Crosstalk Cancellation in Bilateral Bone-Conduction Hearing Devices

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Summary

Bone-Conduction Hearing Devices (BCHDs) are hearing devices used in patients with mixed and conductive hearing loss who would otherwise not benefit from traditional hearing aids. These devices are typically implanted into the skull of patients and can bypass the middle and outer ears to deliver sound to patient's cochleae.

Bilateral implantation outperforms unilateral implantation in most audiological measures, however they do not restore hearing to normal levels, in particular sound localisation remains poor. This is due to the limited attenuation of the signal produced from a BCHD as it vibrates around the skull, resulting in a percept at the contralateral cochlea that would be louder than if that same signal traveled around the head through the air, causing interference.

The purpose of this thesis is to build on previous work from Mcleod and Culling who demonstrated that the crosstalk produced from one bone transmitter (BT) could be cancelled by a contralateral BT. This thesis addresses three questions. First, to determine whether it is possible to cancel the crosstalk from two BTs simultaneously. Second, to replicate the findings from Mcleod and Culling using BCHD-compatible filters in normal-hearing listeners and patients, and finally, whether the crosstalk-cancellation filters remain effective over time.

The first experiment took the crosstalk-cancellation filter design from Mcleod and Culling's experiments and changed it so that these filters could be used with patient's BCHDs. These filters were tested in normal-hearing listeners finding that they were effective at cancelling the crosstalk from two BTs simultaneously.

This filter-creation method was then repeated in patients, finding that while two patients were able to create effective crosstalk-cancellation filters, the filter's crosstalk-cancellation efficacy decreased over time. Future work should focus on collecting more patient data and identify the variables causing this filter deterioration.

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Symbols	Definition	First Appearance
AC	Air Conduction	1
BC	Bone Conduction	2
BCHD	Bone-Conduction Hearing Device	1
BILD	Binaural Intelligibility Level Difference	12
BMLD	Binaural Masking Level Difference	12
BT	Bone Transducer	1
BU	Binaural Unmasking	12
CHL	Cochlear Hearing Loss	1
CROS	Contralateral Routing of Signal	20
ECSP	Ear Canal Sound Pressure	41
FIR	Finite Impulse Response	52
HRIR	Head-Related Impulse Response	4
ILD	Interaural Level Difference	5
IPD	Interaural Phase Difference	5
ISI	Inter-Stimulus Interval	11
ITD	Interaural Time Different	5
MAA	Minimum Audible Angle	23
OAE	Otoacoustic Emission	45
SNR	Signal to Noise Ratio	12
SRM	Spatial Release from Masking	12

Chapter 1 - Introduction

Bone-conduction hearing devices (BCHDs) are hearing devices that transmit sound to the cochlea via the skull using a percutaneous titanium abutment screwed into the skull. They are an effective treatment in patients with conductive hearing loss (CHL), mixed hearing loss, or single-sided deafness (SSD) who can otherwise not benefit from conventional hearing aids. Fitting patients with a BCHD leads to advantages in hearing compared to being unaided (Wazen et al, 2003; Ho et al., 2004; Hol et al., 2009). Fitting bilaterally deaf patients with a second BCHD partially restores binaural hearing (Canale et al., 2022), produces favourable results when compared to unilateral fitting (e.g. Priwin et al., 2004, 2007) and improves patient satisfaction (Dutt et al., 2002; Dun et al., 2010), however hearing is still not restored to a normal-hearing level (den Besten et al., 2020; Caspers et al., 2021). Despite the improvements seen with binaural implantation only 6% of patients with bilateral CHL are fitted with bilateral devices (Cochlear bilateral market survey, 2018) since the benefits of having two BCHDs is limited (Stenfelt and Zeitooni, 2013a).

This limitation is caused by the limited transcranial attenuation. A single BCHD will produce sound both at the ipsilateral cochlea (on the same side as the BCHD) and the contralateral cochlea (at the opposite cochlea to the BCHD), and, due to the limited attenuation, will produce a louder sensation at the contralateral cochlea compared to air-conduction (AC) hearing. This sensation is often referred to as crosstalk, and this crosstalk impairs binaural hearing. Studies have measured the transcranial attenuation of sound across the head finding attenuation of between - 17 (negative attenuation) and 32 dB (Nolan & Lyon, 1981; Håkansson, 1986; Liao, 2010), and that minimal attenuation is seen below 500 Hz (Stenfelt and Goode, 2005; Farrell et al., 2017).

In an attempt to resolve the issues caused by crosstalk in bilaterally implanted patients, researchers have developed methods to create and implement crosstalk-cancellation filters with the aim of isolating the ipsilateral signals at each cochlea. Mcleod and Culling (2019, 2020) adjusted the phase and amplitude of a tone played at one extracutaneous bone transducer (BT) to cancel an identical tone (with unchanged phase and amplitude) played at another. The BTs were RadioEar B71 transducers as used for audiology bone-conduction thresholds. This was done across a range of frequencies and used this to create a crosstalk-cancellation filter. When used in a tone-reception-threshold (TRT) task to cancel contralateral noise, improvements between 11.2 dB and 13 dB were seen.

1.2 Current Research:

The overall aim of the PhD is to explore the viability of creating and using a bilateral crosstalk-cancellation filter in normally hearing participants with the ultimate goal of determining the feasibility of implementing crosstalk-cancellation in patients with bilateral BCHDs. There are several key questions this thesis aims to address:

1. Can a practical filter be designed to effectively cancel bone-conduction (BC) crosstalk? The filter used in Mcleod and Culling (2019, 2020) was a very high resolution filter implemented in the frequency domain and would not have been directly transferrable to a patient's BCHDs, but a 256 – tap finite-impulse-response filter design could be used with a patient's BCHDs.
2. Can the crosstalk-cancellation filter be applied simultaneously on both sides of the head? Previous crosstalk-cancellation research has only used the vibration from one BT to cancel the crosstalk from a contralateral BT at the ipsilateral cochlea. The present study aims to apply the filters at both cochleae and isolate both ipsilateral signals simultaneously.
3. Can patients create effective crosstalk-cancellation filters? The key differences between patients and normal hearing participants are the skin attenuation, the structure of the middle/outer ears and the location of the vibration on the skull. The skin attenuation and ear structure will affect the BC pathways as well as the location of the Vibration being at 'BCHD' position rather than at the mastoid (Röösli et al., 2022). In addition to this the patient's proficiency with BC hearing may allow them to cancel BC crosstalk with greater ease than normally hearing participants.
4. Do the cancellation phase and amplitude at a given frequency remain consistent when a patient removes and re-attaches their BCHDs and do the filters remain effective over time (e.g. after a month). The main limitation with crosstalk-cancellation research using BTs is that the filter's must be recalibrated each time

the BTs are replaced on a participant's head. This is presumed to be caused by repositioning of the BTs altering the BC pathways around the head (Mcleod & Culling, 2020), changing the cancellation phases and amplitudes. In patients with implanted BCHDs, the percutaneous abutment won't move over time so in theory this should not be an issue.

1.3 Thesis Outline:

This thesis starts with a literature review, introducing binaural hearing in normally hearing listeners. Then the BC pathways are explored before summarizing a range of studies that have compared the use of unilateral and bilateral BCHDs in patients with bilateral hearing loss. The introduction finishes by familiarising the reader with the concept of crosstalk cancellation and how this has been implemented in research.

Three experiments are then described. The first experiment (chapter 3) follows on from Mcleod and Culling (2019, 2020), utilising a similar technique but using a filter design that would be able to be used in a patient's BCHDs. These filters were then applied bilaterally to measure normally-hearing participant's TRTs against contralateral noise with and without the crosstalk-cancellation filters applied. Participants were initially trained using simulated crosstalk presented over headphones. The experiment demonstrated the feasibility of this filter design and the ability to apply this filter bilaterally.

This procedure was then used on patients implanted with bilateral BCHDs (chapter 4). Some patients struggled to cancel using the procedure used in chapter 3 so an automated version of the task that automatically sweeps through cancellation phases and levels was implemented instead. Mixed success was seen with participant's cancellation ability, largely due to patients withdrawing from the study due to time constraints. Two patients created successful left-ear crosstalk cancellation filters, improving tone-reception thresholds by 9 dB to 12 dB. The benefit of all filters depreciated to varying degrees after three sessions. Right side cancellation was less successful, finding a 2 dB – 3 dB TRT benefit.

The final experiment (chapter 5) measured the localisation ability of normal-hearing participants when using both AC and BC sound. AC stimulation resulted in fewer errors when localising sound. Additionally, the positioning of the microphones used to record the Head-related impulse responses (HRIRs, ear-canal position vs BCHD position) did not affect accuracy. Feedback improved the accuracy of participants in all conditions, and significant improvement was not seen over time.

Chapter 2 – Background

2.1 Binaural Hearing

Binaural hearing is the ability to detect the differences in sounds arriving at both ears and to use this information to aid in sound perception. Binaural hearing primarily aids sound localisation and hearing a signal when there are competing sound sources, such as being able to hear speech in noise. Two major cues are used in binaural hearing, these are: interaural level differences (ILDs), which are the differences in level of a sound arriving at each ear; and interaural time delays (ITDs), which are the time differences between sound arriving at one ear compared to the other. ITDs are sometimes expressed as Interaural Phase Differences (IPDs) which are the differences in the phase of the sound arriving at each ear. IPDs generally result from ITDs, but they can be directly manipulated by an experimenter. This section discusses the relevant phenomena which contribute to a person's ability to hear binaurally.

It was originally hypothesised that ILDs operate primarily at high frequencies and that ITDs operate at low frequencies (Rayleigh, 1907), however it was later established that they are both important at all frequencies, although ITDs dominate low frequencies and ILDs dominate higher frequencies (Shaw, 1974). The relevant frequency ranges vary across species, determined by the size of the animal's head and the position of the eyes (Heffner and Heffner, 2018).

2.1.1 ITDs

A simple model of ITDs is to model the head as a perfectly spherical object with ears as points on opposite sides of that sphere (Duda & Martens, 1998). Under these assumptions an azimuth of 0° (straight ahead) produces minimal ITDs (0 μs), because the ears will be the same distance from the sound, and a maximal ITD (650 μs) at an azimuth of 90° since the difference in distance between the signal and each ear is at its greatest. This can vary across individuals as head width and ear sizes differ (Middlebrooks 1999). Measurements of ITDs at different frequencies have demonstrated the frequency-dependence of ITDs (Kuhn, 1977). For example, a tone of

500 Hz will have an ITD of 800 μs decreasing to 650 μs for a 4000 Hz tone. These ITD changes are caused by diffraction and are also consistent with a spherically modelled head (Benichoux et al., 2016).

ITDs are primarily used by the human auditory system below 1500 Hz. This is because above 1500 Hz the difference in arrival time of the sound at each ear is greater than the period of the sound wave, causing localisation ambiguities. This transition can be seen in the just-noticeable difference, which refers to the point (in this case the ITD) at which the participant can perceive two identical sounds as being different. In the range of 500-1000 Hz, the just-noticeable difference can be as low as 10 μs (Klumpp & Eady, 1956, Zwislocki & Feldman, 1956). Above 1500 Hz participants are unable to discriminate between two pure tones when an ITD is introduced.

2.1.2 ILDs

Like ITDs, the level difference between each ear is also dependent on the position of the sound source, the frequency of the sound, and the person's individual head shape. According to the inverse square law, as a sound wave travels through the air, energy becomes dispersed over a larger wavefront. Specifically, that the energy loss is inversely proportional to the square of the distance travelled, roughly 6 dB loss per doubling of distance travelled in the free field. This means that if a sound is located to one side of the head, the ipsilateral ear will receive a sound with greater sound pressure level than the contralateral ear. Significant ILDs caused by distance alone are only seen below 1m and increase exponentially the closer to the ear the sound source is, whereas ITDs are largely independent of distance (Brungart and Rabinowitz, 1999).

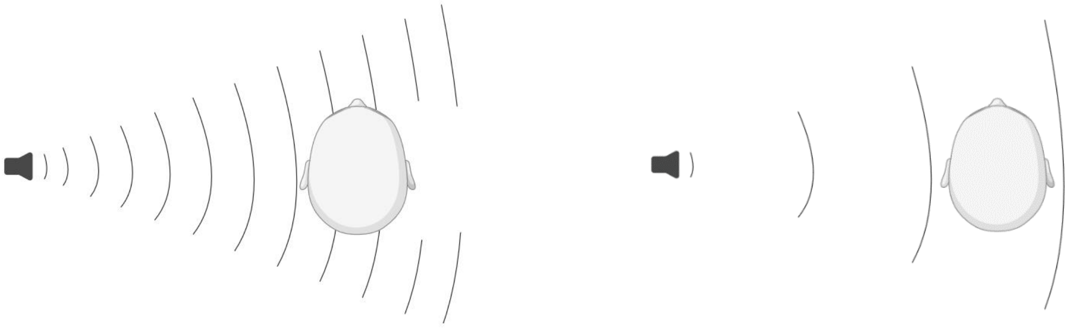


Figure 2.1. Illustration of the Head Shadow Effect. The left image shows a high frequency sound, and the right image shows a low frequency sound.

In addition to the distance affecting the level of a sound, the head itself can act as a physical barrier to a sound wave reaching the contralateral ear; this is called the 'head shadow effect' (Figure 2.1). Head shadow refers to the attenuation applied to sound arriving at the ear when sound is presented from the opposite side of the head. Sound diffracts around the head causing a reduction in sound level and change in spectral pattern. The effect is stronger at higher frequencies, due to the shorter wavelengths being impacted more by the head acting as a physical barrier.

Shaw (1974) reviewed several studies involving a total of 100 participants that looked at the ILDs between the ears at various frequencies and azimuths (Figure 2.2). ILDs at 200 Hz are low and much higher at 1.6k Hz and tend to increase at all azimuths the higher the frequency. The greatest ILD is seen around 110° for both 12k Hz and 7.6 kHz, and at 60° for 3.5 kHz. A double peak at 60° and 120° degrees is seen at 1.6kHz

Listeners can distinguish ILDs of 0.5-1.5 dB over a wide frequency range, however for longer wavelengths (e.g. 500 Hz, which has a wave length of 70 cm), the head blocks a relatively small amount of the sound wave since the wavelength is significantly bigger than the width of the head and so the attenuation would be less than 0.5-1.5 dB. In contrast to this, at high frequencies the head is a more substantial obstacle since the wavelengths are shorter than the width of the head (e.g. 5000 Hz has a wave length of 7 cm).

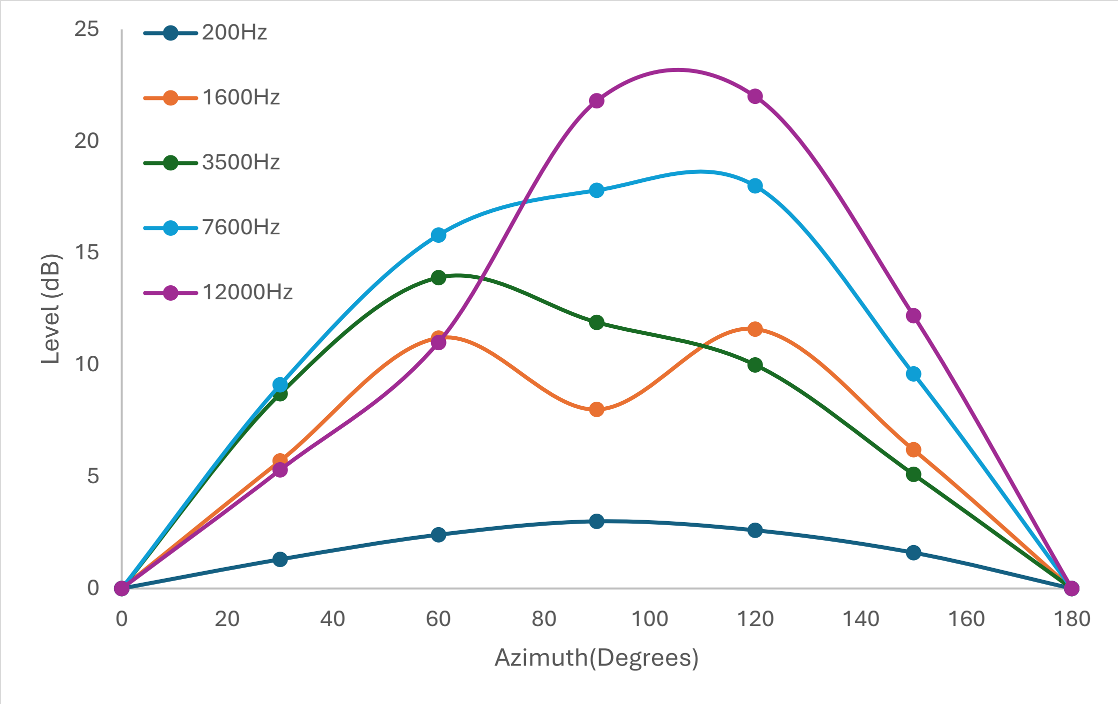


Figure 2.2. Data from Shaw (1974) showing the ILDs of different frequency tones at different azimuths

2.1.3 Head-Related Impulse Responses

ITD and ILD information can be measured by recording a head-related impulse response (HRIR). An HRIR is measured by placing microphones at each ear on a listener (or mannequin) and recording the response to an impulse from a given location (e.g. Gardner & Martin, 1995). The recording from each microphone provides the HRIR which contains the acoustic information of the sound reaching each ear. These HRIRs can then be convolved with an auditory stimulus and played over headphones to mimic the location of the original sound used to measure the HRIRs, avoiding the problem of headphone-presented sound being perceived as coming from inside the listener's head (Wightman & Kistler, 1989). These HRIRs are reliant on the same participant being used to record the HRIRs and when listening to the convolved sound over headphones in order to accurately recreate the original listening experience, however generic HRIRs will still somewhat preserve the auditory space (Wenzel et al., 1993).

2.1.4 Localisation

2.1.4.1 ILDs, ITDs and their Interaction

The ILDs and ITDs can be used to help determine the location of a sound source. A typical research paradigm, the ‘pointing task’, involves a listener adjusting the ITD or ILD of a sound (the pointer) until it matches the location of a second experimenter-controlled sound (e.g. Feddersen et al., 1957; Schiano et al., 1986). When Feddersen (1957) asked participants to match the tone to a noise coming from a 23° azimuth (ITD of 180 μs), 45° (ITD of 369 μs), and 60° (ITD of 486 μs), participant’s manipulation of the ILD of the tone was in line with expectations above 1500 Hz (assuming the head is a perfectly spherical object), although errors were seen at greater azimuths. Below 1500 Hz greater ILDs were needed for the tone to match the location of the noise. This is because ITDs dominate lower frequencies so greater ILDs are required to localise sound. Schiano et al. (1986) used the opposite configuration (adjusting the ILD of noise to match pure tone targets with varying ITDs), using tones between 300 and 1600 Hz. They found that sound became progressively more lateralised between 300 and 1000 Hz before progressively returning to the centre at 1500 Hz. Tones above 1500 Hz were perceived as being straight ahead as there was no effect of ITDs above this frequency.

This transition of using ITDs at low frequencies and ILDs at high frequencies to provide binaural information has been termed the ‘duplex theory’ and was proposed by Rayleigh (1907). Localisation ability for pure tones is strong at lower frequencies, degrading up to 3000 Hz but then increases again up to 10000 Hz where accuracy is regained, which is in agreement with the duplex theory (Stevens and Newman, 1936). This mid-frequency deficit is a symptom of neither phase nor level cues being optimal for sound localisation.

While the duplex theory holds true for pure tones, when complex sounds are presented, the dichotomy between ITDs and ILDs is less clear. Studies have demonstrated the effects of ITDs on high-pass filtered frequency clicks (e.g. David et al., 1959, Harris, 1960, Yost et al., 1971). In addition to this, Hafter and Carrier (1972) conducted an experiment where participants were presented with a tone of fixed time delay and/or level difference and were asked to indicate whether it differed from a diotic stimulus.

Participants were always able to detect the presence of ILDs and ITDs. When a conflicting ITD was applied to a signal with pre-existing ILDs the participant's ability to perceive a difference was lessened, although there didn't seem to be a point at which either cue completely cancelled the other out suggesting that both cues have some influence at all frequencies.

A similar interaction of ITDs and ILDs can be seen in noise where either cue can be used depending on the situation (Lorenzi et al., 1999). Changing only the ITDs of broadband sounds to those of another azimuth will cause the noise to be perceived close to that azimuth, dominating the ILDs, as well as obscuring the elevation and causing front/back confusions (Wightman & Kistler, 1992). Wightman and Kistler (1992) also repeated this with band-limited noise, finding that ITD cues were influential until frequencies below 2.5 kHz were removed, where they started to lose their influence until being eliminated when frequencies below 5 kHz were removed. In addition to this, Macpherson and Middlebrooks (2002) investigated the weightings of ITD and ILD cues in low-pass, high-pass and wide-band noise bursts finding that ILD cues are more heavily weighted in high-pass noise, ITD cues are more heavily weighted in low-pass, and moderate weighting of both cues in wide-band noise, albeit with a heavier weighting on ITDs. These findings, along with those from Wightman and Kistler, suggest that it is the ITDs of low frequency waves that dominate sound localisation in wide-band noise.

Research has since explored complex sounds and the effect of ITDs in various complex waveforms such as noise, two-tone complexes and sinusoidally amplitude modulated tones (Leakey et al., 1958; Henning, 1974a, 1974b; Bernstein & Trahiotis, 1994). Leaky et al. (1958), used a 4000 Hz carrier tone and amplitude modulated it to 200 Hz, and added an ITD to the envelope of the waveform. Participants were able to perceive changes between the two tones when an ITD is added despite the carrier frequency being above the upper limit (1500 Hz) of ITD detection. This suggests that envelopes also carry ITD information, although sensitivity to these high-frequency ITDs is still much lower to low-frequency ITDs where ITD envelope modulation is not effective (Henning, 1981; Henning, 1983).

2.1.4.2 Pinna Cues

The pinna and ear canal are responsible for filtering sound according to its spatial location. Sound will diffract off the head and pinna, creating spectral dips that can then be used to facilitate vertical localisation (Grothe et al., 2010). Pinna cues predominantly aid in elevation and front/back discrimination, although front/back confusions are still made especially with lower bandwidth stimuli (Makous & Middlebrooks, 1990). Batteau (1967) suggested that sound bounced between the ridges of the pinna to create spatial cues by changing the spectrum of the sound. In addition to aiding front/back discrimination, pinna cues also enable vertical localisation. When pinna cues at one ear are impaired azimuth localisation is not impacted, although elevation localisation is impaired particularly at the effected ear and obscuring pinna cues at both ears further impairs vertical localisation (Hofman & Van Opstal, 2003), in addition to impairing horizontal localisation (Muscant & Butler, 1984). During head motion, dynamic cues are used in conjunction with these cues to further aid localisation.

2.1.4.3 Precedence Effect

The precedence effect describes a listener's ability to process sound in a complex, reflective environment where sound distortions and conflicting directional information occurs that is not seen in a free field environment. To parse this complicated sound, listeners will prioritise the interaural cues that arrive at the ears first, this is called the precedence effect and is the reason we can localise the source of a sound rather than the echoes that sounds produces (Hartmann 1997). An example of this is shown in figure 2.3, where despite 4 different clicks being observed, they are perceived as a single event on the listener's left, based on the ITD of the first pair of clicks.

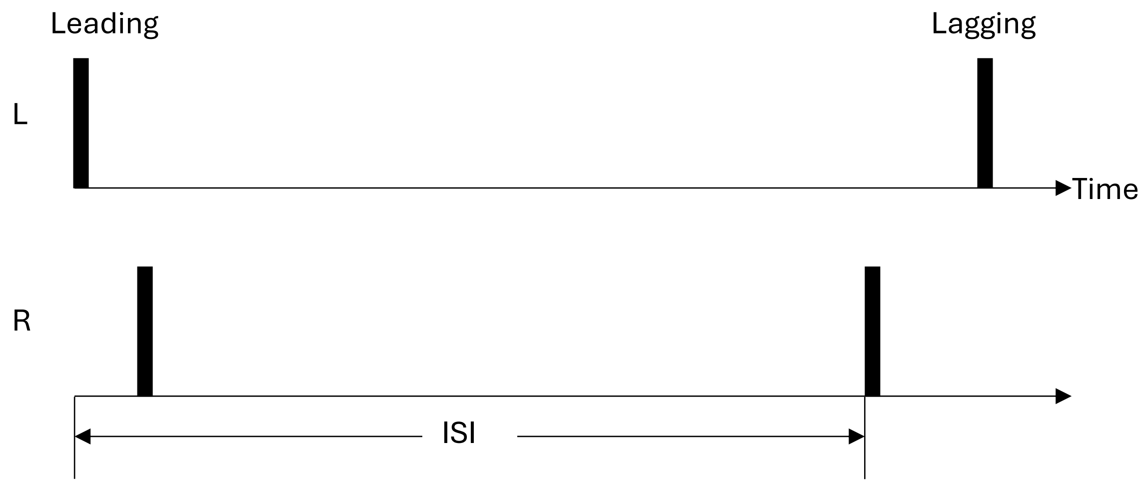


*Figure 2.3***.** Four different clicks are perceived as one since the third click arrives at the ears before the echo threshold. If the inter-stimulus interval (ISI) is increased, the clicks will be heard as different sounds. Adapted from Hartmann, W. M. (2021).

The precedence effect has been demonstrated through studies where sounds are presented in quick succession. A delay of 5-10 ms causes clicks to be perceived as a single sound image whereas delays longer than this are considered two separate sounds (Freyman et al., 1991; Yang & Grantham, 1997; Litovsky 1999). The threshold for speech is about 50ms (Lochner & Burger, 1958). This fusion suppresses the lagging sound and allows listeners to consolidate the reflections of a sound around an environment into a single percept.

2.1.5 Spatial Release from Masking

In the real world interaural differences of a signal and a noise occur due to the signal and noise originating from different locations around the listener, and allow listeners to better detect or identify the signal. This is referred to as spatial release from masking (SRM). Successful models of SRM (e.g. Lavandier and Culling, 2010) include two additional components.

2.1.5.1 Better-ear Listening

Assuming the talker is in front of the listener, the signal to noise ratio (SNR) improves as the masker moves around the head, dipping at 90°, and then increasing up to 120° before returning to near baseline at 180° (Culling & Lavandier, 2021). SRM involves a change in SNR at each ear since the head will act as a sound-attenuating obstacle depending on the source location of the signal and noise, causing an improvement in the SNR at one of the two ears.

2.1.5.2 Binaural Unmasking

Binaural Unmasking (BU) utilises the binaural properties of a sound (ITDs and ILDs) to improve the identification of a signal in noise. BU occurs when the binaural properties of a signal differ to the properties of the noise and result in an improved ability to perceive that signal. For example, if a tone is played at both ears in addition to noise with the same binaural configuration there will be a baseline threshold required to perceive the tone in the noise. If the phase of the tone is shifted by 180° (or π-radians) then the listener's threshold will drop by up to 15 dB. Changing the inter-aural phase of the noise can also improve hearing thresholds, although to a lesser extent than adjusting the signal. This change in threshold is known as the binaural masking level difference (BMLD), and BMLDs observed in speech can also be referred to as binaural intelligibility level differences (BILD, George et al., 2012).

A shortform nomenclature is used to describe the ways sounds are presented to participant's ears in experiments. 'S' refers to the signal and 'N' refers to the noise. A subscript '0' represents a diotic stimulus whereas 'm' refers to a monotic stimulus. 'π' is used to represent a phase change of 180^0, which can also be represented as π radians, and 'τ' represents a time delay. Examples of this nomenclature is shown in table 2.1.

Condition	Definition
N_mS_m	Noise and Signal presented to the same ear
N_0S_0	Noise and Signal identical at the two ears
N_0S_m	Noise presented identical at the two ears, signal presented to only one ear.
N_0S_π	Noise identical at the two ears, signal presented with π radians IPD (180°)

$N_\pi S_0$	Noise presented with π radians IPD (180°), signal identical at the two ears.
$N_t S_0$	Noise presented with a time delay between the two ears, signal identical at the two ears.
$N_{90} S_0$	Noise is presented from a 90° angle, signal is presented from the front.

Table 2.1. Examples of BU nomenclature often seen in BU literature

During one of the first experiments, Hirsch (1948) found that when the interaural phase of a 200 Hz pure tone differed from that of a broadband noise, the TRT was lowered. The largest difference found was 14 dB at lower frequencies, in addition to this he demonstrated that while N_0S_0 and $N_\pi S_\pi$ have the same threshold N_0S_π, $N_\pi S_0$, and N_0S_m have lower thresholds (Fig 2.4). Hirsh and Burgeat (1958) later used broadband noise (200-4000 Hz) to mask tones of different frequencies. They demonstrated that the BMLD reduced with increasing frequency, asymptoting to about 3 dB at 1500 Hz and above.

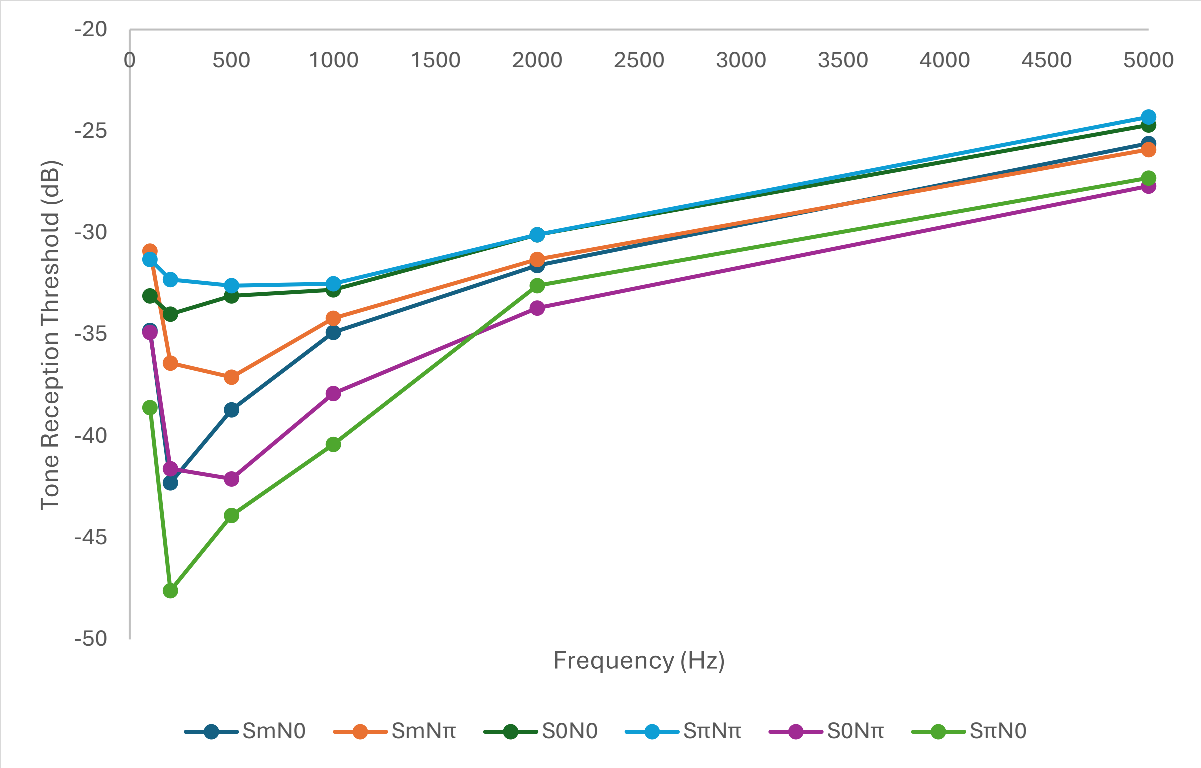


Figure 2.4. Threshold observed when changing the Phase of the tone or noise (Hirsch, 1948)

2.1.5.3 Binaural Unmasking in Speech

BU has also been demonstrated in the intelligibility of speech. The results are similar, although threshold reductions are only 4-5dB in the N_0S_π condition (Schubert, 1956). This may be because BU is most effective at lower frequencies (figure 2.4), but these frequencies are not the most important for speech perception (Fletcher and Galt, 1950). Levitt and Rabiner (1967) later demonstrated this by adding a 180° phase shift above certain frequencies in speech, leaving the lower frequencies unchanged. When a 180° phase shift is applied at all frequencies, speech detection thresholds decrease by 13 dB, although intelligibility thresholds only decrease by 6 dB. The BILD is not reduced when the IPD is set to 0° below 250 Hz, although it halves when an ITD of 0° is applied below 500 Hz and BILDs are eliminated when the ITD is 0° below 1000 Hz. This suggests that for speech, the region where IPDs are most important are between 250 – 1000 Hz and the maximal effects of BU at 250 Hz are not effective for speech perception.

In the real world a listener will often be listening for speech in competing noise from multiple directions rather than trying to detect a tone or speech against a single noise source. An example of this is the 'cocktail party problem' where a listener will try to hear the voice of the person talking to them (signal) in the presence of everyone else's conversations (noise). Masking of speech by speech comes in two varieties. The first is energetic masking, which occurs when the signal and interferer occupy the same temporal space and frequency spectrum, reducing in audibility and intelligibility of the target speech. The second variety is informational masking, which also reduces intelligibility and occurs when there are other similar talkers present to the one you are trying to listen to, causing confusion in knowing which voice to pay attention to (Brungart, 2000; Brungart et al., 2001). The confusion in information masking is due to the content of the noise rather than the signal itself being less audible. When sound sources are in different directions, binaural cues permit a spatial release from masking.

In summary, spatial masking release of energetic masking is caused by two factors. The first is the ILDs between each ear caused by the different spatial locations of each talker and their interaction with the head. The second factor is ITDs. ILDs provide monaural benefits at the single ear through an improved signal-to-noise ratio (better-ear listening) whereas ITDs are compared across the two ears (BU). Spatial release of informational masking probably relies on listeners ability to localise competing sound sources and attribute spectrotemporal components of the competing sound sources to separate spatially located perceptual streams (Bregman, 1980).

2.1.6 Binaural Summation

Binaural summation is the principle that when sound is presented to two ears rather than one, it will be perceived as louder since there is more total neural activity when both cochleae are stimulated. This phenomenon can cause sound to be perceived up to twice as loud, and the effect of binaural summation increases with frequency (Fletcher and Munson, 1933). Listening with two ears rather than one can also improve speech-reception thresholds (SRTs) by up to 5% (Lavandier & Culling, 2008).

2.1.7 Summary

The research discussed in this section demonstrates how listeners use both ITDs and ILDs to determine the source of a sound and aid in speech perception. Interfering with these cues leads to impairments in sound localisation and makes it difficult to attend to a person’s speech in background noise. Restoring this binaural information is important for hearing-impaired listeners to preserve binaural hearing.

2.2 Bone Conduction and Hearing Loss

Hearing pathologies often cause deficits in binaural hearing since they disrupt the hearing of one or both ears. For example, if a person has a conductive hearing loss at one of their ears, they no longer have access to the ITDs and ILDs that would facilitate spatial hearing. Binaural hearing can be restored with hearing aids (Derleth et al., 2021), however traditional air conduction (AC) hearing aids are not effective in patients with the severer forms of conductive hearing loss, such as atresia and cholesteatoma. An alternative to AC hearing aids are Bone-conduction hearing devices (BCHDs), which don't rely on a working middle or outer ear.

2.2.1 Bone-Conduction Hearing Devices

BCHDs (Figure 2.7) are devices which utilise sound vibrations to transmit sounds to the cochlea via the skull, bypassing the outer and middle ear. The first percutaneous implantation was performed by Tjellström and his colleagues (1981) after the concept of osseointegration had been developed by Brånemark in 1977. Patients with percutaneous BCHDs have a screw implanted into their skull for direct skull vibration, and have an abutment that passes through the skin that allows a BCHD to be attached and removed easily. These devices became widely available in the 1980s and were offered to patients suffering from hearing loss caused by conductive or mixed hearing loss, including those with SSD and chronic otitis.

Figure 2.7. A percutaneously implanted BCHD (Picture credit: https://commons.wikimedia.org/w/index.php?curid=32468684)

These devices are separated into two broad categories: percutaneous and transcutaneous. Percutaneous devices (such as the Cochlear™ BAHA® Connect and Oticon Ponto systems) consist of an implantable abutment, an osseointegrated screw, and a transducer. The device is attached directly to the abutment which then vibrates the screw. These devices can deliver high powered vibrations while avoiding the up to 20 dB attenuation that can occur when sound travels through the skin (Verstraeten et al., 2009). The downside of these implants are the skin irritation and other implantation-site issues that can occur after surgery.

Transcutaneous devices also utilise an osseointegrated implant, however a magnet is attached to the implant with the skin closed over the top, a magnet is then attached to the external device and is placed over the subcutaneous magnet. Transcutaneous devices come in two forms: active and passive. Passive devices such as the BAHA® attract system vibrate the magnet and internal device through the skin and while this method of implantation avoids skin complications, the force of the magnet against the skin can cause discomfort after prolonged use and is exposed to attenuation through the skin. Active devices use an external processor to send an electronic signal to an implanted transducer, bypassing the skin. Passive devices include the Bonebridge™ and the Osia® 2 system.

There are also extrinsic devices which do not involve surgical implantation. These 'extrinsic' devices are typically held against the skull using non-surgical methods, such as a softband or bone-conduction glasses. Here an external device holds bone vibration devices against the skin and these vibrations will propagate through the skin to each cochlear without the need for an implant. This results in skin attenuation as well as the discomfort felt from the pressure required to hold the device against the skull. These devices are typically used either in children or prior to implantation to allow patients to test their device(s) before surgery, or can be used commercially in the case of bone-conduction glasses.

BCHDs are typically used when patients have conductive or mixed (a combination of conductive and sensorineural) hearing loss of either one or both ears, where alternative hearing aids are not effective. Air conduction (AC) hearing aids are ineffective when patients have either conductive hearing loss (so sound waves cannot travel through the middle ear effectively), or when patients respond poorly to AC hearing aids, such as those with chronic otitis externa. BCHDs can also be used in patients with SSD as a receiver placed on the deaf side will transmit the signal to the normal hearing cochlear due to the low attenuation of the signal across the skull (Snik et al., 1998, Stenfelt et al., 2000).

2.2.2 BCHDs in SSD

Single-sided deafness is a significant hearing loss in one ear, and effects an estimated 9000 new people in the UK each year and is primarily caused by sudden hearing loss (Dimmlow, 2003). Since patients only have one functioning cochlear, they are unable to benefit from binaural cues and as a result have poorer sound localisation and difficulty understanding speech in noise (Wazen et al., 2003; Welsh et al., 2004; Silverman et al., 2006; Augustine et al., 2013; Agterberg et al., 2014). There are two devices primarily used in the treatment of SSD: contralateral routing of signal (CROS) hearing aids and bone-conduction hearing devices. CROS systems utilise a microphone placed on the worse hearing side and a hearing aid on the better ear to reduce the effect of the head shadow on contralateral sound whereas BCHDs use the cross-head vibrations produced from a single BCHD (which will be discussed in more detail later). In a hearing-in-noise test comparison between BCHDs and CROS systems, patients performed better in the BCHD group regardless of where the signal was presented, in addition to improved response in subjective measures (Wazen et al, 2003; Ho et al., 2004; Hol et al., 2009). Patients with a BCHD also outperform participants with a CROS system in localisation tasks (Hol et al., 2009; Choi et al., 2019), although it's important to note performance is worse than the unaided groups. Overall BCHD performance of SSD patients in objective measures is mixed, but is favourable in subjective measures.

While BCHDs are used in SSD due to their ability to transmit sound vibrations across the skull to the contralateral cochlea, in patients with conductive hearing loss they are used to transmit the sound vibrations to the ipsilateral cochlea. Although many of these patients have bilateral hearing loss, many patients are still fitted with only one BCHD. As few at 6% of patients are fitted with bilateral BCHDs (Cochlear bilateral market survey, 2018). This is despite the benefit of bilateral implantation being demonstrated in 1991 (Hamann et al., 1991) and the process of bilateral implantation being introduced in 1995.

2.2.3 Quantitative Measures of the Bilateral Benefits of BCHDs

There have been numerous studies that have compared the benefit of unilateral and bilateral BCHDs and these will be discussed in the following section.

2.2.3.1 Tone-Reception Thresholds

Priwin et al., (2004, 2007) tested TRTs in quiet in bilateral compared to unilateral implantation, finding an improvement of 2-7 dB in the bilateral condition when the tone is presented to the better hearing side (in unilateral conditions), front, and back. A 5-15 dB benefit was also observed when a tone was presented to the worse hearing side.

2.2.3.2 Speech-Reception Thresholds

Speech reception thresholds (SRTs) were also improved in bilaterally implanted patients. Bosman (2001) and Priwin (2004) found SRT improvements in quiet of 4 dB and 5.4 dB respectively with speech presented from the front when compared to unilateral implantation. Bilateral improvements of 2.5 dB and 3.1 dB were seen when noise was presented to the baffle side (side with a BCHD in the unilateral condition) and 0.8 dB and -1 dB when noise was presented to the head shadow side (side without a BCHD in the unilateral condition).

Canale et al. (2022) compared SRTs of unilateral and bilateral BCHDs in bilaterally implanted patients in 3 conditions: S_0N_0, S_0N_{90}, and $S_{90}N_{-90}$, where the noise was presented to the aided ear in the unilateral condition. Lower thresholds were observed in every condition in every patient when bilateral BCHDs were used (Table 2.2.). An improvement of 4.66 dB was seen in the S_0N_0 condition, 2.24 in the S_0N_{90} condition and 7.50 in the $S_{90}N_{-90}$ condition. Similar results were seen in Canale et al. (2023).

	S_0N_0 Effect (dB)		S_0N_{90} Effect (dB)		$S_{90}N_{-90}$ Effect (dB)	
ID	Monaurally Fitting	Binaurally Fitting	Monaurally Fitting	Binaurally Fitting	Monaurally Fitting	Binaurally Fitting
1	2.6	0.5	1.2	-1.2	1	-3.9
2	1.2	-4.5	-0.5	-3.7	7.2	-5.3
3	3.4	-1	-4.2	-6.4	4.5	-3.4
4	0	-3.3	-3.2	-5.5	0	-1.6
5	-0.2	-3.6	-7.8	-6.4	4.9	-1.8
6	-1.8	-4.1	-6.2	-7.1	-0.7	-1
7	20.9	9.5	16.5	10.4	21.4	2.8

Table 2.2. Results from Canale et al. (2022)

2.2.3.3 Word Score

Dutt et al. (2002a) tested word recognition score (WRS) when speech was presented from the front, and noise to either side of the head between -10 dB to 10 dB SNR. In quiet, bilateral implantation consistently outperformed unilateral implantation, with unilateral catching up at higher intensities (table 2.3). Dutt also tested WRS in noise finding a 4% improvement in bilateral users at +10 and 0 SNR and 2% at -10 SNR. Priwin et al (2007) also found a 14% improvement at +4 SNR and 7% at 0 SNR, although a 5% deterioration in bilateral performance at +6 SNR was also seen. Reuter et al. (1997) also observed better speech comprehension at 65 dB.

Intensity	Mean improvement between conditions
30 dB	4.5%
40 dB	5.5%
50 dB	4%
60 dB	3.5%
70 dB	2%
80 dB	1%

Table 2.3. Mean improvement between of WRS between unilateral and bilateral BCHD patients (Dutt 2002a)

2.2.3.4 Localisation

When unilaterally implanted, patients are only able to lateralise sounds at a chance level and patients perceive sound at the side of the head where their device is located (Snik et al., 1998; Bosman et al., 2001; Priwin 2004, 2007; den Besten et al., 2020; Caspers et al., 2021). Bilaterally implanted patients can lateralise sounds to near normal-hearing levels and while localisation was general better in bilateral conditions, it was still far below normal hearing listeners (Priwin et al, 2007). Snik et al (1998) were the only ones to find a difference in performance at low compared to high frequencies, with participants performing better at noise centred around 2 kHz.

Both den Besten (2020) and Caspers et al. (2021) reported individual paediatric patient's responses with bilateral BCHDs. Localisation ability varied between patients. While most patients showed accurate lateralisation with limited localisation ability, some patients could only lateralise poorly (with one patient perceiving all sound on the left side), while others could localise sound with a high accuracy. It is unknown why this is the case, but poorer performance could be caused by BC threshold asymmetry or age of bilateral implantation (Caspers et al., 2021). In addition, Caspers et al. reported no significant differences in responses across 4 sessions spanning multiple weeks.

Brassington et al. (2023) compared the minimal audible angle (MAA) in bilaterally implanted patients when using one or two BCHDs. For the bilateral condition an minimum of 3.61° was needed to achieve 80% correct lateralisations and in the unilateral condition an minimum of 75.04° was required. Twelve out of 24 unilateral participants were able to consistently lateralise sounds even when they were presented at ±90°.

2.2.4 Qualitative Measures

Bilaterally-aided patients consistently report greater satisfaction and quality of life when using two devices compared to one (Dutt et al., 2002b; Ho et al., 2009; Dun et al., 2010). Bilateral patients reported positive scores in every measure of the Glasgow Children's Benefit Inventory (Dun et al., 2010). Adults preferred using 2 BCHDs in most situations (Dutt et al., 2002b), with high scores in the learning and emotion categories and all children and adults were satisfied with their bilateral BCHDs, and participants preferred the audio quality of bilateral BCHDs better (Canale et al., 2022; Table 2.4).

Study Name	N	Questionnaire Results
Dun et al. (2010)	20	Glasgow Children's Benefit Inventory
		Overall Score: +38 (SD: 8.2)
		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 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)
Canale et al. (2022)	7	Overall Score: +11
		Communication: +23
		Background Noise: +10
		Reverberation: +19
		Aversiveness: +15

Table 2.4. Improvement in questionnaire score when using 2 BCHDs instead of one in bilaterally implanted patients. Adapted from Janssen et al. (2012)

The most commonly reported drawback was that patients reported hearing in noise difficult and found it advantageous to turn off one BCHD in noisy situations where the signal is coming from one side (Priwin et al., 2007; Dun et al., 2010), although in Dutt et al. (2002b) 8 out of 11 patients preferred using 2 BCHDs in noisy situations. When listening to speech in quiet patients prefer 2 BCHDs (Dutt et al., 2002b). Difficulty hearing speakers in noise was also reported in unilaterally implanted patients in addition to difficulty in sound localisation, which was improved when bilateral BCHDs are used (Dutt et al., 2002b; Dun et al., 2010).

2.3 Sound Pathways

2.3.1 Air Conduction

Typically, normally hearing people use AC to receive and hear sound waves. Sound waves travel into the ear canal via the pinna, through the middle ear and ossicles, before reaching the fluid filled cochlea and exciting the auditory pathway.

Once the sound enters the ear canal it reaches the tympanic membrane (ear drum), causing it to vibrate. Beyond the tympanic membrane lie the ossicles. These small bony structures transduce the air vibrations into mechanical energy. The tympanic membrane vibrations move the malleus, which moves the incus, which in turn moves the stapes before causing movements of the oval window of the cochlea.

The cochlea is a tonotopically organized fluid-filled structure which is responsible for transducing the mechanical energy from the middle ear to neural energy at the auditory nerve. The base of the basilar membrane is responsible for high frequency processing while the apex processes low frequency. Sound waves vibrate the basilar membrane at the point at which best resonates with the frequency of the sound, regardless of the method of stimulation.

2.3.2 Bone Conduction

An alternative method of sound transmission is to use bone conduction. This is usually the result of using a device (although it can also occur when exposed to high intensity AC sound, or a person's own voice), to vibrate a part of the skull, which causes sound waves to propagate through the skull bone and tissue. Georg von Békésy (1932) demonstrated that BC stimulates the cochlea in the same way as AC by using an out of phase BC tone to cancel an AC tone, this has been repeated in later experiments (Stenfelt, 2007, Mcleod & Culling, 2017; 2019). Weaver and Bray (1936) further demonstrated this by measuring the response at the guinea pig auditory nerve for both AC and BC sound, finding similarities.

When sound is presented through BC it takes multiple pathways around the skull to the cochlea. Initially it was theorised that conduction occurred solely through vibration of cochlear fluid, bypassing the middle ear (Allen & Fernandez, 1960). It was later suggested however that both the ossicles and soft tissues were also involved in bone conduction (Tonnondorf & Tabor, 1962; Brinkman et al., 1965). Although the exact details of these pathways are not known, Stenfelt suggests five major pathways (Stenfelt and Goode, 2005; Stenfelt 2011, Surendran & Stenfelt, 2022). These major pathways are:

1. Vibrations radiating through the external ear canal (Stenfelt et al., 2003)
2. Inertial movement of the ossicles (Stenfelt et al., 2002; Homma et al., 2009; Dobrev et al., 2020)
3. Inertial movement of the inner ear fluid (Stenfelt, 2015)
4. Compression and expansion of the inner ear and the displacement of cochlear fluid caused by the movement of the petrous bone (Stenfelt, 2015)
5. Changes in the intracranial space, e.g. cerebrospinal fluid (Sohmer, 2000; Sohmer and Freeman, 2004)

Of these five pathways the inertia of the inner ear fluid is speculated to be the most important for BC hearing (Stenfelt, 2016).

2.3.3 Resonance Frequencies

Resonance frequencies are frequencies where upon reaching the cochlea, sound waves that travel across the skull constructively (resonance) or destructively (anti-resonance) interfere. Skull vibration studies report finding no resonance frequencies in the skull below 500 Hz, with the first appearing around 500 Hz (Håkansson et al., 1994). The first global skull resonance frequency often appears around 900 Hz – 1.1 kHz (Eeg-Olafsson et al, 2011). Håkansson et al. (1986) report seeing contralateral head anti-resonances below 500 Hz in human participants, Stenfelt et al., (2000) also noted anti-resonances at the ipsilateral cochlea in a damped dry skull and Stenfelt and Goode (2005) note that more antiresonances are seen when stimulation is further from the cochlea. These anti-resonances cause a decrease in vibration at the ipsilateral cochlea compared to the vibrations caused at the contralateral cochlea, which causes sound to become lateralised to the contralateral side. Surendran and Stenfelt (2023) however report seeing no ipsilateral antiresonances in a perceptual lateralisation task. Variation in resonance frequencies between skulls is large and cannot be predicted as a function of head width, length or circumference, which suggests skull structure influences bone conduction more than simply the shape of the head (Håkansson et al., 1993; Stenfelt et al., 2000).

Antiresonances (frequencies where vibration level at the cochlea is low) are primarily caused when the interaction of the vibrating mass on the skull causes a peak in impedance, and can attenuate sound by 20-40 dB between 50-150 Hz (Stenfelt & Goode, 2005; Eeg-Olofsson et al., 2008; 2011). The frequencies at which these resonances and antiresonances appear differs between skulls, and more are seen when stimulation is on the midline, or at the contralateral side of the head (Stenfelt & Goode, 2005). Ipsilateral antiresonance has also previously been reported (Stenfelt et al., 2000).

2.3.4 Frequency-Dependent Skull Vibration

Several studies have measured the skull's vibration in response to stimuli of various frequencies in dry skulls and cadaver heads (e.g. Stenfelt et al., 2000; Stenfelt and Goode, 2005; Eeg-Olofsson et al., 2008; Eeg-Olafsson et al., 2011). Generally, these produce similar results but there are some differences, such as the ipsilateral antiresonances that are not seen in human participants (Surendran & Stenfelt, 2023). In addition to this, studies of dry skulls find the first resonance frequency around 1.2-1.4 kHz whereas in live participants the first resonance frequency is seen at around 1 kHz (Håkansson et al., 1994).

These studies have observed four patterns of vibration that are associated with specific frequency ranges and are demonstrated in figure 2.8. Below 500 Hz, the skull moves as a rigid body meaning both cochleae will receive the same vibration (Fig. 2.8). Up to 1 kHz, where the first skull resonance occurs, the skull moves to a 'mass spring' system where large parts of the skull move in the same direction (Fig. 2.8). Inverting the signal applied to one bone-vibration device when a second is present, results in improved thresholds since the ipsilateral and contralateral signals will no longer be in opposition to one another (Deas et al., 2010). Above 1 kHz the skull begins transitioning from a mass-spring system to a wave transmission system, and at 2 kHz the skull is dominated by wave transmission. Phase accumulation across the skull increases with frequency, suggesting that wave transmission takes longer to traverse the skull (Stenfelt & Goode, 2005). Mcleod et al. (2018) used laser Doppler vibrometry on a live participant, measuring the traversal of vibrations from a single B71W to verify whether the findings in live participants matched those found in dry skulls and cadaver heads. While at most frequencies the findings were similar, Mcleod et al.'s findings suggest that at 500 Hz the head is transitioning between rigid-body and mass-spring movements, rather than instantly changing. The head became a mass-spring system at 1000 Hz.

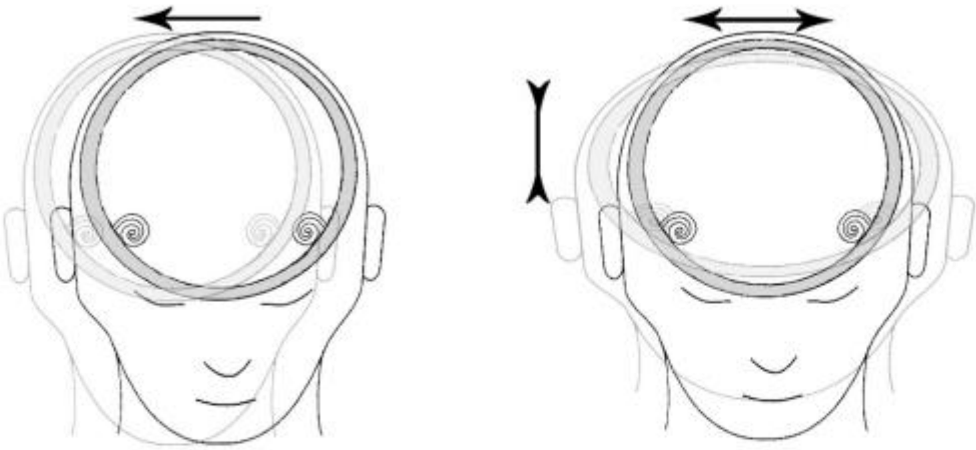


Figure 2.8. Illustration showing the rigid-body movement seen below 500 Hz (left) and the mass-spring motion seen at 1000 Hz (right). Taken from Mcleod and Culling 2018.

2.3.5 The Outer Ear

When the skull is vibrated via BC hearing it moves differently relative to the surrounding air. This compresses and expands the surrounding air and creates airborne sound. A similar phenomenon happens when the ear canal walls are vibrated as a result of BC. The movement of the ear canal walls causes the air inside the ear canal to be expanded and compressed which creates a source of AC sound within the ear canal which stimulates the tympanic membrane in the same way airborne sound waves would. In effect, this creates BC-induced AC sound. This pathway however is only effective if the ear canal is not blocked as with a blocked ear canal there is no air to be expanded or compressed (Stenfelt et al., 2003).

The innermost section of the ear canal is surrounded by bony tissue and can be called the bony ear canal, and the outermost section is surrounded by cartilage and soft tissue, being called the cartilage part of the ear canal. When a BCD is placed near the ear canal (e.g. mastoid position), the cartilage ear canal is effective at transmitting low frequency sounds, although the ear canal is less important at transmitting high frequency BC sound (Stenfelt et al., 2003). The sound from the outer ear BC pathway is 20dB to 40dB lower than other pathways according to BC models (Stenfelt, 2016). While it is considered of low importance compared to the middle and inner ear pathways, some studies suggest that below 800Hz the outer ear becomes the more dominant pathway, producing ear canal sound pressure greater than AC stimulation (Huizing, 1960; Kanna et al., 1976).

2.3.5.1 Occlusion Effect

When the outer ear canal is closed, BC hearing is also affected. This is often referred to as the occlusion effect. When there is incomplete occlusion of the ear canal (meaning the ear canal opening is blocked but the rest of the ear canal is unimpeded) BC hearing can improve by up to 40dB at low frequencies (Stenfelt & Reinfeldt, 2007). This is because incomplete occlusion still allows the ear canal walls to vibrate and create AC sound, however that AC sound no longer leaks out of the ear canal, it is trapped in the ear canal and is transmitted to the middle ear. However, when the ear canal is completely occluded (so that there is no AC stimulation), while ear canal sound pressure increases, the threshold measurements does not improve at the same magnitude, suggesting the outer ear is less influential than other pathways (Goldstein & Hayes, 1971). Reinfeldt et al. (2013) compared the change in ear canal sound pressure to the change in hearing thresholds when the ear canal was occluded (18mm insertion) vs no occlusion (Figure 2.9). Significant differences between the ear canal sound pressure and thresholds were only seen below 500Hz and at 3000Hz and 4000Hz, although a significant difference was seen at 2000Hz for forehead stimulation. There were no significant differences between the higher and lower frequencies. This suggests that while the effect of the ear canal radiation may not be influential at lower and higher frequencies, it has an impact on hearing thresholds at middle frequencies.

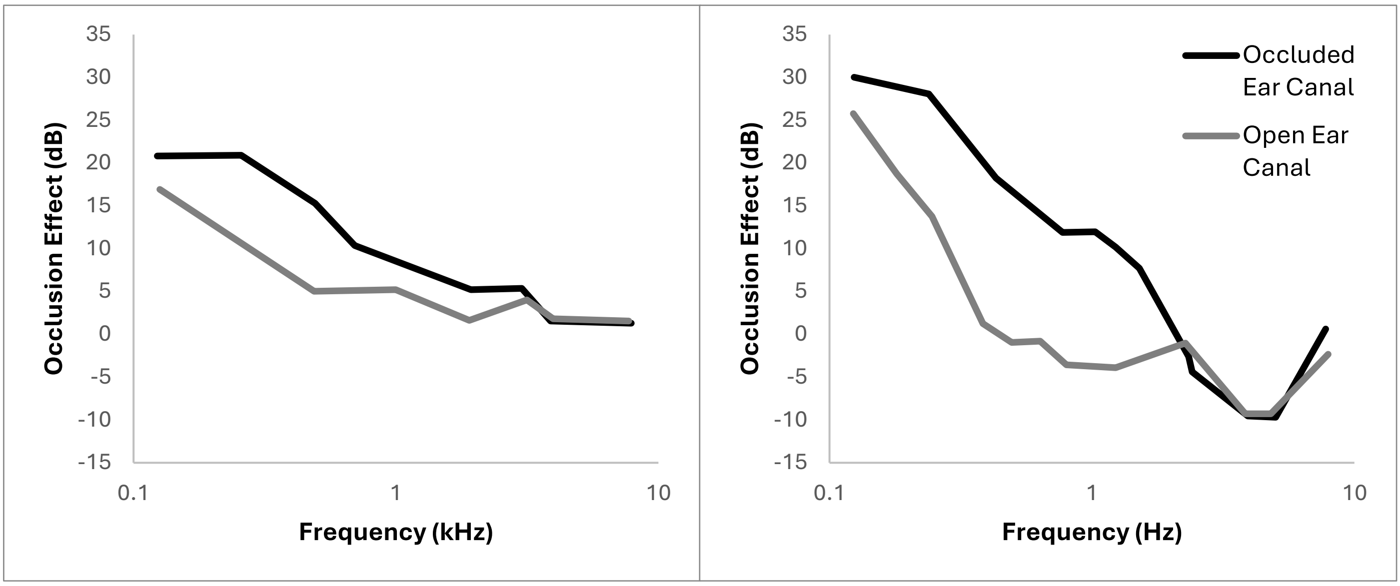


Figure 2.9. Occlusion effect measured when participants are wearing either earplugs (black line) or earmuffs (grey line). The left graph shows participant's subjective measurements of occlusion effect, and the right graph shows microphone-in-ear measurements. Graph reproduced from Reinfeldt et al. (2013).

2.3.6 The Middle Ear

The middle ear is comprised of ossicles - three bony structures: the malleus, the incus, and the stapes, which connect to the oval window, as well as the muscle and tissue that surround these bones. The ossicles are not fixed to the skull and, because of their relatively loose attachment, can vibrate independently of the skull when bone vibration is applied to the skull. This inertial movement will have some impact on the sound produced as a result of skull vibration.

The ossicles are primarily involved when vibration is applied at the mastoid bone (Studebaker, 1962; Dirks & Malmquist, 1969, Goodhill et al., 1970). An individual's ossicles show a high degree of variation compared to others, particularly at high frequencies, but the vibration of the ossicles is 5 dB greater than the vibration of the cochlear fluid (Stenfelt, Hato & Goode, 2002). There is also a negative phase change at higher frequencies (above 3kHz) between the stapes footplate and the promontory bone. Adding occlusion to the ear canal has no effect on either of these differences. At low frequencies the ossicles move in phase with the skull, as the inertial force of the ossicles is not enough to produce movement at these frequencies. At high frequencies (above the resonance frequency of the ossicles: 1.5 kHz), the ossicles become decoupled from the middle ear bones and vibrate independently (Stenfelt & Goode, 2005). Worsened BC thresholds around 2 kHz are seen following otosclerosis of the stapes (Carhart, 1971) and increased BC sensitivity is seen when lowering the resonance frequency of the ossicles (Huizing, 1960).

2.3.7 The Inner Ear

As with the outer and middle ears, the inner ear also provides a source of BC hearing. Skull movement will vibrate the inner ear structures, which in turn creates a wave on the basilar membrane based on the frequency of the sound. The inner ear is the most important contributor to BC hearing, which has been demonstrated through models (e.g. Taschke & Hudde, 2006). In addition to models, studies of patients with otosclerosis which affects the pliability of the round and oval windows only results in a maximal loss of 15-25 dB at 2 kHz (Garcia-Iza et al., 2016), suggesting the vibration directly to the cochlea is the primary BC pathway. This fact is used by audiologists to diagnose conductive hearing loss. AC thresholds can be compared to BC thresholds since the impact of the outer and middle ear on BC thresholds is minor compared to their impact on AC thresholds (Stenfelt, 2013).

One of the ways in which the cochlea can be stimulated is through the vibration of the non-bone structures of the skull. Sohmer et al. (2000) compared the brainstem evoked responses of a transducer on the skull compared to the fontanelle, craniotomy site, and eye, finding no difference in brainstem response between the different sites. They also found that placing the transducer above thinner bone (e.g. temporal) produced better thresholds compared to thicker bone, suggesting that sound penetrates the skull at the location of the vibration source. Studies have shown the impact of the semicircular canal on BC thresholds by studying superior semicircular canal dehiscence (a lesion in the semicircular canal) and have reported 5-15dB lower BC thresholds (Rowoski et al., 2004; Songer and Rowoski, 2010). This is caused by cochlear fluid escaping into the semicircular canal, reducing the overall impedance of the cochlear fluid.

One of the initial pathway theories is that the compression and expansion of the cochlear walls, due to transverse waves across the skull, cause displacement of the cochlear fluid (Chan et al., 1926). When the walls of the cochlea are compressed, the cochlear fluid is displaced since it is incompressible. The displaced fluid causes both the round and oval windows to bulge outwards, and once the compression has ceased, fluid movement is produced. Since the round window has a lower impedance than the oval window it creates fluid motion towards the scala tympani, creating a sound percept in the process. This is also aided by the difference in volume of cochlear fluid between the scala vestibuli and tympani (about a 5:3 volume ratio). However, this pathway may not be impactful at lower frequencies since the head moves as a rigid body, so there is no vibrational wave to compress the cochlear fluid. This causes the inner ear to contribute more to BC hearing as frequency increases (Taschke & Hudde, 2006). In addition to this there are studies which suggest that this pathway may not be a substantial one in BC hearing. When the compliancy of the oval window is compromised (e.g., otosclerosis), it would be expected that BC hearing would be improved since all the fluid would have to be displaced towards the scala tympani and result in increased stimulation of the basilar membrane however the opposite is observed (Carhart, 1971), although despite this, Kim et al. (2013) maintain that compression dominates BC hearing below 750Hz.

There are many pathways into the inner ear other than the round/oval windows and the cochlear wall. These pathways are referred to as third-window structures (and include nerve fibres, veins, cochlear and vestibular aqueducts as well as microchannels into the cochlea (Küçük, 1991, Stenfelt & Goode, 2005)). These compliant third window structures may explain why patients who have problems with their oval window are still capable of BC hearing. Tonnondorf et al. (1996) investigated third window structures in cats finding that while they were involved in BC hearing, there were no third window structures involved in AC hearing. Even though these third window structures might only produce small movements they are still impactful on BC hearing since even a small amount (less than one millionth of total) cochlear fluid is needed to induce hearing of 80-100dB (Stenfelt & Goode, 2005). As the skull vibrates as a rigid body at low frequencies, low-frequency BC hearing cannot be explained by ossicle or cochlear wall movement suggesting cochlear fluid inertia is the primary contributor to BC hearing at low frequencies.

2.3.8 Perception of BC Sound

Stenfelt and Zeitooni (2013b) expanded on previous research by Stenfelt and Hakansson, 2002 who tested the BC level change required to match a 50 dB AC change, finding that at the lowest frequency (0.25 kHz) the largest proportional change was required (0.8dB BC change gave the same perceptual loudness as 1dB AC change). Ratios between 0.9 and 0.93 were found above 1kHz. Results above 1kHz were similar between normal hearing and sensorineural hearing-impaired participants, but below 1kHz BC was perceived to be louder in hearing-impaired participants. Stenfelt tested participant's subjective loudness ratings when listening to BC or AC sound. They found that for AC sound, high frequency and low frequency sounds had the same perceptual loudness for each dB SPL increase, whereas for BC, lower frequencies required a higher vibration force to be perceived as loud as their higher frequency counterparts. The BC condition also had a smaller dynamic range (range of dB to go from inaudible to 'too loud'), for AC a range of 81.2 dB was observed for low-frequency and 80.3 for high-frequency compared to 71.4 dB and 74.1 dB respectively for BC.

2.3.8.1 SRM

Stenfelt and Zeitooni (2013a) demonstrated the SRM deficiencies of BC sound. When sound and noise are presented from the front (S_0N_0) AC and BC SRM was the same. When the noise came from 45° (S_0N_{45}) or 90° (S_0N_{90}) SRM was 4.5 and 4.0 dB respectively for BC, but 8.6 and 7.6 dB during AC presentation. SRM is slightly poorer when the BT is placed at the BCHD implantation site (Zeitooni et al., 2016, shown in Fig. 2.10). The difference in SRM between AC and BC is greatest between 105° and 180°, and SRTs match this pattern (Wang et al., 2024). The poorer performance observed in BC may be caused by the skin acting as a low-pass filter (Stenfelt & Håkansson, 1999).

2.3.8.2 BMLD

Stenfelt and Zeitooni (2013) also measured the BMLD and BILD when using BC sound. BILD benefitted from AC stimulation over BC when the phase of either the noise (S_0N_{180})

or the speech ($S_{180}N_0$) was inverted. However, BC stimulation slightly outperformed AC in the BMLD task using a non-stationary oscillating chirp and BC also performed better when the phase of the noise was inverted (S_0N_{180}). When the signal's phase was inverted ($S_{180}N_0$), AC substantially outperformed BC (11.7 dB and 4.9 dB BMLD respectively). The poor BC performance in the $S_{180}N_0$ condition may be caused by the signal and the crosstalk from the contralateral BC cancelling at the ipsilateral cochlea and vice versa for the contralateral cochlea (Stenfelt & Zeitooni, 2013). Positioning the BT at the BCHD implantation site does not affect either the BMLD or BILD (Zeitooni et al., 2016).

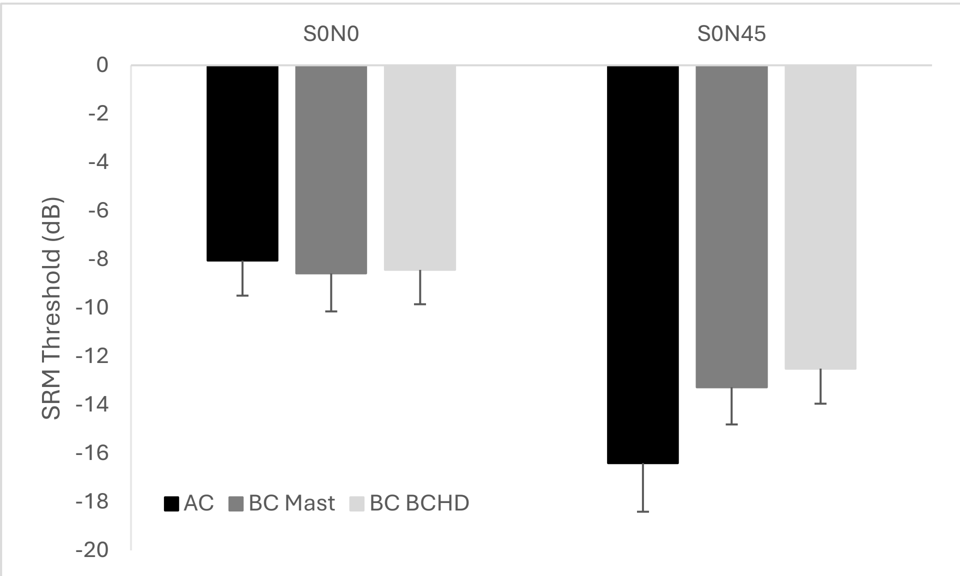


Figure 2.10. The mean SRM threshold at 50% intelligibility for AC (black), BT placed at mastoid (dark grey), and BT placed at BCHD position (light grey). Error bars represent 1 standard deviation. Graph adapted from Zeitooni et al. (2016).

2.3.9 Individual Differences in BC

Although there are individual differences in AC hearing (Surprenant and Watson, 2001; Buss et al., 2007; Kidd et al., 2007; Akeroyd, 2008; Horwitz et al., 2012), many of the physical structures, such as tympanum and ear canal, which effect AC hearing would have little effect on BC hearing, whereas facial anatomy and bone structure would be more influential in BC hearing. McBride et al. (2008), showed little differences between males and females, only finding that females had significantly lower thresholds than male normal hearing listeners at 8000 Hz (Hodges & McBride, 2012), and this difference may be affected by BC placement (Pollard et al., 2015).

Pollard et al. (2017) measured the effect of listener facial morphology on speech intelligibility in BC hearing. Listeners with shorter heads yielded better intelligibility, potentially due to the shorter distance the BC sound has to travel. Listeners with a shorter stature also produced higher intelligibility scores although this is likely because shorter statures were correlated with smaller head sizes. Despite these factors, when vibration responses of different skulls are compared, although deviations are observed, different skulls produce similar vibration in response to BC stimulation (Stenfelt & Goode, 2005).

In a study of skull impedance in BCHD patients, Håkansson et al. (2020) compared the difference in skull impedance across participants finding no difference of gender. Participant's over 60 had a higher skull impedance than those under 60, although this was not statistically significant (p = 0.07 at 250-700 Hz). Attenuation is greatest in young children and decreases as children mature (Mackey et al., 2018), this is potentially caused by the smaller skull size and softer skull bone and tissue.

2.3.10 BC Placement

The placement of the BT/BCHD also has an effect on the force of the vibration reaching the cochlea. Stenfelt and Goode (2005a) found that positioning a BT 3cm either anterior or posterior to the ear canal produced the largest movement of the cochlear promontory and the least phase accumulation, measured using a laser doppler vibrometer. Eeg-Olofsson et al (2008) adjusted the position of a BCHD in cadaver heads and found that vibrations were increased the closer the implant was to the cochlear, becoming larger than the standard implant position. Stenfelt (2012) conducted a similar study where he compared a mastoid-placed BT to one at the regular implantation site and also found that the mastoid placement resulted in an average 2.5 dB threshold improvement across all frequencies, although this was not the case for every participant. In a recent study Dobrev et al. (2016) compared BC at 7 different positions. The only position to produce more vibration than the mastoid was slightly superior to the ear canal, near the mandibular condyle. Small threshold shifts were also observed in normal hearing participants when stimulation was at this position.

2.4 Crosstalk

Crosstalk is an auditory phenomenon where signals from either side of a listener cross over the head and reach the contralateral ear (opposite to the sound source), in addition to the ear ipsilateral (same side as) to the sound source. Figure 2.11 shows how crosstalk occurs in a loudspeaker system. C_{LL} and C_{RR} represent the ipsilateral pathways from the left and right speakers to the left and right ears respectively. C_{RL} and C_{LR} represent the crosstalk from the right speaker to the left ear (hence C_{RL}) and the left speaker to the right ear respectively. These crosstalk signals are problematic when trying to create an immersive 3D listening environment. If ILDs and ITDs were introduced at X_L and X_R to facilitate binaural perception, the crosstalk signals would interfere with this binaural information causing problems in localising sound as well as degrading the signal clarity. In consequence, stereo audio usually uses a technique called amplitude panning. A similar style of crosstalk also occurs when a listener is exposed to BC sound (figure 2.11). As with loudspeakers, the direct signals travel through the skull directly to the ipsilateral cochlea, however cross-talk signals also travel across the skull via the various bone conduction pathways described in the previously to the contralateral cochlea.

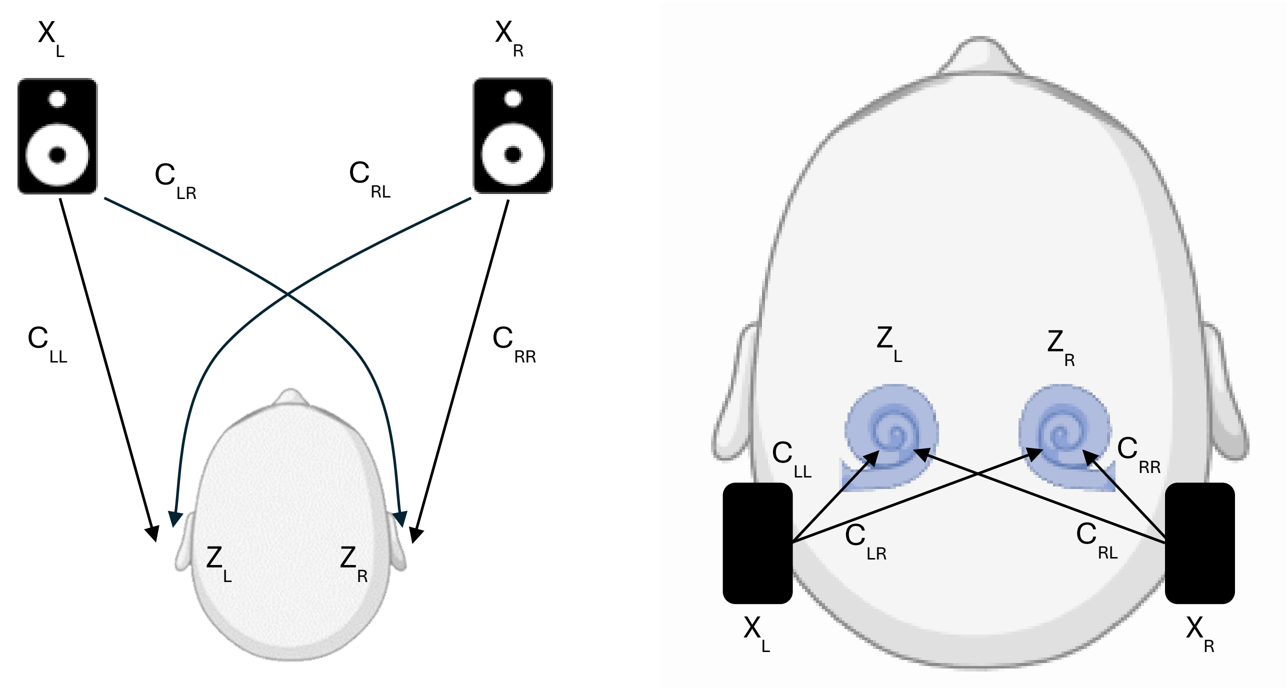


Figure 2.11. Diagrams showing crosstalk in loudspeakers (left) and bone transducers (right). X_L and X_R represent the left and right signals respectively, and Z_L and Z_R indicate the left and right cochlea. C_{LL} and C_{RR} are the left and right ipsilateral pathways while C_{LR} represents the crosstalk from the left source arriving at the right cochlea, and C_{RL} is the crosstalk from the right source arriving at the right cochlea.

2.4.1 Transcranial Attenuation and Delay

Transcranial attenuation (TA) is the reduction in level of a BC signal as it travels around the skull to the contralateral cochlea. Research of TA is typically done on dry skulls, cadaver heads, or patients with SSD (e.g. Håkansson, Carlsson & Tjellström, 1986; Stenfelt et al., 2003; Stenfelt & Goode, 2005; Liao, 2010; Guignard et al., 2013; Snapp et al., 2016), figure 2.12 shows the TA results from a number of these studies. Håkansson (1986) was one of the first to measure the effect of a contralateral BT on the ipsilateral cochlea, finding attenuation between 10 – 28 dB across the skull, noting an anti-resonance between 200 – 400Hz. Stenfelt and Goode (2005) also measured bone vibration of six cadaver heads finding that below 500Hz, phase cycles are similar across the skull, due to the skulls low-frequency rigid-body movement. Above 500Hz few time delays are seen for ipsilateral stimulation, hovering at a consistent 0.2 ms across all frequencies (0.5-10 kHz). Time delays vary between frequencies when the contralateral cochlea is stimulated. A delay of 0 ms is observed at 800 Hz, and a delay of 1 ms is seen at 2.5 kHz, but outside these frequencies the delay varies between 0.3 – 0.6 ms. These varying time delays are caused by the changing wave velocities of BC sound across the skull according to Tonndorf and Jahn (1981), who estimate the wave speed to be about 330 m/s across the living skill and closer to 2000 m/s in dry skulls.

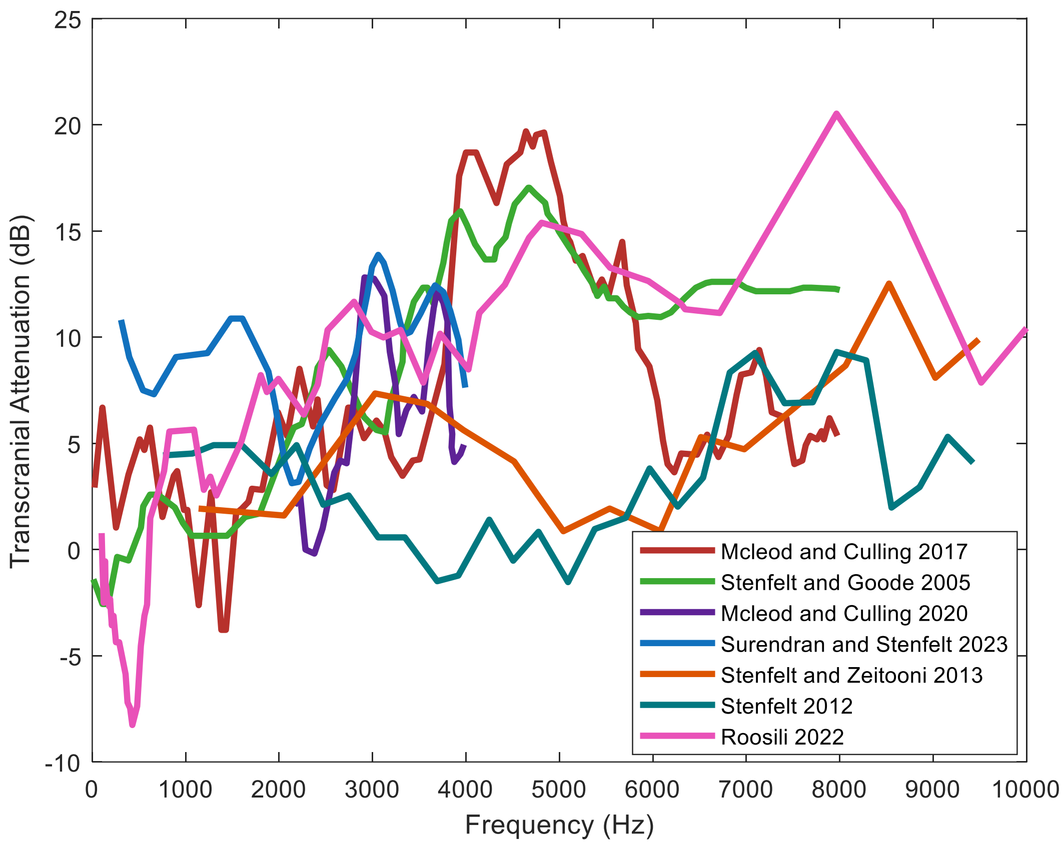


Figure 2.12. A graph comparing TA results gathered from studies of participants (Stenfelt & Zeitooni, 2013; Mcleod & Culling, 2017, 2020, Surendran & Stenfelt, 2023), cadaver heads (Stenfelt & Goode, 2005; Röösli, 2022) and unilaterally deaf patients (Stenfelt, 2012).

Below 500Hz, stimulation produces attenuation of +0-3 dB at the contralateral cochlea at all stimulation positions. When stimulation is close to the cochlea, attenuation is minimal around 500 Hz before increasing to -15 dB at 5 kHz and then returning to -10 dB at 6 kHz (Stenfelt and Goode, 2005; Farrell et al., 2017). Attenuation is at its highest when stimulation is closest to the cochlea, and less attenuation is seen when a dry skull is used compared to a cadaver head (Stenfelt et al., 2000). Vibration at each cochlea is similar below 500 Hz, and above this, attenuation varies between -20 dB and

30 dB, with most of the negative attenuation occurring below 1 kHz.The vibrational similarities between the cochleae at low frequencies are caused by the rigid-head movement described in the previous chapter. The ipsilateral vibration, and the crosstalk, are different when the BT is placed on the other side of the head, suggesting that the head is not symmetrical in terms of BC. This also means that neither the crosstalk (C_{LR} and C_{RL}) or ipsilateral (C_{LL} and C_{RR}) signals are equal, so when calculating attenuation, attenuation from each BT must be calculated individually.

2.4.1.1 Individual Differences in Transcranial Attenuation

Large variation in attenuation between participants has been obersved. When testing unilaterally deaf patients, Nolan and Lyon (1981) used a Radioear B71 bone vibrator and observed hearing threshold differences of up to 50 dB between participants at a given frequency and Stenfelt (2012) found that the average spread at each frequency was 35 dB and that every frequency (between 250 Hz – 10 kHz) revealed large variance between participants. The variation is potentially caused by the antiresonance of the individual skull, affecting the ipsilateral and contralateral vibration pathways differently. This variability means that attenuation cannot be assumed for any one person and must be individually measured, in addition to each side needing to be measured individually (Liao, 2010).

These interindividual differences are also seen in when measuring 3D promontory motion in cadaver heads. It has been shown that this variability increases with increasing frequencies (Dobrev et al., 2019), and that contralateral BC stimulation reveals significant attenuation for high, but not for low-frequency stimulation (Mattingly et al., 2020).

2.4.1.2 Occlusion and Transcranial Attenuation

Occlusion of the ear canal also effects TA (Reinfeldt et al., 2013). The most prominent effect is seen below 1 kHz where occlusion results in a 10-20 dB threshold decrease on the contralateral side, compared to a 5-10 dB threshold decrease on the ipsilateral side, although only 125 Hz, 250 Hz, and 500 Hz differed significantly between the two sides. Above 500 Hz little difference was seen in occlusion’s effect between the ipsilateral and contralateral signals. Occlusion had little effect on either pathway above 2 kHz, although a positive threshold shift of 7 dB was seen at 4 kHz during ipsilateral stimulation. Reinfeldt et al. (2013) suggest that this low frequency effect is due to occlusion of the ear canal causing BC to become reliant on ear canal sound pressure (ECSP), which affects the contralateral pathways more than the ipsilateral pathways during low frequency occlusion. This low frequency effect is likely caused by the inhibition of the soft tissue pathways, which only affect BC below 1.5 kHz (Stenfelt, 2003). The difference between occlusions effect on ipsilateral and contralateral thresholds may be caused by differences in BC pathways, particularly if contralateral stimulation utilizes more soft tissue pathways.

2.4.1.3 Transcranial Attenuation Measurements in SSD

As mentioned in chapter 2 of the introduction, patients with SSD rely on the crosstalk that occurs with BCHDs to enable them to hear sounds coming from their deaf side at their hearing cochlea. Since hearing sound from the deaf side is reliant on the transmission between BCHD and opposite ear it is important to know to what degree intensity is lost across the skull. Nolan and Lyon (1961) measured the TA in 15 unilaterally death patients using a BT, finding an average attenuation of 11 dB which increased from 8.33 dB at 0.25 kHz to 16 dB at 4 kHz. Similar inter-participant variation to Stenfelt (2012) was seen in this study (between 0 – 30 dB). Snapp et al., (2016) utilised a similar methodology in 27 patients using a BT, finding a mean attenuation of only 5.12 dB. This smaller attenuation is due to lower peak TA values, at most frequencies, and the negative attenuation that occurred (TA varied between -17 – 32 dB). A regression analysis between TA and speech-in-noise detection was performed however there was no relationship between the two, suggesting that even if participant's have a high degree of TA they can still benefit from a unilateral BCHD.

A subsequent study by Beros et al. (2022) compared TA in SSD patients with either a B-71 BT,or a BCHD 5 transducer coupled to the skull using a 5 N steel band. The BCHD 5 transducer was also used in a cadaver head. TA for the transcutaneous BCHDs were minimal at the lowest frequency (250 Hz) and gradually increased up to 3 kHz where it plateaued. BCHD use resulted in a 5 – 10 dB higher hearing threshold than the B-71. No significant difference was found between patients and cadaver heads, and an inter-individual variation of up to 50 dB at a single frequency was observed.

2.4.1.4 Transcranial Attenuation Measurements in Normal Hearing Participants

In order to test bone conduction in normal hearing participants, bone conduction headsets are typically used. These are headsets with BTs attached which are placed in contact with the skull and vibrate the skull through the skin. Stenfelt and Zeitooni (2013a) used one such device on 20 normal hearing participants. They found that TA was around 5 dB at around 0.3 to 0.7 kHz, 0db between 0.9 and 1.6 kHz, then a steady increase from 0 dB to around 10 dB at 5 kHz.

The low attenuation of these contralateral signals across the head is problematic since these contralateral signals can interfere with the ipsilateral signals, which would already contain binaural information (such as the ILDs caused by the head shadow). Because there is little attenuation of the crosstalk signals, they would cause significant interference to the pre-existing binaural cues rendering them ineffective, causing the perceptual deficits associated with bilateral BCHD patients discussed in chapter 2.2. In addition to this, at frequencies where there is negative attenuation the sound can be perceived as coming from the side opposite to its source and a signal can be masked by the low or negatively attenuated contralateral crosstalk.

2.4.2 Crosstalk Cancellation

Crosstalk-cancellation is a method used to eliminate the crosstalk from a given sound source so that each signal is only reaching the same-side ear. Bauer (1961) was the first to explore the concept of crosstalk cancellation, but it was Schroeder and Atal (1963) who implemented the first stereo speaker cross-talk cancellation system. This was utilized by Schroeder (1973) who wanted to preserve the stereo separation of a mannequin head recording when played over two loudspeakers. They took the impulse response of the crosstalk (C_{LR} and C_{RL} in figure 2.14) and added the inverse of this to each speaker. Upon reaching each ear the ipsilateral signals (C_{LL} and C_{RR}), with the cancellation wave added, would cancel the crosstalk at each ear, preserving the stereo separation of the recording.

In stereo speaker systems the impulse response between each speaker and the listener's head must be known to cancel the crosstalk to each ear. Inversion of these impulse responses and application to the signal from each speaker allows for stereo speaker crosstalk cancellation (Damaske, 1971), like that used by Shroeder (1973). The addition of this crosstalk signal does not negatively affect hearing since it is masked by the target signal (Damaske & Mellert, 1969). In a method similar to that used in this thesis, Damaske (1971) asked a participant to sit between two loudspeakers, playing identical tones, and adjust the phase and attenuation of one of the sounds until sound could only be heard on one side. This created the cancellation tone for each ear and when reaching the listeners ears these cancellation tones would eliminate the crosstalk from the contralateral loudspeaker and restore spatial separation of the sounds.

The limitation of this method is that it requires the listener to stay stationary as any change in head position will change the transfer function between each speaker and ear, and since the crosstalk cancellation filters are sensitive to even small phase/level changes, these small changes will reduce the effectiveness of the crosstalk-cancellation filters. If the head is moved without the filter being recalibrated the filter will become less effective since the transfer function, and the cancellation phase/level will have changed. This also needs to be calculated for each individual, as head, external ear and torso shape will have an effect on the transfer function (Hamada et al., 1983).

Liao (2010) describes several methods of crosstalk cancellation as well as measuring transfer functions in a dry skull and how this information can be used to design and implement crosstalk-cancellation filters. The four methods of crosstalk cancellation he describes are: ideal, recursive, fast deconvolution and multi-error least mean square. Ideal crosstalk cancellation attempts to find the crosstalk waveform and present the inverted waveform so that the two destructively interfere and result in the remaining ipsilateral signals without any crosstalk. This cancellation is represented in matrix form in equation 1 (using the same notation as seen in figure 2.11), where X represents the signal, C represents the transfer function, C_{LL} and C_{RR} represent the ipsilateral signal paths and C_{RL} and C_{LR} represent the contralateral signal paths, and Z represents the sound that arrives at each cochlea.

$$1)\quad \begin{bmatrix} Z_L \\ Z_R \end{bmatrix} = \begin{bmatrix} C_{LL} & C_{RL} \\ C_{LR} & C_{RR} \end{bmatrix} \begin{bmatrix} X_L \\ X_R \end{bmatrix}$$

For the crosstalk (C) to be eliminated, a second matrix (H) needs to be used, which contains inversions of C_{RL} and C_{LR} (equation 2). When H is multiplied by the crosstalk, it should eliminate the undesired signals leaving only C_{LL} and C_{RR}, effectively restoring stereo hearing.

$$2)\quad \begin{bmatrix} C_{LL} & C_{RL} \\ C_{LR} & C_{RR} \end{bmatrix} \begin{bmatrix} H_{LL} & H_{RL} \\ H_{LR} & C_{RR} \end{bmatrix} = \begin{bmatrix} 1 & 0 \\ 0 & 1 \end{bmatrix}$$

Since matrix H is an inversion of matrix C it can be calculated using the following formula:

$$3)\quad H = \frac{1}{C_{RR}C_{LL} - C_{RL}C_{LR}} \begin{bmatrix} C_{LL} & -C_{RL} \\ -C_{LR} & C_{RR} \end{bmatrix}$$

Ideal crosstalk cancellation does however have limitations. The first is termed the 'ill-condition' by Liao (2010). This occurs at low frequencies when the multiplication of the crosstalk (C_{RL} and C_{LR}) is similar to the multiplication of the same-side signals (C_{LL} and C_{RR}). This is a problem since it would produce a large H value matrix which would require too much amplification for the speakers to produce audible cancellation. A second problem, as mentioned earlier, is that the filters require a stationary head when the sounds are played over speakers. If the head moves then the sound pathway between the sound source and the contralateral ear changes, causing any previously-calibrated filter crosstalk-cancellation filter to become ineffective, and even detrimental to binaural hearing. When the head moves the values of C_{LR} and C_{RL} change, meaning that for ideal crosstalk cancellation to occur H_{LR} and H_{RL} would need to be recalculated.

2.4.2.1 Implementation of Crosstalk Cancellation in Bone Conduction

Various methods have been used to calculate the phase and amplitude shift that need to be applied to a soundwave to cancel the crosstalk from a contralaterally-placed BT. As previously demonstrated by von Békésy (1932), AC and BC both affect the Basilar membrane which means that either an AC or BC waveform can be used to cancel, or amplify, another BC waveform. An AC tone and BC tone of slightly different frequencies can also be used to create a ‘beating’ effect, further demonstrating that BC sound acts on the cochlea in the same way as AC sound (Wever and Lawrence, 1954).

2.4.2.2 Subjective BC Crosstalk Cancellation using AC

Mcleod and Culling (2017) used AC sound from ER-2 insert earphones to cancel the crosstalk produced at a contralateral mastoid-placed BT. They allowed the participant to adjust the phase and level of an AC cancellation tone until the sound is heard only on the BT side. This allowed participants to manually find the correct phase and level shift required to create a cancellation tone to be added at the ipsilateral ear. They referred to this as the “one-BT” method. When compared to Stenfelt and Goode’s (2005) results, Mcleod and Culling found that the phase difference between each frequency interval was larger than Stenfelt and Goode measured in cadaver heads below 4kHz but was similar above this frequency. Large variance in cancellation levels and phase accumulation were also seen between participants, and is a consistent theme across crosstalk-cancellation studies (Stenfelt & Goode, 2005; Rowan & Gray, 2008; Mcleod & Culling, 2017,2019,2020; Surendran & Stenfelt, 2023).

AC sounds can also be used to cancel BT sound at soft tissue sites such as the eye, craniotomy site, or fontanelle (Sohmer et al., 2000; Ito et al., 2011). Chordekar et al. (2011) tested soft-tissue AC cancellation of BC tones in normal hearing participants using a similar method to Mcleod and Culling (2017), finding that total cancellation was possible at the mastoid, neck and eye. This was later confirmed in a head simulation model (Stenfelt and Provanovic, 2022).

2.4.2.3 Objective Crosstalk Cancellation

Objective measures can also be used to obtain crosstalk-cancellation filters. One method is using otoacoustic emissions (OAEs); BC sound can evoke an otoacoustic emission if there is a functioning middle ear and ear canal (Clavier et al., 2010; Usugawa, 2017). Measurement of these emissions can be used to create the cancellation filters needed to perform crosstalk cancellation. Wang et al. (2023), used OAEs to measure the phase and amplitude of the crosstalk from a contralateral BT and utilized this data to create a crosstalk-cancellation filter, a localisation test for sound delivered over two BTs was then conducted with and without the filter applied. Little improvement was seen in participants when the filters were used to cancel broadband crosstalk (mean average errors (MAE) = 49.17° without crosstalk cancellation and MAE = 47.67° with crosstalk cancellation), however substantial improvements were seen in 3 out of 5 participants when narrowband crosstalk was cancelled (MAE = 39° without crosstalk cancellation and MAE = 29.83° with crosstalk cancellation), although this was still worse than AC localisation (MAE = 24.5°). The better performance in the narrowband condition suggests the frequencies which the narrowband targets (2 – 4.5 kHz) were more accurately cancelled by the crosstalk-cancellation filter, meaning the OAE measurements would have also been more accurate at these frequencies.

Ear canal sound pressure (ECSP) can also be used to measure crosstalk. Usagawa (2019, 2020) used a BC microphone in the ear canal to measure the crosstalk from a contralateral BT, and used these measurements to create a crosstalk-cancellation filter. This filter however was only effective below 1.1 kHz. Above 1 kHz the BC microphone recorded little attenuation across the skull, which is inconsistent with other BC studies, suggesting that the ear-canal microphone was not accurately measuring the crosstalk above 1 kHz, and therefore that the ear-canal sound pressure was no longer in phase with the relevant signal at the cochlea. In a later study Otsuka and Nakagawa (2022) used ECSP-measured crosstalk to create crosstalk-cancellation filters and implement them to measures TRTs. When the crosstalk-cancellation filter was implemented, ECSP was lower at all frequencies (250 – 1000 Hz) and most participants showed threshold improvement in the TRT test. This was later used to create a real-time cancellation system where the impulse response at the sensor was taken and run through a training algorithm to find the point of maximal cancellation (Otsuka & Nakagawa, 2023a). Cross-talk cancellation was measured at the sensor and some success was seen in a TRT test, this method suffered from the same problem of cancellation occurring at the sensor rather than the cochlea that their previous study suffered from, as well as being an unviable procedure in many BCHD patients, because their conductive loss lies between the cochlea and the microphone site. Otsuka and Nakagawa (2023b) then used an accelerometer placed behind the ear to measure contralateral vibration. Cancellation was seen at the accelerometer, and a TRT change of 4.2 dB was seen although this was mostly between 250 – 397 Hz. Overall, these methods had limited effectiveness.

While the measurements of OAEs and ECSP, as well as AC masking techniques, can be used to produce crosstalk cancellation filters for low frequency BC sounds they require a working middle ear and ear canal to be performed. This is often not the case in patients who use BCHDs, and the use of an accelerometer placed behind the ear only showed limited effectiveness meaning that alternative methods of measuring the crosstalk need to be used.

2.4.2.4 Subjective BC Crosstalk Cancellation

Tonnondorf and Jahn (1981) conducted an experiment where the vibrations from one BT could be used to cancel the crosstalk from a contralateral BT at the ipsilateral cochlea. The procedure was based on a previous experiment by Zwislocki (1973) who played a tone loud enough into one earphone that it caused that tone to be perceived at the contralateral cochlea via BC. He then asked participants to adjust the phase of a tone played to the contralateral earphone until it cancelled the BC crosstalk from the ipsilateral earphone. As the participant adjusts the phase of the tone it would slowly become lateralised towards one ear before disappearing completely once the participant has selected the correct cancellation value. Phase accumulated linearly with frequency (about 200° per 500 Hz). This occurs because as the frequency increases a larger phase change happens as the sound travels across the skull. Tonnondorf and Jahn repeated the same procedure but using two Radioear B-72 BTs instead of earphones. They found a phase accumulation of around 200° per 1000 Hz, although the phase rapidly increased from around 0° at 500 Hz to 400° at 1000 Hz.

In order to devise a method of crosstalk cancellation that is applicable to patients, Mcleod and Culling (2019) adapted the procedure from Tonnondorf and Jahn (1981) to also include level manipulation of the cancellation tone. Mcleod and Culling (2019) allowed participants to adjust the phase and level produced at an ipsilateral BT to cancel the crosstalk from a contralateral BT, demonstrating the viability of the two-BT method. Values from the two-BT cancellation were the same as those gathered from the one-BT cancellation. Mcleod and Culling (2020), then used this two-BT method to create a crosstalk-cancellation filter which was used to cancel the crosstalk from a contralateral BT in a signal-in-noise task: when a noise was presented at one BT and a signal at the other, threshold was reduced if a cross-talk cancellation signal was added to the signal. Improved thresholds were observed across all frequencies (1200-4035Hz) in both TRTs and SRTs. The main difficulties observed in these experiments were the time-consuming participant training, and needing to recalibrate the filter after each testing session. Recalibration was needed because when the BTs are removed and replaced on the head, they will be in a slightly different position and so the transfer functions between the BT and each cochlea will change. Participants also struggle when cancelling low frequency tones, and cancellation below 1000 Hz is limited due to the similarity in phases between the sound at the ipsilateral and contralateral cochleae. Since the phases of the tones are similar at each cochlea, when the crosstalk-cancellation tone is introduced at the contralateral ear, the crosstalk from that will cancel the signal at the ipsilateral ear. This would make lateralisation of a tone difficult since both the ipsilateral signal and crosstalk are being cancelled simultaneously, and if crosstalk cancellation was applied it would also cancel the signal (Mcleod & Culling, 2017; 2019). Surendran and Stenfelt (2023) worked around this problem by using AC masking at the contralateral ear, changing the task from lateralisation to a cancellation task. This means that rather than listening for a sound to move to disappear at one ear, which can be difficult when it can be heard at the contralateral ear, the participant only needs to listen for the sound the disappear entirely.

2.4.2.5 Limitations of Crosstalk-Cancellation

There are several challenges that currently exist within BC crosstalk-cancellation. The first is the issue of recalibration between sessions. Currently the continual recalibration makes the filters impractical since the time spent recalibrating the filters would exceed the benefit from the crosstalk cancellation. Mcleod and Culling (2020) attribute the changing filter parameters between sessions to the repositioning of the BTs. If this is the case patients with percutaneous abutments or subcutaneously implanted transducers may not see the same changes in crosstalk attenuation over time, because their devices have a fixed location and there would be no need for recalibration of the filters, making it a more practical solution to the crosstalk. The alternative would be an automated measurement of crosstalk, similar to that seen in Otsuka and Nakagawa (2023a), however no solution has been found that would be applicable to BCHD users.

There is the additional obstacle of low frequency cancellation, mentioned previously. While attempts have been made to circumvent this by either using objective measures (Usugawa, 2019, 2020; Otsuka and Nakagawa, 2022, 2023a, 2023b; Wang et al., 2023) or by using masking of the non-test ear to change the experiment from a lateralisation task to a cancellation task (Surendran & Stenfelt, 2023), these methods rely on a working middle ear and ear canal which patients using bilateral BCHDs often do not have.

2.4.3 Skin Attenuation

Skin attenuation provides an additional variable when using BTs (e.g. in normal hearing participants) rather than percutaneous implants which could have an impact on crosstalk cancellation. Håkansson (1986) observed skin attenuation of 0 – 5 dB in patients using BCHDs, although they also found an overall impedance difference of 10 – 30 dB between an implanted screw and an over-the-skin transducer. This was corroborated in Chang and Stenfelt's (2019) human-head model comparing different BC devices; they found that differences in vibration were as high as 20 dB at higher frequencies when a B71 was compared to a BCHD, although little difference was seen below 1.5 kHz. A greater effect was seen for contralateral stimulation, and large phase differences were also seen between the two devices at all frequencies. In Beros et al. (2022) skin attenuation (gathered comparing transcutaneous and percutaneous stimulation using BCHDs) was between 3 dB and 20 dB for patients and increased with frequency. Skin attenuation seen in patients was 3 – 12 dB greater than seen in cadaver heads and inter-individual variation was greater in patients (M = 27.2 dB, up to 40 dB at a single frequency) than cadaver heads (M = 6.6 dB, up to 11.9 dB at a single frequency).

Beros et al., (2022) also compared a B71 BT to a percutaneous and transcutaneous (via steel head band) BCHD 5 transducer, finding that hearing levels were similar between the 3 below 1 kHz but between 1 and 2 kHz the transcutaneous BCHD produced 10 – 20 dB poorer hearing thresholds compared to the other two configurations and between 4 – 6 kHz a percutaneous BCHD resulted in 20 dB lower thresholds compared to the other 2 conditions.

The skin attenuation seen when using a transcutaneous transducer can be affected by the force applied to the transducer against the skin, however as more force is applied the transducers cause more discomfort to the participant. Early on, Corliss and Koidan (1955) found no significant difference in head impedance when the force to a BT exceeded 5 N, although improvements in hearing sensitivity of up to 10 dB have been seen when the force is increased from 1.5 N to 10 N (Nilo 1968; Khanna et al., 1976). In a more recent study by Toll et al. (2011), an improvement of only 1 - 1.5 dB was observed when the static force applied to a B-71 BT increased from 2.4 to 5.4 N, and discomfort ratings were only significantly higher at 5.4 N compared to the other conditions. Incidentally, they also found that BC thresholds were slightly higher at 250 Hz and 1000 Hz when using a Radioear P-3333 steel headband, and that participants rated it a 3 out of 5 on the discomfort scale compared to 1 – 1.5 for the BTs. The findings from Toll et al. (2011) suggest that the optimal force to apply to the BT is 2-4 N for maximum comfort without greatly impacted thresholds, and to avoid using a steel band.

2.5 Current research

The primary goal of this dissertation is to expand on the work done by Mcleod and Culling (2020) in creating a crosstalk cancellation method that can be used in bilateral BCHD users.

The goal of the first experiment was to take the method from Mcleod and Culling (2020), but to use a filter design that could be used in bilateral BCHDs and to apply this filter at both BTs simultaneously. The main changes to the filter design were calibrating the filter in 62.5 Hz intervals, and to use a sample rate of 16000 Hz in order to use a filter design which would be directly translatable to a patient's BCHDs. The filters were then applied to both the BT playing the tone and the BT playing the noise in the TRT test, in theory separating the stimuli so that only the tone is heard at the cancellation ear, and only the noise is heard at the non-cancellation ear.

Once the feasibility of this filter design method had been demonstrated, a similar protocol was used in patients with bilateral BCHDs. The main goals of this research are to test whether patients with bilateral BCHDs are able to create adequate filters, but also to determine whether these filters remain robust over time, which is not the case in normal hearing participants (Tonnondorf & Jahn, 1981; Mcleod and Culling, 2017,2019,2020), but should be in BCHD users if the BT positioning is responsible for the changing crosstalk between sessions. If the latter is proven to be the case, self-calibrated crosstalk-cancellation filters may be an effective solution for crosstalk in patients.

A third experiment testing the localisation ability of participants using either AC or BC sound was also conducted, with both feedback and multiple sessions implemented. The purpose of these interventions were to measure whether improvements were seen after implementing practice or feedback. If successful, practice could be used alongside crosstalk cancellation to aid patients hearing.

Chapter 3 - Simultaneous Crosstalk Cancellation in Normal-Hearing Participants

3.1 Summary

Unilateral crosstalk cancellation has previously been demonstrated in BC hearing by using a pure tone played at one BT, to cancel the crosstalk from a tone played at a contralateral BT. The present study demonstrates that self-calibrated transfer functions can be used to achieve bilateral crosstalk cancellation using finite impulse response (FIR) filters to simultaneously cancel cross-talk from both BTs at both cochleae. Cancellation was performed in 1 kHz wide bands, calibrated at 62.5 Hz intervals. Participants were initially trained to cancel simulated crosstalk presented over headphones. Once proficiency had been demonstrated, participants attempted to cancel crosstalk from a tone at one B71W bone transducer by adjusting the phase and amplitude of an identical tone at a contralateral BT. Most participants were able to cancel the crosstalk, although some participants required substantial practice using the BTs. Individual cancellation phases were similar between sessions although there were frequency ranges where the values from one session varied from another, presumably due to small changes in BT positioning between sessions. Cancellation amplitudes showed more variation between sessions than the cancellation phases. Once crosstalk-cancellation filters had been created for both cochleae, they were applied bilaterally in a masked tone detection task with a tone at one BT and noise at the other. An improvement in threshold of between 5 dB – 15 dB was seen on each side with the least improvement seen at the lowest frequency (1.5 kHz). The results suggest that bilateral crosstalk would be effective in patients with bilateral BCHDs.

3.2 Introduction

Bone-conduction hearing devices are hearing aids that use bone conduction to bypass the ear canals and middle ear to aid patients with conductive or mixed hearing loss who are unable to benefit from AC hearing aids (Tjellström et al., 2001). BCHDs have been an effective treatment and patients report preferring BCHDs to other hearing aids (Dutt et al., 2002). However, patients are typically only fitted with one because bilateral implantation has limited effectiveness due to the crosstalk caused by the vibration of sound waves across the skull. The crosstalk is problematic because of the low levels of transcranial attenuation that occurs when using BC sound (around 10dB according to Snik et al., 1998). The crosstalk interferes with the ipsilateral signal at each cochlea. Despite this crosstalk there are some residual benefits associated with bilateral implantation, although these do not reach AC levels (Bosman, 2001; Priwin et al., 2004, 2007). Crosstalk cancellation is a potential method for reducing crosstalk in bilateral BCHDs and so improving performance across binaural tasks.

3.2.1 Objective Measurement of Skull Transmission

Håkansson (1986) measured the vibration of the skull when a BT (both transcutaneous and percutaneous) is placed on the contralateral side of a patient's head. Attenuation across the skull was 10- 28 dB at all frequencies, and skin attenuation was between 0 dB and 5 dB. Attenuation peaked between 2 kHz and 4 kHz and was lowest at the extremes of the measured range (0.1-10kHz). Stenfelt and Goode (2005) compared 27 different BT placements on a cadaver head by measuring the vibration at the cochlear promontory. When the BT was on the contralateral side Stenfelt and Goode found cochlear vibration to be between 0 and 5 dB greater at 1kHz and between 10 and 20 dB lower at 10kHz. The greatest response with the shortest delay was found when the BT was placed close to the cochlear, although contralateral vibration was not affected by BT placement.

Skull vibration varies between individual skulls so although objective studies can demonstrate trends in attenuation across the skull, the attenuation from one skull cannot be used to predict the attenuation of another and so the crosstalk of each skull must be individually measured. This however is impractical since measurement of the phase and amplitude differences would need to be measured both at the contralateral BT and the ipsilateral cochlea, which would not be practical in living humans since vibration is measured in 3 different directions (x, y and z). This gives 3 transfer functions for the vibration of the skull however the contribution of each of these vibrations to the overall cochlear movement is not known, and without this information cannot be used to create and overall transfer function of the skull.

3.2.2 Subjective Derivation of Cancellation

Rowan and Gray (2008) demonstrate that when tones with varying IPDs (between -180° and 180°) are played over bilateral BTs, participants struggle to accurately lateralise sound. Large variation was seen between participants with one being able to accurately lateralise sound, two having poor sound lateralisation and two lateralising sounds on the side opposite than expected given the IPD. These differences in lateralisation ability are presumed to be because of the differing TAs of each participant's head creating both ITDs and ILDs different to those of the presented tones, causing confusion in spatial perception. Rowan and Gray present a model that suggests that this crosstalk can be cancelled if the phase and level differences between the ipsilateral cochlea and contralateral BTs is known which can be used with the previously mentioned 'ideal cross-talk cancellation' (Liao, 2010). To measure this phase and amplitude difference in human participants, perceptual tasks must be relied on since the previously mentioned methods used in cadaver and dry skull studies are not appropriate for human testing.

Mcleod and Culling (2017) demonstrated how an AC tone could be used to cancel the sound from a single mastoid-placed BT by adjusting the phase and level of the AC tone. This allowed them to calculate the attenuation of a BT tone across the skull. Cancellation phases and levels differed between participants and between cochleae. Sudden trajectory changes in phase accumulation were also seen when the session changed, speculated to be due to repositioning of the BT. Phase accumulation was linear for the contralateral ear and negatively accumulated up to around 3-4 kHz before reversing. Some similarities were seen between participants for level, but they were overall sporadic. Participants were asked to rate how lateralised the sound was following cancellation, perceiving sound as less cancelled at 1 kHz steadily increasing in lateralisation up to 7 kHz where it plateaued. Individual participants reported similar levels of cancellation efficacy. The phase accumulation seen in Mcleod and Culling (2017) was greater than seen in a previous study on cadaver heads (Stenfelt & Goode, 2005). The intracranial pressure and other soft structures (often termed 'third-window structures, Stenfelt & Goode, 2018) could be responsible for this difference. All cancellation levels were negative meaning that there was always some level of attenuation through the skull from BC sound with the largest being around 60 dB, both ipsilaterally and contralaterally. These large level differences between the ears are thought to be the product of the BC sound taking multiple pathways around the head which, upon converging at the basilar membrane, cancel each other out via destructive interference and result in a smaller percept (Eeg-Olofsson et al, 2011; Stenfelt 2005, 2012).

Mcleod and Culling (2017) also measured the inter-session reliability of the cancellation finding that the cancellation levels and phases changed significantly between sessions. This is presumed to be caused by the repositioning of the BT at the start of each session, altering the pathways the BC sound takes around the skull. Stenfelt and Goode (2005) however, report finding no significant difference in attenuation when the position of a contralaterally placed BT was adjusted. In a comparison of standard deviation between ipsilateral and contralateral cancellation Mcleod and Culling (2017) found no significant difference suggesting that BT repositioning impacts both the ipsilateral and contralateral cochlea equally.

Mcleod and Culling (2019) later conducted a study where they used a similar methodology but utilising a second BT to cancel out the first and compared the cancellation values with the values gathered using AC cancellation. The results from the AC-BC cancellation could be used to predict BC-BC cancellation phase and levels, although less accurately at high frequencies. Participants also found BC-BC cancellation more difficult than AC-BC cancellation, sometimes needing multiple attempts over multiple sessions to successfully cancel at a given frequency. Individual participants found different frequencies more difficult.

They also identified two scenarios in which participants struggled to cancel the crosstalk. The first is hypothesised to be a result of the anti-resonances mentioned earlier which cause large level differences between the sound produced by the BT and the sound at the cochlea, making the cancellation level difficult to find. The second occurs at frequencies where there is little phase difference between the signal and the crosstalk at the other ear. This means that cancellation will occur at both ears to an extent when trying to cancel the sound from one of the BTs since phases will be matched at both ears. This second scenario occurs primarily at low frequencies (although Mcleod and Culling also noticed cancellation difficulty at 5 kHz), due to the rigid-body and mass-spring skull movements caused by BC, and would require a greater level of both the signal and cancellation tone so a sufficiently high-amplitude resultant can be heard.

The BC-BC method was then used in a third experiment to achieve unilateral crosstalk cancellation over a wide frequency range using a high-resolution filter (Mcleod and Culling, 2020). Large variation between participant's phase accumulation were seen over the frequency range, although there were some similarities in level, especially when cancelling crosstalk on the left side. Sudden drops in level at sporadic frequencies were noted across all participants, likely due to the antiresonances mentioned earlier. These level drops were often associated with a 180° phase shift caused by the destructive interference.

The effectiveness of the crosstalk cancellation was evaluated by measuring changes in tone-reception thresholds (TRT) with and without unilateral crosstalk cancellation applied. A tone was presented to one side and a noise to the other, and the crosstalk on the tone side was cancelled using the crosstalk-cancellation filter. Improvements were seen at all frequencies between 1 kHz and 5 kHz, and similar improvements were seen in all participants, achieving between 11.2 dB and 13 dB benefit averaged across all frequencies, although the least improvement (9.2 dB) was seen at 1200 Hz. This matches the perceived poorer cancellation reported in Mcleod and Culling (2017). Similar results were seen in a speech reception threshold test, producing a mean threshold improvement of 13.67 dB across all 3 participants.

Surendran and Stenfelt (2023) conducted a similar study using adaptive AC masking of the contralateral ear so that only the signal at the cancellation ear could be heard. This enables cancellation at lower frequencies as, since the non-cancellation ear is masked it becomes a cancellation task rather than a lateralisation task meaning that cancelling both ears at the same time (due to little phase difference between the cochleae) is no longer an issue. Cancellation levels averaged across participants varied between 3 dB and 10 dB over the frequency range (0.25 kHz to 4 kHz), although individual variation was high (up to 75 dB at a single frequency). Level differences were similar between stationary and transient stimuli. Phase cancellation was also similar between stationary and transient stimuli however there were some frequencies where large differences were observed.

The study presented in this chapter follows on from Mcleod and Culling's previous research (2017, 2019, 2020), using BC-BC cancellation. Participants initially completed a screening task where they cancelled simulated crosstalk over headphones, and if successful could continue to the BT study, where they attempted BC-BC cancellation. BC-BC cancellation involved adjusting the phase and level of a tone played to the ipsilateral BT to cancel the cross-talk from a contralateral BT. Doing this reveals the phase and amplitude shift required for a tone to cancel the crosstalk at a given frequency, allowing for the creation of a crosstalk-cancellation filter. After the task was completed at both cochleae, the transfer functions were used to create a digital 256-tap filter which would be used to cancel the cross-talk from both BTs simultaneously in a TRT task comparing thresholds with and without the crosstalk-cancellation filter applied. Transfer functions were recalculated every testing session since taking the headset off would change the position of the BTs and change the transfer function, reported in Mcleod and Culling (2019,2020).

3.3 Experiment 1 - Crosstalk-Cancellation Simulation over Headphones

3.3.1 Method

3.3.1.1 Participants

A total of 18 participants were recruited from Cardiff University, aged 18 – 43 (4 male, 14 female). All participants had self-reported normal hearing.

3.3.1.2 Apparatus

All materials were created using MATLAB®. All sounds were played over Audio-Technica ATH-M50X studio monitor professional headphones and a Logitech G502 Hero Wired mouse was used for the experiment since it includes an "unlockable scrollwheel", allowing participants to rapidly change the phase/level of the cancellation tone. A stereo splitter was used to allow the experimenter and participant to hear each other manipulating the phase and level of the cancellation tone for training purposes.

3.3.1.3 Stimuli

Sinusoidal tones of frequencies between 1 and 7 kHz in 1Khz intervals were used for testing. Two identical tones were played to each ear and were convolved with either simulated crosstalk or a cancellation tone since sound presented over headphones produce minimal crosstalk (Zwislocki, 1953). The simulated crosstalk was a sinusoidal tone whose phase was shifted by a random number between 1° and 360° and a random level shift between 0 dB and 36 dB and the cancellation tone was another sinusoidal wave where the participants controlled the level and phase shift via a mouse scroll wheel. The cancellation levels were chosen based on Mcleod and Culling's previous experiments and pilot data which suggested that most attenuation occurred between these levels.

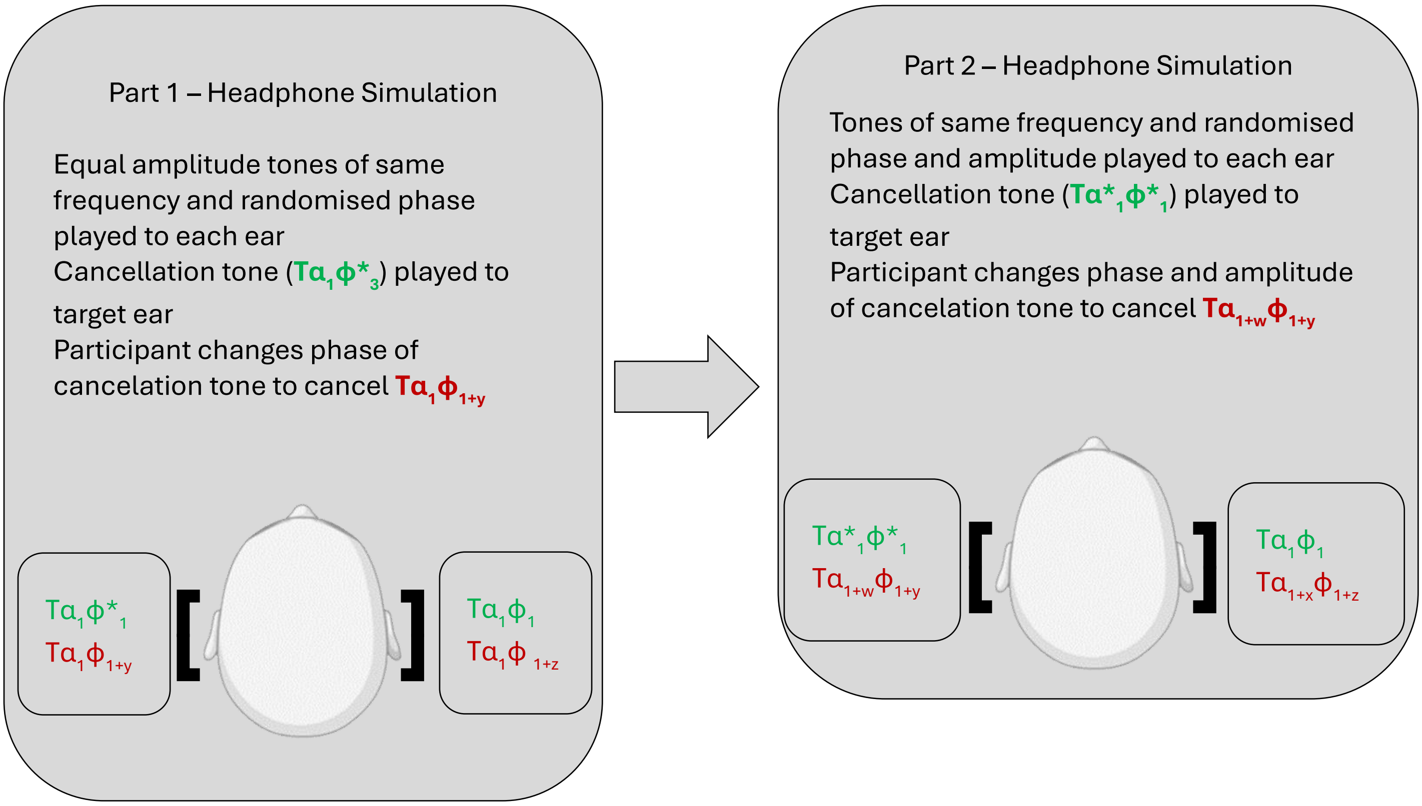


Figure 3.1. A diagram of the headphone-cancellation procedure. The first box shows the phase-only cancellation procedure and the second box shows the phase and level cancellation procedure. α represents the amplitude of the tone and ϕ represents the phase. Asterisks denote the changing phase and/or amplitude of a cancellation tone (green) or cancellation tone crosstalk (red). Green text indicates ipsilateral signals and red text indicates crosstalk. 'W' and 'x' refer to a randomly added amplitude and 'y' and 'z' refer to a randomly added phase shift.

3.3.1.4 Procedure

After wearing the headphones provided by the experimenter, an identical tone at a given frequency (randomised 1 kHz intervals between 1 and 7 kHz) is played to both ears ($T\alpha_1\phi_1$ and $T\alpha_1\phi^*_1$ in figure 3.1). The same tone with random phase shift (between 0° and 360°) was added to the left ear ($T\alpha_1\phi^*_{1+y}$), and another to the right ear ($T\alpha_1\phi^*_{1+z}$). These additional tones simulate the crosstalk that would occur during bilateral BCHD use. Participants were allowed to adjust the phase of the cancellation tone ($T\alpha_1\phi^*_1$) to cancel the simulated crosstalk ($T\alpha_1\phi_{1+y}$). This was done at both ears.

Prior to testing, the experimenter demonstrated how to cancel sound by completing a cancellation over headphones while utilizing a splitter to duplicate the sound to the participant's headphones. This was done to aid the participant's navigation of the searchspace (figure 3.2). After the experimenter demonstration, participants were asked to cancel the simulated crosstalk at the target ear by adjusting the phase of the cancellation tone (in 1° increments) using the scroll wheel of the mouse until the tone could only be heard at the non-test ear. Once the participant perceived the sound to be cancelled, they could save their answer and were given the error (difference between their locked-in phase and the actual phase shift required to cancel the crosstalk) and could move to the next frequency and/or ear (frequency and cancellation side were randomised). The experimenter could listen to the participant cancelling the sound using the splitter to give advice to the participants.

Once the participant had demonstrated proficiency at cancelling the phase across all frequencies, they could move onto the next phase of the experiment. The second phase of the experiment was identical to the first but an additional level shift (between 0 dB and 36 dB) was added to the simulated crosstalk tones($T\alpha_{1+w}\phi_{1+y}$ and $T\alpha_{1+x}\phi_{1+z}$). This means that participants needed to alternate between adjusting the level (in 0.1 dB intervals) and phase until they perceived the sound to be lateralised away from the test ear. After a cancellation attempt participants were given the option to attempt a second cancellation at the same frequency with the same crosstalk but with the ability to switch off the non-cancellation ear. This was done to allow participants to hear when they had perfectly cancelled sound and hear the difference their error made. Participants were given the option to repeat frequencies which they found more difficult to cancel.

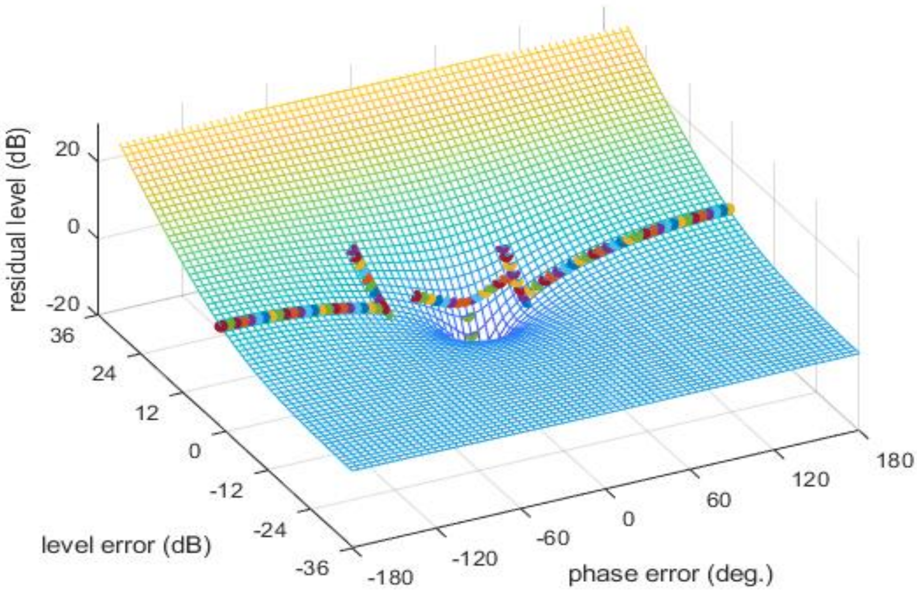


Figure 3.2. An example of a participant navigating the search space in the phase/level crosstalk cancellation experiment over headphones. The dip in the middle shows the point at which the sound is completely cancelled.

3.3.2 Results

Figure 3.3 shows the RMSE of the difference between the cancellation phase and the response phase for crosstalk presented to the left(a) and right(b) ears over headphones. RMSE was chosen to test for the accuracy of participant's localisation. Each block contains 7 cancellation attempts and the RMSE of each block is plotted and averaged across participants. Most participants can accurately cancel on the first attempt and most participants who initially could not, quickly improved over 2-4 more blocks. No effect of side was seen and performance at one ear did not always predict performance of the other. There were 3 participants who did not show improvement in left ear cancellation (pink, green and brown on figure 3.3) and two who did not show improvement in right ear cancellation, although they may have improved with additional practice. These participants would not have been invited to move on to the next part of the experiment. Two participants show sudden increases of error in block 7, both blocks were the first blocks of the next session, suggesting that in some participants time between training affects cancellation ability.

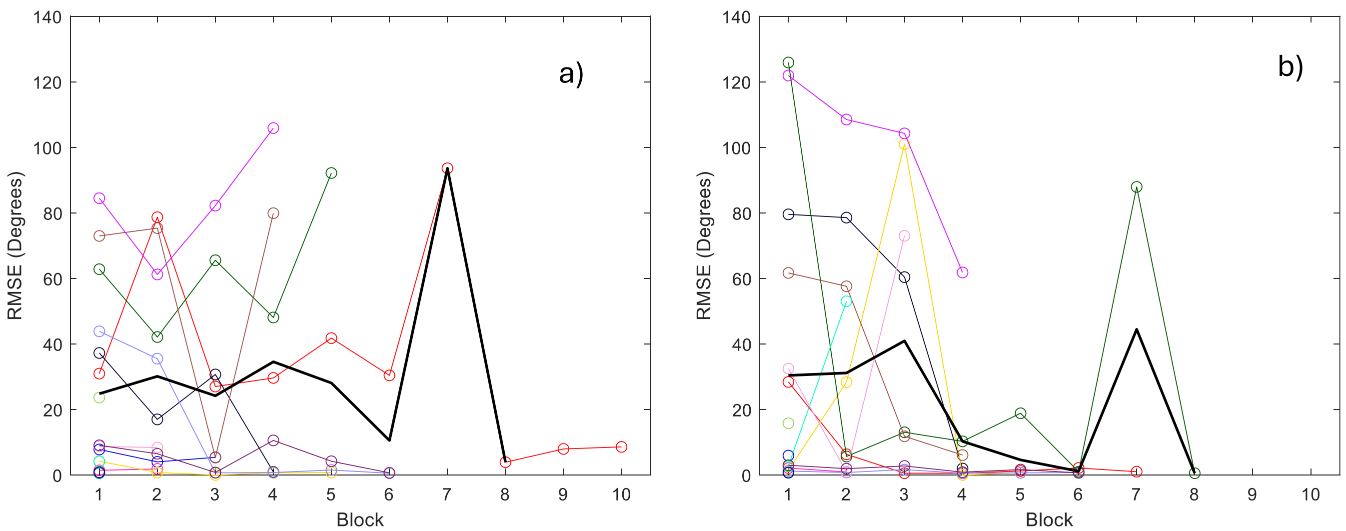


Figure 3.3. Phase RMSE averaged over each block (7 trials) for the left (a) and right (b) ears for the phase only headphone-simulation condition. Each block tested 7 frequencies from 1 kHz to 7 kHz in 1 kHz intervals, although on later blocks only the frequencies participants were unable to cancel were tested. The solid black line shows the mean RMSE for each participant over blocks. Participants are represented by the same colour in both plots.

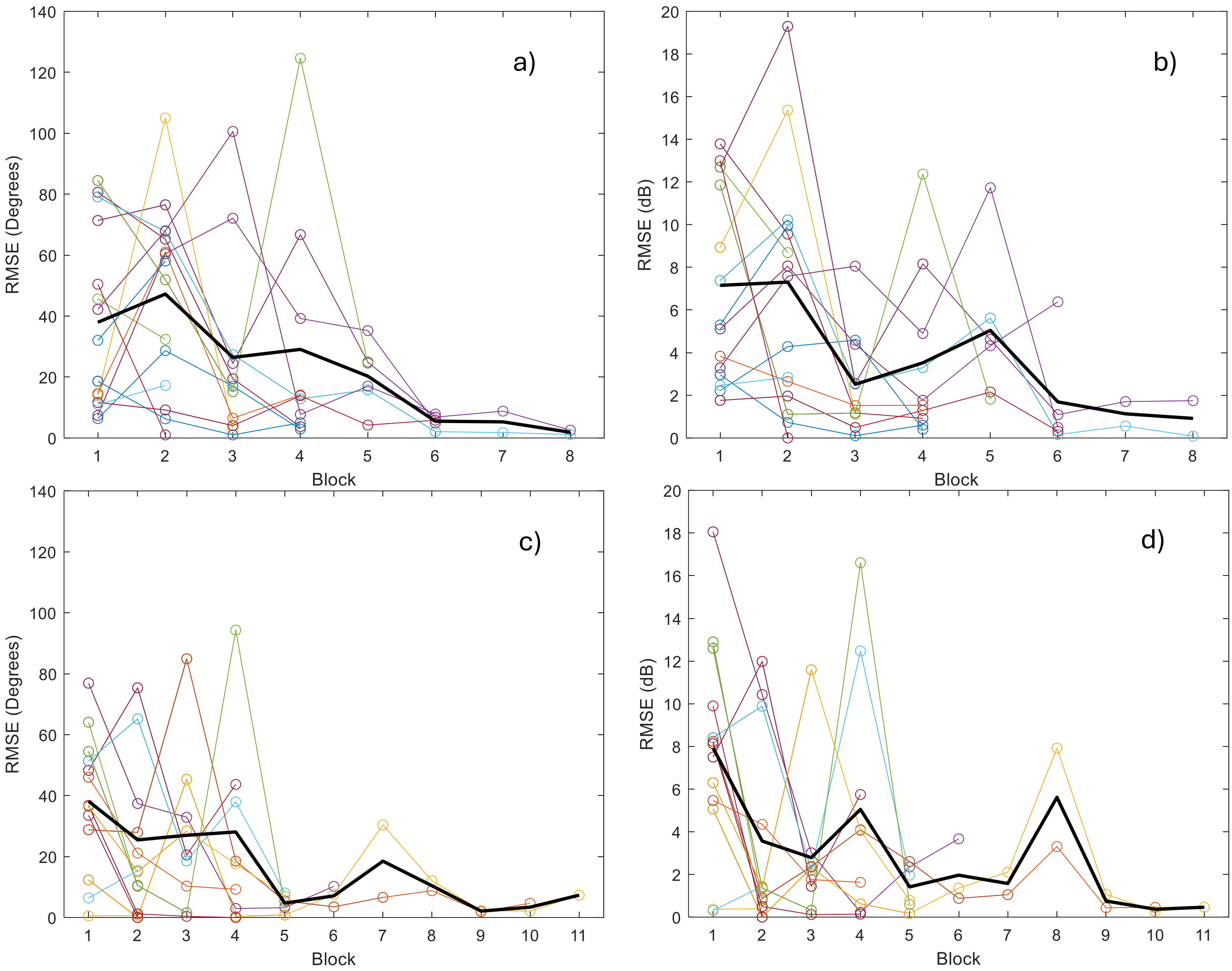


Figure 3.4. RMSE phase error for cancellation on the left (a) and right (c) ears, and RMSE amp error for left (b) and right (d) cancellation in the combined level and phase cancellation task over headphones. The black line represents the mean RMSE over all participants for each block.

Participants initially had poorer cancellation when the simulated crosstalk involved phase and level attenuation. Again, most participants were able to cancel the crosstalk after practice (Figure 3.4), albeit sometimes inconsistently and perfect cancellation was usually not seen. Some participants were unable to achieve a consistent level of cancellation. Repeated measures ANOVAs revealed no significant effect of side was seen for phase ($F(1,35) = 77.79$, $p = .82$) or level ($F(1,35) = 98.23$, $p = .21$). Figure 3.5 shows the individual cancellation levels and phases across each frequency. There was no effect of frequency and participants were able to cancel simulated crosstalk at all frequencies.

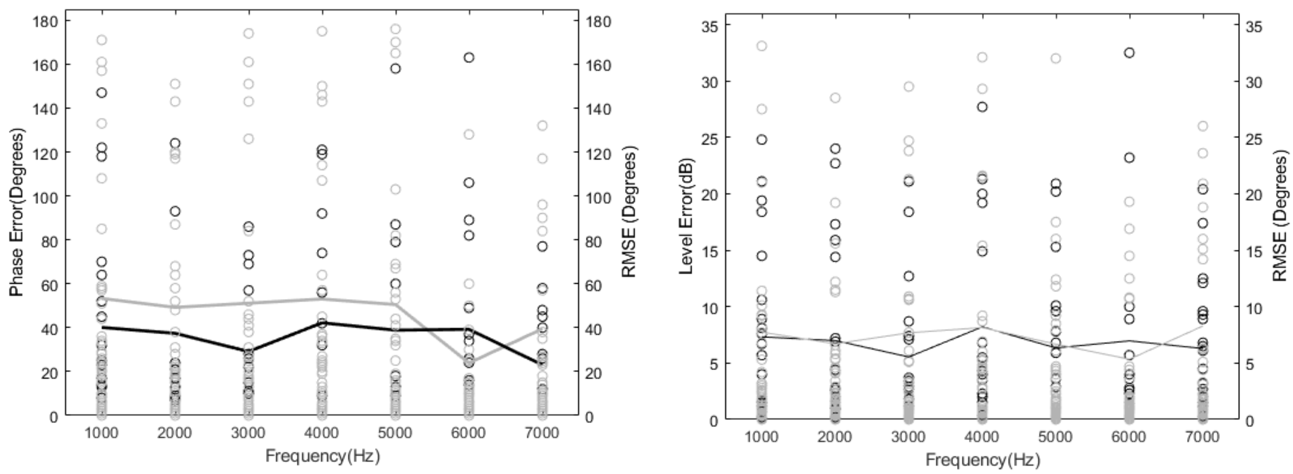


Figure 3.5. Phase (left) and level (right) errors across all frequencies in the phase and level combined cancellation task. Light grey data points show cancellation errors from the left ear and dark grey points show errors from the right ear. The trend lines show the RMSE averaged across each frequency.

3.3.3 Discussion

Participants were able to easily cancel simulated crosstalk when it was presented over headphones and only the phase needed to be changed, either being able to do the task immediately or requiring little training. One participant was unable to accurately cancel sound, but may have been able to with more practice. There were also individual blocks where the participant was unable to cancel on one side for that block, while having accurate cancellation on previous or subsequent blocks, or on the other ear. These instances of sudden poor cancellation may be caused by the starting of the next test session which could require participants to refamiliarize themselves with the sensation and method of sound cancellation.

Adding a second parameter (level) makes navigating the search space more difficult and time consuming since, if one parameter is far away from the cancellation value, changing the other will only result in small perceptual changes. For some participants, this problem resulted in little or no improvement across practice sessions, but most were eventually able to cancel the simulated crosstalk. It is unclear whether more practice would have improved the performance of those who struggled to cancel or if they were unable to do the task. It cannot be an inability to perceive when a sound had become lateralised since they completed the phase-only part of the experiment.

3.4 Experiment 2 – Crosstalk Cancellation in Bilateral Bone Transducers

3.4.1 Methods

3.4.1.1 Participants

Ten paid participants were recruited from Cardiff University although only five (ages 22-43; 3 males, 2 females) finished the study. All but one dropout was either due to the participant not wishing to continue taking part (due to the time investment) or due to poor performance in the task (2 participants). One participant was not invited to take part in the BT experiment due to excessive ear wax. All participants reported normal hearing and had their ear canals examined with an otoscope to check for excessive earwax and tympanic abnormalities. Participants were required to complete the previously mentioned screening task involving cancelling simulated crosstalk over headphones. Participants who were unable to successfully do this were not invited to take part in this experiment.

3.4.1.2 Apparatus

MATLAB® computer software was used to create the apps used in the experiment as well as the pure tones and noise. The pure tones and noise were created with a sampling rate of 16000 kHz since this is the sampling rate of BCHDs. The sounds were driven through an eight channel USB ESI MAYA44 USB+ DAC, into two Radioear™ B71W BTs, which were not verified to be equal. These were held in place by a pair of 3D printed glasses (Figure 3.6), designed in Fusion 360. These glasses were based off those used in Mcleod and Culling (2017) which were created to provide a standardized way to hold the BTs against a participant's skull. Different sizes of glasses were printed to fit different participant's head sizes, so that the B71Ws would be held just behind the participants ear on the mastoid bone. An elasticated headband was used to hold the headset in place and ensure the B71Ws were pressed against the skull, different size head bands were used for different head sizes. A 2-3N pressure was applied to the BTs against the participants head since this is sufficient for good coupling (Toll et al. 2011). Testing was performed in a single-walled sound attenuating booth (Industrial Acoustics Company).

E-A-R-Link 3A foam eartips were used to stop any sound entering through the ear canal and the open tubes were sealed with bluetac. A Logitech G502 wired mouse was used.

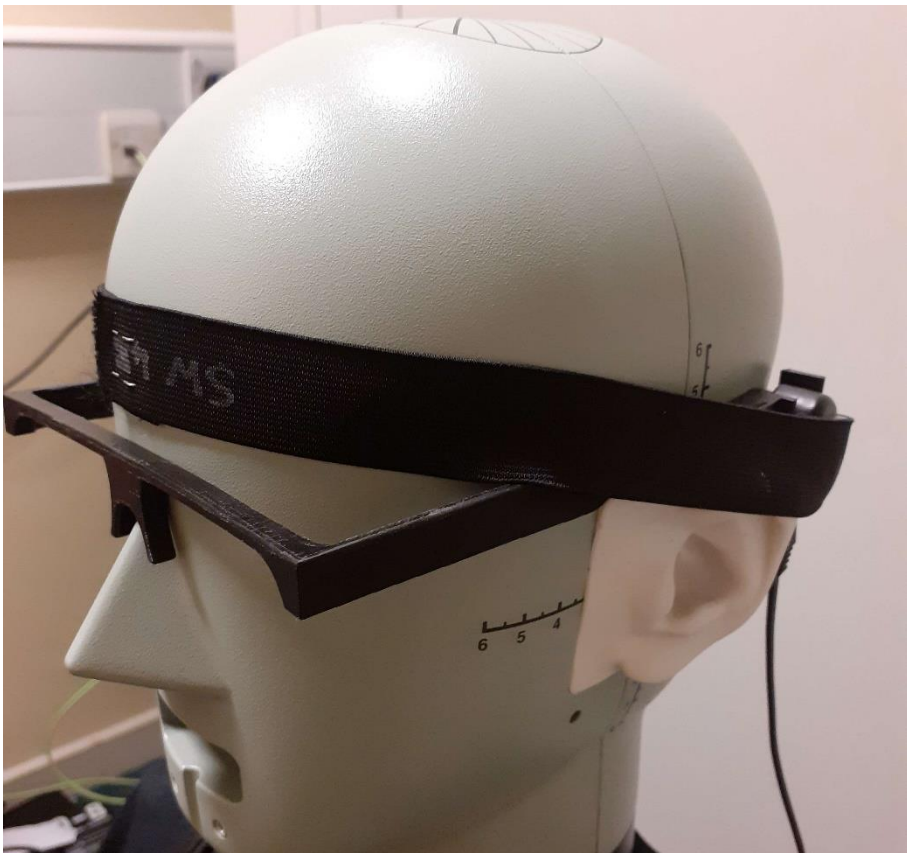


Figure 3.6. The headsets participants wore during the experiment. Multiple sets of both the glasses and the elastic headbands were created to fit different size heads. The glasses were created using Fusion 360 and 3D printed.

3.4.1.3 Procedure

Once participants had demonstrated proficiency at the previous experiement they moved on to this experiment. Since the attenuation of the crosstalk sound is created by the pathways the sound takes around the skull, the transfer function of an individual's skull is unknown, meaning that no feedback could be given. Participants were instructed to wear the BT glasses (Figure 3.6), elastic band and deep insert ear plugs before starting the experiment. Insert length was not measured and the participants were responsible for inserting the ear plugs. Once the experiment began participants were not allowed breaks and were instructed not to move the BT glasses since changing the BT position alters the skulls attenuation and would invalidate all previously cancelled tones in that session (Mcleod and Culling, 2019, 2020). No instruction was given regarding jaw positioning and there was no head-securing device to reduce the discomfort of the participant.

The cancellation procedure was similar to the previous experiment from the participants perspective. Participants were presented with an identical tone at each BT, which would have produced crosstalk on both ears as it travelled across the skull. Participants used the scroll wheel to adjust the phase and amplitude of a tone at the ipsilateral BT until it cancelled the crosstalk from the contralateral BT and no sound was heard at that ear. Participants were tested across 1 kHz bands (between 1.5 kHz to 7.5 kHz) in 62.5 Hz intervals. This was done because cancelling a wider frequency range would be too time consuming and after an hour of using the BTs participants reported discomfort. Participants could freely move between frequencies and cancellation sides to allow them to listen to previously cancelled frequencies and correct them if necessary. Participants chose their starting frequency based on the frequency they reported easiest during the simulated experiment, and proficiency was gained in one frequency before moving on to the next.

When cancelling tones from a previously cancelled frequency, the data from the previous session was pre-loaded so that participants could start with the same cancellation phase and level they previously used. If they increased or decreased the phase or level, the relative changes were applied to every other unadjusted (during the current session) frequency in the frequency range. For example if a participant lowered the cancellation level by 5 dB, 5dB would be subtracted from every cancellation level in the pre-loaded data. If it was the participant’s first time cancelling tones at that frequency range, the first level/phase shift would start at 0, and once they had locked in a phase and level shift the next test frequency would be set to the same cancellation phase and level. This was done because adjacent frequencies would usually only slightly differ in the phase and level shift required to cancel.

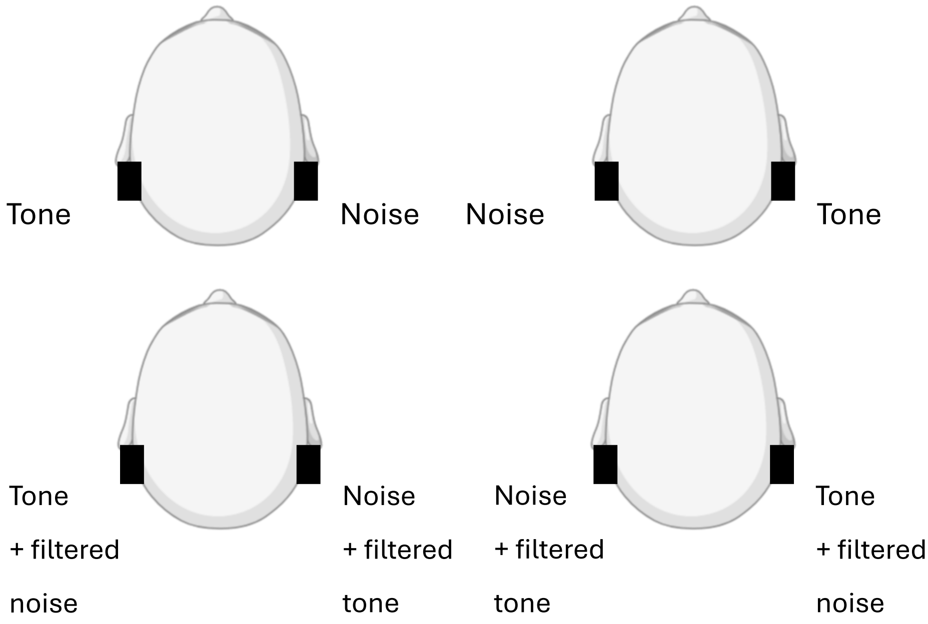


Figure 3.7. The Four different testing conditions presented during the masked tone-reception threshold test. These were presented in a random order and were interleaved. The first two are the control conditions where a tone is presented on one side and noise on the other. The bottom two are the test conditions where the crosstalk-cancellation are signals applied to both sides.

Once the participants were confident that they had cancelled both sides to the best of their ability they moved on to a masked tone-reception threshold test. The cancellation levels and phases gathered from calibration phase of the experiment were then used to create an FIR filter using the host-window method (Abed & Cain, 1984). The test utilised a 2-interval, forced-choice task in four interleaved, 2-down/1-up adaptive staircases (Levitt, 1971) with 12 reversals. The first 2 reversals had a step size of 4 dB and the remaining 10 had a step size of 2 dB. The final 8 reversal levels were averaged to get the threshold measurement. Each trial involved two 0.3 s intervals with a 0.3 s inter stimulus interval. A 0.3 s target tone was played during one of the intervals, which participants needed to indicate using a button press. The interleaving of the staircases was randomised and participants were given feedback. During the threshold measurement a tone was played to one BT while noise was played to the contralateral BT which, due to the crosstalk, would mask the tone. There were four conditions (as shown in figure 3.7) which were randomised, two control conditions where a tone was presented at one BT and noise at the other, and the experimental conditions where a crosstalk-cancellation filter was also applied.

Each frequency range was cancelled and then tested until successful cancellation (> 6 dB TRT improvement with crosstalk cancellation) had occurred on two consecutive sessions or until the participant no longer seemed to improve. Each session lasted about an hour although participants cancelled some frequencies faster than others.

3.4.2 Results

Figure 3.8 shows the phase adjustments required to cancel a contralaterally-played BC tone at the ipsilateral ear. Between adjacent frequencies phase tends to decrease slightly as frequency increases with occasional spikes at certain frequencies (such as that seen at around 2100 Hz in participants 2’s left cancellation), and some sustained phase shifts that last over a wider frequency range (e.g. around 4500-4800 Hz in participant 3). The changes in cancellation phases at adjacent frequencies differed on the participant. Participant 2 had a phase range of about 350° over the entire frequency range whereas participant 1 had a phase range of about 800°. Mcleod and Culling (2017;2020) also found variation in phase ranges between participants, but found overall higher ranges than shown in this study. Cancellation phases between sessions tend to either be similar (as seen in participant 1) or follow a similar trend but all values are shifted by a given phase (e.g. participant 4 between 5500 and 7500 Hz).

The level adjustments (figure 3.9) are less consistent between sessions (different coloured points). There are periods of similarity between sessions such as the cancellation levels seen between 5500 and 7500 Hz in participant 2’s right cancellation. There were also some frequency ranges where the levels were transposed, as seen in the cancellation phases, but there was often no relationship between the cancellation levels at a given frequency during different sessions. In general, the cancellation levels ranged from +5 dB to -35 dB in no particular pattern and while there are some similarities between participants, such as low frequencies tending to require cancellation levels closer to zero, one participant’s level data cannot be used to predict another’s. Throughout a given participant’s level data, sudden level changes occur at certain frequencies which are sometimes accompanied by sudden phase changes (similar to those seen in Mcleod and Culling, 2017, 2020), although frequently there is no abnormal change in phase at that frequency. This phenomenon is not seen at the same frequencies across the two sessions suggesting it may depend on BT placement.

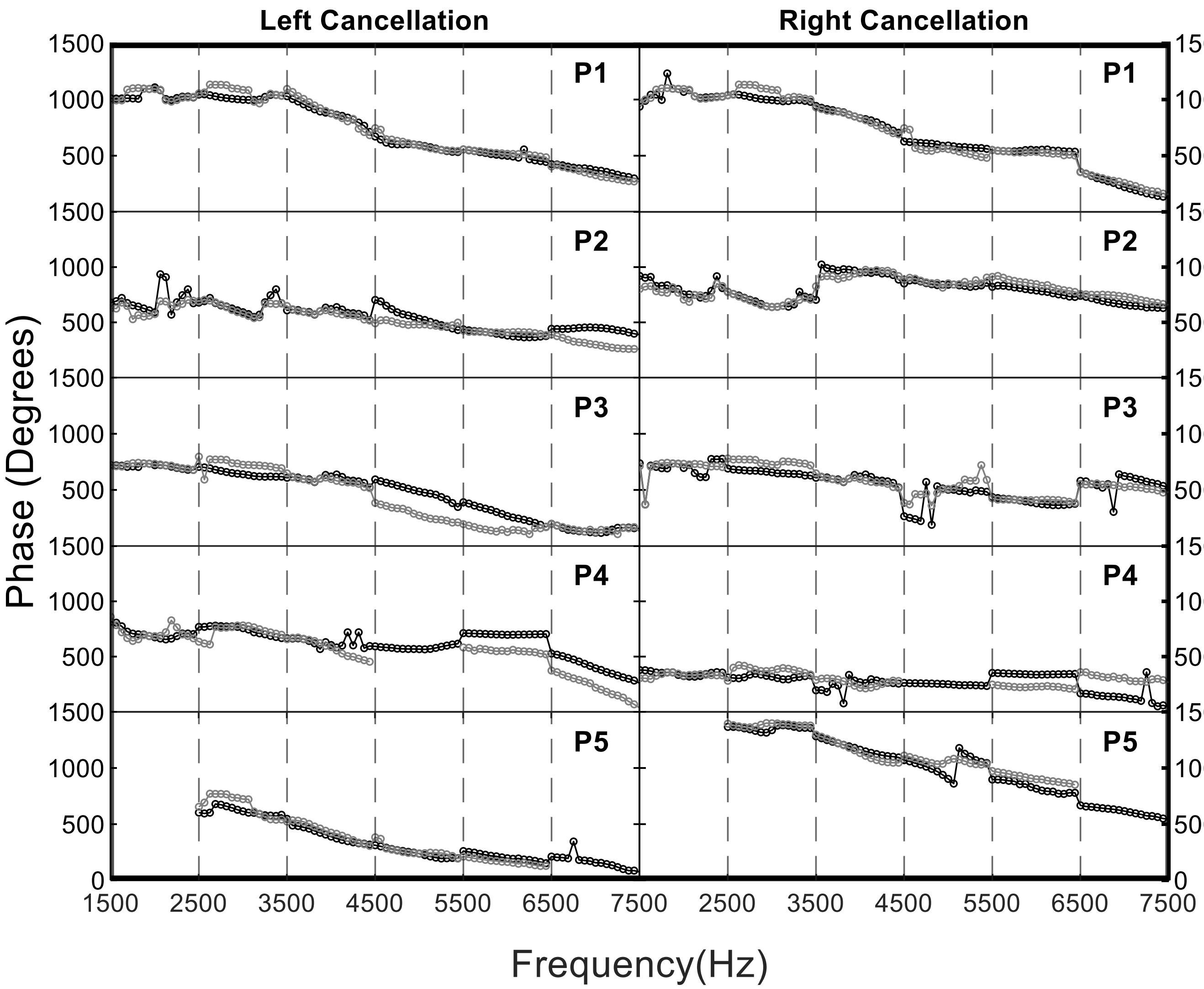


Figure 3.8. The phase adjustments applied to a tone at the ipsilateral BT to cancel the crosstalk from a tone played to a contralateral BT for each participant. Black and grey symbols show the penultimate and final sessions each participant completed for every frequency. Participant 5 did not complete cancellation over the 1.5 kHz – 2.5 kHz range, and only completed one session for the 6.5 kHz to 7.5 kHz range. Participant 1 is the first two plots and the plots descend in order to participant 5.

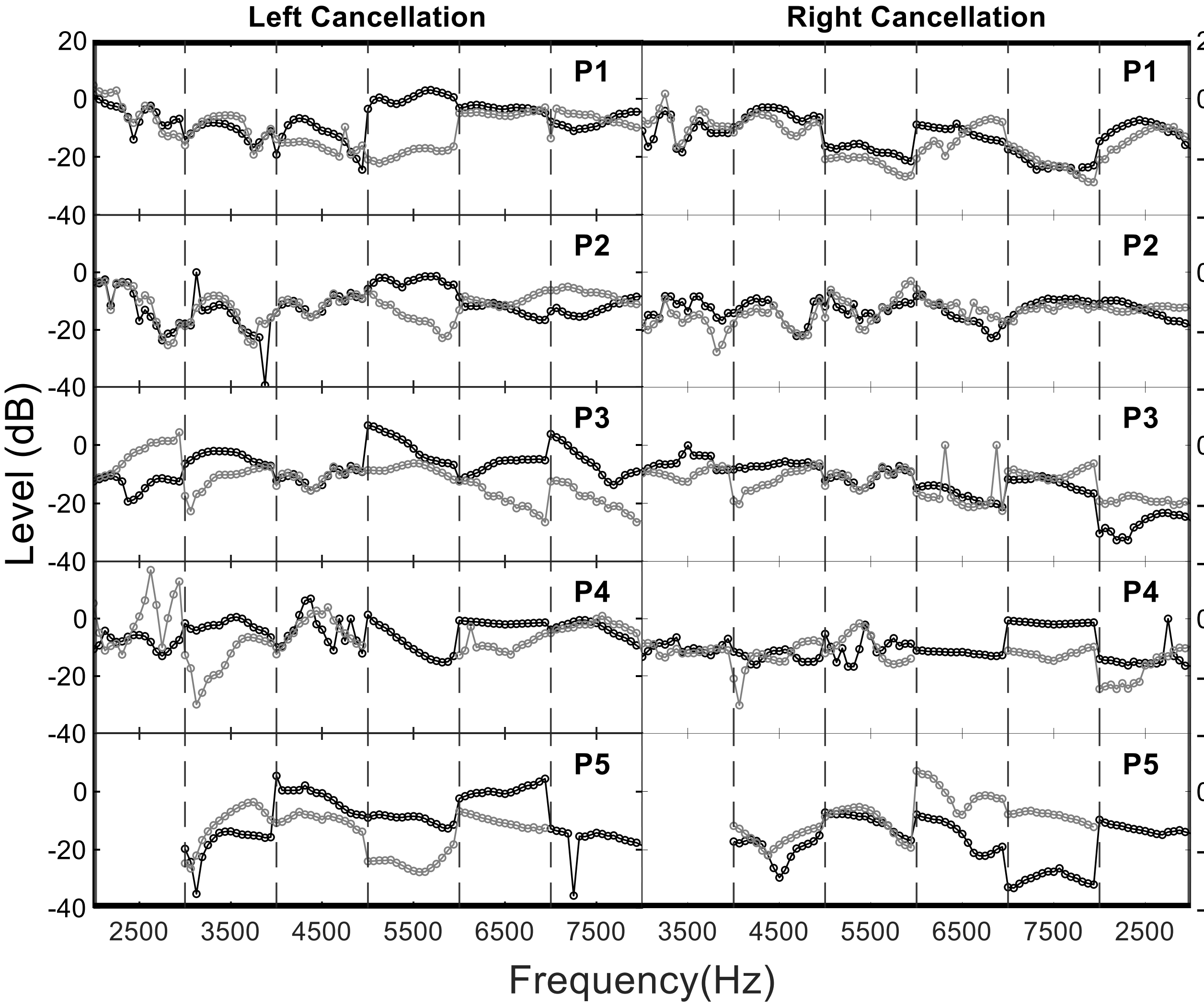


Figure 3.9. The level adjustments applied to a tone at the ipsilateral BT to cancel the crosstalk from a tone played to a contralateral BT. Black and gr show the penultimate and final sessions each participant complete for every frequency. Participant 5 did not complete cancellation over the 1.5 kHz – 2.5 kHz range, and only completed one session for the 6.5 kHz to 7.5 kHz range. Participant 1 is the first two plots and the plots descend in order to participant 5.

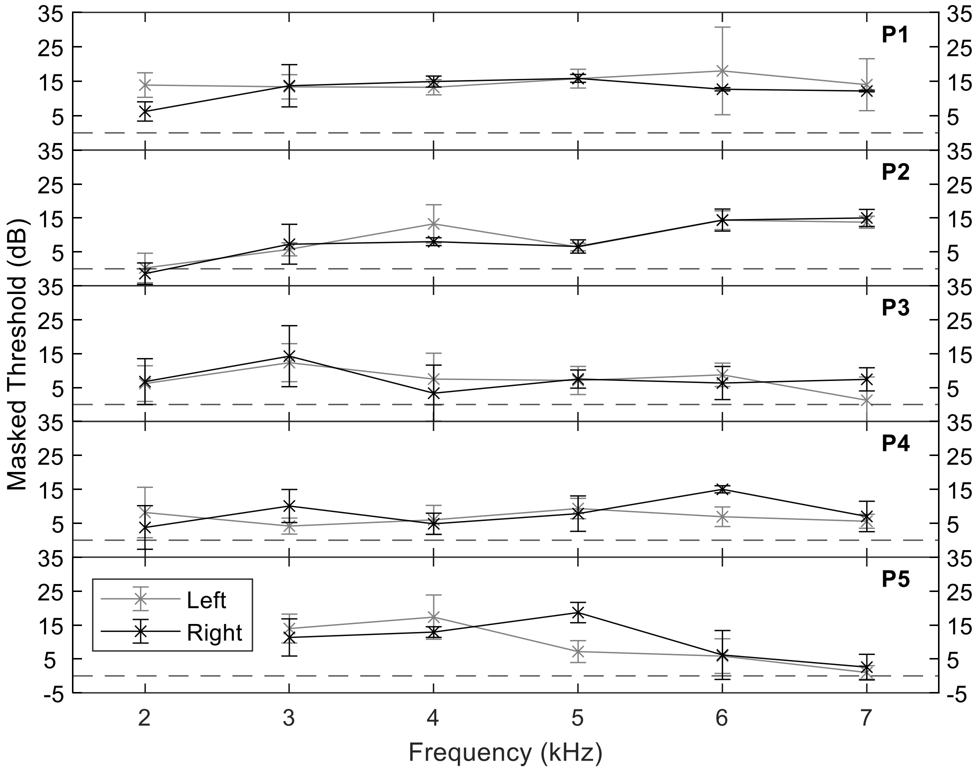


Figure 3.10. Differences in masked tone-reception threshold when the crosstalk-cancellation filter is used compared to no filter being used on both the left and right sides. The black line represents thresholds when a tone is played to the left and the grey line represents right side improvement, and the dashed line shows 0 dB (which indicates no improvement when the filter is used). Each point is averaged across four tests involving 12 reversals each, the last 8 reversals were averaged to get the thresholds.

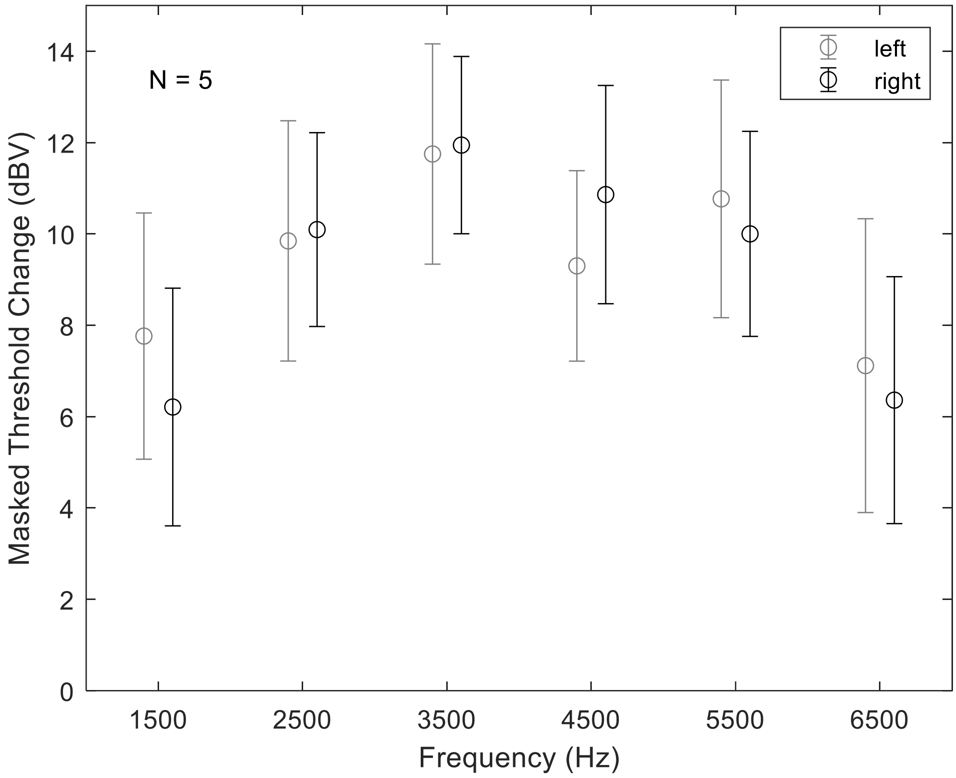


Figure 3.11. Mean masked tone-reception threshold differences between when a cross-talk cancellation filter is used compared to no filter over bilateral BTs. The grey line shows the threshold changes for the left ear and the black line shows the changes for the right ear. Each point is averaged across all 5 participants other than 1500 Hz, which excludes participant 5 since they did not complete this frequency. Error bars show the standard error.

Once the crosstalk-cancellation filtering had been implemented, improvements in masked thresholds were seen across all participants (figure 3.10). A repeated-measures ANOVA revealed a significant effect of filter ($F(1,3) = 35.39$, $p < .05$) and an interaction between filter and side ($F(1,3) = 27.53$, $p < .05$). No effect of side or frequency were observed. Masked thresholds in the left ear improved by 9.31 dB and in the right ear by 9.23 dB (figure 3.11). Participant 1 achieved good cancellation on the left side but otherwise cancellation between 1.5-2.5 kHz was poorer across all other participants. Cancellation was also worse at 6.5-7.5 kHz, with only participants 1 and 2 achieving good cancellation on both sides.

3.5 Discussion

As expected from Mcleod and Culling (2020), the phases and levels required to create the crosstalk-cancellation filter varied between participants. This variation means it is not possible to use one participant's filters to predict another's, so each participant must calibrate their own filters. The difference between a participant's cancellation values from two different sessions are likely the consequence of small positional changes in BT placement. While the headset used in this study was designed to help place the BTs consistently, there will still be small variation in positioning, which may be sufficient to cause changes in the transcranial attenuation of each BT. Inter-session variation between participant's cancellation values also varied. Participants 1 and 5 had similar cancellation phases across both sessions whereas participants 3 and 4 showed more inter-session variance. Differences in variation between the left and right ear are also not always the same. For example, between 5000 Hz and 5500 Hz, participants 1's left-ear cancellation values are nearly identical whereas for the same frequency range on the right side a clear difference can be seen between the two sessions. These differences may be because repositioning of the BT affects different people to different extents. In this experiment it may have been influenced by how well the glasses fit each participant.

Additionally inter-session variation is not consistent across a single frequency range. For example between 3500 Hz and 4500 Hz in participant 2's right cancellation, there was substantial variation between the two sessions up to 4000 Hz, but after 4000 Hz there is little difference between the sessions. While participants were instructed to not move the BTs, it is possible that small positional changes could have occurred as a result of the participant's movements, for example yawning may cause a small change in positioning. It is also likely that over a session the BTs would compress the adjacent skin more and could change the skin attenuation. Participants were given the option to calibrate the filter in any ear or frequency order they wanted so it's possible that participants completed a given frequency at two different points during the experiment and so could have 2 different states of skin compression (e.g. one session they might have started at the highest frequency in the test range, and the next session they started with the lowest). Patterns are difficult to draw from the cancellation levels since there is greater variation in level compared to phase. Cancellation level may be more sensitive to BT repositioning or other factors that affect crosstalk cancellation.

Ability to create a crosstalk-cancellation filter differed between participants. Some participants (participants 1 and 5) were able to create effective filters with relatively little practice, whereas other participants did not show any improvement after hours of practice. Because this experiment was designed to be repeated on patients, no techniques which patients with BCHDs could not use, such as AC masking of the non-cancellation ear (Surendran & Stenfelt, 2023) or the use of OAEs (Usugawa, 2017; Wang et al., 2023) were used to aid participant's cancellation. Two participants showed no improvement in cancellation ability after extensive practice (6 and 8 hours). All other participants who attempted the BT cancellation experiment (after successfully completing the headphones simulation) showed successful crosstalk cancellation, even if they withdrew from the experiment. Physiologically it is unlikely that the crosstalk for the participant's skull was unable to be cancelled, especially considering that studies using masking and OAEs were able to consistently produce cancellation but is more likely to be due to difficulties in navigating the 2d search space or unfamiliarity with the sensation of BC sound. There is also a chance head position or jaw position (e.g. teeth clenching) could have changed over the course of the experiment which may have changed the cancellation values over time.

Participants usually completed both sides of the cancellation in around 40 minutes, leaving 20 minutes for threshold testing. In total participants cancelled 16 frequencies in each ear or 32 in total, allowing for just over one minute per frequency. It is likely that for many participants this was not sufficient for accurate cancellation since finetuning the level and phases can be time consuming (45 minutes per frequency in Surendran & Stenfelt, 2023), but it should allow participants to achieve somewhat accurate cancellation depending on their ability. Session length was limited to an hour as this was the point at which the pressure of the BTs being pressed against the participants head caused discomfort. Since repositioning of the BT alters the required cancellation phase and levels, a break cannot be taken without invalidating the previously cancelled frequencies. Prediction techniques were employed but the same degree of success seen in Mcleod and Culling (2019) was not seen in this study, although a simple prediction based on the participants previous cancellation attempt was utilized.

Future research should focus on repeating this experiment in a clinical population. The experiment was designed to be repeatable in patients with bilateral BCHDs, so the same procedure can be used. Testing patients serves two purposes. First to determine whether the creation of cross-talk cancellation filters is possible in a patient population. The second more pertinent finding would be whether the filters remain robust over time. When using a BT headset (such as the one used in this study), the BTs are placed in a slightly different position at the start of each session, which alters the cancellation levels and phases required to cancel at a given frequency. Since percutaneous BCHDs are implanted into the bone they will not move between sessions meaning that it should be possible to use the same filters over long periods of time. Additionally, the problem of inconsistent cancellation (i.e. variation in participant's cancellation ability between sessions or between adjacent frequencies) would not be as detrimental since only one successful cancellation is required to create an effective filter, which can be fine-tuned over time. There should also be no discomfort when completing the task since the discomfort was caused by the pressure of the elastic band holding the BTs against the skull. If the same filters cannot be used between sessions then the continual recalibration of the filters may make the implementation of them too impractical for everyday use.

3.6 Conclusion

Previous research (Mcleod & Culling, 2020; Surendran & Stenfelt, 2023) has demonstrated the ability to use the vibration from one BT to cancel the crosstalk from a contralateral BT. This study took that knowledge and applied it so that crosstalk from both BTs was cancelled simultaneously and using a filter-creation method that could be easily applied to patients using bilateral BCHDs. Participants required different amounts of practice to become proficient at creating the cancellation filters, but most participants were able to produce effective crosstalk-cancellation filters. These filters were effective at improving masked thresholds between 3 kHz and 7 kHz, however limited improvement was seen at 2 kHz. Future research should aim to replicate the findings from this study in a clinical population using percutaneous bilateral BCHDs.

Chapter 4 - Crosstalk Cancellation in Patients with Bilateral BCHDs

4.1 Summary

Previous studies have demonstrated that a BC tone played at an ipsilateral BT can be used to cancel the crosstalk from a contralateral BT (Mcleod and Culling, 2019, 2020; Surendran & Stenfelt, 2023). The experiment in chapter 3 adapted this procedure so that the filters could be easily integrated into a patients BCHDs, and also tested simultaneous, bilateral BC crosstalk cancellation. Improved thresholds were observed when a crosstalk-cancellation filter was used. However, while the filter was effective, it needed to be recalibrated each session as the cancellation phases and amplitudes changed between sessions. It is believed that the reason for this was that the BT was placed in a slightly different position on the head every session, and even if this difference is small, it affected the bone conduction pathways around the skull and so the cancellation phases and amplitudes. The present study takes the methodology from the BC-BC cancellation in chapter 3 and uses it in patients with bilateral percutaneous BCHDs. Two patients were able to effectively calibrate a crosstalk-cancellation filter, and returned for subsequent testing. The crosstalk-cancellation filters steadily became less effective over time. Future research should attempt to replicate this finding in a variety of patients.

4.2 Introduction

Bilateral bone conduction hearing devices (BCHDs) are a solution for treating patients with conductive or mixed hearing loss who cannot benefit from conventional hearing aids, often due to problems with their external or middle ears (e.g. atresia or cholesteatoma). They allow these structures to be bypassed by using the skull bone and tissue as a conduit for the vibration produced by a transducer.

Unilateral implantation of BCHDs can restore binaural hearing in patients with unilateral deafness (Agterberg et al., 2011; Vogt et al., 2018), however unilateral implantation fails to restore binaural hearing in bilaterally deaf patients (Priwin et al., 2004, 2007).

Bilateral implantation is advantageous over unilateral implantation in bilaterally deaf patients, in terms of audiometric thresholds (Bosman, 2001; Dutt et al., 2002a; Priwin et al., 2004, 2007; Canale et al., 2022), localisation ability (Snik et al., 1998; Bosman et al., 2001; Priwin 2004, 2007; den Besten et al., 2020; Caspers et al., 2021), as well as perceived benefit from the patient (Dutt et al., 2002b; Ho et al., 2009, Dun et al., 2010; Canale et al., 2022). The only situation where a unilateral device outperforms bilateral is when a signal is presented to the unilateral device with noise on the other side of the head, but in these situations the device on the noise side can be turned off or removed (Mcleod et al., 2018).

While the use of bilateral BCHDs produces better results than using only one BCHD, it only results in limited restoration of binaural hearing, failing to achieve AC levels of binaural perception. This is because the crosstalk that occurs when using a BC device interferes with binaural cues (Stenfelt, 2012). In order to fully restore binaural hearing, the crosstalk can be cancelled by presenting an inverse wave (inverse of the crosstalk from the contralateral BT at the ipsilateral cochlea) to the ipsilateral cochlea, which will destructively interfere with the crosstalk and leave only the ipsilateral signals (Rowan & Gray, 2008; Liao, 2010).

Attempts have been made to demonstrate the feasibility of this cancellation in normally hearing participants by presenting a tone at a contralateral BT and allowing participants to subjectively cancel the tone by altering the phase and amplitude of the same tone presented at an ipsilateral BT until the tone can no longer be heard on the ipsilateral side. This gives the phase and amplitude shift required at that frequency to produce crosstalk cancellation at the ipsilateral cochlea. Mcleod and Culling (2020) used this method to create a crosstalk-cancellation filter between 1 – 5 kHz, observing an averaged tone-reception threshold improvement of 12.1 dB when the crosstalk-cancellation filter was implemented. This was repeated in speech finding an improvement of 13.67 dB.

Chapter 3 described an adaptation of this procedure where the same methodology was carried out between 1.5 – 7.5 kHz, with cancellation calculated every 62.5 Hz using digital 256-tap FIR filters, since a crosstalk-cancellation filter with these parameters could be easily implemented a patient’s devices. The crosstalk-cancellation filters were also applied simultaneously at both ears since cancellation at both ears would be necessary to restore binaural hearing in patients. An average tone-reception threshold improvement of 9.27 dB was seen across 5 participants, although participants found cancelling below 2 kHz difficult.

The main limitation of this method of testing is that the attenuation of the signal across the skull changes between sessions. Mcleod and Culling (2019, 2020) presume that this is due to the repositioning of the BTs between sessions, and that even slight repositioning can cause the transfer functions of the BT sound across the skull to change, which can cause the previously designed filters to become ineffective. In theory this would not be a problem in patients with an implanted device since the screw that vibrates is implanted into the skull bone and so will not move between sessions. If this is not the case, the time consuming nature of the cancellation will make it too impractical for patients to realistically implement.

The method used in this study is based on the procedure from the previous experiment (chapter 3) and is performed on patients implanted with bilateral BCHDs, which utilise percutaneous abutments. Patients were screened to ensure they actively used bilateral BCHDs and had appropriate BC thresholds (<40 dB cochlear hearing loss at both ears). Patients were asked to cancel a BC tone from the contralateral BCHD at the ipsilateral cochlea by listening to a cancellation tone with sweeping phases and levels, pressing a button when maximal cancellation was perceived. This was done over a 1 kHz frequency band and a crosstalk-cancellation filter was created using this data. Masked thresholds with and without the crosstalk-cancellation filter applied were measured. The test was repeated in future sessions to determine whether the ideal crosstalk-cancellation filter had changed, and to determine whether the inter-session phase and amplitude changes seen in experiment 2 (as well as Mcleod and Culling, 2019, 2020) disappeared when fixed percutaneous abutments determined the location of stimulation.

4.3 Methods

4.3.1 Participants

Patients were recruited from Queen Elizabeth Hospitals Birmingham and tested in the audiology centre at Nuffield House. Inclusion criteria were: above 18 years old and cochlear hearing loss below 40 dB at both ears. Nine patients were recruited (8F, 1M) between 29 and 66 years old. Participants reported frequent bilateral BCHD use. Information about the patients is shown in table 4.1.

4.3.2 Equipment

Patients performed the experiment on a laptop, using a Logitech G504 mouse with an unlockable scroll wheel. Sound was driven through a MAYA USB+ DAC to a Presonus HP4 amplifier which split the sound to two Cochlear BAHA 5 Superpower transducers. Since the Cochlear transducers had no microphone, a Tonor stand microphone was used to communicate with patients during the experimental session. The microphone signal passed through the laptop to the DAC. The microphone was inactive during testing.

4.3.3 Procedure

Prior to starting the experiment, patients were asked to adjust the volume of a pure tone on the amplifier (via the volume dial) until the tone was near maximal comfort level. A voltmeter was then used to measure the output of this tone so that the same level could be used between testing sessions.

Initially patients followed the same procedure as in the previous experiment where the patients could scroll through phase and level differences until they narrowed in on the point of maximal perceived cancellation. However, some patients found this technically demanding and so a more automated version of the task was used.

Patient	Gender	Age	Age BC implantation (L/R)	Aetiology	BC Thresholds (L/R)	Task Performance	Reason for Withdrawal
102	F	57	25/48	Bilateral microtia Conductive HL	50 – 80* /0 - 35	Performed well, extremely consistent phase and levels between 2875 – 3500 Hz after BCHDs were refitted. No Thresholds measured as patient wanted to end session.	Distance from test site
108	F	48	30	Bilateral meatal atresia Conductive HL	0-30 /10-50*	Some consistency seen between BCHD replacements, bilateral threshold improvements between 4-6 dB seen.	Time commitment
115	F	29	15	CHARGE syndrome Conductive HL	15-35 /10-20	Patient struggled to understand the instructions and was unable to complete the task	Unable to complete task due to comprehension issues
127	F	47	38	Bilateral otosclerosis Mixed HL Moderate AC hearing	25-35 /20-36	Initially performed well at the task achieving 8.3 dB cancellation on the left side. This filter deteriorated down the 2.67 dB and no further calibrations were successful.	Completed Study
140	F	34	29	Chronic ear infections Conductive HL	17-35 /17-35	Inconsistent phase and level differences. Threshold improvements (around 5 dB) were seen, but also inconsistent.	Time commitment
141	F	50	45	Conductive HL Mild Tinnitus	5-25* /0-10	Patient was unable to lateralise the sound due to her tinnitus	Unable to complete task due to tinnitus
142	M	66	62	Conductive HL Moderate AC hearing	25-40 /25-35	Patient was unable to interact with the experiment due to limited technological experience.	Unable to complete task due to unfamiliarity with computers/laptops
145	F	61	53/59	Bilateral tympanic perforations Common Variable Immune Deficiency Mixed HL Moderate AC hearing	20-35 /30-50*	After initially struggling and reporting difficulties lateralising because of tinnitus participant was able to achieve good cancellation (6 – 12 dB) on the left side and limited cancellation (2 – 4 dB) on the right side which persisted over multiple sessions. Phases and levels were also consistent across sessions.	Completed Study
146	F	29	14	Conductive HL	0-25 /20-26	Patient struggled to lateralise the sound, and poor threshold results were seen.	Reason not given

Table 4.2. A table displaying information about the participants demographics, BCHDs and a brief description on their performance during the experiment and the reason for their withdrawal. An asterisk in the BC Thresholds Column indicates a masked BC threshold.

In this automated task the phases and levels of the cancellation tone would automatically sweep. The cancellation phase swept through 0° to 360°, and upon reaching 360° would then reset the cycle, returning to 1°. Patients could stop the sweep by pressing a button on screen and it would save the cancellation phase at that moment. The cancellation level then swept from +10 dB (-10 dB on the non-cancellation side) to -40 dB. Upon reaching -40 dB the sweep would reverse direction increasing up to +10 dB before, continually oscillating between +10 dB and -40 dB. Again patients pressed a button where lateralization was perceived to be maximal. Patients could only stop the sound after 1 full sweep (from 0° to 360° or +10 dB to -40 dB) had been completed and the cancellation phases and levels were always shown on screen.

Once the initial cancellation phase and level had been locked in patients were allowed to fine-tune the cancellation phases and levels using the scrolling method used in the previous experiment, and after saving the fine-tuned phase and levels the frequency decreased by 62.5 Hz and the procedure restarted. Once all frequencies had been tested (from 2500 - 3500 Hz) patients were able to do a final fine-tune where they could increase or decrease the frequency and change the cancellation levels and phases using the previously described fine-tuning method.

Once this calibration was completed, a crosstalk cancellation filter was designed and used in a masked TRT task to determine the TRTs with and without the cancellation filter applied. This followed the same method as the filter design and threshold estimation of Chapter 3, however only 8 reversals were used with the last 6 of each run being averaged. Two threshold measurements (set of reversals) were taken per condition, or more if the first 2 showed high variance. If the cancellation was successful, as demonstrated by a positive threshold difference when the crosstalk-cancellation filter was implemented, it would be repeated at the start of the next session using the same filter design. The left side was cancelled and tested first, and once two left cancellation filters had been created twice and given consistent results, the right side was then cancelled. The right side was tested both in isolation, and with the left filter active simultaneously (so that both the tone and the noise are cancelled at their contralateral cochlea).

Patients loosely followed the same procedure however there was variation according to patient's needs and preferences. For example two patients required the experimenter to adjust the phase and level values during the fine-tuning for them while they indicated verbally when the tone was cancelled.

4.4 Results

Nine patients were recruited for this study, however successful cancellation over multiple sessions was only seen in 2 patients, although 4 patients were able to demonstrate a threshold benefit in a single session. Most patients withdrew from the study after they had produced an effective filter due to time constraints and 7 patients indicated that they would have continued with the study if there had been more time.

Patient 115 had CHARGE, which is defined as Coloboma, Heart defect, Atresia choanae, Retarded growth and development, Genital hypoplasia, and Ear anomalies/deafness (Blake et al., 2006). She was unable to complete the task and would choose round numbers (e.g. -5 dB and 50°) rather than trying to cancel the sound. After the cancellation phases and levels were hidden, the cancellation values became inconsistent between frequencies, suggesting the patient was not cancelling the sound. Patient 142 had not used a computer or laptop before and was unfamiliar with a mouse or any computer interface, and so the experimenter adjusted the phase and amplitude while the patient indicated verbally when the cancellation of the tone occurred. Patient 102 declined to return due to frustration with the task as well as the travel time to the hospital and patient 141 was not invited to return due to concerns with the task worsening her tinnitus. Patients 108, 142 and 146 intended to return for follow-up sessions but could not find the time.

4.4.1 Patient 145

4.4.1.1 Left Cancellation

Patient 145 completed the study over 1 and a half months totalling 5 sessions. During the first session she attempted cancellation, but no benefit was seen during the threshold measurement. During her second session she cancelled the left side over approximately 2 hours achieving a threshold improvement of 7.0 dB (Figure 4.1) and created another on session 3 over 1 hour which achieved an initial threshold improvement of 12.3 dB. The timeline for filter creation and the TRT improvements when using the filter are shown in table 4.2.

Session No.	Cancellation Left 1 (dB)	Cancellation Left 2 (dB)	Cancellation Left 3 (dB)	Cancellation Right 1 (dB)	Cancellation Right 2 (dB)	Cancellation Right 3 (dB)
1 (day 1)	0	-	-	0	-	-
2 (day 15)	-	7.0	-	-	-	-
3 (day 26)	-	7.0	12.3	-	-	-
4 (day 39)	-	5.5	9.3	-	2.5	-
5 (day 44)	-	-	6.3	-	-	3.2

Table 4.2. A table showing which filters were calibrated on which sessions and the TRT improvements that were seen when the appropriate crosstalk-cancellation filter was used. Cancellation Left 1 refers to the improvement in TRT at the first attempt to cancel the left side, cancellation left 2 refers to the second attempt to cancel the left side etc

The cancellation phase and levels can be seen in Figure 4.2. An average level difference of 2.4 dB is seen across all frequencies (although this is reduced to 1.2 dB if you remove the sudden dip at 2937.6 and 2875 Hz) and an average phase difference of 10.4°. In both the previous study and Mcleod and Culling (2020) sudden level differences are associated with a phase reversal due to the antiresonance that can occur at particular frequencies in individual skulls. Since this same phenomenon cannot be seen here, along with the poorer results in the session 1 cancellation compared to session 2 suggests that the cancellation levels at 2937.6 and 2875 Hz may be inaccurate.

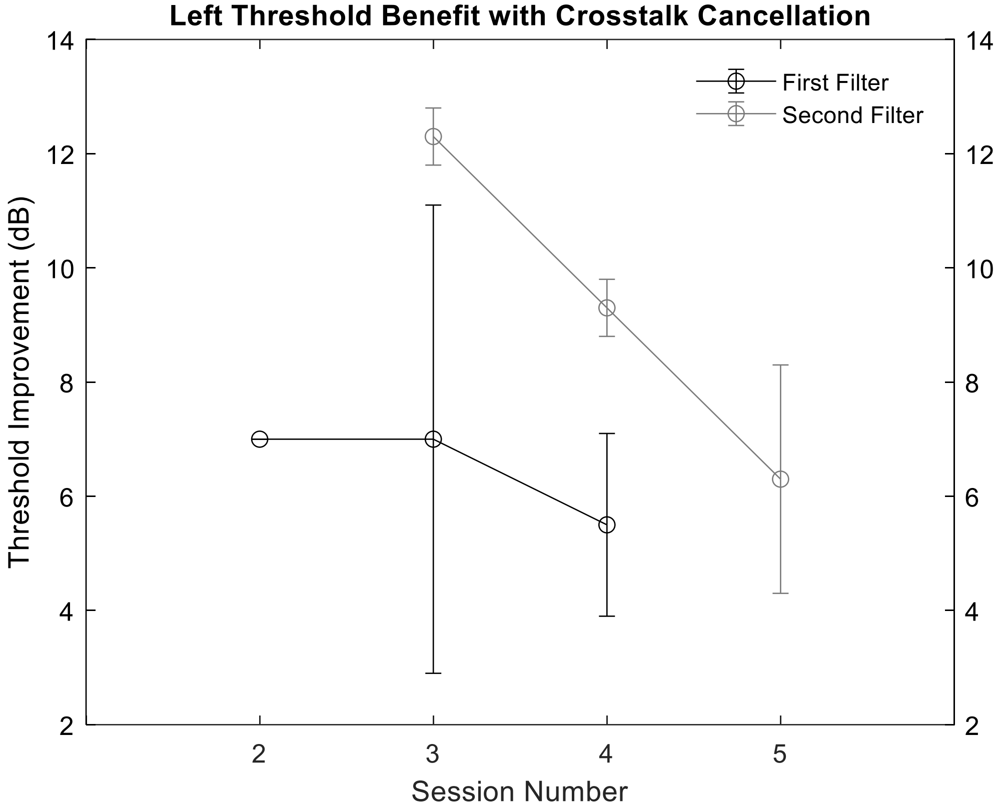
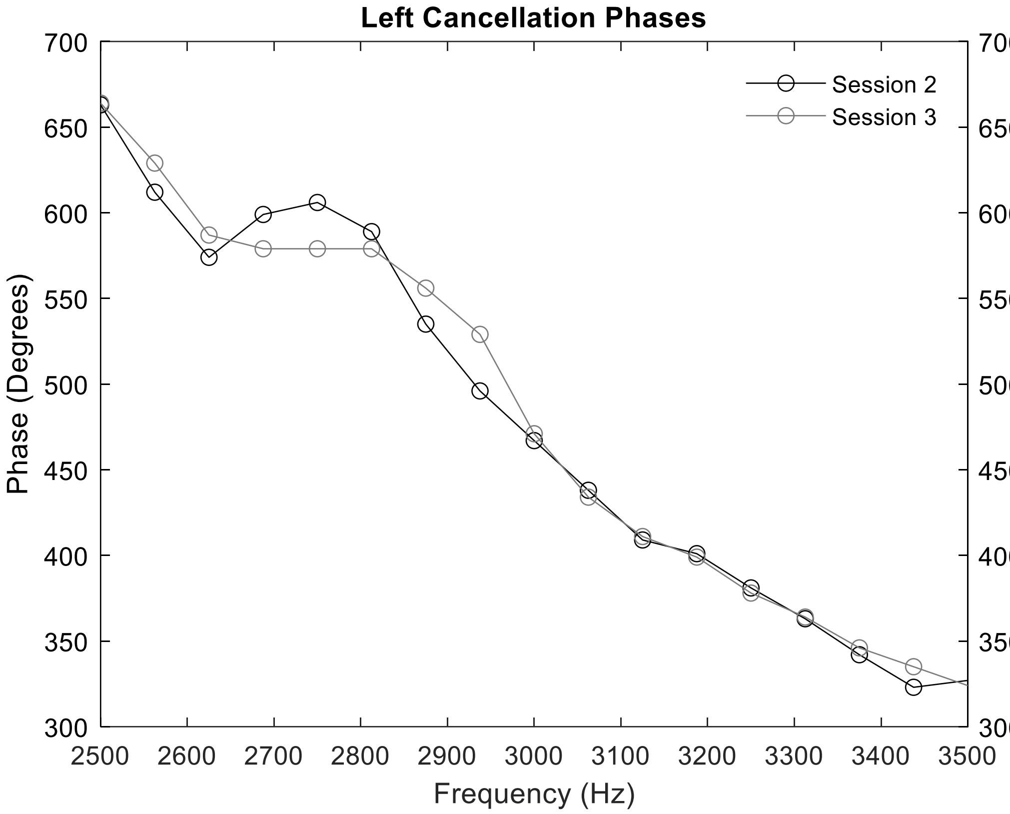


Figure 4.1. A graph showing the left TRT improvement after the left crosstalk-cancellation was implemented.



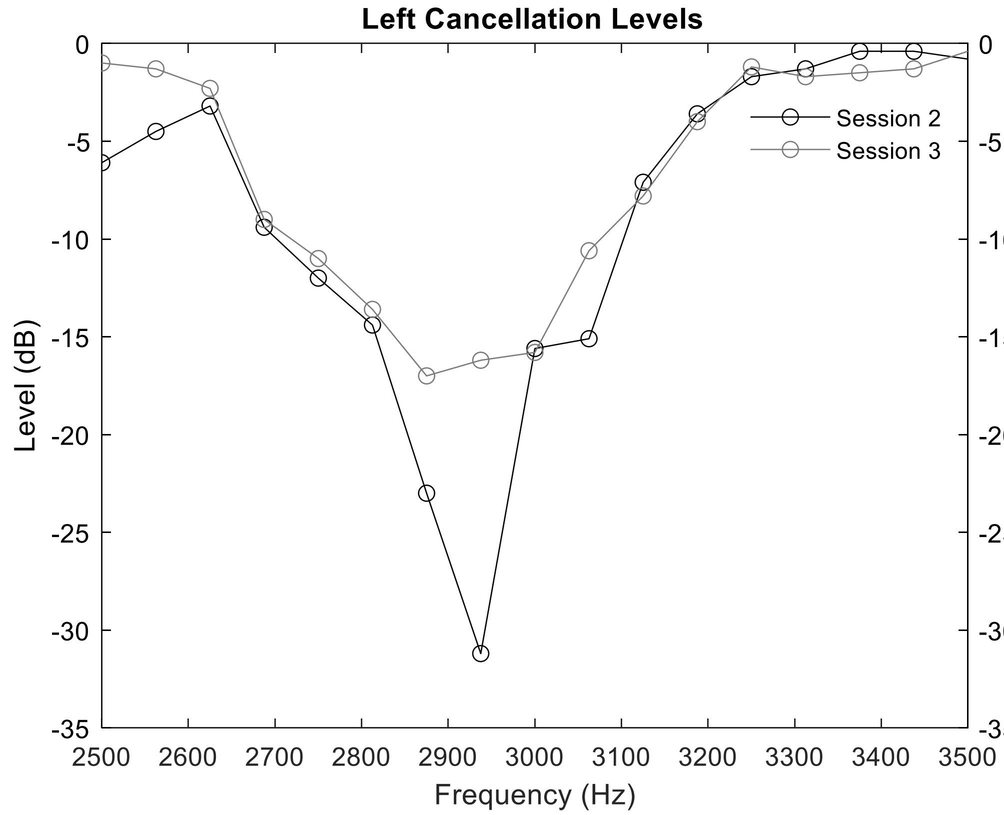


Figure 4.2. Graphs showing the cancellation phase and levels from the successful cancellation attempts on sessions 2 and 3.

4.4.1.2 Right Cancellation

The right calibrations were performed on the final two sessions, after the left side had already been calibrated twice and the cancellation phases and levels are shown in figure 4.3. The initial right side threshold value was 2.5 dB and the second was 3.2dB, both calibration attempts occurred over roughly 1 hour. The cancellation level and phases are shown in figure 4.3. An average level difference of 2 dB and an average phase difference of 35.5° were seen. Overall an improvement of 2.0 dB was seen when the crosstalk-cancellation filter was used ($t(8) = 2.32$, $p < 0.05$) in both right-side and simultaneous TRTs. Right thresholds were not repeated without the left filter also being used simultaneously.

4.4.1.3 Simultaneous Cancellation

Two thresholds were measured on the fourth and fifth sessions (figure 4.4) where both the left (specifically the second successful left calibration) and the right (first successful right cancellation) were used to cancel crosstalk from both BCHDs simultaneously. Thresholds benefit when bilateral crosstalk-cancellation filters were applied degraded by 1dB across sessions 4 and 5 on both the left and right sides.

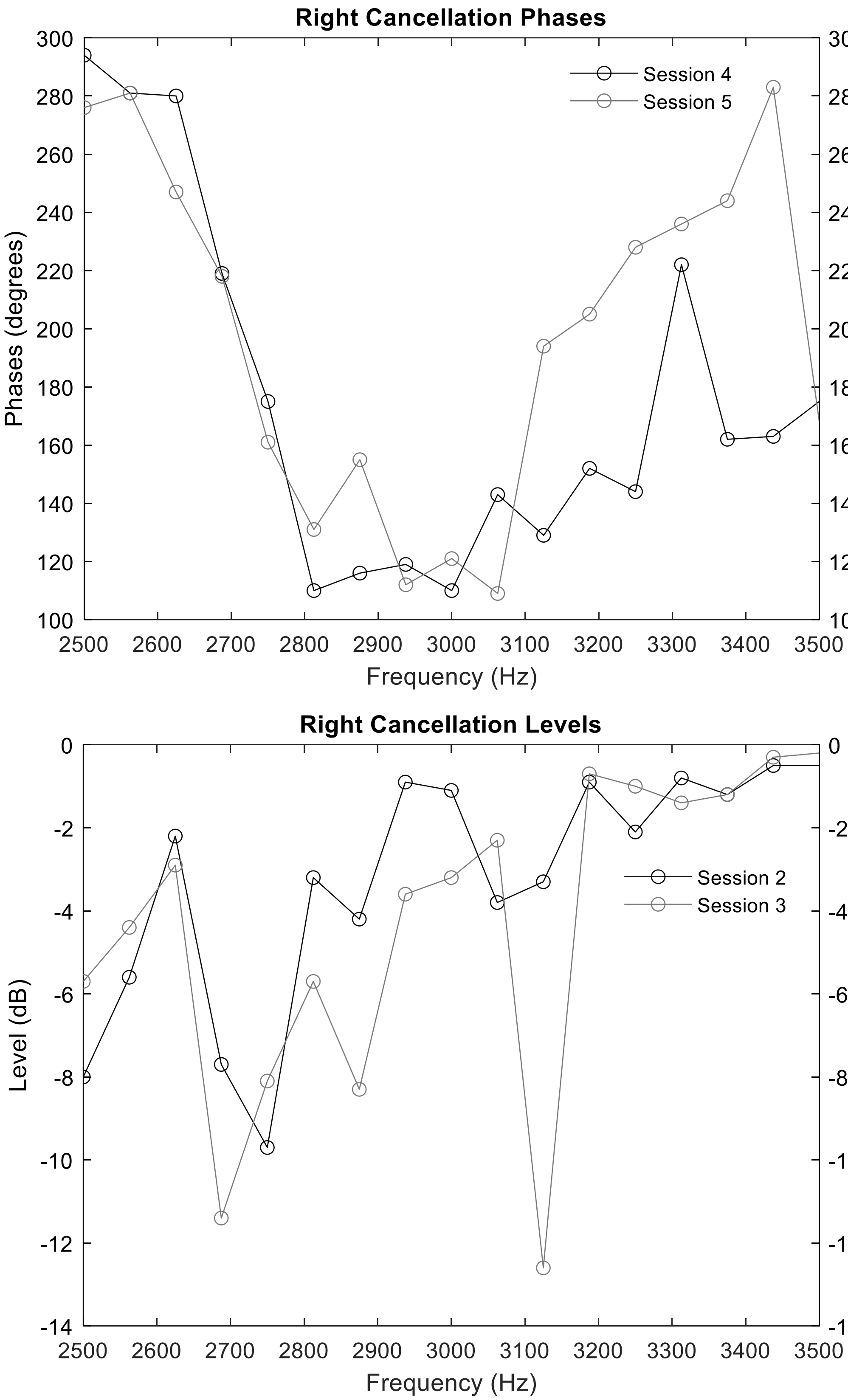


Figure 4.3. Graphs showing the cancellation phase and levels from the cancellation attempts on sessions 4 and 5.

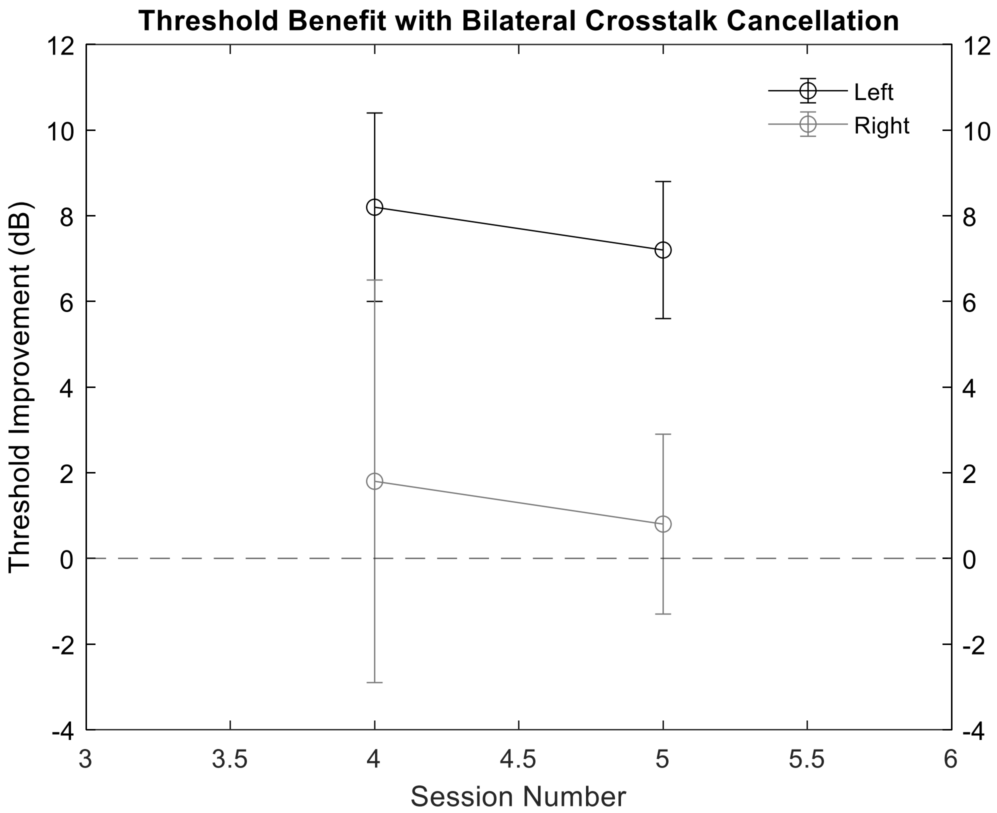
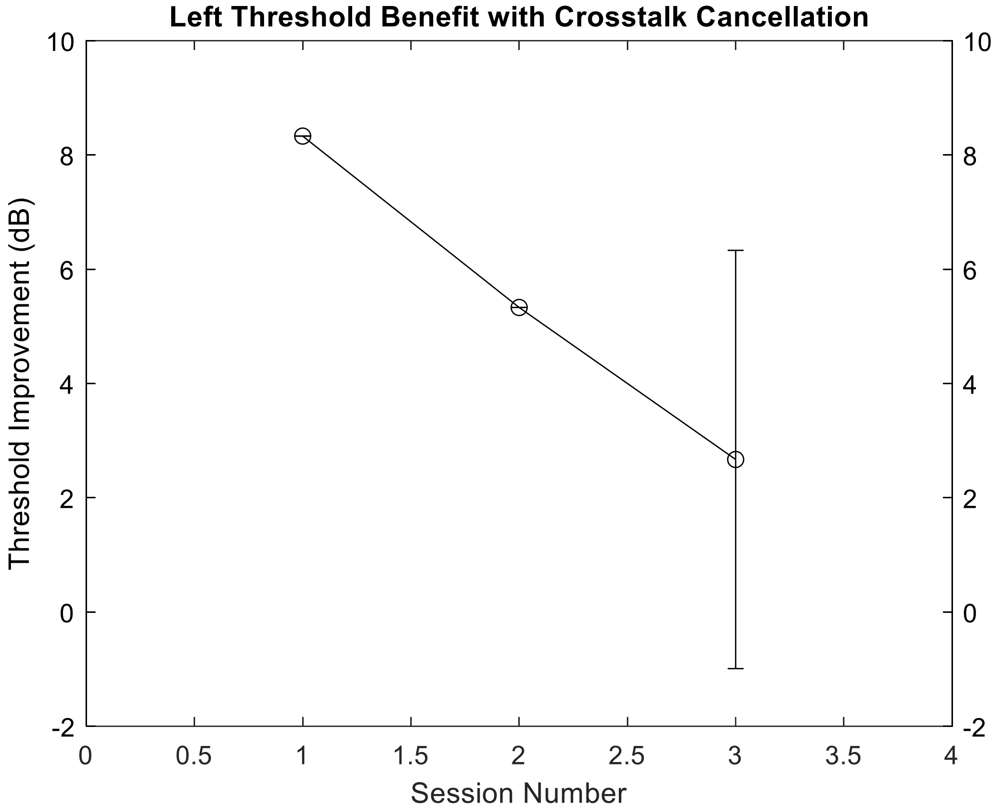


Figure 4.4. A graph showing the TRT improvement when the first right filter and the second left filter are applied simultaneously.

4.4.2 Patient 127

Patient 127 created a successful crosstalk-cancellation filter for their left ear during their first session, with a threshold improvement of 8.3 dB when the filter was applied (Figure 4.5). The filter efficacy deteriorated by 5.63 over 93 days. Left filter calibration was unsuccessful on sessions 2 and 3 and right filter calibration was unsuccessful on session 1 and 2.

Figure 4.5. A graph showing patient 127s TRT improvement when the left side filter created in session 1 was used.



4.5 Discussion

Mixed success was seen during the filter calibration. Only 4 out of 9 patients achieved some benefit when using the crosstalk-cancellation filters, although most patients only attended one or two testing session which may not have been enough to train the patients to cancel sound. Both Mcleod (2020) and the experiment described in chapter 3 trained patients for longer than the 2 – 4 hours some patients in this study had, so it is possible patients may have gained the ability to accurately cancel sound with further training.

Tinnitus was a complaint of patient 141, who found her tinnitus masked the cancellation tone and so was unable to perceive when sound had been cancelled, and of patient 145, who, found that while tinnitus impaired her ability to perceive cancelled sound in the first session was still able to cancel on the left side in following sessions. The tinnitus rate in normal-hearing populations is estimated to be around 10.2%, increasing in older populations (Davis, 1989), and 24% in patients with BCHDs based on a study of 69 patients with mostly unilateral implants (Lekue et al., 2012). This may contribute to making the success rates of cancellation lower in patients than a normal-hearing population.

The consistency of the cancelation phase and levels are reliant on the assumption that the patient had perfectly cancelled the crosstalk. In patient 145's first successful left cancellation there is a sudden decrease in level around 2937.5 Hz, which is unexpected given the typical pattern of cancellation levels of both the other left cancellation attempt and in cancellation from previous studies (such as Mcleod and Culling (2020) and the experiment described in chapter 3 of this thesis). Errors like this would increase the difference in average cancellation phase and level between sessions, not because the values have changed but because the patient has not perfectly cancelled the sound. Despite this fact the average differences were only 2.44 dB and 10.41° which suggests that there is some level of consistency in transcranial attenuation of sound over time, at least in the two sessions it was tested. If the filter was recalibrated in later sessions (such as those where the filter efficacy depreciated) the observed differences may have been greater.

Patients 145 and 127 were the only patients who both achieved accurate cancellation and returned for multiple sessions. All successful cancellations degraded in performance over time, although not so substantially as to eliminate their effectiveness. It is however possible that if the patients were tested over a longer period of time the effectiveness would reduce, either to 0 or to a base level of effectiveness. The better left filter in P145 sharply reduced in effectiveness across sessions from 12.3 dB to 6.3 dB over 18 days, whereas their poorer left filter only decreased from 7 dB to 5 dB over 24 days, suggesting that there could be a base level of cancellation around 5 - 6 dB that a somewhat accurate filter can achieve. However patients 127s filter efficacy rapidly decreased initially, reducing in effectiveness by 3 dB over 21 days, but then only by 2.63 over 93 days suggesting the filters could depreciate at different rates in different patients or that there's individual differences in the filters themselves. Cancellation efficacy of the patient 145s left filter did not change when it was used in the TRT test using simultaneous crosstalk cancellation, however it also reduced in effectiveness between sessions from 8.2 dB to 7.2 dB.

Future research should focus on addressing these issues, as well as generally testing more patients for a longer period of time. This would establish: what percentage of patients would be able to create these filters, whether unsuccessful patients could be trained given enough time, and whether the filters continue to deteriorate over a longer time frame. Variables such as tinnitus severity, learning difficulties and degree of cochlear hearing loss should also be tested to determine whether typical BCHD patients are able to create crosstalk-cancellation filters.

Additionally, there should be a focus on creating an application for calibration that could be distributed to patients that could be completed by patients without a clinician present. The automatic sweeping used in this study could be the basis of this application where the sweeps gradually narrow in range after phases and levels are locked in until the patient until cancellation has been achieved. This was not done in the present study due to its time-consuming nature, but patients would have the ability to perform the cancellation according to their own schedule rather than having to do a certain number of frequencies in a given session.

4.6 Conclusions

This study used a cancellation method based on Mcleod and Culling (2020) to cancel the crosstalk from BCHDs in patients with Bilateral BCHDs. Mixed success was seen in patient's ability to create crosstalk-cancellation filters. This may have been partially due to extenuating factors such as learning difficulties or tinnitus, or because patients did not have sufficient training time. Filter efficacy over time was tested in two patients. One patient created an effective left crosstalk-cancellation filter which improved their left TRT threshold by 12 dB, which decreased to 6.33 dB after 18 days. Another left crosstalk-cancellation filter was created which decreased from 7 dB to 5.5 dB after 24 days. Another patient's left side filter reduced in effectiveness from 8.3 dB to 2.67 dB over 114 days. Right-side cancellation only showed limited effectiveness. Future work should aim to test a variety of patients over a long period of time to establish the ability for patients to create these filters and whether their effectiveness continues to decline in the longer term.

Chapter 5 - Localisation Ability of Bone-Conduction Hearing Compared to Air Conduction

5.1 Introduction

Bone Conduction Hearing Devices (BCHDs) have been an effective treatment for patients who cannot benefit from traditional hearing aids due to having damaged or missing middle and/or external ears. Patients with middle- or outer-ear atresia or conditions which affect facial construction (such as Treacher Collins Syndrome), as well as those with skin problems which cause difficulties in implantation are the target population for BCHDs.

While these hearing devices are effective at restoring participants hearing to a degree, they are not able to restore a participant's sound localisation to normal levels. For bilaterally implanted patients, crosstalk caused by bone conduction (BC) hearing is considered one of the key factors that's limits sound localisation, due to the disruption of binaural cues. For unilaterally implanted patients with SSD, the device is placed on the non-hearing side and, through BC, sound from that side will reach the hearing cochlea. Patients with bilateral conductive hearing loss may also be fitted with a single BCHD due to the limitations in the perceived benefit of bilateral fitting, potential problems caused by the surgery and the cost involved. In this case, sound localisation is dependent on monaural cues available at this single BCHD. The present chapter will be concerned with patients who have been bilaterally fitted and the effects of the crosstalk on localisation.

Patients with bilaterally implanted BCHDs have access to limited binaural cues, which allows them to perform better on listening tests than their unilaterally implanted counterparts (Bosman, 2001; Dutt et al., 2002a; Priwin, 2004; Priwin et al., 2007; Canale, 2022). Bilateral implantation also provides patients with limited sound localisation, compared to unilaterally implanted patients who will always lateralise sound to the implanted side. Brassington (2023) also measured the minimum audible angle (MAA) in bilaterally implanted patients finding a MAA of 3.61° for participants to consistently lateralise sound, whereas an MAA of 75.04° was required for half (12 of 24) of the unilateral patients to consistently lateralise sound, the other half were unable to lateralise.

When looking at accuracy at each individual azimuth using bilateral BCHDs, patients perform poorly when sound is presented from the front, or at ± 90° only accurately localising between 40-50% of sounds within a 30° margin (Fan et al., 2020). When the individual patient data is looked at, performance between each patient shows a high degree of variability ranging from lateralising all sounds to one side to localisation performance within the normally-hearing range (den Besten et al., 2020; Caspers et al., 2021). Most bilaterally-fitted patients can accurately localise sound with some degree of success, although confusion is common around the centre-front azimuths. The existence of one participant in den Besten et al. (2020), who was able to localise sound at a normal hearing level, shows that it is possible to accurately localise sound with bilateral BCHDs. Incidentally a different patient was able to accurately localise sound with one BCHD on only the left side, but could only lateralise sound when bilateral BCHDs were used.

The cause of this variation in localisation ability between patients is unknown but Caspers et al. (2011), suggest that it could be caused by BC threshold asymmetry or age of bilateral implantation. They also tested patients across 4 sessions on different days, and while two participants showed inconsistent performance across the 4 sessions, most participants showed little or no difference between sessions. Caspers et al. also attempted to improve patient’s ability to localise sounds using three methods. The first method used was to adjust the device settings of patients BCHDs. This involved disabling the inbuilt adaptive microphones and noise reduction as well as variable gain and was done in accordance with previous literature that suggested these features could impair sound localisation in bilateral hearing aids (Van de Bogaert et al., 2006). Although no measurable improvement was seen in the localisation tests after these were implemented, 87% of participants preferred their new device settings and they reported improved clarity and loudness of sound. The second method was a short localisation practice session and the third method was instructions patients to practice localisation cues in every day life, for example trying to locate the source of a sound with their eyes closed. Neither of these methods had an impact on localisation scores.

BC localisation has also been tested in normal hearing participants. MacDonald et al. (2006) tested BC thresholds in 4 participants at 45° intervals ± 180° either side of the head using a BC headset, and background noise to mask any AC output from the BTs. They found that BC and AC performed similarly, finding a mean error rate of 17° for the BC condition and 22° for AC. Reversal rates of 7% and 12% were found respectively. Snapp et al., 2020 used an adhesive BCHD attached to the mastoid bone of normal-hearing participants wearing bilateral ear plugs, finding that bilateral BCHD use restored binaural cues for localising broadband noise. Localisation did not match the level of normal hearing (MAE = 8.6° for unimpaired hearing and MAE = 16.5° for aided hearing) and a bias towards the lateral extremes became apparent. Although they found further evidence for bilateral BCHD use benefiting sound localisation, Denanto et al. (2022) found little benefit from bilateral implantation when listening for speech in competing speech and little bilateral benefit in SRM. Denanto et al., note individual variation in bilateral benefit, finding that 5 of 25 participants performed worse when two BCHDs were used. The large difference in test azimuths however mean this experiment is more likely testing lateralisation than localisation.

In the present study, the effects of bone conduction and microphone type/position were separated. Normal-hearing participant's ability to localise sound was tested when speech was presented through both AC and BC. Speech was convolved either Ear canal or BCHD-position head-related impulse responses (HRIRs) in a 2x2 design. Different HRIRs were used to test whether positioning the microphones in a way similar to that of a typical BCHD were necessary for BCHD localisation. Performance with feedback was compared to performance without feedback, and the feedback group was tested again a few days later to test for improvement over time. If the participants show improvement over time in any condition, it suggests that a more extended practice session may be more effective at improving localisation than the short practice session utilised in Caspers et al. (2021).

5.2 Methods

The following experiment received ethical approval from Cardiff University Psychology Department Ethics Committee.

5.2.1 Apparatus

MATLAB™ was used during all testing procedures. An ESI MAYA44 USB+ four channel DAC was used to pass the signal either to a pair of Etymotic ER2 insert earphones with 3M™ E-A-RLink™ foam eartips, or to a pair of Radioear B71W bone transducers. Insert earphones were secured to the participant's clothing using clips so that the earphones would not be moved by the weight of the wires. The B71W bone transducers were placed in a pair of 3D printed glasses (shown in chapter 2), to align them against the participant's mastoid bone. An elastic headband was then placed over the B71Ws to provide slight compression against the skin.

5.2.2 Participants

Thirty participants from Cardiff University, 15 in in the feedback and no feedback groups, were recruited and received either monetary compensation or course credits. Two bilateral BCHD patients were also recruited from the previous study (patients 127 & 145). All normal-hearing participants reported normal hearing with no history of

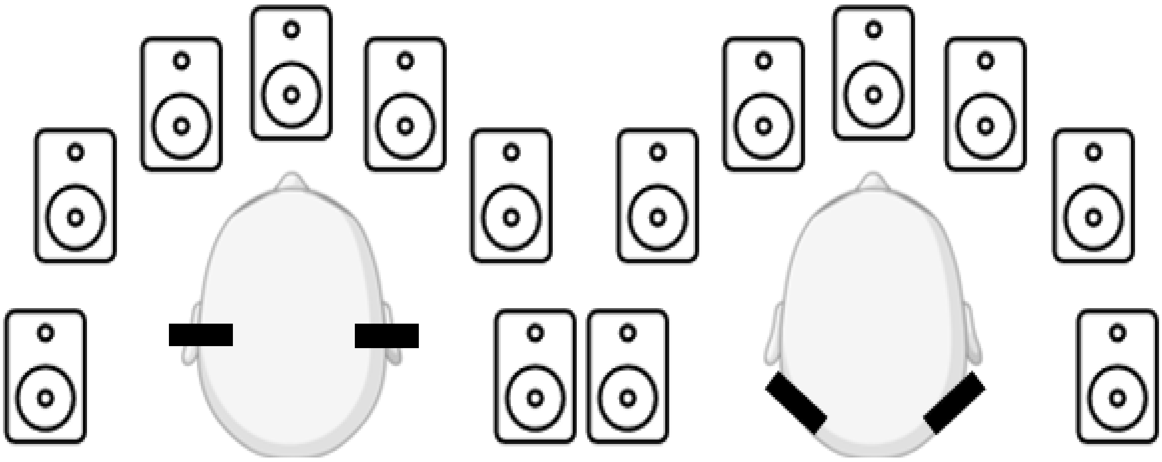


Figure 5.1. Illustration of the method used to record the HRIRs for the Ear-canal condition (left) where microphones were placed inside the ear canals of a KEMAR mannequin, pointing outwards, and the BCHD-position conditions (right) where the microphones were placed behind the head in approximately BCHD position with the microphones pointing outwards. There were 29 speakers used in the study but only 7 are shown here for illustrative purposes.

excessive earwax. One participant in the feedback group was removed from the analysis as they had extensive previous use of BTs.

5.2.3 Stimuli Creation

The stimuli used in the experiment was a male's voice saying 'this direction' since the two unvoiced fricative ensure that this vocalisation contains both high and low frequency components, as well as representing an ecologically valid announcement. This speech was convolved with two sets of HRIRs to simulate free-field hearing with both normal hearing and BCHD hearing. The first set were ear-canal HRIRs which were gathered using a KEMAR mannequin in a sound-treated room. A speaker array (Figure 5.1) consisting of 29 speakers even spaced, at 7.5° intervals between 105° either side of the mannequin were used to gather each transfer function using the log tone sweep method (Muller and Massarani, 2001). A second set of BCHD-position HRIRs (Figure 5.2), were gathered using a mannequin (B&K HATS) with two Behringer C-2 cardioid microphones placed behind each ear, approximately in the same position as a BCHD microphone would be on patients and the B71W bone transducers would be placed on participants during the experiment. The same method was used to obtain both sets of HRIRs.

5.2.4 Stimuli Presentation

The stimuli were presented to participants between ± 90° either side of the participant in the test phase, or ± 105° during the calibration phase. A user interface (figure 5.3) containing boxes spaced 7.5° between ± 105° in an arc, to mimic the directions of the sound sources, were presented to the participants on a laptop screen. ±97.5° and ±105° were included to allow the participant to overshoot the 90°-presented stimuli, but no sounds from those angles were played.

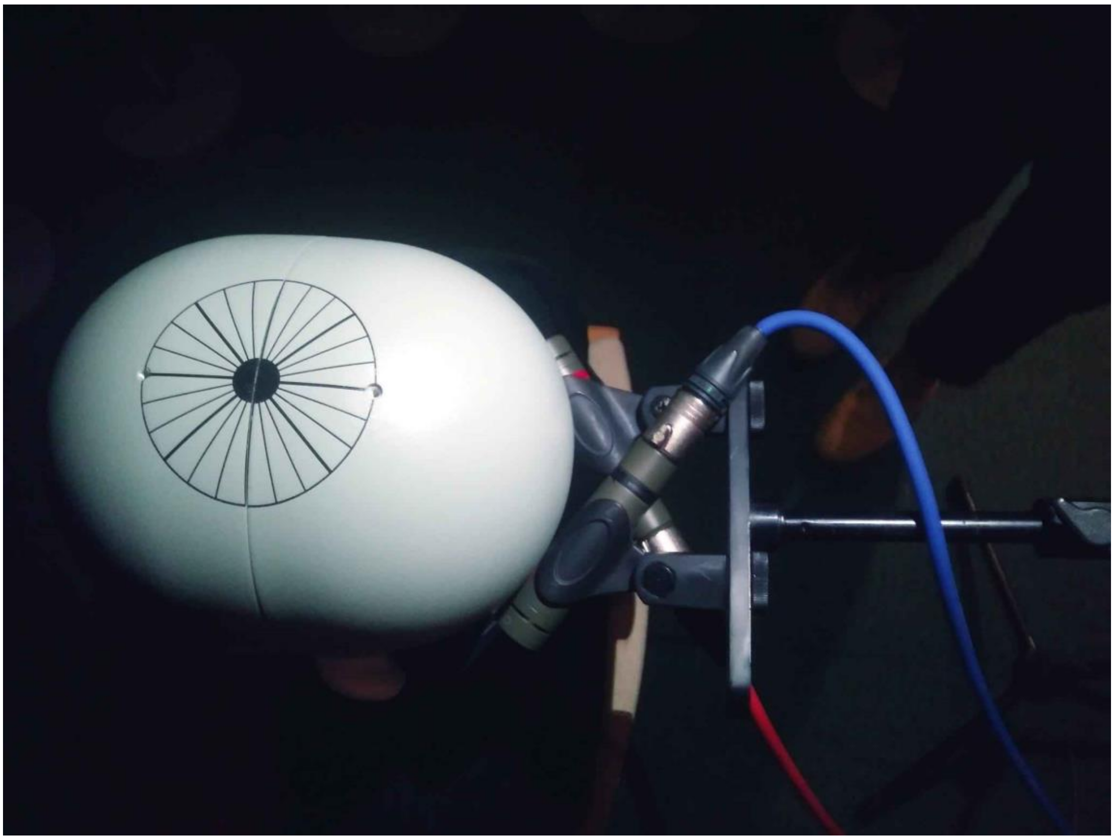


Figure 5.2. A photo taken of the microphone position for the BCHD HRIRs.

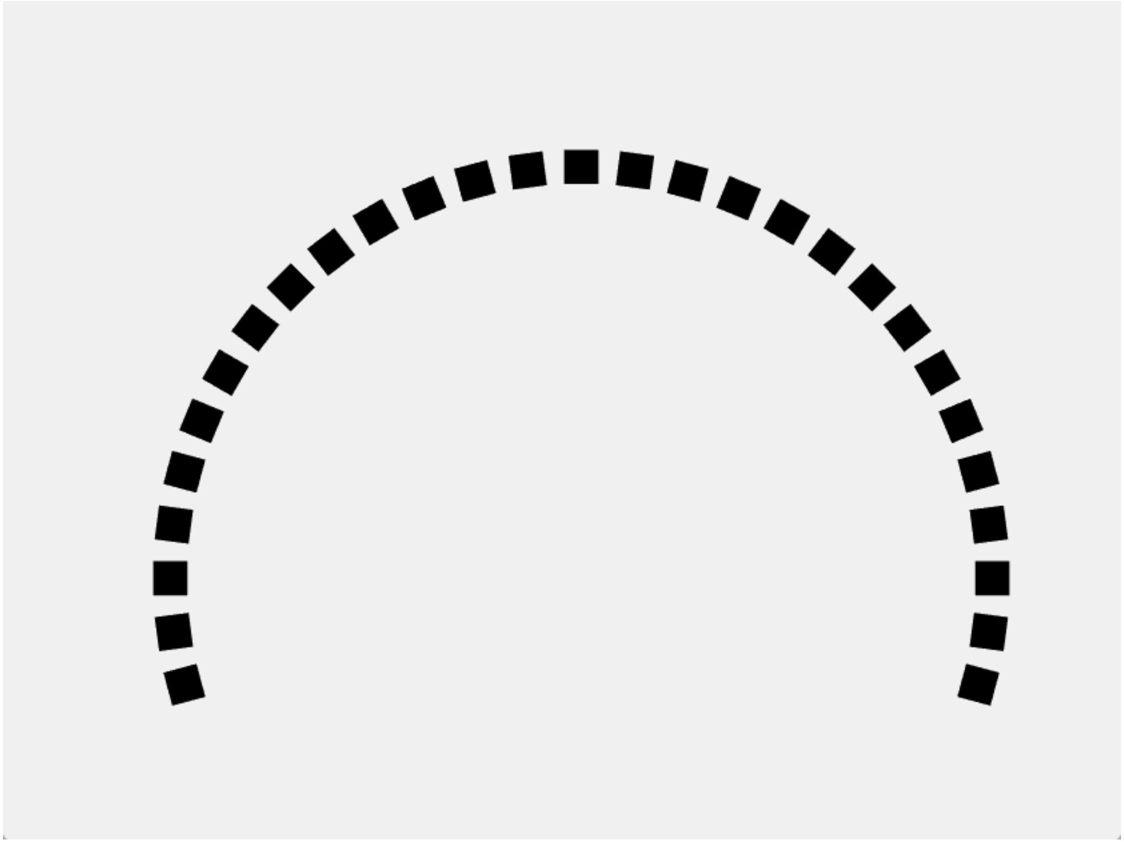


Figure 5.3. The interface participants used to indicate which direction they perceived the signal from.

5.2.5 Procedure

The experiment lasted between 45-65 minutes depending on the participant's response speed. The experiment utilised a 2x2 design (table 5.1): sound presented by air conduction (Using ER2s) with either BCHD-position ($P_{AC}R_{BCHD}$) or ear-canal HRIRs ($P_{AC}R_{EC}$)and sound presented through bone conduction with BCHD-position ($P_{BC}R_{BCHD}$) or ear-canal HRIRs ($P_{BC}R_{EC}$). Condition order was counterbalanced between participants, and conditions with the same sound conduction were always paired (i.e. the two BC-presentation conditions were then followed by 2 AC-presentation conditions or vice versa) to reduce time spent swapping equipment. Each condition started with a 'calibration' phase where participants would listen to the stimuli from each direction, starting at -105° (left side) progressing through each 7.5° increment until it reached 105° (right side) and click on the corresponding box in the response interface. This was to expose participants to the speech from each potential response angle before testing, allowing them to calibrate themselves to the interface. Boxes at 97.5° and 105° were present to prevent the available response set causing edge effects in the data when sounds were presented towards ± 90°.

Following the calibration phase was the test phase. Here speech would be presented from each azimuth once in a random order until all azimuths had been presented and participants were required to respond by selecting a box from the array of boxes described earlier. Once all azimuths had been randomly presented once, the next block would start, randomly cycling through the azimuths again. This happened 10 times, resulting in 10 blocks in each condition, and 10 measurements for each azimuth in each condition. In the first experiment there was no feedback provided. In the second experiment, when participants responded correctly the box they had selected would turn green, when they responded incorrectly the correct box would turn red. Participants were offered breaks between each condition. In the second experiment participants returned a few days later to repeat the experiment.

HRIR Recording (mannequin) / Presentation mode	**BCHD position (B&K HATS)**	**Ear canal (KEMAR)**
AC	$P_{AC}R_{BCHD}$	$P_{AC}R_{EC}$
BC	$P_{BC}R_{BCHD}$	$P_{BC}R_{EC}$

Table 5.1. The names of the conditions in this experiment and their meaning. P is short for presentation mode, and R represents HRIR recording.

5.2.6 Data Analysis

Data analysis was split into a 2x2 repeated measures ANOVA when comparing the first session of the feedback group to the second session of the feedback group and an independent samples t-test when comparing the no-feedback group to the first session of the feedback group.

5.3 Results

Figure 5.4 shows the root-mean-squared-error (RMSE) for each participant in the no-feedback group and Figure 5.5 shows the raw data in a bubble plot showing a bias towards the extreme lateral azimuths. A 2x2x2 factorial ANOVA comparing the non-feedback group to the first session of the feedback group revealed that that was an effect of feedback ($F(1,108) = 77.81$, $p < .001$), shown in Figure 5.6, and whether the stimuli were presented via AC or BC ($F(1,108) = 13.48$, $p < .001$) but not of HRIR ($F(1,108) = 0.26$, $p = .061$). No significant interactions were observed.

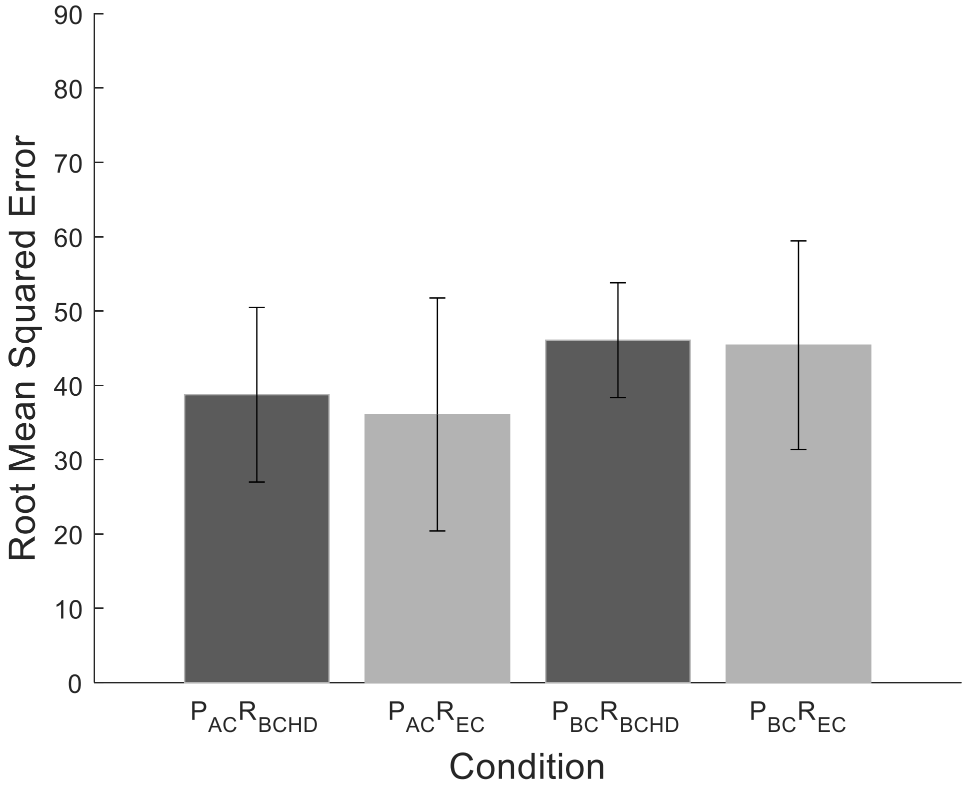


Figure 5.4. Mean RMSE of each condition in the no-feedback group. The error bars represent standard deviation.

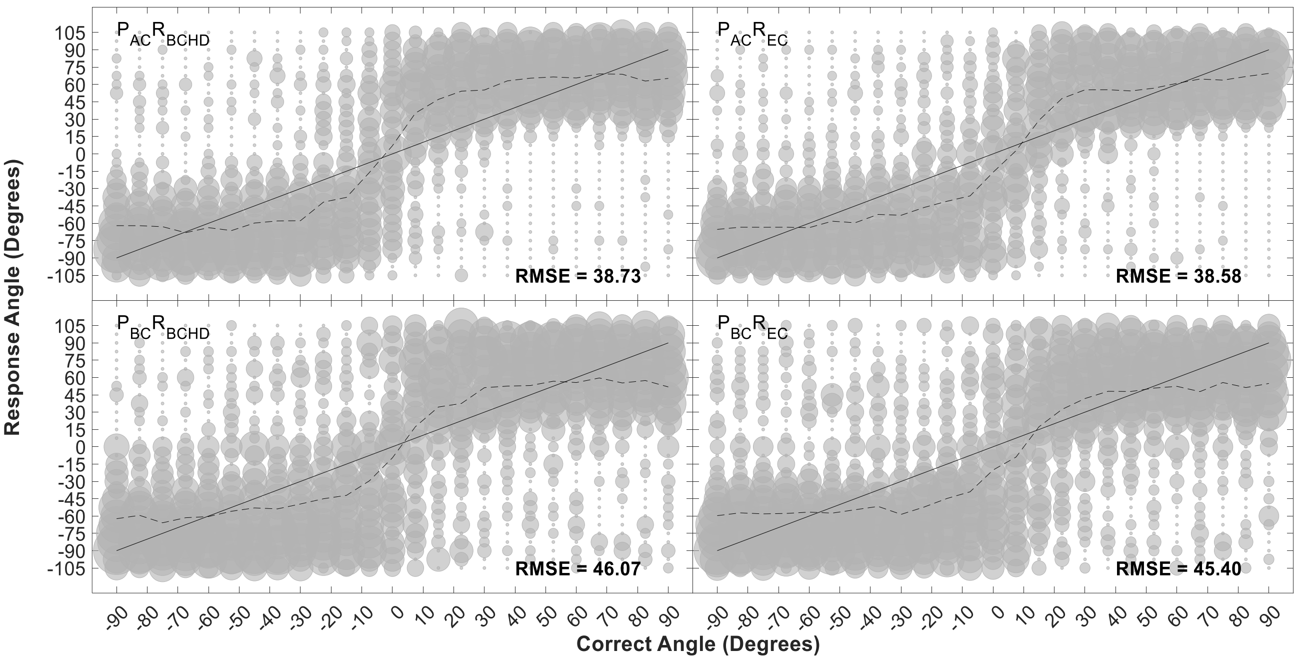


Figure 5.5. A bubble plot showing the raw results for the no-feedback group. Bubbles indicate the participant's responses, bigger bubbles indicate more responses. The dotted line represents the average response, and the filled line represents perfect localisation.

Large variance was seen between participants in both the AC and BC presentation conditions, as seen in previous studies (den Besten et al., 2020; Caspers et al., 2021). Figure 5.7 shows the improvement per block over each condition for the no-feedback group, averaged across all azimuths. A linear regression was used to test whether accuracy could be predicted by block showed that overall RMSE improved over the course of the experiment (F(1,38), = 26.36, p < .001) , accounting for 41% of the variance in RMSE (R^2 = .41, adjusted R^2 = .39). The regression equation indicates that RMSE decreased by 0.154 every block (b = -.154, t(38) -5.13, p < .001). Further line regressions testing whether accuracy could be predicted by block over a single condition (i.e. 10 blocks rather than 40) are reported in table 5.2, showing that only RMSE in the $P_{AC}R_{EC}$ condition could be predicted by block, notably that RMSE increases over each block.

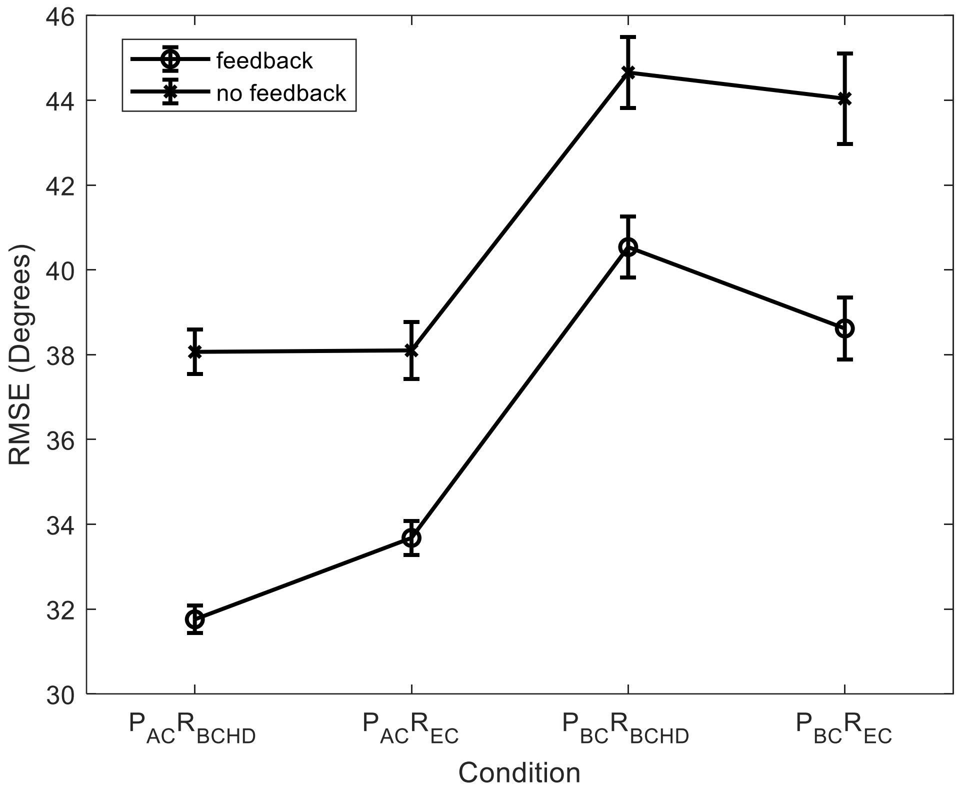


Figure 5.6. RMSE of the no-feedback group compared to the first session of the feedback group. Error bars represent standard error.

Condition	*b*	SE	*t*	*p*	R^2	*F*(df)
$P_{AC}R_{EC}$	0.63	0.20	3.10	<.05	0.55	9.63(1,8)
$P_{AC}R_{BCHD}$	-.08	0.21	-.038	.71	0.018	.14(1,8)
$P_{BC}R_{AC}$	.18	0.19	.96	.37	.10	.91(1,8)
$P_{BC}R_{BCHD}$	-.11	0.21	-.53	.61	.034	.28(1,8)

Table 5.2. A table showing the results of a regression analysis between each condition across both sessions of the feedback group, testing whether RMSE can be predicted as a function of block number.

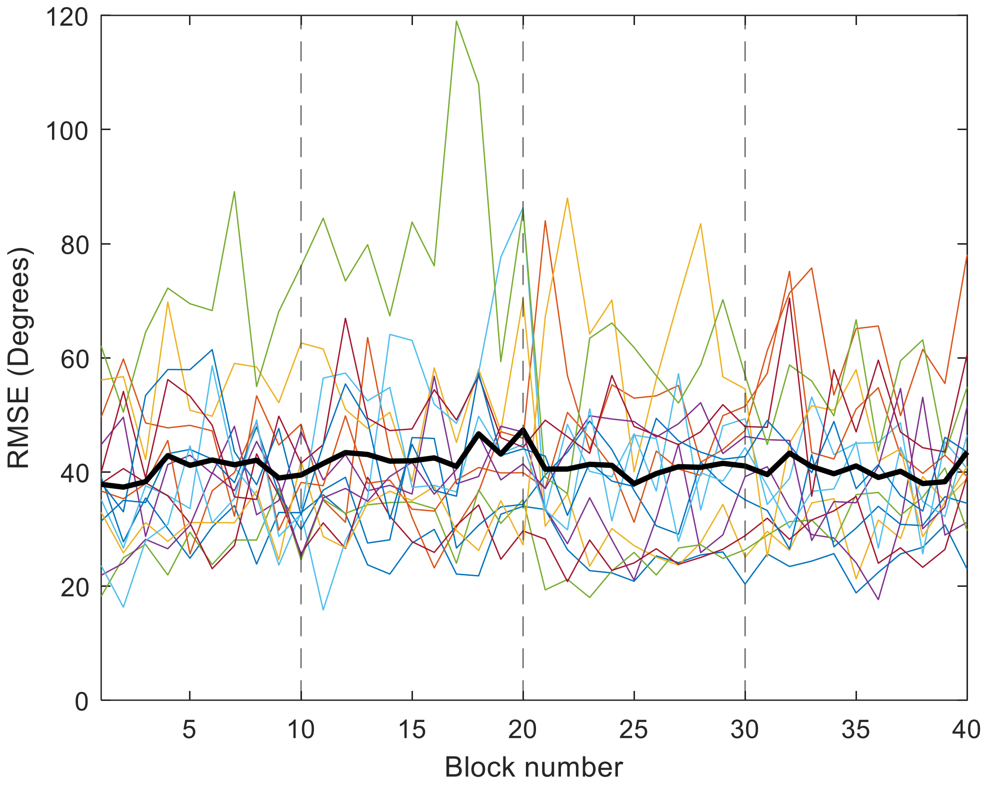


Figure 5.*7.* RMSE in the no-feedback group for each block in the order the participant completed them, regardless of condition. The black line represents the mean RMSE for all participants from each block.

The variance in localisation ability between participants in the feedback group was high in the first session for both the $P_{BC}R_{BCHD}$ (*SD* = 12.57) and $P_{BC}R_{EC}$ conditions (*SD* = 14.36), but became less varied in the second session (*SD* = 8.70, *SD* = 7.09). A linear regression measuring improvement against block over both sessions reports that accuracy can be predicted as a function of block ($F(1,78) = 17.54$, $p < .001$) , accounting for 18.4% of the variance in RMSE ($R^2 = .184$, adjusted $R^2 = .173$). This improvement over time is shown in figure 5.8. The regression equation indicates that RMSE decreased by 0.039 every block ($b = -0.039$, $t(78) = -4.19$, $p < .001$). Table 5.3 shows the data from regression analyses of the four individual conditions and Figure 5.9 shows the difference between sessions 1 and 2 for the feedback experiment in all conditions. In both P_{AC} conditions participants RMSE increased over both sessions of the experiment whereas in both P_{BC} conditions RMSE decreased.

Condition	*b*	SE	*t*	*p*	R^2	*F*(df)
$P_{AC}R_{EC}$	0.169	0.038	4.48	<.001	0.53	20.05(1,18)
$P_{AC}R_{BCHD}$	0.315	0.059	5.38	<.001	0.62	28.94(1,18)
$P_{BC}R_{AC}$	-0.198	0.064	-3.09	<.01	0.35	9.57(1,18)
$P_{BC}R_{BCHD}$	-0.327	0.075	-4.34	<.001	0.51	18.84(1,18)

Table 5.3. A table showing the results of a regression analysis between each condition across both sessions of the feedback group, testing whether RMSE can be predicted as a function of block number.

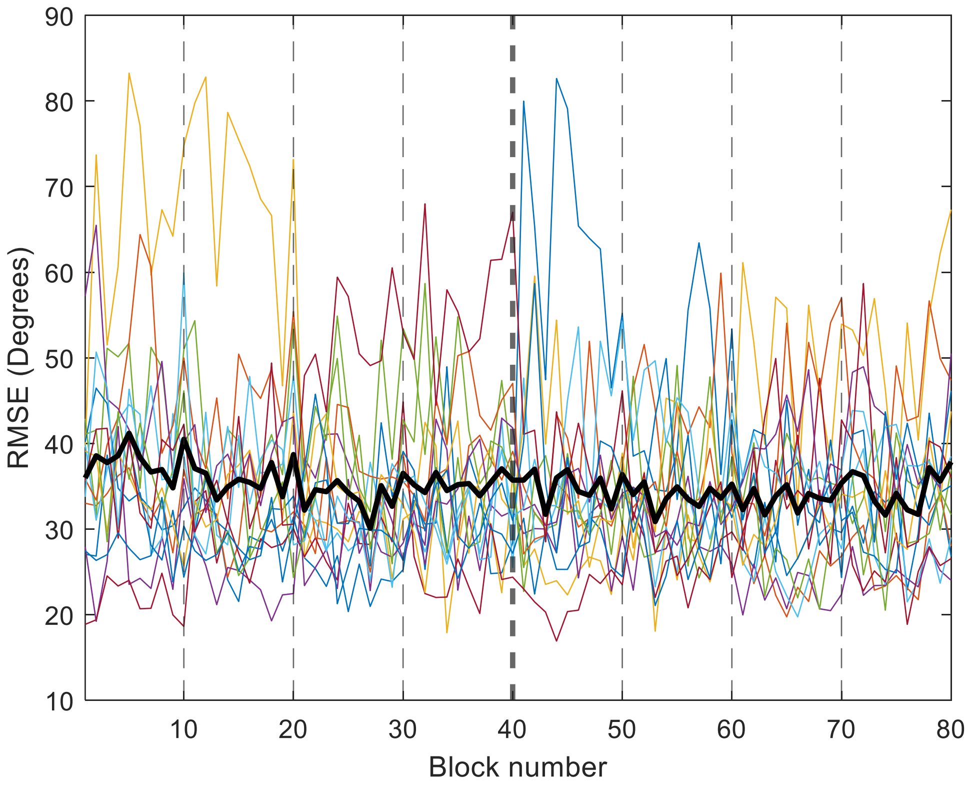


Figure 5.8. RMSE in the feedback group for each block in the order the participant completed them, regardless of condition. The grey dotted line shows the split between session 1 and session 2 and the black line represents the mean RMSE for all participants from each block.

Figure 5.10 shows the RMSE and standard deviation of the no feedback group and session 1 of the feedback group by angle. A three-way ANOVA showed that both presentation mode ($F(1, 192) = 51.83$, $p < .001$)) and feedback ($F(1,192) = 30.67$, $p < .001$)) produced significant changes in RMSE. Standard deviation was similar, finding that presentation mode ($F(1,192) = 81.427$, $p < .001$)) and feedback ($F(1,192) = 62.92$, $p < .001$)).HRIR did not show any differences in either measure, and there were no significant interactions.

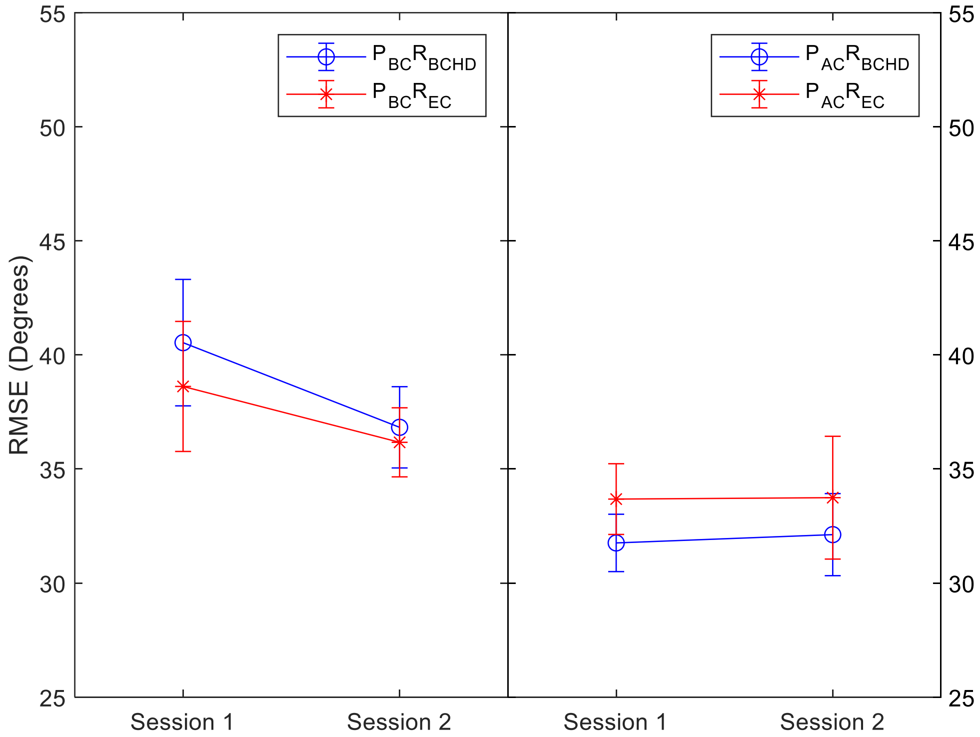


Figure 5.9. Improvement in RMSE for the feedback group from session 1 to session 2 for AC presentation (left) and BC presentation (right). Error Bars show standard error.

A 2x2x2 repeated measures ANOVA comparing results from sessions 1 and 2 of the feedback group reveals that RMSEs were significantly affected by presentation mode ($F(1,24) = 83.91$, $p < .001$)) and session number ($F(1,24) = 7.87$, $p < .05$)). Standard deviation was also influenced by presentation mode ($F(1,24) = 142.35$, $p < .001$) and session number ($F(1,24) = 16.70$, $p < .001$). No effect of HRIR was seen however interactions were seen in RMSE between presentation mode and HRIR ($F(1,24) = 4.66$, $p < .05$) and Presentation and session number ($F(1,24) = 38.90$, $p < .001$)). Additionally significant interactions were seen in standard deviation between presentation mode and HRIR ($F(1,24) = 11.04$, $p < .001$) and presentation mode and session number ($F(1,24) = 45.03$, $p < .001$).

Localisation was less accurate towards the +90° and – 90° azimuths when sound was presented over the BC headset, and feedback improved accuracy at the frontal azimuths. Localisation was less precise in the P_{BC} conditions at the highest and feedback improved precision around -90° as well as the frontal azimuths.

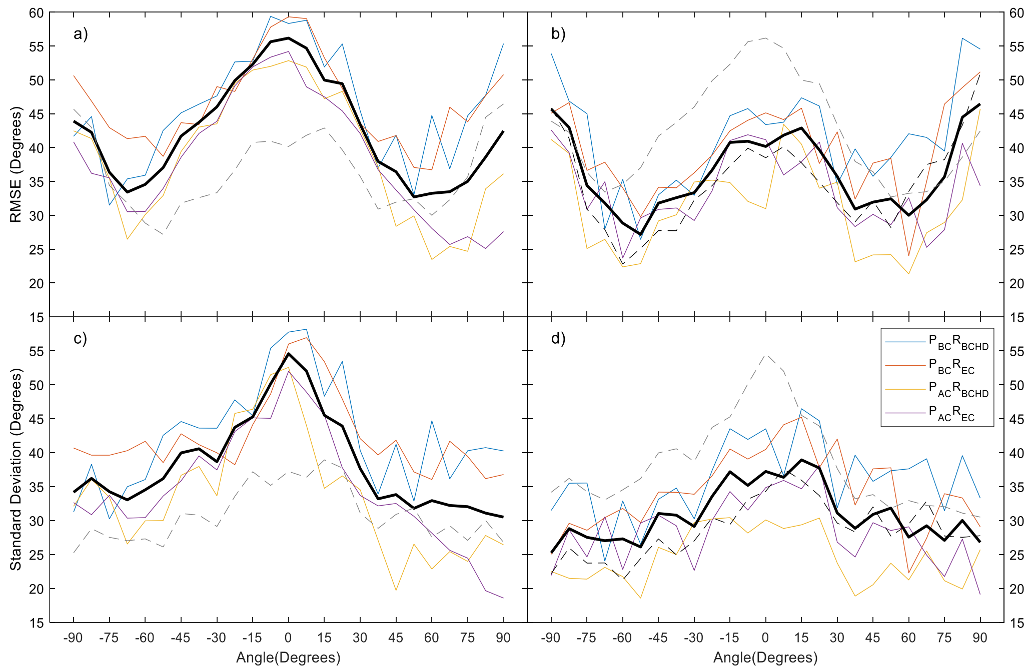


Figure 5.10. A graph showing accuracy (RMSE) and precision (Standard Deviation) per angle for the no-feedback group (a and c) and the first session of the feedback group (b and c). The dashed grey line shows the mean data from the feedback group on the no-feedback graphs and vice versa. The dashed black line shows the means from session 2 of the feedback group.

5.3.1 Localisation in Bilateral BCHD Patients

Two patients with bilateral BCHDs were also tested in the same procedure, their data is shown in a bubble plot (Figure 5.11). Only the $P_{BC}R_{BCHD}$ condition with feedback was used since they had poor AC hearing and the BCHD-position HRIRs would be more appropriate for their BCHDs than the ear-canal HRIRs.

Patient 127 (RMSE = 44.53) performed slightly worse than the normally-hearing participants in the feedback group in the $P_{BC}R_{BCHD}$ condition (RMSE = 40.54 over the first session) and participant 145 performed substantially worse (RMSE = 61.10). Participant 127 could lateralise sound well with some degree of localisation, whereas patient 145 could only lateralise with numerous errors present.

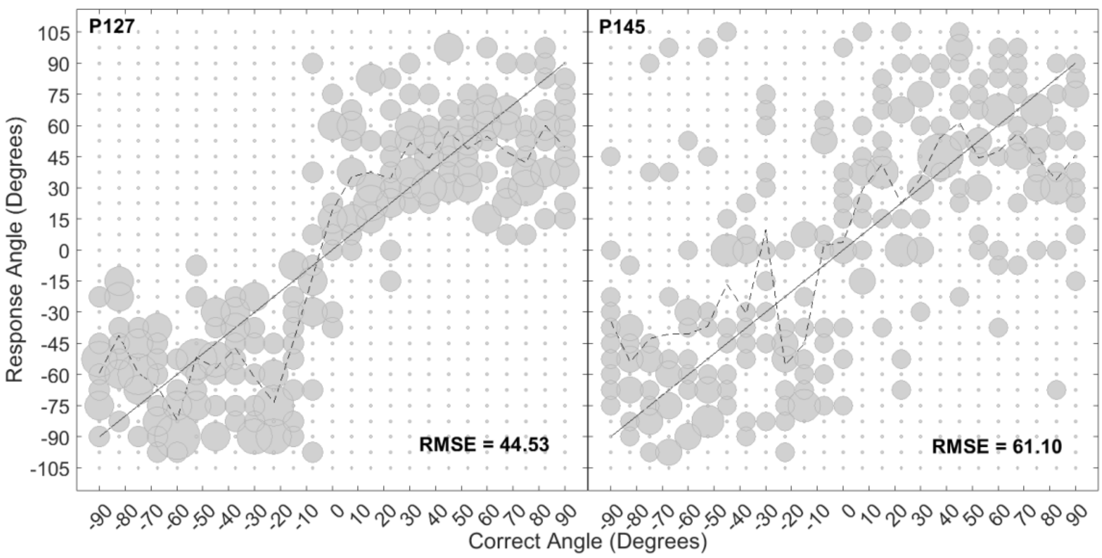


Figure 5.11. A bubble plot showing the raw results of both patients who completed the localisation using two Cochlear BAHA 5 Superpower transducers snapped onto their abutments. Bubbles indicate the patient's responses, larger bubbles indicate more responses. RMSE of both patients is also shown.

5.4 Discussion

Participants performed significantly better in the AC-presentation conditions compared to the BC-presentation conditions, however participants performed better than expected in the BC-presentation conditions compared to the data gathered from patients using bilateral BCHDs (den Besten et al., 2020; Caspers et al., 2021), and from Wang et al. (2023) who observed better noise localisation during AC presentation. The results however are more similar to MacDonald et al. (2006), who saw similar performance differences between the two sound-conduction methods. Although patients with bilateral BCHDs show high variability in their ability to localise sound, in general they can only lateralise sound with a limited degree of localisation. When you look at the individual participant data you see that, although there is a degree of lateralisation, participants are able to localise BC sound at all azimuths to an extent. Patient data is more difficult to compare to previous studies since patient studies have found a wide variety of localisation ability, but the two patients in this study showed less extreme lateralisation. Patients in Caspers et al. (2021) often perceived sounds presented in front of the patient as either left or right, whereas in this study both patients had some degree of frontal localisation

There are numerous factors which differentiate the methods of the present study and those done in patients, particularly the use of a BT instead of a BCHD. All patients in den Besten et al. (2020) and Caspers et al. (2021) were fitted with percutaneous BCHDs, so the vibration would have been driven directly into the skull bone and would have bypassed any attenuation applied by the skin (up to 20 dB according to Verstraeten et al., 2009).

A second key factor is that the normal hearing participants who took part in this study will have intact middle ears. Many patients with BCHDs would not have intact middle or outer ears which would reduce the number of pathways available for BC. The middle ear pathway is particularly important when vibration is applied at the mastoid bone (Studebaker, 1962; Dirks & Malmquist, 1969, Goodhill et al., 1970), which is the case in this study. This pathway could have resulted in increased amplification to the ipsilateral ear, aiding localisation. In addition to this, because the participant's middle ear needed to be blocked, to eliminate AC interference from the BT, the results may be prone to the occlusion effect (Reinfeldt et al., 2013). Reinfeldt et al., report an increased occlusion effect for contralateral stimulation compared to ipsilateral stimulation below 0.8 kHz. This means that the sound would have been louder at the contralateral ear than expected given the signals location which could have negatively influenced participants localisation.

Feedback was effective at improving participant's localisation in all conditions, and exposure to the task improved performance over time for BC presentation, but negatively impacted performance for AC presentaiton. Specifically, feedback and having an additional testing session improved accuracy and precision at the frontal azimuths (Figure 5.9), while leaving the extreme azimuths unchanged. This suggests that feedback can be used to somewhat correct the left/right bias that is observed in BC sound localisation (Caspers et al. 2021).

While an improvement over time is seen, there seems to be an increase in RMSE at the end of both the AC and BC conditions (Figure 5.7), which is fixed when changing perceptual modalities (two AC-presentation conditions are always followed by two BC-presentation conditions or vice versa). This is presumably caused by fatigue which is alleviated by the break that naturally occurs during the equipment swap between conditions (if AC-presentation follows BC-presentation or vice versa). The same uptick in RMSE at the end of each condition can also be seen in the feedback group (Figure 5.8), especially towards the 20th block and at the end of the study. There seems to be in initial period of improvement over the first 20-30 blocks for the feedback group that is not seen in the no-feedback group, but there is no or little improvement after that. Improvement in one condition seems to also effect improvement in the other since there is no increase in RMSE after the participant changes from AC presentation to BC presentation or vice versa. Improvement also carried over between sessions since there was no initial increase in RMSE at the start of the second session.

More improvement however was seen only for the BC-presentation group using BCHD-position HRIRs compared to the ear-canal HRIRs. This suggests that there may have been initial unfamiliarity with both the presentation style (BC) and the binaural cues (BCHD-position vs ear-canal HRIRs) which could be overcome with practice, but presentation or recording style alone did not provide enough of a detriment to be overcome by practice. A potential confound of the feedback is that it makes it obvious that the signal never comes from the 97.5° and 105° boxes and participants in the feedback condition selected them less often, decreasing over the duration of the experiment (participants in the feedback group made on average 2 overshoots per session whereas participants in the no-feedback group averaged 8 overshoots). In addition, the calibration step of both experiments (where participants sequentially listen to the sound from each angle prior to testing) could also be considered a form of feedback which could have inflated scores in the no-feedback group.

RMSEs across all conditions are higher than would be expected from previous studies. Wightman and Kistler (1989) demonstrate the efficacy of using simulated free-field sound presented over headphones and Wenzel et al (1993) showed that generic HRIRs can be used to locate sound accurately. Stevenson-Hoare et al. (2022) discuss the impact of the response method on localisation, for example pointing to posterior azimuths increases error due to the required motor movement. It is possible the on-screen user interface did not translate well into perceived sound direction, and increased participants' errors.

Future work should focus on two areas. The first is to further isolate the variables and to test improvement across sessions in a no-feedback condition to establish whether it is the feedback that causes improvement or the continued exposure to the task (although the findings from Casper et al. 2011 suggest it is the former), and to see whether improvement is seen over further sessions since BC-presentation performance did not reach AC-presentation levels in this experiment. Once this is done the experiment should be repeated in a larger patient population with bilateral BCHDs to test whether the confounds produced by a functioning middle ear, as well as the skin attenuation were essential in observing improvement over time or whether the same improvement can be seen in patients with percutaneous bilateral BCHDs. Ren et al. (2021) suggest that the asymmetries found in patients with skull deformities can significantly alter the ITDs and ILDs that a patient will experience and that this could further inhibit localisation. Additionally, Ren et al. found most participants (13 out of 24) did not improve when using bilateral 'ADHEAR' device compared to unaided, which, like the BTs used in the present study send vibrations through the skin rather than having the transducer implanted into the skull. Previous studies (e.g. den Besten et al., 2020; Caspers et al., 2021) saw improvement with most participants in bilateral compared to unilateral aids, whereas Ren et al. did not find improvement suggesting that skin transduction could have influenced localisation since this was the only obvious difference between the studies. Additionally, patients only had a short exposure period to the ADHEAR device which could have affected their proficiency with it, which may also explain some of the poor performance seen in this study since the participants only had a short exposure to the BTs.

Future work should also attempt to implement a crosstalk-cancellation filter and compare participants ability to localise with and without the filter implemented. Wang et al (2023), utilised OAEs to measure the TA from a contralateral BT. They found minor benefits for localising broadband noise (MAE = 49.17° without crosstalk cancellation, MAE = 47.67° with) and more substantial benefit for localising narrowband noise (MAE = 39° and MAE = 29.83° respectively). Other crosstalk-cancellation methodologies could be explored, such as those used in chapters 3 and 4, to determine whether localisation can be improved further, and whether it can be improved in speech localisation.

5.5 Conclusions

This study has shown that normal hearing participants are able to localise sound better when sound was presented by AC (via ER2 insert earphones) compared to BC (using 2 B71Ws), and that the type of HRIR (BCHD-position vs ear-canal) used to convolve speech had no significant effect on their localisation ability. Feedback was an effective way of improving participants' scores initially and improvement over time was seen in BC conditions. Future research should continue to attempt to develop methods for improving localisation of BC sound, including testing the improvements in clinical populations.

Discussion

6.1 Conclusions and Summary of Findings

The aim of this thesis was to explore bilateral cancellation of BC sound, determining the feasibility of a bilateral crosstalk-cancellation filter in both normal-hearing participants (with a filter designed with BCHD specifications in mind) and in patients with bilateral BCHDs.

Previous studies by Mcleod and Culling (2017, 2019, 2020) have shown that crosstalk caused by a BT could be cancelled at the ipsilateral cochlea by adjusting the phase and amplitude of a cancellation tone to create a crosstalk-cancellation filter. This filter however had only been applied to cancel the crosstalk from one BT at a time, and used a filter design that was incompatible with the processors on existing BCHDs. These filters had also been shown to lose effectiveness after the BTs were removed and refitted, presumably because small adjustments in BT positioning affect the transfer function of the contralateral and ipsilateral sound paths.

A new filter design was used that would allow the same procedures used in normal-hearing participants to be also used with patient's BCHDs. The crosstalk filters were also created and applied bilaterally and simultaneously (i.e. in the tone-reception-threshold test, the noise crosstalk was cancelled on the side that the tone was presented to, and the tone crosstalk was cancelled at the noise side). Tests in normal-hearing participants showed that a tone-reception-threshold benefit of up to 15 dB could be achieved through bilateral cancellation.

This method of crosstalk-cancellation was then tested in patients with bilateral BCHDs finding limited success due to the complicated nature of the task. A version of a task utilising automated sweeps of the phase and level was then implemented and found more success. While some patients were able to engage with the experiment and create successful crosstalk-cancellation filters, others were unable to due to a variety of reasons.

Two patients were able to create two effective left-side crosstalk-cancellation filters after an initial unsuccessful session. These filters improved TRTs in contralateral noise by 12 dB – 8.3 dB once implemented however the benefit of the filters deteriorated over time down to a minimum improvement of 2.67 dB. Right-side cancellation was less successful finding a benefit of around 2 dB in patient 145.

Localisation ability of both AC and BC presented speech was also tested in normal hearing participants. Localisation ability was similar in both conditions which replicates similar findings from MacDonald et al. (2006) but not Wang et al (2023), who found that participants more accurately localised broadband and narrowband noise when it was presented via AC. Additionally, the results contradict the findings from patient studies which consistently report poor bilateral BCHD localisation (den Besten et al., 2020; Caspers et al., 2021).

6.2 Limitation of Crosstalk Cancellation

Throughout these research projects there have been limitations in creating and applying the crosstalk cancellation filters. The most pressing is that the calibration for the filters, i.e. the time spent getting the cancellation phases and amplitudes, can be extremely time consuming. In addition to this, participants can take upwards of 8 hours to train to effectively create these filters. There is also a degree of variance in participant's ability to create these filters, with some being unable to achieve any cancellation in this experiment. With extended training they might have been able to learn to cancel crosstalk but due to the time-consuming nature of this procedure it is impractical for research. Additionally, patients with BCHDs often report problems with tonal tinnitus which can interfere with the experiment. The patient's tinnitus can make it difficult to identify if a tone has been cancelled at one ear since it can cause masking, especially if the tinnitus is at a similar perceived frequency as the cancellation tone.

There are also barriers for some participants and patients in creating these filters. Often patients with BCHDs are elderly and so their familiarity with technology may be limited. One patient in this study had never used a computer mouse before and was unable to adapt to it and interact with the app on the laptop. Some level of proficiency with one potential medium for testing (e.g. phone, pc, tablet) would be needed to do the cancellation at home. Some patients opt for physical devices for controlling their BCHDs over mobile alternatives due to difficulties using the mobile apps. Additionally learning difficulties are common in BCHD patients as a result of congenital disorders (e.g. Down Syndrome, Turner's Syndrome), which could make understanding the experiment difficult as it is a perceptually demanding task.

The issue of low-frequency cancellation is still unsolved. Low-frequency cancellation had been previously tested by Mcleod and Culling (2017, 2020) and while it was possible to cancel BC crosstalk with AC sound, below 1.2 kHz BC crosstalk could not be cancelled with BC. This would limit the effectiveness of any crosstalk-cancellation filter in everyday situations since speech contains these lower frequencies.

6.3 Future Research

Future research should aim to address these limitation as well as answer other questions that have arisen over the course of the thesis.

The most pressing matter would be to try and identify why some participants and patients are able to cancel sound with ease whereas others are unable to despite considerable training. Some patients struggle with the technological aspect of the cancellation which can be alleviated by simplifying or automating the procedure, whereas others struggle to understand the task (which may be the case with patients with cognitive impairment). Other participants seem to struggle to identify the sensory phenomenon that is indicative of cancellation especially in patients with tinnitus, which seems to be the most common barrier to successful crosstalk cancellation. Normally-hearing participants could be exposed to the sensation of perfectly cancelled sound using simulated cancellation over headphones (although the sensation isn't identical to BC cancellation). BCHD patients would require an alternative method to be exposed to this sensation prior to testing.

Since calibrating the filters is time consuming it is unrealistic for an audiologist to supervise patients during calibration at every frequency, this means that patients would need to perform the cancellation themselves. This would likely be via a mobile app that connects to a patient's BCHDs through Bluetooth. The current method would not be suitable since it requires a mouse's scroll wheel for the fine-tuning phases of the experiment. A method that sweeps through phases and levels and allows participants to press a button once the tone is most lateralised, slowly refining the cancellation phase/amplitude through sequential sweeps, would be ideal since it does not require any peripherals and is a relatively simple task. A shorter version of this was tested during this thesis. Patients who struggled with the manual cancellation preferred this method of cancellation and two patients produced good cancellation filters using it. This automated cancellation could easily be recreated as a phone app with the fine-tuning replaced by additional, more precise sweeps.

For example, patients could be given a list of frequencies that they need to cancel (e.g. between 1.5 kHz – 8 kHz), and after selecting one they are presented with the automatic cancellation task which alternates between sweeping the phase and amplitude narrowing the range of values after each sweep and slowing sweep to allow for more accurate cancellation. Once completed, a tone-reception-threshold test can be done and if successful the filters can be saved, with the option to repeat them in the future if the participant wants to improve the filters. An initial session with a clinician may still be needed to familiarize patients with the procedure but the majority of the filter creation can be done by the participants.

Low-frequency cancellation would also be useful for the crosstalk-cancellation filters to be effective in speech perception. The method used in this study is unable to cancel at low frequencies due to the small phase differences between the ears below 0.75 kHz (Mcleod & Culling, 2017). This means that when the crosstalk at the contralateral ear is cancelled, the signal at the ipsilateral ear would also be somewhat cancelled, making detecting cancellation on the contralateral side more difficult. Signal summation via a phase-matched signal could provide an alternative to boosting the power of one of the BTs (Deas et al., 2010; Mcleod and Culling 2017).

6.4 Conclusions

This thesis has shown the feasibility of simultaneous, bilateral crosstalk-cancellation in normally-hearing participants and measured the differences between localisation ability in normal hearing participants when exposed to AC and BC sound. Crosstalk cancellation was also tested in patients with bilateral BCHDs, and success was seen with cancellation in one patient, and these filters retained some effectiveness over multiple sessions. Future research should aim to test more patients to determine whether tinnitus and learning difficulties as well as other factors are barriers to calibrating the filters as well as continually testing the filters to determine whether the deterioration in cancellation efficacy is consistent across all patients. Additionally, a method of filter calibration that patients are able to complete in their own time or one that can be quickly performed in the clinic would be required to make crosstalk-cancellation a practical option for patients with bilateral BCHDs.

References

Abed, A. E., & Cain, G. (1984). The host windowing technique for FIR digital filter design. *IEEE transactions on acoustics, speech, and signal processing*, *32*(4), 683-694.

Agterberg, M. J., Hol, M. K., Van Wanrooij, M. M., Van Opstal, A. J., & Snik, A. F. (2014). Single-sided deafness and directional hearing: contribution of spectral cues and high-frequency hearing loss in the hearing ear. *Frontiers in neuroscience*, *8*, 188.

Allen, G.W, Fernandez, C (1960). The mechanism of bone conduction. Ann Otol Rhinol Laryngol. Mar;69:5-28. doi: 10.1177/000348946006900101. PMID: 13792738.

Augustine, A. M., Chrysolyte, S. B., Thenmozhi, K., & Rupa, V. (2013). Assessment of auditory and psychosocial handicap associated with unilateral hearing loss among Indian patients. *Indian Journal of Otolaryngology and Head & Neck Surgery*, *65*(2), 120-125.

Batteau, D. W. (1967). The role of the pinna in human localization. *Proceedings of the Royal Society of London. Series B. Biological Sciences*, *168*(1011), 158-180.

Batteau, D. W. (1967). The role of the pinna in human localization. *Proceedings of the Royal Society of London. Series B. Biological Sciences*, *168*(1011), 158-180.

Bauer, B. B. (1961). Stereophonic earphones and binaural loudspeakers. *Journal of the Audio Engineering Society*, *9*(2), 148-151.

Baumgartner, Wolf-Dieter; Hamzavi, Jafar-Sasan; Böheim, Klaus; Wolf-Magele, Astrid; Schlögel, Max; Riechelmann, Herbert; Zorowka, Patrick; Koci, Viktor; Keck, Tilman; Potzinger, Peter; Sprinzl, Georg. A New Transcutaneous Bone Conduction Hearing Implant: Short-term Safety and Efficacy in Children. Otology & Neurotology 37(6):p 713-720, July 2016. | DOI: 10.1097/MAO.0000000000001038

Benichoux, V., Rébillat, M., & Brette, R. (2016). On the variation of interaural time differences with frequency. *The Journal of the Acoustical Society of America*, *139*(4), 1810-1821.

Bernstein, L. R., & Trahiotis, C. (1994). Detection of interaural delay in high-frequency sinusoidally amplitude-modulated tones, two-tone complexes, and bands of noise. *The Journal of the Acoustical Society of America*, *95*(6), 3561-3567.

Bernstein, L. R., & Trahiotis, C. (2002). Enhancing sensitivity to interaural delays at high frequencies by using “transposed stimuli”. *The Journal of the Acoustical Society of America*, *112*(3), 1026-1036.

Beros, S., Dobrev, I., Farahmandi, T. S., Veraguth, D., Huber, A. M., & Röösli, C. (2022). Transcutaneous and percutaneous bone conduction sound propagation in single-sided deaf patients and cadaveric heads. *International Journal of Audiology*, *61*(8), 678-685.

Bosman, A. J., Snik, A. M., van der Pouw, C. T., Mylanus, E. A., & Cremers, C. W. (2001). Audiometric evaluation of bilaterally fitted bone-anchored hearing aids: Evaluatión audíométrica de auxiliares auditivos tipo vibrador óseo bilateral. *Audiology*, *40*(3), 158-167.

Blake, K. D., & Prasad, C. (2006). CHARGE syndrome. *Orphanet journal of rare diseases*, *1*, 1-8.

Branemark, P. I. (1977). Osseointegrated implants in the treatment of the edentulous jaw: experience from a 10-year period. *Scad J Plast Reconstr Surg*, *16*, 1-132.

Brassington, W., Parker, R., & Bianchi, F. (2023). Evaluation of the Benefits of Bilateral Fitting in Bone-Anchored Hearing System Users: Spatial Resolution and Memory for Speech. *Ear and Hearing*, *44*(3), 530.

Bregman, A. S., & Steiger, H. (1980). Auditory streaming and vertical localization: Interdependence of “what” and “where” decisions in audition. *Perception & Psychophysics*, *28*(6), 539-546.

Brimijoin, W. O. (2018). Angle-dependent distortions in the perceptual topology of acoustic space. *Trends in Hearing*, *22*, 2331216518775568.

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

Brungart, D. S. (2001). Informational and energetic masking effects in the perception of two simultaneous talkers. *The Journal of the Acoustical Society of America*, *109*(3), 1101-1109.

Brungart, D. S., & Rabinowitz, W. M. (1999). Auditory localization of nearby sources. Head-related transfer functions. *The Journal of the Acoustical Society of America*, *106*(3), 1465-1479.

Brungart, D. S., Simpson, B. D., Ericson, M. A., & Scott, K. R. (2001). Informational and energetic masking effects in the perception of multiple simultaneous talkers. *The Journal of the Acoustical Society of America*, *110*(5), 2527-2538.

Canale, A., Ndrev, D., Sapino, S., Bianchi, C., Bordino, V., & Albera, A. (2022). Speech in noise with bilateral active bone conduction implant for conductive and mixed hearing loss. *Otology & Neurotology*, *43*(9), 1000-1004.

Canale, A., Urbanelli, A., Gragnano, M., Bordino, V., & Albera, A. (2023). Comparison of Active Bone Conduction Hearing Implant Systems in Unilateral and Bilateral Conductive or Mixed Hearing Loss. *Brain Sciences*, *13*(8), 1150.

Carhart, R. (1971). Effects of stapes fixation on bone-conduction response. *Hearing measurement: A book of readings*, 116-129.

Caspers, C. J., Janssen, A. M., Agterberg, M. J., Cremers, C. W., Hol, M. K., & Bosman, A. J. (2022). Sound localization with bilateral bone conduction devices. *European Archives of Oto-Rhino-Laryngology*, *279*(4), 1751-1764.

Chan, W. X., Yoon, Y. J., & Kim, N. (2018). Mechanism of bone-conducted hearing: mathematical approach. *Biomechanics and Modeling in Mechanobiology*, *17*(6), 1731-1740.

Chang, Y., & Stenfelt, S. (2019). Characteristics of bone-conduction devices simulated in a finite-element model of a whole human head. *Trends in Hearing*, *23*, 2331216519836053.

Choi, J. E., Ma, S. M., Park, H., Cho, Y. S., Hong, S. H., & Moon, I. J. (2019). A comparison between wireless CROS/BiCROS and soft-band BCHD for patients with unilateral hearing loss. *PLoS One*, *14*(2), e0212503.

Chordekar, S., Kriksunov, L., Kishon-Rabin, L., Adelman, C., & Sohmer, H. (2012). Mutual cancellation between tones presented by air conduction, by bone conduction and by non-osseous (soft tissue) bone conduction. *Hearing research*, *283*(1-2), 180-184.

Clavier, O. H., Norris, J. A., & Dietz, A. J. (2010). A comparison of the nonlinear response of the ear to air and to bone-conducted sound. *Hearing research*, *263*(1-2), 224-232.

Cochlear bilateral market survey (2018) Bilateral benefits of the BCHD System: A summary of clinical evidence. Report Number BUN713 ISS1 DEC18. December 2018.

Corliss, E. L., & Koidan, W. (1955). Mechanical impedance of the forehead and mastoid. *The Journal of the Acoustical Society of America*, *27*(6), 1164-1172.

Culling, J. F., & Lavandier, M. (2021). Binaural unmasking and spatial release from masking. *Binaural Hearing: With 93 Illustrations*, 209-241.

Damaske, P. (1971). Head-related two-channel stereophony with loudspeaker reproduction. *The Journal of the Acoustical Society of America*, *50*(4B), 1109-1115.

Damaske, P., & Mellert, V. (1969). A procedure for generating directionally accurate sound images in the upper-half space using two loudspeakers. *Acustica*, *22*, 154-162.

David Jr, E. E., Guttman, N., & Van Bergeijk, W. A. (1959). Binaural interaction of high-frequency complex stimuli. *The Journal of the Acoustical Society of America*, *31*(6), 774-782.

Davis, A. C. (1989). The prevalence of hearing impairment and reported hearing disability among adults in Great Britain. *International journal of epidemiology*, *18*(4), 911-917.

Deas, R. W., Adamson, R. B., Curran, L. L., Makki, F. M., Bance, M., & Brown, J. A. (2010). Audiometric thresholds measured with single and dual BCHD transducers: The effect of phase inversion. *International journal of audiology*, *49*(12), 933-939.

den Besten, C. A., Vogt, K., Bosman, A. J., Snik, A. F., Hol, M. K., & Agterberg, M. J. (2020). The merits of bilateral application of bone-conduction devices in children with bilateral conductive hearing loss. *Ear and hearing*, *41*(5), 1327.

Denanto, F. M., Wales, J., Tideholm, B., & Asp, F. (2022). Differing bilateral benefits for spatial release from masking and sound localization accuracy using bone conduction devices. *Ear and Hearing*, *43*(6), 1708-1720.

Derleth P, Georganti E, Latzel M, Courtois G, Hofbauer M, Raether J, Kuehnel V. Binaural Signal Processing in Hearing Aids. Semin Hear. 2021 Aug;42(3):206-223. doi: 10.1055/s-0041-1735176. Epub 2021 Sep 24. PMID: 34594085; PMCID: PMC8463127.

Dimmelow, K. L., Fitzgerald, O. C. A., Johnson, I. J., & McKinney, C. (2003). Hear the other side-a report on Single Sided Deafness Entific. *Medical Systems Newcastle*.

Dirks, D. D., & Malmquist, C. M. (1969). Comparison of frontal and mastoid bone-conduction thresholds in various conductive lesions. *Journal of speech and hearing research*, *12*(4), 725-746.

Dobrev, I., Farahmandi, T. S., & Röösli, C. (2020). Experimental investigation of the effect of middle ear in bone conduction. *Hearing Research*, *395*, 108041.

Dobrev, I., Stenfelt, S., Röösli, C., Bolt, L., Pfiffner, F., Gerig, R., ... & Sim, J. H. (2016). Influence of stimulation position on the sensitivity for bone conduction hearing aids without skin penetration. *International Journal of Audiology*, *55*(8), 439-446.

Duda, R. O., & Martens, W. L. (1998). Range dependence of the response of a spherical head model. *The Journal of the Acoustical Society of America*, *104*(5), 3048-3058.

Dun, C. A., de Wolf, M. J., Mylanus, E. A., Snik, A. F., Hol, M. K., & Cremers, C. W. (2010). Bilateral bone-anchored hearing aid application in children: the Nijmegen experience from 1996 to 2008. *Otology & Neurotology*, *31*(4), 615-623.

Durlach, N. I. (1972). Binaural signal detection-Equalization and cancellation theory.

Dutt, S. N., McDermott, A. L., Burrell, S. P., Cooper, H. R., Reid, A. P., & Proops, D. W. (2002). Speech intelligibility with bilateral bone-anchored hearing aids: the Birmingham experience. *The Journal of Laryngology & Otology*, *116*(S28), 47-51.

Dutt, S. N., McDermott, A. L., Burrell, S. P., Cooper, H. R., Reid, A. P., & Proops, D. W. (2002). Patient satisfaction with bilateral bone-anchored hearing aids: the Birmingham experience. *The Journal of Laryngology & Otology*, *116*(S28), 37-46.

Eeg-Olofsson, M., Stenfelt, S., & Granström, G. (2011). Implications for contralateral bone-conducted transmission as measured by cochlear vibrations. *Otology & Neurotology*, *32*(2), 192-198.

Eeg-Olofsson, M., Stenfelt, S., & Granström, G. (2011). Implications for contralateral bone-conducted transmission as measured by cochlear vibrations. *Otology & Neurotology*, *32*(2), 192-198.

Eeg-Olofsson, M., Stenfelt, S., Tjellström, A., & Granström, G. (2008). Transmission of bone-conducted sound in the human skull measured by cochlear vibrations. *International journal of audiology*, *47*(12), 761-769.

Encke, J., Reimann, D., Hemmert, W., & Völk, F. (2018). On the role of interaural level differences in low-frequency pure-tone lateralization. *Acta Acustica united with Acustica*, *104*(5), 753-757.

Fan, X., Ping, L., Yang, T., Niu, X., Chen, Y., Xia, X., ... & Chen, X. (2020). Comparative effects of unilateral and bilateral bone conduction hearing devices on functional hearing and sound localization abilities in patients with bilateral microtia-atresia. *Acta Oto-Laryngologica*, *140*(7), 575-582.

Farrell, N. F., Hartl, R. M. B., Benichoux, V., Brown, A. D., Cass, S. P., & Tollin, D. J. (2017). Intracochlear measurements of interaural time and level differences conveyed by bilateral bone conduction systems. *Otology & neurotology: official publication of the American Otological Society, American Neurotology Society [and] European Academy of Otology and Neurotology*, *38*(10), 1476.

Feddersen, W. E., Sandel, T. T., Teas, D. C., & Jeffress, L. A. (1957). Localization of high-frequency tones. *the Journal of the Acoustical Society of America*, *29*(9), 988-991.

Fletcher, H., & Galt, R. H. (1950). The perception of speech and its relation to telephony. *The Journal of the Acoustical Society of America*, *22*(2), 89-151.

Fletcher, H., & Munson, W. A. (1933). Loudness, its definition, measurement and calculation. *Bell System Technical Journal*, *12*(4), 377-430.

Freyman, R. L., Balakrishnan, U., & Helfer, K. S. (2001). Spatial release from informational masking in speech recognition. *The Journal of the Acoustical Society of America*, *109*(5), 2112-2122.

Freyman, R. L., Clifton, R. K., & Litovsky, R. Y. (1991). Dynamic processes in the precedence effect. *The Journal of the Acoustical Society of America*, *90*(2), 874-884.

Freyman, R. L., Helfer, K. S., McCall, D. D., & Clifton, R. K. (1999). The role of perceived spatial separation in the unmasking of speech. *The Journal of the Acoustical Society of America*, *106*(6), 3578-3588.

García-Iza, L., Navarro, J. J., Goiburu, M., Pérez, N., & Altuna, X. (2016). Study of the improvement in bone conduction threshold after stapedectomy. *Acta Otorrinolaringologica (English Edition)*, *67*(5), 268-274.

Gardner, W. G., & Martin, K. D. (1995). HRIR measurements of a KEMAR. *The Journal of the Acoustical Society of America*, *97*(6), 3907-3908.

George, E. L., Festen, J. M., & Theo Goverts, S. (2012). Effects of reverberation and masker fluctuations on binaural unmasking of speech. *The Journal of the Acoustical Society of America*, *132*(3), 1581-1591.

Goldstein, D., & Hayes, C. (1971). The occlusion effect in bone-conduction hearing. *Hearing measurement: A book of readings*, 150-157.

Goodhill, V., Dirks, D., & Malmquist, C. (1970). Bone-Conduction Thresholds: Relationships of Frontal and Mastoid Measurements in Conductive Hypacusis. *Archives of Otolaryngology*, *91*(3), 250-256.

Grothe, B., Pecka, M., & McAlpine, D. (2010). Mechanisms of sound localization in mammals. *Physiological reviews*, *90*(3), 983-1012.

Guignard, J., Stieger, C., Kompis, M., Caversaccio, M., & Arnold, A. (2013). Bone conduction in Thiel-embalmed cadaver heads. *Hearing research*, *306*, 115-122.

Hafter, E. R., & Carrier, S. C. (1972). Binaural Interaction in Low-Frequency Stimuli: The Inability to Trade Time and Intensity Completely. *The Journal of the Acoustical Society of America*, *51*(6B), 1852-1862.

Håkansson, B., Brandt, A., Carlsson, P., & Tjellström, A. (1994). Resonance frequencies of the human skull in vivo. *The Journal of the Acoustical Society of America*, *95*(3), 1474-1481.

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

Håkansson, B., Woelflin, F., Tjellström, A., & Hodgetts, W. (2020). The mechanical impedance of the human skull via direct bone conduction implants. *Medical Devices: Evidence and Research*, 293-313.

Hamada, H., Ikeshoji, N., Ogura, Y., & Miura, T. (1985). Relation between physical characteristics of orthostereophonic system and horizontal plane localization. *Journal of the Acoustical Society of Japan (E)*, *6*(3), 143-154.

Hamann, C., Manach, Y., & Roulleau, P. (1991). Bone anchored hearing aid. Results of bilateral applications. *Revue de laryngologie-otologie-rhinologie*, *112*(4), 297-300.

Harris, G. G. (1960). Binaural interactions of impulsive stimuli and pure tones. *The Journal of the Acoustical Society of America*, *32*(6), 685-692.

Hartmann, W. M. (1997). Listening in a room and the precedence effect. *Binaural and spatial hearing in real and virtual environments*, 191-210.

Hartmann, W. M. (2021). Localisation and lateralization of sound. *Binaural Hearing: With 93 Illustrations*, 9-45.

Hartmann, W. M., & Constan, Z. A. (2002). Interaural level differences and the level-meter model. *The Journal of the Acoustical Society of America*, *112*(3), 1037-1045.

Heffner, H. E., & Heffner, R. S. (2018, May). The evolution of mammalian hearing. In *AIP Conference Proceedings* (Vol. 1965, No. 1, p. 130001). AIP Publishing LLC.

Henning, G. B. (1974). Detectability of interaural delay in high-frequency complex waveforms. *The Journal of the Acoustical Society of America*, *55*(1), 84-90.

Henning, G. B. (1974). Lateralization and the binaural masking-level difference. *The Journal of the Acoustical Society of America*, *55*(6), 1259-1262.

Henning, G. B. (1983). Lateralization of low-frequency transients. *Hearing research*, *9*(2), 153-172.

Henning, G. B., & Ashton, J. (1981). The effect of carrier and modulation frequency on lateralization based on interaural phase and interaural group delay. *Hearing research*, *4*(2), 185-194.

Hirsh, I. J. (1948). The influence of interaural phase on interaural summation and inhibition. *The Journal of the Acoustical Society of America*, *20*(4), 536-544.

Hirsh, I. J., & Burgeat, M. (1958). Binaural effects in remote masking. *The Journal of the Acoustical Society of America*, *30*(9), 827-832.

Ho, E. C., Monksfield, P., Egan, E., Reid, A., & Proops, D. (2009). Bilateral bone-anchored hearing aid: impact on quality of life measured with the Glasgow Benefit Inventory. *Otology & Neurotology*, *30*(7), 891-896.

Hodges, M. L., & McBride, M. E. (2012). Gender differences in bone conduction auditory signal processing: Communication equipment design implications. *International Journal of Industrial Ergonomics*, *42*(1), 49-55.

Hofman, P., & Van Opstal, A. (2003). Binaural weighting of pinna cues in human sound localization. *Experimental brain research*, *148*, 458-470.

Hol, M. K., Bosman, A. J., Snik, A. F., Mylanus, E. A., & Cremers, C. W. (2004). Bone-anchored hearing aid in unilateral inner ear deafness: a study of 20 patients. *Audiology and Neurotology*, *9*(5), 274-281.

Hol, M. K., Kunst, S. J., Snik, A. F., & Cremers, C. W. (2010). Pilot study on the effectiveness of the conventional CROS, the transcranial CROS and the BCHD transcranial CROS in adults with unilateral inner ear deafness. *European archives of oto-rhino-laryngology*, *267*(6), 889-896.

Homma, K., Du, Y., Shimizu, Y., & Puria, S. (2009). Ossicular resonance modes of the human middle ear for bone and air conduction. *The Journal of the Acoustical Society of America*, *125*(2), 968-979.

Huizing, E. H. (1960). Bone conduction-the influence of the middle ear. *Acta otolar*, 155.

Janssen, R. M., Hong, P., & Chadha, N. K. (2012). Bilateral bone-anchored hearing aids for bilateral permanent conductive hearing loss: a systematic review. *Otolaryngology--head and neck surgery*, *147*(3), 412-422.

Jones, G. L., & Litovsky, R. Y. (2008). Effects of uncertainty in a cocktail party environment in adults. *J Acoust Soc Am*, *124*, 3818-3830.

Jones, G. L., & Litovsky, R. Y. (2011). A cocktail party model of spatial release from masking by both noise and speech interferers. *The Journal of the Acoustical Society of America*, *130*(3), 1463-1474.

Keating, P., & King, A. J. (2013). Developmental plasticity of spatial hearing following asymmetric hearing loss: context-dependent cue integration and its clinical implications. *Frontiers in systems neuroscience*, *7*, 123.

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

Kim, N., Steele, C., & Puria, S. (2013, June). Carhart's notch: A window into mechanisms of bone-conducted hearing. In *Proceedings of Meetings on Acoustics ICA2013* (Vol. 19, No. 1, p. 050133). Acoustical Society of America.

Klumpp, R. G., & Eady, H. R. (1956). Some measurements of interaural time difference thresholds. *The Journal of the Acoustical Society of America*, *28*(5), 859-860.

Klumpp, R. G., & Eady, H. R. (1956). Some measurements of interaural time difference thresholds. *The Journal of the Acoustical Society of America*, *28*(5), 859-860.

Kuhn, G. F. (1977). Model for the interaural time differences in the azimuthal plane. *the Journal of the Acoustical Society of America*, *62*(1), 157-167.

Lavandier, M., & Culling, J. F. (2008). Speech segregation in rooms: Monaural, binaural, and interacting effects of reverberation on target and interferer. *The Journal of the Acoustical Society of America*, *123*(4), 2237-2248.

Leakey, D. M., Sayers, B. M., & Cherry, C. (1958). Binaural Fusion of Low-and High-Frequency Sounds. *The Journal of the Acoustical Society of America*, *30*(3), 222-222.

Lekue, A., Lassaletta, L., Sánchez-Camón, I., Pérez-Mora, R., & Gavilán, J. (2013). Quality of life in patients implanted with the BAHA device depending on the aetiology. *Acta Otorrinolaringologica (English Edition)*, *64*(1), 17-21.

Levitt, H., & Rabiner, L. R. (1967). Binaural release from masking for speech and gain in intelligibility. *The journal of the acoustical society of america*, *42*(3), 601-608.

Licklider, J. C. R. (1948). The influence of interaural phase relations upon the masking of speech by white noise. *The Journal of the Acoustical Society of America*, *20*(2), 150-159.

Litovsky, R. Y. (1997). Developmental changes in the precedence effect: estimates of minimum audible angle. *The Journal of the Acoustical Society of America*, *102*(3), 1739-1745.

Litovsky, R. Y., Colburn, H. S., Yost, W. A., & Guzman, S. J. (1999). The precedence effect. *The Journal of the Acoustical Society of America*, *106*(4), 1633-1654.

Liu, Y., Wang, Y., Yang, L., Zhu, J., Wang, D., & Zhao, S. (2023). Bilateral adhesive bone conduction devices in patients with congenital bilateral conductive hearing loss. *American Journal of Otolaryngology*, 103923.

Lorenzi, C., Gatehouse, S., & Lever, C. (1999). Sound localization in noise in normal-hearing listeners. *The Journal of the Acoustical Society of America*, *105*(3), 1810-1820.

MacDonald, J. A., Henry, P. P., & Letowski, T. R. (2006). Spatial audio through a bone conduction interface: Audición espacial a través de una interfase de conducción ósea. *International journal of audiology*, *45*(10), 595-599.

Mackey, A. R., Hodgetts, W. E., & Small, S. A. (2018). Maturation of bone-conduction transcranial and forehead attenuation using a measure of sound pressure in the ear canal. *International Journal of Audiology*, *57*(4), 283-290.

Makous, J. C., & Middlebrooks, J. C. (1990). Two-dimensional sound localization by human listeners. *The journal of the Acoustical Society of America*, *87*(5), 2188-2200.

McBride, M., Hodges, M., & French, J. (2008). Speech intelligibility differences of male and female vocal signals transmitted through bone conduction in background noise: Implications for voice communication headset design. *International Journal of Industrial Ergonomics*, *38*(11-12), 1038-1044.

McLeod, R. W. J., Culling, J. F., & Jiang, D. (2018). Advances in the field of bone conduction hearing implants. *Advances in hearing rehabilitation*, *81*, 24-31.

McLeod, R. W. J., Roberts, W. H., Perry, I. A., Richardson, B. E., & Culling, J. F. (2018). Scanning laser Doppler vibrometry of the cranium when stimulated by a B71 bone transducer. *Applied Acoustics*, *142*, 53-58.

Mcleod, R. W., & Culling, J. F. (2017). Measurements of inter-cochlear level and phase differences of bone-conducted sound. *The Journal of the Acoustical Society of America*, *141*(5), 3421-3429.

Mcleod, R. W., & Culling, J. F. (2019). Psychoacoustic measurement of phase and level for cross-talk cancellation using bilateral bone transducers: Comparison of methods. *The Journal of the Acoustical Society of America*, *146*(5), 3295-3301.

Mcleod, R. W., & Culling, J. F. (2020). Unilateral crosstalk cancellation in normal hearing participants using bilateral bone transducers. *The Journal of the Acoustical Society of America*, *148*(1), 63-72.

Middlebrooks, J. C. (1999). Individual differences in external-ear transfer functions reduced by scaling in frequency. *The Journal of the Acoustical Society of America*, *106*(3), 1480-1492.

Middlebrooks, J. C., & Green, D. M. (1991). Sound localization by human listeners. *Annual review of psychology*, *42*(1), 135-159.

Müller, S., & Massarani, P. (2001). Transfer-function measurement with sweeps. *Journal of the Audio Engineering Society*, *49*(6), 443-471.

Musicant, A. D., & Butler, R. A. (1984). The influence of pinnae-based spectral cues on sound localization. *The Journal of the Acoustical Society of America*, *75*(4), 1195-1200.

Nolan, M., & Lyon, D. J. (1981). Transcranial attenuation in bone conduction audiometry. *The Journal of Laryngology & Otology*, *95*(6), 597-608.

Otsuka, S., & Nakagawa, S. (2022). Improved Low-Frequency Crosstalk Cancellation in Bone Conduction Using Bone Transducers and Probe Microphone. *IEEE Access*, *10*, 79201-79212.

Otsuka, S., & Nakagawa, S. (2023a, January). Real-Time Implementation of Unilateral Crosstalk Cancellation Based on Bone Conduction. In *2023 IEEE International Conference on Consumer Electronics (ICCE)* (pp. 1-6). IEEE.

Otsuka, S., & Nakagawa, S. (2023b, July). Canceling Crosstalk Sound in Bone Conduction Hearing: An Accelerometer-Based Approach. In *2023 45th Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC)* (pp. 1-4). IEEE.

Plomp, R. (1976). Binaural and monaural speech intelligibility of connected discourse in reverberation as a function of azimuth of a single competing sound source (speech or noise). *Acta Acustica united with Acustica*, *34*(4), 200-211.

Pollard, K. A., Tran, P. K., & Letowski, T. (2015). The effect of vocal and demographic traits on speech intelligibility over bone conduction. *The Journal of the Acoustical Society of America*, *137*(4), 2060-2069.

Pollard, K. A., Tran, P. K., & Letowski, T. (2017). Morphological differences affect speech transmission over bone conduction. *The Journal of the Acoustical Society of America*, *141*(2), 936-944.

Priwin, C., Jönsson, R., Hultcrantz, M., & Granström, G. (2007). BCHD in children and adolescents with unilateral or bilateral conductive hearing loss: a study of outcome. *International journal of pediatric otorhinolaryngology*, *71*(1), 135-145.

Priwin, C., Stenfelt, S., Granström, G., Tjellström, A., & Håkansson, B. (2004). Bilateral bone-anchored hearing aids (BCHDs): an audiometric evaluation. *The Laryngoscope*, *114*(1), 77-84.

Rayleigh, L. (1907). XII. On our perception of sound direction. The London, Edinburgh, and Dublin Philosophical Magazine and Journal of Science, 13(74), 214-232.

Reinfeldt, S., Stenfelt, S., & Håkansson, B. (2013). Estimation of bone conduction skull transmission by hearing thresholds and ear-canal sound pressure. *Hearing research*, *299*, 19-28.

Ren, L. J., Duan, Y. S., Yu, J. C., Xie, Y. Z., & Zhang, T. Y. (2020). Instant auditory benefit of an adhesive BCHD on children with bilateral congenital microtia. *Authorea Preprints*.

Reuter, W. F., Marks, C., Jovanovic, S., & Gross, M. (1997). Rehabilitation in Franceschetti syndrome: an interdisciplinary approach using bone-anchored hearing aids. *Ear, nose & throat journal*, *76*(6), 402-411.

Röösli, C., Dobrev, I., & Pfiffner, F. (2022). Transcranial attenuation in bone conduction stimulation. *Hearing Research*, *419*, 108318.

Rosowski, J. J., Songer, J. E., Nakajima, H. H., Brinsko, K. M., & Merchant, S. N. (2004). Clinical, experimental, and theoretical investigations of the effect of superior semicircular canal dehiscence on hearing mechanisms. *Otology & Neurotology*, *25*(3), 323-332.

Rowan, D., & Gray, M. (2008). Lateralization of high-frequency pure tones with interaural phase difference and bone conduction. *International journal of audiology*, *47*(7), 404-411.

Sandel, T. T., Teas, D. C., Feddersen, W. E., & Jeffress, L. A. (1955). Localization of sound from single and paired sources. *the Journal of the Acoustical Society of America*, *27*(5), 842-852.

Schiano, J. L., Trahiotis, C., & Bernstein, L. R. (1986). Lateralization of low-frequency tones and narrow bands of noise. *The Journal of the Acoustical Society of America*, *79*(5), 1563-1570.

Schroeder, M. R. (1973). Computer models for concert hall acoustics. *American Journal of Physics*, *41*(4), 461-471.

Schroeder, M. R., & Atal, B. S. (1963). Computer simulation of sound transmission in rooms. *Proceedings of the IEEE*, *51*(3), 536-537.

Schubert, E. D. (1956). Some preliminary experiments on binaural time delay and intelligibility. *The journal of the acoustical society of america*, *28*(5), 895-901.

Senn, P., Kompis, M., Vischer, M., & Haeusler, R. (2005). Minimum audible angle, just noticeable interaural differences and speech intelligibility with bilateral cochlear implants using clinical speech processors. *Audiology and Neurotology*, *10*(6), 342-352.

Shaw, E. A. (1974). Transformation of sound pressure level from the free field to the eardrum in the horizontal plane. *The Journal of the Acoustical Society of America*, *56*(6), 1848-1861.

Silverman, C. A., Silman, S., Emmer, M. B., Schoepflin, J. R., & Lutolf, J. J. (2006). Auditory deprivation in adults with asymmetric, sensorineural hearing impairment. *Journal of the American Academy of Audiology*, *17*(10), 747-762.

Snapp, H. A., Morgenstein, K. E., Telischi, F. F., & Angeli, S. (2016). Transcranial attenuation in patients with single-sided deafness. *Audiology and Neurotology*, *21*(4), 237-243.

Snapp, H., Vogt, K., & Agterberg, M. J. (2020). Bilateral bone conduction stimulation provides reliable binaural cues for localization. *Hearing research*, *388*, 107881.

Snik, A. F., Beynon, A. J., van der Pouw, C. T., Mylanus, E. A., & Cremers, C. W. (1998). Binaural application of the bone-anchored hearing aid. *Annals of Otology, Rhinology & Laryngology*, *107*(3), 187-193.

Sohmer, H., & Freeman, S. (2004). Further evidence for a fluid pathway during bone conduction auditory stimulation. *Hearing research*, *193*(1-2), 105-110.

Sohmer, H., Freeman, S., Geal-Dor, M., Adelman, C., & Savion, I. (2000). Bone conduction experiments in humans–a fluid pathway from bone to ear. *Hearing research*, *146*(1-2), 81-88.

Songer, J. E., & Rosowski, J. J. (2010). A superior semicircular canal dehiscence-induced air-bone gap in chinchilla. *Hearing research*, *269*(1-2), 70-80.

Sprinzl, Georg*; Lenarz, Thomas†; Ernst, Arneborg‡; Hagen, Rudolf§; Wolf-Magele, Astrid*; Mojallal, Hamidreza†; Todt, Ingo‡; Mlynski, Robert§; Wolframm, Mario D.‖. First European Multicenter Results With a New Transcutaneous Bone Conduction Hearing Implant System: Short-Term Safety and Efficacy. Otology & Neurotology 34(6):p 1076-1083, August 2013. | DOI: 10.1097/MAO.0b013e31828bb541

Stenfelt, S. (2005). Bilateral fitting of BCHDs and BCHD® fitted in unilateral deaf persons: Acoustical aspects Adaptación bilateral de BCHD y adaptación de BCHD en sorderas unilaterales: Aspectos acústicos. *International journal of audiology*, *44*(3), 178-189.

Stenfelt, S. (2007). Simultaneous cancellation of air and bone conduction tones at two frequencies: Extension of the famous experiment by von Békésy. *Hearing research*, *225*(1-2), 105-116.

Stenfelt, S. (2011). Acoustic and physiologic aspects of bone conduction hearing. *Implantable bone conduction hearing aids*, *71*, 10-21.

Stenfelt, S. (2012). Transcranial attenuation of bone-conducted sound when stimulation is at the mastoid and at the bone conduction hearing aid position. *Otology & neurotology*, *33*(2), 105-114.

Stenfelt, S. (2015). Inner ear contribution to bone conduction hearing in the human. *Hearing research*, *329*, 41-51.

Stenfelt, S. (2016). Model predictions for bone conduction perception in the human. *Hearing Research*, *340*, 135-143.

Stenfelt, S. P., & Håkansson, B. E. (1999). Sensitivity to bone-conducted sound: excitation of the mastoid vs the teeth. *Scandinavian audiology*, *28*(3), 190-198.

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

Stenfelt, S., & Håkansson, B. (2002). Air versus bone conduction: an equal loudness investigation. *Hearing research*, *167*(1-2), 1-12.

Stenfelt, S., & Prodanovic, S. (2022). Simulation of soft tissue stimulation–Indication of a skull bone vibration mechanism in bone conduction hearing. *Hearing Research*, *418*, 108471.

Stenfelt, S., & Reinfeldt, S. (2007). A model of the occlusion effect with bone-conducted stimulation. *International journal of audiology*, *46*(10), 595-608.

Stenfelt, S., & Zeitooni, M. (2013a). Binaural hearing ability with mastoid applied bilateral bone conduction stimulation in normal hearing subjects. *The Journal of the Acoustical Society of America*, *134*(1), 481-493.

Stenfelt, S., & Zeitooni, M. (2013b). Loudness functions with air and bone conduction stimulation in normal-hearing subjects using a categorical loudness scaling procedure. *Hearing Research*, *301*, 85-92.

Stenfelt, S., Håkansson, B., & Tjellström, A. (2000). Vibration characteristics of bone conducted sound in vitro. *The Journal of the Acoustical Society of America*, *107*(1), 422-431.

Stenfelt, S., Hato, N., & Goode, R. L. (2002). Factors contributing to bone conduction: the middle ear. *The Journal of the Acoustical Society of America*, *111*(2), 947-959.

Stenfelt, S., Wild, T., Hato, N., & Goode, R. L. (2003). Factors contributing to bone conduction: The outer ear. *The Journal of the Acoustical Society of America*, *113*(2), 902-913.

Stevens, S. S., & Newman, E. B. (1936). The localization of actual sources of sound. *The American journal of psychology*, *48*(2), 297-306.

Stevenson-Hoare, J. O., Freeman, T. C., & Culling, J. F. (2022). The pinna enhances angular discrimination in the frontal hemifield. *The Journal of the Acoustical Society of America*, *152*(4), 2140-2149.

Strutt, J. W. (1907). On our perception of sound direction. *Philosophical Magazine*, *13*(74), 214-32.

Studebaker, G. A. (1962). Placement of vibrator in bone-conduction testing. *Journal of speech and hearing research*, *5*(4), 321-331.

Taschke, H., & Hudde, H. (2006). A finite element model of the human head for auditory bone conduction simulation. *Orl*, *68*(6), 319-323.

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

Tjellström, A., Lindström, J., Hallén, O., Albrektsson, T., & Brånemark, P. I. (1981). Osseointegrated titanium implants in the temporal bone: a clinical study on bone-anchored hearing aids. *Otology & Neurotology*, *2*(4), 304-310.

Toll, L. E., Emanuel, D. C., & Letowski, T. (2011). Effect of static force on bone conduction hearing thresholds and comfort. *International Journal of Audiology*, *50*(9), 632-635.

Tonndorf, J. (1966). Bone conduction studies in experimental animals. *Acta Oto-Laryngol.*, *213*, 1-132.

Tonndorf, J., & Jahn, A. F. (1981). Velocity of propagation of bone-conducted sound in a human head. *The Journal of the Acoustical Society of America*, *70*(5), 1294-1297.

Tonndorf, J., & Tabor, J. R. (1962). I Closure of the Cochlear Windows: Its Effect upon Air-and Bone-Conduction. *Annals of Otology, Rhinology & Laryngology*, *71*(1), 5-29.

Usagawa, T. (2017). Application of active control technique on a bone conduction headphone for estimating a cross-talk compensation filter. In *TENCON 2017-2017 IEEE Region 10 Conference* (pp. 3099-3104). IEEE.

Usagawa, T. (2019). Bone conduction microphone measures in the external auditory canal: An application for implementing cross-talk cancellation for bone-conducted sound. *Acoustical Science and Technology*, *40*(5), 356-359.

Usagawa, T. (2020). In-ear microphone measures in the ear canal with bone conduction stimulation: An application for estimating a cross-talk compensation filter. *Acoustical Science and Technology*, *41*(1), 439-442.

Van den Bogaert, T., Klasen, T. J., Moonen, M., Van Deun, L., & Wouters, J. (2006). Horizontal localization with bilateral hearing aids: Without is better than with. *The journal of the acoustical society of America*, *119*(1), 515-526.

Van Wanrooij, M. M., & Van Opstal, A. J. (2004). Contribution of head shadow and pinna cues to chronic monaural sound localization. *Journal of Neuroscience*, *24*(17), 4163-4171.

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

Vogt, K., Frenzel, H., Ausili, S. A., Hollfelder, D., Wollenberg, B., Snik, A. F. M., & Agterberg, M. J. H. (2018). Improved directional hearing of children with congenital unilateral conductive hearing loss implanted with an active bone-conduction implant or an active middle ear implant. *Hearing research*, *370*, 238-247.

Wang, J., Chen, Y., Stenfelt, S., Sang, J., Li, X., & Zheng, C. (2023). Analysis of cross-talk cancellation of bilateral bone conduction stimulation. *Hearing Research*, *434*, 108781.

Wang, J., Xie, S., Stenfelt, S., Zhou, H., Wang, X., & Sang, J. (2024). Spatial release from masking with bilateral bone conduction stimulation at mastoid for normal hearing subjects. *Trends in Hearing*, *28*, 23312165241234202.

Wazen, J. J., Ghossaini, S. N., Spitzer, J. B., & Kuller, M. (2005). Localization by unilateral BCHD users. *Otolaryngology—Head and Neck Surgery*, *132*(6), 928-932.

Wazen, J. J., Spitzer, J. B., Ghossaini, S. N., Fayad, J. N., Niparko, J. K., Cox, K., ... & Soli, S. D. (2003). Transcranial contralateral cochlear stimulation in unilateral deafness. *Otolaryngology--head and neck surgery*, *129*(3), 248-254.

Welsh, L. W., Rosen, L. F., Welsh, J. J., & Dragonette, J. E. (2004). Functional impairments due to unilateral deafness. *Annals of Otology, Rhinology & Laryngology*, *113*(12), 987-993.

Wenzel, E. M., Arruda, M., Kistler, D. J., & Wightman, F. L. (1993). Localization using nonindividualized head-related transfer functions. *The Journal of the Acoustical Society of America*, *94*(1), 111-123.

Wever, E. G., & Bray, C. W. (1937). The nature of bone conduction as shown in the electrical response of the cochlea. *The Laryngoscope*, *47*(1), 61-76.

Wever, E. G., & Lawrence, M. (2015). *Physiological acoustics*. Princeton University Press.

Wightman, F. L., & Kistler, D. J. (1989). Headphone simulation of free-field listening. I: Stimulus synthesis. *The Journal of the Acoustical Society of America*, *85*(2), 858-867.

Wightman, F. L., & Kistler, D. J. (1989). Headphone simulation of free-field listening. II: Psychophysical validation. *The Journal of the Acoustical Society of America*, *85*(2), 868-878.

Wightman, F. L., & Kistler, D. J. (1992). The dominant role of low-frequency interaural time differences in sound localization. *The Journal of the Acoustical Society of America*, *91*(3), 1648-1661.

Wismer, M. G., & O'Brien, W. D. (2010). Evaluation of the vibrational modes of the human skull as it relates to bone-conducted sound. *The Journal of the Acoustical Society of America*, *128*(5), 2792-2797.

Yang, X., & Grantham, D. (1997). Echo suppression and discrimination suppression aspects of the precedence effect. *Perception & psychophysics*, *59*(7), 1108-1117.

Yost, W. A., & Dye, R. H. (1997, November). Fundamentals of directional hearing. In *Seminars in Hearing* (Vol. 18, No. 04, pp. 321-344). Copyright© 1997 by Thieme Medical Publishers, Inc..

Yost, W. A., Wightman, F. L., & Green, D. M. (1971). Lateralization of filtered clicks. *The Journal of the Acoustical Society of America*, *50*(6B), 1526-1531.

Zeitooni, M., Mäki-Torkko, E., & Stenfelt, S. (2016). Binaural hearing ability with bilateral bone conduction stimulation in subjects with normal hearing: Implications for bone conduction hearing aids. *Ear and hearing*, *37*(6), 690-702.

Zurek, P. M. (1993). A note on onset effects in binaural hearing. *The Journal of the Acoustical Society of America*, *93*(2), 1200-1201.

Zwislocki, J. (1953). Acoustic attenuation between the ears. *The Journal of the Acoustical Society of America*, *25*(4), 752-759.

Zwislocki, J. (1953). Acoustic attenuation between the ears. *The Journal of the Acoustical Society of America*, *25*(4), 752-759.

Zwislocki, J., & Feldman, R. S. (1956). Just noticeable differences in dichotic phase. *The Journal of the Acoustical Society of America*, *28*(5), 860-864.