MASTER

Electrophysiological measurements with cochlear implant recipients

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Preface

This report gives information about my master project at the E.N.T.-clinic of the University Medical Centre Nijmegen. During this project I examined the possibility to perform several electrophysiological measurements with cochlear implant recipients using the Nucleus Interface Communicator.

When I started this project I did not have any experience with cochlear implants, I did not even know what they were. But after reading some literature and seeing some users, I became little by little acquainted with the world of cochlear implants. Now, at the end of my project I'm thinking of starting a career in audiology so you could say that the 'new world' really attracted my attention.

I really enjoyed everything I did and can look back on a successful project. For that reason I want to thank some people. First I like to thank my direct supervisors Andy Beynon and Leo Vogten for all their comments, guidance and suggestions. I also want to thank Bas van Dijk and Matthijs Kiliain from Cochlear Technology Centre Europe for their help during my project, as well as Herbert Mauch and Colin Irwin from Cochlear Corp. for their technical support.

Last I want to thank all people from the Otorhinolaryngology department of the University Medical Centre Nijmegen for all their assistance.

Thijs Thielemans
Abstract

When a patient suffers from severe hearing loss and a hearing aid does not lead to a clear improvement, a cochlear implant (CI) could partially restore hearing. This device artificially stimulates the auditory nerve in the cochlea using electrode-signals diverted from the (surrounding) sounds. After intensive habituation and training, these patients can in general communicate reasonably through speech. However, sometimes speech understanding remains insufficient. This can be caused by malfunction of the implant or malfunction of the more central parts in the auditory pathway. To examine this, a software-application is developed at the University Medical Centre Nijmegen with which the implant as well as the central parts in the auditory pathway can be examined.

The implant system (Nucleus 24) is examined in two ways. (1) The voltages on the CI electrodes are measured with EEG electrodes placed on the scalp. Results from five subjects reveal that it is possible to measure CI electrode voltages and identify possibly defect CI electrodes. (2) The tonotopy of the CI electrodes is also examined. In this experiment each individual CI electrode is stimulated in random order and after each stimulus the patient must assign the observed pitch with a value between 0 and 100. Results from one subject show a tonotopic order between all CI electrodes.

The central parts in the auditory pathway are examined by measuring the auditory evoked responses (AER). In this experiment different short sound stimuli are used to evoke typical electric responses in the auditory pathway. The AER-signals are also measured using EEG electrodes on the scalp. Deviant responses can possibly point to central parts malfunction. The stimuli on the CI electrodes as well as the radio-frequent (RF-) signals, used for information transfer to the implant, cause unwanted signals on the EEG electrodes. For that reason the AER-signals are difficult to interpret. A mathematical method is introduced to remove the unwanted artefacts. Results of two subjects show unexpected and inexplicable changes of the DC-component. Therefore, the proposed method can not be used.
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Introduction

The human cochlea (inner-ear) of totally deaf patients can be artificially stimulated by means of a cochlear implant (CI), a device to stimulate the auditory nerve via an electrode array surgically placed in the scala tympani. Sounds are processed and transcutaneously transferred by a high frequency radio carrier (electromagnetic induction with transmitting/receiving coils) and delivered as electrical current pulses to the electrodes.

At the University Medical Centre Nijmegen, considerable experience has been acquired with the 'Nucleus 24-channel' cochlear implant system developed by Cochlear Corp. Recent research with this CI-system has been focused on two aspects: (1) the integrity of the implant and (2) experiments to study the signal processing in the auditory pathway of CI recipients.

The integrity of the implant is examined in two ways. The voltages of the implant electrodes can be measured in order to find electrode malfunctions. The tonotopy of the implant electrodes can be examined in order to determine pitch perception of the recipient for the individual electrode stimulations.

Signal processing in the auditory pathway can be examined by measuring the auditory evoked responses. In these experiments short sound stimuli (e.g. clicks, tones or parts of speech) have been used to evoke typical electrical responses of the cochlea nerve, the brainstem as well as cortical responses. This report focusses on cortical responses. The responses are measured on the scalp of a patient using surface EEG electrodes. However the responses are difficult to identify because the electrical current pulses of the implant electrodes and the radio frequency signal contaminate the measured signal. Although adequate filter paradigms might solve this problem, the impact of these artefacts aggravates a reliable and correct interpretation. A method is proposed to reduce the artefacts.

In order to perform integrity tests and auditory evoked response experiments, a software application is necessary which is able to stimulate individual CI electrodes. The standard Cochlear Corp application is not able to stimulate the electrodes in the desired way, therefore a new clinical software application is developed. Communication between the developed application and the Nucleus '24 channel' implant is performed by the Nucleus Interface Communicator (NIC), a software library recently developed by Cochlear Corp.
The aims of the project are threefold:

- Develop a software application that is able to stimulate the cochlear implant in such way that integrity test and auditory cortical response experiments can be performed.

- Test the developed software by performing experiments with CI recipients, using the application.

- Develop a method to reduce the artefacts occurring in the surface EEG electrode signals, caused by the RF radio carrier and the current pulses of the CI electrodes.

This report consists of five chapters. Chapter one describes the anatomy of the ear as well as the physiology. Chapter two describes the Nucleus 24 cochlear implant system. Chapter three focusses on the integrity tests. Chapter four describes the auditory cortical response experiments. Finally, in chapter five conclusions are drawn and recommendations are made.
Chapter 1

The Ear

The ear is the organ of hearing and equilibrium. It contains both receptors that respond to movements of the head and receptors that convert sound waves into nerve impulses. Impulses from both types of receptors are transmitted via the vestibulocochlear cranial nerve to the brain for interpretation. This chapter describes the anatomy of the ear. Only the functional parts for hearing and the physiology of hearing will be mentioned.

1.1 Anatomy

The ear of a human adult consists of three structural and functional divisions, the outer ear, the middle ear and the inner ear (figure 1.1).

![Figure 1.1: The anatomy of the human ear [1].](image-url)
The outer ear and middle ear are responsible for the conduction of sound stimuli whereas the inner ear takes care of the stimulus processing. All different parts are described in the following sections [2].

1.1.1 The Outer ear

The outer ear consists of the auricle (or pinna) and the external auditory canal (figure 1.1). The auricle is made of a cartilaginous framework covered by skin, and is attached to the underlying temporal bone. The external auditory canal consists of an outer cartilaginous portion and an inner bony portion, and is lined with skin. In the cartilaginous portion the skin lining contains hairs and sebaceous and ceruminous glands (earwax-secreting glands).

The outer ear carries out two physiological functions: an acoustic and a non-acoustic function. The acoustic function allows efficient sound transmission from the environment to the middle ear. It alters the amplitude of the incoming sound wave and, in doing so, provides a mechanism for amplifying sounds within the range of frequencies that make up human speech (from 5 to 20 dB over the frequency range from about 1.5 to 7 kHz). The non-acoustic functions of the ear canal include protection of the middle ear and the maintenance of a clear passage for sound.

1.1.2 The Middle ear

The middle ear (also called tympanic cavity) is a narrow air-filled cavity located in the temporal bones of the skull. It is separated from the external auditory canal of the outer ear by the tympanic membrane (figure 1.2). This membrane is cone-shaped, with its concavity facing downward and outward toward the auditory canal. Attached to the center of the tympanic membrane is the ossicular system consisting of three different ossicles, the malus, the incus and the stapes (figure 1.2).

Figure 1.2: The different parts of the middle ear; Left, the tympanic membrane [3]. Right, the malus (m), the incus (i) and the stapes (s) [4].
1.1 Anatomy

The function of the ossicles is to transmit and amplify sound waves across the tympanic cavity from the tympanic membrane to the oval window, where sound waves are conducted into the inner ear. The ossicles are connected in such a way as to act as a lever system to increase the force of the vibration from the tympanic membrane. In addition the force of vibration is intensified as it is transmitted from the relatively large surface of the tympanic membrane to the smaller surface area of the oval window. The combined effect increases the pressure roughly twenty times [5].

To protect the ear against very intense sound two muscles are attached to the ossicular system, the tensor tympani muscle attaches to the neck of the malleus, and the stapedius muscle attaches to the neck of the stapes.

1.1.3 The Inner ear

The inner ear (also called the labyrinth) contains the organs of both hearing and of equilibrium. It consists of two parts: the outer osseous (bony) labyrinth (figure 1.3) and a membraneous labyrinth contained within the osseous labyrinth and made up of interconnected sacs and tubes.

![Figure 1.3: The osseous bony labyrinth [6]; The semi circular canals are the organs of equilibrium, the cochlea is the organ of hearing.](image)

The space between the the bony labyrinth and the membraneous labyrinth is filled with the perilymph, a fluid secreted by the cells lining the bony canals. This fluid is similar to intracellular fluids. The tubular chambers of the membraneous labyrinth are filled with a second fluid, known as the endolymph. This fluid is similar to extracellular fluid.
1.1 Anatomy

The portion of the inner ear that is responsible for hearing, is called the cochlea. The cochlea is shaped like a snail shell (figure 1.4).

![Figure 1.4: Left, the "snail shell" cochlea; Middle, a cross-section of the cochlea; Right, the cochlea duct (6).](image)

The cochlea winds two and three quarters turns around a central bony axis, the modiolus. Projecting outward from the modiolus is a thin bony plate, the spiral lamina (1) which partially divides the cochlear canal into an upper passageway called the scala vestibuli (2) which originates at the oval window and a lower one called the scala tympani (3) which terminates at the round window. Both of these are filled with perilymph and are separated except at the very narrow apex of the cochlea, an area called the helicotrema. In between these canals there is the triangular passageway called the scala media (6). The scala media is filled with endolymph and terminates at the helicotrema. The roof of the scala media is called the vestibular membrane (4) while its floor is called the basilar membrane (8).

The basilar membrane differs in thickness from the base of the cochlea to its apex. At the cochlear base the basilar membrane is approximately 0.16 mm wide, as at the apex the membrane has broadened to about 0.52 mm [2]. This variation in width results in a systematic change in stiffness from base to apex. Therefore the resonant frequency of the basilar membrane varies along its length. The basilar membrane near the oval window of the cochlea is able to vibrate best (resonance) at a high frequency. The long fibers near the tip of the cochlea vibrate best at low frequency.

On the surface of the basilar membrane lies the Organ of Corti (5). The Organ of Corti (figure 1.5) contains the sound receptors that transduce mechanical vibrations into nerve impulses. It is the functional unit of hearing. The sensory cells are classified as inner hair cells (9) and outer hair cells (10). There are roughly 4 times as many outer hair cells as inner hair cells. The inner hair cells lie in a single row along the length of the basilar membrane. They are completely surrounded by supporting cells. The outer hair cells are cylindrical and lie in 3-5 rows along the basilar membrane. Only their apical and basal surfaces are surrounded by supporting cells; fluid
bathes the medial surfaces. Each hair cell has many stereocilia on its apical surface. Inner hair cells have fewer stereocilia than outer hair cells. The stereocilia become progressively longer on the side away from the modiolus, and the tops of the shorter stereocilia are each attached by a thin filament to the side of its adjacent longer stereocilium. The stereocilia of the outer hair cells are embedded in a gelatinous tectorial membrane (11). This membrane arises from the spiral limbus and extends over the outer hair cells to adjacent supporting cells.

Figure 1.5: The Organ of Corti. The numbers from figure 1.4 correspond with the numbers in this figure. [6]

Although the tectorial membrane extends over the top of the inner hair cells, their stereocilia are free; inner hair cell stereocilia are not embedded in the tectorial membrane. The base of each hair cell synapse with a network of cochlear nerve endings which lead to the spiral ganglion of Corti (7).

1.2 Physiology

Considering the physiology of the ear, the most important part of the ear, namely the inner-ear, must be emphasized. The outer ear and middle ear only transmit and amplify the sound stimuli, whereas the inner-ear fulfills the important task of processing sound stimuli. Therefore the description of the physiology starts at the moment a sound stimulus is transmitted from the stapes at the oval window. When the stapes moves inward against the oval window, the perilymph is pressed away towards the helicotrema and the base of the basilar membrane. However, the elastic tension that is built up in the basilar membrane as it bends towards the round window, initiates a wave that travels along the basilar membrane towards the apex of the cochlea. Each wave is relatively weak on the outset but becomes strong when it reaches that part of the basilar membrane that has a natural resonance frequency equal to the respective sound frequency.
1.2 Physiology

At this point, the basilar membrane will continue to vibrate at this resonance frequency. Consequently, the wave dies out and fails to travel the remaining distance along the basilar membrane. As the basilar membrane differs in resonance frequency along its length, the propagation of the wave is not the same for different frequencies. High frequencies are projected at the basal part of the cochlea while low frequencies are projected at the apical part. In other words, the basilar membrane has tonotopic order. The positions of different resonance frequencies of the basilar membrane measured from the oval window is shown in figure 1.6.

![Figure 1.6: The different positions of resonance frequencies of the basilar membrane](image)

As a result of the vibration of the basilar membrane, the inner and outer hair endings also move upward and downward. This means that the stereocilia make the same movement. Whenever the cilia are bent in the direction of the longer ones, the tips of the smaller ones are tugged outward from the surface of the hair cell. This causes a mechanical transduction that opens conducting channels, allowing rapid movement of positively charged potassium ions into the tips of the stereocilia. This causes depolarization of the entire hair cell membrane. This stimulates the cochlear nerve endings that synapse with the bases of the hair cells. It is believed that a rapidly neurotransmitter is released by the hair cells at these synapses during depolarization. Subsequently, an action potential is transmitted to the brain which has the task of interpreting the signal.

The nerve fibers in the cochlear pathway all the way from the cochlea to the cerebral cortex show spatial organization. In other words, if the nerve fibers at the base of the basilar membrane have been activated (high pitch sound), the action potentials will be sent to different parts of the cortex as when the nerve fibers at the apex of the basilar membrane have been stimulated. In this way, the nervous system is able to determine the positions of the basilar membrane that are stimulated and therefore determine the frequency of a sound. This is called the place principle for determination of sound frequency. Another principle for determination of sound frequency is called the frequency theory. It suggests that the auditory nerve fibers fire at rates...
1.3 Hearing Impairment

proportional to the period of the input signals (sound signals) up to frequencies of 5000 Hz. At low frequencies, individual nerve fibers fire each cycle of the stimulus. At high frequencies, frequency is indicated by the organized firing of groups of nerve fibers.

Loudness is determined in two ways [2]: first, as the sound becomes louder, the amplitude of vibration of the basilar membrane and hair cells also increases, so that the hair cells excite the nerve endings at more rapid rates. Second, as the amplitude of vibration increases, it causes more and more hair cells to become stimulated, thus causing spatial summation.

1.3 Hearing Impairment

Sometimes a person is not able to hear the sounds from the environment and suffers from hearing loss. This hearing loss can be small and some frequencies are not heard very loud but can be observed with some effort. The hearing loss can be severe and sounds with a complete range of frequencies can not be heard at all. Usually hearing loss is divided in two types [7]: (1) conductive loss and (2) perceptive loss. Conductive loss is caused by any affection of the conducting apparatus, the external auditory canal or the middle-ear. Perceptive loss is caused by any affection of the perceiving apparatus, the cochlea or the auditory nerve. Sometimes a mixture of conductive and perceptive loss occurs in one and the same ear.

If a person only suffers from a small hearing loss, a hearing aid could be a solution. This device is able to amplify the sound from the environment so the recipient can hear the sounds he usually hardly could hear. People suffering from severe hearing loss, often called profoundly deaf people, could be helped by a cochlear implant. This device is able to register the sounds and directly stimulates the cochlear nerves. In the next chapter the cochlear implant is described in detail.
Chapter 2

The Nucleus 24 Cochlear Implant System

For centuries, people believed that only a miracle could restore hearing to the deaf. This belief was cleared away about forty years ago when scientists first tried to restore normal hearing by electrical stimulation of the auditory nerve. Unfortunately the first results were negative, but whereas more research on auditory nerve stimulation was performed auditory sensations gradually came closer to sound more like speech. Today, a prosthetic device, called Cochlear Implant (CI), can be implanted in the inner ear and can partially restore hearing of profoundly deaf people. This device makes it possible for some deaf individuals to communicate more easily [8].

The cochlear implant is based on the idea that there are enough auditory nerve fibers left for electrical stimulation. Once the nerve fibers are stimulated, they fire and propagate neural impulses to the auditory cortex. The number of fibers activated is a function of the amplitude and duration of the stimulus current. The loudness of the sound can therefore be controlled by varying the amount of current. The pitch on the other hand is related to the place in the cochlea that is being stimulated. Low pitch sensations are elicited when electrodes near the apex are stimulated, while high pitch sensations are elicited when electrodes near the base are stimulated.

Several cochlear implant devices have been developed over the last years. All the implants try to imitate the normal functioning of hearing. Currently, several companies provide different cochlear implant devices commercially, e.g. Cochlear Corp [9], Advanced Bionics Corp [10], Symbion inc and Med-El Corp [11]. In this report only one type of cochlear implant has been used, i.e. the Nucleus 24 Implant system manufactured by Cochlear Corp. This implant system is used in many hospitals all over the world, including the University Medical Centre Nijmegen (UMCN). This chapter describes in detail the Nucleus 24 implant system.
2.1 Recipient-worn Components

The Nucleus 24 CI system has four components: a microphone, a signal processor, a transmission system and an implant. The sound from the environment is picked up by a sensitive microphone and transmitted to the speech processor. The main function of the processor is to decompose the input sound signal into different frequency components and deliver a coded form of each component to the appropriate electrodes. The CI system sends electrical signals through the skin using a transcutaneous transmission system with external transmitting coil and implanted internal receiving coil, i.e. information is encoded and decoded using a carrier frequency. The external transmitting coil and internal receiver are held in place on the scalp by a magnet. The electrical signal is transmitted to the electrodes of the implant. These electrodes are placed in the scala tympani (figure 2.1) in close proximity with the auditory neurons which lie along the length of the cochlea.

2.1.1 Implant

The CI 24 implant (figure 2.1) consists of a receiver coil, coupling magnet, receiver/stimulator, intra-cochlear electrode array and two extra-cochlear electrodes.

![Diagram of CI 24 implant](image)

Figure 2.1: Left: The CI24 implant consists of a receiver coil, coupling magnet, receiver/stimulator, intra-cochlear electrode array and two extra-cochlear reference electrodes [12]. Right: The intra-cochlear electrode array placed in the scala tympani.

The electrode array consists of 22 platinum electrodes. All 22 electrodes are connected independently to the receiver/stimulator by individual, insulated platinum-iridium wires. The electrodes are spaced equally along 17 mm of the array. The most basal electrode, the electrode closest to the round window, is called Electrode 1 and the most apical electrode is called Electrode 22.
2.2 Stimulus Generation

The CI24 has two extra-cochlear references. One is a small, platinum ball electrode on a separate lead and is referred to MP1. It is placed under the temporal muscle. The second reference is a platinum plate on the body of the receiver/stimulator and is referred to as MP2.

The receiver/stimulator includes an IC that decodes the digital information or data sent across the skin. The IC is inside a hermetically sealed ceramic package, which is surrounded by a hermetically sealed titanium shell protecting the IC.

The receiving coil consists of two turns of platinum wire with a diameter of 32 mm. This coil connects to the receiver/stimulator body and is secured firmly in place by a flexible silicone moulding, pre-formed into a 20° bend. The surgeon will place it against the skull. The coil receives power to power-up the IC.

The CI24 has a coupling magnet that attracts to the external magnet in the transmitting coil. The purpose of the two magnets is to hold the transmitting coil in the head over the internal receiving coil so data or power-up can be sent to the implant.

2.1.2 SPrint Speech Processor

During this project a body worn speech processor has been used, called SPrint™ (figure 2.2). The SPrint has an integrated microphone to pick up any sound from the environment. The microphone is placed behind the ear of the recipient. Also integrated is the transmitting coil of the transmission system.

![Figure 2.2: The SPrint speech processor with the integrated transmitting coil and microphone [12].](image)

2.2 Stimulus Generation

To be able to encode the acoustic signal, i.e. to translate the signal amplitude into electrical stimulation levels, different data parameters are programmed and optimized for each individual CI-recipient. These parameters are stored in the SPrint in a file called MAP. Appendix A shows an example of a patient MAP. The most relevant parameters will be discussed in the next sections.
2.2 Stimulus Generation

2.2.1 Stimulation Modes

Electrical stimulation produces current flow between an active and reference electrode. This pair of electrodes is called channel. The stimulation mode describes the location of the reference electrode relative to the active electrode. The Nucleus 24 cochlear implant system supports two stimulation modes: monopolar (MP) and common ground (CG) stimulation.

Monopolar stimulation

Within the monopolar stimulation mode, three different monopolar modes can be used, (1) MP1, (2) MP2 and (3) MP1+2, shown in figure 2.3.

Figure 2.3: The three monopolar stimulation modes; Top: MP1, Middle: MP2, Bottom: MP1+2. A = Active electrode, Arrow = current.

MP1 indicates that the current flows between the designated active intra-cochlear electrode and the ball electrode. MP2 indicates that the current flows between the designated active intra-cochlear electrode and the plate electrode. MP1+2 indicates that the current flows between the designated active intra-cochlear electrode and both extra-cochlear electrodes.
2.2 Stimulus Generation

Common Ground stimulation

In common ground mode, current flows between the designated active electrode and all other electrodes on the array, which are connected electronically to constitute a single reference electrode.

2.2.2 Electrical Stimulation

To stimulate the Nucleus 24 implant system, a 5.0 MHz Radio Frequency (RF) carrier signal is used to transmit data from the speech processor to the implant. The signal consists of RF frames, each frame including two RF bursts (figure 2.4a).

![Diagram of RF frame structure](image)

Figure 2.4: a) The encoded RF signal from the Nucleus 24 implant system. Each RF frame consists of two RF bursts, each containing three data tokens of 3 bits each.

b) The charge balanced biphasic current pulse, consisting of a negative phase, an interphase-gap and a positive phase. A sequence of two pulses is shown.

Each RF burst contains 3 data tokens and each token contains 3 bits of data. 5 cycles of the 5.0 MHz RF carrier are used to control each bit value. In this way it is possible to encode stimulation mode (5 bits), stimulated electrode (5 bits) and amplitude (8 bits) using a binary code. An example of a binary code of one RF burst containing 9 bits (101011010) is shown in figure 2.4a. The second RF burst is coded in a similar manner. The frequency content of a continuous RF signal is determined by the carrier signal, the RF frame period/frame rate and the gap between two RF bursts.
The transmitted binary code is decoded in the implant which stimulates the nerve using charge-balanced biphasic current pulses sent to the individual intra-cochlear electrodes. Figure 2.4b shows a sequence of two biphasic pulses, called frames. A sequence is defined as a train of pulses repeated several times. Each frame has certain amplitude, phase width and interphase-gap. The phase width is equal to the length of the RF burst, the interphase-gap is equal to the gap between two RF bursts. In this report only default values were used for the phase width and the interphase-gap (24.8 $\mu$s and 8 $\mu$s). During each negative and positive stimulation pulse (e.g. frame N-1), an RF burst, containing information about the upcoming biphasic current pulse (frame N), is transmitted to the implant. Because the pulses are charge balanced, there is an equal amount of charge passing in both directions. Thus at the end of the pulse, no net charge remains within the tissue or electrodes.

Before it is possible to actually stimulate the auditory nerve, the implant must have sufficient power. This power is obtained from the presence of RF. In order to boost the implants power supply, the processor will transmit RF power-up frames to the implant that do not result in any stimulation activity on the electrodes. These frames are also called "null" frames. Cochlear corp. recommends for any stimulation to use 300 power up frames at a frame rate of 1 kHz. Once powered-up, the receiver/stimulator is able to deliver stimulation pulses. During quiet periods (e.g. in between speech) RF power-up frames remain sent to the implant to maintain sufficient power.

### 2.2.3 Speech Coding Strategies

Since the first application of a cochlear implant, many signal processing techniques, called speech coding strategies, have been developed to stimulate the auditory nerve. The Nucleus 24 system implements the spectral peak (SPEAK), continuous interleaved sampling (CIS) and advanced combination encoder (ACE) strategies. These three strategies mainly differ in overall stimulation rate, stimulation rate per channel, number of channels and number of stimulation sites. The differences between SPEAK, CIS and ACE in the Nucleus 24 system are shown in table 2.1.

<table>
<thead>
<tr>
<th>Strategy</th>
<th>Nr of stimulated electrodes</th>
<th>Stimulation rates per channel</th>
<th>Nr of channels stimulated</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPEAK</td>
<td>20</td>
<td>250</td>
<td>6 to 10</td>
</tr>
<tr>
<td>CIS</td>
<td>4,6,8 or 12</td>
<td>900, 1200, 1800, or 2400</td>
<td>4,6,8 or 12</td>
</tr>
<tr>
<td>ACE</td>
<td>22</td>
<td>250, 500, 720, 900, 1200, 1800 or 2400</td>
<td>up to 20</td>
</tr>
</tbody>
</table>

Table 2.1: Number of (maximum) stimulated electrodes, stimulation rates per channel and number of channels stimulated for three coding strategies available in the Nucleus 24 implant system [12].
2.3 Recipient specific MAP parameters

In SPEAK, the audio bandwidth is logarithmically divided into 20 bands and each channel along the array is assigned to one of these frequency bands. The number of channels stimulated is between 6 and 10, but the actual number depends on the spectral composition of the signal. Each channel is stimulated with 250 frames per second. In CIS, the channel selection is fixed. Regardless the spectral content of the incoming signal, 4, 6, 8 or 12 channels are stimulated at a higher stimulation rate per electrode than in SPEAK. In ACE, the number of selected channels can range from 1-20. Typically, the number of stimulated electrodes will be greater than the number of channels so that the stimulated channels are not fixed. However the maximum stimulation rate of the Nucleus 24 system is 14400 Hz. Thus the number of channels is inversely proportional to the stimulation rate per channel.

Using the CIS strategy, rapid temporal variations in speech are better represented. With the SPEAK strategy, in which the spectral information of speech is better preserved. The ACE strategy combines the better temporal resolution with a good spectral resolution.

2.3 Recipient specific MAP parameters

2.3.1 Current Level (CL)

The loudness perceived by the recipient is determined by the amount of electrical charge delivered to the electrode. The product of two parameters, current amplitude and pulse width, controls the actual amount of electrical charge. Increasing the amplitude of the pulse and or widening the pulse width (controlled by the length of the RF bursts) results in more electrical charge being delivered at the electrodes. The greater the charge at the electrodes, the louder the perceived sound for the recipient.

Current levels (CL) are used as unit for amplitude. These levels are expressed in units from 1 to 255 and can be coded into 8 bits. Each unit represents a change of 2 percent (logarithmic scale) in current amplitude (10 μA -1750 μA). The conversion between current level (CL) and Current in μA is defined according equation 2.1:

\[ CL = \log_{10} \frac{Current}{175255} \]

2.3.2 Dynamic Range

The dynamic range of a recipient is determined by the threshold level called T-level and the comfort level called C-level. The T-level is defined as the lowest stimulus current level that elicits a very soft, but consistent hearing sensation. The C-level is defined as the maximum stimulus current level that does produce a comfortable loudness sensation for the recipient. Both levels are determined for every channel.
2.4 Aftercare and research components

After implantation, the speech processor has to be programmed for each recipient individually. The individual parameters, stored in a recipient-specific MAP, have to be optimized using psychophysical tests. To perform these tests with CI 24 implant recipients and to control and program the speech processor, a programming system has been developed by Cochlear Corp called Nucleus programming system. The main components of the Nucleus programming system are: (1) Hardware, (2) Software and (3) a SPrint speech processor (Figure 2.5). This system is also used for research. In the following sections all different components will be described.

![Block diagram of the Nucleus programming system.](image)

**Figure 2.5: Block diagram of the Nucleus programming system.**

### 2.4.1 Hardware

Cochlear Corp provides two hardware systems: the Clinical Programming System (CPS) and the Portable Programming System (PPS). The clinical programming interface system, designed for the clinical environment, consists of the following components (1) a circuit board (IF5 card) placed in an ISA slot of a desktop computer and (2) a processor control interface (PCI) connecting the IF5 card to the speech processor. It also includes a 22 LED display on the front panel, which correspond to the 22 intra-cochlear electrodes. The LEDs indicate which electrodes are stimulated.

The portable programming system consists of a small interface box which connects the speech processor to the computer through the serial communication port. It also has an external trigger output which is controlled by the software. This system is ideal with notebook computers because of its small size.
2.4 Aftercare and research components

2.4.2 Software

To conduct psychophysical testing, research measurements and clinical programming of speech processors, the following software programs and software libraries have been developed by Cochlear Corp.

WinDPS

*Windows®* Diagnostic and Programming System (winDPS) is the main software program. The principal function of the winDPS is to enable an audiologist to measure psychophysical parameters required for a MAP, to select a speech coding strategy, to adjust MAP parameters as required and to write a MAP into the speech processor and the patient database. It is also possible to perform a simple impedance measurement to identify electrode malfunctions. This program is mainly used in a clinical environment.

Nucleus Implant Communicator

The Nucleus Implant Communicator (NIC) is a software library, used for research applications [13]. The library contains approximately hundred functions [26] supporting hardware set-up, generation of stimulation sequences (section 2.2.2) and communication between the PC and the SPrint. Using the NIC library functions and the recipient MAP parameters, the audiologist is able to build a software application in which experiments can be performed that are not possible with winDPS. This allows the audiologist to bypass the clinical programming limitations of winDPS. The library can be implemented in the programming languages C, C++, Delphi and Visual Basic.

NMT

The Nucleus MATLAB Toolbox (NMT) is a software library. It contains approximately 250 MATLAB functions in order to code audio files (wav format), using the recipient MAP, into sequence data which subsequently is streamed directly (on line) into the SPrint. It also facilitates speech-coding and psychophysical experiment development.
Chapter 3

Integrity Tests

When a CI recipient has poor speech understanding after cochlear implantation, an audiologist usually decides to examine the integrity of the implant system. First a standard measurement of intra-cochlear electrode impedances is performed in order to check for shortcut or open circuits.

Additional information is provided by measuring the voltages between implant electrodes [15] produced by the biphasic current pulses. Current represents the energy presented to the auditory nerve. Sometimes hardware failure, e.g. broken wires, causes insufficient or no current to be delivered to the electrodes. As a result no stimulation of the nerves occurs and nothing is heard by the recipient. In this way incomplete spectral information reaches the recipient, causing speech understanding to reduce. Measuring the electrode voltages could provide information about the implant electrodes functioning. The electrode voltages are measured as far-field EEG signals using surface EEG electrodes on the head. To improve signal-to-noise ratio ten EEG signals are averaged, resulting in average electrode voltages (AEVs). Until now, previous studies only describe AEVs in the previous CI 22 system [14]

Integrity of a CI can also be tested by measuring pitch perception of the recipient. Usually, stimulation of implant electrodes with increasing number results in decreasing pitch perception. Stimulation of an apical implant electrode is perceived as a low pitched sound, whereas stimulation of a basal implant electrode results in a high pitched sound. Unfortunately, stimulation of adjacent implant electrodes does not always lead to the correct perception. In such a case one or more implant electrode(s) is (are) disconnected or reversed in order.

The standard electrode impedance measurement is performed with the main software program WinDPS from Cochlear Corp. Unfortunately, stimulus sequences to measure the AEVs and to run pitch perception experiments can not be created with the winDPS program. Therefore a software application is developed using the NIC software library in which it is possible to create these dedicated sequences. A detailed description of the application is given in Appendix B.2. The application is customized for clinical purposes for CI recipients who are implanted at the UMCN.
3.1 Averaged Electrode Voltage Measurements

The AEV measurement comprises two procedures, (1) a sweep procedure and (2) a step procedure. In the sweep procedure the implant electrodes are consecutively stimulated and the EEG signal is measured. When a CI electrode does not induce an EEG potential, the measurement is continued with the step procedure, in which the suspect electrode is stimulated with increasing and decreasing CLs steps as amplitude variations.

3.1.1 Setup

The setup for the AEV test is shown in figure 3.1.

![Figure 3.1: The experimental setup for the AEV test.](image)

The setup consists of a stimulation computer, the portable programming system (PPS), the SPrint speech processor, the implant of the CI recipient. Measuring the electrical activity on the scalp is performed with the Synergy System of Oxford Instruments [16]. The EEG signal was measured with three surface EEG electrodes placed on the scalp, based on the international 10/20 recording system [17] using a conductive water-soluble paste. The active EEG electrode was placed on the forehead at Fz. The reference EEG electrode was placed on the nose and a ground EEG electrode on the cheek. The EEG signal was amplified by a pre amplifier. A trigger signal, controlled by the stimulation software, was connected via the PPS to the Synergy system.

3.1.2 Subjects

Five subjects (n=5) participated in the AEV test. None of the five subjects had uncomfortable hearing sensations using the MAP parameters obtained with the clinical WinDPS sitting software. All five subjects had no complaints about the implant, speech understanding tests were on normal level. No further abnormalities were known. In table 3.1 information about the five subjects is shown. The comfort levels of all 22 electrodes for each subject are shown in Appendix D.
3.1 Averaged Electrode Voltage Measurements

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age [Years]</th>
<th>Month after Implantation</th>
<th>Stimulation Mode</th>
<th>Coding Strategy</th>
<th>Disconnected Electrodes</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>Male</td>
<td>55</td>
<td>9</td>
<td>MP1+2</td>
<td>ACE</td>
<td>Electrode 1, 2</td>
</tr>
<tr>
<td>B</td>
<td>Female</td>
<td>61</td>
<td>24</td>
<td>MP1+2</td>
<td>ACE</td>
<td>Electrode 11, 4</td>
</tr>
<tr>
<td>C</td>
<td>Male</td>
<td>57</td>
<td>10</td>
<td>MP1+2</td>
<td>ACE</td>
<td>Electrode 1, 2</td>
</tr>
<tr>
<td>D</td>
<td>Male</td>
<td>45</td>
<td>13</td>
<td>MP1+2</td>
<td>ACE</td>
<td>Electrode 1, 2</td>
</tr>
<tr>
<td>E</td>
<td>Male</td>
<td>45</td>
<td>12</td>
<td>MP1+2</td>
<td>ACE</td>
<td>Electrode 1, 2</td>
</tr>
</tbody>
</table>

Table 3.1: Information about the five subjects who participated in the AEV test.

3.1.3 Method

To prevent over-stimulation, first C-levels - as obtained with clinical WinDPS fitting software - of the subject were checked via NIC stimulation. When subjective comfort levels appeared to be different, new T and C levels were determined via NIC.

In the sweep procedure a sequence was used consisting of 300 ms power-up and the sweep stimulus, its length depending on the stimulation rate (Figure 3.2a). A stimulation rate of 900 Hz was used, meaning that the stimulus had a length of approximately 24.4 ms. All implant electrodes were stimulated at comfort level.

Figure 3.2: a) Sweep stimulus plus power up; all implant electrodes were stimulated consecutively at comfort level. b) Step stimulus plus power up; in this example electrode 1 was stimulated consecutively with variable CLs, comfort level was equal to 110, in total 20 steps were applied.

In the step procedure, a sequence was used consisting of 300 ms power-up and the step stimulus. Also a stimulation rate of 900 Hz was used, meaning the stimulus length was equal to approximately 23.3 ms. The step stimulus was applied for one single CI electrode. First the CL is decreased with 100 CLs under comfort level, subsequently increasing steps of 10 CLs were applied, until comfort level was reached. Next decreasing steps of again 10 CLs were applied, until the start CL was reached. An example of the step sequence is shown in figure 3.2b; electrode 1 was used, comfort level was equal to 110.
3.1 Averaged Electrode Voltage Measurements

The stimulus properties, phase width, phase gap, stimulation rate, comfort levels and stimulation mode were set according the patient MAP. When an implant electrode was turned off according the recipient's MAP, the CL was zero and only a power-up was applied.

During each single sequence (3 ms before a stimulus was transmitted), a trigger signal was sent to the EEG system to indicate the moment to start the EEG measurement. A detailed description of the triggering is given in appendix B.2.7. The recorded EEG signal was filtered between 1 Hz - 20 kHz, sampled at 50 kHz and averaged over 10 trials. The averaged waveform was exported as an ASCI file from the EEG system into MATLAB for visualization.

3.1.4 Results

Typical examples of the obtained signals during the sweep and step procedure are shown in figure 3.3 a and b. Figure 3.3a shows the average EEG electrode voltages measured on subject A with the sweep procedure.

![Figure 3.3](image)

Figure 3.3: Average of 10 EEG signals (AEV) as a function of time for subject A. a) The 22 CI electrodes are stimulated subsequently at a CL as specified in Appendix D. b) CI electrode 7 is stimulated with increasing and decreasing CLs. Comfort level was equal to 227.

In this signal twenty electrode voltages can be distinguished, produced by the biphasic pulses on implant electrodes 3 to 22. Implant electrode 1 and 2 were not used for this patient, and corresponding the CLs were set to zero. The small negative peaks
3.1 Averaged Electrode Voltage Measurements

are caused by the RF signal, transmitted by the speech processor. As described in section 2.2.2, a 5 MHz radio frequency carrier signal is transmitted. This signal is absorbed by the EEG electrode wires. Although, the 5 MHz component is reduced by a bandpass filtering of 20 kHz, the component of 900 Hz from the frame rate is obvious. Each time single frame information is transmitted to the implant, an RF artefact is produced. RF artefacts appear during power up and stimulation. This means that the EEG electrode signal contains RF artefacts. In the next chapter this will be described in detail.

Figure 3.3b illustrates the average EEG electrode voltages measured on subject A with the step procedure. In this signal twenty-one electrode voltages are distinguished, produced by the biphasic pulses with increasing and decreasing stimulation of electrode number 7. This signal also contains RF artefacts visible as small secondary peaks in the EEG signal.

Figure 3.4a and b illustrate the average electrode voltages of the sweep and step procedure measured on subject B. In this signal also twenty main peaks can be distinguished. Implant electrode 4 and 12 were not used for this patient, due to unknown reasons. Results from the other three subjects are shown in Appendix C. The amplitudes of the AEVs can diverge between all subjects, for example between subject A and E the amplitudes differ a factor 10. The comfort levels for each individual recipient are shown in Appendix D.

Figure 3.4: Average of 10 EEG signals (AEV) as a function of time for subject B. a) The CI electrodes are stimulated subsequently at a CL as specified in App D. b) CI electrode 14 stimulated at increasing and decreasing CLs. Comfort level was 219.
3.1 Averaged Electrode Voltage Measurements

3.1.5 Discussion

Figure 3.3 shows that the EEG electrode voltages do not resemble the shape of the biphasic current pulses on the implant electrodes (figure 2.4b). This is probably caused by capacitive processes in tissue between the EEG electrodes and the implant electrodes, resulting in a somewhat distorted signal. The data also reveals that the AEV vary, caused by the different comfort levels at which the individual implant electrodes are stimulated (Appendix D). It is also caused by the different impedance values for the individual implant electrodes which attenuate the applied current. Unfortunately, the impedances were not measured during this measurement.

Looking at the results of figure 3.3b, one thing is remarkable. The twelfth electrode voltage (at approximately 16 ms) is somewhat smaller than the tenth electrode voltage (at approximately 14 ms), although the same CLs were applied. This is probably caused by the limited time resolution. In order to visualize the electrode voltages, a high sampling frequency is necessary, fast enough to sample at the maximum or minimum value. A sample frequency of 50 kHz (1 sample per 20 µs) is not sufficient for these signals. Therefore the maximum or minimum amplitude of two similar electrode voltages are sometimes different. In Appendix E an example of three electrode voltages with insufficient time resolution is shown. The illustrated peaks correspond with the first three peaks of figure 3.3a.

Figure 3.4a shows AEV results in which no electrode voltage is produced in two implant electrodes of the array. For this subject (subject B) electrode 4 and 12 were disconnected resulting in two gaps at approximately 6 and 15 ms. This indicates that it is possible to detect implant electrode malfunctions (e.g. open circuits) using this setup. Figures 3.4a and b also show the presence of RF artefacts, although much smaller than in the signals from subject A. It is assumed that this increase in RF artefact amplitude is caused by the difference in RF signal absorption between the subjects i.e. turning the transmitting coil causes a change in RF artefact amplitude.

Comparing the comfort levels from Appendix D with the measured electrode voltage amplitudes from the sweep procedures (figure 3.3a, figure 3.4a, figure C.1, figure C.2 and figure C.3), a similar tendency is seen. An increase (or decrease) of the comfort levels along the array, results in an increase (or decrease) in electrode voltage amplitudes. This indicates that the obtained electrode voltages are measured correctly.

The differences in electrode voltages amplitude between the subjects (for example subject A has electrode voltages with amplitudes higher than 200 µV and subject B not higher than 60 µV) may be caused by the different impedances between the implant electrodes and EEG electrodes which is dependent on tissue composition (e.g. skin thickness varies between the five subjects). It is probably also caused by the different comfort levels between the subjects (Appendix D).
3.2 Automated Psychophysical Pitch Estimation

3.1.6 Conclusion

The developed NIC application can stimulate the implant electrodes in such a way that it is possible to measure the electrode voltages in two different procedures, the sweep procedure and the step procedure. The application is easy to use and can be applied in the clinical setting. The results illustrate that it is possible with this configuration to determine whether an implant electrode is disconnected or not.

3.2 Automated Psychophysical Pitch Estimation

Tonotopy can be checked using stimulation of any two adjacent implant electrodes, or using three electrodes, two times stimulation of any one electrode and one time stimulation of another electrode. In these tests, the cooperation of the recipient is necessary. At the UMC Nijmegen pitch perception of recipients is examined with a method described in [18]. The patient has the task to rate and assign the pitch as a number between 0 and 100 while stimulating each individual CI electrode in random order. To reduce variance the rating is repeated four to five times.

3.2.1 Setup

The setup for the tonotopy check is shown in figure 3.5.

![Figure 3.5: The setup for the psychophysical pitch experiment.](image)

The setup contains a stimulation computer running the developed NIC application, the portable programming system (PPS), the SPrint speech processor and an implant worn by the CI recipient.

3.2.2 Subjects

Only one subject (n=1) participated in the pitch perception check. The subject had no uncomfortable hearing sensations using the MAP parameters obtained with the clinical WinDPS fitting software. The subject received the implant one month before the test. No further abnormalities were known. In table 3.2 information about the subject is shown.
3.2 Automated Psychophysical Pitch Estimation

Table 3.2: Information about the subject who participated in the pitch perception check.

<table>
<thead>
<tr>
<th>Gender</th>
<th>Age [Years]</th>
<th>Month after Implantation</th>
<th>Stimulation Mode</th>
<th>Coding Strategy</th>
<th>Disconnected Electrodes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male</td>
<td>69</td>
<td>1</td>
<td>MP1+2</td>
<td>ACE</td>
<td>Electrode 1, 2</td>
</tr>
</tbody>
</table>

3.2.3 Method

To prevent over-stimulation, C-levels - as obtained with clinical WinDPS fitting software - of the subject were checked via NIC stimulation. When subjective comfort levels appear to be different, new T and C levels were determined via NIC.

First the patient had to learn which value he had to assign to each specific sound. Therefore a pitch practise session was performed. Starting with electrode number 22, the stimulus properties phase width, phase gap, comfort levels and stimulation rate were set according the MAP. For this subject a phase width of 25 µs was used, a phase gap of 8 µs and a stimulation rate of 900 Hz. The stimulus duration was set to 500 ms. Implant electrode 22 was stimulated at comfort level, using a 800 ms sequence consisting of 300 ms power up and 500 ms stimulus (Figure 3.6).

![Figure 3.6: The stimulus for the tonotopy check; Implant electrode 22 was stimulated at comfort level using a 800 ms sequence consisting of 300 ms power up and 500 ms stimulus.](image)

It was indicated to the patient that the pitch value of stimulation of electrode 22 should be 0. Then the procedure was repeated for implant electrode 1, telling to the patient that this pitch value should be 100. This practise was repeated for these and other implant electrodes until the patient had an idea about the pitch scales.

After an extensive practise, the main pitch estimation experiment was started. A randomly selected electrode was stimulated and the patient had to assign a pitch value between 1 and 100. This was repeated until all implant electrodes were stimulated and one trial was completed. To reduce variance, each trial for all electrodes
3.2 Automated Psychophysical Pitch Estimation

was repeated four times. All individual pitch estimates were stored into a text file and processed using MATLAB.

3.2.4 Results

Figure 3.7 shows the average estimated pitch values of all trials, plotted as a function of electrode number of subject F. Vertical bars indicate the range of plus and minus one standard deviation. Electrode 1 and 2 were not used.

![Pitch perception graph]

Figure 3.7: Estimated pitch values as a function of the electrode stimulated. Average of four trials is shown. Vertical bars indicate the range of plus and minus one standard deviation.

3.2.5 Discussion

The results from figure 3.7 are similar with the results obtained in [18]. Globally, a tonotopic order is measured along the electrode array. The task is considered to be difficult by the subjects. Although some practice is performed before the actual experiment, most subjects have great difficulties to scale the pitch.
Locally however, the assigned pitch value depends on the order in which the sounds are heard. Suppose for example that electrode 17 is stimulated and the subject assigns value 30 to the sound. If next electrode 5 is stimulated, the subject perceives a much higher pitch, with an estimated value of 100. If in another trial, electrode 4 is stimulated and the subject assigns value 70 and the electrode 5 is stimulated, the subject perceives a slightly lower pitch, with an estimated value 60. In this case two different pitch estimates for one implant electrode are obtained and the standard deviation will be high. This is illustrated in figure 3.7 for implant electrode 12 and 15. It is very important to repeat a trial several times in order to reduce variance. Four trials are recommended, but more trials should be performed.

It must be questioned seriously whether the electrode tonotopy should be examined according this pitch estimate procedure [18]. More efficient stimulation paradigm are probably better alternatives e.g [19]. In this procedure three sounds are presented to the recipient, two sounds with equal pitch and one with different pitch. The recipient has to indicate which sound is perceived as the different pitched sound.

3.2.6 Conclusions

With the developed NIC application it is possible to perform a psychophysical pitch estimate experiment. The application is easy to use and is already applied in the clinical setting. Only one subject participated in this experiment. The average result of four trials shows a global tonotopic order of the electrodes. Because of the difficulty to assign the pitch as described, it must be questioned whether the used method is suitable to examine tonotopy.
Chapter 4

Auditory Late Response Experiments

If integrity tests, as explained in the previous chapter, show no abnormalities and the hardware functions as expected, a recipient still might have poor speech understanding. The problem might be caused by deficiencies somewhere higher in the auditory pathway. In order to examine the auditory pathway, auditory evoked response (AER) experiments are performed [20][22].

In these experiments auditory stimuli are presented to a subject in order to evoke electrical responses in the auditory pathway. Every part of the pathway has characteristic responses. The stimuli may range from clicks to tones and even speech stimuli. The responses are measured on the scalp of the patient using surface EEG electrodes. The responses disclose a series of wave components that are described by their amplitude and latency characteristics [20]. Latency is the time interval (in ms) between a reference point (usually the stimulus onset time) and a specific peak of the AER. The amplitude of the potential can be measured from the voltage baseline to the wave peak (in $\mu$V). Examining the shape of the responses as well as the different latencies may provide information about the functioning or malfunctioning of a particular part of the auditory pathway. In this way it might be possible to locate the problem area.

Acquiring meaningful AERs from cochlear implant recipients is much more difficult than from non-CI clinical populations, because the electrical current pulses of the implant electrodes and the radio frequency signal contaminate the responses. Although adequate filter paradigms might solve this problem partially, the impact of these artefacts aggravates a reliable and correct interpretation.

For non-CI clinical subjects usually a head-phone is used as a stimulator, transmitting sounds directly into the ear and reducing sounds from the environment. A head-phone can not be used when examining cochlear implant recipients, because it is difficult to isolate the microphone between the head-phone. Therefore, a loudspeaker is usually used as stimulus source. A disadvantage of this method is that
4.1 ALR

also sounds from the environment as well as reflected sounds are processed in the speech processor.

With the introduction of NIC and NMT, an alternative for measuring the AERs from Nucleus 24 system CI recipients is introduced. As described in section 2.4.2, NMT is used for creating stimulus sequences from audio data files and streaming of these sequences directly into the speech processor. Unfortunately, the NMT software does not allow the audiologist to control power up properties during streaming and therefore control the RF artefacts. Another limitation is that it is not possible to stream for more than 2 minutes. For these tasks NIC is used. Thus, NMT is used for the creation of a sequence, NIC is used to control power up and sequence properties and transmit the sequence into the speech processor. In this way it is possible to present all kinds of stimuli to a recipient, without processing sounds from the environment and having full control of the stimulation properties. For this reason two software applications are developed, a NIC application and a NMT application. A detailed description of both applications is given in Appendix B.2.

4.1 ALR

In literature [20][22] five classification systems of the AERs are made, according to the latencies of the EEG components and related to their place of origin: (1) Electrocochleography, (2) Auditory Brainstem Response (3) Auditory Middle Latency Response, (4) Auditory Late Response and (5) Auditory P300 Response. This report will only examine the auditory late response (ALR) because the EEG components at approximately 200 ms and 350 ms are not contaminated with artefacts created by the biphasic current pulses during stimulation since only stimulus sequences are used with maximum length of 150 ms. Validation is therefore easier.

The main application of the ALR is for objective assessment of hearing thresholds [20]. It has little application for neurological investigation since the response represents the summation of activity from different areas of the cortex. An advantage of the ALR is that a frequency specific tone burst can be employed which can be equated indirectly to the pure tone frequencies of the subjective audiogram. Secondly, the technique tests the whole auditory pathway and is not just up to the level of the brainstem. A disadvantage is that the response is dependent on the psychological state of the patient and is susceptible to the effects habituation and adaptation.

Because evoked brain activity is so small, two processes are essential for detecting the responses. One process is amplifying the signal; the second is signal averaging. In the latter, all EEG components, evoked by the same stimulus presented repeatedly, are averaged. It is assumed that the pattern of each single response is almost the same. The noise on the other hand is assumed to be random. In this way averaging will reduce the noise and thus improves the signal-to-noise ratio. In [21] and [22] it is recommended to use at least hundred responses for averaging.
4.1 ALR

The EEG components are recorded in a time period from about 50 ms to 400 ms after stimulation onset (t=0 ms). Figure 4.1 shows an example of a typical ALR.

![Diagram of an ALR with labeled components P1, N1, P2, and N2.](image)

Figure 4.1: A typical Auditory Late Response (ALR) obtained from [20]. Number of averaged signals and signal bandwidth are unknown.

The amplitude for the ALR is usually within the 10 µV range. The main components, and their characteristic latency values are P1 (50-80 ms), N1 (100-150 ms), P2 (175-225 ms) and N2 (300-400 ms) [20]. The labels for these peaks refer to the voltage polarity of the response. The N1 and P2 components reflect conscious detection of any discrete change of the auditory environment and presumably arise from the auditory cortex [20].

4.1.1 ALR from a normal hearing person

In order to verify the results from literature, an ALR pilot measurement was performed with one normal hearing subject, female, 23 years old and without hearing impairment. In this measurement the setup was used as described in section 4.2 (PC 2). A set of head-phones was used as stimulus source for both ears. The active EEG electrode was placed on the forehead at Fz. The reference EEG electrode was placed on the nose and a ground EEG electrode on the check.

A 500 Hz tone burst with 20 ms rise time, 80 ms duration and 20 ms fall time was repetitively presented to the subject. Before each tone burst was played, a trigger signal was transmitted from PC 2 to the EEG system to indicate the moment to start the EEG measurement. The EEG signal was band filtered between 1Hz-30Hz and sampled at 50 kHz and then averaged over 120 trials. In order to measure all EEG components, the recording window was set to 400 ms. The averaged EEG signal was exported as an ASCI file to MATLAB for visualization. Figure 4.2 shows the results. The EEG signal from figure 4.2 corresponds with the signal obtained from
4.1 ALR

literature (figure 4.1). The main components of the auditory late response (P1, N1, P2 and N2) can be distinguished easily. This indicates that it is possible to measure an ALR from a normal hearing person with the used experimental setup.

![ALR measurement from a normal hearing person](image)

Figure 4.2: An ALR from a normal hearing person. The EEG signal as a function of time. Stimulus onset was at t=0 ms. Average of 120 trials, bandwidth 1Hz-30Hz.

4.1.2 ALR from a CI recipient: RF and stimulus artefacts

As it is shown that it is possible to measure an ALR from a normal hearing, it must be examined whether it is also possible to measure an ALR from a CI recipient. This is much more difficult than obtaining ALRs from normal hearing persons since artefacts originating from the CI hardware, contaminate the EEG signal. In earlier research at the UMC Nijmegen, such ALR measurements have been performed with CI recipients, using a similar setup as described in section 4.2 (PC 2) but now with a loud-speaker as stimulator. The stimulus was processed in the speech processor using the recipients MAP parameters. This type of configuration is called live mode. The EEG electrode configuration was the same as for a normal hearing person.

A vowel [a] speech stimulus (male voice) with duration 150 ms was repetitively presented to the subject. During each single stimulus sequence, a trigger signal was sent to the EEG system to start the EEG measurement. The recorded EEG signal was filtered between 1 Hz - 125 Hz instead of the usual bandwidth 1 Hz - 30 Hz, due to limitations of bandfilters of the EEG system. Subsequently, the EEG signal was sampled at 50 kHz and averaged over 100 trials.
4.1 ALR

The averaged waveform was exported as an ASCI file from the EEG system into MATLAB for visualization.

Figure 4.3 shows an example of an ALR obtained from a CI recipient. Further information about the subject was not available. In this EEG signal it is difficult to distinguish the typical P1 and N1-components because of the presence of an artefact generated during stimulation by the biphasic pulses from the implant electrodes. This artefact is therefore called stimulus artefact. The stimulus artefact persist as long as the implant is activated by stimulus input. After 150 ms stimulation stops, causing the stimulus artefact to stop as well.

Another artefact is produced by the 5 MHz RF carrier signal. This signal, used for transmitting stimulus information to the implant (figure 2.4), is also received by the EEG electrode. This results in the presence of a RF artefact which consists of frequencies depending on stimulation rate, phase gap and carrier signal frequency (section 2.2.2). The amplitude of the RF artefact is much smaller than the amplitude of the stimulus artefact, in fact the RF artefact is not visible in figure 4.3. Therefore, the RF artefact is of little concern as for the identification of the responses, in contrast with the stimulus artefact.

Both artefacts can be suppressed using appropriate filters. Nevertheless, it is still uncertain whether the distinguished peaks are really originated in the auditory pathway. In order to identify the responses accurately, a method must be developed to completely erase both artefacts.
4.1.3 Proposed artefact reduction

In order to reduce or eliminate both the RF and the stimulus artefacts in the ALR signal, the following procedure is proposed as explained in figure 4.4.

<table>
<thead>
<tr>
<th>Stimulation</th>
<th>EEG Signal</th>
<th>Signal contents</th>
</tr>
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</table>
| a) Normal Stimulation: CL at C-level | ![signal](signal.png) | - brain activity  
- RF artefact  
- stimulus_1 artefact  
- noise_1 |
| b) Inaudible Stimulation: CL below T level | ![signal](signal.png) | - RF artefact  
- stimulus_2 artefact  
- noise_2 |
| c) No Stimulation: only power up signal, CL = 0 | ![signal](signal.png) | - RF artefact  
- noise_3 |
| d) a-k*(b-c)-c              | ![signal](signal.png) | - brain activity  
- noise_4 |

Figure 4.4: Schematic representation of the various EEG signals. a) Normal stimulation: CL at C-level. b) Inaudible stimulation: CL below T-level. c) No stimulation: CL is 0, only power up signal d) Subtraction method to erase the artefacts.

First three different EEG signals must be obtained, schematically shown in figure 4.4 a, b and c. Figure 4.4a illustrates an ALR obtained from a CI recipient using normal stimulation at C-level (analogous to figure 4.3). This EEG signal consists of four components: brain activity (the auditory late response), RF artefacts, stimulus_1 artefact and random noise_1 (EEG noise, physiologic noise and measuring device noise).

Figure 4.4b shows a similar EEG signal but now stimulation is divided by a factor k (k<sub>T</sub>1) in such a way that the (new) highest C-level did not reach the (old) lowest T-level. No sounds are heard by the recipient. This EEG signal consists of three components: RF artefacts, stimulus_2 artefact and random noise_2.

Figure 4.4c shows an EEG signal when no stimulation is performed and only power up is transmitted (CL=0). This EEG signal consists of two components: RF artefacts and random noise_3.

Subsequently, the artefact reduction procedure is performed. Four assumptions are made: (1) The shape of the RF pulses are exactly the same in all three EEG signals. (2) The position of the RF pulses in time are exactly the same in all three EEG
4.2 Setup

The EEG signal of figure 4.4c is subtracted from the EEG signal of figure 4.4b. The obtained signal consists of the stimulus artefact and noise but no RF artefacts are present. This signal is amplified with the factor k, resulting in an analogous stimulus artefact as described in figure 4.4a. Subtraction of this signal from the signal of figure 4.4a results in a signal with brain activity, RF artefacts and noise. When the signal of figure 4.4c is subtracted from the obtained signal, an EEG signal is obtained in which only the brain activity plus noise is present (figure 4.4d). The typical ALR components should then be identified easily.

To verify this, the proposed procedure is experimentally tested using the NMT and NIC applications.

4.2 Setup

The setup for the ALR measurements is shown in figure 4.5.

Figure 4.5: The experimental setup for the auditory late response measurements.

In this setup PC 1 or PC 2 is used as stimulator. When PC 1 is connected, the setup is the same as described in section 3.1.1. If PC 2 is connected, stimulation is done in the so called live mode. A sound is played through the loudspeaker and picked up by the integrated microphone of the SPrint speech processor. In the SPrint the sound is processed using the MAP parameters of the subject and transmitted to the implant electrodes. The distance between the subject and the speaker is approximately 2 meters. Triggering occurs on stimulus onset. The configuration of the EEG electrodes and the EEG system are the same as described in section 3.1.1.
4.3 Subjects

Two subjects (n=2) participated in the ALR measurements. The subjects had no uncomfortable hearing sensations using the MAP parameters obtained with the clinical WinDPS sitting software. No further abnormalities were known. In table 4.1 information about the subjects is shown. The T and C levels of both subjects are shown in Appendix F.

<table>
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<tr>
<th>Gender</th>
<th>Age [Years]</th>
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<th>Stimulation Mode</th>
<th>Coding Strategy</th>
<th>Disconnected Electrodes</th>
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<td>44</td>
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Table 4.1: Information about the subjects who participated in the ALR measurements.

4.4 Method

First a wav-file, containing a 500 Hz tone burst with rise time 20 ms, duration 80 ms and fall time 20 ms, was presented to NMT to create a stimulus sequence according to the individual patient MAP. The sequence was recreated using the NIC application. In order to generate sufficient power the stimulus sequence was headed by 300 ms and tailed by a 580 ms power up sequence (figure 4.6).

<table>
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<tr>
<th>Stimulus rate = 1 kHz</th>
<th>Stimulation rate =10.8 kHz</th>
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<td>power up</td>
<td>ALR stimulus</td>
<td>power up</td>
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<td>300 ms</td>
<td>120 ms</td>
<td>580 ms</td>
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Figure 4.6: The sequence for the ALR measurements consists of 300 ms power-up, 120 ms stimulus and 580 ms power-up.

Power-up is transmitted to the implant with a frequency of 1 kHz. The stimulus frequency is 10.8 kHz. The sequence was repetitively presented with a frequency of 1 Hz to the subject. During each sequence (3 ms before the stimulus was transmitted), a trigger signal was sent to the EEG system to start the EEG measurement. A detailed description of the triggering is given in appendix B.2.7.

The recording window of the EEG was 500 ms in order to measure all responses (P1,N1,P2 and N2). 200 responses were averaged of which the first 100 were band filtered between 1Hz and 30Hz (narrow band filter setting) and the second 100 EEG
4.5 Results

Signals between 1Hz and 20kHz (broadband filter setting). Sampling frequency was again equal to 50 kHz. Both averages were exported from the EEG system to MATLAB for visualization.

For the second run all threshold (T) and comfort (C) levels in the patient MAP were decreased with a factor 0.5. In this way none of the (new) C-levels reached the (old) T-levels leading to sub T-level stimulation for both subjects. For the third run all CLs were set to zero. No auditory cortical responses and implant electrode activity should be measured.

Finally, the ALR measurement was repeated in live mode, using the same 500 Hz tone burst but now presented by a loudspeaker. In the SPrint the sound is processed using the MAP parameters and transmitted to the implant electrodes in the same way as in NMT. Bandfiltering, averaging, sampling frequencies and visualization were the same as in the previous runs.

4.5 Results

Results from subject G are shown in figure 4.7, 4.8, 4.9 and 4.10. The results from subject H have similar shape as shown in Appendix G.

Figure 4.7 shows the two ALR measurements for the two filter settings.

![ALR experiment](image)

Figure 4.7: ALR results for subject G at a stimulus comfort level. Left: narrow band filter setting. Right: broadband filter setting.

In both EEG signals a voltage baseline shift is measured during stimulation (between t=0 ms and t=120 ms). In this period stimulus artefacts are clearly visible in the broadband EEG signal, similar to figure 4.3. In the narrow band EEG signal the stimulus artefacts are suppressed.
4.5 Results

Figure 4.8 shows the two ALR measurements when stimulation is below T-level.

Figure 4.8: ALR results for subject G at a stimulus current level below T-level. Left: narrow band filter setting. Right: broadband filter setting.

In both EEG signals a similar voltage baseline shift is measured during stimulation. The stimulus artefacts are reduced in amplitude. Figure 4.9 shows the two ALR results for the two filter settings when stimulation is at current level 0.

Figure 4.9: ALR results for subject G at a stimulus current level of 0. Left: narrow band filter setting. Right: broadband filter setting.

In the small band EEG signal a similar baseline shift is measured as in the EEG signals from figure 4.8 and 4.7. In the broadband EEG signal a change in baseline during stimulation is obtained but the amplitude is much smaller.
Figure 4.10 shows the two results of two live mode ALR measurements for the two filter settings.

![Graphs showing ALR measurements for two filter settings](image)

**Figure 4.10:** *The ALR measurements for subject G in live mode. Left: narrow band filter setting. Right: broadband filter setting.*

The narrow band EEG signal does not contain obvious P1, N1, P2 and N2 peaks, almost no change in EEG signal is measured. The broadband EEG signal shows that after stimulus onset \( t=0 \) ms the implant electrodes are stimulated for more than 250 ms producing stimulus artefacts. During stimulation a baseline shift is obtained.

### 4.6 Discussion

Comparing the broadband EEG signals with the narrow band EEG signals, provides information about the reproducibility of the measurements. The narrow band EEG signals must also be present in the broadband EEG signals. This is the case in the EEG signals from figure 4.7 and 4.8. Filtering the broadband EEG signals with a bandfilter with cut-off frequencies 1Hz-30Hz, should acquire EEG signals with similar shape as the narrow band EEG signals. The EEG signals from figure 4.9 on the other hand do NOT have an obvious similarity, although a change in baseline is measured during fast power-up (between \( t=0 \) ms and \( t=120 \) ms). NO similarity can be determined with the signals from figure 4.10. Therefore, it can be concluded that the signals from figure 4.9 and 4.10 are unreliable. The difference between the EEG signals with different frequency content could be caused by the change of muscle activity or brain activity in the course of the measurements, it could also be caused by an unattached EEG electrode.

The broadband EEG signal obtained in live mode also shows that the implant electrodes are stimulated for more than 300 ms, although the stimulus was only 120 ms.
This is probably caused by sound reflections in the room since the amplitude of the stimulus artefacts decreases after 120 ms.

None of the produced EEG signals show the typical EEG components illustrated in section 4.1. In figure 4.7, 4.8 and 4.9 a voltage baseline shift is measured during stimulation, causing no typical peak to be seen. In the EEG signal obtained in live mode no baseline shift is measured but also no typical peaks can be seen. This could indicate that the applied method was not suitable for measuring ALRs but the results with the normal hearing person invalidate this. It must also be questioned whether it is possible to measure the late responses for this particular subject. The results also indicates that the baseline shift is probably caused by high frequency stimulation (10.8 kHz) with the NIC application. According Cochlear Corp the stimulation with NIC and the stimulation in live mode occurs in the same way, but the measurements show that this is not the case. The problem is examined further by Cochlear Corp.

4.7 Conclusions

With the NMT application it is possible to create a stimulation sequence with all kind of sound files using the individual patient MAP. The NIC application has the possibility to stream the created sequence in the speech processor. Both applications are easy to use.

ALR measurements were performed using two different configurations: stimulation using the developed NMT and NIC applications and stimulation in live mode. Results show no typical auditory late response wave forms. A voltage baseline shift was measured when stimulating at high frequency with the developed NIC application. Because of the voltage baseline shift, the developed method can not be used. Subtracting the EEG signals according the described method, will not reduce the artefacts.
Chapter 5

Conclusions and Recommendations

During this project a software application is developed for communication with and stimulation of a Nucleus 24 implant system. With this application an electrode voltage experiment and a pitch estimation experiment as well as an auditory evoked response experiment are performed. The application is easy to use and is already applied in the clinical setting.

In the electrode integrity test a sweep and step stimulation were performed, in order to examine the biphasic pulses on the implant electrodes. No abnormalities were measured. The electrode integrity test was performed with five subjects who had no complaints about the implant. Also speech understanding was on normal level. It is recommended to repeat this experiment with subjects having abnormal current spread. For example, subjects suffering otosclerosis have unusual current spread because of the calcified bone composition [23]. The biphasic current pulses do not have a biphasic shape when measured on the scalp as far-field potentials.

In order to examine the tonotopy of the implant electrodes, a psychophysical pitch estimate experiment was performed. The average result of four trials shows a tonotopic order of the different implant electrodes. No conclusions can be drawn since only one subject participated in this experiment. In future research more subjects must be measured. Because of the difficulty to assign the pitch as a number between 0 and 100, it must be questioned whether the used method is suitable to examine tonotopy. Other experiments (e.g. discrimination between two similar pitched sounds and one deviant pitched sound [19]) are much better alternatives.

The auditory late response measurements are performed to examine the signal processing of the brains. Due to the biphasic pulses on the implant electrodes, the obtained responses are contaminated and hard to analyse. A subtraction method is proposed to erase these artefacts. Results show a voltage baseline shift. It is assumed that this baseline shift is generated by high frequency stimulation with the
NIC application. The problem is examined further by Cochlear Corp. As a result of the voltage baseline shift, the proposed subtraction method can not be used. In future research independent component analysis (ICA) could be applied. This method separates the auditory responses into a number of independent components equal to the number of sensors [24]. No conclusions can be drawn since only two subjects participated in this experiment. In future research more subjects must be measured.
Bibliography


# Terminology

<table>
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<tr>
<th>Acronym</th>
<th>Description</th>
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<tr>
<td>AER</td>
<td>Auditory Evoked Response: electrical activity after sound stimuli</td>
</tr>
<tr>
<td>ALR</td>
<td>Auditory Late Response: typical electrical response within 400 ms after stimulus</td>
</tr>
<tr>
<td>CI</td>
<td>Cochlear implant</td>
</tr>
<tr>
<td>CL</td>
<td>Current Level</td>
</tr>
<tr>
<td>C-level</td>
<td>Comfort level: maximum current level that does not produce an uncomfortable loudness</td>
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<tr>
<td>CPS</td>
<td>Clinical Programming System of Cochlear Corp: hardware interface between computer and speech processor</td>
</tr>
<tr>
<td>CVI</td>
<td>C-based programming language application</td>
</tr>
<tr>
<td>Frame</td>
<td>One biphasic pulse</td>
</tr>
<tr>
<td>MAP</td>
<td>Patient MAP: program file in speech processor with individual data parameters</td>
</tr>
<tr>
<td>Live Mode</td>
<td>Measurement configuration in which the sound processed in the speech processor in real time</td>
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<td>NIC</td>
<td>Nucleus Interface Communicator: Cochlear Corp software library which can be implemented in the programming languages C, C++, Delphi and Visual Basic</td>
</tr>
<tr>
<td>NMT</td>
<td>Nucleus Matlab Toolbox: Matlab software library</td>
</tr>
<tr>
<td>Power-up</td>
<td>Radio Frequency frames to the supply the implant with power</td>
</tr>
<tr>
<td>PPS</td>
<td>Portable Programming System of Cochlear Corp: hardware interface between computer and speech processor</td>
</tr>
<tr>
<td>RF</td>
<td>Radio frequency carrier: 5 MHz signal transmission between the implant and speech processor</td>
</tr>
<tr>
<td>Sequence</td>
<td>A train of frames repeated several times</td>
</tr>
<tr>
<td>SPrint</td>
<td>Body worn speech processor</td>
</tr>
<tr>
<td>T-level</td>
<td>Threshold level: lowest current level that elicits a soft but consistent hearing sensation</td>
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<td>University Medical Centre Nijmegen</td>
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<td>WinDPS</td>
<td>Main software program created by Cochlear Corp</td>
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Appendices
Appendix A

Patient MAP

MAP Summary for ****** - MAP 11 Date: dinsdag 27 april 2004
Clinician: A.J. Beynon Software Version: Nucleus R126 V2.0 (2003.4.4.1)

- MAP Details -

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### MAP Channel Details

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Appendix B

Developed Software

To perform integrity and auditory evoked potential experiments with a Nucleus 24 system, a clinician has two possibilities, use the winDPS program from Cochlear Corp. or create a custom software application using the software libraries NIC-STREAM and NIC. If standard test will be performed, the winDPS program is preferred. When special customized tests will be performed, it is better to create a personal software application. In this way specific functions, e.g. controlling power-up and long audio file steaming, can be implemented.

In this project both available software libraries have been used. Through these libraries two different applications have been developed, one MATLAB application which makes use of the NMT and one CVI c-based application which makes use of the NIC library. In the following sections these two applications will be described.

B.1 The developed Matlab applications

As described in section 2.4.2, MATLAB and the NMT can be used for streaming audio data files (wav format) into the speech processor. During this streaming process the software uses the individual patient MAP to create a sequence according the desired audio file. In this sequence, information is stored about which electrode or electrodes must be stimulated at what time. Subsequently the sequence is sent via the speech processor and the transmitting coil to the implant. In this way the patient will "hear" the contents of the wav-file. The advantage of this way of direct stimulation is the fact that no sounds from the environment (noise) are registered, this in contrast to indirect free field measurements.

Unfortunately the NMT does not allow the user to change several stimulus properties; e.g. power-up properties can not be changed. It is also not possible to stream for more than two minutes although this is necessary in some psychophysical tests. Therefore the NMT is not used for direct streaming; it is only used for the creation of sequences.
To be able to do so, a MATLAB application has been written which calls several functions from the NMT. Some NMT functions are originally from the toolbox, some NMT functions have been adjusted and other functions have been created by the author. The original NMT functions are fully described in [25].

The main file is called createseqfile.m. This function is created by the author. The detailed structure of this main file is shown in figure B.1.

**Createseqfile.m**

When this file is executed, the user first has to provide three inputs, (1) the filename of the patient MAP (xls format), (2) the coding strategy (ACE, CIS or SPE(AK)) and (3) the number of sound stimuli used (1 or 2). In this report only tests with one stimulus are performed, the two stimulus tests are already implemented for future research. If one stimulus is used, the user has to provide another input, namely which sound file (wav format) should be used. Using two sound stimuli, two sound files must be provided.

In both cases, the next step is to create the exact sequence (or sequences). This is performed using the adjusted NMT function createmapseq.m. The structure of this file is shown in figure B.2.

Executing this function, the adjusted NMT function read_excel_map.m is called. This function reads the provided patient MAP (xls format) using the MATLAB command xlsread and extracts the following patient-specific MAP parameters; active electrodes, stimulation modes, threshold levels, comfort levels, channel stimulation rate, number of active bands, number of maxima, Q-value, base level and phase gap. In order to create a xls format patient MAP, the MAP from the winDPS program must first be saved as a text file before it can be read and saved in Excel.
B.2 The NIC application using Labwindows CVI

Subsequently, it is determined which coding strategy must be used. The sound file is read by the MATLAB command wavread. Next the default non-patient-specific MAP parameters are loaded using one of the three coding strategy MAP functions from the NMT, ACE_map.m, CIS_map.m or SPEAK_map.m [25]. Then the function seq.m is called. This simple function is created by the author and calls two NMT functions, process.m and plot_sequence.m [25]. The first function calculates the sequence properties using the correct MAP parameters and sound file(s). The second function plots the sequence as a spectrogram which is a representation of the stimulated electrodes as a function of time.

Createmapseq.m

MATLAB closes the function createmapseq.m and continues with the main function createseqfile.m. Subsequently eight text files will be opened for writing. The sequence properties, stimulated electrodes, current levels, number of sounds, phase width, phase gap, sequence length, stimulation mode and stimulation period are then written in separate text files. Last the text files are closed. In this way the sequence can be read into another program to regenerate the exact sequence.

B.2 The NIC application using Labwindows CVI

As described in section 2.4.2, the NIC software library functions can be used in (software) applications which have to perform specific experiments with Nucleus cochlear implants. For using the NIC functions [26] [27], the developed application must be programmed in one of these four programming languages C, C++, Delphi or Visual Basic.
B.2 The NIC application using Labwindows CVI

As a result of past experiences in the clinical setting, the application for this research has been written using National Instruments Labwindows CVI. This program is an integrated C environment with a virtual instrumentation possibility (Graphical User Interface). This Windowslike user interface can be constructed with buttons, numeric indicators, binary switches and text windows. Using these units or combination of these units, specific actions can be performed. The specific actions are also called callback functions. The code for all units can be created automatically, although the code for the results when one of the units has been used, has to be written by the user. In this way an user-friendly application can be created which can be used for many purposes.

In this project an application has been made in which the user has the opportunity to control each stimulus property of the Nucleus 24 system and in this way perform the following experiments; pitch perception experiment [18], impedance experiment [28], electrodes integrity experiments [23] and auditory cortical response experiments [21] and [29]. The structure of this application (NICthijs.c), is shown in figure B.3.

![Figure B.3: NICthijs.c; the five different parts of the Labwindows CVI application.](image)

When executed the application runs some header files in which first the used C functions are defined. These C functions are specific C program language commands. Secondly the NIC library functions are defined. Next all parameters and all void functions are defined. The word void is C terminology for a function whose only purpose is to carry out certain instructions. Subsequently the main function is executed, this means the user interface is opened. Figure B.4 shows a screen-shot of the user interface of the developed CVI application.

The user interface contains indicators and callback functions (17 in total). As described earlier, the callback functions are used to perform specific actions. The indicators do not perform special actions and are simply indicating stimulus information. The values of the indicators are used in the callback functions. All callback functions can be divided into six groups, stimulus callback group, pitch callback group, stream callback group, impedance callback group, easy use callback group and the various callback group. In the following sections each group will be described.
B.2 The NIC application using Labwindows CVI

Figure B.4: A screen-shot of the graphical user interface of the developed CVI application.

B.2.1 Stimulus callback group

The stimulus callback group consists of five callback functions. Using these callbacks according numeric order, communication between computer and speech processor will be established, sequences can be created, stimulation of the implant electrodes can be started and the stimulation can be stopped.

The first callback in this group is the InitCallback function. This callback can be executed pressing the Init button. When pressed, two void functions will be called, (1) Init.Hardware and (2) Init.Communication. The first void function sets-up the hardware and creates a power-up frame. The second void function sets-up the communication between the computer and the speech processor. Both functions contain NIC library functions.
B.2 The NIC application using Labwindows CVI

The second callback in this group is the Seqcallback function. This callback can be executed pressing the Sequence button. When pressed, three void functions will be called, (1) Get.GUI.struct.parm, (2) Set.User.Sequence and (3) Clear.sequence. The first void function reads the indicators, active/inactive electrodes, electrodes current levels, phase gap, phase width and stim(ulation) rate, from the user interface. The second void function is able to create a stimulation sequence ¹. Figure B.5 shows the structure of this function.

Figure B.5: The Set.User.Sequence function; this function has the ability to create two different sequences.

First three (numeric) indicators from the user interface are read namely, stimulation modes, number of steps and stepsize. Next the binary switch step CL is read. If this switch is set to 'off', a sequence (sweep) is created in which all active electrodes will be stimulated continuously with indicated current levels. When the switch is set to 'on', a sequence (current levels step) is created in which all active electrodes will be stimulated continuously with increasing and decreasing current level. The number of increased and decreased steps is determined by number of steps, the step width is determined by stepsize. The third void function will delete all sequence information. All functions contain NIC library functions.

The third callback is the Writecallback function. This callback can be executed pressing the Write button. When pressed, the void function write_sequence is called. This function calls a NIC library function which writes the created sequence into the speech processor.

The forth callback is the Startcallback function. This callback can be executed pressing the Start button. When pressed, the void function start_sequence is called. This function calls a NIC library function which starts the sequence and therefore starts stimulating the electrodes of the cochlear implant.

The fifth callback is almost the same as the forth callback. This callback, called Stopcallback, does not start the sequence but on the other hand stops the sequence by calling the void function stop_sequence.

¹sequence = power up + stimulus; all sequences consist of 300 ms (1000 Hz) power-up, as recommended by Cochlear Corp, and the stimulus.
B.2.2 Pitch callback group

The pitch callback group consists of three callback functions. Using these callbacks according numeric order, the user can perform a complete pitch experiment. This implies that the user can perform a test, an experiment and process the generated data.

The first callback in this group is the Practise callback function. This callback can be executed pressing the Practise button. Before it can be pressed, first the initialization of the hardware and communication must be performed. This can be done by using the init button. When both buttons are pressed, the user will perform a practice in which the patient can exercise the pitch experiment. The detailed structure of the callback is shown in figure B.6.

The second callback in this group is the Pitchexperiment callback. This callback can be executed pressing the Experiment button. When pressed, the user will perform a pitch experiment. In this experiment, the patient has to give a number (0-100) for pitch percept every time an (random) electrode of the array is stimulated. If all electrodes have been stimulated, one experiment is performed. The detailed structure of the callback is shown in figure B.7.

The function starts with opening a text file in which all data will be written. Next an array of 22 numbers is created ranging from the value 1 to 22. These values correspond with the electrodes which will be stimulated. During a pitch experiment every electrode must be stimulated, so none of the numbers in the array are the same. When the array is full, this means 22 random numbers have been created, some properties are read from the user interface; the current levels of all electrodes, the phase width, the phase gap, the stim(ulation) rate and the stimulus dur(ation).
Subsequently the initialization of the hardware and communication is performed using the void functions Init_hardware and Init_communication.

Figure B.7: The pitch experiment callback function; the patient has to give a number (0-100) for pitch percept every time a random electrode of the array is stimulated, in this way the tonotopy of the electrodes can be examined.

Then, the first value of the array is used to create a sequence. In other words if the value is equal to e.g. 14, a sequence is created in which electrode 14 is stimulated. This sequence is written into the speech processor, started and stopped using the void functions write_sequence, start_sequence and stop_sequence. Between the start and stop sequence command, a delay is implemented the same way as in the practise callback function.

The callback will now wait for input from the user. This means, the patient has to give a value between 0 to 100 for pitch percept. When electrode 22 is stimulated, the sound should be experienced as 'low' and the patient should give a value of 0 for pitch percept. If electrode 1 is stimulated, the sound should be experienced as 'high' and the patient should give a value of 100 for pitch percept etc. In this way, the tonotopy of the electrodes can be examined. After user input, the next value in the array is used to create a new sequence. This continues until all values from the array have been used, that is all electrodes have been stimulated. The generated data is written into the text file. Now the user has the possibility to perform
another experiment but can also choose to stop. If stopped, the text file is closed and the experiment is ended. To increase accuracy, it is recommended to repeat the experiment five times.

The third callback in this group is the Excelcallback. This callback can be executed pressing the Excel button. When pressed, Microsoft Excel is opened. Using this program the generated data can be processed in three steps (figure B.8).

![Figure B.8: The three steps to be performed to process the generated data file from the pitch experiment.](image)

First the generated data text file has to be renamed according number of performed pitch experiments. When one pitch experiment is performed, the file should be named PE1E.txt, if two pitch experiments are performed, the file should be named PE2E.txt etc. Next the (renamed) data file must be opened in Excel using the text import wizard. In step 1 the user should choose delimited, step 2 and step 3 can be skipped pressing the next button. Subsequently one of the five created macro’s, depending on the number of experiments, must be pressed to process the data. All data is organized and a plot according specified requirements, is created.

### B.2.3 Stream callback group

The stream callback group consists of two callback functions. Using these callbacks according numeric order, the user can perform a complete stream experiment. This implies that the user can perform a test and a stream experiment.

The first callback in this group is the Streampraccallback function. This callback can be executed pressing the Streampractise button. When pressed, the user will perform a practice in which the patient can become accustomed to the stimulus (or stimuli) of the stream experiment. The callback only works when the MATLAB application (section B.1) has already been executed. As described, the output of the MATLAB application consists of several text files containing information about the created sequence. In the streamprac callback these text files will be used to recreate the sequence and stimulate to the electrodes of the implant. This process is also called streaming.

The callback is divided into two parts, the first part is executed when one stimulus has been used, the second part is executed when two stimuli have been used in the MATLAB application. The main structure of both parts is the same, except for the
B.2 The NIC application using Labwindows CVI

fact that the number of sequence variables is doubled and the length of the sequence is approximately twice as long (both stimuli are streamed consecutively) when two stimuli have been used. The structure of the callback is shown in figure B.9.

The function starts with opening the text files created with the MATLAB application. The sequence information is read from the text files and initialization is performed using both void functions Init.Hardware and InitCommunication. Subsequently the sequence is recreated, written into the speech processor, started and stopped in the same way as in the pitch experiment. The length of the delay is determined using the void function UserOnSequenceEndFunction. This NIC function is called when the sequence has ended. Last the text files are closed and the callback stops.

The second callback in this group is the Streamcallback function. The callback can be executed pressing the Stream button. When pressed, the user will perform a stream experiment. In this experiment the electrode array is several times stimulated with the sequence(s) created in the MATLAB application. The callback only works when the MATLAB application has already been executed. Again the callback can be divided into two parts, the first part is executed when one stimulus has been used, the second part is executed when two stimuli have been used.

The structure of the first part is shown in figure B.10. The function starts with opening the text files created with the MATLAB application. The sequence information, sound frequency and number of sounds are read and initialization is performed using both void functions Init.Hardware and InitCommunication. The sequence is recreated, whereafter the internal computer clock is set to zero and the counter variable is increased by one. The counter has now value one. The sequence is written into the SPrint and is started by the void functions Write.sequence and Start.sequence. The internal computer clock is read and compared to 1 divided by the sound frequency. If the internal clock is not equal to this value, the clock is read again until it has the same value. In this way a flexible delay is created. When it answers this condition, the sequence is stopped using the void function Stop.sequence.

Subsequently, a similar construction is implemented, the counter value is compared to the number of sounds. If the counter value is unequal to the number of sounds, a
new sequence is streamed. Otherwise the streaming will not continue and the text files are closed. In this way the user has the possibility to adjust the number of sounds streamed.

The structure of the second part is shown in figure B.11. The function starts with opening the text files created with the MATLAB application. The sequence information, sound frequency and number of sounds are read and initialization is performed using both void functions Init.Hardware and Init.Communication. Both sequences are recreated, whereafter a matrix is set. This 2D-matrix, consisting of twenty rows and ten columns, contains the values 1 and 2 corresponding with both sequences. One column has a 0.85 ratio of the value 1 and 2, this means three times the value 2 and seventeen times the value 1. In between two values 2, at least two values 1 are present. All rows differ from each other.

Next the start position in the matrix is determined; the row index (counter) is always equal to zero, the column index is determined randomly. Subsequently the row index is increased by one, implying that the index has the value one at this point. After increasing, the value of the matrix at that specific position is determined. If this value is equal to 1, the first sequence must be used, if the value is equal to 2, the second sequence must be used. Then, the functions of figure B.10 are executed, this means the sequence is written into the SPrint, started and stopped as described. When the counter value is unequal to the number of sounds, the row value (counter) is increased and the second row value is used for streaming. This continues until one complete row, which means twenty values, is used. Because of the length of the rows, the minimum value of the number of sounds is equal to 20 and can be increased in
multiples of twenty. In this way, during each multiple a random row can be used. If the last sequence is streamed, the text files are closed and the callback stops.

![Diagram](image)

**Figure B.11:** The structure of the second part of the stream callback function.

### B.2.4 Impedance callback group

The impedance callback group consists of two callback functions. Using these callbacks according numeric order, the user can measure the electrode impedances.

The first callback in this group is the Createcallback function. This callback can be executed pressing the Create imp. measurement file button. When pressed, the function creates an impedance measurement file in which is specified which electrodes must be used to measure impedance on. The function first reads the number of measured electrodes from the user interface. When 22 (default) is used, electrode 1 to 22 is measured. If 10 is used, electrode 1 to 10 is measured. Next the stimulation mode is read from the user interface. The user has the possibility to choose MP1, MP2, MP1+2, or CG. Subsequently, a file is created according Cochlear Corp specifications [30].

The second callback in this group is the Impcallback function. This callback can be executed pressing the Impedance measurement button. When pressed, the file
impedance.exe, provided by Cochlear Corp, is executed [30]. This application measures the impedance present for each electrode specified in the electrode data file and outputs the results to an output file. The callback function visualizes some of the output information in a MS-DOS prompt window.

B.2.5 Easy use callback group

The easy use callback group consists of three callback functions, SetGlobalCurrentcallback, Setelectrodes callback and comfort callback. These callbacks can be used to facilitate the application use.

The first callback in this group is the SetGlobalCurrentCallback. This callback can be executed using the Electrode CL indicator from the user interface in combination with the set electrode current binary switch from the user interface. This means if the binary switch is set to 'all', the current level value (0-255) from the indicator will be applied in all current level indicators of the electrodes from the user interface. When the binary switch is set to 'one', the callback function is not active.

The second callback in this group is the Setelectrodescallback function. This callback can be executed using the set electrodes binary switch from the user interface. When set to 'all', all electrodes will become active. If set to 'none', all electrodes will become inactive.

The third callback in this group is the Comfortcallback function. This callback can be executed using the comfort button from the user interface. When pressed, all patients comfort levels will be applied in the current level indicators of the electrodes. This callback only works when the MATLAB application has already been executed and all comfort levels are extracted from the patient MAP.

B.2.6 Various callback group

The various callback group consists of two callback functions. The first callback is the Democallback function. This callback can be executed by pressing the Demo button. When pressed, the executable NICWorkshopDemo is called. This Laax Workshop Demo is used for demonstrations and is provided by Cochlear Corp [30].

The second callback is the Quitcallback function. This callback can be executed by pressing the Quit button. When pressed, two void functions are called (1) Disconnect and (2) Clear_sequence. The first void function disconnects the computer and the speech processor using a NIC library function, the second void function clears all sequences. Next the graphical user interface is closed and the CVI application is terminated.
B.2.7 Triggering

In all sequences except for the impedance sequence and the pitch sequence, a trigger was necessary in order to indicate the start time when using an external measuring device. The triggering is controlled in the CVI application [27]. The exact way of triggering is shown in figure B.12.

![Diagram of triggering](image)

Figure B.12: The triggering

When using a sequence, the trigger is transmitted at the same time when power frame 298 (of total 300 frames) is sent (figure B.12). After all 300 power frames, the stimulus is transmitted starting with the first frame.
Appendix C

Results of the AEV measurements of subjects C, D and E

Figure C.1: Average of 10 EEG signals (AEV) as a function of time for subject C. a) All CI electrodes stimulated subsequently at a CL as specified in Appendix D. b) CI electrode 4 stimulated at increasing and decreasing CLs. Comfort level was equal to 195.
Figure C.2: Average of 10 EEG signals (AEV) as a function of time for subject D. 
a) All CI electrodes stimulated subsequently at a CL as specified in Appendix D. b) CI electrode 22 stimulated at increasing and decreasing CLs. Comfort level was equal to 232.

Figure C.3: Average of 10 EEG signals (AEV) as a function of time for subject E. 
a) All CI electrodes stimulated subsequently at a CL as specified in Appendix D. b) CI electrode 22 stimulated at increasing and decreasing CLs. Comfort level was equal to 174.
Appendix D

The current levels applied in the sweep procedure

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Figure 3.3 | Figure 3.4 | Figure C.1 | Figure C.2 | Figure C.3
Appendix E

Average Electrode Voltages (AEVs) in detail

Figure E.1: Detail of the result of figure 3.3, for the first three electrodes
Appendix F

The T and C-levels of subjects G and H

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Appendix G

Results of the ALR Measurements from Subject H

Figure G.1: ALR results for subject H at stimulus comfort level (Appendix F). Left: narrow band filter setting. Right: broadband filter setting.
Figure G.2: ALR results for subject H at a stimulus current level below T-level. Left: narrow band filter setting. Right: broadband filter setting.

Figure G.3: ALR results for subject H at a stimulus current level of 0. Left: narrow band filter setting. Right: broadband filter setting.
Figure G.4: ALR results for subject H in live mode. Left: narrow band filter setting. Right: broadband filter setting.
Appendix H

Abstract (in Dutch)

Wanneer het gehoor van een patient in het geheel niet meer functioneert en een hoortoestel niet leidt tot een duidelijke verbetering, zou een cochleair implantaat (CI) een oplossing kunnen bieden. Hierbij wordt de gehoorzenuw kunstmatig direct in de cochlea gestimuleerd door elektrodesignalen die zijn afgeleid van het (omgevings)geluid. Na intensieve gewenning en training kunnen deze patiënten doorgaans redelijk communiceren via spraak. Soms blijft het spraakverstaan echter onvoldoende. De oorzaak daarvan kan liggen bij zowel het implantaat als bij een disfunctioneren van meer centrale delen in het auditieve systeem van de patiënt. Om dit te onderzoeken is in het UMC Nijmegen een software-applicatie ontwikkeld waarmee zowel het functioneren van het implantaat als het functioneren van de centrale delen in het auditieve systeem onderzocht kunnen worden.

Het implantaat is onderzocht op twee manieren. (1) De spanningen op de CI-elektrodes zijn geregistreerd met EEG-oppervlakte-elektrodes aangebracht op de schedel. Resultaten van vijf proefpersonen laten zien dat het mogelijk is om met behulp van de ontwikkelde applicatie de CI-elektrodespanningen te registreren en eventueel defecte CI-elektrodes te identificeren. (2) De subjectieve toonhoogte-verdeling van de CI-elektrodes is eveneens onderzocht. Hierbij worden afzonderlijke CI-elektrodes in willekeurige volgorde gestimuleerd en moet de patiënt bij elke stimulus een getalwaarde toekennen aan de waargenomen toonhoogte. Resultaten van een proefpersoon laten een subjectieve toonhoogteverdeling zien over de gestimuleerde CI-elektrodes.

Het functioneren van de centrale delen in het auditieve systeem is onderzocht met behulp van de 'auditory evoked responses' (AER). Hierbij worden verschillende korte geluidstimuli gebruikt die typische elektrische responsies veroorzaken in het auditieve systeem. De AER-signalen worden eveneens gemeten met behulp van EEG-oppervlakte-elektrodes op de schedel, waarbij eventueel afwijkende responsies kunnen duiden op disfunctioneren. Echter zowel de stimuli op de CI-elektrodes als de radio-frequente (RF-) signalen gebruikt voor informatieoverdracht naar het implantaat, veroorzaken ongewenste signalen op de EEG-elektrodes. Daardoor zijn de gemeten AER-signalen moeilijk te interpreteren. Een mathematicische methode is geventroduceerd om deze artefacten te verwijderen. Resultaten van twee proefpersonen laten echter onverwachte en nog onverklaarde veranderingen van de DC-component zien, waardoor de voorgestelde methode niet zonder meer gebruikt kan worden.