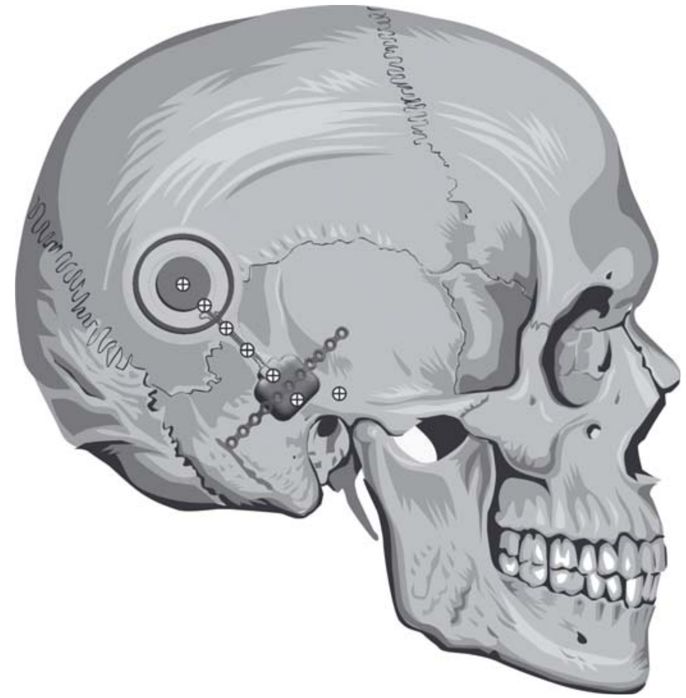
Transmission of bone-conducted sound in the human skull based on vibration and perceptual measures



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ABSTRACT

For patients who are rehabilitated with bone conduction (BC) hearing aids, the position on the skull of the hearing aid is critical for the perception of the sound. The aim of this work was to describe the vibration of the cochlea from BC sound stimulation at different positions on the skull. The relevance of the vibration velocity of the cochlea as a perceptual measure was also investigated.

In human cadavers vibration stimulation was applied at eight positions on each side of the skull with a frequency range of 0.1-10 kHz. The resulting velocity of the cochlear vibration was measured by a laser Doppler vibrometer from both ipsilateral and contralateral stimulation. A prototype of a novel bone conduction implant (BCI), positioned approximately 5 mm behind the ear canal, was tested with the same methodology. In live human subjects vibration stimulation was applied at four positions on the head. The resulting vibration velocity of the otic capsule was measured with a laser Doppler vibrometer. Bone conducted hearing thresholds in the same subjects were compared to the otic capsule vibration results.

With vibration stimulation on the ipsilateral side there was an increased magnitude response of the cochlear vibration with shorter distance between the stimulation position and the cochlea. When the bone conducted stimulation was on the contralateral side the change in magnitude of the cochlear vibration between positions was limited. BC stimulation at a position close to the ipsilateral cochlea increased the response magnitude difference between the cochleae. The results of stimulating with a BCI and a transducer were similar. The influence of the squamosal suture on BC sound transmission was not clear but indications of a small damping effect were found. With simultaneous bilateral stimulation at the low frequencies correlated signals were added constructively or destructively while non-correlated signals gave a 3 dB sound energy increase. Time separation between ipsilateral and contralateral stimulation was found to be largest at positions close to the cochlea. The velocity response at the otic capsule from BC stimulation was similar between human cadavers and live humans. In live humans the correlation between vibration of the otic capsule and hearing perception was low at the individual level, while median data showed similar trends between the two methods.

When BC sound stimulation is applied at a smaller distance between the stimulation position and the cochlea, sound transmission improves to the ipsilateral cochlea and is decreased to the contralateral cochlea. Measures of the vibration of the otic capsule from BC sound stimulation as an estimation of BC hearing perception was investigated and the results indicate that the method is valid. A patient with a hearing loss where there is an indication for BC hearing aids

can likely benefit from increased ipsilateral stimulation, and also an improved binaural hearing from bilateral stimulation, when the hearing aid is applied close to the cochlea. The BCI is a realistic alternative to other BC hearing aids.

LIST OF PAPERS

This thesis contains the following papers, which can be referred to in the text by their Roman numerals, and as number in the reference list. Paper I: 122; paper II: 66; paper III: 68.

I Eeg-Olofsson M, Stenfelt S, Tjellström A, Granström G. Transmission of bone-conducted sound in the human skull measured by cochlear vibrations. International Journal of Audiology 2008;47:761-769. Copyright © 2008 British Society of Audiology, International Society of Audiology, and Nordic Audiological Society.

II Eeg-Olofsson M, Stenfelt S, Granström G. Implications for contralateral bone-conducted transmission as measured by cochlear vibrations. Otology & Neurotology 2011;32:192-198. Copyright © 2011 Otology & Neurotology, Inc.

III Håkansson B, Reinfeldt S, Eeg-Olofsson M, Östli P, Taghavi H, Adler J, Gabrielsson J, Stenfelt S, Granström G. A novel bone conduction implant (BCI): Engineering aspects and pre-clinical studies. International Journal of Audiology 2010;49:203-215. Copyright © 2010 British Society of Audiology, International Society of Audiology, and Nordic Audiological Society.

IV Eeg-Olofsson M, Stenfelt S, Taghavi H, Reinfeldt S, Håkansson B, Finizia C. Transmission of bone conducted sound – correlation between hearing perception and cochlear vibration. Manuscript.

Papers I-III are printed with permission from the publishers.

LIST OF ABBREVIATIONS

ABR	Auditory brainstem response
AC	Air conduction
BAHA	Bone anchored hearing aid
BC	Bone conduction
BCI	Bone conduction implant
BEST	Balanced electromagnetic separation transducer
BMLD	Binaural masking level difference
C-BEST	Capsuled Balanced electromagnetic separation transducer
CSF	Cerebrospinal fluid
dB	Decibel
dbc	direct bone conduction
F	Force
HL	Hearing level
Hz	Hertz
ISQ	Implant stability quotient
LDV	Laser Doppler vibrometer
LSCC	Lateral semicircular canal
MAPP	Mastoid surface area that attaches to the petrous part of the temporal bone
PBCD	Percutaneous bone conduction device
RFA	Resonance frequency analysis
rms	root mean square
SNR	Signal-to-noise ratio
SPL	Sound pressure level
SSD	Single sided deafness
v	Velocity
Z	Impedance

CONTENTS

Preface

This thesis is based on measures of cochlear vibration as a response to bone conducted sound stimulation on the human skull. Results from a thorough investigation of the cochlear velocity response from bone conducted stimulation at eight positions on both sides of the human skull, are described. Even if the correlation between vibration of the cochlea and hearing perception is supported from other studies, no direct comparison has previously been made on the same subjects. Results from such a correlation are presented in this thesis, as well as a novel bone conduction implant. The overall goal of the work presented in this thesis is to contribute to the understanding of bone conduction sound physiology of the human skull, and that the increased knowledge can provide improvements in hearing rehabilitation for patients with hearing losses.

Introduction

Bone conduction physiology

Trying to understand the concept of bone conduction (BC) physiology has occupied many researchers during the 20th century. Von Békésy demonstrated by his famous cancellation experiment that the basilar membrane of the cochlea was stimulated in the same way by air conducted (AC) sound and bone conducted (BC) sound (1). This finding has been repeated and extended by others (2-5). It was also concluded that the direction of the travelling wave of the basilar membrane always goes from the base to the apex (6, 7). Other similarities are that two tone distortion products can be generated from both AC tones and BC tones (8). By measuring Auditory Brainstem Response (ABR) from BC stimulation and the Jewett wave V, Schratsenstaller *et al.* (9) confirmed von Békésy´s (7) theory that the basilar membrane travelling wave always goes from the stiffer part in the base of the cochlea to the apex. He discovered that the latency of Jewett wave V was delayed more for BC stimulation than for AC stimulation when the level was decreased. There are also other dissimilarities reported. BC evoked oto-acoustic emissions have a different level response than AC evoked oto-acoustic emissions (10) do. Low frequency loudness growth is different for BC sound compared to AC sound (11). Another difference is that ultrasonic sound up to 100 kHz can be heard by BC (12). No generally agreed explanation for this phenomenon exists.

Linearity of bone conducted sound of the human skull

Linearity of the vibration transmission of the human skull is the foundation of our understanding of BC sound propagation. The fact that sound transmission of the skull is linear has been described in detail (3, 13-17). Some authors have described a non-linear system (2, 18) and the reason has been explained by the chosen methodology. It is now generally accepted that BC sound transmission in the human skull is linear, at least for frequencies between 0.1-10 kHz and up to 77 dB HL (14).

Mechanical point impedance

The dynamic mechanical properties of the human skull have been of interest in many fields of science. In this thesis the main focus is on a BC sound perspective and hearing perception. Another area of interest is skull trauma. Especially resonance frequencies of the human skull were investigated to better understand how a blow against the head would affect the appearance of fractures (16, 17, 19). An additional purpose was to construct better head protection devices like helmets. Many authors have investigated the dynamic properties of the human skull by measuring the magnitude and phase of the mechanical point impedance, sometimes in combination with the transfer function of BC sound (2, 13, 15, 20-22).

Mechanical impedance (*Z*) is the "structure's resistance to vibration velocity when an excitation force is applied" (15) and it is defined as the quotient between excitation force (F) and response velocity (v), $Z=F/v$. When measuring mechanical point impedance an impedance head can be used. A transducer applies the vibrating force and is attached to the impedance head. The impedance head measures the force applied simultaneously with the acceleration. In the post processing the acceleration is transformed to velocity. From the impedance Z both the magnitude and the phase angle can be obtained. When applying a force against a body (this body can be anything with a certain mass, not necessarily a human body) it can be completely rigid at the point of measure, or it can flex. For the human skull in low frequencies below 100-150 Hz there is no flexion of the skull bone when stimulated by a BC hearing aid transducer. Instead the head is moving as a whole. We call this a rigid body motion and the following can be stated:

1. With increasing frequency the mechanical point impedance magnitude will increase by approximately 6 dB/octave.
2. The force will lead the velocity by 90° ($\pi/2$ radians).

Eventually with increasing frequency the skull surface will start to flex for the force applied on the whole skull mass. The stimulation point will with increasing frequency gradually be decoupled from the head mass. When it is decoupled we call this a stiffness controlled movement. The skull surface now acts as a spring:

1. With increasing frequency the impedance decreases by 6 dB/octave.
2. The force lags the velocity by 90° ($-\pi/2$ radians).

Mechanical impedance can provide information about rigid body motion or stiffness controlled motion. Conclusions can be drawn about resonance frequencies from the interaction between magnitude and phase curves. It does not reveal information about how motion in the stimulation point affects other points, for example the cochlea. It seems reasonable that a different stiffness at an excitation point on the skull can result in alteration of the sound transmission to the cochlea. Parts of the above paragraph can be found in Haughton (23).

Skull resonances

It is important to differ between forced and free resonances. A forced resonance occurs due to an external influence from for example a vibrating transducer on the skull and is caused by the interaction between the transducer and the skull (24). A forced resonance does not oscillate after excitation. You can have both forced anti-resonances and forced resonances. A free resonance depends on the structural properties of the skull and you can see a free oscillation after excitation. A free resonance can be either a resonance or an anti-resonance. There have been numerous of reports on skull resonances. In the early research only a limited range of frequencies was investigated. The vast majority have been made on dry skulls or cadaver heads. The study of resonance frequencies on live subjects before percutaneous titanium implants were available was a problem due to the skin and soft tissues. Von Békesy found the first resonance frequency on a living person at 1800 Hz (1). Franke (20), using mechanical impedance measurements, reported the first resonance frequency in a dry skull to be 820 Hz, which was lowered to 500 Hz when filled with gelatine. The same author also reported the two first resonances of a cadaver head to be 600 Hz and 900 Hz. Franke did not manage to obtain skull resonances from a live subject. Gurdjian *et al.* (19) reported the first skull resonance to be 880 Hz in a dry skull using mechanical impedance measurements. He obtained a somewhat lower resonance frequency when the dry skull was filled with gelatine. Håkansson *et al.* (24) measured the acceleration frequency response on six subjects with bilateral titanium implants. By doing so it was possible to study the skull resonances on live subjects with direct contact to the skull with the skin and soft tissues in place. Up to 19 resonance frequencies were found with a large individual variation between subjects. The resonances were damped and were likely to have a limited impact on hearing perception. On the other hand sharp anti-resonances were found. These anti-resonances were estimated to have a possible impact on hearing perception.

Sound waves

General facts about sound waves can be found in Haughton (23). Different kinds of sound waves can exist in sound propagation in the human skull. When we refer to sound waves we think of longitudinal sound waves in air, or in other media such as fluids. These are also called compressional waves where the particles move back and forth, in line with the direction of the wave propagation. The second type of wave is the transversal wave, or shear wave, where the particles move in a perpendicular direction to the wave propagation. The term "shear" means a change in shape but not a change in volume. The third type of wave is the bending wave, or flexural wave. Bending waves exist in plates or beams. To confuse the reader a bit more a plate wave (which is not a bending wave in a plate) is a mixture of longitudinal and transversal waves, also called a quasilongitudinal wave (25). By a vibration stimulus on the human skull all of these wave types can propagate and eventually lead to vibration of the basilar membrane. Bending waves are dispersive which means that their velocity changes with frequency. Longitudinal and transversal waves can have different velocities in the same medium and at the same frequency. Plate waves are also in general dispersive (21). This means that when we stimulate with vibrations on the human skull at a certain frequency, there is a possibility that all of the above mentioned types of sound waves propagate along the human skull at different velocities, and affect the cochlea at different latencies from the time of stimulation. Therefore, when calculating the time between stimulus onset at one position and the response at another position, a group delay function must be used. The group delay estimates time from excitation to response of the information transmission, from the phase function of the sound transmission. The group velocity estimates are derived from the group delay with the addition of the distance between the two positions. A sharp anti-resonance appears at low frequency vibration stimulation of the human skull. An anti-resonance means that there is very limited movement at the response position for a certain stimulation frequency. The group delay is only defined for smooth phase responses and does not exist in the presence of resonances or anti-resonances. Therefore group delay calculation of the human skull (see paper II) was only considered above a stimulation frequency of 1 kHz. Certainly there are anti-resonances above this frequency region but the frequency where they occur differs between subjects. Due to averaging of the results from several subjects the influence of high frequency anti-resonances are reduced enabling the calculation of group delay.

Velocity of bone conducted sound of the human skull

Since there is still uncertainty about what kind of wave propagation composing BC sound of the human skull, both group velocity and phase velocity have been presented in the literature. Primarily live subjects have been used (1, 20, 25, 26) but also dry skulls and cadavers (21, 27). Different methods have been applied including tone cancellation (25, 26), recording with pickups (1, 20) and frequency response using an accelerometer or a laser Doppler vibrometer (LDV) (21). Von Bekesy (1) reported the phase velocity to be 540 m/s and Zwislocki (26) 260 m/s. Both of them came to the conclusion that the velocity was frequency independent. Franke (20) reported the group velocity to be 80-300 m/s where the highest velocity increase was just above 500 Hz. At 1500 Hz the velocity was fairly constant. Tonndorf *et al.* (25) reported the phase velocity to be 55-330 m/s where the higher velocities were found at frequencies above 2 kHz. Tonndorf suggested that transversal waves were predominantly present in the low frequency range and plate waves in the higher frequency range. Wigand *et al.* (27) and Stenfelt *et al.* (21) reported that the velocity was frequency dependent. Further, both Wigand *et al.* and Stenfelt *et al.* investigated if the position of stimulation influenced the velocity of BC sound of the human skull. They both found that the velocity was higher at the skull base and somewhat lower at the cranial vault. Wigand *et al.* reported on the velocity of the skull base to be 3000 m/s and of the cranial vault 420 m/s. Stenfelt *et al.* described the phase velocity of the skull base to be 400 m/s and primarily frequency independent, while the phase velocity of the cranial vault had a dispersive wave motion where the phase velocity ranged from 250 m/s to 300 m/s. Even lower values of 100 m/s were found.

Pathways of bone conducted sound of the human skull

The puzzle of BC hearing has not yet been solved. One often used reference in the subject of BC pathways is Tonndorf 1966 (28). He describes seven modes of BC stimulation of the basilar membrane in cats:

1. Ossicular inertia
2. Middle ear cavity compliance
3. Pure compressional effect
4. Oval window release (mobility of the oval window)
5. Round window release (mobility of the round window)
6. Inner ear fluid inertia
7. Cochlear aqueduct effect

Listed below are the pathways which today are regarded as the most important for BC hearing (29, 30).

The outer ear component - the occlusion effect

BC stimulation of the human skull leads to a motion of the bony part and the cartilage part of the ear canal. This motion produces a sound pressure in the ear canal. The sound pressure acts on the tympanic membrane and the vibration is transferred further to the middle ear ossicles and to the inner ear. When the ear canal is occluded the ear canal sound pressure increases enough to affect BC hearing (31). Tonndorf (28) and Huizing (32) have described the theory explaining the occlusion effect. Tonndorfs' theory is that the open ear canal functions as a high pass filter. When the ear canal is occluded the filter effect is removed and thus low frequency sounds will dominate (28). Huizing explained the occlusion effect by altered resonance of the ear canal when occluded (32). According to Stenfelt *et al.* (30) both are right, but Tonndorf regarding low frequencies, and Huizing regarding higher frequencies. There have been theories that the occlusion effect is due to surrounding masking noise being eliminated from the ear canal. This is wrong since the occlusion effect is the same in both noisy surroundings as well as in a sound isolated room (32). During occlusion the ear canal sound pressure is generally decreasing above 2 kHz where other pathways dominate (31). The exact increase and frequency range of the occlusion effect is dependent on the type (33), and how the ear canal is occluded, provided that the tympanic membrane and the middle ear are normal. The occlusion effect can be eliminated by inserting a plug into the bony part of the ear canal (28, 31). The role of the mandible in contributing to a raised sound pressure level in the ear canal when occluded has been investigated (1, 20, 28, 31, 34, 35). Even if the transmission of BC sound via the mandibular joint is possible, the sound pressure level of the occluded ear canal did not change whether the joint was present or not (31, 34).

Middle ear ossicles inertia

The middle ear ossicles resonate at around 1.5 kHz (36). From approximately 1.5 kHz to 3.1 kHz the ossicular inertia pathway is dominating, or has at least the same influence on BC hearing as the inner ear fluid inertia (37). In case of absence or obstruction of the oval window, the BC thresholds are hardly affected (38, 39). In the case of otosclerosis (40) when the stapes is fixed the middle ear ossicles contribution decreases around 1.5 kHz, hence the BC thresholds around this frequency are elevated (36). When the middle ear is affected from chronic ear disease or effusion the BC thresholds are generally elevated (41). The BC thresholds with stimulation at the mastoid are more sensitive to lesions of the middle ear than the forehead position, especially if the stapes is still mobile. In the case of immobile stapes both forehead stimulation and mastoid stimulation are affected in a similar way (42-44). It was early discovered that the ossicular chain had influence on BC hearing. By adding small masses to the tympanic membrane or the ossicles the resonance frequency was lowered and contributed to improved BC thresholds in the low frequency range (1, 28, 32, 45, 46). If the stapes is glued with no motion relative to the temporal bone, no contribution from the ossicles inertia exists. If the incudo-stapedial joint is severed the stapes contributes to BC hearing but to a lesser extent than a normal middle ear. If the malleus is fixed to the temporal bone, the stapes moves with higher velocity in the high frequency range, but contributes less to BC hearing (36).

Inner ear fluid inertia

This pathway is regarded to be the most important pathway in BC hearing from low frequencies up to 4 kHz. The prerequisite for inner ear fluid inertia to stimulate the basilar membrane is a fluid flow between the vestibular side and the tympanic side of the cochlea (30). A pressure gradient between the scala vestibuli and the scala tympani produces a fluid flow and a travelling wave on the basilar membrane. To achieve the fluid flow there must be one compliant structure on each side of the basilar membrane, the oval window and the round window. Even if one of the windows is obstructed BC hearing is only marginally affected (38-40, 47-50). If the pressure of the cerebrospinal fluid (CSF) is increased, the same pressure increase has been measured in both the endolymphatic and perilymphatic space (51, 52). The cochlear aqueduct is believed to be the main connection for the perilymphatic space, and the endolymphatic sac and duct for the endolymphatic space (52). These connections can function as a “third window” that allows for fluid flow despite obstructions of either the oval window or the round window.

Compression and expansion of the cochlea

By compressing the cochlear walls the oval window and the round window bulge simultaneously. The compliance of the round window is estimated to be 20 times as compliant as the oval window (53). The volume of the scala vestibuli is larger than the volume of the scala tympani (ratio 5:3) and the area ratio is approximately 3:2 (28). When the cochlea is compressed more fluid is moved from the scala vestibuli to the scala tympani. Stenfelt *et al.* (54) has measured the fluid displacement in temporal bones at the oval window and the round window. For

BC the fluid displacement was 5-15 dB greater at the round window below 2 kHz. Above 2 kHz the fluid displacement was greater at the oval window. The impedance of the third window connections of the inner ear is larger for fluid flow at high frequencies (47). The theory of cochlea compression is based on threshold measurements with lesions of the middle ear. For an obstruction of the oval window as in otosclerosis it has been mentioned before that the BC hearing is only marginally affected. If compression of the cochlea would be responsible for the fluid flow the thresholds would be lower since fluid is forced to the tympanic side of the basilar membrane. Such an improvement of the BC hearing threshold is not seen. On the other hand if inner ear fluid inertia was the main pathway above 4 kHz a travelling wave of the basilar membrane would, according to the theories above, be obstructed by the increased impedances of the "third window". When doing a fenestration operation the thresholds are restored or even slightly improved (40). According to the compression theory this operation would deteriorate the thresholds by an outflow possibility on the vestibular side. Since the thresholds are not improved above 4 kHz, an assumption is that the compression theory of basilar membrane stimulation could be valid above 4 kHz (30). Further, if the limit to achieve an effective excitation of the cochlea from compression response is set to a wavelength less than 10 times the size of the cochlea (cochlea diameter approximately 10 mm), the compression pathway would be possible above 4 kHz (29).

Cerebrospinal fluid pathway

The CSF pathway has been extensively investigated by Sohmer, Freeman and colleagues (55-57). There is an apparent possibility of stimulating the cochlea with stimulation through the CSF but the importance of this pathway for BC hearing is still unknown.

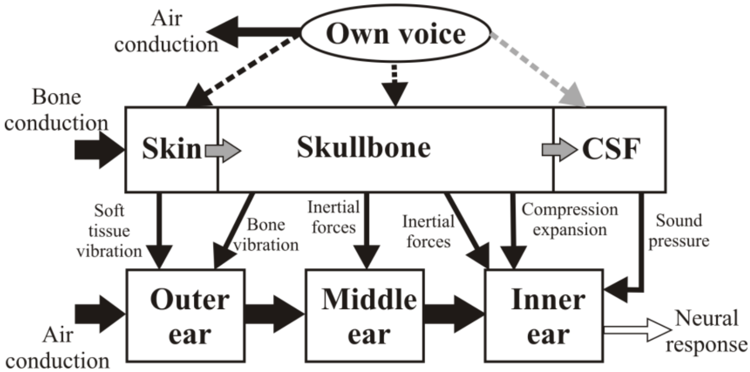


Figure 1. A model illustrating pathways for hearing BC sounds. Reprinted from Stenfelt S, 2011 (29) with permission from S. Karger AG, Basel.

Different positions of bone conducted stimulation on the human skull

BC hearing sensitivity has been studied from multiple locations of the skull. One of the purposes was to investigate the threshold difference between the mastoid and the forehead BC stimulation. Another purpose was to decide the most suitable placement for BC threshold testing (42-44, 58, 59). McBride *et al.* (60) tested 11 stimulation positions of the head for BC hearing threshold and found the zygomatic process to be the most sensitive position, followed by the mastoid, the vertex and the temple. Ito *et al.* (61) measured the acceleration of the front teeth from BC stimulation of the ipsilateral and contralateral mastoid and temporal region, and also from vibration stimulation of the eye. Ito concluded that BC thresholds are not directly related to vibration response of the teeth. Stenfelt *et al.* measured the cochlear vibration of the cochlear promontory of cadaver heads using an accelerometer and a LDV (21). One of the findings in the comprehensive study was an increased vibration response on the ipsilateral side and a decreased response on the contralateral side with the stimulation position close to the cochlea. Moreover the mechanical point impedance at the stimulation position, the vibration directions of the cochlea, the transcranial attenuation, the time delay from stimulation to the vibration response of the cochlea and the velocity of BC sound in the human skull was also investigated. In papers I and II the distance 55 mm and closer to the cochlea, as well as the zygomatic process, were investigated by measuring the cochlear vibration from BC stimulation at 8 positions, both on the ipsilateral and the contralateral side.

Transcranial attenuation

The transcranial attenuation (the quotient between ipsilateral and contralateral cochlear response from corresponding stimulation positions) has been of interest in BC physiology, masking in BC audiometry and in hearing rehabilitation with BC hearing aids. The results have been varying, with large standard deviations and a transcranial attenuation of 0-15 dB (62-64). In a recent study by Stenfelt 2012 (65) tone audiometry measurements were done in individuals with unilateral deafness. In this patient group masking errors do not influence the results. It was shown that the transcranial attenuation varied within individuals depending on frequency. Results could differ with as much as 20 dB in the same individual between frequencies. The result between individuals also had a large variance. The median transcranial attenuation at the mastoid stimulation position was 3-5 dB in the low frequency range up to 0.5 kHz, close to 0 dB from 0.5-1.8 kHz, 10 dB from 3-5 kHz and 4 dB at 8 kHz. The transcranial attenuation at the position for BC hearing aids was 2-3 dB less than at the mastoid position. Measurements of transcranial attenuation have been done in dry skulls and cadavers (21, 22, 66-68). The general conclusion from these studies was that the transcranial attenuation increased monotonically with increased frequency. The transcranial attenuation also increased with shorter distance between the stimulation position and the cochlea. At the BC hearing aid position (55 mm behind and slightly above the ear canal) the transcranial attenuation was approximately 0 dB, increasing to 10-20 dB close to the cochlea and at high frequencies.

Cochlear sensitivity to stimulation direction

It is not known today if the cochlea is more sensitive to a specific direction. Stenfelt (21, 22) has in a dry skull and in cadaver heads measured the cochlear vibration response from stimulation at several locations of the head. It was shown that the cochlea moves in all space dimensions. At low frequencies below the first skull resonance the direction of stimulation dominated the vibration response. At higher frequencies the vibration response for the medial-lateral direction (x-direction), approximately in line with the ear canal, was equal to the other vibration directions (y and z). The x-direction, the y-direction (anterior-posterior) and the z-directions (superior-inferior), here named as in Stenfelt *et al.* (21, 22). The x-direction is important for the following reasons: (1) When measuring with a LDV (paper I-IV) the laser beam is approximately in line with the x-direction. (2) BC stimulation on the mastoid or at the BC hearing aid position is also approximately in the x-direction. Listed below are findings supporting that BC hearing perception can be estimated from the vibration response of the x-direction:

1. There is an 8-12 dB difference in hearing perception between forehead versus mastoid stimulation position for BC (ISO 389-3 (1994)). A similar difference is found for the x-direction in the same comparison, especially below 3.0 kHz (21).
2. The transcranial transmission (the quotient between contralateral and ipsilateral response from corresponding stimulation positions) measured as relative BC threshold measurement in live humans, is similar to the transcranial transmission relative cochlear vibration results for the x-direction in cadavers (65, 69).

Binaural hearing in bone conducted sound

The prerequisite for directional hearing and the ability to focus on a certain sound in noisy environments are separate inputs to the two cochleae. For AC stationary sounds (pure tones) directional hearing depends on time differences in low frequencies below 1.0 kHz, and on level differences above 1.0 kHz. The pinna and the head modify the spectra of the sound and thus form spectral cues for localization in the vertical plane (70). For complex sounds such as speech all these cues for localization may be available simultaneously (70). Information about temporal-, intensity-, and spectral cues are analyzed in the auditory nervous system central to the cochlea. The ability to extract binaural cues from bilateral BC sound stimulation has been shown (71-75), although not to the same extent as for AC hearing. In contrast to AC sounds, BC sounds from a stimulation position on the skull reach both cochleae. The information differences from each cochlea to the central auditory system thus are reduced. Jahn *et al.* (76) described a mechanical interference for pure tones at the cochlea to be either destructive or additive. Rowan *et al.* (77) showed that alteration of phase input even at higher frequencies could lead to directional hearing. The explanation given was that the alteration of phase leads to level differences due to constructive or destructive addition at the cochlear level. This interference finding has also been reported by Eeg-Olofsson *et al.* (66).

The Bone Anchored Hearing Aid (BAHA)

The beginning of the BAHA

When Brånemark by serendipity found that titanium had the property of attaching very tight to bone tissue, a new era of osseointegration in the dental-, and the craniofacial area started (78). The osseointegration between titanium and bone was explored by many authors, for example Tjellström, Albrektsson, Brånemark and Linder (79-83). Hallén suggested that a conventional BC hearing aid could be attached to a molar titanium fixture of the maxilla, and the result was improved hearing (78). Bo Håkansson (Chalmers University of Technology, Göteborg, Sweden) is the inventor of the BAHA, and with an interdisciplinary collaboration with Anders Tjellström (Sahlgrenska University Hospital, Göteborg, Sweden) the first patients received the BAHA in 1977. The titanium implant was installed 55 mm behind the ear canal opening. The actual hearing aid was connected to a permanently skin penetrating abutment that was attached to the titanium implant. Vibrations from the hearing aid were transmitted to the skull and the cochlea by direct bone conduction (*dbc*). The concept of *dbc* implied improved sound transmission and sound quality compared to the transcutaneous BC hearing aid. It was also cosmetically more appealing. The first patients were followed and investigated carefully (15, 80, 84-91). Today the number of patients who have been rehabilitated with the BAHA is uncertain but is probably over 80.000 (92). Over the years simplifications have been done to the surgical technique (93, 94) and a new wider implant with a medium rough surface and small-sized threads at the implant neck is also under evaluation (95). The actual hearing aid has also been improved by new generations of the BAHA (93), now available from two hearing companies, Cochlear Bone Anchored Solutions and Oticon Medical.

Indications for the BAHA

The main indications for the BAHA are unilateral or bilateral conductive-, or mixed hearing losses and single sided deafness (SSD) (96). A consensus statement on the BAHA system was published by Snik *et al.* 2005 (97). There are no exact hearing threshold guidelines for when to choose the BAHA and when to choose other hearing aids. In addition to hearing test results an important guidance is the patients' experience from wearing the BAHA on a head band for a period of time. The information obtained from hearing tests and wearing a head band, together with a holistic and open view of the patients' needs and wishes, often leads to a well-founded choice of hearing rehabilitation. A well-functioning teamwork between different professions, such as ENT-surgeons, audiological physicians and audiologists, is also important for a successful hearing rehabilitation.

BAHA complications

Although the BAHA is a success there has been continuous reporting of complications. One weak spot for the BAHA is the titanium implant which can lose osseointegration, or be lost due to trauma. Further the percutaneous solution exposes the soft tissues around the implant to surrounding dirt and bacteria which can lead to skin irritations and infections. Complaints are sometimes raised from numbness of the skin around the surgical site and the lack of hair around the abutment. A comprehensive list of complications is found in Hobson *et al.* (98). Publications of BAHA complications (88, 90, 91, 94, 98-116) do vary in results and in many other aspects. As mentioned above the development of surgical techniques, material, and indications is still ongoing. A grading according to Holgers *et al.* (103) has been valuable for the comparison of adverse skin reactions in different materials. De Wolf *et al.* has published a thorough and detailed overview (table 4 (94)) where it is obvious that it is a complex task to summon the diversified material to a general approximation of BAHA complications. Hobson *et al.* (98) suggested an overall complication rate of 23.9% and the rate of revision surgery of 12.1%.

Implantable bone conduction hearing aid

The BCI

Even if the BAHA complication rate is fairly low a percutaneous implant requires daily care of the wearer to avoid complications. The abutment sticking out from the head surface makes the titanium implant more vulnerable to traumatic contact situations. Further the fact that a screw is sticking out from the head surface cannot be accepted by some patients due to the stigma. These examples are part of the reason to an ongoing development of a novel BCI (67, 68, 117). The BCI is developed by Bo Håkansson (Chalmers University of Technology, Göteborg, Sweden). The generic feature of the BCI is that the skin is intact, and that the sound signal is transmitted over the skin (transcutaneously) using a magnetic induction system. The signal is transmitted from an external sound processor to an internal receiver and to the implanted transducer which is secured approximately

20 mm behind the ear canal. The vibrations are still transmitted to the cochlea by *dbc* but the position is closer to the cochlea than the standard BAHA position. The transducer in the BCI is using the Balanced Electromagnetic Separation Transducer (BEST) principle (118). The BEST is smaller than the BAHA transducers. As shown in paper III (Figure 5), where the naked BAHA transducer and the capsuled BEST (C-BEST) were driven electrically and tested on a skull simulator, the C-BEST had a higher output force level in the high frequency range (around 3 kHz) due to a high frequency resonance. The roll-off above 3 kHz was steeper compared to the BAHA transducers.

Aims

The aims of the thesis were:

- To investigate the influence from ipsilateral and contralateral BC stimulation at different distances from the cochlea as measured by cochlear vibration.

- To investigate the effect from bilateral BC stimulation on one cochlea as measured by cochlear vibration.

- To investigate the correlation between the vibration of the otic capsule and hearing perception.

- To describe a prototype of a novel BCI.

- To investigate the cochlear vibration from stimulation with the BCI prototype compared to commercially available BC hearing aids.

Method and materials.

The studies in this thesis are approved by the Regional Ethical Review board, Göteborg

Summary of methods

Paper I and II

Seven human cadavers were used. The same cadavers were used in both paper I and II. Four mm titanium fixtures (Cochlear Bone Anchored Solutions AB, Mölnlycke, Göteborg, Sweden) were attached at 8 positions on both sides of the skull. Position 1 to 6 was attached from 55 mm behind and slightly above the ear canal (position 1) in a straight row to position 6, which was situated 5 mm behind the ear canal opening. Position 7 was in the zygomatic root, and position 8 close to, or in contact with, the otic capsule. Each position was stimulated with BC sound from the same transducer (normally used in a Baha® Classic 300) with the frequency range 0.1-10 kHz. The resulting velocity response including both amplitude and phase for a given input force level of 1 Newton, was measured with a LDV on the ipsilateral and the contralateral cochlear promontory. The stability of the fixtures to the skull bone was measured with resonance frequency analysis (RFA) (Osstell transducer and Osstell instrument; Integration Diagnostics AB, Göteborg, Sweden) and mechanical point impedance.

Paper III

Three human cadavers were used. These were other cadavers than in paper I and II. A 4 mm titanium fixture was attached 55 mm behind and slightly above the ear canal opening (position A). Its stability was measured with RFA. At a position approximately 5 mm behind the ear canal opening (position B) a square shaped recess of 16 x 16 x 8 mm was drilled. BC stimulation with a constant input voltage was made at position A with a BEST (118), and in position B with a C-BEST that was secured in the recess (Figure 10 in paper III). The electrical input voltage was 0.5 Volt rms. Sound field stimulation was performed with the Baha® Intenso and the Baha® Classic (complete devices) at position A. For the sound field stimulation at position B the C-BEST was connected to a Vibrant Soundbridge® (Vibrant Med-El, Innsbruck, Austria). This setup implied that the signal was transmitted over intact skin. The sound field pressure levels presented to the devices' microphone were 60, 70 and 90 dB SPL. The resulting velocity response was measured with a LDV on the ipsilateral and the contralateral cochlear promontory.

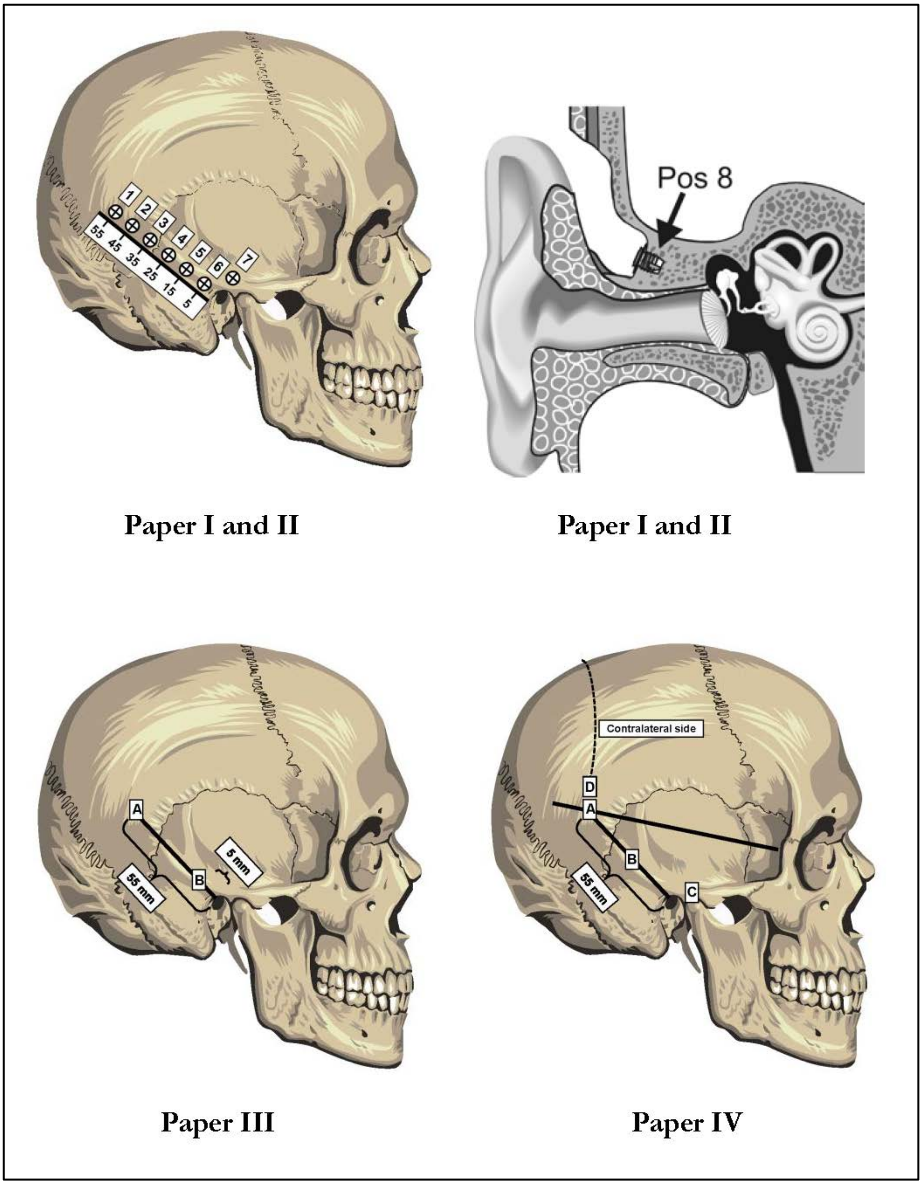


Figure 2. Stimulation positions in paper I-IV

Paper IV

Sixteen live human subjects with a single sided common cavity of the ear (ipsilateral side, test ear) were included. The transducer used was a B-71 transducer (Radioear corp., USA). Masked BC warble tone audiometry was conducted at 16 frequencies between 0.25-8.0 kHz with 1/3 octave resolution from four positions of the head. Positions A, B and C corresponding to positions 1, 4 and 7 in paper I and II. Position D was corresponding to position A, although on the contralateral side of the head. Measurements of the velocity response from BC stimulation at the same four positions were conducted with a LDV. The laser beam was aimed at the cochlear promontory and the lateral semicircular canal (LSCC) where both measuring points were covered with skin.

Subjects

Human cadavers

Human cadavers have been used in three of the four including papers. Through the years of BC research human cadavers and dry skulls have commonly been used. It is known that the damping effect of the live human skull is high. This effect implies that resonance frequencies are damped and do not significantly influence hearing (24). In dry skulls the resonances and anti-resonances are undamped which can make vibration response hard to interpret (22). The difference in vibration response from BC stimulation on a human cadaver skull and a live human skull is assumed to be small, but no such result has been published. In paper I-III neither the skull size nor the thickness of the skull was measured. It is a general belief that the dimensions of the skull affect vibrations of the skull. Khalil (16) reported that smaller skulls had higher resonance frequency and larger skulls a lower resonance frequency. Håkansson *et al.* (24) could not see such a relation, but suggests that thickness and stiffness, and also head size can have a role in skull vibration.

Live subjects

In paper IV 21 patients from a register labeled “due for cleaning of a radical cavity” were collected. For practical reasons only a limited amount of patients could participate in the study. An even spread in age (ranging from 24-70 years) and gender was sought for. A power calculation was done based on the ipsilateral relative vibration velocity of the cochlear promontory from paper I. The velocity response at two positions, positions 4 and 7 from paper I (corresponding to positions B and C in paper IV), relative to position 1 (position A in paper IV) was included in the power calculation. The level of the power was set to 80% and the power calculation was divided in octave bands between 0.1-10 kHz. For position B relative to position A the suggested number of subjects from the power calculation was 14, why this was taken as guidance for the number of subjects needed for a reliable statistical analysis. For position C the highest suggested number of subjects was 10.

Methods

Resonance frequency analysis

In the field of implant osseointegration the stability of which the implant is attached to the bone is important, both for survival and function. Several methods are available to measure the stability of the implant. Removal torque is mainly used in research since it is an invasive method (119). Other methods are the Periotest® which consists of an electromagnet that accelerates a metal slug towards the object of measure. The metal slug is in contact with the object for a certain time which is measured by an accelerometer. The time in contact depends on the stability of the object (119). RFA was used in paper I and II. The method measures the implant stability with bending force. A bar consists of two piezo-ceramic elements. One of the elements is a transducer and stimulates the implant with a frequency sweep from low to high frequencies. The other element measures the response to the stimulation. The first resulting resonance frequency of the bar is measured, and the value is converted to an implant stability quotient (ISQ). The ISQ value depends on the stiffness of the implant and the surrounding bone, the width of the implant and the length of the implant above the bony crest (120). With RFA the implant is loaded by a bending mode and the stability of the implant is measured in two opposite directions. To include the orthogonal directions the implant was turned 90° and measured again. No normative data for craniofacial implant RFA is available but Sennerby *et al.* (121) has proposed an ISQ value between 65-75 to be regarded as a stable implant in the dental region.

Mechanical point impedance

In the introduction section the mechanical point impedance was described as a method to provide information about the mechanical properties of the skull. In paper I and II, mechanical point impedance was used to confirm the stability of the implants. While the RFA uses bending force in an orthogonal direction to the length axis of the implant, the mechanical point impedance exerts a force in line with the implant. In case of a loose implant the impedance will drop dramatically.

Laser Doppler vibrometry

The LDV (HLV-1000, Waldbronn, Germany) uses a laser beam which is reflected on a surface. The laser beam is reflected back giving information about a change in motion of the surface. The change in position leads to a Doppler shift for the laser beam frequency. The output of the LDV provides a voltage that is proportional to the velocity of the surface. A great advantage with an LDV is that it can measure vibration without touching the object, and only limited space is required. LDV has been used for measuring the vibration of the cochlea in many recent studies (21, 36, 37, 54, 66-68, 122-124). Another often used method is the accelerometer (21, 22, 37, 125, 126). A disadvantage with the LDV used in paper I-IV is that it only measures the vibration in one direction. An accelerometer can be built up by three orthogonal accelerometers in the same housing and therefore measure the vibration in three orthogonal directions, which can be of interest when measuring vibration of the human cochlea. On the other hand the accelerometer requires large space, which makes it difficult to use when measuring live human cochlear vibration. Also, the accelerometers will add mass to the measuring point, and that mass will interfere with the skull and may create resonance phenomena that can give erroneous result, especially in the high frequency range. In order to get good reflection from the laser beam it is advantageous to enhance the reflection. If the reflection is good the measurement time is reduced and the results are also more reliable. In paper I, II and IV small glass spheres have been used. The method has been proven to be reliable (21, 36). In paper III a small piece of reflective tape was glued onto the cochlear promontory. No validation of this method was done. In a non-published study (127) comparison was made of vibration response from BC stimulation between a reflective tape on the promontory, and the naked promontory. These measurements were done during surgery in live humans. The vibration response was similar. This finding indicates a reliable response using the glued reflective tape in paper III. The reflective tape in the unpublished study was adhesive to the wet bone. A bonding between the tape and skin (as in paper IV where the promontory and LSCC had a thin skin surface) is not deemed to be as tight as a tape against wet bone. The method using glass spheres was therefore chosen as reflectors in paper IV.

When measuring the velocity of the promontory vibration the motion amplitude can be very small. A vibration amplitude that is too small can imply that the noise floor of the LDV is measured instead of the vibration. In paper I and II continuous registration of the response variance of the LDV revealed a low signal-to-noise ratio (SNR) in the low frequency range below 0.2 kHz. In paper III no such noise floor estimations were conducted. Instead comparative measures from different levels of acoustical input were made. From these curves it seems likely that the results are close to the noise floor below 0.4 kHz and above 5.0 kHz, especially for the BCI and Baha ® Classic measurements at 60 dB SPL input level at the contralateral side. In paper IV the noise floor was controlled by measuring the velocity response from reflectors of naked skin with the LDV without any stimulation. The safe margin above the noise floor was decided to be 10 dB. With this value chosen the results included the important part of the frequency range 0.3-5.0 kHz for speech.

Transducers

Transmission of BC sound in the human skull is assumed to be linear (see introduction). Results from vibration measurements can be misleading if the transducer in certain frequency ranges, or due to low input voltage, does not provide enough output to overcome the noise floor of the instrument measuring the resulting velocity response (i.e. LDV in paper I-IV). Another issue is feed-back from the transducer to the device microphone mainly caused by radiation of sound from the skull added by reverberant acoustical conditions in the measurement room. The latter was an issue in paper III with the cadavers lying on a stainless steel table in a room with tile walls and stone floor. In paper III feed-back was avoided by adjusting the devices volume control settings. In a previous similar study (67) the total harmonic distortion from the devices (Baha® Classic and BEST) was measured during ipsilateral stimulation. It was found that the responses below 0.4 kHz and above 7.0 kHz should be interpreted with caution due to limited transducer output.

Audiometry

Two sessions of tone audiometry were conducted in paper IV. The main reason for a second session was a procedural error when measuring the masked tone thresholds for position D, where the masking was in the wrong ear. For the second session adjustments were made (see paper IV). Since two sessions of tone audiometry were conducted, these could serve as a test re-test procedure. However the methodology between the sessions was changed and therefore no such test re-test was included. One concern was the radiation of sound from the transducer, especially in position C which is the position closest to the test ear, just in front of the tragus. Sound radiated into the ear canal of the test ear can be heard from AC stimulation if the sound pressure level is high enough. The procedure is explained in paper IV. The chosen frequencies for the sound radiation test were 1.0 kHz and 4.0 kHz based on a report by Lightfoot (128). In these measurements we found that AC sound from transducer radiation did not affect the BC thresholds. An obvious complement to the tone threshold measurements is speech audiometry. A speech in noise test was done after the first session of tone audiometry but was not reported in paper IV due to suspicion of method error. In the 5 word speech in noise test according to Hagerman (129) the aim was to compare SNR thresholds between the positions. The signal was at the forehead, an adaptive noise at the four positions, and a masking noise in the non-test ear. The relative SNR results were indicating a worse speech perception when approaching the cochlea. The exception was position D. Furthermore the difference in SNR from one position to the other could in some subject be as high as 15-25 dB which is hard to explain. The speech in noise test will be re-tested in a near future.

Main results

Paper I

Vibration of the cochlear promontory from ipsilateral BC stimulation

A general trend of increasing velocity of the cochlear promontory when the stimulation position was approaching the cochlea ($p<0.001$) was found (Figure 4 in paper I). The velocity response showed large differences between individuals. In the low frequency range the skull moved as a rigid body. All positions except position 8 showed a low frequency anti-resonance followed by a steep increase in velocity response. The closer the stimulation position was to the cochlea the lower the anti-resonance frequency. Above the first resonance frequency the difference in the median absolute velocity response with stimulation at positions 1, 2 and 3 was limited. There was a velocity increase comparing positions 3 to 4, and 4 to 5. Positions 5-7 showed approximately the same velocity response. Position 8 showed an increased response compared to position 7. To enable comparisons between the skulls the median absolute velocity response with stimulation at positions 2-8 were related to position 1. The same pattern appeared when the data were analyzed in this way. It was seen that the velocity response was affected by the squamosal suture. The BC sound transmission was 1-4 dB greater when it did not pass through the squamosal suture compared to when the suture was part of the pathway.

Paper II

Vibration of the cochlear promontory from contralateral BC stimulation

As in the ipsilateral velocity responses there were large individual differences in the velocity response of the cochlear promontory from BC stimulation. The velocity response from stimulation at positions 1-5 were similar for frequencies up to 1.0 kHz. In the same frequency range positions 6 and 7 had lower, and position 8 higher velocity response. Above 1.0 kHz the differences between positions were small with a general tendency of a lower velocity response when the stimulation positions were closer to the cochlea. When positions 2-8 were related to position 1 the described pattern was more obvious. The transcranial transmission was generally decreasing with positions closer to the contralateral cochlea. However the opposite situation was shown for positions 1-5 where the contralateral velocity response was dominating between 0.6-0.8 kHz, an effect that can be ascribed the anti-resonances of the ipsilateral transmission for stimulation at positions 1-5. The velocity response at one cochlea from bilateral stimulation compared to unilateral stimulation was calculated both for equal stationary signals and uncorrelated or non-stationary signals. With equal stationary signals the result was lower velocities in the low frequency range. In the higher frequencies the velocity response was either a constructive or a destructive addition. For uncorrelated non-stationary signals the velocity response showed an equal contribution of energy from all positions in the low frequency range up to 0.3 kHz. From 0.3-1.0 kHz the ipsilateral anti-resonance had the main influence of the summed energy. Above 1.0 kHz the stimulation contribution from the contralateral side was decreasing with positions closer to the cochlea. The median time delay from contralateral stimulation compared to ipsilateral stimulation to one cochlea was greater for positions closer to the cochlea (1.0 ms) than positions far from the cochlea (0.5 ms) at a frequency range 1.0-5.0 kHz. Above 5.0 kHz the difference between positions was limited and approximately 0.3-0.4 ms.

Paper III

Vibration of the cochlear promontory from acoustical and electrical BC stimulation

The ipsilateral average velocity response of the cochlear promontory to acoustical stimulation between position A and B was equal or higher for the BCI (position B, 5 mm behind the ear canal) compared to both the Baha® Classic and the Baha® Intenso (both at position A, 55 mm behind and slightly above the ear canal) in the frequency range 0.7-7.0 kHz. The difference was 5-10 dB and 0-5 dB respectively. The contralateral response was generally lower for the BCI compared to the BAHA devices. The difference between electrically driving the BEST at position A and the C-BEST at position B was 10-20 dB higher with BC stimulation at position B compared to A, in the same frequency range as above (0.7-7.0 kHz). The contralateral velocity response was similar for the two positions between 1.0-4.0 kHz. Below 1.0 kHz and above 4.0 kHz the average contralateral velocity response was higher for position A. The transcranial attenuation was 5-25 dB for stimulation at position B (0.5-9.0 kHz) and around 0 dB or slightly negative for position A.

Paper IV

Tone thresholds and vibration measures. Difference between positions.

The velocity responses from the LSCC and the promontory were highly correlated ($r>0.8$, $p<0.001$). Even though there were large individual differences of the responses at the otic capsule, Wilcoxon's signed rank test showed significant differences of the velocity response at the otic capsule from stimulation at positions B and C in relation to position A: at 1.0-1.6 kHz for position B ($p<0.05$) and at 0.315-1.0 kHz and 2.5 kHz for position C ($p<0.05$). No significant differences were found when position D was compared with position A. Above 2.0 kHz the velocity responses were more similar between positions. For the tone audiometry there was a general tendency of lower hearing thresholds when the stimulation position approached the cochlea. Using Wilcoxon's signed rank test significant differences between positions B, C and D in relation to position A were found between 0.5-1.6 kHz, and at 8.0 kHz for position B ($p<0.05$), at 0.315, 0.4, 2.5, 3.15 and 8.0 kHz for position C ($p<0.05$), and at no frequency for position D. The correlation between the two methods was low but Wilcoxon's signed rank test only showed sporadic significant differences. When comparing the LDV results in paper IV with paper I and paper II (66, 122) the pattern of the different curves were similar but there were differences in the magnitude of the velocity response of the otic capsule, especially for position C.

Discussion

Vibration of the cochlea - a valid estimation of hearing perception?

Hearing by BC is dependent on skull anatomy and mechanical properties, different pathways for stimulation of the basilar membrane and possibly different vibration directions. Hearing by BC is therefore a complex phenomenon and is still not understood to its full extent. The vibration of the cochlea is one way to approach BC hearing. By measuring the cochlear vibration the end organ for hearing by BC is studied (1-5). Further BC transmission in the human skull is linear (3, 13-17) at least for the frequency range 0.1-10 kHz and up to 77 dB SPL (14).

One of the aims of this thesis was to investigate the relation between vibration of the otic capsule (including the cochlear promontory and the LSCC) and hearing perception. This method is based on several research findings (see introduction for details). (1) The inner ear fluid inertia is the most important pathway for the stimulation of the basilar membrane. (2) For BC stimulation at the mastoid the vibration response for the lateral-medial direction (x-direction) is dominating in low frequencies when the stimulation direction coincides with the x-direction, and is equal or higher than the other directions at higher frequencies. (3) Perceptual measures of the transcranial transmission and the threshold difference between BC stimulation at the forehead and the mastoid correlate on average well with the same measures on dry skulls or cadavers measured as cochlear vibration (21, 69).

Inner ear fluid inertia

By measuring the vibration of the cochlea we assume that the inner ear fluid inertia is the most important BC sound pathway. What are the bases for this assumption? In order to achieve a travelling wave of the basilar membrane there must be a pressure gradient and a fluid flow over the basilar membrane. This fluid flow can occur from inertial forces due to the compliance of the oval window and round window membranes (30). Also, there are outlets for both the endolymphatic and perilymphatic space (see introduction). Stenfelt *et al.* (54) reported that fluid displacement at the round window was 5-15 dB greater than at the oval window below 2.0 kHz. Above 2.0 kHz the fluid displacement was changing with frequency to become 5-15 greater at the oval window above 7.0 kHz.

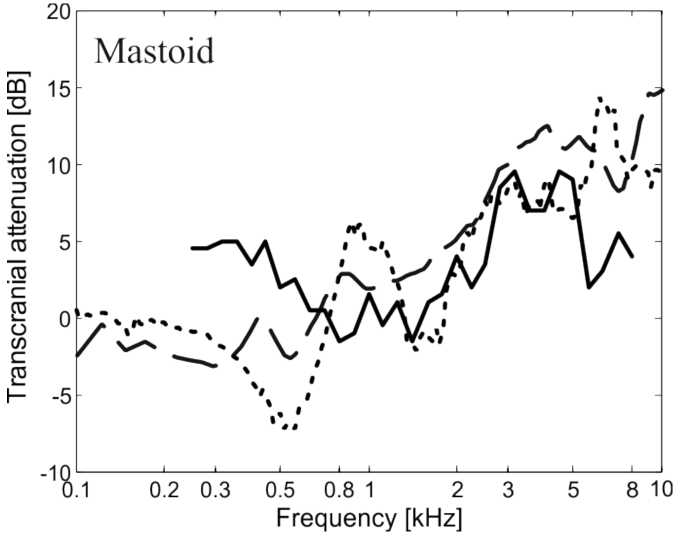
At low frequencies below 0.3 kHz the skull moves as a rigid body when stimulating with BC sound (15, 21). No sound waves are produced that affect the cochlear space. The middle ear ossicles are also moving in phase with the surrounding bone at this low frequency (36) and are not a contributing factor. The outer ear component has no significance below 0.4 kHz (31). Thus when the skull moves as a whole in response to BC stimulation at this low frequency, inertial forces can produce a travelling wave of the basilar membrane.

Above this frequency the skull bone starts to flex and the stimulation position is gradually changing from a mass controlled to a stiffness controlled motion. Sound waves can propagate through the skull bone and affect the cochlear motion. If the outer ear is occluded and the middle ear ossicles are intact in a normal ear, the outer ear component and the middle ear component can be the dominant contributors to BC hearing up to 4 kHz. If the condition of the outer ear and the middle ear do not allow for these BC components to contribute to BC hearing, for example in conditions where the oval window is immobile (38-40) or the round window is obliterated (47-50), the BC thresholds are only marginally affected. In other words, the inner ear fluid inertia can still, and despite lesions mentioned above, stimulate the basilar membrane.

Transcranial attenuation and standard threshold difference between forehead and the mastoid

Comparisons have been made between the vibration of the cochlea and threshold measurements for transcranial attenuation and standard threshold difference between the forehead and the mastoid. There is an agreement between these methods, but not at low frequencies, and to some degree not at high frequencies. Stenfelt (65) has described comparisons between laser and threshold measurements of the transcranial attenuation. This is displayed in Figure 3. As can be seen there is a quite low agreement in the low frequency range where the vibration transcranial attenuation of the cochlea is lower than the perceived transcranial attenuation. In mid frequencies the agreement is better, but is worse again in the high frequency range. The latter is especially evident for the mastoid stimulation position, corresponding to position 4 in paper I and II. Reinfeldt (69) made the same comparison with a similar resulting difference between the two methods. The comparison is also described by Stenfelt *et al.* (21) concerning forehead and mastoid difference where a difference is noted especially above 3 kHz.

A



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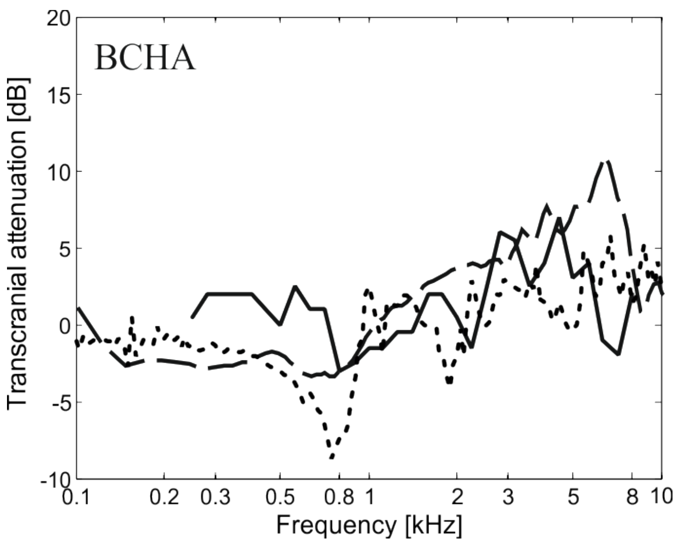

Figure 3. Transcranial attenuation measurements from Stenfelt 2012 (65) (solid line) measured with BC pure tone thresholds, Eeg-Olofsson et al. 2011 (66) (dotted line) measured with laser Doppler vibrometer in 1 direction and Stenfelt et al. 2005 (21) (dashed line) measured with triaxial accelerometer in 3 directions. (A): Results from stimulation at the mastoid (corresponding to position 4 in paper I and II, and position B in paper IV. (B): Results from stimulation at the bone conduction hearing aid (BCHA) position, corresponding to position 1 in paper I and II, and position A in paper III and IV. From Stenfelt 2012 (65) reprinted with permission.

Direction sensitivity

Skull resonances are damped in cadaver heads and live human heads. It is suggested that these resonances have no significant effect on BC hearing perception (21, 24). Sharp anti-resonances can have an effect on BC perception (24) and might depend on the cochlear sensitivity to different vibration directions (21). Such a sensitivity is today unknown. There is a possibility that by using a single direction measure (as with the LDV) information is lost from other vibration directions. Comparisons have been made on cadaver heads between the quadratic summation from all directions (the vibratory energy transmitted to the cochlea from BC stimulation) and the single x-direction response. Close to the cochlea (20 mm and 40 mm behind the ear canal) the x-direction is similar to the quadratic summation (vibratory energy) of all directions during BC stimulation. For the forehead-mastoid threshold comparison the x-direction corresponds better below 3 kHz, and the quadratic summation above 3 kHz (21). In the high frequency range the cochlea moves in all directions why the directional sensitivity should have low impact. According to the present theory of BC pathways above 4 kHz, compression and expansion of the cochlear shell is the main contributor to basilar membrane stimulation (30). This might be an explanation for the worse agreement between threshold and vibration results from measures of the transcranial attenuation at the high frequency range (see Figure 3).

Comparison between vibration of the otic capsule and thresholds in the same individuals (paper IV)

The vibration response of the cochlear promontory from position 7 relative to position 1 in paper I was 5-10 dB higher than for the corresponding measure of the otic capsule in paper IV. The reason for this discrepancy has been discussed in paper IV. It is worth noting that the same measure for positions B (position 4) and D (position 1 contralateral) were similar comparing the results in paper I and II, and paper IV. It was speculated if a common cavity, which all subjects in paper IV had, could influence the sound transmission. If so, only position C was affected. It is more likely that other factors, such as the contact between the transducer and the zygomatic root, played an important role for the discrepancy. Position C varied substantially between subjects in the sense that the zygomatic root had different anatomical features, as well as being covered by skin and soft tissue of various thicknesses. The zygomatic root also has quite a thin ridge which the transducer was placed upon. This ridge could imply both different contact and different direction of stimulation. Moreover, position 7 in paper I was positioned slightly posterior to position C. The anchorage with a 4 mm titanium fixture in paper I did secure the contact and the direction of stimulation in a more controlled manner than in paper IV.

The vibration of the otic capsule is not a valid measure to estimate BC hearing for an individual subject. The median result comparing vibration of the otic capsule and BC thresholds for all participants, indicates that vibration measures in the x-direction is a valid estimation of hearing perception. The large standard deviation in both vibration and threshold measures implies that the number of subjects must increase to better understand the individual correlation between vibration and threshold. The low frequency discrepancy does support previous findings between threshold and vibration measures (65). The reason for this discrepancy is unknown.

Human skull sound transmission

Mechanical point impedance and velocity response

In paper III the impedance data from paper I was investigated further. At positions on the mastoid surface area that attaches to the petrous part of the temporal bone (MAPP) the mechanical point impedance was higher than outside the MAPP-area. A higher impedance value (with the same mass assumed for the different positions) means that the bone at the stimulation position is stiffer. Thus when a position with higher mechanical point impedance (see Figure 2 in paper I) is stimulated with a BC transducer the bone starts to flex at higher frequencies (only by approximately 0.1 kHz). It is important to distinguish mechanical point impedance results from the transfer function between stimulation position and the measuring point. In the transfer function the first anti-resonance is a forced phenomenon meaning that it is not only dependent on structural properties of the skull but depends on the position of the applied force too, and also where the measuring point is positioned. If either are moved the anti-resonance frequency changes. In Figure 4 and Figure 5 in paper I it is shown that the forced anti-resonance occurs at a lower frequency for positions on the MAPP than positions outside the MAPP. Even if the mechanical point impedance and the transfer function are separate measures, the frequency of the maximum impedance of positions close to the cochlea (positions 6 and 7) coincides with the frequency for their first anti-resonance of the transfer function. A plausible explanation to this could be that the temporal bone, or a part of the temporal bone, moves separately to the rest of the skull, at least up to the first skull resonance. Another indication of a separate moving temporal bone is the contralateral stimulation at position 8 (see below). At position 1 the forced anti-resonance occurs at a higher frequency which is close to the first skull resonance. The anti-resonance is probably a decoupling effect of position 1 from the temporal bone and the cochlea. The low frequency raise in velocity response of the cochlear promontory from the first anti-resonance to the first skull resonance for positions on the MAPP can imply a higher stimulation to the cochlea at a lower frequency compared to BC stimulation outside the MAPP. For position 8 which is very close to, or actually in contact with the otic capsule, there is no clear anti-resonance in the ipsilateral stimulation. Due to the mastoidectomy the anatomical structure is changed when stimulating at position 8 compared with stimulation at the other positions. This change might influence the response.

When investigating the contralateral response the forced anti-resonance frequencies are more spread and therefore it is not seen in the median data (Figure 4, paper II). The difference in velocity response is not as evident as for the ipsilateral stimulation. Close to the ear canal there is a general drop in velocity response for positions 6 and 7. This might be due to a rotational motion which reduces the response on the cochlear promontory. Position 8 on the other hand has a higher velocity response from 0.4-2.0 kHz compared to the other positions. It is speculated that this increase can be due to a part of the temporal bone moving separately from the rest of the skull. No mechanical point impedance data was possible to obtain from position 8 due to limited space. Even though the contralateral response from position 8 is relatively high in the low and mid frequencies the transcranial transmission (the quotient between contralateral and ipsilateral response from corresponding stimulation positions) is low. In general, the closer the stimulation position is to the cochlea, the lower the transcranial transmission. Hence the major difference in transcranial transmission is due to the ipsilateral transmission.

Bilateral stimulation and time delay

Time delays and velocity results in BC transmission of the human skull have varied between studies (see introduction). In paper II the contribution to one cochlea from simultaneous bilateral BC stimulation compared to BC stimulation only on one side depends on both frequency and stimulation position. Below approximately 250 Hz the two transducers would work in opposite directions and the resulting velocity of the cochlear promontory was estimated to be lower. Above 1.0 kHz the contribution from bilateral stimulation decreases with frequency. It is shown in Figure 7 in paper II for equal stationary signals where the amplitude of the constructive and destructive addition gets lower, and for uncorrelated non-stationary signals where the sum of the sound energy decreases with increased frequency. The effects from bilateral BC stimulation described above were enhanced with shorter distance between the stimulation position and the cochlea. The time delay between contralateral and ipsilateral BC stimulation was greatest for stimulation at positions close to the cochlea in the mid frequencies (1-3 kHz). No results were obtained from lower frequencies due to the group delay function not being defined in the presence of anti-resonances and resonances. Even if the difference in the amplitude of the velocity response of one cochlea from bilateral BC stimulation is small in lower frequencies, there is also a time difference in this frequency range. Thus it might be possible to extract binaural cues in the time domain at low frequencies from bilateral BC stimulation. From a clinical perspective many studies indicate that binaural cues from BC stimulation cannot be extracted to the same extent as in AC stimulation. The reason is believed to be crossover of BC sound to both cochleae reducing the information difference between the two cochleae. The binaural masking level difference (BMLD) tests have been conducted using BC stimulation (71, 73) and there has been evidence of the ability of binaural hearing from BC stimulation. Dichotic stimulation with phase alteration leads to a masking level difference. From Rowan *et al.* (77) the alteration of phase of BC stimulation between the skull sides at high frequencies is suggested to lead to an amplitude difference between the cochleae. In paper II the result from bilateral BC stimulation of stationary signals at one cochlea is either a constructive or a destructive addition. Even if time differences between the cochleae can be extracted at the brain stem level, the BMLD test might be a result of addition of stationary signals.

BCI

In paper III the results from stimulating at different positions of the human skull (paper I and II) were repeated in a slightly different manner. To measure frequency response (sensitivity) difference between position A and B the BEST and the C-BEST were driven with a constant input voltage which means that the transducer performance is included. The purpose was to investigate the sensitivity differences as well as the transducer differences. The sensitivity difference between positions A and B was similar to the same comparison in paper I and II (positions 6 relative to position 1). Besides the similar result it was concluded that the flat surface by which the C-BEST is in contact with the underlying bone can replace the titanium screw implant. Further a complete device comparison was made between a prototype of the BCI in position B, and two BAHA devices. The BCI prototype was a Vibrant Soundbridge® device where the C-BEST replaced the floating mass transducer. The maximum power output was chosen for comparison between the devices. At maximum power output the full capacity of the device is measured. An output below the maximum power output can imply that the amplifier setting decides the output of the device which can blur differences between the devices. It was shown that the BCI had an equal or higher output in mid-, and high frequencies for ipsilateral stimulation, and a generally lower output for contralateral stimulation measured as cochlear vibration, compared to the two BAHA devices. The higher output around 3 kHz is likely related to the high frequency boost of the BCI around 3 kHz.

The BCI is currently entering the stage of long term human trials. It is a project with complicated engineering challenges of all parts included. From a medical point of view the implant has to be put in place by a minor surgical intervention. As in the majority of cochlear implants and middle ear implants the signal information over the skin and soft tissues to the receiver unit is accomplished using an inductive link. The surgical procedure is anticipated to be both easier and faster compared to cochlear implants and middle ear implants since the mastoid cavity will be left intact, except for a 4 mm deep seating for the transducer. An incision through the skin and soft tissues of the mastoid and through the periost will be made at the level of, or slightly behind and above the posterior border of the pinna. The pinna will be pushed and secured forward. The transducer size is 13.5 x 13.5 x 7 mm with rounded corners. A 4 mm deep seating for the transducer is drilled according to the size some 20 mm from the ear canal opening. The internal magnet part with the receiving coil will be placed under the periost in the posterior aspect of the incision and the transducer is put in place in its seating. The transducer is then secured with orthopedic screws (1 x 4 mm) through a metal bar that holds it in place. The procedure will be done under general anesthesia for the first patients. Depending on the outcome and evaluation of the surgical procedure it might be possible to place the BCI under local anesthesia.

Besides what is already mentioned in paper III important questions remain. Despite the results from the papers included in this thesis, the actual audiometric outcome of the BCI in place in a live human is unknown. A few both published and non-published studies add information regarding implant-bone interface, speech audiometry and feedback issues.

Animal study

An animal study was undertaken (130) on three sheep where a circular titanium cylinder with a diameter of 6 mm (the flat surface in contact with the bone) was placed in the mastoid bone, where the mastoid cells were visible in the pre-drilled seating. The titanium cylinder was secured by a metal bar exerting a pressure on the cylinder top. An impedance head and a transducer were attached to the titanium cylinder for measurement of the mechanical point impedance. An accelerometer was attached to the titanium cylinder on the contralateral side for measurement of the transfer function. Measurements were conducted at installation and after a healing period of 8 months. The mechanical point impedance and the transfer function measured at three input levels varied between the sheep. The left implant of sheep 3 did not heal and could therefore not be stimulated. The left implant of sheep 1 was not rigidly attached after the healing period. The mechanical point impedance and the transfer function of the remaining implants were, after 8 months, similar to the first measurements. The seatings were smooth and there were no signs of resorption. A thorough analysis of the results is ongoing as well as histologic examinations of the implant-bone interfaces.

Speech audiometry in live humans

A non-published study investigated tone thresholds and speech audiometry in subjects with titanium fixtures for both an auricular prosthesis, and a BAHA (131). The BAHA was positioned posterior to the fixtures for the auricular prosthesis with an average distance of 28 mm. The speech audiometry was conducted according to Hagerman (129), with fixed noise and adaptively changing the speech signal at the two positions. There was no significant difference in tone threshold between the two positions (n=10, contralateral ear masked in 6 subjects). The SNR improvement for the speech audiometry was on average 8.7 dB (range 5.2-10.5 dB) (n=4, contralateral ear masked in 3 subjects). The results indicate that speech perception improves when the stimulation is closer to the cochlea than the standard BAHA position.

Feedback analysis

Taghavi *et al.* (132) conducted an investigation of feedback from the BCI and from a percutaneous bone conduction device (PBCD) (Baha ® Classic) on a dry skull. The main cause of feedback in a PBCD is that sound radiates from the skin close to the device, and also from the housing. The inherent limitation for the output force level in the PBCD is the gain headroom (the extra gain that can be provided before the device oscillates) which amounts to 3-4 dB in linear devices with feedback suppression in off mode. The PBDC were positioned at the standard BAHA position 55 mm behind the ear canal. The BCI was positioned 10-15 mm behind the ear canal. It was shown that the BCI had 10-30 dB more gain headroom when the mechanical output was normalized at the cochlear level. The latter means that the input was adjusted so that the acceleration at the cochlea from BC stimulation with the PBCD and the BCI was equal. More specifically the improvement in gain headroom at the critical frequency (lowest gain headroom for the BCI) was 17 dB. One reason for the increased gain headroom for the BCI can be the higher mechanical point impedance at a position close to the ear canal (see paper I and III). A higher mechanical point impedance probably leads to lower radiation from the skull bone for a given force stimulation level. Another, and maybe, more important reason is that the transducer in the BCI is completely encapsulated in the transducer casing and separated from the microphone in the sound processor. The feedback from the transducer to the microphone is thereby reduced and is most likely less than in the PBCD, where the transducer and the microphone is placed in the same housing. Such gain headroom could imply that the amplification in the high frequencies can be increased compared to the PBCD, which can be of importance, especially for patients where a sensorineural hearing loss needs to be compensated.

BCI considerations

Since there are no live human results at this point of time one can only speculate around this matter. In paper I, II and III the measurements have been conducted in cadavers with intact mastoids. The planned position for the BCI is 20 mm behind the ear canal which requires an intact mastoid. In many patients where there is an indication for a BC hearing aid, surgery of the mastoid has been executed. This might imply that the BCI has to be applied further back and probably the efficiency of sound transmission to the ipsilateral cochlea decreases. From an audiological point of view this alteration of position is a minor problem, even if a patient with a sensorineural hearing loss might benefit slightly less from the BCI at a position further back. From a surgical point of view a 4 mm deep seating on the cranial vault can lead to a thin bone plate above the dura, or actually contact with the dura. The consequences for both hearing perception and medical complications for such a solution will be a matter of extended research. Another interesting situation is if the BCI can be applied to the planned position 20 mm behind the ear canal in obliterated mastoid cavities.

Doing MRI scans with the BCI implanted can damage both the magnetic components of the BCI and distort the MRI image so it is not useable. A solution is to temporarily remove the implanted part. This action requires an operation including surgical risks and a possible risk of destroying the implant, and/or the implant site. Research in this field is ongoing where alterations of implanted magnetic parts can overcome obstacles concerning MRI scans.

A common question regarding the BCI is if it is a necessary alternative to the percutaneous solutions. The answer is obviously no. The BAHA is, and has for the last 30 years, been a success. The complication rate is a concern for BAHA wearers but not a sole reason for developing an implantable alternative. Even if the BAHA complication rate was very low an implantable solution would probably be developed. The cost effectiveness will certainly be scrutinized. If the overall complication rate is low and the BCI will be robust in its construction and long term function, the cost effectiveness will be an important factor for the future existence of the BCI.

A

B

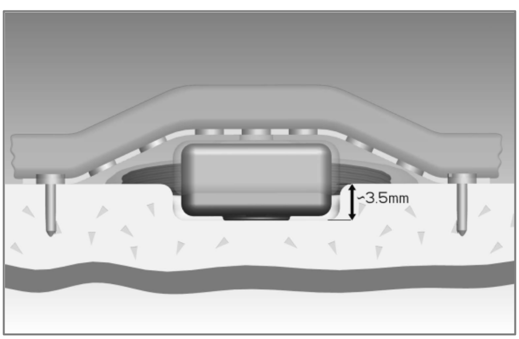

Figure 4. The principle design of the BCI is shown in (A). To the upper left the retention magnet and the receiving coil. To the bottom right the transducer with the metal bar attached, to secure the transducer in its seating. In (B) a drawing of the transducer in place in its seating is shown. Note the flat contact between the transducer and the bone.

The significance of the results in this thesis

It is important to emphasize that the conclusions drawn from the results in paper I, II and III are based on cadaver measurements and not live humans. Even if the results in paper IV indicate that hearing perception from BC stimulation can be estimated by measuring the vibration of the cochlea, at least for frequencies above 1.0 kHz, a more thorough validation of these results require a larger study population.

Paper I-IV add pieces to the knowledge of the physiology of BC sound of the human skull. By stimulating at 8 close positions (paper I-II), novel information has been gained providing understanding of the transfer function from BC sound stimulation at the parietal bone and the temporal bone. Paper III is more clinically related where complete BC devices stimulate at different positions. Paper IV is focused on validation of studies using the vibration of the cochlea as a perceptual estimate, including Papers I-III.

In paper I and II BC stimulation at a position closer to the cochlea improves sound transmission to the ipsilateral cochlea and separates the sound between the ipsilateral and the contralateral cochlea, as measured by the cochlear promontory velocity. The measurements were based on constant force stimulation which is not clinically relevant. How much stimulation that in the end reaches the cochlea depends on (1) the incoming sound, (2) the hearing aids processing of sounds, (3) the properties of the hearing aid transducer, and (4) the transfer function between the skull surface and the cochlea. In paper I and II we investigated only the latter. On the other hand, if the comparison is made with the same BC hearing aid stimulated with the same incoming sound, the results show the sensitivity difference between the complete devices. The possibility to increase the BC stimulation on the ipsilateral side by placing the BC stimulation closer to the cochlea could be beneficial for patients with a sensorineural hearing loss. The improved sensitivity also gives a larger gain headroom reducing feed-back problems (132). This is also important if large gain is needed to compensate for a large sensorineural component.

As stated above, binaural hearing is possible from bilateral BC stimulation. In paper II and III it is shown that sound transmission separation between the two cochleae can increase with BC stimulation at positions with short distance to the cochlea compared with positions far from the cochlea. An increased sound separation could imply improved binaural hearing.

Findings concerning the BCI are encouraging and the project will proceed by implantation of the full device in live patients. A risk analysis is ongoing. The high frequency boost has been adjusted from 3.0 kHz to 4.5-5.0 kHz. This adjustment covers the high frequency speech sounds and could be important for patients with a sensorineural hearing loss in the high frequencies.

Conclusions

- BC sound transmission improves with shorter distance between stimulation position and the cochlea, measured as cochlear vibration.

- BC sound transmission is more isolated from the contralateral cochlea with a shorter distance between stimulation position and the ipsilateral cochlea, as measured by cochlear vibrations.

- The method of measuring the vibration of the otic capsule from BC sound stimulation is a valid estimation of hearing perception. A BC hearing aid is therefore likely to be more efficient applied at a stimulation position close to the cochlea.

- Bilateral application at a position close to the cochlea could be beneficial for binaural hearing.

- The BCI is a realistic alternative to other BC hearing aids.

Populärvetenskaplig sammanfattning på svenska

Bakgrund

Ljud som vi hör går via ytteröronen, mellanörat, hörselsnäckan och sedan in till hjärnan. En del av det ljud vi hör kommer också via skallbenet (benlett ljud), vidare till öronbenet och hörselsnäckan, och sedan till hjärnan. Ett klassiskt exempel på detta är när vi lyssnar på vår röst när den spelats in på band. Då låter den främmande. Det beror på att ungefär 50 % av ljudet från vår röst går via benledning och låter därför annorlunda än det ljud som går ut via munnen.

Anledningen till att benlett ljud i skallbenet är så studerat är dess betydelse för hörselmätningar, olika hörselskador och hörhjälpmedel. Individer som har kroniska öroninflammationer eller missbildningar på hörselgången och mellanörat är ofta i behov av hörhjälpmedel. Ibland är det svårt att anpassa en vanlig hörapparat vid dessa tillstånd. Fram till i början av 1980-talet var det vanligt att man satte en benledningshörapparat mot huden bakom örat. En benledningshörapparat tar upp ljud via en mikrofon. Ljudet omvandlas i hörapparaten till vibrationer som i sin tur förs vidare över hud och mjukvävnad, till skallbenet, och sedan med ljudvågor i skallbenet till hörselsnäckan. Problemet med dessa apparater var att de hölls på plats med glasögonbågar eller en stålbåge på huvudet, vilket var obekvämt och inte alltid estetiskt tilltalande. Vidare var ljudkvaliteten dålig, mest beroende på att huden dämpar ljudöverföringen. När vibrationer leds över hud och mjukvävnad måste vibratorn och elektronik med batteri också klara av att skapa den extra kraft som krävs. Det skulle medföra en betydligt större hörapparat vilket skulle vara än mindre tilltalande för användaren. En avgörande förbättring av situationen prövades i slutet av 1970-talet av Anders Tjellström på ÖNH-kliniken, Sahlgrenska Universitetssjukhuset. Forskningsresultat framtagna under ledning av Per-Ingvar Brånemark (Professor emeritus i Anatomi, Göteborgs Universitet) hade visat att titan hade en unik egenskap att fästa hårt mot ben. Modellen hade testats med stor framgång på många patienter som fått titanimplantat fastskruvade i tandlösa käkar, och sedan fått konstgjorda tänder monterade på titanimplantatet. I samarbete med Professor Bo Håkansson (Chalmers Tekniska Högskola) fäste Anders Tjellström en benledningshörapparat på ett titanimplantat som man förankrat i benet bakom örat. Den benförankrade hörapparaten (BAHA) var född. Sedan dess räknar man med att över 80 000 individer har rehabiliterats med BAHA. BAHA är en succé. Trots denna succé så finns det avigsidor med BAHA. Operationen när man sätter skruven på plats innebär att man tar bort all mjukvävnad utom tunn hud, inklusive hårsäckar, på ett ca 6 cm^2 område runt skruven. Skruven bakom örat går igenom ett hål i huden. Det gör hudområdet känsligt och exponerat för smuts och bakterier. För att undvika inflammationer och infektioner måste användaren sköta sin skruv varje dag genom enkel men noggrann hygien. Ibland hjälper det inte med god hygien utan hudproblem uppkommer ändå vilket kan kräva sjukhuskontakt, och i svåra fall även en ny operation. Skruven sitter i allmänhet hårt fast men kan tappa greppet av okänd anledning. Den kan också lossna vid trauma. Skruven är diskret men uppfattas ibland som kosmetiskt besvärande.

Benlett ljud i skallbenet är ett komplicerat fenomen. Med t.ex. en benledningshörapparat förs vibrationer vidare till hörselsnäckan genom skallbenet. Genom att hörselsnäckan vibrerar stimuleras basilarmembranet i hörselsnäckan vilket leder till att sinnesceller för hörseln aktiveras. Man har visat att basilarmembranet stimuleras på samma sätt av benlett ljud som av ljud som går via ytterörat och till hörselsnäckan (luftlett ljud). Med andra ord är det samma målorgan för luftlett och benlett ljud. Det benledda ljudet når hörselsnäckan på olika sätt beroende på ljudfrekvens. I låga frekvenser rör sig hela skallen som en helhet. Hörselsnäckan rör sig också med hela skallen men vätskan i hörselsnäckan rör sig inte i samma takt, och kan då stimulera basilarmembranet i hörselsnäckan. Detta beror på innerörevätskans ”tröghet”. I högre frekvenser sker en vågutbredning från stimuleringspunkten som får hörselsnäckan att vibrera. Återigen kan den trögare vätskan i hörselsnäckan stimulera basilarmembranet. Andra vägar bidrar också till att stimulera basilarmembranet vid benlett ljud, t.ex. hörselbenens tröghet vid vissa frekvenser och att hörselsnäckan komprimeras och utvidgas vid andra frekvenser. Om vi täpper igen hörselgången kan bidraget av ljud från vibrationer i hörselgångens ben och brosk föras vidare via trumhinna och hörselben till hörselsnäckan och ge en ljudupplevelse. Skallens anatomi bidrar också till att benlett ljud är komplicerat. Vidare är skallbenet sammanfogat av bensömmar som kan påverka ljudvågsutbredningen över skallen. Som nämnts ovan vibrerar hörselsnäckan vid benlett ljud. Den vibrerar åt alla olika håll och förmodligen är hörseln mer känslig för vibrationer i vissa riktningar och vid vissa frekvenser.

Det finns många studier som pekar mot att uppmätta vibrationer av hörselsnäckan motsvarar det vi uppfattar av det inkomna ljudet. Många av dessa studier är gjorda på torra skallar och skallar på lik. Fördelen med att använda torra skallar och lik är att man kan göra omfattande undersökningar av benlett ljud, vilket hade varit omöjligt på en levande människa. Dock har inte vibrationer av hörselsnäckan på levande människa gjorts, och därigenom inte heller säkerställt en koppling mellan uppmätt hörsel och vibrationer av hörselsnäckan på samma individ.

Bone Conduction Implant (BCI) är en benledningshörapparat som är under utveckling i samarbete mellan Chalmers Tekniska Högskola och Sahlgrenska Universitetssjukhuset. Det som skiljer BCI från andra benledningshörapparater är att den opereras in under huden, och kommer efter läkning fungera under intakt hud. Den placeras också i direkt kontakt med öronbenet närmare hörselgången än

BAHA. Flera studier (inklusive studie I och II i denna avhandling) har visat att man får en förbättrad benledd ljudöverföring till samma sidas hörselsnäcka om avståndet mellan stimuleringspositionen och hörselsnäckan är liten. Ljudöverföringen till den motsatta sidans hörselsnäcka blir däremot försämrad vilket separerar ljudet mellan snäckorna mer. Detta är positivt eftersom det kan innebära att man lättare lokaliserar en ljudkälla om man har hörapparater på båda sidor. Förhoppningen med BCI är en minst lika effektiv benledningshörapparat som BAHA, men då BCI innebär att huden är intakt kan komplikationsrisken bli mindre.

Syfte

- Att mäta hur hörselsnäckan vibrerar vid benledd stimulering från olika positioner på skallen.
- Att mäta hur samtidig stimulering på båda sidor av skallen påverkar en hörselsnäcka.
- Att undersöka kopplingen mellan vibration av hörselsnäckan och uppmätt hörsel.
- Att beskriva en ny typ av implanterbar benledningshörapparat (BCI).
- Att undersöka hörselsnäckans vibration vid stimulering med BCI jämfört med befintliga benledningshörapparater.

Metod

Hörselsnäckans vibrationshastighet är mätt med en laser Doppler vibrometer. Detta är gjort på lik, men också på patienter där ena sidan är speciellt lämplig för denna typ av mätning efter en tidigare genomförd öronoperation. På liken var det 8 stimuleringspositioner på varje sida av skallen (studie I och II), eller två stimuleringspositioner på skallens ena sida (studie III) där BCI jämfördes med BAHA. På patienterna var det tre stimuleringspositioner på samma sida som vibrationsmätningarna, och en position på motsatt sida. På patienterna utfördes också hörselmätningar som kunde jämföras med vibrationsresultaten.

Resultat

Ju närmare hörselsnäckan man stimulerar med benlett ljud, desto mer rör den sig medan den motsatta hörselsnäckan rör sig mindre. Vid samtidig dubbelsidig benledd ljudstimulering adderas ljudsignalen från två rena toner och resultatet blir antingen större eller mindre än om signalen bara hade kommit från en sida. Vid dubbelsidig ljudstimulering med komplexa signaler (tal t.ex.) så blir den sammanlagda energin på en hörselsnäcka bara större. Påverkan från dubbelsidig stimulering är lägst vid positioner nära hörselsnäckan och vid höga frekvenser. Kopplingen mellan vibrationer och uppmätt hörsel är tydligast om man betraktar medelvärden av resultaten från de deltagande patienterna. Jämför man kopplingen för varje individ är det mer otydligt. BCI ger jämfört med andra befintliga benledningshörapparater en större stimulering av hörselsnäckan vid viktiga ljudfrekvenser för uppfattning av tal, och en liknande stimulering i andra frekvenser utom i de högsta och lägsta frekvensregistren där BCI är något sämre.

Slutsats

Benledd ljudöverföring till samma sidas hörselsnäcka ökar vid en stimuleringsposition nära hörselsnäckan, men minskar till den motsatta hörselsnäckan. Att stimulera bägge sidor samtidigt kan vara bra för att kunna lokalisera ett ljud, och för att kunna urskilja ett ljud i bakgrundsbrus. Att mäta vibrationerna från hörselsnäckan överensstämmer med upplevt ljud. BCI är ett realistiskt alternativ till befintliga benledningshörapparater.

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