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Cochlear Implant Impedance Telemetry Measurements and Model Calculations to Estimate Modiolar Currents

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Acknowledgments

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Abstract

The use of impedance and neural response telemetry measurements through stimulation and recording of electrical signals can facilitate device fitting and parameter adjustments, especially in young children. However, the detailed configuration of electrical impedances and current distributions around the electrode array is unknown and has not been able to be determined using standard impedance telemetry measures. We therefore attempted to improve the impedance measurement procedure by applying a more detailed model of electrical impedances of stimulation and recording electrodes within the cochlea and by developing more sophisticated measurement protocols to identify the model parameters in a given implant subject. In particular, the modiolar currents, the portion of the currents assumed to be responsible for activating the neural elements, are determined and evaluated.

Their predictive value for device fitting parameters is being investigated and first results have been obtained by a series of postoperative and intraoperative measurements, which will allow us to estimate changes of the modiolar current distributions over the first weeks after implantation.

This approach is mainly based on a published patent, and makes use of the Nucleus Matlab Toolbox and the Nucleus Implant Communicator (NIC) software. The ultimate goals of a refined model and more specific impedance measurements are semiautomatic fitting and of programming parameter update procedures for CI users with varying electrical stimulation conditions.

Part I.

Background

1. Cochlear Implant

1.1. Overview

Hearing

Hearing is one of the five senses and describes the ability to perceive sounds by the ear.

Sound in the form of an acoustic pressure wave is received by the outer ear, carried on through the auditory canal and converted into mechanical vibrations in the middle ear. The cochlea, a spiral-shaped cavity, turning around its axis, the modiolus, constitutes the inner ear. The ossicles of the middle ear transmit the mechanical vibrations onto the oval window, where it generates pressure waves in the fluid of the cochlea, the perilymph, corresponding to the frequency of the acoustic signal. This leads to displacements of the basilar membrane, which deflects the hair cells attached to the basilar membrane. The deflection of the hair cells induces a cascadeof signals, which generates a neural impulse, which is in turntransmitted to the auditory nerve, and finally, the auditory cortex.

If the transmission of acoustic sound into mechanical vibrations into acoustic pressure waves into neural impulses is somehow defective, the person suffers to some extent from deafness. For example, damage of the hair cells by diseases, certain drug treatments or congenital disorders prevent the pressure waves in the perilymph to be converted into neural impulses, and auditory signals can no longer be perceived. [6]

Cochlear Implants (CI) provide partial hearing by stimulating the auditory nerve cells, and thereby circumventing malperformance of the transmission of acoustic sound into neural impulses.

History of Cochlear Implants

The first CI surgery in Switzerland took place in Zurich in 1977. Since then, cochlear implants have provided partial hearing to more than 600 patients in Zurich and to more than 200'000 patients worldwide. Being the most successful neural prosthesis to date, the cochlear implant is the only implant capable of acting as a replacement of a sense organ.

Currently, most cochlear implants are provided by three major manufacturers,

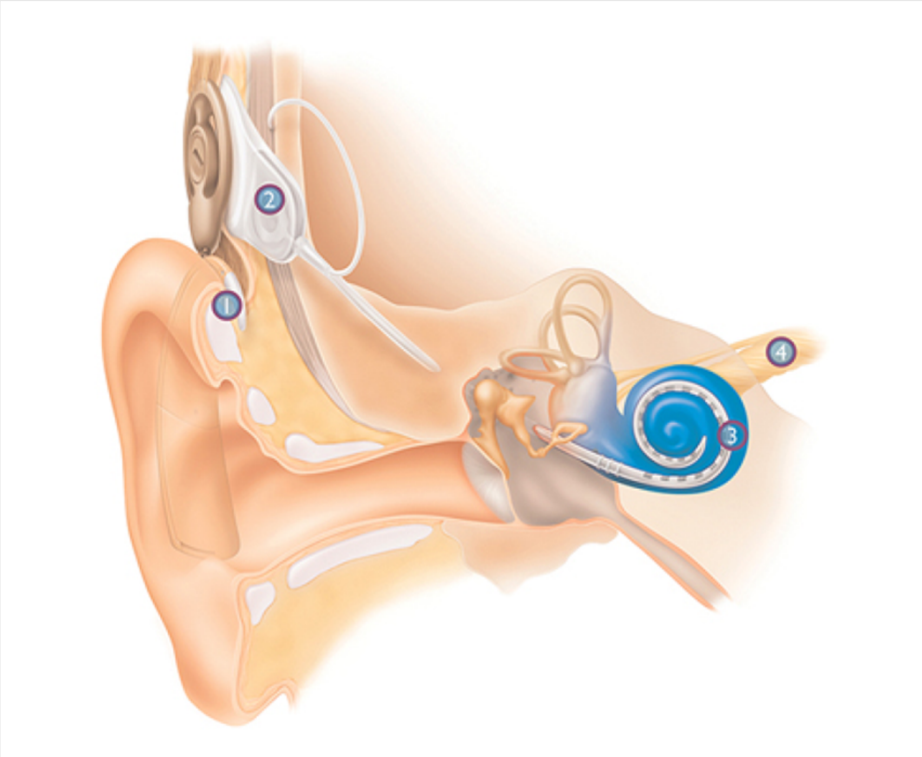

Figure 1: Components of a Cochlear Implant

including Cochlear (www.cochlear.com), Med-el (www.medel.com) and Advanced Bionics (www.advancedbionics.com).

Cochlear Implant Operation

Cochlear Implants consist of the following components which can be seen in Figure 1: Sound is picked up by one or several microphones attached to the ear and transmitted to a speech processor ((1) in Fig. 1), where the acoustic signal is digitized and encoded. These signals are then transmitted by the sending external coil to a receiving coil ((2) in Fig. 1) implanted beneath the skin. Subsequently, electrical impulses are sent to the array of electrodes ((3) in Fig. 1) inserted into the cochlea. The electrodes stimulate the auditory nerve fibers ((4) in Fig. 1) within the cochlea by inducing an electrical field through biphasic current pulses. [6, 5, 9, 12]

The specific auditory sensation is regulated by the stimulation frequency and the position of the stimulated electrode. Depending on which auditory nerves are stimulated, a different auditory sensation is perceived by the brain.

1.2. Current Distribution

When a current is applied to an intracochlear electrode, the current is distributed in different directions, as is displayed in Figure 2. One part of the current flows longitudinally along the low impedance fluid within the cochlear cavity. The other part of the current flows radially in the direction of the modiolus.

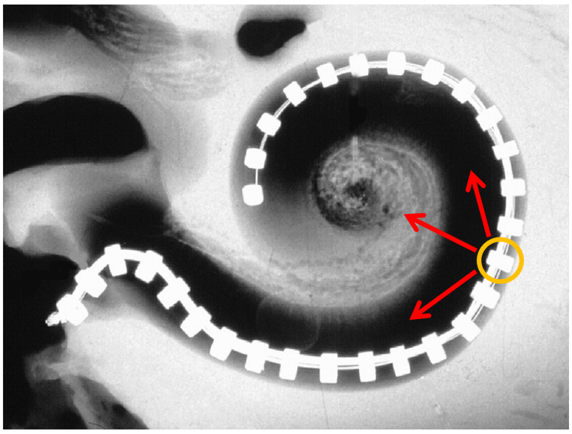

Figure 2: Cochlea: Current Distribution

Figure 3 displays a close-up of the cochlea with an inserted electrode array. The array of electrodes winds itself around the axis of the cochlea, the modiolus. In the modiolus, the auditory nerve cells gather and form the cochlear nerve. The current flowing in the direction of the modiolus is understood to be the current to stimulate the adjecent auditory nerve cells.

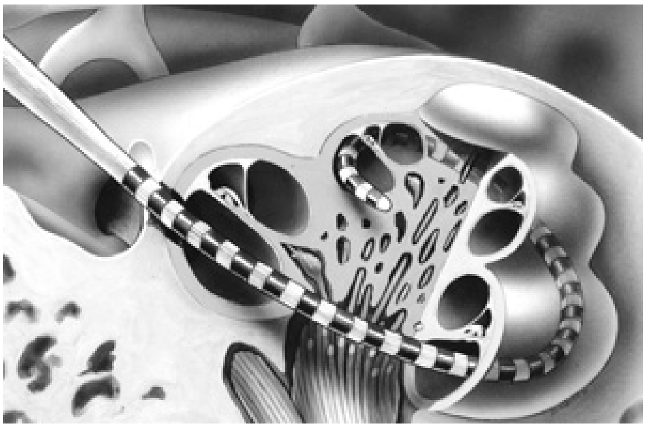

Figure 3: Array of Electrodes

1.3. Impedance Measurements

A cochlear implant is a precisely controllable device, and its way of stimulation is determinable. Still, the attributes of the electrodes are subject to change, and therefore have to be checked on a regular basis. Evaluating the electrodes functionality is facilitated by various analysis tools, as for example the clinical software of the major cochlear implant companies. Besides different fitting parameters, the clinical software provides the possibility to measure and calculate electrical impedances in the cochlea. The impedances allow an enhanced comprehension of the setting of the CI in the cochlea and of its performance. By measuring for example the monopolar impedance (compare with chapter 4.1.1) of each electrode, we can investigate the electrode's overall functionality, and detect problems such as wheater a short-circuit or a break exists or not.

In this study, the analysis of well-established impedance measurements is further persued, and by means of additional impedance calculations, the cochlear implant's functionality is investigated.

2. Tools/Methods

2.1. NIC

To communicate with the CI and to enable the measuring of implant and electrode properties, Cochlear developed the Nucleus Implant Communicator (NIC). NIC is a software interface aiming to provide methods to define and deliver stimulation patterns and other operations. NIC is platform independent and has been used in C, Python, Matlab and Delphi/Visual Basic.

NICs main feature, to define and deliver stimulation patterns and to record the resulting measurements, was also used in this work. Its application is limited to Freedom and L34 speech processors only and to CI24M/R and CI24RE implants. Therefore, in this study, all implants are of the implant type CI24RE Nucleus Freedom from Cochlear.

Measuring and collecting the data was implemented in Python. For further information about NIC and its functionalities, consult [4] or the program files in Appendix B.

2.2. Voltage Telemetry

To gather information about the implant, we can send information to the implant, and receive specific information from the implant. Sending information is called Streaming, receiving information Telemetry. By voltage telemetry specific voltages between electrodes can be measured, passed onto the receiver and sender coil and collected. We can define any combination of two electrodes as a channel to measure the voltage between them. It's possible to use an intracochlear or an extracochlear electrode or common ground as reference.

Thanks to NIC, we can define stimulation and timing parameters. With a stimulus command, a specific stimulus can be delivered to the electrodes. Here we used a constant current biphasic stimulus (Fig. 6), as CIs generally stimulate the auditory nerve with a series of short biphasic electrical pulses. The pulses are biphasic and charge-balanced because the net current through the tissue should be zero to avoid unwanted long-term electrochemical effects. [7] The amplitude of the biphasic stimulus is defined by the applied current level.

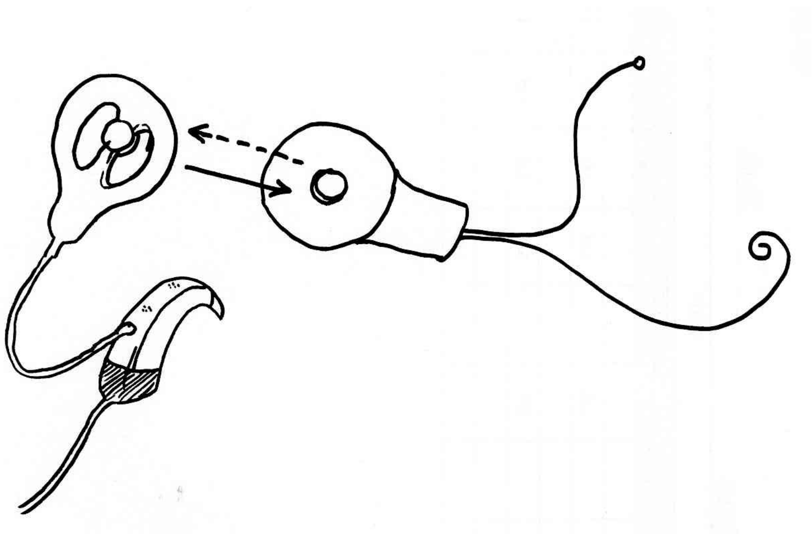

Figure 4: Streaming and Telemetry

The fact that the amplitude of the current level has an effect on the signal to noise ratio, (the higher the current level is, the lower the signal to noise ratio) has an impact on our choice of current level. (chapter 7.3)

2.3. Impedance Calculation

The impedance between a pair of electrodes in tissue can be modeled by a three-element model consisting of an ohmic and a capacitive part. (Fig. 5)

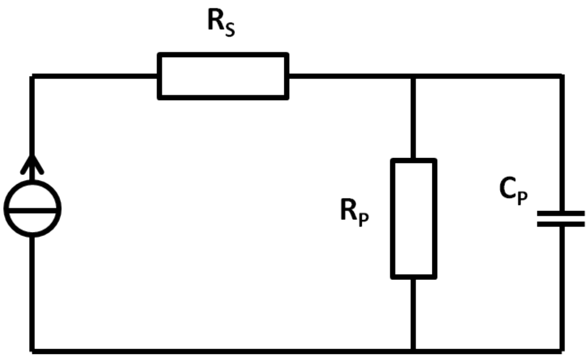

Figure 5: Impedance equivalence Circuit

As input we use a biphasic stimulus (as explained in chapter 2.2). The voltage response to such a biphasic stimulus and the effect of the resistive and capacitive component of the electrode/tissue interface impedance can be seen in the waveform of Figure 7. With NIC we can choose the exact sample time for the calculation of the voltage to current ratio. If we measure at the leading edge of the stimulus' fist phase we measure the voltage across R_S, as the capacitor is fully conductive at current onset. Measuring at the trailing edge of the stimulus' first phase we measure the voltage across R_S plus R_P. We chose the measuring point at the trailing edge, as is shown in Figure 7, where the highest voltage change can be measured. From that, we calculate the R_S plus R_P applying Ohm's Law (Appendix (A.1)). To allow the use of Ohm's Law we assume that at the point of measurement at the trailing edge of the stimulus' first phase the capacitor is fully charged and therefore nonconductive, which may not be tha case.

Further discussion about the impedance and its capacitive part can be found in [10, 2].

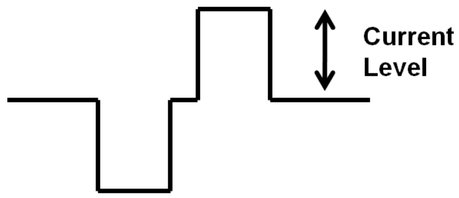

Figure 6: Biphasic Stimulus

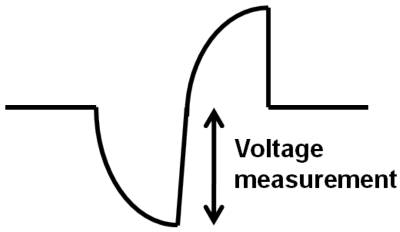

Figure 7: Voltage Response

2.4. Implant Load

The implant load is designed to connect to a CI24M implant in a box, and thereby provide conditions for measuring and testing impedances, for example. All specifications, as for example the circuitry and impedances of the implant load network, are known.

In this study we used the implant load to test and verify our procedures.

For further information consult [3].

Part II.

Modiolus Currents Model

3. Modiolus Currents Model

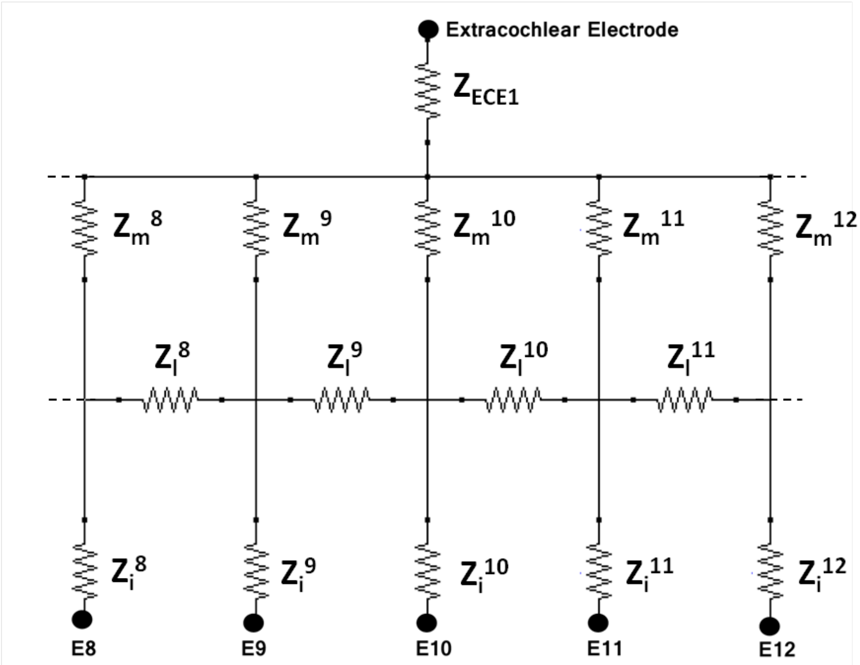

Figure 8: Modiolus Currents Model

The Modiolus Currents Model was developed and patented [11] by Kostas I. Tsampazis, Paul M. Carter and Herbert Mauch, members of Cochlear cooperation. It describes methods to determine intracochlear impedances, some of which are described in the following chapter.

This master thesis project attempts to improve the impedance measurement procedure by applying this more detailed model of electrical impedances of stimulation and recording electrodes within the cochlea, based on this model, and developing more sophisticated measurement protocols to identify the model parameters in a given implant subject. In particular, the main interest consists of identifying the modiolar currents thought to be responsible for neural excitation and separating them from longitudinal shunting currents, which do not contribute to the generation of action potentials.

In the following chapters the impedances are referred to as follows. The impedances adjacent to the intracochlear electrodes are the interface impedanes Z_i, characterizing the electrode/tissue interface. Z_l are the impedances in longitudinal direction from the electrode, along the liquid in the cochlea, Z_m is the impedance in direction of the modiolus and Z_{ECE1} is the interface impedance of the extracochlear electrode.

4. Calculations and Results

To reach an understanding of the distribution of the currents in the cochlea, we need to measure voltages generated by the CI in the cochlea. We use Voltage Telemetry (chapter 2.2) to measure the voltage of a determined channel while stimulating with a predefined current (chapter 7.3) between two electrodes.

In the following chapters, different measurements are described, as well as the calculations applied to gain new information about the network.

4.1. Impedances

4.1.1. Monopolar Impedance Measurement

First, we apply a current between each intracochlear electrode (e.g. Electrode 9 in Fig. 9) and the extracochlear electrode, and measure the voltage between the two electrodes. From this the monopolar impedance is easily calculated. As the extracochlear electrode we used ECE 1.

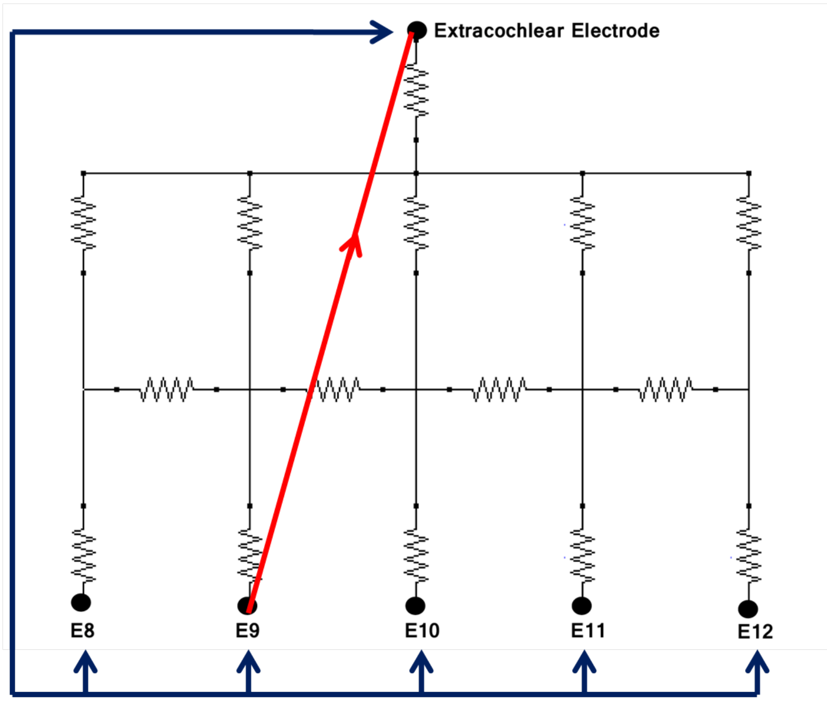

Figure 9: Monopolar Impedances.
Explanation:
Red Arrow: Current Stimulation.
Blue Arrow: Measure Voltages.

The monopolar impedance has been used in clinical software such as Customn Sound to evaluate the condition of the intracochlear electrodes.

Looking at the Modiolus Currents Model, we assert that the monopolar impedance is primarily composed of the interface impedance of the intracochlear electrode, the interface impedance of the extracochlear electrode and the impedance in the direction of the modiolus adjecent to the intracochlear electrode. In addition, the longitudinal impedances along the fluid in the cochlea (4p impedance measurement), and the impedances in the direction of the modiolus that are further away from the stimulated electrode contribute to the monopolar impedance.

We use the monopolar impedance, on the one hand, for compliance check. On the other hand, during each monopolar stimulation, (between the extracochlear and an intracochlear electrode) we measure not only the voltage on the stimulated intracochlear electrode, but also on all the other electrodes, with the extracochlear electrode acting as reference. The utility of these voltages is described in chapter 4.2.

4.1.2. Four Point Impedance Measurement

Similarly we can calculate further impedances.

For the four point impedance (Fig. 10), we apply a current between two intracochlear electrodes, spaced three electrodes apart, as for a bipolar +2 impedance calculation. In to Fig. 10 these electrodes are for example electrode 8 and electrode 11.

Voltage telemetry allows us not only to select the electrodes we want to stimulate, but also the pair of electrodes to measure the voltage. Thus we can measure the voltage present between two other electrodes aside from the pair of electrodes on which the current is applied. For the calculation of the impedance in longitudinal direction, we measure the voltage between the two electrodes (e.g. electrode 9 and 10) located between the electrodes where the currrent is applied (e.g. electrode 8 and 11).

We assume that no currents flow apart from the direct connection between the two stimulation electrodes, as is marked in Fig. 10. (About the assumptions: chapter 5.2) Therefore, the measured voltage also drops over the impedance along the fluid in the cochlea between the two electrodes, on which we measure the voltage (e.g. electrode 9 and 10). Consequentially, we can calculate this impedance, referred to

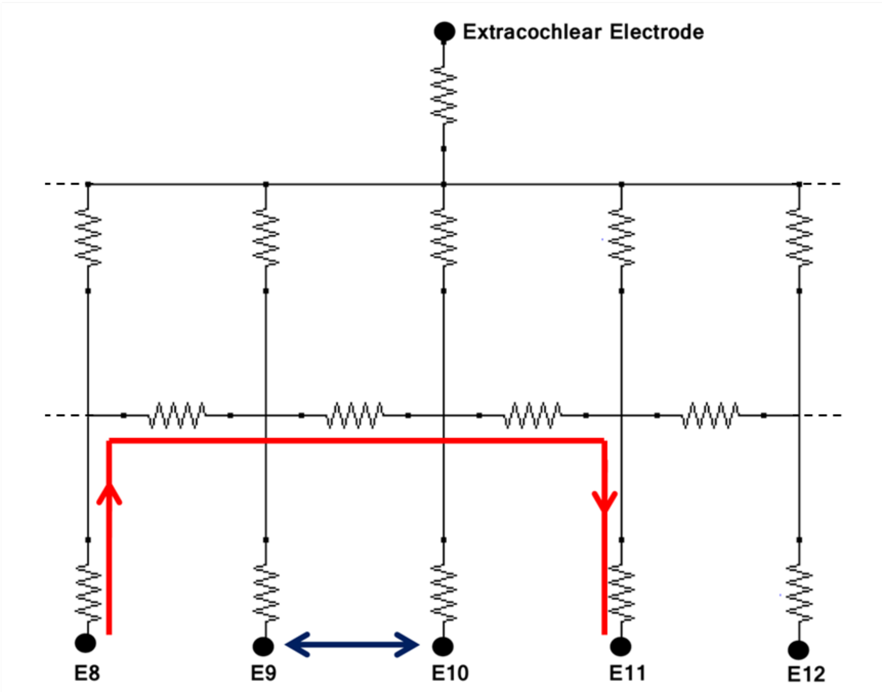

Figure 10: 4 Point Impedance Measurement Model

as "impedance in longitudinal direciton" or "4 point impedance" .

In Figure 11, the distribution of 4 point impedances along the array of electrodes from a cochlear implant is plotted. Figure 12 shows the 4 point impedance data collected from 16 subjects. It can be observed that most values amount to a few hundred Ohms.

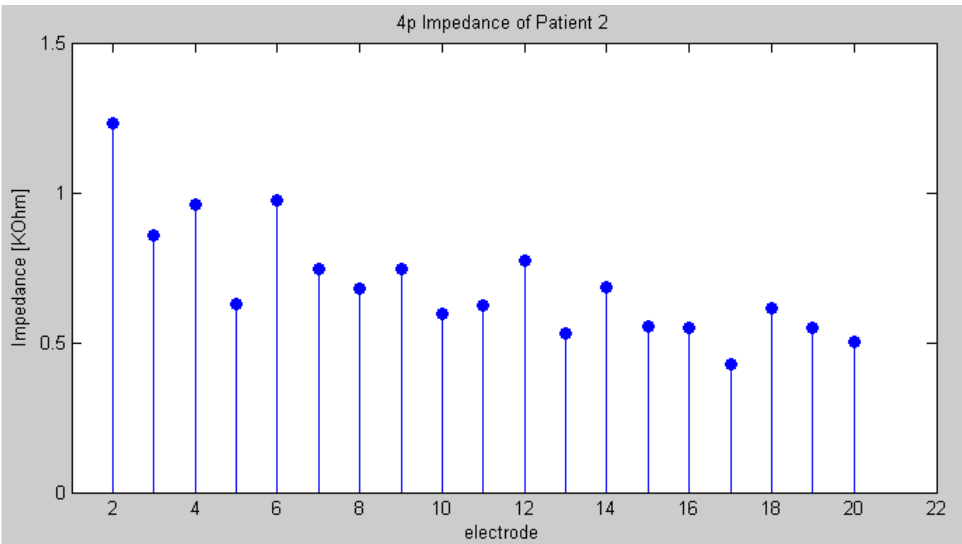

Figure 11: 4 Point Impedance Measurement Data

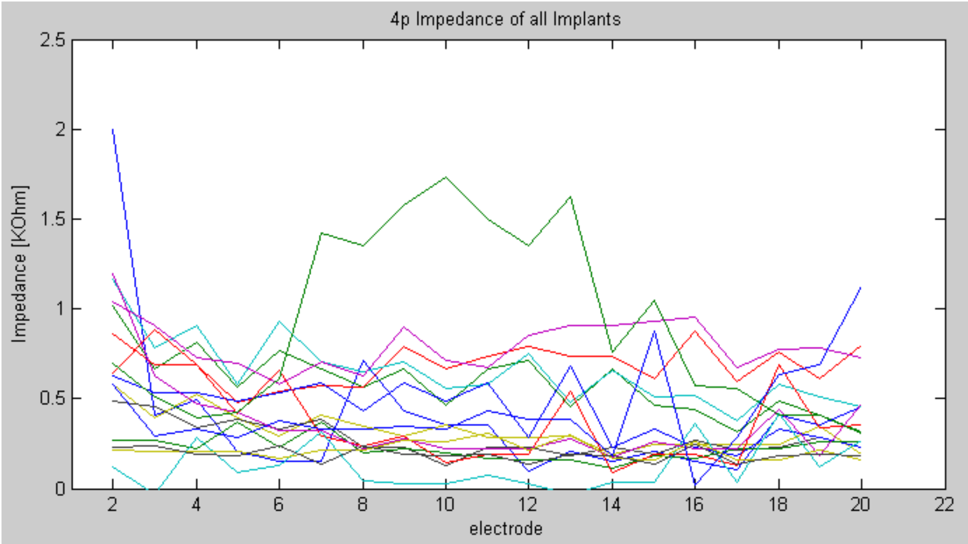

Figure 12: 4 Point Impedance Measurement Data from 16 subjects.
Each graph in this Figure displayes the amplitudes of one 4 point impedance measurement like the one in Figure 11.

4.1.3. Three Point Impedance Measurement

The measurement for the 3 point impedance is similar to the 4 point impedance measurements. The current is applied between two intracochlear electrodes spaced two electrodes apart, (see in Fig. 13: electrode 9 and 11) and the voltage measured between one of the electrodes (e.g. electrode 9) and the intermediate electrode (e.g. electrode 10). The measured voltage can be assumed to drop across the interface impedance (in Fig. 13 this is the interface impedance of electrode 9 and the surrounding tissue) and over the 4 point impedance between two electrodes (e.g.between electrode 9 and 10). From this voltage, we can calculate the sum of the two impedances. A sample distribution of the 3 point impedances along the array of electrodes is plotted in Fig. 14.

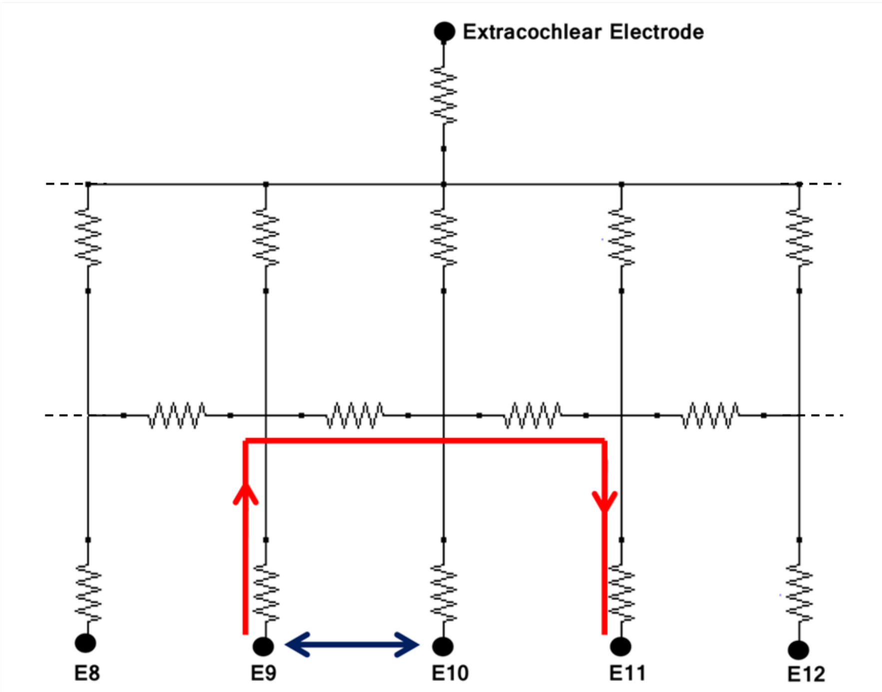

Figure 13: 3 Point Impedance Measurement Model

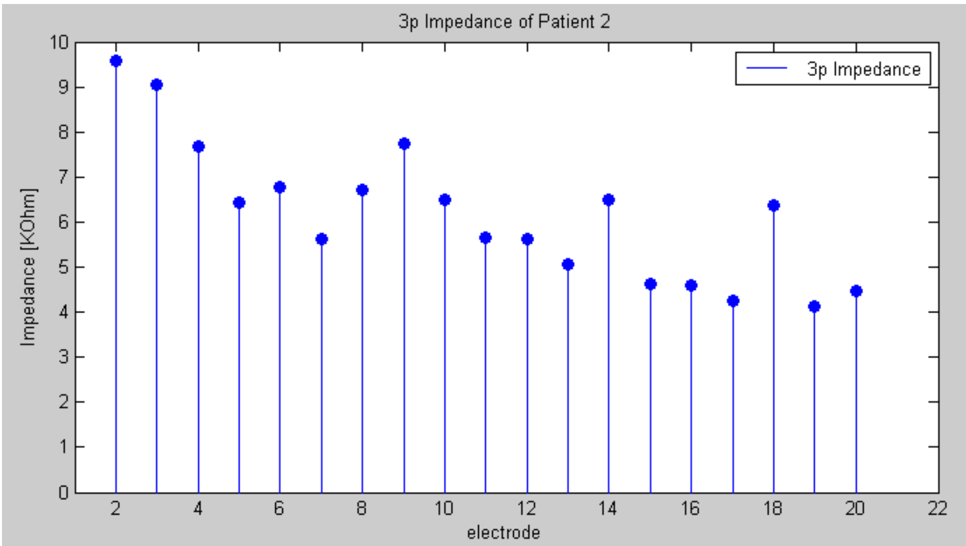

Figure 14: 3 Point Impedance Measurement Data

The 4 point impedance measurement yields the impedances along the fluid in the cochlea, the 3 point impedance measurement the sum of the 4 point impedance and the interface impedances. Consequentially, we can calculate the interface impedances, an examplary distribution being plotted in Fig. 15.

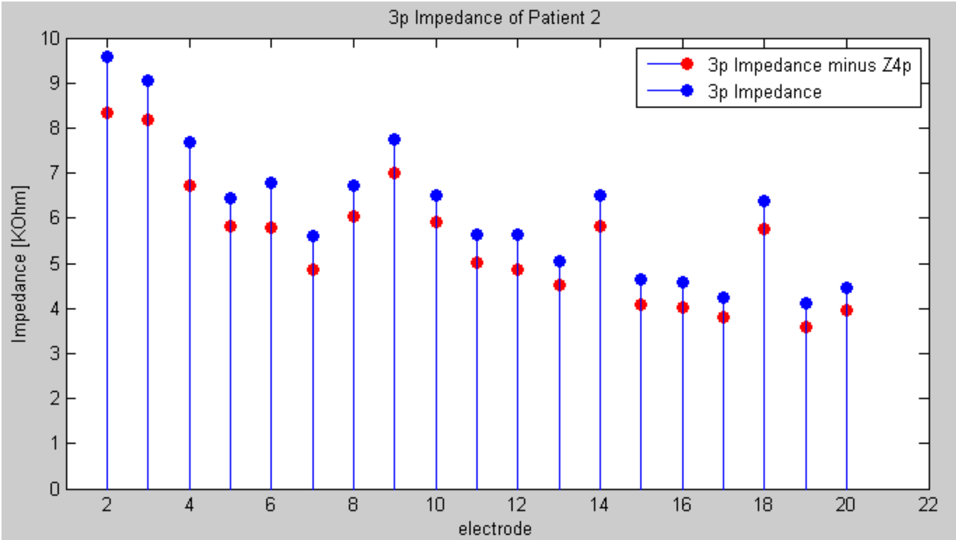

Figure 15: Interface Impedance

4.2. Currents

After measuring and calculating various impedances, we turn towards the calculation of the currents.

4.2.1. Currents in Longitudinal Direction

For the calculation of the current in longitudinal direction, we use the 4 point impedances (chapter 4.1.2), the interface impedances (chapter 4.1.3) and the measurements of all voltages during monopolar stimulation (chapter 4.1.1).

First, we calculate the voltages applied at the internal nodes of the network (colored orange in Fig. 16). From the measured voltages between the intracochlear electrodes and the extracochlear electrode, (colored blue in Fig. 16) we subtract the voltage drop over the respective interface impedance (colored green in Fig. 16). The voltage equals the current through the impedance, multiplied by the impedance. The current is the stimulation current at the stimulated electrode, (if stimulation at electrode n) and is 0 at any other electrode (if stimulation at any other electrode aside from electrode n). The impedance is the interface impedance of the respective electrode.

$$Vnode(n) = Vmono(n) - I(n) * Zinterface(n) \tag{1}$$

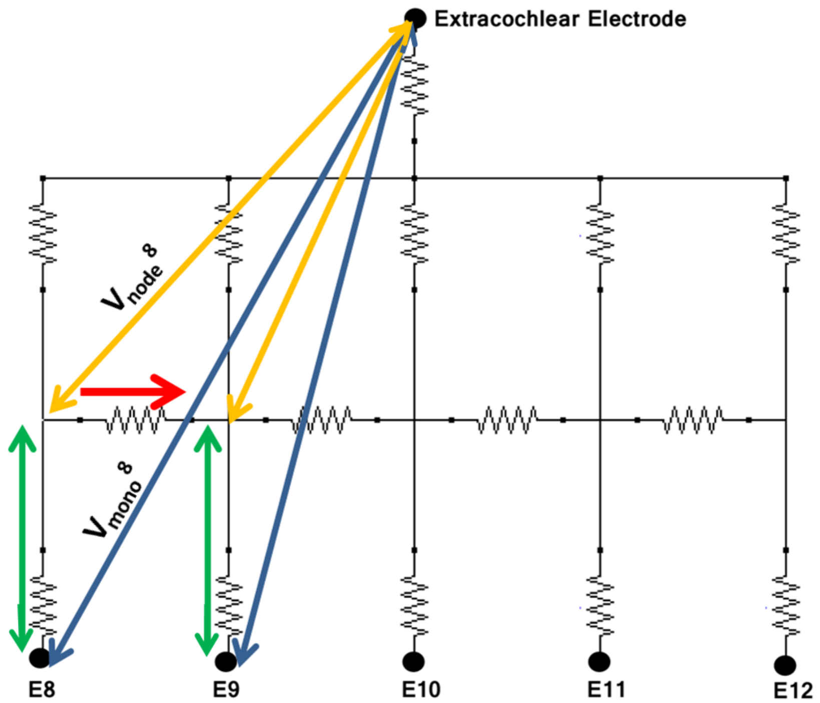

Figure 16: Model for Calculation of Longitudinal Currents

By iterating equation 1 over all electrodes and over all stimulation pairs, we calculate a 22x22 matrix Vnodes.

Secondly, from the voltages at the nodes, (colored in orange in Fig. 16) we calculate the voltages across each impedance in the longitudinal direction (colored red in Fig. 16). This voltage is then divided by the respective longitudinal impedance, which results in the respective current, according to equation 2.

$$I - longitudinal(n) = \frac{Vnode(n+1) - Vnode(n)}{Zinterface} \tag{2}$$

Problems with calculation

The calculation of the currents in longitudinal direction is subject to various inaccuracies.

On the one hand, the accuracy of the measured voltages is unsatisfactory. The gain of the measurements not only defines the resolution of the measurement, which we want as high, as possible, but also the measurement range. The range of the measured voltages depends on the current level (the higher the current level is, the

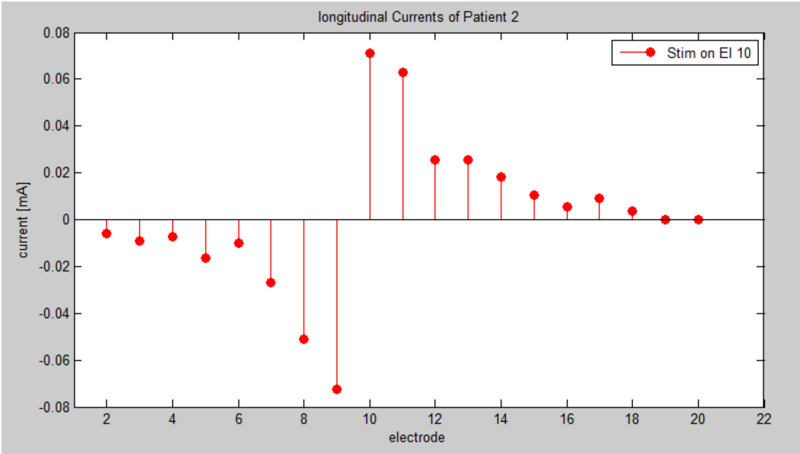

Figure 17: Longitudinal Currents

higher the measured voltages are), which we want to choose high due to suppression of measurement noise. Unfortunately, there is no current level which fulfills all demands and therefore, the accuracy of the voltage measurements is not sufficient. For further explanations read chapter 7.3.

On the other hand, the assumptions proposed in [11] and made by us (both are discussed in chapter 5) may have a larger influence on the impedance calculations than anticipated.

Another source of error may arise from digitalization and discretization during the streaming and telemetry process.

4.2.2. Currents in Direction of the Modiolus

From the distribution of the current in longitudinal direction, (chapter 4.2.1) the currents in the direction of the modiolus can easily be calculated. At each internal node, the currents from the surrounding edges are added up. The afferent minus the efferent current in longitudinal direciton (calculated as described in chapter 4.2.1 and both colored blue in Fig. 18) plus the applied current (colored green in Fig. 18) result in the current in the direction of the modiolus (colored yellow in Fig. 18). This calculation is based on Kirchhoff's circuit law. (appendix A.2)

The above calculation is applied to E(n), producing in the end an array of modiolus currents. Note that for monopolar stimulation at each electrode, the incoming currents at the other electrodes are zero.

Eventually, we obtain the distribution of the currents in the direction of the modi-

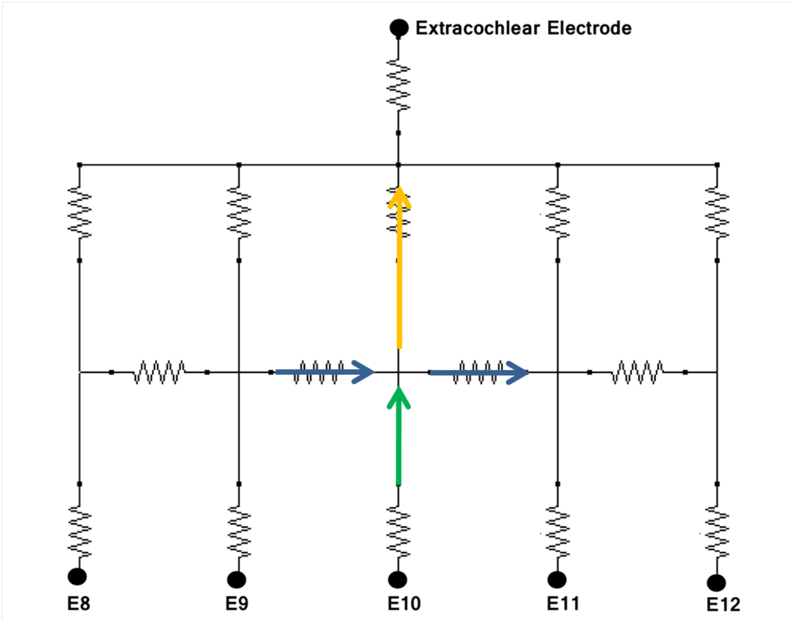

Figure 18: Model for Calculation of Radial Currents in the Direction of the Modiolus

olus dependant on the stimulation mode. Combined with the distribution of the currents in longitudinal direction, we know the distribution of currents during monopolar stimulation of each electrode separately.

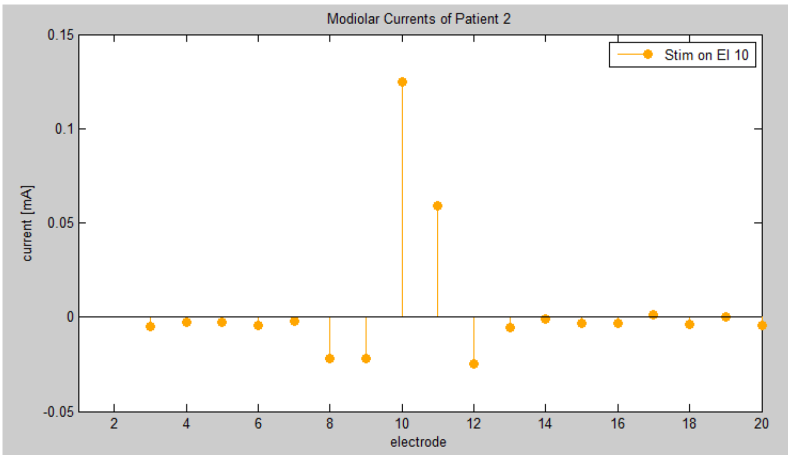

Figure 19: Currents in the Direction of the Modiolus

As the modious current distribution is calculated from the currents in longitudinal direction, we cannot make a useful statement about the ratio of currents in longitudinal versus in modiolus direction.

4.3. alternative Calculations

An alternative approach to estimate impedances and currents was proposed as follows.

Calculation of the impedances (longitudinal impedance and interface impedance) calculated as described in chapter 4.1.2 and 4.1.3.

For each electrode, the equivalent network for the neighboring four electrodes was calculated. To make this calculation feasible we made some simplifications:

- All 4 point impedances are set to be the same, namely the average of the calculated 4 point impedances.
- All intracochlear electrode interface impedances are set to be the same, namely the average of the calculated interface impedances,
- The interface impedance of the external electrode is estimated to be one hundredth of the average intracochlear interface impedance. This estimate is based on the assumption that the interface impedance is inversely proportional with the surface area of the electrode.
- For the calculation of each modiolus impedance, we further assume that the adjoining modiolus impedances are all the same.

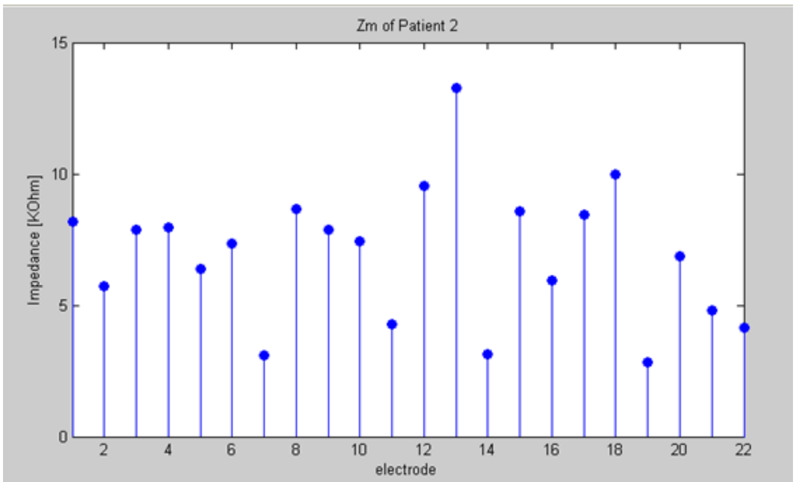

Figure 20: Impedance in the Direction of the Modiolus

With these assumptions, we can establish a set of equations to calculate the modiolus impedances by equating the calculated equivalent network with the measured monopolar voltages. Some results from this calculation are plotted in figure 20. These results on their own are not of great significance due to all the simplifications, but they could serve as starting point for a series of iterative calculations to estimate more precise values for the monopolar impedances. With these impedances, the whole network would be determined ,and we could easily calculate the current distribution.

5. Use of Assumptions

In the Modiolus Currents Model (chapter 3), which is based on [11], and in the subsequent calculations, several assumptions and simplifications have been made, some of which are described in the following chapter.

5.1. Distribution of Currents

For the calculation of the 4 point and interface impedances, (chapter 4.1.2 and 4.1.3) we assume that the current flows only on the indicated path. (marked in red, both in Fig. 10 and 13). This assumption is based on the fact that the impedances in the direction of the modiolus and the interface impedances are much larger than the impedances along the fluid in the cochlea, (ratio $\sim$ 100:1) and this has been verified by network calculations. However, the model represents a simplification of the implant in the cochlea, and therefore, a larger amount of current seems to spread. As our calculation of the impedances does not take this inaccuracy into consideration, the impedances are subject to inaccuracies. The importance of this inaccuracy has not been evaluated.

5.2. Impedance Calculation

According to chapter 1.3, our measurement of the impedances takes into account only the linear, ohmic part of the complex impedance and makes the assumtion that the capacity is fully charged at the instant of measurement. For one, this assumption is inexact. Also, the impedances are nonlinear. The nonlinearities of the impedance originate from the following mechanisms. One factor is the temporary ionic displacement, which induces a capacitive component [8]. Other factors may have a bearing, which are still subject to investigation. The nonlinearity of the calculated impedances leads to a dependance of the impedance values on the current level, and on the temporal dynamics. Therefore, the measured impedances may be subject to change depending on the instant of measurement.

The assumption that the impedances consist only of their ohmic part, leads to some errors. A possible approach to determine the capacitive part is discussed in chapter 9.

6. Experiment Setup

To ensure simple and efficient conduction of the necessary measurements, we implemented an executable python program.

The program, which is based on the NIC (2.1) software packet, is designed to work as a stand-alone program on any computer connected to the network of the University Hospital Zurich. A GUI (graphical user interface, see Fig. (21)) provides the possibilities for the user to enter the patient's name, the side of the implant (left or right ear) and choose the current level for stimulation. Pushing the button "Manual Run" starts the measurement at the chosen current level, pushing "Automatic Run" starts measuring at current level 60, 80 and 100 consecutively.

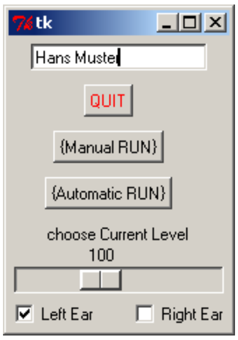

Figure 21: Python GUI

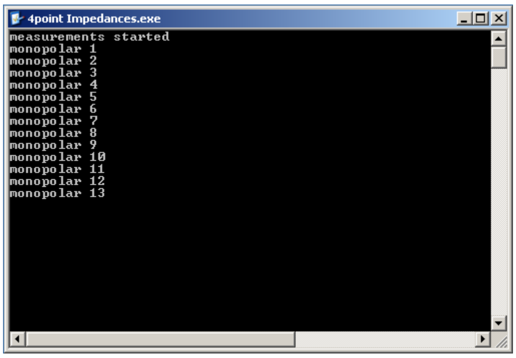

Figure 22: Screenshot of Measurement Interface

The program communicates with the implant by using the Freedom Programming Pod as programming interface, which in turn is connected to the freedom processor of the test subject, as can be seen in Figure (23). For further information about the NIC software constraints, consult [4].

A measurement session with each test subject consists of a series of measurement

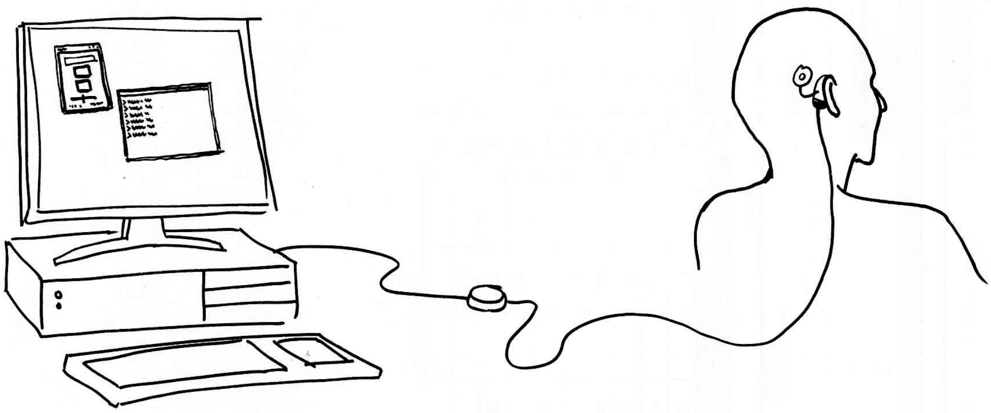

Figure 23: Experiment Setup

runs at different current levels, the usual current level being beteween the values of 60 and 140. At the moment, one measurement run takes about 160 seconds.

Part III.

Results

7. Experiment Results and Discussion

With the illustrated experiment setup, we performed measurement cycles with ten patients with a total of 16 implants, thereof one measurement cycle was performed intraoperatively and one implant subject was tested on two different dates. A list of all measurements is attached in Appendix C.

Each measurement cycle consisted of (at least) three measurements at three different current levels. Each measurement consists of the following measurements:

- Construction of a monopolar Matrix: measure the voltage between each electrode and the extracochlear electrode and repeat this while applying a current between each electrode and the extracochlear electrode separately.
- Measure the voltage between the stimulated electrode and the extracochlear electrode once again with a different gain to ensure that the voltage is in the measurement range.
- Measure and calculate the 4 point impedances.
- Measure the 3 point impedances and then calculate the interface impedances.

Hereafter, all data is saved in text files for further analysis.

7.1. Matlab GUI

To facilitate the analysis of the measurements, a Matlab Gui was implemented. Immediately after each measurement, the Matlab Gui loads the data files, processes them if necessary, and displays the data as graphs in various modes.

In Figure 24 a screenshot of the Matlab GUI, displaying two different graphs, is shown.

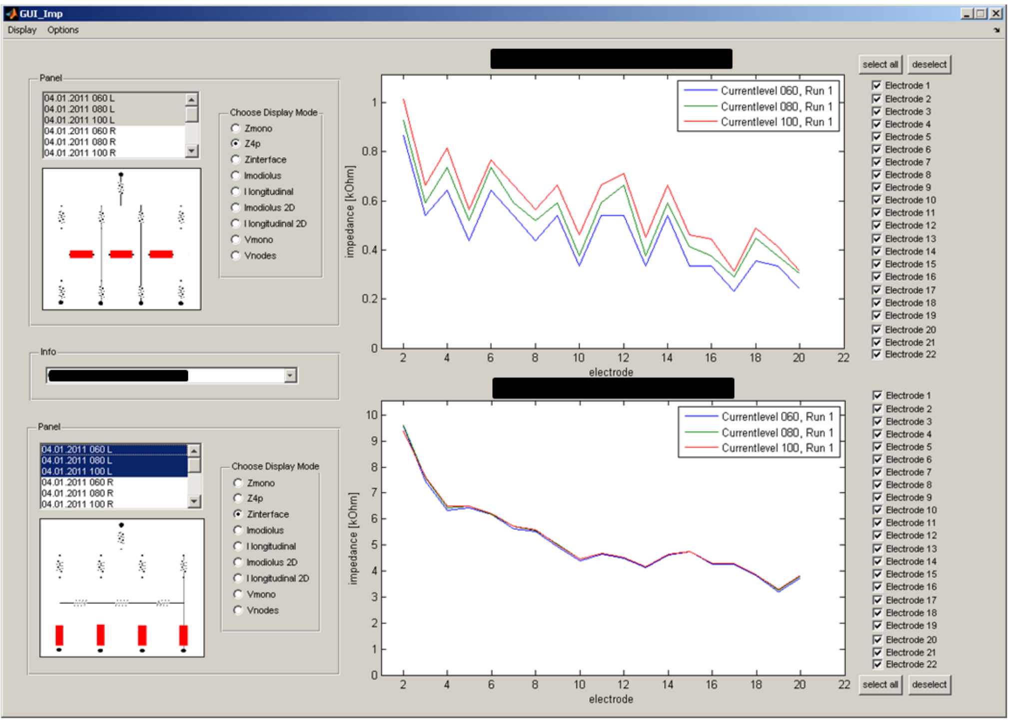

Figure 24: Matlab GUI

7.2. Case Studies

Out of all 10 implant subjects, we want to inspect two cases exemplarily. The first subect (in this chapter referred to as Patient 1) shows rather anticipated values and value distribution, while the other subject (referred to as Patient 2) has particularly unusual characteristics. We will discuss both the anticipated measurements and the possible explanations of observed irregularities.

When viewing the monopolar impedance measurements of Patient 2 (Fig. 26), the impedance values of electrode 6, and also those of electrode 1 and 2, stand out. While most other impedance values are very similar, the monopolar impedance of electrodes 1, 2 and 6 are all above 20kΩ. In contrast, the monopolar impedances of Patient 1 (Fig. 25) all reside in the same range of values, between 7 and 16 kΩ.

Similarly, the longitudinal impedances of Patient 1 (Fig. 27) distributed evenly. In Figure 28 we notice that the impedances around electrode 6 are rather low, while the impedances around electrode 1 and 2 are rather high.

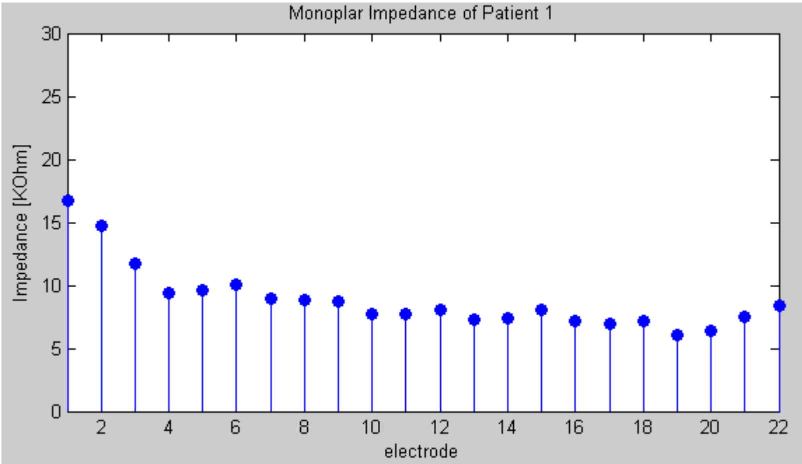

Figure 25: Monopolar Impedance of Patient 1

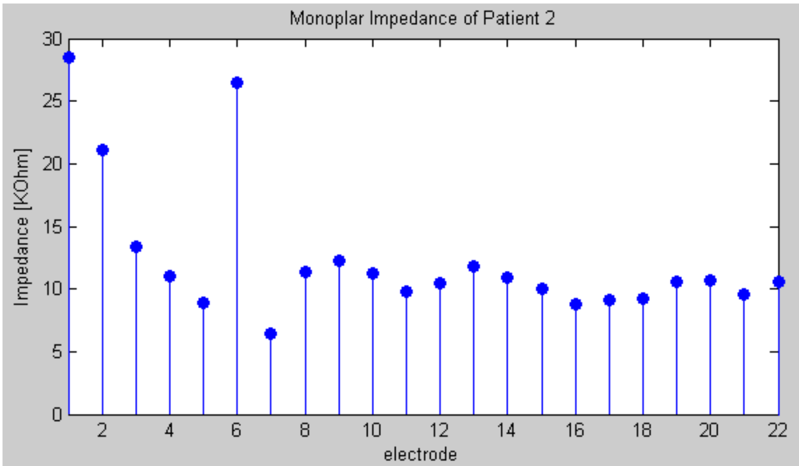

Figure 26: Monopolar Impedance of Patient 2

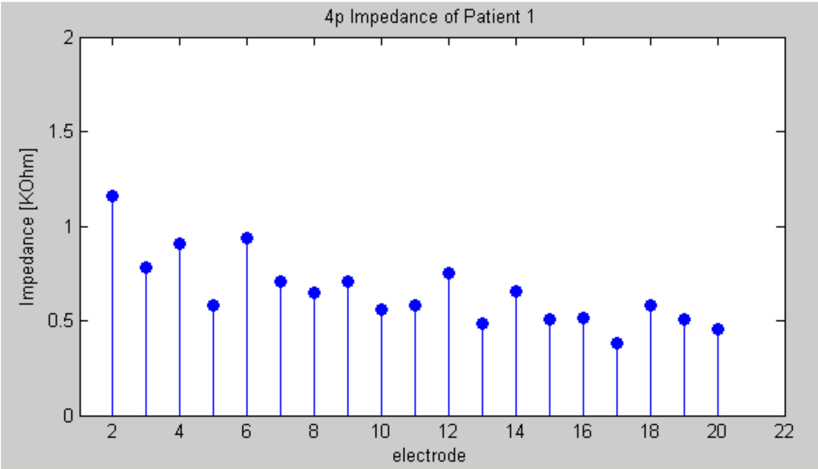

Figure 27: 4 Point Impedance Measurement of Patient 1

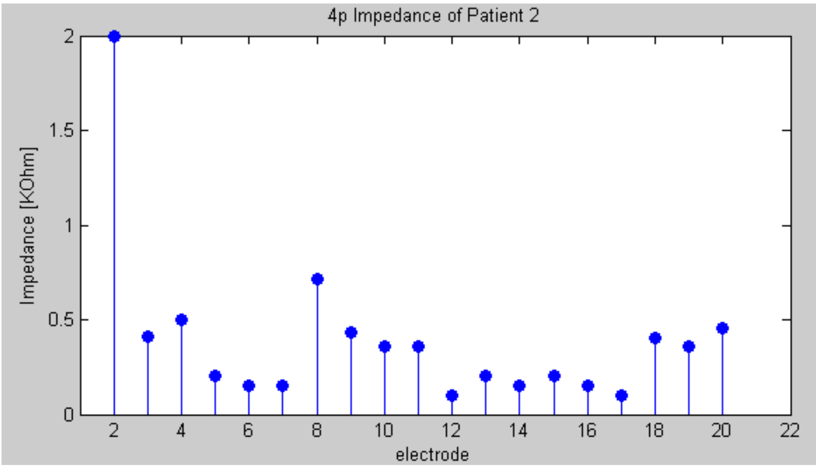

Figure 28: 4 Point Impedance Measurement of Patient 2

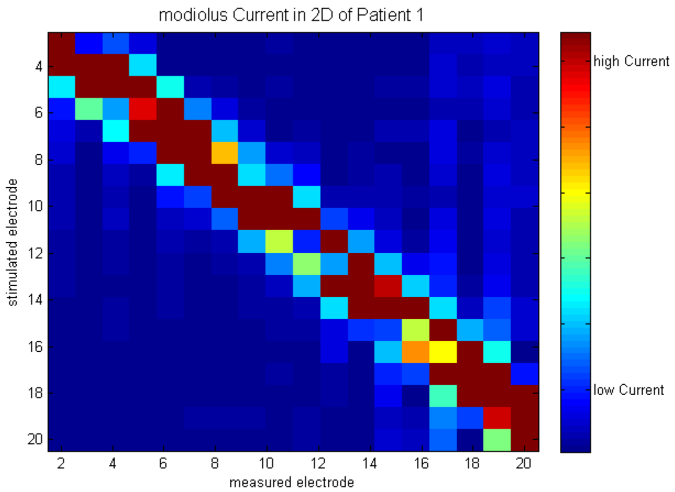

Figure 29: Modiolus Currents of Patient 1

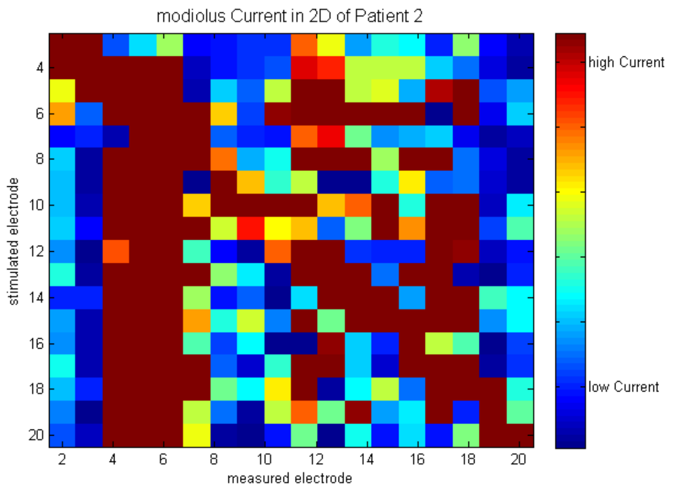

Figure 30: Modiolus Currents of Patient 2

Comparing the radial currents in direction of the modiolus of Patient 1 (Fig. 29) and Patient 2 (Fig. 30) the differences are very obvious. Patient 1 has a very evenly allotted current distribution. The current is applied to each electrode successively (denoted as "stimulated electrode" on the y-axis, each row corresponds to one stimulation pair) and each time the current distribution is calculated (denoted as "measured electrode"). We expect the highest current to be on the diagonal and that the amplitude of the currents declines to both sides, creating a diagonal pattern, as can be seen in Figure 29. In Figure 30 we still see the expected diagonal pattern but there is also a lot of unexpected noise. On one hand, we see high current levels seemingly randomly distributed. On the other hand, there is a distict accumulation of unexpected high current levels at electrode 6. Electrode 6 seems to act as a sink, as a lot of current flows in the direction of the modiolus at electrode 6 regardless of the stimulation electrode. Prior our measurements Patient 2 had wished to turn off electrode 6 what improved the patients comfort.

7.3. Current Level Determination

NIC (chapter 2.1) introduces a system to select a specific current for electrode stimulation. A scale of currents in amperes is assigned nonlinearly to values between 1 and 255, 1 being the lowest current, according to equation 3.

$$I[amperes] = 17.5 * 100^{\frac{current-level}{255}} * 10^{-6} \tag{3}$$

There are various contraints to keep in mind if we consider which current level suits our purposes best. For one, the signal to noise ration is augmented at higher current levels, which is favorable. Also, we must consider the acceptable level for the patient. Futhermore, the choice of gain is connected to the choice of current level, which prohibits a high current level.

Through our measurements we can conclude as follows.

- At current levels below 80, (corresponds to 74.2 microamperes) the signal is predominated by noise, and therefore, less favourable.
- Current level 140 (corresponds to 219.3 microamperes) is still acceptable for all patients, however the stimuli are audible for most patients. In intraoperative measurements, higher current levels can be applied.
- The measured voltage equals the current multiplied by the impedance. As the impedance is constant, the voltage increases proportionally with the current level, therefore the higher the selected current, the higher we need the measurement range to be. The range and the accuracy of the measurement are both regulated by the gain, but are inversely proportional. We tried to avoid this problem by applying different gains for different measurements at different current levels (as can be seen in the program in Appendix B).

Thus, we mostly used current levels between 80 and 140, modifying the gain accordingly. For instance, a current level of 140 and a gain of 0.4 (for monopolar impedance), 1.0 (for 3 point measurement) and 2.0 (for 4 point measurement and monopolar voltage matrix) is a viable calibration.

7.4. Current Calculation

As already mentioned, (see chapter 4.2.1) the major problem we encountered is in the accuracy of the values of the longitudinal currents, and consequently, also the values of the currents in direction of the modiolus. The current in longitudinal direction have values that seem to be 1-fold to 10-fold of the expected values.

For the calculation of the current in longitudinal direction we use equation 2 on page 18. In the following paragraph the error is described and its origin analyzed by looking at all components of the calculation seperately.

$\Delta V = V_2 - V_1$ The accuracy of the voltage measurements underlies limitations of the voltage telemetry precision. If we deduct one of these voltages from the other (as is done in equation 2) , these limitations become influential. Let's say Vnode underlays an error of 0.5% of its value, which corresponds to between 0.001 and 0.01 Volts. The calculated difference between two adjecent monopolar voltages is between 0.001 and 0.01 Volts. Therefore we can conclude if

$$V_{measured} = V_{accurate} \pm Error$$

and "Error" is in one case + 0.01 Volts and in the other -0.01 Volts, then

$$\Delta V_{measured} = (V_{accurate-2} + 0.01V) - (V_{accurate-1} + 0.01V) = \Delta V_{accurate} + 0.02V$$

In this case the error is at least in the same range as the accurate $\Delta V_{accurate}$. Therefore the resulting $\Delta V_{measured}$ is $\Delta V_{accurate}$ multiplied by a factor of up to 20.

Z_l We calculate Z_l by using Ohm's Law $R = \frac{U}{I}$. U is measured by voltage telemtry and, as before, assumed to underlie an error of 0.5%. I is the applied current. As is explained in chapter 23 the actual current flowing trough the resistance may be lower than the applied current due to dispersion of the current. Consequently, if 20% of the current disperses

$$R_{measured} = \frac{U(1 \pm 0.005)}{I(1 \pm 0.2)}$$

leading to $R_{measured}$ being 0.83 times the actual value $R_{accurate}$.

We conclude that

$$I_{measured} = \frac{\Delta V}{Z_l} \leq 24 * I_{accurate}$$

which exceeds the error we observe in our measurements by far.

7.5. Changes over Time

As impedances change strongly after the first stimulations of the electrodes ([8]), comparing intraoperative measurement with measurements made later on could reveal correlation between the change in impedances, fitting parameters and the current distributions. In the scope of this study, only one intraoperative measurement has been done, and the first postoperative measurement will take place after this master thesis has been completed.

Another possibility to analyze changes over time is to schedule experiments with the same test subject on a regular basis. The duration of this study did not allow many repetitions of experiments after a couple of months, therefore, measurements at two different dates were done only with one subject, Patient 0 with a two month period between the two measurements.

For the evaluation of the changes in the measured impedaces we compared the mean and standard deviation of the change in the monopolar impedance versus the mean and standard deviation of the change of the 4 point impedance measurement. The analysis revealed no significant differences in variation between the two impedance measurements, the monopolar and the 4 point measurement. The mean of the variation of the monopolar impedance is 1481 Ω, which equates and avearaged 14% change, the mean of the 4 point impedance measurement variation is 116, which equates an averaged 14.75% change. The standard deviations meet 0.62% (monopolar impedance) and 0.63% (4 point measurement). For comparison: the averaged mean deviation of two measurements on the same day is approximately 0.1-5%.

Although the small number of iterated patient measurements does not allow a conclusive statement, we can conclude that the changes over time in impedances other than the monopolar impedance seem to be consistent with the changes in the monopolar impedance. Therefore the change in measurements appears to be due not to measurement errors but to changes in the cochlea.

8. Conclusion

In this study, we succeeded in improving the impedance measurement procedure by applying a more detailed model of electrical impedances within the cochlea [11]. A more sophisticated measurement protocol was developed to identify the model parameters in a given implant subject and to reach conclusions about its properties. In particular, we succeeded in calculating the impedances along the liquid in the cochlea of an implant subject, and in evaluating the range of values of such an impedance. Also the impedance of the tissue/electrode interface have been calculated. And finally, the modiolus curents, the portion of the currents assumed to be responsible for activating the neural elements, were determined.

We came across different challenges, several being based on the undeterminability of the model network. Therefore, the calculation had to be based on a series of assumptions and simplifications. This leads to imprecision, for instance, in the calculation of the currents in longitudinal direction, a problem that has not been solved as yet.

Further, an experiment setup was implemented, whereby measurements with 15 implants were carried out postoperatively and with one implant intraoperatively. This allowed us to investigate and analyze the data and to draw first conclusions. In the next chapter, various suggestions concerning continuative research are made.

9. Future Work

The study at hand explains how new measurements concerning impedance and currents in the cochlear have been made for the first time, and first results were obtained. This sets the starting point for further investigations, a few of which are proposed followingly.

For instance, more measurements could be carried out over time investigating the changes and their significance. By obtaining measurements on a regular basis during several months or years, a statement about the relation between different quantities and distributions could be made. This approach could be combined with investigating measurements over different stimuli.

To reach an understanding about the capacitive nature of the impedances, a different measuring point could be selected. Instead of measuring at the trailing edge of the simulus' first phase, as is done in this study, we could measure at the leading edge. The difference of these two voltages gives evidence about the capacitiv part of impedances. [7]

In the future, the deepened understanding of the changes in the proposed (4 point and 3 point) impedances and current distributions (in longitudinal direction and in direction of the modiolus) may lead to the quantitative knowledge about their predictive value for device fitting parameters. The most crucial parameter of fitting for a cochlear implant is the establishment of the lowest and highest usable stimulation level for each electrode in the array. [1]The lowest stands for the minimum stimulation level that the subject can detect, referred to as T(Threshold)-Level. The highest stands for the loudest comfortable stimulus level, referred to as C(Comfort)-Level. Classical fitting can be very time-consuming, and hence, the alleviation, that future research discussing the current distribution could provide (already proposed in [11]), possesses very high significance.

Part IV.

Appendix

Bibliograpy

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A. Equations and Laws

A.1. Ohm's Law

Ohm's law states that the current through a conductor between two points is directly proportional to the potential difference or voltage across the two points, and inversely proportional to the resistance between them.

The mathematical equation that describes this relationship is

$$I = \frac{V}{R}$$

while I is the current through the conductor in units of amperes, V is the potential difference measured across the conductor in units of volts, and R is the resistance of the conductor in units of ohms.

(ref: (Robert A. Millikan and E. S. Bishop (1917). Elements of Electricity. American Technical Society. p. 54.)

A.2. Kirchhoff's Current Law

The current law from Kirchhoff describes that at any node in an electrical circuit, the sum of currents flowing into that node is equal to the sum of currents flowing out of that node. Or that the algebraic sum of currents in a network meeting at a point is zero.

This principle can be stated as:

$$\sum_{k=1}^{n} I_k = 0$$

while n is the total number of branches with currents flowing towards or away from the node.

C. Test Subjects

Initials	Implant Type	Implant Side	Date of Implantation	Number of Measurements
CH	CI512	L	14.10.2010	3
GZ	Freedom CA	R	21.10.2005	3
GZ	Freedom C	L	13.05.2003	3
RS	Freedom CA	R	12.05.2006	3
RS	Freedom CA	L	14.09.2007	3
EBu	Freedom CA	R	14.12.2005	3
Ebu	Freedom C	L	02.04.2003	3
HR	Freedom C	R	25.08.2004	5
WL	Freedom CA	R	27.12.2005	4
KJ	Freedom CA	R	10.08.2004	6
KJ	Freedom CA	L	02.08.2007	3
EBo	CI512	R	03.12.2010	5
GB	CI512	L	25.2.2011	3
CE	CI512	R	25.11.2009	3+4
CE	Freedom CA	L	21.09.2004	3+4

E. Master Thesis Files

E.1. Project Description

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Institute for Biomedical Engineering

MASTERS THESIS
D-ITET, HS 2010/FS 2011
Rahel von Rohr

COCHLEAR IMPLANT IMPEDANCE TELEMETRY MEASUREMENTS AND MODEL CALCULATIONS TO ESTIMATE MODIOLAR CURRENTS

1. INTRODUCTION

Today's cochlear implants can generate stimulation patterns and record electrical signals through the same implanted electrodes (Dillier et al., 2002; Patrick et al., 2006; Zeng et al., 2008). The use of impedance and neural response telemetry measurements can facilitate device fitting and parameter adjustments, especially in young children. It is well known, however, that subjective hearing thresholds and objective electrophysiological thresholds are poorly correlated and that variations of these measures may occur over time whose origins are not well understood.

Variations of electrical impedances over time have frequently been observed which may indicate changes of the effective electrical field distribution within the cochlea and subsequently changes in neural excitability due to modified current flow in the vicinity of auditory nerve fibers (Paasche et al., 2006). The detailed configuration of electrical impedances and current distributions around the electrode array is unknown and has not been able to be determined using standard impedance telemetry measures.

This master thesis project therefore attempts to improve the impedance measurement procedure by applying a more detailed model of electrical impedances of stimulation and recording electrodes within the cochlea and developing more sophisticated measurement protocols to identify the model parameters in a given implant subject.

In particular, the main interest consists of identifying the modiolar currents thought to be responsible for neural excitation and separating them from longitudinal shunting currents which do not contribute to the generation of action potentials. The approach is mainly based on a published patent (Tsampazis et al., 2009) and makes use of the Nucleus Matlab Toolbox and the Nucleus Implant Communicator (NIC) software. The ultimate goal of a refined model and more specific impedance measurements would be to allow semiautomatic fitting and updates of speech processor map parameters for CI users with varying electrical stimulation conditions.

2. TASK LIST

- Write a detailed time-table of the work to be performed
- Review the relevant literature
- Evaluate potential solutions/methods to address the aims of the project
- Choose and implement one solution/method using Matlab/NIC

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- Perform a validation or verification of the obtained results using theoretical and experimental methods
- Write a detailed report of the project

3. LITERATURE

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Masters Thesis, HS 2010/ FS 2011

Start of Project: 01.10.2010
End of Project: 01.04.2011

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