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In vitro investigations of a mastoid anchored transcutaneous bone conduction hearing system

Master of Science Thesis

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Abstract

A conventional “Percutaneous Bone Anchored Hearing Aid” (Baha®) is an important rehabilitation alternative for patients suffering from conductive or mixed hearing loss. Even if these percutaneous implants have a reasonably low complication rate, there are some drawbacks reported.

A system with transcutaneous energy transmission has the main advantage that no percutaneous plug is needed as the implanted transducer is permanently placed in the temporal bone close to the cochlea. The signal is transmitted with an induction link through the intact skin.

The aim of this study was to investigate the feasibility of MED-EL Vibrant® Soundbridge® with a Balanced Electromagnetic Separation Transducer (BEST®) attached instead of the floating mass transducer. This transcutaneous hearing system’s performance was compared with a conventional Baha® Classic 300 on a skullsimulator and a cranium.

Results from the skullsimulator show that the transcutaneous system's output is about the same as the percutaneous system’s output at frequencies below 1 kHz. Above 1 kHz the response is about 10 dB lower than for the transcutaneous system. However, results from the semi dry skull shows that the behaviour for the two systems is about the same at frequencies above 1 kHz, but the percutaneous system has a response about 10 dB higher than the transcutaneous system below 1 kHz.

Based on the results, it is concluded that the transcutaneous system can be a feasible alternative to the BAHA®. Nevertheless to have higher response than the percutaneous system on the semi dry skull and on the skullsimulator in the whole frequency range, the transcutaneous system needs to be optimized. However, result from measurement on a cadaver head shows that optimization is probably not needed.

Keywords: Hearing system, bone conduction, mastoid, bone anchored, percutaneous transmission, transcutaneous transmission, bone conduction implant.

Acknowledgements

We would like to thank MED-EL for providing us with a Vibrant® Soundbridge® to use during this study. Further we would like to thank Professor Bo Håkansson for guidance and a pleasant collaboration. Finally, many thanks to our fellow students Maria Stegberg and Magnus Smith for joyful company and teamwork, helpful advices and for sharing their results with us.

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1 Introduction

Within the educational program to get a Master of Physics diploma there is a compulsory thesis all students have to do. This master thesis is performed at the end of the education and during the thesis, the students are expected to do a more thorough scientific study that should be linked to courses studied and by using knowledge achieved during the education. The master thesis shall be presented both in a written report and with an oral presentation. This is the written report.

In the authors' chase for a topic to do for their master thesis, contact was taken with a former teacher, Professor Bo Håkansson. He proposed a subject he thought could be of interest, which it also was. This report presents the subject, the results and the conclusions drawn from the study.

1.1 Background

The perceptions of sound differ from person to person. The hearing is to the highest degree subjective, what someone thinks is a comfortable sound could be an unpleasant sound for another person. One thing everyone has in common is how the auditory organs are constructed. The most important part of the hearing organs is the cochlea. The cochlea is exposed to vibrations from the surrounding; these vibrations stimulate the emitting of electrical signals, which the brain transforms into what is interpreted as sound. [1]

The vibrations that the cochlea is exposed to, can be of two different kinds. Firstly, the sound can be caught by our external ear. Via the eardrum and all the small bones in the middle ear the sound gets amplified and reaches the cochlea at the oval window. This way of hearing is called air conducted hearing or AC hearing. Secondly, the vibrations can arise due to the fact that the skull bone around the cochlea is put into motion. This will also be interpreted as sound and is called bone conducted hearing or BC hearing.

The skull bone can be set into vibrations in various ways. Sound waves which hit the head will be dampened by the overlaying soft tissue, but some of them will reach all the way to the bone that finally will cause vibrations in the cochlea. The perhaps most common way our skull is set into motion is through our own speech. When speaking, the vibrations will propagate from the oral cavity to the skull bone and in that way cause vibrations which are interpreted as sound by the brain. This is why your own voice does not sound familiar on a sound recording.

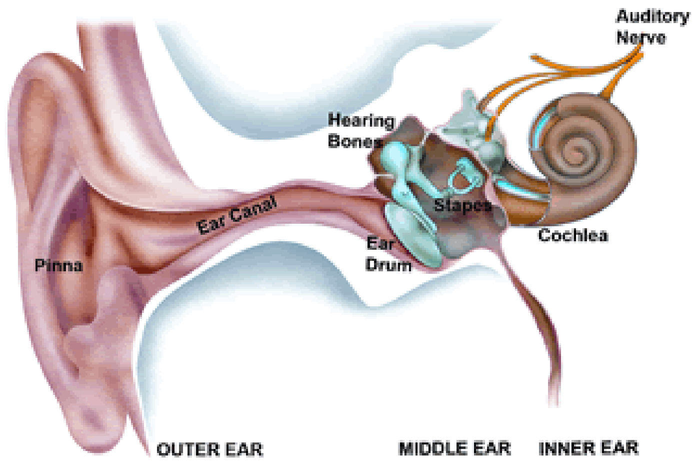

Figure 1. Anatomy of the human hearing system. [2]

Injuries and diseases in the middle or outer ear can lead to the AC hearing getting impaired and even to deafness. It is however possible to construct hearing systems that induces vibrations to the skull bone. These vibrations will also propagate to the cochlea. One commonly used system is the Baha® Classic 300 which is designed to induce vibrations in the parietal bone via a transducer and gives the patient BC hearing. Another system is the MED-EL Vibrant® Soundbridge® which causes vibrations directly in the small bones in the middle ear. Each of those systems has their strengths and weaknesses. The advantages with the Baha® system is that it helps patients who cannot use middle ear implants but its weakness is that it is a percutaneous system which increases the risk for infections. The strength of the MED-EL system is the transcutaneous system which avoids the infection risk. The disadvantages are that the system cannot help some of the patients with defective middle ear and that the procedure to implant it, is more complicated, compared to the Baha® system.

1.2 Question formulation

This thesis will investigate if it is possible to use the strengths of the two existing systems, the transcutaneous MED-EL Vibrant® Soundbridge® to avoid infections and the Baha® Classic 300 to help patients who cannot use the middle ear implant. Measurements of relevance will be performed with a MED-EL Vibrant® Soundbridge® with its floating mass transducer replaced with a BEST® transducer. Another presumed advantage with this combined system, is that the BEST transducer can be placed closer to the cochlea, with resection of the mastoid, compared to where the percutaneous Baha® Classic 300 system is placed. The expectation is that, by comparing the results from the measurements of the combined system and the Baha® Classic 300 system, it will be possible to answer if it is possible to construct an implantable hearing system that could be a realistic alternative, for the concerned patients.

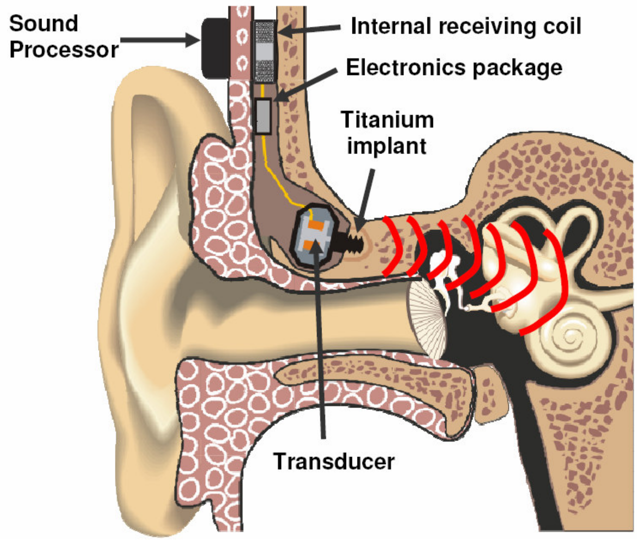

Figure 2. The transcutaneous hearing system with the implanted transducer [3].

1.3 Objectives

The objectives of this thesis are: to do accurate and relevant measurements; to do calculations on the data collected; to visualize the results in well describing figures; and to evaluate which of the two systems undergoing measurements is the most preferable, with respect to certain criteria.

1.4 Notation

In the figures, the notation dB(X) is frequently used and is referring to conventional use of decibel with the reference level of one X. If the reference contains power, the square of the amplitude is specified, otherwise the square is implied. The sound level in dB SPL is given relative 20 µPa.

2 Equipment

2.1 The Agilent

The single most important equipment during the measurements was the Agilent 35670A 4-channel FFT Dynamic Signal Analyzer. It is a powerful instrument which does not only measure signals; it also works as a signal generator. This signal analyzer device was used in two different modes, the FFT mode and the Swept-sine mode.

The fast Fourier transform, FFT, makes it possible to almost instantly achieve measurement data. This makes the FFT mode a good choice for those situations where one wants fast results. However, in the measurements done; this instrument mode was used only a few times since it is impossible to control the source in a desirable way, i.e. to keep the amplitude fixed in one of the input channels.

Where the FFT driving mode fails, the swept-sine mode proves to be a strong alternative. While using swept-sine mode it is possible to keep the incoming signal constant at different frequencies, on a desirable channel.

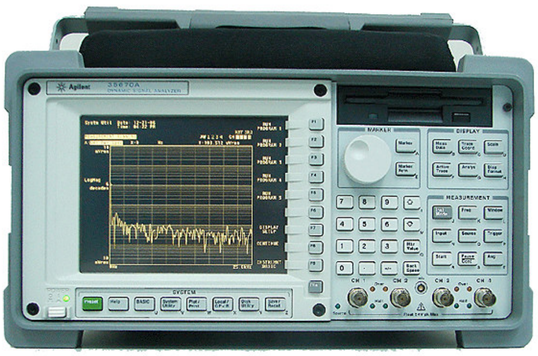

Figure 3. A front view of Agilent 35670A 4-channel FFT Dynamic Signal Analyzer [4].

The data from the measurements can be exposed to various mathematic transforms before it is plotted on the screen of the Agilent. The data can also be saved onto ordinary floppy discs for later analysis. There are many features that can be used in this powerful tool, however, for a more thorough description of the analyzer the manufacturer's homepage should be visited. [4]

2.2 The Skullsimulator

The skullsimulator TU-1000 (Figure 4) was used in several of the measurements. It is designed to fit into most soundproof boxes and with impedance which should mimic the load of a skull. It is used to measure the force output from various transducers. The force is converted into a voltage which can be measured with for example the Agilent. The impedance of the simulator is created by a mass of 50 g which is ~10 times more than the counterweight of the transducer. How the two masses of the transducer and the mass of the skullsimulator affect the output force is seen in Figure 14.[5]

Figure 4. The skullsimulator TU-1000 to the right and its power supply unit to the left.

2.3 The Condenser microphone

To measure sound pressure levels a condenser microphone; Brüel & Kjær Condenser Microphone Type 4134, which is connected to a power supply; Brüel & Kjær Microphone Power Supply Type 2804, was used, see Figure 5. The voltage caused by the sound is measured with the Agilent. By knowing the relation between the sound pressure level and the voltage, it is possible to calculate the sound pressure level. However, to do this calculation a calibration is needed. This was done by using Brüel & Kjær Calibrator Type 4230 which provides a source with known and very accurate frequency and sound pressure level.

Figure 5. The condenser microphone with its power supply.

In some cases, where the source from the Agilent was controlled by keeping the voltage from the microphone constant, i.e. when a measurement with constant sound pressure level is wanted, an amplification of the voltage from the microphone, for low sound pressure levels, was needed. This is due to limitations in how low the desirable input-signal of the Agilent could be set.

The problem was solved by building an amplifier with an operational amplifier and a high-pass filter to remove the DC-component from the microphone signal. See Figure 6 for how it looked.

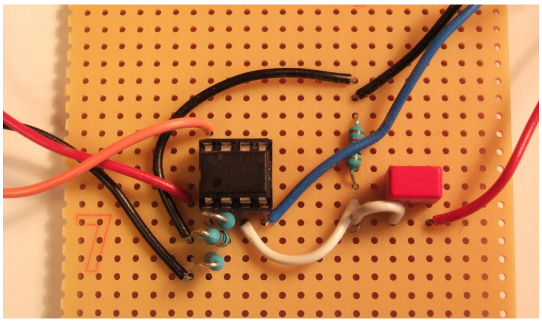

Figure 6. An OP-Amplifier with a high-pass filter, designed to amplify the signal ten times and to remove DC-voltages from the condenser microphone amplifier.

2.4 The Anechoic chamber

To avoid noise and sound from the surrounding a Brüel & Kjær Anechoic Test Chamber Type 4222 (Figure 7) was used. By placing the object undergoing measurement in the chamber, more accurate data was received. In the ceiling inside the chamber there is a loudspeaker mounted which was used in several of the measurements.

Figure 7. Brüel & Kjær Anechoic Test Chamber Type 4222.

2.5 The Hearing Laser Vibrometer

To measure vibrations on the skull a Polytec HLV-1000 which is a laser Doppler vibrometer, LDV, was used. The principle behind this instrument is the Doppler effect. When the laser light hits a moving object the reflected light receives a frequency shift depending on the velocity of the moving object. This reflected light is collected and superpositioned with a reference light beam inside the LDV. From this signal it is possible to extract the velocity of the moving object. The velocity measured by the LDV is converted to a voltage that is measured by the Agilent.

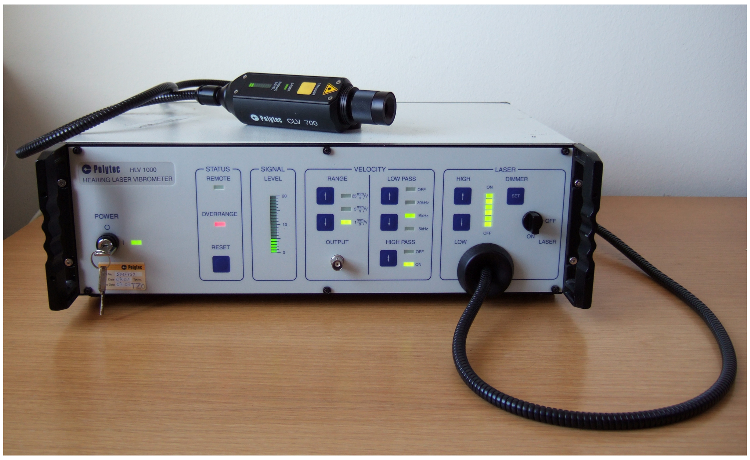

Figure 8. The Polytec HLV-1000.

2.6 The Skull

In the later measurements an intact skull (cranium) was used. The inside of the cranium is covered with damping material to better mimic the properties of a skull. However, the damping material was applied ten years ago and it is reasonable to assume that it has gotten stiffer. Probably the same is true for the bone itself and its joints. But the skull is still a useful complement to measurements on human cadaver heads. On the skull there were several positions where the transducers could be mounted onto, see Figure 9 below.

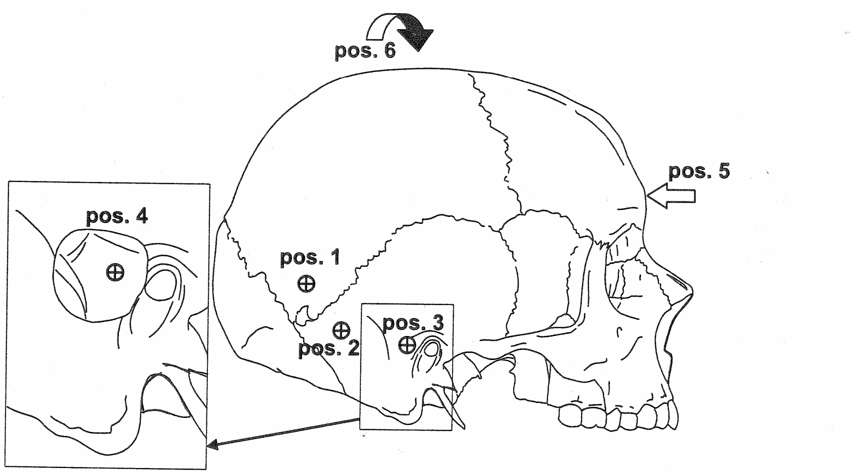

Figure 9. This figure shows some positions that can be used to attach transducers on. Position 1 is the normal place for the BAHA system and position 4 is the planned position for the BEST transducer. To be able to attach BEST to position 4 one has to do a resection of the mastoid to reach the planned place [5].

Of specific interest were positions 1 and 4. Position 1 is on the parietal bone and is the normal spot for the BAHA transducer. Position 4 is placed inside the temporal bone and is reached after resection of the mastoid. This was the planned, and in this thesis investigated, position for the BEST transducer. In Figure 10 it can be seen how the skull truly looks like and how the laser is measuring vibrations on the promontory in the ear channel. This spot is close to the cochlea and the bone there is the second hardest structure in the body, after enamel, and behaves as a rigid body.

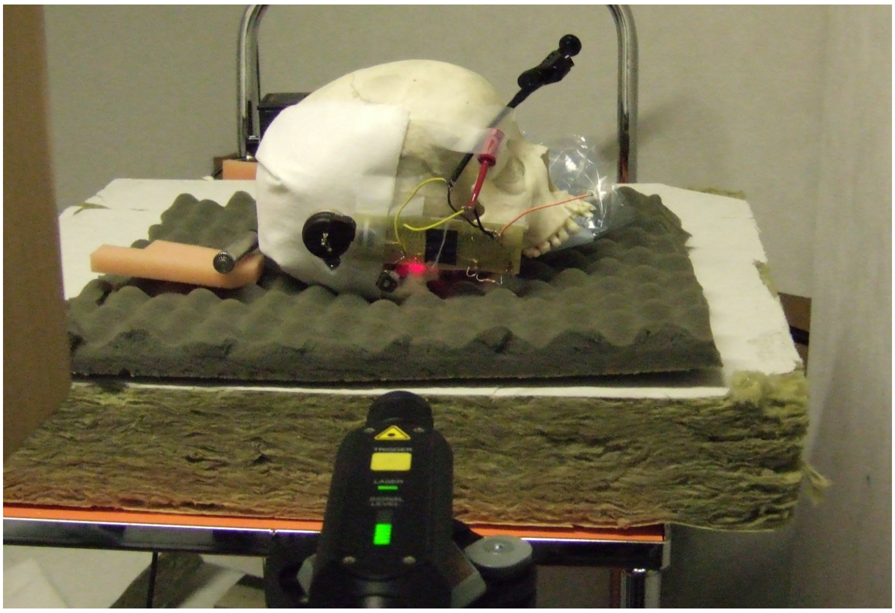

Figure 10. The experiment set-up to measure the acceleration in the promontorium, caused by the BAHA and MED-EL systems, which transforms the sound from the speaker (left in the picture) to vibrational energy in the skull.

2.7 Additional sound equipment

Apart from the equipment mentioned above, the use of some other equipment is worth mentioning. When the Agilent was used as a source it had problem to supply the speakers with enough power at higher sound pressure levels. Because of that, a normal sound amplifier was used; namely ROTEL Six Channel Power Amplifier RB-976MkII.

When performing measurements on the skull, the anechoic chamber was not large enough to hold the hearing laser vibrometer. A sound insulated room was used instead. The sound from the surrounding was insulated so that ambient noise of about 40 dB SPL never was exceeded. Instead of the speaker in the anechoic chamber, another speaker had to be used; HECO Odeon 100.

3 Devices under test

3.1 The Baha® Classic 300

In this thesis Baha® Classic 300 are referred to as BAHA, which is an acronym of Bone Anchored Hearing Aid. The BAHA system serves as a reference, to compare the result from the real device under test, the MED-EL system together with a BEST transducer. Good knowledge, the long clinical experience of the BAHA and the fact that the two systems are planned to help the same group of patients makes BAHA to an almost perfect reference. This system consists of a microphone, an amplifier supplied by a battery and a transducer which is snapped into a percutaneous titanium screw fixed to the parietal bone. Major drawbacks are related to the perforation of cutis and the fact that the microphone is close to the transducer, which can cause acoustic feedback.

Figure 11. The BAHA Classic 300 system. In the left picture the front side of the BAHA is shown, in the middle picture the back side with the snap coupling and in the right figure [7] the generic design of the percutaneous system.

3.2 The BEST

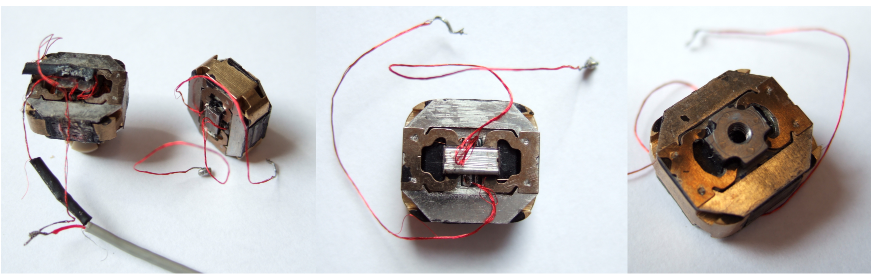

Figure 12. The BEST transducer from different angles.

The Balance Electromagnetic Separation Transducer, BEST, contains two main parts. The middle part, see Figure 13, is a paramagnetic body with a coil around it which is attached to, and transfers vibrations to, the load i.e. the skull bone. The second part surrounds the middle part and contains ferromagnets and paramagnets to create and conduct the magnetic flux. It also works as counterweight with a mass, m. The two parts are connected to each other with blade springs with compliance, C. The current in the coil produces an electromagnetic field which interacts with the magnetic field and creates a force between the two parts. Due to Newton’s second law the force gives rise to acceleration and in this case of interest, vibrations in the skull bone. The springs’s task is to hold the surrounding part in place around the middle part to prevent the air gap from collapsing.

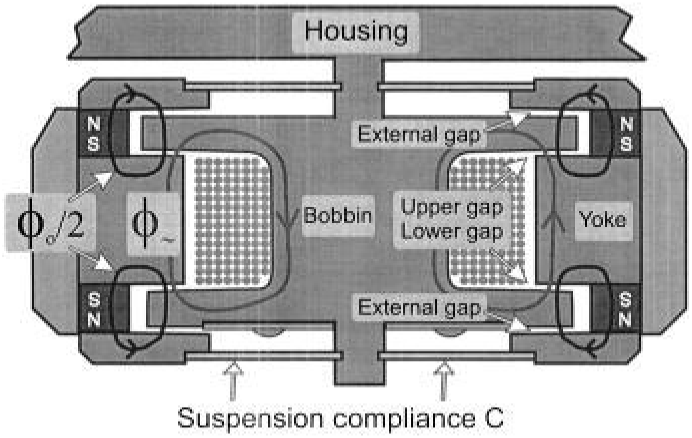

Figure 13. The principle sketch of the BEST transducer. [8]

The stiffness of the springs reduces the response in the low frequency range just like a high pass filter. The construction with double air gaps keeps the air gaps’ total size constant. In normal cases one would like to have stiff springs to avoid distortion. However, this construction will allow the use of softer springs and at the same time still have distortion on an acceptable level. Softer springs allow the use of a smaller counterweight which means that the entire construction gets smaller.[8]

A model is also developed to describe the performance of the BEST, which is helpful for understanding how the different parameters such as the masses and springs affect the output force, see Figure 14 below. [9]

In this thesis three different BEST transducers have been used. They are quite similar to each other; the main difference between them is the thickness the wire in the coil is. The three transducers were; BEST70, BEST90 and BEST150 with wire diameters of 70, 90 and 150 µm respectively.

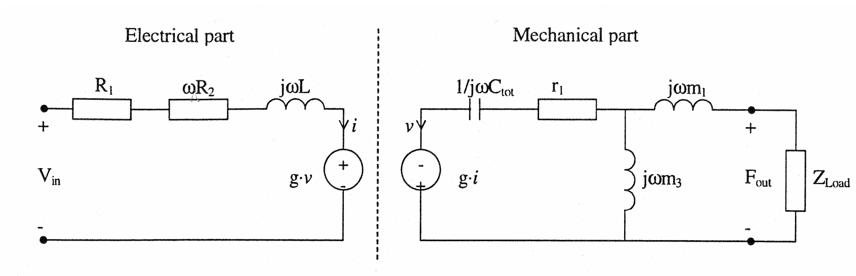

Figure 14. Electrical model describing the BEST transducer. In the model: R_1 is the DC resistance in the coil, R_2 is the losses due to magnetic hysteresis and eddy currents, L is the inductance of the coil, g is a constant that describe the electromechanical transmission between the two parts of the transducer, C_{tot} is the total compliance of the blade springs (in the text denoted as C), r_1 is the damping of the springs, m_3 is the counterweight (in the text denoted as m), m_1 is the mass of the middle part together with the mass of the coupling to the load, Z_{load} is the effective mass of the load (in the skullsimulator that mass is 50 g). The voltage over Z_{load} is the output force and the current through Z_{load} is the velocity of the load. [9]

3.3 MED-EL's audio processor, inductive link and envelope detector

MED-EL Vibrant® Soundbridge® is a middle ear implant with transcutaneous energy transmission consisting of an inductive link.

During the measurements a MED-EL Vibrant® Soundbridge®, without the floating mass transducer, i.e. middle ear transducer was used. The audio processor (AP), the inductive link and the envelope detector described here can be used to drive any electromagnetic transducer. In this thesis MED-EL Vibrant® Soundbridge® without the floating mass transducer are referred to as MED-EL, MED-EL system or inductive link.

Figure 15. To the left is the audio processor, AP, with a microphone and a transmitting coil. To the right are the receiver coil and the envelope detector. The two coils are together called the inductive link. Below the receiver coil is the floating mass transducer can be seen, which in this thesis is replaced with a BEST [10].

For an overview of the system see Figure 2. The sound is captured with a microphone in the AP (Figure 15) and transformed into an electric signal. The AP then creates a current in the transmitting coil which is an amplitude modulated version of the electric signal. As carrier frequency is either the resonance frequency or a multiple of the resonance frequency of the circuit usually chosen. The current in the coil creates an electromagnetic flux through soft tissues to an implanted receiving coil. In the implanted coil the electromagnetic flux gives rise to a current which is demodulated in the envelope detector (Figure 16). The signal from the envelope detector can then be used to drive either the floating mass transducer or, as in this thesis, an implanted BEST.

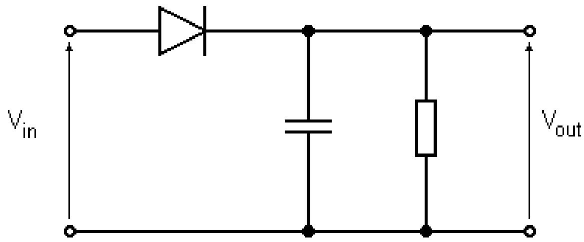

Figure 16. The principle schematic of an envelope detector. v_{in} is the incoming signal from the implanted coil and v_{out} is the outgoing signal which will drive the transducer.

4 Measurements

4.1 Output impedance of MED-EL's inductive link

The output impedance from MED-EL's inductive link was measured in two steps using the set-up in Figure 17. The source voltage, v_s, the voltage over the resistor, v_R, the voltage over BEST90 with the skullsimulator as load, v_Z, and the voltage over the inductive link, v_{il}, were all measured with the Agilent.

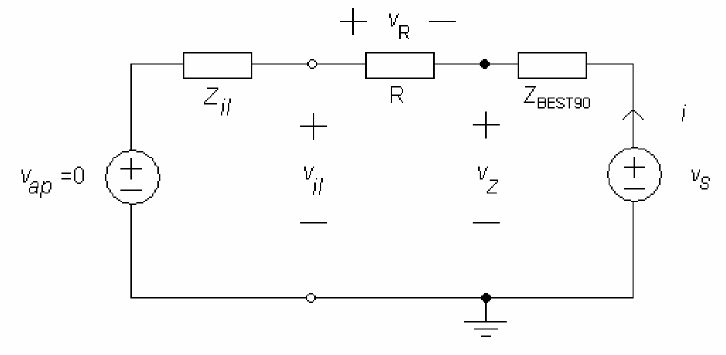

Figure 17. Measurement set-up for measuring the impedance of the inductive link. Two sets of measurements were done, one with varying *R* and constant v_s and one without Z_{BEST90} and constant R but varying v_s.

The BEST90 transducer was placed in series with a resistor *R* to simulate working conditions for the inductive link. As mentioned the measurements were carried out in two steps. First the resistor, *R*, was varied between some values to see if the impedance of the inductive link, Z_{il} depends on the load. The calculation to derive Z_{il} was done with the Agilent's built-in math functions and the expression used was

$$Z_{il} = \frac{v_{il}}{i} = \frac{v_{il}}{v_R/R} = \frac{v_{il}}{v_{il} - v_Z} R = \frac{v_{il}/v_s}{v_{il}/v_s - v_Z/v_s} R \tag{1}$$

In the second measuring step the source voltage was altered and the transducer was also removed. The resistance, *R*, was kept constant. The expression to calculate Z_{il} was also slightly changed

$$Z_{il} = \frac{v_{il}}{i} = \frac{v_{il}}{v_R/R} = \frac{v_{il}}{v_R} R \tag{2}$$

4.2 Impedance of the BEST70, BEST90 and BEST150

The measurements of the electrical impedance of the different transducers with different coil wire diameter were made using the set-up below

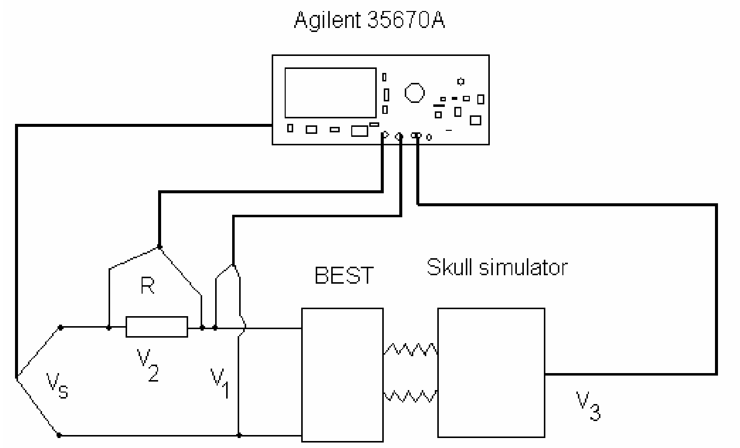

Figure 18. Measurement set-up for measuring the impedance of BEST70, BEST90 and BEST150. Also the set-up for measuring the force output from the BEST70 and BEST90 transducers.

The transducer was mounted onto the skullsimulator to mimic the true load. In series there was a resistor, denoted *R*, connected. To this, the source was connected and swept with a sine wave with constant voltage over each frequency. To calculate the impedance of the transducer, the following expression was used

$$Z_{BEST} = \frac{v_1}{v_2/R} = \frac{v_1}{v_2} R \tag{3}$$

where u_{BEST} (or v_1) is the voltage over the transducer and i_{BEST} is the current through the transducer. The current is calculated by measuring the voltage, v_2, over the resistor.

The inductive impedance of each BEST was also calculated using

$$L_{BEST} = \frac{\mathrm{Im}\{Z_{BEST}(f = 5\ kHz)\}}{2\pi f} \tag{4}$$

The reason for using the frequency 5 kHz is that it is far above the resonance frequency of the BEST.

4.3 Frequency response function for BEST70 and BEST90

During the impedance measurement above, the signal from the skullsimulator was captured at the same time. That made it possible to calculate a frequency response function. Two different frequency response functions were calculated. One as the force output from the skullsimulator divided by the current flowing through the transducer, i.e. BEST70 or BEST90 and the second as the force output divided by the voltage over the transducer. The expressions for the frequency response functions are

$$FRF_i = \frac{F_{BEST}}{i_{BEST}} = \frac{k_F v_3}{v_2 / R} = k_F R \frac{v_3}{v_2} \quad \text{, and} \tag{5}$$

$$FRF_u = \frac{F_{BEST}}{u_{BEST}} = k_F \frac{v_3}{v_1} \tag{6}$$

where k_F is the conversion factor to convert the voltage from the skullsimulator into a force. The quotients, v_3/v_2 and v_3/v_1, were measured with the signal analyzer.

4.4 Force output from MED-EL with BEST and BAHA

The transducers with their respective driving units were mounted onto the skullsimulator and placed inside the anechoic chamber. Inside the anechoic chamber a condenser microphone was also placed. A sweeping sine wave was then sent to the built-in speaker in the chamber from the Agilent. The sound was captured by the microphone and sent back to the signal analyzer. By doing so, it was possible to keep the sound level at a constant desirable level. The levels used in all sweeps were 50 dB SPL up to 90 dB SPL with 5 dB intervals. For each sweep, the signal from the skullsimulator and the microphone was collected with the Agilent. The force produced by the transducer was calculated by multiplying the signal from the skullsimulator with a conversion factor. The procedure described above was carried out for both the BAHA system and the MED-EL system, the latter with both BEST70 and BEST90 attached respectively.

During the measurements of the force output from the transducers both the force and the sound levels were saved. By dividing these signals with each other, the frequency response function, *FRF*, for each system was calculated.

$$FRF = \frac{Output\ force\,level}{Sound\ pressure} = \frac{F_{out}}{SP_{in}} \tag{7}$$

4.5 Force output with modified impedance

These measurements were performed in the same way as for the force measurements described above. However, a resistor parallel connected with a capacitor was put in series with the BEST90 transducer, see Figure 19. The number of sound levels, on which measurements was carried out was also reduced.

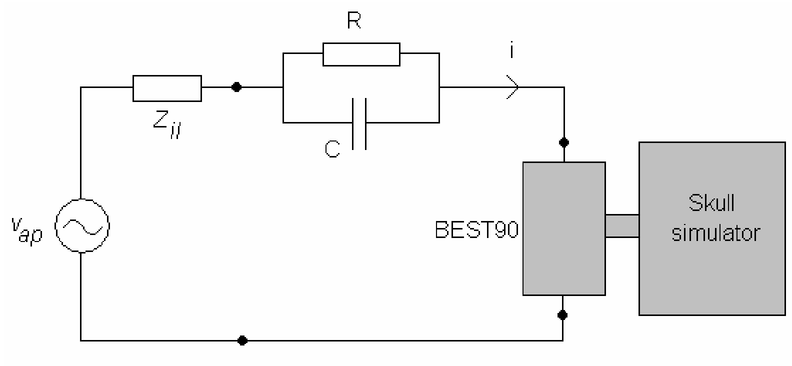

Figure 19. An impedance network consisting of a resistor and a capacitor in series with the BEST90 transducer to try to increase the force output at higher frequencies.

The R and C should be selected so that the current through BEST90 increases in a desired frequency range without, at the same time, decreasing the current outside that frequency range to much. To increase the current through BEST90, the magnitude of the total impedance of the circuit, shown in Figure 19, should be decreased. The problem is that the impedance of the inductive link depends on the rest of the circuit connected to it. However, to get some guidance, the impedance of the inductive link is approximated to be only resistive with a magnitude of 65 Ω.

In Figure 20 the calculated magnitude of the total impedance (of the inductive link, the impedance network and the BEST) of the circuit shown in Figure 19 is shown. According to it a good choice is R = 200 Ω to get an extra gain at 2 kHz. It should be remembered that this calculation is an approximation and as seen later in the results section it is not good to chose R = 200 Ω to get extra gain at 2 kHz. Measurements were also done with R = 20 Ω which should result in worse gain, but it turns out that this is not the case. However the calculations give a hint in what range R should be in.

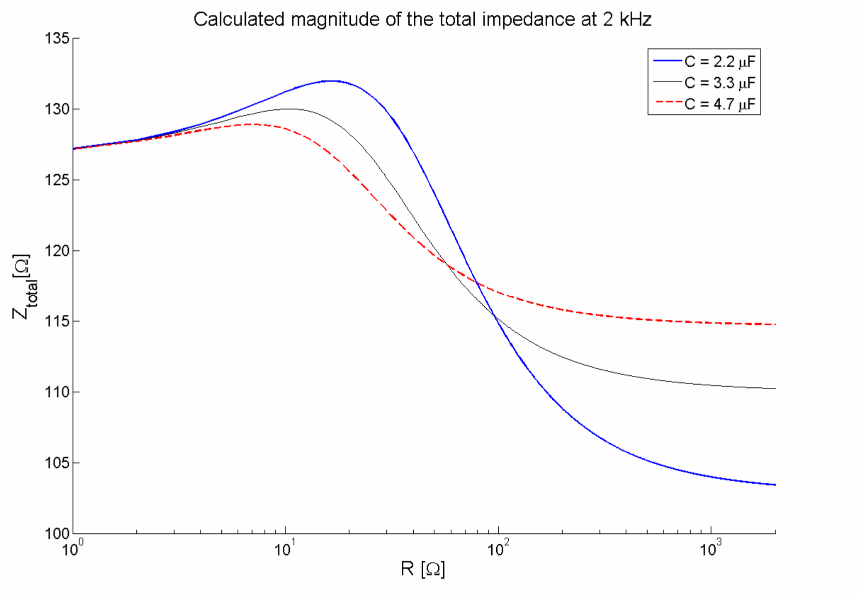

Figure 20. Calculated magnitude of the total impedance with different R at 2 kHz. In the calculations the impedance of the inductive link is approximated to be only resistive with a magnitude of 65 Ω.

4.6 Acceleration on a dry skull

The measurements were carried out inside a room designed to reduce acoustic noise. A dry skull was placed onto a soft material to further reduce vibrations. Onto the skull the transducers and driving systems were placed on their designated places. The sound from a speaker was used to generate sound waves which, through the hearing systems were transformed into vibrations in the skull. The speaker was hanged in wires from the ceiling to avoid vibrations to propagate to the measuring system. A laser doppler vibrometer specially designed to measure velocity of moving objects where used to detect these vibrations. A condenser microphone was also placed close to the driving systems' microphones and connected to the Agilent. By keeping the signal from the microphone at a constant level, it was possible to measure the velocity in the dry skull caused by a specific sound level. The acceleration was then calculated by multiplying the measured velocity obtained by the laser, v_{rms}, with $j\omega$ according to the equations below.

$$\begin{aligned} & v(t) = v_{rms}\sqrt{2}\exp\{j\omega t\} \\ & \Rightarrow a(t) = \dot{v}(t) = v_{rms}\sqrt{2}\, j\omega\exp\{j\omega t\} \\ & \Rightarrow a_{rms}(j\omega) = j\omega v_{rms} \end{aligned} \tag{8}$$

The measurements were carried out twice, once with MED-EL's system with BEST90 at position 4 and once with BAHA on position 1. To be able to mount BEST90 to its desired position, an extension peg had to be used. See Figure 9 for where each position is placed.

4.7 Acceleration on a semi dry skull

The same measurement procedure was used as for the measurements on the dry skull. The only thing that did differ was the fact that the skull had been filled with a water solution of gelatine. This was done to mimic a real skull with a brain, which alters the mechanical properties of the skull.

Exactly as with the dry skull the measurement was carried out with BAHA on position 1 and MED-EL with BEST90 on position 4. MED-EL with BEST90 was also placed at position 1 to be able to compare the relative gain or attenuation by moving the stimulation point closer to the cochlea.

4.8 Distortion from MED-EL with BEST and BAHA

Distortion of the BAHA and BEST90 transducers was measured both with the skullsimulator and with the real skull filled with gelatine as load. To be able to measure the distortion, the FFT mode on the Agilent was used, with a flat top weight function. While this mode is selected, it is possible to use a fixed sine signal as output from the Agilent. This output signal was sent to a loudspeaker and the sound level was measured with the condenser microphone. In the case of measurements on the skull, the laser registered the vibrations in the skull caused by the transducers. In the case of measurements on the skullsimulator, the force was directly measured with the signal analyzer as a voltage. The sound levels were chosen so that the output levels were 10 dB below the saturation levels for the MED-EL system (the same sound levels were used when measuring on the BAHA system). The fundamental frequencies studied were f_1 = 500, 750, 1000, 1500, 2000 and 3000 Hz to be able to compare with earlier studies [8]. The distortion calculated is then the sum of the powers of harmonics of the main frequency divided by the power of the main frequency itself; this is also called total harmonic distortion, *THD*.

$$THD = \frac{\sum_{m=2}^{n} |u(mf_1)|^2}{|u(f_1)|^2} \tag{9}$$

Where n decides the total numbers of harmonics included. While measuring on the skull, the THD was calculated from the velocity output, and when measuring on the skullsimulator the THD was calculated from the force output. Those two measurements cannot be directly compared since the force is proportional to the acceleration, which is the derivative of the velocity, if the mass can be assumed to be independent of the frequency. In Figure 21, a typical appearance of harmonic distortion is shown.

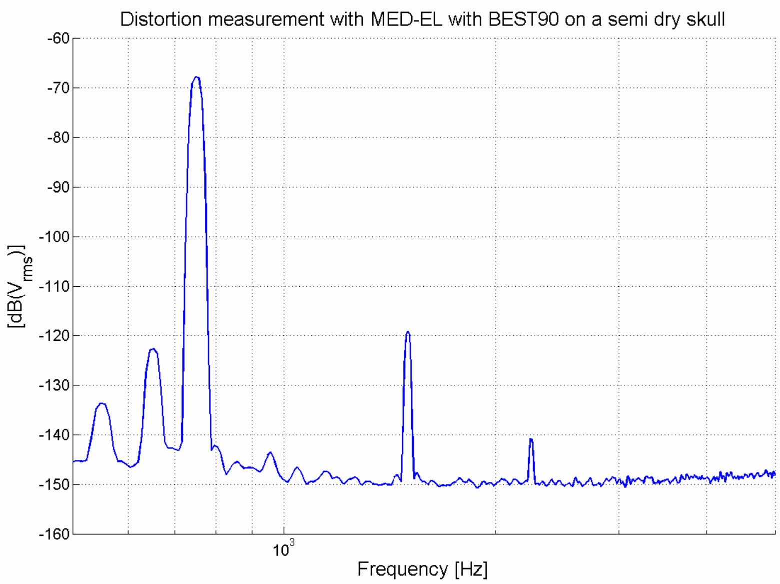

Figure 21. The velocity has been measured with a fundamental frequency of 750 Hz. The harmonics at 1500 Hz and 2250 Hz is clearly visible. The two peaks below 750 Hz are artefacts from the hearing laser vibrometer.

5 Results

To better visualize the results all data in the figures in this section have been slightly smoothed.

5.1 Output impedance of MED-EL's inductive link

The output impedance of MED-EL's inductive link was measured as described in the measurement section. In Figure 22 and Figure 23 the load is varied between different values. In Figure 24 and Figure 25 the load is kept constant and the source level is varied.

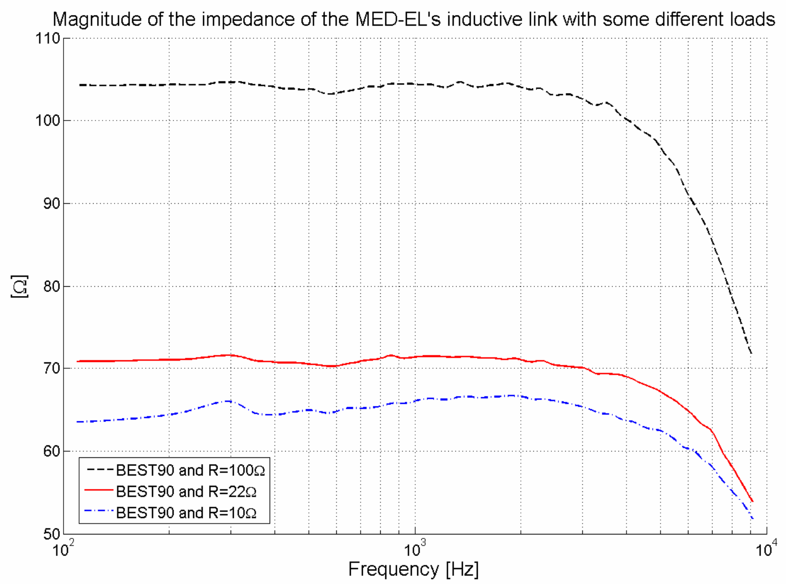

Figure 22. The magnitude of the output impedance of MED-EL's inductive link plotted for different loads.

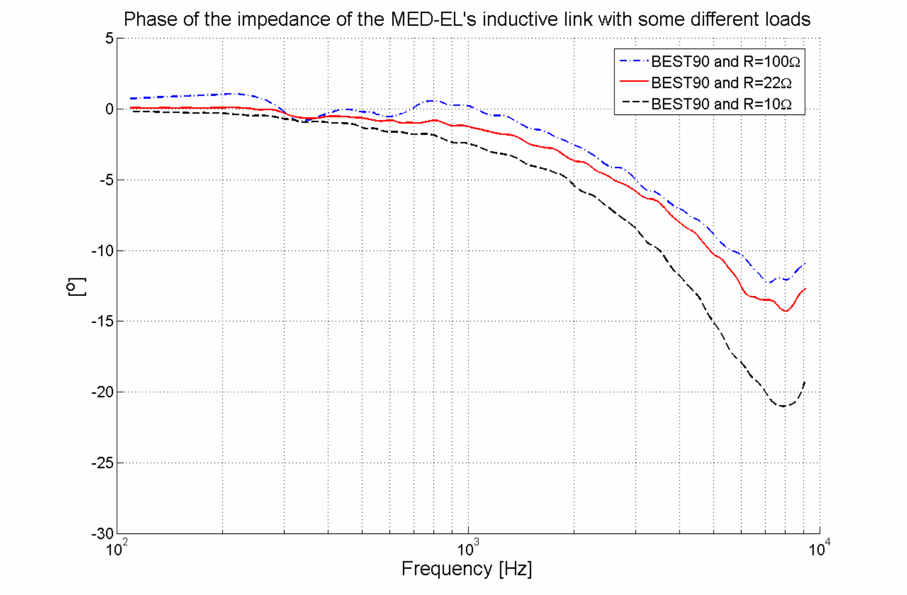

Figure 23. The phase in degrees of the output impedance of MED-EL's inductive link plotted for different loads.

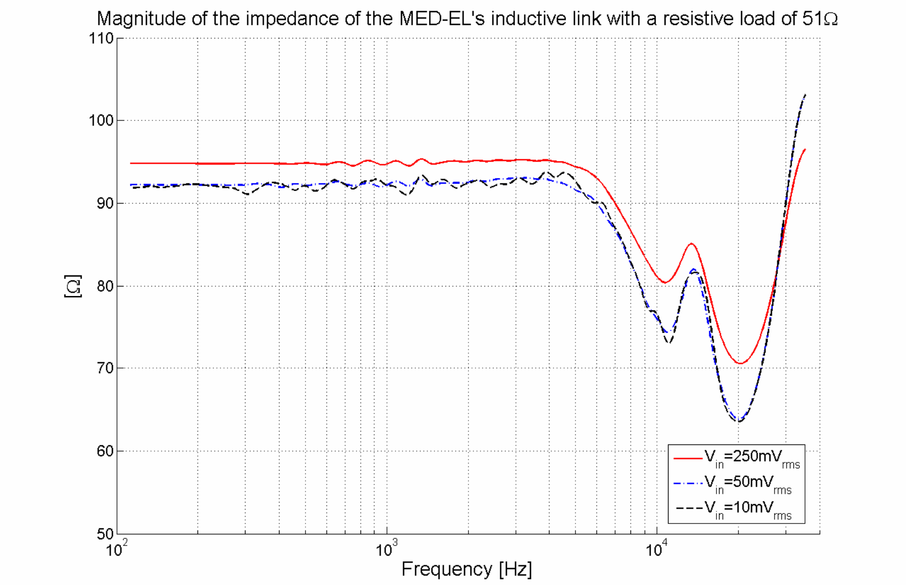

Figure 24. The magnitude of the output impedance of MED-EL's inductive link plotted for different source voltages.

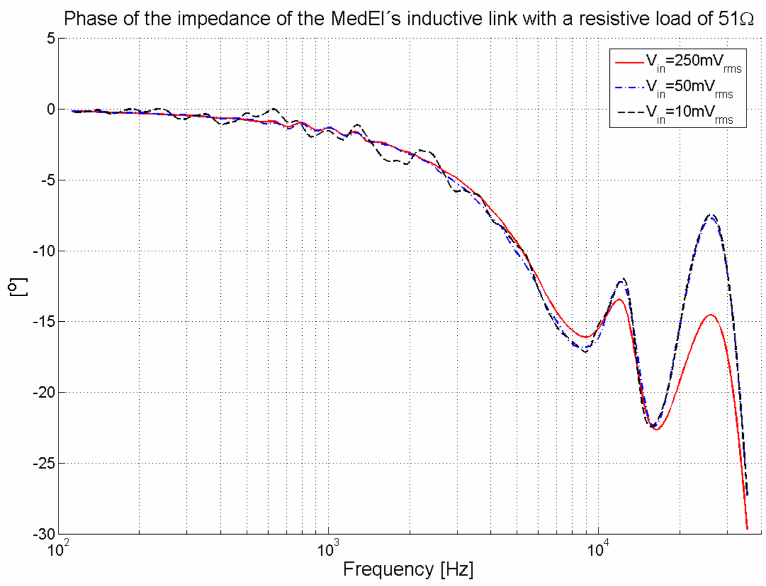

Figure 25. The phase of the output impedance of MED-EL's inductive link plotted for different source voltages.

5.2 Impedance of the BEST70, BEST90 and BEST150

In the following three figures below the impedance of BEST70, BEST90 and BEST150 transducers is plotted with magnitudes, phases, real parts and imaginary parts. In Figure 29 the magnitude for each of the transducers are plotted in the same graph.

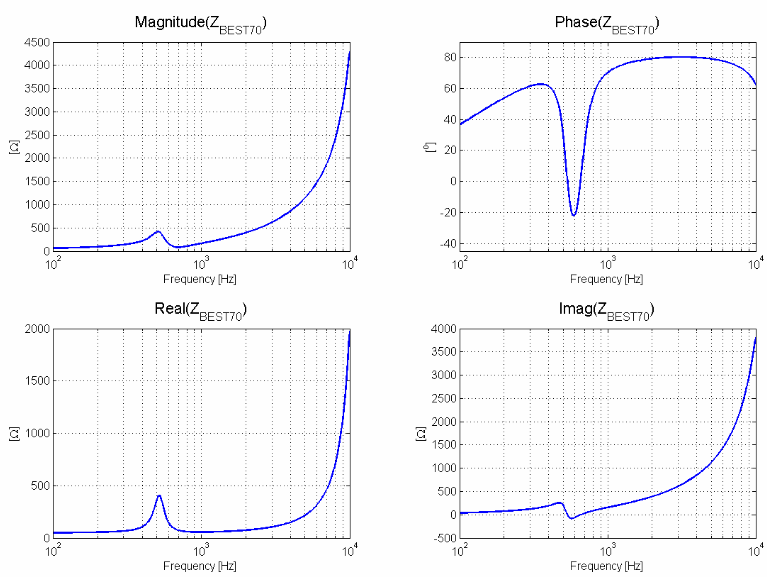

Figure 26. The impedance of BEST70.

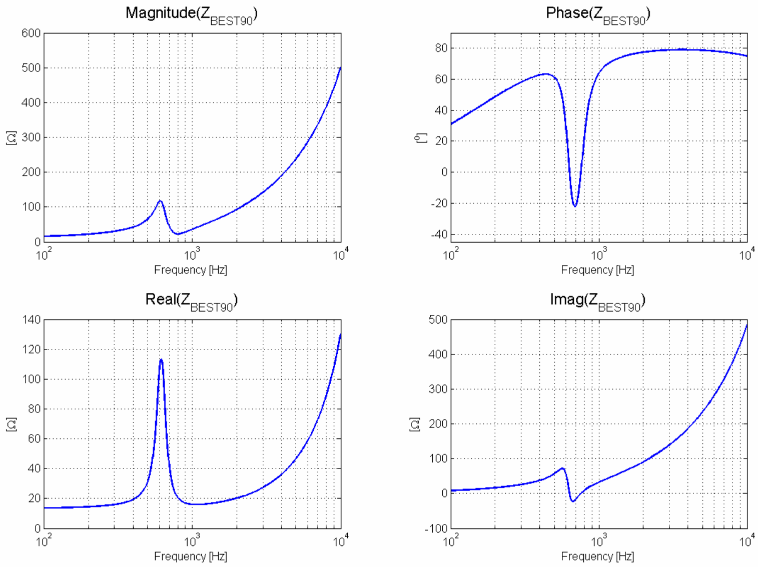

Figure 27. The impedance of BEST90.

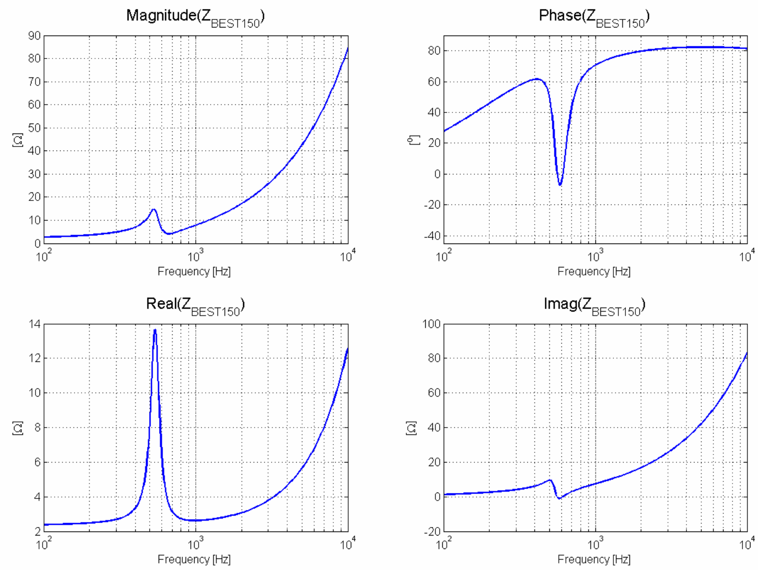

Figure 28. The impedance of BEST150.

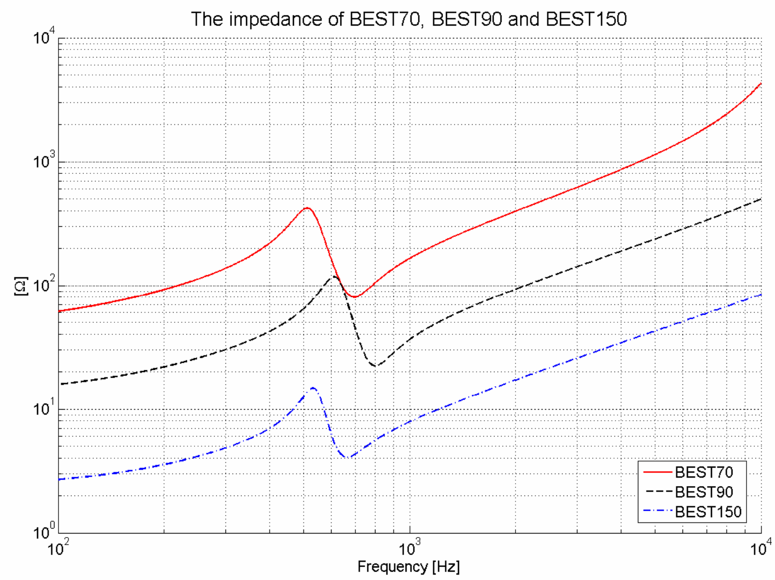

Figure 29. The magnitude of the impedance for BEST70, BEST90 and BEST150.

The inductance for each transducer was calculated as follow:
$L_{BEST70} = 35\ mH$, $L_{BEST90} = 7.4\ mH$ and $L_{BEST150} = 13\ \mu H$.

5.3 Frequency response functions for BEST70 and BEST90

The two frequency response functions were calculated and the result is plotted in the two figures below.

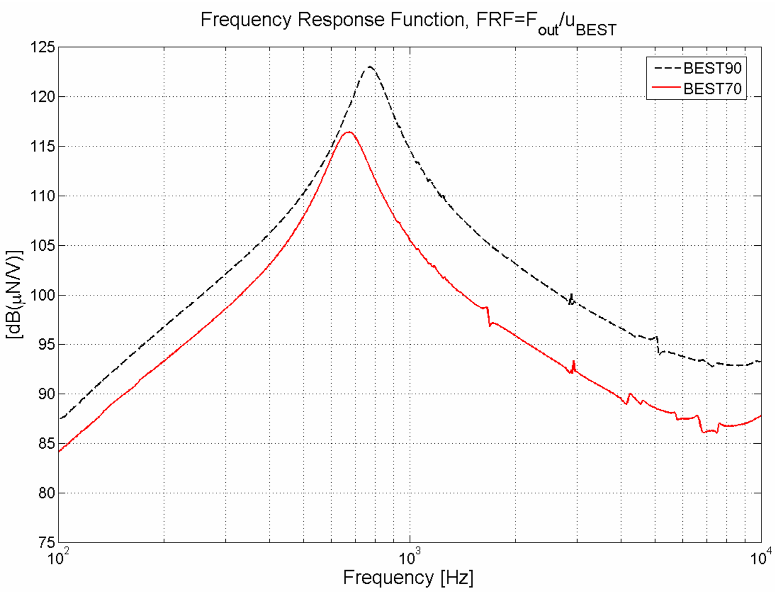

Figure 30. The force output from the BEST70 and BEST90 transducers connected to the skullsimulator divided by the voltage over the transducers, also called the voltage to force frequency response function, $FRF_u(\omega)$.

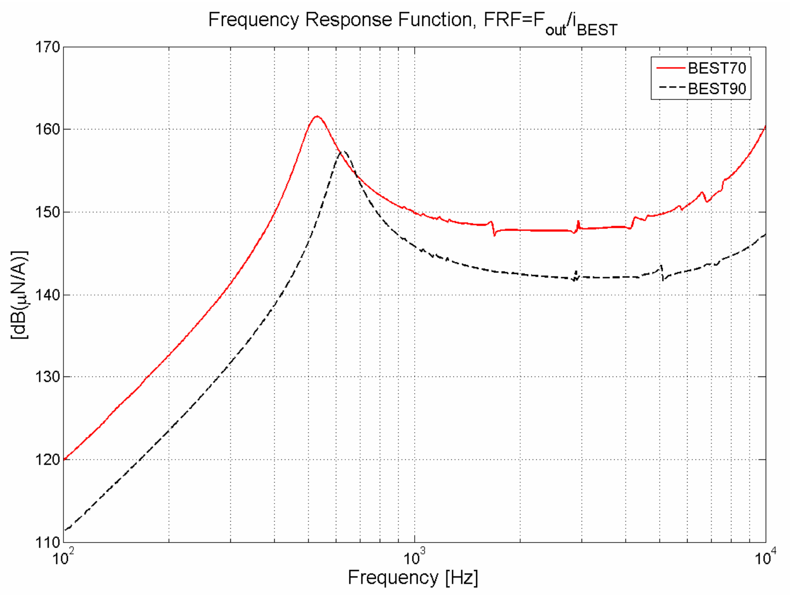

Figure 31. The force output from the BEST70 and BEST90 transducers connected to the skullsimulator divided by the current through the transducers, also called the current to force frequency response function, $FRF_i(\omega)$.

5.4 Force output from MED-EL with BEST and BAHA

The output force levels, OFL, from the BAHA Classic 300 transducer is plotted in Figure 32 and Figure 33 below. In the first plot with the BAHA the volume setting is set to 2 and in the second plot the volume setting is set to the maximum 3. There are several curves in each plot and in each curve the force is measured at a specific sound level ranging between 50 dB SPL and 90 dB SPL.

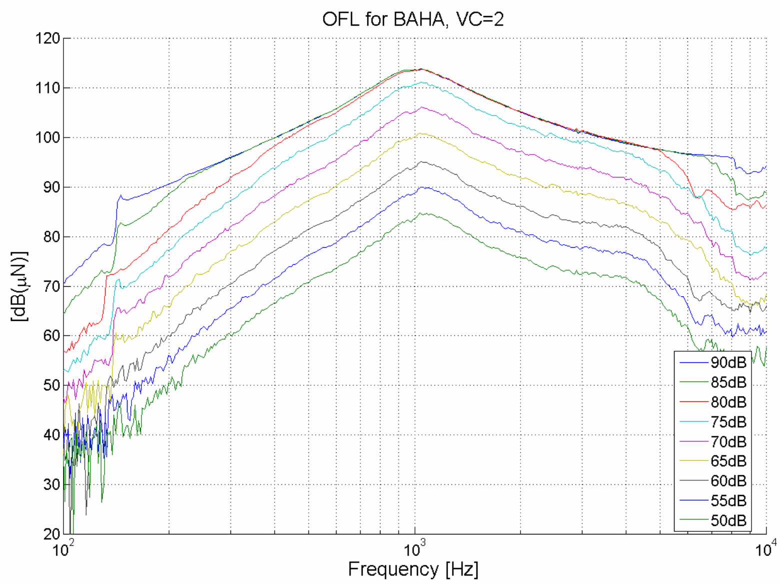

Figure 32. Output force level, OFL, for BAHA with volume setting 2 for nine different sound levels ranging between 50 dB SPL and 90 dB SPL.

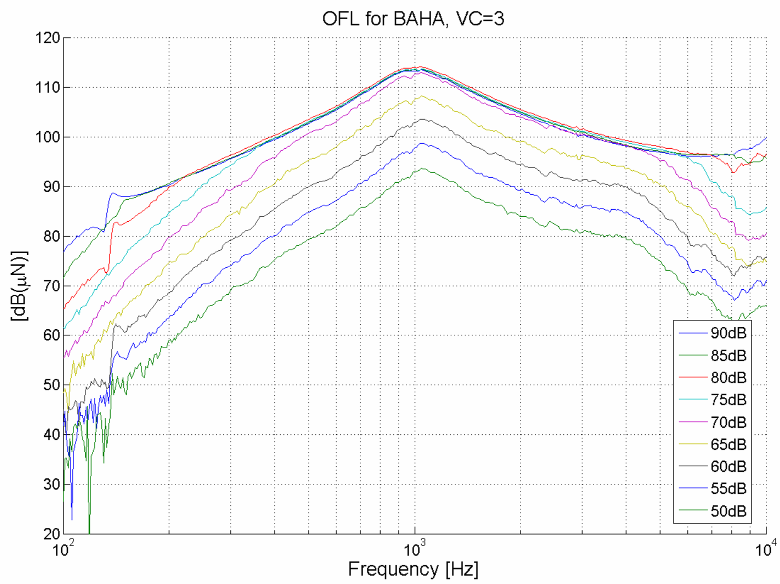

Figure 33. Output force level, OFL, for BAHA with volume setting 3 for nine different sound levels ranging between 50 dB SPL and 90 dB SPL.

In the following two figures the OFL for MED-EL with BEST70 and BEST90 respectively is plotted.

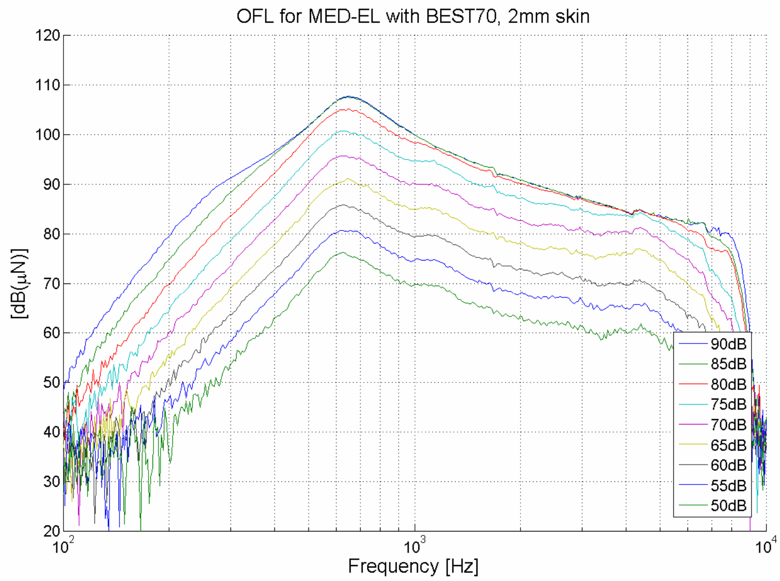

Figure 34. Output force level, OFL, for MED-EL with BEST70 for nine different sound levels ranging between 50 dB SPL and 90 dB SPL.

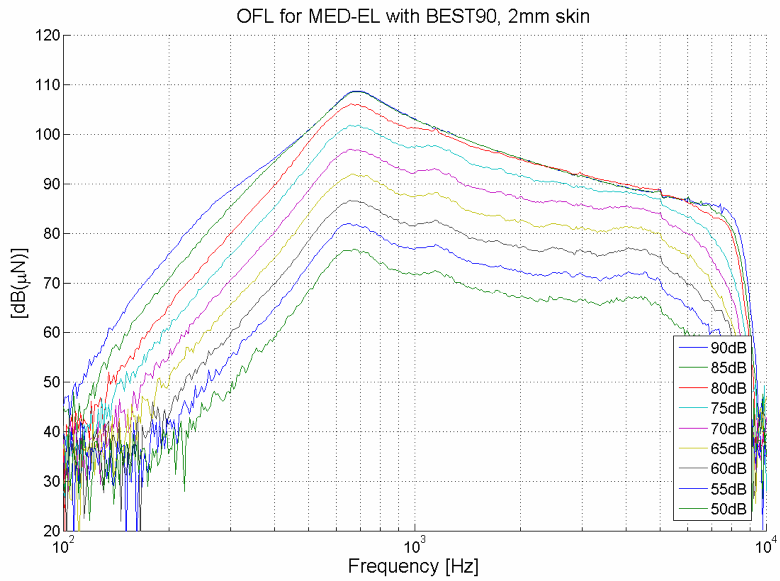

Figure 35. Output force level, OFL, for MED-EL with BEST90 for nine different sound levels ranging between 50 dB SPL and 90 dB SPL.

In Figure 36 and Figure 37 below the OFL for all systems is plotted in the same plot. In the first the OFL at 60 dB SPL is shown and in the second, the OFL at 90 dB SPL. The OFL at 90 dB SPL is also the maximum OFL.

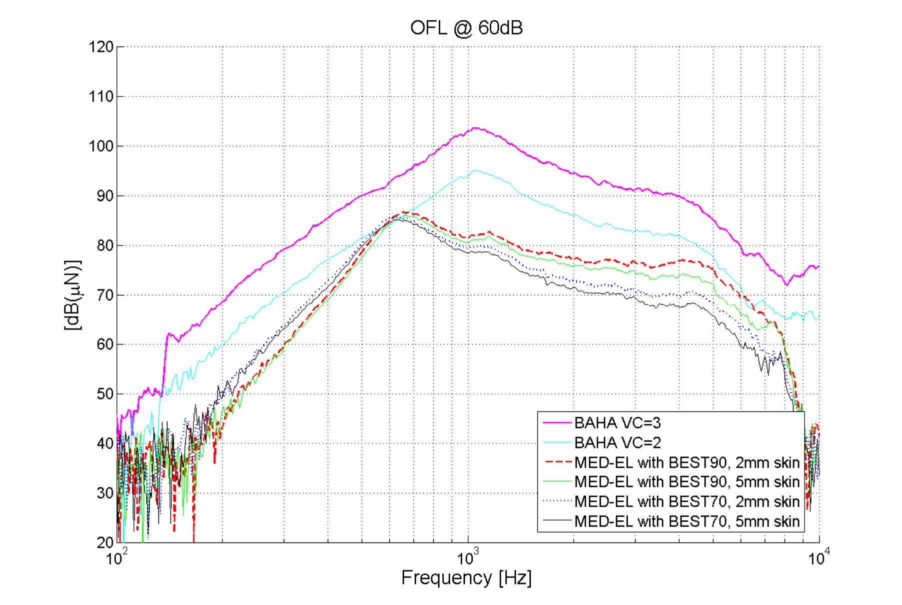

Figure 36. Output force level, OFL at 60 dB SPL for some different systems.

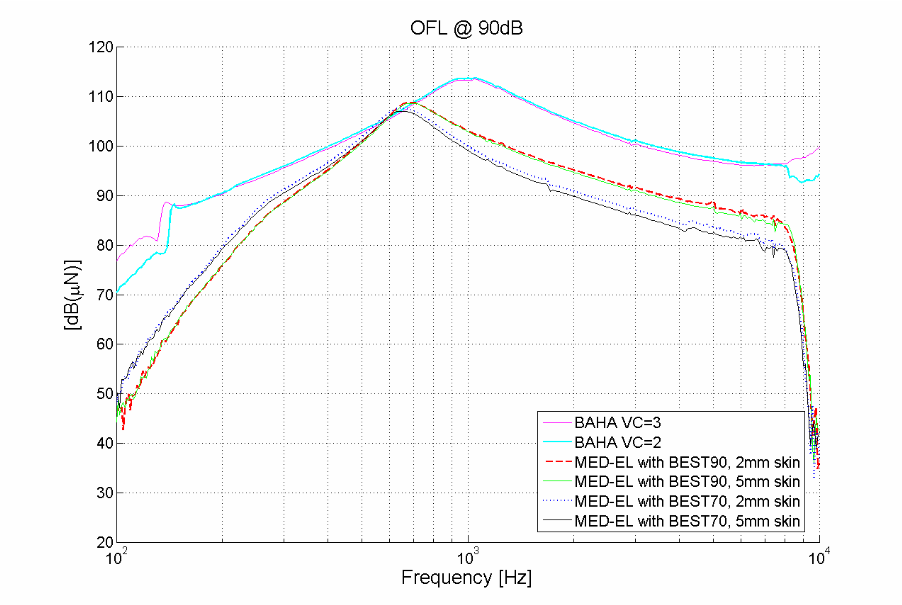

Figure 37. Output force level, OFL at 90 dB SPL for some different systems.

The frequency response function, FRF, is calculated and plotted in Figure 38 below for the different systems: BAHA with volume settings 2 and 3, and MED-EL with BEST70 and BEST90 respectively.

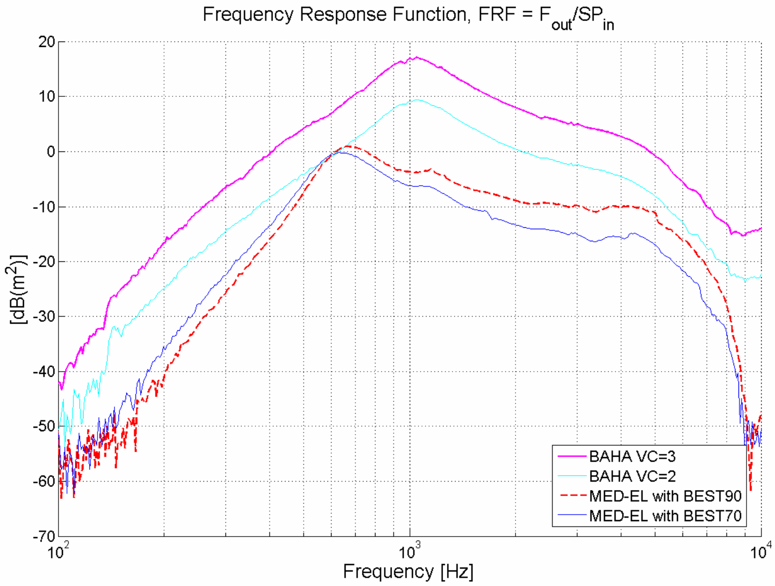

Figure 38. The frequency response function, FRF. For the BAHA system with volume setting on both 2 and 3. The FRF is also plotted for the MED-EL with BEST system with both BEST70 and BEST90.

5.5 Force output with modified impedance

As described in the measuring chapter above the output force level, OFL, is measured once again from the MED-EL with BEST90. The only difference is that impedance consisting of a resistor and a capacitor is placed in series with BEST90. The OFL at 90 dB SPL is shown in Figure 39.

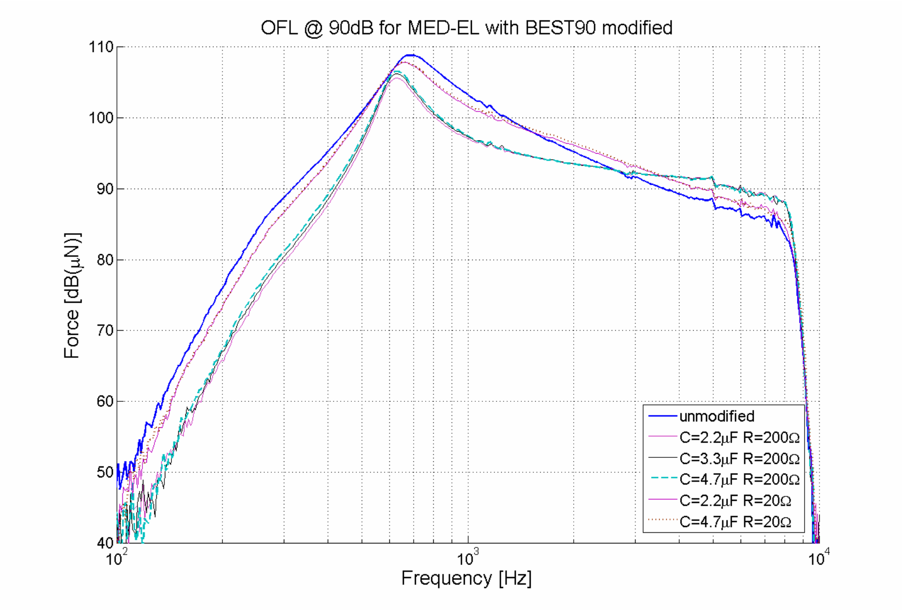

Figure 39. Output force level from the MED-EL with BEST90 system at 90dB SPL. The impedance of BEST90 is altered with another impedance in series; namely a resistor, R, parallel connected with a capacitor, C. The values of C and R are varied between some different values.

5.6 Acceleration on a dry skull

The acceleration was calculated from the data obtained by the hearing laser doppler vibrometer and the result was plotted in Figure 40 to Figure 44. Figure 40 shows the acceleration response at the promontory from the BAHA system at position 1 of the skull and Figure 41 shows the acceleration response from the MED-EL system attached at position 4.

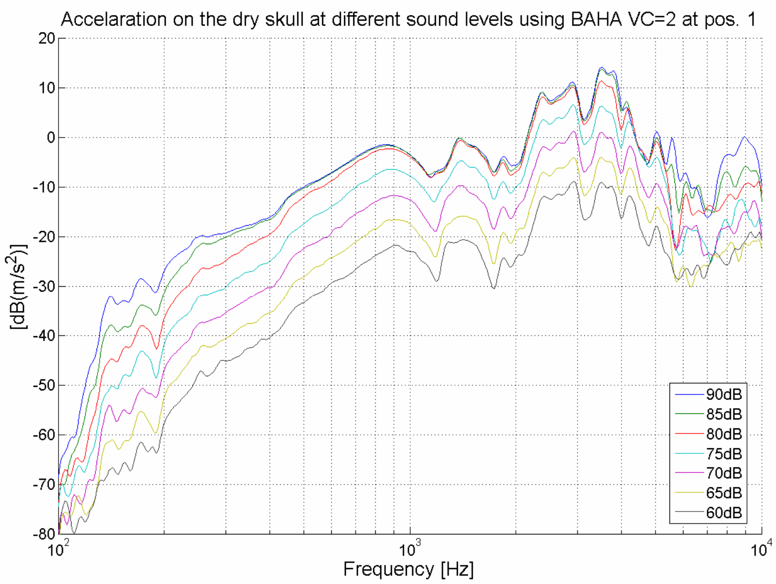

Figure 40. The acceleration on the dry skull with BAHA with volume setting 2 placed on position 1. The acceleration was plotted for seven different sound levels ranging from 60 dB SPL to 90 dB SPL.

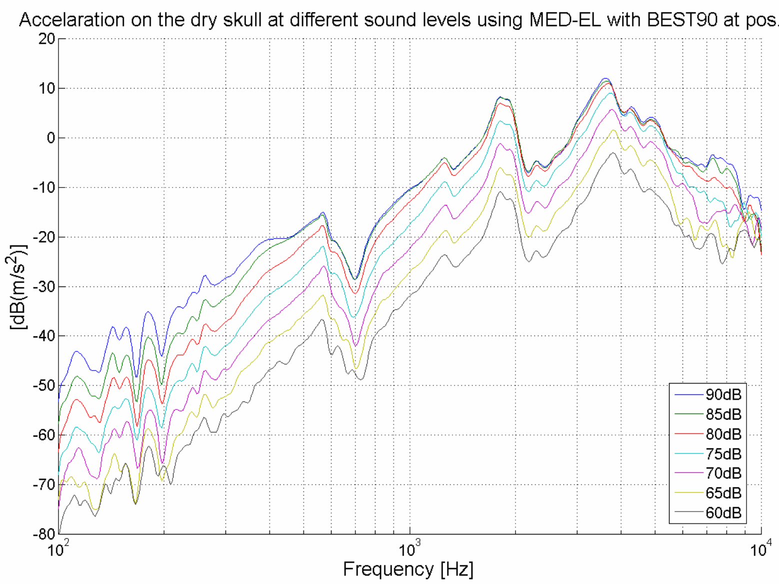

Figure 41. The acceleration on the dry skull with MED-EL's system with BEST90 placed on position 4. The acceleration was plotted for seven different sound levels ranging from 60 dB SPL to 90 dB SPL.

In Figure 42 and Figure 43 below, the acceleration is plotted for the BAHA system with both volume setting 2 and 3 at position 1, and the MED-EL system with BEST90 at position 4, all measured at the sound level of 90 dB SPL. Both figures contains the same information, however, the data in Figure 43 have been exposed to some more smoothing to better visualize the appearance of the acceleration.

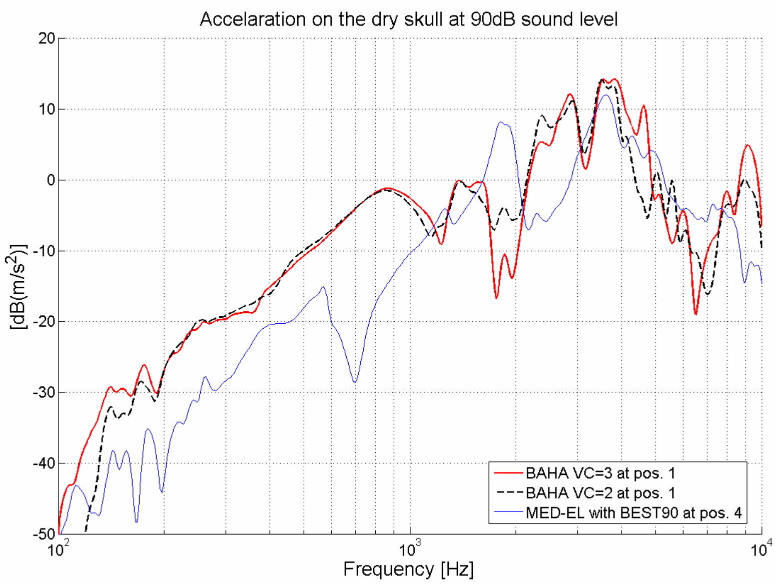

Figure 42. The acceleration is plotted for BAHA with volume setting 2 and 3 respectively at position 1 and MED-EL with BEST90 at position 4. All acceleration responses are obtained with constant sound level of 90 dB SPL.

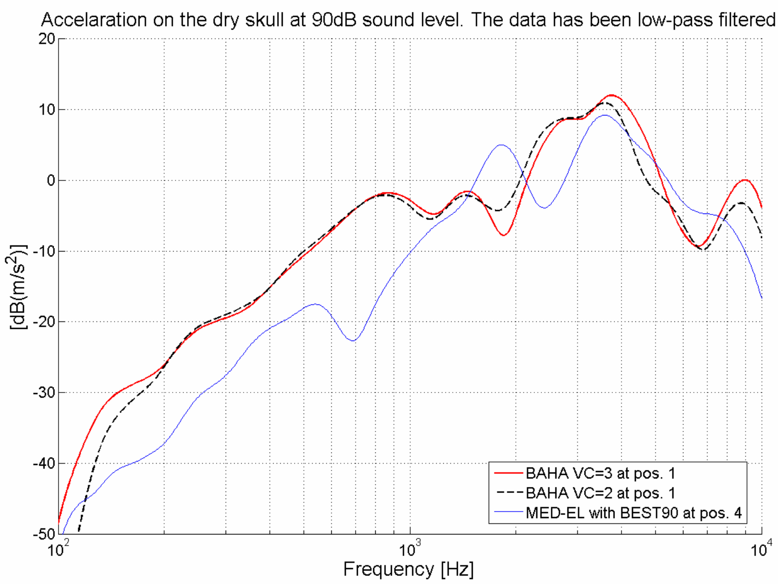

Figure 43. The acceleration is plotted for BAHA with volume setting 2 and 3 respectively at position 1 and MED-EL with BEST90 at position 4. All accelerations are plotted at a constant sound level of 90 dB SPL. The data have also been extra low-pass filtered.

The frequency response function, FRF, in acceleration over sound pressure for the different systems is plotted below in Figure 44.

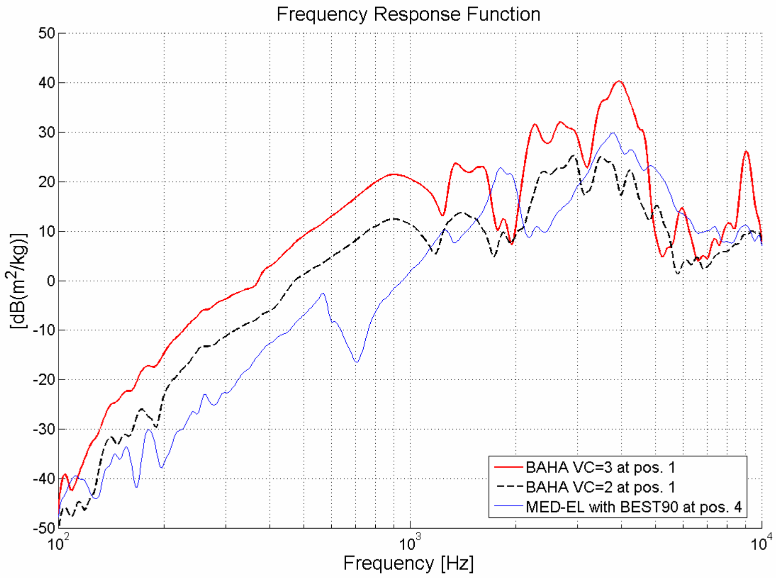

Figure 44. The frequency response function, FRF, is plotted for the three different adjustments: BAHA with volume setting 2 at position 1, BAHA with volume setting 3 at position 1 and MED-EL with BEST90 at position 4.

5.7 Acceleration on a semi dry skull

The acceleration was measured in the same way as on the dry skull, which now is filled with gelatine, and plotted below. The first two figures display the acceleration for the BAHA system and the MED-EL with BEST90 system, respectively.

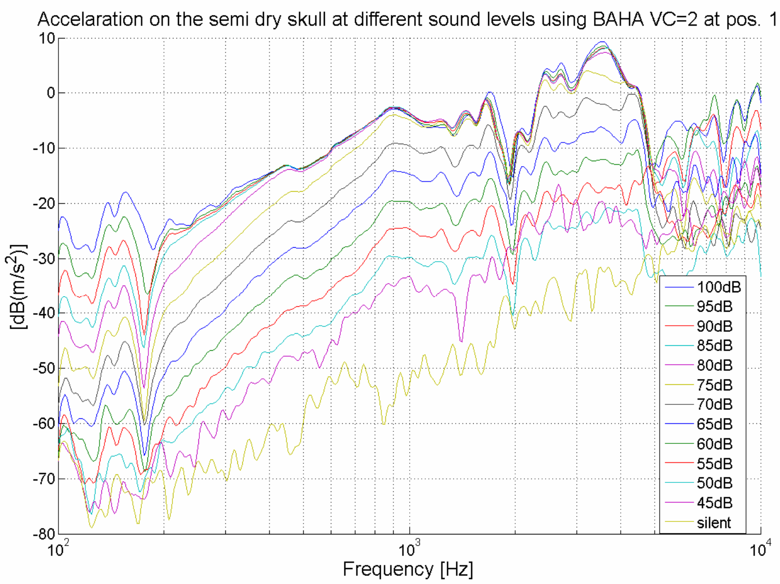

Figure 45. Acceleration on the semi dry skull using the BAHA at position 1 plotted for sound levels between 45 dB SPL and 100 dB SPL with 5 dB intervals. A measurement without any sound from the speaker was also made to find the noise floor from the driving system and laboratory environment.

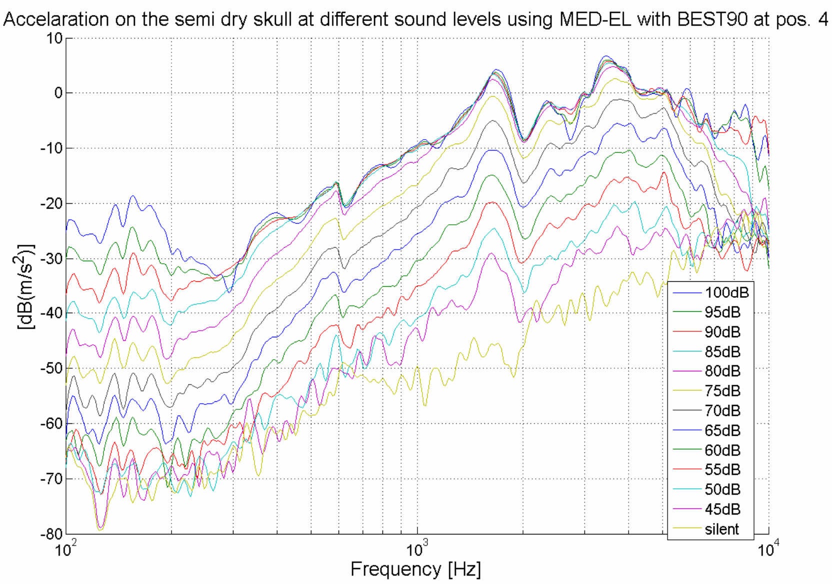

Figure 46. Acceleration on the semi dry skull using the MED-EL with BEST90 at position 4 plotted for sound levels between 45 dB SPL and 100 dB SPL with 5 dB intervals. A measurement without any sound from the speaker was also made to find the noise floor from the driving system and laboratory environment.

To be able to easier compare the results, the acceleration is plotted at 90 dB SPL for the BAHA system at position 1 and the MED-EL system with BEST90 at both position 1 and position 4 see Figure 47. The acceleration response without any transducers attached was also measured, to justify that all vibrations measured came from the transducer.

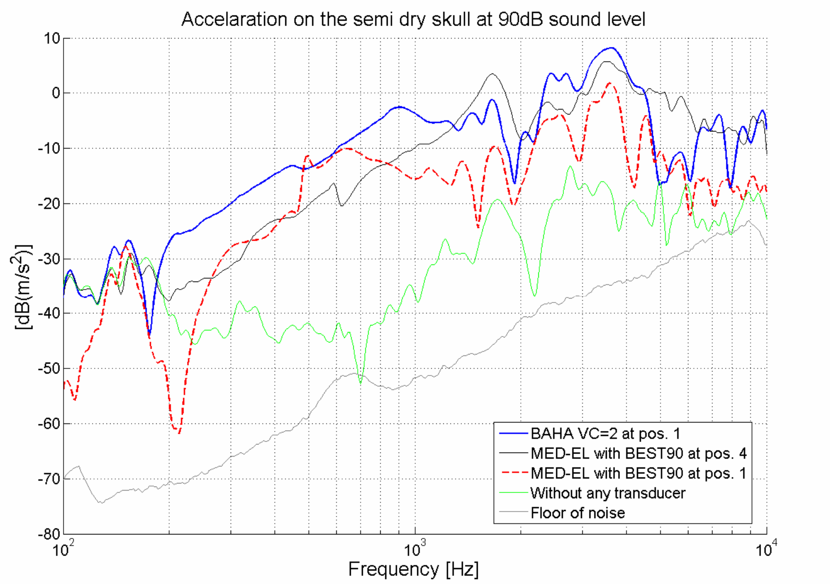

Figure 47. Acceleration measured at 90 dB SPL on the semi dry skull. The BAHA was positioned at position 1, the MED-EL with BEST90 was positioned at both position 1 and 4. The acceleration with only the speaker causing vibrations on the skull is also plotted.

Figure 48 below shows the frequency response function of the different systems, MED-EL with BEST90 at position 1 and at position 4 and BAHA at position 1.

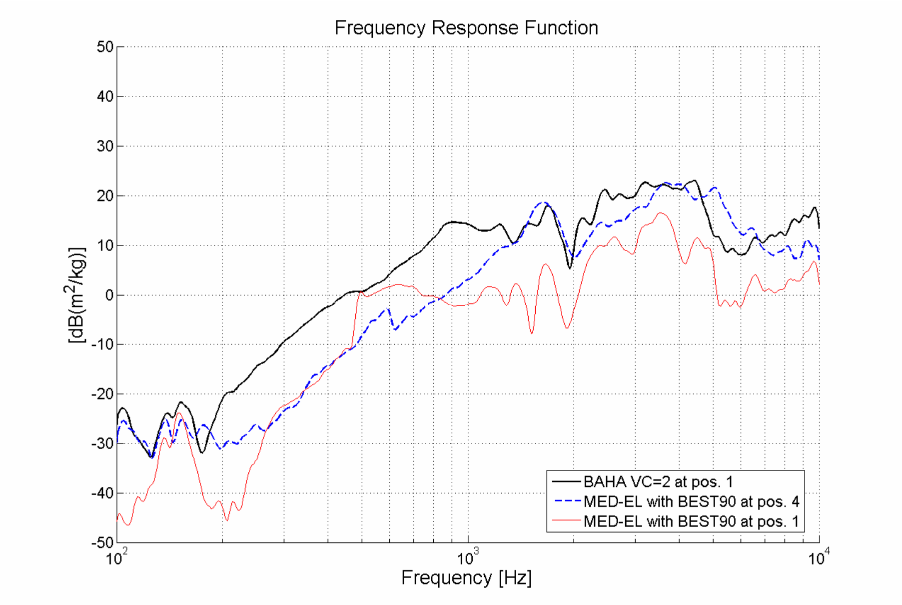

Figure 48. The frequency response function for the BAHA system at position 1 and the MED-EL with BEST90 at both position 1 and 4.

5.8 Distortion from MED-EL with BEST and BAHA

The distortion of each system was measured and the result is given in Table 1 to

Table 5.

Table 1. MED-EL with BEST90 on the semi dry skull. THD measured with different numbers of harmonics. THD is calculated from the velocity signal from the hearing laser doppler vibrometer.

Fundamental frequency	# harmonics	THD(%)
500	2	3.09
750	2	4.82
1000	2	1.72
1500	1	0.96
1500	2	1.05
2000	1	2.66
2000	2	2.94
3000	1	1.40
3000	2	1.67

Table 2. BAHA on the semi dry skull. THD is calculated from the velocity signal from the hearing laser vibrometer.

Fundamental frequency	THD(%)
500	0.00
750	0.00
1000	1.30
1500	1.48
2000	1.78
3000	1.24

Table 3. MED-EL with BEST90 placed onto the skullsimulator. The THD is calculated from the signal directly from the inductive link (the electric input to the BEST). Measurements with reduced number of harmonics were done due to an artifact at 10 kHz. The number of harmonics was then chosen to not include the artifact.

Fundamental frequency	# harmonics	THD(%)
500	10	0.63
750	10	4.05
1000	10	2.79
1000	8	2.40
1500	10	1.34
1500	5	1.26
2000	10	2.27
2000	5	2.01
3000	10	2.73
3000	5	2.62

Table 4. MED-EL with BEST90 placed onto the skullsimulator. The THD is calculated from the signal from the skullsimulator (the mechanical output of the BEST). Measurements with reduced number of harmonics were done due to an artifact at 10 kHz. The number of harmonics was then chosen to not include the artifact.

Fundamental frequency	# harmonics	THD(%)
500	10	0.96
750	10	0.63
1000	10	0.64
1000	8	0.63
1500	10	0.52
1500	5	0.50
2000	10	0.99
2000	5	0.93
3000	10	1.49
3000	5	1.41

Table 5. BAHA placed onto the skullsimulator. The THD is calculated from the signal from the skullsimulator. Measurements with reduced number of harmonics were done due to an artifact at 10 kHz. The number of harmonics was then chosen to not include the artifact.

Fundamental frequency	# harmonics	THD(%)
500	10	1.62
750	10	0.18
1000	8	0.20
1000	10	0.22
1500	5	0.30
1500	10	0.37
2000	3	0.33
2000	10	0.47
3000	2	0.56
3000	10	0.59

6 Discussion

6.1 Output impedance of MED-EL's inductive link

During the measurements it was noticed that the output impedance of the inductive link was dependent on the load and the source level. That is why several measurements of varying load and varying source level was made, with the result in Figure 22 to Figure 25. By changing the resistor value in series with BEST90, the total load impedance was varied. The source level was varied between some values of relevance, i.e. voltages within the operational range of the inductive link.

Conclusively, the source level does have an effect on the output impedance; however the effect is small. The effect of the load, on the output impedance, is on the other side much bigger. As the load is increases, so does the output impedance of the inductive link. The output impedance of the inductive link increases with increasing load R and decreases when BEST90 is added to the circuit. In fact, it seems as if the inductive properties of BEST90 are in favor to get a higher output from the inductive link. From the data in Figure 22 and Figure 24 a simple and rough approximation of the output impedance of the inductive link can be constructed, valid in the frequency range of 100 Hz to 5 kHz, as shown below:

$$|Z_{il}| = 0.4 \cdot (R - 20\delta_{BEST90}) + 70\Omega \tag{10}$$

where δ_{BEST90} is equal 1 Ω one when BEST90 is added to the circuit and equal to 0 when BEST90 is not in the circuit.

To get better knowledge of the output impedance of the inductive link, more systematic measurements are needed to be carried out.

6.2 Impedance of the BEST70, BEST90 and BEST150

The results were somewhat predicted with the impedance higher for BEST70 and lower for BEST150 compared to BEST90. The diameter of the wire in e.g. BEST90 is larger than in BEST70 (90 µm compared to 70 µm). The length of the wire in BEST90 is at the same time shorter than in BEST70 since the diameter of the wire does not allow as many loops to winded of the wire in the coil. The results were also compared to those found in an earlier study done by Cortés [9] with relatively good resemblance. A straight comparison could not be made since different transducers of different design have been used by Cortés; however the behavior is more or less the same.

From the results of the measurements of the MED-EL's inductive link, it could be expected that BEST90 is better to use for maximal output power; due to the fact that the impedance of BEST90 is the one closest to the inductive link's impedance. This was also confirmed in the force output measurements.

6.3 Frequency response functions for the BEST70 and BEST90

The result plotted in Figure 30 and Figure 31 are useful to be able to calculate how much force the transducers produce for a given voltage or current. The results also give a guidance which requirements are desirable on the inductive link.

With respect to heat generation in the transducer element, the input power needs to be limited. The ratio of the force squared and input power is of interest, which is presented in Figure 49. From Figure 49 the maximum output force, which is limited by the maximum input power, can be calculated by simply adding the maximum input power in dB(W).

Important to note is that all calculations in this section only is applicable in the transducers dynamic ranges.

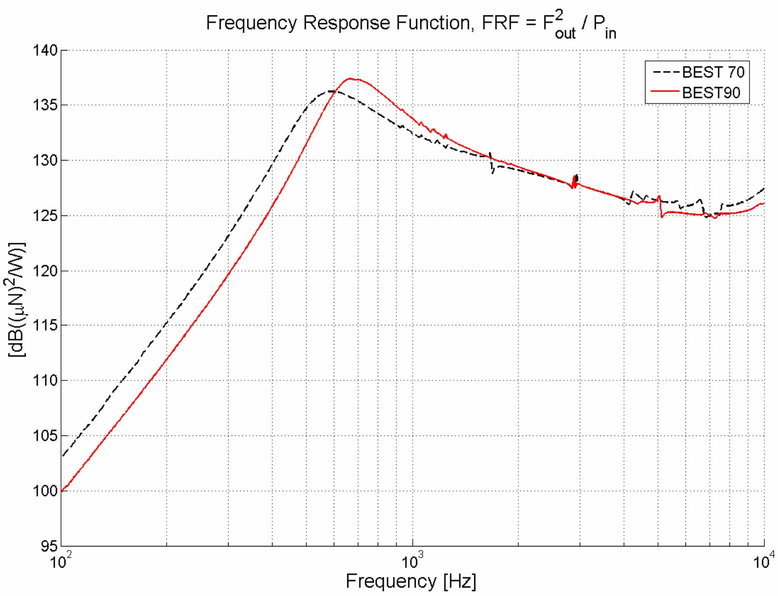

Figure 49. This figure shows the frequency response function F^2 / P and is created from the same data used for Figure 26, Figure 27 and Figure 31.

6.4 Force output from MED-EL with BEST and BAHA

It is apparent from Figure 32 and Figure 33 that the maximum output force level, OFL, for the BAHA system is the same for both volume settings as expected. However, the maximum OFL is reached for a lower sound level when the volume setting of the BAHA transducer is set to 3. Below the maximum OFL is reached in either plot, the OFL is linear dependent of the sound level input, as expected. Within this interval of linearity between the output force and the incoming sound the BAHA system is within its dynamic range.

Similar to the BAHA system, the MED-EL with a BEST is linearly increasing the OFL with increasing sound level until the maximum OFL is reached; see Figure 34 and Figure 35. Within this range the MED-EL system is in its dynamic range.

In Figure 36 and Figure 37 the OFL for all systems are plotted in the same plot. In the first the OFL at 60 dB SPL is shown and in the second the OFL at 90 dB SPL. The OFL at 90 dB SPL is also the maximum OFL. In Figure 32 to Figure 35 several curves lie on top of each other. This means that the driving system's maximum output force has been reached and thereby OFL at 90 dB SPL is the maximum OFL. The plots also show how the OFL of the MED-EL systems are affected by changing the distance in the inductive link from 2 mm to 5 mm. The distance in the inductive link is altered by inserting some plastic separators of 1 mm thickness, which are supposed to mimic the skin of a person. It was observed that the OFLs are only affected slightly by changing the distance from 2 mm to 5 mm. This fact was used in all later measurements where the distance in the inductive link always was kept at 2 mm.

One result that can be drawn from Figure 36 and Figure 37 is that the OFL , measured on the skullsimulator, from the BAHA system is considerably higher than for the MED-EL system. When on the other hand the implants are attached to the skull position 1 and 4 respectively, the latter is more "sensitive" with regard to cochlear vibrations. In this study the author did not have access to changing the gain or output power from the audio processor of the MED-EL's system.

The frequency response function, FRF, in Figure 38 is calculated when each system is in its dynamic range. In this case the sound levels used during the measurements were 60, 65 and 70 dB SPL. The FRFs makes it clear and visible that the gain in the BAHA is considerably higher then in the MED-EL system. It is also notable that by changing the volume setting from 2 to 3 on the BAHA, the gain is increased more or less equally at all frequencies.

6.5 Force output with modified impedance

As described in the measuring chapter above, the output force level, OFL, is measured once again from the MED-EL with BEST90. The only difference is that impedance consisting of a resistor and a capacitor is placed in series with BEST90. The OFL at 90 dB SPL is shown in Figure 39 and the increase or decrease of the OFL by adding impedance in series with BEST90 is shown in Figure 50 below.

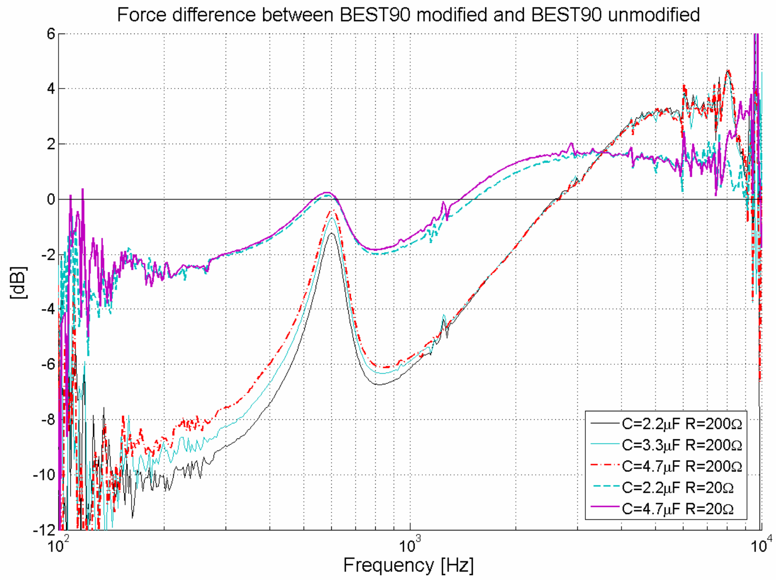

Figure 50. The force difference between an unmodified impedance of BEST90 and modified impedance. A value above zero means that a larger force is produced in the case where there is impedance in series with the BEST90. The measurements were carried out at 90 dB SPL.

The two figures show that by adding an impedance network it is possible to raise the force from the MED-EL system in a higher frequency range. However, by doing so the OFL is significantly lowered in lower frequency ranges. The gain by doing the modification to the circuit was not as big as expected.

6.6 Acceleration on a dry skull

The result drawn from Figure 40 and Figure 41 is that the acceleration for each transducer has a dynamic range, similar to when the force was measured with the skullsimulator. For the MED-EL system there is a unexpected dip in the curves around 700 Hz. If the results from the force measurements with the skullsimulator are kept in mind, one could expect a local maximum of the acceleration and not the opposite. However, if the result is compared to measurements performed by Stenfelt [5], see Figure 51 below, it is understandable why this minimum of the curve occurs. The thick and solid curve in the figure shows how the acceleration, measured in the same direction as done in this study, depends on the skulls properties. This curve shows anti-resonance, but at the frequency 580 Hz. The reason why the minima do not occur at the same frequency could be that the measurements carried out by Stenfelt are about ten years old. During that time the properties of the skull could have changed a lot (such as changes in the stiffness of the bone, its joints and the damping material inside the skull) and hence so also the frequency properties.

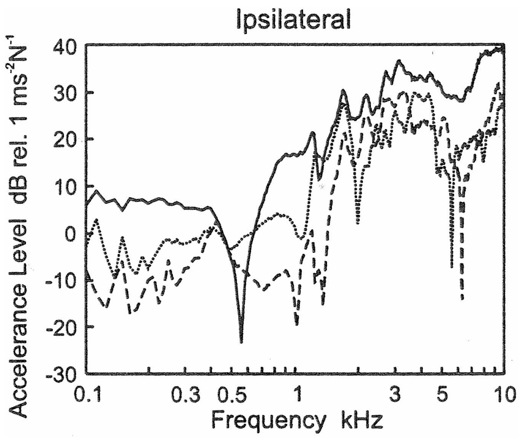

Figure 51. The acceleration plotted for different directions. The thick line corresponds to the same direction as the hearing laser vibrometer is measuring in. The accelerations in this figure are caused by a transducer placed at position 4. [5]

Another detail of interest in Figure 41 for the MED-EL system is the maximums at the frequency 1840 Hz and twice that at 3680 Hz. A third but small maximum can also be found at four times the first frequency at 7360 Hz. Since they are multiple of each other it is reasonable to conclude that they have to do with the resonance of the skull. Those maximums can be found in Figure 40 for the BAHA system as well; even at three times the frequency of the first a peak can be seen. This further implies that it is a property of the skull that causes those fluctuations in the output level.

Interesting in Figure 42 and Figure 43 is that the acceleration of MED-EL with BEST90 actually lies above the BAHA system in the frequency intervals about 1.5 kHz to 2.1 kHz and 4.8 kHz to 7.8 kHz. The two systems actually lie in more or less the same level in the interval from 1 kHz and above. This was not seen at all when comparing the two systems maximum OFL, on the skullsimulator. In the maximum OFL measurements, the MED-EL system was about 10 dB below the BAHA system. However, the fact that the MED-EL system at position 4 has a larger output than the BAHA system at position 1 at higher frequencies was expected, because the point of stimulation lies closer to the promontory, which is the measuring point targeted by the laser through the ear canal.

The frequency response function, FRF, for the different systems plotted in Figure 44 is calculated from when each system is in their dynamic range. The dynamic ranges for the two systems have been decided from Figure 40 and Figure 41, respectively.

6.7 Acceleration on a semi dry skull

It should be mentioned that measurements with the BAHA system with volume control setting 3 have been made. However, there were problems with acoustic feedback and because of that the results were not included. Instead all measurements on the semi dry skull was made with volume control setting 2.

In the acceleration plots for the semi dry skull (Figure 45 and Figure 46) the noise floor was also plotted. The data for the noise floor was achieved by having the speaker absolutely quiet. In that way the only thing causing the vibrations is the inherent microphone and amplifier noise and possibly environmental noise from the surrounding laboratory environment.

Figure 47 was made to compare the results of the acceleration at 90 dB SPL. In the figure the acceleration without any transducers attached was also plotted, which means that the vibrations measured are caused only by the sound waves from the speaker. In the range from 100 to 200 Hz these vibrations are totally dominating. In the range from 200 to 1 kHz, the MED-EL system in position 4 is about 10 dB below the BAHA system. To correct this, the power output from the MED-EL's driving system has to be increased with a factor of ten or the voltage output with a factor square root of ten. Above frequencies of 1 kHz, the MED-EL system is close to or even above the BAHA systems' acceleration. The result is comparable to the results achieved on the dry skull. Interesting to notice is also the two curves for the MED-EL system at the two different locations. From them the relative gain or attenuation done by changing attachment from position 4 to position 1 can be calculated. This is illustrated in Figure 52 below.

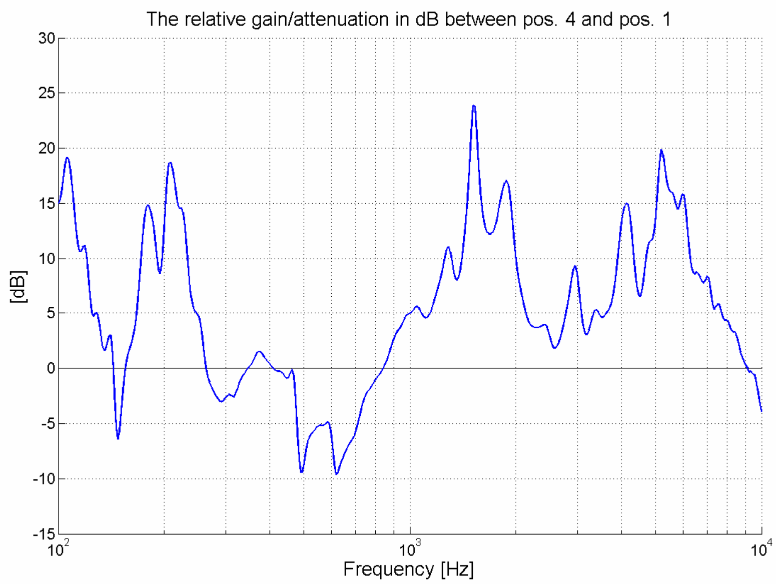

Figure 52. The MED-EL with BEST90 was placed on both position 4 and 1. The difference in acceleration is then plotted. Values above zero indicate that the acceleration from position 4 is larger than from position 1.

The curve in Figure 52 was constructed from the mean value of many measurements ranging from 50 dB SPL to 90 dB SPL. However, the gain or attenuation by changing attachment position was almost the same for each SPL. The greatest gain is obtained in the frequency range above 900 Hz and the only attenuation worth mentioning is between about 500 Hz and 800 Hz. This attenuation is due to the anti-resonance of the skull, discussed in the section "Acceleration on a dry skull".

The frequency response function, FRF, in Figure 48 is practically in the same level as the FRF for the dry skull. However there are some differences such as the FRF for the semi dry skull is smoother than for the dry skull. Some of the peaks which was seen with the dry skull have disappeared or been suppressed. The smoothness is due to the fact that the skull is filled with gelatine, which makes the system more dampened.

On the dry skull there were some peaks that occurred due to the properties of the skull (see "Acceleration on a dry skull"). The gelatine filled skull showed the same behavior in the frequency response function, however not as legibly. There are peaks found at many frequencies such as 1640, 3400, 5040, 6600, 8200 and 9700 Hz. They are all roughly multiples of the 1640 Hz frequency but the interval between each maximum decreases with increasing frequency. This can be explained by the fact that at higher frequencies a smaller mass has to be moved and thereby the frequency interval between two maximums decreases with increasing frequency. Or in other words, the skull is not a perfect rigid body.

6.8 Distortion from MED-EL with BEST and BAHA

While measuring on the semi dry skull a few numbers of harmonics were used for the calculations of the total harmonic distortion, THD (Table 1 and Table 2). In the case of the BAHA system, the number of harmonics is limited down to one and for the MED-EL system one or two (measurement dependent). Figure 21 shows how it looks for the MED-EL system when two harmonics can be found from a sinus with fundamental frequency of 750 Hz. The reason why not more harmonics can be found is that the noise floor is greater than the harmonics themselves. However, the result from the measurements on the semi dry skull seen in the Table 1 and Table 2, show that the THD is smaller for the BAHA system.

The result from Table 4 and Table 5 shows that the THD is overall larger for the MED-EL system, except at the frequency 500 Hz. The reason for the relative high THD for BAHA at 500 Hz is that the BAHA system has its resonance peak at approximately 1000 Hz (see Figure 38) and thus the first harmonic is emphasized. If the THD results from the MED-EL system from Table 1 and Table 4 is compared, the THD at 750 Hz is relative high when measuring on the skull but relative low when measuring on the skullsimulator. This is understood by analyzing Figure 38 and Figure 48. Figure 38 shows the frequency response function when measuring it on the skullsimulator and Figure 48 shows the FRF on the semi dry skull. On the semi dry skull there is a maximum close to the first overtone of 1500 Hz, which will increase the THD from the skull. The FRF from the skullsimulator does not have any maximum at this frequency and thus the THD will not receive as much boost at this harmonic.

One reason why the overall THD is so much smaller when analyzing the results from the semi dry skull for the BAHA system, is that the FRF is about 10 dB higher for the BAHA system compared with the MED-EL system in the frequency range below 1 kHz (see Figure 48). For the BAHA system this will make the fundamental tone stronger and the overtones weaker when compared to the MED-EL system.

Table 3 gives the THD from MED-EL's inductive link and Table 4 gives the THD from the skullsimulator with the MED-EL with BEST90 attached. The THD is lower from the skullsimulator than from the inductive link. This can be explained by the fact that BEST90 attached to the skullsimulator acts like a band-pass filter (see Figure 30). For frequencies over 700 Hz the overtones will be relatively reduced more than the fundamental tone. The THD for 500 Hz on the other hand will not be decreased in the same way.

The THD is overall lower for the BAHA system even when measuring on the skullsimulator. Inspecting the FRF in Figure 38 shows that the BAHA system's FRF descend faster with higher frequencies than the MED-EL system's FRF, thereby giving a lower overall THD for the BAHA system.

The only exception is at the frequency 500 Hz where the THD for the BAHA system is higher. This depends on the fact that the first overtone occurs at the BAHA system's FRF's peak.

It was expected that the THD is higher for the MED-EL system compared to the BAHA system due to the fact that the modulation and demodulation circuits cause distortion.

7 Error analysis

The calibration of the microphone was done one single time and the same conversion factor was used during the entire work and for all calculations. The amplification in microphone power supply should be equal to unity and the condenser element of the microphone should give 11.7 mV/Pa according to calibration chart [11]. However the conversion factor after the power supply was measured to 10.2 mV/Pa and that imply a deviation of 1.2 dB. The reference microphone cannot be placed at the same position as the microphone inside the hearing aid. That can cause the sound levels at the two different positions to deviate, especially at higher frequencies where standing waves can cause large differences, see Figure 53 below. In these graphs, it can be seen that the choice of microphone position affects the result, especially at lower sound levels. However, the difference is seldom more than 2 to 3 dB.

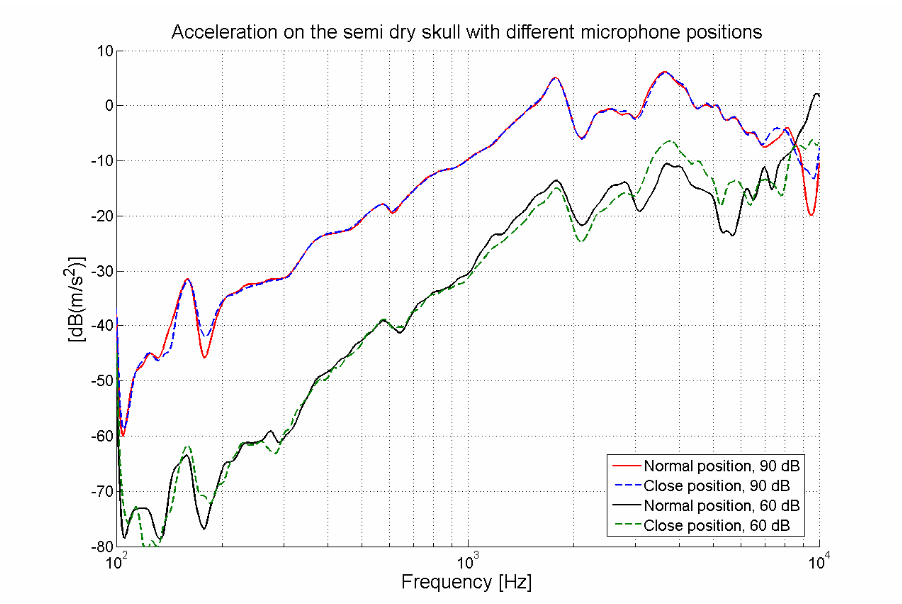

Figure 53. The acceleration measured on the semi dry skull with the MED-EL with BEST90 placed at position 4. The acceleration is measured at 60 and 90 dB SPL and with the condenser microphone at two different positions. One very close to MED-EL's AP and one placed at the normal measuring position seen in Figure 10.

The skull used during the measurements has been discussed in several places in this thesis; however, some additional things are worth mentioning. Every human skull is unique and deviate from the average. It is not considered how representative the skull, used during the present measurements, is. The fact that the dry skull does not have brain tissue and soft tissue makes it most likely to deviate from a real human head. Even when the skull is filled with gelatine it still lacks its soft tissue and attachment to the rest of the body. During the measurements, the ventilation of the test room was turned off to reduce noise but that increased the temperature up to about 30 °C. Some measurements were carried out when the gelatine solution had become old. Temperature and aging is suspected to affect the gelantine solution's consistency so it will change to be more water-like than like brain substance.

The Hearing Laser Vibrometer was calibrated last time in July 2001. The HLV is only measuring vibrations in the beam direction. For that reason the HLV will not give the complete picture over the vibrations in the promontory. However, the measurements give a fair overview of the vibrations in the promontory, especially at lower frequencies, since the stimulation and measuring of vibrations is done in the same direction. It is important to point out that vibrations in one direction are not the same as hearing perception.

The measurement errors caused by the Agilent are estimated to be negligible. There are two possible reference point settings to choose between, floating and grounded input. The used option during all measurements has been floating. The reason for this choice was that it was convenient and provides good enough measurement performance.

Unluckily two slightly different BAHA was available and it is possible that different BAHA have been used for different measurements. The differences of the two individuals are however very small and do not affect the results. The two individuals are old and the resonance frequency is slightly high compared to the nominal. They were however tested carefully before the measurements and were found to fulfill every standard with margins.

The product from MED-EL might not be a device for custom use and there is no information on how representative the example used in the measurements is. But still it is most likely that the example used is close to examples for custom use.

8 Conclusions

The conclusions are the combination of the MED-EL system and the BEST transducer is feasible solution as an alternative to the present BAHA system. Based on measurements of the dry and semi dry skull, the maximum output from the present MED-EL system with BEST is 10 dB lower than the percutaneous BAHA system at frequencies below 1 kHz. However, when these measurements were repeated on a cadaver the two systems showed almost equal performance at all frequencies except in the range 300 to 500 Hz [12].

The output impedance of the MED-EL system is load dependent and is measured to be in the size of 50 – 100 Ω in the relevant frequency range and for a relevant load.

The BEST90 is more suitable than BEST70 and BEST150, because its impedance is closest to the output impedance of the MED-EL's inductive link. However, the difference between BEST90's and BEST70's output response is small. It is first with increasing frequency that the difference starts to grow and reaches a maximum of 6 dB. That depends on the fact that the impedance of BEST70 deviates even more than BEST90 from the impedance of the inductive link at higher frequencies.

It is possible to increase the output response at higher frequencies by modifying the impedance of the transducer, but at the cost of decreasing response in the low and middle frequency ranges are too high. As it looks today it is not of immediate interest to modify the impedance by adding capacitors or resistors in series with the BEST driven by the present amplitude modulated output driver stage.

When measuring on the skullsimulator the maximum output force of the MED-EL system was 10 dB below the BAHA system at frequencies above 1 kHz. However when measuring the maximum output acceleration on the dry and semi dry skull with each system on their proposed positions, the MED-EL system is in parity with the BAHA system above 1 kHz. Below 1 kHz the MED-EL system has dropped 10 dB in relation to the BAHA system. The fact that the output of the MED-EL system is that much below on the semi dry skull indicates that it could be a problem.

There is more distortion in the MED-EL with BEST system relative the BAHA system, measured on both the skull and on the skullsimulator. The distortion arises in the MED-EL and is attenuated by the BEST. The properties of the semi dry skull also influence the size of the total harmonic distortion.

9 Future work

There are two obvious ways to proceed for future studies. The first and most simple way is to accept the performance and to start with more measurements on cadaver heads. The measurements need to be done on several subjects to get a more generally applicable result. A suitable casing also needs to be developed. The other way is to try to improve the performance, perhaps by building the system on a laboratory board and try to improve and to optimize all parts. It would, for example be interesting to test pulse width modulation instead of amplitude modulation. By using pulse width modulation it would be possible to build a system which is chargeable when there is no need for amplification of sound (e.g. during sleep or in silent environments). It would also be possible with a pulse width modulation system to get a higher output force level from the BEST transducer.

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