A Thesis for the degree of Doctor of Philosophy

Detection of binaural processing in the human brain

Sami Azam

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Declaration

I hereby declare that the work herein, now submitted as a thesis for the degree of Doctor of Philosophy of the Charles Darwin University, is the result of my own investigations, and all references to ideas and work of other researchers have been specifically acknowledged.

I hereby certify that the work embodied in this thesis has not already been accepted in substance for any degree, and is not being currently submitted in candidature for any other degree.
Abstract

Binaural hearing is the ability to combine the information from both ears in order to detect the location of a sound source or to distinguish a sound from background noise. It may be impaired in people who have suffered from prolonged hearing loss as children, for example due to otitis media. Currently, there is no objective method to detect binaural hearing. Auditory evoked potentials (AEPs), the brain’s response to auditory stimuli, may be used as an objective measure of hearing and auditory processing. Auditory evoked potentials can be recorded using electroencephalography (EEG), but are difficult to detect and distinguish from other brain activity due to their small amplitudes.

The aim of this research is therefore to develop, test and evaluate an approach towards building an objective methodology to detect binaural processing in the human brain using auditory evoked potentials.

To achieve this, the time averaged EEG responses of normal hearing subjects to repeated auditory stimuli were analysed. The stimuli, 500 Hz or 1000 Hz Blackman windowed pure tones, were presented as homo-phasic (the same phase in both ears) or anti-phasic (180 degree phase difference between the two ears) and mixed with various noise conditions. Auditory evoked potentials for homo-phasic and anti-phasic conditions were obtained by averaging 500 trials of in-phase and 500 trials of out-phases of each EEG epoch.
When the auditory evoked potentials were analysed, it was found that the amplitude of the dominant frequency component in the 20 - 50 Hz range of the MLR (middle latency response) of the brain was larger for the anti-phasic condition than for the homo-phasic condition for both 500 Hz and 1000 Hz stimuli. It was also noted that the normalised amplitude differences were larger when phase shifts occurred every few seconds than when phase shift only occurred once per epoch. The effect was more pronounced when the stimuli were embedded in noise, resulting in a higher mean value of the Normalised amplitude difference than for stimuli of pure tones without noise. These results are likely to relate to the psychoacoustic phenomenon known as binaural masking level difference which finds that the detection of a signal in a background of noise is easier when the signal has a different inter-aural phase difference than the noise.

The overall findings of this research indicate that the amplitude of the dominant peak in the 20 - 50 Hz frequency range of the MLR second peak may be used as a marker for binaural processing in the human brain.
List of publications

Sami Azam, Travis Brown, Mirjam Jonkman, and Friso De Boer, *An acquisition method for the MLR of auditory evoked potentials*, Biomedical Engineering and Informatics (BMEI), 4th International Conference IEEE, 2011.


Sami Azam, Travis Brown, Mirjam Jonkman, and Friso De Boer, *Effect of 500 Hz stimulus phase reversal on the 20-40Hz frequency component of the AEP*, International Conference on Biomedical Engineering and Biotechnology (BEB2011), Shanghai, China, 2011.

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<tr>
<td>ABR</td>
<td>Auditory Brainstem Response</td>
</tr>
<tr>
<td>AEP</td>
<td>Auditory Evoked Potential</td>
</tr>
<tr>
<td>BMLD</td>
<td>Masking Level Difference</td>
</tr>
<tr>
<td>CDU</td>
<td>Charles Darwin University</td>
</tr>
<tr>
<td>ECG</td>
<td>Electrocardiogram</td>
</tr>
<tr>
<td>EEG</td>
<td>Electroencephalogram</td>
</tr>
<tr>
<td>EMG</td>
<td>Electromyogram</td>
</tr>
<tr>
<td>EOG</td>
<td>Electrooculogram</td>
</tr>
<tr>
<td>ERP</td>
<td>Event Related Potential</td>
</tr>
<tr>
<td>g.DAQsys</td>
<td>Guger Technologies Data Acquisition System</td>
</tr>
<tr>
<td>g.USBamp</td>
<td>Guger Technologies USB Biosignal Amplifier</td>
</tr>
<tr>
<td>ICA</td>
<td>Independent Component Analysis</td>
</tr>
<tr>
<td>ILD</td>
<td>Interaural Level Difference</td>
</tr>
<tr>
<td>IID</td>
<td>Interaural Intensity Difference</td>
</tr>
<tr>
<td>ITD</td>
<td>Interaural Time Difference</td>
</tr>
<tr>
<td>IPD</td>
<td>Interaural Phase Difference</td>
</tr>
<tr>
<td>MLR</td>
<td>Middle Latency Response</td>
</tr>
<tr>
<td>LLR</td>
<td>Late Latency Response</td>
</tr>
<tr>
<td>PT</td>
<td>Pure Tone</td>
</tr>
<tr>
<td>SLM</td>
<td>Sound Level Meter</td>
</tr>
<tr>
<td>SPL</td>
<td>Sound Pressure Level</td>
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SNR  Signal to Noise Ratio
SVR  Slow Vertex Response
$S_0$  Signal is presented to both ears in-phase.
$S_0S_\pi$  Signal is in-phase in one ear and $180^\circ$ out-of-phase in the other ear
$S_0N_0$  Signal and noise are presented to both ears in-phase.
$S_\pi N_0$  Signal is $180^\circ$ out-of-phase in one ear and the noise is in-phase.
$S_0N_\pi$  Signal is in-phase and the noise is $180^\circ$ out-of-phase in one ear
$S_\pi N_\pi$  Signal and noise both $180^\circ$ out-of-phase to one ear compared to the other ear.
### Symbols

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Definition</th>
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<tr>
<td>dB</td>
<td>decibel</td>
</tr>
<tr>
<td>Hz</td>
<td>hertz</td>
</tr>
<tr>
<td>kHz</td>
<td>kilohertz</td>
</tr>
<tr>
<td>µV</td>
<td>microvolt</td>
</tr>
<tr>
<td>nV</td>
<td>nanovolt</td>
</tr>
<tr>
<td>ms</td>
<td>milliseconds</td>
</tr>
<tr>
<td>µsec</td>
<td>microseconds</td>
</tr>
<tr>
<td>ml</td>
<td>milliliters</td>
</tr>
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</table>
Chapter 1: Introduction

1.1 Background

The human brain is capable of receiving and processing sounds from two ears. Binaural hearing is the ability to simultaneously listen to tones and sounds with two ears [1]. Specific properties of the sound that enter both ears facilitate binaural hearing and the auditory processing in the brain. This helps to accurately detect, localise, separate and identify sound sources in space [1, 2].

Binaural hearing and auditory processing assist a listener in determining where a signal is coming from (sound localisation). This allows the listener to focus on a single sound source and ignore other sound sources and background noise, i.e. to separate sound from noise [3, 4]. It is vital for this process that both ears function correctly. If the organs of the inner ear do not correctly convert the sound to electrical signals for the brain to receive, the other stages of processing and sound interpretation will not occur [5, 6].

Sound is perceived in the auditory cortex of the brain, an area which is also directly involved in speech perception and language processing [7, 8]. Normal functioning of the auditory sense organs and the auditory pathways are prerequisites for the normal development of speech and language in
children [9] and if any of these elements does not function properly the ability to interpret sound deteriorates.

Otitis media or middle ear infection is a common disease in childhood that can adversely affect the ability to hear normally [10]. High rates of otitis media have been found in Aboriginal children [11] which if left untreated may lead to long lasting conductive hearing loss [12-14].

During the first few years of a child’s development, the brain develops rapidly. By the age of three, a child’s brain has reached nearly ninety per cent of its adult size. The growth in each region of the brain largely depends on the reception of stimulation. Auditory stimulation spurs activity in the auditory cortex and provides the necessary conditions for learning [15, 16]. Children who have otitis media for a prolonged period during the critical years of rapid brain development are therefore at risk of developing auditory and speech processing disorders, which may in turn affect their cognitive development [10].

Conductive hearing loss is hearing loss due to any condition that interferes with the transmission of sound through the outer and middle ear to the inner ear [17]. It has been suggested that the conductive hearing loss due to otitis media causes an imbalance in the sound perception in both ears [18]. As a consequence binaural hearing (hearing with both ears), i.e. the auditory processing by the brain involved in the comparison of the sounds received by one ear with the sounds received by the other ear, may be impaired. This
leads to a decreased ability to detect sounds in a noisy environment [19], which may affect the processing and understanding of speech. Binaural hearing may be permanently impaired in people who have suffered from hearing loss as children.

To understand the impact of binaural hearing on the auditory processing of the brain it is important to look at the relationship between the binaural hearing and the brain signals. Electroencephalography is a non-invasive measurement of the brain’s electrical activity. The signal that is measured is called the electroencephalogram (EEG), which is the brain’s on-going electrical activity measured from the scalp.

Event-related potentials (ERPs) are brain responses related to some "event". This event may be a sensory stimulus (such as a visual flash or an auditory sound) or a mental event (such as the recognition of a specified target stimulus). ERPs reflect the voltage changes in the on-going EEG that are related to specific events. By analysing these voltage changes it is possible to understand how different types of information are processed by the brain [20, 21].

Evoked potentials (EPs) are responses to transient physical stimuli and are a subclass of the event related potentials (ERP). These potentials are very small, in the range of 0.5 µV to 100 µV. They are usually recorded using EEG equipment.
Auditory evoked potentials (AEPs) are responses to auditory stimuli. Their low voltage combined with relatively high background electrical noise requires the use of highly sensitive amplifiers and averaging algorithms [22, 23].

One of the applications of AEPs is the evaluation of the functioning of the human hearing system. Generally, a hearing test involves the subject providing a physical response when a sound of a particular frequency is heard. Testing of the hearing system using AEPs allows for more accurate testing and is independent of an individual’s voluntary response. It is useful for testing infants or children with language disorders who may not give a reliable response. Brainstem auditory evoked potentials have been used as an effective tool in the analysis of hearing loss and to diagnose the functionality of the lower brain or the brain stem [24-26]. An objective testing facility or methodology, which can detect binaural processing carried by the upper brain stem or the auditory cortex of the brain however is not known. To address this issue, a process to determine electrophysiological evidence in the binaural processing of the brain has to be developed. This should initially be developed for subjects who can hear normally, as subjects with hearing defects may not exhibit the electrophysiological characteristics to the same extent.
1.2 Aim of Research

The aim of this research is to propose an approach towards building an objective methodology to detect binaural processing in the human brain using auditory evoked potentials.

1.3 Approach of Research

To address the aim of this research the following steps will be taken:

- Experimental facilities will be designed, tested and commissioned;
- Stimuli will be designed to evoke the event-related potentials related to binaural hearing;
- Standardized audiometric testing will be conducted to select normal hearing subjects;
- Electroencephalography which is a non-invasive measurement of the brain’s electrical activity will be carried out;
- To detect the presence of the binaural processing in the brain, auditory evoked potentials (AEPs) elicited due to binaural stimuli will be recorded and analysed. Initially the auditory evoked potentials in the EEG will be processed to minimise the influence of artifacts. Thereafter, signal characteristics will be analysed; and
- An approach to detect binaural processing in the human brain will be developed, based on the characteristics of the signals. Results will be analysed and conclusions drawn.
1.4 Structure of Thesis

This thesis is divided into six chapters. Chapter 1 is the Introduction and presents the background, aim and approach taken for this research.

Chapter 2 provides the structure and functionalities of the brain, background on electroencephalography (EEG) signals and the human ear, different types of hearing tests, hearing loss, binaural hearing, evoked potentials and artifacts.

Chapter 3 focuses on the experimental setup and the methodology used in conducting the experiments. It includes a description of the audiometric testing environment used for this research as well as subject preparation. This chapter also describes the hardware, software and other experimental facilities for conducting EEG experiments. Furthermore, it describes the EEG test protocol and the signal conditioning used for this research.

Chapter 4 starts with explaining the initial experiments conducted to validate the experimental setup. In this chapter, the preliminary experiments of this research are discussed and analysed. These include the design of the auditory stimuli used in the preliminary experiments, the development of algorithms to investigate changes in the response of the brain elicited by the different auditory stimuli and a comparative analysis of different experiments are conducted. An overall approach towards an objective way of quantifying
the changes in the auditory response, in particular the middle latency response (MLR), is analysed and presented.

Chapter 5 presents the final experiments based on the results of the preliminary experiments. Data of the final experiments are analysed and overall findings are presented.

Chapter 6 contains the conclusions followed by recommendations for future research.
1.5 Summary

Binaural hearing is the ability to simultaneously listen to tones and sounds with two ears allowing the brain to detect, localize, separate, and identify sound sources. Binaural hearing ability can be adversely affected due to number of causes such as otitis media. Currently no objective method is available which can detect the binaural processing of the brain using auditory evoked potentials. The aim of this research is to propose an approach towards building an objective methodology to detect binaural processing in the human brain using auditory evoked potentials. A step by step approach to achieve this aim is described.
2.1 EEG (Electroencephalogram)

Electroencephalography is a non-invasive measurement of the brain’s electrical activity. The nerve cell or neuron is the basic functional unit of the nervous system and, when brain cells are activated, local current flows are produced [21]. Large populations of the active neurons generate electrical activity which can be measured on the surface of the head.

Figure 2.1 A pyramidal neuron in the hippocampus [27]
In the nervous system, a synapse is a structure that permits a neuron to pass an electrical or chemical signal to another cell (neural or otherwise). Pyramidal neurons are also known as pyramidal cells having a pyramidal shaped cell body (soma) with two distinct dendritic trees as shown in Figure 2.1.

Electric potential differences are generated due to summed postsynaptic graded potentials from pyramidal cells resulting in the formation of electric dipoles between the body of a neuron and apical dendrites [27, 28]. These electric dipoles create an electric field that can be measured over the scalp using EEG systems [26]. Recording of this EEG potential is achieved by attaching two or more electrodes on the scalp which are connected to an amplifier. EEG signals are collected from different electrode locations and are significantly amplified to around 100 to 10,000 times and displayed on either a computer screen or oscilloscope appearing as waveforms, or stored on storage disks [21, 29]. During the recording of the EEG signals, the raw EEG signal may also reflect changes in unrelated electrical activity that occurs at the same time. Unwanted signals, that do not have a cerebral origin, are known as artifacts, sometimes also referred to as noise during the experiment. Common types of artifacts include power line artifacts, and artifacts related to muscle activity such as eye movement, eye blinking and movement of the head. EEG due to muscle artifacts is normally of much higher amplitude than AEPs. In order to have a good quality EEG signals for analysis it is important that artifacts should be avoided or minimized during the experiment and filtered out during the pre-processing phase [30].
2.2 History of EEG

Carlo Matteucci (1811-1868) and Emil Du Reymond (1818-1896) were the first to record electrical signals generated from muscle nerves. They introduced the field of neurophysiology. Hermann Von Helmholtz (1821-1894) introduced the concept of action current. He accurately measured the velocity of nerve conduction [34], which had been vastly overestimated up to that point in time [26]. In 1875, Richard Caton demonstrated that electrical signals in the micro-volt range can be recorded on the cerebral cortex of rabbits and dogs. In 1928, Hans Berger, who began his study of human EEG in 1920, developed a standardized methodology for the recording and visualisation of EEG signals. He recorded for the first time the electrical brain signals by attaching electrodes to the human scalp, and demonstrated the change in shape of time-varying signals from location to location on the scalp. He described the alpha rhythm, one of the major components of EEG, in 1929. The experiments conducted by Berger became the foundation of electroencephalography [31]. In 1930 Hans Berger also conducted the first EEG experiment on a sleeping subject. He also reported the effects of epilepsy, lack of oxygen and a number of diffuse and localised brain disorder [32]. During this time the research group led by Kornmuller in Berlin, built the first biomedical device for recording brain potentials. The importance of multichannel recordings, using a large number of electrodes, was first recognised by Kornmuller [33]. In 1932 the Rockefeller Foundation in the USA manufactured a differential amplifier for recording EEG signals. As the pioneers of electroencephalography in North America, Hallowell and Pauline Davis were the first in the USA to investigate the characteristics of EEG

during human sleep [34]. In 1934 Hallowell Davis used a cathode ray oscilloscope to conduct a study of peripheral nerve potentials and illustrated a good quality alpha rhythm of the brain.

The American EEG Society was founded in 1947. The First International EEG Congress was also held in the same year in London, United Kingdom. The application of EEG expanded significantly throughout the 1950s. During this time the first surgical operation for removing the epileptic foci was done. Microelectrodes with diameters of less than 3 µm were also invented for EEG measurements. In the 1960s and thereafter EEG signal analysis was further developed to analyse the brain development of full-term and premature newborns [35]. Analysis of visual evoked potentials, commonly used for epilepsy tests progressed during the 1970s. Nowadays, high end computing machines are used to record EEGs. Current EEG machines can capture data at high sampling rate and support powerful signal processing tools for data analysis. In recent years, the study of brain function has also progressed using imaging techniques such as positron emission tomography (PET), single photon emission computed tomography (SPECT) and magnetic resonance imaging (MRI). These technologies can produce two or three dimensional images with good spatial resolution. EEG recordings may be combined with other neuroimaging systems such as functional magnetic resonance imaging (fMRI). EEGs can be recorded invasively from the cortex using sophisticated needle-type electrodes to avoid attenuation and nonlinearity effects produced by the skull [34, 36].
2.3 Characteristics of EEG Signals

The electrical activity of the cerebral cortex is typically described in terms of its rhythmic activity. The unprocessed EEG is continuous in time and the amplitude of the EEG signal can vary between 40 to 100 µV with frequencies in the range of 0.5 - 100 Hz [37]. The amplitude of the EEG signal may vary based on the placement of the scalp electrodes and is also highly dependent on the potential distribution over the scalp surface. It is important to note that scalp potentials depend on the nature and the location of the underlying current sources as well as the conductive and geometric properties of the head [31]. EEG signal measurements have been accomplished over a maximum of 124 discrete locations on the scalp surface. In practice, due to time constraints and lack of hardware support, standardised recording techniques limit the number of scalp electrodes to 16 or 32 [38].

Frequency bands provide a clinically useful categorization of different rhythms. Generally, brain rhythms can easily be recognized by visual inspection of the EEG signal, but signal processing is sometimes also required to analyse the EEG signal. The amplitude and the frequency of these brain rhythms represent different cognitive functions and vary with age, health and the level of arousal [39].
The EEG signal or the brain rhythms are commonly categorized into five spectral frequency bands known as alpha, beta, delta, theta and gamma, see Figure 2.2. These frequency bands will be discussed in more detail in the next section.

Figure 2.2 Brain Rhythms with their frequency bands [34]
2.4 EEG Frequency Bands

The EEG rhythm will be described in order of their frequency bands, starting with the lowest frequency band, the delta rhythm. Table 2.1 shows the summary of the EEG rhythms along with their frequency and amplitude.

Table 2.1 EEG Amplitude and Frequency Bands

<table>
<thead>
<tr>
<th>Signal</th>
<th>Frequency</th>
<th>Amplitude</th>
</tr>
</thead>
<tbody>
<tr>
<td>Delta</td>
<td>0.5 – 4 Hz</td>
<td>Less than 200 µV</td>
</tr>
<tr>
<td>Theta</td>
<td>4 – 8 Hz</td>
<td>Less than 100 µV</td>
</tr>
<tr>
<td>Alpha</td>
<td>8 - 13 Hz</td>
<td>Less than 50 µV</td>
</tr>
<tr>
<td>Beta</td>
<td>13 - 30 Hz</td>
<td>Less than 30 µV</td>
</tr>
<tr>
<td>Gamma</td>
<td>&gt; 30 Hz</td>
<td>Less than 2 µV</td>
</tr>
<tr>
<td>Spiked</td>
<td>3 Hz</td>
<td>Less than 200 µV</td>
</tr>
</tbody>
</table>

**Delta Rhythm**

The delta rhythm has a frequency spectrum of 0.5 - 4 Hz and amplitude of less than 200 µV. These types of waves occur during sleep and infancy but may also indicate serious brain disease or cerebral damage. In addition, a number of artifacts may cause confusion with this rhythm such as artifacts caused by the large muscles of the neck and jaw. The delta waves are found in the central cerebrum, mostly in the parietal lobes [40].
**Theta rhythm**

The theta rhythm has frequencies in the range of 4 - 8 Hz and amplitudes of less than 100 µV. These waves occur during certain stages of sleep and may also be observed in children who are awake. Theta waves can be correlated with the level of arousal. Changes in the rhythm of the theta waves have been associated with access to unconscious material, creative inspiration, deep meditation, rapid eye movement (REM) sleep, brain damage and recollection [34, 41, 42]. They are mainly observed in the parietal and temporal regions.

**Alpha Rhythm**

The frequency of the alpha rhythm ranges from 8 - 13 Hz with amplitudes of less than 50 µV. Alpha waves commonly appear as sinusoid shaped signals. They usually occur in normal subjects who are awake and in a relaxed state with their eyes closed. The activity is suppressed when the eyes are open. The majority of the alpha waves are found in the posterior half of the head, usually in the occipital region of the brain [26].

**Beta Rhythm**

The frequency of the beta rhythm ranges from 13 - 30 Hz and the amplitude is less than 30 µV. Beta waves are associated with mental alertness and active concentration and occur when the subject is in a state of arousal. The majority of the beta waves can be observed on the central and frontal regions of the scalp [34].
**Gamma Rhythm**

Gamma waves have frequencies above 30 Hz (mainly up to 45 Hz) with amplitudes of less than 2 µV. Gamma waves are considered the only rhythm that can be detected in every lobe of the brain. These waves allow the simultaneous processing of information in different sections of the brain. Gamma waves may also be useful in detecting event-related synchronization of the brain [43]. The gamma rhythm can indicate the locus for finger, toe and tongue movement [34, 44].

It is important to note that the change in continuous electrical activity, while recording, depends on the level of an individual’s consciousness. It is not possible to detect any cerebral activity from a patient with complete cerebral death [45]. Cerebral death is a condition when the brain is dead and the signal captured is a flat line, refer to Figure 2.3.

Most of the above rhythms may continue up to several minutes, while others, such as the gamma rhythm, occur only for a few seconds. There are often multiple rhythms present at the same time.

**Spikes and sharp waves**

Spikes and sharp waves are also a part of EEG signals and occur in an irregular and unpredictable temporal pattern. These types of waves are often found in patients suffering from epileptic seizures [26]. A spike is differentiated from a sharp wave by its duration; a spike has duration in the range 20 - 70 ms, while a sharp wave is 70 - 200 ms long.
Sleep rhythm

Out of three functional states of the brain, (1: awake, 2: sleep without rapid eye movement (REM), and 3: sleep with REM), the sleep state can be further subdivided into three distinct stages related to the level of depth of sleep. The three distinct stages of sleep and their brain activity are illustrated in Figure 2.3. During deep sleep the brain generates a rhythm of high amplitude and low frequency whereas in light sleep the waves are smaller in amplitude with a higher frequency. Sleeping with rapid eye movement increases the brain activity causing rhythms of the highest frequencies during the sleeping state.

![Figure 2.3 Relationship of EEG Waveforms and Level of Consciousness [34]](image-url)
2.5 International 10-20 Electrode Position System

In 1958, the International Federation of Societies for Electroencephalography and Clinical Neurophysiology recommended the conventional electrode placement for 21 electrodes called the “10-20 electrode placement system” [46], as depicted in Figure 2.4.

![Figure 2.4 Labels for points according to 10-20 electrode placement](21)

This system standardized the physical placement and designations of electrodes on the scalp. Often the reference electrodes are connected to the left and right earlobes (A1 and A2). The naming of this system is coined thus as each location is either 10% or 20% of the total distance measured from the established reference points (nasion and inion) of the skull. It must be noted that percentages are employed because individuals have different
sized skulls. Furthermore, these positions are pre-defined for the acquisition and measurement of electrical activity of the brain.

Electrode placements are labelled according to brain areas: F (frontal), C (central), T (temporal), P (posterior), and O (occipital). The letters are accompanied by numbers; odd numbers (1,3,5,7) refer to the left hemisphere whereas the even numbers (2,4,6,8) corresponds to the right hemisphere as shown in Figure 2.4. Left and right are considered by convention from the point of view of a subject. It should also be noted that the 'z' is used to classify all electrode sites along the midline and that the numbers become progressively larger away from the midline [21]. Different areas of the brain have different functions, e.g. F7 is located near centres for rational activities, Fz near intentional and motivational centres, F8 close to sources of emotional impulses. The cortex around the C3, C4, and Cz locations deals with sensory and motor functions. Locations near P3, P4, and Pz contribute to perception and differentiation. Emotional processors are located near T3 and T4, while certain memory functions occur at T5 and T6. Primary visual areas can be found at O1 and O2. However it is important to mention that the scalp electrodes may not reflect these particular areas of cortex due to the non-homogenous properties of the skull, different orientation of the cortex sources, coherences between the sources, etc. With modern instrumentation, the location of a ground electrode plays no significant role in the measurement [47]. Location such as the forehead (Fpz), the hand or the leg may be used as reference [21, 48]. The combination of any active
electrode with a reference and ground electrode is considered to be the general configuration for EEG.

2.6 Applications of the EEG

The signals acquired from EEG experiments proved to be a valuable diagnostic tool in the field of clinical medicine, particularly in neurology, neurosurgery and in psychiatry. The EEG can be used to monitor the functionality of the brain. With the current advancement in computer processing, EEG signals can now be captured at a very high speed which aids in recording complex patterns of neural activity within fractions of a second after a stimulus has been delivered [26]. With the current technology it is also possible to record multiple channels of EEG simultaneously which can then be processed with advanced algorithms and signal processing techniques [21, 26].

2.6.1 Epilepsy

Electroencephalography (EEG) has become an essential tool in the study and management of epilepsy [49]. A person with epilepsy suffers from seizures where the normal patterns of the brain activity are disrupted. Epilepsy can bring changes in awareness, behaviour and sensation. People with epilepsy may suffer from muscle spasms and sometimes lose consciousness depending on the origin of the seizure and the way the different areas of the brain were triggered during a seizure. It is important to
note that the duration of seizures varies and that some seizures are very difficult to observe.

Seizures can be broadly categorized into two groups [26]: partial seizures and primary generalized seizures, depending on duration, symptoms and the location at which the seizure starts. Partial seizures start in a restricted (focal) area of the brain, while generalized seizures involve the entire brain from the onset. An EEG can be used to diagnose epilepsy and gather information about the type and location of seizure. For the purpose of diagnosing epilepsy, measurements are conducted in a dark room and the subjects are instructed to keep their movements minimal in order to reduce the impact of artifacts. A standard recording of 1 to 2 hours is considered to be sufficient for recording epilepsy. During this period two activation methods are commonly used to stimulate the patient [50]; hyperventilation where the subject needs to breathe readily and deeply and photic simulation where the subject is stimulated visually with a strobe light flashing at a rate of 1 - 25 Hz.

There are different types of EEG recordings available for a better assessment of the seizures. One involves simultaneous video recording of the patient and the EEG which allows the neurologist to correlate EEG finding to visual findings. This may improve the assessment of seizures. Another, more convenient and less expensive method, is to record the EEG during normal, everyday conditions by a small, digital recording device attached to a belt around the patient’s waist. This type of recording is known as “ambulatory EEG” and is usually done in their home for a period of 24
hours or more and therefore includes both walking and sleeping cycles [36, 51].

Automatic spike and seizure detection are the most important parameters to take into consideration in EEG signal interpretation for seizures. The design of such detection algorithms involves several signal processing considerations regarding the mathematical characterization of EEG waves and epileptic seizures [52-54]. Noise and artifacts rejection technique are another important part of the processing.

2.6.2 Sleep Disorders

Another common type of clinical issue where the EEG plays a significant role is the analysis of sleep disorders. Sleep disorders are quite common and may be caused by medical or psychological conditions. Sleep disorders can be broadly be classified into categories known as insomnia, hypersomnia, circadian rhythm disorders, parasomnia and sleep apnea-hypopnea syndrome (SAHS) [26].

The effect of insomnia can be characterized by difficulty falling asleep or staying asleep, waking up too early or having poor quality sleep [55].

Hypersomnia is a disorder that causes excessive sleep. People suffering from hypersomnia may sleep for more than 10 hours per day and they may sleep very deeply. An example of hypersomnia is narcolepsy. Narcolepsy is a
disorder that is characterized by uncontrollable daytime sleep attacks while night-time sleep is fairly normal [36, 56].

Circadian rhythm disorders are a family of sleep disorders affecting the timing of sleep. Humans have biological rhythms, known as circadian rhythms, which are controlled by a biological clock and work on a daily time scale. Change in body temperature, alertness, appetite, hormone secretion and sleep timing are affected due to this circadian clock. The biological clock is affected by a change in life routine which in turn affects the sleep-wake schedule and disappears within a week [57].

Parasomnia describes disruptive sleep-related disorders. They are characterized by undesirable physical or verbal behaviour or experiences. Parasomnia occurs in association with sleep, specific stages of sleep, or sleep-aware transitions [58].

The Sleep Apnea-Hypopnea Syndrome (SAHS) is another type of sleeping disorder that can be categorised as:

- Obstructive Sleep Apnea-Hypopnea Syndrome (OSAHS) where the patient suffers complete or partial obstruction of the upper airway (UA) during sleep. Patients might have partial or partial blocking of the throat (pharynx or UA) which leads to throat vibration during sleep and consequently snoring [59]. In addition, breathing is reduced and even stops for up to a minute or more causing drop in the oxygen level. This
might be followed by a short interruption to sleep for up to 3 seconds allowing breathing to start again resulting in disruption in sleep. These episodes of interruption might happen multiple times during sleep and the subject should be aware that this is a medical condition which requires medical attention [60].

- Central Sleep Apnea Syndrome (CSAS) where the patient suffers central respiratory abnormalities. This is caused by a decrease or instability of the central ventilator drive. CSAS is a result of a disease affecting the area of the central nervous system which control breathing and occurs mostly in the non-REM stage [61].

EEG signal patterns and characteristics can be used to classify the different types of sleep disorder. In order to properly diagnose sleeping disorders it is important to quantitatively determine how the patterns of sleep change. This type of information is collected by having the patient stay overnight with electrodes attached to the scalp and other parts of the body. Physiological signals such as the electrooculogram (EOG), electromyogram (EMG) and electrocardiogram (ECG) are also recorded to diagnose sleep disorders. Polysomnography is a diagnostic test during which a number of physiological variables are measured and recorded during sleep. Several algorithms have been developed for analysing sleep disorders [26].
2.6.3 Other EEG Applications

Apart from assessing epilepsy and sleeping disorders, the EEG can be clinically used for the following [21, 62]:

**Brain death**

Brain death is the clinical diagnosis of irreversible total loss of brain function. Brain death is characterised by electrocerebral inactivity of the EEG.

**Dementia**

Dementia is a syndrome that shows the decline in intellectual and cognitive abilities. It affects the normal social activities and the relationship and interaction with other people [63]. The effect of dementia is often studied with the help of the EEG. In most cases, such as in primary degenerative dementia like Alzheimer’s disease, the delta and theta wave activities increase whereas the activity of the alpha rhythm EEG decreases [26, 34].

**Head injury**

EEG can be used to locate areas of damage following head injury, stroke or a tumor. Symmetry of alpha activity within hemispheres can be monitored. In cases of restricted lesions such as a tumour, a hemorrhage, or trombosis, the cortex normally generates low frequency waves [21].
**Effects of drugs**

Drugs cause significant changes in the EEG signal patterns especially when using for the treatment and suppression of various mental and central nervous system abnormalities. Therefore it is important to know the effects of these drugs that causes changes to EEG waveforms and their patterns [26, 34]. The effect of administration of drugs for anaesthesia on EEGs is of interest to clinicians. Several studies have attempted to find the correlation between the EEG changes and the stages of anaesthesia [64]. It is found that during the initial stage of anaesthesia a high frequency frontal activity occurs in the EEG. This activity gets slower with greater depth of anaesthesia, and generate a high amplitude, low frequency EEG signal. In cases of acute intoxication, the EEG patterns are similar to those of anaesthesia intoxication [26, 65].

**Brain-Computer interface**

The design of a brain-computer interface is another EEG application. It enables a subject to communicate and control an external device through EEG activity [66]. This is an active area of research and under this type of mechanism messages are conveyed by the brain rather than using physical movement [5, 67].
2.7 The Brain

The human brain is the most complex part of the human body. Practically it controls all the other body’s parts and their functions such as body temperature, blood pressure, heart rate and breathing, processing of sensory information and the control of physical motions. It constantly processes streams of sensory data which makes the human aware of the surrounding. [68]. During the first few years of life the development of the brain takes place at a very rapid rate. Stimulation causes activity in different regions of the brain and helps the brain to grow [16, 69]. Before moving into the details of the brain structure and its functions, an overview of the brain development is given in the next section.

2.7.1 Overview of Brain Development

The elementary building block of the brain is the nerve cell or neuron. By the time a baby is born it has all the neurons it will ever have. There are about 80 to 120 billion [68, 70] neurons that serve the function of gathering and transmitting the electrochemical signals. Neurons start to develop during foetal development and they migrate to form various parts of the brain. Their migration into forming different parts of the brain differentiates them and they begin to specialize in responding to chemical signals [71, 72]. By the age of 3, a baby’s brain has reached almost 90% of its adult size. Each region of growth needs training in developing their functionalities and largely depends on receiving stimulation, which spurs activity in that region [16]. This
stimulation provides the foundation for learning. Brain development or learning is the process of building links or connections between the neurons, called synapses. The different experiences that a child encounters will contribute to the formation of neural circuits or groups of neurons that communicate with each other. Organization of the brain is done by forming pathways to connect different parts of the brain through synapses [71] and these different parts of the brain are engaged in different functions discussed further in Section 2.7.2. The development of synapses occurs very quickly during children’s early years, in response to the young child’s experiences. At its peak, the cerebral cortex of a healthy toddler may create 2 million synapses per second [73]. By the time children reach three years of age, their brains have approximately 1,000 trillion synapses [69].

2.7.2 Structure and Functionalities of the Brain

The brain consists of several major components: the cerebrum, cerebellum, diencephalon and brain stem. It has a weight of approximately 1.5 kg with an average size of 1260 cubic cm for males and 1130 cubic cm in females. About 80 to 120 billion [68, 70] neurons serve the function of gathering and transmitting electrochemical signals. Figure 2.5 shows the composition of the brain.
The cerebral cortex is the outermost layer of the cerebrum and has a thickness of around 2 to 3 mm. It is associated with complex functions such as sensations, voluntary actions, reasoning, planning and problem solving [36]. The cortex consists of two symmetrical hemispheres which are separated by the deep sagittal fissure. The cortex of each cerebral hemisphere is divided into four lobes known as frontal, parietal (middle rear), temporal (side), and occipital (rear) as shown in Figure 2.6 [75]. The cerebral cortex consists of grey matter which forms the surface layer of the cerebrum, whereas the white matter composed of myelinated fibres lies under the cortex and occupies the majority of the cerebrum mass [76].
The Cerebrum constitutes of around 85% of the human brain. It is involved in the interpretation of sensory impulses, in controlling voluntary motor responses and in intellectual processes [75]. In general the cerebrum has three major types of functional areas: sensory, motor, and association areas. Sensory areas, which can be found in several cerebral lobes, transform the sensory receptors’ impulses into sensations. For example, the occipital lobe is mostly used for visual processing, and is therefore often called the visual cortex. Occipital lobes are used for processing images from the eyes and link that information with images stored in memory. Language recognition, spatial visualization, and the processing of sound are examples of functions of the temporal lobes [78]. Areas interpreting sensations from skin stimulation and sensory areas for taste are located in the parietal lobes of the brain. The sensory areas for smell are located in the inferior part of the frontal lobe. Ascending sensory fibres cross over from one side to the other in the spinal
cord or brain stem. Thus, sensory areas in the left cerebral hemisphere receive impulses from the right side of the body, and vice versa [76].

Motor areas, located in the frontal lobe, have two sections. One helps control skeletal muscles and the other is involved with learned activities, such as writing, problem solving, and planning. Descending motor fibres cross over from one side to the other in the brain stem. Thus the left side of the cerebrum controls skeletal muscles on the right side of the body, and vice versa [76].

Association areas, which play critical roles in the interrelationships of sensations, memory, will and the coordination of motor responses, interrelate sensory inputs and motor outputs and occur in each cerebral lobe [79].

Higher cognitive functions and many complex motor activities depend on the cortex, and it is responsible for the perception and conscious understanding of all sensations.

The diencephalon, although small in size, plays an important role. It lies between the brain stem and the midbrain and consists of two major components: the thalamus and the hypothalamus.

The thalamus consists of two lateral masses of neural tissue that are joined together by a narrow isthmus of neural tissue called the intermediate mass. Sensory impulses coming from the lower regions of the brain and the spinal cord are first received by the thalamus. The thalamus provides a general, but
nonspecific, awareness of sensations such as pain, pressure, touch and temperature. It is believed that it associates sensations with emotions, although it is the cerebral cortex that interprets the precise sensation [80]. The thalamus also serves as a relay station for motor impulses descending from the cerebral cortex to lower brain regions [76].

The primary function of the hypothalamus is to control all the internal systems of the human body, and this is accomplished through its regulation of body temperature, mineral and water balance, appetite and digestive processes, heart rate and blood pressure, sleep and wakefulness, emotions of fear and rage, and secretions of hormones by the pituitary gland. It is located inferior to the thalamus and anterior to the midbrain. It communicates with the thalamus, cerebrum, and other parts of the brain [78]. The hypothalamus is the major control centre of the autonomic nervous system. The hypothalamus is also the connecting link between the brain and the endocrine system, which produces chemicals (hormones) that affect every cell in the body [80].

In evolutionary terms, the **brain stem** is the oldest part of the brain. It is the posterior part of the brain and connects the cerebrum with the spinal cord. Motor and sensory neurons travel through the brainstem allowing for the relay of signals between cerebrum and the spinal cord. The brain stem consists of the mid-brain, pons, and medulla oblongata [76] (see Figure 2.7).
The midbrain occupies the most superior portion of the brainstem and is located posterior to the hypothalamus and superior to the pons. It contains reflex centres for head, eye, and body movements in response to visual and auditory stimuli [82]. The pons lies between the midbrain and the medulla oblongata and is recognizable by its bulb-like anterior portion. It consists primarily of nerve fibres. Longitudinal fibres connect lower and higher brain centres, and transverse fibres connect with the cerebellum. With the medulla oblongata it controls the rate and depth of breathing. The medulla oblongata is the most inferior portion of the brain and it is the connecting link with the spinal cord. Ascending and descending fibres extending between the brain and the spinal cord cross over to the opposite side of the brain within the
medulla. The medulla contains three control centres that are vital for homeostasis. The respiratory control centre works with the pons to regulate the rate and depth of breathing. It is also involved in associated reflexes such as coughing and sneezing. The cardiac control centre regulates the rate of heart contractions and the vasomotor centre regulates blood pressure and blood flow by controlling the diameter of blood vessels [83].

**Cerebellum:** Behind the brainstem, at the base of the brain, lies the cerebellum, which is the second largest component of the brain. The cerebellum contains hundreds of millions of neuron for data processing and relays information between areas of the cerebral cortex that are involved in motor control. It may also be involved in cognitive functions such as attention and language and in regulating fear and pleasure responses [76, 84], but movement related functions are the main role. The next couple of sections (2.8 and 2.9) of this chapter deals with the physiology of hearing and describes how the sound is traversed from the outer ear to the inner ear and then finally processed by the brain.
2.8 The Human Ear and Hearing

2.8.1 Sound

In order to understand the physiology of hearing a basic understanding of the physics of sound is required. A sound is a wave that is introduced in a medium by a vibrating object. The medium can be air, water or solid material and the vibrating object can be anything that has the ability to cause vibrations in the particles or the molecules of the medium. When vibrating air molecules hit our ear drum we can hear sounds. A tuning fork sending out a sound wave is a classic example as shown in Figure 2.8 and 2.9 [85].

Figure 2.8 Tuning Fork (Phase: a – Neutral Position and Phase: b – Prong move towards right) [85]
The vibrating tuning fork sends out a travelling pressure wave which causes the air molecules to vibrate around their mean positions. The waves created due to this action are classified as longitudinal waves that follow special rules of propagation. As shown in Figure 2.8 (a) when the tuning fork is in the neutral position it does not cause any disturbance so no differences can be seen in the movement of the air particles.

As the prong ‘A’ moves towards right it compresses the air particles near it forming a wave of compression as shown in Figure 2.8 (b). This moves forward due to vibrating air layers. When the prong ‘A’ goes back to its original position the pressure on its right decreases and forms a wave of rarefaction.

As the tuning fork continues to vibrate, waves consisting of alternated compressions and rarefactions spread in air as shown in Figure 2.9 (d) and the direction of motion of the sound waves is same as that of air particles.
Some important attributes need to be considered in a sound wave. One is the frequency of the waves, measured in cycles per second or Hertz (Hz). High-pitched sounds have higher frequencies than lower-pitched sounds. The second important attribute of the wave is its amplitude or intensity, which is related to the magnitude of the movements produced. The higher the amplitude of the wave the louder the sound is. Another important attribute of the wave is the phase or the time difference between the waves coming to the ear. Amplitude and phase are the two important cues for sound detection and localization.

### 2.8.2 The Ear

Human ears are sensitive detectors capable of detecting the fluctuations in air pressure that hit the eardrum. The human ear is capable of detecting sound waves with a wide range of frequencies, ranging from approximately 20 Hz up to 20000 Hz [86]. Sound waves with frequencies of less than 20 Hz are known as infrasound and are below then human audible range whereas sound waves of more than 20,000 Hz are known as ultrasound and are above the human audible range. The human ear, like that of other mammals, also plays a vital role in the sense of balance and body position. Although the ear is the sense organ that recognizes sound, it is the brain and the central nervous system that process the sound. Sound waves are perceived in the auditory system of the brain. The ear acts as a device for collecting sound and then behaves as a transducer converting sound (mechanical energy), to a nerve impulse which is then transmitted to the brain [5]. The ear's ability to
do this allows us to perceive the pitch of a sound depending on the sound wave's frequency, the loudness of a sound which is related to the wave's amplitude and the timbre of the sound caused by various frequencies which make up a complex sound wave [87]. Anatomically the ear has three distinguishable parts known as outer ear, middle ear and the inner ear. Figure 2.10 shows the major components of the outer, inner and the middle ear.

Figure 2.10 Anatomy of the Human Ear [88]
2.8.2.1 The Outer Ear

![Outer Ear Diagram]

Figure 2.11 Composition of the Outer Ear [89]

The outer ear is the most external portion of the ear and consists of a partially cartilaginous flange called the pinna, which includes a resonant cavity called the concha. In addition there is an ear canal through which sound waves pass to the ear drum. The ear drum, also called the tympanic membrane, is a barrier that separates the outer ear from the middle ear [90]. In detecting the incoming sound, the outer ear plays a very important role. It influences the resonances of the sound on the tympanic membrane and it provides directionality cues that help in sound localization.

The pinna is angled in such a way that it can capture sound from the front more than from behind, which aids in localizing sound. In localizing the sound the most important cues are differences in intensity and phase of the sound waves in the two ears. This will be discussed in more detail in Section 2.10.

The raised ridges of the pinna and the concha reflect sound waves into the ear canal in a way that depends on the direction and elevation of the sound.
source [91]. The complex shape of the pinna causes multiple reflections which contribute to detecting the source of the sound using both the azimuth (direction in the horizontal plane) and the elevation of the source [92, 93]. The directional selectivity in the reception of the sound is increased when the wavelength of the sound is shorter than the dimension or the area of the pinna.

Once the external ear collects sound waves with the aid of the pinna and concha, the sound is channelled down through the auditory canal to the ear drum. The dimensions of the auditory canal, which is a slightly bent tube of approximately 2.5 centimetres long and about 7 millimetres in diameter are important in determining the auditory canal's resonant frequency, which is known as the frequency at which air in the canal is most easily set into motion. The human auditory canal has a resonant frequency around 2.5 KHz. As a consequence sound containing frequencies near or equal to the resonant frequency will have a gain at around 10 dB [94], which is one reason why the sensitivity to sounds is best around this frequency. This sound pressure then comes in contact with the ear drum or tympanic membrane. The tympanic membrane is a thin cone-shaped membrane which separates the external ear from the middle ear in humans and transmits sound from the air to the ossicles inside the middle ear.
2.8.2.2 The Middle Ear

The middle ear is an air filled space connected to the back of the nose with the help of a long thin tube called the Eustachian tube. The Eustachian tube is usually closed because the air in the middle ear chamber must be completely still for the optimal vibration of the middle ear bones. Occasionally, when we yawn or sneeze, it opens briefly to allow an exchange of air. This equalizes the air pressure in the middle ear to the air pressure outside. In the middle ear there are three small bones which connect the ear drum and the oval window. These bones transmit sound from the middle ear to the inner ear.

The first bone, the hammer, also known as the malleus, is attached to the lining of the eardrum and the other end of the hammer is tightly bound by ligaments to the second bone, the anvil or incus. The anvil is secured to the
stirrup or stapes, the third bone, whose footplate is in turn anchored against the oval window, as shown in Figure 2.12.

The inner mechanism of the middle ear transfers vibrations from the large eardrum down to a small flexible window in the wall of the cochlear, known as the oval window and the sound energy is effectively amplified. Once the tympanic membrane vibrates, it pushes the tip of the malleus and, since the first two bones are joined comparatively rigidly, the bones rotate together and transfer the force via the stapes to the oval window [90, 96].

Other than transferring acoustic energy from the external ear to the inner ear another important function of the middle ear is to control the sound energy level. In the middle ear, the volume of sound is increased or decreased as needed. If the sound is relatively soft, the ossicles vibrate rapidly to boost the volume. But if the incoming sound is very loud, two muscles attached to the ossicles in the middle ear lower the amplitude of vibrations by contracting. These muscles are called the tensor tympani and the stapedius [97]. As they contract, they pull on the three bones, reducing their ability to vibrate. This action helps prevent loud noises from damaging the ears.

2.8.2.3 The Inner Ear

The inner ear has a series of complicated passageways and contains the organ which is responsible for hearing, the cochlear. It also has a sense organ that is attuned to the effects of gravity. The cochlear which starts at the
oval window has a shape that looks like a snail shell as shown in Figure 2.13. The volume of cochlear is about 0.2 ml and the cochlear contains hundreds of specialised cells that are attached to the nerve fibres. Tiny hairs line the curves of the cochlear [90].

Figure 2.13 Cross-sectional view of the cochlear [98]

Figure 2.13 shows that the cochlear consists of three chambers which are separated from each other by membranes. The scala vestibuli and the scala tympani are continuous with each other at the apex of the cochlear. The cochlear duct extends nearly to the apex of the cochlear and is separated from the scala vestibule and the scala tympani by two special membranes as shown in Figure 2.14. The vestibular membrane separates the cochlear duct and the scala vestibule whereas the basilar membrane separates the cochlear duct and the scala tympani [76, 99].
The vestibular and the tympanic canals are continuous with each other and they contain the same fluid, which is similar in composition to spinal fluid. On the other hand the cochlear duct, which is running in between the two canals vestibular and tympanic, contains fluid of a different chemical composition. This chemical difference between the two fluids plays a crucial role in the initial stage of hearing process [80, 99]. The organ of Corti containing the sensitive receptor cells for sound stimuli lays on the upper surface of the basilar membrane within the cochlear duct as shown in Figure 2.15. The receptor cells of the organ of Corti, known as hair cells, are mechanoreceptors and transform the pressure waves in the cochlear into receptor potentials [101]. The inner pressure waves in the cochlear cause the basilar membrane to vibrate and the movement of the basilar membrane stimulates these hair cells.
Once the basilar membrane vibrates it causes the hairs of the hair cells to contact the tectorial membrane, which stimulates the formation of action potentials by the hair cells. The cochlear branch of the vestibulocochlear (VIII) nerve then transmits the signal to the auditory centres of the temporal lobes of the cerebrum, where the sensation is interpreted. Some of the nerve fibres cross over to the opposite side of the brain so that the auditory cortex in each temporal lobe interprets impulses originating in each ear [76].

The vestibular system is another component of the inner ear. It responds to changes in position and movement of the head. Hairs cells are also found in the vestibular system, which detects these changes. Figure 2.16 shows that the vestibular system consists of three membranous semicircular canals and two saclike swellings, the utricle and saccule, all of which lie inside the tunnels of the temporal bone on each side of the head.
Figure 2.16 Composition of the Vestibular System [103]

The semicircular canals contain membranous labyrinths which contains the receptors that are used to detect the motion of the head. The semicircular canals are positioned in such a way that they can detect angular acceleration during rotation of the head along three perpendicular axes. The three perpendicular axes can be thought of as corresponding to nodding the head up and down, shaking the head from side to side, and tipping the head so the ear touches the shoulder. The receptor cells of the semicircular canals contain hair like stereo-cilia which are enclosed by a gelatinous mass known as cupula. The cupula extends across the lumen of each semicircular canal at the ampulla which is a slight bulge in the wall of each duct.

Due to movement of the head, the semicircular canal within its bony enclosure of the bony labyrinth and the attached bodies of the hair cells move. In addition to that, the fluid filling the duct also vibrates but due to inertia it tends to retain its original position [83]. Thus the moving ampulla is pushed against the stationary fluid, which causes the bending of the hairs of
the hair cells, which then stimulates the formation of impulses carried to the brain via the vestibular branch of the vestibule cochlear (VIII) nerve [76].

2.9 Auditory Pathways

The cochlear converts sound energy (pressure waves) into neural impulses. This neural information is carried out in the inner ear by the auditory nerve, which consists of approximately 30,000 nerve fibers (individual axons) arising from each ear. The auditory nerve branches into several different pathways that eventually go to the auditory cortex. Various pathways seem to be specialized for processing different aspects of auditory information [104]. The auditory system contains specialized neural analysers for locating and identifying sound sources within the auditory environment. Auditory information from the ear is sent to the auditory nerve, cochlear nucleus, superior olivary complex, inferior colliculus, medial geniculate (midbrain) and then to auditory cortex for further processing, as shown in Figure 2.17.

Figure 2.17 Illustration of human auditory pathway and evoked potential loci of origin [105]
The auditory nerve carries the signal to the brainstem and synapses in the cochlear nucleus. From the cochlear nucleus, auditory information is split into two streams: the ventral stream and the dorsal stream.

![Auditory Pathways](image)

**Figure 2.18 Auditory Pathways [106]**

Auditory nerve fibres connect to the ventral cochlear nucleus synapse with giant, hand-like terminals, as shown in Figure 2.18. The ventral cochlear nucleus cells then project to a collection of nuclei in the medulla, called the superior olive. It is believed that the superior olive is the region where the difference in timing and the intensity of the sound in each ear are compared and these are the two binaural cues that aid in localizing the sound [106, 107]. The superior olive then projects up to the inferior colliculus via a fibre tract called the lateral lemniscus.

The second stream of the auditory information starts in the dorsal cochlear nucleus. The dorsal cochlear nucleus is believed to differentiate the sounds, detecting differences in frequency [108]. The complex circuitry of the dorsal
cochlear nucleus takes part in analysing the quality of the sound. This pathway also projects directly to the inferior colliculus, via the lateral lemniscus.

![Auditory Pathways to auditory cortex](image)

**Figure 2.19 Auditory Pathways to auditory cortex [106]**

From the inferior colliculus, both streams of information proceed to the sensory thalamus. The medial geniculate nucleus is the auditory nucleus of thalamus. The medial geniculate projects to the primary auditory cortex located in the temporal lobes as shown in Figure 2.19. This is the part of the brain where the sensory information is processed and sound is perceived.
Signals at the cortical region are much larger in amplitude than signals in other regions of the skull [7].

### 2.10 Binaural Hearing

The term binaural hearing refers to the auditory processing involved in the comparison of the sounds received by one ear with the sounds received by the other ear (the word "binaural" means "two ears"). Binaural hearing enables us to find the direction of a sound source and to determine the meaning and contents of those sounds [109, 110]. These tasks are relevant to children who spend many hours in noisy environments such as classrooms and adults who have to operate in similarly complex situations in the workplace or in their everyday interactions. A common problem, known as the cocktail party effect, was first described by Colin Cherry in 1953. It is the ‘ability to focus one’s attention on a single speaker among a cacophony of conversations and background noise’ [3]. The listener processes and “tunes in” or “locks in” to one particular conversation, thus cancelling out and treating the remaining other conversations as noise. People with normal hearing are equipped with basic auditory mechanisms that make it possible to do so [111, 112]. Their normal binaural hearing ability allows the auditory system to work out the direction and distance of sound sources and to detect certain sounds at much lower intensity levels than if only one ear could be used [19]. In other words, when sound reaches the ears from a particular location in space, certain properties of the sound facilitate the binaural hearing system to accurately detect, localize, separate and identify sound
sources. In addition to that, if the signals are contaminated or masked with noise the threshold of signal detection can sometimes be lower when listening with two ears rather than one. This is demonstrated by the phenomenon of the binaural masking level difference, which will be discussed in detail in section 2.10.2. When listening in a noisy environment to relevant information the listener faces two distinct challenges: analysis and synthesis of sound and detecting the direction of sound.

**Analysis and Synthesis**

The analysis process involves segmentation of an incoming auditory signal to individual channels or streams [113]. Sounds in an auditory scene are all summed together to generate the signal that enters the ear. This mixture of sound contains unnecessary information which is of no use to the listener. The auditory system filters and the synthesis process involves the reconstruction of individual sound waveforms from the separated sound streams.

**Direction of the Sound**

The second challenge is that of directing attention to the sound source of interest while ignoring others, and of switching attention between sources, as when intermittently following two conversations. Most of our cognitive processes can operate only on one source at a time, so we typically select a particular sound source on which to focus [114].
Given one sound source, the two ears receive slightly different sound patterns due to a finite delay produced by their physically separated locations. Slight differences in the acoustic cues, such as interaural time difference (ITD) and interaural intensity difference (IID) enable the brain to identify the location and direction of the incoming sound waves. The threshold of detecting a signal masked in noise can also be affected by these cues [115-118].

2.10.1 Binaural Cues

In the first stage of processing the auditory chain, information from the two ears remains segregated. Sound delivered to just the left ear (stimulation that is monaural, meaning “one ear”) will activate cells in the left cochlear nucleus, but has no effect on those in the right cochlear nucleus. Exactly the opposite is true for monaural stimulation delivered to the right ear. However, at processing stages beyond the cochlear nucleus (superior olive, inferior colliculus, and higher), the auditory system becomes binaural (“both ears”), meaning that these later stages receive neural input from the left ear and the right ear. Some neurons can be activated by sound presented to either ear; others may be activated by sound presented to one ear but inhibited by sound presented to the other ear [119]. The binaural cells in the brain register information about the location of a sound source relative to the head. To understand how these binaural cells function, we first need to be familiar with the two main cues available for sound localization. In both of these cues, the sounds received by the two ears are compared, which is why the cues are
called binaural cues. When a person hears sounds in the environment there are two important tasks that the auditory system accomplishes; determining what the person hears and where the sound is coming from [120].

Sound localization refers to the process of determining the location or source of a sound. Specific properties of the sound, such as intensity, spectral, and timing cues, that enter both ears facilitate binaural hearing and the auditory processing of the brain [121]. When the sound reaches the ears from a particular location in space, the shape of the listener’s head plays a vital role in determining the source and the direction of the sound. The shape of the head and its properties of sound enable the auditory system to create a set of important acoustic markers. In the horizontal plane the localization of the sound sources predominantly depends on the inter-aural time difference (ITD) and the inter-aural level difference (ILD) [122].

**Inter-aural time difference (ITD)** occurs when the sound reaches one ear earlier than the other ear depending on which side of the head is closer to the source of the sound. The time difference varies with the angle of the sound source. It is the primary cue used for detecting the sound direction in the low frequency spectrum [123]. For example a signal originates from a sound source on the left the signal needs to travel further to reach the right ear than the left ear. This difference in distance between the ears causes a time difference between the arrival of the sound in the left and right ear and acts as a cue to the auditory system of the brain to localize the sound.
ITDs can easily be created with headphones, simply by delaying the signal at one ear relative to the other. The perceived position or the lateralization of the sound is dependent upon its ITD. Sound with ITD of 0 are perceived at the centre of the head and sounds with larger ITDs will be perceived on one side of the head depending on the value of ITD [124]. In addition, lateralization of the sound also depends on the frequency of the tone. The measurements of Schiano et. al’s [125], showed that the lateralization position moves slightly further out from the centre as frequency was increased from 300 to 1000 Hz, then progressively returned to the centre of the head as frequency was further increased up to about 1500 Hz. It was also concluded that pure tones of yet higher frequency are perceived at the centre of the head and do not depend on the value of the ITD.

There are different methods and stimuli used to measure interaural time difference. For example abrupt stimulus such as a click is used to measure onset ITD (the time difference between the onset of the signal reaching two ears). If the stimulus used is not abrupt but periodic then ongoing ITDs are measured. This is where the waveforms reaching both ears can be shifted to create a particular phase difference and the size of this shift is recorded. The shift recorded is known as the interaural phase difference (IPD) and can be used for inputs such as pure tones and amplitude modulated stimuli [123, 125].
**Inter-aural level difference (ILD)** occurs when a sound source is closer to one side of the head than the other resulting in a difference in intensity of the sound with respect to both ears. This intensity difference occurs because the head acts as an “acoustic shadow” that blocks sound waves from reaching the side of the head that is further from the sound source [1]. Thus, for a lateralized sound source, a portion of its sound energy will be blocked from reaching the further ear because the head is in between. The head produces what is called a sound shadow, a weakening of the intensity of sound at the more distant ear. This binaural difference in sound intensity provides another cue for sound localization. It is also important to note that when the source is located equidistant from the two ears, the interaural intensity difference is zero. Neurons have been discovered that respond best when the two ears receive slightly different intensities, some preferring the stronger intensity in the right ear and others preferring the stronger intensity in the left ear [126, 127]. These binaural neurons, incidentally, are found at different neural sites from those registering interaural time differences. Moreover, binaural cells sensitive to interaural intensity tend to fall into the excitatory/inhibitory category, meaning that the input from one ear is excitatory and the input from the other is inhibitory. In fact, the algebraic subtraction of inhibition from excitation - an operation easily performed by a binaural cell - accounts for a given cell’s preferred interaural intensity difference [128]. ILD measurements at each ear have been reported by many researchers. Shaw conducted experiments in 1974 to investigate ILDs for 100 people at different frequencies and the maximum ILDs calculated were 3, 10, 17, and 21 dB at frequencies of 0.2, 1, 5, and 10 kHz respectively [111, 123, 129].
The general effects can be understood from the theory of the interaction of sound with a sphere. The shadowing effect or the depth of the shadow is dependent on how much the head blocks the sound, which in turn depends on the relation of the wavelength to the size of the head. For sound with a frequency of 500 Hz the wavelength is about 70 cm. In this case the shadowing effect is minimal as diffraction ensures that the head offers only a minor obstacle to the sound. On the other hand if the sound has a frequency around 5000 Hz the wavelength is about 7 cm. The diffraction is minimal, and the acoustic shadow is correspondingly deeper. Minor deviations result from the interfering effects of sounds reflecting from the torso or shoulders, both generally at low frequencies. Substantial deviations occur at frequencies above about 2 KHz. They, too, result from interference from reflected sounds, but here the reflections are mostly from the folds and cavities of the outer ear [123, 130].

**ITDs and ILDs together**

All real-world sounds generate both ITDs and ILDs, but that does not mean that the binaural system always uses both to determine the direction of a sound source. Sounds with frequency below or equal to 1 KHz can be localized based on the inter-aural time and phase differences. Sounds with high frequencies can be localized due to intensity differences [131, 132]. In the case of frequency below or equal to 1 KHz, the dimension of the head is smaller than the half wavelength of the sound waves. In this case the auditory system can determine phase lags. On the other hand for sounds with frequencies greater than 2 KHz the dimensions of the head are greater
than the length of the sound waves. Under this circumstance sound localization based on interaural phase difference alone is not possible. However the interaural level differences become larger and these level differences are evaluated by the auditory system. In the case of identical ITDs and ILDs the auditory system filters the sound based on sound direction which also helps the listener in localizing sound directions [133]. Interaural time difference and the interaural level differences in both the horizontal and vertical plane are important auditory processing tools to distinguish signals from the noise.

2.10.2 Binaural Masking Level Difference

The binaural masking level difference (BMLD) is an important psychoacoustic phenomenon that demonstrates and facilitates the use of two ears in detection of signals in noise. It occurs when the cues are presented with interaural level and phase differences. The phase or level difference of the signal at the two ears are not the same as the masker [134].

In a standardized BMLD experiment, the signal threshold is determined by generating a signal (S) together with noise (N). Initially the signal and the noise will be played monoaurally, which serves as the reference condition known as monotonic condition (SmNm). The configuration such as phase, amplitude, duration, bandwidth of the signal and noise are then manipulated at the two ears in various conditions, and the masked threshold of the signal that is still detectable is then determined [134]. In 1979 Sever & Small [135]
demonstrated that the masked threshold level of a signal is the same in both the monotonic (signal and the noise played monaurally) and in the dichotic (signal and noise 180 degree out of phase - the noise or the signal will have 180 degree phase shift compare to the other) condition, thus either the monotonic or the dichotic condition may be used as the reference condition to calculate the BMLD [135].

2.10.2.1 Binaural Modelling

Binaural modelling can be said to have begun in 1948 with Jeffress’s research [136] suggesting a neural coincidence mechanism to detect interaural time difference.

He proposed a model which localizes low frequency sounds by using interaural time differences. The model shows a network of neurons, known as “coincidence detectors,” at the level of the midbrain that fire only when the inputs from the two ears reach the same neuron at the same time. A specific coincidence neuron will respond to inputs with a specific interaural time delay.

In that same year Hirish and Licklider, investigated the effects of the binaural masking level difference (BMLD) and binaural hearing thresholds [137, 138]. They proposed that the detection of a signal with background noise is much easier when the signal has a different inter-aural time difference than that of the noise. In this scenario, the masker is the noise and makes the signal
more difficult to be heard. The level of masking is measured in decibels (dB) and is defined as the difference required to make the sound audible. The value of the BMLD increases with increasing interaural signal level difference, with increasing masker intensity level, and/or with a more narrow masker bandwidth [134].

Many of the effects of frequency and of the interaural configuration can be explained by Durlach’s “Equalization-Cancellation” model [139] as shown in Figure 2.20. Durlach proposed in 1963 a theory of binaural masking which is still popular in designing BMLD experiments. In his model he suggests that the auditory system adjusts the masking signals from the two auditory channels until they become equal. During this equalisation process the target signals are subjected to the same transformation. The auditory system then subtracts the total signal of one channel from the other. Provided there are interaural differences between either the masking stimulus or the target stimulus, the masking components will become nullified leaving a residual signal (target signal) for perceptual analysis. According to that model, there is an equalization in level and internal delay of the signals at the two ears, so that a subsequent subtraction of one from the other will cancel as much of the masking noise as possible. There is a resulting gain in target-to-masker ratio over that found at either ear; hence, there is a gain in detectability of the target. But as the gain in detectability is not infinite, but only, at most, about 15 dB, the cancellation must be imperfect. Durlach developed a mathematically simple form of imperfection which was sufficient to predict much of the experimental data. This analytic “black box” approach is still
used [123, 140], and the equalization-cancellation (E-C) principle has also been incorporated into a computational model. According to Colburn & Durlach, 1965 [141], a larger BMLD can be obtained when the phase of either the signal or the masker is inverted. Three different masking level conditions can be introduced based on the combination of the phase of the signal and the noise. In general, a particular stimulus is described using the symbols S (for signal) and N (for noise), each followed by a suffix to indicate the relative phase in the two ears; '0' for same phase (so-called homophasic) and 'π' for a 180 degree (pi radians) phase difference (so-called antiphasic). For example, S₀N₀ means that the signal and noise both have the same phase in each ear, and SₜN₀ means that the noise has the same phase, but the signal is 180 degrees out of phase.

The Three Conditions for Binaural Masking Level Difference (BMLD) tests are:

- Signal in phase, noise in phase at two ears (S₀N₀) homophasic (diotic) condition.
- Signal in phase, noise 180 degree out of phase (S₀Nₜ) at two ears, antiphasic (dichotic) condition.
- Signal 180 degree out of phase, noise in phase at two ears (SₜN₀) antiphasic (dichotic) condition.
When the same tone in noise is played to both ears, the tone is harder to detect than when one ear either does not get the tone, or has the tone at a different phase [142].

Figure 2.20 Durlach's Equalization and Cancellation model [118]

Durlach’s model can explain this result by assuming that the brain can subtract the signals at the two ears.

With $S_0N_0$, subtracting or adding the two ears signal does not assure separating the signal from the noise.

Subtraction: $(N+S) - (N+S) = 0$;

Addition: $(N+S) + (N+S) = 2(N+S)$ [118]

But with $S_\pi N_0$, since the signal is positive in one ear and negative in the other, subtraction gives double the intensity.

Subtraction: $(N+S) - (N-S) = 2S$ [118]
On the other hand for pure-tone signals presented over headphones in a wideband masking noise, another common experimental design, three general effects are of particular importance.

1. The BMLD depends upon the interaural configurations of the target and the masker, being largest when the noise is in-phase across the ears (i.e., its ITD is 0 ms) and the tone is out-of-phase across the ears (i.e., its ITD is equal to one-half of the tone’s period), but is close to 0 dB when both are in-phase or out-of-phase thresholds [137, 138].

2. The BMLD is largest at low frequencies, being about 15 dB at 500 Hz, and reducing to about 3 dB for frequencies above about 1500 Hz [143].

3. The BMLD can also be observed for other stimuli in addition to pure tones, such as complex tones, clicks, and speech sounds [144]. Other classes of stimuli also give BMLDs, but the values may differ. For instance, the gain in the detectability of speech is about 13 dB, although the gain in intelligibility is only 6 dB [145]. The BMLD has been exploited experimentally to study other aspects of binaural hearing. Examples include the existence of long internal delays [146], the width and the resolution of the binaural auditory filter [147].
The differences in detectability of the various BMLD conditions are believed to be related to the fact that our auditory system is capable of making use of binaural cues to detect signals in noise, mainly the interaural temporal difference cues for low-frequency signals [148]. Culling et. al. [149] showed that an elaboration of the Equalization-Cancellation model could also account for many of the “dichotic pitches”. These are sensations of pitch that are created by the binaural interaction of specially-crafted noise stimuli. They can only be heard binaurally; the noise stimulus at each ear, when presented (using headphones) monaurally in isolation, gives no pitch, but, when presented binaurally, there is a clear, definite pitch, which results from uncancelled ITD disparities in localized bands of the noise [149].

2.11 Hearing Loss

Hearing loss, or deafness, is the partial or total inability to hear sound in one or both ears. Hearing loss can result if there is a problem at any point in the hearing pathway; that is in the outer, middle or inner ears, or in the auditory pathways in the brain. Hearing loss can also be defined in terms of the process of the development of the speech. A pre-lingual hearing loss is where the hearing is lost before a child has completely developed speech and language. It may be congenital or acquired in the first few years of life, and can affect how well a child learns to speak. A post-lingual hearing impairment means the hearing loss is acquired after speech and language have developed. Post-lingual hearing loss is more common than pre-lingual hearing loss [150]. There are different types of hearing loss, depending on which part of the hearing pathway is affected.
**Conductive hearing loss** is due to any condition that interferes with the transmission of sound through the outer and middle ear to the inner ear [17]. Conductive hearing loss leads to a decreased loudness and can often be aided by medical or surgical treatment. Some of the causes of the conductive hearing loss are [150]:

- Blockage of the ear canal by impacted wax or foreign objects;
- Partial or complete closure of the ear canal (known as atresia);
- Outer ear infection (sometimes a result of swimming);
- Chronic ear infection such as otitis media, a common problem in young children;
- Perforated ear drum, possibly from a middle ear infection or a loud explosion; and
- Otosclerosis, an hereditary condition where bone grows around the tiny stirrup bone (stapes) in the middle ear.

**Sensorineural hearing loss** results from damage in the inner ear, the cochlear, the auditory nerve or damage in the brain. The cochlear has approximately 30,000 hearing nerve endings (hair cells). The hair cells in the large end of the cochlear respond to very high-pitched sounds, and those in the small end (and throughout much of the rest of the cochlear) respond to low-pitched sounds. Hair cells degenerate with age and as a result hearing deteriorates. It is estimated that more than 50% of the population over the age of 70 has impaired hearing. Impaired hearing is the most common physical handicap in the industrialized world. Another common reason for
hearing loss due to hair cell damage is noise-induced hearing loss. These types of hearing loss are often most pronounced in the high frequency range. They often interfere with speech understanding, as consonant sounds are in the high frequency range, which are most important, especially in noisy surroundings. Head trauma, ear infections, tumours and ototoxic drugs such as gentamycin are other potential causes of sensorineural hearing loss [151].

Sensorineural hearing loss usually leads not only to a loss of perceived loudness but also to a lack of clarity of the sound so both the quantity and the quality of the sound are affected. This can sometimes limit the benefit that a hearing aid can offer as sounds may be loud enough but distorted.

Sensorineural hearing is usually irreversible and hearing devices are often recommended [17]. Hearing aids can help most people with mild to moderate sensorineural hearing loss in both ears. For more severe levels of hearing loss, even good quality hearing aids may not be adequate as the perceived sounds may be too distorted. If the inner ear is severely damaged, a cochlear implant may provide a solution. A cochlear implant is an electronic device that is surgically implanted. It bypasses the damaged inner part of the ear to stimulate the hearing nerve directly. Unlike hearing aids which simply amplify the sound, cochlear implants convert sound waves to electrical impulses in a way that mimics the natural hearing [152].
Yet another type of hearing loss is **central hearing loss**, where the central nervous system is affected. Patients with central hearing loss typically have inconsistent auditory behaviour. Interpreting speech is a complex task. Some people can hear perfectly well but have difficulty interpreting or understanding what is being said. A condition called central auditory processing disorder frequently leads people to think they have hearing loss when their hearing is actually normal. This is because central hearing loss affects the ability to filter out competing sounds and noises [153]. Despite the fact that this problem is extremely common and present in many highly successful people, it is classified as a learning disability. People with central auditory processing disorders have difficulties that may include difficult hearing when there are several conversations going on (cocktail party problem), inability to read or study with the radio or television on and generally missing the first sentence from people talking to them if they are involved in an auditory attention task.

**Functional hearing loss** is a type of hearing loss that involves a psychological or emotional problem, rather than physical damage to the hearing pathway [17]. Individuals with this type of hearing loss do not seem to hear or respond; yet, in reality, they have normal hearing. The most important challenge for physicians is to identify this condition properly.

**Mixed hearing loss** is a hearing loss where there is a problem in both the conductive pathway (i.e. in the outer or middle ear) and in the sensorineural pathway. An example of a mixed hearing impairment is when there is a
conductive loss due to a middle ear infection plus a sensorineural loss due to the ageing process. This term is used when both conductive and sensorineural hearing losses are present in the same ear.

### 2.12 Ranking of Hearing Loss

The severity of a hearing impairment is ranked according to the additional intensity above a nominal threshold that a sound must be before being detected by an individual. It is (measured in decibels (dB) of hearing loss (HL)). Hearing impairment may be ranked as mild, moderate, moderately severe, severe or profound as defined below [154-156]:

**Mild hearing loss** is defined as between 25 and 40 dB hearing loss. The result of mild hearing loss is difficulty with hearing quiet speech. People who suffer from mild hearing loss have some difficulties keeping up with conversations, especially in noisy backgrounds. One-on-one conversations are fine but it becomes hard to listen to every word in the presence of background noise.

**Moderate hearing loss** is defined as between 41 and 54 dB hearing loss. The result is difficulty in understanding conversational speech. A person may understand about 50% of speech, most vowels and louder consonants, and has difficulty with listening if the environment is noisy. Hearing aids may be of benefit and the person may use lip reading to assist with understanding.
**Moderately severe hearing loss** is defined as between 55 and 70 dB hearing loss. This results in difficulty with speech and decreased speech intelligibility.

**Severe hearing loss** is defined as between 71 and 90 dB hearing loss. In this case a person may hear loud voices close to the ear but will not understand speech without a hearing aid. Sign language may be beneficial to assist with language learning.

**Profound hearing loss** is 91 dB hearing loss or more. The effect is that the person may be able to hear sounds, but not identify them through hearing aids. People who suffer from profound hearing loss are very hard of hearing and rely mostly on lip-reading, and/or sign language.

**Totally deaf** means no hearing at all.

### 2.13 Hearing Tests

The purpose of a hearing test is to determine the hearing thresholds of a person for particular frequencies, or a frequency range. There are a number of hearing tests available to evaluate the hearing of subjects of different ages as discussed below.

**Pure tone audiometry**, commonly used to assess hearing and hearing loss for adults. With the help of the pure tone audiometry it is possible to measure
how well someone can hear sounds of a different pitch and volume. It is a subjective test as it involves the response or the feedback from the subject [157, 158]. This type of testing should be conducted in a soundproof laboratory with the operation of a calibrated audiometer and earphones that present sounds of varying intensity and frequency to each ear separately. Pure tones of rising decibel intensities are transmitted to one ear until the subject responds by pressing a button or raising their finger [158]. This process is then repeated for all the frequencies that are to be tested. Since the pure tones will be presented to the subject through circum-aural headphones, the form of hearing that is to be tested is referred to as air conduction hearing. This form of testing requires the subject to respond to events and is therefore only suitable for adults and older children who can understand the instructions. Once the testing for one ear is completed the whole process needs to be repeated for the other ear. The frequencies that are used in this type of test usually include 500, 750, 1000, 1500, 3000, 5000, 6000, and 8000 Hz [158]. Once the threshold at a given frequency is determined the results are plotted with a specific symbol on the audiogram (Symbol O for the right ear and Symbol X for the left ear). Results from the audiogram are used to make the initial diagnosis of hearing impairment (normal/abnormal hearing sensitivity). In the case of abnormal hearing, the results would help in the diagnosis of the type and degree of hearing impairment [159]. Audiometric testing can be conducted according to Australian Standard AS/ NZS1269.4:2005 using computer based audiometric software. Pure tone audiometry will be used for this research to select
subjects with normal hearing. Normal hearing for the purpose of the study was defined as having no more than 20 dB hearing loss [160].

**Visual Reinforcement Audiometry** (VRA) is a hearing test which is typically used for children over six months of age. The test is a behavioural test. Sounds are presented through a loudspeaker or earphones, the child responds by turning his/her head and is rewarded with a visual stimulus [161]. Frequency-specific thresholds using warble tones or narrow band noise can be determined using this test.

**Conditional Play Audiometry** hearing test is conducted as a game. Sounds at different volumes and pitches are played into the child's ears through headphones and the audiologist will ask the child to do something with a toy (i.e., touch a toy, move a toy) every time the sound is heard. This test relies on the cooperation of the child, which may not always be given [162].

**Tympanometry** is a testing methodology that is used to evaluate the function of the middle ear. Tympanometry measures the movement of the eardrum. It is not strictly a hearing test, but rather a measure of energy transmission through the middle ear. The results of this test should always be viewed in conjunction with pure tone audiometry. A tone of 226 Hz is generated by the tympanometer into the ear canal and some of this sound is reflected back by the tympanic membrane and picked up by the instrument. The instrument measures the reflected sound and plots the results on a chart known as a tympanogram. The tympanogram provides a graphic representation showing
the relationship of air pressure in the external ear canal to the impedance (resistance to movement) of the ear drum and middle ear system. The maximum compliance of the middle ear system which is represented by the highest peak of the graph occurs when the pressure in the middle ear cavity is equal to the pressure in the external auditory canal. This is a non-invasive measurement and does not require any response from the subject [162, 163].

**Otoacoustic emissions (OAEs)** are sounds given off by the inner ear when the cochlear is stimulated by a sound. When sound stimulates the cochlear, the outer hair cells vibrate. The vibration produces a nearly inaudible sound that echoes back into the middle ear. The sound can be measured with a small probe inserted into the ear canal. This type of test is usually carried out with newborn babies. Sound is delivered through a tiny flexible plug inserted into the baby's ear. A microphone in the plug records the otoacoustic emissions (responses) of the inner ear in reaction to this sound. The test is painless and is usually completed within a few minutes, while the baby sleeps [31]. This test can detect blockage in the outer ear canal, as well as the presence of middle ear fluid and damage to the outer hair cells in the cochlear. However, OAEs do not check the auditory nerves or the brain's response to sound [157, 162].

The **auditory brainstem response (ABR)** test is a hearing test which measures the response of the brainstem to an auditory stimulus. This type of test can be used to diagnose the functionality of the inner ear and auditory nerve as well as the lower brain and is commonly conducted on infants.
Clicking sounds are made through tiny earphones in the baby's ears and the EEG signals are recorded from electrodes which are attached to the baby's scalp [164]. The ABR will be discussed in more detail in Section 2.15.

### 2.14 BMLD Test

The binaural masking level difference test is a type of test that is used to find the ability to listen to tones and sounds with two ears. It is a test of auditory processing [165]. Ira Hirsh was a pioneer researcher in investigating the effects of the binaural masking level difference and binaural hearing thresholds [137, 138]. The test is based on diotic listening when a signal or the masking noise are phase shifted when presented to both ears. Such listening is connected with the so-called “cocktail party” effect, where the signal and masking sounds have different source locations in space. In this type of test a pulsating or constant tone is used where the tone or the signal are played to both ears while a masking noise is also provided to both ears. The auditory threshold for the signal is found based on the presentation of the tone and the noise. The tone and noise can be presented in the same phase (homophasic) or they can be presented in the opposite phase (antiphasic) [166]. In homophasic conditions, the signal and noise are presented in the same phase to both ears ($S_0N_0$), whereas in antiphasic conditions one of the two signals is shifted 180° out of phase, while the second sound remains in phase between the ears, e.g. $S_0N_\pi$. The difference in auditory thresholds for the signals obtained for the ($S_0N_\pi - S_0N_0$) conditions is determined as BMLD. The auditory threshold in antiphasic
conditions should be lower. This hearing improvement after shifting one of the sounds by 180° between the ears is referred to as release from masking [115, 139].

2.15 Objective Auditory Processing Tests

A test based on auditory evoked potential may provide an objective measure of auditory processing. The ABR measures the electrical response of the inner ear and the auditory nerve and can be used for diagnosing the functionality of the lower brain stem. This type of test is often used for babies. The test cannot be used to detect auditory processing disorders in other parts of the brain.

Currently, there is no objective method to detect auditory processing related to binaural hearing, and disorders can therefore not objectively be determined. Conductive hearing loss in children can adversely affect the normal hearing ability and the development of the brain. As a result children may develop long term speech and language problems [10]. It is therefore important to develop an objective methodology which can detect binaural processing. Auditory evoked potentials will be used to develop such a methodology in this thesis. Auditory evoked potentials will be discussed in more detail in the next section.
2.16 Evoked Potentials

An evoked potential (EP) is an event-related activity, which occurs due to the electrical response from the brain or the brainstem to various types of sensory stimulation. The non-invasive method of recording such type of electrical potentials may be very useful in detecting abnormalities in sensory pathways, brain diseases and disorders related to language and speech [36]. Individual evoked potentials however have low amplitudes, ranging from 0.5 µV to 100 µV. They can be recorded using an EEG. Since the amplitude of an EP is similar to that of spontaneous neural activity, it is difficult to observe an EP. The target signal (EP) must be extracted from the "noise" of other spontaneous neural activity. The most commonly used method for EP extraction is the technique of averaging [21]. Since the spontaneous neural activity can be considered a near random process of positive and negative values, the averaging of these values over time will approach zero. The averaging of a sufficient number of repeated evoked potential sweeps, therefore, results in the relative elimination of noise. Several conditions must be met to successfully employ this technique:

- the raw signal is a linear sum of the target signal (EP) and noise (spontaneous activity);
- consecutive evoked potentials are similar in shape;
- individual components of consecutive EPs appear at the same time interval after stimulation;
- noise is of random distribution and is not in temporal relation to the stimulating impulse [23, 25].
Evoked potentials consist of a sequence of deflections or waves, each characterized by their specific latency, amplitude and other features [167]. Analysis of the evoked potential signals can provide information about the presence of disease or degeneration, and can help to determine the location of nerve lesions.

2.16.1 Visual Evoked Potentials

A visual evoked potential (VEP) is an electrical response originating from the brain which can be recorded from the occipital region of the scalp. It can be used to evaluate the functionality of the visual pathway. Visual evoked potentials have amplitudes which are considerably larger than the AEP, ranging up to 20 µV, and their spectral components lie in the range of 1 to 300 Hz. The visual evoked potential is the only type of evoked potential that can be observed directly in the EEG without prior noise reduction. The normal visual evoked potential waveform can be characterized by 3 peaks. An initial negative peak (N1 or N75), occurring about 75 ms after stimulus, followed by a large positive peak at 100 ms after stimulus (P100) which is followed by another negative peak occurring at about 145 ms post stimulation (N145) [26]. In eliciting the visual evoked potential two different types of stimuli are used: pattern reversing and flashing [168, 169]. The choice of the stimuli depends on the suspected disorder and the ability of the subject to cooperate during the recording procedure.
**Pattern Reversing** involves recording of the VEP based on reversing the stimulus pattern. The stimuli are generated in the form of a chessboard pattern displayed on a video screen. During the experiment the subject is instructed to focus on a point in the centre of the screen while the black-and-white squares are reversed at a fixed repetition rate (e.g., 2 reversals per second) so that the white squares become black and vice versa. The recorded VEP waveform amplitude and latency depend on a number of factors that include the size of the chessboard squares, the luminance and contrast of the squares, and the repetition rate.

**Flashing stimuli** are used when the patient cannot focus or maintain the level of fixation required for pattern reversal stimulation. The flash stimulus is delivered at a rate of five to seven flashes per second. Although the eyes are closed during this procedure, a sufficient amount of light will pass through the eyelids to activate the retina.

Visual evoked potentials are used to investigate ocular and retinal disorders and to detect visual field defects and optic nerve pathology. Visual evoked potential testing is also used for the diagnosis of epilepsy and multiple sclerosis. The latency of the VEP can indicate whether the neural pathways are transmitting signals properly. For example in case of multiple sclerosis, a disease which causes damage to multiple areas to the myelin sheath within the brain and the spinal cord [170], it takes a longer time for signals to be conducted from the eyes to the visual cortex, resulting in an abnormal VEP [170].
2.16.2 Auditory Evoked Potentials

Auditory evoked potentials are very small electrical voltage potentials in response to auditory stimuli such as tones, chirps, clicks etc. They originate from the brain and are recorded from the scalp. Auditory evoked potentials consist of negative and positive deflections that follow the stimulus in a time locked manner. The ability to record the AEPs on the scalp mainly depends on the four following factors [171, 172].

- The number of cells activated by the stimulus;
- The degree of synchronization of this activation (the more synchronous the activation, the larger the response);
- The geometry of the structure activated (depending on the arrangement of cells, the microscopic dipoles produced by the activation of each neuron may sum up or cancel each other); and
- The ability of surrounding tissues (bone, muscle, glia) to conduct electricity.

Auditory evoked potentials are further classified as either a transient or a steady state response. Transient AEPs are seen when the stimulus delivery is slow enough for the brain response to settle down completely between stimuli whereas steady state AEPs can be seen when the rate of stimulus delivery creates overlap of the individual transient responses. Transient responses are characterized by their latency (time between stimulus onset and the AEP) and their amplitude. Steady-state responses on the other hand
can be investigated in the frequency domain [172]. The amplitude of an AEP is in the range of 0.5 -10 µV. Since the AEP signals are of low voltage, the signals need to be captured using highly sensitive amplifiers. Because they are combined with relatively high background electrical noise, signals need to be averaged to increase the signal to noise ratio. The complexity and challenges in designing a system to capture AEPs stem from the fact that these signals are so minute in amplitude. An AEP acquisition system needs to be of high quality in order to avoid introducing more noise into a signal where the signal to noise ratio (SNR) is naturally low. The objective of this type of system involves capturing the continuous recording of the brain at an appropriate sampling frequency, synchronising the AEP with the stimulus and averaging the trials to obtain a higher SNR. The averaging technique is commonly used to improve the signal to noise ratio before any further processing and analysis is performed [21].

The stimulus may be presented in the form of tone bursts, chirps, tone pips, sinusoidal tones, clicks, virtual acoustics and speech [122, 134, 142, 173-175]. The stimulus delivered must have a higher intensity than the hearing threshold, but not so loud to be unpleasant for the subject. The scalp location where the AEP amplitude is recorded is usually the vertex (Cz) [176, 177] and earlobe and mastoid are considered to be the ground and reference for AEP acquisition [172].
The auditory evoked response can be divided into three different intervals according to latency (latency is the time between stimulus onset and the AEP [178]), as described in the next four sections of this chapter.

2.16.2.1 Auditory Brainstem Response

The auditory brainstem response is an auditory evoked potential of very low amplitude ranging from 0.1 to 0.5 µV. It occurs between 0 to 10 ms after the stimulus. The auditory brain stem responses (ABRs) can be measured using scalp electrodes that pick up electrical potentials generated by the synchronous activity of populations of neurons in the brain stem [179]. The auditory brainstem response of a normal subject can be characterized by 5-7 peaks generated by various neural structures in the auditory pathways. The conventional method for labelling these peaks consists of using Roman numerals I to VII [180] as shown in Figure 2.21. The waveform represents specific anatomical points along the auditory neural pathway. The auditory structures that generate the auditory brainstem response are believed to be as follows [181].

![ABR Response](image)

Figure 2.21 ABR Response [181]
Wave I – generated by the peripheral portion of the auditory vestibular nerve

Wave II – generated by the central portion of the auditory vestibular nerve

The auditory vestibular nerve is the eighth of twelve cranial nerves, and is responsible for transmitting sound and equilibrium (balance) information from the inner ear to the brain.

Wave III – generated by the cochlear nucleus. The cochlear nuclei (CN) are the two heterogeneous collections of neurons in the mammalian brainstem that receive input from the cochlear nerve, which carry sound information from the cochleare.

Wave IV – generated by the superior olivary complex/lateral lemniscus, a collection of brainstem nuclei that function in multiple aspects of hearing. The superior olivary is an important component of the ascending and descending auditory pathways.

Wave V – generated by the lateral lemniscus/inferior colliculus. The inferior colliculus (IC) (Latin, lower hill) is the principal midbrain nucleus of the auditory pathway and receives input from several more peripheral brainstem nuclei in the auditory pathway, as well as inputs from the auditory cortex [182].
Due to the low amplitude of the auditory evoked potential (AEP), several thousands of auditory brainstem epochs need to be averaged in order to reduce the noise level [180]. Most of the spectral content of the AEP lies in the frequency range of 100 Hz to 1.5 kHz [180, 183].

Interpretation of the ABR is primarily done by looking at the amplitude and the latency of the signal. The auditory brainstem response has been used for the evaluation of hearing loss, diagnosis of certain brainstem disorders and monitoring to prevent neurological damage during surgery.

### 2.16.2.2 Middle Latency Response

The middle latency response (MLR) is an ongoing brain signal that follows the auditory brainstem response. It usually occurs from 20 ms post stimulation [172, 184-186]. It consists of a series of positive and negative peaks originating in the upper brainstem and/or auditory cortex. The MLR follows the auditory brainstem response (ABR) and precedes the late latency response (LLR) of the brain. The middle latency response consists of three main peaks Na, Pa and Nb which occur between 20 to 80 ms after stimulation time. These components represents the activity of subcortical and primary auditory cortical neural activities which are related to auditory information processing [187]. Peaks labelled with an N are negative peaks and P indicates positive peaks. Peaks with ‘a’ designations precede peaks with ‘b’ designations in latency.
The first prominent peak of the MLR waveform is Na, which occurs at a latency of 20 ms relative to stimulus onset, depending on the type of stimulus used. Both subcortical and cortical structures appear to contribute to the peak Na [189]. The Na component is recorded reliably in individuals with normal auditory systems and serves as a visual marker for the onset of the MLR.

The second prominent peak of the MLR waveform is Pa, which occurs at a latency of approximately 25 milliseconds and is believed to arise from the temporal lobe [190]. The amplitude of this peak decreases slightly with reductions in arousal, implying a minor contribution of the eye movements [190]. It is also affected by stimulus intensity, frequency and rate. Generally Pa will decrease in amplitude and increase in latency if the stimulus intensity or frequency are reduced [190].
The latency of Nb generally falls in the range of 35 to 45 ms relative to the onset of the stimulus. It is slightly less reliable than the Na and Pa components because it is influenced more by arousal and attention. Because Nb is so variable among individuals, it is not as well studied as Na and Pa. Thus less is known about the influence of stimulus characteristics (e.g. frequency, intensity, and rate) on the amplitude or latency of response [190].

The MLR has been found to depend on a number of parameters that include the subject state of arousal, age and gender. These subject factors can alter the amplitude, latency, and morphology of the MLR waveforms in normal-hearing subjects. Stimulus characteristics such as signal amplitude and frequencies are also considered to be the factors that produce changes in amplitudes and latencies of MLR waveforms. These subject and stimulus factors appear to have a more pronounced effect in MLR recordings in young children [191-195]. Tucker et. al. [196] have conducted experiments where they have represented the effect of stimulus rate and gender on the auditory middle latency response. Their findings suggest that Pa and Pb amplitudes decreases with increasing stimulus rate, and that the Pa latency significantly increases with increasing stimulus rate. No significant differences were seen in the Pb latency. Gender had a significant effect on the Pa latency and Pa amplitude. Pa latencies were longer in male subjects, and Pa amplitudes were larger in female subjects. Gender did not have a significant effect on the Pb waveform [196]. This finding is consistent with the finding of Woods and Clayworth showing larger Pa amplitude in females [193]. The waveform characteristics of the MLR can be used as a possible indicator for depth of
anaesthesia. Evidence suggests that general anaesthetics increases the latency and decreases the amplitude of the MLR depending on the depth of anaesthesia [197]. However, the response is very small (typically 1 µV) and it can therefore be difficult to measure this in a clinical environment where the presence of artifacts can prevent acquisition of a good quality MLR [198]. Artifacts will be further discussed in Section 2.17.

Clinical uses of the MLR include the electrophysiological determination of low frequency hearing thresholds, the assessment of cochlear implant functionality, the assessment of auditory pathways, and the localization of auditory pathway lesions [185].

The possible origin of the middle latency responses is discussed by several researchers. It is believed that the MLR represent the response of the firing of the neurons in the primary auditory cortex in response to an acoustic stimulus [198]. It has been reported by Goto, T and Lee, Y in [199, 200] that the direct cortical recordings of auditory responses to auditory stimulations indicate generators located in the temporal lobes of the brain. Considering latency variations for different species, similar activation was also found in the primary auditory cortex of cats and monkeys [201]. In addition, intracranial recordings in cats, rats and guinea pigs indicate that the peaks of the MLR are normally generated by the auditory cortex [202].

Since the MLR is a response of the brain due to auditory stimuli, this type of response cannot be evoked from subjects who are profoundly deaf. If the
subject has mild or moderate hearing loss, a given auditory stimulus will generally sound quieter [198]. This is particularly the case for conductive hearing losses, where sound is attenuated in the outer or middle ear before reaching the cochlear. These types of hearing loss can result in a reduction in the amplitude of the auditory evoked potential. Consequently, the signal-to-noise ratio (SNR) of the MLR will be reduced.

2.16.2.3 ABR and MLR – Performance Analysis

Both the auditory brainstem response and the middle latency response have been advocated for estimating the pure tone audiogram [203]. But conflicting results are reported in literature on which response is better in estimating the hearing thresholds, especially in response to low-frequency stimuli of 500 Hz or 1 KHz [190, 204, 205]. Some researchers claimed that ABR testing using 0.5, 1, 2 and 4 KHz tone have shown better results in estimating hearing threshold for both normal hearing and hearing impaired infants and adults [206, 207]. However other researchers mentioned that ABR recordings for 0.5 and 1 KHz tones sometimes produce unacceptable elevated thresholds of more than 25 dB. This limits the usefulness of the ABR for measuring hearing thresholds [204, 208]. Since the ABR may produce conflicting results in measuring hearing thresholds for low frequencies, the MLR has been investigated as an objective electrophysiologic measure of low frequency hearing [190, 209]. Some researchers concluded that MLR predicts hearing thresholds for 500 Hz and 1 kHz tones for normally-hearing adults and children [204, 210] better than the ABR. Musiek et. al. [211] conducted
experiments and found that MLR performs better than ABR for threshold estimation. The experiment was conducted on 15 normal hearing adults where clicks were used to obtain the ABR and the MLR response. Maurizi et. al. [186] has also conducted similar experiments on adults by investigating the middle latency response to clicks and tones (0.5 and 1 KHz). Results indicate that the MLR was a reliable indicator when using tones. 20 normal subjects in the age range of 26 to 33 participated and their middle latency response shows the conventionally labelled Na, Pa, and Nb waves present. It is also found that the latencies of these waves tend to be greater than those of the corresponding waves elicited by clicks [186]. It is also important to note that ABR requires a high degree of neural synchrony which is known to be synchronous oscillations of membrane potentials in a network of neurons. As a result it is possible that individuals with certain neurological disorders may exhibit no ABRs despite hearing normally. For the same reason, a lack of sufficient synchrony in response to low frequency signals often makes the MLR superior to ABR in assessing low-frequency hearing [209, 212]. However the MLR does not reach its mature morphology until adolescence and, especially with children, the sleep state has a significant influence on this [209]. Kraus et. al. conducted a study to find age-related changes in the MLR. Their findings suggest that the detectability of both Na and Pa increased significantly with age [213] and their results were consistent with other researcher’s findings [171]. As described by Yakovlev and Lecours, age-related changes in MLR may be related with the maturation of the non-specific thalamocortical pathways and sensory cortex and is complete only by the onset of puberty [214].
For low frequency tones the timing difference between the left ear and right ear tones also play a very important role for sound detection. Animal research has shown that in reptiles whose hearing range generally extends from about 0.1 to 5 kHz, the auditory system can resolve phase at frequencies below about 1 kHz [215, 216] whereas birds can detect phase differences until about 2 kHz above which they then becomes undetectable [217]. In humans and in all other mammals it has been found that interaural timing, or interaural phase, of low-frequency sounds (<4000 Hz) is an important cue for localizing sounds [215]. The auditory system follows the mechanism of combining the signals from the two ears and forwards that information to the central auditory system [132].

### 2.16.2.4 Late Latency Response

The late latency response (LLR) or the slow vertex response is another component of the auditory evoked potential. It has a response which is of very low frequency and a voltage range between three and ten microvolt. The LLR signal is usually characterized by 4 peaks containing two positive (P1 and P2) and two negative peaks (N1 and N2) [218, 219]. N and P refer to the sign of the potential (positive and negative) at the vertex compared to the potential at the reference electrode. The response can be evoked by the onset of a tone or it can be triggered by a significant change in intensity and frequency of a tone.
The evoked potentials resulting from the auditory stimulation have important clinical applications. The amplitude of the N1-P2 response can be up to 25 $\mu$V for moderate to high intensity stimuli and knowledge about the characteristics of the response will provide useful information about the individual's hearing threshold. The N1-P2 complex is a cortically generated, auditory evoked potential that can be recorded using either acoustical or electrical stimulation. In adults, the latency of N1 is about 110 ms and followed by a positive potential (P2) that has a latency of approximately 175 ms. McEvoy LK et. al. [221] conducted experiments to look into the effects of stimulus parameters on human evoked potentials to shifts in the lateralization of a noise. Twenty subjects in the age range of 21 to 46 participated and the stimulus frequency was kept below 200 Hz and is mediated through the middle and the apical regions of the cochlea. Their finding suggests that changing in the interaural time difference of a continuous binaural noise causes a shift in the perceived lateralization of the noise. In this case the N1-P2 peaks occurred at 130 ms and 220 ms respectively. In addition Ungan P et. al. [222] also looked into the N1 response of the LLR by introducing both
the IID and the ITD to the binaural stimuli at a regular interval of 2 seconds. 100 clicks per second were dichotically presented with either interaural time delay of 1ms or an interaural intensity difference of 20 dB (HL) at 2 seconds interval. Their results show that the N1 response evoked by IID and ITD are binaurally processed in different ways and/or in different areas in auditory cortex.

One of the main clinical applications of this response is to investigate the auditory hearing threshold for adults [223]. Measurements of the amplitude and latency of the averaged evoked potential are compared with normative values in order to distinguish between healthy subjects and subjects with neurological impairment. In addition, the LLR can be used to detect the hearing ability of hearing impaired children. Suzuki et. al. [224] conducted experiments in order to find the reliability of the late latency response audiometry in young children with sensorineural hearing loss. They conducted experiments on 6 children aged 2 to 4 years who have sensorineural hearing loss of 60-75 dB for the test frequency of 1000Hz. It was found that the amplitude of the response increased as the stimulus intensity was increased from 60 to 80dB which indicates that the response detectability in children with sensorineural hearing loss sharply increased at the level of the stimulus intensity close to their auditory thresholds.

The middle and late auditory evoked potentials have significantly larger amplitudes than the brainstem auditory evoked potential. Hence only several hundred trials are required for averaging of MLR and LLR signals [36, 178, 225].
Although the LLR can be used to measure the hearing threshold, this type of test is not widely used. Since the LLR response goes up to 500 ms after stimulation [198] and several hundreds of trials are required for averaging, the total experimental time gets bigger. It becomes more difficult for the subject to maintain their concentration, which causes response degradation over time [226]. When recording the mid-latency and late latency responses (MLR and LLR respectively), it is important that the subject should stay awake and alert. The fact due to long experimental time for LLR acquisition, subjects tend to feel drowsy which is known to affect especially the late latency response [227]. As a result it causes diminution of late latency response amplitude N1-P2 over time [228]. The MLR therefore seems to be the preferred candidate at this stage for this research.

2.17 Auditory Stimuli

Selection of the auditory stimuli depends on the type of auditory evoked response and the purpose of the test. The following variables need to be considered during the design stage of any auditory stimuli:

- Number of trials (or sweeps) – the number of times that the stimulus sequence is reiterated in order to reach the desired SNR;
- Duration of stimuli;
- Duration of silence – the period of silence between the stimuli;
- Frequency – the frequency of the tone which is presented;
- Sound Pressure Level – logarithmic measure of sound intensity;
- Rise and/or fall time – the length of time of a linear or sinusoidal rise and/or fall in amplitude of the tone;
- Plateau time – the length of time the amplitude of the tone is constant;
- Noise application – duration and phase of the masker noise if applicable; and
- Sampling frequency.

The most common type of stimuli are acoustic clicks (or rectangular pulses) tone bursts (or tone pips) and pure tones [229]. The ABR and MLR occur at a very short period of time after the stimulus. The duration of the stimulus therefore also needs to be short in order to prevent the overlapping of the stimulus and the response [225]. The damping time of the headphones needs to be taken into account as well [36]. The sound pressure level is another stimulus parameter that needs to be considered. The value of parameter usually ranges from 50 dB to 70 dB. Sounds which are lower than 50 dB may become difficult to detect while sounds exceeding 70 dB may produce artifacts [230]. From an experimental perspective in order to keep this parameter constant, the headphones (or any other sound producing device) need to be calibrated with a sound level meter.

Stimuli can also be delivered in a fashion where the stimuli in one ear will be out of phase with the other ear. In addition noise can be added in order to implement the concepts of Durlach’s equalization model [139]. The combinations of the stimuli that can be derived are:
- $S_0$ – signal is presented to both ears in-phase.
- $S_0S_\pi$ – signal is in-phase in one ear and $180^\circ$ out-of-phase in the other ear;
- $S_0N_0$ – signal and noise are presented to both ears in-phase.
- $S_\piN_0$ – signal is $180^\circ$ out-of-phase in one ear and the noise is in-phase.
- $S_0N_\pi$ – signal is in-phase and the noise is $180^\circ$ out-of-phase in one ear; and
- $S_\piN_\pi$ – signal and noise both $180^\circ$ out-of-phase to one ear compared to the other ear.

Stimulus selection for this research will be further discussed in Chapter 4.

### 2.18 Artifacts

The presence of artifacts has always been a problem for the analysis and the interpretation of EEG recordings. Artifacts are unwanted electrical signals that are almost always present in EEG signals. Artifacts are introduced to a EEG signals by a patient’s interference or movements and by sources outside the patient’s body [231]. The presence of artifacts significantly reduces the quality of the EEG signal. This reduces the clinical usefulness of the EEG signals and makes both manual and automatic analysis difficult. In some cases it is impossible to filter out the artifacts from the signals due to the similarity between the artifacts and signals [232].
The artifacts in EEG signals can be broadly classified into two categories: physiological artifacts and extraphysiological artifacts.

**Physiologic artifacts** (i.e. from the patient's body) can occur due to eye movement, muscle activity, perspiration and respiration, heart beat and tongue and jaw movement.

**Extraphysiological artifacts** (i.e. due to environment, equipment and electronic connections) can occur due to unstable and/or faulty electrodes, unstable and/or faulty hardware, dirty and/or scratched electrode surfaces, 50 or 60 Hz power line noise and other environmental factors creating external noise.

### 2.18.1 Physiological Artifacts

The electrooculogram (EOG) generates EEG signals due to eye movements and/or blinks. This is found to be the most significant and common physiological artifact of EEG signals [233, 234]. Muscle activity, cardiac activity pulsating vessels and respiration may also induce artifacts.

#### 2.18.1.1 The electrooculogram (EOG)

The electrooculogram (EOG) is the electrical activity that is caused by movement of the eyes. It is one of the noncerebral artifacts which is
significant enough to be clearly visible in the EEG signal [36]. EOG artifacts are generally higher in amplitude and lower in frequency than the EEG signal and affect mainly the low frequency of the EEG signals. The EOG reflects the potential difference between the retina (negative) and the cornea (positive) which changes during eye movement. The strength of the interfering EOG signal will depend on the direction of the eye movement and the proximity of the electrode to the eye. EOG artifacts can sometimes be confused with slow EEG activity, e.g., theta and delta activities [36]. Eye movement is not only present when the subject is awake, but may also interfere with the EEG when rapid movements occur during sleep.

Another common type of artifact is caused by eyelid movement (“blinks”) which also influences the corneal-retinal potential difference. The blinking artifacts usually produce potentials of larger amplitudes and can also be seen easily in the EEG recording. They usually produce larger peaks and more abruptly changing waveforms than eye movement and also contain high frequency components [235].

Several methods have been suggested in literature for the minimization of the EOG artifacts from contaminated EEG signals [232, 233]. One of the most regularly practiced procedure for EOG artifacts correction is rejection [236]. This method involves discarding portions of EEG data that possess attributes characteristic for eye movement and blinking, such as amplitude, peak, variance and slope that surpass established criterion thresholds. A disadvantage of this technique is that it can lead to a significant loss of data.
It is also difficult to identify all artifact occurrences and the remaining segments of information may no longer be suitable for study or examination. The deletion of specific trials means that more trials need to be completed in order to obtain enough results. Schlogl et. al. [237] used a regression-based analysis for the removal of fast and slow EOG related artifacts. A cascaded spatio-temporal algorithm to eliminate EOG artifacts was developed by Liu, T. and D. Yao, [234].

Other artifact minimization techniques based on independent component analysis (ICA), principal component analysis (PCA) and neural networks are discussed in various publications [238, 239]. Independent component analysis (ICA) is a computational method for separating a multivariate signal into additive subcomponents by assuming that the subcomponents are non-Gaussian signals and that they are all statistically independent from each other. ICA is a special case of blind source separation. When the statistical independence assumption is correct, blind ICA separation of a mixed signal gives good results. A simple application of ICA is the "cocktail party problem", where the underlying speech signals are separated from a sample data consisting of people talking simultaneously in a room. Usually the problem is simplified by ignoring time delays or echoes [240].

Principal component analysis (PCA) is another mathematical algorithm that reduces the dimensionality of the data while retaining most of the variation in the data set. It accomplishes this reduction by identifying directions, called principal components, along which the variation in the data is maximal. By
using a few components, each sample can be represented by relatively few
numbers instead of by values for thousands of variables. Samples can then
be plotted, making it possible to visually assess similarities and differences
between samples and determine whether samples can be grouped [241].

2.18.1.2 Muscle Activity

The electrical activity caused by muscle movement of the body is another
common type of non-cerebral artifact. It can be measured from the body
surface and is called the electromyography (EMG). It occurs when the
subject makes certain movements such as swallowing, chewing, talking,
frowning, etc [36]. Generally the potentials generated by these muscles are
of shorter duration and may have amplitudes which are several times larger
than the EEG signal, generally between 10 µV - 2 mV with a frequency in the
range of 30-500 Hz [235]. The overall shape of the EMG signal depends on
the degree of muscle contraction; a weak contraction produces spikes of low
amplitude whereas strong contraction produces large spikes. The muscle
artifacts can be considerably reduced by allowing the subject to relax or
sleep. Filters are commonly used to remove these artifacts. Independent
component analysis (ICA) can also be used to separate the EEG from EMG
artifacts into statistical independent components [242].
2.18.1.3 Cardiac Activity

The electrical activity of the heart as reflected by the ECG can also affect the EEG signal. Although the amplitude of the ECG signals measured from the scalp is low compared to the EEG amplitude, it can still affect the EEG signals at certain electrode positions. Generally, people with short and wide necks have the largest ECG artifacts on their EEG [233]. Individual variations in the amount and persistence of ECG artifacts are related to the field of the heart potentials over the surface of the scalp. Depending on the circumstances, the magnitude of ECG artifacts varies from 80 µV to as high as 2 mV in the frequency range of 0.5-50 Hz [235]. The ECG may be acquired independently by one or several electrodes and this may be used in cancelling the ECG activity that is found in the EEG recording [235].

2.18.1.4 Pulse and Respiration Artifacts

A pulse artifact occurs when the electrode is placed on a pulsating vessel. It can induce slow waves in the EEG recording. Respiration can also cause artifacts. These can take two forms, one in the form of slow and rhythmic activity, synchronous with the body movements of respiration and mechanically affecting the impedance of (usually) one electrode, the other one can be slow or sharp waves that occur synchronously with inhalation or exhalation of the patient [243].
2.18.2 Extraphysiological artifacts

2.18.2.1 Alternating current (50 / 60 Hz) artifact

Alternating current devices can also cause artifacts. The 50 / 60 Hz signal is one of the sinusoid shape artefact that interferes with the actual EEG signal. This interference from e.g. power lines (including the harmonics of the wave) occurs because of the influence of the electric and magnetic field on the electrode leads. The body generates potentials which are added to the signal [244]. This interference can be eliminated by using a notch filter around 50 or 60 Hz and their harmonics. However, this filtering method should be performed with caution since there may be a loss of data if the frequency range of the AEP overlaps with the notch filter. To minimise this effect, electrical appliances and devices must be positioned as far away as possible from the electrodes and cables.

2.18.2.2 Electrode conditions and movement

It is important to have high quality electrodes in order to capture good quality EEG signals. Scratched and dirty electrode surfaces are poor receptors and results in low quality EEG signals. A smooth surface helps to keep the impedance low. Unstable electrodes frequently cause sharp waves in EEG recordings. Any movement of electrodes will cause a change in the conductivity and create a spike in the EEG recording. Movement of electrodes causes changes in the DC contact potential at the electrode-skin interface which produces an artifact commonly known as the ‘electrode-pop’
artifact [243]. The electrode impedance is a direct measure of how well the electrode cup is conducting. It is recommended that the impedance is lower than 5 kΩ at all times. This can be achieved by applying the conductive gel and adjusting the electrode until the impedance is lower than 5 kΩ. The electrode leads should not curl together and should not touch each other as this would increase the chance of interference.

2.18.2.3 Stimulus Artifacts

The presentation of the stimuli can also cause artifacts in the EEG signal. This type of artifact is known as stimulus cross-talk. To record the auditory evoked potentials, the stimuli are presented to the subject through headphones. The electrodes for capturing EEG signals are very sensitive and can easily pick up the electromagnetic waves generated by the headphone. The magnetic field produced by the headphone can couple with the electrodes, causing a signal of a similar shape as the stimulus to appear on the EEG channel. These stimuli related artifacts are present in every trial and are not of a random nature. This means that they will not cancel out during the averaging process. The amplitude of the artifact is dependent on the magnitude of the stimulus and the distance between the headphone and electrode.

2.18.2.4 Movements in the environment

Movement of other persons around the patient can generate artifacts, usually of capacitive or electrostatic origin. This type of artifact should be avoided as much as possible [243].
2.18.2.5 Other Equipment

There is a whole range of electrical equipment which is capable of causing artifacts. This is especially true in a clinical environment where machinery such as respirators, perfusion pumps, and other (mechanical) actuators such as flush devices, cutting, drilling, suctioning, rubbing and washing can cause (rhythmic) artifacts. Other examples include power cables, transformers, and antenna-equipped devices where the electric field can possibly be picked up on electrode leads. Therefore it is important to take careful consideration of the placement of the EEG equipment and the surrounding machinery [243].

For the recording of the auditory evoked potentials it is of vital importance to avoid and minimise artifacts as much as possible.
2.19 Summary

This chapter is the literature survey for the research and lays the ground for the ideas in the thesis. The chapter introduces the concepts relevant to the thesis' experiments and describes related research. An overview is given of electroencephalography (EEG), including historical development, recording methods, characteristics of EEG signal and common applications. The anatomical structure of the ear and the brain are described as they are relevant for hearing. Since the research is to investigate the brain response based on binaural hearing, the literature search was extended further into binaural hearing and the binaural cues used by the brain to localize sound sources. A series of hearing tests are compared and a need for objective way of measuring auditory processing related to binaural hearing is proposed. Auditory evoked potentials may be used to develop such methodology. Evoked potentials and the classifications of the evoked potentials are discussed in detail. Past researches on analysing the three different types of brain response (ABR, MLR and LLR) elicited due to auditory stimuli are summarised and the response that will be further analysed in this research was chosen. Finally, the chapter explains the different configurations of the auditory stimuli that can be used for auditory evoked potentials and artifacts which may pose a problem for the analysis and the interpretation of EEG recordings are described.
3.1 Experimental Overview

The aim of this research is to propose an approach towards building an objective methodology to detect binaural processing in the human brain using auditory evoked potentials. Normal hearing subjects in the age range of 18 to 35 were selected for this research. This age range was chosen as they were people who could consent to the research while minimizing the negative effects of age on hearing [245]. In order to evaluate a subject’s hearing range for this research, a pure tone audiometric hearing test was conducted according to the relevant Australian and International Standards. The research focusses on the middle latency response (MLR). As discussed in section 2.15.1.3, previous research demonstrated that, contrary to the ABR, its response can be used as an objective electrophysiological measure of low frequency hearing [190, 209], while being less time consuming and less affected by difference in attention than the LLR. The relevant ethic clearance (H12043) was obtained to conduct experiments on human subjects.

A series of homophasic and antiphasic stimuli are developed to evoke the event related potentials to detect phasic effects in the binaural processing of the brain. The subjects are presented with these stimuli and the neural activity is captured over a sufficiently large number of trials using the EEG equipment. The auditory evoked potentials are measured from the cortical
position Cz [176, 177] with a reference electrode placed on the left earlobe and ground electrode located on the forehead [172]. The collected data is then pre-processed (averaged) to reduce the influence of artifacts, after which the signals can be analysed.

### 3.2 Audiometric Testing Laboratory

For both the hearing test and electroencephalography, the testing environment must be subjected to particular standards. According to Australian Standard AS/NZS 1269.4:2005 (Occupational Noise Management), the maximum acceptable background noise level for audiometric testing is 30 dB for both A and C frequency weightings as measured by a sound level meter [246]. It is a requirement that the testing should be conducted in an electrically shielded soundproof laboratory. The lights in the laboratory are also to be switched off in order to reduce 50 Hz power artifacts. A drawing of the soundproof room used for the research is presented in Figure 3.1.
Figure 3.1 Drawing of Soundproof Laboratory used for the research [247]

All experiments conducted in the soundproof room are controlled from another room adjacent to the soundproof room. The purpose of this control room is for monitoring of the subject and the data acquisition. Two computers are being used, one to present the stimulus and the other computer to capture the raw signal data. Both are located in the control room to minimize 50 Hz interference and to avoid noise from the hard disk drives and cooling fans of the computer. There are two doors between the soundproof studio and the control room, each has acoustic door stop seals on either side. Fibre glass infills and acoustic tiles further enhance the sound insulation characteristics of the room. The photographs in Figures 3.2 and 3.3 show the arrangement of the experimental test equipment outside and inside the soundproof laboratory respectively.
Several arrangements were considered in the soundproof room in order to have an efficient setup for EEG experiments. An adequate space was provided, with a chair and table set up under the observation window. All the subjects are seated in a comfortable chair with their back to the double doors. The bio-signal amplifier is placed on a small table behind the subject near the entrance as shown in Figure 3.3. The lighting in the soundproof room is switched off during the test to minimize electrostatic interference and unwanted artifacts. Lighting would also acts as a visual stimulus for the subject which is not acceptable for auditory EEG experiments. Detailed specifications of the systems can be found in Appendix A.1.
3.3 Hearing Test Standards and Protocols

A pure tone audiometric hearing test was used to evaluate the hearing thresholds for each subject. Pure tone audiometry is the principal hearing test that is most often carried out to identify the hearing threshold levels of an individual at various frequencies. This method was chosen as the evaluation of the hearing thresholds as it utilises air-conduction hearing. This is the same method which is used to evoke the auditory event-related potentials.

The hearing test was designed to conform to the relevant Australian Standards. The Standards which applied to the hearing test were [246, 248, 249]:

- AS IEC 60645-1:2000 Electroacoustics – Audiological Equipment (Part 1: Pure-Tone Audiometers);

According to these Standards, the conventional frequency range for audiometric testing is 500 Hz to 8 kHz. The hearing test used for this research conforms to this requirement and the examination was conducted by transmitting pure tones at frequencies of 500, 750, 1000, 1500, 3000, 5000 and 8000 Hz to a single ear. The amplitude of each tone varied from -20 dB up to 100 dB to determine the hearing threshold. The subject
responded to each frequency by pressing the space bar when they could just hear the tone. The test completes the frequency sequence in the left ear first, followed by the right ear. The hearing test is conducted using a computer with specialised software and circumaural headphones. Digital Recordings ‘Digital Audiometer Professional’ [250] was the software application utilized to conduct the hearing test. The software aligns well with the requirements of the Australian Standards for audiological equipment. Appendix B.1 has detailed specifications of the audiometer that was employed for this research. A circumaural AKG-K271 magnetically headphone shielded as shown in Figure 3.4 was used for the hearing test and EEG recording. This headphone meets the requirements of the standards. It also covers a much broader frequency range that may be useful for the stimuli of the EEG recording and the frequency response of the headphone is given in Figure 3.5. It is also a light headphone (240g), contributing to subject comfort. Specification of the headphone can be found in Appendix B.2.

Figure 3.4 AKG K271 Studio Headphones [251]
A sound level meter was used to calibrate the headphones and to measure the sound pressure level of the testing room. It could measure both A and C frequency weighting characteristics and claims a good degree of accuracy. Specifications of the sound level meter can be found in Appendix A.3.

Figure 3.5 AKG K271 Studio Headphones [252]

Figure 3.6 Radio Shack Sound Level Meter [253]
Before commencing with experiments, the headphone and the audiometer are calibrated at 60 dB in accordance with the standards. This ensures that for every frequency each tone will be presented at the correct decibel level. This step must also be followed to meet safety requirements to ensure that test subjects’ hearing cannot be adversely affected. The frequency dependent volume calibration is also necessary to ensure consistency between experiments, i.e. that all subjects hear the same volumes. Details of the calibration method can be found in Appendix C.1.

The environment for the conduction of the test is also governed by the Australian Standards. The maximum allowable background noise level for audiometric testing is 30 dB for both A and C frequency weightings (as measured on the sound level meter). This ensures that the soundproof facility is acceptable for an audiometric test and the extraphysiologic artifacts due to the acoustical environment are maintained to a tolerable level. The background noise in both A and C weightings in the sound proof room remained below 25 dB.

The environment temperature is required to be within the range of 15 degrees Celsius to 35 degrees Celsius. The soundproof laboratory, as described in Section 3.2, conforms to these requirements. The room temperature was maintained at 25 degrees Celsius.

The Standards also require that particular details be recorded and that certain instructions are provided to subjects prior to conducting the hearing
test. It is important to make sure that the participant has signed the consent form (refer to Appendix D.1) and understood all components. This ensures that the subject is participating willingly and does not feel pressured. The completion of the Hearing Test Checklist for Audiologists/Professionals (refer to Appendix D.2) further illustrates that the equipment setup and subject preparation conform to the standards.

The standards also specify that subjects must avoid significant noise exposure 24 hours prior to audiometric testing as it can temporary elevate thresholds. Prior to the hearing test, subjects’ were informed to avoid, for at least 24 hours, unnecessary or significant noise exposure as this can temporarily elevate their thresholds.

According to the Australian Standards, there are also specific rules concerning the preparation of each subject. These rules pertain to the secure fit of the headphone, and the unobstructed path between the headphone and the auditory canal. In order to abide by these rules, the following steps are taken:

- The headphones are carefully aligned with the entrance to the ear canal and adjusted to provide a secure fit to each subject.
- Any object such as jewellery or hair which obstructs the placement of the headphones and path between the headphone and the ear is repositioned, and the researcher ensures that the left and right positions for the headphones correspond with the correct ears.
• The subjects are reminded to keep their movement to a minimum and not to touch the headphone while the hearing test is conducted.

A maximum testing time of 20 minutes was chosen, with a break between the testing of ears. This testing time and schedule enables the subjects to concentrate on the test. Appendix E details all the procedures established for conducting a hearing test.

3.4 Stimulus Construction

Stimuli are constructed using MATLAB R2012b to evoke the middle latency response related to binaural hearing. All the stimuli were constructed as 60 dB Blackman windowed sinusoidal waves with frequencies of 500 Hz and 1000 Hz and a duration of 18 ms. Phasic and anti-phasic signals are applied to the ears. Limited bandwidth and broad bandwidth noise was added to a number of signals. EEG signals were captured from 20 ms until 100 ms after the start of the stimuli, the region of the MLR. To facilitate averaging for removal of artifacts, the stimuli are repeated at regular periodic intervals. The above choices are further motivated in the sections below.

3.4.1 Frequency of Stimuli

The choice of 500 Hz and 1000 Hz frequencies for the stimuli depends on a number of factors. Both the frequencies are in the normal hearing range and
are widely used in the standardized hearing tests [246, 248, 249]. In addition, 500 and 1000 Hz frequencies stimuli are commonly used to evaluate the auditory pathway [160, 204, 210, 254-256].

3.4.2 Windowing of Stimuli

The Blackman window was applied to ensure a smooth acoustic transition between the silence duration and the start or end of the stimulus. It is also used to reduce spectral splatter of the signal [257]. If no windowing is used the abrupt starting and ending of the stimuli will result in a distorted sound, usually perceived as a click.

3.4.3 Duration of Stimuli

The duration of the stimuli needs to be long enough so that a subject can clearly identify the difference between in-phase and out of phase signals. This will ensure that binaural processing has occurred, and that this may therefore be present in the EEG. Stimulus duration of 18 ms was selected. At a stimulus frequency of 500 Hz this results in 9 cycles of a sinusoid.
3.4.4 Homophasic and Antiphasic Stimuli

As discussed in Chapter 2, there are two important cues for binaural processing:

- inter-aural time difference (ITD) also known as inter-aural phase difference (IPD) for periodic signals; and
- inter-aural level difference (ILD).

The inter-aural phase difference (IPD) was selected for this research to develop, test and evaluate an approach to detect binaural processing in the human brain objectively. 500 and 1000 Hz were selected as frequencies for the stimuli. As discussed in Section 2.10.1, due to the physics of the process, the IPD is a better marker for binaural processing in the region up to 2000 Hz than the ILD. Furthermore, IPD can be easily implemented for ±180 degrees of phase difference through simple reversal of the signal and/or noise in one of the channels [116, 258]

Another reason implementing a phase shift ±180 degrees is that it enables the replication of the configurations stated by Durlach’s “Equalization-Cancellation” model used to determine the BMLD. In his model he suggests that the auditory system subtracts the total signal of one channel from the other. When an interaural phase difference of ±180 degree is applied between the left and right stimulus, the resulting components should gain detectability [123, 140].
3.4.5 Presentation of Stimuli

The stimuli are presented in three configurations:

- Blocks of 10 antiphase and 10 homophase tones, 50 times each.
- 500 repetitions of antiphase tones followed by 500 repetitions of homophase tones.
- Randomized antiphase and homophase tones of 1000 times in total.

These three different configurations of the stimuli all result in 500 trials of homophase stimuli and 500 trials of antiphase stimuli. The three configurations of stimuli will inform whether the order of presentation of stimuli has an impact on the auditory evoked potentials.

3.4.6 Influence of Noise

The influence of noise on the AEP is also investigated. The stimuli were masked by noise with a Gaussian amplitude distribution and a bandwidth of 100 to 900 Hz, 20 to 8000 Hz and 20 to 20000 Hz respectively. Gaussian noise was selected as it enables easier detection of the binaural cues [259]. The noise band of 100 to 900 Hz was selected based on the findings of Bourbon [260], who conducted the first band-widening experiment utilizing monaural and binaural processing conditions. He found that the detection thresholds in the N₀Sₐ condition increased 3 dB per doubling of bandwidth till the masker was about 400 Hz wide. For wider bandwidths of the masker, the
threshold remains essentially constant as bandwidth was increased further [139, 260]. The 20 to 8000 Hz band noise was selected as a comparison, using a broadband spectrum covering the important frequencies for speech. The range is further extended to include the full spectrum of human hearing, from 20 Hz to 20 kHz, for the final experiments of this research. The signal to noise ratio for all stimuli with noise is chosen as 5 dB as it was sufficient for the subject to hear the tone in the background noise.

### 3.4.7 Measurement Location

All auditory evoked potential signals are measured from the cortical position Cz [176, 177] with a reference electrode placed on the left earlobe and ground electrode located on the forehead [172]. The cortical position Cz is chosen as the signals at the cortical region are much larger in amplitude than the signals in other regions of the skull [7].

### 3.5 Stimulus Generation

This research is focused on the auditory evoked potential in the middle latency response (MLR). The MLR occurs from 20 ms post-stimulation until 80 - 100 ms after the onset of an acoustic stimulus [172, 184-186]. To be able to average the signals to remove artifacts from the EEG, the starting point of each epoch must be accurately known. A trigger signal is therefore generated along with the stimuli, which can be captured together with the EEG signals. In this case synchronization between the two auditory stimuli
and the trigger signal is very important. To achieve this, a Creative Sound Blaster Audigy 4 sound card was selected for this research. This is a professional soundcard capable synchronising multiple channels [261]. It can support up to 96 KHz sampling frequency with low latency. The specifications of the soundcard can be found in the Appendix A.4.

![Creative Sound Blaster Audigy 4](image)

**Figure 3.7 Soundcard-Audigy 4 [261]**

An audio cable was built to connect the output of the sound card to the headphone and EEG-USB amplifier. Since the maximum input into the amplifier is 250 mV whereas the maximum output of the soundcard is 2.5 volts peak to peak, a voltage divider was used to step down the voltage ten times.
3.6 Methodology of Electroencephalography

After completion of the hearing test (procedure for conducting hearing test is given in appendix C, D and E), the next phase of the research is focussed on eliciting and recording auditory evoked potentials relating to binaural hearing. The stimulus which has been discussed in the previous section will be used to elicit these potentials. In order to measure these phenomena, it was necessary to develop a method for acquisition. In order to conduct the EEG experiments on normal hearing subject, a series of steps need to be completed to ensure accurate and reliable data acquisition. All EEG experiments are performed in a soundproof laboratory and the maximum background noise level for EEG testing is kept below 25 dB in order to minimize the noise artifacts. It is a requirement that the testing be done in a soundproof laboratory [262]. The lights in the laboratory are also switched off in order to reduce 50 Hz artifacts. Scalp-electrode impedance levels must be kept below 5 kΩ to prevent poor quality EEG data [251, 263]. In order to obtain low impedance measurements the following method was used:

- A cotton bud was used to gently move the subjects’ hair out of the way to achieve better contact;
- Prior cleaning of scalp site with Theodor-Korner- Apotheke abrasive electrode gel;
- Electrodes were checked for any damage or surface imperfections; and
- Nihon-KohdenElefix conductive paste was applied on the electrode cups.
Three different sized skullcaps are available and care should be taken to choose an appropriate size to fit the subjects head. The correct sized head cap was fitted carefully for each subject as this assists in achieving lower impedance levels.

After positioning the headphones over the cap, the impedance levels are monitored to check whether they were acceptable for EEG collection. If the values are in excess of 5 kΩ, the headphones are removed and the electrodes are repositioned until the impedance was low enough. The electrode placement is done according to the 10-20 international electrode placement system [22]. The EEG signals are measured from the cortical position Cz and the forehead is used as the ground with the reference electrode attached to the left earlobe. Once all the electrodes are placed on the skull, they are connected to the amplifier. If they exhibit a low impedance reading, the EEG experiments can start. The subject preparation is similar to the hearing test as described previously in this chapter. A detailed description of the process of subject preparation for electroencephalography can be found in Appendix F.

### 3.7 Data Acquisition

Since the auditory evoked potentials are very small in amplitude, a data acquisition system that has minimal interference from external noise sources is required. The signal also needs to be amplified to around 1000 times [21, 22]. In order to develop a suitable data acquisition system, it is important to
select both the hardware and the software that satisfies the reliability and the accuracy requirements for capturing the data [264]. A diagram displaying the final setup used for the experiments is shown in Figure 3.7.

![Diagram of Data Acquisition Set-up](image)

**Figure 3.8 Data Acquisition Set-up**

Computers which are used for stimulus delivery and EEG has the same specifications with Intel(R) Core (TM) i7 CPU 870 @ 2.90 GHz 3.07 GHz processing ability. Specifications of the computer can be found in appendix A.1 Data acquisition and processing of the EEG data is accomplished using MATLAB based software. EEG recording is accomplished using a g.tec USB biosignal amplifier (g.USBamp). The amplifier has the capability of recording 16 24 bits channels simultaneously. The device can be connected directly to the PC via USB [265]. It consists of four blocks with four channels each (refer
to Figure 3.8) and a ground and reference for each block to eliminate the interference from each recorded signal.

Figure 3.9 g.USBamp biosignal amplifier [260]

The amplifier works with passive and with active electrodes. Internal digital bandpass and notch filters are available. Digital inputs and outputs allow the recording of trigger channels together with the biosignal channels to facilitate the analysis of the results. It has a built-in calibration unit and impedance checking facility as well. The input range is ± 250 mV and this ensures that EEG signals can be recorded without saturating. It has a short-cut input that allows connecting the amplifier inputs quickly to ground potential to protect it against overflows. Detailed specifications of the USB amplifier can be found under the product specification of g.USBamp and can be retrieved from the manufacturer website [265]. An overview of the specifications is given in appendix A.6.

A major challenge with the measurement of auditory evoked potentials is ensuring channel synchronization. It is not often that processes in Microsoft windows function in parallel due to the multi-tasking nature of the operating
system and thus timing is a major technical challenge. If the presentation of the stimuli is delayed by the operating system, an incorrect segment of the EEG may be extracted. Trigger-synchronized averaging is the most important factor in extracting the auditory evoked potential. Averaging as shown in Figure 3.9 is a common signal processing technique used in the area of event related potentials. It is an important step in revealing the underlying auditory evoked potential from the spontaneous electrical activity that is also recorded. In this application it is designed to improve the signal to noise ratio of the AEP. The raw EEG signal is the resultant of many brain processes occurring simultaneously, however, only the signal relating to the auditory stimuli is to be considered. Therefore all other signals in the EEG apart from the AEP are regarded as noise. The magnitudes of the auditory evoked potentials are extremely small relative to the random bioelectric activity. These potentials are also time-locked based on the onset of stimulus whereas; the spontaneous bioelectric activity is not. As unwanted signals or noise occur randomly, the averaging process will ensure that the amplitude tends towards zero leaving the evoked potential [266]. Averaging is to be taken over a sufficiently number of trials [267], and 500 was selected for this research [36].
Coupling a trigger signal to the delivery of the stimulus is one method of determining the starting points. Accurate trigger-synchronized averaging of a particular event potential can only be valid if the peaks of the trigger signals are detected correctly and the soundcard channels are well synchronized. The initial trigger signal was a rectangular pulse train with a pulse duration equivalent to the stimulus duration. This pulse duration was difficult to achieve as the soundcard was not a reliable function generator. Constant voltage over the required pulse duration was not achievable due to a gradual decay. This produced distortion (pseudo-peaks) in the trigger signal which at times were incorrectly assumed to be the beginning of the trigger signal. This is evident in Table 3.1 and 3.2 where the same trigger signal was played simultaneously through four channels of the soundcard. The peak of each trigger was recorded in sample points and determined in MATLAB based on the maximum values of the function. Based on the stimuli and sampling rate used, the number of sample points between stimuli was 9600. The perfect result would mean that the distance between peaks for all peaks should be
9600 sample points and the difference in peak position between all channels should be zero. Table 3.1 demonstrates that the mismatch between soundcard channels for trigger detection of the longer duration trigger rectangular pulse ranges between 9.5% and 26%. To improve the reliability of the trigger detection, the trigger signal was redesigned. Distortion was found to significantly decrease with shorter pulse duration. The trigger signal was redesigned to a short pulse duration representing the start of each stimulus. The redesign produced a significantly cleaner trigger signal. Table 3.1 and 3.2 also show the difference between the onsets of each trigger signal in all channels. Automatic detection was performed using a difference method. The use of MATLABs “diff()” function enabled an approximation of the discrete derivative of the trigger channel. The maximum derivatives were then found which correlate with the impulse trigger signals. These peak detected points were used as the starting points of the stimuli for each trial. Table 3.2 demonstrates that the shorter impulse trigger signal is more reliable in terms of correct peak detection than the initial longer duration trigger. The method of using a short duration rectangular pulse train as a trigger signal proved to be more reliable showing only 4% mismatch between two channels. The high correlation between the simultaneous trigger onsets in the four channels shows the high synchronization between separate channels.

After the peaks are successfully detected, the data are segmented into individual epochs based on the peak positions. The epochs are then
averaged to produce a better signal to noise ratio for both antiphasic and homophasic cases.

**Table 3.1 Peak Detection of Long Duration Trigger Channel for 200 Trials**

<table>
<thead>
<tr>
<th>Peak</th>
<th>Channel1</th>
<th>Channel2</th>
<th>Channel3</th>
<th>Channel4</th>
<th>Difference between peaks (Channel1)</th>
<th>Difference Between Channels</th>
</tr>
</thead>
<tbody>
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<td>56017</td>
<td>56017</td>
<td>56017</td>
<td>56016</td>
<td>9598</td>
<td>0   1  0</td>
</tr>
<tr>
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<td>65615</td>
<td>65617</td>
<td>65615</td>
<td>9599</td>
<td>0   2 -2</td>
</tr>
<tr>
<td>3</td>
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<td>75214</td>
<td>75214</td>
<td>75214</td>
<td>9600</td>
<td>0   0  0</td>
</tr>
<tr>
<td>4</td>
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<td>84814</td>
<td>84814</td>
<td>84814</td>
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Total Non-Zero Elements: 33 or 19 or 52

16.5% 9.5% 26%
Table 3.2 Peak Detection of Short Duration Trigger Channel for 200 Trials

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<th>Channel3</th>
<th>Channel4</th>
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<th>1 and 2</th>
<th>3 and 4</th>
<th>2 and 3</th>
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</tbody>
</table>

Total Non-Zero Elements 8 or 1 or 8 4% 0.5% 4%

All recorded EEG data are stored on an external hard-drive. Signal processing is performed offline. The software package used to process the data is MATLAB R2012b. Once the experiments are completed all the electrodes are unplugged from the USB bio-signal amplifier and gently taken out individually.
3.8 Signal Processing

After the neural activity has been captured from the electroencephalogram, the next phase is to analyse the signals. The interface between the biosignal amplifier and the computer enables data to be saved in such a way that it can be easily imported into MATLAB R2012b for processing.

The captured data, including the captured EEG channels, the trigger channel and the stimulus channels are saved in MATLAB format. Stimulus channels are captured to ensure that the stimulus is consistently delivered in the correct configurations to both the left and right ears. A trigger channel is used as part of the trigger-synchronised averaging process of EEG and Channel Cz data are captured to record relevant neural activity.

Once the EEG signal along with the stimuli and the trigger signal are saved for analysis, the next part of the signal processing is to find the starting point of each individual epoch from actual EEG signal. In order to accurately determine the location of each of the elicited potentials associated with in-phase and outphase, a trigger signal is delivered to correspond to the delivery of the stimuli. The location of the narrow rectangular pulses used as trigger signal is determined using a peak detection algorithm in MATLAB. The identification of the peaks is then confirmed visually to ensure the position was accurately determined.
The location of the start of each delivered stimulus is determined based on the trigger input and then the window for each epoch is constructed. The window is based on the total of the signal and silence duration. Each window has a duration of 80 ms and the recorded neural activity is segmented from 20 ms to 100 ms post-stimulation to achieve the middle latency response of the brain that lies in this time frame of the evoked potential. After dividing the recorded neural data into epochs, artefact rejection is implemented. Epochs which contained significant artefacts are rejected, epochs containing absolute amplitudes in excess of ±150 µV are excluded from the analysis [268]. After these epochs are excluded, the remaining epochs are averaged. This method increases the signal to noise ratio of the auditory evoked potential.

Many algorithms have been developed for processing EEG signals. The operations include, but are not limited to, time domain analysis, frequency domain analysis, wavelet analysis, neural network modelling etc. Also, several algorithms have been developed to visualize the brain activity from EEG images. A common technique used in processing EEG signals is the Fourier Transform. The Fast Fourier Transform was chosen based on some of the reasons stated below:

- Fourier analysis easily calculates the frequencies and the amplitudes of those frequencies which make up the signal which provides a broad overview of the characteristics of the signal. Dominant frequencies of the signal can easily be seen using FFT. It is a numerically efficient analysis technique for finding the different frequency components of the EEG signal [269, 270].
• Fourier analysis is very effective in problems dealing with frequency location.

• Processing with Fast Fourier Transformation (FFT) is faster. A Fast Fourier Transform is only one of the many available algorithms and techniques for achieving the transformation with a considerably smaller amount of mathematical operations and hence is much faster to implement.

![Figure 3.11 EEG signal and the Frequency components](image)

Figure 3.11 EEG signal and the Frequency components [271]

The FFT allows us to see clearly what frequency components dominate the EEG, as can be seen in figures 3.11.
3.9 Summary

This chapter describes the experimental methodology. The testing environment used for this research was developed according to Australian standards and the experiments were conducted in an electrically shielded soundproof laboratory. Hearing test standards and protocols were adopted for the experiment and pure tone audiometric testing was used to evaluate the hearing thresholds for each subject. Stimuli were constructed to evoke the middle latency response related to binaural hearing. Choices of variable such as, sound level in dB, frequency of stimuli, windowing of stimuli, duration of stimuli, phasic difference of the stimuli, order of presentation of stimuli, and choice of background noise embedded in the stimuli were explained. The subjects are presented with the stimuli and the EEG signal is captured. The method for performing electroencephalogram is described in detail in this chapter. The chapter concluded by giving an overall explanation of the algorithm and the techniques developed for EEG signal processing.
Chapter 4: Preliminary Experiments

In this chapter, the preliminary experiments of this research are discussed and analysed. Experiments are carried out to discover any electrophysiological evidence of the binaural processing in the human brain.

Section 4.1 describes the experiments to validate the experimental setup. In Section 4.2, stimuli are selected, the design of which was discussed in the Section 3.4. Section 4.3 describes the development of algorithms to extract markers for binaural processing. Initial results and the accuracy of the algorithms are also discussed. Statistical tests of the preliminary results are described in Section 4.4. The algorithm is further refined in Section 4.5 with the introduction of a normalisation step. The chapter is concluded with a discussion of results in Section 4.6.

4.1 Watermelon Experiments

To determine if there is any interference produced by the hardware, software or the testing environment, an experiment was conducted by testing an EEG on a watermelon as they are devoid of brain activity but have a high water content to allow the signals to travel and communicate freely. The experiment replicates the experimental setup used to collect an EEG in practice. The signal from the electrode was amplified with the g.USBamp biosignal
amplifier, and Matlab was used to record the signals and the data capturing rate was set at 4800 Hz. The experimental setup can be seen in Figure 4.1.

Figure 4.1 EEG on Phantom Head (Watermelon)

Figure 4.1 shows the arrangement of the experimental setup and the signal was captured from the electrode attached (named as Channel Cz) on the watermelon. The amplitude of the signal captured was in the nano-volt range (approximate 20-30 nV) and was generated from the internal circuitry of the system. The signal which was sampled at 4.8 KHz showed very minute interference with no EEG which demonstrates that the testing facility does not exhibit any major interference.
4.2 Stimuli for Preliminary Experiments

Twenty subjects in total participated in the preliminary experiments and the subjects were distributed in three separate groups of size 8 (Group A), 8 (Group B) and 4 (Group C). In total 8 experiments were conducted initially to analyse the middle latency response based on binaural stimuli. Detailed description of the stimulus used for each experiment is presented below.

**Group A: Experiment 1, 2, 3 and 4**

In Group A there are 4 experiments conducted on eight subjects. For experiment 1 and 2 the stimuli were made of a pure tone of 500 Hz frequency having a sound pressure level of 60 dB. The pure tone was then passed through a Blackman window in order to reduce spectral splatter in the frequency lobes. In experiment 1 a block of 10 antiphasic stimuli followed by 10 homophasic stimuli were presented for a total of 1000 trials and in experiment 2 the stimuli were presented as a block of 500 antiphasic stimuli followed by 500 homophasic stimuli. In both cases there are 1000 trials of binaural stimuli used out of which 500 antiphasic and 500 homophasic related event potentials are captured for averaging.

In experiment 3 and 4 the stimuli used were also a 60 dB 500 Hz Blackman windowed pure tone masked with two different conditions. In both the cases the stimuli were embedded with continuous Gaussian noise and a total of 1000 trials were conducted as blocks of 10 antiphasic and 10 homophasic
trials. In experiment 3 the tone was masked by Gaussian noise having bandwidth in the range of 100 and 900 Hz whereas in experiment 4 the tone was masked by a Gaussian noise having bandwidth in the range of 20 Hz and 8 kHz. The signal to noise ratio for the stimuli in both the cases was 5 dB.

**Group B: Experiment 1 and 2**

In Group B there are 2 experiments, conducted on eight subjects. The stimuli were a 60 dB Blackman windowed pure tone and the frequency of the tone was changed to 1000 Hz. In the first experiment, pure tones as stimuli were presented whereas in the second experiment the pure tone was masked by Gaussian noise with a bandwidth in the range of 20 Hz to 8000 Hz. The signal to noise ratio for the stimuli was 5 dB. In both cases a block of 10 antiphasic stimuli followed by 10 homophasic stimuli were presented for a total of 1000 trials. This provided 500 antiphasic and 500 homophasic related event potentials to be captured for averaging.

**Group C: Experiment 1 and 2**

In Group C another two experiments were conducted on another group of 4 normal hearing subjects. The relationship of the in-phase and out-phase AEP was further explored by randomising the presentation of homophasic and antiphasic tones in the presence of noise. In the first experiment, a Blackman windowed pure tone of 500 Hz frequency was masked by Gaussian noise
with a bandwidth in the range of 100 and 900 Hz whereas in the second experiment, a Blackman windowed pure tone of 500 Hz frequency was masked by Gaussian noise with a bandwidth in the range of 20 to 8000 Hz. In both cases the signal to noise ratio for the stimuli was 5 dB and the stimuli were presented randomly for a total of 1000 trials. These trials included 500 homophasic and 500 antiphasic tones with randomised presentation.

All experiments presented a total of 1000 stimuli for signal averaging and the auditory evoked potential was measured from the cortical position Cz [176, 177] with a reference electrode placed on the left earlobe and ground electrode located on the forehead [172]. Table 4.1 below shows the summary of all the stimuli used for preliminary experiments.
Table 4.1 Group A Stimuli

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<tr>
<th>Group A Stimuli Description</th>
<th>Experiments 1 and 2</th>
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</tr>
</thead>
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<tr>
<td>Number of Subjects</td>
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<td></td>
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<tr>
<td>Pure tone</td>
<td>Frequency - 500Hz</td>
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</tr>
<tr>
<td>Pure tone duration</td>
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<tr>
<td>Silent Duration</td>
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<tr>
<td>Windowing Technique</td>
<td>Blackman Window</td>
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<td>Sound Pressure Level</td>
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<td>Single Trial Duration</td>
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</tr>
<tr>
<td>Number of trials</td>
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<tr>
<td>Sequence of Trials for Experiment 1</td>
<td>Blocks of 10 antiphasic stimuli (180 degree out of phase) and blocks of 10 homophasic stimuli (in same phase), 50 times each.</td>
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<tr>
<td>Sequence of Trials for Experiment 2</td>
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<table>
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<td>Pure tone</td>
<td>Frequency - 500Hz masked with noise</td>
</tr>
<tr>
<td>Pure tone duration</td>
<td>18 ms</td>
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<tr>
<td>Silent Duration</td>
<td>200 ms</td>
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<td>Windowing Technique</td>
<td>Blackman Window</td>
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<td>Sound Pressure Level</td>
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<tr>
<td>Single Trial Duration</td>
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<tr>
<td>Number of trials</td>
<td>1000</td>
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<tr>
<td>Sequence of Trials</td>
<td>Blocks of 10 antiphasic stimuli (180 degree out of phase) and blocks of 10 homophasic stimuli (in same phase), 50 times each.</td>
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<td>Noise Bandwidth for Experiment 4</td>
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<tr>
<td>Signal to Noise Ratio</td>
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Table 4.2 Group B Stimuli

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<td>Pure tone duration</td>
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<td>Number of trials</td>
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<tr>
<td>Sequence of Trials 1</td>
<td>Blocks of 10 antiphasic stimuli (180 degree out of phase) and blocks of 10 homophasic stimuli (in same phase), 50 times each.</td>
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</tbody>
</table>

| Number of Subjects          | 8                                                                           |
| Pure tone                   | Frequency - 1000Hz masked with noise                                        |
| Pure tone duration          | 18 ms                                                                       |
| Silent Duration             | 200 ms                                                                      |
| Windowing Technique         | Blackman Window                                                             |
| Sound Pressure Level        | 60 dB                                                                       |
| Single Trial Duration       | 218 ms                                                                      |
| Number of trials            | 1000                                                                        |
| Sequence of Trials          | Blocks of 10 antiphasic stimuli (180 degree out of phase) and blocks of 10 homophasic stimuli (in same phase), 50 times each. |
| Noise type                  | Gaussian Noise                                                              |
| Noise Bandwidth for Experiment 4 | 20 – 8000 Hz                     |
| Interaural Phase Delay of noise | 0                          |
| Signal to Noise Ratio       | 5 dB                                                                        |
### Table 4.3 Group C Stimuli

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<td>Pure tone duration</td>
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<td>Silent Duration</td>
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<td>Blackman Window</td>
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<td>Sequence of Trials</td>
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<td>Noise Bandwidth for Experiment 1</td>
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<tr>
<td>Noise Bandwidth for Experiment 2</td>
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<tr>
<td>Interaural Phase Delay of noise</td>
<td>0</td>
</tr>
<tr>
<td>Signal to Noise Ratio</td>
<td>5 dB</td>
</tr>
</tbody>
</table>
4.3 Data Processing Algorithm

An example of the auditory stimuli, the trigger signal and the EEG response elicited by the stimuli captured by the amplifier is shown in the Figure 4.2.

Figure 4.2 Example of a captured stimuli, trigger and EEG response.

Once the auditory evoked response elicited by homophasic and antiphasic stimuli were captured, the EEG signals captured from 20 ms to 100 ms post stimulation were extracted. Epochs which contained significant artefacts were rejected, epochs which contained absolute amplitudes in excess of ±150 µV were excluded from the analysis and then the remaining epochs are averaged to obtain a good SNR. Further discussion of segmenting, rejecting and averaging EEG epochs can be found in Section 3.7. Figure 4.3 shows a
sample of averaged EEG epoch based on homophasic and antiphasic auditory stimuli.

Figure 4.3 Sample of averaged EEG epoch - homophasic and antiphasic stimuli

Figure 4.3 shows difference in shape between the averaged homophasic EEG and the averaged antiphasic EEG [272] along with the peaks (Na, Pa, Nb and Pb) of the middle latency response. Please refer to sec 2.16.2.2 for further information on the components of the MLR. A Fast Fourier Transformation was conducted over the averaged epochs for both homophasic and antiphasic auditory evoked potentials for each subject of Experiment 1 to find the energy distribution of the frequency potentials of the averaged EEG signal in the frequency spectrum. Figure 4.4 to Figure 4.11 below shows the frequency spectrum for both averaged in-phase
(homophasic) and out-phase (antiphasic) EEG epochs of all the eight subjects of Group A, Experiment 1. FFT was computed on the EEG data to look into the amplitude and phase of the spectral peak frequency in the first 200 Hz of the EEG signal with a resolution of 2.3 Hz.

Figure 4.4 Frequency spectrum of averaged in-phase/out-phase EEG epoch – Subject 1, Group A Experiment 1
Figure 4.5 Frequency spectrum of averaged in-phase/out-phase EEG epoch – Subject 2, Group A Experiment 1

Figure 4.6 Frequency spectrum of averaged in-phase/out-phase EEG epoch – Subject 3, Group A Experiment 1
Figure 4.7 Frequency spectrum of averaged in-phase/out-phase EEG epoch – Subject 4, Group A Experiment 1

Figure 4.8 Frequency spectrum of averaged in-phase/out-phase EEG epoch – Subject 5, Group A Experiment 1
Figure 4.9 Frequency spectrum of averaged in-phase/out-phase EEG epoch – Subject 6, Group A Experiment 1

Figure 4.10 Frequency spectrum of averaged in-phase/out-phase EEG epoch – Subject 7, Group A Experiment 1
Findings:

- Based on visual inspection of the graphs displayed above, EEG signals, elicited by homophasic and antiphasic stimuli, have most of their energy in frequencies up to 100 Hz. With the exception of figure 4.5 (subject 3) and figure 4.10 (subject 8), energy of the EEG signal above 100 Hz is small. This corresponds with the findings that the EEG signal of a human being lies in the range of 0.5 to 100 Hz [37].

- A process need to be developed to separate in-phase and out-phase based on the amplitude in the frequency spectrum. As can be seen in some figures the amplitudes of the in-phase are above, while in other figures the amplitudes of the out-phase are above for certain frequency ranges.
The amplitude spectrum exhibits “lobes”, suggesting that there may be a limited number of dominant frequencies driving the processes, while variations around these dominant frequencies occur. These dominant frequencies may be characterised by the peak value of each lobe. The amplitude and phase of this peak may capture the dynamics of the underlying process.

It is therefore suggested that the peak values of the spectrum is used for further analysis. The Matlab function `findpeaks()` was used to find the amplitude and the frequency of the spectral peaks in the frequency domain. The function returns local maxima or peaks of the input data. The function works by comparing each element of data to its neighbouring values. If an element of data is larger than both of its neighbours or equals infinity, the element is a local peak. Another Matlab common function `angle()` was used to calculate the phase angle of the spectral peaks.

### 4.3.1 First 15 peaks in the frequency spectrum

The first 15 peaks found in the frequency spectrum of the averaged EEG epoch for both homophasic and antiphasic stimuli are selected to extract the amplitude, phase and the frequency of those peaks. The purpose of choosing 15 peaks is to ensure that a wide range of frequencies is covered, including the important frequencies of up to 100 Hz. Amplitude and phase for all the 15 spectral peaks from each subjects’ averaged EEG epoch (homophasic/antiphasic) are plotted individually having frequencies of all the
15 peaks as the horizontal axis common against the vertical axis of amplitude and phase of the peaks as shown below.

Figure 4.12 Magnitude/Phase of first 15 peaks of averaged in-phase/out-phase EEG epoch - Subject 1, Group A Experiment 1
Figure 4.13 Magnitude/Phase of first 15 peaks of averaged in-phase/out-phase EEG epoch - Subject 2, Group A Experiment 1

Figure 4.14 Magnitude/Phase of first 15 peaks of averaged in-phase/out-phase EEG epoch - Subject 3, Group A Experiment 1
Figure 4.15 Magnitude/Phase of first 15 peaks of averaged in-phase/out-phase EEG epoch - Subject 4, Group A Experiment 1

Figure 4.16 Magnitude/Phase of first 15 peaks of averaged in-phase/out-phase EEG epoch – Subject 5, Group A Experiment 1
Figure 4.17 Magnitude/Phase of first 15 peaks of averaged in-phase/out-phase EEG epoch – Subject 6, Group A Experiment 1

Figure 4.18 Magnitude/Phase of first 15 peaks of averaged in-phase/out-phase EEG epoch – Subject 7, Group A Experiment 1
**Findings:**

- A trend can be found regarding the magnitude of the spectral peak frequency for all the plots shown in Figure 4.12 to Figure 4.19. The amplitude decreases with increasing frequency whereas there is no consistent trend found for the frequency or phase;

- The peak frequency for in-phase and out-phase may be different, with some following each other closely, while others are more spread out; and;

- In the higher frequency range, in particular above 80 - 100 Hz, several peaks seem to be “missed” by either the in-phase or out-phase. This is probably due to the low energy level at these higher frequencies.
The first four peaks from the frequency spectrum were selected for further analysis. The spectral peaks of these dominant frequencies carry most of the energy of the signal.

4.3.2 Analysis of the first four dominant frequency peaks

From this section of this chapter, a number of tables will be used for data presentation of experiments. Table 4.4 below shows the abbreviated notation, along with the explanation, used in the various tables. The reader can refer to this table when needed.
Table 4.4 Notations used for data table

Subject: Subjects are numbered 1 to 8.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Notations Used for Data Table</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak 1</td>
<td>A1Out: Antiphasic Amplitude [µV] for Peak 1</td>
</tr>
<tr>
<td></td>
<td>F1Out: Antiphasic Frequency [Hz] for Peak 1</td>
</tr>
<tr>
<td>Peak 2</td>
<td>A2Out: Antiphasic Amplitude [µV] for Peak 2</td>
</tr>
<tr>
<td></td>
<td>F2Out: Antiphasic Frequency [Hz] for Peak 2</td>
</tr>
<tr>
<td>Peak 3</td>
<td>A3Out: Antiphasic Amplitude [µV] for Peak 3</td>
</tr>
<tr>
<td></td>
<td>F3Out: Antiphasic Frequency [Hz] for Peak 3</td>
</tr>
<tr>
<td>Peak 4</td>
<td>A4Out: Antiphasic Amplitude [µV] for Peak 4</td>
</tr>
<tr>
<td></td>
<td>F4Out: Antiphasic Frequency [Hz] for Peak 4</td>
</tr>
<tr>
<td>Peak 1</td>
<td>PH1Out: Antiphasic Phase [degrees] for Peak 1</td>
</tr>
<tr>
<td></td>
<td>PH1In: Homophasic Phase [degrees] for Peak 1</td>
</tr>
<tr>
<td>Peak 2</td>
<td>PH2Out: Antiphasic Phase [degrees] for Peak 2</td>
</tr>
<tr>
<td></td>
<td>PH2In: Homophasic Phase [degrees] for Peak 2</td>
</tr>
<tr>
<td>Peak 3</td>
<td>PH3Out: Antiphasic Phase [degrees] for Peak 3</td>
</tr>
<tr>
<td></td>
<td>PH3In: Homophasic Phase [degrees] for Peak 3</td>
</tr>
<tr>
<td>Peak 4</td>
<td>PH4Out: Antiphasic Phase [degrees] for Peak 4</td>
</tr>
<tr>
<td></td>
<td>PH4In: Homophasic Phase [degrees] for Peak 4</td>
</tr>
</tbody>
</table>

Peak 1 A1Out-A1In: Amplitude Difference between the first antiphasic and the first homophasic peak.
Peak 2 A1Out-A1In: Amplitude Difference between the second antiphasic and the second homophasic peak.
Peak 3 A1Out-A1In: Amplitude Difference between the third antiphasic and the third homophasic peak.
Peak 4 A1Out-A1In: Amplitude Difference between the fourth antiphasic and the fourth homophasic peak.
Table 4.5 and 4.6 show the amplitude, frequency and phase of the first four spectral peaks extracted from the frequency spectrum of all 8 subjects’ averaged EEG epoch for both antiphasic and homophasic conditions.

**Group A Experiment 1**

Table 4.5 Amplitude, Frequency, Phase of the first 4 spectral peaks [Antiphasic AEP]

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 A1Out [µV]</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 3 A3Out [µV]</th>
<th>Peak 4 A4Out [µV]</th>
<th>Peak 1 F1Out [Hz]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 3 F3Out [Hz]</th>
<th>Peak 4 F4Out [Hz]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.2385</td>
<td>0.4697</td>
<td>0.6363</td>
<td>0.6430</td>
<td>9.3750</td>
<td>23.4375</td>
<td>35.1563</td>
<td>51.5625</td>
</tr>
<tr>
<td>2</td>
<td>0.7426</td>
<td>0.1443</td>
<td>0.2223</td>
<td>0.0969</td>
<td>9.3750</td>
<td>37.5000</td>
<td>49.2188</td>
<td>65.6250</td>
</tr>
<tr>
<td>3</td>
<td>0.9176</td>
<td>0.2862</td>
<td>0.1938</td>
<td>0.2142</td>
<td>9.3750</td>
<td>35.1563</td>
<td>46.8750</td>
<td>63.2813</td>
</tr>
<tr>
<td>4</td>
<td>0.4533</td>
<td>0.3171</td>
<td>0.2758</td>
<td>0.0722</td>
<td>9.3750</td>
<td>25.7813</td>
<td>46.8750</td>
<td>77.3438</td>
</tr>
<tr>
<td>5</td>
<td>1.1984</td>
<td>0.8331</td>
<td>0.3913</td>
<td>0.4505</td>
<td>7.0313</td>
<td>23.4375</td>
<td>35.1563</td>
<td>49.2188</td>
</tr>
<tr>
<td>6</td>
<td>0.3987</td>
<td>0.2318</td>
<td>0.1974</td>
<td>0.1429</td>
<td>11.7188</td>
<td>30.4688</td>
<td>46.8750</td>
<td>65.6250</td>
</tr>
<tr>
<td>7</td>
<td>0.3863</td>
<td>0.1471</td>
<td>0.2117</td>
<td>0.2413</td>
<td>9.3750</td>
<td>21.0938</td>
<td>35.1563</td>
<td>51.5625</td>
</tr>
<tr>
<td>8</td>
<td>1.0055</td>
<td>0.4664</td>
<td>0.3291</td>
<td>0.2155</td>
<td>9.3750</td>
<td>23.4375</td>
<td>46.8750</td>
<td>60.9375</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 PH1Out [degrees°]</th>
<th>Peak 2 PH2Out [degrees°]</th>
<th>Peak 3 PH3Out [degrees°]</th>
<th>Peak 4 PH4Out [degrees°]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>-54.6574</td>
<td>-29.3734</td>
<td>-81.8349</td>
<td>-162.9210</td>
</tr>
<tr>
<td>2</td>
<td>-27.3947</td>
<td>-96.8243</td>
<td>-116.9690</td>
<td>-175.4290</td>
</tr>
<tr>
<td>3</td>
<td>106.0257</td>
<td>143.4579</td>
<td>-162.6980</td>
<td>87.5004</td>
</tr>
<tr>
<td>4</td>
<td>-26.1033</td>
<td>-48.3330</td>
<td>-76.9247</td>
<td>-129.8750</td>
</tr>
<tr>
<td>5</td>
<td>170.7757</td>
<td>102.7913</td>
<td>102.2994</td>
<td>90.3124</td>
</tr>
<tr>
<td>6</td>
<td>-60.7430</td>
<td>-87.3732</td>
<td>-138.0350</td>
<td>153.6715</td>
</tr>
<tr>
<td>7</td>
<td>149.8964</td>
<td>83.3756</td>
<td>154.7833</td>
<td>111.9598</td>
</tr>
<tr>
<td>8</td>
<td>-50.8316</td>
<td>-70.6572</td>
<td>-45.2042</td>
<td>-52.2343</td>
</tr>
</tbody>
</table>
To determine whether these first 4 spectral peaks can indeed approximate the characteristics of the original signal, the amplitude, frequency and phase of the first four peaks (antiphasic and homophasic) tabulated above are then used to generate approximated EEG signals. The amplitude (A), frequency (f) and phase (φ) of each of the first four spectral peaks (antiphasic and homophasic) are used to generate signals using equation 4.1.

\[ y(t) = A \cdot \cos(2\pi ft + \phi) = A \cdot \cos(\omega t + \phi) \] (4.1)
A summated signal is then derived by adding the four signals for both the in-phase and the out-phase signals. All the four signals and the added signal are plotted individually on the same plot along with the averaged AEP epoch in the time domain as shown below. For each subject there are two graphs, representing the in-phase EEG epoch with its first four spectral peak frequency signal plus summated signal and the out-phase EEG epoch with its first four spectral peak frequency plus summated signal.

Figure 4.20 EEG Epoch (In-phase) with dominant frequency signal – Subject 1, Group A Experiment 1
Figure 4.21 EEG Epoch (Out-phase) with dominant frequency signal – Subject 1, Group A Experiment 1

Figure 4.22 EEG Epoch (In-phase) with dominant frequency signal – Subject 2, Group A Experiment 1
Figure 4.23 EEG Epoch (Out-phase) with dominant frequency signal – Subject 2, Group A Experiment 1

Figure 4.24 EEG Epoch (In-phase) with dominant frequency signal – Subject 3, Group A Experiment 1
Figure 4.25 EEG Epoch (Out-phase) with dominant frequency signal – Subject 3, Group A Experiment 1

Figure 4.26 EEG Epoch (In-phase) with dominant frequency signal – Subject 4, Group A Experiment 1
Figure 4.27 EEG Epoch (Out-phase) with dominant frequency signal – Subject 4, Group A Experiment 1

Figure 4.28 EEG Epoch (In-phase) with dominant frequency signal – Subject 5, Group A Experiment 1
Figure 4.29 EEG Epoch (Out-phase) with dominant frequency signal – Subject 5, Group A Experiment 1

Figure 4.30 EEG Epoch (In-phase) with dominant frequency signal – Subject 6, Group A Experiment 1
Figure 4.31 EEG Epoch (Out-phase) with dominant frequency signal – Subject 6, Group A Experiment 1

Figure 4.32 EEG Epoch (In-phase) with dominant frequency signal – Subject 7, Group A Experiment 1
Figure 4.33 EEG Epoch (Out-phase) with dominant frequency signal – Subject 7, Group A Experiment 1

Figure 4.34 EEG Epoch (In-phase) with dominant frequency signal – Subject 8, Group A Experiment 1
**Findings:** The signal constructed by using the first four spectral peaks can follow the original EEG signal closely, but without the “noisy” features of the original signal, i.e. the first four spectral peaks can be used to describe the low frequency characteristics of the original signals. An exception in this case is Subject 8 EEG epoch which is quite noisy and the signal constructed by using the first four spectral peaks had a poor fit with the original EEG signal. Since the spectral peaks of these dominant frequencies carry most of the energy of the signal therefore the first four spectral peaks are selected for further analysis.
4.3.3 Comparative analysis of the first four peaks

In order to investigate whether there is any difference in the energy carried by the first four spectral peaks of the AEP evoked by homophasic and antiphasic stimuli, amplitude differences of the spectral peaks are calculated. Table 4.3 and 4.4 already showed the amplitude and the phase of the first four spectral peaks for both the cases. Amplitude differences between the first four antiphasic and homophasic spectral peaks along with frequency and phase of those spectral peaks are now shown below in tabular format.

Table 4.7 Amplitude difference of the first 4 spectral peaks

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak1 A1Out-A1In [µV]</th>
<th>Peak2 A2Out-A2In [µV]</th>
<th>Peak3 A3Out-A3In [µV]</th>
<th>Peak4 A4Out-A4In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.1993</td>
<td>0.0727</td>
<td>0.2375</td>
<td>0.5430</td>
</tr>
<tr>
<td>2</td>
<td>0.4794</td>
<td>0.0176</td>
<td>-0.0432</td>
<td>-0.0031</td>
</tr>
<tr>
<td>3</td>
<td>0.2353</td>
<td>0.1456</td>
<td>0.0215</td>
<td>0.1568</td>
</tr>
<tr>
<td>4</td>
<td>-0.4140</td>
<td>-0.1787</td>
<td>0.0327</td>
<td>-0.0386</td>
</tr>
<tr>
<td>5</td>
<td>-0.3731</td>
<td>0.1254</td>
<td>0.0035</td>
<td>0.2770</td>
</tr>
<tr>
<td>6</td>
<td>-0.4524</td>
<td>0.0049</td>
<td>0.0549</td>
<td>0.0625</td>
</tr>
<tr>
<td>7</td>
<td>-0.1782</td>
<td>0.0075</td>
<td>0.0082</td>
<td>0.0667</td>
</tr>
<tr>
<td>8</td>
<td>0.1645</td>
<td>0.0749</td>
<td>0.1207</td>
<td>0.0758</td>
</tr>
</tbody>
</table>
Table 4.8 Frequency and Phase of the first 4 spectral peaks [Antiphasic]

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 F1Out [Hz]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 3 F3Out [Hz]</th>
<th>Peak 4 F4Out [Hz]</th>
<th>Peak 1 PH1Out [degreesº]</th>
<th>Peak 2 PH2Out [degreesº]</th>
<th>Peak 3 PH3Out [degreesº]</th>
<th>Peak 4 PH4Out [degreesº]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>9.3750</td>
<td>23.4375</td>
<td>35.1563</td>
<td>51.5625</td>
<td>-54.6574</td>
<td>-29.3734</td>
<td>-81.8349</td>
<td>-162.9210</td>
</tr>
<tr>
<td>2</td>
<td>9.3750</td>
<td>37.5000</td>
<td>49.2188</td>
<td>65.6250</td>
<td>-27.3947</td>
<td>-96.8243</td>
<td>-116.9690</td>
<td>-175.4290</td>
</tr>
<tr>
<td>3</td>
<td>9.3750</td>
<td>35.1563</td>
<td>46.8750</td>
<td>63.2813</td>
<td>106.0257</td>
<td>143.4579</td>
<td>-162.6980</td>
<td>87.5004</td>
</tr>
<tr>
<td>5</td>
<td>7.0313</td>
<td>23.4375</td>
<td>35.1563</td>
<td>49.2188</td>
<td>170.7757</td>
<td>102.7913</td>
<td>102.2994</td>
<td>90.3124</td>
</tr>
<tr>
<td>6</td>
<td>11.7188</td>
<td>30.4688</td>
<td>46.8750</td>
<td>65.6250</td>
<td>-60.7430</td>
<td>-87.3732</td>
<td>-138.0350</td>
<td>153.6715</td>
</tr>
<tr>
<td>7</td>
<td>9.3750</td>
<td>21.0938</td>
<td>35.1563</td>
<td>51.5625</td>
<td>149.8964</td>
<td>83.3756</td>
<td>154.7833</td>
<td>111.9598</td>
</tr>
<tr>
<td>8</td>
<td>9.3750</td>
<td>23.4375</td>
<td>46.8750</td>
<td>60.9375</td>
<td>-50.8316</td>
<td>-70.6572</td>
<td>-45.2042</td>
<td>-52.2343</td>
</tr>
</tbody>
</table>

Table 4.9 Frequency and Phase of the first 4 spectral peaks [Homophasic]

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 F1In [Hz]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak 3 F3In [Hz]</th>
<th>Peak 4 F4In [Hz]</th>
<th>Peak 1 PH1In [degreesº]</th>
<th>Peak 2 PH2In [degreesº]</th>
<th>Peak 3 PH3In [degreesº]</th>
<th>Peak 4 PH4In [degreesº]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>9.3750</td>
<td>30.4688</td>
<td>46.8750</td>
<td>75.0000</td>
<td>-26.5682</td>
<td>-57.8831</td>
<td>-115.0570</td>
<td>131.0123</td>
</tr>
<tr>
<td>2</td>
<td>9.3750</td>
<td>30.4688</td>
<td>51.5625</td>
<td>77.3438</td>
<td>-66.4877</td>
<td>100.5076</td>
<td>-132.1340</td>
<td>177.0746</td>
</tr>
<tr>
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<td>9.3750</td>
<td>25.7813</td>
<td>42.1875</td>
<td>58.5938</td>
<td>-28.6111</td>
<td>-96.3807</td>
<td>-107.4660</td>
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<tr>
<td>5</td>
<td>9.3750</td>
<td>23.4375</td>
<td>37.5000</td>
<td>49.2188</td>
<td>139.0748</td>
<td>93.2393</td>
<td>81.5948</td>
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<td>28.1250</td>
<td>42.1875</td>
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<td>-39.6916</td>
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<td>11.7188</td>
<td>28.1250</td>
<td>44.5313</td>
<td>70.3125</td>
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<td>143.8102</td>
<td>31.5662</td>
</tr>
<tr>
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<td>9.3750</td>
<td>25.7813</td>
<td>46.8750</td>
<td>60.9375</td>
<td>-31.3902</td>
<td>-53.6235</td>
<td>-47.1871</td>
<td>-66.8672</td>
</tr>
</tbody>
</table>

Findings: It is found from Table 4.7 that the amplitude of the second spectral peak of the antiphasic condition is for 7 of the 8 subjects higher than the peak of the homophasic condition. The amplitude of the third spectral peak of the antiphasic condition is also in most cases higher than the homophasic case, again 7 out of 8. It should be noted however, that the exception is not the same subject (subject No 4 and 2 respectively). Regarding the phase with its corresponding frequency, no consistent trends could be found for any of the peaks analysed.
4.3.4 Comparative analysis of the first four peaks - Group A

Experiment 2

Using the same processing techniques, the first four peaks were extracted from the Fast Fourier Transform of the averaged out-phase and in-phase EEG epoch of Group A Experiment 2 data. For experiment 2, Blackman windowed pure tones of 500 Hz frequency with a sound pressure level of 60 dB were presented as a block of 500 antiphasic stimuli followed by 500 homophasic stimuli, totalling 1000 trials. Amplitude, frequency and phase of the first four spectral peaks of the averaged EEG epochs are tabulated for both antiphasic and homophasic conditions. In addition, the amplitude differences of the spectral peaks are also tabulated as shown in Table 4.10.
Table 4.10 Amplitude, Frequency, Phase of the first 4 spectral peaks
[Antiphasic] Group A Experiment 2

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 A1Out [µV]</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 3 A3Out [µV]</th>
<th>Peak 4 A4Out [µV]</th>
<th>Peak 1 F1Out [Hz]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 3 F3Out [Hz]</th>
<th>Peak 4 F4Out [Hz]</th>
</tr>
</thead>
<tbody>
<tr>
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<td>0.3202</td>
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<td>23.4375</td>
<td>37.5000</td>
<td>51.5625</td>
</tr>
<tr>
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<td>1.4473</td>
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<td>49.2188</td>
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<td>0.6416</td>
<td>0.1068</td>
<td>9.3750</td>
<td>30.4688</td>
<td>49.2188</td>
<td>67.9688</td>
</tr>
<tr>
<td>5</td>
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<td>0.3745</td>
<td>0.1182</td>
<td>0.0899</td>
<td>21.0938</td>
<td>39.8438</td>
<td>56.2500</td>
<td>67.9688</td>
</tr>
<tr>
<td>6</td>
<td>0.9818</td>
<td>0.2473</td>
<td>0.3042</td>
<td>0.1595</td>
<td>9.3750</td>
<td>28.1250</td>
<td>49.2188</td>
<td>67.9688</td>
</tr>
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<td>0.5341</td>
<td>0.1086</td>
<td>0.2010</td>
<td>7.0313</td>
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<td>39.8438</td>
<td>60.9375</td>
</tr>
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<td>0.2921</td>
<td>0.1263</td>
<td>0.0783</td>
<td>23.4375</td>
<td>49.2188</td>
<td>77.3438</td>
<td>93.7500</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 PH1Out [degrees°]</th>
<th>Peak 2 PH2Out [degrees°]</th>
<th>Peak 3 PH3Out [degrees°]</th>
<th>Peak 4 PH4Out [degrees°]</th>
</tr>
</thead>
<tbody>
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<td>-107.8160</td>
<td>90.9461</td>
</tr>
<tr>
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<td>-26.9153</td>
<td>-84.6999</td>
<td>-50.0161</td>
<td>-101.1860</td>
</tr>
<tr>
<td>3</td>
<td>-55.8418</td>
<td>-58.6262</td>
<td>-31.9598</td>
<td>-82.0141</td>
</tr>
<tr>
<td>4</td>
<td>-14.4415</td>
<td>-83.4463</td>
<td>-71.2925</td>
<td>-168.4770</td>
</tr>
<tr>
<td>5</td>
<td>126.8630</td>
<td>26.6873</td>
<td>-129.2430</td>
<td>-41.3483</td>
</tr>
<tr>
<td>6</td>
<td>-48.9908</td>
<td>-111.9210</td>
<td>178.9485</td>
<td>135.4892</td>
</tr>
<tr>
<td>7</td>
<td>-179.1730</td>
<td>81.8978</td>
<td>53.0459</td>
<td>93.0719</td>
</tr>
<tr>
<td>8</td>
<td>-123.5100</td>
<td>-161.6330</td>
<td>174.7567</td>
<td>120.8985</td>
</tr>
</tbody>
</table>
Table 4.11 Amplitude, Frequency, Phase of the first 4 spectral peaks [Homophasic] Group A Experiment 2

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 A1In [µV]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 3 A3In [µV]</th>
<th>Peak 4 A4In [µV]</th>
<th>Peak 1 F1In [Hz]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak 3 F3In [Hz]</th>
<th>Peak 4 F4In [Hz]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.2264</td>
<td>0.4630</td>
<td>0.1574</td>
<td>0.2638</td>
<td>11.7188</td>
<td>25.7813</td>
<td>42.1875</td>
<td>53.9063</td>
</tr>
<tr>
<td>2</td>
<td>0.6670</td>
<td>0.2704</td>
<td>0.1915</td>
<td>0.2129</td>
<td>9.3750</td>
<td>25.7813</td>
<td>37.5000</td>
<td>51.5625</td>
</tr>
<tr>
<td>3</td>
<td>0.4398</td>
<td>0.2632</td>
<td>0.2229</td>
<td>0.1040</td>
<td>9.3750</td>
<td>25.7813</td>
<td>35.1563</td>
<td>53.9063</td>
</tr>
<tr>
<td>4</td>
<td>0.3251</td>
<td>0.2335</td>
<td>0.2636</td>
<td>0.1539</td>
<td>28.1250</td>
<td>51.5625</td>
<td>63.2813</td>
<td>77.3438</td>
</tr>
<tr>
<td>5</td>
<td>0.5527</td>
<td>0.5769</td>
<td>0.3512</td>
<td>0.2820</td>
<td>7.0313</td>
<td>21.0938</td>
<td>37.5000</td>
<td>51.5625</td>
</tr>
<tr>
<td>6</td>
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<td>0.1738</td>
<td>0.2196</td>
<td>0.0808</td>
<td>11.7188</td>
<td>30.4688</td>
<td>53.9063</td>
<td>82.0313</td>
</tr>
<tr>
<td>7</td>
<td>0.9955</td>
<td>0.4736</td>
<td>0.2785</td>
<td>0.5727</td>
<td>7.0313</td>
<td>23.4375</td>
<td>35.1563</td>
<td>49.2188</td>
</tr>
<tr>
<td>8</td>
<td>0.6090</td>
<td>0.2127</td>
<td>0.4147</td>
<td>0.1097</td>
<td>9.3750</td>
<td>25.7813</td>
<td>49.2188</td>
<td>65.6250</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 PH1In [degrees°]</th>
<th>Peak 2 PH2In [degrees°]</th>
<th>Peak 3 PH3In [degrees°]</th>
<th>Peak 4 PH4In [degrees°]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>-9.9261</td>
<td>7.5367</td>
<td>-77.2361</td>
<td>-96.3604</td>
</tr>
<tr>
<td>2</td>
<td>-49.5453</td>
<td>-123.2280</td>
<td>-61.5370</td>
<td>-151.6010</td>
</tr>
<tr>
<td>3</td>
<td>159.8374</td>
<td>123.0597</td>
<td>91.9581</td>
<td>97.4106</td>
</tr>
<tr>
<td>4</td>
<td>-160.5320</td>
<td>152.8675</td>
<td>105.9747</td>
<td>64.6303</td>
</tr>
<tr>
<td>5</td>
<td>165.0450</td>
<td>141.1020</td>
<td>64.1088</td>
<td>47.8421</td>
</tr>
<tr>
<td>6</td>
<td>-75.5819</td>
<td>-154.3160</td>
<td>-176.6900</td>
<td>95.8242</td>
</tr>
<tr>
<td>7</td>
<td>175.4710</td>
<td>123.9320</td>
<td>121.7609</td>
<td>104.6683</td>
</tr>
<tr>
<td>8</td>
<td>-51.1734</td>
<td>-87.6057</td>
<td>-38.9878</td>
<td>-88.7757</td>
</tr>
</tbody>
</table>
Table 4.12 Amplitude difference of the first four spectral peaks
Group A Experiment 2

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak1 A1Out-A1In [µV]</th>
<th>Peak2 A2Out-A2In [µV]</th>
<th>Peak3 A3Out-A3In [µV]</th>
<th>Peak4 A4Out-A4In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.1416</td>
<td>0.2961</td>
<td>0.3908</td>
<td>-0.1641</td>
</tr>
<tr>
<td>2</td>
<td>-0.1059</td>
<td>0.0398</td>
<td>-0.0097</td>
<td>0.1073</td>
</tr>
<tr>
<td>3</td>
<td>1.0075</td>
<td>0.1433</td>
<td>0.0811</td>
<td>0.2799</td>
</tr>
<tr>
<td>4</td>
<td>0.1635</td>
<td>-0.1297</td>
<td>0.3779</td>
<td>-0.0471</td>
</tr>
<tr>
<td>5</td>
<td>-0.1957</td>
<td>-0.2024</td>
<td>-0.2330</td>
<td>-0.1921</td>
</tr>
<tr>
<td>6</td>
<td>0.5293</td>
<td>0.0736</td>
<td>0.0847</td>
<td>0.0788</td>
</tr>
<tr>
<td>7</td>
<td>-0.4630</td>
<td>0.0605</td>
<td>-0.1699</td>
<td>-0.3717</td>
</tr>
<tr>
<td>8</td>
<td>-0.3432</td>
<td>0.0794</td>
<td>-0.2883</td>
<td>-0.0314</td>
</tr>
</tbody>
</table>

A comparative analysis of the first four spectral peak amplitude differences as shown in Table 4.12 shows that unlike Group A Experiment 1 where the second and the third dominant frequency peak amplitude difference showed the maximum number of positives, in Group A Experiment 2 only the second spectral peak amplitude difference shows the highest number of positive amplitude differences. 6 out of 8 subjects showed higher second spectral peak amplitude when evoked with antiphasic stimuli. Regarding the phase angle, again no consistent results can be obtained from the data.
Findings:

- Based on the results of experiment 1 and 2, no marker can be extracted from the first four dominant frequencies to indicate in-phase or out-phase with regards to frequency or phase for any of the subjects.
- The difference in amplitude of the second spectral peak may be an indicator of detecting the impact of homophasic and antiphase stimuli in the binaural processing of the human brain [273].

4.4 Processing of the remaining preliminary experiments

The above mentioned processing techniques will be used for all the other experimental data to search for trends and consistency. Amplitude differences of the second spectral peak frequency will be investigated to look for a correlation, which has already been suggested in the results of Group A, Experiment 1 and 2.

This section contains all the remaining preliminary experimental data presented in tabular format. Tables 4.13 to 4.18 contains second spectral peak frequency along with the amplitude of the peaks. In addition the table also contains the amplitude difference between the second spectral peak in the middle latency response evoked due to antiphase and homophasic stimuli.
Group A Experiment 3

In experiment 3, the stimuli used were a 60 dB 500 Hz Blackman windowed pure tone masked with continuous Gaussian noise having bandwidth in the range of 100 and 900 Hz. A total of 1000 trials were conducted as blocks of 10 out-phase and 10 in-phase trials and the signal to noise ratio for the stimuli was 5 dB.

Table 4.13 Group A Exp-4 Exp-3 Second spectral peak frequency and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.3351</td>
<td>28.1250</td>
<td>0.1490</td>
<td>51.5625</td>
<td>0.1861</td>
</tr>
<tr>
<td>2</td>
<td>0.0620</td>
<td>28.1250</td>
<td>0.0873</td>
<td>30.4688</td>
<td>-0.0253</td>
</tr>
<tr>
<td>3</td>
<td>0.6058</td>
<td>23.4375</td>
<td>0.4590</td>
<td>23.4375</td>
<td>0.1468</td>
</tr>
<tr>
<td>4</td>
<td>0.4638</td>
<td>23.4375</td>
<td>0.2584</td>
<td>28.1250</td>
<td>0.2054</td>
</tr>
<tr>
<td>5</td>
<td>0.3433</td>
<td>23.4375</td>
<td>0.2116</td>
<td>25.7813</td>
<td>0.1317</td>
</tr>
<tr>
<td>6</td>
<td>0.0822</td>
<td>51.5625</td>
<td>0.1404</td>
<td>30.4688</td>
<td>-0.0582</td>
</tr>
<tr>
<td>7</td>
<td>0.4470</td>
<td>21.0938</td>
<td>0.3721</td>
<td>21.0938</td>
<td>0.0748</td>
</tr>
<tr>
<td>8</td>
<td>0.2183</td>
<td>23.4375</td>
<td>0.1285</td>
<td>25.7813</td>
<td>0.0898</td>
</tr>
</tbody>
</table>

There is a trend that can be seen again in experiment 3 and in this case 6 out of 8 subjects’ perceived antiphasesic stimuli better than homophasesic stimuli.
Group A Experiment 4

In experiment 4, 60 dB 500 Hz Blackman windowed pure tone masked with continuous Gaussian noise having bandwidth in the range of 20Hz and 8 kHz and a total of 1000 trials were presented as blocks of 10 out-phase and 10 in-phase trials. The signal to noise ratio for the stimuli in this case was again 5 dB. The second spectral peak along with amplitude difference is presented in tabular format as shown below and in this case 5 out of 8 subjects showed the positive amplitude difference.

Table 4.14 Group A Exp-4 Second spectral peak frequency and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.2367</td>
<td>25.7813</td>
<td>0.1101</td>
<td>25.7813</td>
<td>0.1266</td>
</tr>
<tr>
<td>2</td>
<td>0.1055</td>
<td>25.7813</td>
<td>0.1152</td>
<td>21.0938</td>
<td>-0.0097</td>
</tr>
<tr>
<td>3</td>
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<td>23.4375</td>
<td>0.2690</td>
<td>23.4375</td>
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<td>-0.0678</td>
</tr>
<tr>
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<td>0.1793</td>
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<td>0.0142</td>
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<tr>
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<td>0.1186</td>
</tr>
<tr>
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<td>0.2585</td>
<td>35.1563</td>
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<tr>
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<td>25.7813</td>
<td>0.1896</td>
<td>44.5313</td>
<td>0.1192</td>
</tr>
</tbody>
</table>
Group B Experiment 1

In experiment 1, the stimuli were a 60 dB Blackman windowed pure tone and the frequency of the tone was changed to 1000 Hz. In experiment 1, pure tones as stimuli were presented as a block of 10 antiphasic stimuli followed by 10 homophasic stimuli for a total of 1000 trials. This provided 500 antiphasic and 500 homophasic related event potentials to be captured for averaging. In this case 6 out of 8 subjects showed the positive amplitude difference.

Table 4.15 Group B Exp-1 Second spectral peak frequency and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.3061</td>
<td>28.1250</td>
<td>0.4365</td>
<td>23.4375</td>
<td>-0.1304</td>
</tr>
<tr>
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<td>23.4375</td>
<td>0.3216</td>
<td>42.1875</td>
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</tr>
<tr>
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<td>23.4375</td>
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</tr>
<tr>
<td>5</td>
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<td>0.2386</td>
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<td>0.0233</td>
</tr>
<tr>
<td>6</td>
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<td>30.4688</td>
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<td>0.0315</td>
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<td>1.0033</td>
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<td>0.3529</td>
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</table>
**Group B Experiment 2**

In experiment 2, the stimuli were a 60 dB Blackman windowed 1000 Hz pure tone masked by Gaussian noise with a bandwidth in the range of 20 Hz to 8000 Hz. The signal to noise ratio for the stimuli was 5 dB. A block of 10 antiphase stimuli followed by 10 homophase stimuli were presented for a total of 1000 trials. In this case again 6 out of 8 subjects showed the positive amplitude difference.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.1003</td>
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<td>0.9221</td>
<td>18.7500</td>
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</tr>
<tr>
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</tr>
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<td>0.3088</td>
<td>37.5000</td>
<td>0.2603</td>
<td>30.4688</td>
<td>0.0485</td>
</tr>
<tr>
<td>4</td>
<td>0.2580</td>
<td>25.7813</td>
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<td>21.0938</td>
<td>0.0367</td>
</tr>
<tr>
<td>5</td>
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</tr>
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<td>-0.0819</td>
</tr>
<tr>
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<td>0.1462</td>
<td>25.7813</td>
<td>0.0428</td>
</tr>
<tr>
<td>8</td>
<td>0.2747</td>
<td>30.4688</td>
<td>0.3245</td>
<td>25.7813</td>
<td>-0.0499</td>
</tr>
</tbody>
</table>
**Group C Experiment 1 and 2**

The relationship of the homophasic and the antiphasic auditory EEG was further explored by randomising the presentation of homophasic and antiphasic tones in the presence of noise. In the first experiment of Group C, a Blackman windowed pure tone of 500 Hz frequency was masked by Gaussian noise with a bandwidth in the range of 100 and 900 Hz whereas in the second experiment of Group C, a Blackman windowed pure tone of 500 Hz frequency was masked by Gaussian noise with a bandwidth in the range of 20 to 8000 Hz. In both the cases the signal to noise ratio for the stimuli was 5 dB and the stimuli were presented randomly for a total of 1000 trials. These trials included 500 in-phase and 500 out-phase tones with randomised presentation.

**Table 4.17 Group C Exp-1 Second dominant frequency peaks and amplitude difference**

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.6523</td>
<td>49.2188</td>
<td>0.4160</td>
<td>23.4375</td>
<td>0.2363</td>
</tr>
<tr>
<td>2</td>
<td>1.0187</td>
<td>21.0938</td>
<td>0.5521</td>
<td>23.4375</td>
<td>0.4666</td>
</tr>
<tr>
<td>3</td>
<td>0.7950</td>
<td>49.2188</td>
<td>0.4126</td>
<td>30.4688</td>
<td>0.3824</td>
</tr>
<tr>
<td>4</td>
<td>0.2512</td>
<td>51.5625</td>
<td>0.2232</td>
<td>44.5313</td>
<td>0.0280</td>
</tr>
</tbody>
</table>
Table 4.18 Group C Exp-2 Second dominant frequency peaks and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak 2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.4701</td>
<td>32.8125</td>
<td>0.3998</td>
<td>25.7813</td>
<td>0.0703</td>
</tr>
<tr>
<td>2</td>
<td>0.5272</td>
<td>23.4375</td>
<td>0.4906</td>
<td>25.7813</td>
<td>0.0366</td>
</tr>
<tr>
<td>3</td>
<td>0.3258</td>
<td>25.7813</td>
<td>0.2186</td>
<td>23.4375</td>
<td>0.1071</td>
</tr>
<tr>
<td>4</td>
<td>0.1910</td>
<td>28.1250</td>
<td>0.1768</td>
<td>32.8125</td>
<td>0.0142</td>
</tr>
</tbody>
</table>

In both experiments it is found that all 4 subjects, when presented with randomised stimuli, exhibited a larger amplitude of the second spectral peak in the case of out-phase stimuli.

### 4.5 Statistical Test of the preliminary results

The overall preliminary experiments so far indicate that a higher amplitude of the second spectral peak frequency is found when elicited by antiphasic stimuli of both 500 and 1000 Hz frequency. 44 out of 56 experiments showed positive amplitude difference between the second spectral peaks extracted from the antiphasic and homophasic middle latency response. To investigate whether the results obtained is not a coincidence a couple of statistical tests are conducted at this stage. In this case there are two possible outcomes - either higher antiphasic second spectral peak or higher homophasic second spectral peak. If the chance of a higher antiphasic second spectral peak is just as high as the chance of a lower antiphasic second spectral peak, this
corresponds to a binomial probability distribution [274] for which there are two possible outcomes with fixed probabilities \((p = 0.5)\). Table 4.19 shows all the preliminary experiments along with the number of higher antiphasic second spectral peak found under each experiment. For example in Group A Experiment 1 the amplitude of the second spectral peak of the antiphasic condition is for 7 of the 8 subjects higher than the peak of the homophasic condition. Binomial probability of the occurrence of the event higher antiphasic second spectral peak for each experiment is calculated using equation 4.2 [274] and added to the table.

\[
\binom{n}{r} \cdot p^r \cdot (1 - p)^{n-r} \tag{4.2}
\]

- \(n\) = number of subjects
- \(r\) = number of higher antiphasic second spectral peak
- \(p = 0.5\) probability that the event higher antiphasic peak will occur
- \((1-p) = 0.5\) probability that the event higher homophasic peak will occur

A matlab function binopdf() was used to calculate the probability of the event.

Table 4.19: Binomial probability of the occurrence of higher antiphasic peak

<table>
<thead>
<tr>
<th>Group</th>
<th>Experiment</th>
<th>Number of Subjects (n)</th>
<th>Number of Higher Antiphasic Peak (r)</th>
<th>Binomial Probability (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>1</td>
<td>8</td>
<td>7</td>
<td>3.12</td>
</tr>
<tr>
<td>A</td>
<td>2</td>
<td>8</td>
<td>6</td>
<td>10.94</td>
</tr>
<tr>
<td>A</td>
<td>3</td>
<td>8</td>
<td>6</td>
<td>10.94</td>
</tr>
<tr>
<td>A</td>
<td>4</td>
<td>8</td>
<td>5</td>
<td>21.87</td>
</tr>
<tr>
<td>B</td>
<td>1</td>
<td>8</td>
<td>6</td>
<td>10.94</td>
</tr>
<tr>
<td>B</td>
<td>2</td>
<td>8</td>
<td>6</td>
<td>10.94</td>
</tr>
<tr>
<td>C</td>
<td>1</td>
<td>4</td>
<td>4</td>
<td>6.25</td>
</tr>
<tr>
<td>C</td>
<td>2</td>
<td>4</td>
<td>4</td>
<td>6.25</td>
</tr>
</tbody>
</table>
Table 4.19 shows that other than Group A Experiment 4, where the probability of the occurrence of the higher second spectral antiphasic peak is around 22%, in all the other experiments the percentage is below 11% which means roughly 1 out of 9 subjects will be successful in having higher antiphasic second spectral peak. In addition after calculating the binomial probability based on the total number of the occurrence of higher antiphasic second spectral peak in the total number of preliminary experiments (44 out of 56) it is found that the probability is very small. In this case the value is 0.00078% which is almost negligible. Having a low value of the probability of occurrence of higher antiphasic second spectral peak indicates that the results obtained so far are not due to coincidence.

### 4.5.1 z-Test Approximation of the Binomial Test

Further test is conducted on the binomial distribution to signify the results achieved so far. Since 44 out of 56 experiments generate higher second spectral antiphasic peak comparing to second spectral homophasic peak, a z-test is conducted to find if the null hypothesis: “that both the antiphasic and homophasic higher second spectral peak are equally likely to occur (i.e., \( P \) (higher antiphasic peak) = \( P \) (higher homophasic peak) = 0.5) at an alpha level of 0.05” can be rejected or not. Because the measurements are binary (either antiphasic or homophasic), the null hypothesis is binomially distributed with the following parameters: \( n=56, p=0.5 \). Approximating the distribution with a normal distribution (since \( n > 30 \)) with a mean (\( \mu \)) of 28 and standard deviation (\( \sigma \)) of 3.74 a z –test is conducted as shown in Table 4.20.
Equation 4.3 and 4.4 is used to calculate mean and standard deviation of the distribution [274]

$$\mu = np \quad (4.3)$$

$$\sigma = \sqrt{np(1-p)} \quad (4.4)$$

- $n$ = number of experiments (56)
- $p = 0.5$ probability that the event higher antiphasic peak will occur
- $(1-p) = 0.5$ probability that the event higher homophasic peak will occur

Table 4.20 : Hypothesis Testing using Z score and P value

<table>
<thead>
<tr>
<th>Null Hypothesis</th>
<th>$\mu=28, \sigma=3.74$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Alternative Hypothesis</td>
<td>$\mu &gt; 28$</td>
</tr>
<tr>
<td>Tail of Test</td>
<td>Upper tailed</td>
</tr>
<tr>
<td>Type of Test</td>
<td>z-test</td>
</tr>
<tr>
<td>Alpha level</td>
<td>$\alpha = 0.05$</td>
</tr>
<tr>
<td>Critical Values of Test Statistic using z table</td>
<td>1.65 (because just less than 5% of the area under the standard distribution lies between it and positive infinity)</td>
</tr>
<tr>
<td>Test Statistic of Data</td>
<td>4.27 obtained after converting the number of higher second antiphasic peak, $44(x)$, into a z-score using the formula: $z = \frac{x-\mu}{\sigma}$ [274]</td>
</tr>
<tr>
<td>p-value of Data</td>
<td>$p=0.00003$</td>
</tr>
<tr>
<td>Conclusion</td>
<td>Reject the Null Hypothesis</td>
</tr>
</tbody>
</table>

Since the z-score of the sample exceeds the critical z-score, the null hypothesis that both the antiphasic and homophasic higher second spectral
peak are equally likely to occur at an alpha level of 0.05 is rejected. To find whether these 44 out of 56 is significant, the p-value of the sample is calculated and is found to be 0.003%. This indicates that this is a significant result.

4.6 Comparison between experiments - second spectral peak amplitude

Further analysis of the results is conducted in order to compare between the experiments conducted so far. In this case second spectral peak amplitude difference for each subject under each experiment was divided by the average of second spectral homophasic and antiphasic peak amplitudes. The purpose of dividing with the average value of homophasic and antiphasic spectral peak amplitude was to normalize the amplitude difference of the second spectral peaks by including a division with the overall impact of the second spectral dominant EEG frequencies. This may account for variability in amplitudes of subjects or the experimental conditions of that particular experiment. A formula was written in excel to normalize the amplitude difference.

Once the second spectral peak difference was normalised a mean value of all the normalised peak difference for each experiment were calculated and the comparison between the experiments were conducted based on the magnitude of the mean value achieved for each experiment.
Equation 4.5 calculates the mean value of all the normalised peak difference of eight (8) subjects.

\[
Mean = \frac{1}{8} \sum_{i=1}^{8} \frac{(A2:OUT_i - A2:IN_i)}{(A2:OUT_i + A2:IN_i) * \frac{1}{2}}
\] (4.5)

4.6.1 500 Hz frequency pure tone stimuli

This section presents tables which show the normalised value of the second spectral peak amplitude difference for all the subjects of the preliminary experiment of Group A and C. In addition the mean of all 8 normalised amplitude differences under each experiment is also presented in Table 4.19. It is important to note that all the preliminary experiments of Group A and C used 500 Hz pure tone as a stimuli embedded with and without noise and for this reason this section is made only for comparing auditory evoked responses elicited by 500 Hz frequency stimuli. The first table (Table 4.19) shows the normalised value of the second spectral peak difference for all the experiment (1, 2, 3 and 4) of group A.
Table 4.21 Normalised amplitude difference for Group A Experiment 1, 2, 3 and 4

<table>
<thead>
<tr>
<th>Experiment 1 10 Block Out – 10 Block In</th>
<th>Experiment 2 500 Block Out – 500 Block In</th>
<th>Experiment 3 10 Block Out – 10 Block In (100 Hz to 900 Hz – masked noise)</th>
<th>Experiment 4 10 Block Out – 10 Block In (20 Hz to 20000 Hz – masked noise)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject</td>
<td>AmpDiff/AvgAmp</td>
<td>AmpDiff/AvgAmp</td>
<td>AmpDiff/AvgAmp</td>
</tr>
<tr>
<td>1</td>
<td>0.1678</td>
<td>0.4846</td>
<td>0.7689</td>
</tr>
<tr>
<td>2</td>
<td>0.1303</td>
<td>0.1369</td>
<td>-0.3384</td>
</tr>
<tr>
<td>3</td>
<td>0.6825</td>
<td>0.4278</td>
<td>0.2758</td>
</tr>
<tr>
<td>4</td>
<td>-0.4396</td>
<td>-0.7690</td>
<td>0.5689</td>
</tr>
<tr>
<td>5</td>
<td>0.1628</td>
<td>-0.4254</td>
<td>0.4748</td>
</tr>
<tr>
<td>6</td>
<td>0.0213</td>
<td>0.3494</td>
<td>-0.5232</td>
</tr>
<tr>
<td>7</td>
<td>0.0525</td>
<td>0.1201</td>
<td>0.1828</td>
</tr>
<tr>
<td>8</td>
<td>0.1746</td>
<td>0.3145</td>
<td>0.5175</td>
</tr>
<tr>
<td><strong>MEAN</strong></td>
<td><strong>0.1190</strong></td>
<td><strong>0.0799</strong></td>
<td><strong>0.2409</strong></td>
</tr>
</tbody>
</table>

Figure 4.35 below shows the mean difference graphically. The largest mean value is found for Experiment 3 where the subjects’ auditory response was evoked with stimuli of pure tone embedded with narrow bandwidth Gaussian noise of 100 to 900 Hz.
The second largest mean value is found for experiment 4 where the stimulus was embedded with band noise of 20 Hz to 8 kHz followed by the other two experiments where the stimuli were delivered as a pure tone with no noise and were delivered either in 10 block sequence (Experiment 1) or 500 block sequence (Experiment 2). These results possibly indicates that a narrow band noise masker may acts as a cue for the subject to perceive tone better in noise thus eliciting a larger mean value of normalised amplitude difference in the second spectral peak frequency component.

The relationship of the in-phase and out-phase AEP was further explored by randomising the presentation of homophasic and antiphasic tones in the presence of noise. Experiment 1 of Group C presented a 100-900 Hz band Gaussian noise masking a 500Hz pure tone sinusoid whereas the second experiment of Group 3 presented a 20 Hz - 8000Hz wide-band Gaussian
noise masking a 500Hz pure tone sinusoid. In both cases the stimuli were presented randomly for a total of 1000 trials. Table 4.20 and Figure 4.36 below show the normalised amplitude difference.

Table 4.22 Normalised amplitude difference for Group C Experiment 1 and 2

<table>
<thead>
<tr>
<th>Subject</th>
<th>AmpDiff/AvgAmp</th>
<th>AmpDiff/AvgAmp</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.4424</td>
<td>0.1617</td>
</tr>
<tr>
<td>2</td>
<td>0.5941</td>
<td>0.0720</td>
</tr>
<tr>
<td>3</td>
<td>0.6333</td>
<td>0.3936</td>
</tr>
<tr>
<td>4</td>
<td>0.1182</td>
<td>0.0772</td>
</tr>
<tr>
<td>MEAN</td>
<td>0.2814</td>
<td>0.2482</td>
</tr>
</tbody>
</table>

Figure 4.37 Mean of Normalised Amplitude Difference – Group C Experiment 1 and 2
The mean value of the normalised amplitude difference between the two cases demonstrates that a narrower band Gaussian noise masker elicits a larger mean value of the normalised second peak amplitude difference in the second spectral peak frequency component of the MLR. Further discussion about the results will be presented in the results discussion part of this chapter.

4.6.2 1000 Hz frequency pure tone

In Group B Experiments 1 and 2, 1000 Hz is used as a signal frequency. The plotted graph below shows a different result and does not follow the trend observed with 500 Hz frequency experiments. In this case Blackman windowed pure tone presented as a block of 10 antiphasic stimuli followed by 10 homophasic stimuli yields a larger mean value of normalised amplitude difference comparing to experiment 2 where pure tone masked with continuous Gaussian noise having bandwidth in the range of 20 and 8000 Hz. Table 4.21 and Figure 4.37 below represents the normalised amplitude difference.
Table 4.23 Normalised amplitude difference for Group B Experiment 1 and 2

<table>
<thead>
<tr>
<th>Subject</th>
<th>Experiment 1</th>
<th>Experiment 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>-0.3513</td>
<td>0.1763</td>
</tr>
<tr>
<td>2</td>
<td>-0.0822</td>
<td>0.2203</td>
</tr>
<tr>
<td>3</td>
<td>0.0883</td>
<td>0.1705</td>
</tr>
<tr>
<td>4</td>
<td>0.4940</td>
<td>0.1533</td>
</tr>
<tr>
<td>5</td>
<td>0.0930</td>
<td>0.0945</td>
</tr>
<tr>
<td>6</td>
<td>0.1244</td>
<td>-0.1607</td>
</tr>
<tr>
<td>7</td>
<td>0.3678</td>
<td>0.2551</td>
</tr>
<tr>
<td>8</td>
<td>0.2991</td>
<td>-0.1665</td>
</tr>
<tr>
<td>MEAN</td>
<td><strong>0.1291</strong></td>
<td><strong>0.0929</strong></td>
</tr>
</tbody>
</table>

Figure 4.38 Mean of Normalised Amplitude Difference Group B Experiment 1 and 2
4.7 Discussion of preliminary results

To achieve the aim of this research it is important to look for evidence in the changes of the binaural processing of the brain evoked by a series of homophasic and antiphasic stimuli. Evidence is likely to indicate the presence of the binaural hearing in auditory processing mechanism. An approach towards quantifying brains’ response elicited due to homophasic and antiphasic stimuli has been developed based on evaluating spectral frequency peaks in the frequency spectrum.

Upon spectral analysis of the data using the Fourier Transform, a higher amplitude of the second spectral peak frequency is found when elicited by antiphasic stimuli of both 500 and 1000 Hz frequency [275]. 44 out of 56 experiments showed positive amplitude difference between the second spectral peaks extracted from the antiphasic and homophasic middle latency response. The inter-aural phase difference (IPD) between the stimuli acts as a binaural cue to the brain for identifying the direction of the sound source [123]. Majority of the subjects’ binaural processing performs better when the signal arrives antiphasic comparing to homophasic. This could possibly relate to finding higher response in the middle latency response (MLR) as the frequencies of the second peak correspond with this response (20 – 50 Hz). It should be noted that there is differences in the frequencies of the second spectral peaks in respect to homophasic and antiphasic evoked response. This variance in frequency could be attributed to the individual’s characteristics of the EEG. On one hand, the subjective effect, such as the
concentration level of the listener, could be the result of reactivity of the brain oscillation and it could be varied from one person to another [276]. Additionally, the individual’s variability of his/her brain responses to stimuli could be attributed to inherent characteristics as well as anatomical structures of the individual brain [277]. There is also evidence from past research that individuals show a high variability in the auditory processing of stimuli or listening to the music [276, 278].

44 experimental results out of 56 indicate that the subjects’ middle latency response is higher when the tones are presented with a 180 degree phase shift between the right and left ear tone subjects. This supports the theory behind Durlach’s “Equalization-Cancellation” for the 180 degree phase shift. In his model he suggested that the auditory system of the brain can subtract the signals at the two ears. Providing there are 0 or 180 interaural phase differences between the left and right stimulus, when the stimulus are 180 degree out of phase the resulting components after subtraction will have a gain in detectability. Since the signal is positive in one ear and negative in the other, subtraction gives double the intensity, which means theoretically human will hear well when the stimulus are presented out of phase [123, 140].

Based on the preliminary experiments, a dominant frequency between 20 Hz to 50 Hz may be used to detect the differences between auditory evoked responses based on antiphasic and homophasic auditory stimuli.
It should be noted however that the ABR may still be present in the first 8 ms of the signal used for analysis. The ABR takes up to 10 ms which results in an affected time of up to 28 ms for the stimuli. The MLR starts from 20 ms, hence the possible overlap of 8 ms. However, the ABR is much smaller in amplitude than the MLR [36, 178, 225]. Furthermore, the main frequency content of the ABR auditory stimuli is above 100 Hz [180, 279, 280]. The dominant frequency behaviour of the MLR is below 100 Hz [281-283] and the ABR is therefore unlikely to have a noticeable influence on the results.

Between each stimulus, 200 ms silence was placed to ensure that the LLR had to a large extent damped out [219, 284]. Furthermore, the dominant frequency content of the second peak (20 – 50 Hz) is unlikely to overlap with the LLR (below 20 Hz) [283, 285, 286]. A possible remainder of the previous LLR on the next response will therefore have little influence on the results.

As both the ABR and the LLR are unlikely to have a significant influence on the second peak of the frequency spectrum, it is expected that the second peak of the frequency response is linked to the behaviour of the MLR.

As shown in Table 4.19, the mean value of the normalised second peak amplitude difference for the first two experiments of Group A are 0.1190 and 0.0799 respectively. In the first two experiments, where the tone was not masked with noise, it is found that experiment 1 elicited a larger mean value of normalised amplitude difference compared to experiment 2. One possible reason can be the number of the phase changes occurring between the
stimuli. With the experiment 1 stimuli, the phase changed every 10 intervals. Thus the neural processing of the binaural cues may not have the time to become locked onto the stimulus so the change between the two conditions is much more defined. By having a block of 500 sequential tones of both in-phase and out-phase conditions, the subjects level of concentration may deteriorate as they are listening to the same tone with no change for 500 times. This may be a factor resulting in lower mean value of normalised evoked response differences. Experiments 1 and 2 demonstrate that maybe a short sequence of in-phase and out-phase pure tones will elicit a larger mean value of normalised amplitude difference in the 20-45Hz frequency component of the middle latency response whereas blocks of larger number of tones elicit a smaller difference, possibly because the brain enters into a steady state response to the stimuli.

From Table 4.19 it can be concluded that, where the stimuli are masked with noise, a larger mean value of normalised second peak amplitude difference results. The mean value of the normalised second peak amplitude difference for Group A of experiment 3 and 4 are 0.2409 and 0.2109 respectively. This suggests that background noise may provide additional binaural cues for binaural processing. By introducing noise to the stimuli, a context is established to the pure tone such as when interpreting speech surrounded by background noise. In addition by introducing noise as a masker to the signal, a binaural masking level difference can be created. It is also reported by Ira and Licklider that the detection of a signal in a background of noise is much
easier when the signal has a different inter-aural time or phase difference than that of the noise [137, 138].

Among all four experiments it is found that the middle latency response elicited by a 500 Hz pure tone embedded with 100 Hz to 900 Hz Gaussian noise yields the maximum mean value of normalised amplitude difference. Again this may be due to an increase in binaural masking level difference with increasing signal inter-aural level difference, with increasing masker intensity level, and/or with a more narrow masker bandwidth [134].

A similar trend can be seen in the case where the stimuli were randomly generated and the two cases tested in the Group C experiment 1 and 2 were to confirm again the impact of Gaussian noise on the second spectral component. As seen in Table 4.20 the mean value of the normalised second spectral peak amplitude difference between the two cases demonstrates that in experiment 1 a narrower band Gaussian noise masker elicits a larger mean value (0.2814) compared to the mean value (0.2482) found from experiment 2. Having randomised presentation of stimuli may not be a good option as this does not replicate the real environment where it is rare to have sounds with such a random shift.

For all the experiments with 500 Hz stimuli it is found that a pure tone embedded with noise yields the higher mean value of normalised spectral peak amplitude differences compared to the values achieved for pure tone with no noise.
However, no conclusions can be made for Group B experiments where 1000 Hz is used as a signal frequency. The pure tone stimuli with no noise in experiment 1 produced a larger mean value (0.1291) of normalised second spectral peak amplitude difference compared to the mean value (0.0929) obtained in experiment 2 where a pure tone was masked with continuous Gaussian noise having bandwidth in the range of 20 and 8000 Hz.
4.8 Summary

This chapter presents the preliminary experiments which were carried out during the course of the thesis in order to find evidence of any electrophysiological changes in the middle latency response of the brain. A concise description of the experiments to validate the setups was presented, followed by the selection of the stimuli and description of the design process. The chapter also presented the development of the algorithm which helped in extracting markers that can be used towards building an objective methodology to detect binaural hearing in the human brain. Finally the chapter provided a detailed description of the experimental results. Based on the preliminary experiments, a dominant frequency between 20 Hz to 50 Hz may be used to detect the differences between auditory evoked responses based on antiphasic and homophasic auditory stimuli.
Chapter 5: Final Experiments

5.1 Changes after the Preliminary Experiments

The preliminary experiment results indicate that the second spectral peak, most likely related to the MLR, can be used as a detector to indicate the presence of binaural hearing in the binaural processing of the homophasic and antiphasic stimuli. Data collected from final experiments are analysed to compare again second peak response based on homophasic and antiphasic stimuli. However certain changes were made before conducting the final experiments which address some of the limitations found in the preliminary experiments. The changes are:

- The number of experiments conducted for each group in the preliminary stage varied, thus the total experimental time was also different. During conducting preliminary experiments all the experiments were conducted in one sitting. However, insufficient breaks or pauses were included, resulting in tired subjects. During the final experiments, six experiments are conducted in total on all of the same subjects and sufficient breaks were given between the experiments to refresh subjects.
• Selection of the bandwidth of the noise was reconsidered for the final experiments. Stimuli were made again of 500 Hz and 1000 Hz frequency. Experiments with a noise bandwidth of ±400Hz were included to maintain consistency with the preliminary experiments. In addition a broader bandwidth of 20 to 20 KHz noise was taken instead of 20 to 8 KHz in order to include all the frequencies within human hearing range.

• The number of subjects was increased from 8 to 12 to make the results statistically more significant.

• Due to limitations in processing power during the preliminary experiments, sampling rates of the EEG signals were limited to 4.8 KHz. During the final experiments a computer with improved processing power was employed, and the capture rate of the data was therefore increased to 19.2 KHz. However in both cases higher sampling rate was used to look into the other effects such as ABR and LLR response of the brain.

• During the preliminary experiments, recordings of the EEG signals were not closely monitored for artifacts and as a result some of the experimental data had to be discarded. During the final experiments, special care was taken to monitor EEG signals and if major artifacts were seen new measurements were conducted. In addition, the experiment was sometimes interrupted by the subject himself, e.g. if
the subject needed a toilet break or wanted a glass of water. Sometimes subjects were distracted and their movements caused a change in electrode conductivity which also meant that the experiment needed to be repeated.

5.2 Stimuli for final experiments

Experiment 1 (similar to Group A preliminary Experiment 1)

For Experiment 1 the stimulus chosen was the same as the one used in Group A preliminary Experiment 1. The stimuli were made of a pure tone of 500 Hz frequency having a sound pressure level of 60 dB. The pure tone was then passed through Blackman window and the stimuli were presented as a block of 10 antiphasic stimuli followed by 10 homophasic stimuli for a total of 1000 trials which provided 500 antiphasic and 500 homophasic related event potentials.

Experiment 2 (similar to Group A preliminary Experiment 3)

The next stimulus that was selected from Group A preliminary experiment 3 was a 60 dB 500 Hz Blackman windowed pure tone masked with continuous Gaussian noise having bandwidth in the range of 100 and 900 Hz and a total of 1000 trials were conducted as blocks of 10 out-phase and 10 in-phase trials. The signal to noise ratio for the stimuli remains the same as 5 dB.
Experiment 3 (500 Hz, noise bandwidth to cover higher frequencies)

For Experiment 3 the stimulus used was a 60 dB 500 Hz Blackman windowed pure tone masked with continuous Gaussian noise, having a bandwidth of 20 and 20000 Hz, and a total of 1000 trials were conducted as blocks of 10 out-phase and 10 in-phase trials. The signal to noise ratio for the stimuli remains the same at 5 dB.

Experiment 4 (similar to Group B preliminary experiment 1)

In Experiment 4, the stimuli used were also a 60 dB Blackman windowed pure tone and the frequency of the tone was changed to 1000 Hz. A block of 10 antiphase stimuli followed by 10 homophase stimuli were presented for a total of 1000 trials which provided 500 antiphasic and 500 homophasic related event potentials to be captured for averaging.

Experiment 5 (1000 Hz, choice of new narrow band noise)

In Experiment 5, stimulus used was a 60 dB 1000 Hz Blackman windowed pure tone masked with narrow band noise of 600 Hz to 1400 Hz bandwidths with SNR of 5 dB. The stimuli were presented as a block of 10 antiphase stimuli followed by 10 homophase stimuli for a total of 1000 trials. Again the noise in this case was made of ± 400 Hz bandwidth around the centre frequency of 1000 Hz to maintain consistency with the bandwidth used for 500 Hz frequency pure tone.
Experiment 6 (1000 Hz, noise bandwidth to cover higher frequencies)

For the last experiment the stimulus used as a 60 dB 1000 Hz Blackman windowed pure tone masked with continuous Gaussian noise having again wider bandwidth in the range of 20 and 20000 Hz and a total of 1000 trials were conducted as blocks of 10 out-phase and 10 in-phase trials. The signal to noise ratio for the stimuli was 5 dB.

Again the duration of the stimulus in all the experiments was 18 ms. A 200 ms period of silence between each presentation of stimulus was used ensure no response of the middle latency response remains. All experiments presented a total of 1000 stimuli for signal averaging and the auditory evoked potential was measured from the cortical position (Cz) [176, 177] with a reference electrode placed on the left earlobe and ground electrode located on the forehead [172]. Table 5.1 on the next page shows the summary of all the stimuli used for final experiments.
Table 5.1 Final Stimuli – 500 Hz Frequency

<table>
<thead>
<tr>
<th></th>
<th>Experiment 1</th>
<th>Experiments 2 and 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of Subjects</td>
<td>12</td>
<td>12</td>
</tr>
<tr>
<td>Pure tone</td>
<td>Frequency - 500Hz</td>
<td>Frequency - 500Hz masked with noise</td>
</tr>
<tr>
<td>Pure tone duration</td>
<td>18 ms</td>
<td>18 ms</td>
</tr>
<tr>
<td>Silent Duration</td>
<td>200 ms</td>
<td>200 ms</td>
</tr>
<tr>
<td>Windowing Technique</td>
<td>Blackman Window</td>
<td>Blackman Window</td>
</tr>
<tr>
<td>Sound Pressure Level</td>
<td>60 dB</td>
<td>60 dB</td>
</tr>
<tr>
<td>Single Trial Duration</td>
<td>218 ms</td>
<td>218 ms</td>
</tr>
<tr>
<td>Number of trials</td>
<td>1000</td>
<td>1000</td>
</tr>
<tr>
<td>Sequence of Trials</td>
<td>Blocks of 10 antiphase stimuli (180 degree out of phase) and blocks of 10 homophase stimuli (in same phase), 50 times each.</td>
<td>Blocks of 10 antiphase stimuli (180 degree out of phase) and blocks of 10 homophase stimuli (in same phase), 50 times each.</td>
</tr>
<tr>
<td>Noise type</td>
<td>Gaussian Noise</td>
<td></td>
</tr>
<tr>
<td>Noise Bandwidth for Experiment 2</td>
<td>100 – 900 Hz</td>
<td></td>
</tr>
<tr>
<td>Noise Bandwidth for Experiment 3</td>
<td>20 – 20000 Hz</td>
<td></td>
</tr>
<tr>
<td>Interaural Phase Delay of noise</td>
<td>0</td>
<td></td>
</tr>
<tr>
<td>Signal to Noise Ratio</td>
<td>5 dB</td>
<td></td>
</tr>
</tbody>
</table>
Table 5.2 Final Stimuli – 1000 Hz Frequency

<table>
<thead>
<tr>
<th>Final Stimuli – 1000 Hz Frequency</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Experiment 4</strong></td>
<td></td>
</tr>
<tr>
<td>Number of Subjects</td>
<td>12</td>
</tr>
<tr>
<td>Pure tone</td>
<td>Frequency - 1000Hz</td>
</tr>
<tr>
<td>Pure tone duration</td>
<td>18 ms</td>
</tr>
<tr>
<td>Silent Duration</td>
<td>200 ms</td>
</tr>
<tr>
<td>Windowing Technique</td>
<td>Blackman Window</td>
</tr>
<tr>
<td>Sound Pressure Level</td>
<td>60 dB</td>
</tr>
<tr>
<td>Single Trial Duration</td>
<td>218 ms</td>
</tr>
<tr>
<td>Number of trials</td>
<td>1000</td>
</tr>
<tr>
<td>Sequence of Trials</td>
<td>Blocks of 10 antiphasic stimuli (180 degree out of phase) and blocks of 10 homophasic stimuli (in same phase), 50 times each.</td>
</tr>
</tbody>
</table>

| Experiments 5 and 6              |   |
| Number of Subjects              | 12 |
| Pure tone                        | Frequency - 500Hz masked with noise |
| Pure tone duration               | 18 ms |
| Silent Duration                  | 200 ms |
| Windowing Technique              | Blackman Window |
| Sound Pressure Level             | 60 dB |
| Single Trial Duration            | 218 ms |
| Number of trials                 | 1000 |
| Sequence of Trials               | Blocks of 10 antiphasic stimuli (180 degree out of phase) and blocks of 10 homophasic stimuli (in same phase), 50 times each. |
| Noise type                       | Gaussian Noise |
| Noise Bandwidth for Experiment 2 | 600 – 1400 Hz |
| Noise Bandwidth for Experiment 3 | 20 – 20000 Hz |
| Interaural Phase Delay of noise  | 0 |
| Signal to Noise Ratio            | 5 dB |
5.3 First four dominant frequency peaks of final experiment 1

As already established after analysing the preliminary experimental data, the second spectral peak is the important parameter to find the difference in the auditory evoked potentials elicited by homophasic and antiphasic stimuli. The focus will remain on this parameter for analysing the final experimental data. The same positive trend in the difference between the antiphasic and homophasic spectral peak are expected. However, before focusing on the second spectral peaks of the final EEG data, a quick check carried out on Experiment 1 to determine whether the second peak again generates the maximum number of positive amplitude difference. For this reason the Fourier Transformation was carried again on all of the 12 subject's experimental data. The tables (Table 5.3 to 5.5) below shows the amplitude, frequency and phase of the first four peaks extracted from the frequency spectrum of all 12 subject's averaged EEG epoch for both antiphasic and homophasic conditions.
## Experiment 1

Table 5.3 Amplitude, Frequency, Phase of the 4 peaks [Antiphasic AEP]

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 A1Out [µV]</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 3 A3Out [µV]</th>
<th>Peak 4 A4Out [µV]</th>
<th>Peak 1 F1Out [Hz]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 3 F3Out [Hz]</th>
<th>Peak 4 F4Out [Hz]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.7907</td>
<td>0.1397</td>
<td>0.2540</td>
<td>0.1440</td>
<td>8.5333</td>
<td>23.4667</td>
<td>36.2667</td>
<td>48.0000</td>
</tr>
<tr>
<td>2</td>
<td>0.2641</td>
<td>0.2827</td>
<td>0.2074</td>
<td>0.0872</td>
<td>8.5333</td>
<td>20.2667</td>
<td>36.2667</td>
<td>51.2000</td>
</tr>
<tr>
<td>3</td>
<td>0.3207</td>
<td>0.3492</td>
<td>0.1067</td>
<td>0.0913</td>
<td>8.5333</td>
<td>24.5333</td>
<td>51.2000</td>
<td>60.8000</td>
</tr>
<tr>
<td>4</td>
<td>0.2383</td>
<td>0.4460</td>
<td>0.2061</td>
<td>0.3120</td>
<td>8.5333</td>
<td>26.6667</td>
<td>51.2000</td>
<td>54.4000</td>
</tr>
<tr>
<td>5</td>
<td>0.6044</td>
<td>0.3132</td>
<td>0.2473</td>
<td>0.1366</td>
<td>8.5333</td>
<td>23.4667</td>
<td>36.2667</td>
<td>50.1333</td>
</tr>
<tr>
<td>6</td>
<td>0.9029</td>
<td>0.2857</td>
<td>0.1752</td>
<td>0.1455</td>
<td>9.6000</td>
<td>24.5333</td>
<td>39.4667</td>
<td>57.6000</td>
</tr>
<tr>
<td>7</td>
<td>0.9564</td>
<td>0.5153</td>
<td>0.1189</td>
<td>0.2455</td>
<td>8.5333</td>
<td>24.5333</td>
<td>39.4667</td>
<td>51.2000</td>
</tr>
<tr>
<td>8</td>
<td>0.7970</td>
<td>0.3307</td>
<td>0.2156</td>
<td>0.1888</td>
<td>10.2667</td>
<td>20.2667</td>
<td>54.4000</td>
<td>73.6000</td>
</tr>
<tr>
<td>9</td>
<td>0.3025</td>
<td>0.1968</td>
<td>0.1400</td>
<td>0.0972</td>
<td>16.0000</td>
<td>46.9333</td>
<td>61.8667</td>
<td>76.8000</td>
</tr>
<tr>
<td>10</td>
<td>0.4961</td>
<td>0.2477</td>
<td>0.1906</td>
<td>0.1580</td>
<td>8.5333</td>
<td>35.2000</td>
<td>52.2667</td>
<td>62.9333</td>
</tr>
<tr>
<td>11</td>
<td>0.6366</td>
<td>0.4052</td>
<td>1.1821</td>
<td>0.2215</td>
<td>17.0667</td>
<td>32.0000</td>
<td>49.0667</td>
<td>66.1333</td>
</tr>
<tr>
<td>12</td>
<td>1.3171</td>
<td>0.7373</td>
<td>0.5628</td>
<td>0.1729</td>
<td>12.8000</td>
<td>29.8667</td>
<td>44.8000</td>
<td>68.2667</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 PH1Out [degrees°]</th>
<th>Peak 2 PH2Out [degrees°]</th>
<th>Peak 3 PH3Out [degrees°]</th>
<th>Peak 4 PH4Out [degrees°]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>-36.5904</td>
<td>-22.1292</td>
<td>-47.9353</td>
<td>-115.9950</td>
</tr>
<tr>
<td>2</td>
<td>176.9004</td>
<td>110.8171</td>
<td>81.1348</td>
<td>23.7287</td>
</tr>
<tr>
<td>3</td>
<td>-21.2523</td>
<td>-109.1240</td>
<td>-153.0830</td>
<td>0.5542</td>
</tr>
<tr>
<td>4</td>
<td>-134.9890</td>
<td>153.9447</td>
<td>48.0639</td>
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<tr>
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<td>-29.1243</td>
<td>-92.6657</td>
<td>-39.4946</td>
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</tr>
<tr>
<td>6</td>
<td>-70.2196</td>
<td>-73.8532</td>
<td>-158.5050</td>
<td>40.0514</td>
</tr>
<tr>
<td>7</td>
<td>164.5831</td>
<td>114.6312</td>
<td>112.0033</td>
<td>118.8704</td>
</tr>
<tr>
<td>8</td>
<td>-177.6150</td>
<td>163.4087</td>
<td>123.8643</td>
<td>96.7175</td>
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<tr>
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<td>42.6920</td>
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<td>120.6300</td>
<td>119.5156</td>
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<td>125.9532</td>
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</table>
Table 5.4 Amplitude, Frequency, Phase of the 4 peaks [Homophasic AEP]

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 A1In [µV]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 3 A3In [µV]</th>
<th>Peak 4 A4In [µV]</th>
<th>Peak 1 F1In [Hz]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak 3 F3In [Hz]</th>
<th>Peak 4 F4In [Hz]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.6807</td>
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<td>23.4667</td>
<td>35.2000</td>
<td>51.2000</td>
</tr>
<tr>
<td>2</td>
<td>0.2782</td>
<td>0.2364</td>
<td>0.1065</td>
<td>0.0274</td>
<td>19.2000</td>
<td>35.2000</td>
<td>50.1333</td>
<td>62.9333</td>
</tr>
<tr>
<td>3</td>
<td>0.0823</td>
<td>0.1603</td>
<td>0.0887</td>
<td>0.0744</td>
<td>8.5333</td>
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<td>57.6000</td>
<td>80.0000</td>
</tr>
<tr>
<td>4</td>
<td>0.0965</td>
<td>0.1594</td>
<td>0.1832</td>
<td>0.0735</td>
<td>6.4000</td>
<td>21.3333</td>
<td>48.0000</td>
<td>70.4000</td>
</tr>
<tr>
<td>5</td>
<td>0.0525</td>
<td>0.2983</td>
<td>0.1493</td>
<td>0.0653</td>
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<td>27.7333</td>
<td>49.0667</td>
<td>76.8000</td>
</tr>
<tr>
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<td>0.7561</td>
<td>0.2221</td>
<td>0.0933</td>
<td>0.1119</td>
<td>9.6000</td>
<td>26.6667</td>
<td>39.4667</td>
<td>54.4000</td>
</tr>
<tr>
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<td>0.2413</td>
<td>0.2270</td>
<td>0.0659</td>
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<td>75.7333</td>
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<td>8</td>
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<td>0.2666</td>
<td>0.2956</td>
<td>0.2567</td>
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<td>46.9333</td>
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<td>74.6667</td>
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<tr>
<td>9</td>
<td>0.3240</td>
<td>0.1509</td>
<td>0.3406</td>
<td>0.1184</td>
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<td>33.0667</td>
<td>50.1333</td>
<td>67.2000</td>
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<td>0.1708</td>
<td>0.1186</td>
<td>0.1763</td>
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<td>68.2667</td>
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<td>0.3593</td>
<td>0.0901</td>
<td>9.6000</td>
<td>27.7333</td>
<td>52.2667</td>
<td>68.2667</td>
</tr>
<tr>
<td>12</td>
<td>1.2667</td>
<td>0.7311</td>
<td>0.5004</td>
<td>0.3410</td>
<td>11.7333</td>
<td>28.8000</td>
<td>43.7333</td>
<td>57.6000</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 1 PH1In [degrees°]</th>
<th>Peak 2 PH2In [degrees°]</th>
<th>Peak 3 PH3In [degrees°]</th>
<th>Peak 4 PH4In [degrees°]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
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<td>-73.7150</td>
<td>-172.1050</td>
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<td>45.9374</td>
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<tr>
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<td>114.0131</td>
<td>-34.9715</td>
<td>74.8818</td>
</tr>
<tr>
<td>4</td>
<td>-11.5651</td>
<td>-37.3929</td>
<td>-117.0310</td>
<td>15.6680</td>
</tr>
<tr>
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<td>-3.9924</td>
<td>-146.2980</td>
<td>86.9636</td>
<td>24.8773</td>
</tr>
<tr>
<td>6</td>
<td>-52.2898</td>
<td>-104.6410</td>
<td>-169.7960</td>
<td>74.5970</td>
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<tr>
<td>7</td>
<td>-91.6771</td>
<td>135.1081</td>
<td>-179.7380</td>
<td>-130.1830</td>
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<td>-172.1370</td>
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</tr>
<tr>
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</tr>
<tr>
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<td>-126.4860</td>
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<td>135.2038</td>
<td>52.5877</td>
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</tbody>
</table>
Table 5.5 Amplitude difference of the first four peaks

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak1 A1Out-A1In [µV]</th>
<th>Peak2 A2Out-A2In [µV]</th>
<th>Peak3 A3Out-A3In [µV]</th>
<th>Peak4 A4Out-A4In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.1101</td>
<td>0.0513</td>
<td>0.0518</td>
<td>-0.0990</td>
</tr>
<tr>
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<td>-0.0142</td>
<td>0.0462</td>
<td>0.1008</td>
<td>0.0599</td>
</tr>
<tr>
<td>3</td>
<td>0.2384</td>
<td>0.1889</td>
<td>0.0180</td>
<td>0.0169</td>
</tr>
<tr>
<td>4</td>
<td>0.1418</td>
<td>0.2866</td>
<td>0.0229</td>
<td>0.2385</td>
</tr>
<tr>
<td>5</td>
<td>0.5519</td>
<td>0.0149</td>
<td>0.0981</td>
<td>0.0714</td>
</tr>
<tr>
<td>6</td>
<td>0.1468</td>
<td>0.0636</td>
<td>0.0819</td>
<td>0.0335</td>
</tr>
<tr>
<td>7</td>
<td>0.7620</td>
<td>0.2740</td>
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<td>0.1796</td>
</tr>
<tr>
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<td>0.0641</td>
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<td>-0.0680</td>
</tr>
<tr>
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<td>0.0459</td>
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<td>-0.0213</td>
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<tr>
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<td>0.0769</td>
<td>0.0720</td>
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<tr>
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<td>0.8228</td>
<td>0.1315</td>
</tr>
<tr>
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<td>0.0504</td>
<td>0.0062</td>
<td>0.0624</td>
<td>-0.1681</td>
</tr>
</tbody>
</table>

**Findings:** Table 5.5 again shows that the amplitude difference between the second spectral antiphase peak and second spectral homophasic peak have the highest number of positive differences compared to the difference between amplitudes of the first, the third and the fourth antiphase and homophasic second spectral peaks frequency. All the subjects in the study showed higher second spectral peak amplitude when presented with antiphase stimuli. Regarding the phase and its corresponding frequency, no consistent results could again be found for any of the peaks analysed.
Four signals are again generated based on the data of the first four peaks and a summed signal was derived from adding those four signals. All the four signals and the added signals are plotted individually on the same plot along with the averaged AEP epoch (20 ms to 100 ms post stimulation) in the time domain as shown in the figures below. For each subject there are two graphs, representing the in-phase EEG epoch with its first four spectral peak frequency signal plus summed signal and the out-phase EEG epoch with its first four spectral peak frequency plus summed signal. Below shows the graphs of the first two subject’s experimental data.

Figure 5.1 EEG Epoch (In-phase) with dominant frequency signal – Subject 1
Figure 5.2 EEG Epoch (Out-phase) with dominant frequency signal – Subject 1

Figure 5.3 EEG Epoch (In-phase) with dominant frequency signal – Subject 2
Looking at those graphs it is again found that the first four dominant frequency peaks of the signal can capture the main characteristics of the original signal.

### 5.4 Data Analysis for Final Experiments

The second peak in the frequency spectrum of the averaged epoch homophasic and antiphase auditory evoked potentials of the MLR showed a positive trend in the amplitude differences of the second frequency peak for the majority of the preliminary experiments and for all the subjects in the first final experiment. It is also found that the second spectral peak falls around the EEG frequency range of 20 to 50 Hz, indicating that this difference occurs
during the middle latency response (MLR). In order to check whether the achieved results show the same trend for the remaining final experiments the second peak along with the frequency and the amplitude difference for each subject under each experiment are calculated and tabulated as shown below.

**Experiment 2**

Table 5.6 Second dominant frequency peaks and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.1320</td>
<td>29.8667</td>
<td>0.0604</td>
<td>25.6000</td>
<td>0.0716</td>
</tr>
<tr>
<td>2</td>
<td>0.1627</td>
<td>30.9333</td>
<td>0.0655</td>
<td>34.1333</td>
<td>0.0972</td>
</tr>
<tr>
<td>3</td>
<td>0.1661</td>
<td>22.4000</td>
<td>0.0683</td>
<td>35.2000</td>
<td>0.0978</td>
</tr>
<tr>
<td>4</td>
<td>0.1719</td>
<td>21.3333</td>
<td>0.1441</td>
<td>30.9333</td>
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</tr>
<tr>
<td>5</td>
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<td>29.8667</td>
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<td>28.8000</td>
<td>0.0600</td>
<td>26.6667</td>
<td>0.0847</td>
</tr>
<tr>
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<td>0.2919</td>
<td>52.2667</td>
<td>0.2020</td>
<td>36.2667</td>
<td>0.0899</td>
</tr>
<tr>
<td>8</td>
<td>0.1882</td>
<td>30.9333</td>
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</tr>
<tr>
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<td>29.8667</td>
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</tr>
<tr>
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<tr>
<td>11</td>
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</tr>
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<td>0.1819</td>
<td>33.0667</td>
<td>0.0469</td>
</tr>
</tbody>
</table>
## Experiment 3

Table 5.7 Second dominant frequency peaks and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.3227</td>
<td>24.5333</td>
<td>0.2596</td>
<td>26.6667</td>
<td>0.0631</td>
</tr>
<tr>
<td>2</td>
<td>0.1431</td>
<td>36.2667</td>
<td>0.0666</td>
<td>34.1333</td>
<td>0.0765</td>
</tr>
<tr>
<td>3</td>
<td>0.4286</td>
<td>24.5333</td>
<td>0.4212</td>
<td>27.7333</td>
<td>0.0073</td>
</tr>
<tr>
<td>4</td>
<td>0.3572</td>
<td>34.1333</td>
<td>0.0815</td>
<td>22.4000</td>
<td>0.2757</td>
</tr>
<tr>
<td>5</td>
<td>0.4161</td>
<td>32.0000</td>
<td>0.3587</td>
<td>29.8667</td>
<td>0.0574</td>
</tr>
<tr>
<td>6</td>
<td>0.2665</td>
<td>28.8000</td>
<td>0.1900</td>
<td>25.6000</td>
<td>0.0765</td>
</tr>
<tr>
<td>7</td>
<td>0.3562</td>
<td>24.5333</td>
<td>0.2246</td>
<td>29.8667</td>
<td>0.1316</td>
</tr>
<tr>
<td>8</td>
<td>0.1644</td>
<td>34.1333</td>
<td>0.1227</td>
<td>43.7333</td>
<td>0.0418</td>
</tr>
<tr>
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</tr>
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<td>10</td>
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<td>28.8000</td>
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</tr>
<tr>
<td>11</td>
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<td>32.0000</td>
<td>0.1845</td>
<td>29.8667</td>
<td>0.0061</td>
</tr>
<tr>
<td>12</td>
<td>0.5058</td>
<td>27.7333</td>
<td>0.5046</td>
<td>29.8667</td>
<td>0.0012</td>
</tr>
</tbody>
</table>
### Experiment 4

**Table 5.8** Second dominant frequency peaks and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.2049</td>
<td>28.8000</td>
<td>0.1344</td>
<td>29.8667</td>
<td>0.0706</td>
</tr>
<tr>
<td>2</td>
<td>0.1737</td>
<td>29.8667</td>
<td>0.0709</td>
<td>35.2000</td>
<td>0.1028</td>
</tr>
<tr>
<td>3</td>
<td>0.5064</td>
<td>24.5333</td>
<td>0.4893</td>
<td>26.6667</td>
<td>0.0171</td>
</tr>
<tr>
<td>4</td>
<td>0.3959</td>
<td>26.6667</td>
<td>0.2154</td>
<td>39.4667</td>
<td>0.1805</td>
</tr>
<tr>
<td>5</td>
<td>0.4230</td>
<td>26.6667</td>
<td>0.3937</td>
<td>30.9333</td>
<td>0.0293</td>
</tr>
<tr>
<td>6</td>
<td>0.2953</td>
<td>25.6000</td>
<td>0.2648</td>
<td>26.6667</td>
<td>0.0305</td>
</tr>
<tr>
<td>7</td>
<td>0.4013</td>
<td>28.8000</td>
<td>0.2792</td>
<td>29.8667</td>
<td>0.1222</td>
</tr>
<tr>
<td>8</td>
<td>0.3545</td>
<td>24.5333</td>
<td>0.2514</td>
<td>26.6667</td>
<td>0.1032</td>
</tr>
<tr>
<td>9</td>
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<td>21.3333</td>
<td>0.2691</td>
<td>32.0000</td>
<td>0.1194</td>
</tr>
<tr>
<td>10</td>
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<td>25.6000</td>
<td>0.3627</td>
<td>24.5333</td>
<td>0.4579</td>
</tr>
<tr>
<td>11</td>
<td>0.3234</td>
<td>28.8000</td>
<td>0.2935</td>
<td>27.7333</td>
<td>0.0299</td>
</tr>
<tr>
<td>12</td>
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<td>28.8000</td>
<td>0.7089</td>
<td>28.8000</td>
<td>0.1833</td>
</tr>
</tbody>
</table>
## Experiment 5

Table 5.9 Second dominant frequency peaks and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.1529</td>
<td>21.3333</td>
<td>0.1423</td>
<td>25.6000</td>
<td>0.0106</td>
</tr>
<tr>
<td>2</td>
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<td>32.0000</td>
<td>0.1415</td>
<td>36.2667</td>
<td>0.0039</td>
</tr>
<tr>
<td>3</td>
<td>0.3043</td>
<td>48.0000</td>
<td>0.1879</td>
<td>23.4667</td>
<td>0.1163</td>
</tr>
<tr>
<td>4</td>
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<td>43.7333</td>
<td>0.1459</td>
<td>23.4667</td>
<td>0.0490</td>
</tr>
<tr>
<td>5</td>
<td>0.3600</td>
<td>26.6667</td>
<td>0.0845</td>
<td>36.2667</td>
<td>0.2754</td>
</tr>
<tr>
<td>6</td>
<td>0.1728</td>
<td>25.6000</td>
<td>0.1086</td>
<td>26.6667</td>
<td>0.0642</td>
</tr>
<tr>
<td>7</td>
<td>0.3467</td>
<td>23.4667</td>
<td>0.2564</td>
<td>35.2000</td>
<td>0.0903</td>
</tr>
<tr>
<td>8</td>
<td>0.3487</td>
<td>21.3333</td>
<td>0.1435</td>
<td>43.7333</td>
<td>0.2052</td>
</tr>
<tr>
<td>9</td>
<td>0.2789</td>
<td>29.8667</td>
<td>0.1407</td>
<td>25.6000</td>
<td>0.1383</td>
</tr>
<tr>
<td>10</td>
<td>0.3008</td>
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<td>0.2029</td>
</tr>
<tr>
<td>11</td>
<td>0.2172</td>
<td>29.8667</td>
<td>0.1104</td>
<td>35.2000</td>
<td>0.1068</td>
</tr>
<tr>
<td>12</td>
<td>0.3149</td>
<td>41.6000</td>
<td>0.2787</td>
<td>30.9333</td>
<td>0.0363</td>
</tr>
</tbody>
</table>
## Experiment 6

Table 5.10 Second dominant frequency peaks and amplitude difference

<table>
<thead>
<tr>
<th>Subject</th>
<th>Peak 2 A2Out [µV]</th>
<th>Peak 2 F2Out [Hz]</th>
<th>Peak 2 A2In [µV]</th>
<th>Peak 2 F2In [Hz]</th>
<th>Peak2 A2Out-A2In [µV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.2250</td>
<td>29.8667</td>
<td>0.2062</td>
<td>33.0667</td>
<td>0.0188</td>
</tr>
<tr>
<td>2</td>
<td>0.1147</td>
<td>35.2000</td>
<td>0.0496</td>
<td>43.7333</td>
<td>0.0652</td>
</tr>
<tr>
<td>3</td>
<td>0.2859</td>
<td>26.6667</td>
<td>0.1734</td>
<td>27.7333</td>
<td>0.1125</td>
</tr>
<tr>
<td>4</td>
<td>0.2673</td>
<td>33.0667</td>
<td>0.1008</td>
<td>24.5333</td>
<td>0.1665</td>
</tr>
<tr>
<td>5</td>
<td>0.4592</td>
<td>24.5333</td>
<td>0.4221</td>
<td>34.1333</td>
<td>0.0371</td>
</tr>
<tr>
<td>6</td>
<td>0.2834</td>
<td>23.4667</td>
<td>0.1054</td>
<td>37.3333</td>
<td>0.1779</td>
</tr>
<tr>
<td>7</td>
<td>0.3227</td>
<td>22.4000</td>
<td>0.1510</td>
<td>28.8000</td>
<td>0.1717</td>
</tr>
<tr>
<td>8</td>
<td>0.1317</td>
<td>39.4667</td>
<td>0.1316</td>
<td>35.2000</td>
<td>0.0001</td>
</tr>
<tr>
<td>9</td>
<td>0.3681</td>
<td>50.1333</td>
<td>0.0730</td>
<td>33.0667</td>
<td>0.2951</td>
</tr>
<tr>
<td>10</td>
<td>0.1697</td>
<td>45.8667</td>
<td>0.0949</td>
<td>35.2000</td>
<td>0.0748</td>
</tr>
<tr>
<td>11</td>
<td>0.3685</td>
<td>25.6000</td>
<td>0.0523</td>
<td>33.0667</td>
<td>0.3163</td>
</tr>
<tr>
<td>12</td>
<td>0.2509</td>
<td>25.6000</td>
<td>0.1539</td>
<td>27.7333</td>
<td>0.0970</td>
</tr>
</tbody>
</table>

**Findings:** As seen from Table 5.6 – 5.10, there is a clear positive amplitude difference between the antiphasic and homophasic AEPs of the MLR in the second spectral frequency peaks. In addition there are differences to be not for each subject in the frequency value of the second spectral peaks in respect to homophasic and antiphasic evoked response. All the subjects for each experiment showed higher second spectral peak amplitude when
presented with antiphasic stimuli. The positive amplitude difference apparent in Table 5.6 to 5.10 indicated again that antiphasic stimuli yields a higher second dominant frequency peak in the spectrum of the averaged AEP in the middle latency of the brain.

5.5 Statistical Test of the Final results

The final experiments also indicate that higher amplitude of the second spectral peak frequency is found when elicited by antiphasic stimuli of both 500 and 1000 Hz frequency. In this case all the experiments show higher antiphasic second spectral peak in the middle latency response. A binomial probability of the occurrence of the higher antiphasic peak for each experiment is calculated again using the equation 4.2 defined in 4.4. In this case also there are two possible outcomes – either higher antiphasic second spectral peak or higher homophasic second spectral peak with fixed probability of 0.5 is chosen. Since all the 12 subjects for all the six experiments showed higher second antiphasic peak therefore the binaural probability of the occurrence of higher antiphasic second peak is calculated as 0.024% for each experiment. Having a low value of the probability of occurrence of higher antiphasic second spectral peak indicates that the results obtained so far are not due to coincidence.

In this case since all the 72 experiments generate higher second spectral antiphasic peak comparing to second spectral homophasic peak, z-test was not conducted to testify the null hypothesis that: "both the antiphasic and homophasic higher second spectral peak are equally likely to occur (i.e., P
(higher antiphase peak) = P (higher homophase peak) = 0.5) at an alpha level of 0.05” can be rejected or not since the probability of occurrence of equally like event is near zero in this case.

### 5.6 Comparison between experiments - second spectral peak amplitude.

Further analysis of the results is conducted in order to compare between the experiments conducted so far. In this case the second peak amplitude difference for each subject under each experiment is normalised again as was done for the preliminary experiments. Normalisation is based on the average amplitude of second spectral homophase and antiphase peak. The formula is:

\[
\text{Normalised amplitude difference} = \frac{\text{Amplitude difference}}{\text{Average amplitude}}
\]

Once the second spectral frequency peak difference was normalised a mean value of all the normalised amplitude difference for each experiment was calculated and a comparison between the experiments was conducted based on the magnitude of the mean value achieved for each experiment. Table 5.11 shows the normalised value of the second spectral frequency peak amplitude difference for all the subjects of the final experiment 1, 2 and 3.
In addition the mean of all 12 normalised amplitude differences under each experiment is also calculated based on equation 5.1 and added to the table.

\[
Mean = \frac{1}{12} \sum_{i=1}^{12} \frac{(A2:OUT_i - A2:IN_i)}{(A2:OUT_i + A2:IN_i) \cdot \frac{1}{2}}
\]  

(5.1)

The first three final experiments are presented in Table 5.11 as their stimuli were all 500 Hz frequency Blackman Windowed pure tone signals.

Table 5.11 Normalised amplitude difference for Final Experiment 1, 2 and 3

<table>
<thead>
<tr>
<th>Subject</th>
<th>Experiment 1 10 Block Out – 10 Block In</th>
<th>Experiment 2 10 Block Out – 10 Block In (100 Hz to 900 Hz – masked noise)</th>
<th>Experiment 3 10 Block Out – 10 Block In (20 Hz to 20000 Hz – masked noise)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.4495</td>
<td>0.7447</td>
<td>0.2166</td>
</tr>
<tr>
<td>2</td>
<td>0.1781</td>
<td>0.8519</td>
<td>0.7294</td>
</tr>
<tr>
<td>3</td>
<td>0.7417</td>
<td>0.8342</td>
<td>0.0173</td>
</tr>
<tr>
<td>4</td>
<td>0.9467</td>
<td>0.1764</td>
<td>1.2571</td>
</tr>
<tr>
<td>5</td>
<td>0.0486</td>
<td>0.5955</td>
<td>0.1482</td>
</tr>
<tr>
<td>6</td>
<td>0.2505</td>
<td>0.8273</td>
<td>0.3352</td>
</tr>
<tr>
<td>7</td>
<td>0.7244</td>
<td>0.3640</td>
<td>0.4530</td>
</tr>
<tr>
<td>8</td>
<td>0.2146</td>
<td>0.3781</td>
<td>0.2910</td>
</tr>
<tr>
<td>9</td>
<td>0.2639</td>
<td>0.1982</td>
<td>0.9987</td>
</tr>
<tr>
<td>10</td>
<td>0.3676</td>
<td>0.4783</td>
<td>0.2451</td>
</tr>
<tr>
<td>11</td>
<td>0.1980</td>
<td>1.0879</td>
<td>0.0325</td>
</tr>
<tr>
<td>12</td>
<td>0.0084</td>
<td>0.2283</td>
<td>0.0023</td>
</tr>
<tr>
<td>MEAN</td>
<td>0.3660</td>
<td>0.5637</td>
<td>0.3939</td>
</tr>
</tbody>
</table>

A graphical representation of the mean normalised amplitude difference for the final experiment 1, 2 and 3 is shown in the Figure 5.5.
Findings: Looking at the mean value of the normalised second peak amplitude difference of the first three experiments it is quite clear that experiment 2 yields the largest mean of the normalised amplitude difference compared to the other two. In this experiment the 500 Hz pure tone stimulus was masked with noise of a bandwidth of 100 Hz to 900 Hz. This again demonstrates that narrow band noise maskers elicits a larger mean value of the normalised amplitude difference in the second spectral peak frequency component upon presentation of in-phase and out-phase stimuli. The second largest difference is found for experiment 3 where the stimulus was embedded with band noise of 20 Hz to 20 kHz. In experiment 1 where stimulus was a pure tone of 500 Hz yields the smallest difference. The last three final experiments uses stimuli made of 1000 Hz frequency pure tone. Table 5.12 shows the normalised value of the second spectral
frequency peak amplitude differences for all the subjects of the final Experiment 4, 5 and 6. In addition the mean of all 12 normalised amplitude mean differences are also calculated and added to the table.

Table 5.12 Normalised amplitude difference for Final Experiment 4, 5, 6

<table>
<thead>
<tr>
<th>Subject</th>
<th>Experiment 4</th>
<th>Experiment 5</th>
<th>Experiment 6</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>AmpDiff/AvgAmp</td>
<td>AmpDiff/AvgAmp</td>
<td>AmpDiff/AvgAmp</td>
</tr>
<tr>
<td>1</td>
<td>0.4158</td>
<td>0.0716</td>
<td>0.0872</td>
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<td>0.0269</td>
<td>0.7935</td>
</tr>
<tr>
<td>3</td>
<td>0.0343</td>
<td>0.4728</td>
<td>0.4899</td>
</tr>
<tr>
<td>4</td>
<td>0.5906</td>
<td>0.2877</td>
<td>0.9045</td>
</tr>
<tr>
<td>5</td>
<td>0.0719</td>
<td>1.2394</td>
<td>0.0841</td>
</tr>
<tr>
<td>6</td>
<td>0.1089</td>
<td>0.4560</td>
<td>0.9153</td>
</tr>
<tr>
<td>7</td>
<td>0.3590</td>
<td>0.2995</td>
<td>0.7251</td>
</tr>
<tr>
<td>8</td>
<td>0.3405</td>
<td>0.8338</td>
<td>0.0011</td>
</tr>
<tr>
<td>9</td>
<td>0.3631</td>
<td>0.6591</td>
<td>1.3378</td>
</tr>
<tr>
<td>10</td>
<td>0.7739</td>
<td>1.0177</td>
<td>0.5657</td>
</tr>
<tr>
<td>11</td>
<td>0.0969</td>
<td>0.6518</td>
<td>1.5033</td>
</tr>
<tr>
<td>12</td>
<td>0.2290</td>
<td>0.1222</td>
<td>0.4794</td>
</tr>
<tr>
<td>MEAN</td>
<td><strong>0.3520</strong></td>
<td><strong>0.5115</strong></td>
<td><strong>0.6572</strong></td>
</tr>
</tbody>
</table>
The graphical representation of the normalised amplitude difference for the last three final experiments is shown in Figure 5.6.

![Mean of Normalised Amplitude Difference Final Experiment 4, 5 and 6](image)

Figure 5.6 Mean of Normalised Amplitude Difference Final Experiment 4, 5, 6

**Findings:** The mean value calculated for the last three final experiments show a similar trend as found in the first three final experiments of the 500 Hz tone. In both the cases tones embedded with noise yield a larger mean difference comparing to the tones without noise. Experiment 5, where the tone was masked with 600 to 1400 Hz band noise, yields a larger mean of the normalised second spectral peak amplitude difference compared to experiment 4 where the stimuli generated were only a pure tone of 1000 Hz. However contrary to the first three experiments, final Experiment 6, where the tone is masked with wide band of 20 Hz to 20 kHz noise, yields a larger mean than Experiment 5 where narrow band noise was applied.
5.7 Discussions of results of the final experiments

The purpose of conducting final experiments was to verify whether the conclusions which were drawn after conducting the preliminary experiments remain valid or not.

After analysing the final experimental data it is found that indeed the second spectral peak that lies in the frequency band gap of around 20 to 50 Hz can be used to find the difference in the middle latency response of the auditory evoked response elicited by homophase and antiphase stimuli. Spectral analysis of the data using the Discrete Fourier Transform demonstrated a higher amplitude of the second frequency peak is found for the AEP epoch (20ms to 100 ms post stimulation averaged epoch) when elicited by antiphase stimuli. All the subjects showed positive amplitude difference between the second spectral peaks extracted from antiphase and homophase MLR as shown in Table 5.5 to 5.10.

The frequency range of the second peak indicates that this effect possibly relates to the middle latency response (MLR).

The higher second spectral peak amplitude found in the auditory response evoked due to antiphase stimuli supports Durlach’s “Equalization and Cancellation” theory.
For both 4.8 KHz and 19.2 KHz sampling rates the second dominant spectral peak was found in the 20 to 50 Hz frequency band which indicates that sampling rate of 4.8 KHz of the EEG signal will be sufficient for this process.

It is again found that a pure tone embedded with noise yields the higher mean value of normalised second spectral peak amplitude differences compared to the values achieved for pure tone with no noise. However, no conclusions could be reached on the difference between narrow band noise and broad band noise as the results of experiments 2 and 3 were contradicted by experiments 5 and 6. This is an area for further research and will be included in recommended future work.

The findings of this research are likely to relate the psychoacoustic phenomenon known as binaural masking level difference (BMLD). The overall results indicate that the detection of a signal in a background of noise is much easier when the signal has a different inter-aural phase difference than that of the noise, as was to be expected from literature [137, 138].

Finally the amplitude difference between the spectral peaks may be used to build an objective methodology to detect electrophysiological changes of binaural processing in the middle latency response. The results achieved so far suggest that this difference indicates the presence of binaural processing in the human brain.
5.8 Summary

The chapter introduced the final experiments designed based on the results of the preliminary experiments, which indicated the presence of binaural hearing in the binaural processing of the homophasic and antiphasic stimuli. Several changes and adjustments were made for the final experiment. The second spectral peak responses based on homophasic and antiphasic stimuli were analysed. The final experimental data again demonstrates higher amplitudes of the second spectral peak in the middle latency response when elicited by antiphasic stimuli with and without noise. The overall findings of this research indicate that the amplitude of the second spectral peak in the 20 - 50 Hz frequency band of the middle latency response can be used as a marker for building an approach in detecting the binaural processing of the brain.
Chapter 6: Conclusions and Future Work

The aim of this research was to develop, test and evaluate an approach towards building an objective methodology to detect binaural processing in the human brain using auditory evoked potentials.

Auditory evoked potentials (AEPs) were recorded and analysed to investigate the electrophysiological effect in the neural activity based on the phase reversal of the binaural stimuli. AEPs for both homophasic and antiphasic conditions were obtained by averaging 500 trials of in-phase and 500 trials of out-phases of each EEG epoch. Stimuli consisted of 500 Hz and 1000 Hz pure tones of 18 ms duration, mixed with various noise conditions. The AEPs were then analysed in the frequency domain.

Two series of experiments were carried out, consisting of 56 preliminary experiments to test the equipment and approach and develop a signal processing methodology and the final 72 experiments to validate the developed signal processing methodology.
6.1 Conclusions

After analysing preliminary and final experimental EEG data it was found that the amplitude of the second spectral peak of the MLR was larger for the antiphasic condition than the homophasic condition. The peak occurred in the range of 20 – 50 Hz. This held for 79% of the preliminary experiments, and all of the final experiments.

The results indicate that the amplitude of the second peak may be used as a marker for binaural processing in the human brain. It is hoped that this may lead to an objective technique to assist with the diagnosis of auditory processing disorders in the future.

The behaviour of the second peak indicates changes in the middle latency response (MLR) as the frequencies of the second peak correspond with this response (20 – 50 Hz).

The results support Durlach’s “Equalization-Cancellation” theory which suggests that humans will hear stimuli better when they are presented out of phase [123, 140].

It was also found that pure tone stimuli embedded with noise yield higher mean values of normalised second spectral peak amplitude differences than stimuli of pure tones without noise. This supports the findings of Ira and Licklider that the detection of a signal in a background of noise is much
easier when the signal has a different inter-aural time or phase difference than that of the noise [137, 138]. However, no conclusive results could be derived whether narrow band or broad band noise is better for the detection of binaural activity.

The findings of this research are likely to relate to the psychoacoustic phenomenon known as binaural masking level difference (BMLD). The overall results indicate that the detection of a signal in a background of noise is easier when the signal has a different inter-aural time difference than that of the noise, as was expected from literature [137, 138].

It was also noted that the normalised amplitude differences were more pronounced when phase shifts occurred every few seconds than when phase shift only occurred once per epoch. This may be due to the brain entering into a steady state response to the stimuli. Random phase shifting did not result in improved detection.

Finally the findings of this research indicate that the proposed approach may lead towards an effective way of building an objective methodology to quantify a person's binaural hearing ability using the middle latency response of the brain.
6.2 Future Work

The development of an objective method for measuring binaural processing might be a step towards understanding auditory and speech processing disorders which are linked to reduced binaural hearing ability. In order to further investigate and improve the developed method, the following recommendations for future research are made to progress this research.

6.2.1 Stimuli & Measurement

A 180 degree phase difference between the left ear and the right ear tone played an important cue and the impact of this was quite evident in the second frequency peak of the EEG signal. However, future research could further enhance the stimuli for binaural hearing by investigating other cues:

- inter-phase difference (IPD) with phase changes of signals or noise other than ±180 degrees;

- inter-aural level difference (ILD) for signals above 2 kHz. Signals below 2 kHz should also be considered to validate the assumption that the IPD is a better marker in this region than the ILD.; and

- inter-aural time difference (ITD) instead of inter-aural phase difference (IPD). ITD changes both the starting point and the phase of the stimuli. Inter-phase difference with phase changes of signals or noise other than ± 180 degrees could be investigated;
The best stimuli for this research consisted of sequences of block of 10 antiphasic tones followed by block of 10 homophasic tones. Block size and composition could be further investigated to further optimise the results.

Binaural masking level difference (BMLD) experiments using the same stimuli could be compared. Sound stimuli based on the best BMLD results obtained can be developed for EEG recordings.

The total time of experimentation should be reduced to avoid subject fatigue. The time to prepare a subject can be reduced by using an electrode montage cap instead of the current cap. The time of experiments may be further reduced by minimising the duration of the audio signal, as well as minimising the pause between audio signals.

All experiments were done at the cortical position Cz of the skull. Measuring only one electrode location limits the understanding of relation between different skull positions. Measurement at multiple locations of the skull may also provide insight about the propagation of the signal in the brain. This may lead to a better placement of the electrode(s).
6.2.2 Processing Methodology

More advanced artifacts rejection methods could be investigated to minimise the impact of artifacts.

Advanced feature extraction algorithms can be developed and the data can be processed based on a time-frequency varying technique. The potential of wavelet analysis and statistical approach such as PCA, ICA, in analysing the EEG signals can be explored. Use of Box and Whisker plots can be another option to look into the distribution of the data and for the representation and enrichment of the results. In addition, analysing the spectral power of the EEG signals can be another parameter to investigate.

The auditory brainstem response and the late latency response in the auditory evoked potential can be investigated in the future.

6.2.3 Subjects

All subject of this research had normal hearing. The research could be expanded to include subjects with abnormal hearing.

Subject age should be expanded to investigate whether the developed methodology can be applied to young children. This may simplify the detection of children with binaural hearing defects.
6.3 Summary

This chapter includes the conclusions and the future work of the thesis. Novel findings were summarised. A number of the findings match with the expectations from existing research based on binaural hearing. The main conclusion is that the amplitude of the second peak may be used as a marker for binaural processing in the human brain. This may lead to an objective technique to assist with the diagnosis of auditory processing disorders in the future. In addition, recommendations for future work were proposed such as expanding the research to people with hearing disorder, looking at other possible cues for binaural hearing, comparing the results with BMLD experiments and using more advanced feature extraction algorithms and statistical procedures.
Appendix A. Hardware Specifications

A.1 Computer Specifications

<table>
<thead>
<tr>
<th>Hardware</th>
<th>Properties</th>
</tr>
</thead>
<tbody>
<tr>
<td>CPU</td>
<td>Intel(R) Core (TM) i7</td>
</tr>
<tr>
<td></td>
<td>CPU 870 @ 2.90 GHz 3.07 GHz</td>
</tr>
<tr>
<td>Hard Disk</td>
<td>80 Gigabyte</td>
</tr>
<tr>
<td>RAM</td>
<td>8.00 GB</td>
</tr>
<tr>
<td>Operating system</td>
<td>Windows 7 Enterprise Service Pack 1</td>
</tr>
<tr>
<td>Soundcard</td>
<td>Sound Blaster Audigy 4, Creative</td>
</tr>
</tbody>
</table>
## A.2 Headphone AKG K271 Specifications

<table>
<thead>
<tr>
<th>Specification</th>
<th>Details</th>
</tr>
</thead>
<tbody>
<tr>
<td>Circum-aural, closed-back headphones with self-adjusting headband and auto-shut-off feature</td>
<td></td>
</tr>
<tr>
<td>Applications: Professional stereo-studio-headphones for live sound and studio monitoring</td>
<td></td>
</tr>
<tr>
<td>Frequency range: 16–28.000 Hz</td>
<td></td>
</tr>
<tr>
<td>Sensitivity: 91 dB at 1 mW / 104 dBV</td>
<td></td>
</tr>
<tr>
<td>Impedance: 55 ohms per channel</td>
<td></td>
</tr>
<tr>
<td>Power handling capability: 200 mW</td>
<td></td>
</tr>
<tr>
<td>Cable: 3 m (OFC) single-sided, plug-in, easily replaceable cable with mini-XLR connector and with hard-gold plated, screw-able 1/4” jack plugs</td>
<td></td>
</tr>
</tbody>
</table>
### A.3 Radio Shack SLM Specifications

<table>
<thead>
<tr>
<th>Specification</th>
<th>Details</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Microphone type:</strong></td>
<td>Electret Condenser</td>
</tr>
<tr>
<td><strong>Range:</strong></td>
<td>30 to 130 dB</td>
</tr>
<tr>
<td><strong>Accuracy:</strong></td>
<td>± 2 dB @ 94 dB SPL</td>
</tr>
<tr>
<td><strong>Reference 0 dB = 0.0002 MicroBar</strong></td>
<td></td>
</tr>
<tr>
<td><strong>Weighting:</strong></td>
<td>A and C</td>
</tr>
<tr>
<td><strong>Frequency:</strong></td>
<td>32 to 10,000Hz</td>
</tr>
<tr>
<td><strong>Display Response:</strong></td>
<td>Fast and Slow</td>
</tr>
<tr>
<td><strong>Signal output:</strong></td>
<td>1 Volt (peak-peak min.)</td>
</tr>
<tr>
<td><strong>Impedance:</strong></td>
<td>10 kOhms (min. load)</td>
</tr>
<tr>
<td><strong>Distortion:</strong></td>
<td>Less than 2% at 1 kHz.</td>
</tr>
<tr>
<td><strong>Power:</strong></td>
<td>9 Volts battery</td>
</tr>
<tr>
<td><strong>Dimensions (HWD):</strong></td>
<td>$5{\frac{3}{4}} \times 2{\frac{1}{5}} \times 1{\frac{2}{5}}$ inch (145 × 55 × 35 mm)</td>
</tr>
<tr>
<td><strong>Weight:</strong></td>
<td>4.6 oz - 130 grams</td>
</tr>
</tbody>
</table>
### A.4 Soundcard Specifications

<table>
<thead>
<tr>
<th>INTERFACE/BUS- PCI 32-bit bus-mastering AUDIGY 4 PROCESSOR</th>
</tr>
</thead>
<tbody>
<tr>
<td>64-voice hardware wavetable synthesizer</td>
</tr>
<tr>
<td>24-bit Digital-to-Analog conversion</td>
</tr>
<tr>
<td>Sampling rates of 8, 11.025, 16, 22.05, 24, 32, 44.1, 48 and 96 kHz in 7.1 modes and up to 192 kHz in stereo mode.</td>
</tr>
<tr>
<td>24-bit Analog-to-Digital conversion during recording in 8, 16 or 24-bit at sampling rates of 8, 11.025, 16, 22.05, 24, 32, 44.1, 48 and 96 kHz.</td>
</tr>
<tr>
<td>Supports Sony/Philips Digital Interface (SPDIF) format of up to 24-bit/96kHz quality.</td>
</tr>
<tr>
<td>Low latency multitrack recording with ASIO2 support.</td>
</tr>
<tr>
<td>AUDIGY 4 ON-BOARD CONNECTORS- Line level out (Front/Side/Rear/Centre) Line In-Microphone-In Digital I/O (for stereo SPDIF output to Digital I/O Module).</td>
</tr>
</tbody>
</table>
### A.5 Specifications of USB amplifier

<table>
<thead>
<tr>
<th>Feature</th>
<th>Details</th>
</tr>
</thead>
<tbody>
<tr>
<td>16 DC-coupled wide-range input channels per unit, 4 independent grounds, record any type of signal (EEG/ECoG/ECG/EMG/EOG/...)</td>
<td>connect various sensors, stack units for 32/48/64/... channels.</td>
</tr>
<tr>
<td>24-bit resolution with simultaneous sampling of all channels with up to 38.4 kHz, digital signal filtering and pre-processing, connect via USB 2.0.</td>
<td>Works with passive and with active electrodes, 8 digital trigger inputs/unit, 4 digital outputs/unit, new simplified synchronization of units.</td>
</tr>
<tr>
<td>Internal digital bandpass and notch filters, built-in calibration unit and impedance checking.</td>
<td>Easy configuration and setup via the software, high-speed online data processing for SIMULINK and for LabVIEW available, recommended by BCI2000</td>
</tr>
<tr>
<td>Driver package/API available</td>
<td></td>
</tr>
<tr>
<td>CE-certified and FDA-approved medical device, safety class: II, conformity class: Ila, type of applied part: CF</td>
<td></td>
</tr>
</tbody>
</table>
Appendix B. Software Specifications

B.1 Digital Audiometer (Professional)

<table>
<thead>
<tr>
<th>Test signal type: Pure Tones and Warble Tones</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frequency range: 20 Hz to 20,000 Hz (Professional version) and 125 Hz to 8,000 Hz (Screening version)</td>
</tr>
<tr>
<td>Frequency resolution: 24 frequencies (Professional version) and 11 frequencies (Screening version)</td>
</tr>
<tr>
<td>Frequency stability: +/- 0.0001 Hz (at 1000 Hz)</td>
</tr>
<tr>
<td>S/N = 96 dB (for 16 bit systems)</td>
</tr>
<tr>
<td>Amplitude (SPL or HL) range: -20 dB to 80 dB</td>
</tr>
<tr>
<td>Amplitude resolution: 1 dB or 5 dB (user-selectable)</td>
</tr>
<tr>
<td>Amplitude stability: +/- 0.0003 dB</td>
</tr>
<tr>
<td>No harmonic and inter-modulation distortions</td>
</tr>
<tr>
<td>Calibration modes: Aural with ± 3 dB precision and SLM-based with ± 0.1 dB precision</td>
</tr>
<tr>
<td>Test modes: Automatic and manual with instant display of threshold values. In automatic mode one can pre-select test frequencies for fast testing (for example 500 Hz, 1 kHz, 2 kHz, 4 kHz)</td>
</tr>
</tbody>
</table>
Appendix C. Calibration

C.1 Audiometer Calibration

Open the file Digital_Audiometer.html to display the Digital Audiometer panel in the Internet browser as shown in Figure C.1. Once the software was successfully installed the next phase of the experiment is to calibrate the audiometer.

1. Once the Digital Audiometer panel is displayed in the Internet browser, the license key need to be entered in the field on the upper left corner of the Digital Audiometer panel and do not press the ‘Return’ or ‘Enter’ key on the keyboard.
2. Click the ‘START Calibration’ button


4. Attach the acoustic coupler at the microphone end of the SLM and ensure that it is properly secured as shown in Figure C.2.

![Figure C. 2 DR1-R collar and the SLM](image)

5. Gently put the DR1-R on the microphone of the SLM. Push and rotate back and forth until seated as shown in Figure C.3.

![Figure C. 3 SLM with acoustic coupler](image)
6. The lights in the sound proof should be switched off before the calibration starts. Insert the SLM with DR1-R between the L & R headphones. According to Figure C.4, the SLM’s microphone should point toward the L headphone. There should be a good seal between the headphone and the DR1-R acoustical coupler (air should not be able to flow via gaps).

![Figure C. 4 Setup of Headphones, SLM and Acoustic Coupler](image)

7. Set the SLM to ‘C’ and ‘Fast Response’ and set the scale to 60 dB so that the audiometer can be calibrated to 60dB.

8. The sound card mixer level and the volume slider of the Digital Audiometer panel should be adjusted to achieve calibration in the upper range (-6.0 dB – 0 dB) in order to permit the most effective SNR performance.

9. A minimum time of 80 seconds is required for the warm up of the SLM and an accurate reading.

10. Calibration is considered achieved when the SLM displays a reading of 60 dB.

11. Record the volume slider decibel reading (for example -2.2 dB)
12. Repeat the aural volume calibration process for the right headphone. Do not change the sound card mixer volume as this will ruin the calibration for the left panel of the headphone.

13. Ensure that the signal volume is identical in both channels and if necessary, move the volume calibration slider on the Digital Audiometer panel.

14. Ensure that the difference between the two channels should not exceed ±0.1 dB.

15. Once satisfied, click the ‘STOP Calibration’ button.

The software and headphones are now calibrated. It is not necessary to recalibrate the headphones each time a test is required, if one has not changed the mixer settings on the PC and the recorded values of calibration were correct. It is advisable to check these before testing whenever possible however.
C.2 Settings on Digital Audiometer

1. There are two types of testing mode available for hearing test and they are
   Pure Tone (PT) test mode
   Warble Tone (WT) test mode.
   The difference between the pure tone and warble tone test mode is the type of tone played to conduct the hearing test and any one of the testing mode can be selected. Since we are conducting pure tone audiometric testing so pure tone (PT) test mode was selected in this case.

2. The test can be run manually or automatically. Auto test was selected by selecting the ‘AUTO Test’ radio button on the upper right side of the Digital Audiometer panel in order to avoid the user interference.
3. There are two types of SPL resolution and can be set to either 1 or 5 dB. In order to minimise the testing time SPL resolution of 5dB was selected.

4. The [min] and [max] buttons located next to the vertical axis on the audiogram allow the adjustment of max and min SPL (or HL) levels in the -20 dB to 80 dB range. Hearing tests are conducted having the full range from -20 dB to 80 dB.

5. The HL (hearing loss) scale is the default setting and the setting should not be changed.

6. Frequencies that need to be tested can be selected by pressing the square button on the horizontal axis of the digital audiometer. The conventional frequency range for audiometric testing is 500 Hz to 8 kHz. This research tested the a frequency range with the following intervals: 500 Hz, 750 Hz, 1kHz, 1.5 kHz, 3 kHz, 5 kHz and 8 kHz.

7. For each selected frequency, the hearing test will commence at -20 dB and increase in steps of 10 dB. Once the subject signals a response, the test level will drop by 20 dB and then increase in steps of 5 dB and the subject will be tested with the same frequency again in order to ensure reliability.

8. The coloured circles that appear beneath each frequency statistically represent the accuracy and reliability of the results and are explained below.
The reliability scale is as follows –

<table>
<thead>
<tr>
<th>COLOUR</th>
<th>DISCREPANCY RANGE</th>
<th>MEASUREMENT RELIABILITY</th>
</tr>
</thead>
<tbody>
<tr>
<td>GREEN</td>
<td>0 - 5 dB</td>
<td>Good</td>
</tr>
<tr>
<td>ORANGE</td>
<td>5 - 10 dB</td>
<td>Acceptable</td>
</tr>
<tr>
<td>RED</td>
<td>10 - 15 dB</td>
<td>Objectionable</td>
</tr>
<tr>
<td>BLACK</td>
<td>&gt; 15 dB</td>
<td>Not reliable</td>
</tr>
</tbody>
</table>

9. The ‘STAT’ button and the DISP button should be turned on and the setting should not be changed. Both the button is used for displaying coloured circles and the subject’s for each frequency (i.e. it presents a visual assessment of the test reliability for the chosen number of resolutions)

10. The test rate (i.e. time interval between tests) is set to default at 1 second – do not change this setting
Appendix D Forms and Checklist

D.1 Consent Form

**TITLE OF STUDY:** Detection of binaural processing in the human brain / Standard Hearing Test

**INVESTIGATORS:** Sami Azam

**Procedure:**
Initially you will be asked to have a hearing test as this project requires the subjects to have normal hearing ability. It is a computer-based test and you will be seated in a sound-isolated laboratory. Tones of increasing frequency will be generated and you are required to respond to those tones by pressing the space bar on the keyboard as soon as you hear the tones. The hearing test will take no longer than 20 minutes.

If you are deemed to have normal hearing, you will then be asked to participate in the next phase of the experiment. The electroencephalogram (EEG) recording requires the placement of a cap and electrodes at specific locations on your head. This involves the process of placing gel into each electrode cup using a cotton bud, the gel is then gently worked into the hair and scalp. Once this preparation is complete, the experiment will commence. You will be asked to remain still and calm whilst you will hear certain sounds and your brain activity is recorded.
Risks and Inconveniences:

The risks involved in the study are minimal. All equipment that will come into direct contact with your skin will be sterile, and at no point will the skin on your scalp be broken. The greatest inconvenience to you will be a small amount of electrode gel in your hair or minor skin irritation after the cap is removed. The gel can easily be rinsed out of your hair with soap and water at the conclusion of the experiment.
I hereby give my consent to be the subject of your research. You have given me:

A – An explanation of the procedures to be followed in the project, including an identification of those that are experimental.

B – Answers to any inquiries that I have made.

I understand that:

A – My participation is voluntary and I may withdraw my consent and discontinue participation in the project at any time. My refusal to participate will not result in any penalty.

B – By signing this agreement, I do not waive any legal rights or release Charles Darwin University, its agents or you from liability for negligence.

Name:

Gender:

Age:

Contact Number:

_________________________________
Signature

_____________________
Date
1. Title of research:

   **Detection of binaural processing in the human brain**

2. Questions or queries you may have about this research, please contact:

   Mr. Sami Azam, Research Student
   School of Engineering and Information Technology
   Ellengowan Drive, Casuarina Campus
   Charles Darwin University
   Darwin   NT   0909
   Australia
   Telephone: 08 8946 7646
   Fax: 08 8946 6680
   e-mail: sami.azam@cdu.edu.au
D.2 Hearing Test Checklist

1. Does the software used comply with the requirements of Australian Standards? Yes / No

2. Is the software calibrated in accordance with the requirements of headphone and sound level meter? Yes / No

3. Has a listening check been carried out on the day of the test? Yes / No

4. Has a basic calibration of the software been carried out every time before the test? Yes / No

5. Have the background noise levels been measured? Yes / No

6. Are the background noise levels suitable for the combination of earphone and software? Yes / No

7. Has the test subject been made fully aware of the reason why audiometry is being made available? Yes / No

8. Has the test subject been encouraged to have positive motivation towards the test? Yes / No

9. Has the condition of the test subject’s hearing protectors been examined? Yes / No

10. Does the test subject wear the hearing protectors correctly? Yes / No

11. Are the circumstances in which audiometry is carried out such that the test subject will not be disturbed or distracted by events unrelated to the test procedure nor by people in the surroundings? Yes / No

12. Has the test subject informed of the results of the audiometry? Yes / No

13. Has the specified minimum amount of information been obtained, recorded and reported, where appropriate, for each test subject tested? Yes / No

14. Have the specified test procedures been followed? Yes / No
D.3 Hearing Test Questionnaire

1. Have you ever had hearing trouble? □ Yes / □ No
2. Has anyone ever suggested that you have hearing trouble? □ Yes / □ No
3. Have you even worked in a noisy environment? □ Yes / □ No
4. Do you currently work in a noisy environment? □ Yes / □ No
5. Have you ever had ear problem as a child or adult? □ Yes / □ No
6. If yes, did you have these problems frequently or for lengthy amount of time? □ Yes / □ No
7. Have you ever had ear infections? □ Yes / □ No
8. Is it difficult for you to follow a conversation when many people are in the same room? □ Yes / □ No
9. Do you have problems to follow a normal conversation with several people?
   □ Yes / □ No
10. Do you get confused about where sounds come from? □ Yes / □ No

Name: __________________________
Date: __________________________

______________________________
Signature
Appendix E Hearing Standards

Audiometers

The audiometer that is going to be used for conducting hearing test should have the ability to play test tones of 500 Hz, 1 kHz, 1.5 kHz, 2 kHz, 3 kHz, 4 kHz, 6 kHz, 8 kHz. The conventional frequency range for audiometric testing is 500 Hz to 8 KHz. The software application Digital Audiometer Professional (DAP) was selected to conduct the hearing test. DAP can test a frequency range between 0 and 22,000 Hz with an amplitude testing range for each frequency of -20 dB to 100 dB.

Acoustical Environment for Audiometry

The maximum acceptable background noise level for testing is 30 dB for both A and C frequency weightings (as measured on the sound level meter, SLM). This will ensure that the acoustical environment (i.e. soundproof facility) is acceptable for both the audiometric and EEG test but also that noise Artefacts are minimized. The soundproof room was inspected and the background sound pressure level of the room was below 25 dB.
Room temperature
According to Australian standard the room temperature of the sound proof lab should be within the range of 15 deg C to 35 deg C. The audiometric environment where the experiments were conducted had a constant temp of 25 deg C.

Frequency Range
As already stated that the conventional frequency range for audiometric testing should be in the range from 500 Hz to 8 kHz, this research tested the frequency range: 500 Hz, 750 Hz, 1 kHz, 1.5 kHz, 3 kHz, 5 kHz and 8 kHz.

Sound level Meter: Frequency Weighting and Accuracy
The sound level meter Radio Shack that was used for measuring background noise level and for headphone calibration had both A and C frequency weighting characteristics and claims a good degree of accuracy.

Recording of Results
According to the Australian standard the following details shall be recorded before conducting the hearing test:

- General Details that includes name of tester, date of test, make, model identification number and type of headphone used.
- Subject’s name, address, contact number, date of birth and sex.
- All subject’s history related to hearing was initially screened by asking them to fill the self-assessment hearing questionnaire. The
questionnaire will provide indication if the participant has a history of any otological disease or condition.

- Certify the participant has signed the consent form and understood all components.

**Instruction to subject**

- All the subject should avoid significant noise exposure (e.g. attendance at a night club within the past 24 hours) prior to audiometric testing should be avoided as this can temporarily elevate thresholds

- All the subjects need to be asked to arrive at least 10 minutes before the hearing test.

- The subject should be asked about any recent noise exposure, which should be noted as a comment on the audiogram form.

- Ask for any queries if the subject have, and then provide a satisfactory response.

**Subject Preparation**

According to the Australian Standards there are specific rules that need to be followed during the preparation of the subject. It is important to place the headphones on the subject in such a way that the headset is secure and the caps are level with the entrance to the ear canal. It is also important to assure that there is a proper seal and the subject is also comfortable with the headphone placement. If any object or hair is in the path between the headphone and ear, or jewellery is interfering with the placement of
headphones, all the obstructions need to be cleared. Ensure that the headphone L and R position are placed properly on the associated ears. During the experiment all the subject should be reminded of keeping their movement to minimum and should not touch the headphone while the experiment is running.

Test time
In order to conduct a reliable testing it is important to keep the test time less than 20 minutes as the long testing process might cause the subject lose their attention. Subject may benefit from short break once their one ear is tested.
Appendix F EEG testing

Instruction to the subject

- Subjects must arrive at least ten (10) minutes before the scheduled commencement of testing with the purpose of minimizing and/or eliminating inaccuracies that may exist due to physical exertion.
- Subjects must not expose themselves to loud noises before the EEG test as it might affect their hearing ability.
- All the subjects need to be explained about the EEG test and answer their queries if they have any before the experiment.
- The subjects need to be assured that the EEG experiment is totally a non-invasive experiment and will cause no harm to the subject.
- During the preparation of the subject, the electrodes need to be placed with conducting gel for better conductivity. It is important to assure that the gel can be easily washed off after the experiments.
- The subjects need to caution if they experience any sort of pain or discomfort whilst the electrode skullcap and/or electrodes are positioned.
Subject sitting position

All the subjects are instructed to sit on a chair in an upright position. Although several research tests have been conducted where subjects lay on the floor and/or are encouraged to sleep [133, 254] depending on the type and the aim of the experiments, in this research the participants are not permitted to fall asleep for the following reasons:

- If the subject falls asleep then the response to modulated tones at frequencies less than 70 Hz may be considerably affected [134].
- The early phase of auditory processing will also be affected across the different stage of sleep [228].
- EOG artefacts due to rapid eye movement will also increase if the subject fall asleep and the cortical responses are profoundly dependent on mental alertness.

Electrode head cap and strap

Three different sized skullcaps are available and care should be taken to choose an appropriate size to fit the subject’s head. Inappropriate selection of the electrode skullcap will result in incorrect positioning of the electrodes. An appropriately sized skullcap will also assist in achieving lower impedance.
Electrode Placement

For the purpose of the research, three electrode positions were selected as listed below and were connected to the biosignal amplifier ‘g.USBamp’ in the following arrangement:

Table F.1 Electrode Positions

<table>
<thead>
<tr>
<th>AMPLIFIER SOCKET</th>
<th>ELECTRODE SITE</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (RED)</td>
<td>Cz</td>
</tr>
<tr>
<td>GROUND (YELLOW)</td>
<td>FPZ</td>
</tr>
<tr>
<td>REFERENCE (BLUE)</td>
<td>Left earlobe</td>
</tr>
</tbody>
</table>

The positions of Cz (positive input), mastoid (negative input) and FPZ (ground) are used by many researchers for EEG signal acquisition [172, 176, 177].

Measurement of the electrode positions

1. The distance from the nasion to the inion was measured using a measuring tape. Let the distance between the nasion and inion be: $D_{NI}$
2. Cz is 50% of $D_{NI}$.
3. Measure the ear-to-ear distance, $D_{EAR-EAR}$. The measurement should be taken exactly over the vertex position Cz.
4. Divide the distance $D_{EAR-EAR}$ by five to calculate the value $D_{20\%}$
   - $F_{Z}$ is $D_{20\%}$ toward the nasion from Cz and FPZ is $D_{20\%}$ toward the nasion from $F_{Z}$
Electrode Selection and Preparation

Two types of electrodes were used for the experiments:

1. Screwable electrodes:

![Screwable Ag/AgCl EEG Electrodes](image)

Screwable electrodes were used in the Cz position. The electrode was plugged at the desired position into the electrode ring. Once the electrode touched the skull surface, the electrode was fixed by turning it into 90 degrees.

2. Gold Cup electrodes:

![Gold Cup Electrodes](image)
Gold cup electrodes were used as they provide a good contact with the skin surface. Attaching gold cup electrodes on the skin surface is less time consuming and offers lower electrode impedance level. Gold cup electrodes were used for the position FPz and the earlobe. They are attached with scotch tape on the skin surface.

3. Fill the electrodes with conductive electrode gel for providing a good contact between the skull and the electrode. It is important that electrodes selected should not have any scratches or dirt on their surface, otherwise it will result in low quality EEG recordings.

4. Connect the electrode cable in a way that they don’t tangle with each other.

5. In order to obtain low impedance measurements the following method was used:
   - A cotton bud was used to gently move the subjects hair out of the way to achieve better contact.
   - Prior cleaning of scalp site with Theodor-Korner- Apotheke abrasive electrode gel.
   - Nihon-KohdenElefix conductive paste was applied on the electrode cups.
   - Electrodes were checked for any damage or surface imperfections.

6. Once the experiment is the electrodes then need to be soaked in lukewarm water for up to five minutes. The electrodes are then cleaned with the help of a cotton bud to remove all the gel and rinsed with clean water.
Impedance Checks

1. Once the electrodes are connected to the amplifier, the impedance of the mounted electrodes needs to be checked. It is important to have a low scalp-electrode impedance level to prevent poor quality EEG data. The amplifier measures the impedance of each electrode in kilo-ohm and different colour indicates different ranges of the impedance values. The table below shows the colour used for different ranges of impedance.

<table>
<thead>
<tr>
<th>Colour</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Green</td>
<td>Impedance &lt; 5 kΩ</td>
</tr>
<tr>
<td>Yellow</td>
<td>Impedance 5 to 7 kΩ</td>
</tr>
<tr>
<td>Red</td>
<td>Impedance &gt; 7 to 100 kΩ</td>
</tr>
<tr>
<td>Blue</td>
<td>Electrode not connected (NC)</td>
</tr>
</tbody>
</table>

In order to have a good EEG recording the impedance value should be lower than 5 kΩ. Impedance below 5 kΩ are optimal for EEG recording [21, 263].

2. If the impedance is more than 5 kΩ then the electrode-to-skin contact needs to be checked again and the preparation needs to be done again. It is important to note that the ground electrode is also important in this case, if the ground electrode has high impedance all the remaining electrodes will be high due to that.
Appendix G. Data acquisition

Data acquisition is done using a g.USBamp biosignal amplifier. Below is the step by step procedure for acquiring data using this amplifier.

1. Once all the electrodes are connected to the amplifier, the amplifier that is connected to the data capturing computer through USB port, needs to be switched on.

2. Start the data acquisition software from Matlab as shown in Figure G.1.:

![Figure G.1 DAQ Recorder Panel](image)

3. In order to activate the g.USBamp board we need to select ‘DAQ Boards’ from the Setting menu.
4. The software will automatically recognize the amplifier if the amplifier is connected and switched on. In this case g.USBamp is connected to the PC and the following window appears:

![Figure G.2 DAQ board]

The window shows that g.USBamp has 16 analog 24 bit input channels with the default sampling frequency set to 256 Hz. In addition the serial number of the hardware is also displayed.

5. Before starting the data acquisition couple of more configurations need to be completed as listed below:
The number of channels that needs to be recorded should be selected. Sampling rate should be changed to 4.8 or 19.2 KHz.

NO Highpass, Lowpass and Notch filter should be applied.

Common ground and Common reference should be selected.

The mode needs to be changed from 'Test Signal' to 'Measure'.

Press OK to confirm the settings.

6. Select 'Amplifier Settings' from the 'Header' menu and set the scaling unit to $\mu$V. The Scaling dialog box allows scaling each channel individually.
The scaling settings do not affect the calibration of the data. Click on Scaling from the menu bar. The following window appears:

![Scaling window](image)

**Figure G.4 DAQ Amplifier settings**

7. Select ‘Channel Configuration’ from the ‘Header’ menu and ensure that all channels are of type EEG.

8. Select ‘Impedance Check’ from the ‘Tools’ menu and press ‘Start’. Impedance levels of less than 5 kΩ are sought for each channel.
9. Once all the initial settings are completed the file name where the data is going to be stored needs to be specified by clicking on ‘Specify path and filename’ button in the lower section of the recorder panel.

10. Click on ‘record’ button in the lower section of the recorder panel to start the recording of the data.
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