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Utilization of the chirp stimulus in auditory brainstem response measurements

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The electrical activity originating from the auditory structures of the brainstem is called the auditory brainstem response (ABR). The ABR is used widely for objective assessment of the hearing system. Whereas traditional pure tone audiometry requires the test subject to respond to probe tones in a predefined manner, the ABR can be recorded without any active participation of the subject. Recently, a new kind of stimulus has been developed for ABR measurements. This stimulus, called chirp, aims at increasing the ABR amplitude by promoting synchronous firing of the hair cells in the inner ear. 1,5–2 times higher ABR amplitudes have indeed been recorded using the chirp when compared to traditional stimuli.

In this thesis, traditional click and tone burst stimuli were compared to corresponding chirp variants; the broadband chirp and frequency-specific narrowband chirps. Also, the ABR results were compared to those obtained by another objective method, the auditory steady-state response (ASSR). Responses were obtained from six normal-hearing subjects and five hard-of-hearing subjects using the Eclipse platform from Interacoustics. In normal-hearing subjects, the ABR amplitudes were compared between traditional and chirp stimuli. In all subjects, ABR and ASSR thresholds were compared to the behavioural thresholds.

Results failed to show significant differences between traditional stimuli and corresponding chirp variants. Also, comparison of and ASSR thresholds could not be done reliably because of the small number of obtained thresholds.

It was concluded that the results were influenced by the type of the insert earphones used in this study. Previous results in favour of the chirp have been obtained using a different type of earphones. Also, it was suggested that the influence of hearing loss on chirp-elicited ABRs be investigated in the future.

Keywords: chirp, ABR, auditory evoked potential, auditory threshold

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Välittömästi äänistimulusta seuraavaa, aivorungosta lähtöisin olevaa sähköistä aktiviteettia kutsutaan auditoriseksi aivorunkovasteeksi (ABR, *auditory brainstem response*). ABR:ää käytetään laajalti kuulon objektiiviseen testaamiseen. Siinä missä perinteisessä äänesaudiometriassa koehenkilön tulee vastata testiääniin ennalta määrättyllä tavalla, ABR voidaan mitata ilman koehenkilön aktiivista osallistumista. Hiljattain ABR-mittauksia varten on kehitetty uudenlainen stimulus: chirp. Chirpin tarkoituksena on tuottaa suurempi ABR-amplitudi lisäämällä sisäkorvan karvasolujen aktiviteetin yhtäaikaaisuutta. 1,5–2-kertaisia parannuksia onkin raportoitu verrattuna perinteisiin stimuluksiin.

Tässä työssä perinteisiä click- ja tone burst -stimuluksia on vertailtu vastaaviin laajakaistaiseen chirpiin ja taajuusspesifeihin kapeakaistaisiin chirpeihin. ABR-kynnystasoja vertailtiin myös toisella objektiivisella metodilla, ASSR:llä saatuihin kynnystasoihin. Vasteet mitattiin kuudelta normaalikuuloiselta ja viideltä huonokuuloiselta koehenkilöltä Interacousticsin Eclipse-järjestelmällä. Perinteisten ja chirp-stimulusten eroja tutkittiin vertailemalla normaalikuuloisilta mitattujen vasteiden amplitudeja. Myös kaikilta koehenkilöiltä mitattuja kynnysarvoja verrattiin äänesaudiometrialla määriteltyihin kuulokynnyksiin.

Tulosten perusteella perinteisten stimulusten ja vastaavien chirp stimulusten välillä ei löytynyt tilastollisesti merkitseviä eroja. ASSR- ja ABR-kynnysarvojen luotettava vertailu ei myöskään ollut mahdollista johtuen saatujen kynnysarvojen pienestä määrästä.

Johtopäätöksissä todettiin, että tuloksiin on todennäköisesti vaikuttanut tutkimuksessa käytettyjen kuulokkeiden tyyppi. Aiemmissa, chirpin paremmuuden kannalla olleissa tutkimuksissa, mittaukset tehtiin käyttäen toista kuulokemallia. Myös kuulonaleneman vaikutukset chirp stimuluksella suoritettuihin ABR-mittauksiin ovat huonosti tunnettuja, antaen aiheita jatkotutkimuksille.

Avainsanat: chirp, ABR, aivorunkovaste, kuulokynnys

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Units and abbreviations

Units

dB	Decibel
dB HL	Hearing level in decibels
dB SL	Sensation level in decibels
dB SPL	Sound pressure level in decibels
Hz	Frequency in Hertz
Ω	Electrical resistance in ohms

Abbreviations

ABR	Auditory Brainstem Response
ASSR	Auditory Steady-State Response
CANS	Central auditory nervous system
CN	Cochlear nucleus
DPOAE	Distortion product otoacoustic emission
EEG	Electroencephalography
HL	Hearing loss, also hearing level
HI	Hearing impairment
IC	Inferior colliculus
IHC	Inner hair cell
ILD	Interaural level difference
ITD	Interaural time difference
LDL	Loudness discomfort level
LL	Lateral lemniscus
LIF	Latency-intensity function
MGB	Medial geniculate body
NB	Narrowband
OHC	Outer hair cell
OAE	Otoacoustic emission
SD	Standard deviation
SOAE	Spontaneous otoacoustic emission
SOC	Superior olivary complex
TB	Tone burst
TOAE	Transient otoacoustic emission

1 Introduction

The sense of hearing is a complex system which transforms small pressure fluctuations of the air into neural signals and further into sensations and perception. Because of its complexity, the behaviour of the hearing system can often be observed only indirectly. Audiometric tests assess the function of the hearing system via the subjective perception of sounds by the person being tested. The most familiar audiometric test determines hearing thresholds across different frequencies by having the subject respond to pure tone sounds and finding the lowest sound level at which the subject responds. These tests are in a way the most truthful, since they give information about the actual sensing of sounds, i.e. hearing. But before sounds are perceived consciously somewhere in the auditory cortex and other related cortical areas of the brain, they are processed by the lower levels of the hearing system (Middlebrooks, 2009). Malfunction at the lower levels can result in alterations or a complete loss of auditory perception. Non-auditory factors can also affect the hearing; for example the processing of sounds differs depending on whether attention is paid to a sound or not (Melcher, 2009). Also, schizophrenic patients may have auditory hallucinations.

The behavioural testing of hearing also demands that the subject is able to show a certain behaviour in response to sounds (ISO 8253-1:2010). In the case of pure tone audiometry, this behaviour would be to push a button each time a tone is heard. But there are many cases where this kind of response is impossible to obtain. For example small children and mentally retarded persons may not be able to respond correctly. Studying the hearing of sleeping or anaesthetized subjects is naturally impossible if the only way to gain insight into the function of the auditory system is to have the subjects respond consciously.

In hope to find solutions to the challenges in assessing the function of the hearing system, objective methods have been developed, which do not rely on the subject reacting to sounds. Electric and magnetic fields generated by the nervous system can be observed without any input from the subject. An auditory brainstem response (ABR) is the electrical activity generated by the auditory parts of the brainstem in response to sound stimuli (Kokkonen, 2007). The ABR differs from other auditory evoked potentials in that the level of consciousness, anaesthesia or most sedative medications do not affect it (Stone *et al.*, 2009). Because the ABR originates from the brainstem, it gives information about the functioning of only the first stages of the auditory pathway. A clear ABR does not necessarily mean that a person can hear normally. It is, however, a good indicator of the ability of the hearing system to transform pressure fluctuations into neural signals. In other words, it gives valuable information about the condition of the inner ear and the auditory nerve. ABR is used widely in newborn hearing screening as well as in identifying lesions of the auditory nerve, monitoring of patients during surgery and scientific research (Musiek *et al.*, 1994).

In the recent years, a new way of improving the accuracy of ABR measurements has been suggested. Traditionally, the ABR is recorded in response to a short 80–100- μ s click, whereas the new idea is to use a so-called chirp stimulus in eliciting the

ABR. The chirp is a transient sound, where the higher frequencies are delayed with respect to the lower frequencies, resulting in a very short (~ 5 ms) upward-gliding tone (Dau *et al.*, 2000). The chirp aims at producing more synchronous activity of the hair cells in the inner ear, thereby increasing the response amplitude and making it easier to detect even at low sound intensities. Studies have shown a 1,5–2-fold gain in chirp-elicited ABRs compared to click ABRs (Elberling *et al.*, 2010).

The purpose of this thesis is to evaluate the usefulness of the chirp stimulus in ABR measurements. Although growing evidence in favour of the chirp can be found in the literature, experiments on commercially available systems need to be done in realistic clinical conditions in order to find the true value of the new approach. Another goal is to compare the ABR method with another objective approach, the auditory steady-state response (ASSR). The aim of the thesis is to answer the following questions:

- Can the quality of ABR measurements be improved by using the chirp stimulus?
 - Does the chirp stimulus elicit higher response amplitudes than the click?
 - Does the use of chirp stimulus result in more accurate estimates of the subject’s behavioural hearing thresholds?
- How do the click- and chirp-elicited ABR measurements compare with ASSR in terms of test duration and accuracy?

The structure of the thesis is as follows: in section 2, an overview of the function of the hearing system is given, starting from the outer parts of the ear and following the auditory pathway. Some relevant higher-level functions of hearing are also introduced. In section 3, objective electromagnetic measures of the hearing system are discussed. The fundamentals of ABR and ASSR as well as their uses in clinical diagnosis are presented. Also, the anatomy of the chirp stimulus is discussed. Section 4 presents the methodology used in order to find answers to the research questions. Further, the organization of the experiments is described. Results of the experiments are presented and discussed in section 5. Finally, section 6 summarizes the experimental results and concludes the thesis.

2 The sense of hearing

2.1 Anatomy and physiology of hearing

Here, the different parts of the ear are introduced. The anatomy is explained starting from the external and visible parts and continuing further inside the ear.

2.1.1 Outer ear

The outer ear consists of the earlobes (pinnae) and the ear canal, ending at the tympanic membrane (Fig. 1). The outer ear modifies the sound as it is guided to the tympanic membrane, giving relative rise to the frequencies in the region from 2 kHz to 7 kHz (Pickles, 1982, p. 13). This is especially due to the dimensions of the ear canal: averaging 22,5 mm in length and 7,5 mm in diameter according to Karjalainen (2008, p. 77) or 35 mm in length and 1 cm in diameter according to Gacek (2009). The increase in sound pressure is in the order of 15–20 dB in the frequency region from 2 kHz to 7 kHz and is caused mostly by the resonances in the ear canal and the concha of the pinna (the large depression of the earlobe, closest to the ear canal). Naturally, there is no actual amplification taking place in the ear canal since the outer ear is a passive structure which cannot introduce any additional sound energy — only the pressure of certain frequency bands is increased relative to other bands. (Pickles, 1982; Lonsbury-Martin *et al.*, 2009)

Another task of the outer ear is to provide cues for sound localization by modifying the frequency content of the sound. When the sound is coming from the side, the auditory system uses the interaural time difference (ITD) and interaural level difference (ILD) determining the direction of the sound source. However, ITD and ILD cannot distinguish between front and back or above and under. For example, a sound source directly in front of a person results in ITD and ILD values of 0 (since the distance to both ears is the same and there is no shading caused by the skull), but the values are actually the same for all directions in the median plane. The spectral modulation provided by the earlobes can be used as a cue for sound localization by the auditory system. These cues are useful only at higher frequencies, approximately in the 3 kHz to 6 kHz region. (Pickles, 1982, p. 13)

2.1.2 Middle ear

The middle ear is responsible for transmitting the sound energy from the tympanic membrane to the inner ear. This is achieved by a chain of three small bones (see Fig. 1), known as the ossicles: malleus, incus and stapes (Gacek, 2009). The malleus is attached to the tympanic membrane and the stapes connects to the oval window of the cochlea. The joint between the malleus and incus is relatively rigid, so they do not move relative to each other almost at all (Lonsbury-Martin *et al.*, 2009). However when very loud low-frequency sounds are presented to the ear, there can be some bending of the malleoincudal joint (Karjalainen, 2008, p. 78). Also, at high sound levels (100–110 dB SPL) the vibrational mode of the ossicular chain changes into one that no longer transmits energy effectively to the cochlea, thus providing

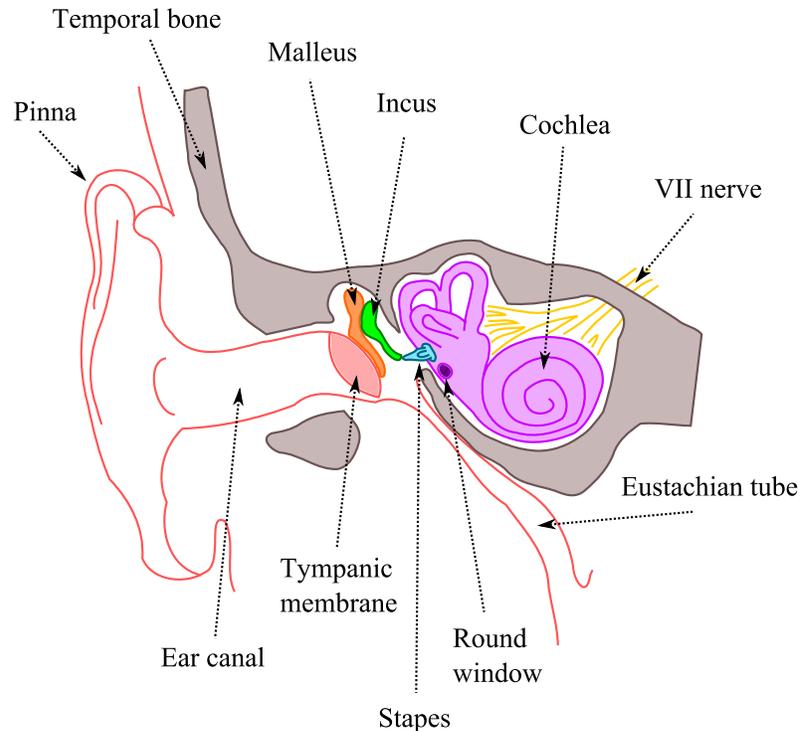


Figure 1: Anatomy of the ear. Figure redrawn from (Schnupp and Israel King, 2010, p. 52).

additional protection against loud noise (von Békésy, 1960, pp. 112–113). There are also two muscles binding to the ossicles: the tensor tympani and the stapedius muscle. The latter is responsible for the acoustic reflex—or the stapedius reflex—which occurs at high sound levels and is assumed to be a mechanism for protecting the inner ear (Gacek, 2009). Other functions have also been suggested for the middle ear muscles, such as reducing the perception of self-produced sound and selective attenuation of low frequency masking noise (Pickles, 1982).

There are at least two good reasons why the ossicles are particularly apt for transducing vibrations to the inner ear. One is the need for impedance matching between air and the cochlear fluids. If the sound waves were to enter the cochlea directly through the oval window, the impedance mismatch between air and the fluid would cause a significant part of the sound energy to reflect at the interface (Pickles, 1982). However, with the introduction of impedance matching, a bigger fraction of the energy of the sound wave is transmitted to the cochlea. The impedance matching is achieved by three ways (Pickles, 1982):

1. The area of the tympanic membrane is large relative to that of the oval window
2. The lever action of the ossicles
3. The buckling motion of the tympanic membrane

The effect of the relative areas of the two endpoints of the middle ear is quite clear: as pressure is defined as force per area, the same force applied on a smaller area

creates a larger pressure. The pressure gain is equal to the ratio of the two areas, resulting in a ratio of 20:1 in humans. The effective ratio is only 14:1, since the tympanic membrane does not vibrate as a whole. The lever action occurs, because the arm of the incus is shorter than that of the malleus. This results in an increase in force and a decrease in velocity at the stapes. The impedance ratio of the lever is approximately 1,3. The third mechanism of impedance matching is achieved by the conical shape of the eardrum, which allows a buckling action to take place, increasing the effectiveness of energy transfer by a factor of 4. Adding all the parts together, the impedance ratio of the middle ear is 73:1. (Pickles, 1982; Lonsbury-Martin *et al.*, 2009)

The other benefit the ossicles deliver is the fact that the pressure is applied only to the oval window, while the pressure at the round window at the other end of the cochlea remains constant. This pressure gradient in the cochlear fluids enables the flow that initiates the movement of the basilar membrane. If the pressure fluctuations were presented to both ends of the cochlea, the difference in pressure between the oval window and the round window would be zero and the cochlear fluids would remain at equilibrium, or at least the effect would be considerably weaker. (von Békésy, 1960; Pickles, 1982)

2.1.3 Inner ear

The inner ear is situated inside the temporal bone of the skull and it consists of two parts: the vestibular system and the cochlea. The vestibular system is responsible for sensations related to balance and the cochlea for hearing.

The human cochlea is an approximately 35 mm long, spiral-shaped structure of 2,5 to 2,75 turns that is responsible for transforming acoustic pressure fluctuations into neural activity (Gacek, 2009). The cochlea is subdivided into three longitudinal canals: scala vestibuli, scala tympani and scala media (Fig. 2). The scala media contains endolymph, which is a fluid with a high concentration of K^+ and a low concentration of Na^+ and Ca^{2+} ions. The lateral boundary (the side wall) of the scala media is formed by the stria vascularis. Scala tympani and scala vestibuli are filled with perilymph, a fluid similar to cerebrospinal fluid. Also, the scala vestibuli and scala tympani are connected via a small opening at the apex of the cochlea, called helicotrema. Through the helicotrema, the motion of fluid in one compartment is coupled to the other. The movement of the stapes initiates motion in the fluids of the scala vestibuli, which is further coupled to the scala tympani. At the end of the scala tympani is the round window — an opening covered by a membrane. The pressure of the fluid in the scala tympani pushes the membrane of the round window to the middle ear space in order to balance the pressure inside the cochlea. The three compartments are separated by different structures (Fig. 2). Between the scala vestibuli and scala media there is the Reissner's membrane and between the scala media and scala tympani there is the basilar membrane (BM), which is spanned between a bony shelf—called the osseous spiral lamina—and the spiral lamina on the lateral wall of the cochlea. The BM is a long and narrow sheet on top of which, in the scala media, lies the organ of Corti. The mechanical properties

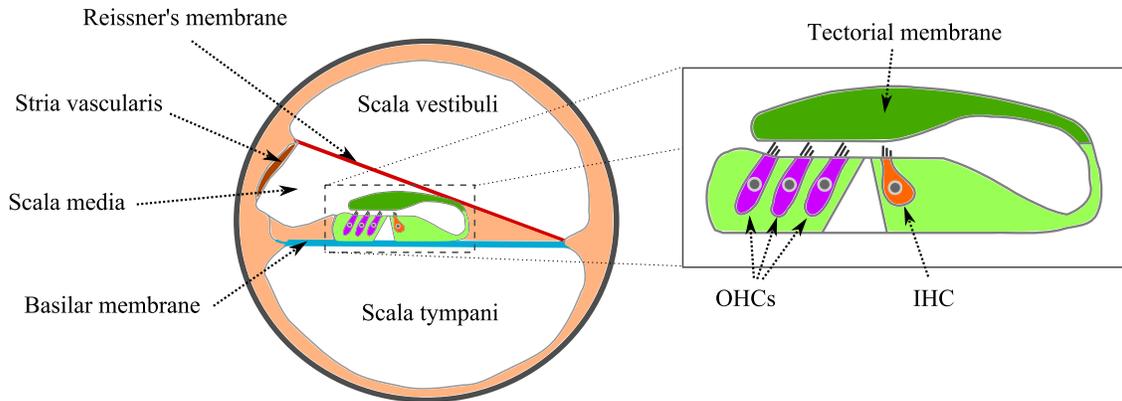


Figure 2: Cross-section of the cochlear duct. Figure redrawn from (Schnupp and Israel King, 2010, p. 65).

of the BM vary as a function of the distance from the oval window. Near the basal end of the cochlea, the membrane is narrow, and light, whereas near the helicotrema it is wider and floppier. The gradient of mass and stiffness along the BM enables the pressure fluctuations in the fluid to be transferred into a traveling wave on the BM (Fig. 3). The displacement pattern of the traveling wave depends on the frequency of the stimulating tone, leading to a certain place of maximum displacement for a given frequency. Vice versa, the frequency which causes the maximum displacement at some given place is called the characteristic frequency (CF) of that location on the BM. Frequency selectivity along the BM is further enhanced by the hair cells, which appear to be tuned to a certain frequency. This phenomenon can be seen from the tuning curves of individual hair cells. Tuning curves are obtained by choosing the sound amplitude at different frequencies in such a way that the neural activity stays constant (Karjalainen, 2008). Since the hair cell is most sensitive to a sound at its CF, the amplitude needed for a given level of activity is at its lowest. (see e.g. Pickles, 1982; Salvi *et al.*, 2000; Gacek, 2009)

The organ of Corti has two kinds of hair cells: outer hair cells (OHC) and inner hair cells (IHC). The hair cells extend to the surface of the organ of Corti towards the tectorial membrane. The apical ends of the hair cells are covered with bundles of stereocilia, which are tall and stiff poles made up from polymerized actin filaments, and are able to bend at their base. The stereocilia are arranged in rows such that the height of the stereocilia increases in each row towards the lateral wall of the cochlea. The stereocilia in a bundle are interconnected along their shafts by side links and also by fine fibers called tip links, which connect the mechano-electrical transduction (MET) ion channels of the shorter stereocilia to the side of its taller neighbor (Fettiplace and Hackney, 2006). In the IHCs, the rows of stereocilia are slightly bowed, whereas in the OHCs they are formed in a shape that resembles the letter 'W'. In the human cochlea, there are approximately 3000 IHCs, arranged in one row along the basilar membrane, and 12000 OHCs in three rows. (see e.g. Lim, 2001; Gillespie, 2001; Lonsbury-Martin *et al.*, 2009)

The mechanism, by which the hair cells transform mechanical movement into

neural signals is very elaborate and many parts of the process are still unknown. Although both inner and outer hair cells have sensory capabilities, the inner hair cells seem to be responsible for the actual detection of sound, whereas the outer hair cells serve as a mechanism for providing positive feedback in order to enhance the frequency-specificity and sensitivity of hearing. As the basilar membrane moves in response to the traveling wave, so does also the organ of Corti and this movement causes the bundles of stereocilia to bend. Deflecting the stereocilia bundle towards the tallest row causes the tip links to open the MET channels, whereas bending them to the opposite direction closes the channels. As the membrane potential of the hair cell is negative relative to the surrounding endolymph in the scala media, opening the MET channels causes positive ions to surge in, causing depolarization of the hair cell. Bending the stereocilia towards the shortest row closes more channels and the cell is hyperpolarized. Some of the MET channels are open even if there is no bending in either direction. This results in a sort of bias, which exists even when the hair cells are not excited.

For IHCs, the cell potential controls the release of neurotransmitters at the base of the cell. The neurotransmitters in turn bind to receptors on the nerve terminal, triggering an action potential in the auditory nerve. (Gillespie, 2001; Fettiplace and Hackney, 2006; Lonsbury-Martin *et al.*, 2009) Even when the IHC is at its resting potential (-50 to -70 mV), neurotransmitters are released, resulting in steady firing of the auditory nerve. Thus, depolarization and hyperpolarization of the IHC modulate the release of neurotransmitters (Geleoc and Holt, 2009). Each fiber in the auditory nerve connects to one specific IHC, but each IHC connects on average to 10 different fibers, resulting in a total of approximately 30000 nerve fibers (Gacek, 2009). The neurons of the auditory nerve are bipolar, stretching from the cochlea to the brainstem, and the cell bodies are grouped in the spiral ganglion.

OHCs on the other hand respond by conformational changes, shortening in response to depolarization and extending for hyperpolarization. This way, the OHCs can locally boost the vibration of the basilar membrane and improve the sensitivity and

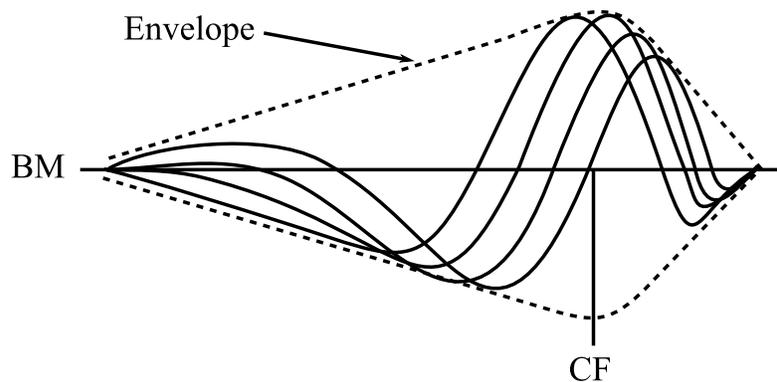


Figure 3: A schematic drawing illustrating the principle of the traveling wave. The place of maximum excitation along the basilar membrane (BM) is individual for each frequency. The characteristic frequency for the shown waveform is marked by CF.

selectivity of hearing. The boosting phenomenon is called the cochlear amplifier and it is not linear, since it is more pronounced at lower sound levels and weakens as the sound level increases. This seems to agree with measurement data, which show that the sensitivity and selectivity of hearing are worse for loud sounds. (Fettiplace and Hackney, 2006; Irvine, 2009) It is also possible to record the sound made by the resonating OHCs with a sensitive microphone and even in some cases heard without any amplification. This sound produced by the cochlear amplifier is called otoacoustic emission (OAE). There are many kinds of OAEs, but the three most frequently measured of them are spontaneous otoacoustic emissions (SOAE), transient otoacoustic emissions (TOAE) and distortion product otoacoustic emissions (DPOAE). In brief, SOAEs are generated in the cochlea without any sound stimulus, i.e. the cochlea spontaneously generates sound. TOAEs are recorded by stimulating the cochlea with a short sound and then recording the response from the ear canal. DPOAEs take advantage of the nonlinearities of the cochlea to record an emission from a frequency that corresponds to the distortion product of two pure tone signals.

2.1.4 Central auditory nervous system

After the ear has finished its task of converting the acoustical signals into neural signals, the information is carried to the many different parts of the central auditory nervous system (CANS). The ear itself doesn't interpret the sound it senses in any way and it is the responsibility of the CANS to process the information provided by the ears. Having a coarse idea about the structure of the CANS helps in understanding different psychoacoustic phenomena such as sound localization, but is even more important when discussing the auditory brainstem response later in chapter 3.2.

The CANS starts from the cochlear nucleus (CN) at the brainstem and ends at the auditory cortex of the cerebrum. An illustration describing the auditory pathways of the brain in Fig. 4. The neural signal from the cochlea is carried to the CN by the auditory nerve, which is part of the VIII cranial nerve, also called the vestibulocochlear nerve. The rest of the fibers in the VIII cranial nerve originate from the semicircular canals, which take care of the sense of balance. The vestibular system has its own pathways in the brainstem and it terminates to other locations than the auditory nerve. The CN is located in the brainstem near the place, where the medulla, pons and the cerebellum meet. It comprises three parts: the dorsal cochlear nucleus (DCN), the anterior ventral cochlear nucleus (AVCN) and the posterior ventral cochlear nucleus (PVCN). Each fiber of the auditory nerve splits into three and connects to all three cochlear nuclei. A tonotopic organization – meaning that the fibers carrying information from a specific frequency range are mapped close to each other – is preserved throughout the auditory pathway, also in the CN. The CN contains neurons of multiple classes, all of which respond in a distinctive way to different kinds of sounds. Some are more sensitive to temporal patterns and others to changes in frequency content. The purpose of this early processing could be the detection of the onset of a sound or deriving preliminary

information for localizing the sound source from spectral and temporal cues. The CN projects to higher parts of the pathway via three major fiber bundles: the trapezoid body, which contains fibers destined for the superior olivary complex or the inferior colliculus (IC); the intermediate acoustic stria, which contain fibers projecting to the contralateral IC and nuclei of the lateral lemniscus (LL); and the dorsal acoustic stria, containing fibers headed for the contralateral IC and nuclei of the LL. (see e.g. Musiek and Oxholm, 2000; Middlebrooks, 2009)

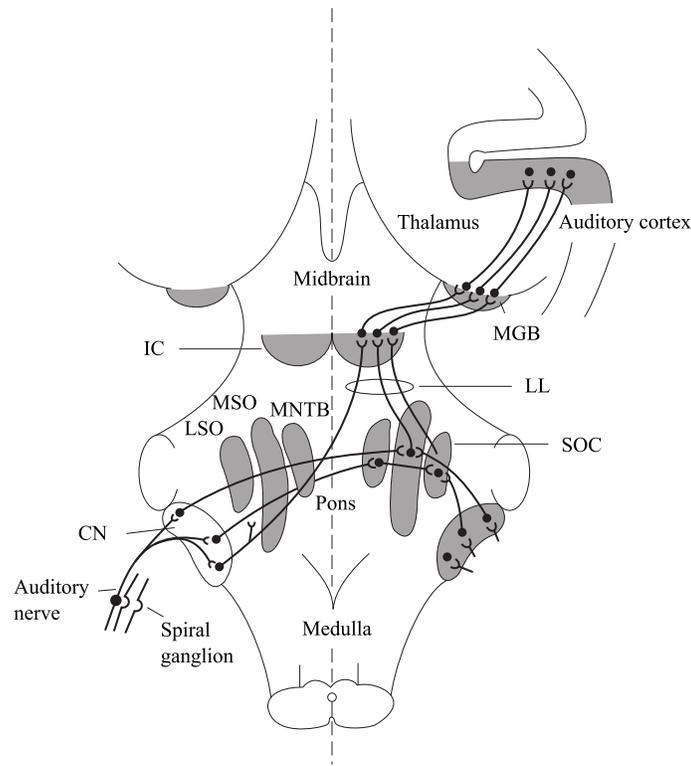


Figure 4: Auditory pathways. Figure adapted from (Middlebrooks, 2009).

The superior olivary complex (SOC) is positioned ventral and medial to the CN in the caudal portion of the pons. (Musiek and Oxholm, 2000) The SOC is the first point, where notable binaural processing takes place. The SOC contains many nuclei, the three most prominent being the medial nucleus of the trapezoid body (MNTB), the medial superior olive (MSO) and the lateral superior olive (LSO). The MSO and LSO seem to be specialized in horizontal localization of the sound source using interaural time difference (ITD) and interaural level difference (ILD) as measures. The role of the MNTB appears to be to provide inhibitory inputs to the MSO and LSO based on projections from the contralateral side. The SOC also contains multiple periolivary nuclei containing efferent neurons, which are able to control the stapedius muscle and decrease the electromotility of the OHCs. The purpose of these descending paths is to provide a way to control the dynamic range of hearing. By doing this, the ear can be protected against loud sounds, but also the masking by noise can possibly be reduced. (Brown, 2009)

The different ascending paths on both sides of the brainstem come together in a

large fiber bundle, called the lateral lemniscus (LL). There are also numerous nuclei in the LL, the most distinguishable being the ventral, intermediate and dorsal nuclei of the lateral lemniscus, abbreviated VNLL, INLL and DNLL, respectfully. VNLL and INLL are innervated mainly by the contralateral structures, whereas DNLL combines inputs from both ears and shows, similarly to the MSO, sensitivity to ITD. (Middlebrooks, 2009; Musiek and Oxholm, 2000)

One of the largest and most identifiable auditory structures of the brain stem is the IC, located on the dorsal surface of the midbrain. (Musiek and Oxholm, 2000) The main parts of the IC are the lateral nucleus, the dorsal cortex, the dorsomedial nucleus and the central nucleus (ICC), where almost all ascending auditory pathways synapse. The IC combines ascending and descending information from auditory and somatosensory and provides the superior colliculus (SC) with auditory inputs. The SC is responsible for the orienting movements of the eyes and head and it uses visual, somatosensory and auditory information in doing this. (Middlebrooks, 2009)

After now having passed through the brainstem and the midbrain, the auditory pathway reaches the forebrain, where the medial geniculate body (MGB) of the thalamus and the cortical areas in the temporal lobe are responsible for the processing of auditory information. The MGB has three divisions: ventral, dorsal and medial. The ventral part has cells with quite sharp tuning curves, whereas the dorsal division contains more broadly tuned neurons. The medial part gathers information from various sources, including somatosensory inputs. The auditory cortex has three hierarchical levels: core, belt and parabelt. The functions of the auditory cortex are very complex and the relations between the neural activity and the auditory perception is a subject of constant research. (Middlebrooks, 2009)

2.2 Psychoacoustics

2.2.1 Hearing threshold

Hearing threshold is the level of the most quiet sound that a person can hear. The hearing threshold for different sounds can vary based on, for example, their duration and spectral content. The ISO standard determining norms for audiometric test methods defines hearing threshold as follows:

“lowest sound pressure level or vibratory force level at which, under specified conditions, a person gives a predetermined percentage of correct detection responses on repeated trials” (ISO 8253-1:2010)

The hearing threshold is determined with a clinical procedure known as audiometry, where the hearing is tested with pure tones of specific frequencies ranging normally from 500 Hz to 8000 Hz. Although the exact thresholds are subjective and thus differ from person to person, it is possible to determine “standard” hearing thresholds by measuring the hearing of a large number of people and using the average of the results as the standard value.

The average hearing thresholds are used as reference values for two essential measures used in acoustics and audiology: the sound pressure level and the hearing

level. Sound pressure level (SPL) is given in decibels with respect to the reference value $20 \mu Pa$, which corresponds to the sound pressure of a 1 kHz tone at the average hearing threshold. Hearing level (HL) is the deviation from the average hearing threshold, measured in decibels. This means that the sound pressure level corresponding to the average hearing threshold on any frequency has the HL value of 0 dB. For example, 25 dB HL is 25 decibels above the normal-hearing threshold for that frequency. Also a third measure is frequently used in audiology: the sensation level. Sensation level (SL) gives the sound level with respect to a person's own hearing threshold. So if a person has a threshold of 20 dB HL, a sound at 15 dB SL corresponds to a sound level of 35 dB HL.

2.2.2 Masking

When the threshold for detecting a test sound is increased by the presence of another sound, the latter is said to be masking the former. The test sound and the masker can be presented to the ear simultaneously or one after the other. When studying simultaneous masking, the effects of the spectral content of the signals are of special interest. When the durations of the sounds are not overlapping, the phenomenon of increasing the detection threshold of the test sound is known as temporal masking.

In simultaneous masking, the test sound is usually a pure tone. The masking sound can be anything from broadband noise to pure tones. The effects of masking are usually visualized by means of masking audiograms, where the threshold of detecting the test sound is plotted against frequency for different levels of the masking noise. Using white noise results in masking audiograms, which stay constant up until approximately 500 Hz after which they start to incline with a slope of 3 dB/oct. If a constant masking on all frequencies is wanted, the spectrum of the masking noise can be filtered by a lowpass filter with a slope of -3 dB/oct above 500 Hz. (Karjalainen, 2008) Masking with narrowband noise results in a masking audiogram which has a slightly asymmetrical shape. The amount of asymmetry increases with the level of the masking noise, extending the masking towards higher frequencies. This phenomenon of raising the threshold of frequencies higher than the masking tone is called upward spread. One way to understand the reason for upward spread is to look at the envelope of the traveling wave on the basilar membrane (Fig. 3). The amplitude of the traveling wave increases as it moves towards the place of maximum excitation and then falls down very quickly. So the amplitude decreases more slowly towards the basal end of the basilar membrane, which corresponds to the observed upward spread, since higher frequencies lie towards the basal end of the basilar membrane from the place of maximum excitation. It has been proposed that masking may not be caused solely by the spread of excitation, but also by nonlinear neural suppression mechanisms. (Oxenham and Plack, 1998) By using pure tones for masking, the masking audiogram looks very similar to that obtained by narrowband noise, with the exception of additional frequency regions where the masking tone and the test tone interfere in a special way.

Temporal masking occurs before the onset (forward masking) and after the offset (backward masking) of the masking sound. Forward masking takes place approxi-

mately 5–10 ms before the onset of the masker and backward masking extends up to 150–200 ms after the masking sound. (Karjalainen, 2008, p.103)

2.3 Audiology

Audiology is a science, which examines the function and dysfunction of the hearing system. It is an interdisciplinary field, combining biomedicine, technology and behavioral sciences (Jauhiainen, 2008). According to the definition by the European Federation of Audiology Societies (EFAS), the audiological services provided to hearing impaired people deal with (EFAS):

1. Diagnostics
2. Auditory (re)habilitation
3. Communication / communication disorders
4. Prevention
5. Research and training

Compared to the medical branches of otology and neurotology—the former examining the normal and pathological anatomy and physiology of the ear and the latter studying the neurological function of the ear—audiology is perhaps more concentrated on the sense of hearing as a whole.

Biomedical engineering plays an essential role in audiology as technological solutions—such as hearing aids, cochlear implants and diagnostic equipment—are becoming more and more common. The branch of technical audiology focuses on applying an engineering approach to audiological problems. Technical audiologists can have varying backgrounds in e.g. electrical engineering, measuring technology, biophysics, computer sciences, mathematics, physics or acoustics. They also deal with different kinds of duties, like earpiece manufacturing, calibration of hearing aids and cochlear implants, test equipment maintenance and clinical testing.

Different characteristics of hearing require different diagnostic methods. At a very high level, the methods can be divided into two groups: subjective and objective test methods. Subjective methods try to determine how the patients experience the sound presented to them. This means that psychoacoustic phenomena are always present in the measurements. Problems that can arise when doing subjective tests, include non-cooperative patients and the difficulty to train the test procedure to the patients. Apart from very simple listening tests—such as the determination of hearing threshold which is discussed in the next section—the patient has to be a “good listener” in order to provide usable results. Objective methods aim at collecting information about the hearing system directly, without having to rely on the patient’s experience or opinion. Such methods are very useful when the patient is unable to respond, such as during sedation or anaesthesia. Other situations, when an objective test is preferred over a subjective one, are found for example in paediatrics where infant hearing screening is performed. Auditory brainstem response

and other electrophysiological measurements of the hearing system provide objective data about the neurological function of the hearing system. Other objective test methods include measuring OAEs and tympanometry (measuring the acoustic impedance of the ear drum).

2.3.1 Audiometry and the audiogram

Audiometry is the testing and measurement of the sense of hearing. Although the objective of a hearing test can be to determine issues such as perceived loudness, just noticeable differences in frequency or the level for the acoustic reflex, the most common audiometric procedure is pure-tone audiometry. In pure-tone audiometry, the hearing thresholds for different frequencies are determined using sinusoidal pure tones. The test signal can be presented either by air conduction—meaning headphones or loudspeakers—or by bone conduction, in which case the vibrations generated by a special bone vibrator placed over the mastoid bone are transmitted to the inner ear through the skull.

The complete procedure for conducting a pure-tone audiometry is described in the ISO 8253-1:2010 standard, where it is suggested that either an ascending method or a bracketing method be used in determining the hearing threshold. In the ascending method, the sound level is decreased after an initial response at a high enough level in steps of 10 dB until no response occurs. Then the level is ascended in 5 dB steps until there is again a response. The level, at which a response occurs three times out of a maximum of five ascents, is defined as the hearing threshold level. In the bracketing method, the level of the test tone is increased by 5 dB after each response, after which the level is decreased in 5 dB steps until no response occurs. Then the level is decreased 5 dB once more and a new ascent begins at this level. This is continued until three ascents and three descents have been completed. The testing begins at 1 kHz and continues upwards, followed by the lower frequency range, in descending order. Before testing the other ear, a repeat test is carried out at 1 kHz on the ear tested first. (ISO 8253-1:2010)

When hearing in one ear is significantly worse than in the other, there is a chance that the test signal presented to the worse ear is heard in the contralateral ear. This is called cross-hearing, and can be battled with masking noise, presented to the contralateral ear. ISO 8253-1:2010 suggests that the use of a contralateral masking signal should be considered if the threshold level measurements result in a hearing level of 40 dB or more.

The result of a pure-tone audiometry is called an audiogram and it gives a graphical presentation of the frequency-specific hearing thresholds. Thresholds obtained by air conduction in the unmasked case are marked with a circle and a cross for the right and left ear, respectfully. Adjacent points in the audiogram are connected with straight, solid lines. Also, if colors are used, red symbols and lines correspond to the right ear and blue markings to the left ear. An example of an audiogram is given in Fig. 5. Often, the type of hearing loss is described by the visual appearance of the audiogram. For example a situation, where the hearing thresholds in all frequencies are the same, is usually called 'flat', referring to the shape of the audiogram.

Similarly, a 'dip' means that there is a higher threshold on some frequency interval, resulting in an audiogram with a downward notch at those frequencies.

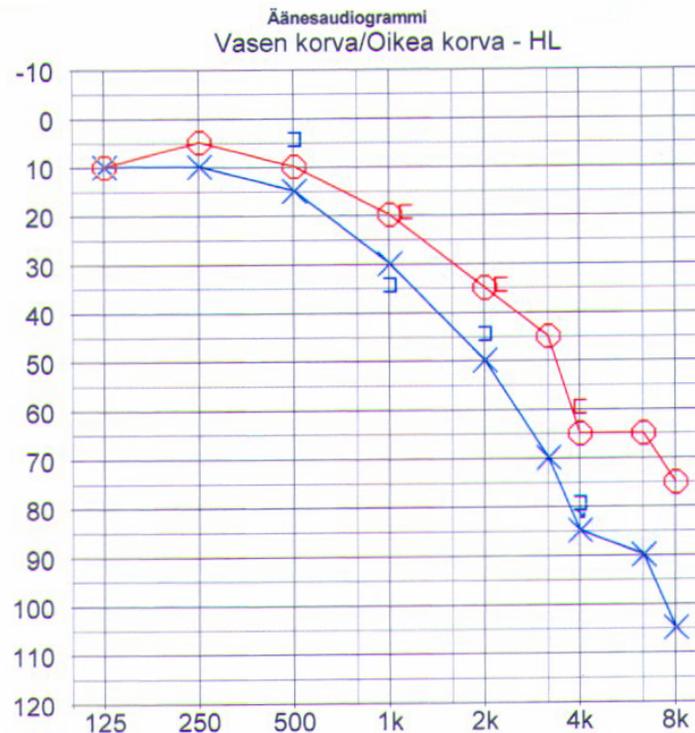


Figure 5: Example of an audiogram with a high-frequency sloping curve in both ears. Red circles and blue crosses are used for air conduction thresholds in the right and left ear, respectively. The square brackets correspond to bone conduction thresholds.

2.3.2 Hearing impairment

Hearing impairment (HI) refers to both complete and partial loss of the ability to hear, whereas deafness refers to the complete loss of hearing in one or both ears (WHO Fact sheet N°300). HI affects over 250 million people globally, making it the most frequent sensory deficit in humans (European Commission Heidi Wiki – Hearing loss). Uimonen *et al.* (1999) studied the northern Finnish population and found that 3,8% had mild HI; 1,3% moderate; 0,4% severe; and 0,1% had profound hearing impairment according to the WHO classification (see Appendix A for a table of grades of HI).

There are many ways for HI classification, including the type and cause of the impairment. Depending on the part of the ear which is affected, the HI falls into either of two types: conductive or sensorineural HI (WHO Fact sheet N°300). Conductive HI is caused by a problem in the outer or middle ear, such as in a chronic middle ear infection. Conductive HI can often be treated medically or surgically (WHO Fact sheet N°300). Another way to classify a HI is by its cause. Some causes

can be present already at birth and are called congenital causes. These include hereditary hearing loss, low birth weight, baby’s lack of oxygen during birth and the use of ototoxic drugs during pregnancy (WHO Fact sheet N°300). Two thirds of congenital HI cases are caused by hereditary genetics (Arlinger *et al.*, 2008). Those causes which can lead to HI at any age are called acquired causes. Especially age related hearing loss (presbycusis) causes the number of HI cases to increase as the population ages. The effect of age related hearing loss is described in the ISO 7029:2000 standard. In this standard, an equation is given for the calculation of the median value of the hearing threshold for otologically normal persons of a specific age. Examples of the resulting data are shown in Fig. 6. An otologically normal person is defined in the standard as follows:

“person in a normal state of health who is free from all signs or symptoms of ear disease and from obstructing wax in the ear canals, and who has no history of undue exposure to noise.” (ISO 7029:2000)

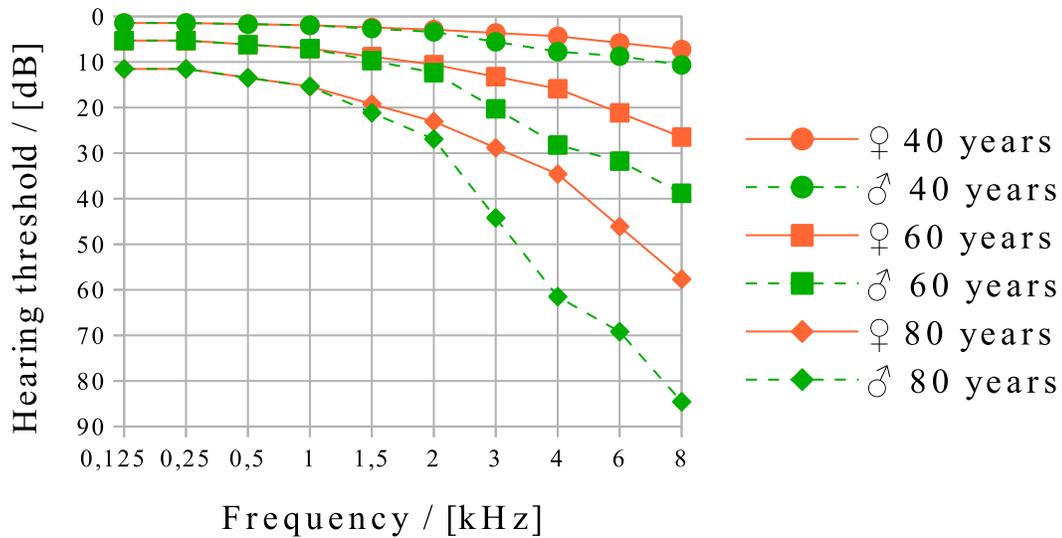


Figure 6: Median hearing thresholds as a function of age. Data from (ISO 7029:2000).

Exposure to noise is another acquired cause for hearing loss. It is considered that the risk of a noise-induced hearing impairment is increased when the continuous A-weighted equivalent sound level exceeds 85 dB. This kind of hearing loss develops gradually, whereas very loud impulsive sounds can lead to immediate HI. The peak sound pressure level has to reach 140 dB in order to present a considerable risk of inducing a permanent HI. Other reasons for hearing impairment include some diseases—such as otitis media, Ménière’s disease and otosclerosis—the use of ototoxic drugs and head or ear injury. (Arlinger *et al.*, 2008)

The way in which a hearing impairment reveals itself is not necessarily limited to the increase of hearing threshold. Sensorineural HI can also lead to the diminishing of the dynamic range of hearing. This means that although the most silent sound a person with a HI can hear is louder than for a person with normal hearing, the upper limit of the dynamic range of hearing can stay unchanged. The upper limit of the dynamic hearing range is considered to be the loudness discomfort level (LDL) and it is usually in the range of 100–120 dB. (Arlinger *et al.*, 2008) So if a person has a hearing threshold of 40 dB due to hearing loss, but the LDL remains at 110 dB, the dynamic range of hearing is diminished from 110 dB to 70 dB. Some people suffer from over-sensitivity to loud sounds, a condition called hyperacusis. In this case the LDL is reduced, decreasing the dynamic range from the other end. Hyperacusis can also occur in persons with normal hearing thresholds (Arlinger *et al.*, 2008).

3 Auditory evoked potentials

The electrical activity of the brain can be measured with scalp electrodes, resulting in a recording called an electroencephalogram (EEG) (Geddes and Baker, 1989, p. 717). The central nervous system generates spontaneous electrical activity which has characteristics reflecting the subject's level of consciousness. For a normal relaxed adult the dominant frequency—or the alpha rhythm—of the EEG ranges from 8 Hz to 13 Hz, whereas for example the delta waveform is associated with normal sleep and has a frequency of 1–3,5 Hz (Geddes and Baker, 1989, pp. 721–722). When a sense organ is stimulated, fluctuations related to the stimulus can be observed in the EEG waveform. This series of fluctuations is called an evoked potential (EP) (Ferraro and Durrant, 1994). When studying the sense of hearing, the stimuli are sounds played to the ear and the corresponding response is called an auditory evoked potential (AEP). As the auditory information travels in the neural system from the cochlea through the brainstem and up to the auditory cortex, different functional parts of the auditory pathway are active at different times, contributing to the distinct peaks of the AEP. An approximate representation of the AEP waveform is shown in Fig. 7. The AEP produced by a brief stimulus is generally divided into three parts (Melcher, 2009):

1. The first part of the AEP, up to approximately 40 ms after the sound stimulus, is called the auditory brainstem response (ABR). The ABR will be discussed in section 3.2.
2. Middle latency response (MLR) begins right after the ABR and continues for approximately 40 ms.
3. The late latency response (LLR) begins 50–100 ms after the stimulus

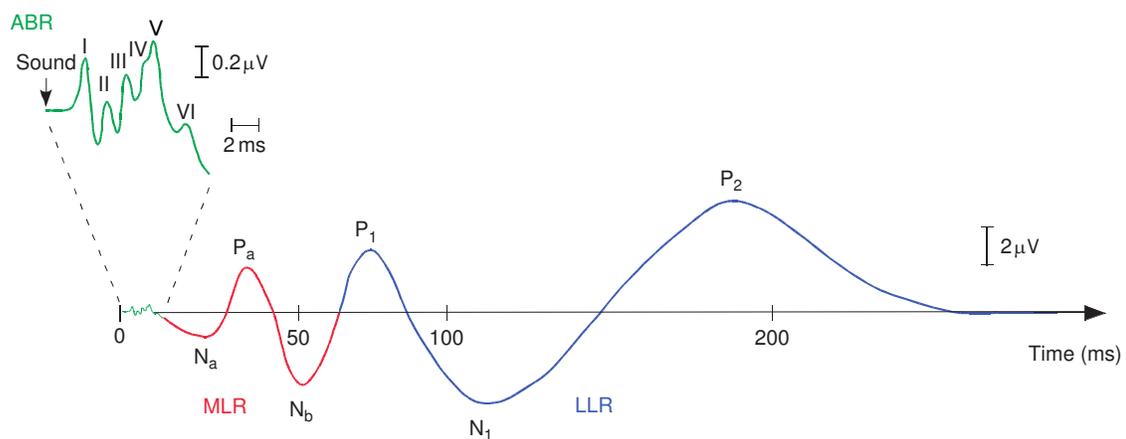


Figure 7: Stylized auditory evoked potential (AEP) in humans. The stimulus is presented at 0 ms. The AEP consists of the ABR (in green), MLR (in red) and LLR (in blue). Adapted from (Melcher, 2009).

All parts of the AEP share some characteristics, such as the amplitude of the response depending on the intensity and rate of presentation of the stimulus. Increasing the intensity of the stimulus causes the amplitude of the response to increase, whereas increasing the rate of stimulus presentation decreases it. A higher stimulus intensity also usually leads to a smaller latency, meaning that the response peaks are observed earlier than with a lower intensity stimulus. (Melcher, 2009)

The parts of the central nervous system contributing to the MLR include the thalamo-cortical pathway, the mesencephalic reticular formation and the IC. The detectability of the MLR also increases with age, suggesting that the structures responsible for generating the MLR develop with age. This may be associated with the development of the temporal lobe and the auditory thalamocortical pathway before puberty. (Kraus *et al.*, 1994) As for the LLR, it originates from cortical areas (Ferraro and Durrant, 1994; Melcher, 2009) and is therefore closely associated with the actual auditory perception of a sound (Elberling and Osterhammel, 1989). Both the MLR and LLR depend on the subject's level of consciousness, so that the response can change across different stages of normal sleep as well as under anesthesia. The LLR also has components that are sensitive to whether the subject is concentrating on the listening task or not (Melcher, 2009).

3.1 Potentials in the cochlea

Electrical activity in and around the cochlea can also be measured, although in this case the electrodes are placed in the ear rather than on the scalp. The goal is to capture the phenomena that occur in the cochlea during a time window of 1–10 ms after the stimulus. The term for the method of recording potentials generated in the cochlea is electrocochleography (ECoChG). Cochlear potentials are caused by the activity in the hair cells and in the cochlear nerve. (Elberling and Osterhammel, 1989)

In ECoChG, the electrodes are either trans-tympanic or extra-tympanic. A trans-tympanic electrode is typically a thin and flexible metallic needle, which is placed on the promontory of the medial part of the middle ear (Elberling and Osterhammel, 1989). This requires the electrode to carefully puncture the tympanic membrane, hence the term trans-tympanic. Extra-tympanic electrodes are placed in the ear canal, which makes it possible to use other electrodes than just thin needles. Although there are differences in the recordings obtained by trans- or extra-tympanic electrodes, both techniques have proven to be useful in practice (Elberling and Osterhammel, 1989).

The three potentials that can be obtained by ECoChG are the cochlear microphonic (CM), the summing potential (SP) and the action potential (AP). The CM has a waveform which closely resembles that of the stimulating signal. Some evidence suggests, that the CM may be responsible for stimulating the electromotile activity of the OHCs (Ferraro, 2000). The SP is a direct current response—as opposed to the CM and AP, which are both alternating—persisting for the duration of the evoking stimulus. The SP actually reflects the shape of the stimulus envelope, rather than its waveform. The SP is a complex response, and its origins remain

somewhat unknown, although it is thought that at least some components of the SP are generated by the various nonlinearities in the cochlea (Ferraro, 2000). The AP, also known as the auditory compound action potential (ACAP), is the action potential generated by the auditory nerve. The AP represents the total activity from thousands of individual nerve fibers. The characteristic deflections of the AP correspond to the first waves of the ABR. The first large negative wave of the AP, called N_1 is identical to wave I of the ABR. A subsequent, smaller wave N_2 corresponds to wave II of the ABR. The AP recorded by ECochG is actually used in enhancing wave I of the ABR in patients with significant hearing loss. (Ruth, 1994)

3.2 Auditory brainstem response

The auditory brainstem response (ABR) is the electrical signal which originates from the brainstem immediately after a sound stimulus. The ABR is normally regarded as the activity during the first 10 ms after the stimulus (Arnold, 2000), meaning that activity from the higher levels of the brain is not included in the ABR. The ABR has been given many names, such as the brain stem-evoked response (BSER), brain stem-auditory evoked potential (BAEP) and brain stem-auditory evoked response (BAER).

The ABR is a very popular in clinical audiology for various reasons. One of them is that the ABR waveform is quite similar between individuals, making it easier to be identified. Another advantage of the ABR is that the level of awareness doesn't affect the ABR, nor do most medications. For example small children can be tested reliably during natural or sedation-induced sleep (Arnold, 2000). Another popular use of the ABR is in neurosurgery, where the condition of the auditory system is monitored during an operation such as the removal of a tumor from the vestibulocochlear nerve (Melcher, 2009).

When viewed as an electrical signal, the ABR consists of five waves, which all correspond to specific structures in the brainstem. The waves are numbered with Roman numerals from I to VII, following the convention chosen by Jewett and Williston (1971) in their research on auditory far fields. The ABR waveform can be seen in (Fig. 8). It is believed that the waves I and II both originate from the auditory nerve; wave I generated by the distal portion and wave II by the proximal end of the auditory nerve. Wave III is assumed to originate from the cochlear nucleus (CN), wave IV from the superior olivary complex (SOC) and wave V from the lateral lemniscus (LL) as it terminates in the inferior colliculus (IC) (Arnold, 2000). Wave V is the most prominent wave of the ABR and is visible even at levels near the hearing threshold. This makes wave V a very popular target for audiological measurements and analysis.

3.2.1 ABR stimuli

The stimuli used the most in ABR measurements are the click stimulus and the tone-burst stimulus. The click stimulus is a 50–200 μ s-long rectangular pulse (Arnold, 2000). Until quite recently, it was believed that the ABR is generated primarily by

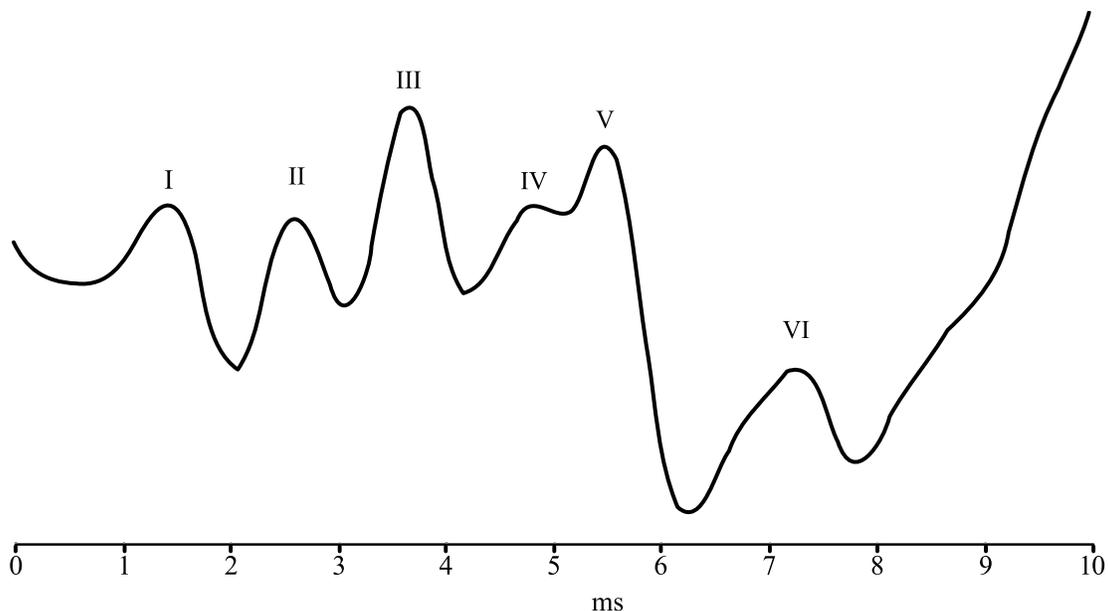


Figure 8: ABR waveform. Figure adapted from (Kokkonen, 2007).

the *onset* of a sound stimulus and so the click appeared to be the ideal stimulus for evoking the ABR (Ferraro and Durrant, 1994). However Dau *et al.* (2000) found that a chirp stimulus produced a bigger wave V and deduced that this was because of improved neural synchrony. The chirp stimulus is discussed later on in section 3.4. Although the click is considered to be a broadband stimulus (a nearly flat magnitude response up to 10 kHz), the results obtained by using the click correlate strongly with the average behavioral threshold only in the frequency range 1–4 kHz (Ferraro and Durrant, 1994; Bauch and Olsen, 1988).

The broadband click can be filtered in order to measure responses from lower frequencies or certain frequency bands. Another way to produce frequency-specific recordings is to use a tone-burst stimulus: a brief tone with rise/fall times of only a few milliseconds ($< 5ms$) and brief or no plateau duration (Arnold, 2000). This can also be seen as windowing a pure tone with a very narrow temporal window. Due to this windowing, the spectrum of the stimulus spreads considerably when compared to a long sine tone which contains only one frequency component. Different windowing functions can be used in order to reduce the amplitudes of the spectral sidelobes. Thus, although the tone-burst stimuli are not as frequency-specific as the pure tone stimuli used in traditional audiometry, they are more confined to a certain frequency band than a click.

The frequency-specificity of the ABR can be further improved by using notched noise for masking. In this approach wideband noise is filtered with a notch filter, resulting in a noise spectrum which covers all other frequencies, except the frequency band of interest. If the noise level is high enough to mask the unwanted frequency components of the stimulus, the activity from the masked frequency will be completely uncorrelated with the stimulus. When the samples are averaged, this activity will appear as random noise and have no effect on the wanted response.

The polarity of the stimulus refers to the initial direction of the transducer diaphragm (Ferraro and Durrant, 1994), which also determines the initial direction of the tympanic membrane movement. For a rarefaction stimulus, the initial direction of the transducer is outwards and for a condensation stimulus it is inwards. If the transducer produces an electromagnetic artifact that interferes with the measured EEG signal, it is possible to suppress this component by using alternating polarity (Ferraro and Durrant, 1994). Using alternating polarity also eliminates the cochlear microphonics from the recording (Elberling and Osterhammel, 1989).

In ABR recordings, choosing the stimulus repetition rate is a tradeoff between response quality and test duration. If the stimulus rate is too high, the response will be corrupted by the activity caused by the preceding stimulus. With higher stimulus rates, there is also neural adaptation taking place in the auditory system (Ferraro and Durrant, 1994). This affects the shape and amplitude of the response and should be avoided. On the other hand, lower stimulus rates result in a greater data collection time. For screening purposes, stimulus rates up to 30–40 Hz are acceptable, but for better wave morphology rates should not greatly exceed 10 Hz (Ferraro and Durrant, 1994; Elberling and Osterhammel, 1989). One more thing to consider while determining the stimulus rate is that if the frequency of the mains electricity (~ 50 Hz) is a multiple of the stimulus rate, it will be phase-locked with the stimulus and cause interference in the measurements. This is why the stimulus rate should not be an integer submultiple of the mains frequency (Ferraro and Durrant, 1994; Arnold, 2000).

3.2.2 Recording the ABR

In clinical EEG the electrodes are placed on the scalp according to the 10-20 system, which designates places for 21 electrodes. This system has later been extended to 74 electrode locations with the 10-10 system. The electrode locations of the 10-10 system are shown in (Fig. 9).

In ABR recordings there are usually four electrodes: one vertex electrode (Cz), or if that is not possible, then this electrode is placed on the upper forehead, just below the hairline (Fz); two electrodes are placed over the mastoid bones — one behind each ear (TP9 and TP10); (Geddes and Baker, 1989, p. 731) and a ground electrode which can be placed on the chin or on the cheek. The EEG signal is captured by using a technique called differential recording, where the difference between the signals from two electrodes, namely the Cz and TP9/10, is recorded (Ferraro and Durrant, 1994). After the signal has been picked up by the electrodes, it is first processed by the preamplifier where the low voltage ($< 1\mu V$) of the ABR is amplified about 50 dB and bandpass filtered with a passband from 30–100 Hz to 3000 Hz (Arnold, 2000).

The electrodes capture naturally much more than just the activity generated by the auditory system. The background noise originating from other parts of the brain, muscles and from the environment (50 Hz component from the mains electricity) totals to a much higher signal amplitude than the ABR alone — in technical terms, the signal-to-noise-ratio (SNR) is extremely low. Filtering the

signal can improve the situation slightly, but none of the preprocessing measures are enough to reveal the ABR. This is why the ABR is obtained by averaging measurements from numerous runs. The quality of the response can be assessed by comparing two averages, each consisting of half of the recorded waveforms. The similarity of the ABR waveforms is used as a measure of the response quality of the two averages and the measured signal as a whole. The response is usually measured 1000–2000 times and then averaged over all individual recordings. If the background noise is assumed to be independent of the ABR signal and to have a zero mean, the average of the background noise approaches zero as more samples are added to the average. On the other hand, since the ABR is time-locked to the stimulus—meaning that the ABR waveform is the same for each run—waves occur at the same time instances relative to the stimulus. So as opposed to the random background noise, the ABR component of the signal adds up constructively. The improvement in the SNR is proportional to the square root of the number of averaged samples, meaning that increasing the number of samples from 1000 to 2000 improves the SNR by only 3 dB. (Ferraro and Durrant, 1994)

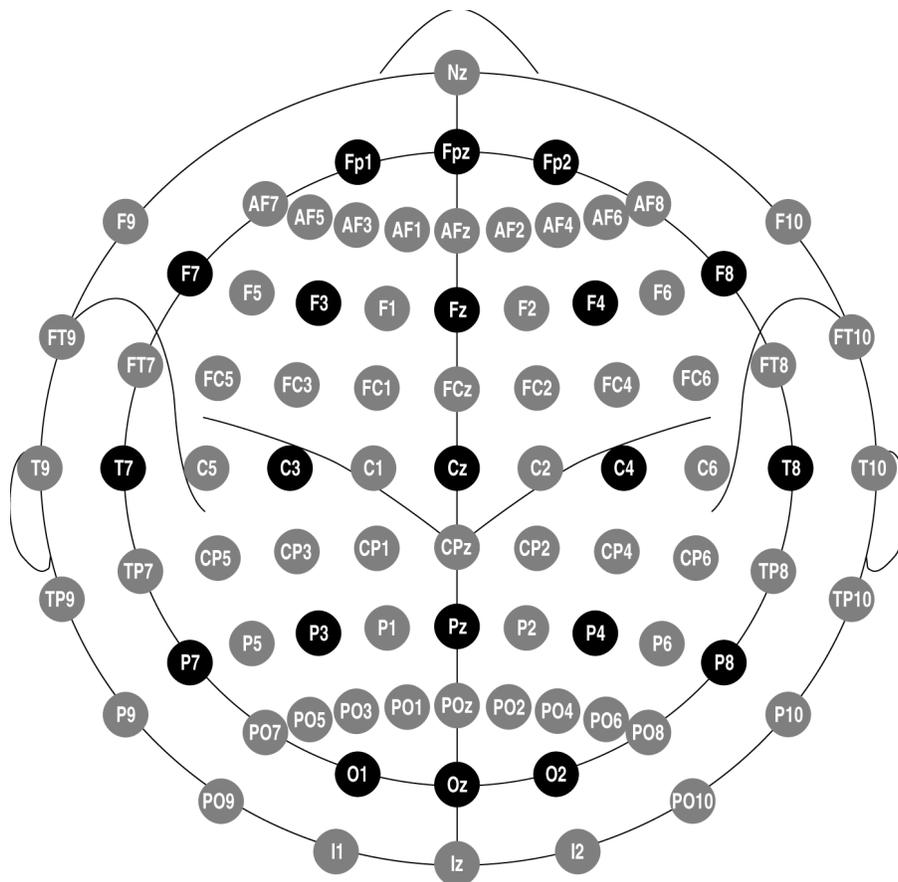


Figure 9: The 10-10 system for EEG electrode placement

3.2.3 ABR latency

When a stimulus is transmitted to the cochlea, there is a delay before the corresponding neural response can be observed. Most of the delay is caused by the time it takes for the traveling wave to reach its place of maximum excitation on the basilar membrane, but it has been proposed that also other mechanisms, such as ones taking place on the neural level, might also explain a fraction of the total latency (Neely *et al.*, 1988; Ruggero and Temchin, 2007). Neely *et al.* (1988) studied the wave V latency and suggested that the latency could be separated into mechanical and neural components. The mechanical latency corresponds to the travel time in the cochlea and the neural latency can further be separated into the synaptic delay of the IHC and the neural propagation delay from the IHC synapse to the brainstem structure responsible for generating the wave V.

The total amount of latency depends on various characteristics of the stimulus. In addition to the spectral content, the intensity of the stimulus has a clear impact on the wave V latency. By measuring the amount of wave V latency with respect to intensity it is possible to obtain a latency-intensity function (LIF). Normally the latency decreases as intensity increases, producing a LIF with a negative slope (Arnold, 2000). This phenomenon is illustrated in Fig. 10, where the ABR waveform is shown for decreasing stimulus intensities. It can also be seen that although the response amplitude decreases with the stimulus intensity, it is still visible at intensities near the hearing threshold. As will be discussed later in section 3.5, the LIF is a powerful diagnostic tool often used in audiology.

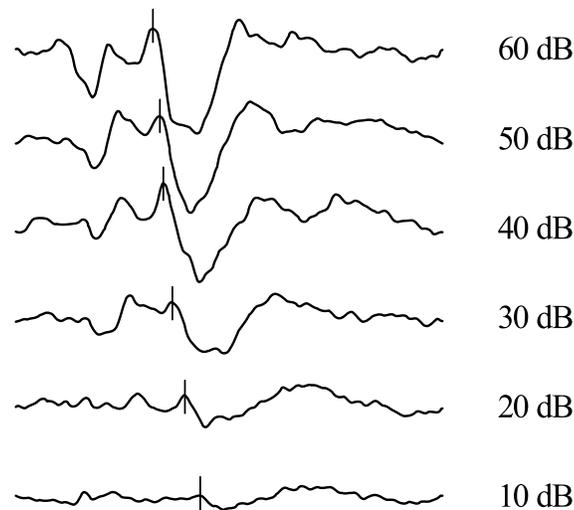


Figure 10: The dependency of ABR amplitude and wave V latency (wave V marked with a vertical dash) on stimulus intensity. Figure adapted from (Kokkonen, 2007).

The varying amount of latency experienced by different frequencies is a phenomenon often called temporal dispersion, i.e. different frequency components of a signal are separated in time. In signal processing terms, the cochlea can be seen as a system with a nonconstant group delay. Temporal dispersion in the cochlea causes the neural response of a transient broadband stimulus—such as the click

stimulus—to spread in time. This happens because the click signal consists of multiple frequency components, all of which enter the cochlea at the same time but have differing ABR latencies. A more synchronous activation of the cochlea results in a sharper and larger response and is therefore favourable. Expressing the latency with respect to frequency results in the latency-frequency function (LFF). One way to obtain an estimate of the LFF is to measure the ABR latency for frequency-specific stimuli. For example, Neely *et al.* (1988) measured the latency of otoacoustic emissions and brainstem responses using narrowband tone-burst stimuli.

Another way to get an estimate of the LFF is to use masking noise on certain frequency bands in order to suppress the response from the corresponding areas of the basilar membrane. An approach originally used by Teas *et al.* (1962) is to use high-pass filtered noise and lower the cut-off frequency in steps. According to the same principle of noise masking which was explained in section 3.2.2, only the unmasked portion of the basilar membrane affects the response. When these masked responses are subtracted one-by-one from an unmasked response, starting from the response with the highest cut-off frequency, the result is the ABR activity from a specific frequency band. This approach is called the derived-band ABR. The LFF can be estimated by looking at the latency of each derived-band ABR waveform. If the derived-band ABRs are again added together, the result is similar to the unmasked response. (Teas *et al.*, 1962; Elberling, 1974)

3.2.4 Compensating for the latency

Because of the temporal dispersion taking place in the cochlea, the peak amplitude of the ABR decreases and the response is smeared in time. In order to produce a clearer and more easily detectable response, some form of compensation for the latency can be applied.

The terms input compensation and output compensation refer to the two ways this balancing can be attained (Elberling *et al.*, 2007). Input compensation means that the latency characteristics are taken into account in the design of the input signal, i.e. the stimulus. If the problem is again viewed as a signal processing task, the objective is to apply the inverse of the group delay to the input signal before feeding it to the system. This means that when input compensation is used, there has to be an estimate of the LFF. It is possible to use the individual LFF but that becomes quite laborious, since it requires the LFF to be measured first, then fitting the input signal to the data and only after that measuring the actual response. This is why the input signal is usually designed against some general model which corresponds to the average behaviour of the inner ear.

Output compensation makes use of the ABR waveform in the derivation of the synchronous response. An example of output compensation is the method called Stacked ABR (Don, 2005), which is based on the same idea as the derived-band ABR by Teas *et al.* (1962) with the exception that the wave V peaks of the responses from different frequency bands are aligned before adding the waveforms together. One advantage of output compensation is that there is no need to estimate the LFF. On the other hand, measuring frequency-specific responses and adjusting them

individually takes more time (Don *et al.*, 2009).

3.3 Auditory steady-state response

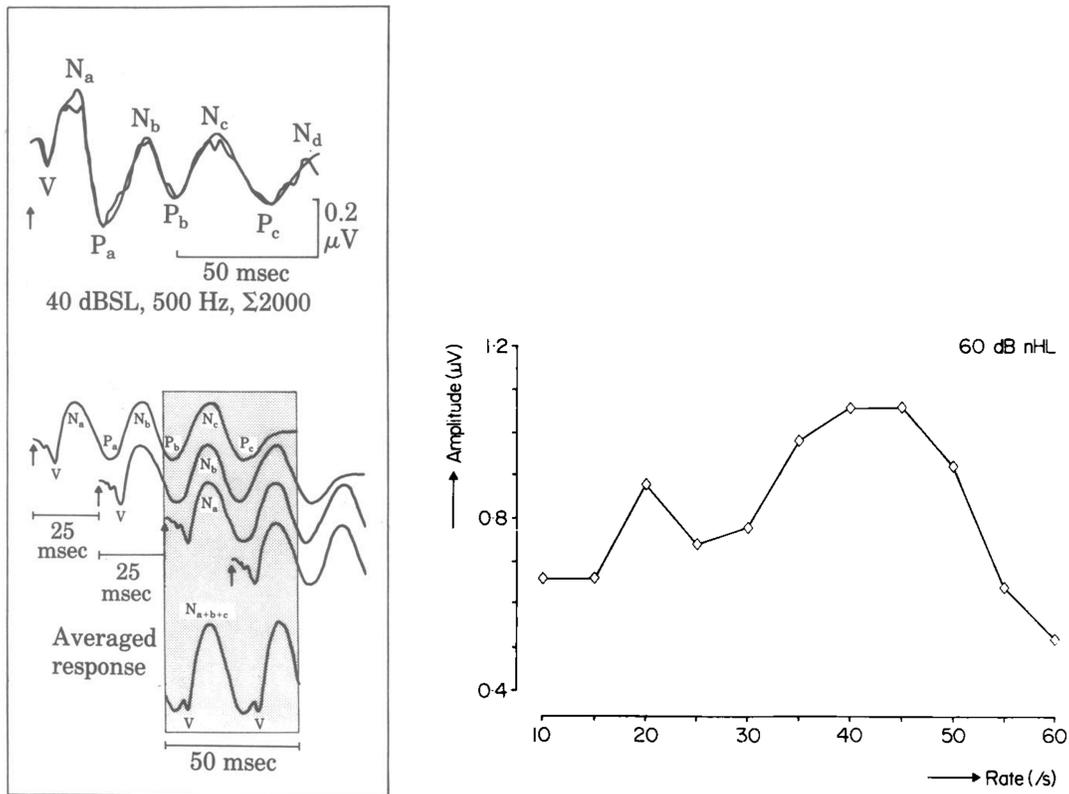
The auditory steady-state response (ASSR) is quite similar to the ABR in that it is derived from the EEG signal and measured with similar equipment. Also, both ASSR and ABR represent the neural activity of the auditory system in response to an auditory stimulus. However, further comparison to ABR reveals that the ASSR is actually very different from the ABR in many ways. A comparison of ABR and ASSR methods is given in Table 1, at the end of this section.

A steady-state response is created, when the stimulus signal is presented at such a high pace that the evoked response of the next stimulus cycle overlaps with the response from the previous cycle. In general, a steady-state signal can be described as a signal whose frequency components remain constant over a long period of time (Picton *et al.*, 2003). By choosing the interstimulus interval so that the positive and negative peaks of the AEP overlap and add up constructively (shown in Fig. 11a), the response becomes easier to detect. This effect is strongest for an interstimulus interval of ~ 25 ms, which corresponds to the time interval between successive MLR waves. This so-called “40-Hz response” was first discovered by Galambos *et al.* (1981), who used monaural clicks and tone bursts to evoke the response.

3.3.1 ASSR stimuli

The most used stimulus in ASSR measurements is a modulated pure tone. The modulation can be amplitude modulation (AM), frequency modulation (FM), or a combination of both, in which case the term mixed modulation (MM) is often used. The advantage of using a modulated pure tone is that this kind of stimulus is confined to a narrow frequency band centered around the frequency of the modulated tone—or the carrier frequency. This makes it possible to have good frequency-specificity in ASSR measurements. Other stimuli—such as clicks and tone bursts—have wider frequency spectra and are thus less often used in ASSR measurements.

The rate of stimulus presentation has a direct influence on the response amplitude. In the case of a modulated pure tone, the stimulus rate is equivalent to the modulating frequency. Stimulus rates up to a few hundred Hz have been studied and for adults the strongest response has been found to occur at a stimulus rate of 40 Hz (Galambos *et al.*, 1981). Despite resulting in the strongest response, 40 Hz is not always the optimal choice for the stimulus rate, like in the case of sleeping patients or young children. Sleep reduces the amplitude of the 40-Hz response significantly, whereas the responses recorded at stimulus rates over 70 Hz are not affected by the level of arousal and are therefore favoured when measuring sleeping patients (Cohen *et al.*, 1991). Higher stimulation rates of over 70 Hz have also been used successfully for measuring the ASSR in infants and young children. The maximum amplitude of the ASSR in these patients is achieved at approximately 20 Hz (Stapells *et al.*, 1988). It seems that the ASSR differences between young and adult patients has something to do with the development of the hearing system. Pethe *et al.* (2004)



(a) Illustration of the constructive interference of succeeding MLR responses. Adapted from (Galambos *et al.*, 1981). (b) ASSR amplitude as a function of stimulus presentation rate (Stapells *et al.*, 1984).

Figure 11: The principles underlying the dependency of the ASSR amplitude on stimulus rate.

studied stimulation rates of 40 Hz and 80 Hz in young children (up to 14 years) and concluded that the optimal stimulation rate changed from 80 Hz to 40 Hz by the age of 13.

The chirp stimulus has also been used in ASSR measurements and has been shown to result in a larger response amplitude when compared to the click (Elberling *et al.*, 2007). Although the chirp is a transient waveform, the response obtained by presenting them at a fast rate is in fact a steady-state response. A train of chirp stimuli can also be seen as a continuous periodic signal, meaning that it is possible to construct it from a sum of sine waves with different amplitudes and phases.

It is not clear, which parts of the auditory system contribute to the ASSR. The fact that the 40-Hz response is influenced by sleep suggests that the response is generated somewhere at the cortex, whereas the response to a higher stimulus rate could originate from lower levels, such as the brainstem. When viewing the ASSR amplitude as a function of the stimulus rate, the local maxima at 40 Hz and 90 Hz are probably a consequence of constructive interference between the AEP waves of successive stimulus cycles.

3.3.2 Detecting the ASSR

The ASSR recording is usually analyzed in frequency domain, rather than in time domain as is the case in ABR analysis. As was described previously, the frequency components of a steady-state response remain constant during the observation period. For example the 40 Hz response—which was elicited by successive click stimuli presented at a repetition rate of 40 Hz—contained a strong frequency component at 40 Hz. In the case of the 40 Hz response it was quite easy to see the correspondence between the MLR waveform peak interval and the enhanced response for the same interstimulus interval. However, in order to better understand how the steady-state response is created at the frequency corresponding to the stimulus repetition rate, the signal processing of the cochlea needs to be reviewed.

As was described in section 2.1.3, the bending of the IHCs causes them to depolarize and hyperpolarize, depending on the direction to which they are bent. Depolarization and hyperpolarization modulate the release of neurotransmitters, which in turn trigger an action potential in the auditory nerve. Although there is a steady release of neurotransmitters even when the IHC is at its resting potential (Geleoc and Holt, 2009), a simplified model can be used, which assumes that action potentials in the auditory nerve are triggered only by the depolarization of the IHC. Also, larger depolarization results in faster firing of the nerve, making it possible to view the cochlea as a filterbank in series with a compressing half-wave rectifier (Lins and Picton, 1995). This means that the neural response to a stimulus can be modelled as a compressed and rectified version of itself. Just as in a crystal radio there is a rectifying diode for envelope detection of an AM signal, the rectification of the cochlea makes it possible to detect the envelope of the stimulus from the steady-state response. In ASSR, where the stimulation rate is kept constant, the envelope is a periodic signal with a fundamental frequency corresponding to the stimulation rate. The most straightforward case is an AM signal where the modulating signal and thus also the envelope is a pure sine, meaning that the envelope contains only one frequency component; for other stimuli the envelope spectrum can be more complicated. The response spectrum is also distorted because of the nonlinearities, such as compression, in the cochlea. The principle behind ASSR detection is shown in Fig. 12.

Because the ASSR is analyzed in frequency domain, it becomes possible to measure responses to multiple carrier frequencies simultaneously. The principle is to use individual modulating frequencies for each carrier. As was described above, the response spectrum contains a frequency component at the modulating frequency; the same applies for multiple stimuli, so the response contains multiple components corresponding to the individual modulation frequencies of the stimuli. This way, the modulating frequency acts as a label for a given stimulus and its response. The use of multiple stimuli is illustrated in Fig. 13. The possibility to measure multiple responses combined with the narrow spectra of the stimuli makes ASSR a very attractive method for audiogram estimation (Lins *et al.*, 1996).

The decision of whether a response is present or not can be done objectively using statistical methods. One way to test the presence of a response at the frequency

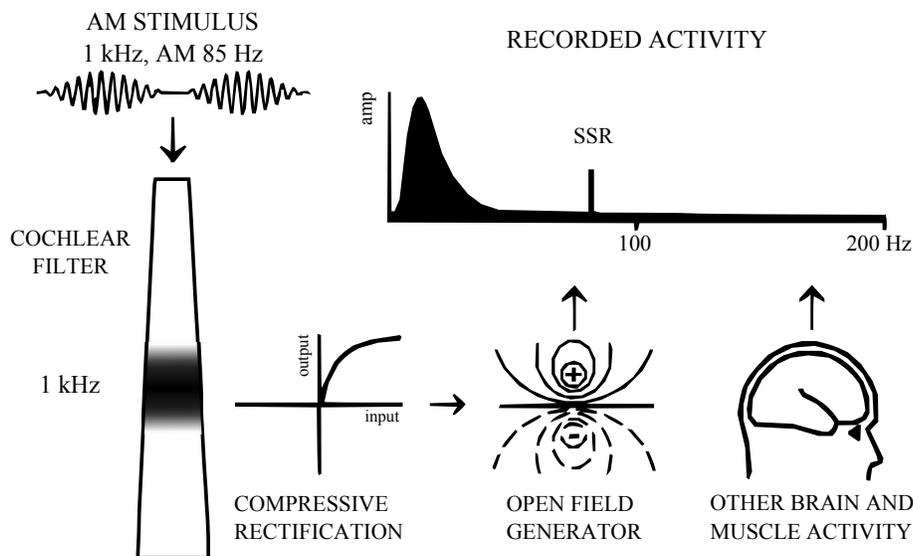


Figure 12: Steady-state response (SSR) detection in frequency domain. Figure adapted from (Lins *et al.*, 1996).

corresponding to the stimulus rate is to make multiple measurements and then combine the results to obtain a test statistic. Hotelling's T^2 test, the circular T^2 test and the magnitude squared coherence can all be used to inspect the two-dimensional distribution of the real and imaginary parts of the frequency components from the different measurements. If the mean of the distribution is at the origin of the complex plane, then there is no response present at that frequency. Using the test statistic, it is possible to test whether the hypothesis of a zero-meaned distribution can be rejected or not. It is also possible to concentrate only on the phase of the frequency component by using e.g. phase coherence or the Rayleigh test. (Picton *et al.*, 2003)

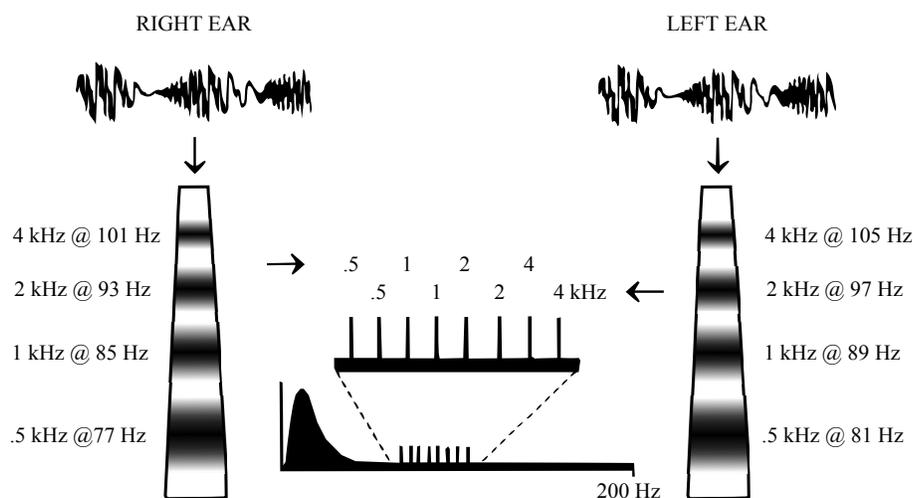


Figure 13: Multiple ASSR responses with AM. Figure modified from (Lins *et al.*, 1996).

Another way to recognize the presence of a response is to perform the measurement only once and make the decision by comparing the frequency of interest to other frequencies. A popular approach is the so-called F-test, where a ratio of the signal power at the stimulation frequency to the average power in N neighbouring frequency bins is calculated. If a response is present, the test should reveal that the signal power at the stimulus frequency is higher than the power of the background noise. (Picton *et al.*, 2003)

Table 1: Comparison of ABR and ASSR

	ABR	ASSR
Signal analysis	Time domain	Frequency domain
Response detection	Manual	Automated
Stimulation rate	$\sim 10\text{--}30$ Hz	~ 40 Hz or > 80 Hz
Typical stimuli	Click, tone bursts	AM pure tones
Frequency specificity	Fair	Good
Both ears simultaneously?	No	Yes
Multiple frequencies simultaneously?	No	Yes
Affected by sleep or sedation?	No	Depends on stimulation rate
Response quality measure	Noise level, wave reproducibility	Test statistic value

3.4 The chirp stimulus

The chirp stimulus can be described as a short sound, where the frequency of a sine wave glides quickly upwards. Another way to characterize the chirp would be to say that it is a transient sound, where the higher frequencies are delayed with respect to the lower frequencies. The purpose of the chirp stimulus is to achieve synchronous neural activity in the cochlea by compensating for the cochlear delay. The estimated frequency dispersion caused by the cochlear delay is taken into account when designing the stimulus signal, meaning that this approach provides input compensation. It has been shown that by using the chirp in ABR recordings, the response amplitude can be increased by a factor of 1,5–2,0 when compared to the click stimulus (Elberling and Don, 2008). The chirp is designed according to a delay model of the cochlea, which can be expressed as a power function (Elberling and Don, 2010):

$$\tau = k \cdot f^{-d}, \quad (1)$$

where τ is the delay in seconds, f is the frequency in Hz and k and d are constants, whose values can be determined by for example otoacoustic emissions or tone-burst ABR (Fobel and Dau, 2004); and derived-band ABR measurements (Elberling and

Don, 2008). The amplitude spectrum of the chirp is usually designed to match that of the click for better comparison of the effects of different delay models on the response (Fobel and Dau, 2004). The chirp waveform is shown in Fig. 14 with the click waveform. The amplitude spectrum—ideally being the same for both stimuli—is also shown.

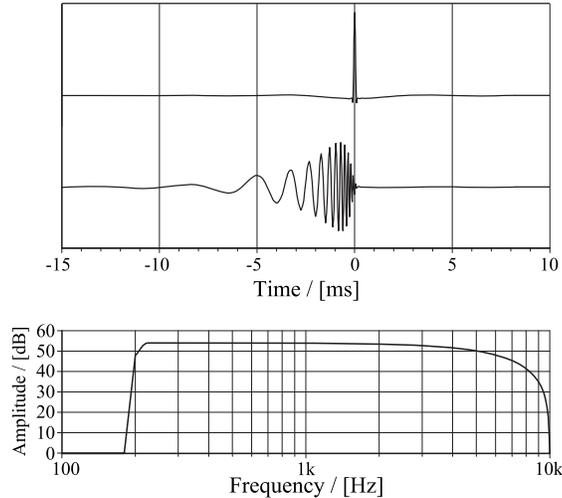


Figure 14: Click and chirp waveforms. Figure adapted from (Elberling and Don, 2008).

It is also possible to measure frequency-specific responses to chirp stimuli. This can be done either by masking with notched noise or by designing a chirp with frequency components spanning only over a restricted band (Wegner and Dau, 2002).

The chirp implementation used in the Interacoustics Eclipse platform is called a CE-chirp, named after Claus Elberling. This chirp differs from many previous implementations in that the amplitude spectrum is designed to be flat within five octave bands from 350 Hz to 11,300 Hz. There are also four narrowband CE chirps having center frequencies at 500 Hz, 1000 Hz, 2000 Hz and 4000 Hz. The waveforms of the CE-chirps are shown in Fig. 15. The broadband chirp can be constructed by summing the narrowband stimuli. The individual levels by which the narrowband chirps contribute to the broadband chirp are also shown in the figure. (Elberling and Don, 2010)

3.4.1 Previous research on the chirp

The advantage of using a chirp rather than a click in ABR measurements was shown in a study by Dau *et al.* (2000). They constructed their chirp based on a linear basilar-membrane model. Ten normal-hearing subjects were chosen for the experiments, which consisted of comparing the responses of the following three stimuli: a $80\mu\text{s}$ click, a broadband chirp and a reversed chirp. The results showed a significantly larger wave V amplitude when using the chirp rather than the click for the levels 20–40 dB SL. The reversed chirp had a significantly lower wave V amplitude

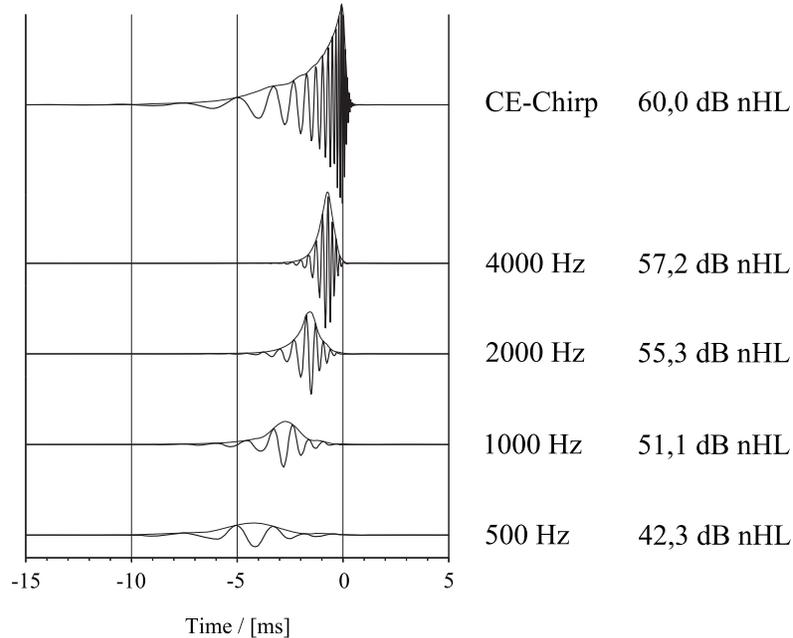


Figure 15: The waveform of the broadband CE-chirp and four narrowband chirp stimuli (Elberling and Don, 2010).

than the chirp for levels 10–40 dB SL and also lower than for the click for levels 20–40 dB SL. The reason for not getting a significant difference between the click and the chirp at the 10 dB level was because for four subjects there was no clear response at this level, and the number of remaining subjects was not enough to obtain a significant difference in the wave V amplitude.

In another study by Fobel and Dau (2004), three different chirp stimuli were constructed. One of the chirps was the one used by Dau *et al.* (2000) (called the M-chirp in the paper of Fobel and Dau). Another chirp, called the A-chirp, was obtained by fitting a model to tone-burst-evoked ABR wave V data. The third chirp, named the O-chirp, was based on measurement data from stimulus-frequency otoacoustic emissions (SFOAE). The resulting ABRs were compared with each other and with the response from the traditional click stimulus. The result was that the A-chirp performed better than the O- and M-chirps, especially at low stimulus levels, whereas there were no great differences between the O- and M-chirps at low and medium stimulus levels. Nevertheless, all chirps performed better than the click stimulus when comparing the wave V amplitudes.

The results of both studies suggested that the characteristics of the optimal stimulus vary with the level of the stimulus. In the study by Dau *et al.* (2000), the chirp did produce a larger wave V amplitude than the click also for the levels 50–60 dB SL, but the difference was not significant any more. The authors suspected that the decrease in performance at high levels was caused by cochlear upward spread. The A-chirp by Fobel and Dau (2004) was fitted to measurement data from multiple frequencies and levels. As the level was increased, the temporal structure of the was A-chirp changed. More specifically, the louder the stimulus, the shorter

the chirp. The result that the duration of the best-performing chirp decreases as the level of stimulation increases was confirmed by Elberling *et al.* (2010) when they experimented on five chirps of different durations at three levels of stimulation. Their results showed that shorter chirps produced the largest wave V amplitudes for higher stimulus levels.

The claim that the chirp would provide increased neural synchrony was questioned by Petoe *et al.* (2010), who suggested that the increase in wave V amplitude might be caused by recruitment of a greater number of neurons instead of better synchrony. Observations supporting their theory include the absence or diminishing of the earlier waves I–IV and the larger inter-subject variation in wave V latency when compared to the click stimulus. This doesn't necessarily decrease the usefulness of the chirp when determining the hearing threshold, since the most important factor there is the wave V amplitude. However, Petoe *et al.* (2010) question the suitability of the chirp to more complicated diagnostic procedures which would require less variance in the wave V latencies.

3.5 ABR and ASSR as diagnostic tools

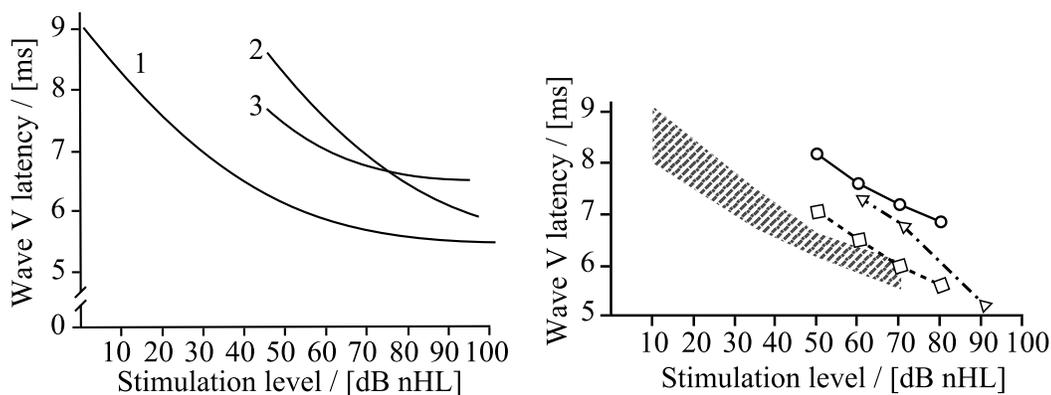
When assessing hearing loss, pure tone audiometry (see section 2.3.1) is always preferred to other methods, but in some cases this approach cannot be used. Such difficult-to-test populations include the testing of infants, non-cooperative subjects and subjects with impairments that prevent pure tone audiometry. As opposed to behavioural testing, where the subject has to respond to sound stimuli, objective testing with ABR and ASSR provides a way to obtain information about the subject's hearing without the subject having to contribute in any way.

Behavioural thresholds are estimated by finding the ABR threshold or the ASSR threshold, i.e. the smallest stimulus intensity eliciting a detectable response. When the ABR or ASSR threshold is subtracted from the behavioural threshold, a measure called difference score is obtained. ABR threshold for a click stimulus correlates best with the average behavioural threshold in the frequency range from 1 kHz to 4 kHz (Ferraro and Durrant, 1994; Bauch and Olsen, 1988) and is usually 10 dB to 20 dB higher than the best behavioural threshold within that range in subjects with normal hearing or cochlear hearing loss (Arnold, 2000). The difference score can be as small as 5 dB in subjects with cochlear hearing loss (Arnold, 2000). Similarly, ASSR thresholds are usually higher than the behavioural thresholds with reported mean difference scores of -3.72 dB to 17 dB in normal hearing subjects and within ~ 8 –13 dB in subjects with sensorineural hearing loss (Korczak *et al.*, 2012).

In addition to hearing threshold estimation, the ABR is further used in distinguishing between different types of hearing loss (see section 2.3.2) by examining the latency-intensity function (LIF). In conductive hearing loss, the sound is attenuated in the middle ear before reaching the cochlea. The LIF resulting from a cochlear hearing loss typically has a normal slope, but because of the attenuation, it is shifted to higher intensities (curve 2 in Fig. 16a) (Arnold, 2000). A smaller horizontal shift can be seen in cochlear hearing loss (curve 3 in Fig. 16a), where there is also a vertical shift when compared to the normal LIF (van der Drift *et al.*, 1988). In

the case of sloping high-frequency cochlear hearing loss the LIF has a steeper slope, having increased latencies near the hearing threshold but nearly normal latencies at high intensities. This can be explained by the fact that the place of initial response affects the wave V latency greatly. So when the broadband stimulus intensity is high enough to stimulate the damaged high-frequency fibers near the oval window, the overall latency of wave V diminishes rapidly, as illustrated in Fig. 16b. (Arnold, 2000)

The interpeak latency between waves I and V can also be used in detecting tumors and other lesions in the auditory pathway. A prolongation in neural conduction time between the structures generating waves I and V can be caused by a lesion. (Musiek *et al.*, 1994; Arnold, 2000)



(a) Idealized LIF curves for normal hearing (curve 1), conductive HL (curve 2) and cochlear HL (curve 3) from (van der Drift *et al.*, 1988).

(b) Examples of LIF curves from (Arnold, 2000). The shaded area shows the normal LIF; circles: conductive HL; triangles: sloping high-frequency sensorineural HL; diamonds: flat sensorineural HL.

Figure 16: The effects of different types of hearing loss on the LIF curves.

4 Methodology and experiments

4.1 Subjects

Ten volunteering subjects took part in the experiments: six (3 male, 3 female) having normal hearing (hearing threshold less than 15 dB HL at 125–8000 Hz) and five (2 male, 3 female) having differing types of hearing loss. If possible, those subjects who had had a pure tone audiometry done by a professional audiologist no more than six months ago, were preferred. Mean ages of the two groups were 24,8 years in the normal-hearing group (SD 2,8) and 75,8 in the hard-of-hearing group (SD 6,69). The subjects are referred to by a letter–number-combination with N1–N6 referring to the normal hearing subjects and H1–H5 referring to the hard-of-hearing subjects.

4.2 Stimulus parameters

ABR measurements were carried out using six different stimuli: broadband chirp (referred to as *chirp* from now on), 80- μ s click (*click*), 500 Hz tone burst (*500 Hz TB*), 4 kHz tone burst (*4 kHz TB*), 500 Hz narrowband chirp (*500 Hz NB chirp*) and 4 kHz narrowband chirp (*4 kHz NB chirp*). Responses to each stimulus were obtained at four intensities and from both ears, resulting in a total of 48 response waveforms per subject. For the normal-hearing subjects the intensities were 60 dB nHL, 40 dB nHL, 20 dB nHL and 10 dB nHL. For the hard-of-hearing subjects the intensities were selected individually based on their audiogram, starting above hearing threshold and reducing the intensity towards the threshold.

Common practice in clinical trials is to use a stimulus repetition rate of approximately 11 per second. The recommended value on the Eclipse platform for precise measurements is 11,1 stimuli per second (Eclipse Operation Manual) and this was used in the experiments as well. In general, the measurement should be repeated until the averaged waveform is of adequate quality. As a tradeoff had to be made between the quality of the responses and the overall duration of the experiments, it was decided to average 1000 repetitions for each measurement.

Stimulus polarity was set to alternating. Input signals were filtered at the preamplifier from 33 Hz to 1500 Hz for narrowband stimuli and from 100 Hz to 3000 Hz for broadband stimuli. Epochs with a level exceeding $\pm 40\mu$ V were rejected and thus not included in the average. The "Bayesian weighting" option of the Eclipse platform was used. Bayesian weighting weights individual recording epochs differently when calculating the average, assigning more weight to the epochs with less noise. The tone burst stimuli were 3 sine waves of duration and were temporally windowed using a Blackmann window.

In the ASSR measurements, the Eclipse platform tests multiple frequencies simultaneously and moves automatically to lower intensities as the accuracy of the result reaches a significant level. Thus, the number of stimulus repetitions is not constant, but depends on the quality of the response instead. If an adequate response has not been obtained by a predetermined time (defaults to 6 minutes on the Eclipse platform), it is concluded that no reliable response could be obtained

for the specified frequency and intensity. (Eclipse Operation Manual)

4.3 Preparations

Electrodes were attached to the skin: one on the forehead just below hairline (Fz), one over the mastoid bone on each side (TP9 and TP10), and a grounding electrode on the chin, or cheek depending on where it was the easiest to attach. Prior to attaching the electrodes, the skin was prepared by scrubbing the skin with fine sandpaper and applying degreaser in order to minimize the electrical impedance of the contact between the electrode and the skin. Too high a contact impedance causes the amount of noise to increase. The objective is to achieve a contact impedance below $3\text{ k}\Omega$. Also, the interelectrode impedance difference should not exceed $1,5\text{ k}\Omega$. With proper preparation of the skin, it is possible to achieve a contact impedance of only $0,8\text{ k}\Omega$ (Kronlund, 2007). The electrodes used in the experiments were disposable electrodes which were pre-gelled, meaning no additional contact gel was required. After the electrodes were correctly placed and connected to the EEG preamplifier, the impedances were checked with the integrated meter of the EEG preamplifier. If the contact between skin and the electrode was not good enough, the electrode was removed and the procedure of attaching the electrode was repeated.

ER-3A insert earphones were placed into the ear canal. The sound of the insert earphones is generated in a separate transducer and is carried to the ear by a plastic tube. A punctured earplug is used to hold the tube in its place and to provide additional insulation against external noise. The transducers can be attached for example to the shirt collar with a clip to avoid discomfort or earphone displacement due to the pull of the transducer weight.

4.4 Organization of the experiments

The experiments were carried out by the author at the Department of Audiology of the Helsinki University Central Hospital during the spring and summer of 2012. The experiments were started at 9 a.m., if possible. The subjects were advised to wake up early to induce sleepiness during the measurements.

The test were performed using the Interacoustics Eclipse platform. A software running on the laptop controlled the Eclipse unit, recorded the measurements and applied necessary signal processing methods to the measurement signals. Insert headphones and an EEG preamplifier were connected to the Eclipse unit. Electrodes were further connected to the EEG preamplifier. All preparations and experiments were conducted by the author.

If the subject didn't have an audiogram, one was obtained by audiometry, conducted by the author. Audiometry was done in a soundproof booth and specialised equipment available at the Department of Audiology.

The experiments were conducted in a room with adequate insulation from external noise. Lights were completely switched off during the experiments in order to both prevent electrical interference and to help the subject relax. A chair with adjustable tilt was used and the subjects could choose a comfortable setting of the

chair, varying between horizontal and semi-supine from subject to subject. In order to minimize interference caused by muscle tension and other brain activity, the subjects were instructed to sit or lie comfortably with their eyes closed and to sleep if possible.

After the preparations, described above, the ABR recordings were performed in the following sequence: click, chirp, 500 Hz TB, 500 Hz NB chirp, 4 kHz TB, 4 kHz NB chirp. After the ABR recordings, the ASSR measurements were done. The time used for the ASSR recordings was limited to 30 minutes. In total, the duration of the session was 2,5–3 hours.

For two subjects, the organization of the experiments deviated from the one described above. For subject N6, the number of averages was 2000. Further, instead of measuring the 4-kHz narrowband responses, the 2-kHz responses were measured instead. Because of the increased test duration, only one ear was tested at the 2 kHz frequency. Also, no ASSR measurements were done. For subject H1, no ASSR measurements were done. Instead, the 2-kHz narrowband responses were measured. This was because the subject had an interesting shape of the audiogram, showing better hearing thresholds around 2 kHz than on surrounding frequencies. For two subjects (N2 and H1), not all electrode contact impedances were within acceptable limits. In both subjects the faulty electrodes were the ones placed over the left mastoid. No improvement in the contact impedances could be detected even after replacing the electrodes twice. Measurements were done regardless of the less-than-ideal contact impedances and good results were obtained from both subjects.

4.5 Data analysis

After the measurements, the ABR recordings were inspected by a clinical audiologist, who annotated the wave V peak if a response was detectable. The wave V peak-to-peak-amplitude was measured from the wave V peak to the next trough. Wave V latency was determined by the wave V peak. Amplitude and latency measurements had to be done with the EP25 module of the Eclipse platform, as it was not possible to export the data to any other software. Statistical analysis of the amplitude and latency data was carried out with IBM[®] SPSS[®] Statistics, version 20. It was assumed that responses from each ear of a subject could be treated independently, so the results from right and left ears were pooled, resulting in measurements from $N = 22$ ears.

Results were analysed at the 5% level of significance. Nonparametric methods had to be used for statistical analysis, since the data did not satisfy the ANOVA assumptions, namely the normality of the residuals and homoscedasticity. Normality was tested using the Shapiro-Wilk test and the equality of variances tested with the Levene's test.

The effect of stimulus intensity on wave V amplitudes and latencies was tested in order to assess the quality of the measurements. It is known that the ABR amplitude and latency have a dependency of the stimulus intensity, and thus if the measurement data can be shown to follow that dependency, the measurements can be seen to represent valid wave V characteristics. The dependency of responses on

stimulus intensity was done using the Friedman test for repeated measures. The effect of stimulus intensity was inspected individually for every stimulus type, each test consisting of four repeated measures corresponding to the four different stimulus levels. If a significant effect was detected, post-hoc analysis was carried out using pairwise Wilcoxon signed-rank tests with the Bonferroni correction for multiple comparisons, resulting in a level of significance of $p = 0,05/6 = 0,0083$ for the pairwise comparisons.

Comparison of traditional stimuli to chirp stimuli in normal-hearing subjects was done by comparing the wave V amplitudes and latencies of the ABR. The click was compared to the broadband chirp and the tone burst stimuli were compared to the NB chirps of the same frequency. Comparison was done between two stimuli of the same intensity—separately for each intensity—using the matched-pair Wilcoxon signed-rank test.

In hard-of-hearing subjects, the ABR responses were analysed case-by-case and no statistical analysis was done because of the difficulty in comparing data from subjects with differing hearing conditions. Instead, the LIF curves were inspected and compared to normative data, as was explained in section 3.5.

Also, the relationships between ABR, ASSR and behavioural thresholds were studied by comparing the ABR and ASSR difference scores.

5 Results

A summary of the ABR and ASSR recordings is shown in appendix B.

5.1 Comparison of click and chirp stimuli in ABR recordings

In Fig. 17, responses to chirp and click stimuli for one of the normal-hearing subjects are shown. Recordings have been done for both ears at four different intensities. In the Eclipse platform, it was not possible to combine recordings from different subjects into a grand average waveform. Neither was it possible to export the waveform data into a numeric format for processing with other analysis tools.

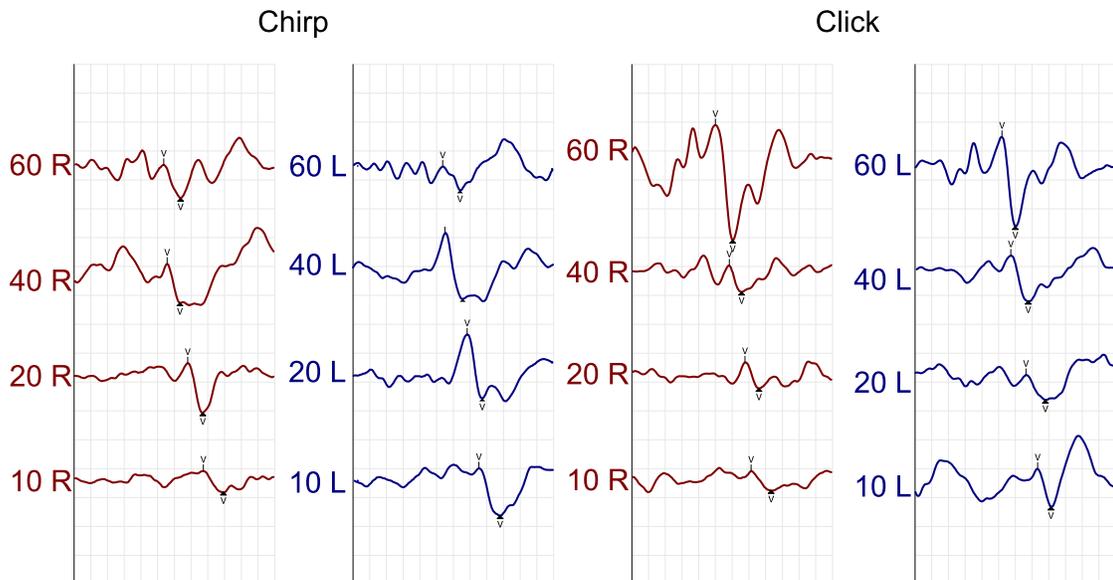


Figure 17: Averaged waveforms from one subject (N4). The two curves pm the left are responses to the broadband chirp stimulus and the two curves on the right to the click stimulus. For each stimulus, both ears are tested at four intensities: right and left ear at 60 dB nHL, 40 dB nHL, 20 dB nHL and 10 dB nHL. The wave V peak and trough have been marked for the determination of wave V latency and amplitude.

5.1.1 Normal hearing subjects

Results for the normal hearing subjects in response to broadband and 4 kHz narrowband stimuli are shown in Fig. 18 and Fig. 19, respectively.

For chirp, click and the 4 kHz NB chirp, there was a significant difference in the wave V amplitudes depending on the stimulus intensity (Chirp: $\chi^2(3) = 7,92$; $p = 0,048$, click: $\chi^2(3) = 16,2$; $p = 0,001$, 4 kHz NB chirp: $\chi^2(3) = 14,2$; $p =$

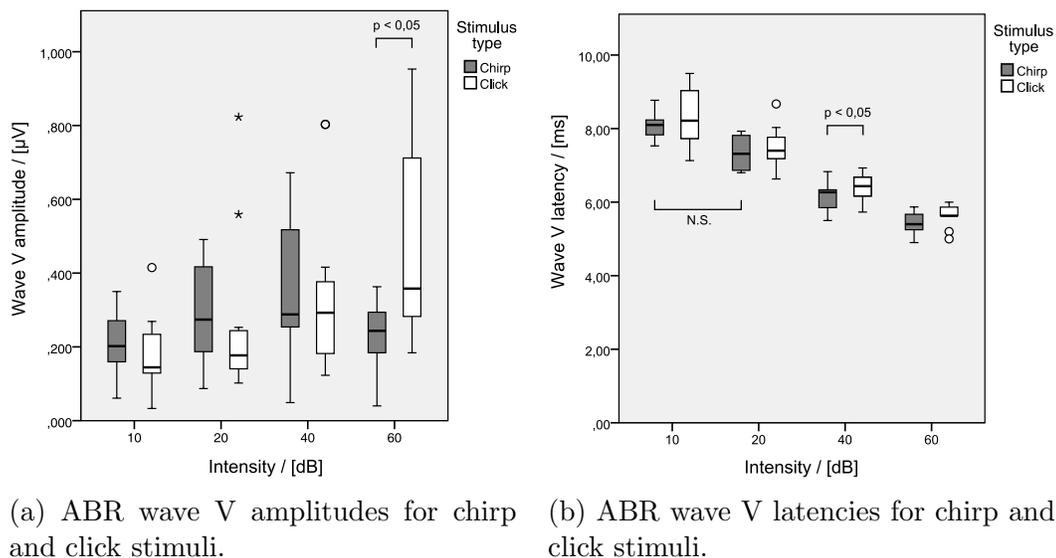


Figure 18: ABR wave V characteristics for chirp and click stimuli, measured from normal hearing subjects.

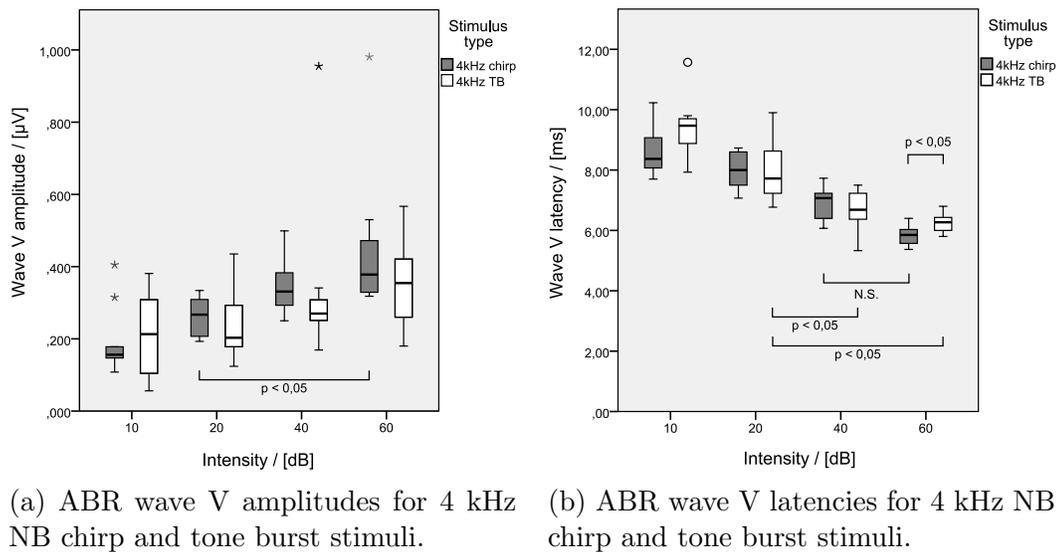


Figure 19: ABR wave V characteristics for 4 kHz NB chirp and tone burst stimuli, measured from normal hearing subjects.

0,003), whereas for the 4 kHz TB there was no significant difference ($\chi^2(3) = 5,229$; $p = 0,156$). Post-hoc analysis revealed a significant difference between 20 and 60 dB for the 4 kHz NB chirp. The post-hoc analysis failed to show a significant difference between any pair of intensities for the click and chirp stimuli.

The difference in wave V amplitudes between a traditional stimulus and the corresponding chirp stimulus of the same intensity was significant only in the case of 60 dB click and chirp ($Z = -2,275$; $p = 0,023$). Other differences did not reach significance. By visual inspection of the data in (Fig. 18) and (Fig. 19), there seems to be a slight trend towards higher amplitudes in the chirp stimuli in the majority of cases.

For all stimuli, there was a significant difference in the wave V latency depending on the stimulus intensity (chirp: $\chi^2(3) = 30,0$; $p < 0,0005$, click: $\chi^2(3) = 33,0$; $p < 0,0005$, 4 kHz NB chirp: $\chi^2(3) = 25,933$; $p < 0,0005$, 4 kHz TB: $\chi^2(3) = 19,348$; $p < 0,0005$). Post-hoc analysis revealed significant differences between all intensities, except for the 4 kHz TB, where the 10 dB intensity didn't show a significant difference to any other intensity and a nonsignificant difference between 10 and 20 dB NB chirps.

The differences in wave V latencies between a traditional stimulus and the corresponding chirp stimulus of the same intensity were significant only in two cases: between 40 dB click and chirp ($Z = -2,045$; $p = 0,041$); and between 60 dB 4 kHz TB and NB chirp ($Z = -2,295$; $p = 0,022$).

5.1.2 Hard-of-hearing subjects

In the hard-of-hearing subjects, responses could not be detected in the majority of recordings. Especially, responses to the broadband chirp and the 4 kHz NB chirp could only be detected in one of the five subjects. The summary of the ABR recordings in table A2 reveals that the traditional click and tone burst stimuli were more successful in eliciting a detectable response.

The clinical value of the results was assessed by plotting the LIF curves from the detected click responses. As was explained in section 3.5, the LIF curves have certain characteristics that are specific to different types of hearing loss. The LIF curves of three hard-of-hearing patients—H1, H2 and H3 (see table A2)—are presented in (Fig. 20). The types of the subjects' hearing loss are:

- H1: Slightly sloping high-frequency hearing loss (from 30/25 dB at 125 Hz to 65/45 dB at 8 kHz, for left/right ear) with a peak around 2 kHz (10/35 dB at 2 kHz). Also bone conduction thresholds have been determined; they match the air conduction thresholds precisely.
- H2: Mixed conductive and sensorineural hearing loss in both ears. Air conduction thresholds produce a flat audiogram in the left ear (~ 50 dB) and a slightly sloping audiogram in the right ear (from 55 dB at 125 Hz to 85 dB at 8 kHz). Bone conduction thresholds produce audiograms sloping towards high frequencies (from 5/5 dB at 250 Hz to 30/35 dB at 4 kHz).

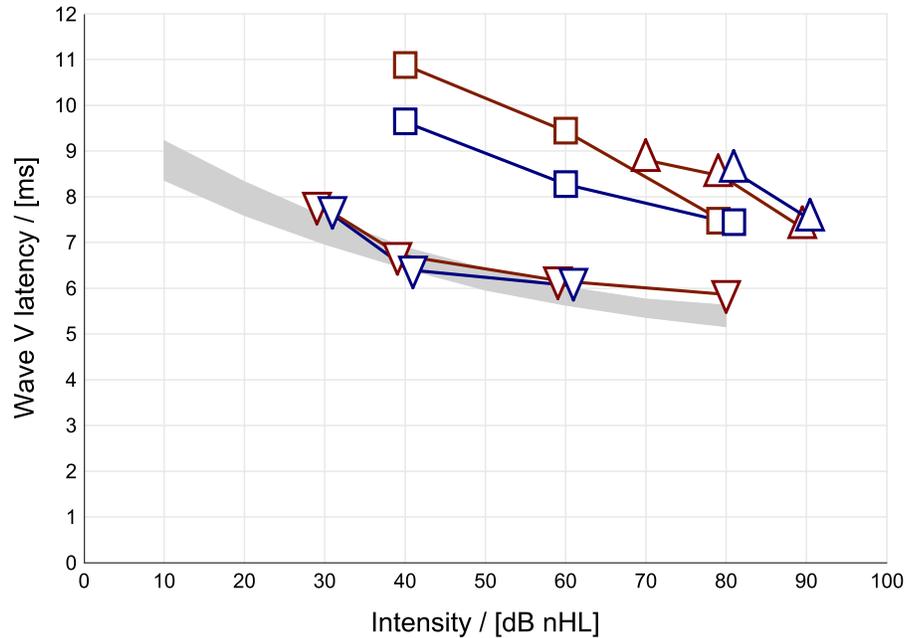


Figure 20: LIF curves of subjects H1 (downward triangles), H2 (squares) and H3 (upward triangles). Red curves represent measurements from the right ear and blue curves from the left ear. The shaded area corresponds to the LIF of an average normal-hearing subject.

- H3: Sloping high-frequency hearing loss (from 25/70 dB at 125 Hz to 90/100 dB at 8 kHz)

The LIF curves in Fig. 20 correspond very well to the ones in Fig. 16. For subject H1, the peak in the audiogram around 2 kHz seems to counterbalance the effect of the slight sloping towards the high frequencies, producing an LIF similar to that of a flat sensorineural HL. For subject H2, the conductive part of the HL is evidenced by the vertical shift of the curves for both left and right ear. The left ear has a flat HL, so the LIF is similar to a purely conductive HL, whereas the right ear has an additional sloping high-frequency sensorineural component. This is shown in the LIF by the elevated curve at 40 dB and 60 dB returning to a “normal” LIF of a conductive HL at 80 dB. The same shape is seen in the LIF curve of subject H3, which is also identical to the sloping high-frequency sensorineural HL curve in (Fig. 16b).

Based on the results from normal-hearing subjects where the chirp latencies were either approximately the same or smaller than the click latencies, the chirp recordings were reviewed by the author. The chirp waveforms were compared to click waveforms and peaks occurring at the same latency or before the click wave V peak were marked as “possible” responses (see Table A2). Possible responses were found from the recordings of subject H3. The LIF curves from subjects H1 and H3, based on chirp responses, are shown in Fig. 21. It can be seen that the LIF curve of subject H1 has shifted vertically by approximately 4 ms. This kind of LIF curve would suggest a conductive HL, but judging by the subject’s audiogram, there is no

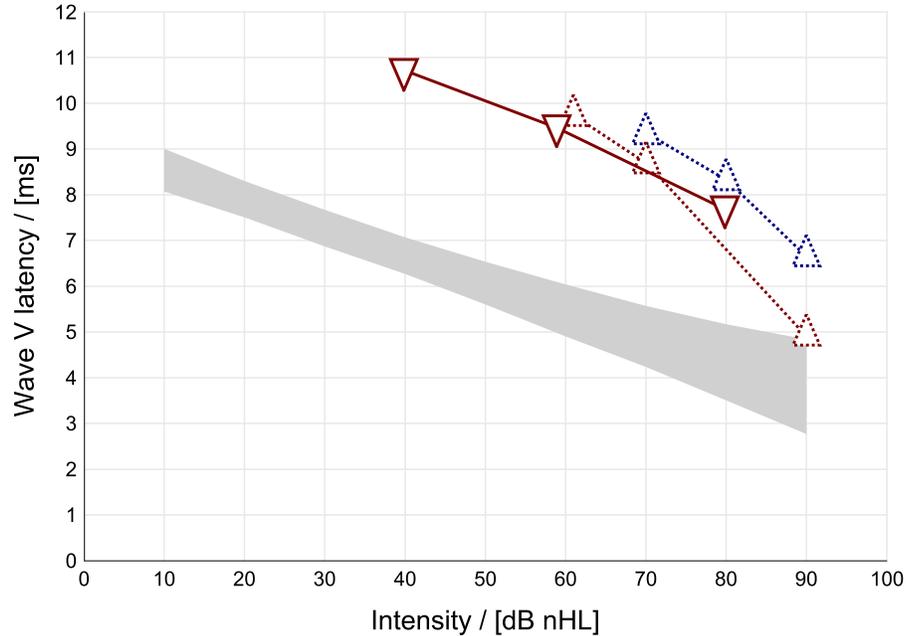


Figure 21: LIF curves in response to the chirp stimulus from subjects H1 (upward triangles) and H3 (downward triangles). The responses from subject H3 are uncertain, and thus marked with dashed lines. The shaded area corresponds to the LIF of an average normal-hearing subject.

reason to suspect a conductive HL, giving room for speculation about whether the response was correctly identified. For subject H3, the LIF curves based on possible responses show similar characteristics to those obtained with click stimuli, but with a steeper slope towards the higher intensities.

5.2 ABR and ASSR thresholds vs. subjective thresholds

For the hard-of-hearing subjects the ABR measurements were done on intensities both above and below their behavioural thresholds. This enables determining the ABR threshold, i.e. the smallest intensity at which a response can still be detected. For the normal-hearing subjects, the measurements were done only down to 10 dB, which was above the ABR threshold for most subjects.

The click ABR threshold was compared to the average of behavioural thresholds from 1 kHz to 4 kHz. ABR thresholds could be determined for both ears in three subjects (H1, H2 and H3) and further for one ear in one subject (H5). A difference score was calculated by subtracting the ABR threshold from the average hearing threshold. Difference scores from left and right ears were pooled, resulting in seven difference scores. The average difference score was -10 dB ($N = 7$, SD 15,7 dB).

For the 4 kHz TB, the average difference score as measured from four hard-of-hearing subjects was 18 dB ($N = 8$, SD 22 dB). For the 500 Hz TB, the average difference score from three hard-of-hearing subjects was -2 dB ($N = 5$, SD 9,08 dB).

The 4 kHz chirp response could only be detected in one subject. For this subject

(H1), the ABR thresholds were in many cases below the behavioural thresholds. The average difference score across all frequency-specific stimuli (500 Hz, 2 kHz and 4 kHz) were 3,33 dB for the TB stimuli and 8 dB for the chirp stimuli. This is most likely an underestimate of the difference scores, since for this subject the responses could in some cases be detected even for the lowest tested intensity. The true ABR thresholds could thus be even lower and the difference scores higher.

ASSR thresholds were determined for 500 Hz, 1 kHz, 2 kHz and 4 kHz. Difference scores were calculated by subtracting the ASSR threshold from the behavioural threshold of the corresponding frequency. Difference scores from both ears were again pooled. The average difference scores for the normal-hearing subjects were -43 dB at 500 Hz (N = 6, SD 23 dB), -24 dB at 1 kHz (N = 8, SD 14 dB), -24 dB at 2 kHz (N = 8, SD 13,7 dB) and -21 dB at 4 kHz (N = 8, SD 12 dB). ASSR measurements were conducted also on four hard-of-hearing subjects (H2–H5). ASSR thresholds could be determined for both ears at all frequencies in one subject (H5), for the right ear at all frequencies in one subject (H4) and for the left ear at one frequency in one subject (H3). For one subject (H2) no thresholds could be determined. Adding the difference scores of these subjects to the normal-hearing subjects, the average difference scores for all subjects were -44 dB at 500 Hz (N = 10, SD 22 dB), -29 dB at 1 kHz (N = 11, SD 15 dB), -27 dB at 2 kHz (N = 11, SD 12,7 dB) and -16 dB at 4 kHz (N = 11, SD 15 dB).

5.3 Discussion

5.3.1 Comparison of wave V amplitudes

Based on the results from both normal-hearing and hard-of-hearing subjects, no clear advantage could be shown in using the chirp and NB chirp stimuli in ABR recordings, compared to the traditional click and tone burst stimuli. Wave V amplitudes and latencies were compared in normal-hearing subjects, but significant differences between corresponding chirp and traditional stimuli were found only in three cases: in amplitudes between the 60 dB chirp and click; in latencies between the 40 dB chirp and click; and in latencies between the 60 dB 4 kHz NB chirp and TB. Shorter latencies have been reported for the chirp stimulus (Elberling and Don, 2008) so this result is in line with previous work, but remarkably, the difference in amplitudes was in the opposite direction as was anticipated; the click elicited significantly higher wave V amplitudes than the chirp at the 60 dB intensity level. In the study by Dau *et al.* (2000) no significant difference between click and chirp stimuli at the levels 50–60 dB was reported. It was suggested, that this was caused by the upward spread of excitation, leading to decreased synchrony at higher stimulus levels. Similar results have been reported in (Elberling and Don, 2008) and (Elberling *et al.*, 2010).

To the author’s knowledge, no reports exist of ABR recordings using the chirp stimulus on hearing impaired subjects. Although the click performed better in the recordings of the hard-of-hearing subjects, no reliable conclusions can be made about the usefulness of the chirp in ABR measurements of hearing impaired patients based

on the results in this study.

Based on visual inspection of the data in Fig. 18 and Fig. 19, there seems to be a slight trend towards higher amplitudes and smaller latencies for the chirp responses. It is important to ask, whether the reason for this trend not reaching statistical difference could be attributed to the sample size, i.e. would a bigger sample have shown better results. Using the data from Elberling and Don (2008), where the chirp was shown to elicit 1,5–2 times higher wave V amplitudes than the click, an estimate for the sample size can be calculated using the equation (2) (Uhari and Nieminen, 2001).

$$n = \frac{2SD^2}{(\bar{x}_1 - \bar{x}_2)^2} f(\alpha, \beta) \quad (2)$$

With the values $\alpha = 0,05$ for the test significance level and $\beta = 0,8$ for the test power, the function f gets a value of 7,9. Further, taking the mean values and the worse of the two standard deviations for click and chirp amplitudes at 60 dB from (Elberling and Don, 2008), the sample size needed to obtain a significant difference would be $n = 5,9$, which would mean that the sample size in the current study is indeed adequate. However, the calculated standard deviations of the results in this study are higher than the ones in (Elberling and Don, 2008), leading to the requirement of a bigger sample in order to find a significant difference with the difference in amplitudes reported in (Elberling and Don, 2008). It is therefore possible that the sample was not of adequate size, taking into account the accuracy of the experiments. The high variance in measurements is discussed in section 5.3.2.

Recently, Elberling *et al.* (2012) compared ER-2 and ER-3A earphones and found that ER-2 insert earphones elicited slightly larger click-ABRs and considerably larger chirp-ABRs than those generated by the ER-3A. The authors concluded that because of its acoustical properties, the ER-2 earphone is a better choice than the ER-3A when using chirp stimuli in normal-hearing adults at stimulus levels below 60 dB. More precisely, the comparison of the amplitude responses of the two earphones reveals that the ER-3A has a much more limited frequency range (Fig. 22). The ER-2 earphones have been used in the majority of those previous experiments that have indicated larger chirp-elicited responses. A reasonable explanation for the lack of evidence in favour of the chirp in this study could be that the earphones used in this study were of the type ER-3A. In the 10–40 dB range, the chirp amplitudes seem to have a slight tendency to be higher than for the click, but as the earphones are different than in Elberling and Don (2008), a 1,5–2-time gain in amplitude is not achieved. The effect of the earphone is even more evident for the 4 kHz narrowband measurements, where no significant differences were found between TB and NB chirp wave V amplitudes. As can be seen from the amplitude response of ER-3A, the earphone behaves like a passband filter from 1 kHz to 4 kHz with a remarkably steep slope after the 4 kHz cut-off frequency. The TB and NB chirps are narrowband stimuli with their energy centered *around* 4 kHz. This means that some of the stimulus energy above 4 kHz is lost because of the filtering effect. It might be, that any possible gain that could be achieved by using the NB chirp instead of the TB, is rendered insignificant by the ER-3A earphones.

Although it seems plausible that the earphone acoustics had a significant influ-

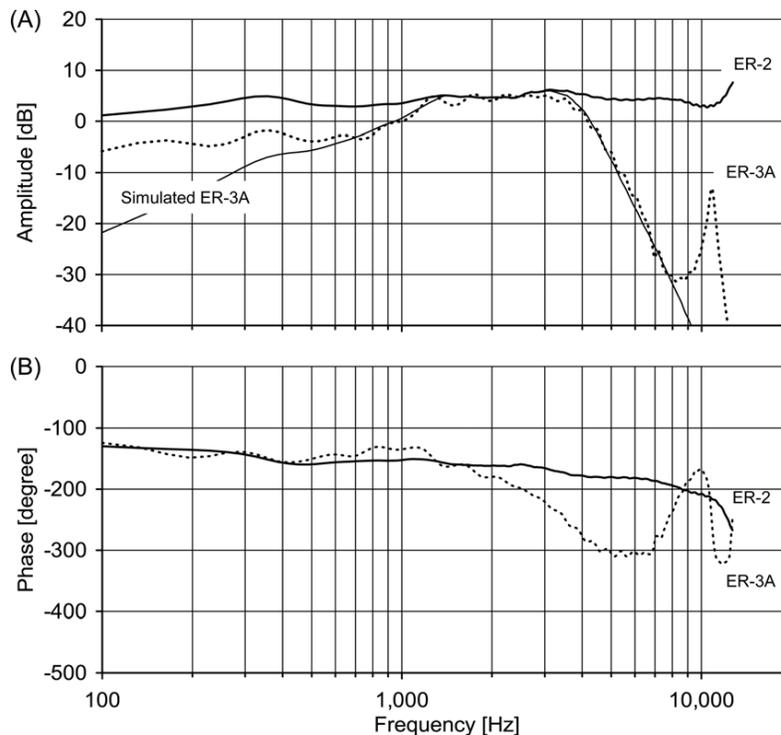


Figure 22: Amplitude (A) and phase (B) responses of ER-2 and ER-3A insert earphones. (Elberling *et al.*, 2012)

ence on the results in this study, some questions remain. The CE-chirp implemented in the Eclipse platform has a flat amplitude response, whereas previous implementations of the chirp had an amplitude response that followed the click stimulus' response. Would a similar effect of earphone selection be seen if the click-matched chirp stimuli was used? Also, (Elberling *et al.*, 2012) suggest that the main reason for the ER-2 earphones eliciting higher ABR amplitudes is the added contribution from higher frequency regions as compared to the ER-3A. If indeed the increased wave V amplitudes of the chirp are caused by high-frequency activity, then is there any advantage in using the chirp stimulus in subjects with sloping high-frequency hearing loss? Further, if the reason for the chirp eliciting a higher wave V amplitude is in the increased synchrony across the basilar membrane, then should this effect not be visible when the response is generated mainly by the lower frequency regions, as is the case in high-frequency hearing loss?

5.3.2 Quality of the ABR recordings

As can be seen in Fig. 18a and Fig. 19a, the variance in wave V amplitudes is large. The same applies for both click and chirp responses, so it does not seem to be characteristic to just one stimulus type. Exemplary ABR waveforms, as presented in books and product brochures, have often very clear morphologies and easily detectable responses down to the ABR threshold. Although waveforms as the one in Fig. 8 can on occasion be recorded, often even determining whether a recording con-

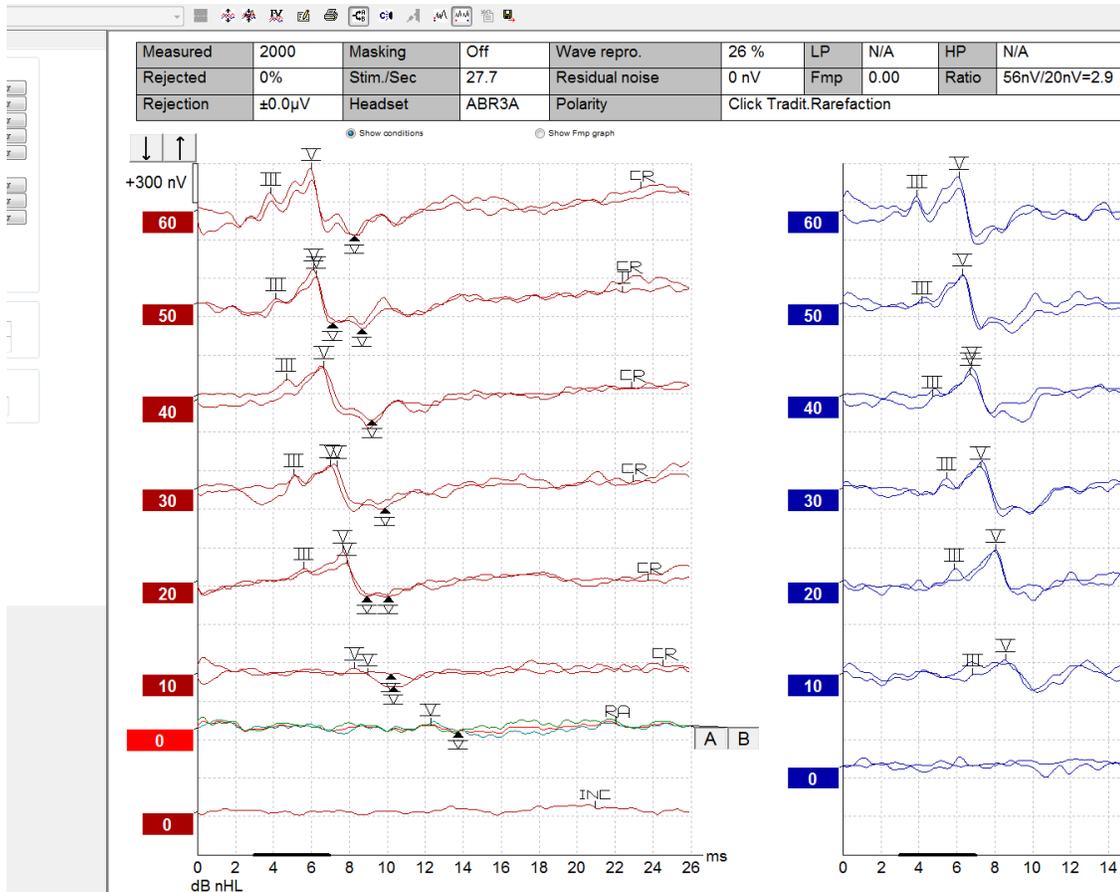


Figure 23: Ideal ABR recordings as presented in the Eclipse manual (Eclipse Operation Manual).

tains a response at all can be a challenging task. In Fig. 23, there is a figure taken from the Eclipse platform operation manual showing clear ABRs with the waves III and V marked. The next figure (Fig. 24) contains the click responses from one of the normal-hearing subjects in this study. Although the responses from the left ear of the subject (rightmost blue curves) are quite clear, the responses from the right ear (red curves) contain a significant amount of interference. The possible origins of interference in ABR measurements were discussed in section 3.2.2 and included muscle activity and other brain activity. In very noisy recordings, the determination of the wave V amplitude becomes cumbersome, since rather than showing up as a clear peak, the wave V can be buried under a large fluctuation, as is visible in Fig. 24. The same applies for the following negative trough, which is used in determining the peak-to-peak amplitude.

One way to reduce the amount of interference in measurements is to increase the number of averaged recording epochs. A common number of averaged repetitions in clinical trials is 2000. In the current study 1000 epochs were recorded in order to keep the duration of the experiments reasonable. Doubling the number of repetitions to 2000 would have improved the signal-to-noise-ratio by merely 3 dB, while doubling

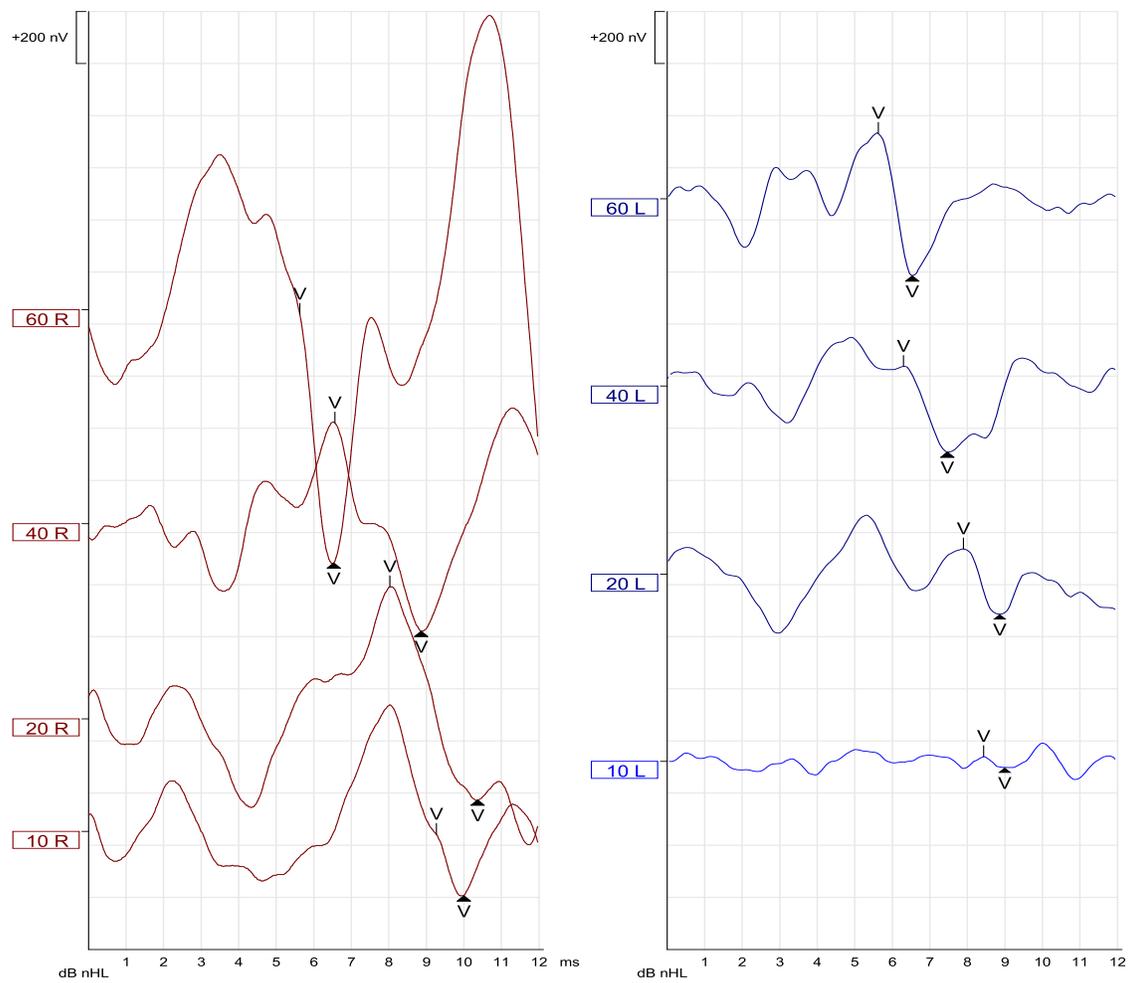


Figure 24: An ABR recording with considerable background noise.

the time needed for the experiments. With the given stimulus rate (11,1 / s) and number of repetitions (1000), the time needed to obtain a single averaged response was 1,5 minutes. In total, 48 responses were obtained (6 stimuli at 4 intensities from both ears), adding up to 72 minutes for the duration of the ABR experiments. Adding to this the time needed for preparations and the ASSR measurements, the experiments took between 2,5 to 3 hours. The aforementioned 3 dB improvement in SNR would have meant a total duration of 4 hours for the session, which was not considered acceptable. Because the subjects were not under anaesthesia or sedation, some had trouble relaxing during the whole duration of the experiments. This was shown as increased interference caused by muscle tension and eye movements. The gains in increasing the number of recordings might have been superseded by an increased level of interference towards the end of the session.

A professional audiologist can detect the possible wave V peaks even from very noisy recordings by knowing exactly where to look, i.e. at which latencies the peaks normally occur. Although this is often successful in normal-hearing subjects, it cannot be ruled out whether there really is a response behind the interference or if there just happens to be a random peak caused by interference with a suitable latency. It might also be that some pathology of the auditory chain causes increased ABR latencies, leading to the possibility of misinterpreting a noisy ABR response. Combining information from measurements done on different intensities reduces the probability of error, since the latency of wave V is known to increase as intensity decreases. This way, it is possible to see if a certain peak in the response follows the same intensity dependence as wave V. The results indicated a significant dependency of both wave V amplitude and latency on the stimulus intensity for chirp, click and 4 kHz NB chirp stimuli. Also in the 4 kHz TB, there was a significant relation between latency and intensity, whereas no significant dependency of the amplitude on intensity was found. These results suggest that the wave V peaks and troughs were correctly identified, as they followed the known wave V behaviour. Also, the click-elicited responses from hard-of-hearing subjects were successful in showing characteristic LIF curves that matched those presented in earlier studies.

Although the responses were analysed by a professional audiologist, the actual measurements were conducted by the author who had no previous experience of ABR measurements. It is obvious that this inexperience had an effect on the obtained results, especially evident in the recordings of hard-of-hearing subjects. In many cases the stimulus intensities used were too low for the hard-of-hearing subjects, either making it very difficult to detect a response or even in a few cases making it impossible by beginning below the ABR threshold. The effect of the person conducting the experiments is, however, likely to influence all measurements equally, so the anticipated difference between click and chirp stimuli should have been visible regardless of that. Especially in the hard-of-hearing subjects the failure of the chirp to elicit detectable responses is likely to be caused by some other factor, since responses were found in the same subjects using the click.

5.3.3 ASSR and ABR thresholds

ASSR and ABR thresholds were compared to the behavioural thresholds. ASSR measurements were conducted on four normal-hearing subjects and four hard-of-hearing subjects. Likely because of a high level of interference from muscular activity, the ASSR recordings were less successful in the hard-of-hearing subjects. The difference scores were obtained by subtracting the ASSR threshold from the behavioural threshold. The difference scores were higher than those observed in previous studies (Lins and Picton, 1995; Vander Werff, 2009). The higher difference scores can be caused by the decision to limit the ASSR measurement duration to 30 minutes. It is likely that more accurate estimates could have been obtained if the time had not been limited. Nevertheless, at the 500 Hz frequency the ASSR thresholds are remarkably imprecise. The ABR click thresholds were compared to the average hearing thresholds at 1–4 kHz in three hard-of-hearing subjects. Difference scores were calculated and found to be smaller (-10 dB) than the average ASSR difference scores. Although also in this case the accuracy could have been improved by increasing the number of measured intensities, this result is within the range that is normally expected for ABR measurements (Arnold, 2000).

In some hard-of-hearing subjects, the Eclipse ASSR was unable to find responses because of interference caused probably by muscular activity. The ASSR program kept rejecting recording epochs that had too high a level. Although the time limit in the Eclipse platform for finding a reliable steady-state response is 6 minutes, rejected epochs do not advance the timer. In some subjects the ASSR test did not advance at all during the majority of the experiment because epochs were rejected. In the same subjects, ABR responses could be detected, although noisy epochs are rejected similarly. It seems that the Eclipse ASSR is more sensitive to interference than the ABR.

Interestingly, the 4 kHz TB thresholds were on average below the behavioural thresholds in hard-of-hearing subjects. The 500 Hz TB thresholds were relatively close to the behavioural thresholds. These results have to be interpreted cautiously because of the small number of successful recordings. However, even as a single case example, the ABR thresholds of subject H1 are interesting, because the average difference score across the three measured frequency-specific stimuli are positive for both TB and NB chirp stimuli. This might be explained by the audiogram of the subject, which has a peak around 2 kHz with hearing thresholds of 10 dB in the right ear and 35 dB in the left ear. It should also be remembered, that even the frequency-specific stimuli are not ideal in that their frequency response spans over a certain bandwidth. It is therefore possible that portions of the stimulus energy excite the region with better hearing thresholds, thus creating a response even at intensities below the hearing threshold at 4 kHz. Nevertheless, the same trend could be detected in other hard-of-hearing subjects as well, since the average difference score across four subjects was positive. Although smaller difference scores can be expected from subjects with sensorineural hearing loss (Arnold, 2000), another possible explanation could be that the behavioural thresholds are not accurate.

Comparison of the frequency-specific ASSR and ABR thresholds cannot be done

reliably, since both ASSR *and* ABR thresholds could only be determined for the right ear of one subject(H5). Comparison of the average difference scores of ABR and ASSR does not give any accurate information, either, because the averages were obtained from different subjects, the ASSR difference scores consisting mainly of those measured from normal-hearing subjects and the ABR difference scores consisting only of those measured from hard-of-hearing subjects.

Unfortunately, the question of whether the chirp could be used to provide more accurate estimates of the behavioural hearing threshold, could not be answered because of the small amount of detected responses to the chirp in hard-of-hearing subjects. Also, in normal hearing subjects the levels were above the ABR threshold in most cases, so the comparison of ABR thresholds is not possible in those subjects, either.

In conclusion, the accuracy of the ASSR and ABR thresholds cannot be reliably determined based on the results of this study. In many cases the thresholds were either below or above the measured intensity range, meaning that for many subjects the thresholds could not be determined. The number of measured intensities could not be increased, since this would have meant significant increases in the test duration.

6 Summary and conclusions

In this thesis, the use of a new chirp stimulus in ABR measurements was studied. First, the theoretical background was reviewed starting from the hearing organs and following the auditory chain to the auditory cortex. Psychoacoustic phenomena underpinning the chirp were also addressed. Also, a brief look at the field of audiology was taken. Section 3 discussed auditory evoked potentials and their use in objective testing of the hearing system. The design of the chirp stimulus was described in section 3.4. A description of the methodology and experiments of the thesis were given in section 4. Results of the experiments were presented and their meaning discussed.

Clinical tests were conducted on six normal-hearing subjects and five hard-of-hearing subjects using both traditional stimuli: the click and tone burst; and the corresponding chirp and narrowband chirp stimuli. Also auditory steady-state responses were measured. The quality of the ABR recordings was assessed in terms of the ABR amplitude and the accuracy of the estimates of behavioural hearing thresholds. Also, the overall quality of the ABR recordings was discussed.

The results did not show a significant difference between the amplitudes of chirp- and click-elicited ABR responses. This is in contrast with earlier studies of the chirp stimulus, where the chirp has been found to result in ABR wave V amplitudes 1.5–2 times higher than for the click stimulus. A likely explanation for the result is that the ER-3A earphones used in this study have a more limited amplitude response than the ER-2 earphones used in many of the previous chirp studies. The accuracy of ABR thresholds could not be reliably determined because of the small number of obtained thresholds. In most cases, the ABR thresholds were either above or under the measured intensity range. The ASSR thresholds could be determined for four normal-hearing subjects and for one hard-of-hearing subject. For two hard-of-hearing subjects, the ASSR thresholds could be partly determined. The ASSR threshold results were higher than those reported previously, but this was likely caused by the limited time (30 min) allowed for the ASSR recordings.

Future work on the subject of the chirp stimulus should include studies on hearing-impaired subjects. All published research this far has been done on normal-hearing subjects, leaving open the question of the validity of the chirp in ABR measurements on hearing-impaired subjects. The chirp and click stimuli should also be compared in ABR measurements using a higher stimulus rate. Threshold estimation is often done using a higher stimulus rate than the one used in scientific papers and accurate clinical studies. Comparing responses between different stimulus rates would be of high value when judging the value of the chirp stimulus. It would also be interesting to use the chirp stimulus in ASSR measurements and compare the results to traditional ASSR stimuli.

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A WHO table of grades of hearing impairment

Table A1: WHO table of grades of hearing impairment. (WHO/PDH/91.1)

Grade of impairment	Corresponding audiometric ISO value (Average of 500, 1000, 2000, 4000 Hz) of better ear	Performance	Recommendations
0 – No impairment	25 dB or better	No or very slight hearing problems. Able to hear whispers.	
1 – Slight impairment	26–40 dB	Able to hear and repeat words spoken in normal voice at 1 metre.	Counselling. Hearing aids may be needed.
2 – Moderate impairment	41–60 dB	Able to hear and repeat words using raised voice at 1 metre.	Hearing aids usually recommended.
3 – Severe impairment	61–80 dB	Able to hear some words when shouted into better ear.	Hearing aids needed. If no hearing aids available, lip-reading and signing should be taught.
4 – Profound impairment including deafness	81 dB or greater	Unable to hear and understand even a shouted voice.	Hearing aids may help understanding words. Additional rehabilitation needed. Lip-reading and sometimes signing essential.

