compared infants of different age groups. One exception was Yang, Rupert and Moushegian (1987) who measured transcranial attenuation from the contralateral channel of ABRs from bone-conduction click-stimuli for neonates, 1 year olds, and adults. They predicted that adults had 0-10 dB of attenuation, 1-year olds had 15-25 dB of attenuation and neonates had 25-35 dB of attenuation. The current study was the first to measure the maturation of transcranial attenuation for infants and children of different ages with frequency-specific stimuli, using a physical measure instead of a physiological or behavioural measure. The amount of overall attenuation for infants was substantially lower than values predicted from physiological measures of hearing (e.g., Small & Stapells, 2008b; Yang, Rupert, & Moushegian, 1987); however, findings from the current study support previous trends that young infants under 11 months of age show the greatest attenuation.

Results from this study support the prediction that physical properties of the infant skull, such as immature sutures and fontanelles, contribute to the large transcranial attenuation of bone-conducted sound for infants. Sohmer, Freeman, Geal-Dor, Adelman and Savion (2000) found that 14 dB of acceleration was lost between the fontanelle and the adjacent temporal bone, which provided evidence that the energy might dissipate across fontanelles. In addition, attention must be paid to the differences in structure of the skull bone. The speed of sound has been shown to be faster in compact bone, compared to spongy diploë present in adult skulls (O'Brien & Liu, 2005). One speculation is that the unilaminar compact bone may allow vibratory energy to pass through the bone to the cochlea, while the spongy diploë in adult bone limits the vibratory energy passing through the bone, and instead allows for lateral transmission across the skull.

For the current study, the sound pressure measured with the transducer on the ipsilateral temporal bone was subtracted from sound pressure measured from each of the two other skull positions (forehead and contralateral temporal bone). Therefore, any infant-adult differences in ear canal properties did not contribute to the results of this experiment. It is also important to note that because the measurement was in the ear canal, the results do not provide information regarding the energy reaching the cochlea, nor do they provide information on the mechanisms of bone-conduction hearing, but only how bone-conducted sound is traveling across the skull. These measurements provide a gross estimation of the vibratory energy that is reaching the temporal bone. Therefore, infant-adult differences in the anatomy and function of the tympanic membrane, middle ear ossicles or vibration of the cochlea were not likely contributing to this measure of transcranial attenuation. Even when these factors were excluded from the measurement, differences in transcranial attenuation between infants and adults are significant, which provides more support that skull properties contribute significantly to infant-adult differences in transcranial attenuation. Although findings from the current experiment cannot directly explain infant-adult differences in bone-conduction sensitivity, age-dependent trends observed from measures of transcranial attenuation are similar to the age-dependent trends observed throughout maturation of bone-conduction hearing (Small & Stapells, 2008a). Therefore, a link between skull properties, transcranial attenuation, and bone-conduction sensitivity can be theorized.

One potential confounding factor regarding infant-adult differences in sound pressure relates to the properties and the status of the middle ear in young infants. As mentioned previously, sound pressure in the ear canal from air-conduction stimulation has shown to be higher among children with otitis media with effusion (Martin, Westwood & Bamford, 1996) and otitis media is more common among infants in their first year of life compared to older infants (Paradise et al., 1997). In addition, developmental changes in the middle ear impedance may affect sound pressure measures, particularly among the youngest infants in Group A. As noted earlier, Group A included 10 infants younger than 6 months of age. It is likely that the middle ear status and developmental differences contributed to the raw measurements of sound pressure. However, because the attenuation values were difference calculations with the probe tube in the same ear across all transducer locations, middle ear contributions were likely cancelled out, if sound pressure and middle ear effects are linearly related.

Results of frequency-specific trends from the current study revealed that the most overall attenuation was present for 1000 Hz, while the least attenuation was present for 500 and 4000 Hz. From the contralateral temporal bone, 500 Hz had less attenuation than the other three frequencies across all groups, which is consistent with previous studies on transcranial attenuation of frequency-specific bone-conducted sounds in adults (Stenfelt, 2012). The interaction between frequency and age group was not significant in the present study, and therefore frequency-specific trends corresponding to differences in bone-conduction sensitivity between infants and adults cannot be explained by transcranial attenuation measured through sound pressure in the ear canal. A combination of factors makes comparisons between frequencies difficult for a measure of sound pressure in the ear canal, particularly with sound originating from the contralateral temporal bone. For instance, Stenfelt (2012) commented that the convergence of bone-conducted sound on the contralateral cochlea is a complicated manner involving a combination of three orthogonal vibration directions. In addition, large intersubject variability in skull properties can partly be attributed to anti-resonances in the skull, which may produce attenuation at frequencies that cannot be predicted (Eeg-Olofsson, Stenfelt, & Granstrom, 2011; Stenfelt, 2012). Behavioural thresholds for adults with unilateral hearing loss have shown that transcranial attenuation is greater for high frequencies compared to low frequencies (Kirikae, 1959b; Nolan & Lyon, 1981). Additionally, there are methodological concerns with sound pressure measures in the ear canal, such as airborne radiation of sound and the interaction with the noise floor for frequencies above 2500 Hz (Huizing, 1960a). Finally, when measuring sound pressure in the ear canal, infant-adult differences in ear canal volume and resonances may alter the absolute intensity of sound radiating in the ear canal originating from a pure-tone bone-conduction stimulus (Small & Hu, 2011). These factors in combination make it difficult to make direct comparisons of frequency-specific trends in transcranial attenuation related to sound pressure in the ear canal to frequency-specific infant-adult differences in bone-conduction sensitivity.

3.2.2 Mechanical impedance of the skin-covered skull

This study is the first to investigate the mechanical impedance of the skin-covered skull of infants and young children, and we now have a better understanding of the factors that contribute the most to maturational changes in bone-conduction sensitivity. The significant age-related differences in mechanical impedance of the skin-covered skull found in Experiment 2 provide strong evidence to support that infant-adult differences in bone-conduction thresholds are related to properties of the immature skull. Mechanical impedance magnitude measured both at the forehead and temporal bone increased systematically with age for low frequencies and at resonant frequency. Infants up to 2 years of age had significantly lower mechanical impedance magnitude compared to adults for low frequencies, and children up to age 3 years had lower impedance than adults at resonant frequency. Young infants (1-10 months) had the lowest impedance magnitude, with increasing magnitude throughout maturation. Notably, this trend of increasing impedance with increasing age group was only present for low frequency and resonant frequency data. Although an effect of age was not significant for high frequencies, an interaction was present between age and skull position, showing that the youngest infant group had significantly higher impedance magnitude values than groups with those age 2 years and older for the temporal bone only. This trend was in the opposite direction from that observed for low frequency and resonant frequency data.

To review, mechanical impedance is the amount that a system opposes an applied force. Mechanical impedance is a frequency-dependent measure, in that, as frequency is increasing, a negative slope signifies that impedance is driven by the property of stiffness, and a positive slope signifies that impedance is driven by mass. The frequency at which the slope equals zero is the resonant frequency. Here, the stiffness and mass properties are equal. The impedance magnitude at resonant frequency is representative of the resistance, or friction, of the system. Håkansson, Carlsson, and Tjellstrom (1986) developed a model of the mechanical impedance of the skin and skull by measuring impedance of the skin-covered skull and impedance of the skull through the BAHS abutment. They found that the impedance of the skull is substantially higher than the impedance of the skin, and therefore a measurement of the impedance of the skin-covered skull primarily represents the impedance of the skin and subcutaneous tissue, and the skull bone is not incorporated into the measurement. The authors developed a cascade model, predicting that when vibratory acceleration is applied transcutaneously, the energy is shunted through the skin and the resulting force is applied to the skull. Therefore, mechanical impedance of the skin can influence the magnitude of force that is outputted from the transducer (Lundgren, 2010).

Findings from the current study indicate that for low frequencies, the adult skull has higher impedance compared to the skull of infants up to 11 months of age when measured at the temporal bone and higher impedance compared to infants and children up to 4 years of age when measured at the forehead. For the high frequencies, a different pattern emerged. Infants up to 11 months had higher impedance than adults (and all other age groups), but only at the temporal bone.

Because the mechanical impedance of the skin is lower for infants than adults for low frequencies, it is speculated that a larger force is applied to the skin for infants. An important factor to consider is the difference in skin and subcutaneous tissue properties between infants and adults, most notably the difference in thickness and density. Adult skin contains more collagen (Vitellaro-Zuccarello, Cappelletti, Rossi, & Sari-Gorla, 2005), and elastin fibres in adult skin contain a denser matrix (Seidenari, Giusti, Bertoni, Magnoni, & Pellacani, 2000). Infant skin also bind and retain water easier than adult skin, allowing infant skin to have higher compressibility (Nikolovski, Stamatas, Kollias, & Wiegand, 2008; Seidenari, Giusti, Bertoni, Magnoni, & Pellacani, 2000; Stamatas, Nikolovski, Mack, & Kollias, 2011). The epidermis of infants also contains fewer skin cells overall. Infant skin and subcutaneous tissue is thinner than adult skin and subcutaneous tissue (Stamatas, Nikolovski, Mack, & Kollias, 2011). These infant-adult differences in skin properties may contribute to the difference in both stiffness and damping, which is apparent in the impedance magnitude for low frequencies and at resonant frequency. Although findings of the current study cannot provide evidence regarding how energy is distributed beyond the skin and subcutaneous tissue, the speculation that a greater amount of force is passing through the skin and subcutaneous tissue leads to the prediction is that a larger force may be transmitted to the ipsilateral cochlea. These results correspond well with frequency-specific trends of the maturation of bone-conduction sensitivity (Casey, 2012; Foxe & Stapells, 1993; Hulecki & Small, 2011; Nousak & Stapells, 1992; Small & Stapells, 2008a; Small & Stapells, 2006; Stapells & Ruben, 1989; D. W. Swanepoel, Ebrahim, Friedland, Swanepoel, & Pottas, 2008; Vander Werff, Prieve, & Georgantas, 2009).

The difference in impedance between infants in Group A and Groups C, D and adults for high frequencies at the temporal bone may also contribute to infant-adult differences in bone-conduction sensitivity. One explanation that may account for these differences is that the compressibility of the infant skin may allow for easier coupling of the full area of the contact plate with the skin-covered temporal bone. Although the flattest part of the temporal bone was used for data collection for all participants, older children and adults have a more rigid structure to their temporal bone, where complete coupling could be more difficult. As stated previously, impedance for high frequencies, dominated by mass, is sensitive to changes in contact area (Håkansson, Carlsson, & Tjellstrom, 1986), and a larger contact area produces a larger mass component. Although this result is not explained by physical differences in skin properties, the implication for contact area still has significant consequences to infant-adult bone-conduction threshold differences and may perhaps contribute to poorer bone-conduction sensitivity at higher frequencies.

Finally, for the resonant frequency, infants in Group A (1-10 months) have lower resonant frequencies than for children 2 years of age and older and adults. Resonant frequency for Group B (11-24 months of age) falls between that for young infants and children age 2-4 years, but is not significantly different from either group. This finding is likely due to the large amount of skin maturation that occurs within the first year of life (Stamatas, Nikolovski, Mack, & Kollias, 2011). Overall, the damping of low frequency stimuli is lower for the infant skin and subcutaneous tissue compared to adults, which furthers strengthens the hypothesis that a connection lies between the properties of the infant skin and subcutaneous tissue and the greater sensitivity to bone-conduction stimuli.

3.2.3 Implications for calibration of the bone-conduction transducer

It is reasonable to conclude that based on the differences in mechanical impedance between infants and adults, some of the infant-adult differences in threshold are likely due to biases in calibration. The artificial mastoid is a tool used for calibrating bone-conduction transducers. To calibrate transducers for bone-conduction audiometry, reference-equivalent threshold force levels (RETFLs) have been determined and standardized for each audiometric frequency. RETFLs are conversion factors between the average force level measured at threshold from a large sample of normal hearing adults and 0 dB HL measured on the artificial mastoid. The artificial mastoid was constructed to contain the properties of an adult skull, including the mechanical impedance of the skin-covered mastoid; however, Cortes (2002) found that the mechanical impedance from the artificial mastoid did not fall within the confidence intervals from her average impedance recordings across 30 adult skulls. In addition, Lundgren (2010) calculated a bias of 0-6 dB re 1 µN/V of output force between the median adult skin-covered skull and the artificial mastoid due to differences in mechanical impedance. Although there is an overall bias error evident from bone-conduction calibration, all transducers are calibrated to one standardized tool, and therefore, the overall difference in impedance is not a significant problem as long as there is minimal variability in impedance across subjects. This is not the case, as shown by Flottorp and Solberg (1976), who found that deviation of mechanical impedance from their adult group contributed to 5-6 dB of output force variation, with the greatest deviation occurring at the resonant peaks of the bone-conduction transducer. In addition, Lundgren (2010) investigated whether individual adult variability of impedance for the skin-covered skull would affect the variability of output force from the B-71 bone-conduction transducer. He calculated that a standard deviation of 2.4 dB for adult mechanical impedance accounts for a standard deviation of 0-5 dB for force and acceleration output from the B-71 transducer, depending on the frequency measured. Lundgren (2010) concluded that individual variability of the impedance of an adult mastoid could account for significant variability of force, acceleration, and apparent power. Because all bone-conduction transducers are calibrated to one reference (i.e., the artificial mastoid), the variability in mechanical impedance is a more pertinent factor to bone-conduction audiometry than the bias error between impedance of artificial mastoid and the average adult skull.

To explore the implications from the current study, data from the temporal bone at 5.4 N was examined. As mentioned, the youngest infant group had lower mean mechanical impedance compared to adults across selected low frequencies from 125-1000 Hz, with differences decreasing from 10.8 to 3.9 dB (calculated from Table 2.3). Across selected high frequencies from 1995 to 10000 Hz, differences in mean mechanical impedance between the youngest age group and adults were 0.1, -0.4, -0.3, -2.0, -6.2, and -3.4 dB (calculated from Table 2.5). Therefore, output force from a bone-conduction transducer calibrated using an artificial mastoid with adult RETFL values could produce a significantly different output force when it is applied to an infant head.

Not surprisingly, the standard deviation for infants and children was generally higher compared to adults (Tables 2.3 and 2.5), which may partly explain the large variability observed in both physiological and behavioural measures of threshold estimation for infants (Casey, 2012; Hulecki & Small, 2011; Small & Stapells, 2006). In order to better predict normal bone-conduction thresholds for infants, alternate RETFLs are required to account for the potential differences in output force due to the differences in mechanical impedance.

A fundamental concept for bone-conduction calibration is that output force levels correspond to hearing levels. However, it is important to realize that that mechanical impedance may not influence output force in a linear way. RETFLs calculated for direct (percutaneous) bone conduction standardized on a skull simulator were 2-10 dB different from the RETFLs calculated for transcutaneous bone conduction on an artificial mastoid (Carlsson, Håkansson, & Ringdahl, 1995). Behavioural thresholds were lowered by 10-20 dB when the signal was delivered percutaneously compared to transcutaneously (Håkansson, Tjellström, & Rosenhall, 1984), and as stated previously, there are considerable differences in mechanical impedance between the skull and the skin-covered skull (Håkansson, Carlsson, & Tjellstrom, 1986). Some researchers have predicted that output force is not greatly influenced by properties of the skin and subcutaneous tissue; whereas, measures of acceleration are largely influenced by the skin and subcutaneous tissue (Carlsson, Håkansson, & Ringdahl, 1995; Håkansson, Tjellström, & Rosenhall, 1985). Based on the results of the present study, there are two possible approaches to correct for infant-adult differences in RETFLs. First, because there are significant differences in mechanical impedance between infants and adults, one method could be to calculate infant RETFLs by standardizing infant force thresholds to a specialized artificial mastoid with properties of the infant skin-covered skull. However, because impedance values increase with maturation, this would involve creating a unique artificial mastoid for each age group until impedance values reach adult levels. A second method could be to use infant force thresholds calibrated on the standard artificial mastoid designed with adult skull properties. This would essentially equate to a correction factor between infant and adult thresholds in dB HL. Therefore, it may be just as helpful for bone-conduction audiometry for researchers to create a correction factor between infant bone-conduction thresholds and adult thresholds as it would be to develop a unique artificial mastoid for infants. However, as mentioned by Lundgren (2010), the overall difference between the artificial mastoid and the mean impedance of the adult population is not as significant as the variability of mechanical impedance within the population, and therefore even after computing a correction factor for infants, individual differences within the infant populations will likely create variation in the corrected output force amounting to variation in bone-conduction thresholds.

3.3 Implications for Fitting and Verification of the BAHS

There are significant limitations to current methods used to verify that the output from the BAHS is sufficient for hearing speech at a normal conversational level. The technique most widely used clinically is a measure of aided thresholds in the soundfield. It is well established that aided thresholds for both air-conduction hearing aids and BAHS are neither reliable nor valid methods of assessing performance of the hearing device. Specifically, the interaction with noise floor and the non-linear processors produce a measure that is not a true aided threshold (Nicholson, Christensen, Dornhoffer, Martin & Smith-Olinde, 2011). In addition, measuring threshold cannot predict the BAHS performance at a typical conversational level (Dillon, 2001).

As described in Chapter 1, methods for BAHS verification that are under investigation include measurements of sound pressure in the ear canal during stimulation from the BAHS, and acceleration or output force from the BAHS (Hodgetts, Håkansson, Hagler, & Soli, 2010). The procedure using sound pressure in the ear canal as a verification tool began with a probe tube measurement in the ipsilateral ear when the BAHS transducer was stimulated at the individual's threshold and loudness discomfort levels. Then, sound pressure for the long-term average speech spectrum for average speech was measured to verify that the output is situated within the dynamic range of the individual. The results from the current study suggest that sound pressure in the ear canal as a verification tool cannot accurately predict sound that is reaching the cochlea, and also cannot provide accurate frequency-specific information. This conclusion was based on the differences observed between the measure of sound pressure in the ear canal from the current study and the predicted attenuation for infants using physiological measures of hearing. Other factors that may affect the sound pressure measurement include interactions with the noise floor (Hodgetts, Håkansson, Hagler, & Soli, 2010), pure-tone resonances of the ear canal (Kruger, 1987; Kruger & Ruben, 1987), the nature of how the ear canal radiates bone-conducted sound differently across frequencies (Huizing, 1960a), and the interference with airborne sound radiating from the bone-conduction transducer (Lightfoot, 1979). For the current study, 4000 Hz was particularly difficult to measure in the ear canal in adult pilot subjects, and stimulus intensity was increased to 60 dB HL for that reason. Clinically, the unequal measurements across frequencies will affect the calculated gain applied from a non-linear prescription formula. In conclusion, based on the results from the current study, as well as other factors listed in Chapter 1, this method does not seem to be a viable option for verification of output from the BAHS, particularly for children and infants. Also, as mentioned earlier, many infants and young children who need a BAHS have atresia, which would not allow probe tube measures in the ear canal.

A verification method with more clinical potential for all ages is a measure of output force from the BAHS. This technique can be completed with the BAHS directly on the skull of the user; however, this would require specialized and expensive equipment that is not widely available and technical skills that would require extensive training (Hodgetts, Håkansson, Hagler, & Soli, 2010; Hodgetts, 2011). An alternate approach is to measure output force through the BAHS when attached to an artificial mastoid or skull simulator (Håkansson & Hodgetts, 2009). This procedure for verifying aided output from the BAHS begins with collecting *in situ* force thresholds with the BAHS, and targets are calculated using a prescription formula in reference to those thresholds. The BAHS in then connected to a skull simulator in the test box of a hearing aid analyzer, and output force is measured to verify that targets are met (Håkansson & Hodgetts, 2009; Hodgetts, 2010b). The implications for using this technique for infants and young children from the current findings are discussed in the following section.

3.3.1 Skull simulator and artificial mastoid

One aim of the current study was to determine whether the BAHS verification tools should be used for measuring output force from the soft band BAHS for infants and young children. The skull simulator was developed as a tool for verifying the output from the BAHS when it is intended for percutaneous use. Unlike the artificial mastoid, the skull simulator was not designed to contain properties of an adult skull, yet is meant to simulate the skull so that output force from BAHS on the skull simulator is similar to output force from the BAHS on an adult skull. Specifically, because the mechanical impedance of the skull is substantially greater than the impedance of the BAHS transducer, and as long as the impedance of the skull simulator is similarly large, the exact impedance value of the skull simulator is irrelevant. In addition, any intersubject variability of the impedance of the skull becomes negligible (Håkansson & Carlsson, 1989).

In contrast, variability of the skin-covered skull affects the output force from a bone-conduction transducer, as described earlier. Therefore, in contrast to skull simulators, the precise impedance of the artificial mastoid is relevant for collecting measurements of force output. Skull simulators should not be used independently for verifying the BAHS for transcutaneous use; however, a mini-artificial mastoid has been developed for use in conjunction with the TU-1000 (Stenfelt & Håkansson, 1998). When the BAHS is attached to the artificial mastoid, a measure of output force is recorded with reference to the impedance magnitude of the artificial mastoid. If the BAHS is then coupled to the head of the user, any individual variability in output force could largely affect the bone-conducted sound reaching the cochlea of the individual. For air-conduction hearing aids, when verifying that the aided output from the hearing aid in a 2-cc coupler is reaching calculated targets, a real-ear to coupler difference (RECD) is applied to the aided output (Tharpe, Sladen, Huta, & McKinley Rothpletz, 2001). For bone-conduction, this could be called a real-head to artificial mastoid difference.

As mentioned previously, it is not feasible that the skin-covered skull impedance is collected for each individual fitted with a BAHS, but even if calculating an individual correction factor is not possible, the bias error between the mean mechanical impedance and the artificial mastoid (Lundgren, 2010) still warrants the use of a correction factor based on means, similar to the average RECD used for fitting air-conduction hearing aids (Bagatto, Scollie, Seewald, Moodie, & Hoover, 2002). Findings from the current study have shown that differences in mean impedance between Group A and adults ranged from -6.2 to 10.8 dB depending on the frequency. Therefore, it is suggested that these differences are taken into account through a correction factor when measuring output force from the BAHS when coupled to the artificial mastoid.

3.3.2 BAHS placement on the skull

As previously discussed, the mastoid is the preferred and standardized location for the transducer placement for bone-conduction audiometry (American National Standards Institute [ANSI], 1996). Physiological measures of infant hearing also revealed that wave V latencies were longer (Stuart, Yang, & Stenstrom, 1990) and ASSR thresholds were poorer (Small, Hatton, & Stapells, 2007) when the bone-conduction transducer was positioned on the forehead compared to the ipsilateral temporal bone or mastoid. Although studies have repeatedly shown that bone-conduction hearing is most sensitive with the transducer on the temporal bone or mastoid compared to the forehead, manufacturer guidelines for fitting soft band BAHS have recommended that parents position the device on the forehead if necessary.

Based on findings from Experiment 1, infants and children generally had more attenuation of bone-conducted sound when the transducer was placed on the forehead compared to the contralateral temporal bone. However, adults did not show differences between skull locations, nor did infants in Group A using this measurement of transcranial attenuation. For Groups B, C, and D, the greater attenuation from the forehead compared to the contralateral temporal bone is consistent with a model proposed by Stenfelt, Håkansson and Tjellström (2000) and Stenfelt and Goode (2005b). They stated that acceleration on the cochlea was driven by vibration from three orthogonal directions, but was dominated by vibration from the medial (x) direction for both ipsilateral and contralateral stimulation. They found that attenuation was minimal for frequencies below 2000 Hz, and 5-10 dB for high frequencies, similar to attenuation studies using behavioural thresholds (Kirikae, 1959b). In contrast, cochlear vibrations with the transducer on the forehead originated primarily from the cranial direction (i.e., from the top of the head), and efficient transmission of sound is poorest when the transducer is stimulating the midsagittal plane of the skull, encompassing the forehead (Stenfelt, Håkansson, & Tjellström, 2000; Stenfelt & Goode, 2005b). It is possible that the cochlear sensitivity to a certain plane of vibration is different between infants, children and adults, explaining some of the large differences in transcranial attenuation found in previous studies of hearing. From the current findings, age-dependent patterns related to transducer position could not be completely teased out from this measure of attenuation. It is likely that smaller than expected estimates of differences in transcranial attenuation between positions for infants and adults did not fully account for all the energy crossing the skull due to methodological limitations.

Skull position was also investigated for measures of impedance, and it was observed that the largest effect of skull position was for high frequencies, which is driven by differences in mass. These findings are in contrast to those reported by Corliss and Koidan (1955), Smith and Suggs (1976) and Flottorp and Solberg (1976) who found negligible differences of impedance between forehead and temporal bone. Flottorp and Solberg (1976) commented that comparing impedance between skull positions should be made with caution because it is more difficult to ensure stable contact with the temporal bone compared to the forehead, and therefore, an effect of position on impedance may be due to methodological differences instead of anatomical differences. For the current study, contact with the head was also typically easier at the forehead location for most age groups. The forehead had higher impedance than the temporal bone for the mass-dominated high frequencies, but not for the stiffness-dominated low frequencies, which may be explained from the contact plate making full contact with the forehead but not for the temporal bone. As described previously, contact area has a larger effect on high frequencies for both bone-conduction hearing (Khanna, Tonndorf, & Queller, 1976; Watson, 1938) and impedance (Håkansson, Carlsson & Tjellström, 1986). Interestingly, a difference between skull positions for the high frequencies was not present for Group A, for whom coupling of the entire surface area of the contact plate was likely better achieved due to the compressible nature of the skin covering the head and a flatter temporal bone surface. In contrast, older children and adults had more rigid and uneven temporal bones, making complete contact more difficult.

In conclusion, results from the current study have indicated that, even when using sound pressure in the ear canal as a measure of transcranial attenuation, attenuation from the forehead is greater than from the contralateral temporal bone for most young children. Mechanical impedance was also higher for most of the infant groups for the high frequencies. Therefore, the findings of the study in combination with the physiological data suggest that the forehead should be avoided for soft band BAHS and the temporal bone, with careful placement to optimize contact between the BAHS and the head, is the preferred position.

3.3.3 Contact force of BAHS soft band

Measures of mechanical impedance across different contact forces indicated that more prominent effects of force were observed for low frequencies compared to high frequencies. Notably, the differences across force for high frequencies were only present at the temporal bone between 5.4 and 4 N and between 5.4 and 2 N. Similar to the discussion on skull position and impedance, differences between contact forces for the high frequencies may be due to increased surface area of coupling for 5.4 N compared to lower forces on the temporal bone. Because it was easier to achieve complete contact on the forehead, no differences between contact forces were observed for high frequencies.

For low frequencies, a higher impedance magnitude was observed for increasing contact force. Although the trend observed in the current study is consistent with previous studies of impedance (Cortes, 2002), an increase in impedance with increasing contact force does not explain the improvement in bone-conduction hearing observed from previous studies of behavioural and physiological measures (Lau, 1986; Watson, 1938; Whittle, 1965; Yang, Stuart, Stenstrom, & Hollett, 1991). In addition, Hodgetts, Scollie and Swain (2006) found negligible effects of contact force on measures of BAHS output force. Therefore, the effect observed for mechanical impedance may not be large enough to influence the output force from the transducer. Flottorp and Solberg (1976) commented that the static force of coupling to the head could cause a shift in the amount of liquid underneath the skin altering the values of the mechanical impedance. However, it is unknown whether this shift causes a change in hearing threshold. More research is required to assess the correlation between measures of threshold and mechanical impedance.

Previous studies manipulating the contact force of a bone-conduction transducer have only been completed on sleeping infants, adults, or artificial mastoids. This was the first study to investigate contact force on alert infants and young children. An important observation was that a larger number of movement artifacts was present for the lower contact forces compared to 5.4 N. In addition, larger standard deviations were generally observed for 4 and 2 N across age groups for many of the frequencies in Tables 2.3 and 2.5. Therefore, temporary changes in mechanical impedance for lower contact forces may indicate that a less stable output force could be delivered to the active child. Therefore, findings from the current study support recommendations from previous studies that have suggested the soft band should be tightened so that it is secure without causing discomfort (Hodgetts, Scollie, & Swain, 2006; Hodgetts, 2010b).

3.4 Candidacy for the BAHS Soft Band

As discussed in Chapter 1, the candidacy criteria for fitting an individual with a BAHS have changed as technology advances. Originally, the BAHS was developed for individuals with a bilateral conductive or mixed hearing loss with a mild-moderate sensorineural component at most. Recent investigations have inquired about the benefits of a BAHS for individuals with unilateral losses, including both severe-profound sensorineural hearing loss [single-sided deafness (SSD)] and conductive hearing loss. In addition, studies have started to analyze the potential benefits from bilateral BAHS.

It has now been established that infants have greater transcranial attenuation than adults based on the physical measures of sound pressure in the ear canal from the current study, and physiological measures from previous studies (Foxe & Stapells, 1993; Small & Stapells, 2008b; Stapells & Ruben, 1989; Yang, Rupert, & Moushegian, 1987). These findings have implications regarding how the BAHS will transmit bone-conducted sound to the contralateral ear.

For individuals with SSD, the BAHS is worn on the poor ear, and sound is intended to be contralaterally routed to the normal ear. The objective for individuals with SSD is that the contralateral routing of the signal will reduce the head shadow effect to improve understanding speech in noise and localization of sounds. For individuals with unilateral conductive losses, the BAHS is worn on the poor ear, and sound is directed to the normal functioning ipsilateral cochlea. The controversial aspect of this fitting is that sound could also be routed across the head to the ear without any conductive loss, possibly creating interference and a detriment to binaural hearing (Snik et al., 2005; Snik, Leijendeckers, Hol, Mylanus, & Cremers, 2008; Stenfelt, 2005). The same concern is apparent for bilateral BAHS fittings (e.g., Stenfelt, 2005). Stenfelt (2012) remarked that interfering signals causes binaural bone-conduction hearing to be less effective than binaural air-conduction hearing, and individuals with higher transcranial attenuation are likely to have the most success with bilateral BAHS. This speculation regarding transcranial attenuation is also appropriate for candidates with unilateral conductive hearing loss, where those with higher transcranial attenuation will likely receive more benefit from the BAHS because the signal will not interfere with the contralateral normal hearing ear. For candidates with SSD, the BAHS will likely provide more benefit to individuals with lower transcranial attenuation because the contralateral routing of the signal will be more effective.

The data from the current study, in combination with previous studies, using physiological measures of hearing suggest that infants with unilateral conductive hearing loss will receive less contralateral interference at the normal hearing ear from the BAHS compared to adults; however, infants with SSD may not benefit as well as adults from the contralateral routing of the signal. In addition, bilateral BAHS are probably more effective for infants with bilateral conductive or mixed hearing loss compared to adults, due to their greater transcranial attenuation. Because these outcomes are only speculative based on measures of transcranial attenution, more research is required to assess outcome measures for fitting of the BAHS for these potential candidates.

3.5 Conclusions

The current study has contributed important findings to the field of bone-conduction research, particularly to our understanding of the maturation of bone-conduction hearing. As mentioned, previous studies have only measured the impedance of the skin-covered skull for children as young as 9 years of age during the collection of impedance measurements for the purpose of developing an artificial mastoid (Flottorp & Solberg, 1976). Mechanical impedance data are now available for individuals age 7 years and younger and give a more complete picture of how mechanical impedance can change across the lifespan. These findings contribute to an explanation for infant-adult frequency-specific sensitivity differences to bone-conduction stimuli and will eventually aid in developing an optimal BAHS fitting and verification protocol for infants and young children. The portable hand-held tool developed for the purpose of this study was essential in collecting impedance values for infants and young children and will be helpful for collecting impedance measures in future research.

The present findings showed that transcranial attenuation decreased throughout maturation, which is consistent with previous physiological studies (e.g., Small & Stapells, 2008b). Therefore, properties of the skull are likely contributing to the large transcranial attenuation found in infants. In addition, the maturation of bone-conduction sensitivity may be explained, at least in part, by mechanical impedance. Findings from the current study showed an increase in impedance throughout maturation for the low frequencies, which relates to increasing stiffness of the skin and subcutaneous tissue. It can be speculated that a lower impedance results in a greater magnitude of output force from the transducer and subsequently, lower thresholds. The bias in calibration of the bone-conduction transducer to an artificial mastoid developed with adult properties in mind could be responsible for the infant-adult differences observed in low-frequency bone-conduction sensitivity. Additionally, for high frequencies, infants in the youngest age group had higher impedance at the temporal bone compared to adults and children age 2 years and older. These findings may explain the poorer bone-conduction sensitivity for infants observed at 2000 Hz (Small & Stapells, 2006; 2008a). In conjunction with previous measures of infant physiological thresholds for bone-conducted stimuli, these results suggest that a correction factor should be applied to the RETFLs used for calibrating the bone-conduction transducer for adults.

These findings have implications for the fitting and verification of the BAHS soft band for infants and young children. Similar to the correction factor that will be necessary for bone-conduction audiometry, when using an artificial mastoid as a verification tool for the BAHS, a correction factor should be applied to account for substantial differences in mechanical impedance. If an artificial mastoid is used to verify that the output force from the BAHS matches calculated targets, a risk of over-amplification for low frequencies may occur when applying the device to the head an infant with lower mechanical impedance.

Second, the results of the current study on infants and young children are consistent with previous studies on adults showing that attenuation of bone-conducted sound was greater when the transducer was placed on the forehead compared to the contralateral temporal bone (Stenfelt, Håkansson, & Tjellström, 2000; Stenfelt & Goode, 2005b). However, results from the present study also suggest that poor coupling of the device to the head could affect impedance. Therefore, it is recommended that the BAHS soft band be positioned on the temporal bone, ensuring that the plastic plate from the soft band makes complete contact with the head.

Third, an increasing contact force resulted in increasing mechanical impedance for low frequencies, which is consistent with previous studies on adults (Cortes, 2002). More variability in impedance was also observed for the lower contact forces, which may affected the output force from the BAHS. Therefore, it is recommended that the soft band be tightened to a level where a consistent signal is delivered to the head and the output force is not greatly affected by the active lifestyle of a young child.

Finally, results from the study provided insight regarding fitting the soft band BAHS for infants and young children with unilateral conductive hearing loss or unilateral severe-profound sensorineural hearing loss, as well as implications for fitting bilateral BAHS for individuals with bilateral conductive or mixed hearing loss. Transcranial attenuation was greatest for young infants and decreased throughout maturation; therefore, fitting procedures that make use of the contralateral routing of the signal, such as a BAHS CROS for SSD, would be less effective for infants compared to adults. However, when the signal is targeting the ipsilateral cochlea and contralateral signals could cause interference to binaural hearing, such as bilateral BAHS or a BAHS for an individual with a unilateral conductive hearing loss, these fittings would be more effective for infants who have greater transcranial attenuation of bone-conducted sound compared to adults.

3.5.1 Future studies

The current study was a preliminary investigation of the mechanisms responsible for infant-adult differences in bone-conduction hearing. More research is necessary to determine the precise mechanism by which sound is transmitted across the skull. Future research on transcranial attenuation for infants could involve measures of the acceleration of the cochlea, which is often used to estimate bone-conduction hearing and investigate bone-conduction mechanisms in adults (Eeg-Olofsson, Stenfelt, Tjellström, & Granström, 2008; Eeg-Olofsson, 2012; Eeg-Olofsson, Stenfelt, & Granstrom, 2011; Reinfeldt, Stenfelt, Good, & Håkansson, 2007; Stenfelt, Håkansson, & Tjellström, 2000; Stenfelt, 2012; Stenfelt & Goode, 2005b). Physiological studies of masking or behavioural studies using infants and young children with severe-profound unilateral hearing loss could also provide valuable insight to the maturation of transcranial attenuation. Research is necessary to understand the connection between the mechanical impedance differences between infants and adults and the output force presented at the head to provide more insight to how the differences in impedance affect the sensitivity to pure-tone and speech stimuli. In addition, a larger normative sample is important for determining the correction factor in mechanical impedance between infants of different ages, adults, and the artificial mastoid used for verifying aided output from the BAHS. Mechanical impedance measurements on the temporal bone of children and infants with craniofacial anomalies would also be important for determining relevant differences in the fitting and verification protocol for infants and children typically fit with a soft band BAHS. Finally, more research is needed to determine the benefit of bilateral BAHS for infants and children of different ages, and the effectiveness of the BAHS for young children with unilateral conductive hearing loss or SSD. In the end, the development of a fitting and verification protocol that uses objective and standardized verification tools of a similar caliber to those standards and tools available for fitting air-conduction hearing aids is desired. As fitting protocols and verification tools for the BAHS become more readily available in the clinic, it is important that additional protocols are specifically developed for infants and young children with the BAHS soft band.

REFERENCES

Abdala, C. (2001). Maturation of the human cochlear amplifier: Distortion product otoacoustic emission suppression tuning curves recorded at low and high primary tone levels. *The Journal of the Acoustical Society of America, 110*(3), 1465-1476. doi: 10.1121/1.1388018

Abdala, C., & Keefe, D. H. (2012). Morphological and functional ear development. In L. A. Werner, R. R. Fey & A. N. Popper (Eds.), *Human auditory development* (pp. 19-59). New York: NY: Springer Science+Business Media.

Abdala, C., & Sininger, Y. S. (1996). The development of cochlear frequency resolution in the human auditory system. *Ear & Hearing, 17*(5), 374-385.

Abdala, C., Sininger, Y. S., Ekelid, M., & Zeng, F. G. (1996). Distortion product otoacoustic emission suppression tuning curves in human adults and neonates. *Hearing Research, 98*(1), 38-53. doi: 10.1016/0378-5955(96)00056-1

Adell, R., Lekholm, U., Rockler, B., & Brånemark, P. I. (1981). A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. *International Journal of Oral Surgery, 10*(6), 387-416. doi: 10.1016/S0300-9785(81)80077-4

American Academy of Audiology [AAA]. (2004). Pediatric amplification protocol. Retrieved November/2, 2012, from http://www.audiology.org/resources/documentlibrary/Documents/pedamp.pdf

American National Standards Institute [ANSI]. (1987). *Mechanical coupler for measurement of bone vibrators (ANSI S3.16-1987)*. New York, NY: American National Standards Institute.

American National Standards Institute [ANSI]. (1996). *American national standard specifications for audiometers (ANSI S3.6-1996)*. New York, NY: American National Standards Institute.

Anson, B. J., & Donaldson, J. A. (1981). *Surgical anatomy of the temporal bone*. Philadelphia, PA: W.B. Saunders Company.

Bagatto, M. P. (2001). Optimizing your RECD measurements. *The Hearing Journal, 54*(9), 32-36. doi: 10.1097/01.HJ.0000293458.15919.d6

Bagatto, M. P., Moodie, S., Scollie, S., Seewald, R., Moodie, S., Pumford, J., & Liu, K. P. R. (2005). Clinical protocols for hearing instrument fitting in the desired sensation level method. *Trends in Amplification, 9*(4), 199-226. doi: 10.1177/108471380500900404

Bagatto, M. P., & Scollie, S. (2010). Current approaches to the fitting of amplification to infants and young children. In R. Seewald, & A. M. Tharpe (Eds.), *Comprehensive handbook of pediatric audiology* (pp. 527-552)

Bagatto, M. P., Scollie, S., Seewald, R. C., Moodie, K. S., & Hoover, B. M. (2002). Real-ear-to-coupler difference predictions as a function of age for two coupling procedures. *Journal of the American Academy of Audiology, 13*(8), 407-415.

Bagatto, M. P., Seewald, R. C., Scollie, S., & Tharpe, A. M. (2006). Evaluation of a probe-tube insertion technique for measuring the real-ear-to-coupler difference (RECD) in young infants. *Journal of the American Academy of Audiology, 17*(8), 573-581.

Bárány, E. (1938). Chapter II: Critical survey of some previous investigations. *Acta Oto-Laryngologica, 26*, 6-26. doi: 10.3109/00016483809122310

Bárány, E. (1938). Chapter XIII: Inertia movements of the chain of ossicles and of the liquid in the labyrinth, when the skull is rotated. *Acta Oto-Laryngologica, 26*, 114-120. doi: 10.3109/00016483809122321

Beentjes, B. I. J. (1972). The cochlear aqueduct and the pressure of cerebrospinal and endolabyrinthine fluids. *Acta Oto-Laryngologica, 73*(2-6), 112-120. doi: 10.3109/00016487209138919

Bentler, R. A. (1989). External ear resonance characteristics in children. *Journal of Speech and Hearing Disorders, 54*(2), 264-268.

Bosman, A. J., Hol, M. K. S., Snik, A. F., Mylanus, E. A. M., & Cremers, C. W. R. J. (2003). Bone-anchored hearing aids in unilateral inner ear deafness. *Acta Oto-Laryngologica, 123*(2), 258-260. doi: 10.1080/000164580310001105

Breisch, E., Haas, E. A., Masoumi, H., Chadwick, A. E., & Krous, H. F. (2010). A morphometric analysis of the infant calvarium and dura. *Forensic Science, Medicine, and Pathology, 6*(4), 249. doi: 10.1007/s12024-009-9136-2

Brinkman, W. F. B., Marres, E., & Tolk, J. (1965). The mechanism of bone conduction: An experimental study. *Acta Oto-Laryngologica, 59*(2-6), 109-115. doi: 0.3109/00016486509124544

Brown, A. M., Sheppard, S. L., & Russell, P. T. (1994). Acoustic distortion products (ADP) from the ears of term infants and young adults using low stimulus levels. *British Journal of Audiology, 28*(4-5), 273-280. doi: 10.3109/03005369409086577

Burrell, S. P., Cooper, H. C., & Proops, D. W. (1996). The bone anchored hearing aid—The third option for otosclerosis. *The Journal of Laryngology & Otology, 110*(21), 31-37. doi: 10.1017/S0022215100136242

Byrne, D., Dillon, H., Ching, T., Katsch, R., & Keidser, G. (2001). NAL-NL1 procedure for fitting nonlinear hearing aids: Characteristics and comparisons with other procedures. *Journal of the American Academy of Audiology, 12*(1), 37-51.

Calandruccio, L., Fitzgerald, T. S., & Prieve, B. A. (2006). Normative multifrequency tympanometry in infants and toddlers. *Journal of the American Academy of Audiology, 17*(7), 470-480. doi: 10.3766/jaaa.17.7.2

Carhart, R. (1950). Clinical application of bone conduction audiometry. *Archives of Otolaryngology—Head & Neck Surgery, 51*(6), 798-808. doi: 10.1001/archotol.1950.00700020824003

Carlsson, P., & Håkansson, B. (1997). The bone-anchored hearing aid: Reference quantities and functional gain. *Ear & Hearing, 18*(1), 34-41.

Carlsson, P., Håkansson, B., & Ringdahl, A. (1995). Force threshold for hearing by direct bone conduction. *The Journal of the Acoustical Society of America, 97*, 1124-1129. doi: 10.1121/1.412225

Casey, K. (2012). *Comparisons of auditory steady-state response and behaviour air- and bone-conduction thresholds in infants and adults with normal hearing.* (Unpublished Master of Science). University of British Columbia, Vancouver, B.C.

Christensen, L., & Dornhoffer, J. L. (2008). Bone-anchored hearing aids for unilateral hearing loss in teenagers. *Otology & Neurotology, 29*(8), 1120-1122. doi: 10.1097/MAO.0b013e31818af398

Christensen, L., Smith-Olinde, L., Kimberlain, J., Richter, G. T., & Dornhoffer, J. L. (2010). Comparison of traditional bone-conduction hearing aids with the Baha system. *Journal of the American Academy of Audiology, 21*(4), 267-273. doi: 10.3766/jaaa.21.4.5

Cochlear. (2011). Baha 3 A bone-conduction hearing solution: Fitting guide. Retrieved from http://professionals.cochlearamericas.com/sites/default/files/resources/BUN027%20ISS1%20Baha%20fitting%20guide.pdf

Cochlear. (2011). Cochlear Baha 3 A bone-conduction hearing solution: Candidate selection guide. Retrieved November/10, 2012, from http://professionals.cochlearamericas.com/sites/default/files/resources/BUN070%20ISS1%20DEC11%20Candidate%20Selection%20Guide.pdf

Cooper, H. R., Burrell, S. P., Powell, R. H., Proops, D. W., & Bickerton, J. A. (1996). The Birmingham bone anchored hearing aid programme: Referrals, selection, rehabilitation, philosophy and adult results. *The Journal of Laryngology and Otology.Supplement, 21*, 13-20. doi: 10.1017/S0022215100136229

Corliss, E. L. R., & Koidan, W. (1955). Mechanical impedance of the forehead and mastoid. *The Journal of the Acoustical Society of America, 27*, 1164-1172. doi: 10.1121/1.1908150

Cortes, D. (2002). *Bone conduction transducers: Output force dependency on load condition.* (Report No. 122966). Göteborg, Sweden: Chalmers University of Technology.

Dadson, R., Robinson, D., & Greig, R. (1954). The mechanical impedance of the human mastoid process. *British Journal of Applied Physics, 5*(12), 435-442. doi: 10.1088/0508-3443/5/12/304

Dahm, M. C., Shepherd, R. K., & Clark, G. M. (1993). The postnatal growth of the temporal bone and its implications for cochlear implantation in children. *Acta Oto-Laryngologica Supplement, 505*, 1-39. doi: 10.3109/00016489309128539

Davids, T., Gordon, K. A., Clutton, D., & Papsin, B. C. (2007). Bone-anchored hearing aids in infants and children younger than 5 years. *Archives of Otolaryngology—Head & Neck Surgery, 133*(1), 51-55. doi: 10.1001/archotol.133.1.51

de Wolf, M. J. F., Hol, M. K. S., Huygen, P. L. M., Mylanus, E. A. M., & Cremers, C. W. R. J. (2008). Nijmegen results with application of a bone-anchored hearing aid in children: Simplified surgical technique. *Annals of Otology, Rhinology & Laryngology, 117*(11), 805-814.

Declau, F., Cremers, C., & Heyning, P. V. (1999). Diagnosis and management strategies in congenital atresia of the external auditory canal. *British Journal of Audiology, 33*(5), 313-327. doi: 10.3109/03005369909090115

Dickinson, W. W. (2010). Verification of Baha fitting for single-sided deafness: How, what, where and why do we measure? *Seminars in Hearing, 31*(4), 350-365. doi: 10.1055/s-0030-1268034

Dillon, H. (2001). *Hearing aids*. New York, NY: Thieme.

Eby, T. L., & Nadol, J. B. (1986). Postnatal growth of the human temporal bone: Implications for cochlear implants in children. *Annals of Otology, Rhinology, and Laryngology, 95*(4), 356-364.

Eeg-Olofsson, M. (2012). *Transmission of bone conducted sound in the human skull based on vibration and perceptual measures.* (Unpublished Doctor of Philosophy). Institute of Clincial Sciences. Department of Otorhinolaryngology, Göteborg, Sweden. (2077/28248)

Eeg-Olofsson, M., Stenfelt, S., Tjellström, A., & Granström, G. (2008). Transmission of bone-conducted sound in the human skull measured by cochlear vibrations. *International Journal of Audiology, 47*(12), 761-769. doi: 10.1080/14992020802311216

Eeg-Olofsson, M., Stenfelt, S., & Granstrom, G. (2011). Implications for contralateral bone-conducted transmission as measured by cochlear vibrations. *Otology & Neurotology, 32*(2), 192-198. doi: 10.1097/MAO.0b013e3182009f16

Eggermont, J. J., Brown, D. K., Ponton, C. W., & Kimberley, B. P. (1996). Comparison of distortion product otoacoustic emission (DPOAE) and auditory brain stem response (ABR) traveling wave delay measurements suggests frequency-specific synapse maturation. *Ear & Hearing, 17*(5), 386-394.

Egolf, D. P., Feth, L. L., Cooper, W. A., & Franks, J. R. (1985). Effects of normal and pathologic eardrum impedance on sound pressure in the aided ear canal: A computer simulation. *The Journal of the Acoustical Society of America, 78*(4), 1281-1285. doi: 10.1121/1.392897

Federspil, P. A., Tretbar, S. H., Bohlen, F. H., Rohde, S., Glaser, S., & Plinkert, P. K. (2010). Measurement of skull bone thickness for bone-anchored hearing aids: An experimental study comparing both a novel ultrasound system (SonoPointer) and computed tomographic scanning to mechanical measurements. *Otology & Neurotology, 31*(3), 440-446. doi: 10.1097/MAO.0b013e3181d2775f

Feigin, J. A., Kopun, J. G., Stelmachowicz, P. G., & Gorga, M. P. (1989). Probe-tube microphone measures of ear-canal sound pressure levels in infants and children. *Ear & Hearing, 10*(4), 254-258.

Flottorp, G., & Solberg, S. (1976). Mechanical impedance of human headbones (forehead and mastoid portion of the temporal bone) measured under ISO/IEC conditions. *The Journal of the Acoustical Society of America, 59*(4), 899-906. doi: 10.1121/1.380949

Ford, E. H. R. (1958). Growth of the human cranial base. *American Journal of Orthodontics, 44*(7), 498-506. doi: 10.1016/0002-9416(58)90082-4

Foxe, J. J., & Stapells, D. R. (1993). Normal infant and adult auditory brainstem responses to bone-conducted tones. *International Journal of Audiology, 32*(2), 95-109. doi: 10.3109/00206099309071860

Franke, E. K. (1956). Response of the human skull to mechanical vibrations. *The Journal of the Acoustical Society of America, 28,* 1277-1284.

Freeman, S., Sichel, J. Y., & Sohmer, H. (2000). Bone conduction experiments in animals-evidence for a non-osseous mechanism. *Hearing Research, 146*(1-2), 72-80. doi: 10.1016/S0378-5955(00)00098-8

Gilman, S., & Dirks, D. D. (1986). Acoustics of ear canal measurement of eardrum SPL in simulators. *The Journal of the Acoustical Society of America, 80*(3), 783-793. doi: 10.1121/1.393953

Granström, G., Bergström, K., Odersjö, M., & Tjellström, A. (2001). Osseointegrated implants in children: Experience from our first 100 patients. *Otolaryngology—Head and Neck Surgery, 125*(1), 85-92. doi: 10.1067/mhn.2001.116190

Gravel, J. S. (1989). Behavioral assessment of auditory function. *Seminars in Hearing, 10*(3), 216-228.

Håkansson, B., Brandt, A., Carlsson, P., & Tjellström, A. (1994). Resonance frequencies of the human skull in vivo. *The Journal of the Acoustical Society of America, 95*, 1474-1481. doi: 10.1121/1.408535

Håkansson, B., & Carlsson, P. (1989). Skull simulator for direct bone conduction hearing devices. *Scandinavian Audiology, 18*(2), 91-98. doi: 10.3109/01050398909070728

Håkansson, B., Carlsson, P., Tjellström, A., & Liden, G. (1994). The bone-anchored hearing aid: Principal design and audiometric results. *Ear, Nose, & Throat Journal, 73*(9), 670-675.

Håkansson, B., & Hodgetts, W. E. (2009). *Fitting and verification procedure for direct bone conduction hearing devices* (381/60 ed.). MA/US: H04R 29/00.

Håkansson, B., Lidén, G., Tjellström, A., Ringdahl, A., Jacobsson, M., Carlsson, P., & Erlandson, B. (1990). Ten years of experience with the Swedish bone-anchored hearing system. *The Annals of Otology, Rhinology & Laryngology, Supplement 151*, 1-16.

Håkansson, B., Tjellström, A., & Rosenhall, U. (1984). Hearing thresholds with direct bone conduction versus conventional bone conduction. *Scandinavian Audiology, 13*(1), 3-13. doi: 10.3109/01050398409076252

Håkansson, B., Tjellström, A., & Rosenhall, U. (1985). Acceleration levels at hearing threshold with direct bone conduction versus conventional bone conduction. *Acta Oto-Laryngologica, 100*(3-4), 240-252. doi: 10.3109/00016488509104786

Håkansson, B., Tjellström, A., Rosenhall, U., & Carlsson, P. (1985). The bone-anchored hearing aid: Principal design and a psychoacoustical evaluation. *Acta Oto-Laryngologica, 100*(3-4), 229-239. doi: 10.3109/00016488509104785

Håkansson, B. (2003). The balanced electromagnetic separation transducer a new bone conduction transducer. *The Journal of the Acoustical Society of America, 113*(2), 818-825. doi: 10.1121/1.1536633

Håkansson, B., Carlsson, P., & Tjellstrom, A. (1986). The mechanical point impedance of the human head, with and without skin penetration. *The Journal of the Acoustical Society of America, 80*(4), 1065-1075. doi: 10.1121/1.393848

Henry, P., & Letowski, T. R. (2007). *Bone conduction: Anatomy, physiology, and communication.* (DTIC Document No. ADA468459). Aberdeen, MD: Army Research Laboratory.

Hill, T., & Lewicki, P. (2006). *Statistics: Methods and applications*. Tulsa, OK: StatSoft.

Hodgetts, W. E. (2010, November 30). BCAL webinar. Message posted to http://www.irsm-canada.com/research/bone-conduction-amplification.htm

Hodgetts, W. E. (2010). Other implantable devices: Bone-anchored hearing aids. In R. Seewald, & A. M. Tharpe (Eds.), *Comprehensive handbook of pediatric audiology* (pp. 585-598). San Diego, CA: Plural Publishing.

Hodgetts, W. E. (2011). BAAD to the bone: An update of bone anchored amplification devices. *Canadian Academy of Audiology Conference*, Victoria, BC.

Hodgetts, W. E., Håkansson, B., Hagler, P., & Soli, S. (2010). A comparison of three approaches to verifying aided Baha output. *International Journal of Audiology, 49*(4), 286-295. doi: 10.3109/14992020903142676

Hodgetts, W. E., Scollie, S., & Swain, R. (2006). Effects of applied contact force and volume control setting on output force levels of the BAHA® softband. *International Journal of Audiology, 45*(5), 301-308. doi: 10.1080/14992020600582133

Hodgetts, W. E., Hagler, P., Håkansson, B., & Soli, S. D. (2011). Technology-limited and patient-derived versus audibility-derived fittings in bone-anchored hearing aid users: A validation study. *Ear & Hearing, 32*(1), 31-39. doi: 10.1097/AUD.0b013e3181f0b685

Hol, M. K. S., Bosman, A. J., Snik, A. F., Mylanus, E. A. M., & Cremers, C. W. R. J. (2005). Bone-anchored hearing aids in unilateral inner ear deafness: An evaluation of audiometric and patient outcome measurements. *Otology & Neurotology, 26*(5), 999-1006. doi: 10.1097/01.mao.0000185065.04834.95

Hol, M. K. S., Cremers, C. W. R. J., Coppens-Schellekens, W., & Snik, A. F. (2005). The BAHA softband: A new treatment for young children with bilateral congenital aural atresia. *International Journal of Pediatric Otorhinolaryngology, 69*(7), 973-980. doi: 10.1016/j.ijporl.2005.02.010

Hol, M. K. S., Snik, A. F., Mylanus, E. A. M., & Cremers, C. W. R. J. (2005). Does the bone-anchored hearing aid have a complementary effect on audiological and subjective outcomes in patients with unilateral conductive hearing loss? *Audiology and Neurotology, 10*(3), 159-168. doi: 10.1159/000084026

Hol, M. K. S., Spath, M. A., Krabbe, P. F., van der Pouw, C. T., Snik, A. F., Cremers, C. W., & Mylanus, E. A. M. (2004). The bone-anchored hearing aid: Quality-of-life assessment. *Archives of Otolaryngology—Head & Neck Surgery, 130*(4), 394-399. doi: 10.1001/archotol.130.4.394

Holte, L., Margolish, R. H., & Cavanaugh, R. M. (1991). Developmental changes in multifrequency tympanograms. *International Journal of Audiology, 30*(1), 1-24. doi: 10.3109/00206099109072866

Huizing, E. H. (1960). Bone conduction-the influence of the middle ear. chapter V: The sound pressure in the auditory canal. *Acta Oto-Laryngologica, Supplement 155*, 46-54. doi: 10.3109/00016486009122352

Huizing, E. H. (1960). Bone conduction-the influence of the middle ear. chapter VIII: Theory of bone conduction II. *Acta Oto-Laryngologica, Supplement 155*, 71-75. doi: 10.3109/00016486009122355

Hulecki, L. R., & Small, S. A. (2011). Behavioral bone-conduction thresholds for infants with normal hearing. *Journal of the American Academy of Audiology, 22*(2), 81-92. doi: 10.3766/jaaa.22.2.3

Ikui, A., Sando, I., Haginomori, S. I., & Sudo, M. (2000). Postnatal development of the tympanic cavity: A computer-aided reconstruction and measurement study. *Acta Oto-Laryngologica, 120*(3), 375-379. doi: 10.1080/000164800750000595

Ikui, A., Sando, I., Sudo, M., & Fujita, S. (1997). Postnatal change in angle between the tympanic annulus and surrounding structures. computer-aided three-dimensional reconstruction study. *The Annals of Otology, Rhinology, and Laryngology, 106*(1), 33-36.

International Electrotechnical Commission [IEC] 373. (1990). *Mechanical coupler for measurements on bone vibrators* (2nd ed.). Geneva, Switzerland: International Electrotechnical Commission.

Jacobsson, M., Tjellstrom, A., Fine, L., & Andersson, H. (1992). A retrospective study of osseointegrated skin-penetrating titanium fixtures used for retaining facial prostheses. *The International Journal of Oral & Maxillofacial Implants, 7*(4), 523-528.

Janssen, R. M., Hong, P., & Chadha, N. K. (2012). Bilateral bone-anchored hearing aids for bilateral permanent conductive hearing loss. *Otolaryngology—Head and Neck Surgery, 147*(3), 412-422. doi: 10.1177/0194599812451569

Jiang, Z. D., Zheng, M. S., Sun, D. K., & Liu, X. Y. (1991). Brainstem auditory evoked responses from birth to adulthood: Normative data of latency and interval. *Hearing Research, 54*(1), 67-74. doi: 10.1016/0378-5955(91)90137-X

Joint Committee on Infant Hearing. (2007). Year 2007 position statement: Principles and guidelines for early hearing detection and intervention programs. *Pediatrics, 120*(4), 898-921.

Keefe, D. H., Bulen, J. C., Arehart, K. H., & Burns, E. M. (1993). Ear-canal impedance and reflection coefficient in human infants and adults. *The Journal of the Acoustical Society of America, 94*(5), 2617-2638. doi: 10.1121/1.407347

Keefe, D. H., Bulen, J. C., Campbell, S. L., & Burns, E. M. (1994). Pressure transfer function and absorption cross section from the diffuse field to the human infant ear canal. *The Journal of the Acoustical Society of America, 95*(1), 355-371. doi: 10.1121/1.408380

Khanna, S. M., Tonndorf, J., & Queller, J. E. (1976). Mechanical parameters of hearing by bone conduction. *The Journal of the Acoustical Society of America, 60*(1), 139-154. doi: 10.1121/1.381081

Kirikae, I. (1959). Chapter I: General survey of previous theories on fundamental bone conduction mechanism. an experimental study on the fundamental mechanism of bone conduction. *Acta Oto-Laryngologica.Supplementum, 145,* 2-3. doi: 10.3109/00016485909124884

Kirikae, I. (1959). Chapter V: The vibration pattern of the head and the hearing acuity of bone conduction. An experimental study on the fundamental mechanism of bone conduction. *Acta Oto-Laryngologica.Supplementum, 145*, 27-39. doi: 10.3109/00016485909124888

Kirikae, I. (1959). Chapter VII: Mechanism of inertia bone conduction (II) tile effect of loading the tympanic membrane upon bone conduction. an experimental study on the fundamental mechanism of bone conduction. *Acta Oto-Laryngologica.Supplementum, 145*, 52-71. doi: 10.3109/00016485909124890

Kirikae, I. (1959). Chapter XI: Experimental study on bone conduction with fixation of the oval and round window an evaluation of compression bone conduction theories. an experimental study on the fundamental mechanism of bone conduction. *Acta Oto-Laryngologica.Supplementum, 145*, 91-104. doi: 10.3109/00016485909124894

Kriewall, T. J. (1982). Structural, mechanical, and material properties of fetal cranial bone. *American Journal of Obstetrics and Gynecology, 143*(6), 707-714.

Kruger, B. (1987). An update on the external ear resonance in infants and young children. *Ear & Hearing, 8*(6), 333-336.

Kruger, B., & Ruben, R. J. (1987). The acoustic properties of the infant ear. A preliminary report. *Acta Oto-Laryngologica, 103*(5-6), 578-585.

Kuczmarski, R. J., Ogden, C. L., Guo, S. S., Grummer-Strawn, L. M., Flegal, K. M., Mei, Z., . . . Johnson, C. L. (2002). 2000 CDC growth charts for the United States: Methods and development. *National Center for Health Statistics, Vital Health Stat 11*(246)

Kunst, S. J., Leijendeckers, J. M., Mylanus, E. A. M., Hol, M. K. S., Snik, A. F., & Cremers, C. W. (2008). Bone-anchored hearing aid system application for unilateral congenital conductive hearing impairment: Audiometric results. *Otology & Neurotology, 29*(1), 2-7. doi: 10.1097/mao.0b013e31815ee29a

Lau, C. C. (1986). The effect of coupling force on bone conduction audiometry. *British Journal of Audiology, 20*(4), 261-268. doi: 10.3109/03005368609079025

Lavigne-Rebillard, M., & Pujol, R. (1988). Hair cell innervation in the fetal human cochlea. *Acta Oto-Laryngologica, 105*(5-6), 398-402. doi: 10.3109/00016488809119492

Lightfoot, G. R. (1979). Air-borne radiation from bone conduction transducers. *British Journal of Audiology, 13*(2), 53-56. doi: 10.3109/03005367909078877

Lundgren, H. (2010). *Bone conduction transducers and output variability.* (Master of Science Thesis in Biomedical Engineering No. EX042/2010). Göteborg, Sweden: Chalmers Institute of Technology.

Lupin, A. J., & Gardiner, R. J. (2001). Scalp thickness in the temporal region: Its relevance to the development of cochlear implants. *Cochlear Implants International, 2*(1), 30-38. doi: 10.1002/cii.38; 10.1002/cii.38

Macnamara, M., Phillips, D., & Proops, D. W. (1996). The bone anchored hearing aid (BAHA) in chronic suppurative otitis media (CSOM). *The Journal of Laryngology & Otology, Supplement 21*, 38-40. doi: 10.1017/S0022215100136254

Margulies, S. S., & Thibault, K. L. (2000). Infant skull and suture properties: Measurements and implications for mechanisms of pediatric brain injury. *Journal of Biomechanical Engineering, 122*, 364-371. doi: 10.1115/1.1287160

Marsella, P., Scorpecci, A., Pacifico, C., Presuttari, F., & Bottero, S. (2012). Pediatric BAHA in italy: The "Bambino gesù" Children's Hospital's experience. *European Archives of Oto-Rhino-Laryngology, 269*(2), 467-474. doi: 10.1007/s00405-011-1689-5

Martin, H. C., Westwood, G. F. S., & Bamford, J. M. (1996). Real ear to coupler differences in children having otitis media with effusion. *British Journal of Audiology, 30*(2), 71-78. doi: 10.3109/03005369609077934

McBride, M., Letowski, T., & Tran, P. (2008). Bone conduction reception: Head sensitivity mapping. *Ergonomics, 51*(5), 702-718. doi: 10.1080/00140130701747509

McDermott, A. L., Sunil, N. D., Reid, A. P., & Proops, A. W. (2002). An intra-individual comparison of the previous conventional hearing aid with the bone-anchored hearing aid: The nijmegen group questionnaire. *The Journal of Laryngology & Otology, 116*(28), 15-19. doi: 10.1258/0022215021911293

Meyer, S. E., Jardine, C. A., & Deverson, W. (1997). Developmental changes in tympanometry: A case study. *British Journal of Audiology, 31*(3), 189-195. doi: 10.3109/03005364000000021

Moore, J. K., & Linthicum, F. H. (2007). The human auditory system: A timeline of development. *International Journal of Audiology, 46*(9), 460-478. doi: 10.1080/14992020701383019

Moore, J. K., Ponton, C. W., Eggermont, J. J., Wu, B. J. C., & Huang, J. Q. (1996). Perinatal maturation of the auditory brain stem response: Changes in path length and conduction velocity. *Ear & Hearing, 17*(5), 411-418.

Moser, P. J. (2009). *Electronics and instrumentation for audiologists*. New York, NY: Psychology Press, Tayler & Francis Group.

Mylanus, E. A. M., Snik, A. F., & Cremers, C. W. R. J. (1994). Influence of the thickness of the skin and subcutaneous tissue covering the mastoid on bone-conduction thresholds obtained transcutaneously versus percutaneously. *Scandinavian Audiology, 23*(3), 201-203. doi: 10.3109/01050399409047509

Mylanus, E. A. M., van der Pouw, K. C. T. M., Snik, A. F., & Cremers, C. W. R. J. (1998). Intraindividual comparison of the bone-anchored hearing aid and air-conduction hearing aids. *Archives of Otolaryngology—Head & Neck Surgery, 124*(3), 271-276. doi: 10.1001/archotol.124.3.271

National Deaf Children's Society [NDCS]. (2010). *Bone anchored hearing aids for children and young people: Guidelines for professionals*. London, England: National Deaf Children's Society.

Nicholson, N., Christensen, L., Dornhoffer, J., Martin, P., & Smith-Olinde, L. (2011). Verification of speech spectrum audibility for pediatric Baha softband users with craniofacial anomalies. *Cleft Palate-Craniofacial Journal, 48*(1), 56-65. doi: 10.1597/08-178

Nikolovski, J., Stamatas, G. N., Kollias, N., & Wiegand, B. C. (2008). Barrier function and water-holding and transport properties of infant stratum corneum are different from adult and continue to develop through the first year of life. *The Journal of Investigative Dermatology, 128*(7), 1728-1736. doi: 10.1038/sj.jid.5701239

Niparko, J. K., Cox, K. M., & Lustig, L. R. (2003). Comparison of the bone anchored hearing aid implantable hearing device with contralateral routing of offside signal amplification in the rehabilitation of unilateral deafness. *Otology & Neurotology, 24*(1), 73-78.

Nolan, M., & Lyon, D. J. (1981). Transcranial attenuation in bone conduction audiometry. *The Journal of Laryngology & Otology, 95*(06), 597-608. doi: 10.1017/S0022215100091155

Nousak, J. M., & Stapells, D. R. (1992). Frequency specificity of the auditory brain stem response to bone-conducted tones in infants and adults. *Ear Hear, 13*(2), 87-95.

O'Brien, W. D., & Liu, Y. (2005). *Evaluation of acoustic propagation paths into the human head.* (Technical report No. ADA437351).North Atlantic Treaty Organization. Retrieved from http://ftp.rta.nato.int/public//PubFullText/RTO/MP/RTO-MP-HFM-123///MP-HFM-123-15.pdf

Opperman, L. A. (2000). Cranial sutures as intramembranous bone growth sites. *Developmental Dynamics: An Official Publication of the American Association of Anatomists, 219*(4), 472-485. doi: 10.1002/1097-0177(2000)9999:9999<::AID-DVDY1073>3.0.CO;2-F

Oticon Medical. (2011). Genie medical fitting guide. Retrieved November/12, 2012, from http://www.oticonmedical.com/~asset/cache.ashx?id=8255&type=14&format=web

Papsin, B. C., Sirimanna, T. K., Albert, D. M., & Bailey, C. M. (1997). Surgical experience with bone-anchored hearing aids in children. *The Laryngoscope, 107*(6), 801-806. doi: 10.1097/00005537-199706000-00015

Paradise, J. L., Rockette, H. E., Colborn, D. K., Bernard, B. S., Smith, C. G., Kurs-Lasky, M., & Janosky, J. E. (1997). Otitis media in 2253 Pittsburg-area infants: Prevalence and risk factors during the first two years of life. *Pediatrics, 99*(3), 318-333. doi: 10.1542/peds.99.3.318

Parry, G., Hacking, C., Bamford, J., & Day, J. (2003). Minimal response levels for visual reinforcement audiometry in infants. *International Journal of Audiology, 42*(7), 413-417. doi: 10.3109/14992020309080050

Perez, R., Adelman, C., & Sohmer, H. (2011). Bone conduction activation through soft tissues following complete immobilization of the ossicular chain, stapes footplate and round window. *Hearing Research, 280*(1), 82-85. doi: 10.1016/j.heares.2011.04.007

Picton, T. W., Durieux-Smith, A., & Moran, L. M. (1994). Recording auditory brainstem responses from infants. *International Journal of Pediatric Otorhinolaryngology, 28*(2), 93-110. doi: 10.1016/0165-5876(94)90001-9

Ponton, C. W., Moore, J. K., & Eggermont, J. J. (1996). Auditory brain stem response generation by parallel pathways: Differential maturation of axonal conduction time and synaptic transmission. *Ear & Hearing, 17*(5), 402-410.

Priwin, C., Jönsson, R., Hultcrantz, M., & Granström, G. (2007). BAHA in children and adolescents with unilateral or bilateral conductive hearing loss: A study of outcome. *International Journal of Pediatric Otorhinolaryngology, 71*(1), 135-145. doi: 10.1016/j.ijporl.2006.09.014

Priwin, C., Stenfelt, S., Granström, G., Tjellström, A., & Håkansson, B. (2004). Bilateral Bone-anchored hearing aids (BAHAs): An audiometric evaluation. *The Laryngoscope, 114*(1), 77-84. doi: 10.1097/00005537-200401000-00013

Puria, S., & Rosowski, J. (2012). Békésy's contributions to our present understanding of sound conduction to the inner ear. *Hearing Research, 293*(1-2), 21-30. doi: 10.1016/j.heares.2012.05.004

Raine, C. H., Lee, C. A., Strachan, D. R., Totten, C. T., & Khan, S. (2007). Skin flap thickness in cochlear implant patients–a prospective study. *Cochlear Implants International, 8*(3), 148-157. doi: 10.1002/cii.335

Ramakrishnan, Y., Davison, T., & Johnson, I. (2006). How we do it: Softband®–management of glue ear. *Clinical Otolaryngology, 31*(3), 224-227. doi: 10.1111/j.1749-4486.2006.01166.x

Rance, G., & Tomlin, D. (2006). Maturation of auditory steady-state responses in normal babies. *Ear & Hearing, 27*(1), 20-29. doi: 10.1097/01.aud.0000191944.03797.5a

Reinfeldt, S., Stenfelt, S., Good, T., & Håkansson, B. (2007). Examination of bone-conducted transmission from sound field excitation measured by thresholds, ear-canal sound pressure, and skull vibrations. *The Journal of the Acoustical Society of America, 121*, 1576-1587. doi: 10.1121/1.2434762

Rotteveel, J. J., de Graaf, R., Colon, E. J., Stegeman, D. F., & Visco, Y. M. (1987). The maturation of the central auditory conduction in preterm infants until three months post term. II. the auditory brainstem responses (ABRs). *Hearing Research, 26*(1), 21-35. doi: 10.1016/0378-5955(87)90033-5

Salamy, A. (1984). Maturation of the auditory brainstem response from birth through early childhood. *Journal of Clinical Neurophysiology, 1*(3), 293-329.

Sanford, C. A., & Feeney, M. P. (2008). Effects of maturation on tympanometric wideband acoustic transfer functions in human infants. *The Journal of the Acoustical Society of America, 124*(4), 2106-2122. doi: 10.1121/1.2967864

Saunders, J. C., Kaltenbach, J. A., & Relkin, E. M. (1983). The structural and functional development of the outer and middle ear. In R. Romand, & M. R. Romand (Eds.), *Development of auditory and vestibular systems* (pp. 3-25). New York, NY: Academic Press.

Schillinger, R. (1939). Pneumatization of the mastoid: A roentgen study. *Radiology, 33*(1), 54-67. doi: 10.1148/33.1.54

Scollie, S., Seewald, R., Cornelisse, L., Moodie, S., Bagatto, M. P., Laurnagaray, D., . . . Pumford, J. (2005). The desired sensation level multistage input/output algorithm. *Trends in Amplification, 9*(4), 159-197. doi: 10.1177/108471380500900403

Seemann, R., Liu, R., & Di Toppa, J. (2004). Results of pediatric bone-anchored hearing aid implantation. *The Journal of Otolaryngology, 33*(2), 71-74.

Seidenari, S., Giusti, G., Bertoni, L., Magnoni, C., & Pellacani, G. (2000). Thickness and echogenicity of the skin in children as assessed by 20-MHz ultrasound. *Dermatology, 201*(3), 218-222. doi: 10.1159/000018491

Simms, D. L., & Neely, J. G. (1989). Growth of the lateral surface of the temporal bone in children. *The Laryngoscope, 99*(8), 795-799. doi: 10.1288/00005537-198908000-00005

Sininger, Y. S., & Abdala, C. (1996). Hearing threshold as measured by auditory brain stem response in human neonates. *Ear & Hearing, 17*(5), 395-401.

Sininger, Y. S., Abdala, C., & Cone-Wesson, B. (1997). Auditory threshold sensitivity of the human neonate as measured by the auditory brainstem response. *Hearing Research, 104*(1), 27-38. doi: 10.1016/S0378-5955(96)00178-5

Small, S. A., Hatton, J. L., & Stapells, D. R. (2007). Effects of bone oscillator coupling method, placement location, and occlusion on bone-conduction auditory steady-state responses in infants. *Ear & Hearing, 28*(1), 83-98. doi: 10.1097/01.aud.0000249787.97957.5b

Small, S. A., & Hu, N. (2011). Maturation of the occlusion effect: A bone conduction auditory steady state response study in infants and adults with normal hearing. *Ear & Hearing, 32*(6), 708-719. doi: 10.1097/AUD.0b013e31821de1b0

Small, S. A., & Stapells, D. R. (2008). Maturation of bone conduction multiple auditory steady-state responses. *International Journal of Audiology, 47*(8), 476-488. doi: 10.1080/14992020802055284

Small, S. A., & Stapells, D. R. (2008). Normal ipsilateral/contralateral asymmetries in infant multiple auditory steady-state responses to air-and bone-conduction stimuli. *Ear & Hearing, 29*(2), 185-198. doi: 10.1097/01.aud.0000305157.83012.f4

Small, S. A., & Stapells, D. R. (2006). Multiple auditory steady-state response thresholds to bone-conduction stimuli in young infants with normal hearing. *Ear & Hearing, 27*(3), 219-228. doi: 10.1097/01.aud.0000215974.74293.b9

Smith, J. B., & Suggs, C. W. (1976). Dynamic properties of the human head. *Journal of Sound and Vibration, 48*(1), 35-43. doi: 10.1016/0022-460X(76)90369-2

Snik, A. F., Jorritsma, F. F., Cremers, C., Beynon, A. J., & Van den Berge, N. W. (1992). The super-bass bone-anchored hearing aid compared to conventional hearing aids. audiological results and the patients' opinions. *Scandinavian Audiology, 21*(3), 157-161. doi: 10.3109/01050399209045997

Snik, A. F., Mylanus, E. A. M., & Cremers, C. W. R. J. (1995). The bone-anchored hearing aid compared with conventional hearing aids. audiologic results and the patients' opinions. *Otolaryngologic Clinics of North America, 28*(1), 73-83.

Snik, A. F., Dreschler, W. A., Tange, R. A., & Cremers, C. W. R. J. (1998). Short-and long-term results with implantable transcutaneous and percutaneous bone-conduction devices. *Archives of Otolaryngology—Head & Neck Surgery, 124*(3), 265-268. doi: 10.1001/archotol.124.3.265

Snik, A. F., Leijendeckers, J., Hol, M. K. S., Mylanus, E. A. M., & Cremers, C. (2008). The bone-anchored hearing aid for children: Recent developments. *International Journal of Audiology, 47*(9), 554-559. doi: 10.1080/14992020802307354

Snik, A. F., Mylanus, E. A. M., Proops, D. W., Wolfaardt, J. F., Hodgetts, W. E., Somers, T., . . . Mylanus, E. A. M. (2005). Consensus statements on the BAHA system: Where do we stand at present? *The Annals of Otology, Rhinology & Laryngology.Supplement, 195*(12), 2-12.

Sockalingam, R. (2012). Consensus on bone-anchored hearing aid fittings. Retrieved October/20, 2012, from http://www.audiologyonline.com/ask-the-experts/consensus-on-bone-anchored-hearing-7032

Sohmer, H., Freeman, S., Geal-Dor, M., Adelman, C., & Savion, I. (2000). Bone conduction experiments in humans-a fluid pathway from bone to ear. *Hearing Research, 146*(1-2), 81-88. doi: 10.1016/S0378-5955(00)00099-X

Sperber, G. H., Guttman, G. D., & Sperber, S. M. (2001). *Craniofacial development*. Hamilton, Ontario: BC Decker Inc.

Stamatas, G. N., Nikolovski, J., Mack, M. C., & Kollias, N. (2011). Infant skin physiology and development during the first years of life: A review of recent findings based on in vivo studies. *International Journal of Cosmetic Science, 33*(1), 17-24. doi: 10.1111/j.1468-2494.2010.00611.x

Stapells, D. R., & Ruben, R. J. (1989). Auditory brain stem responses to bone-conducted tones in infants. *The Annals of Otology, Rhinology, and Laryngology, 98*(12 Pt 1), 941-949.

Steele, D. G., & Bramblett, C. A. (1988). *The anatomy and biology of the human skeleton*. College Station, Texas: Texas A&M University Press.

Stenfelt, S. (2005). Bilateral fitting of BAHAs and BAHA fitted in unilateral deaf persons: Acoustical aspects. *International Journal of Audiology, 44*(3), 178-189. doi: 10.1080/14992020500031561

Stenfelt, S. (2012). Transcranial attenuation of bone-conducted sound when stimulation is at the mastoid and at the bone conduction hearing aid position. *Otology & Neurotology, 33*(2), 105. doi: 10.1097/MAO.0b013e31823e28ab

Stenfelt, S., & Goode, R. L. (2005). Bone-conducted sound: Physiological and clinical aspects. *Otology & Neurotology, 26*(6), 1245-1261.

Stenfelt, S., & Håkansson, B. (1998). A miniaturized artificial mastoid using a skull simulator. *Scandinavian Audiology, 27*(2), 67-76. doi: 10.1080/010503998420306

Stenfelt, S., & Håkansson, B. (2002). Air versus bone conduction: An equal loudness investigation. *Hearing Research, 167*(1-2), 1-12. doi: 10.1016/S0378-5955(01)00407-5

Stenfelt, S., Håkansson, B., & Tjellström, A. (2000). Vibration characteristics of bone conducted sound in vitro. *The Journal of the Acoustical Society of America, 107*, 422-431. doi: 10.1121/1.428314

Stenfelt, S., Hato, N., & Goode, R. L. (2002). Factors contributing to bone conduction: The middle ear. *The Journal of the Acoustical Society of America, 111*, 947-959. doi: 10.1121/1.1432977

Stenfelt, S., Hato, N., & Goode, R. L. (2004). Fluid volume displacement at the oval and round windows with air and bone conduction stimulation. *The Journal of the Acoustical Society of America, 115*(2), 797-812. doi: 10.1121/1.1639903

Stenfelt, S., Puria, S., Hato, N., & Goode, R. L. (2003). Basilar membrane and osseous spiral lamina motion in human cadavers with air and bone conduction stimuli. *Hearing Research, 181*(1), 131-143. doi: 10.1016/S0378-5955(03)00183-7

Stenfelt, S., & Reinfeldt, S. (2007). A model of the occlusion effect with bone-conducted stimulation. *International Journal of Audiology, 46*(10), 595-608. doi: 10.1080/14992020701545880

Stenfelt, S., Wild, T., Hato, N., & Goode, R. L. (2003). Factors contributing to bone conduction: The outer ear. *The Journal of the Acoustical Society of America*, *113*, 902-913. doi: 10.1121/1.1534606

Stenfelt, S., & Goode, R. L. (2005). Transmission properties of bone conducted sound: Measurements in cadaver heads. *Journal of the Acoustical Society of America, 118*(4), 2373-2391. doi: 10.1121/1.2005847

Stenfelt, S., & Håkansson, B. (1999). Sensitivity to bone-conducted sound: Excitation of the mastoid vs the teeth. *Scandinavian Audiology, 28*(3), 190-198. doi: 10.1080/010503999424761

Stuart, A., Yang, E. Y., & Botea, M. (1996). Neonatal auditory brainstem responses recorded from four electrode montages. *Journal of Communication Disorders, 29*(2), 125-139. doi: 10.1016/0021-9924(95)00018-6

Stuart, A., Yang, E. Y., & Stenstrom, R. (1990). Effect of temporal area bone vibrator placement on auditory brain stem response in newborn infants. *Ear & Hearing, 11*(5), 363-369.

Studebaker, G. A. (1967). Clinical masking of the nontest ear. *The Journal of Speech and Hearing Disorders, 32*(4), 360-371.

Swanepoel, D. W., Ebrahim, S., Friedland, P., Swanepoel, A., & Pottas, L. (2008). Auditory steady-state responses to bone conduction stimuli in children with hearing loss. *International Journal of Pediatric Otorhinolaryngology, 72*(12), 1861-1871. doi: 10.1016/j.ijporl.2008.09.017

Takahara, T., & Sando, I. (1987). Mesenchyme remaining in temporal bones from patients with congenital anomalies. A quantitative histopathologic study. *The Annals of Otology, Rhinology, and Laryngology, 96*(3 Pt 1), 333-339.

Takahara, T., Sando, I., Hashida, Y., & Shibahara, Y. (1986). Mesenchyme remaining in human temporal bones. *Otolaryngology—Head and Neck Surgery, 95*(3 Pt 1), 349-357.

Tharpe, A. M., Sladen, D., Huta, H. M., & McKinley Rothpletz, A. (2001). Practical considerations of real-ear-to-coupler difference measures in infants. *American Journal of Audiology, 10*(1), 41-49. doi: 10.1044/1059-0889(2000/006)

The Pediatric Working Group. (1996). Amplification for infants and children with hearing loss. *American Journal of Audiology, 5*(1), 53-68.

Tietze, L., & Papsin, B. C. (2001). Utilization of bone-anchored hearing aids in children. *International Journal of Pediatric Otorhinolaryngology, 58*(1), 75-80. doi: 10.1016/S0165-5876(00)00472-9

Tjellström, A., Rosenhall, U., Lindström, J., Hallén, O., Albrektsson, T., & Brånemark, P. (1983). Five-year experience with skin-penetrating bone-anchored implants in the temporal bone. *Acta Oto-Laryngologica, 95*(5-6), 568. doi: 10.3109/00016488309139444

Tjellström, A., Håkansson, B., & Granström, G. (2001). Bone-anchored hearing aids: Current status in adults and children. *Otolaryngologic Clinics of North America, 34*(2), 337-364.

Tjellström, A., Håkansson, B., Lindström, J., Brånemark, P. I., Hallen, O., Rosenhall, U., & Leijon, A. (1980). Analysis of the mechanical impedance of bone-anchored hearing aids. *Acta Oto-Laryngologica, 89*(1-2), 85-92. doi: 10.3109/00016488009127113

Tonndorf, J. (1968). A new concept of bone conduction. *Archives of Otolaryngology—Head and Neck Surgery, 87*(6), 595-600. doi: 10.1001/archotol.1968.00760060597008

Tonndorf, J., Greenfield, E. C., & Kaufman, R. S. (1966). The occlusion of the external ear canal: Its effect upon bone conduction in cats. *Acta Oto-Laryngologica, 61*(S213), 80-104. doi: 10.3109/00016486609120801

Tonndorf, J. (1962). Compressional bone conduction in cochlear models. *The Journal of the Acoustical Society of America, 34*(8), 1127-1131. doi: 10.1121/1.1918259

Tsai, V., Ostroff, J., Korman, M., & Chen, J. M. (2005). Bone-conduction hearing and the occlusion effect in otosclerosis and normal controls. *Otology & Neurotology, 26*(6), 1138-1142.

Valente, M., Abrams, H., Benson, D., Chisolm, T., Citron, D., Hampton, D., . . . Sweetow, R. (2006). Guidelines for the audiologic management of adult hearing impairment. *Audiology Today, 18*(5), 32-37.

Van Maanen, A., & Stapells, D. R. (2009). Normal multiple auditory steady-state response thresholds to air-conducted stimuli in infants. *Journal of the American Academy of Audiology, 20*(3), 196-207. doi: 10.3766/jaaa.20.3.6

Vander Werff, K. R., Prieve, B. A., & Georgantas, L. M. (2009). Infant air and bone conduction tone burst auditory brain stem responses for classification of hearing loss and the relationship to behavioral thresholds. *Ear & Hearing, 30*(3), 350-368. doi: 10.1097/AUD.0b013e31819f3145

Vanniasegaram, I., Bradley, J., & Bellman, S. (1994). Clinical applications of transcranial bone conduction attenuation in children. *The Journal of Laryngology & Otology, 108*(10), 834-836. doi: 10.1017/S0022215100128269

Verhagen, C. V., Hol, M. K. S., Coppens-Schellekens, W., Snik, A. F., & Cremers, C. W. (2008). The Baha softband. A new treatment for young children with bilateral congenital aural atresia. *International Journal of Pediatric Otorhinolaryngology, 72*(10), 1455-1459. doi: 10.1016/j.ijporl.2008.06.009

Verstraeten, N., Zarowski, A. J., Somers, T., Riff, D., & Offeciers, E. F. (2009). Comparison of the audiologic results obtained with the bone-anchored hearing aid attached to the headband, the testband, and to the "snap" abutment. *Otology & Neurotology, 30*(1), 70-75. doi: 10.1097/MAO.0b013e31818be97a

Vitellaro-Zuccarello, L., Cappelletti, S., Rossi, V. D. P., & Sari-Gorla, M. (2005). Stereological analysis of collagen and elastic fibers in the normal human dermis: Variability with age, sex, and body region. *The Anatomical Record, 238*(2), 153-162. doi: 10.1002/ar.1092380202

Von Békésy, G. (1960). In Wever E. G. (Ed.), *Experiments in hearing* (E. G. Wever Trans.). New York: NY: McGraw Hill.

Watson, N. A. (1938). Limits of audition for bone conduction. *The Journal of the Acoustical Society of America, 9*(4), 294-300. doi: 10.1121/1.1915936

Wazen, J. J., Spitzer, J., Ghossaini, S. N., Kacker, A., & Zschommler, A. (2001). Results of the bone-anchored hearing aid in unilateral hearing loss. *The Laryngoscope, 111*(6), 955-958. doi: 10.1097/00005537-200106000-00005

Wazen, J. J., Spitzer, J. B., Ghossaini, S. N., Fayad, J. N., Niparko, J. K., Cox, K., . . . Soli, S. D. (2003). Transcranial contralateral cochlear stimulation in unilateral deafness. *Otolaryngology—Head and Neck Surgery, 129*(3), 248-254. doi: 10.1016/S0194-5998(03)00527-8

Weston, W. L., Lane, A. T., & Morelli, J. G. (1996). *Color textbook of pediatric dermatology* (2nd ed.). Philadelphia, PA: Mosby Elsevier.

Weston, P. B., Gengel, R. W., & Hirsh, I. J. (1967). Effects of vibrator types and their placement on bone-conduction threshold measurements. *The Journal of the Acoustical Society of America, 41*(4), 788-792. doi: 10.1121/1.1910408

Whittle, L. S. (1965). A determination of the normal threshold of hearing by bone conduction. *Journal of Sound and Vibration, 2*(3), 227-248. doi: 10.1016/0022-460X(65)90110-0

Widdowson, E. M. (1968). Growth and composition of the fetus and newborn. *Biology of Gestation, 2*, 1-49.

Woelflin, F. (2011). *The mechanical point impedance of the skin-penetrated human skull.* (Master of Science No. 143059). Göteborg, Sweden: Chalmers University of Technology.

World Health Organization. (2007). Child growth standards: Head circumference-for-age [data file]. Retrieved October/4, 2012, from http://www.who.int/childgrowth/standards/hc_for_age/en/index.html

Yang, E. Y., Rupert, A. L., & Moushegian, G. (1987). A developmental study of bone conduction auditory brain stem response in infants. *Ear & Hearing, 8*(4), 244-251.

Yang, E. Y., Stuart, A., Stenstrom, R., & Hollett, S. (1991). Effect of vibrator to head coupling force on the auditory brain stem response to bone conducted clicks in newborn infants. *Ear & Hearing, 12*(1), 55-60.

Yoshida, M., & Uemura, T. (1991). Transmission of cerebrospinal fluid pressure changes to the inner ear and its effect on cochlear microphonics. *European Archives of Oto-Rhino-Laryngology, 248*(3), 139-143. doi: 10.1007/BF00178923

Yoshinaga-Itano, C. (2003). Early intervention after universal neonatal hearing screening: Impact on outcomes. *Mental Retardation and Developmental Disabilities Research Reviews, 9*(4), 252-266. doi: 10.1002/mrdd.10088

Yoshinaga-Itano, C., Sedey, A. L., Coulter, D. K., & Mehl, A. L. (1998). Language of early- and later-identified children with hearing loss. *Pediatrics, 102*(5), 1161-1171.

Young, R. W. (1957). Postnatal growth of the frontal and parietal bones in white males. *American Journal of Physical Anthropology, 15*(3), 367-386. doi: 10.1002/ajpa.1330150316

Zeitoun, H., De, R., Thompson, S. D., & Proops, D. W. (2002). Osseointegrated implants in the management of childhood ear abnormalities: With particular emphasis on complications. *The Journal of Laryngology and Otology, 116*(2), 87-91. doi: 10.1258/0022215021909935

APPENDIX A: Individual Subject Inclusion/Exclusion Information

Individual subject inclusion/exclusion for each analysis

Subject #	Age Group	BC Attenuation	Low Frequency Impedance	High Frequency Impedance	Resonant Frequency
CH185	A	*	I	I	I
CH139	A	I	I	I	I
CH123	A	I	I	I	I
CH162	A	I	I	I	I
CH161	A	I	I	I	I
CH130	A	I	I	I	I
CH187	A	I	*	*	*
CH177	A	I	I	I	I
CH149	A	I	I	I	I
CH197	A	I	I	I	I
CH141	A	I	I	*	I
CH175	A	I	I	I	I
CH189	A	I	*	*	*
CH146	A	I	I	I	I
CH160	A	I	I	I	I
CH109	A	I	I	I	**
CH131	A	*	*	*	*
CH173	A	*	*	*	*
CH134	B	I	I	I	I
CH122	B	I	I	I	I
CH165	B	I	I	I	I
CH179	B	*	I	I	I
CH114	B	I	I	I	I
CH196	B	I	I	I	I
CH135	B	I	I	I	I
CH115	B	I	I	I	I
CH154	B	I	I	I	I
CH157	B	I	I	I	I
CH136	B	I	I	I	I
CH152	B	I	I	I	I
CH124	B	I	I	I	I
CH117	C	I	I	I	I
CH129	C	I	I	I	I
CH159	C	I	I	I	I
CH138	C	I	I	I	I

A "*" indicates subject did not complete all conditions, "**" indicates subject data was excluded due to noise, and "***" indicates that data could not be interpreted.

Individual subject inclusion/exclusion for each analysis

Subject #	Age Group	BC Attenuation	Low Frequency Impedance	High Frequency Impedance	Resonant Frequency
CH156	C	I	I	I	I
CH168	C	I	I	I	I
CH119	C	I	I	I	I
CH153	C	**	I	I	I
CH193	C	I	I	I	I
CH199	C	*	*	*	*
CH167	C	I	I	I	I
CH143	C	I	I	I	I
CH147	C	I	I	I	I
CH172	C	I	I	I	I
CH116	C	I	I	I	I
CH126	D	I	I	I	I
CH191	D	I	I	I	I
CH188	D	I	I	I	I
CH184	D	I	I	I	I
CH186	D	I	I	I	I
CH144	D	I	I	I	I
CH107	D	I	I	I	I
CH140	D	I	I	I	I
CH174	D	I	I	I	I
CH101	D	I	I	I	I
CH112	D	I	I	I	I
CH145	D	I	I	I	I
CH113	D	I	I	I	I
CH169	D	I	I	I	I
CH118	D	I	I	I	I
CH195	D	I	I	I	I
CH102	D	I	I	I	I
CH125	D	I	I	I	I
CH133	D	I	I	I	I

A "*" indicates subject did not complete all conditions, "**" indicates subject data was excluded due to noise, and "***" indicates that data could not be interpreted.

Individual subject inclusion/exclusion for each analysis

Subject #	Age Group	BC Attenuation	Low Frequency Impedance	High Frequency Impedance	Resonant Frequency
AD001	Adult	I	I	I	I
AD002	Adult	I	I	I	***
AD003	Adult	I	I	I	I
AD004	Adult	I	I	I	I
AD005	Adult	I	I	I	I
AD006	Adult	I	I	I	I
AD007	Adult	I	I	I	I
AD008	Adult	I	I	I	I
AD009	Adult	I	I	I	I
AD010	Adult	I	**	**	**
AD011	Adult	I	I	I	I
AD012	Adult	I	I	I	I
AD013	Adult	I	I	I	I
AD014	Adult	I	I	**	I
AD015	Adult	I	I	I	I
AD016	Adult	I	I	I	I
AD017	Adult	I	I	I	I

A "*" indicates subject did not complete all conditions, "**" indicates subject data was excluded due to noise, and "***" indicates that data could not be interpreted.

APPENDIX B: Individual Measures of Transcranial Attenuation

Individual measures of transcranial attenuation (dB)

Subject #	Age Group	Tymp Result	Ipsilateral-Forehead 500 Hz	Ipsilateral-Contralateral 500 Hz	Ipsilateral-Forehead 1000 Hz	Ipsilateral-Contralateral 1000 Hz	Ipsilateral-Forehead 2000 Hz	Ipsilateral-Contralateral 2000 Hz	Ipsilateral-Forehead 4000 Hz	Ipsilateral-Contralateral 4000 Hz
CH185	A	--	18.4	11.2	4.1	-12.9	--	--	--	--
CH139	A	--	-1.1	5.1	8.5	11.2	7.9	5.9	10.8	5.7
CH123	A	Normal	3.3	-5.4	6.6	19.0	4.4	-10.4	-0.6	-3.2
CH162	A	Normal	12.8	10.0	11.9	4.1	-1.1	1.8	7.8	9.8
CH161	A	Flat	4.4	0.0	18.0	-0.4	19.7	10.4	1.3	4.8
CH130	A	Normal	17.9	8.8	12.1	-7.4	-0.6	0.0	1.6	2.2
CH187	A	Normal	-4.9	-11.5	6.6	-3.1	7.8	5.4	9.5	6.4
CH177	A	Normal	3.8	3.1	12.7	3.0	3.5	-0.4	4.4	2.6
CH149	A	Normal	5.0	5.1	7.8	6.1	19.6	16.1	3.0	9.6
CH197	A	Normal	4.1	8.5	27.3	12.6	11.2	9.8	0.7	6.3
CH141	A	Normal	10.4	6.8	4.9	0.7	12.2	9.9	7.6	9.1
CH175	A	Normal	8.1	-0.4	10.1	17.3	-9.8	-4.9	0.5	1.6
CH189	A	Normal	22.5	11.8	12.8	9.2	4.7	3.9	9.5	9.6
CH146	A	Normal	5.2	-10.0	13.6	10.4	8.3	13.8	4.6	6.2
CH160	A	Flat	6.9	3.1	25.1	5.0	3.1	7.4	3.9	2.0
CH109	A	Normal	0.6	24.1	4.7	21.8	11.6	12.7	4.9	3.0
CH131	A	Normal	--	--	--	--	--	--	--	--
CH173	A	Normal	26.5	17.0	16.6	27.3	--	--	--	--
CH134	B	Normal	0.2	-2.5	8.0	-1.3	9.5	-3.3	-2.2	0.5
CH122	B	Normal	19.6	5.6	29.7	13.9	3.2	-1.2	4.3	4.3
CH165	B	Normal	2.0	-7.2	8.7	-1.4	5.2	2.6	4.2	7.8
CH179	B	Normal	7.7	-2.3	17.9	8.1	8.1	6.7	5.2	

A "--" indicates these data were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of transcranial attenuation (dB)

Subject #	Age Group	Tymp Result	Ipsilateral-Forehead 500 Hz	Ipsilateral-Contralateral 500 Hz	Ipsilateral-Forehead 1000 Hz	Ipsilateral-Contralateral 1000 Hz	Ipsilateral-Forehead 2000 Hz	Ipsilateral-Contralateral 2000 Hz	Ipsilateral-Forehead 4000 Hz	Ipsilateral-Contralateral 4000 Hz
CH114	B	Normal	12.3	0.0	27.0	19.1	0.1	-4.4	4.2	2.9
CH196	B	Normal	15.6	12.2	14.6	1.5	9.8	2.2	5.2	6.3
CH135	B	Normal	6.6	2.6	7.7	-0.4	1.0	-6.1	8.5	8.6
CH115	B	Normal	5.8	5.6	22.8	4.8	12.3	6.0	10.5	4.9
CH154	B	Normal	7.1	8.2	7.8	3.0	14.4	1.2	1.5	-0.3
CH157	B	Normal	6.2	-1.0	3.2	0.6	7.4	9.5	5.7	6.6
CH136	B	Normal	6.1	2.0	25.9	3.3	10.8	17.6	8.4	8.6
CH152	B	Normal	0.2	-0.5	5.4	8.4	9.5	9.5	10.1	7.6
CH124	B	Normal	0.7	-3.7	15.7	3.3	1.7	-0.6	5.1	1.3
CH117	C	Normal	6.1	1.4	11.0	4.1	11.7	12.2	11.0	5.3
CH129	C	Normal	-5.3	-4.0	1.6	-0.6	10.4	3.0	5.8	12.8
CH159	C	Normal	5.5	1.9	9.8	0.1	8.1	4.6	3.3	5.2
CH138	C	Normal	6.5	-0.3	9.0	-4.4	-3.1	-2.9	4.0	5.5
CH156	C	Normal	1.9	8.1	4.0	11.7	0.9	-0.6	6.7	-0.9
CH168	C	Normal	9.0	3.3	6.1	4.7	8.8	5.4	10.1	2.7
CH119	C	Normal	4.5	0.3	6.7	16.5	-0.1	-11.0	12.9	4.9
CH153	C	Normal	--	--	--	--	5.5	-0.3	11.4	10.8
CH193	C	Normal	13.7	10.1	9.1	5.6	-2.9	-4.8	-0.6	3.7
CH199	C	Normal	--	--	--	--	--	--	--	--
CH167	C	Normal	0.4	0.4	8.8	-2.5	11.0	-1.3	0.3	7.7
CH143	C	Normal	4.2	-9.1	24.8	2.2	16.0	12.1	4.1	4.1
CH147	C	Normal	4.6	-4.6	5.8	-9.9	1.2	0.5	3.6	3.6

A "--" indicates these data were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of transcranial attenuation (dB)

Subject #	Age Group	Tymp Result	Ipsilateral-Forehead 500 Hz	Ipsilateral-Contralateral 500 Hz	Ipsilateral-Forehead 1000 Hz	Ipsilateral-Contralateral 1000 Hz	Ipsilateral-Forehead 2000 Hz	Ipsilateral-Contralateral 2000 Hz	Ipsilateral-Forehead 4000 Hz	Ipsilateral-Contralateral 4000 Hz
CH172	C	Normal	6.3	4.0	16.0	10.9	8.9	3.4	4.3	1.0
CH116	C	Normal	28.3	-9.1	-11.1	-20.6	16.1	15.4	6.2	8.1
CH126	D	Normal	8.3	-0.4	15.0	19.4	3.6	2.1	6.4	5.1
CH191	D	Normal	1.0	-2.5	5.5	2.0	8.9	5.0	6.2	5.8
CH188	D	Normal	14.8	6.6	14.0	10.4	1.8	4.8	1.1	3.4
CH184	D	Normal	4.6	2.1	3.6	5.1	12.4	7.7	4.8	5.7
CH186	D	Normal	12.0	4.0	6.6	5.5	7.8	-0.4	4.9	6.2
CH144	D	Normal	12.3	7.1	11.8	0.2	5.1	0.4	8.6	8.2
CH107	D	Normal	19.9	11.7	13.0	19.5	11.6	7.7	10.9	5.5
CH140	D	Normal	0.3	-5.8	9.0	-10.2	6.0	5.6	-3.6	-4.7
CH174	D	Normal	-3.6	-6.8	-4.7	-11.1	5.3	4.9	2.1	4.8
CH101	D	Neg Press	6.3	-7.7	-0.6	2.2	13.7	5.9	4.6	4.9
CH112	D	Normal	3.9	-8.3	-3.8	-11.3	5.4	-2.5	6.5	3.7
CH145	D	Normal	2.8	-0.8	5.0	1.2	14.5	10.3	0.9	3.5
CH113	D	Normal	-9.1	-6.3	15.3	14.5	13.5	3.7	0.1	-2.8
CH169	D	Normal	4.1	5.0	9.8	-2.5	1.8	-3.7	10.3	8.0
CH118	D	Normal	2.1	1.5	12.5	3.7	2.1	-1.2	4.5	2.2
CH195	D	Normal	7.2	7.4	25.5	9.9	2.7	0.7	9.3	17.1
CH102	D	Normal	10.9	3.1	10.4	5.1	0.3	1.6	6.6	4.4
CH125	D	Normal	7.6	3.6	-4.1	-6.5	11.7	4.8	-1.3	-1.0
CH133	D	Normal	9.4	-2.1	12.9	12.7	13.4	9.0	-5.6	0.9

A "--" indicates these data were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of transcranial attenuation (dB)

Subject #	Age Group	**Tymp Result**	**Ipsilateral-Forehead 500 Hz**	**Ipsilateral-Contralateral 500 Hz**	**Ipsilateral-Forehead 1000 Hz**	**Ipsilateral-Contralateral 1000 Hz**	**Ipsilateral-Forehead 2000 Hz**	**Ipsilateral-Contralateral 2000 Hz**	**Ipsilateral-Forehead 4000 Hz**	**Ipsilateral-Contralateral 4000 Hz**
AD001	Adult	Normal	-3.5	-9.1	3.5	-1.6	8.6	4.6	6.3	7.9
AD002	Adult	Normal	-2.6	1.6	3.1	3.9	7.7	6.8	3.7	6.2
AD003	Adult	Normal	1.6	-0.3	0.2	-3.6	3.3	4.7	5.7	3.5
AD004	Adult	Normal	14.9	11.5	14.9	1.5	-2.1	-4.8	-0.9	1.2
AD005	Adult	Normal	-12.8	-6.7	-3.0	-8.0	5.2	3.5	4.7	2.2
AD006	Adult	Normal	4.7	6.6	6.4	1.0	8.2	4.2	4.3	6.7
AD007	Adult	Normal	6.7	2.4	4.8	4.4	9.4	6.7	8.5	9.1
AD008	Adult	Normal	0.3	-10.5	8.7	0.9	7.8	5.7	6.4	5.6
AD009	Adult	Normal	-6.3	-4.5	4.6	-1.3	7.0	5.3	-1.8	-10.9
AD010	Adult	Normal	6.0	9.5	3.8	0.6	3.2	1.3	-3.5	2.2
AD011	Adult	Normal	2.9	-0.8	9.1	2.8	5.3	8.7	3.7	-6.2
AD012	Adult	Normal	11.6	5.9	11.4	6.4	-2.0	4.4	5.7	0.2
AD013	Adult	Normal	6.0	2.6	0.7	-1.6	4.8	7.1	3.4	5.3
AD014	Adult	Normal	3.2	7.7	7.8	7.7	6.7	2.1	-4.5	-4.6
AD015	Adult	Normal	4.2	3.0	5.5	9.8	0.9	1.0	1.2	-1.3
AD016	Adult	Normal	-10.2	-10.4	6.1	1.2	-1.4	2.8	0.2	-16.3
AD017	Adult	Normal	3.2	-0.4	6.7	1.7	3.0	1.8	3.6	-3.5

A “--“ indicates these data were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

APPENDIX C: Individual Measures of Impedance Magnitude for Low Frequencies

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Age Group	Mastoid 5.4 N					Mastoid 4 N					Mastoid 2 N				
		125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz
CH185	A	33.38	37.05	29.69	26.84	27.33	31.22	30.10	27.56	24.03	26.97	31.84	32.94	23.44	26.27	25.55
CH139	A	14.91	33.99	27.66	27.26	24.87	43.69	32.78	28.68	21.72	24.11	40.47	31.41	29.77	29.18	23.96
CH123	A	43.23	33.88	26.21	27.47	24.05	42.66	35.97	26.50	20.55	22.98	37.09	29.00	26.06	26.15	25.56
CH162	A	32.57	32.94	26.13	25.65	23.75	39.08	33.65	30.04	28.17	23.00	41.45	27.23	25.27	23.59	23.24
CH161	A	44.07	34.27	31.16	26.25	28.97	40.86	37.38	32.89	32.04	28.87	38.34	37.61	30.47	27.09	28.23
CH130	A	39.54	33.28	27.44	28.33	25.52	31.43	22.62	28.45	27.28	24.45	32.54	20.89	25.60	28.16	26.24
CH187	A	38.75	30.17	26.46	23.75	23.80	21.04	24.21	27.81	30.04	26.94	33.19	26.30	26.94	23.22	24.51
CH177	A	32.83	26.56	24.60	27.73	26.09	36.84	28.49	23.77	19.33	18.34	40.73	30.52	24.41	21.15	21.27
CH149	A	34.53	22.85	28.46	25.78	23.75	34.86	27.22	26.27	23.58	28.51	27.50	22.21	25.15	25.93	28.34
CH197	A	44.41	38.33	35.52	31.40	31.02	46.24	35.92	30.74	27.86	27.78	42.77	33.77	29.82	24.58	24.24
CH141	A	39.84	37.17	34.19	29.45	29.50	31.09	31.33	31.43	26.87	25.68	33.58	33.69	30.28	25.52	26.11
CH175	A	39.03	30.70	26.56	26.63	25.33	42.77	30.13	28.68	27.83	20.89	35.90	30.42	27.19	21.10	23.31
CH189	A	53.44	35.82	31.29	31.52	28.74	45.70	37.75	31.89	29.67	27.91	38.57	34.41	26.74	22.01	24.98
CH146	A	38.39	29.60	27.93	26.15	27.75	46.18	32.08	23.60	23.53	25.83	34.91	27.30	22.67	21.38	20.31
CH160	A	40.38	35.49	27.68	30.62	29.82	29.66	32.97	25.03	26.63	18.36	34.36	26.39	22.11	17.44	18.44
CH109	A	34.95	30.64	28.69	20.87	26.96	37.20	32.84	32.43	26.25	25.66	34.97	27.78	25.23	20.93	26.24
CH131	A	--	--	--	--	--	--	--	--	--	--	--	--	--	--	--
CH173	A	--	--	--	--	--	--	--	--	--	--	--	--	--	--	--
CH134	B	40.36	34.76	29.97	27.39	22.27	37.39	30.03	22.44	19.00	18.95	32.98	25.91	22.92	18.35	17.73
CH122	B	39.02	37.08	33.27	30.87	28.82	40.18	18.47	25.87	25.50	21.74	33.00	49.60	24.27	24.82	21.43
CH165	B	44.83	41.08	36.54	33.26	32.94	20.21	39.90	31.72	29.78	26.02	44.44	34.65	31.55	25.12	26.19
CH179	B	46.80	39.05	33.21	34.59	29.59	34.83	34.87	29.95	27.12	26.27	39.63	33.32	29.03	19.74	12.98

A "--" indicates these data were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Forehead 5.4 N					Forehead 4 N					Forehead 2 N				
	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz
CH185	37.41	33.48	25.42	19.71	20.57	39.10	32.25	26.09	24.01	18.91	38.21	31.16	24.16	21.63	19.05
CH139	44.77	38.58	32.60	28.28	25.98	35.46	37.31	31.76	27.39	22.81	38.48	33.96	29.85	26.93	18.55
CH123	41.00	31.99	30.15	23.35	21.57	38.78	37.61	29.01	28.85	24.13	40.91	31.90	30.33	25.37	13.41
CH162	25.45	34.45	28.74	27.00	22.30	37.00	29.60	26.64	23.19	19.65	38.03	29.51	25.42	17.31	15.00
CH161	47.78	36.63	26.80	26.52	29.01	36.70	34.25	28.80	20.70	21.53	37.11	35.50	26.27	24.96	17.64
CH130	39.69	36.66	33.50	24.61	20.56	44.45	36.83	30.50	23.54	19.10	39.46	34.18	24.97	22.64	20.88
CH187	41.06	33.32	24.64	21.62	20.03	32.01	27.18	22.79	18.62	17.06	--	--	--	--	--
CH177	42.56	35.98	28.09	22.80	17.45	38.90	33.02	24.05	20.38	17.64	33.84	33.49	26.13	14.51	13.46
CH149	37.61	33.53	27.33	23.73	26.50	42.26	35.56	31.11	27.37	24.51	36.62	31.96	27.47	22.73	20.87
CH197	47.94	39.14	36.55	32.04	28.62	50.53	38.61	32.57	28.76	23.93	42.08	37.08	26.30	18.22	18.97
CH141	39.94	31.25	27.93	22.69	21.58	39.78	31.51	27.22	23.46	21.44	34.67	28.04	24.71	18.05	18.68
CH175	39.29	33.76	27.14	24.89	20.10	38.76	33.13	27.31	21.00	14.56	37.56	32.59	22.78	14.75	16.28
CH189	--	--	--	--	--	--	--	--	--	--	--	--	--	--	--
CH146	38.32	29.10	26.87	26.01	21.41	40.37	35.38	24.74	16.61	19.47	18.06	29.60	22.91	12.63	19.59
CH160	40.10	37.41	31.61	26.40	23.27	43.44	34.30	25.23	20.87	18.38	45.41	35.15	28.15	19.31	18.99
CH109	47.31	39.03	34.47	27.98	25.74	43.09	36.57	30.07	21.65	23.49	36.30	37.40	30.84	25.48	20.90
CH131	--	--	--	--	--	--	--	--	--	--	--	--	--	--	--
CH173	--	--	--	--	--	--	--	--	--	--	--	--	--	--	--
CH134	45.64	39.48	32.58	27.94	23.21	40.69	35.10	28.83	20.30	18.50	38.19	30.17	25.40	17.83	15.38
CH122	49.72	40.57	34.73	33.33	30.12	50.07	32.16	31.17	25.93	24.17	47.03	40.50	30.55	28.63	20.81
CH165	54.54	38.61	35.61	32.99	29.84	43.87	41.21	36.13	32.20	28.38	51.57	40.67	35.96	30.34	27.69
CH179	52.56	43.01	26.57	33.84	29.28	47.29	39.20	32.47	27.83	18.51	42.64	34.14	30.21	20.28	16.43

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Age Group	Mastoid 5.4 N					Mastoid 4 N					Mastoid 2 N				
		125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz
CH114	B	41.66	36.67	30.97	29.74	27.19	46.29	37.42	30.99	29.16	26.29	42.67	32.62	28.03	25.22	25.22
CH196	B	45.30	35.70	31.88	27.57	23.96	30.68	33.59	27.64	23.55	21.68	43.66	30.22	30.26	27.35	21.59
CH135	B	46.32	41.36	35.79	33.99	30.89	38.45	36.07	31.10	26.68	21.91	45.85	34.39	31.45	25.53	24.36
CH115	B	47.99	40.47	36.88	33.79	32.34	50.40	38.63	35.96	33.39	31.58	43.35	38.10	33.19	29.22	28.72
CH154	B	43.91	38.95	32.88	34.28	31.97	48.26	40.74	32.67	31.18	27.20	26.62	37.79	31.81	28.65	27.48
CH157	B	48.68	43.69	38.09	35.77	32.81	46.21	40.63	37.05	34.76	34.56	44.21	39.46	33.01	30.53	33.08
CH136	B	53.30	39.41	30.45	29.58	27.66	43.35	32.92	27.97	21.92	28.06	43.34	35.34	27.92	26.40	25.83
CH152	B	43.82	38.78	32.46	28.88	27.18	49.43	39.14	33.67	26.31	27.57	38.32	38.38	31.02	27.38	23.23
CH124	B	46.38	40.75	35.18	31.04	28.75	36.57	36.76	31.60	26.80	23.80	41.46	32.63	27.52	20.94	22.36
CH117	C	46.19	41.90	35.99	33.51	33.12	45.62	41.64	30.80	34.82	30.44	32.76	39.06	33.37	32.61	29.53
CH129	C	42.00	42.09	34.65	31.23	30.83	41.03	36.85	31.76	28.15	26.87	30.48	34.25	29.19	23.61	21.56
CH159	C	31.52	41.35	37.24	32.84	31.57	47.38	39.54	34.33	31.27	29.68	43.62	33.01	27.39	22.88	19.68
CH138	C	44.43	37.89	30.52	32.06	29.22	36.76	35.56	31.14	27.51	26.92	40.48	33.19	26.32	24.70	21.92
CH156	C	40.40	40.41	35.09	31.09	28.42	44.57	38.60	34.91	31.45	28.55	43.67	37.00	32.00	25.55	24.89
CH168	C	47.05	41.64	36.41	32.86	32.47	49.48	42.17	37.75	34.40	34.06	43.70	39.81	34.37	30.74	28.96
CH119	C	52.62	41.38	36.86	34.88	32.19	45.59	38.05	34.18	32.98	30.82	43.27	38.42	30.13	32.76	30.25
CH153	C	29.65	41.64	37.65	34.34	32.98	42.06	38.89	32.78	29.60	25.96	36.50	34.51	29.68	27.45	24.08
CH193	C	45.22	39.56	33.71	29.25	26.80	40.50	36.60	31.63	28.30	24.07	39.91	36.26	30.68	27.01	25.64
CH199	C	49.76	36.48	30.35	28.57	24.69	45.12	37.69	30.70	31.02	29.07	--	--	--	--	--
CH167	C	52.14	44.31	37.52	33.97	32.66	41.55	38.30	29.76	28.68	27.07	43.61	42.09	30.42	26.25	22.44
CH143	C	46.80	41.59	35.96	31.17	31.50	41.16	35.90	30.68	20.92	25.50	35.09	35.91	28.99	24.96	23.21
CH147	C	45.29	41.54	35.38	30.98	30.01	42.73	36.28	29.57	26.69	24.69	40.10	35.78	30.76	28.26	24.99

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Forehead 5.4 N					Forehead 4 N					Forehead 2 N				
	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz
CH114	46.18	40.86	32.54	29.14	25.84	30.51	33.91	28.19	24.09	22.01	42.66	34.77	26.78	20.69	20.20
CH196	43.32	38.01	33.54	30.60	27.91	39.74	37.03	31.14	23.14	23.07	35.80	33.29	28.83	26.58	21.04
CH135	50.44	42.44	36.01	31.23	27.58	53.24	40.65	30.42	25.33	27.94	38.33	27.78	26.55	24.83	18.37
CH115	49.03	40.71	34.58	29.87	24.39	50.76	37.64	33.02	27.13	21.91	42.94	38.13	27.18	20.65	21.72
CH154	47.38	44.51	38.43	34.79	31.01	46.99	37.56	27.98	27.01	23.53	38.54	37.11	27.24	28.20	22.41
CH157	52.87	38.48	36.18	32.18	30.70	47.89	41.73	34.71	29.38	28.03	42.71	38.15	32.93	28.69	22.08
CH136	44.02	42.33	37.76	33.09	31.73	42.47	40.83	34.51	31.18	26.37	42.70	37.22	32.46	29.71	24.73
CH152	44.78	40.29	33.87	28.05	22.24	42.73	37.40	34.50	26.47	12.08	22.55	32.44	25.09	20.54	17.14
CH124	41.79	39.70	32.65	29.87	25.22	40.37	33.60	30.22	25.68	19.69	40.19	27.75	27.57	16.37	20.80
CH117	49.56	44.56	39.34	34.24	29.33	41.75	40.82	35.45	31.33	26.26	39.66	37.28	31.24	26.92	20.60
CH129	47.55	41.50	35.11	31.79	28.13	48.12	40.86	34.52	31.40	25.16	39.92	32.97	28.76	25.14	17.79
CH159	41.40	39.72	35.11	33.35	26.81	42.91	37.67	29.36	24.15	25.90	38.25	39.74	30.27	24.45	20.26
CH138	45.58	41.39	34.87	28.20	23.02	42.63	34.80	29.13	23.15	23.12	40.86	32.02	30.20	24.51	23.07
CH156	57.65	43.78	38.58	35.84	32.40	51.10	35.05	28.93	27.37	24.72	40.86	32.02	30.20	24.51	23.07
CH168	57.61	45.54	39.69	36.51	34.32	32.23	45.40	38.69	31.89	28.47	45.71	37.13	33.44	31.32	28.84
CH119	52.20	44.43	38.74	34.92	31.81	47.16	44.90	37.40	32.08	29.42	43.41	40.87	35.30	29.15	25.83
CH153	47.69	45.57	39.56	35.98	33.50	41.66	42.99	37.65	34.24	28.47	47.65	40.27	34.78	31.52	28.43
CH193	46.88	39.01	33.26	30.39	28.18	47.41	39.13	31.48	28.24	26.13	40.84	33.82	29.96	22.77	18.99
CH199	48.14	39.97	30.60	30.44	24.77	41.22	36.14	25.77	28.65	25.56	34.18	35.23	28.58	24.68	23.74
CH167	49.53	40.75	35.40	28.97	26.81	48.53	39.94	33.06	25.82	16.32	46.75	38.14	31.53	29.49	24.12
CH143	50.72	44.53	36.04	31.02	29.09	44.03	40.20	34.52	24.60	23.92	25.50	35.72	29.02	25.35	22.31
CH147	49.60	43.18	36.35	32.39	27.99	45.39	40.06	34.14	28.68	23.92	43.76	35.78	30.63	27.30	22.39

A “--“ indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Age Group	Mastoid 5.4 N					Mastoid 4 N					Mastoid 2 N				
		125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz
CH172	C	49.77	44.70	39.64	36.45	32.89	45.74	41.94	36.52	33.04	29.73	42.66	40.70	33.99	29.25	27.58
CH116	C	27.80	41.58	35.71	32.35	29.90	44.56	40.17	35.58	32.51	30.75	44.86	38.34	32.03	28.78	28.57
CH126	D	26.82	42.40	39.20	35.59	32.23	53.06	42.34	37.11	33.72	30.84	45.94	41.28	36.19	31.98	30.36
CH191	D	45.07	45.87	37.58	34.48	32.34	45.34	40.47	31.89	32.38	31.75	40.34	36.07	26.24	28.14	26.80
CH188	D	46.59	40.45	34.86	31.47	28.66	45.24	39.42	33.48	29.37	27.56	41.21	39.80	33.87	30.27	27.13
CH184	D	53.71	45.23	39.53	37.13	33.89	52.30	40.55	35.27	30.20	28.95	41.55	41.77	35.05	33.40	30.72
CH186	D	50.38	40.81	36.98	34.01	32.55	46.40	40.22	34.56	30.73	25.40	39.44	37.78	30.27	28.17	28.77
CH144	D	51.06	44.26	37.55	33.27	31.69	45.49	39.50	32.74	30.32	27.83	33.14	36.86	32.42	28.84	23.12
CH107	D	59.75	43.82	36.54	34.37	31.95	46.78	40.00	34.52	30.73	26.24	46.82	38.86	29.16	27.97	25.01
CH140	D	44.37	40.92	32.93	32.78	26.70	39.89	34.77	29.47	25.67	22.83	37.32	33.50	27.46	22.57	17.54
CH174	D	60.55	47.22	41.75	38.06	35.16	45.45	40.44	35.75	31.66	28.50	45.24	41.24	38.13	35.54	32.64
CH101	D	49.78	43.49	35.46	32.94	31.07	49.61	40.48	34.17	31.17	29.77	40.59	38.43	30.91	31.75	28.69
CH112	D	50.42	41.01	33.81	33.69	30.65	47.81	40.07	35.90	33.54	31.07	44.61	37.47	30.91	27.88	30.01
CH145	D	50.96	44.05	35.86	35.54	34.79	49.00	43.97	38.87	35.30	32.67	48.42	36.55	32.75	31.90	30.60
CH113	D	40.55	35.07	26.48	24.24	20.54	28.60	40.46	34.92	30.85	29.24	41.86	37.76	30.49	30.95	26.77
CH169	D	47.78	41.02	34.72	29.67	26.74	47.37	37.77	29.56	29.17	24.15	43.80	32.90	29.36	26.26	23.38
CH118	D	46.15	38.78	32.46	28.06	25.67	35.74	31.93	25.91	18.95	20.64	29.06	35.21	28.71	24.26	22.54
CH195	D	33.66	44.02	37.78	33.70	31.64	47.20	39.70	32.17	30.43	30.03	41.96	37.93	31.94	26.54	20.97
CH102	D	51.74	43.35	37.97	35.10	33.40	45.81	41.52	35.62	29.61	31.05	16.74	39.09	33.97	29.97	25.91
CH125	D	47.94	41.14	34.37	33.86	29.82	43.98	38.14	32.76	31.57	28.06	22.35	39.63	32.22	31.66	26.91
CH133	D	49.67	45.18	37.74	33.92	34.19	41.78	39.80	35.27	31.58	28.30	43.28	35.94	31.54	30.20	27.02

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Forehead 5.4 N					Forehead 4 N					Forehead 2 N				
	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz
CH172	62.05	47.89	41.74	38.57	34.19	51.42	44.01	38.37	34.42	31.66	55.21	40.26	34.64	31.97	25.80
CH116	56.47	41.99	35.79	29.70	25.92	30.22	37.35	28.16	23.72	21.45	44.21	29.47	26.29	18.51	18.96
CH126	48.59	44.41	38.34	34.34	30.55	51.19	44.35	38.66	34.87	31.12	41.95	37.44	26.56	26.05	17.39
CH191	54.88	47.17	40.84	36.91	35.74	48.73	42.71	38.38	32.56	32.79	40.24	31.71	29.94	31.03	26.30
CH188	58.24	47.11	40.15	36.51	34.54	48.28	43.20	36.04	31.96	29.28	47.28	41.14	35.03	28.00	26.88
CH184	27.96	45.77	40.28	36.90	33.00	50.99	43.39	37.68	34.16	33.20	45.79	39.90	34.41	29.52	28.43
CH186	45.01	40.58	35.06	26.48	28.56	45.20	38.89	33.71	28.34	21.78	41.57	36.91	27.40	23.50	18.43
CH144	50.02	41.82	38.90	35.70	33.08	42.38	37.21	32.84	26.11	21.55	39.95	36.79	30.19	27.95	23.28
CH107	48.84	44.98	38.98	35.41	32.66	43.64	39.31	33.68	22.03	23.06	47.18	39.98	32.51	29.35	23.98
CH140	52.59	45.65	39.81	35.67	32.00	53.20	41.39	37.72	30.90	28.06	44.93	38.86	34.26	31.02	27.36
CH174	47.78	46.39	39.61	36.91	34.05	52.46	45.02	38.89	34.51	30.39	49.26	42.17	35.50	31.94	26.74
CH101	35.11	44.03	35.62	32.55	32.95	48.92	42.31	35.09	30.78	31.59	47.67	40.14	36.20	31.22	30.36
CH112	52.03	46.40	39.18	36.06	33.80	51.22	41.78	37.80	33.96	29.46	46.69	38.51	35.88	27.54	26.59
CH145	41.49	40.05	36.69	30.24	27.03	50.89	40.04	34.73	26.59	26.76	43.52	38.70	32.58	28.33	26.26
CH113	57.26	46.17	40.15	36.37	33.91	51.88	41.17	36.67	29.23	25.91	44.56	38.26	35.04	20.59	26.91
CH169	46.19	39.96	32.55	30.43	25.85	35.65	35.73	31.38	24.61	21.66	39.79	41.66	27.18	21.44	23.14
CH118	48.64	45.02	40.48	37.06	34.11	37.96	38.93	33.34	28.55	23.98	43.59	31.55	32.90	28.71	23.17
CH195	53.21	48.36	42.06	39.14	34.41	47.86	46.31	41.30	36.45	32.00	46.31	43.12	35.71	32.82	28.62
CH102	55.67	45.93	38.89	34.22	28.32	38.37	43.92	36.79	34.05	30.58	31.24	39.34	34.70	29.10	25.74
CH125	44.48	42.15	35.63	31.47	28.99	34.17	40.49	35.50	30.38	26.35	44.28	41.10	33.33	27.26	22.77
CH133	45.51	46.10	41.27	36.44	33.29	35.42	42.61	35.20	32.45	31.79	48.56	41.87	35.93	32.51	29.82

A “--“ indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Age Group	Mastoid 5.4 N					Mastoid 4 N					Mastoid 2 N				
		125 Hz	**251 Hz**	**501 Hz**	**758 Hz**	**1000 Hz**	**125 Hz**	**251 Hz**	**501 Hz**	**758 Hz**	**1000 Hz**	**125 Hz**	**251 Hz**	**501 Hz**	**758 Hz**	**1000 Hz**
AD001	Adult	48.29	40.59	35.83	32.91	29.25	49.25	37.95	34.35	30.73	26.65	50.09	37.77	30.85	24.57	25.55
AD002	Adult	52.34	42.91	38.85	35.51	34.08	47.93	42.89	40.75	38.94	37.38	49.63	42.32	36.87	34.39	29.28
AD003	Adult	51.97	43.13	37.32	32.80	30.82	41.34	35.15	29.33	22.81	24.18	37.04	33.92	29.61	22.29	25.22
AD004	Adult	44.89	39.86	36.12	30.94	30.08	39.37	35.37	30.89	28.00	27.68	41.01	39.02	36.36	32.85	29.09
AD005	Adult	49.77	39.80	35.74	31.55	29.07	42.59	43.39	37.06	32.52	29.88	41.33	36.56	30.33	25.38	22.77
AD006	Adult	44.18	43.15	38.28	34.85	32.02	48.27	40.33	34.92	33.52	30.16	43.40	40.14	36.42	34.15	34.42
AD007	Adult	34.22	45.52	40.48	36.52	33.73	43.00	42.70	36.75	33.39	32.02	53.22	45.11	41.23	38.45	35.46
AD008	Adult	46.46	43.82	39.05	35.45	33.03	32.96	42.94	37.35	34.23	31.63	46.69	42.12	35.00	31.40	29.75
AD009	Adult	52.32	44.70	39.33	36.35	34.19	44.96	40.01	35.17	32.74	30.74	34.12	41.65	35.64	32.45	27.02
AD010	Adult	--	--	--	--	--	--	--	--	--	--	--	--	--	--	--
AD011	Adult	47.79	39.44	35.22	31.42	28.37	39.74	33.00	28.22	27.72	24.49	35.38	36.98	33.52	29.17	26.45
AD012	Adult	48.82	44.54	38.25	35.87	33.38	48.93	41.81	34.24	32.91	29.50	40.21	34.41	23.66	27.30	25.65
AD013	Adult	47.59	41.26	35.17	29.63	29.53	46.21	38.55	31.82	30.71	27.55	43.28	38.84	31.87	28.78	25.97
AD014	Adult	52.27	41.90	35.98	32.89	28.81	36.86	36.73	30.73	26.87	24.97	45.88	33.59	27.70	23.20	18.15
AD015	Adult	47.14	42.35	37.26	34.11	29.71	48.72	41.85	36.92	32.63	28.84	38.67	33.25	31.02	29.44	27.90
AD016	Adult	41.17	35.93	33.01	26.84	25.19	40.02	34.38	28.92	22.50	22.13	37.23	31.48	29.29	24.80	21.40
AD017	Adult	49.67	40.67	34.76	31.49	30.46	44.59	36.55	31.60	27.89	26.13	37.74	32.72	27.96	26.55	26.20

A “--“ indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for low frequencies

Subject #	Forehead 5.4 N					Forehead 4 N					Forehead 2 N				
	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz	125 Hz	251 Hz	501 Hz	758 Hz	1000 Hz
AD001	50.51	47.76	41.85	37.23	35.35	53.22	44.87	39.16	34.43	32.34	41.57	40.20	32.76	30.46	26.91
AD002	55.12	46.78	41.00	37.99	35.68	43.47	45.58	39.30	35.27	32.23	49.42	40.87	35.17	31.15	28.36
AD003	44.98	44.96	38.20	33.83	30.78	42.85	42.74	37.77	33.49	28.26	45.27	39.37	34.23	25.64	27.09
AD004	58.75	45.94	40.00	34.92	32.56	43.96	41.97	37.00	33.69	30.06	48.15	41.64	36.70	31.64	26.15
AD005	45.73	44.88	38.37	35.59	32.19	47.80	41.26	40.27	34.70	32.62	46.25	36.79	34.90	27.61	24.84
AD006	51.66	44.56	40.08	34.97	32.08	48.02	44.01	40.05	35.53	30.79	45.11	41.31	35.26	32.27	30.93
AD007	53.22	45.11	41.23	38.45	35.46	43.45	36.89	33.17	31.20	25.95	39.48	40.54	34.18	30.18	24.91
AD008	48.00	47.47	41.63	37.98	35.54	44.54	43.82	39.34	35.64	32.86	43.18	43.70	35.45	34.53	27.26
AD009	52.42	44.80	39.68	35.24	32.51	45.29	42.10	36.91	32.68	31.82	48.11	42.24	37.00	33.84	28.85
AD010	--	--	--	--	--	--	--	--	--	--	--	--	--	--	--
AD011	54.64	45.11	40.64	36.82	33.09	33.24	45.10	40.60	36.82	34.31	41.67	39.36	33.67	30.84	26.61
AD012	51.44	45.34	40.94	34.38	32.63	54.37	45.50	37.68	35.83	31.97	48.04	42.67	39.05	33.22	28.38
AD013	50.24	42.35	38.29	32.56	27.62	50.23	39.67	34.75	30.70	26.36	54.13	37.04	31.85	28.18	26.69
AD014	56.41	49.32	43.98	40.42	37.72	50.67	45.80	39.88	35.28	31.52	29.92	42.22	34.98	34.75	31.26
AD015	31.83	46.25	37.34	36.80	34.03	46.75	43.98	38.09	33.03	31.02	47.78	40.23	30.08	31.14	28.13
AD016	49.26	41.39	34.56	31.96	27.00	40.47	35.65	30.26	28.09	24.00	34.32	37.53	30.83	27.76	24.11
AD017	50.52	45.24	38.02	34.44	28.75	30.25	42.03	36.04	33.18	31.73	44.13	40.08	34.85	31.67	29.52

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

APPENDIX D: Individual Measures of Impedance Magnitude for High Frequencies

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	**Age Group**	**Mastoid 5.4 N**						**Mastoid 4 N**					
		1995 Hz	**3019 Hz**	**3981 Hz**	**6025 Hz**	**7943 Hz**	**10000 Hz**	**1995 Hz**	**3019 Hz**	**3981 Hz**	**6025 Hz**	**7943 Hz**	**10000 Hz**
CH185	A	25.63	30.67	32.85	26.55	34.27	42.64	24.00	28.30	32.96	38.90	36.55	28.84
CH139	A	23.05	27.28	29.27	34.42	38.20	40.50	27.13	30.33	31.75	37.20	40.04	35.81
CH123	A	29.67	25.30	32.10	37.03	39.01	42.90	23.13	24.00	31.07	33.39	35.52	39.71
CH162	A	22.31	22.57	26.35	30.78	29.64	28.43	19.68	23.39	23.40	30.00	35.38	25.62
CH161	A	22.16	22.52	23.52	30.38	33.59	27.80	23.66	19.16	23.11	33.35	14.92	31.17
CH130	A	25.22	24.95	31.99	37.31	42.12	45.69	27.57	29.78	32.24	25.59	23.83	29.61
CH187	A	22.80	28.06	30.83	35.31	36.75	40.76	23.89	28.78	31.71	34.22	37.31	39.26
CH177	A	25.74	29.41	33.09	39.72	37.12	32.16	23.11	28.57	29.06	35.03	42.98	41.50
CH149	A	25.49	26.13	27.23	30.63	35.10	32.35	26.47	25.60	26.80	27.77	29.28	27.74
CH197	A	25.03	23.97	24.97	32.00	30.65	33.01	22.94	27.57	30.76	39.92	38.64	40.52
CH141	A	23.82	28.02	31.46	36.65	37.77	40.54	25.08	29.03	31.97	34.35	37.49	37.42
CH175	A	24.30	29.45	33.77	37.45	40.14	39.60	25.21	30.15	33.22	36.82	40.11	43.91
CH189	A	27.21	26.47	29.38	32.38	20.85	15.46	25.86	30.01	29.99	34.15	34.54	32.30
CH146	A	23.72	29.42	32.06	37.71	36.39	43.55	26.45	30.05	32.62	37.41	41.40	45.52
CH160	A	23.42	28.12	32.68	39.76	41.80	43.46	23.17	28.07	31.04	31.95	21.06	20.14
CH109	A	28.51	28.16	25.22	36.92	34.73	24.44	25.19	29.22	32.32	37.73	40.54	40.10
CH131	A	--	--	--	--	--	--	--	--	--	--	--	--
CH173	A	--	--	--	--	--	--	--	--	--	--	--	--
CH134	B	26.31	29.05	31.97	35.50	35.26	33.91	19.88	25.94	28.04	33.76	33.19	30.16
CH122	B	25.67	27.06	30.44	35.38	36.12	43.30	19.15	15.65	23.61	25.68	29.69	26.51
CH165	B	26.22	21.18	25.93	30.52	31.16	37.20	23.98	25.10	23.32	31.56	14.55	26.70
CH179	B	25.12	18.74	16.22	24.63	23.56	22.47	24.25	27.67	25.33	30.50	35.28	30.97

A “--“ indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

	Mastoid 2 N						Forehead 5.4 N					
Subject #	**1995 Hz**	**3019 Hz**	**3981 Hz**	**6025 Hz**	**7943 Hz**	**10000 Hz**	**1995 Hz**	**3019 Hz**	**3981 Hz**	**6025 Hz**	**7943 Hz**	**10000 Hz**
CH185	27.21	31.23	34.59	36.27	37.36	39.17	20.34	29.59	32.75	34.45	36.24	29.69
CH139	25.45	30.31	32.81	36.18	39.02	44.21	24.08	30.63	32.22	36.72	37.65	41.11
CH123	23.56	28.52	33.51	27.92	31.35	35.45	23.78	30.02	32.54	35.38	38.02	41.61
CH162	24.12	30.57	32.95	37.33	39.81	40.99	24.11	28.54	33.44	36.88	36.48	41.31
CH161	19.80	25.46	28.94	26.92	28.10	26.63	22.27	28.21	31.24	36.22	36.69	36.48
CH130	24.82	26.08	31.55	36.12	38.08	36.76	24.27	30.06	32.41	35.58	38.93	42.38
CH187	26.73	31.32	32.65	39.48	43.74	42.20	26.01	31.64	33.32	37.98	39.44	41.66
CH177	20.12	23.48	28.26	35.09	36.66	29.17	25.77	30.28	32.13	38.12	38.67	26.70
CH149	27.41	20.81	25.67	35.82	27.37	26.52	22.82	29.15	31.92	38.52	39.16	42.53
CH197	22.65	27.36	22.92	34.50	33.57	33.43	20.13	28.14	30.14	36.07	38.09	37.91
CH141	22.83	28.33	32.36	38.75	29.45	28.10	--	--	--	--	--	--
CH175	24.68	30.12	32.30	37.92	37.40	38.91	26.38	29.94	33.53	36.06	38.15	41.26
CH189	23.77	27.63	29.08	39.27	31.95	29.24	--	--	--	--	--	--
CH146	25.84	28.98	31.05	37.03	36.11	34.53	24.37	30.07	32.89	36.01	31.86	38.50
CH160	18.33	23.85	22.26	36.95	22.85	27.97	23.81	28.15	32.95	32.18	17.77	27.71
CH109	25.47	29.87	32.74	37.41	40.49	39.68	23.28	27.99	31.49	34.90	37.46	39.28
CH131	--	--	--	--	--	--	--	--	--	--	--	--
CH173	--	--	--	--	--	--	--	--	--	--	--	--
CH134	25.90	29.40	31.04	35.70	34.71	31.18	22.78	28.82	30.84	34.74	39.88	38.67
CH122	17.28	20.83	26.28	23.40	30.23	27.86	22.80	29.01	31.87	37.18	40.85	42.12
CH165	23.46	24.70	26.51	35.20	33.27	37.05	21.52	28.44	34.30	38.55	38.77	42.06
CH179	21.30	20.07	13.69	31.96	27.32	24.87	20.82	29.19	31.37	37.00	39.38	42.15

A “--“ indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Forehead 4 N 1995 Hz	Forehead 4 N 3019 Hz	Forehead 4 N 3981 Hz	Forehead 4 N 6025 Hz	Forehead 4 N 7943 Hz	Forehead 4 N 10000 Hz
CH185	26.08	29.87	33.36	37.24	39.19	42.40
CH139	21.65	30.43	31.81	35.76	38.13	41.65
CH123	19.02	27.60	32.83	33.90	31.67	35.61
CH162	25.79	30.28	32.99	37.39	37.64	41.36
CH161	25.19	29.77	31.41	39.12	40.67	43.86
CH130	24.58	30.55	31.07	35.55	39.44	42.76
CH187	26.44	28.99	31.36	33.51	39.57	40.78
CH177	24.67	28.22	32.63	34.45	39.08	32.47
CH149	18.89	24.57	30.29	33.39	33.88	35.71
CH197	23.17	28.81	32.27	35.69	38.57	39.88
CH141	21.17	28.43	29.89	39.19	38.24	39.35
CH175	26.37	30.34	32.12	35.08	39.84	41.19
CH189	--	--	--	--	--	--
CH146	24.10	27.51	31.80	35.52	38.94	40.65
CH160	20.17	28.49	30.42	37.18	36.28	41.00
CH109	23.08	28.33	32.30	34.97	37.48	39.72
CH131	--	--	--	--	--	--
CH173	--	--	--	--	--	--
CH134	26.15	29.89	32.39	31.22	37.80	42.31
CH122	21.88	29.05	33.29	39.29	41.47	43.08
CH165	24.29	30.60	34.30	38.89	38.94	42.37
CH179	25.28	30.71	33.36	29.55	39.94	42.58

Subject #	Forehead 2 N 1995 Hz	Forehead 2 N 3019 Hz	Forehead 2 N 3981 Hz	Forehead 2 N 6025 Hz	Forehead 2 N 7943 Hz	Forehead 2 N 10000 Hz
CH185	25.42	29.23	32.42	34.86	37.80	40.28
CH139	26.51	29.85	32.93	34.95	34.17	36.85
CH123	23.29	26.25	34.72	28.58	32.10	18.84
CH162	25.14	30.19	33.50	35.12	38.73	42.48
CH161	26.47	30.70	32.17	35.65	36.52	42.37
CH130	24.58	29.39	32.73	36.10	39.89	41.93
CH187	--	--	--	--	--	--
CH177	16.39	24.19	31.83	29.58	31.80	35.35
CH149	23.86	29.03	31.80	34.71	37.98	40.56
CH197	24.27	28.48	33.94	35.84	36.64	40.24
CH141	21.76	26.39	28.50	36.83	37.82	35.24
CH175	25.51	29.89	31.74	35.91	38.76	37.54
CH189	--	--	--	--	--	--
CH146	26.87	31.17	32.67	37.38	38.20	41.54
CH160	23.36	27.68	30.02	34.27	39.71	45.57
CH109	24.86	29.74	32.79	33.98	35.11	40.71
CH131	--	--	--	--	--	--
CH173	--	--	--	--	--	--
CH134	26.63	30.95	33.92	35.15	37.14	42.15
CH122	24.44	28.93	32.79	38.25	38.67	41.01
CH165	19.30	26.35	31.35	32.96	33.73	36.81
CH179	24.11	30.41	31.37	36.66	37.03	39.86

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Age Group	Mastoid 5.4 N						Mastoid 4 N					
		1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz
CH114	B	24.29	28.83	32.33	38.34	35.32	38.79	24.77	31.11	33.84	39.04	36.00	35.49
CH196	B	24.76	23.92	28.83	33.25	33.84	35.38	21.48	25.04	25.38	28.95	38.00	31.54
CH135	B	23.63	25.71	31.20	34.78	31.75	36.03	23.54	27.46	26.77	32.60	35.34	36.53
CH115	B	26.57	21.64	22.93	23.46	27.98	24.68	25.47	23.96	27.24	30.17	24.77	28.40
CH154	B	25.03	26.24	28.30	27.93	26.84	31.39	22.85	21.83	22.48	32.33	31.02	33.61
CH157	B	22.88	26.86	31.59	33.35	34.56	37.94	26.21	25.75	30.39	33.36	35.60	36.07
CH136	B	28.28	25.90	24.37	23.28	33.05	31.65	24.51	21.51	19.77	30.73	26.78	27.14
CH152	B	24.59	27.73	29.30	35.93	37.79	36.61	21.97	26.22	29.18	42.10	38.00	37.04
CH124	B	23.84	31.07	34.18	39.04	41.90	42.32	25.23	31.25	32.18	25.05	38.76	38.86
CH117	C	19.58	21.20	23.05	32.75	24.41	31.12	22.29	20.24	21.55	32.31	25.93	31.96
CH129	C	20.13	19.34	21.80	28.57	28.00	30.03	20.35	20.69	21.85	33.79	24.83	26.27
CH159	C	23.44	24.31	27.08	35.06	26.55	31.33	20.59	21.11	23.94	32.15	25.05	28.54
CH138	C	20.61	21.78	28.13	29.97	27.95	28.32	22.38	23.79	24.39	26.86	28.19	26.84
CH156	C	26.07	25.38	29.21	31.27	27.39	31.16	21.10	25.34	25.11	23.36	26.96	31.65
CH168	C	25.04	23.23	26.07	28.59	30.57	31.88	26.64	27.53	29.37	35.07	36.04	34.90
CH119	C	27.60	26.88	28.87	32.72	24.64	28.72	23.59	22.72	23.91	34.82	23.15	33.62
CH153	C	25.39	26.99	30.04	31.63	29.11	30.20	18.21	12.49	17.35	13.86	22.44	27.17
CH193	C	26.04	29.98	28.97	33.85	34.48	33.55	25.53	28.40	28.09	36.90	30.64	33.95
CH199	C	22.96	22.04	24.09	29.90	21.08	25.55	24.84	27.81	32.46	36.13	28.62	27.83
CH167	C	24.31	22.68	23.29	33.91	32.37	31.75	24.53	23.06	28.04	29.18	33.59	32.47
CH143	C	26.21	27.31	28.71	36.28	37.74	38.95	19.22	22.98	23.37	24.88	28.01	24.14
CH147	C	24.00	29.94	31.03	36.66	38.39	37.30	22.14	25.48	30.96	32.67	32.43	31.78

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Mastoid 2 N						Forehead 5.4 N					
	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz
CH114	14.44	27.36	27.84	36.85	35.45	37.02	24.30	30.13	32.66	37.78	39.63	43.47
CH196	18.87	26.40	20.97	31.99	34.24	35.17	25.84	29.11	32.61	37.28	39.09	28.71
CH135	24.38	27.18	30.45	32.21	32.87	35.46	24.16	30.62	33.66	37.24	38.95	42.40
CH115	20.53	20.71	27.74	31.85	16.84	27.41	22.65	29.54	32.51	35.16	37.90	37.06
CH154	23.80	19.59	18.35	28.26	29.10	27.24	20.54	28.24	32.15	34.63	39.26	41.08
CH157	21.52	19.73	23.99	18.15	30.39	29.66	25.97	33.02	36.20	34.81	42.54	36.20
CH136	24.71	15.84	17.97	32.10	24.18	29.39	21.35	27.60	33.90	36.04	39.47	42.47
CH152	22.73	28.51	32.25	39.03	30.22	36.83	23.95	29.95	34.26	37.58	39.27	43.58
CH124	24.26	28.85	31.29	31.72	37.95	37.24	24.91	31.56	34.18	36.96	38.91	41.80
CH117	23.15	19.60	17.19	35.31	25.53	30.21	22.57	30.40	35.00	37.37	39.46	42.26
CH129	21.24	25.98	26.55	28.46	20.15	34.03	23.64	30.10	33.46	37.08	39.60	42.60
CH159	21.76	27.66	30.84	26.65	28.85	32.56	24.29	30.61	32.51	38.63	40.98	44.57
CH138	16.03	18.81	24.61	27.78	22.49	23.11	25.06	29.91	33.76	36.70	39.08	42.33
CH156	22.50	24.04	20.01	27.40	22.07	24.80	25.49	29.58	34.12	37.43	40.78	43.42
CH168	21.88	21.99	25.11	25.40	36.31	30.02	23.34	27.17	32.02	36.28	37.60	41.75
CH119	22.88	19.11	28.51	33.00	28.32	32.51	21.69	29.81	32.73	38.49	40.48	44.01
CH153	19.10	21.69	20.81	22.06	17.17	22.29	23.24	29.32	33.74	36.49	40.50	41.84
CH193	24.76	25.13	28.21	32.50	14.70	30.05	24.78	30.17	33.98	38.45	40.02	42.48
CH199	--	--	--	--	--	--	23.85	30.71	34.00	35.87	39.17	40.43
CH167	12.04	13.62	17.54	21.32	22.53	25.95	23.13	30.17	33.71	30.23	38.61	39.06
CH143	24.34	27.59	23.35	36.44	30.93	30.19	24.44	29.41	33.12	35.40	39.99	43.31
CH147	25.18	30.17	32.06	31.35	29.31	31.97	22.11	28.08	33.20	37.51	39.25	41.85

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Forehead 4 N						Forehead 2 N					
	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz
CH114	26.50	31.14	33.23	38.13	39.59	42.85	26.85	31.38	32.67	40.09	41.42	45.23
CH196	23.04	30.07	31.72	35.00	36.80	40.19	20.35	28.38	33.15	35.60	34.96	36.55
CH135	21.39	28.92	26.93	37.33	39.81	42.42	23.92	27.43	33.43	35.97	39.00	43.78
CH115	24.19	30.11	33.57	35.46	38.94	41.98	23.74	30.90	33.26	36.46	38.10	42.54
CH154	24.06	29.96	33.25	39.72	39.34	42.84	26.38	30.13	33.98	36.08	39.83	43.01
CH157	24.76	30.92	34.37	36.77	39.99	40.39	24.35	30.40	33.92	37.79	41.10	43.16
CH136	24.32	30.07	34.81	35.74	39.86	42.42	25.73	30.13	34.18	36.36	39.80	42.77
CH152	26.52	30.65	34.00	37.09	39.36	43.10	26.83	30.68	32.74	37.69	40.35	42.73
CH124	25.65	30.71	33.82	35.96	40.05	42.87	26.72	30.64	32.81	37.58	40.33	42.92
CH117	24.60	31.11	32.74	39.64	39.20	41.46	21.25	28.67	31.56	33.20	36.32	41.50
CH129	25.06	30.53	32.64	37.94	31.76	34.41	27.24	32.02	33.43	37.34	39.89	42.93
CH159	21.93	29.18	32.33	37.33	38.31	43.38	26.25	30.64	34.03	34.32	39.44	41.90
CH138	25.65	30.28	32.62	36.35	38.94	42.46	26.37	30.59	33.40	36.33	38.92	42.44
CH156	24.07	29.60	32.17	37.04	40.82	40.20	26.37	30.59	33.40	36.33	38.92	42.44
CH168	24.18	30.68	34.49	38.41	40.35	43.72	23.94	30.25	34.44	37.53	43.14	44.41
CH119	21.69	28.41	33.02	39.12	38.41	42.26	23.70	30.04	33.81	37.77	39.38	42.61
CH153	21.52	30.34	33.70	37.46	39.67	42.50	25.42	30.23	32.85	37.15	39.96	42.65
CH193	25.04	30.97	32.35	35.61	39.63	43.25	27.19	31.34	32.00	36.27	39.92	43.36
CH199	23.36	30.46	34.56	39.14	40.78	42.24	24.13	30.08	32.46	36.06	39.31	42.62
CH167	25.83	30.71	32.92	37.11	38.78	41.14	26.55	31.33	34.35	37.71	40.68	42.95
CH143	24.97	30.67	34.45	37.79	40.76	43.29	26.24	31.48	33.00	37.50	40.87	43.70
CH147	20.40	26.12	28.75	35.18	38.68	39.18	25.42	31.46	34.74	36.93	39.56	43.34

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Age Group	Mastoid 5.4 N						Mastoid 4 N					
		1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz
CH172	C	31.20	26.61	25.63	34.04	36.74	36.86	26.81	24.93	15.58	33.59	28.45	28.03
CH116	C	24.73	29.90	32.88	37.03	38.68	39.19	25.00	26.87	32.05	38.16	34.20	33.85
CH126	D	22.50	28.17	31.78	34.82	30.32	36.00	23.42	26.13	26.81	36.23	28.88	36.07
CH191	D	26.33	22.37	24.93	25.34	17.61	28.08	25.61	20.36	17.35	28.84	23.51	22.61
CH188	D	21.72	20.06	19.16	18.91	18.92	26.92	22.68	14.02	10.30	23.78	21.33	31.14
CH184	D	27.46	27.97	32.09	36.97	33.80	32.88	22.92	25.86	22.92	30.53	14.84	23.00
CH186	D	26.02	26.55	25.78	33.33	32.25	29.74	24.57	28.35	30.17	33.73	36.42	35.44
CH144	D	23.33	25.48	30.56	30.00	33.97	36.31	21.31	16.49	17.69	25.51	25.36	23.22
CH107	D	26.19	27.59	28.10	37.42	39.44	35.21	27.42	30.12	30.96	39.66	35.05	31.55
CH140	D	19.67	21.25	23.89	26.76	27.58	32.09	24.59	30.58	31.09	33.54	32.04	33.45
CH174	D	28.86	23.04	23.81	30.56	26.03	29.92	22.47	19.09	22.32	25.07	16.49	28.23
CH101	D	23.75	21.16	24.74	26.01	19.66	33.33	21.64	23.37	25.58	26.03	27.53	33.58
CH112	D	22.16	26.32	28.96	32.74	25.36	34.09	24.90	26.09	29.56	26.05	22.17	31.36
CH145	D	27.55	26.03	30.81	37.21	32.87	37.71	25.33	27.67	31.18	36.68	34.25	36.80
CH113	D	14.11	19.51	25.55	27.38	20.61	28.07	22.71	25.99	28.14	32.34	28.01	31.43
CH169	D	24.39	29.62	32.61	35.50	34.48	34.17	25.75	29.13	31.38	34.20	33.21	32.59
CH118	D	25.42	28.81	28.26	35.88	34.02	35.14	27.52	29.66	29.87	34.54	32.10	30.95
CH195	D	26.61	25.35	25.50	34.18	33.22	34.78	23.42	22.29	26.36	33.37	29.62	30.06
CH102	D	29.17	25.10	22.56	24.65	28.72	29.85	25.90	20.96	17.86	26.13	28.50	26.90
CH125	D	24.06	19.35	17.81	30.78	27.00	32.00	18.64	13.38	21.18	26.10	17.23	29.45
CH133	D	27.97	27.19	22.09	27.98	24.60	30.80	23.84	22.30	23.38	31.98	20.16	24.63

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

	Mastoid 2 N						Forehead 5.4 N					
Subject #	**1995 Hz**	**3019 Hz**	**3981 Hz**	**6025 Hz**	**7943 Hz**	**10000 Hz**	**1995 Hz**	**3019 Hz**	**3981 Hz**	**6025 Hz**	**7943 Hz**	**10000 Hz**
CH172	24.18	23.44	25.39	21.20	27.12	28.83	31.50	25.83	27.10	38.01	39.50	40.17
CH116	24.04	25.21	29.96	36.15	26.66	30.10	24.68	29.97	34.46	36.79	40.64	42.77
CH126	23.92	29.06	33.45	36.26	35.70	36.67	23.17	29.97	33.05	37.93	39.60	43.82
CH191	16.04	13.66	16.94	31.35	17.13	25.77	24.61	28.89	34.49	36.91	39.23	42.14
CH188	21.64	23.19	18.55	20.34	23.07	23.38	22.89	29.26	32.85	37.72	40.30	42.83
CH184	26.79	28.29	31.58	31.43	28.82	31.82	23.77	28.99	32.90	38.60	41.53	44.37
CH186	24.92	30.97	32.81	33.37	31.27	28.82	24.56	29.91	33.00	36.28	39.44	42.63
CH144	22.48	26.26	28.58	34.14	28.76	32.18	22.23	28.54	32.84	36.70	39.44	43.15
CH107	20.86	22.93	25.20	29.51	28.30	24.37	20.99	27.95	31.51	37.56	38.51	41.15
CH140	14.26	17.93	22.43	32.42	21.95	30.83	23.21	29.24	33.27	37.43	41.13	43.83
CH174	22.63	21.38	19.84	27.78	13.96	24.66	22.61	26.96	33.64	37.38	38.76	42.93
CH101	22.52	18.81	23.31	23.44	10.63	30.17	24.10	29.30	33.37	39.58	41.23	44.17
CH112	19.68	19.71	18.89	27.21	26.94	27.31	23.04	28.41	33.19	37.79	40.48	43.32
CH145	21.61	18.45	20.22	24.35	21.44	27.37	23.61	30.19	33.41	37.65	40.41	43.85
CH113	18.84	21.74	24.27	31.92	23.45	30.43	25.13	27.01	31.49	37.13	39.84	43.37
CH169	23.30	25.90	28.50	32.13	31.79	35.42	23.55	29.75	32.37	37.27	39.85	42.55
CH118	25.98	26.54	28.01	37.39	33.83	31.91	25.35	28.39	30.77	36.87	38.92	42.56
CH195	23.31	28.63	28.30	32.30	29.03	24.98	24.34	28.25	33.65	36.88	39.80	42.56
CH102	21.65	13.66	20.68	22.16	25.26	27.14	22.87	28.61	32.21	36.48	40.08	43.25
CH125	20.73	23.52	24.44	35.71	27.37	30.66	24.73	29.82	32.12	36.74	38.80	42.63
CH133	23.33	18.75	23.66	29.27	24.94	31.10	24.03	27.08	31.13	37.41	40.82	43.44

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Forehead 4 N						Forehead 2 N					
	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz
CH172	25.14	29.16	32.41	37.72	41.33	41.75	25.20	26.19	30.49	31.19	29.40	34.33
CH116	26.10	31.57	35.06	38.45	40.50	42.63	27.14	31.75	34.43	36.04	38.58	42.74
CH126	22.01	29.47	33.13	37.57	40.42	43.91	26.79	32.15	35.19	38.31	41.24	44.03
CH191	24.19	29.92	33.57	38.15	40.85	44.37	25.87	32.51	34.92	37.11	36.48	40.26
CH188	22.84	29.16	34.60	35.46	38.49	41.27	24.74	30.42	34.74	35.80	41.72	43.58
CH184	23.85	23.80	29.60	33.69	34.13	37.94	22.14	26.01	34.60	38.65	41.79	44.88
CH186	26.04	31.70	33.02	38.32	40.21	36.92	25.36	30.59	33.17	34.23	38.56	40.63
CH144	25.31	29.36	31.60	37.02	38.58	43.83	24.77	30.55	32.18	37.87	40.94	43.47
CH107	25.26	31.29	33.04	40.50	39.99	37.27	26.98	31.08	33.44	42.92	40.16	43.30
CH140	22.47	26.11	27.12	33.35	30.78	33.98	24.81	32.17	33.71	38.30	43.69	44.01
CH174	22.11	27.85	34.82	37.18	39.87	43.45	20.27	27.31	31.07	36.42	37.38	42.05
CH101	23.01	29.15	32.46	36.35	37.28	40.78	22.95	29.94	34.24	37.05	40.50	44.36
CH112	24.16	29.46	33.37	37.74	41.78	43.84	26.03	31.21	33.96	38.21	41.88	44.83
CH145	24.08	31.43	32.65	36.46	39.77	42.63	25.80	31.97	35.67	36.93	40.64	43.75
CH113	25.36	26.92	31.92	36.02	36.97	41.37	23.54	29.88	34.11	37.96	39.45	42.98
CH169	22.66	28.38	33.27	36.59	39.25	41.41	23.90	30.71	33.11	34.75	36.35	35.28
CH118	26.14	30.65	33.16	37.50	40.50	43.61	24.33	30.66	32.95	40.42	40.12	44.25
CH195	24.78	28.33	32.62	36.77	39.34	41.90	19.49	25.09	30.79	37.20	26.10	41.93
CH102	23.38	24.89	30.42	36.29	40.44	36.34	19.02	29.48	31.38	34.21	39.53	41.32
CH125	24.67	30.41	32.29	39.31	39.08	42.67	25.86	30.58	34.82	37.43	38.44	42.79
CH133	20.73	29.29	32.95	37.08	40.12	43.28	24.90	29.96	34.08	37.10	40.43	42.87

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Age Group	Mastoid 5.4 N						Mastoid 4 N					
		1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10000 Hz
AD001	Adult	23.51	21.11	24.47	29.03	22.44	26.63	22.61	20.94	21.91	31.40	23.33	25.63
AD002	Adult	27.20	28.47	30.02	36.02	38.28	41.19	37.86	35.00	33.55	30.09	31.39	31.33
AD003	Adult	23.41	26.37	30.15	34.35	35.23	38.29	20.93	24.08	26.50	30.18	25.27	33.08
AD004	Adult	24.59	27.85	30.81	32.59	28.81	31.53	21.79	20.97	23.30	25.22	25.23	24.73
AD005	Adult	23.15	27.08	30.62	31.97	30.73	33.22	23.25	25.66	30.64	31.78	33.21	31.81
AD006	Adult	25.85	25.53	26.87	31.15	30.74	31.21	26.98	28.04	28.95	30.51	30.31	29.51
AD007	Adult	27.36	23.52	27.60	33.65	26.29	33.64	25.93	19.89	24.44	32.37	18.47	32.46
AD008	Adult	26.82	27.41	30.46	34.45	32.67	35.44	25.27	23.40	25.55	37.12	22.95	33.94
AD009	Adult	26.56	25.52	29.37	32.65	29.96	33.04	23.28	21.96	20.16	31.93	18.77	31.97
AD010	Adult	--	--	--	--	--	--	--	--	--	--	--	--
AD011	Adult	22.95	24.79	25.81	28.85	30.87	31.56	25.55	23.92	24.50	23.03	24.82	23.22
AD012	Adult	26.39	32.48	35.32	36.94	36.03	36.59	24.82	31.22	32.34	33.02	28.95	32.09
AD013	Adult	24.58	29.32	31.20	35.56	33.72	36.41	22.06	25.17	22.80	32.84	25.21	27.03
AD014	Adult	26.90	31.03	33.66	30.95	27.65	27.13	26.02	25.81	19.13	25.15	--	23.67
AD015	Adult	23.49	23.93	28.56	30.96	28.32	30.22	25.55	28.26	30.33	30.86	24.97	29.72
AD016	Adult	24.75	26.82	28.19	29.16	19.69	28.09	26.56	23.55	22.17	25.36	8.05	26.84
AD017	Adult	24.44	25.73	29.46	33.62	29.82	32.86	24.57	26.65	27.26	30.84	23.66	24.77

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Mastoid 2 N						Forehead 5.4 N					
	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10 000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10 000 Hz
AD001	19.86	21.63	23.73	26.33	13.16	30.55	26.49	27.29	32.83	38.28	39.12	42.33
AD002	24.81	30.58	31.72	36.72	32.64	34.30	25.12	29.40	32.14	39.79	41.34	42.97
AD003	16.10	19.66	21.46	28.51	29.78	32.61	23.48	30.82	33.30	37.22	41.44	43.88
AD004	26.16	25.49	30.06	33.89	30.08	31.50	25.12	28.59	31.01	38.01	37.56	41.64
AD005	27.31	29.74	29.07	29.11	20.15	27.24	26.99	29.44	33.83	36.77	38.99	41.26
AD006	25.85	28.01	31.76	36.17	38.96	39.03	24.25	28.48	32.37	35.96	40.97	42.90
AD007	28.46	32.16	20.40	35.26	31.95	36.38	28.46	32.16	20.40	35.26	31.95	36.38
AD008	24.80	26.86	27.15	32.79	28.76	32.62	25.95	26.46	33.17	38.33	40.73	43.51
AD009	23.32	21.25	24.73	31.12	25.83	29.99	24.93	28.14	32.69	37.31	40.21	44.23
AD010	--	--	--	--	--	--	--	--	--	--	--	--
AD011	28.37	31.19	31.53	36.54	29.57	30.91	26.86	29.55	33.85	37.22	41.46	44.47
AD012	21.98	21.97	24.56	32.82	15.63	28.93	23.77	27.22	29.27	36.21	37.52	42.24
AD013	20.84	24.98	27.26	28.84	26.35	29.06	23.96	30.11	33.89	37.18	41.29	43.00
AD014	19.71	17.19	15.43	28.05	--	21.72	27.71	26.91	32.36	37.08	40.27	43.64
AD015	21.02	22.59	25.99	33.16	23.42	28.92	23.24	28.23	32.61	35.91	39.40	43.04
AD016	24.15	24.70	24.05	28.71	23.27	24.15	24.03	30.14	33.18	38.02	41.12	44.58
AD017	26.00	25.08	23.71	30.71	22.68	25.44	29.18	28.82	32.60	39.96	42.65	43.39

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for high frequencies

Subject #	Forehead 4 N						Forehead 2 N					
	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10 000 Hz	1995 Hz	3019 Hz	3981 Hz	6025 Hz	7943 Hz	10 000 Hz
AD001	23.28	27.43	32.42	35.40	37.15	36.99	24.33	29.95	34.89	36.95	40.31	43.67
AD002	25.71	29.05	32.87	36.16	39.41	42.81	24.82	30.41	33.12	40.53	41.12	44.15
AD003	25.12	31.08	34.08	39.17	44.53	43.36	27.58	31.86	33.36	38.22	42.76	44.92
AD004	26.12	30.34	33.81	35.96	38.44	42.54	24.68	29.52	31.95	39.27	38.61	42.29
AD005	25.71	29.82	32.51	38.31	40.49	45.20	25.66	29.96	33.67	37.54	39.47	42.38
AD006	24.58	28.34	33.74	36.98	41.80	44.73	24.40	28.62	34.32	40.00	40.74	44.85
AD007	24.98	30.48	33.89	38.41	40.51	44.91	24.09	30.47	32.68	39.65	40.54	45.64
AD008	23.75	28.79	32.64	37.87	39.68	44.04	24.26	28.40	32.17	37.00	39.36	43.18
AD009	23.97	29.47	32.51	37.56	39.81	43.98	24.83	31.29	33.85	38.00	40.24	44.03
AD010	--	--	--	--	--	--	--	--	--	--	--	--
AD011	25.55	29.85	33.40	35.92	41.75	44.54	26.17	30.74	33.14	38.10	41.33	44.10
AD012	23.04	29.81	34.02	38.81	40.41	43.96	24.10	30.57	33.72	38.13	41.07	43.35
AD013	23.51	28.92	32.72	36.34	39.47	42.46	20.05	24.67	28.07	35.33	37.10	36.41
AD014	24.15	29.17	33.42	36.59	41.10	44.42	24.34	29.43	33.09	38.45	38.77	44.66
AD015	23.60	28.11	31.75	36.50	39.79	43.42	24.18	29.50	31.49	35.62	39.15	43.10
AD016	24.15	28.14	30.75	36.72	39.05	41.64	25.47	30.21	33.83	37.09	41.24	43.22
AD017	26.07	28.17	31.38	37.21	41.01	42.14	24.73	30.15	33.65	38.98	39.49	43.53

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

APPENDIX E: Individual Measures of Impedance Magnitude for Resonant Frequency

Individual measures of impedance magnitude (dB re 1 Ns/m) for resonant frequency

Subject #	Age Group	Mastoid 5.4 N	Mastoid 4 N	Mastoid 2 N	Forehead 5.4 N	Forehead 4 N	Forehead 2 N
CH185	A	24.31	19.02	22.66	14.43	17.60	16.79
CH139	A	17.54	21.73	19.68	17.23	18.00	15.32
CH123	A	20.36	18.20	17.22	11.78	17.12	13.41
CH162	A	15.61	14.78	18.04	15.64	12.03	14.99
CH161	A	18.70	16.61	17.06	19.43	16.96	14.34
CH130	A	18.10	23.76	21.89	15.67	14.85	15.90
CH187	A	19.01	20.51	22.35	18.47	16.92	--
CH177	A	23.00	17.68	15.82	15.48	17.64	11.93
CH149	A	21.22	24.44	19.54	18.61	17.68	16.67
CH197	A	17.87	21.63	20.07	17.05	18.14	18.48
CH141	A	18.00	17.15	14.64	17.84	18.94	15.47
CH175	A	21.79	18.33	19.85	13.35	13.24	8.17
CH189	A	24.82	23.19	21.11	--	--	--
CH146	A	20.91	20.81	20.18	18.66	16.05	11.20
CH160	A	22.27	17.67	13.13	15.73	16.60	14.10
CH109	A	--	22.44	22.82	17.54	14.37	17.71
CH131	A	--	--	--	--	--	--
CH173	A	--	--	--	--	--	--
CH134	B	16.70	15.80	15.30	18.41	18.50	14.96
CH122	B	23.19	12.44	16.83	21.66	19.94	19.20
CH165	B	21.18	22.99	21.74	20.74	18.24	15.42
CH179	B	16.30	21.38	14.09	19.40	14.05	16.43

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for resonant frequency

Subject #	Age Group	Mastoid 5.4 N	Mastoid 4 N	Mastoid 2 N	Forehead 5.4 N	Forehead 4 N	Forehead 2 N
CH114	B	22.51	19.30	14.68	18.87	18.89	20.16
CH196	B	22.57	15.42	17.72	15.03	16.52	13.75
CH135	B	23.61	19.02	18.71	19.53	18.30	18.02
CH115	B	17.94	22.05	20.35	13.86	16.33	17.43
CH154	B	24.50	18.43	17.85	20.25	14.54	13.79
CH157	B	22.88	24.25	18.68	21.45	14.55	17.54
CH136	B	17.98	15.51	15.84	20.13	20.21	16.58
CH152	B	21.25	20.05	19.17	17.15	12.08	15.69
CH124	B	17.64	20.48	21.31	19.71	18.09	12.83
CH117	C	21.20	18.29	18.79	19.59	20.33	13.11
CH129	C	19.72	18.82	17.90	21.21	21.20	14.82
CH159	C	22.15	18.89	14.00	19.60	20.10	16.52
CH138	C	19.77	19.09	14.97	21.44	21.22	20.00
CH156	C	22.71	19.50	21.29	23.34	22.77	21.51
CH168	C	21.45	25.70	21.59	22.16	23.54	19.21
CH119	C	22.33	19.96	16.65	21.69	21.36	18.53
CH153	C	23.90	14.30	17.52	24.42	21.20	19.60
CH193	C	25.36	22.55	22.21	20.53	18.10	18.37
CH199	C	21.13	23.71	--	19.67	17.20	17.57
CH167	C	20.93	23.06	10.50	20.55	20.94	18.82
CH143	C	26.15	17.73	21.20	23.01	21.86	21.75
CH147	C	23.84	18.98	22.06	18.98	18.27	21.78

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for resonant frequency

Subject #	Age Group	Mastoid 5.4 N	Mastoid 4 N	Mastoid 2 N	Forehead 5.4 N	Forehead 4 N	Forehead 2 N
CH172	C	25.11	23.10	18.10	24.48	27.91	21.40
CH116	C	24.73	24.62	22.03	23.20	20.19	18.19
CH126	D	22.11	23.06	22.19	18.32	21.26	18.44
CH191	D	21.55	19.40	9.86	24.30	23.25	24.86
CH188	D	19.73	16.44	16.07	20.29	21.36	20.28
CH184	D	25.92	22.83	25.69	23.77	22.10	20.27
CH186	D	22.85	22.94	19.84	21.08	20.36	18.14
CH144	D	22.18	11.59	20.37	21.50	20.30	19.46
CH107	D	25.29	21.82	19.56	20.99	22.16	17.47
CH140	D	18.36	21.86	13.90	22.98	22.47	20.00
CH174	D	22.14	16.21	18.21	21.10	21.95	18.04
CH101	D	17.86	20.43	16.46	22.13	15.51	20.79
CH112	D	21.86	23.36	17.37	22.67	24.10	17.80
CH145	D	24.72	25.10	14.46	20.13	19.21	18.82
CH113	D	14.12	22.27	18.84	23.19	21.03	22.53
CH169	D	22.77	19.36	22.11	17.40	18.81	18.58
CH118	D	22.88	18.43	22.18	23.87	18.50	20.96
CH195	D	22.87	20.64	19.23	23.38	21.23	19.49
CH102	D	17.84	18.16	13.66	22.68	19.55	18.64
CH125	D	15.10	13.38	20.15	20.88	17.08	13.99
CH133	D	25.79	19.42	20.00	23.09	22.34	23.94

A “--“ indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual measures of impedance magnitude (dB re 1 Ns/m) for resonant frequency

Subject #	Age Group	Mastoid 5.4 N	Mastoid 4 N	Mastoid 2 N	Forehead 5.4 N	Forehead 4 N	Forehead 2 N
AD001	Adult	20.54	19.49	18.99	24.94	23.17	19.91
AD002	Adult	26.13	--	19.24	24.84	24.89	21.78
AD003	Adult	23.07	19.64	15.46	22.77	20.86	17.62
AD004	Adult	23.44	19.02	25.49	24.89	25.41	23.68
AD005	Adult	22.04	19.32	19.50	26.73	25.57	21.49
AD006	Adult	23.66	26.98	21.09	23.88	24.10	23.71
AD007	Adult	20.95	18.35	20.40	20.50	17.82	19.96
AD008	Adult	25.54	23.18	24.24	24.88	23.66	23.36
AD009	Adult	24.37	20.15	21.25	23.85	20.37	21.71
AD010	Adult	--	--	--	--	--	--
AD011	Adult	22.75	21.59	19.79	26.46	25.54	21.86
AD012	Adult	25.86	23.47	20.71	23.05	20.80	22.84
AD013	Adult	20.77	22.29	16.90	22.33	20.49	19.57
AD014	Adult	21.47	23.19	17.58	23.68	23.81	23.44
AD015	Adult	22.00	25.02	20.10	22.30	23.13	23.69
AD016	Adult	22.36	20.48	20.93	21.67	21.37	20.87
AD017	Adult	23.93	19.55	24.23	26.04	25.58	23.69

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

APPENDIX F: Individual Resonant Frequency Measures

Individual measures of impedance magnitude (dB re 1 Ns/m) for resonant frequency

Subject #	**Age Group**	**Mastoid 5.4 N**	**Mastoid 4 N**	**Mastoid 2 N**	**Forehead 5.4 N**	**Forehead 4 N**	**Forehead 2 N**
CH185	A	24.31	19.02	22.66	14.43	17.60	16.79
CH139	A	17.54	21.73	19.68	17.23	18.00	15.32
CH123	A	20.36	18.20	17.22	11.78	17.12	13.41
CH162	A	15.61	14.78	18.04	15.64	12.03	14.99
CH161	A	18.70	16.61	17.06	19.43	16.96	14.34
CH130	A	18.10	23.76	21.89	15.67	14.85	15.90
CH187	A	19.01	20.51	22.35	18.47	16.92	--
CH177	A	23.00	17.68	15.82	15.48	17.64	11.93
CH149	A	21.22	24.44	19.54	18.61	17.68	16.67
CH197	A	17.87	21.63	20.07	17.05	18.14	18.48
CH141	A	18.00	17.15	14.64	17.84	18.94	15.47
CH175	A	21.79	18.33	19.85	13.35	13.24	8.17
CH189	A	24.82	23.19	21.11	--	--	--
CH146	A	20.91	20.81	20.18	18.66	16.05	11.20
CH160	A	22.27	17.67	13.13	15.73	16.60	14.10
CH109	A	--	22.44	22.82	17.54	14.37	17.71
CH131	A	--	--	--	--	--	--
CH173	A	--	--	--	--	--	--
CH134	B	16.70	15.80	15.30	18.41	18.50	14.96
CH122	B	23.19	12.44	16.83	21.66	19.94	19.20
CH165	B	21.18	22.99	21.74	20.74	18.24	15.42
CH179	B	16.30	21.38	14.09	19.40	14.05	16.43

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual resonant frequency measures (Hz)

Subject #	Age Group	Mastoid 5.4 N	Mastoid 4 N	Mastoid 2 N	Forehead 5.4 N	Forehead 4 N	Forehead 2 N
CH114	B	1862	1660	1585	1445	1479	977
CH196	B	1380	1585	2042	1479	1288	1175
CH135	B	2042	1660	1622	1175	1738	955
CH115	B	3236	2512	2188	1479	1445	1259
CH154	B	3236	2399	2818	2042	1380	1259
CH157	B	1995	2344	2951	1778	1318	1380
CH136	B	3890	3715	3020	1698	1549	1622
CH152	B	1778	1820	1738	1660	1000	977
CH124	B	1820	1413	1549	1660	1122	912
CH117	C	3020	2570	2692	1778	1413	1413
CH129	C	1950	2291	1479	1549	1660	1023
CH159	C	2455	2570	1820	1820	1413	1230
CH138	C	2884	3311	2570	1380	1479	1445
CH156	C	1738	2042	1862	1820	1479	1585
CH168	C	3802	2399	2818	2188	1905	1514
CH119	C	2239	2455	3162	1995	1905	1445
CH153	C	2399	3162	1905	1698	1820	1622
CH193	C	1096	1096	1514	1380	1230	1259
CH199	C	2951	1778	--	1413	1380	1288
CH167	C	2951	3020	2089	1862	1259	1202
CH143	C	3311	2399	1622	1905	1622	1023
CH147	C	1862	2291	1738	1549	1905	977

A “--“ indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual resonant frequency measures (Hz)

Subject #	Age Group	Mastoid 5.4 N	Mastoid 4 N	Mastoid 2 N	Forehead 5.4 N	Forehead 4 N	Forehead 2 N
CH172	C	3802	2754	2630	2630	2692	2291
CH116	C	1995	2089	1698	1230	1096	955
CH126	D	1820	1820	1778	1905	1905	1349
CH191	D	2951	3162	2951	2291	1514	1622
CH188	D	3715	3802	2399	1778	1820	1349
CH184	D	2570	1905	2630	1995	2344	1660
CH186	D	2512	1738	1698	1549	1230	1023
CH144	D	2138	3890	1698	2042	1175	1514
CH107	D	2399	1820	2291	1995	1549	1514
CH140	D	2344	1380	2291	2188	1995	1514
CH174	D	3631	3162	2884	2239	2089	1738
CH101	D	3311	1820	3090	1778	1778	1820
CH112	D	2188	1738	2754	2188	1905	1259
CH145	D	2818	1820	3388	1259	1413	1349
CH113	D	1995	2188	1995	2138	1622	1950
CH169	D	1862	1660	1660	1738	1230	1585
CH118	D	1585	832	1072	2188	1660	1148
CH195	D	3162	3388	1549	2291	2512	1995
CH102	D	3715	3236	3020	2138	2455	2042
CH125	D	3890	3020	2138	1479	1413	1380
CH133	D	2455	3467	3467	2291	1698	1905

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.

Individual resonant frequency measures (Hz)

Subject #	Age Group	Mastoid 5.4 N	Mastoid 4 N	Mastoid 2 N	Forehead 5.4 N	Forehead 4 N	Forehead 2 N
AD001	Adult	2692	2754	2692	2291	2291	1660
AD002	Adult	2138	--	1622	1862	2188	1445
AD003	Adult	2138	1318	2239	1862	1820	1413
AD004	Adult	2399	2951	3020	1622	1820	1950
AD005	Adult	1622	1445	1288	2042	1862	1230
AD006	Adult	2344	1995	1950	2089	2188	1905
AD007	Adult	3548	3090	3981	1738	1585	1698
AD008	Adult	1862	3310	3631	2188	1905	1778
AD009	Adult	2692	3981	3020	2239	1549	1778
AD010	Adult	--	--	--	--	--	--
AD011	Adult	2042	2344	1380	2188	1995	1738
AD012	Adult	1905	1479	2344	2188	1660	1820
AD013	Adult	1862	1445	1349	2042	1698	1905
AD014	Adult	1622	1445	1380	2512	1905	1738
AD015	Adult	2399	1380	2089	2089	2188	1905
AD016	Adult	1380	851	1023	1514	1479	1622
AD017	Adult	2138	1288	2399	2512	2188	1318

A "--" indicates data for this subject were not available. An italicized subject number indicates data for that subject were not included in the analysis. The inclusion/exclusion details are provided in Appendix A.