Psychoacoustic Measurements of Bone Conducted Sound

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for the degree of Doctor of Philosophy

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Contents

Thesis submitted to Cardiff University i

Declaration iii

Statement 1 iii

Statement 2 iii

Statement 3 iii

Statement 4: iii

List of Figures x

List of Tables xiii

List of Symbols xiv

Acknowledgements xvi

Summary xvii

1 Introduction 1

1.1 Bone Conduction 1

1.2 Mechanisms of Bone Conduction 2

1.2.1 Inner ear fluid 3

1.2.2 Middle ear 5

1.2.3 The outer ear 5

1.2.3.1 Occlusion effect and the outer ear 6

1.3 Summary of relative contributions 7

1.4 Transcranial attenuation and interaural level difference 7

1.5 Advantages of hearing with two ears 8

1.5.1 Sound localisation 9

1.5.2 Head shadow effect 10

1.5.3 Binaural Summation 10

1.5.4 Binaural Squelch 11

1.5.5 Binaural unmasking 11

1.6 Spatial release from masking 11

1.6.1 Energetic masking 12

1.6.2 Informational masking 12

1.7 Hearing loss 13
1.8 Bone Conduction Hearing Aid 13
1.8.1 Device Components 14
1.8.2 Clinical Indications 14
1.8.3 Clinical indication for Children 15
1.9 BCHA in single sided deafness 15
1.10 Bilateral BCHAs 19
1.10.1 Subjective Outcomes 19
1.10.2 Listening Preferences 20
1.10.3 Audiometric measurements 20
1.11 Crosstalk 22
1.11.1 Ideal crosstalk cancellation 24
1.12 Bilateral BCHA and limitations due to crosstalk 25
2 Pilot Study investigating the use of a scanning laser Doppler vibrometry of the cranium when stimulated by a B71 bone transducer 27
2.1 Summary 27
2.2 Introduction 27
2.3 Materials and Methods 29
2.3.1 Ethical Approval 29
2.3.2 Participant 29
2.3.3 Laser Vibrometers 29
2.3.4 Data collection 30
2.3.5 Data Analysis 31
2.4 Results 32
2.5 Discussion 34
2.5.1 Effects of frequency 34
2.5.2 Limitations 35
2.5.3 Potential Applications 36
2.6 Conclusion 37
3 Measurements of transcranial attenuation and phase of bone-conducted sound. 38
3.1 Summary 38
3.2 Introduction 38
3.3 Methods 41
3.3.1 Apparatus 41

3.3.2 Calibration 41
3.3.3 Exclusion and inclusion criteria 42
3.3.4 Participants 42
3.3.5 Testing procedure 42
3.3.6 Experiment 1 45
3.3.7 Experiment 2 46
3.4 Results 46
3.4.1 Experiment 1 data exclusion 46
3.4.2 Experiment 1 phase data 47
3.4.3 Experiment 1 Level Data 47
3.4.4 Transcranial Attenuation and Transcranial Delay 48
3.4.5 Experiment 2 Across-sitting variability 49
3.4.6 Experiment 2 Grade Data 51
3.5 Discussion 51
3.5.1 Cancellation Phase change with frequency 51
3.5.2 Cancellation Level changes with frequency 52
3.5.3 Test re-test standard deviation 52
3.5.4 Grading 53
3.5.5 Future research 54
3.6 Conclusion 55
4 Psychoacoustic measurement of phase and level necessary for cross-talk cancellation using bilateral bone transducers. 56
4.1 Summary 56
4.2 Introduction 56
4.3 Method 59
4.3.1 Apparatus 59
4.3.2 Participants 59
4.3.3 Testing procedure 59
4.3.4 Calculations 62
4.3.5 Data comparison methodology 64
4.4 Results 65
4.4.1 Number of attempts needed at each frequency 65
4.4.2 Level difference between techniques 65
4.4.3 Phase difference between techniques 66
4.5 Discussion 67
4.5.1 Ipsilateral level and phase 67

4.5.2 Contralateral level and phase 69
4.6 Conclusion 70
5 Unilateral Cross-talk cancellation in normal hearing participants using Bilateral Bone Transducers 71
5.1 Introduction 71
5.2 Method 73
5.2.1 Apparatus 73
5.2.2 Participants 74
5.2.3 Experiment A: Initial Phase and Level Prediction and data fitting 74
5.2.4 Experiment A: Secondary Phase and Level Prediction and data fitting in preparation for tone and speech reception thresholds 75
5.2.5 Experiment A: Cross-talk cancellation data collection method 76
5.2.6 Experiment B: Tone Reception Thresholds (TRT) with and without cross-talk cancellation. 77
5.2.6.1 Stimuli Creation 77
5.2.6.2 Procedure 78
5.2.7 Experiment C: Use of cross-talk cancellation results in order to improve speech reception thresholds. 78
5.2.7.1 Stimuli Creation 78
5.2.7.2 Procedure 78
5.3 Results 79
5.3.1 Experiment A results 79
5.3.2 Experiment B results 83
5.3.3 Experiment C 84
5.4 Discussion 85
5.4.1 Experiment A Cancellation results 85
5.4.2 Experiment B 86
5.4.3 Experiment C 86
5.5 Conclusion 87
6 The relative benefits of interaural time and level differences when sound is presented via air or bone conduction 88
6.1 Introduction 88
6.2 Method 90
6.2.1 Apparatus 90
6.2.2 Participants 90
6.2.3 Stimuli 91

6.2.4 Stimuli 91
6.2.5 Spatial release from masking 92
6.2.6 Prediction method 93
6.2.7 Data analysis 95
6.3 Results 96
6.4 Discussion 97
6.5 Conclusion 97
7 General Discussion and Future Work 98
7.1 The "Take-Home" Message 98
7.2 Summary of Findings 98
7.3 Opportunities for further research 99
7.3.1 Phase and level measurement 99
7.3.1.1 Data collection speed and ease 99
7.3.1.2 Real time bilateral cross-talk cancellation 100
7.3.2 Potential limitations of the technique 101
7.3.2.1 Processing limitations 101
7.3.2.2 Non-linearity 102
7.3.2.3 Intracranial pressure 102
7.3.3 Technological research 103
7.3.3.1 Audio streaming capability 103
7.3.3.2 BCHA orientation 104
7.3.4 Final Remarks 104
References 105

List of Figures

Figure 1 Schematic of the relative contributions of bone conduction hearing and how this can interact with air conduction. A box represents an anatomical part of the head. An intact line represents a transmission of energy. A long dashed line represents stimulation from of tactile sensors in the skin and bone. A dotted line represents nerve impulses. Reproduced from Stenfelt and Goode (2005). 3

Figure 2 Ear canal pressure of BC sound with an open and closed ear canal averaged over 9 ears. Error bars indicate the standard deviation from 61 measurements as reproduced from Stenfelt and Goode, (2005). 7

Figure 3 Panel depicts a sound source presented at 45° from the listener. Panel B shows the impulse response from the left (thin line) right (thick line). Panel C shows the spectra of the amplitude for the same sound source (Litovsky 2012) 9

Figure 4 (a) Image depicting lack of head shadow effect at low frequency (b) Showing head shadow at high frequencies (Figure reproduced from Shannan, (2010)). 10

Figure 5 Average sound localisation results from 12 participants with a single BCHA placed on the best and shadowed side as well as a bilateral fitting (reproduced from Priwin et al 2004) 22

Figure 6 Showing acoustic transfer functions between two loudspeakers and a participant (Reproduced from Liao, 2010) 22

Figure 7 Block diagram of crosstalk cancellation algorithm (reproduced from Liao, 2010) 24

Figure 8 Three PolytecTM PSV-500 vibrometers focusing on participants right temporoparietal region (contralateral to the B71 bone transducer). 30

Figure 9 (a) showing relative transverse displacement at 250 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (b) maximal contralateral displacement at temporoparietal region at 250 Hz (c) showing relative transverse

displacement at 500 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (d) maximal contralateral displacement at temporoparietal region at 500 Hz (e) showing relative transverse displacement at 1000 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (f) maximal contralateral displacement at temporoparietal region at 1000 Hz (g) showing relative transverse displacement at 2000 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (h) maximal contralateral displacement at temporoparietal region at 2000 Hz. 33

Figure 10 Visual representation of rigid-body motion (a) and mass-spring motion (b). Illustrations based on Stenfelt (2011)..... 35

Figure 11 Overview of primary sound pathways via bone conduction, adapted from (Stenfelt, 2011). 39

Figure 12 Illustration of the psychophysical procedure for cancelling a bone-conducted sound at both ears and providing an effectiveness rating. 43

Figure 13 The raw cancellation phase (unwrapped) and level results from ER2 earphones needed at different frequencies to cancel a single B71W BT tone in 3 participants. 47

Figure 14 Mean inter-cochlear phase and inter-cochlear level data, Zwislocki (1953) phase data (using a loud tone at one ear rather than a BT), and Stenfelt & Goode (2005a) inter-cochlear phase and inter-cochlear level difference (derived from accelerometer measurements of cadaver heads). 49

Figure 15 The standard deviation of the inter-trial cancellation level and phase for each of the participants. Results from 8 different testing sessions. 50

Figure 16 The mean cancellation grade (n=8) for each of the participants. Error bars show one standard deviation. 50

Figure 17 Panels (a) and (b) illustrates sound cancellation at the cochlea by interaction of the two BTs by destructive interference (black arrows). Panel (a) showing cancellation at the left cochlea and (b) at the right cochlea. The resultant signal of the two

BT signals (grey arrows) is then cancelled with ER2 earphones at the opposite ear. Panels (c) and (d) illustrate cancellation of sound at the cochlea from a left sided (c) and right sided (d) bone transducer using ER2 earphones. 58

Figure 18 Image of lens-less glasses with attached B71 bone transducers. 60

Figure 19 Illustrations of cross-talk cancellation modified from Rowan and Gray (2008) and Zurek (1986). (a) Model of cross talk cancellation using two BTs (see text for details). (b) Model of left-BT stimulation with cancellation at the left and right ear. (c) Model of right-BT stimulation with cancellation at the left and right ear. (d) Model of two-BT stimulation with cancellation at the left cochlea and the two signals interacting to give a phase and level at the contralateral (right) cochlea. 64

Figure 20 Difference between the predicted level and phase using the one- and two-BT techniques for each participant. Error bars show the standard deviation of the differences between the two techniques (n=6 per frequency result)..... 66

Figure 21 Outline of the experimental order 74

Figure 22 Showing two conditions a) shows pure tone on one BT and noise on the contralateral BT b) shows the addition of cancellation noise at the BT with the tone. .. 78

Figure 23 Showing the phase difference needed between bilaterally placed bone transducers to cancel perceived sound at the left and right cochlea on 5 different sittings in three different participants. Line of best fit created using spline fitting method..... 81

Figure 24 Showing the level difference needed between bilaterally placed bone transducers to cancel perceived sound at the left and right cochlea on 5 different sittings in three different participants. 82

Figure 25 Tone reception threshold with an without cross-talk cancellation in three participants (n=3 per condition) error bars show standard deviation. 83

Figure 26 Mean tone reception thresholds with and without cross-talk cancellation (n=9 per condition) error bars show standard deviation. 83

Figure 27 Box plot showing the SRT with and without the use of cross-talk cancellation in three participants. 84

Figure 28 HRIR impulse responses that were convolved with the speech sentences to simulate each condition. 92

Figure 29 Schematic of the Lavandier and Culling (2010) model. 94

Figure 30 SRM for noise at 0° vs. 90° (n=8). Error bars show standard deviation. Predicted values are produced using Lavandier and Culling, 2010 binaural speech intelligibility model..... 96

List of Tables

Table 1 Summary of study findings investing the use of BCHA for SSD treatment *Benefits only when noise front and speech lateralised to bad ear, †Benefits were found however HINT only performed with speech from deaf side, ‡Study was of SSD patients with contralateral hearing loss. HINT = Hearing in noise testing, SRT = Speech reception threshold, SL = Source localisation APHAB = Abbreviated Profile of hearing aid benefit, SSDQ = Single-sided deafness questionnaire, GHABP = Glasgow hearing aid benefit profile, SSQ = Speech, spatial and qualities of hearing scale, GBI = Glasgow benefit inventory 18

Table 2 Glasgow Benefit inventory from Janssen et al. (2012) 19

Table 3 Showing Tone and speech thresholds in quiet as adapted from Janssen et al., (2012) * Speech Reception Thresholds (SRT) 21

Table 4 Difference in tone thresholds in quiet between bilateral and unilateral BCHA as sourced from Janssen et al., 2012 21

Table 5 Grading system post attempted cancellation 45

List of Symbols

AC	Air Conduction	1
APHAB	Abbreviated Profile of hearing aid benefit	13
BAHA	Bone Anchored Hearing Aid	12
BC	Bone Conduction	1
BCHA	Bone Conduction Hearing Aids	7
BMLD	Binaural Masking Level Differences	9
BT	Bone Transducer	1
CI	Cochlear Implant	16
CROS	Contralateral Routing of Signals	7
DSP	Digital Singal Processor	11
EM	Energetic Masking	10
FIR	Finite Impulse Response	86
GBI	Glasgow benefit inventory	15
GHABP	Glasgow hearing aid benefit profile	15
HINT	Hearing in Noise Test	13
HRIR	Head Related Impulse Response	87
ICLD	Inter-Cochlear Level difference	7
ICP	Intracranial Pressure	51
ICPD	Inter-Cochlear Phase Difference	7
ILD	Interaural Level difference	7
IM	Informational Masking	10
IPD	Interaural Phase Difference	7
ITD	Interaural Time Difference	9
JND	Just Noticeable Difference	84
LDV	laser Doppler vibrometer	23

MAA	Minimum Audible Angle	84
MRI	Magnetic Reasonance Imaging	22
PTA	Pure Tone Audiogram	17
QOL	Quality of Life	16
SD	Standard Deviation	42
SL	Source Localisation	15
SNR	Signal to Noise Ratio	9
SRM	Spatial Release from Masking	10
SRT	Speech Reception Threshold	9
SSD	Single Sided Deafness	7
SSQ	Speech, spatial and qualities of hearing scale	15
TA	Transcranial Attenuation	7
TM	Tympanic Membrane	5
TRT	Tone Reception Thresholds	72

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Summary

Bone-conduction hearing aids (BCHAs) are a widely used method of treating conductive hearing loss, but the benefit of bilateral implantation is severely limited due to interaural cross-talk. In theory two BCHAs could deliver improved stereo separation using cross-talk cancellation. Sound vibrations from each BCHA would be cancelled at the contralateral cochlea by an out-of-phase signal of the same level from the ipsilateral BCHA.

In order to achieve this the phase and level of sound at each cochlea needs to be known. A method to measure the level and phase required for these cancellation signals was developed and cross-validated with a second technique that combines air- and bone-conducted sound in normal hearing subjects. Levels measured with each method differed by <1 dB between 3-5 kHz. The phase results also corresponded well for the cancelled ear (11° mean difference). The newly developed method using only bone transducers is potentially transferable to a clinical population.

To demonstrate cross-talk cancellation tone and speech reception thresholds (TRT and SRT) were investigated with and without unilateral cross-talk cancellation. Band limited noise was emitted from one BT whilst signal +/- cancellation signal was produced by the other. Benefits of cross-talk cancellation under this atypical listening situation were found to be 12.08 and 13.7 dB for TRT and SRT thresholds.

In order to estimate the potential benefits of cross-talk cancellation in spatially realistic environments, phase and level elements of impulse responses from a BAHA 4 were convolved with speech. This found that cross-talk cancellation had the potential to lower SRTs in a clinical population by approximately 4.4 dB. Future work will focus on real-time processing and examine using a clinical population.

1 Introduction

1.1 Bone Conduction

Bone conduction (BC) is broadly defined as the process by which vibration of skull bones result in a sound percept (Stenfelt, 2013). However, this description can be misleading, since BC sound also involves the transmission of vibrations through cartilage and soft tissue. For example, when a Bone Transducer (BT) is placed on the mastoid over intact skin for audiological testing, the vibration also has to pass through that skin. Stenfelt, (2013) states that transmission may be more accurately referred to as "body conduction" as this would take account of the addition of soft tissues.

The concept of BC has been known for several centuries (Stenfelt & Håkansson 2002). It was utilised in the Weber and Rinne test in the nineteenth century in order to distinguish between a sensorineural and conductive hearing loss (Turner 1990). These tests are still taught within medical school and utilised when formal audiological assessment is unavailable. The Weber test uses a tuning fork placed firmly on the centre of the forehead whilst vibrating. Patients with conductive hearing impairment such as otitis media with effusion (glue ear), experience lateralisation towards the affected ear, whilst those with a sensorineural defect experience lateralisation away from the affected ear. Although the pathological ear is generally known, often previous hearing loss is not. Thus, if lateralisation was experienced, then the Rinne test is also required in order to differentiate pathology at either ear, since lateralisation towards one ear could indicate a conductive loss at that ear or a sensorineural loss at the contralateral ear (Turner 1990). This is performed by placing a tuning fork firmly on both mastoids and comparing the level experienced when the fork is held 2.5 cm away from the ear canal. If the level experienced was greater when placed on the mastoid as well as the Weber test lateralising towards the ear then a conductive loss can be diagnosed. When these tests were devised the mechanism by which they stimulated the cochlea was not understood.

Von Békésy was the first to address the fundamental question of BC hearing (von Békésy 1960). He hypothesised that the basilar membrane could be stimulated by either Air Conduction (AC), which is to say through the movement of the tympanic membrane, or by BC. That is, that both sensations resulted from stimulation of the cochlea. If this hypothesis was correct, then he reasoned it would be possible to cancel a BC tone with an out-of-phase AC tone of the same amplitude so that they did not hear anything in the target ear. He was able to achieve this in one subject at 400 Hz (Stenfelt & Håkansson

2002). This approach of cancelling one sound with another is used extensively in this thesis.

1.2 Mechanisms of Bone Conduction

Following von Békésy's studies, research was driven primarily by the growth of otological surgery as well as the need to be able to predict success rates of surgery for conditions such as tympanic membrane (ear drum) perforations and destructive middle ear conditions such as cholesteatoma. Early studies by Wever and Lawrence (1954) and von Békésy indicated the primary method of energy transfer was via an osseous route due to a transverse waves being conducted along the skull. Later, Allen and Fernandez (1960) theorised (incorrectly) that BC sound was entirely due to the vibrations in the cochlear fluids and that the middle ear ossicles (malleolus, incus and stapes) were not involved. Soon after, Tonndorf and Tabor (1962) showed energy may also be conducted via soft tissues as a pressure wave before being transferred to an osseous or fluid route. Brinkman et al. (1965) suggested that the inertia of the cochlear fluid and ossicular chain could also affect BC thresholds. Since then it has become clear that there is no one mechanism which allows BC transmission. Instead, it is comprised of a number of pathways each having a differential effect depending on frequency, BT position and anatomical differences (Stenfelt & Goode 2005a). One method of categorising the different contributory factors is by anatomical vectors such as the outer ear canal, middle ear ossicles and inner ear fluid (Stenfelt & Goode 2005a). These vectors can be involved in multiple conduction routes (as shown in Figure 1) the different vectors and their associated conduction routes will be discussed in turn.

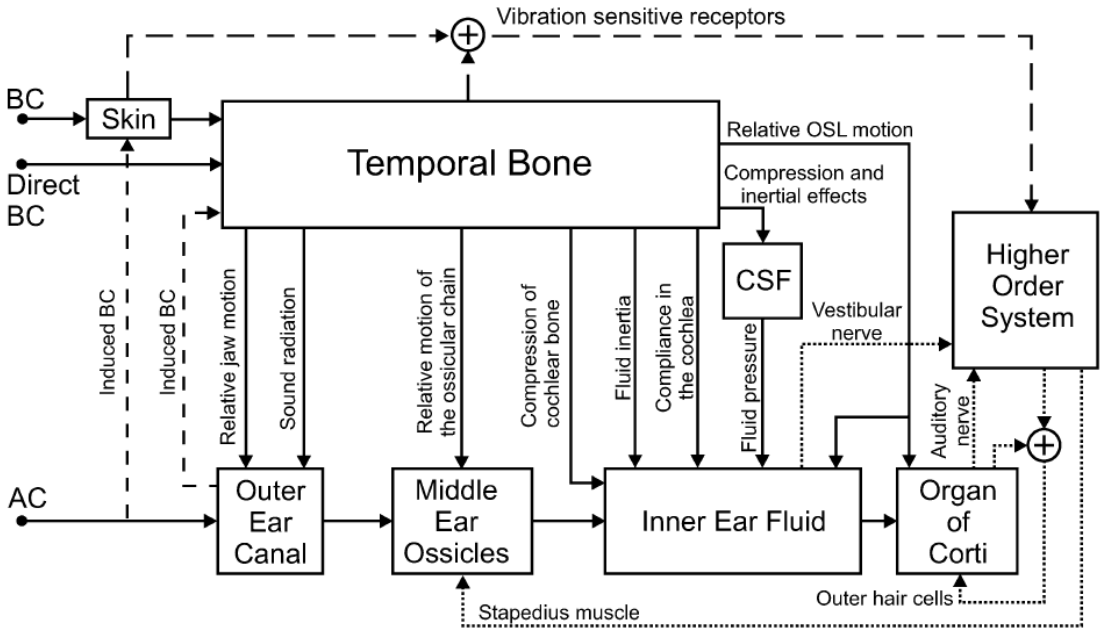


Figure 1 Schematic of the relative contributions of bone conduction hearing and how this can interact with air conduction. A box represents an anatomical part of the head. An intact line represents a transmission of energy. A long dashed line represents stimulation from of tactile sensors in the skin and bone. A dotted line represents nerve impulses. Reproduced from Stenfelt and Goode (2005).

1.2.1 Inner ear fluid

The cochlear is divided along its length into three fluid compartments: the scala media, scala vestibuli and scala tympani. The scala tympani and scala vestibuli are separated by the scala media throughout most of the length of the cochlea, but are joined at the apex by a small opening called the helicotrema. At the base of the scala tympani is the round window and at the base of the scala vestibuli is the oval window where the stapes is attached. The two windows are needed within a bony labyrinth in order to allow fluid movement. This happens reciprocally, i.e. if the oval window is compressed via the stapes then it causes the round window to bulge. The fluid movement in the cochlear generates a travelling wave along the scala media within which the organ of Corti is located. This is the receptor organ of hearing and is tonotopically organised. Receptors on the organ of Corti close to the oval window (at the base) are stimulated by high frequency sounds whilst receptors closer to the helicotrema (at the apex) of the cochlear are stimulated by low frequency sounds.

Although pressure waves are primarily transferred via the reciprocal movement of the oval and round windows there are a number of other pathways by which fluid pressure waves may be introduced into the cochlear. These are collectively referred to as third

window structures (Stenfelt & Goode 2005a). These third window pathways include the vestibular and cochlear aqueducts as well as veins and nerve fibres (Küçük et al. 1991). Tonndorf et al.'s (1966), investigated these pathways in cats finding that the cochlear aqueduct acts as a third window in the transmission of BC but not for AC signals.

The importance of third window structures as well as cochlear fluid movement has been aided by research into otosclerosis. This is an inherited disease in which there are areas of bony remodelling that primarily affect the stapes footplate, impairing its movement. The round window can also be affected, thus further impairing the mechanism of AC conduction. These two pathologies can cause a significant deterioration in AC hearing thresholds by up to 60 dB (García-Iza et al. 2016). BC thresholds meanwhile are relatively unaffected with maximal loss usually diagnosed of 15 to 25 dB at 2 kHz. Since there is little or no transmission from the outer or middle ear to the cochlear this means that the BC stimulation must primarily act directly on the cochlear (Stenfelt & Goode 2005a).

A further route by which the inner ear fluid can be affected is by compression and expansion of the cochlear walls. When the cochlear is compressed it causes the round and oval windows to bulge. Since the round window is 20 times more compliant than the oval window the round window is the main point of displacement (Kirikae 1959). This method of moving inner ear fluid is different to that of AC where the primary driving force is from the stapes footplate at the oval window. Stimulation of the organ of Corti is also aided by differences in volume ratio of 5:3 between the scala vestibuli and scala tympani (Tonndorf et al. 1966). This results in more fluid being displaced from the scala vestibuli than tympani, thus aiding the stimulation of hair cells in the organ of Corti.

1.2.2 Middle ear

The ossicles, which consist of the malleus, incus and stapes, are suspended in the middle ear cavity by ligaments. In addition to the ligamentous attachments there are also muscular ones. These comprise of the tensor tympani, which attaches to the malleus, and the stapedius muscle, which attaches to the stapes. These muscles function, firstly, to protect the cochlear from loud sounds by contracting and thus limiting the movement of the ossicles, and secondly, to reduce perceived level of sounds such as an individual's own voice as well as actions such as chewing (Bance et al. 2013).

At low frequencies the ossicles move in phase with the skull vibrations, so there is no relative motion (Stenfelt & Goode 2005a). However, at higher frequencies the inertia of the ossicles becomes greater than the stiffness of the ligaments and muscles attached to them. This causes the ossicles to move independently from the skull, and induces motion at the stapes footplate. Displacement of the footplate becomes very complex, primarily due the axis of rotation of each of the ossicles being different. At some frequencies the ossicles can vibrate in phase thus constructively interfering and increasing perceived level. In addition, the ossicles and the vibrations of the skull can vibrate in phase further increasing the level at the cochlear (Stenfelt et al. 2000). This complex relationship means that the ossicles can have a significant impact on level which is highly BT-placement and frequency specific. This is potentially one of the reasons why there are large differences in BC threshold between individuals (Walker et al. 2005).

1.2.3 The outer ear

During BC the skull vibrates producing a relative motion when compared to the surrounding air. This causes the air in the outer ear canal to produce a pressure wave, both due to the deformation of the canal walls as well as movement of the skull as a whole. This wave can act on the tympanic membrane and produces a sound percept via the same route as AC sound. Jaw motion has a potential impact upon this radiated sound. This is due to the close proximity of the tempomandibular (Jaw) joint to the auditory canal. When the jaw moves, the elasticity of the ear canal tissue and canal shape is modified, potentially changing the acoustic properties of the canal (Pinto 1962). Cadaveric studies removing cartilaginous parts of the ear canal have found that it is the outer third of the ear canal (made of soft tissues) that are primarily responsible for sound radiation rather than the inner bony two-thirds of the canal (Stenfelt et al. 2003; Tonndorf et al. 1966).

The relative contribution of the outer ear canal varies significantly with frequency as well as with occlusion of the ear canal (such as when using earphones for testing). For

frequencies below the first resonance of the skull (approximately 0.7 kHz), the skull moves as a whole. This results in no radiated sound at these frequencies in an open ear canal (Håkansson et al. 1994). Open ear studies comprising of patients with conductive hearing loss due to otosclerosis have supported these findings identifying no change in BC threshold at low frequencies, whilst up to 60 dB loss in AC threshold (Bagger-Sjöbäck et al. 2015). Since patients with otosclerosis have a fixed stapes it is not possible for sound radiated from the ear canal to be transferred to the cochlea. This indicates low-frequency BC sounds are not transferred via canal radiation.

However, the relative contributions of BC hearing are very different when the ear canal is occluded. During ear canal occlusion sound radiation becomes the major contributor to BC hearing (Stenfelt & Reinfeldt 2007). This is particularly clear at frequencies below 1 kHz as shown by Stenfelt et al. (2003). This is also supported by Khanna et al. (1976) using a closed ear canal, who found that, when a low-frequency BC tone is subjectively cancelled by an earphone AC tone, there is a simultaneous significant drop in sound level measured in the ear canal. This indicates destructive interference is occurring both in the ear canal as well as at the cochlea, suggesting that at low frequencies the majority of the contribution of BC sound is comprised of radiated sound from the ear canal when the ear canal is closed.

1.2.3.1 Occlusion effect and the outer ear

In order to explain the occlusion effect Tonndorf et al., (1966) theorised that the inertia and compliance of the air in the ear canal and tympanic membrane (TM) produce a high pass filter. When the canal is closed this filter no longer functions resulting in increased low-frequency sound. An alternative theory of the occlusion effect was given by Huizing, (1960) who suggested that the effect was due to a change in resonance properties of the ear canal when it is open and closed. Much akin to a change in resonance when tube is open or closed. Tonndorf's theory was later found to be true at low frequencies where the mass effect and compliance of the ear canal is critical in determining the acoustic properties. Whilst Huizing's theory holds true at high frequencies (above 2 kHz) where the resonance and antiresonance properties are key (Stenfelt & Goode 2005a). Figure 2 shows the impact of an open and closed ear on sound pressure level.

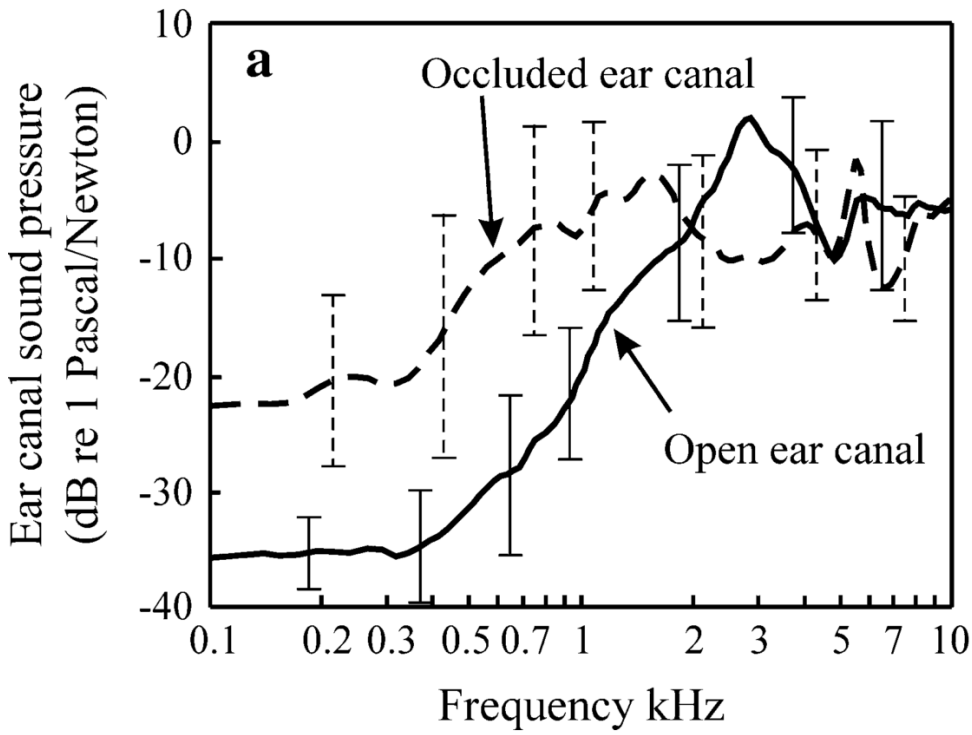


Figure 2 Ear canal pressure of BC sound with an open and closed ear canal averaged over 9 ears. Error bars indicate the standard deviation from 61 measurements as reproduced from Stenfelt and Goode, (2005).

1.3 Summary of relative contributions

Overall the inertia of fluid in the inner ear is the most important factor in BC hearing, particularly at low frequencies (in the normal ear). However, it becomes less of a factor at higher frequencies. The inertia of the middle ear is also of importance for BC hearing at mid-frequencies. At higher frequencies the compression and expansion of the cochlear is likely an important contributory factor. Whilst sound radiated from an open outer ear canal has a limited contribution toward BC hearing in an open ear, in an occluded ear, radiated sound from the outer canal dominates BC hearing at frequencies of 1 kHz or less. The transmission of pressure waves via third window contact of cerebrospinal fluid is thought to have a minimal effect on BC thresholds at all normal hearing frequencies (Stenfelt & Goode 2005a). All the main contributory factors are outlined in Figure 1.

1.4 Transcranial attenuation and interaural level difference

In addition to the complex interaction of the outer, middle and inner ear that occur at the ipsilateral cochlea there is also a similar interaction at the contralateral cochlea.

However, due to the additional distance the sound waves travel, there is a degree of attenuation, often referred to as Transcranial Attenuation (TA). Stenfelt, (2012) defined TA as "the threshold difference between an ipsilaterally transmitted and contralaterally transmitted BC sound when the stimulation is at a similar position at the two sides of the cranium." There is a high degree of variability of TA with frequency as well as between participants (Nolan & Lyon 1981). Despite great variation mean TA over 28 subjects found that at 0.5 kHz TA was approximately 4 dB, at 0.5-1.8 kHz attenuation was close to 0 dB and at frequencies between 3-5 kHz attenuation was 10 dB.

The relatively low TA means that Bone Conduction Hearing Aids (BCHA) have been widely used for patients with Single Sided Deafness (SSD) instead of a CROS aid (which mixes sound from microphones on both sides of the head and presents them to the patient's one good ear). However, the relatively low attenuation which makes BCHAs an ideal method for a CROS aid replacement causes significant problems when patients have two functioning cochleae. This is because the crosstalk (defined by the signal reaching the contralateral cochlea) limits signal separation from the left and right side. In those with bilateral BCHAs and two functioning cochleae, signal transmitted from the left side would combine at each cochlea with that from the right side. Mixing happens at both cochleae but to differing amounts at each frequency and destroys many of the interaural level and phase difference (ILD, IPD) cues which would normally provide many of the signal processing benefits that binaural hearing provides. The underlying mechanisms of binaural processing are outlined below.

1.5 Advantages of hearing with two ears

The major advantages of binaural hearing are primarily related to sound localisation and to the detection and identification of sounds when there are competing sources. The latter benefit can be subdivided into several distinct effects/processes:

- Head shadow effect
- Binaural summation
- Binaural squelch
- Binaural unmasking

When speech and noise are presented from different directions, the head-shadow and unmasking processes operate in concert to improve intelligibility over a situation in which they both lie in the same direction (Hawley et al., 2004).

1.5.1 Sound localisation

Sound localisation can both exploit monaural information from each individual ear (Litovsky 2012) and it can compare the sound at both ears to give binaural cues. Figure 3 shows how a click from 45° to the left of the listener in the horizontal plane (panel A) appears at the two ears (panels B and C).

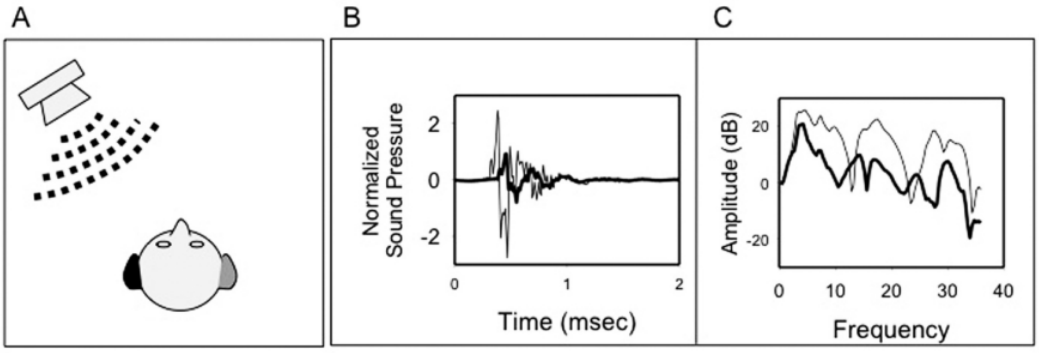


Figure 3 Panel depicts a sound source presented at 45° from the listener. Panel B shows the impulse response from the left (thin line) right (thick line). Panel C shows the spectra of the amplitude for the same sound source (Litovsky 2012)

Since the source is not equidistance from both ears (i.e. not 0° or 180°) there is a time difference between the ears. These binaural clues are known as interaural time differences (ITD) which can also be represented as IPD. The head also acts as an acoustic obstacle to sound, resulting in interaural level differences (ILD). For nearby sounds, the difference in distance to the source can also contribute to ILDs. Although the auditory system uses both cues in order to predict the location of sound, it is thought that ITD information is dominant in estimating direction over ILD (Witghman and Kistler,1998). The auditory system is particularly sensitive to ITD cues below 1.5 kHz whilst above this, ITD cues are of less use. ILD cues are of particular use above frequencies of 4 kHz. This results in relatively poor sound localisation between pure tones of 2-4 kHz as both ITD and ILD cues are relatively limited.

Although binaural cues give accurate discrimination between left and right, they do not give information about the elevation of a sound source, nor about whether the sound is from in front or behind. Movement of the head during presentation can help discriminate front from back (Perrett & Noble 1997), although this process is much slower than normal localisation, taking approximately 0.5 seconds.

Pinna cues can aid sound localisation by assisting elevation and front/back discrimination. The pinna limits sound from behind the listener which results in a level reduction of approximately 2-3 dB at frequencies above 2 kHz (Middlebrooks & Green

1991). Not only this, but sound reflections from the pinna and ear canal, which are dependent on the angle of sound, result in spectral changes which can aid in localisation as shown in Figure 3, panel C (Musicant & Butler 1984).

1.5.2 Head shadow effect

The head shadow effect describes the reduction in level between ipsilateral and contralateral ears when sound is presented from one side. However, there are additional effects other than a purely level modification. These include a change in spectral pattern and a change in phase, both caused by diffraction of the sound wave around the head.

High frequency sounds are predominately affected by the head shadow effect. This is due to the relative wavelength size when compared to the head. This can result in a 15 dB ILD at high frequencies whilst at low frequencies the ILD can be negligible (as demonstrated in Figure 4).

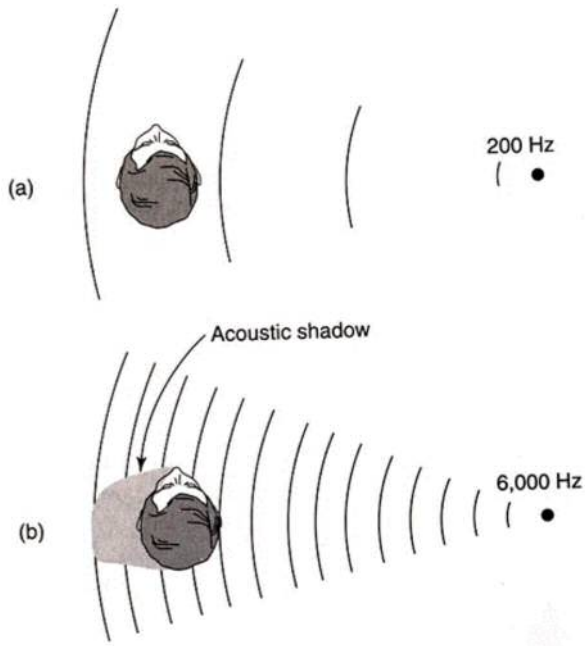


Figure 4 (a) Image depicting lack of head shadow effect at low frequency (b) Showing head shadow at high frequencies (Figure reproduced from Shannan, (2010)).

1.5.3 Binaural Summation

Binaural summation is the effect of improved speech reception thresholds (SRT) in noise when the same sound is presented to the two ears rather than just to one. Comparisons of SRT using earphones with monaural and binaurally presented speech found that there was a mean 5% improvement when sound was presented binaurally (Lavandier and Culling, 2008).

1.5.4 Binaural Squelch

A further phenomenon called binaural squelch is related to the central processing ability of the brain to be able to use ILD and ITD signals as already outlined above. However, instead of these signals being purely used to calculate directionality of sound it can also be used in to separate noise and target sources (Carhart 1965). This is achieved by central processing of the target and noise (Gray et al. 2009). Specifically "binaural squelch describes an improvement in speech intelligibility in noise due to the addition of a second acoustic input at the contralateral ear with a poorer SNR that in the first ear "(Senn et al. 2005; Gray et al. 2009). In normally hearing listeners, binaural squelch is probably largely caused by binaural unmasking.

1.5.5 Binaural unmasking

This describes the improvement in threshold of a signal in noise when either the level or phase difference of the signal at the two ears are not the same as the noise signal. A common experimental format includes finding the threshold when noise and signal are presented without a phase difference between the two ears. The signal is then presented out of phase between the ears (π-radian phase difference) whilst the noise remains unchanged. The change in threshold that results, the binaural masking level difference (BMLD), can be in the region of 15 dB when compared to the unchanged result. Similarly, when the phase of the noise is changed then there is also significant improvement (approximately 13 dB) although the change is not as great as that of the signal modification (Moore 2003).

1.6 Spatial release from masking

Spatial release from masking (SRM) is broadly defined as improvement in the ability to understand speech in noise, when the sound sources come from different directions, rather than the same direction. In addition to understanding the fundamental mechanisms in normal hearing individuals, the SRM has been researched specifically in those with hearing impairment, including those with conductive hearing loss. SRM is generally quantified in two ways. In the first method, the Signal to Noise Ratio (SNR) is varied adaptively to a pre-set level of speech reception, usually for 50% words correct, and the adapted level is defined as the speech reception threshold (SRT). The second method uses the percentage correct at a fixed SNR in order to assess SRM. A series of fixed SNRs can also be presented to give the full psychometric function. A common experimental setup includes finding the difference in thresholds when target speech and a masker are co-located from the front and comparing the relative difference in threshold

when the target speech is moved (commonly 90°). This can result in an improvement in threshold of up to 12 dB (Bronkhorst & Plomp 1988; Litovsky 2005). SRM may reflect the release of at least two different types of masking: energetic and informational.

1.6.1 Energetic masking

Energetic masking (EM) is used to describe the reduction in audibility which occurs to the target signal when the masking signal compete for representation over the same peripheral neural units (Best et al. 2011). Signals can overlap peripherally both in time and frequency. When the target and masker are separated more than half of the release of EM comes from "better ear listening". This occurs due to attenuation of the masker caused by the listeners head, this results in one ear having an improved SNR compared to the other The rest is attributable to binaural unmasking (Zurek 1993; Jelfs et al. 2011).

1.6.2 Informational masking

Informational masking (IM) is used to describe the interference that the masker introduces which cannot easily be explained via peripheral auditory processing. This is primarily associated with confusion between masker and target or the reduction in ability to be able to focus attention on the target sound (Best et al. 2011). IM can be influenced by factors such as the similarity of competing voices and uncertainty on where to direct attention during speech (Kidd et al. 2005; Brungart et al. 2001). For example, if the target was a male voice with a fundamental frequency of 125 Hz and masker was a female voice with a fundamental frequency of 250 Hz then the SRT might be better than if both target and masker were both males. Similarly, if the target and masker had similar speech materials or carry similar meaning the task becomes more difficult to perform.

The extent of masking and the relative weighting of each of the EM and IM is dependent on both target and masker. For example, speech mixtures will vary in how they overlap in frequency and time, thus impacting upon the amount of EM that is present (Best et al. 2011).

1.7 Hearing loss

Hearing loss is an extremely common impairment, particularly in older adults. Studies have found the prevalence of hearing loss (defined as >25 dB at 0.5-4 kHz) in the better ear to be 47% in people between 57-89 years old (Moscicki et al. 1985). In a paediatric population, conductive hearing loss makes up approximately 4% of the total diagnoses of hearing loss (excluding otitis media), with an additional 5% of diagnoses comprised of mixed sensorineural/conductive losses (Parving 1983). It is these two groups of patients that are most likely to benefit from a Bone Conduction Hearing Aid (BCHA). This is of great importance since patients with untreated hearing loss have been shown to report higher rates of anxiety, depression as well as being less likely to participate in organised social activities when compared to those people who wear hearing aids (Hagr 2007; Kochkin & Rogin 2000; Seniors Research Group 1999). Hearing aid use has been shown to significantly improve the 'health triangle' of social, physical and mental health in those with all degrees of hearing loss (Kochkin & Rogin 2000; Seniors Research Group 1999). However, traditional AC hearing aids are of limited effectiveness in those with significant conductive loss. Air-borne hearing aids are also inappropriate in some patients as they can cause recurrent attacks of otitis externa, so the use of a BCHA is one potential option when aiming to treat such patients effectively.

1.8 Bone Conduction Hearing Aid

Brånemark's (1960) discovery that titanium could be reliably osseointergrated was crucial in the development of dental implants, and also spawned the development of the majority of modern BCHAs (Brånemark et al. 1977). In 1977, a collaboration made up of Sahlgrenska University Hospital, Chalmers University of Technology, and Brånemark Osseointegration Center started the design of the first BCHA (Håkansson et al. 1985). This design consisted of a titanium screw, percutaneous abutment and transducer/sound processor. It had many advantages over its predecessors, including improved transmission of sound (particularly high frequencies) and elimination of discomfort occurring due to pressure on the skin via Softband devices (von Békésy 1960). Between 1985 (when the first 10 cases were reported by Håkansson (Håkansson et al. 1985)) and 2014, there have been approximately 150,000 patients treated with osseointergrated devices, (Reinfeldt et al. 2015) with 6000-7000 of those performed within the United Kingdom (Colquitt et al. 2011).

1.8.1 Device Components

The 'classical' direct drive devices are broadly composed of three main parts. Firstly, the titanium implant (3-4mm in length) which is usually placed 55 mm behind the pinna into the mastoid, where it osseointergrates. Secondly, an abutment which connects to the implant and protrudes through the skin in order to connect the sound processor and transfers vibrations. Lastly, the sound processor comprises of a microphone, digital signal processor (DSP), battery and force transducer. The processor can be detached from the abutment when not in use by the patient. In some cases, where strong amplification is required the microphone and transducer are separated so that the microphone is body worn. This allows a more powerful transducer to be placed on the abutment without feedback.

In addition to these core components modern bone-conduction hearing implants commonly have the ability to connect to other devices in order to adjust settings and stream sound from a device. One such example is the BAHA 5's ability to connect to Apple devices via 2.4 GHz wireless technology licenced from GN ReSound (AudiologyOnline 2011).

1.8.2 Clinical Indications

When compared to traditional hearing aids BCHAs are a relatively expensive method of restoring hearing. Thus, there must be good evidence for benefits in the quality of life in selected patients (Arunachalam et al. 2001). Broadly, current indication criteria include:

- Congenital malformations of the middle or external ear
- Chronically discharging ears or ears that prevent the use of traditional hearing aids such as those which are highly susceptible to otitis externa (Hagr 2007)
- Patients with bilateral conductive hearing loss due to ossicular disease (not surgically correctable) and not appropriate for traditional hearing devices (Berger 1976)

In addition to the traditional indications related to conductive hearing loss patients are also being fitted with a BCHA for single sided deafness (Pfiffner et al. 2011). Chronic suppurative otitis media and recurrent otitis externa are the most common indication for BCHA fitting in adults, as they make the use of traditional hearing aids impractical (Priwin & Granström 2005).

1.8.3 Clinical indication for Children

The majority of the paediatric population fitted with a BCHA have a conductive hearing loss related to craniofacial malformations (Priwin & Granström 2005), with the most common indication due to Treacher Collins syndrome, where patients can suffer from aural atresia or purely abnormalities in the ossicular chain (Yellon 2007; Papsin et al. 1997). BCHAs within this group have been shown to be beneficial in a number of different areas including the facilitation of normal language development (Verhagen et al. 2008), improvements of >40 dB in pure tone audiometry (Nicholson et al. 2011), and advantages in aesthetic outcomes (Granström & Tjellström 1997). They also deliver more reliable audiological outcomes when compared to reconstructive surgery with can be challenging (Marres et al. 1995). The major limiting factor is the relative thickness of the paediatric skull which means shorter screw lengths are required. In addition to this challenge is that patients who require such an intervention often have abnormal skull contour or suboptimal bone quality which increases failure of osseointergration rates up to 15% (Granstrom 2000; Tjellström & Granström 1995; Granström & Tjellström 1997). Thus while it has been suggested that children as young as 3 can be fitted with a BCHA (Colquitt et al. 2011) the devices are indicated for children aged ≥5 (Spitzer et al. 2002; Davids et al. 2007). In addition to treatment of congenital atresia, children with Down's syndrome can also be successfully treated with a BCHA when traditional hearing aids have failed (Sheehan & Hans 2006).

1.9 BCHA in single sided deafness

There are thought to be approximately 9,000 new diagnoses of Single Sided Deafness (SSD) in the UK each year, with 24 % of sufferers having to give up work as a result (Dimmelow et al. 2003). Commonly reported issues from SSD patients include: social exclusion, associated with difficulties keeping up with group conversations; dangers as a pedestrian; as well as increased problems at the workplace (Dimmelow et al. 2003). Since sufferers have only one working cochlea they cannot process any interaural cues and so lack the ability to derive any of the normal benefits of binaural processing, such as improved sound localisation and understanding of speech in background noise (Wazen et al. 2003). Treatment options of SSD include education related to the best sitting positions in noisy environments to make the most use of monaural cues as possible. Hearing devices are also available in the form of a CROS (Contralateral Routing of Signal) hearing aid and a BCHA. Several countries have also approved cochlear implantation as a treatment option (Agterberg et al. 2014).

In 2002 the FDA approved the use of BCHA for SSD. Since then, there has been an increasing numbers of patients being treated with this method (Pai et al. 2012). Although binaural hearing cannot be restored, the BCHA makes use of the relatively limited transcranial attenuation to allow sound from the deaf side to be transferred to the contralateral side. This aims to improve the detections of sound when the sound is laterally projected to the deaf side (Bovo et al. 2011).

Patients with congenital SSD are an important subgroup, which are at greater risk of learning problems at school when compared to those with normal hearing (Lieu 2013). Prevalence rates of congenital SSD are estimated to be in the region of 0.1 % (Schmithorst et al. 2014). It was theorised by Colburn, (1982) that patients with congenital SSD may not benefit as much from treatment with either BCHA or CROS, because such patients have always relied on a monaural strategy of processing pinna-induced spectral-shape cues. However no evidence exists to support this, with Agterberg et al., (2014) finding that there was no difference in localisation ability between those with congenital or acquired SSD. Instead Agterberg et al. found that localisation ability was dependent on high-frequency cues.

Despite much published literature, evidence for the benefit of BCHA in SSD is inconsistent with most subjective studies showing benefit, while most objective studies do not. A recent review concluded that there was a lack of high level evidence for both BCHA and CROS (Peters et al. 2015). A summary of both subjective and objective results is presented in Table 1.

In the subjective studies, measures such as the APHAB have been widely used. Studies consistently found significant improvements in ease of communication and in the background noise domains. In contrast, scores tended to be poorer in the assessment of aversiveness to loud noise (Linstrom et al. 2009; Yuen et al. 2009; Dumper et al. 2009; Gluth et al. 2010; Hol et al. 2010; Lin et al. 2006). The single-sided deafness questionnaire is a further commonly used assessment tool which found that the majority of patients use their devices between 4-7 hours a day and that there was a perceived benefit in quality of life and in hearing (particularly in quiet) (Hol et al. 2010; Gluth et al. 2010; Linstrom et al. 2009).

However, several objective studies have investigated BCHA via the use of the hearing in noise testing (HINT), the majority of which have found benefits only when speech is directed to the deaf ear. In this experimental configuration there is an improvement in thresholds due to compensation of the head shadow effect. However in

the converse condition where speech is directed to the hearing ear and noise the deaf ear, the hearing threshold worsens due to head shadow being compensated for (Martin et al. 2010; Lin et al. 2006; Linstrom et al. 2009; Dumper et al. 2009; Yuen et al. 2009). Thus, overall there is no benefit. Lin et al. 2006 reported that the increases in the beneficial condition were greater than the decreases in the detrimental condition, but these differences were not significant. One study of twenty-one patients with SSD and contralateral hearing loss did find significant improvements in speech recognition in both quiet and noise. Since the benefit in noise occurred regardless of a spatial separation between speech and noise, this benefit must be related to improvement in the stimulus audibility, and so might not be sustained if the stimulus were presented at a higher sound level.

Future devices may be able to improve HINT in BCHA users via communication with an additional microphone on the normal hearing side. Scene classification technology, currently used to benefit cochlear implant users (Wolfe et al. 2015) could then adaptively turn the BCHA on and off depending how favourable the signal to noise ratio is on the deaf side.

Authors	Objective Measures			Subjective Measures					Study Conclusions
	HINT	SRT	SL	APHAB	SSDQ	GHABP	SSQ	GBI	
Lin et al. (2006)	×*		×	✓					Benefit
Linstrom et al. (2009)	×*			✓	✓				Benefit
Yuen et al. (2009)	×†			✓		✓			Benefit
Arndt et al. (2011)			×				×		No benefit
Dumper et al. (2009)	×			✓			×		Limited benefit
Martin et al. (2010)	×						×		No benefit
Wazen et al. (2010)	✓‡							✓	Benefit
Gluth et al. (2010)				✓	✓	✓			Benefit
Hol et al. (2010)			×	✓	✓		✓		Benefit
Bovo et al. (2011)		×							No benefit
Pai et al. (2012)							✓		Benefit

Table 1 Summary of study findings investing the use of BCHA for SSD treatment *Benefits only when noise front and speech lateralised to bad ear, †Benefits were found however HINT only performed with speech from deaf side, ‡Study was of SSD patients with contralateral hearing loss. HINT = Hearing in noise testing, SRT = Speech reception threshold, SL = Source localisation APHAB = Abbreviated Profile of hearing aid benefit, SSDQ = Single-sided deafness questionnaire, GHABP = Glasgow hearing aid benefit profile, SSQ = Speech, spatial and qualities of hearing scale, GBI = Glasgow benefit inventory

1.10 Bilateral BCHAs

There are relatively few studies investigating the benefits of bilateral BCHAs. A literature review by Janssen et al (2012) identified eleven relevant studies. These studies' findings regarding subjective outcomes, listening preferences and audiometric measurements are outlined below.

1.10.1 Subjective Outcomes

There are several significant benefits identified using different subjective quality of life measures after the fitting of bilateral BCHAs. Dun et al., (2010) used the Glasgow benefit inventory (GBI), finding that there were positive scores in all sections of the Glasgow children's inventory. The greatest subjective gains were in the emotion and learning domains. Ho et al., (2009) also used the GBI in 93 bilateral-BCHA patients, finding benefit scores were higher than those fitted with a unilateral BCHA and that scores were close to those seen after cochlear implantation. Dutt et al., (2002) compared the benefit in 15 patients after fitting of a second BCHA. Assessment was via a number of different methods including the GBI which found that there was a significant increase in quality of life (QOL) in patients after fitting with a unilateral BCHA and that QOL further increased with bilateral fitting but the benefit was not so great as those seen after the first fitting. Glasgow benefit inventory results are shown in Table 2.

Study	No	Questionnaire Results of Bilateral Bone- Anchored Hearing Aid (BAHA)
Dun et al. (2010)	20	Glasgow Children's Benefit Inventory
		Overall score +38 (SD 18.2)
		Learning score: +56 (SD 25.4)
		Emotional score +45 (SD 24.4)
		Physical score +17 (SD 12.5)
		Vitality score +32 (SD 18.1)
Ho et al. (2009)	71	Glasgow Benefit Inventory
		Overall benefit score +38 (95% CI, 33-44)
		General domain +50 (95% CI, 43-57)
		Physical health +18 (95% CI, 11-25)
		Social support +14 (95% CI, 8-21)

Table 2 Glasgow Benefit inventory from Janssen et al. (2012)

1.10.2 Listening Preferences

Dun et al., (2010) investigated bilateral BCHA usage in 20 children who found that 90% of children were using both BCHAs 7 days a week whilst the other subjects were using the BCHA at least 5 days a week for the majority of the day. Children were found to be more likely to remove their BCHAs in background noise or turn one BCHA off if noise was coming from one side. Dutt et al., 2002b had comparable findings within 11 adult subjects who all used both aids for at least 8 hours a day, 7 days a week. Similarly to the Dun et al., (2010), study, they also found that three participants preferred to only use one aid in noisy situations. Nine (from twenty) patients thought that they could localize sounds better with bilateral BCHAs over unilateral fitting. Snik et al., (1998) tested three experienced bilateral BCHA users and found that correct identifications of sound localization improved by 53% over the unilateral condition.

1.10.3 Audiometric measurements

Bilateral BCHAs were shown to produce improvements in Speech Reception Thresholds (SRTs) (4-5.4 dB) in quiet when compared to unilateral fitting (Priwin et al. 2004; Priwin et al. 2007; Hamann et al. 1991; Bosman et al. 2001) as shown in Table 3.There was also evidence for improvement in pure tone audiogram (PTA) with bilateral fitting of a BCHA under a number of different conditions (as shown in Table 4) (Priwin et al. 2004; Priwin et al. 2007). Improvement in PTA when sound was presented in front was 4 dB (consistent with SRT results). Evidence for benefits listening to speech in noise were not clear, with patients with a unilateral BCHA benefiting from head shadow when noise was from the unaided side. When this side was then aided the additional amplification caused the head shadow advantage to be lost.

In addition to SRTs, Bosman et al., (2001) also measured Binaural Masking Level Differences (BMLDs). Bosman et al., (2001) found that, in bilaterally fitted patients, there was a significant release from masking at 125, 250 and 500 Hz of between 6-6.6 dB, but there was no significant BMLD at higher frequencies. Priwin et al., (2004) also investigated BMLD at 250, 500 and 1000 Hz, finding smaller changes in threshold of 3-5 dB. Although it was not clear if these results were significant, as there was high variation between the participants tested. BMLDs were much smaller than those usually identified in participants with normal AC hearing, for whom BMLDs can be 11 dB +/- 2 dB (Jerger et al. 1984) in the 125-500 Hz frequency range. The deficit was thought to be due to crosstalk limiting the ability for central binaural processing to make use of phase difference cues.

Study	No	Mean Improvement in SRT with Bilateral versus Unilateral BCHA
Bosman et al. (2001)	25	4.0 dB ($p < .001$)
Hamann et al. (1991)	23	4.0 dB
Priwin et al. (2004)	12	5.4 dB ($p = .001$)

Table 3 Showing Tone and speech thresholds in quiet as adapted from Janssen et al., (2012) * Speech Reception Thresholds (SRT)

Study	No	Mean Improvement in PTA with Bilateral versus Unilateral BCHA
Priwin et al. (2004)	12	2-7 dB with sound from front, all around, or to aided side 5-15 dB with sound to unaided side
Priwin et al. (2007)	3	4 dB with sound from front

Table 4 Difference in tone thresholds in quiet between bilateral and unilateral BCHA as sourced from Janssen et al., 2012

Lastly the benefit to sound localisation was investigated by Priwin et al (2004) using a circular 12 speaker array set 30° apart (shown in Figure 5). Unilateral fitting results were found to be close to chance level (8.3 % correct and 25 % correct ± 30°) with significant improvements with bilateral fittings at 0.5 and 2 kHz. Priwin suggested that these improvements identified at low and high frequencies gave evidence that bilateral fitting allowed use of ITD and ILD clues not available in unilateral fitting. Priwin later performed a similar study comparing unilateral and bilateral BCHA fitting in children (Priwin et al 2007). On this occasion finding an improvement in localisation in those with bilateral severe conductive loss, however this was not significant. A significant improvement in lateralisation ability was identified. Bosman et al., (2001) and Priwin et al. (2004, 2007) both concluded that the improvements identified in BMLD and localisation ability showed that bilaterally fitted BCHA patients do benefit (to some extent) in binaural hearing but that these benefits were limited due to crosstalk.

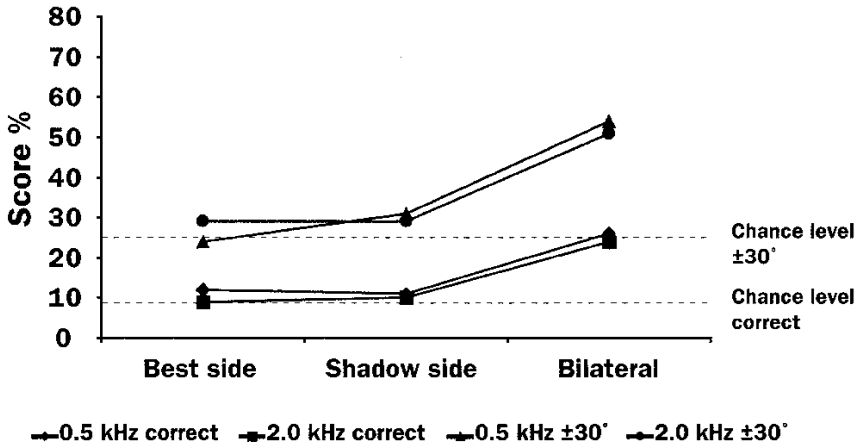


Figure 5 Average sound localisation results from 12 participants with a single BCHA placed on the best and shadowed side as well as a bilateral fitting (reproduced from Priwin et al 2004)

1.11 Crosstalk

Crosstalk was originally described as problematic during production of binaural signals with two loudspeakers. This is illustrated in Figure *6*, where sound from the left speaker (X_L) has a path ($\mathbf{C_L^L}$) to the left ear (Z_L) as well as a crosstalk path ($\mathbf{C_R^L}$) to the right ear (Z_R). The same is true of the right speaker and ear.

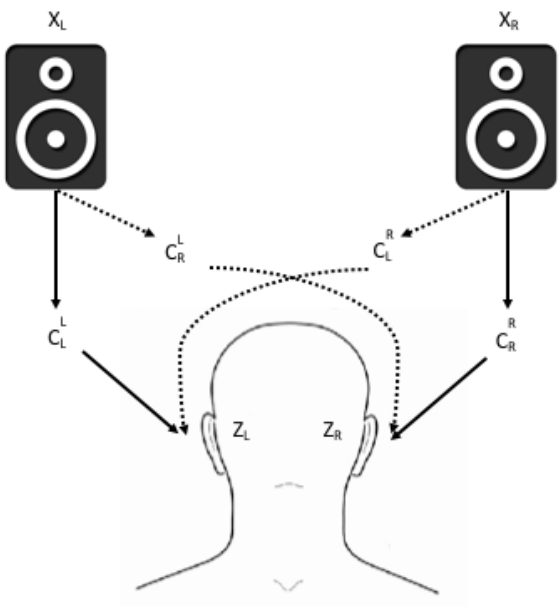


Figure 6 Showing acoustic transfer functions between two loudspeakers and a participant (Reproduced from Liao, 2010)

In order to mitigate the effect of the crosstalk signal, crosstalk cancellation is required. The ultimate aim is to be able to reproduce at listeners' ears the individual stereo channels in the recording. The crosstalk cancellation technique requires the cancellation of crosstalk signals from each speaker to the opposite ear. This method was initially proposed by Bauer in 1961 before Schroeder and Atal (1963) employed the methodology. Later, Schroeder used the technique in comparative studies in reverberant spaces as well as concert halls (Schroeder 1973; Schroeder 1969). The primary method of constructing a cross talk cancellation filter is to invert the head responses obtained by modelling or direct measurement of the crosstalk signal. This can be achieved either with microphones in the ears of participants or via measurements from a KEMAR manikin or similar (Gardner 1998). Schroeder used a dummy head microphone to construct an inverse filter or crosstalk canceller. This was then convolved with impulse responses from a concert hall and presented in an anechoic room with and without crosstalk cancellation. He described the effect as "nothing less than amazing" (Schroeder 1973).

One of the primary problems in crosstalk cancellation is that sound waves coming from two different sources produce interference patterns. Depending on the distance between and from the ears, the distance between the loudspeakers as well as the frequency, the interference patterns might cause the signal to be destructive, complementary or constructive (Choueiri 2008). In a perfect crosstalk cancellation system (defined by infinite crosstalk cancellation over the audio band) frequencies where destructive interference happens to occur can be compensated. However this requires a level boost at the loudspeakers not just to cause the crosstalk to be cancelled but also for the frequency spectrum to be reconstructed perfectly at both ears with no spectral coloration (Choueiri 2008). However at these frequencies, the level boosts are required are very prone to small errors this causes small movements in head orientation/position to not only loose crosstalk cancellation but to have the additional interference of undesired acoustic artefacts (Nelson & Rose 2005; Takeuchi & Nelson 2002).

There are several different methods of crosstalk cancellation including: ideal, adaptive, recursive and fast deconvolution methods (Liao 2010). Liao (2010) attempted each method in order to perform crosstalk cancellation of BC sound in a dry skull.

1.11.1 Ideal crosstalk cancellation

The below equations refer to labels on Figure *6*. Equation 1 is expressed in the frequency domain where $\mathbf{C_R^R}$ and $\mathbf{C_L^L}$ vectors of the direct paths and $\mathbf{C_R^L}$and $\mathbf{C_L^R}$ vectors of the crosstalk paths. Z is the resultant signal at both of the ears.

$$\begin{bmatrix} \boldsymbol{Z_R} \\ \boldsymbol{Z_L} \end{bmatrix} = \begin{bmatrix} \boldsymbol{C_R^R} & \boldsymbol{C_R^L} \\ \boldsymbol{C_L^R} & \boldsymbol{C_L^L} \end{bmatrix} \tag{1}$$

The aim of crosstalk cancellation is to reproduce the original left sided speaker signal at the left ear and vice versa. In order for the signal from the left and right speaker to reproduced a crosstalk matrix H is needed (superscript signifying the side signal origin and subscript indicating the . Where the crosstalk cancellation matrix equals Equation 2.

$$\begin{bmatrix} \mathbf{1} & \mathbf{0} \\ \mathbf{0} & \mathbf{1} \end{bmatrix} = \begin{bmatrix} \boldsymbol{C_R^R} & \boldsymbol{C_R^L} \\ \boldsymbol{C_L^R} & \boldsymbol{C_L^L} \end{bmatrix} \begin{bmatrix} \boldsymbol{H_R^R} & \boldsymbol{H_R^L} \\ \boldsymbol{H_L^R} & \boldsymbol{H_L^L} \end{bmatrix} \tag{2}$$

This matrix can be rewritten to include crosstalk cancellation filters as well as signal reaching both ears as shown in Equation 3 and schematically demonstrated in Figure 7.

$$\begin{bmatrix} \boldsymbol{Z_R} \\ \boldsymbol{Z_L} \end{bmatrix} = \begin{bmatrix} \boldsymbol{C_R^R} & \boldsymbol{C_R^L} \\ \boldsymbol{C_L^R} & \boldsymbol{C_L^L} \end{bmatrix} \begin{bmatrix} \boldsymbol{H_R^R} & \boldsymbol{H_R^L} \\ \boldsymbol{H_L^R} & \boldsymbol{H_L^L} \end{bmatrix} \tag{3}$$

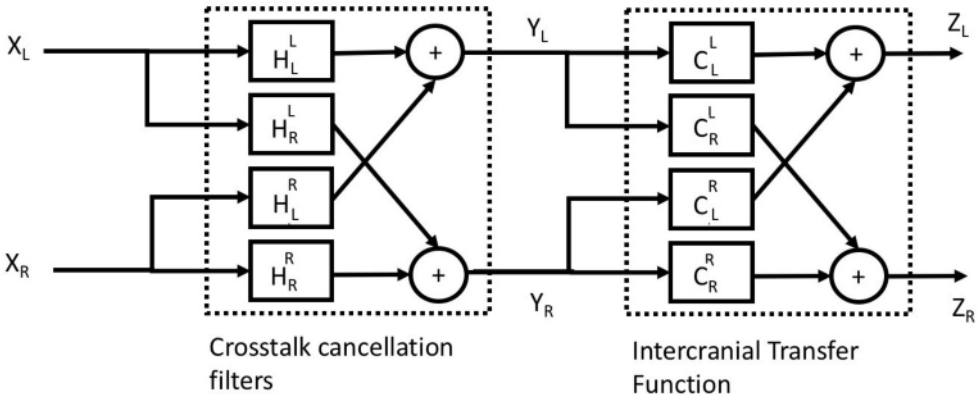


Figure 7 Block diagram of crosstalk cancellation algorithm (reproduced from Liao, 2010)

Since in ideal crosstalk cancellation the H matrix is an inversion of the C matrix it is possible to calculate the H matrix via Equation 4.

$$\boldsymbol{H} = \frac{\mathbf{1}}{\boldsymbol{C_R^R C_L^L - C_L^R C_R^L}} \begin{bmatrix} \boldsymbol{C_R^R} & \boldsymbol{-C_R^L} \\ \boldsymbol{-C_L^R} & \boldsymbol{C_L^L} \end{bmatrix} \tag{4}$$

A particular problem at low frequencies is that the direct path of multiplication ($\mathbf{C_R^R \bullet C_L^L}$) can be almost the same as the product of multiplication of ($\mathbf{C_L^R \bullet C_R^L}$). This "ill condition" creates a large H value matrix which in real application should be avoided since the signal level required would be too high for the speaker. This is a problem when attempting to correct the level. In addition to the ill condition there is also a problem when the phases become very similar as a very small movement of the listener would cause the crosstalk to not only stop working but also be detrimental to the signal when compared to not performing cancellation at all (Liao 2010). Thus ideal crosstalk cancellation cannot be achieved in the real world both due to movement of listeners as well as limitations in speaker output. Instead several other strategies are used which make a compromise to ideal crosstalk cancellation in order to make the cancellation method more robust and practical in a real world setting.

1.12 Bilateral BCHA and limitations due to crosstalk

Bilateral BCHA do show some significant advantages, especially with regards to amplification, However, these benefit come with a high cost, with a single BAHA fitting costing £7304.80 in an adult and £9330.17 in a child in the first year. Although precise costings do not exist for bilateral fittings there would be an additional cost in the region of £5000-7000. This greater cost is even harder to justify given that each new iteration of BCHA design has a more powerful transducer. This means that replacement of the device for a newer device would, in the long term, likely be of greater amplification benefit than having bilateral fittings, as well as being significantly cheaper. In order for bilateral BCHA to be of greater benefit to patients there needs to be additional benefits other than amplification. These include improvements in sound localisation and speech in noise threshold. Research from Priwin and Bosman have shown limited benefits in bilateral BCHA in sound localisation and SRT, but the use of a crosstalk cancellation system has the potential to restore patients' binaural processing ability significantly. Bilateral BCHA are particularly suited to being utilised in combination with crosstalk cancellation. This is due to the fixed point of the abutments which means movements of a listener's head (which can be a problem when using speakers for crosstalk cancellation) do not need to be compensated for.

All of the different forms of crosstalk cancellation require accurate measurements of phase and level to be effective. For speaker crosstalk cancellation this can be performed via microphone measurements from the ear. Bone-conduction measurements cannot be performed in such a way. Liao (2010) used accelerometer measurements to perform

crosstalk cancellation in real time in a dry skull, but this had the limitation of cancelling variation at the abutment rather than at both cochleae, as well as missing the role of the soft tissues. Thus, the primary aim throughout this dissertation is firstly to develop a method which can accurately measure the phase and level of crosstalk vibrations at the cochlea, and, secondly, to use the measured crosstalk signal in order to create a fixed filter which can be implemented in a bilateral BCHA fitting. A model will also be used to investigate the theoretical benefits from a crosstalk cancellation algorithm in a patient population. Other items which will be explored and developed are the need to construct a methodology which is robust and practical to perform in a real clinical setting or, preferably, at home by the patient or patient's family. Lastly, the method ideally should be as automated as possible since many patients currently fitted with bilateral BCHA have complex medical needs, including cognitive impairments. Thus, having a method which is fast and easy to perform will have the greatest impact on the largest number of BCHA users.

For crosstalk cancellation to be possible in a patient population there are also technological challenges which need to be overcome. Primarily the requirement to stream audio from one BCHA to the other and vice versa in real time and with low latency. This could place considerable strain on the battery requirements of the BCHA. However, this will not be a focus of research within this dissertation.

2 Pilot Study investigating the use of a scanning laser Doppler vibrometry of the cranium when stimulated by a B71 bone transducer

2.1 Summary

Scanning laser Doppler vibrometry (LDV) has been used extensively to investigate the movement of the middle and inner ear, but has never been used to measure vibrations from a bone transducer travelling over the skin, subcutaneous tissue and cranium in a live subject. Using three scanning laser Doppler vibrometers we measured the displacement of the cranium in 3D in a live subject when stimulated by a B71 bone transducer placed 55 mm posterior to the external auditory canal. Four pure tones (250 Hz, 500 Hz, 1000 Hz, 2000 Hz) were presented separately via the bone transducer. The displacement of the scalp was imaged in four different areas (Ipsilateral to the bone transducer in the temporoparietal region, contralateral temporoparietal region, occipital region and vertex) and linked to the phase of stimulation. Measured scalp motion was consistent with expected displacement of the underlying cranium. Rigid-body motion was the dominant mode of vibration at 250 Hz. At 1000 Hz a mass-spring effect was seen. At 500 Hz there was a transition frequency between vibration as a rigid-body and as a mass-spring. Higher frequencies (2000Hz) showed that wave transmission was the primary vibrational mode of sound transmission over the cranium. These results broadly support previous research studies but open up potential areas of interest in the investigation of differing skull resonance frequencies.

2.2 Introduction

The routes by which bone-conducted sound travels to the inner ear have been extensively investigated and modelled in both dry skulls and cadavers heads (Håkansson et al. 1986; Wismer & O'Brien 2010; Rowan & Gray 2008). The major routes by which bone conduction signals reach the inner ear as outlined by Tonndorf are thought to be via (a) inertial excitation of the ossicular chain, (b) air-borne sound generated by movement of the ear canal walls (Stenfelt et al. 2003), (c) direct excitation of the cochlea (Tonndorf & Khanna 1968; Zwislocki 1981; Berger et al. 2003), and (d) pressure wave transmission via cerebrospinal fluid (Guignard et al. 2013). One or more of these modes of sound transmission can facilitate hearing in patients with bone-anchored hearing aids (BAHAs).

Bone-conducted sound can also be problematic in extremely loud environments such as an aircraft carrier flight deck or an MRI machine, where it bypasses conventional noise-protection equipment (Homma et al. 2010). A greater understanding of how bone-conducted sound is transferred to the cochlea at different frequencies may allow improved design of hearing-protection devices (HPDs) as well as allowing optimisation of sound transfer to the inner ear via bone-anchored hearing aids (BAHAs).

Investigations of skull vibration have primarily focused on mechanical point impedance (Stenfelt et al. 2000; Håkansson et al. 1986) and resonance frequencies (Håkansson et al. 1993; Khalil et al. 1979). These studies have employed the use of accelerometers and have found that there are no resonance frequencies in the skull below 500 Hz. They found large variations in the resonance frequencies of individuals that could not be explained by head width, length or circumference alone. Due to these findings the consensus is that differences in skull resonance frequency is likely due to geometrical differences within the skull structure (Håkansson et al. 1993).

Several studies have investigated resonances through their effect on lateralisation of bone-conducted sound in patients. Anti-resonance was identified at frequencies between 100 and 350 Hz by Håkansson et al. (1986). Anti-resonance reduces the vibration around the cochlea either directly via bone conduction or via the ossicular chain via destructive interference from different vibrational paths. This causes a marked drop in sound level at the cochlea. Stenfelt et al., (2000) also showed that anti-resonance was present at low frequencies at the cochlea ipsilateral to the Bone Transducer (BT), causing the sound to be lateralised to the side contralateral to the point of stimulation. Håkansson et al., (1993) concluded that one of the major causes of lateralisation was the anti-resonance seen at different frequencies and that the resonance frequencies were likely to be less important.

Previous studies have been primarily focused around the vibrational characteristics of specific areas of the skull (usually close to the cochlea). These have primarily used dry bones or cadavers, although it is unknown how the vibrational characteristics of the skull may differ between a cadaver head and a live subject. It may be that the reduction in intracranial pressure present in a cadaver head could affect the surrounding skull's vibrational modes. Our focus was around how the skull as a whole vibrates in a live subject in 3D. The investigation employed a scanning laser Doppler vibrometer (LDV). This has been used extensively to investigate the physiology of the middle ear and tympanic membrane (Voss et al. 2000; Stenfelt et al. 2004; Homma et al. 2009; Eeg-

Olofsson et al. 2008; Guignard et al. 2013). However, these investigations generally used temporal bones and only employed a single laser vibrometer. Use of a single vibrometer limited vibrational mode data collection to a single plane. In order to capture the vibrational modes of the skull in 3D we employed three scanning laser vibrometers focused on the same area simultaneously. This builds on the research by McKnight et al., 2013 who also used three scanning laser vibrometers to investigate the vibratory response over a range of audible frequencies in dry skulls. Their research into the admittance frequency response and the 3D velocity of the skull surface found that the spherical shell model of skull motion best fitted their findings. This model proposes that the skull can be represented as a spherical shell which supports twisting and elongation parallel to the skull surface but is incompressible in the thickest direction.

The aims of the present study were as follows:

- Evaluate the possibility of using three vibrometers simultaneously in the investigation of bone-conducted sound though skin-surface vibration. This measurement technique has been used previously in live subjects (Kitamura 2012) but not to investigate bone-conducted sound. Where the technique has been used to evaluate bone-conducted sound, dry skulls were employed (McKnight et al. 2013). Thus, it is unclear if the use of a living participant will be possible as there will be interference from head movement as well as limitations on the time taken for measurements.
- Measure the skull vibration of the whole skull in 3D. This is possible since the system used is able to link the stimulus with the phase of the displacement at any point. This facility allowed several different areas of the skull to be imaged separately and then reconstructed and linked to the same point in the phase cycle.

2.3 Materials and Methods

2.3.1 Ethical Approval

The following experimental procedure was approved by the Psychology Ethics committee and the Engineering health and safety department of Cardiff University.

2.3.2 Participant

A bald male (30 years of age) with no previous history of hearing problems participated in the measurements.

2.3.3 Laser Vibrometers

Data was collected via the Polytec^{TM} PSV-500-3D system. This comprised of three scanning laser-Doppler vibrometer units. Each unit consists of a Class 2 He-Ne Laser with an output power of 1mW and an optical transducer that senses frequency shift of

reflected light (Rohrbaugh & Polytec Gmbh 2010). On the basis of the Doppler shift the frequency shift is used to determine the velocity and displacement of a point on an object. All three laser Doppler vibrometers were mounted on tripods and were focused on the target in different, but non-orthogonal, planes (Figure 8). The simultaneous use of three laser Doppler vibrometers in different planes allowed the calculation of displacement and velocity in three orthogonal dimensions. The three lasers were focused on various reference points of interest before measurements were taken. This allowed a 3D reconstruction of the area of interest to be made.

Figure 8 Three PolytecTM PSV-500 vibrometers focusing on participants right temporoparietal region (contralateral to the B71 bone transducer).

2.3.4 Data collection

A B71 (RadioEar^{TM}) bone transducer was placed on the left temporal bone 55 mm behind the external auditory canal. This is the recommended surgical placement position for a bone-anchored hearing aid (Stenfelt et al. 2000). The participant wore laser-protective goggles at all times during testing. Laser measurements were made on the ipsilateral and contralateral temporoparietal region, occipital region and vertex. Testing was performed at four different frequencies (250, 500, 1000, 2000 Hz). All excitation frequencies were presented at one head position before the participant was repositioned to image a different region. The participant lay supine for all tests except for measurements of the occipital region for which he lay on his right side. For each test the orientation of the participant was altered so that all three lasers were able to have a clear line of sight on the area of interest. A single tone was played on the BT by production of a sine wave with a sampling frequency of 44.1 kHz via Matlab^{TM}. Signals were sent to the B71 transducer via ESI Maya A44 audio interface and to the Polytec PSV unit. The

signal input was used by the Polytec PSV to synchronise the displacement and velocity to the phase cycle. Data collection at each viewing angle at each frequency was performed for 15 minutes.

2.3.5 Data Analysis

The 3D displacements relative to the phase of sound were reconstructed for each of the four areas of skull imaged. 3D reconstructions were initially generated perpendicular to the area imaged. In order to visualise displacement of all imaged areas along the axial plane, these images were rotated in 3D. The imaged occipital region was rotated 90 degrees towards the coronal plane. Both temporoparietal images were rotated towards the sagittal plane. The vertex image was not rotated as it was already imaged in the axial plane. Each of the four imaged areas were then merged and linked to the same phase cycle of one frequency in order to show displacement of the head as though looking down on the head from above. Reconstructions were performed at each of the four frequencies of excitation. Images of the displacement of the skull at each frequency were reconstructed every 10 degrees. Images were selected where displacement was at positive maximum (depicted in blue) and negative maximum (depicted in red) along the axial plane for each of the four imaged areas.

2.4 Results

Surface vibration of the skin in response to vibration of the underlying bone was detectable by our system at frequencies from 250 Hz (lowest tested) to 2 kHz. Figure 9 show the spatial pattern of displacement of the skin surface at various maxima for different frequencies.

Figure 9 (a) shows that at 250 Hz the left temporoparietal region (ipsilateral to the B71 BT) is maximally positively displaced when the right is negatively displaced. 180 degrees later in the phase cycle the right temporoparietal region is maximally positively displaced and the left is negatively displaced, as shown in Figure 9 (b).

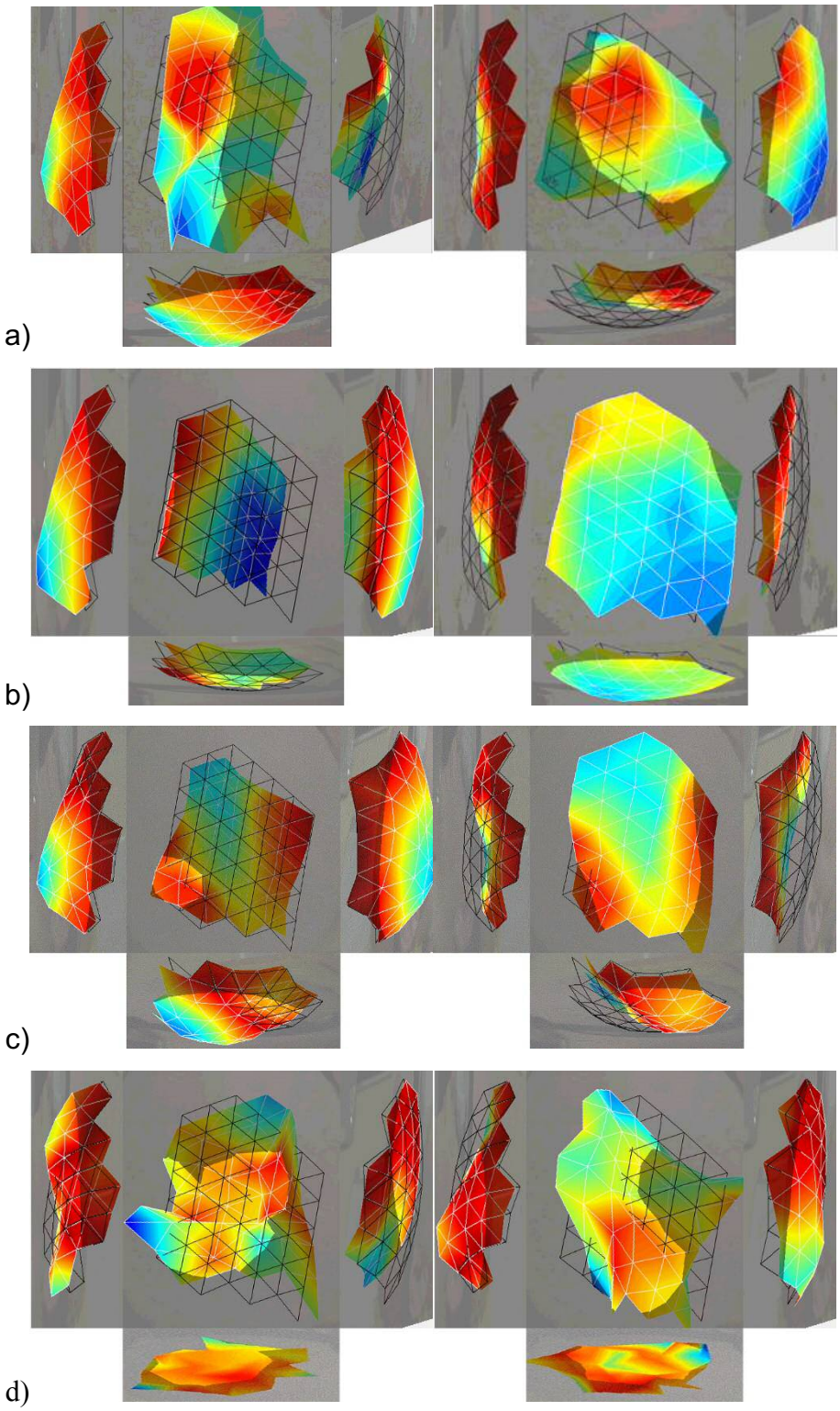


Figure 9 (a) showing relative transverse displacement at 250 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (b) maximal contralateral displacement at temporoparietal region at 250 Hz (c) showing relative transverse displacement at 500 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (d) maximal contralateral displacement at temporoparietal region at 500 Hz (e) showing relative transverse displacement at 1000 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (f) maximal contralateral displacement at temporoparietal region at 1000 Hz (g) showing relative transverse displacement at 2000 Hz when the ipsilateral temporoparietal region is at maximum positive displacement (h) maximal contralateral displacement at temporoparietal region at 2000 Hz.

Figure 9 (c) shows that at 500 Hz the left temporoparietal region is at maximal displacement when the right side is almost at maximal displacement. Similarly when the left side is negatively displaced the right side is almost at maximal negative displacement (as shown in Figure 9 (d)). A surface wave displacement wave can also be visualised travelling over the vertex of the skull as well as across the occipital region. During phase cycles when the temporoparietal regions are positively displaced the vertex and occipital regions are negatively displaced, and vice versa.

At 1000 Hz maximal positive and negative displacements are seen simultaneously at both temporoparietal regions (as shown in Figure 9 (e) and (f)). During maximal positive displacement at the temporal region the occipital region and vertex are negatively displaced. Conversely during maximal negative displacement at the temporal region the vertex and occipital region are positively displaced.

The displacement pattern at 2000 Hz, shown in Figure 9 (g-h), is less clear with no large displacement wave in the temporoparietal regions, which were seen at all the lower frequencies. Instead, a complex displacement wave is visualised travelling across the vertex. No clear travelling wave is visible in the occipital region.

2.5 Discussion

2.5.1 Effects of frequency

Our measurements of skin displacements suggest that at 250 Hz the whole skull is moving from left to right, as shown in Figure 10 (a). This finding is consistent with more direct measurements of the vibrational modes of the skull (Stenfelt & Goode 2005b; Stenfelt 2011; von Békésy 1960). These studies found that the vibrational behaviour of the skull below frequencies of 400Hz is a rigid-body motion.

At 500 Hz we found that large areas of the skull appeared to move into positive and negative displacement together. However, the displacements on both sides were not in unison. Previous studies using cadaver heads have concluded that at frequencies between 500 Hz and 1000 Hz the skull acts as a mass- spring system as shown in Figure 10 (b). Our findings in a live participant seem to show a transition between rigid body motion and a mass-spring system at this frequency.

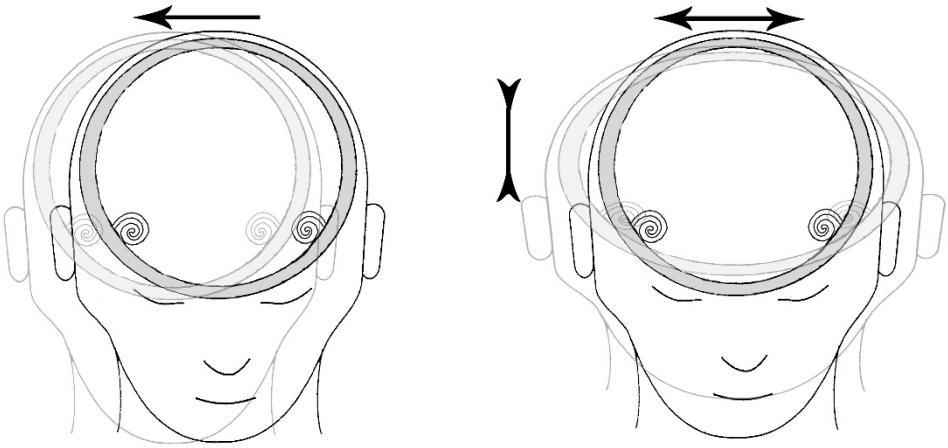


Figure 10 Visual representation of rigid-body motion (a) and mass-spring motion (b). Illustrations based on Stenfelt (2011).

At 1000 Hz both temporoparietal regions move synchronously in positive and negative displacement. The vertex and occipital regions can be seen to be positively displaced when the temporoparietal regions are negatively displaced and vice versa. This again is consistent with findings from Stenfelt (2011) that the skull is acting as a mass-spring system at these frequencies.

Displacement findings at 2000 Hz are more difficult to interpret. There is no longer synchronisation between the displacements of the skull at different areas. However, there are visible transmission waves over the vertex. The displacement pattern over the temporoparietal regions and occipital regions are not so clear. However, these findings are consistent with previous research which found that between 1000 Hz and 2000 Hz the skull vibration transitioned from mass spring motion to wave transmission (Stenfelt 2011).

An alternative method of defining the vibrational modes which have been identified is via modal behaviour. This is particularly clear at 500 Hz, for which the skull vibration could be compared to the lowest vibrational mode of a sphere (Russell 2010).

2.5.2 Limitations

We have shown that, using three vibrometers, it is possible to measure skin displacement, presumed to reflect that of the underlying skull at low frequencies (up to 2 kHz). At higher frequencies a clear pattern of wave transmission was not clear. This is likely related to the localised effect of the vibrational wave at high frequencies. It may be that there were not enough measurement points in order to have a high enough resolution to detect these signals. Future studies using this methodology should consider increasing

the number of scanning points at higher frequencies. Although this will substantially increase the time taken to collect data.

One potential concern when performing this study was that our imaging would not detect skull movement but only movement of the skin and underlying soft tissues. Our measurements have assumed that the displacements detected were of the underlying skull as well as skin. It is possible that this may be a confounding factor when assessing the movement of the skull, as these displacements could be due to pressure waves over the soft tissues of the skull rather than the bony skull itself. Consistency with the literature leads us to believe that our findings at frequencies between 250 Hz and 1000 Hz are unlikely to be due to anything other than the movement of the underlying skull in rigid body motion or a mass spring like behaviour. However, our findings of wave transmission at 2000 Hz could be explained by pressure wave transmission in the skin and soft tissue alone. Nonetheless, all our findings support previous research which has found the same vibrational modes in cadaver heads at similar frequencies.

Another potential concern is that the participant's head was resting against a rigid object. This in itself could change the vibrational characteristics of the head. However, if the participant was allowed to sit up, movement from the participant would have likely made data collection impossible as well as causing increased tension from the muscles attaching to the occipital region of the head which may have affected the skulls vibration characteristics.

2.5.3 Potential Applications

This new methodology opens up potential research areas for sound transmission pathways in individuals who receive a high proportion of bone-conducted sound, such as divers, individuals wearing hearing protection, as well as those with conductive hearing losses.

There are potential research opportunities for developing bone-conducted-sound cancellation techniques whereby bone-conducted sound and its phase can be monitored in real time and cancelled via a BT via a feedback mechanism. This could be key in overcoming the bone conduction limit, at which hearing protection devices cease to be effective. This limit exists due to the vibration of the skull causing a very high level of bone-conducted sound to be transferred. It is a particular problem on aircraft carrier flight decks.

Future research using a similar technique could present a sine sweep to the BT and image different areas of the skull. The scanning LDV then has the ability to extract the differing modes of vibration at each frequency. This allows the potential to visualise areas of the temporal bone where there are clear anti-resonance frequencies. Pure tones at those frequencies can then be presented to see if they result in strong lateralisation to the opposite ear. This could allow a clearer understanding of the relative importance of anti-resonance frequencies of the skull compared to that of the ossicles in lateralisation.

LDV also allows the potential to investigate how two BTs on each mastoid affect the movement of the skull at difference frequencies. This could allow calibration of a cross talk cancellation system whereby sound from one BT is cancelled at the contralateral cochlea by a contralateral BT (Liao 2010).

2.6 Conclusion

We have shown that LDV applied to the scalp is a viable method for measuring displacement of the underlying skull when stimulated by a bone transducer. Our findings in a live participant correspond well with previous research on dry skulls and cadaver heads (Stenfelt et al. 2000). 3D reconstructions demonstrated rigid body motion of the skull at 250 Hz and a transition to a mass spring system at 500 Hz. At frequencies of 1000 Hz we showed clear evidence of a mass-spring effect. However, the number of points used for each scan did not allow a high-enough resolution to identify clear evidence of a wave transmission at 2000 Hz via localised compression of parts of the skull.

This methodology allows further research into methods of utilising two BAHAs in a cross-talk cancellation system. This could be achieved by measuring cranial displacements over the temporal bones using LDV at differing frequencies. This would allow one BAHA to be calibrated in order to cancel the vibrations detected from the LDV by the other BAHA.

3 Measurements of transcranial attenuation and phase of bone-conducted sound.

3.1 Summary

Bone-anchored hearing aids are a widely used method of treating conductive hearing loss, but the benefit of bilateral implantation is limited due to interaural cross-talk. The present study measured phase and level at each cochlea in individuals with normal hearing. In principle, the technique could be used to implement a cross-talk cancellation system in those with bilateral bone conductors. The phase and level of probe tones over two insert earphones was adjusted until it cancelled sound from a bone transducer (i.e. resulting in perceived silence). Testing was performed via placement of a single bone transducer on one mastoid and stimulated at frequencies between 0.25-8 kHz. Probe phase and level results were used to calculate inter-cochlear level and phase differences. The inter-cochlear phase differences of the bone-conducted sound were similar for all three participants showing a relatively linear increase between 4 and 8 kHz. The attenuation characteristics were highly variable over the frequency range as well as between participants. This variability was thought to be related to differences in anti-resonance properties between the ears. Repeated measurements of cancellation phase and level of the same frequency produced good consistency across session from the same participant.

3.2 Introduction

Bone-anchored hearing aids (BAHAs) generate vibrations, which travel through and around the cranium as well as the surrounding tissues (Stenfelt, 2012). For certain patients (particularly those with middle ear defects) this method of sound transfer can offer significant benefits over air conduction (AC) (Gripper et al., 2007). The exact mode by which bone conduction (BC) stimulation is audible was initially addressed by von Békésy (1932). He discovered that it was possible to cancel a 400 Hz tone transmitted via a Bone Transducer (BT) with binaural earphones after the AC sound phase and level was carefully altered. This led to the hypothesis that the initial sound-transfer paths were different for AC and BC but both culminate in stimulating the basilar membrane. Tonndorf, et al. (1966) later described several possible methods by which BC sound transfers to the basilar membrane. However, the relative contributions of these pathways has been disputed. Röösli et al. (2012) suggested that there were four major components

which make up BC sound. These were (a) the inertial movement acting on the ossicles (b) inertia of the inner ear fluid (Stenfelt and Goode, 2005a); (c) sound radiated into the external ear from the ear-canal wall and from soft tissue vibration and skull (Stenfelt et al., 2003; Stenfelt and Reinfeldt, 2007); (d) compression of the petrous bone and sound pressure transfer from the cerebro-spinal fluid (Puria and Rosowski, 2012). These are outlined in Figure 11, adapted from Stenfelt (2011). Stenfelt later demonstrated that the most important mode of transmission was the effect of fluid inertia within the cochlea (Stenfelt et al., 2003; Stenfelt and Goode, 2005a).

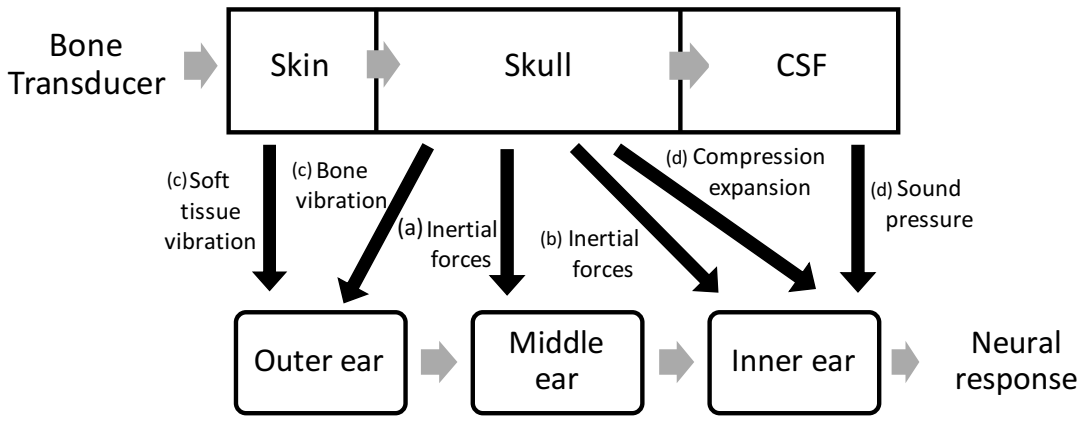


Figure 11 Overview of primary sound pathways via bone conduction, adapted from (Stenfelt, 2011).

Stenfelt (2012) defined Transcranial attenuation (TA) of BC sound as "the difference in sensitivity between an ipsilaterally transmitted and contralaterally transmitted BC sound when the stimulation is at a similar position at the 2 sides of the cranium". Nolan and Lyon, (1981) raised the question of whether sound transmitted by BC is received at the same intensity at the ipsilateral and contralateral cochleae and how transcranial attenuation varies with frequency. Studies which have focused on the transcranial attenuation properties of the skull have used various method with many investigating the difference in hearing threshold in a single-sided deafness (SSD) group when placing a BT on the mastoid bone of the hearing side and the deaf side (Nolan and Lyon, 1981; Stenfelt, 2012). Studies such as this have shown that there is considerable variability in attenuation from -23 dB to 37 dB across listeners and frequencies (Stenfelt, 2012). Stenfelt's (2012) comparison of attenuation in unilaterally deaf patients found that median attenuation was 3-5 dB for frequencies up to 0.5 kHz and 0 dB for frequencies between 0.5-1.8 kHz. Attenuation was much greater (10 dB) at higher frequencies (3-5 kHz). One limitation of these studies is they only give an estimation of transcranial attenuation since they assume skull symmetry and therefore symmetrical attenuation.

However, asymmetry in a complex shape such as the skull is well known from CT imaging (Wismer and O'Brien, 2010). Thus this research is mainly beneficial in investigating appropriate masking thresholds for BC sound rather than giving precise transcranial attenuation properties for an individual subject's skull.

The present study made accurate psychoacoustic measurements of phase and level differences at both cochleae from a single BT. We can then use these values to calculate the TA or transcranial delay (TD). Which we define as the difference between contralateral cochlea measurements when compared to ipsilateral. It is our hypothesis that it may be possible to build on this methodology in order to accurately measure the phase and level of sound from patients with bilateral BAHAs. This would allow the creation of a cross-talk cancellation system by which cross-talk from one BAHA can be cancelled at the contralateral cochlea by a matched level and opposite phase from the other BAHA. This would then allow patients to make use of interaural level difference cues, improving speech in noise thresholds. The current study only investigates a unilaterally placed BT but an extension of the method would allow bilateral data collection to be able to collect the necessary phase and level data for a cross-talk cancellation system.

Measurements of phase and level have been made previously at the cochlea using similar techniques. However, they have been limited to relatively few frequencies (Clavier et al., 2010; Puria and Rosowski, 2012; Stenfelt, 2007). Studies which have investigated vibration and phase characteristics of the skull over a wide frequency range have used holographic interferometry (Dörheide and Hoyer, 1984; Hoyer and Dörheide, 1983) or laser-Doppler-vibrometer and accelerometer measurements (Stenfelt and Goode, 2005a). The present work is the first study to measure the level and phase of BC sound reaching both cochleae over a wide frequency range. There were of two experiments. The first investigated a narrow frequency range on each experimental sitting in order to identify the 'fine structure' of phase and level changes. It also allowed investigation to see if common patterns of cancellation level and phase between participants could be identified. The second experiment tested a wide frequency range on different occasions in each participant with the aim of elucidating the variation in results of cancellation phase and level between each sitting of the same participant. This informs how much variation in results may be due to slight variation in BT placement position and coupling which is impossible to avoid without attaching the BT to an abutment. Both

experiments utilised the same experimental methodology with the only variation being the frequencies which were tested.

3.3 Methods

The following experimental methodology was approved by Cardiff University Psychology Department Ethics Committee.

3.3.1 Apparatus

Matlab™ was used to generate tones at a sampling rate of 44.1 kHz over three channels with the ability to vary the level and phase of each channel independently. An 8-channel Echo Darla 24/96 DAC passed signals through an 8-channel Behringer Powerplay Pro-8 Headphone Amplifier to a pair of Etymotic ER2 insert earphones and a B71W (Radioear) BT for BC mastoid stimulation. ER2 earphones with ER1-14B eartips attached were used to present the AC sound. These earphones where selected over open-ear headphones to prevent contamination from air-borne sound produced by the BT from affecting the signal at the cochlea. To minimise differences between experimental sittings of the same participant and between different participants, specially adapted lens-less glasses were used. These comprised of a highly flexible plastic attachment which held the B71W in position whilst causing very limited vibration of the glasses themselves. The glasses allowed lower variation in B71W placement, because the superior portions of both pinnae as well as the bridge of the nose are effectively used as a fixed-point reference tripod for the glasses to rest on. The attachment for the B71W onto the glasses aimed to position the BT 55 mm behind the opening of the external auditory canal. This is the recommended surgical placement position for a BAHA abutment (Battista and Ho, 2003). Testing was performed in a single-walled Industrial Acoustics Company (IAC) sound attenuating booth within a sound treated room.

3.3.2 Calibration

In order to stimulate the BT at an appropriate level one participant performed cancellation of BC sound using AC using the method described in section 3.2.5 at 1,3,5,7 kHz. Once cancellation was achieved the corresponding level of both ER2 and BT was varied by + 5 dB and - 5 dB. No noticeable change in cancellation quality was identified by the participant indicating linear output from both BT and ER2 over this level range. The presentation level of the BT was then set to the initial presentation level for all participant testing.

3.3.3 Exclusion and inclusion criteria

Participants with self-reported normal hearing and no previous history of otitis externa or ear surgery were included. Otological examination was performed on participants to check for earwax. Participants with the potential for wax impaction following deep insertion of ER1-14B eartips were excluded.

3.3.4 Participants

Six participants were recruited, however, following ear examination, two were excluded and one further participant experienced temporary otalgia following one testing session and did not take part in further sessions. Therefore, three participants completed testing (age range 22-29 years old).

3.3.5 Testing procedure

After deep insertion of eartips (approximately 22 mm in the ear canal), the bone vibrator was placed on the left mastoid and held in place by the adapted lens-less glasses (the left side will subsequently be referred to as the ipsilateral side and the right the contralateral side). A softband was then placed over the participant's head and B71W bone vibrator in order to maintain a good acoustic coupling with the skull. The band was adjusted in order to achieve a static pressure of 2.5-3 N as previously described by Reinfeldt et al., (2010). Participants performed 8 separate 1-hour testing sessions. Each session comprised a testing of several frequencies in a narrow band range (different on each session) and testing over a wide frequency range (repeated on each session). The testing of a high density of frequencies over a narrow band was designed to identify the 'microstructural' changes in level and phase over this band. For clarity we have called this experiment 1. Retesting the same frequencies on each sitting revealed the variations caused by replacement of the BT over a number of experimental sittings. This will be referred to as experiment 2.

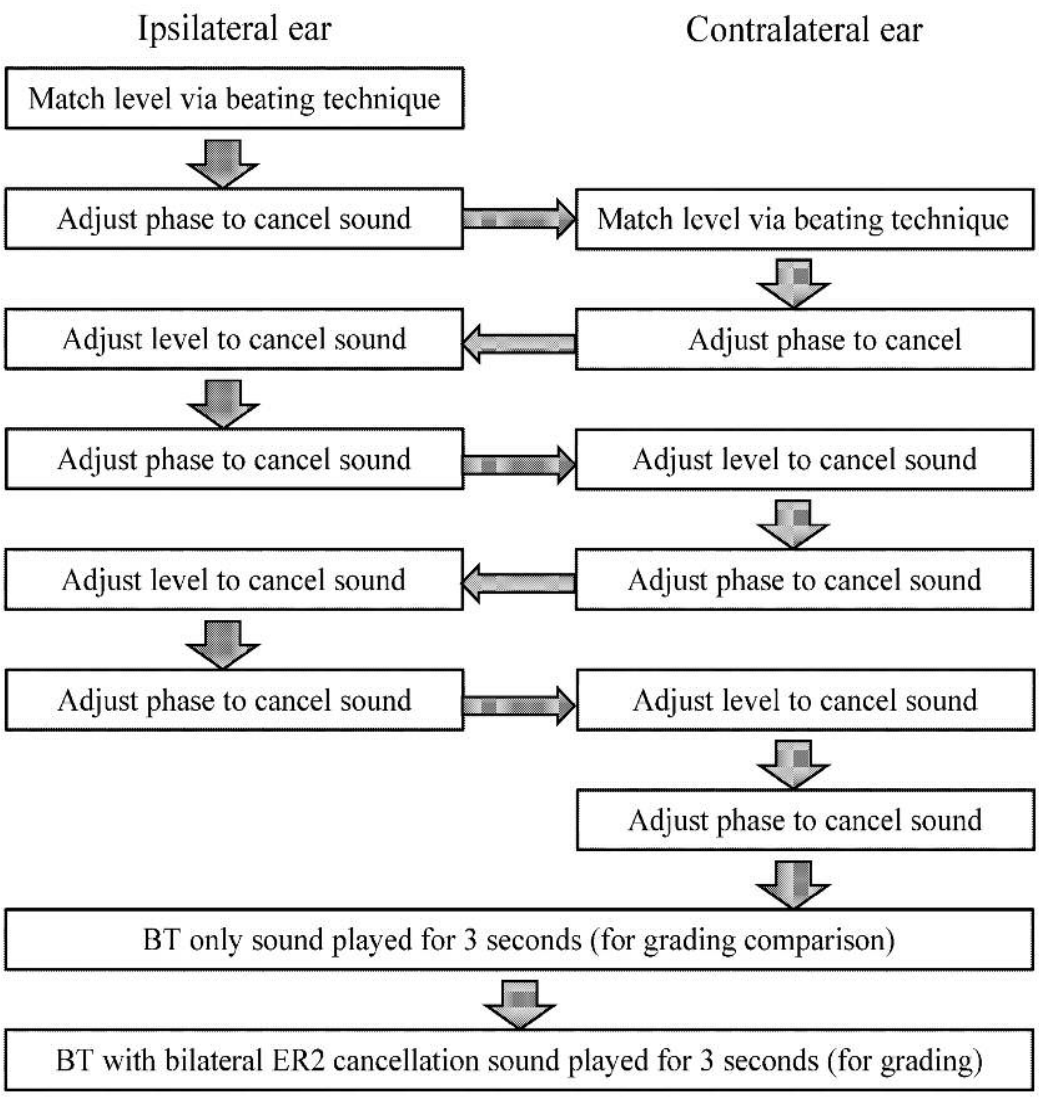


Figure 12 Illustration of the psychophysical procedure for cancelling a bone-conducted sound at both ears and providing an effectiveness rating.

Figure 12 shows the testing procedure undertaken at each frequency. During each test, a single target tone was presented via the BT at the same level and a 1-Hz higher tone presented via the ipsilateral ear over the ER2 earphones. No contralateral masking was used throughout the procedure. The participant was asked to vary the level of the ER2-presented tone in order to maximise the perceived beating effect as the two signals constructively and destructively interfered. Beating is known to be maximal when the levels of the signals at the basilar membrane are equal (Wever and Lawrence, 1954). Level changes were made using the scroller on a computer mouse. Each step of the scroller changed the level by 0.2 db. Once the participant had selected an appropriate level, the same levels were presented again but using the same frequency in both the left

ER2 earphone and the BT. Since the level of the AC and BC sound should be matched at the ipsilateral basilar membrane, the participant could then be asked to change the phase of the ER2-presented tone so as to minimise the perceived sound in the left ear. This procedure was intended to determine the phase at which destructive interference would occur and was again achieved using the mouse scroller, with each step of the scroller changing the phase by 2 degrees. To cancel the signal going to the contralateral ear, the same processes of level adjustment followed by phase change were repeated on the right ER2 while the cancellation signal was simultaneously played on the left ER2. Participants were then asked to perform two further iterations of changing the level and phase in both ears in order to minimise the sound perceived at each ear.

Following the adjustments to the phase and level of both ears, the initial un-cancelled signal from the BT was presented for 3 seconds. This was then followed by the cancellation signal from the ER2 earphones with the BT signal for 3 seconds. Using table I the participants were then asked to rate from 1-5 how well they had achieved cancellation (i.e. the reduction in loudness achieved). The purpose for this grading system was twofold. Firstly, it indicated the relative difficulty of achieving cancellation at different frequencies. Secondly it allowed results in which the participant had not achieved good cancellation to be identified. Phase and level results from tests where grades of 1 were recorded were excluded from final analysis.

Grade	Description
1	As loud as start of task
2	Slightly quieter than bone transducer alone
3	Much quieter than bone transducer alone
4	Only slightly audible
5	Total cancellation (nothing audible)

Table 5 Grading system post attempted cancellation

Participants performed eight one-hour testing sessions where half the experimental time was taken performing experiment 1 and the other half performing experiment 2. During each experimental session the BT position was not adjusted.

3.3.6 Experiment 1

Participants performed testing every 50 Hz between 0.25 kHz and 8 kHz with no frequency presented more than once. Testing was split into 8 sessions, which focused around a 1-kHz frequency band (for example 3.05-4 kHz in 50 Hz increments). The order at which each frequency band was tested was randomised across participants as was the order of the increments (up or down) during each testing session. This aimed at reducing practice affects which might have biased the results. In order to unwrap the phase results for the three participants the phase result from the highest frequency (e.g. 2 kHz from a 1.05-2 kHz testing session) was compared to the lowest frequency for the adjacent set frequency (e.g. 2.05 kHz from a 2.05-3 kHz testing session). The phase results for the whole testing session were shifted by the same number of phase cycles to have the smallest difference in phase between the two separate testing sessions. The phase results for both the ipsilateral and contralateral sides were shifted by a multiple of 360º so that the phase displayed was as small as possible between testing sessions. The unwrapped phase results at the ipsilateral and contralateral cochleae were then compared at the lowest test frequency (250Hz) and anchored at this point where there is little phase difference. This was necessary in order to calculate the number of phase cycles between the cochlea to calculate TD.

Calculation of the TD and level TA was performed by subtracting the ipsilateral and contralateral unwrapped phase and the level. TA was calculated by subtracting the ipsilateral cancellation level from the contralateral. A positive value indicating the ipsilateral level was greater than the contralateral. We compared our psychoacoustic

measurements of TA and TD to Zwislocki's (1953) as well as physical measurements from Stenfelt & Goode (2005). Zwislocki's psychoacoustic method focused on the relative contribution of BC sound from earphones. He applied a loud tone to a right sided earphone in order to generate a BC input which was then cancelled at the contralateral ear via level and phase adjustment of the same tone via a further earphone. Using his measurements, it is possible to estimate the TD through the skull between 250 Hz - 2.6 kHz (the frequencies that were investigated). Stenfelt and Goode employed accelerometers, which were attached close to the cochlea in severed cadaver heads. Several accelerometer positions were examined however the third occipital position was thought to be the closest to the recommended abutment placement position. Thus the mean accelerometer measurements placed in the third occipital position were used from six cadaver heads as a comparator. Stenfelt & Goode's collected data related to phase and level of vibrations in all three planes. However, the relative contribution to hearing of each plane of transmission are not known therefore the dominant plane of transmission was used as a comparator. This was referred to by Stenfelt and Goode as the x-axis, where vibrations were parallel to the sagittal plane. The difference between the phase and level measurements from the ipsilateral and contralateral x-axis accelerometers in the third occipital plane were then used to estimate the TA and TD.

3.3.7 Experiment 2

Experiment 2 utilised the same testing procedure as already outlined in Figure 12. Participants performed 8 testing sessions of the same 8 frequencies (every 1 kHz between 1-8 kHz). The order in which the frequencies were presented was randomised in order to minimise practise effects. The primary purpose of this experiment was to investigate the effect of small placement differences and coupling between the BT and skull on phase and level results needed for cancellation. In order to achieve this, the mean and standard deviation for each test frequency were calculated for the ipsilateral and contralateral phase and level.

3.4 Results

3.4.1 Experiment 1 data exclusion

Participant 1 had 3 grading scores of 1. Participant 2 had 5 grading scores of 1 and participant 3 had 6 grading scores of 1. The ipsilateral and contralateral phase and level results related to these scores were excluded from further analysis. The majority of the results from these scores did not align with phase and level results for closely related

frequency results and therefore this method was deemed appropriate for excluding outlier results.

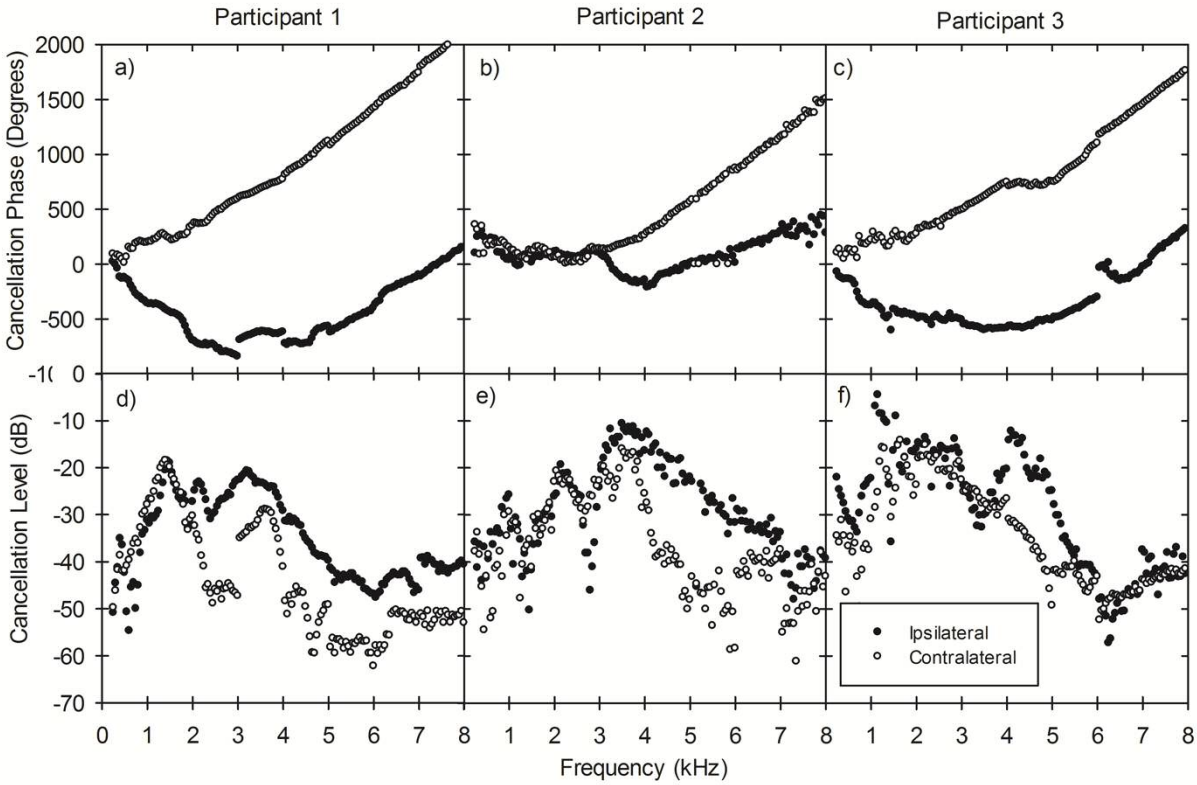


Figure 13 The raw cancellation phase (unwrapped) and level results from ER2 earphones needed at different frequencies to cancel a single B71W BT tone in 3 participants.

3.4.2 Experiment 1 Phase Data

Figure 13 a-c shows the raw unwrapped phase data for the ipsilateral and contralateral ER2 for the 3 participants. The data would include the effects of the transducers and their coupling to the head or ear, so the absolute values are not meaningful. However, it is clear from the results that, within each testing session, trends in phase seem to be very consistent, while between sessions there can be discontinuity in phase (clearest in participant 1 between the 3-4 kHz testing session). This shows that, although using lens-less glasses as a method for B71W positioning provides considerable reliability, it has not completely resolved the problem of phase consistency between testing sessions. Calculation of the difference between ipsilateral and contralateral ER2 phase results can be used to calculate the transcranial delay (as discussed in 3.3.4).

3.4.3 Experiment 1 Level Data

Figure 13 d-f show the ipsilateral and contralateral levels for cancellation using ER2s. The reference scale level is arbitrary but instead is of use to compare the relative

levels needed for cancellation at each cochleae as well as be able to identify changes over frequency. As was evident in the phase results there are several discontinuities in level. These are most clear in participant 1 in the contralateral ER2 at 3 kHz and participant 3 in the ipsilateral ER2 at 1 kHz. These jumps signify the intersection between different testing sessions and are likely due to alterations in coupling or position between the skull and transducer despite the use of glasses to minimise transducer placement error. Additional possible reasons for discontinuities between testing sessions maybe related to changes in softband positioning.

3.4.4 Transcranial Attenuation and Transcranial Delay

Figure 14 shows the mean TA and TD calculated from the three participants using data from experiment 1. TD increases relatively rapidly at frequencies below 4 kHz before a gradual linear increase at higher frequencies. However, there were significant differences in the number of phase cycles that were needed to cancel sound over the same frequency range, participant 1 requiring 2 full cycles more than participants 2 and 3. This may have been due to different skull sizes or skull widths, but this was not investigated within this study.

Additionally, the large discontinuities in level (seen between experimental sessions) that are apparent in Figure 13 are no longer visible, indicating that these differences are primarily due to coupling variability which does not affect TA. Since cancellation thresholds are equally impacted on both sides, the effects of the changes in coupling etc. are removed when the raw data at one ear is subtracted from that of the other to give the transcranial differences. Patterns in TA are not as consistent across participants as those seen in TD. All participants had results with highly negative TAs although these were at different frequencies. Negative TA results indicate that a greater level is needed for cancellation at the ear contralateral to the BT when compared to the ipsilateral ear. Participant 2 showed the greatest negative TA (-16.2 dB) at 2.8 kHz, this was primarily due to a large reduction in ipsilateral cancellation level.

For each participant, there were large drops (>10 dB) in the level needed to cancel BC sound over a relatively narrow frequency range (0.5 kHz). In participant 1, this was most marked at the contralateral ER2 at frequencies of 2, 4 and 4.5 kHz. In Participant 2 and 3, the ipsilateral ER2 showed the most prominent acute reductions in cancellation level. These were identified at 1.5 kHz and 2.8 kHz in participant 2 and 3.3 kHz and 6.3 kHz in participant 3.

The greatest positive TA was identified in participant 2 at 5.75 kHz, where the ipsilateral ear was 25.9 dB higher than contralateral. Other frequencies where a large TA was identified were 4.85 kHz in participant 2, 2.9 kHz and 4.6 kHz in participant 1 and 4.9 kHz in participant 3.

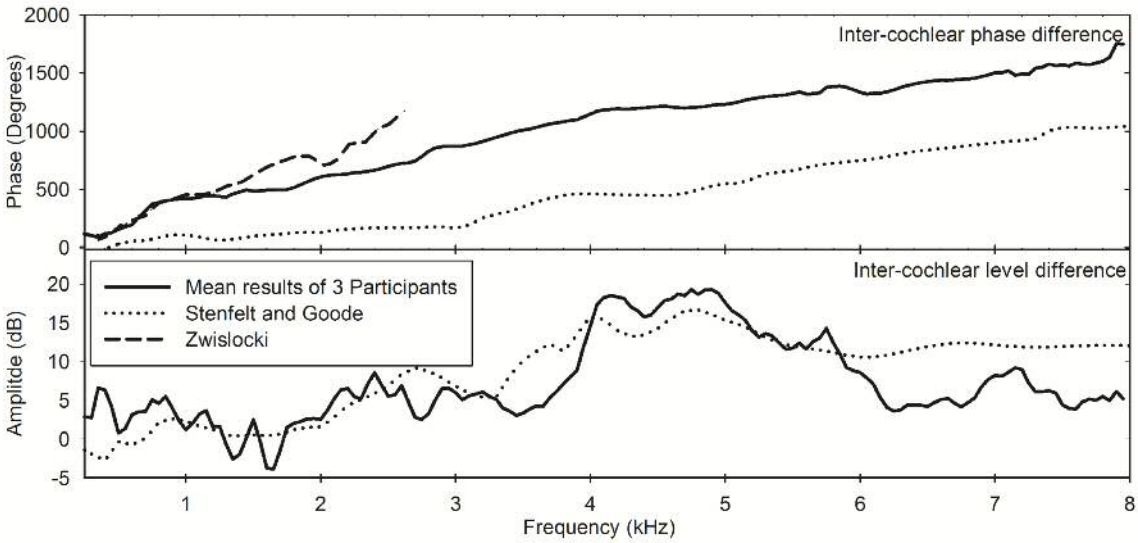


Figure 14 Mean inter-cochlear phase and inter-cochlear level data, Zwislocki (1953) phase data (using a loud tone at one ear rather than a BT), and Stenfelt & Goode (2005a) inter-cochlear phase and inter-cochlear level difference (derived from accelerometer measurements of cadaver heads).

3.4.5 Experiment 2 Across-sitting variability

Figure 15 shows the standard deviation of the ipsilateral and contralateral level and phase results for each participant (n=8 for each participant). Repeated measures analysis of variance was performed on the standard deviation results of level needed for cancellation comparing the ipsilateral and contralateral ER2s for each of the three participants (Participant 1: F=1.016, p=0.347, Participant 2: F=0.297, p=0.603, Participant 3: F=0.01, p=0.974) as well as overall (F=0.767 p=0.39) This found that there was no difference in standard deviation between the ipsilateral and contralateral levels. Repeated measures analysis of variance was also performed on the standard deviation of the cancellation phase comparing the ipsilateral and contralateral ER2 results (Participant 1: F=10.126, p 0.015, Participant 2: F=1.933 p=0.207, Participant 3: F=0.734, p=0.42, overall F=5.967, p=0.023) Overall and in participant 1 there was a significantly greater standard deviation in the contralateral ER2 when compared to the ipsilateral. The highest frequencies had the greatest standard deviation at the contralateral ER2.

Paired T-tests were performed to investigate if there were consistent differences in level between the ipsilateral and contralateral sides for each participant at each frequency. In one instance the contralateral level was found to be significantly greater than the

ipsilateral side. This was identified at 1 kHz in participant 1 where the mean difference between ipsilateral and contralateral side was 3.8 dB ($t=-2.97$, $p<0.01$). At all other repeated frequencies in all participants the mean level needed for cancellation was higher at the ipsilateral side. However, this difference was not significant in participant 1 at 2, 3 and 8 kHz and in participant 3 at 2, 3, 6, 7 and 8 kHz.

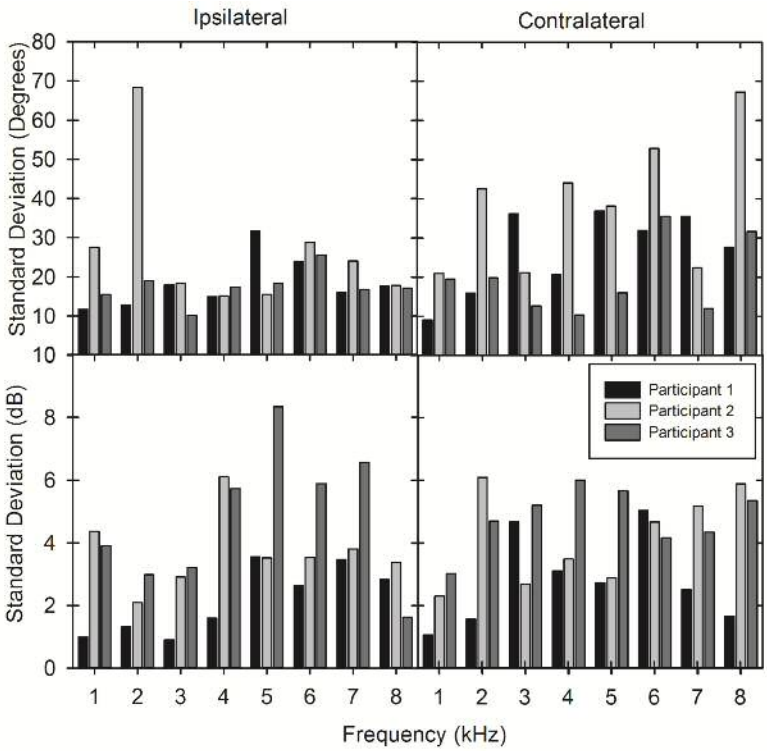


Figure 15 The standard deviation of the inter-trial cancellation level and phase for each of the participants. Results from 8 different testing sessions.

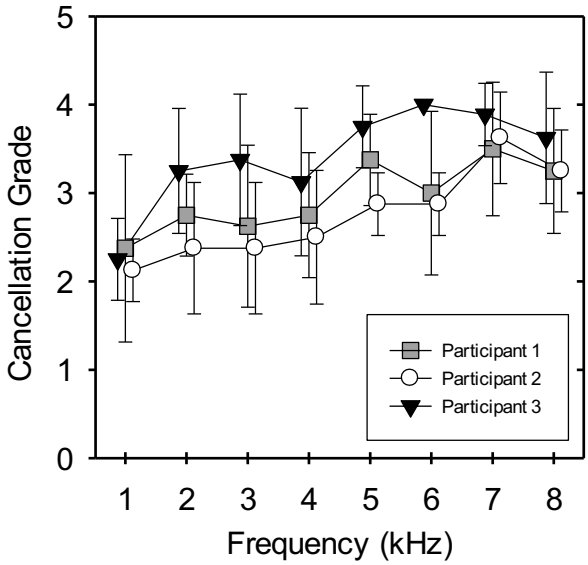


Figure 16 The mean cancellation grade (n=8) for each of the participants. Error bars show one standard deviation.

3.4.6 Experiment 2 Grade Data

Figure 16 shows the mean grade results of the 8 testing sessions. Grading scores are lowest in all of the participants at the lowest frequency of 1 kHz. There is a general increase in grades between 1 kHz and 7 kHz reflecting perceived better cancellation at higher frequencies.

3.5 Discussion

3.5.1 Cancellation Phase change with frequency

There were consistent findings in all three participants for both ipsilateral and contralateral phase results. Linear increase in phase of the ipsilateral ER2 where identified at all but the lowest frequencies (<1 kHz). The contralateral phase reduced with increasing frequency up to approximately 3 kHz before a relatively linear increase. The linear incline in the contralateral phase was not as great as for the ipsilateral cancellation phase.

Our TD findings (shown in Figure 14) correspond very well with frequencies below 2.7 kHz which Zwislocki (1953) examined showing comparable TD progression over this range. All three participants had greater TD when comparing findings with Stenfelt and Goode's (2005) measurements from accelerometer measurements of cadaver heads. Although TD overall was greater in all three participants when compared to Stenfelt and Goode's (2005) work. Phase progression (as defined as the change in phase over a frequency range) was similar at frequencies above 4 kHz where wave motion predominates. Our results found a greater accumulation of phase cycles than Stenfelt and Goode, indicating a larger TD. It is unclear why there are differences between the two experiments.

One possible explanation for the difference relates to intracranial pressure (ICP). This is defined as the mean arterial pressure minus the cerebral perfusion pressure. Normal values are between 5-15 mmHg (Dunn, 2002). Von Békésy (1932) described how a cancelled BT signal is audible during coughing. It is also known that ICP increases during coughing. Thus, ICP may play an important role in phase velocity. Stenfelt & Goode's laser Doppler measurements were performed on cadaver heads within which the ICP would likely be 0 mmHg. Since Stenfelt has previously shown that fluid inertia within the cochlea is the main mode of sound transmission in bone-conducted sound this could have implications for the Doppler results. Further research is needed to investigate the role of ICP in cancellation results. Manipulation of the ICP would allow the investigation

of how ICP changes the phase at different frequencies. This may then explain differences between Stenfelt's phase findings and our own.

3.5.2 Cancellation Level changes with frequency

TA results were less consistent between participants when compared to phase. This fits well with previous studies (Nolan and Lyon, 1981; Stenfelt, 2012; Stenfelt and Goode, 2005b). Further investigation of TA (shown in Figure 14) found that in the majority of cases participants had a positive TA indicating greater perceived level at the ipsilateral cochlea. This is to be expected as dampening will be experienced over a greater distance in the contralateral cochlea when compared to the ipsilateral resulting in greater energy dissipation (Stenfelt, 2012). However, at some frequencies there were large negative TA. One example of this was identified within participant 2's results at 2.7 kHz. This was due to a significant drop in level needed for cancellation at the ipsilateral cochlea at 2.7 kHz without any change at the contralateral cochlea. These strong lateralisation effects have been reported previously and are thought to be due to resonance and anti-resonance (Håkansson et al., 1996; Stenfelt et al., 2000; Tonndorf et al., 1966). Anti-resonance is thought to occur when sound pathways take different routes, which causes them to destructively interfere at or before the basilar membrane. Previous studies have concluded that the anti-resonance frequencies, which usually occur at the ipsilateral ear, may explain the large differences in the literature on TA properties (Eeg-Olofsson et al., 2011; Stenfelt and Goode, 2005; Stenfelt, 2012; Stenfelt et al., 2000). Further research is needed to investigate whether the anti-resonance found at 2.7 kHz was primarily due to the intrinsic properties of the ossicles, cochlea or temporal bone, or if it is also dependent on BT position or occlusion effect.

Our measurements corresponded well with Stenfelt's TA data. With low or negative (indicating a higher level on the contralateral side) TA at low frequencies. In both studies TA increased to around 10-15 dB from 4 kHz before a small reduction in TA in frequencies above 6 kHz.

3.5.3 Test re-test standard deviation

There was no difference in standard deviation (SD) between the ipsilateral and contralateral cancellation levels. This may indicate that small variations in coupling and position which occur with repeated placement of the BT affect both sides equally and that the task is of equal difficulty irrespective of the side of the BT.

The SD of the phase from the contralateral ER2 was found to be significantly greater than the ipsilateral ER2. This indicates that variations in BT placement are more

critical to the phase at the contralateral cochlea. The SD at the contralateral ER2 was found to be higher with increasing frequency. This indicates that small variations in placement positions can make large differences in phase in the contralateral cochlea when compared to the ipsilateral cochlea and that these differences increase with frequency. It was unclear why participant 2 was found to have a large increase in standard deviation at the 2 kHz level. It maybe that the participant wasn't achieving good cancellation at this frequency on two separate occasions but maintained a good grading so that the results were not excluded from analysis. An alternative explanation is that at 2 kHz there are two transmission pathways which are of similar level and that they are interacting to cause large changes in phase despite only small changes in placement position.

3.5.4 Grading

All Participants reported that cancellation was 'best' achieved between 4 and 7 kHz. The most difficult frequencies to cancel were the lowest frequencies (under 1 kHz). All participants had a similar pattern of grading with a steady increase in grade from 1 kHz up to 7 kHz, which corresponds with greater perceived cancellation at the cochlea before a slight fall at 8 kHz.

During collection of grading data, the degree of lateralisation (indicating if only one side was poorly cancelled) was not considered. Thus a poor cancellation grading at one frequency would result in both the ipsilateral and contralateral phase and level results being excluded. This method was used as it was felt that if the ipsilateral cochlea cancellation was performed poorly then the contralateral cochlea cancellation would have been very difficult to perform accurately and therefore would have been equally inaccurate.

The lowest mean grading was identified at 1 kHz, this was likely due to the greater influence in lateralisation due to interaural phase difference (Clavier et al., 2010; Rowan and Gray, 2008). This makes the initial task of accurately detecting maximal beating more challenging, impacting the accuracy of the rest of the task. Additionally the skull is thought to act in as a mass spring between 0.3 kHz and 1kHz (Stenfelt, 2011). At these low frequencies the two temporal bones will be vibrating approximately out of phase. It could therefore be presumed that the psychoacoustic effect of this would be that participants experienced a beat at each ear at opposite points in the phase cycle. If this was indeed occurring, then it would make the level adjustment even more challenging.

3.5.5 Future research

This study has shown consistent, progressive patterns in the phase and level of BC sound within the same participant. Thus, future studies employing a similar technique should find it possible to significantly speed up the testing procedure by using level and phase results for one frequency to predict the level and phase of similar frequencies. By doing this, a number of experimental steps can be avoided. This might allow the same spectral sampling to be achieved over a single one-hour testing session as opposed to eight one-hour testing sessions. The speed at which such data can be collected could be critical as it could make potential applications for phase and level values more appealing. Such values could be key in the creation of cross-talk cancellation systems for bilateral BAHA users whereby two BAHAs are employed with the sound from one BAHA reaching the contralateral cochlea cancelled by appropriately filtered sound delivered by the ipsilateral BAHA. In order to achieve this, the level and phase of sound from each BAHA needs to be known accurately at each frequency. This method could allow the creation of such a system (Liao, 2010). To study if this is possible we will investigate if the use a single-bone-transducer method can be used to accurately predict the inter-cochlear level difference (ICLD) and inter-cochlear phase difference (ICPD) between two BTs in order to cancel sound at one cochlea. Other potential uses include the creation of hearing protection systems where despite protection of AC sound the BC pathways are still high enough to cause sensorineural hearing loss, such as on an aircraft carrier flight deck (Puria and Rosowski, 2012). The limit of hearing protection via conventional means is commonly known as the BC threshold (Reinfeldt et al., 2007). In theory knowledge of the phase and level of sound reaching each cochleae could be used to create an out of phase sound of the same level to cancel the BC sound. Thus, overcoming the BC threshold barrier which currently limits hearing protection devices.

3.6 Conclusion

Using a single BT and ER2 headphones, we have demonstrated that it is possible to achieve repeatable phase results in both the ipsilateral and contralateral ears in binaurally hearing participants on multiple separate testing sessions over a large frequency range from 0.25-8 kHz. There was a general linear increase in phase in both the ipsilateral and contralateral ear above 4 kHz. In the contralateral ear, the phase for cancellation reduced with increasing frequency in all three participants between 0.25 and 4 kHz. The test retest phase SD was found to be greater in the contralateral cochlea when compared the ipsilateral and the SD increased with increasing frequency. This indicates that small variations in BT position affect phase at the contralateral cochlea more than the ipsilateral cochlea. This could be due to several different vibrational pathways interacting to stimulate the contralateral cochlea, whilst the ipsilateral cochlea may be more likely to have a 'dominant' vibrational pathway and is thus less likely to be affected by a small change in position.

There were significant variations in ipsilateral and contralateral levels needed for cancellation. These were both frequency and participant dependent and corresponded well with the existing literature (Pfiffner et al., 2011; Stenfelt, 2012), but are seen in much greater detail in our data. There were multiple large increases in ICLD of >10 dB identified over a relatively narrow frequency range (<0.5 kHz). Similarly, large negative ICLDs were identified where the contralateral cochlea required 10 dB or more sound to cancel than the ipsilateral cochlea. The large increases and decreases over a narrow frequency range are thought to be due to resonance and anti-resonance (Stenfelt, 2012). Despite great variation in cancellation levels at different frequencies, there was no significant difference in variation between testing sessions when comparing the standard deviation of the ipsilateral and contralateral sides. This indicates that small changes in position and coupling appear to affect both cochleae equally.

In future studies we plan to investigate if it is possible to simultaneously use two BTs (one on each mastoid) and for the phase and level of one to be adjusted in order to cancel sound at one cochlea. If such a method is possible then the cancellation phase and level results from a single BT and ER2 eartips should be able to predict them.

4 Psychoacoustic measurement of phase and level necessary for cross-talk cancellation using bilateral bone transducers.

4.1 Summary

Two bone-conduction hearing aids (BCHAs) could deliver improved stereo separation using cross-talk cancellation. Sound vibrations from each BCHA would be cancelled at the contralateral cochlea by an out-of-phase signal of the same level from the ipsilateral BCHA. A method to measure the level and phase required for these cancellation signals was developed and cross-validated with an established technique that combines air- and bone-conducted sound. Three participants with normal hearing wore bone transducers (BTs) on each mastoid and insert earphones. Both BTs produced a pure tone and the level and phase were adjusted in the right BT in order to cancel all perceived sound at that ear. To cross-validate, one BT was stimulated with a pure tone and participants cancelled the resultant signal at both cochleae via adjustment of the phase and level of signals from the earphones. Participants achieved cancellation using both methods between 1.5-8 kHz. Levels measured with each method differed by <1 dB between 3-5 kHz. The phase results also corresponded well for the cancelled ear (11° mean difference) but poorly for the contralateral ear (38.4° mean difference). The first method is transferable to patients with middle-ear dysfunction, but covers a limited frequency range. Extrapolation to other frequencies may be guided by data from the second method.

4.2 Introduction

One problem with bone-conducted (BC) stimulation is that there is little interaural attenuation between signals at the two cochleae (Stenfelt 2012; Rowan & Gray 2008). This can be useful in patients fitted with a bone conduction hearing aid (BCHA) for single sided deafness (SSD), but can be problematic in patients with two working cochleae where the aim is to restore the benefits of binaural hearing (Rowan & Gray 2008). If two bone transducers (BTs) are used to stimulate right and left mastoids simultaneously, signals from each BT reaches both the right and left cochlea. This cross-talk can be further complicated by the possible interaction with air-borne sound at the basilar membrane with the BC sound (Jahn & Tonndorf 1982; Zurek 1986; Stenfelt & Goode 2005b; Rowan &

Gray 2008). Rowan and Gray (2008) proposed a model which showed that if the phase and level of sound arriving at each cochlea from both BTs is known then this would allow for the potential development of a cross-talk cancellation system. A system such as this could be used in bilateral BCHA patients to restore the interaural level difference, a key component for effective binaural hearing (Liao 2010; Majdak et al. 2013). The ability to achieve cross-talk cancellation relies on an increased understanding of phase and level across frequency, as well as understanding how this varies between patients (Zurek 1986).

A common method for investigation of level is via threshold measurements in patients with SSD to calculate transcranial attenuation (TA) (Nolan and Lyon, 1981). Transcranial attenuation can be defined as the difference in thresholds between ipsilateral and contralateral BT placement in an SSD patient (Stenfelt 2012). This method of calculation makes several assumptions, including assuming equal coupling and positioning on both mastoids, as well as skull symmetry with the same resonance and antiresonance properties on both sides. However, it is well known that there can be significant asymmetry in the skull on the right and left sides (Wismer & O'Brien 2010). Therefore, these assumptions maybe useful for elucidating appropriate bone conduction masking levels in audiological testing, but not for calculating the precise interaural level difference in a single patient. Since level can be higher at the contralateral cochlea it can be misleading to describe relative sound levels as attenuation, so we will use the term inter-cochlear level difference (ICLD). Previously we demonstrated that accurate psychoacoustic measurements of the ICLDs and inter-cochlear phase differences (ICPDs) reaching the cochleae could be made in participants with normal binaural hearing.

The present study compared this technique with a psychoacoustic method that employs only bone-conducted sound. The new method employs two bone transducers (BTs) with sound cancelled at one or other cochlea by varying the level and phase of one BT, resulting in a strongly lateralised percept at the contralateral cochlea (Figure 17 a,b). We will call this the "two-BT" method. This two-BT method could be used in a clinical population with conductive hearing loss. The effectiveness of cancellation was assessed by using an additional cancellation signal from the ipsilateral (uncancelled) earphone. If this signal could be adjusted in level and phase such that very little sound was heard, cancellation at the contralateral ear was deemed successful. The comparison method uses a single BT at a time with sound emitted from it cancelled at the cochlea via Etymotic ER2 earphones (Figure 17c, d). This "one-BT" method is not applicable in a clinical population. The two procedures were performed for each of the two techniques, as shown

in Figure 17. The results of phase and level using the one-BT method were then used to calculate expected results from the two-BT method. Expected and actual results were then compared.

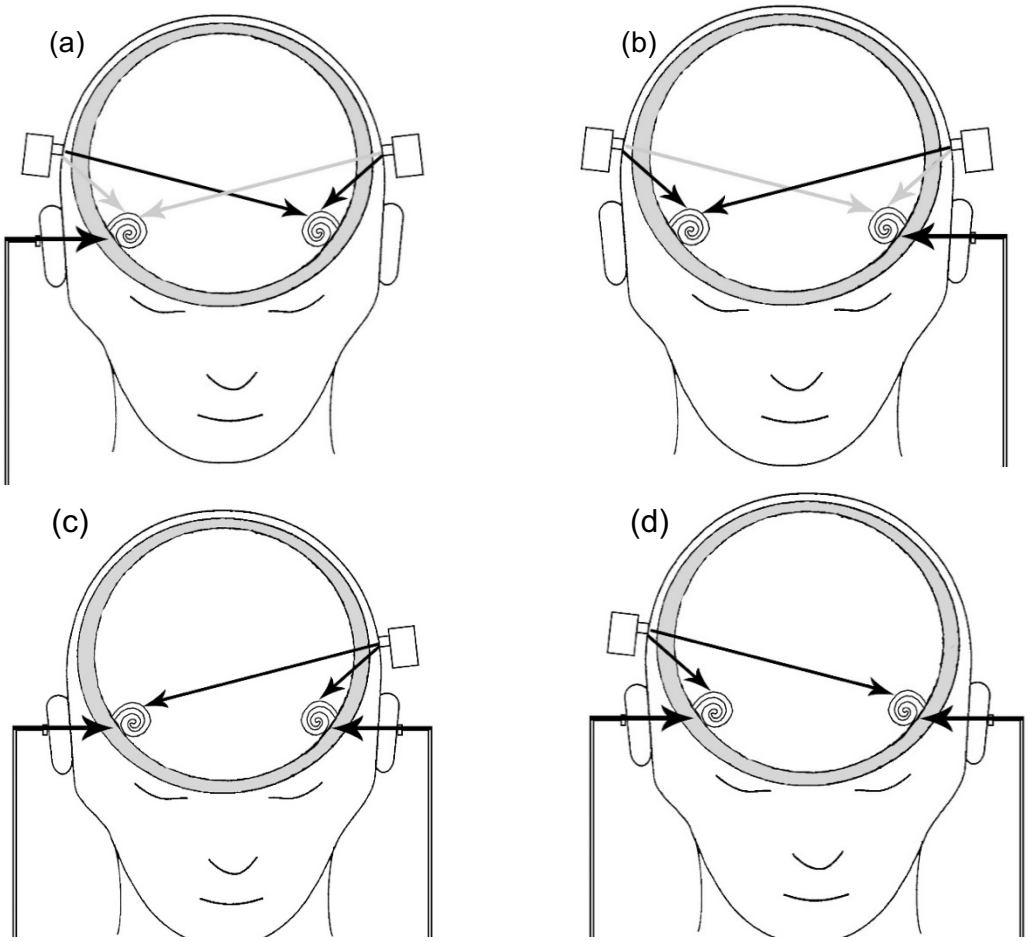


Figure 17 Panels (a) and (b) illustrates sound cancellation at the cochlea by interaction of the two BTs by destructive interference (black arrows). Panel (a) showing cancellation at the left cochlea and (b) at the right cochlea. The resultant signal of the two BT signals (grey arrows) is then cancelled with ER2 earphones at the opposite ear. Panels (c) and (d) illustrate cancellation of sound at the cochlea from a left sided (c) and right sided (d) bone transducer using ER2 earphones.

4.3 Method

4.3.1 Apparatus

Matlab™ 2012 software was used to generate pure tones at a sampling rate of 44.1 kHz over four channels with the ability to vary the level and phase of each channel independently. An 8-channel Darla Echo 24/96 DAC passed signals through an 8 channel Behringer Powerplay Pro-8 Amplifier to Etymotic ER2 insert earphones and two Radioear™ B71 BTs for BC mastoid stimulation. To minimise differences in BT placement between experimental sittings for the same participant and between different participants, specially adapted lens-less glasses were used which had attachments behind the ears holding both BTs in position. The glasses allowed lower variation in BT placement as the superior portions of both pinnae and the bridge of the nose were effectively used as a fixed-point reference tripod for the glasses to rest on. The attachment for the BT onto the glasses positioned the BT 55mm behind the opening of the external auditory canal. This is the recommended surgical placement position (Stenfelt et al. 2000; Battista & Ho 2003). Testing was performed in a single-walled sound attenuating booth (Industrial Acoustics Company) within a sound deadened room.

4.3.2 Participants

Three participants were used (age range 22-29 years old) with normal hearing and no previous history of otitis externa or ear surgery. In order to prevent wax impaction, otological examination was performed on participants before deep insertion of ER1-14B eartips connected to the ER2 earphones. ER2 earphones where selected over open ear headphones to prevent air-borne sound emitted by the bone transducer from reaching the cochlea.

4.3.3 Testing procedure

The following experimental methodology was approved by Cardiff University Psychology Department Ethics Committee. Prior to performing the outlined testing procedure, each participant undertook at least 8 hours of practise sessions. In these sessions participants practised cancellation of a pure-tone signal from a one-BT with ER2 earphones via adjustment of the phase and level of each earphone independently. Participants also attempted multiple frequencies between 0.5-8 kHz using the two-BT technique described below. The aim of this was twofold. Firstly, it was used to determine at which frequencies participants could reliably perform the task and secondly for the participants to be familiar with the task so that results of cancellation were reliable. It emerged that participants found the two-BT task very challenging at frequencies below

1.5 kHz and consequently this was the lowest test frequency chosen for the data collection sessions.

After deep insertion of ER2 earphones, the two BTs were placed on the left and right mastoids, and held in place by adapted lens-less glasses as shown in Figure 18. An elasticated material band was then placed over the participant's head and the BTs.

Figure 18 Image of lens-less glasses with attached B71 bone transducers.

The one-BT method was used first. A pure tone was presented via the BT. A second pure tone, 1 Hz higher than the tone from the BT was presented via the ipsilateral earphone. The participant was asked to vary the level of the earphone-presented tone in order to maximise the perceived beating effect as the two signals constructively and destructively interfered. Beating is known to be maximum when the level of the signals at the basilar membrane are equal (Wever, E.G., Lawrence 1954). Beating maximisation was achieved by changing the level of the earphone-presented sound. Adjustment was made by using a scroller on a computer mouse. Each step of the scroller changed the level by 0.2 dB. This method allowed the level of the two presented tones to be matched at the cochlea. Once the participant had selected a maximal beating level, the cancellation phase could be estimated. The same levels were presented again but using the same frequency in both the earphone and the BT simultaneously. The participant was asked to change the phase of the ER2 presented tone so as to minimise the perceived sound in that ear. Phase adjustment was performed using the mouse scroller, with each scroll changing the phase by 2°. To cancel the signal going to the contralateral ear, the same two processes of level adjustment followed by phase change were repeated using the contralateral earphone

while the level and phase modified cancellation signal was simultaneously maintained on the ipsilateral earphone. In this way, the bone-conducted sound at both ears could be cancelled. Participants could then make further refinements *ad libitum* to the level and phase of the earphones signals at each ear in order to continue reducing the perceived sound as shown in Figure 17a. A graphical user interface allowed the participant to switch between any of the four parameters for adjustment or to indicate that they were satisfied that the perceived sound could be reduced no further. The resulting phases and levels from the earphones needed for cancellation in both ears were recorded for a given BT signal. The same method was repeated with stimulation of the opposite BT at the same frequency as shown in Figure 17b.

Following completion of the one-BT task the two-BT task was performed. Both BTs presented the same pure tone at the same level and phase. Participants were asked to adjust the phase of the right BT in order to minimise the perceived sound in the left ear. Phase adjustment was performed via mouse scroller with a 2° step size, as previously. Participants were then asked to adjust the level (with a 0.2 dB step size) of the right BT in order to minimise the perceived signal at the left ear. Participants could make as many adjustments to the level and phase as deemed necessary to minimise the left ear signal.

At some frequencies participants did not find that there was a variation in perceived level when changing the phase. It was thought that this happens when there is a large level difference at the cancellation cochlea between the two BTs, preventing detection of destructive interference. In such cases the level of the right BT was decreased by 3 dB in order to reduce the level difference and then phase adjustment was re-attempted. If this was unsuccessful a 3 dB increase on the original BT signal was made and phase readjusted. This step down and step up by 3 dB level adjustment was repeated until variation in perceived level in the left ear was achieved.

Once signal cancellation was completed in the left ear using two BTs, the sound at the right ear was cancelled using the earphone in the right ear. This was performed by first matching the earphone level with that of the combined BT signals using the beating technique. Level and phase were then adjusted over the earphone in order to cancel the signal (using the same method as outlined in chapter 3). If participants had achieved cancellation throughout, then no signal was would be audible at either ear, despite both bone transducers and a single earphone producing a pure tone.

Each condition was attempted four times by the three participants. This was performed at eight different frequencies (1.5 kHz and in 1 kHz step between 2-8 kHz)

with both left- and right-sided cancellation, and using both one- and two-BT techniques. Thus at total of sixteen conditions were performed for each frequency per participant. Each testing session lasted approximately 45 min and only tested one frequency. The order at which each frequency was attempted was counterbalanced between subjects in order to minimise practise effects. In seven testing sessions, participants could not achieve cancellation using the two-BT technique. On these occasions a different frequency was attempted and the participant reattempted the failed frequency on the next occasion. This required differing numbers of attempts for some participants. In order for data from an experimental sitting to be included four complete sittings were needed (with left and right sided cancellation, using the one- and two-BT techniques)

4.3.4 Calculations

Mathematical models have been produced showing how two-BT sounds can interact (Zurek 1986; Rowan & Gray 2008). In our equations (which focus on left sided cancellation only), lower-case Greek symbols represent phase shift and gain values at the left or right cochlea (which are directly measured in the one-BT method), while corresponding upper-case Greek symbols represent adjusted values of input signals in the two-BT method. Superscripts R and L refer to the side of the BT and subscripts to the side of the cochlea. Symbols without a superscript correspond to differences between the two-BTs or cochleae at the defined subscript. For instance sound from left BTs arrive at the left cochlea with a resultant phase difference (φ_L^L) and level difference (α_L^L). The diagram on

Figure 19a represents this condition (where squares represent phase and triangles represent level). Similarly the right side BT signal will arrive at the left cochlea with a phase (φ_L^R) and level difference (α_L^R) as shown in

Figure 19b. In order to achieve full signal cancellation at the left cochlea using the two-BT method (as shown in

Figure 19c), the 'source' interaural level difference (A_L) of the BTs must complement the difference in transmission gain to the left cochlea between the two transducers. As shown by:

$$\alpha_L^L - \alpha_L^R = A_L \quad (5)$$

Similarly the 'source' interaural phase difference (Φ_L) must compensate and oppose the phase difference between the sounds reaching the left cochlea from both bone transducers, as shown by:

$$\varphi_L^L - \varphi_L^R + \pi = \Phi_L \quad (6)$$

The resultant level and phase of sound at the right cochlea after left-cochlea cancellation (as shown in Figure 19d) can by predicted from the one-BT method by addition of the two individual BT results with the phase (Φ_L) and level (A_L) shifted signal. Equation 6 shows that the level of the left BT needed for cancellation is $\alpha_L^L - \alpha_L^R$. Thus the gain from the left BT to the right cochlea in that case can be given by:

$$\alpha_R^R + \alpha_L^L - \alpha_L^R = \textit{Source gain} \quad (7)$$

The required phase shift of sound at the left BT for cancellation at the right cochlea is $\varphi_L^L - \varphi_L^R + \pi$ thus the phase shift from the left microphone to the right cochlea in that case is given by:

$$\varphi_R^R + \varphi_L^L - \varphi_L^R + \pi = \textit{Source phase shift} \quad (8)$$

The signals from left BT which have been shifted by phase (Φ_L) and level (A_L) can be combined with the unchanged signal from the right BT at the right ear by vector summation to give the predicted phase and level of the resultant signal at the right ear. Calculation of the x, y components of the resultant vector are shown in Equations 5 and 6.

$$cos(\varphi_R^R + \varphi_L^L - \varphi_L^R + \pi) \times 10^{\frac{\alpha_L^L - \alpha_R^L - \alpha_R^R}{20}} + cos(\varphi_R^L) \times 10^{\frac{\alpha_L^R}{20}} = x \quad (9)$$

$$sin(\varphi_R^R + \varphi_L^L - \varphi_L^R + \pi) \times 10^{\frac{\alpha_L^L - \alpha_R^L - \alpha_R^R}{20}} + sin(\varphi_L^R) \times 10^{\frac{\alpha_L^R}{20}} = y \quad (10)$$

The level of the resultant signal at the right cochlea after cancellation at the left cochlea is calculated by:

$$log10(\sqrt{x^2 + y^2}) = \alpha_R \quad (11)$$

The predicted phase at the right cochlea is given by arctangent of the x, y components, where atan2 is a commonly used programming function, which returns the four-quadrant acrtangent.

$$atan2(x, y) = \varphi_R \qquad (12)$$

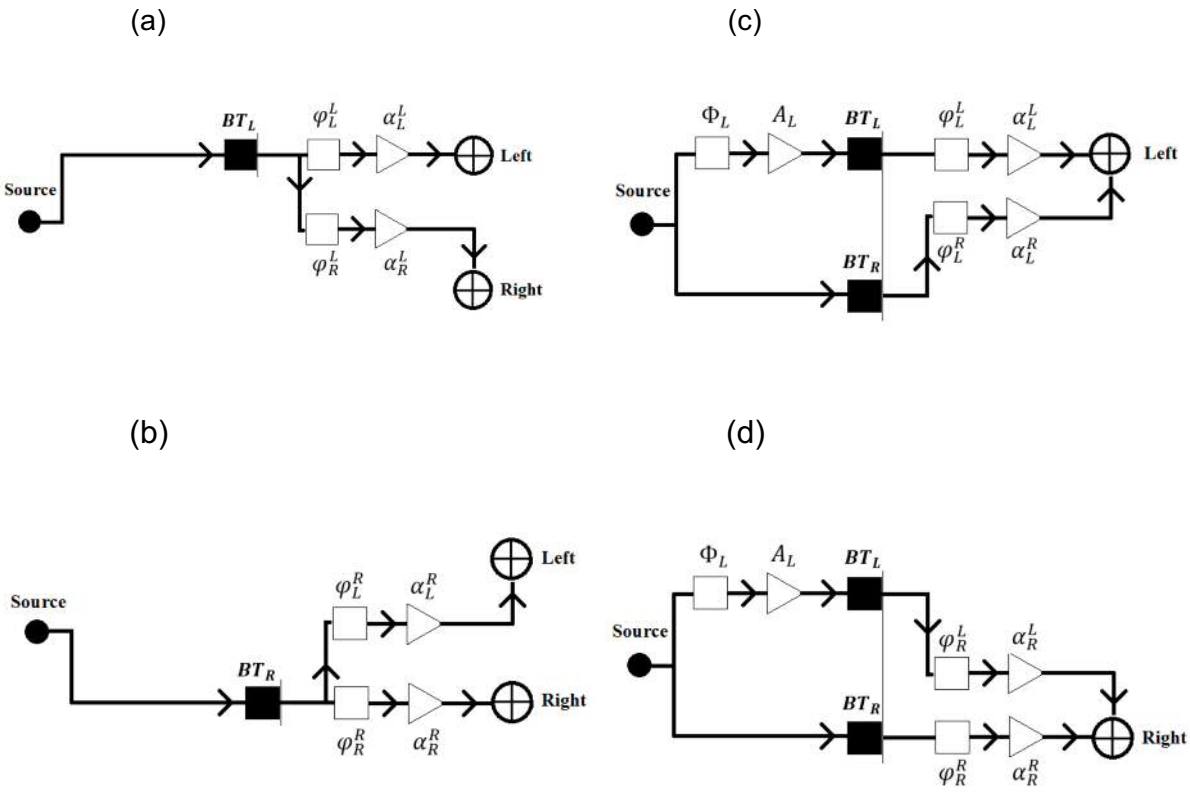


Figure 19 Illustrations of cross-talk cancellation modified from Rowan and Gray (2008) and Zurek (1986). (a) Model of cross talk cancellation using two BTs (see text for details). (b) Model of left-BT stimulation with cancellation at the left and right ear. (c) Model of right-BT stimulation with cancellation at the left and right ear. (d) Model of two-BT stimulation with cancellation at the left cochlea and the two signals interacting to give a phase and level at the contralateral (right) cochlea.

4.3.5 Data comparison methodology

The one- and two-BT phase and level results were compared via differences between pairs of one- and two-BT results of the same frequency. In order to avoid averaging of positive and negative results (which would likely identify a mean of no difference between the techniques) only absolute differences were recorded.

To minimise the effect of participant error on the evaluation of the equivalency of the two techniques, possible erroneous results were filtered. This was primarily performed due to the difficulty of the two-BT task, which meant that on some occasions participants could hear the tone again at the target cochlea after the contralateral sound

was cancelled by the ER2 earphones. Filtering was achieved via calculation of the median phase from the four results at the cancellation cochlea in the two-BT technique. The three results closest to the median where then included for further analysis. A similar method was also performed in the one-BT technique in order to filter spurious results (although they were less common than in the two-BT technique). This was achieved via the same method of median phase calculation from four results and selection of the closest 3 results to the median. Thus, six results, (three from each side in the one-BT task) and a further three results from the two-BT method were available for comparison at each of the test frequencies for the three participants. The one-BT method results were then paired (one left BT and one right BT). The paired phase and level results were utilised in Equations 1-8 in order to predict the two-BT phase and level results necessary for cancellation at the left and right cochlea. The difference between predicted results was then compared to measured results. The mean difference from six results (three from left and three from right cancellation) was then calculated for each participant at each frequency.

4.4 Results

4.4.1 Number of attempts needed at each frequency

For the two-BT cancellation task participants 1 and 2 required two attempts at 1.5 kHz. Participant 2 also required four attempts at 3 kHz before being able to achieve cancellation and participant 3 required three attempts at 6 kHz.

4.4.2 Level difference between techniques

8 results were used to calculate the mean difference in phase and level between the one- and two-BT techniques at different frequencies. The standard deviation of the difference was also calculated. Figure 20a shows the mean difference in techniques for each of the three participants at the cancellation cochlea in the two-BT technique (ipsilateral). The smallest mean difference overall between techniques was found at frequencies between 3 and 5 kHz where there was a mean difference of 0.93 dB. The mean difference in level at the ipsilateral cochlea over all frequencies was 1.81 dB. The highest frequencies had the greatest difference between techniques.

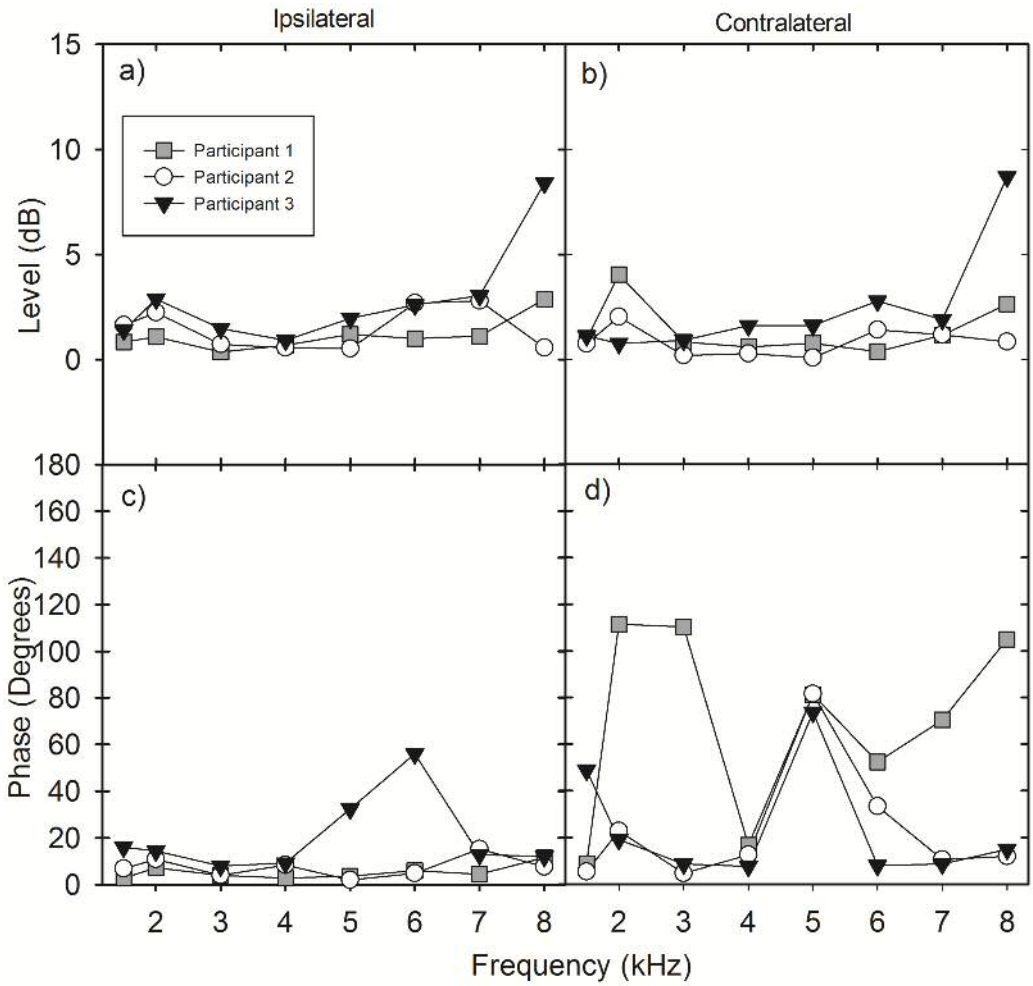


Figure 20 Difference between the predicted level and phase using the one- and two-BT techniques for each participant. Error bars show the standard deviation of the differences between the two techniques (n=6 per frequency result).

Figure 20 b and show the level differences between the two techniques of the contralateral cochlea from cancellation. The highest correspondence between techniques was again at 3-5 kHz. The mean difference was 0.77 dB within this range and 1.14 dB over all the test frequencies. An analysis of variance (ANOVA) was performed on the all signed difference level results. This found a significantly greater variance contralateral level difference results when compared to ipsilateral level results ($F = 6.292$, $p<0.01$).

4.4.3 Phase difference between techniques

Figure 20 c and show the difference between techniques in phase at the ipsilateral cochlea. The phase differences between participants appeared more variable than level results. However, differences in technique were again smallest over the 3-5 kHz range. The mean difference was 8.3° within this range and 11° over all the tested frequencies.

The phase-difference results in the contralateral cochlea had the greatest variation (Figure 20 d). All participants were found to have a large difference in results from the

two techniques at 5 kHz when compared to other frequencies. There was a mean difference of 78.8° at this frequency and 38.4° overall. An analysis of variance (ANOVA) was performed on the all signed difference results of phase. This found a significantly greater variance in contralateral phase results when compared to ipsilateral results ($F = 3.744$, $p<0.05$).

4.5 Discussion

4.5.1 Ipsilateral level and phase

We have shown that it is possible to perform psychoacoustic measurements of phase and level in order to measure the cross-talk signal using one- and two-BT methods. There was a high degree of concordance between techniques at the cancellation cochlea in both phase and level as indicated by mean differences of 1.14 dB and 11°. This error if introduced to two originally matching signals would result in a 8.17 dB attenuation of the combined signals. Thus, in the two-BT technique we have shown that participants are able to detect lateralisation from ICLDs between frequencies of 1.5 and 8 kHz. Phase and level differences between techniques was smallest at frequencies between 3-5 kHz. The increased differences between techniques at <2 kHz were likely caused by the greater influence of ICPD (Zhang & Hartmann 2006), which introduces potentially conflicting cues. However, the greatest differences were found at higher frequencies. One possible explanation for these findings may be related to the reduced output level from the B71 BT at higher frequencies. It maybe that distortion may have occurred at higher frequencies, which would mean the single BT technique could have underestimated levels necessary for cancellation. This would explain why the differences increase with greater frequency.

Participants found the two-BT technique more challenging than the one-BT task with some participants requiring reattempts of particular frequencies on a different sitting. Participant 2 had three attempts at 3 kHz before on the forth sitting being able to produce reliable results. Participant 3 also had two attempts at 6 kHz before successfully completing the task on the third attempt. There was no apparent agreement between participants as to which frequencies were hard to perform except at 1.5 kHz where participants 1 and 2 both had two attempts.

There are two possible explanations for why some participants found the task difficult at particular frequencies. In chapter 3, it was found that over different frequencies there may be up to 20 dB variation in level at the cochlea over a 0.3 kHz frequency range. Stenfelt et al (2000) described the frequencies over which these large variations occur as

areas of antiresonance. If one of these antiresonance frequencies was close to the test frequency, then this would cause a large disparity in levels reaching the target cancellation cochlea from each of the BTs. The large level difference makes the task significantly harder to achieve, as level matching has to occur before phase changes between the two BTs will cause enough destructive interference to induce lateralisation. Another situation in which the two-BT task can be challenging is when there is little or no ICPD between the two cochleae for each BTs. Thus, when one cochlea is cancelled there is also a degree of cancellation at the opposite cochlea. This makes the task difficult as very small change in phase can cause lateralisation to change from one cochlea to the other. The most challenging situation to encounter in the two-BT task is a combination of a small ICPD and large level difference.

In chapter 3 we showed that it was possible to accurately measure the phase and level at the ipsilateral cochlea using the one-BT technique. However, the ultimate aim of accurate measurement of phase and level is to allow the creation of a cross-talk cancellation system for bilateral BCHA users (as outlined in the introduction). This rules out the use of earphones because most patients with bilateral BCHAs are prescribed them due to conductive hearing loss, which obstructs airborne sound from reaching the cochlea. Thus, in order for this technique to be clinically applicable, a BCHA-only measurement technique needs to be employed. Within this study we have shown that the two-BT method can give equivalent results between 1.5-8 kHz to the one-BT method. Further research is now needed in several areas.

First, the effect on cancellation results of an open external auditory cancel when compared to closed needs to be investigated. To those with normal hearing it may be difficult to perform cancellation with open ears, as airborne sound from the transducer will interfere with the bone-conducted signal. However, for those with conductive hearing loss it may not significantly affect the task. Additionally, we have found that the two-BT task is difficult to perform at low frequencies (<1.5 kHz), so for a cross-talk cancellation system to be as effective as possible a different method of calculating the phase and level needed for cancellation might be required at these lower frequencies. One possible way of achieving this is to use knowledge gained via our laser-Doppler vibrometry. However, this method revealed rigid body motion at frequencies <0.5 kHz. Under these circumstances, there should be no ICPD. Thus, any cross-talk cancellation would also cancel the desired signal. This would mean measurement of the phase and level at these low frequencies would be of no clinical value. Since modal vibration can be clearly seen

within our Doppler results it might be possible to use the head circumference measurements from normal hearing participants to predict the changes in vibration mode at frequencies between 0.5 kHz (where rigid body motion predominates) and 1 kHz (where a mass-spring effect is observed). This would then allow the creation of a simpler two-BT task that only requires adjustment of level to perform cancellation at one cochlea, removing the difficulty that phase adjustment adds to the task.

4.5.2 Contralateral level and phase

At the cochlea contralateral from cancellation in the two-BT technique, there was high concordance between techniques with regard to the level (mean difference 0.77 dB) but poor correspondence for the phase (mean difference 38.3°). Having an accurate method of predicting or measuring level at the cochlea contralateral from cancellation is of importance as it will be key in correcting sound level in a full cross-talk cancellation system. This is particularly important when there is relatively little ICPD difference (<30°) as part of the desired signal will be cancelled. In order to correct for this the, level of both sides needs to be increased. A further instance when modification of level may be necessary is when the ICPD is close to being out of phase. This results in a degree of signal summation causing the audible signal to be greater than desired. In chapter 2 and 3 we showed that at low frequencies (<0.75 kHz) there is little or no ICPD. Therefore, signal summation is greatest over this frequency range. Since cross-talk cannot be performed if the ICPD is small (a cross-talk ill condition) it has been suggested that it may be of benefit to match the phase in order to cause maximal signal summation (Deas et al. 2010). This could have potential clinical benefits, since many patients with bilateral BCHAs do not have a pure conductive loss, (Bosman et al. 2001). In such instances summation could be desired in order to make the signal louder (Deas et al. 2010). Further work needs to be performed to investigate how often contralateral cancellation and summation happens between 0.25-8 kHz.

We have shown that there are large differences in the predicted contralateral phase results when compared to the predicted ipsilateral phase results. We believe this is primarily caused by frequencies where there is little ICPD. In such instances, small discrepancies between the cancellation results of the one and two-BT techniques can result in large changes in the phase at the contralateral cochlea. This is because the level of destructive interference causes the phase to change significantly. Fortunately, knowledge of the contralateral phase of the resultant signal after cross-talk cancellation is of less functional use. Since it is the ICLD signal, which is the target of modification.

We have already shown that attempted manipulation of the phase differences at frequencies lower than 1.5 kHz may be of limited benefit.

4.6 Conclusion

These findings show that cross-talk signals can be measured accurately using the two methods to give equivalent results. This is significant since accurate measurements of phase and level at the cochleae over a wide frequency range have not been previously possible. It is these values that are required for cross-talk cancellation.

Participants found the two-BT method more challenging to perform when compared to the one-BT method. Although it is potentially applicable in a clinical population with conductive hearing loss as it does not employ earphones. A further drawback of the two-BT method is that it can be very challenging to perform reliably at frequencies less than 1.5 kHz. This is likely due to two reasons. Firstly, the ICPD between the cochleae can be negligible at low frequencies. This means a small change in phase can cause lateralisation to change from one cochlea to the other. Secondly, the ICPD is the primary method of sound localisation at these low frequencies and the method relies on the ability of the participant to able to use level cues to perform cancellation. Having the level and phase cues competing can make the procedure challenging.

The one-BT technique (similar to the method Békésy described in 1947) can be used over the full frequency spectrum but is not clinically applicable to a conductive hearing loss population (since earphones are required) and takes longer to perform than the two-BT method. Further research is needed to investigate methods of making the two-BT procedure easier and faster to perform as well as allowing the successful collection of data at lower frequencies. One possible method to attempt to overcome this challenge is to perform cancellation at a frequency which is relatively easy to perform (for example 2 kHz) and then to use the results to make small reductions in frequency using the previous results. This means that the participants' phase and level measurements will be close to the actual cancellation phase and level. This method is employed in the following chapter. If the approach works, then results could be used to create a fixed filter needed for cross-talk cancellation. If employed in bilateral BCHA users, this could have significant benefits in terms of speech understanding in background noise as well as sound localisation.

5 Unilateral Cross-talk cancellation in normal hearing participants using Bilateral Bone Transducers

5.1 Summary

Following demonstration of the equivalency of the single and two bone transducer (BT) techniques we now aim to show that the phase and level measurements can be used to improve hearing thresholds by using the phase and level values to create a single sided in cross-talk cancellation system. In order to be able to achieve this a faster data collection method was created. This uses the previously measured phase and level results to interpolate likely results for new test frequencies. Be doing this the testing time to collect the necessary phase and level values was reduced to approximately 20 min. This was necessary since participants were normal hearing volunteers without and abutments, meaning testing tone and speech reception thresholds with and without cross-talk cancellation had to be performed without adjustment of the BT.

The inter-cochlear phase difference (ICPD) and inter-cochlear level difference (ICLD) were consistent between experimental sittings in the same participant but different between participants. Use of cross-talk cancellation showed a significant improvement in tone and speech reception thresholds (12.1dB and 13.67dB) when compared to no cross-talk cancellation.

5.2 Introduction

There are relatively few studies which have investigated the benefits of bilateral BCHAs. Those which have found improvements in sound field average tone thresholds in adults (2-15 dB) (Bosman et al.; Priwin et al., 2004) and speech recognition thresholds in quiet (4.2 dB) (Bosman et al. 2001) when compared to unilateral fitting. However, these benefits may be purely due to amplification from two hearing aids rather than increased ability to process sound binaurally. In order to investigate true binaural processing advantages Binaural Masking Level Differences (BMLD) have been used. These have shown significant benefit (6-6.1 dB) at low frequencies (125-500 Hz) but little benefit at higher frequencies (Bosman et al. 2001; Priwin et al. 2004). Sound localisation judgements have also been shown to improve from 22.2 % and 24.3 % at 0.5 kHz to 2 kHz to 41.8 % and 45.3 % respectively (Bosman et al. 2001). This shows that there is a true binaural advantage although it is severely limited due to cross-talk (Deas et al. 2010).

Cross-talk cancellation was originally theorised by Bauer in 1961 in order to more accurately reproduce binaural recorded signals from two loudspeakers. This was later put into practice by Schroeder and Atal in 1963 to overcome the problem of signal crossing from one speaker to be audible at both ears (as discussed in chapter 1). Several different methods of cross-talk cancellation have been developed. However, they all attempt to implement the theoretical "ideal cross-talk cancellation" taking into account real world limitations such as the dynamic range of the amplifier or transducer. This is problematic because ideal cross-talk cancellation has the potential to require high output levels in order to cancel signal when the two signals are close to being in phase in order to present them at the correct level. This is because destructive interference will occur to a large proportion of the signal. In this 'ill-condition' where the signal phases are close it can leave the system very prone to small measurement inaccuracies as well as head movement. Thus over frequencies where there is little interaural phase difference cross-talk cancellation cannot be achieved reliably.

In chapter 4 we showed how the phase and level using a two-BT method give equivalent phase and level results at the cochleae when compared to measuring them separately via earphones. Within this study we aim to show a faster method of calculating the phase and level results necessary for cross-talk cancellation. This allows participants to cancel a pure tone at one frequency at the target ear. Once achieved participants can then adjust the frequency in 20 Hz increments using a mouse scroller. When the frequency is adjusted the previously cancelled signal again becomes audible the participant can then make further adjustments to the phase and level difference in order to cancel the new frequency tone. This is advantageous as the phase and level needed for cancellation only need to be varied by a small amount to optimise the cancellation rather than starting from an unknown point (as was the case in the one-BT technique outlined in chapter 3). This also means that there are not frequencies that cannot be cancelled without multiple attempts on different sittings (due to large interaural level differences) as were encountered in chapter 4. In addition to this, a data fitting algorithm which predicts the level and phase change based on data already inputted can also be utilised. By making these changes we aim to show:

- That the two-BT cancellation using the frequencies in close proximity method can greatly increase the speed of data gathering as well as the accuracy of cancellation results.

- Despite slight variations in BT positioning and coupling to the skull that there is a strong concordance between level and phase results on multiple sittings in the same participant.
- Show the cross-talk measurement values can be used to give an improvement in signal to noise ratio as shown via an improvement in tone detection thresholds at a number of different frequencies.
- Show that the cross-talk values can also be used to improve speech reception thresholds.

The two-BT technique is the highly preferred method since data collection speed is greatly increased. This is desired since normal hearing participants are employed for testing (who have no abutments). In chapter 2 we showed that if the BT is removed and then replaced this are small differences in coupling and position which means that phase and level measurements previously made are no longer valid. Therefore, phase and level measurements as well as the creation of a cross-talk cancellation system and its testing must be performed in one experimental sitting. This means that time of data collection is critical, firstly because a single long testing period may cause participants to fatigue but also that wearing a Softband with two-BT for a long period can be uncomfortable (Verhagen et al. 2008). Although in a normal hearing population measuring reliable phase and level measurements is challenging we believe that if this same approach was implemented in a patient population the same time limitations and would not apply. This is because an abutment would likely give a constant position and coupling even if removed and replaced.

5.3 Method

The outlined methodology was approved by Cardiff University Psychology Department Ethics Committee.

5.3.1 Apparatus

Sound presentation and data calculation was performed with the use of Matlab®. A USB ESI MAYA44 USB+ four-output-channel DAC was used in conjunction with an 8-channel Behinger Powerplay Pro-8 Headphone amplifier to pass audio signals to two B71 (Radioear) bone transducers (BT). A pair of Etymotic ER2 insert earphones with ER1-14B eartips were inserted into the ears of the participants to prevent air born sound radiated from the BTs interfering with the cross-talk cancellation results. BT placement was the same as outlined in chapter 3 with no adjustment of the BT positioning once measurements of phase and level had begun. All testing was performed in a single-walled

Industrial Acoustics Company (IAC) sound attenuating booth within a sound treated room. A computer screen was visible outside the booth window with a keyboard and mouse inside the booth for participants to adjust phase and level differences as well as input transcripts.

5.3.2 Participants

Three participants aged between 21 and 29 years old were recruited from Cardiff University and were paid for each testing session. All had previous experience with psychoacoustic experiments, were primary English speakers and had self-reported normal hearing with no previous history of ear pathology. Otoscopic examination prior to testing was normal. All participants had performed at least 5 hours of testing in order to cancel sound via the adjustment of phase and level in either the single BT task (outlined in chapter 3) or bilateral and unilateral task (outlined in chapter 4). Participants underwent five trials of experiment A (with cancellation on the right and left side) in order to obtain phase and level data for use in a prediction algorithm. Following this, participants performed experiment A with the aid of a prediction algorithm. This allowed the speed of data collection to be increased greatly. Immediately after completing 'fast' experiment A they underwent one trail of experiment B or C with use of the cancellation phase and level results they had just completed. The experimental order is outlined in Figure 21.

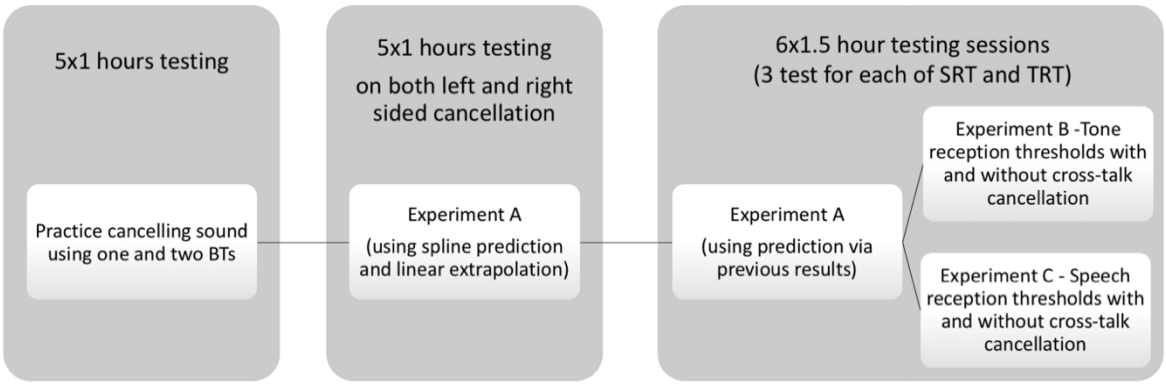


Figure 21 Outline of the experimental order

5.3.3 Experiment A: Initial Phase and Level Prediction and data fitting

Two methods were employed in order to perform data fitting and prediction. In the initial five sittings result prediction employed a cubic spline interpolation utilised from Matlab® curve fitting toolbox. This was used to predict the phase and level of cancellation between two or more known frequencies. Spline interpolation is a numerical analysis method which fits input data to a piecewise polynomial. It is particularly suitable for data fitting related to level difference which can fluctuate considerably over a narrow

frequency band as well as having a variable number of peaks and troughs that can occur. Spline interpolation was used instead of other data fitting methodologies such as via high order polynomials as they would encounter the problem of the Runge's phenomenon (Tolm 2014) whereby large errors occur in prediction between the known cancellation values. Data fitting via a moving average would also not be appropriate as it would underestimate the cancellation levels during frequencies where signal summation or destruction are occurring. Phase and level results were analysed separately using the same spline prediction methodology. If a participant found that a cancellation result was suboptimal they could reattempt it and only the most recent result for the attempted frequency would be included for data fitting.

The data prediction methodology for frequencies greater or less than those already attempted used linear interpolation of the closest three frequencies in order to predict the target frequency. Additionally, safety mechanisms were built in so that if the predicted level was above a loudness threshold the algorithm would present the mean level of the three values instead of the level predicted via linear interpolation. This was necessary to prevent very loud tones being presented if there was an increasing level slop in the previous values.

By employing the outlined prediction techniques, the data collection time could be reduced to approximately 50 min. If a similar technique had been used as that described in chapter 3 via a one BT technique the method would have taken approximately 16 hours for each sitting.

5.3.4 Experiment A: Secondary Phase and Level Prediction and data fitting in preparation for tone and speech reception thresholds

In order to further increase the speed of phase and level data collection a different data prediction algorithm was used prior to Tone and Speech reception threshold testing. This was necessary due to the discomfort of wearing a relatively tight headband for a long period of time. The mean phase and level was calculated every 20 Hz between 1-5 kHz using results from the five separate cancellation attempts. The participant would attempt cancellation using the predicted phase and level value results. Adjustments to the phase and level differences between the two-BT could then be made via the use of a mouse scroller to find the true values. The participant would then change the frequency and using the mean phase and level results for cancellation from the initial five trials the algorithm would change the phase accordingly. For example, if the participant attempted 3 kHz and found the phase difference to be 20° and the mean change between 3 kHz and 3.1 kHz

3 from five sittings was 200° then the computer would present a phase difference of 220° at 3.1 kHz. This could then be adjusted by the participant using the same mouse scroller method. If no sound was perceived at the cancellation cochlea the frequency could be further adjusted until sound is heard at the target cochlea.

Following phase adjustment, the participant would then adjust the level. Once again the algorithm would use the mean change (using the five previous the mean results) between the previously attempted frequencies to predict the new frequency. This method allowed to correct for differences in coupling between testing sessions as well as in differences in starting phase. Via implementation of this methodology accurate cancellation results can be achieved in 15 minutes.

5.3.5 Experiment A: Cross-talk cancellation data collection method

Participants were initially asked to cancel audible sound at the left or right cochlea using the two-BT technique already described in chapter 4. The initial starting frequency was 3 kHz. If participants could not cancel sound at this frequency, then initial attempt frequency was increased by 200 Hz until cancellation was possible. Once an initial cross-talk cancellation result had been achieved the participant increased the presentation frequency by 200 Hz and again attempted cross-talk cancellation. The phase and level difference from the previous attempted frequency remained the same. Thus, usually only a small adjustment in phase and level was needed in order to achieve cancellation. This considerably reduced the difficulty and time for participants to again achieve cancellation. During this process the values of level and phase difference as well as the frequency were displayed on the screen. Participants were told that in most cases an increase in frequency would result in an increase in phase difference. A further iteration of increasing the frequency by 200 Hz and keeping the previous phase and level difference settings was performed. Once the cancellation program had at least three phase and level results from different frequencies it could start predicting the phase and level needed for cancellation based on the previous results (as outlined above). Participants were asked to continue to cancel audible sound at the cancellation cochlea at least every 200 Hz up to 5 kHz. Once cancellation had been attempted between 3-5 kHz participants were asked to cancel frequencies at least every 100 Hz starting at 2.9 KHz down to 2 kHz. From 2 kHz down to 1 kHz participants attempted a cancellation frequency at least every 60 Hz.

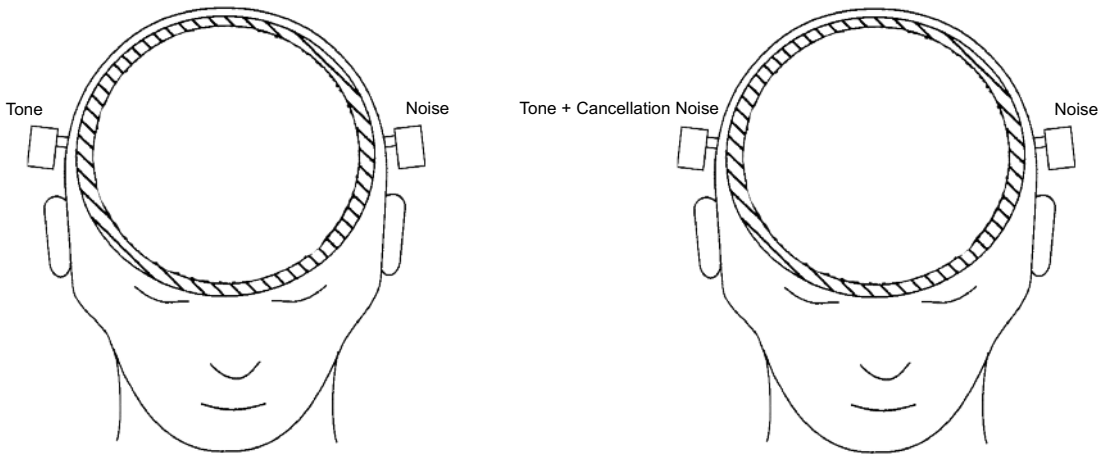
Once frequencies had been attempted between 1-5 kHz participants could use the mouse scroller to change the frequency and the prediction algorithm would simultaneously present what it viewed as the likely level and phase difference needed for

cancellation. Participants then had the opportunity to attempt further frequencies where the tone became audible. If a frequency had previously been attempted only the most recent level and phase would be used in the prediction algorithm. This gave a method for correcting mistakes by the participant. Participants were told to perform a sweep from 1-5 kHz and back down to 1 kHz without the need for corrections before the testing session finished.

5.3.6 Experiment B: Tone Reception Thresholds (TRT) with and without cross-talk cancellation.

5.3.6.1 Stimuli Creation

Speech shaped noise interferers were made via filtering Gaussian noise with a 512-point finite impulse response which was matched to long term excitation pattern of speech (Moore & Glasberg 1983; Lavandier & Culling 2010). The 4 second length of noise was then band-pass filtered to match the frequency over which cancellation had been performed (1-5 kHz). All the interferers had a greater length than the targets. In the noise only condition (without cross-talk cancellation) twenty individual monaural noise recording were prepared and used at random in the threshold task.



Within the cancellation noise condition the noise was converted into the frequency domain to obtain the real phase and level components. The phase and level differences from the two-BT cancellation task (which the participant had just completed) were used to alter the level and phase by the cancellation values. The new 'cancellation noise' was then converted back into the time domain. A binaural recording of the noise and cancellation noise was then generated. Twenty such paired noise and cancellation noise recordings were prepared and used at random in the threshold task described below.

Figure 22 Showing two conditions a) shows pure tone on one BT and noise on the contralateral BT b) shows the addition of cancellation noise at the BT with the tone.

5.3.6.2 Procedure

Each participant performed 12 runs of detection thresholds (two conditions at six frequencies) which lasted approximately 45 minutes. In order to assess how effective cross-talk cancellation can be at different human filter bands, pure tones were tested approximately every 20 equivalent band widths (ERBs) between cross-talk cancellation frequencies (1-5 kHz). Test frequencies were 1200,1530, 1945, 2475, 3150 and 4035 Hz.

Each run utilised a 2-down/1-up adaptive threshold measurement task (Levitt 1971), with 12 reversals. A 4 dB step size was used for the initial two reversals and 2 dB in subsequent reversals. The average signal level from the last eight reversals was recorded as the threshold level. Each trial consisted of a two interval, forced choice task. Each interval lasted 2 seconds with a 0.5 second pause between intervals. The target tone contained within one of the intervals was 0.5 seconds in duration. The participant indicated via button press on a computer terminal which interval contained the target tone. Intervals with and without a target tone were presented in a random order and trial-by-trail feedback was given. The conditions (as shown in Figure 22) as well as the order of frequencies attempted were randomised to minimise practise affects.

5.3.7 Experiment C: Use of cross-talk cancellation results in order to improve speech reception thresholds.

5.3.7.1 Stimuli Creation

Gaussian speech shaped noise which was then band limited between 1-5 kHz was produced using the same method as the tone reception thresholds. As in the previous experiment within the noise only condition twenty individual monaural noise recording were prepared and used at random in the threshold task. Similarly, twenty stereo recordings were made with noise on one channel and cancellation noise on the other channel.

5.3.7.2 Procedure

Target speech was from a male voice ("CW") from MIT recordings of the Harvard sentence list (Rothauser et al. 1969). The target speech sentences were band limited between the previously performed cancellation frequencies (1-5 kHz). A modified version of Plomp's (1986) 1-up/1-down adaptive threshold task was undertaken to obtain

speech reception thresholds (SRTs) where ten sentences were used to examine each condition. To achieve this semantically and grammatically unpredictable sentences were employed. An example of such a sentence include "PLUCK the BRIGHT ROSE WITHOUT LEAVES" where keywords are highlighted in capitals. The task aimed to ascertain the signal to noise ratio where there is 50 % intelligibility.

At the start of each condition the initial SNR for the first target sentence was highly unfavourable. On the initial payback the listener was instructed to press the "return' key on the keyboard in order for the stimulus to be repeated, this time at a 4 dB higher SNR. If the participant judged that they could elude two or more target words, then they would enter the proposed transcript into the computer program via the keyboard. If one or more of the predicted target words matched the target, then the program would display the target sentence on the screen. The participant would then self-mark the transcript before moving on to the next target sentence. If no words were the same as the target the same target sentence would be presented at a 4 dB more favourable SNR. The remaining nine sentences were presented only once where the threshold increased by 2 dB if three or more target words were correctly identified, otherwise the SNR for the following sentence would decrease by 2 dB. The average threshold from the last eight SNR results were used as the SNR for the examined condition. To ensure validity of results the typed transcriptions with individual self-scoring results were also recorded to ensure the participant complied with instructions.

5.4 Results

5.4.1 Experiment A results

Figure 23 shows the phase differences necessary to cancel perceived sound at the left and right cochlea in three participants between 1-5 kHz on five separate experimental sittings. Figure 24 shows the level differences needed for cancellation at the left and right cochlea on the same five experimental sittings.

It is clear that within the same participant there are similar patterns of phase progression on different sittings. In addition to this there are large reductions in the level necessary for cancellation over a narrow frequency bands. This is most pronounced on the right side in participant 1 at 3.2 kHz and on the left side in at 2 kHz in participant 2. A reduction is also visible on the left side in participant 3 at 1.7 kHz. During all of these instances there is an associated change in the phase progression where the phase decreases by 180° before resuming the previous phase progression rate.

Although difficult to quantify there are clear patterns of level variation on the same side within the same participant on multiple different sittings. The pattern of level variation also appears to correspond poorly when comparing left and right sides in the same participant. This indicates that the level variations which are seen are likely due to local signal summation/destruction or resonance/antiresonance rather than whole skull resonance/antiresonance which have a greater role at frequencies less than 1kHz.

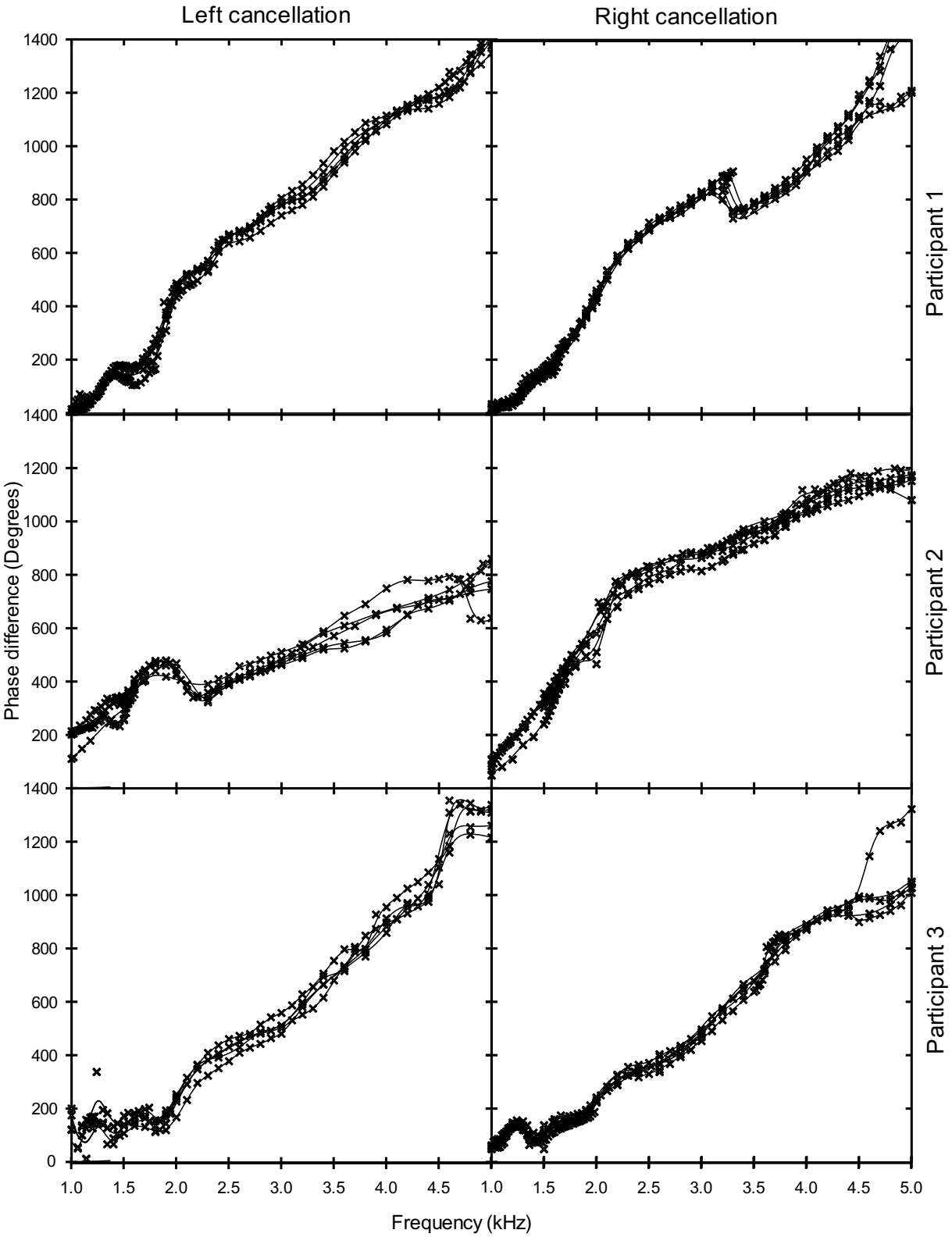


Figure 23 Showing the phase difference needed between bilaterally placed bone transducers to cancel perceived sound at the left and right cochlea on 5 different sittings in three different participants. Line of best fit created using spline fitting method.

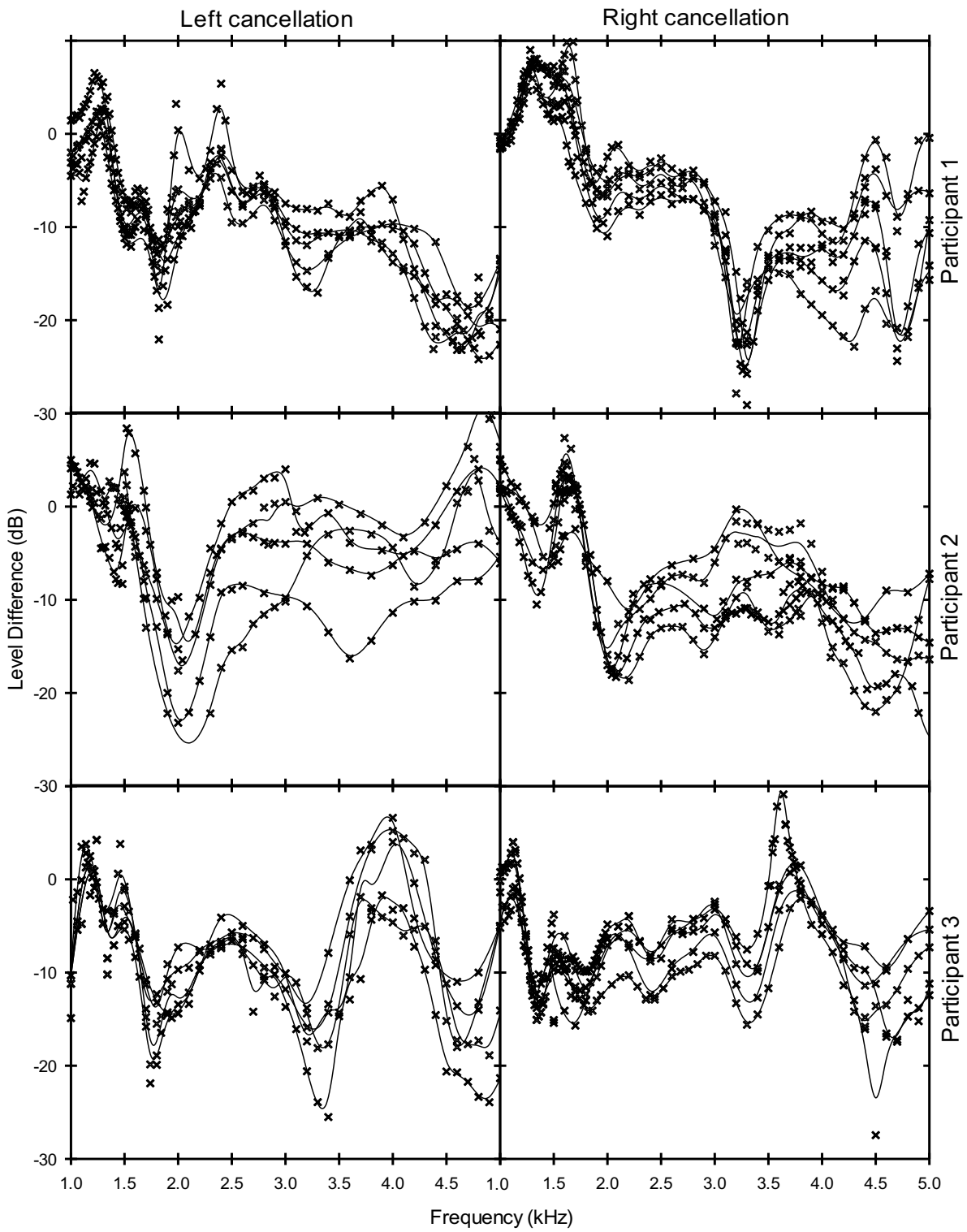


Figure 24 Showing the level difference needed between bilaterally placed bone transducers to cancel perceived sound at the left and right cochlea on 5 different sittings in three different participants.

5.4.2 Experiment B results

Figure 25 shows the mean TRT with and without crosstalk cancellation. A one-way repeated measured analysis of variance (ANOVA) was conducted to investigate if there was a significant difference in tone reception thresholds with and without the addition of cancellation noise (N=54) The results of the ANOVA indicated a significant improvement in thresholds overall (Wilks Lambda = 0.163, F = 246.6, p <0.0001).

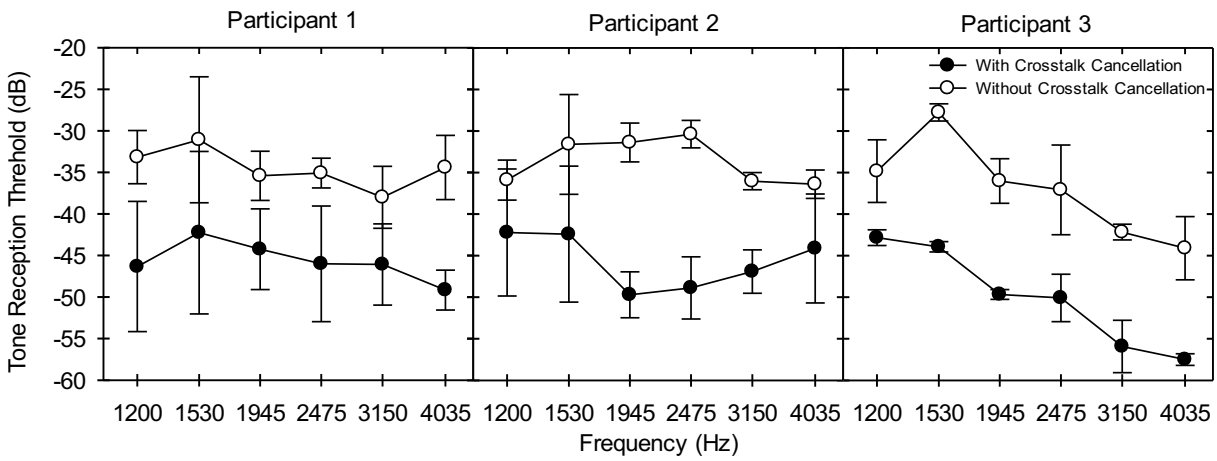


Figure 25 Tone reception threshold with an without cross-talk cancellation in three participants (n=3 per condition) error bars show standard deviation.

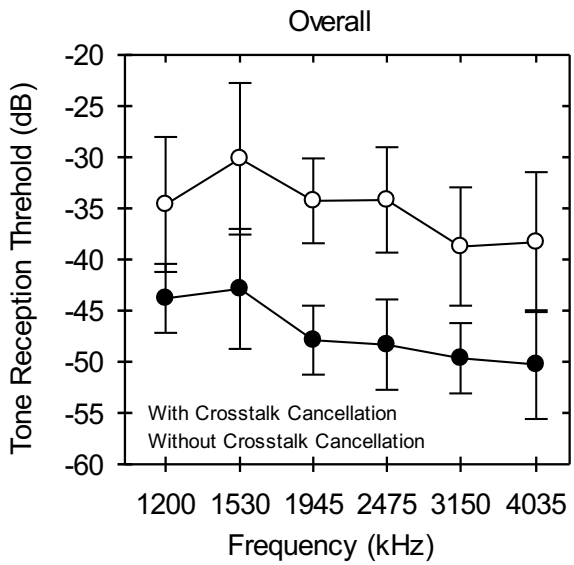


Figure 26 Mean tone reception thresholds with and without cross-talk cancellation (n=9 per condition) error bars show standard deviation.

Figure 26 shows the mean TRT overall. There was a mean improvement in TRT of 12.1 dB with the addition of a cross-talk cancellation signal. The greatest relative improvement in TRT was at 2475 Hz where there was a 14.1 dB difference. The smallest difference was found at the lowest frequency of 1200 Hz where a 9.1 dB difference was found between conditions.

5.4.3 Experiment C

Figure 27 shows a box plot of the SRT with and without the use of cross-talk cancellation. A one-way repeated measured analysis of variance (ANOVA) was again conducted to investigate if there was a significant difference with and without the use of cross-talk cancellation. The results of the ANOVA indicated a significant improvement in thresholds overall (Wilks Lambda = 0.039, F = 792.65, p <0.0001). Participants had a mean improvement of 13.67 dB (n=33) with the addition of cross-talk cancellation noise.

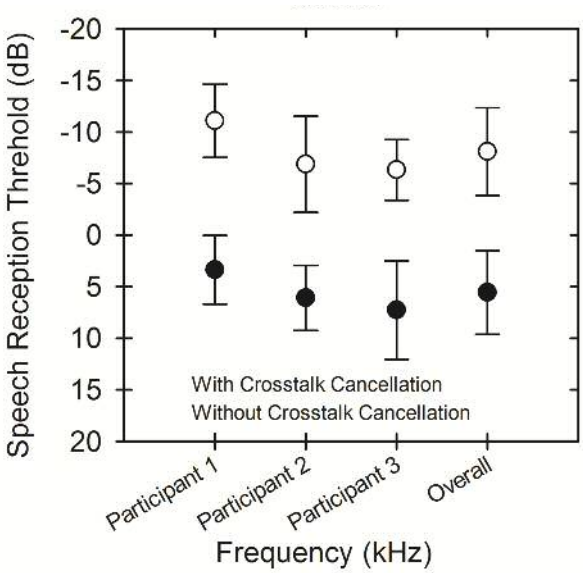


Figure 27 Box plot showing the SRT with and without the use of cross-talk cancellation in three participants.

5.5 Discussion

5.5.1 Experiment A Cancellation results

All participants phase progression had similar elements with frequency. There was greater test retest variability at high and low frequencies when compared to mid (2-4 kHz) frequencies. All participants phase progression was non-linear between 1-1.5 kHz (as was identified in chapter 3). Participant 1's right sided cancellation results showed the phase 180° change in phase between 3.2-3.4 kHz. This is encountered when two signals destructively interfere. In this case the phase progression from both BTs was different (as shown in chapter 3) and must have caused destructive interference not only at the cancellation cochleae but also the cochleae where the signal is aiming to be preserved. This is supported by fact that there is a corresponding reduction in cancellation level over the same frequency spectrum. This is an example of an ill condition over which cross-talk cancellation could not successfully be undertaken.

Prior to experimentation it was theorized that phase progression with frequency will likely be approximately the same between the left and right side. This is because (as found in chapter 3) phase progression is relatively linear (at higher frequencies where model variations are not occurring) and is likely related to head size. Thus if BTs are symmetrically placed phase progression should be similar on both sides. This is seen in participant 2 and 3 where phase progression between 2.5-4.5 kHz was approximately 370° in participant 2 bilaterally and 550° in participant 3. Participants 1's phase progression was 560° for left cancellation and 400° for right cancellation. The difference between sides (160°) being due to the 180° phase inversion already discussed.

There were clear consistencies in test retest level cancellation results within the same participant. These results indicate that the despite small differences in BT placement that impact upon reliability that there is still are large degree of concordance in results. However, as was identified in chapter 3 as well as in previous studies there was great variation between sides and between participants (Håkansson et al. 1986; Khalil et al. 1979; Stenfelt & Goode 2005b; Håkansson et al. 1994)

All participants within this study had self-reported normal hearing, thus the number of possible bone conduction routes are greater than that of a patient with a conductive hearing loss (as shown in chapter 1). Examples of possible interactions are via inertia of the ossicles, radiation into the ear canal, inertia of the inner ear fluid and compression and expansion of the cochlea walls. Thus it could be theorised that if patients with conductive hearing undertook the same cross-talk cancellation experiments that they may have

potentially have less level variation as many of the bone conduction routes still rely on normal anatomy of ear canals, tympanic membrane and ossicles. All of which can be absent/abnormal in patients fitted with bilateral BAHAs. This could have implantations for further improvements in the speed of performing the cross-talk task.

5.5.2 Experiment B

TRT were performed at different ERBs. This was in order to more fully assess how accurately experiment A can be performed as well as give an indication of the possible benefits of cross-talk cancellation at different bandwidths. Cross-talk cancellation was only performed on a single side. Although it would have been possible to construct a bilateral cross-talk cancellation method this would have meant additional signal at the contralateral BT. This additional signal would make evaluation if cross-talk was working more difficult. Thus a noise and a cross-talk cancellation noise experimental setup was constructed. All participants had similar reductions in TRT with the addition of crosstalk noise with benefits of 11.15 dB, 13.00 dB and 12.09 dB respectively. The smallest mean gain in TRT was at the lowest test frequency of 1200 Hz where a 9.16 dB improvement in TRT was identified with addition of crosstalk noise. The frequency with the greatest benefit in TRT with crosstalk noise was found at 2475 Hz. It is likely that the differences in results are due to the accuracy of the phase and level measurements performed in experiment A. The lower TRT benefits at lower frequencies indicating the greater difficulty of performing the two-BT cancellation task over this range.

5.5.3 Experiment C

We have shown that in an artificial situation where noise is directed only to one cochlea and speech to the other then there can be very large benefits with the addition of cross-talk cancellation noise. However, there are several limitations to the study. Firstly, noise and speech in a real life scenario are very rarely completely separated. It is therefore difficult to show how much of the changes in SRT can be transferred to a real world scenario. In addition to this the speech was band limited to cover the same frequency spectrum as the cross-talk cancellation measurements. Thus our results overestimate any real potential benefits but show that the outlined methodology can be used to create a working cross-talk system. One other potential limitation for real word use is that the our current cross-talk design only cancels sound from the ‘first reflection’. In other words, it does not cancel sound which is reflected from the contralateral side of the skull back to the ipsilateral cochlea. It is also unknown if this reflection theory would be an issue or if the degree of attenuation from vibrations travelling from one side of the skull and then

back would mean that the energy dissipation would be large to not be significant in normal hearing situations the majority of the time.

Future work needs to focus around several areas. Firstly, if the assumption is made that perfect cross-talk cancellation can be achieved to restore ICLD how much benefit in SRT can be gained in more realistic listening scenarios and how well does this fit will binaural prediction models. Secondly what are the benefits in SRT when performing bilateral cross-talk cancellation over the same frequency range with and without band filtering the speech to match the measurement frequencies. Additionally, there are further challenges on how this method can be implemented in real time, since in the outlined scenario all audio was prepared prior to its use. Future research will focus on how low latency full audio transfer bilaterally might be possible in order to make cross-talk cancellation possible.

5.6 Conclusion

Within this study we have shown that it is possible to collect phase and level differences with considerably improved speed when compared to data collection times needed in chapter 3 (although this was not the focus of investigation and thus times were not formally recorded). In addition to this we have shown that these values can be successfully used in a fixed filter to create a cross-talk cancellation algorithm. Within this study only cross-talk cancellation was performed on either the left or right side and never bilaterally. Future work will also perform bilateral cancellation.

We showed that by employing unilateral cross-talk cancellation of band limited noise there was a significant benefit in TRT (12.08 dB) as well as in SRT (13.7 dB, 95% CI 8.09-19.24 dB). Future research should focus on ascertaining the potential benefits to patients with bilateral BCHAs as well as attempting to perform bilateral cross-talk cancellation in real time.

6 The relative benefits of interaural time and level differences when sound is presented via air or bone conduction

6.1 Introduction

Bone conduction hearing aids (BCHAs) have been shown to be an effective tool primarily for the treatment of patients with conductive hearing loss (including patients with craniofacial abnormalities and canal atresia), as well as for patients where hearing aids are contraindicated, such as patients with recurrent otitis externa or canal eczema (Snik et al. 2005; Stenfelt & Zeitooni 2013). Previously it has been shown that the cross-talk of bone conducted (BC) sound varies significantly across frequency as well as patients (Nolan & Lyon 1981; Stenfelt 2012; Snyder 1973; Vanniasegaram et al. 1994). The degree of cross-talk, cost and additional surgical risks associated with a further procedure (if not performed simultaneously) have meant that the majority of patients receive one BCHA even though they may have been diagnosed with a bilateral conductive hearing loss (Stenfelt & Zeitooni 2013). As a result of the cross-talk, patients fitted with bilateral BCHAs are not be able to make full use of sound localisation and spatial unmasking (Stenfelt and Zeitooni, 2013). The fundamental mechanisms by which effective binaural hearing is disrupted are that cross-talk at each cochlea disturbs the interaural time differences (ITDs) and reduces the interaural level differences (ILDs) (Liao 2010; Majdak et al. 2013).

However, the limited number of studies which have compared unilateral and bilateral BCHA fitting show that residual binaural processing is possible. These studies have shown that there are benefits to bilateral fitting, although the degree of benefit is highly dependent on listening situation. Sound localisation studies comparing bilateral and unilateral BCHA have shown that there are significant improvements with bilateral fitting and that this is particularly evident at higher frequencies (van der Pouw et al. 1998; Priwin et al. 2004; Bosman et al. 2001). The same studies also investigated speech reception thresholds (SRT) in quiet and found an improvement of 4 dB for bilateral fitting compared to unilateral. This modest improvements in SRT in quiet is thought to be primarily due to bilateral loudness summation (Bosman et al. 2001). Bosman et al. (2001) and Priwin et al. (2004) both performed binaural masking level differences (BMLD) measurements for pure tones in broadband noise, Bosman et al., (2001) finding benefits of 6.1, 6.0 and 6.6 dB at 125, 250 and 500 Hz (Bosman et al., 2001).

Other studies that have investigated bilateral fitting with BCHA have included research by Deas et al. (2010), who performed phase inversion between bilateral BCHAs. This uses knowledge that at low frequencies (<750 Hz) the skull vibrates in phase. This is particularly evident at frequencies below 500 Hz where the skull moves in rigid body motion (Stenfelt and Goode, 2005). Stimulation via two bone transducers at low frequencies produce vibrations which are out of phase at the two cochleae. Deas et al. (2010) found that by inverting the phase of one of the bone transducers, the signals at the two cochleae summed, and there was a significant improvement in hearing thresholds at frequencies below 750 Hz. This shows that it is possible to cause signal summation at low frequencies to increase hearing threshold in patients.

In our previous study (chapter 5) we showed that ICLD and ICPD (inter-cochlear phase difference) can be accurately measured in order to create a cross-talk cancellation system within the 1-5 kHz frequency range. In addition to this, the cross-talk measurements were used to significantly improve speech reception thresholds (SRTs) through cross-talk cancellation. However, the previous study started out with speech and noise completely separated. The total separation of signals at each cochlea is not a listening condition which would naturally occur. Within this study we aim to assess the potential benefits for a cross-talk cancellation system if implemented in bilateral BCHA patients in realistic conditions. This work builds on previous studies performed by Hausler et al. (1983) and Stenfelt and Zeitooni, (2013).

Hausler et al. (1983) investigated 17 subjects which included both participants with unilateral conductive or bilateral hearing loss of at least 35 dB between 0.25-4 kHz, spatial discrimination was investigated in a free-field using the Minimum Audible Angle (MAA) in both the horizontal and vertical plane. Just Noticeable Difference (JND) were also used to investigate lateralisation via headphones using changes in ILD and ITD. They found that participants with a conductive loss of greater than 35 dB over a number of frequencies had near-normal ILD JNDs, but abnormal ITD JNDs (when AC sound was presented at 90 dB). They also found BC ILDs were similar in both normal hearing participants and those with a conductive loss. They concluded that normal-hearing participants primarily have access to ILDs rather than ITDs when listening to BC sound. Stenfelt and Zeitooni, (2013) compared spatial release from masking (SRM) with BC and AC sound in normal-hearing participants wearing headphones and bilateral bone transducers. They found that SRM was approximately halved (in dB) when speech and noise were separated by 90° and sound presented via BC rather than AC.

In the present study, we measured SRM for BCHAs using appropriate head related impulse responses (HRIRs) that possessed only ITDs or only ILDs. By convolving the HRIRs with speech and presenting them via earphones and bone transducers we investigated the relative contributions of ITD and ILD SRM for BC compared to air conduction (AC). This procedure allowed us to identify the difference in SRM between AC and BC sound using only ILDs. The difference in SRM under the ILD condition will give an indication of how much benefit cross-talk cancellation could have if implemented bilaterally. This is because a cross-talk cancellations system has the potential to correct ILD but not ITD. Thus the difference in SRM between AC and BC using an ILD-only HRIR can be viewed as the maximum possible benefit of a cross-talk cancellation system for bilateral BCHAs.

6.2 Method

The experimental methodology outline was approved by Cardiff University Psychology Department Ethics Committee.

6.2.1 Apparatus

All testing procedures were performed with the use of Matlab®. An ESI MAYA44 USB+ four-output-channel DAC was used to pass signals through an 8-channel Behinger Powerplay Pro-8 Headphone amplifier to two B71 (Radioear) bone transducers and a pair of Etymotic ER2 insert earphones with ER1-14B eartips attached. To reduce bone transducer placement variation the bone transducers were attached to lens-less glasses which were then worn by each participant (as shown in the chapter 3). The glasses aimed to place the bone transducers approximately 55 mm behind the opening of the auditory canal, as this is the recommended surgical placement position (Battista & Ho 2003). To increase coupling between the bone transducers and the skull an elasticated soft band was used. There was no adjustment of either bone transducer or ER2 earphones once testing had commenced. Testing was performed in a single-walled Industrial Acoustics Company (IAC) sound attenuating booth within a sound treated room. A computer screen was visible outside the booth window with a keyboard inside the booth for participants to input transcripts.

6.2.2 Participants

Eight paid participants, age 18-28 years old were recruited from Cardiff University (2 males and 6 females). All participants were primary English speakers with self-reported normal hearing and no previous history of otitis externa or ear surgery. An otoscopic examination was used to visualise the ear canal and tympanic membrane of

each participant to check for a macroscopically normal ear. No formal pure-tone audiometry was performed. Participants were excluded if they had a large amount of wax in the ear canal which could become impacted by deep insertion of ER1-14B eartips.

6.2.3 Stimuli recording

HRIRs were used to simulate free-field conditions. HRIRs were provided by Cochlear Ltd., and had been collected at the House Ear Institute employing methods outlined by Chan et al. (2008). HRIRs were recorded and represented by a 150-point finite impulse response (FIR) filter (at 0° and 90° azimuths) at a sampling rate of 24 kHz. A cochlear bone anchored solutions BP100 microphone was modified in order to allow electrical output to be measured directly. The omnidirectional output from the microphone was amplified using a Tascam M-06 mixer. The BP100 was attached to the right side of a Knowles Electronics Manikin for Acoustic Research (KEMAR) approximately 55 mm behind and slightly above the ear-canal opening (the recommended implant position) (Battista & Ho 2003; Reinfeldt et al. 2015) . It was assumed that there was head and room symmetry, such that measurements from the right side would be identical to those from the same angle on the left side.

A prerecording was made without the KEMAR in the booth using a B&K 4133 microphone at the location where the KEMARs head would lie. The microphone was then used in combination with a pre-amplifier (B&K 2639) and a measuring amplifier (B&K 2609) to produce a linearly amplified signal. Broadband white noise was presented over a 2 second period to create a finite impulse response (FIR) via a time-domain Wiener filter method (Wiener 1949). Resultant HRIRs consisted of two FIRs (one for each ear).

6.2.4 Stimuli creation

The speech was always simulated to come from the front. The two conditions measured were at 0° azimuth (S_0N_0) and noise at 90° azimuth (S_0N_{90}). In order to investigate the relative benefits of ITD and ILD when speech and noise are spatially separated by 90°, the two HRIRs were first converted into the frequency domain. The phase spectrum from the 90° azimuth was combined with the level spectrum from the 0° azimuth to create a HRIR with only ITD differences. To create a HRIR with ILD only properties from 90°, the phase spectra from 0° azimuth was combined with the level spectra from 90° azimuth. An inverse Fourier transform was then performed to derive time-domain HRIRs. These HRIRs that were used in the experiment are shown in Figure 28.

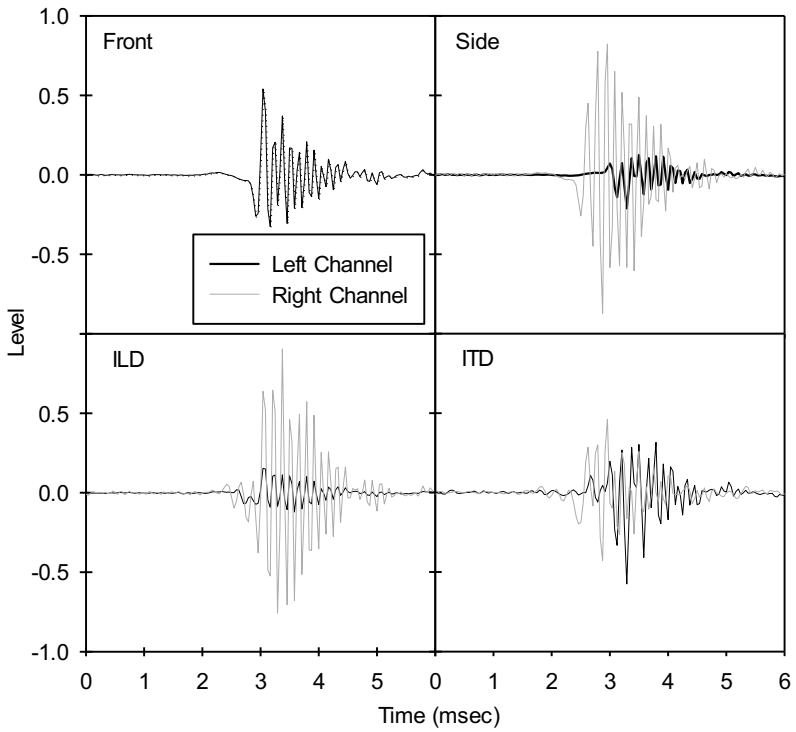


Figure 28 HRIR impulse responses that were convolved with the speech sentences to simulate each condition.

6.2.5 Spatial release from masking

The participants undertook approximately 1 hour of testing, during which 8 SRTs were measured. SRTs were recorded in each of eight conditions. The conditions were 4 spatial conditions (collocated, separated, ILD-only and ITD-only) × 2 presentation modes (AC and BC). In the ILD-only and ITD-only conditions, the HRIRs used for the noise were processed, as described above, in order to isolate the individual binaural cues.

Speech and speech-shaped noise interferers were convolved with the HRIRs described above to produce stereo stimuli. Before data collection, participants completed four practise SRTs. The practise tasks included presentation via ER2 earphones (AC) and bilaterally placed B71 bone transducers (BC). Two conditions (collocated speech and noise at 0° azimuth (S_0N_0) and noise at 90° azimuth (S_0N_{90})) were both played using the AC and BC presentation methods. The order of these four practise conditions matched the sequence for the main experiment. Within the main experiment four participants started with BC testing initially followed by AC testing. The other four participants started the main testing procedure with AC testing. Although testing was performed in blocks of either AC or BC, the four sub conditions (S_0N_0, S_0N_{90}, S_0N_{ITD}, S_0N_{ILD}) were rotated to minimise practise effects.

Target speech was from a male voices ("CW") from MIT recordings of the Harvard Sentence lists (Rothauser et al. 1969). An adaptive threshold task was performed whereby participants transcribed these grammatically and semantically unpredictable English sentences mixed within speech shaped noise. An example of such a sentence used within the current study is: "RICE is OFTEN SERVED in ROUND BOWLS" with keywords in capitals. The task aimed to identify the signal-to-noise ratio at which there was 50% intelligibility. This was performed via an 1-up/1-down adaptive threshold task (Plomp and Mimpen, 1979). Each SRT measurement used ten sentences and a different speech-shaped noise interferer. The speech-to-noise ratio was initially low and on the first target sentence the participant could attempt to enter a transcript via the computer keyboard or replay the same stimuli, but with an improved signal-to-noise ratio. If the participant chose to replay the stimuli the target level was increased by 4 dB. This increase in target level could be repeated until the participant successfully identified at least two target words. Participants were asked to only attempt a transcript if they could identify three or more words. Once the computer algorithm matched two words, the first target sentence would appear on the screen with the keywords in capital letters. Participants would then score themselves on how many of the keywords they had correctly identified. All subsequent target sentences were only presented once. If the participant correctly identified three words in the target sentence the following target sentence level would increase by 2 dB. If less than three words were identified then the level would decrease by 2 dB. The SRT result was calculated via the mean of the speech-to-noise ratios calculated after the last eight sentences.

The practise sentences were the same order for all participants. Within the primary task the order of the conditions was rotated while the order of the sentence lists remained the same, so that each condition was presented once at each serial position, thus counterbalancing any effect which may arise purely due to the speech material, or to practise/fatigue.

6.2.6 Prediction method

The Jelfs et al. (2011) model of spatial release from masking (SRM) in normal hearing listeners was used in order to predict the relative benefits for each of the four testing parameters. The model is comprised of two perceptual components, better-ear listening and binaural unmasking (Culling et al. 2012). Better-ear listening employs the theory that listeners can focus on the ear with the better SNR and that this can be performed at different frequencies independently. This arises because two sound or speech sources occupy different locations and this gives rise to a variation in attenuation

due head shadow (Lavandier et al. 2012; Long et al. 2006). Binaural unmasking relies on differences in the waveform of sound at each ear using cues such as ITD and ILD in order to detect the presence of the target signal (Culling 2011).

The Jelfs et al. (2011) model processes HRIRs within frequency channels via two separate separate paths (as outlined in

Figure 29).

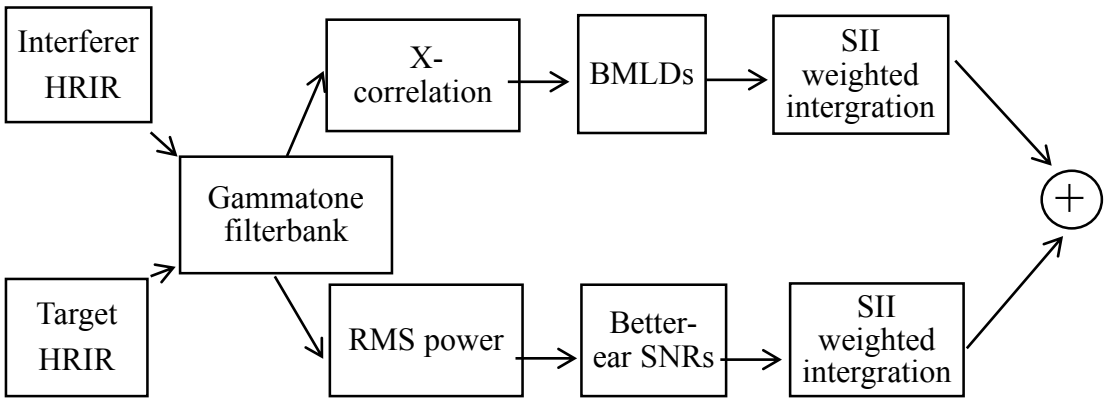


Figure 29 Schematic of the Lavandier and Culling (2010) model.

The inputs were the original and manipulated HRIRs from a BP100 bone transducer described above. The HRIRs for speech and noise were passed separately through a gammatone filterbank (Patterson et al. 1987) with filters distributed at one per Equivalent Rectangular Bandwidth (Moore & Glasberg 1983) up to 9.5 kHz. Using cross-correlation of the filtered HRIRs, the IPD of both the interferer and the target were calculated for each different frequency band as well as the interaural coherence of the interferer. The coherence was calculated as the maximum of the cross correlation, while the phase difference was calculated via multiplication of the delay corresponding to this maximum by the centre frequency of the band of interest. The BMLD was then calculated using the following formula (Culling et al., 2005, 2004) adapted from Durlach (1972)

$$BMLD = 10\, log_{10}\left[\frac{k - cos\,(\emptyset_T - \emptyset_I)}{k - \rho}\right] \qquad (13)$$

where

$$k = (2 + \sigma_\varepsilon^2) exp\,(\omega^2 \sigma_\delta^2) \qquad (14)$$

and $\emptyset_T$ and $\emptyset_I$ are the IPD of the target and interferer, ρ is the interaural coherence of the interferer, ω is the centre frequency in radians/s of the band, σ_δ= 0.000105, and σ_ε

=0.25 (Durlach 1972). The resulting BMLDs were integrated across frequency using SII weightings (ANSI 1997) to yield the predicted contribution of binaural unmasking to SRM.

In order to evaluate the better-ear listening component the model proposed by Zurek (1993) was employed to define head shadow in anechoic conditions. The rms power of the filtered speech and noise HRIRs were separately calculated for each ear, and their ratio calculated to produce SNRs in each frequency channel for each ear. The higher of the SNRs at each ear was taken as the SNR for the "better-ear". The better-ear values where then similarly integrated across frequency with SII weighting to calculate the contribution of better-ear listening to spatial unmasking.

The Jelfs et al. (2011) prediction model then adds the BMLD and better-ear listening benefits in order to give a relative SRT under different noise source directions. This was used to give a prediction of the relative change in SRT using HRIR of ITD, ILD and noise from 90° when compared to collocated speech and noise from 0°.

6.2.7 Data analysis

Mean and standard deviation was calculated for each of the 8 conditions from the 8 participants. A one-way repeated measured analysis of variance (ANOVA) was conducted to investigate if there was a significant difference between conditions in both AC and BC testing. Pairwise comparisons found that there was a significant difference between for each of the AC conditions ($p<0.05$). Pairwise comparisons of the BC conditions showed that there were significant differences all conditions except between 0_90 and ILD as well as 0_0 and ITD.

6.3 Results

Figure 30 shows the mean improvement in threshold for 50 % intelligibility (SRT) when compared to collocated speech and noise from 8 participants using AC and BC, the SRM. The prediction of SRM is based on the Jelfs et al. (2011) model. The air-conduction data is in good agreement with the model. The bone-conduction data shows a deficit in SRM consistent with disruptive effect of crosstalk. There were large differences in SRTs between conditions as well as between participants as indicated by error bars indicating standard deviation. There was once instance where during BT testing the ITD result was worse than the 0_0 condition. This did not occur in any other instance.

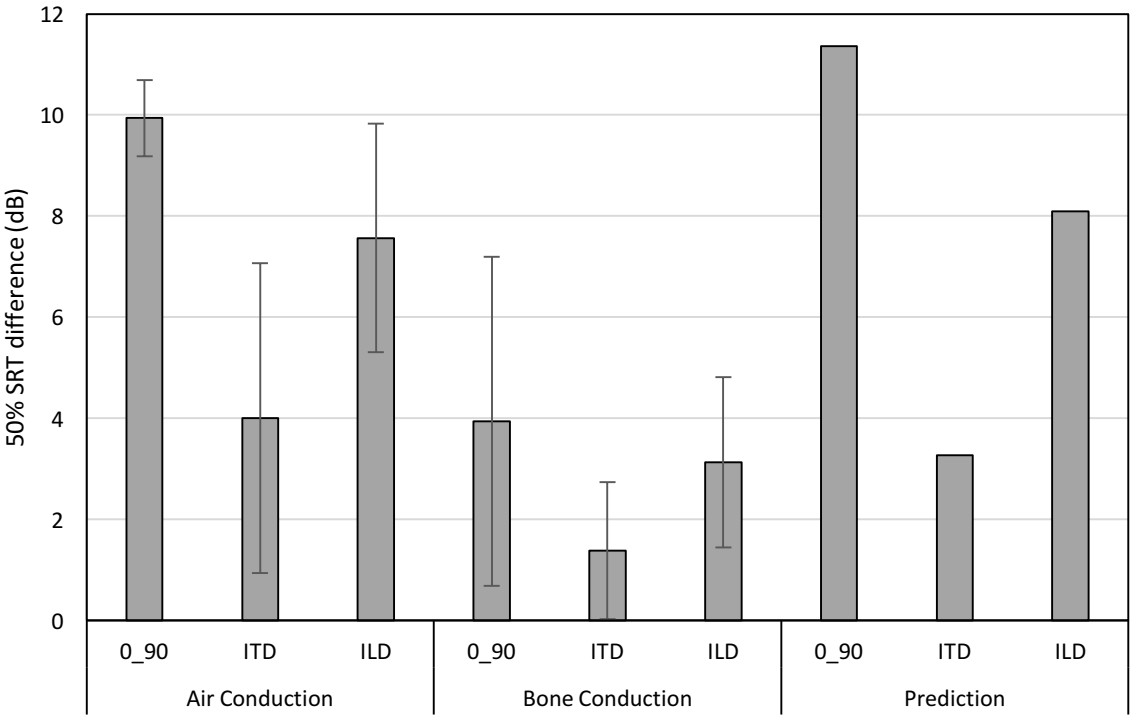


Figure 30 SRM for noise at 0° vs. 90° (n=8). Error bars show standard deviation. Predicted values are produced using Lavandier and Culling, 2010 binaural speech intelligibility model.

6.4 Discussion

The Jelfs et al. (2011) model gave good predictions in all test conditions for air conduction (as it was designed for). The model was least accurate when estimating the threshold increase between 0° and 90° (predicting 11.8 dB benefit when compared to 9.9 dB which was actually identified). ITD-only and ILD-only estimations for air-conduction SRM were good (-0.74 dB and 0.53 dB difference between measured results respectively).

BC SRM was significantly reduced when compared to AC SRM (by 6 dB, 2.6 dB and 4.4 dB in the 90°, ITD and ILD conditions respectively). This reflects a 60 % decrease in SRM, agreeing well with the findings of Stenfelt & Zeitooni (2013) who performed binaural intelligibility level differences using AC and BT presentation. They found that, compared to speech and noise presented via headphones, there was a 50% decrease in SRM when when sounds were presented via bilateral BTs. This shows that although ITD and ILD clues are still present in BC sound they are severely limited due to cross-talk.

The primary purpose for the present study was to investigate the difference in SRT benefit between AC and BC in the ILD-only condition. This is of significance since any cross-talk cancellation algorithm corrects the ILD. Therefore, the potential benefit of a cross-talk cancellation system under ideal conditions is the difference in AC BC thresholds in the ILD-only condition, which was measured here to be 4.4 dB. This benefit is significantly smaller than that measured in Chapter 5 which identified a 13.6 dB benefit in SRT with unilateral cross-talk cancellation. However, a 4.4 dB improvement in SRT could still be highly beneficial to patients with two BAHAs. Although it is still currently unknown if a 4.4 dB is achievable in a real world setting since cross-talk cancellation cannot be performed at all frequencies (as already discussed in chapter 5). In addition, there may also be limitations on the number of filter channels available when performing cross-talk cancellation in real time. Thus further research is needed to investigate the potential benefit of cross-talk cancellation under even more realistic conditions.

6.5 Conclusion

Comparison of AC and BC thresholds showed a 4.4 dB reduction in the ILD condition. This reduction indicates the potential benefits of ideal cross-talk cancellation. Future research should focus on more realistic scenarios which take account of frequencies where cross-talk cancellation cannot be achieved and the limitations of filter bands when processing sound in real time.

7 General Discussion and Future Work

7.1 The "Take-Home" Message

The main conclusions that can be drawn from this thesis are that it is possible to gather accurate psychoacoustic measurements of the phase and level of sound reaching the cochleae from bone transducers. Additionally, we have demonstrated that in principle this information can be used to create a fixed filter that can be used in a cross-cancellation network to significantly improve speech reception thresholds. This scheme could potentially be implemented in bilateral BCHAs.

7.2 Summary of Findings

The results of the experiments reported here have shown the following:

- Laser Doppler measurements of a participant's skull can be used to show the modal vibration of the skull when stimulated by a bone transducer (BT) at low frequencies (<1.5 kHz).
- Using 3D reconstruction the skull can be seen to move in phase when comparing both temporal regions at frequencies of 0.25 kHz (indicating rigid body motion). At 1 kHz the displacement for both temporal regions was out of phase indicating mass-spring motion.
- A single-BT and two-earphone (ER2) method can be used to cancel sound emitted from a BT at both cochleae. A major advantage of the method was that it could be performed over a wide frequency spectrum (0.25-8 kHz)
- High variability was found in the level of air-conducted (AC) sound needed to cancel the bone-conducted (BC) sound using the single-BT method over the frequency range. However reproducible patterns of level needed for cancellation were identified. Additionally, it was found that at low frequencies some participants had a higher cancellation level at the contralateral cochlea compared to the ipsilateral. This was thought to be related to resonance and antiresonance of the skull and ossicular structures.
- Using the Single-BT method phase was found to decrease (suggesting reduced velocity) at lower frequencies (0.25-3 kHz) in the ipsilateral cochlear. At higher frequencies (>4 kHz) phase was found to linearly increase on both the ipsilateral and contralateral sides.
- The ICPD produced by a single BT varied across participants. This may have been related to different skull sizes. The mean ICPD of all participants was found to be highly similar to the psychoacoustic measurements of Zwislocki (1953). However, the measured ICPDs were greater than those measured by Stenfelt and Goode (2005).
- ILD measurements were very consistent to those of Stenfelt and Goode (2005).

- For the two-BT method, grading of cancellation showed that the perceived best cancellation was achieved at higher frequencies and the worst at the lower frequencies. Below about 1.2 kHz the task became too difficult to collect data. This was thought to be related to the greater influence of phase cues at lower frequencies.
- It is possible to use a single-BT method to predict the cancellation phase and level results in a two-BT methodology. This may be key in future work since the single-BT method can be used at all frequency ranges whilst the two-BT method becomes challenging at frequencies lower than 1.5 kHz. However, the single-BT method would not be applicable in a clinical setting since it relies on normal ear canals.
- Use of the two-BT technique in combination with a prediction algorithm greatly increased the speed of phase and level measurements so that on an initial trial could be performed in approximately 50 min and a recalibration could be performed in approximately 15 min.
- Single-sided cross-talk cancellation using the phase differences between the two BTs showed that it could be used to significantly benefit tone thresholds in noise (12.1 dB) and SRT (13.67 dB). These benefits are greater than those that would be seen in real application since the noise and speech was band limited and the speech and noise were presented on separate BTs.
- The potential benefits of cross-talk cancellation, based on measurements using HRIRs from a BP100 are about 4 dB improvement in SRT.

7.3 Opportunities for further research

There are several significant challenges which still need to be addressed before cross-talk cancellation can be used in a patient population. These are broadly categorised into the phase and level measurement and technological challenges.

7.3.1 Phase and level measurement

7.3.1.1 Data collection speed and ease

One of the primary aims of this thesis was to achieve at least a 10 dB benefit from cross-talk cancellation. It is possible to predict the attenuation (dB) which will be achieved via Equation 16 where α is level error (dB) and φ is phase error (degrees).

$$10\ log_{10}\left(10^{\frac{\alpha}{20}-cos\,\pi(\frac{\varphi}{180})}\right) = A \qquad (15)$$

In chapter 5 we identified SRT benefits of 13.7 dB when noise and speech is band-limited to the same frequencies for which cross-talk cancellation was calculated (1-5 kHz). This corresponds to a mean measurement error over the frequency range of approximately of 5° and 0.35 dB. This shows that the participants were able to perform the psychoacoustic task with a very high degree of accuracy over the test frequencies.

The use of the prediction algorithm (introduced in chapter 5) allowed the collection of phase and level data both faster and more accurately. However, the method still requires some skill on the part of the participant in order to obtain optimal measurements. Since many patients who currently have bilateral BCHA also have complex medical needs which often including learning difficulties, the development of a psychoacoustic technique which is as easy as possible is of high priority.

We envisage two further improvements to the data collection methodology.

Firstly, a new experimental technique could be used whereby slightly different frequencies are presented via the two BTs to produce beating. Beating was employed in chapter 3 to level match between BC and AC sound. However, with this new technique beats would occur to differing degrees at both cochleae. The participant would be asked to mouse click left when sound is lateralised to the left and click right when sound is lateralised to the right. Within the new bilateral beating task, the timing of when a participant perceives sound to be maximally lateralised to the left and right will give an indication of the ICPD. The level difference will be varied between the BTs at the same frequency. The frequency will then change automatically. By investigating the change in phase at the same frequency under different level presentations it will be possible to predict the cancellation phase and level at these frequencies. These values will then be used to make the previously employed technique easier to perform.

The second method of speeding up data collection is to use the change in phase from cancellation at one cochlea to predict the phase change at the contralateral cochlea. This technique may be possible, as it was noted in chapter 5 that the phase progression was very similar on the left and right side (especially at higher frequencies >4 kHz).

7.3.1.2 Real time bilateral cross-talk cancellation

Future research will aim to perform real time cross-talk cancellation. In order to do this, the two-BT data collection technique would be performed bilaterally. At frequencies where cross-talk cancellation cannot be performed due to small ICPD differences (0.5-1 kHz) the best approach may be to invert the phase. Since the signals from the two BTs would normally arrive out of phase, inversion will cause signal summation over this frequency band, improving the amplification provided by the system. This inversion strategy was found to be beneficial by Deas et al. (2010).

The cross-talk signal processing cannot be performed with low enough latency within a windows, mac or android operating system. Thus, a specialist unit Tucker Davis Technology (TDT) real-time processor (RP3), which is designed for low-latency, real-

time filtering will be used for experimental development. A possible experimental procedure may be to collect the necessary phase and level measurements with improvements described above and then upload these onto the TDT RP3 unit as a fixed filter. The BTs would remain in place but they would instead be connected to the TDT RP3 unit. The participant could then be placed in a speaker array where tone detection, speech intelligibility and sound localisation measurements can be made with and without cross-talk cancellation.

7.3.2 Potential limitations of the technique

7.3.2.1 Processing limitations

With more advanced processing including feedback and noise reduction not to mention cross-talk cancellation there are increasing delays. This is a major problem in AC hearing aids since direct and amplified sound interact when the day is longer than 10 ms and can cause spectral ripples (Dillon 2012). However, in those with severe conductive hearing loss there will be very little direct sound reaching the cochleae. This could potentially allow for somewhat greater delays since problems such as spectral ripples will not be an issue. A potential acceptable delay maybe 40 ms which is agreed as the target for the international telecommunication union as an acceptable delay in order to avoid significant "lip sync effect" (Galster 2010).

There are two main forms of signal processing. One employs FFT to processes a group of samples together. The other method employs filters which divide the signal into channels and process them channel by channel. Both processing methods suffer from the same fundamental issue that as the spectral resolution is improved the temporal resolution is degraded. This means that if cross-talk cancellation was implemented there would need to be a trade-off between the number of filterbank channels which can be used to increase the accuracy of the cross-talk cancellation and the time it takes to perform this action. Commonly 32 coefficient filters are used in many hearing aids using an FFT. Using an FFT method of signal processing means that the filter channels are equally distributed with frequency. However, if an alternative method of filtering is used then it is possible to optimise the cancellation by placing filters are frequencies where there are large changes in cancellation level or phase. If this were to be used further research needs to be performed to investigate at what frequencies the filters are placed in order to achieve the optimum cross-talk cancellation within the given number of filters. It is likely there will be considerable variability between participants as this will need to be tailored to frequencies where there are large changes in level over a small frequency range. In addition to this, the filters must also be take account of the relative importance of the

relative importance of different frequency bandwidths in order understand speech (Apoux & Healy 2009).

7.3.2.2 Non-linearity

Previous studies which have compared AC and BC stimulation have found non-linarites in BC when compared to AC presented sound (Khanna et al. 1976). Additionally, otoacoustic emissions have been found to increase emission amplitude if BC-evoked when compared to those from a AC source (Rossi et al. 1988). If true, this could represent a serious challenge to cross-talk implementation. However, this contradicts findings from Håkansson et al. (1986) which identified no non linearity's in skull vibrations over a range which would be expected from a BCHA. Håkansson et al. (1996) states that the previous findings were likely due to methodological reasons such as not measuring the mechanical input delivered. Khanna et al (1976) give possible explanations of looseness of the coupling between the vibrator and the skull as well as the underlying tissues not acting perfectly elastically. Additionally, from the experiments already conducted it is unlikely that if nonlinearity was encountered over normal BT vibrational range that the cross-talk cancellation system would have had such an impact on SRT.

7.3.2.3 Intracranial pressure

It is not currently clear how much of impact intracranial pressure has on BC sound propagation in the skull. Bekesy (1948) remarked during cancellation of BC sound that yawning caused the sound to once again become audible. It is not clear if this was due to a change in jaw movement or if there were temporary changes in intracranial pressure which can occur when taking a deep breath (Bloomfield et al. 1997). It maybe that changes in intracranial pressure could cause a change in phase and level of BC transmitted sound. One possible mechanism via which this could occur is via the finding that one pathway for BC sound transmission to the cochlea is via the cerebrospinal fluid (CSF) (Freeman et al. 2000; Sohmer et al. 2000). A pressure change within the CSF could therefore impact the degree of fluid movement and thus the level of sound. However, there are several important factors at work. Firstly, the Monro-Kellie model of intracranial pressure states "the sum of the intracranial volumes of blood, brain, CSF and other components is constant and that the skull is considered an enclosed inelastic container" (Dunn 2002). Therefore, an increase in one volume is offset via another. Under normal conditions this results in the intracranial pressure being very tightly controlled with small changes in sitting and standing which are a result of increased venous drainage which is then compensated for via an increased arterial blood pressure (Williams 1981). Thus,

although there are still changes in intracranial pressure they may or may not be significant enough to affect the phase and level of BC conducted sound.

A further possible factor which needs to be considered is that level may be affected (due to differences in ease of CSF fluid movement) but phase may not be since the speed of sound is not affected by fluid pressure, but only by its elasticity and density. If this is the case, then when it is likely that level of BC sound reaching the cochleae from both BTs would be equally altered, and so would not impact the degree of cross-talk cancellation achieved. During preparation of the two BT technique some exploratory testing was performed whereby cancellation was achieved over the 1-5 kHz range (whilst sitting) as described in chapter 5. The participant then used the mouse scroller to change the frequency whilst the computer automatically adjusted the contralateral BT in order to present the previously inputted phase and level of cancellation signals. On this occasion the participant was standing. The participant did not find any clear alteration in the quality of cross-talk cancellation. Future research should aim to perform the two-BT task when standing and sitting independently in order to investigate if there are differences in cross-talk signal.

7.3.3 Technological research

7.3.3.1 Audio streaming capability

In order to perform cross-talk cancellation audio needs to be streamed from one BCHA to the other and vice versa. To be of use, it also needs to be performed with low latency. Currently, the BAHA 5 does have the ability to stream sound from devices like the mini microphone. However, this is performed with the 2.4 GHz frequency band and with a latency too high to be appropriate for cross-talk cancellation. 2.4 GHz is the most widely used band for wireless data transmission for small devices. This is primarily because its signals pass though air with little signal degradation. However, the short wavelength means that it does not propagate well through soft tissues such as the head and body (Cho et al. 2007). This means that limited signal settings can be synced between two BAHAs using this method a full audio signal would not be an efficient method of transmitting the data.

We envisage two possible solutions to this problem. Firstly, the easiest way to achieve low latency with little extra battery requirements would be to have a wired connection. However, this is unlikely to be a very acceptable to patients due to the aesthetic impact a wire would have. A further method which may be appropriate would be employing Near Field Magnetic Conduction (NFMC) (Galster 2010). The typical transmission frequency range is between 3-15 MHz. The primary benefit of NFMI is that

it propagates well though and around the human head. Additionally, it has been widely used for technologies such as the telecoil. NFMI is not commonly used for signal streaming from specialist microphone devices or music streaming devices because as the distance increases the signal degrades quickly. . However, in this application of NFMI, the distance needed to transfer sound will only be from one mastoid bone to the other.

7.3.3.2 BCHA orientation

Further research also needs to be performed to investigate if and how the orientation of the BCHA may affect phase and level needed for cross-talk. This is because it is unknown if the new BC transducer for the BAHA 5 the BCDrive ® (Flynn MC. 2015) causes vibrations purely in the same plane as the abutment or if there is a small amount of angular vibration which also occurs. Thus although the abutment position does not change a change in orientation of the BCHA may impact the level of vibrations and thus cross-talk effectiveness. If orientation is affected, then it maybe that a new abutment clip design needs to be developed to standardise the BCHAs orientation.

7.3.4 Final Remarks

This thesis has successfully developed a proof of concept methodology to gain measurements of phase and level at the cochlea. We have shown these psychoacoustic measurements can be performed to a high degree of accuracy. Additionally, we have demonstrated that the measurements can be successfully applied in a cross-talk cancellation system to give significant improvements in TRT and SRT. Future developments will focus on further reducing the time taken to collect measurements as well as making the psychoacoustic task easier to perform. Spectral/temporal limitations and small ICPDs at certain frequencies will mean that cross-talk cancellation will not be optimal at all frequencies. However, accurate measurement combined with the use of NFMC technology have a potential to give patients with bilateral BCHAs significant improvements in sound localisation and SRTs as we have demonstrated by our results with single-sided crosstalk cancellation.

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