Cochlear Implants: bridging the gap between computational model and clinic

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Chapter 1

General introduction

Cochlear Implants

Cochlear Implants (CIs) are medical devices that have been widely used in the past decades to restore hearing for people with certain types of substantial hearing loss. Their basic working principle is to evoke sound perception by directly stimulating the auditory nerve in the cochlea with electrical pulses, bypassing the need to transfer sound through the outer and middle ear to the inner ear mechanically. They are currently the most successful hearing prosthetic for individuals who, for whatever reason, cannot meaningfully understand or even detect (amplified) sound, but whose auditory nerves are still present and healthy enough to convey signals to the brain.

There are several different manufacturers of CIs, each producing multiple different CI models, but all CIs presently available follow the same general design. Figure 1.1 illustrates an overview of the basic setup of a CI in a patient. The implant itself is completely subdermal, with no components or wires going through the skin and the device does not contain an internal battery, so the CI is dependent on external hardware for both signal and power transmission. The external devices consist of a microphone and speech processor, which capture and convert sound into a digital signal that is transmitted to the CI through a pair of sending/receiving coils (one external, the other subdermal), using magnetic induction at radio frequencies. The external coil is usually located in a so-called headpiece, which is connected to the speech processor by wire and is held in place over the location of the subdermal coil using a pair of ferromagnets placed in the coils' centres, though in some designs, the head piece and speech processor are integrated into a single unit.

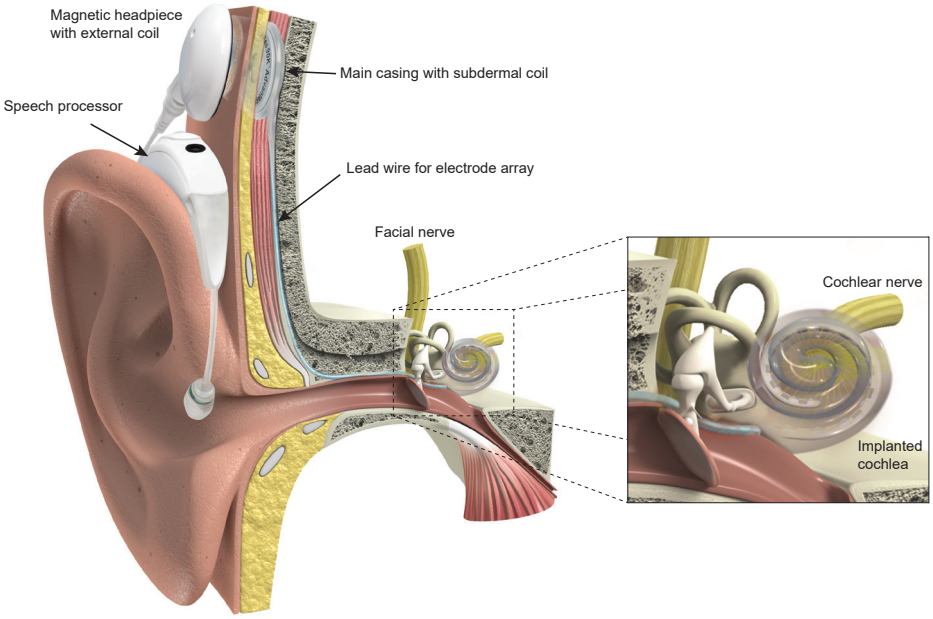

Figure 1.1. *Overview of a cochlear implant and its external components (image courtesy of Advanced Bionics). The inset on the right shows a blow-up of the cochlea implanted with the electrode array.*

The subdermal coil is part of the main casing of the CI, which is embedded in the skull of the patient in a surgically drilled-out bed. The casing also contains the electronic components necessary to process the received signal and activate the intracochlear electrode array, which is connected to the casing by a lead wire that runs through a drilled-out tunnel through the temporal bone that surrounds the cochlea. The electrode array is essentially a thin silicone tube with a number of exposed platinum electrode contacts spaced out along its length, which is inserted into the cochlea through the round window or via a cochleostomy.

The cochlea is, in simple terms, a snail shell-like structure inside the skull, a spiralling tunnel that is divided into three main compartments: the scala tympani, the scala media and the scala vestibuli (figure 1.2). The scala tympani is separated from the other two scalae by the osseous spiral lamina and basilar membrane/organ of Corti, while the scala media and scala vestibuli are separated by Reissner's membrane. In normal acoustic hearing, sound transferred to the cochlea through the outer and middle ear causes the basilar membrane to oscillate, which generates movement of the stereocilia of the outer and inner hair cells in the organ of Corti. The hair cells of these stereocilia respond to this mechanical motion by inducing an electrical signal in the auditory nerve fibres that carry this signal from the organ of Corti, through the osseous spiral lamina into the cochlear modiolus, where all the nerve fibres in the cochlea bundle up and proceed towards the brain.

The basilar membrane's oscillating behaviour is frequency dependent and as such it can be considered a kind of mechanical Fourier transformer. High frequency sound will cause the basilar membrane to resonate mainly at the base of the cochlea and lowering the frequency of the sound will move the area of maximum resonance towards the apex. Since there are auditory neurons spread out along the entire length of the basilar membrane/organ of Corti, this means that each auditory neuron normally only responds to a sound frequency (or frequency range) that corresponds to its position along the length of the basilar membrane. This principle is referred to as the tonotopic organisation of the auditory neurons and it is an important way for the auditory system to convey pitch information to the brain.

CIs take advantage of this tonotopic organisation due to the fact that their electrode arrays have multiple electrode contacts spaced out along the length of the electrode array. When inserted as intended into the scala tympani, each electrode contact will be located at a different depth in the cochlea and will electrically stimulate a different subpopulation of auditory neurons, which will typically be the nerve fibres that are closest to the electrode contact in question. Due to the tonotopic organisation, each electrode contact is therefore expected to invoke a different pitch percept, corresponding to how deeply it is located in the cochlea.

In clinical application, the CI's electrode contacts usually only inject current into the cochlea in so-called monopolar mode. This stimulation mode is so named because only one contact on the electrode array is activated, while the return electrode is located

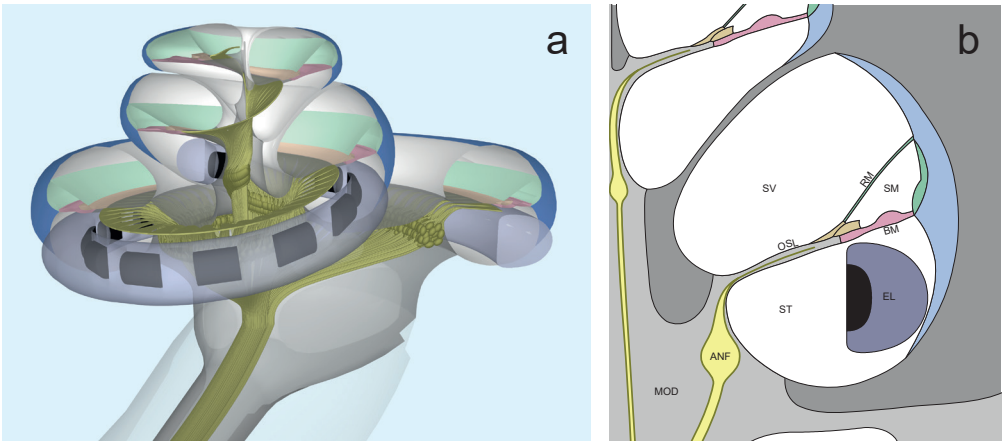

Figure 1.2. *Schematic overview of select structures inside the (human) cochlea. Panel a shows a partial cross-section of a cochlea from the computational model central to this thesis. Panel b shows a line drawing of the cross-section at the beginning of the second turn of the cochlea. The main compartments denoted are the scala vestibuli (SV), scala media (SM) and scala tympani (ST), the latter of which contains the electrode array (EL). These compartments are separated by the osseous spiral lamina (OSL), basilar membrane (BM) and Reissner's membrane (RM). Also depicted are the auditory nerve fibres (ANF), which run though the osseous spiral lamina and the cochlear modiolus (MOD).*

relatively far away from the cochlea (either on the CI's casing itself or through a separate electrode with its own lead wire, placed outside the skull underneath the temporalis muscle), meaning that the electrical field inside the cochlea is by approximation equal to that of a hypothetical monopole.

The stimulus applied to the electrode typically consists of symmetric biphasic pulses, illustrated in figure 1.3a. These pulses are charge-balanced, meaning the same amount of electrical charge passes through the electrode contact during the cathodic phase as during the anodic phase. This is a requirement for safe electrical stimulation of the auditory nerves; failure to charge-balance the pulses could lead to a charge build-up inside the cochlea, which could produce ototoxic reactions and damage the auditory neurons. The duration of the pulse is generally set to a fixed value, while the amplitudes of the pulses are used to increase the intensity of the stimulus (i.e., the loudness perceived by the patient), which is determined dynamically by the external speech processor when it codes sound into electrode array stimuli using a preconfigured speech coding strategy.

In modern speech coding strategies, output is split over a number of separate 'channels', each assigned to a different auditory frequency range, analogous to the natural tonotopic organisation of the cochlea described above. Generally, each channel corresponds to an individual contact along the electrode array, stimulating in monopolar mode, though channels can also be defined as multipolar configurations instead. The output stimuli are arranged so that no two channels are ever stimulated simultaneously, as demonstrated in

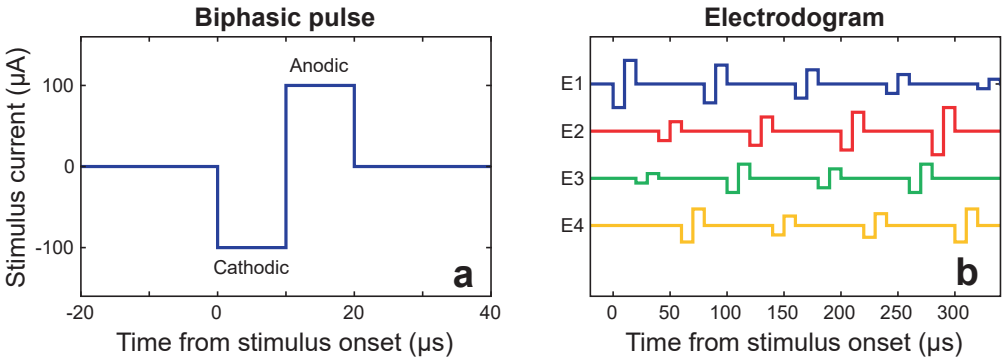

Figure 1.3. *Illustration of biphasic pulses typically used in cochlear implant stimulation. Panel a shows a single, cathodic-first symmetric biphasic pulse with a phase width of 10 μs and an amplitude of 100 μV, plotted over time. Panel b shows a mock stimulation sequence for some hypothetical stimulus on an electrode array with four electrodes/channels (E1-E4). The order of stimulation for the electrodes is staggered; none of them are ever stimulated simultaneously.*

figure 1.3b. This is done to prevent electrical interaction between the individual channels/electrodes, as stimulating multiple intracochlear contacts haphazardly will cause perceptual problems for the patient, due to the electrical fields generated by each active contact unpredictably adding to or subtracting from each other at the level of the auditory nerves.

Though CIs are remarkably successful medical devices, they are not without their limitations. For instance, CI-induced hearing is not comparable to natural hearing; the human cochlea has roughly 30,000 auditory nerve fibres which normally relay signals to the brain more or less independently from each other, while CIs are not able to stimulate these fibres with any finesse or precision. Modern CIs have only 12 to 24 electrode contacts and each contact stimulates a relatively large and wide subpopulation of neurons simultaneously with each stimulus. Though this is sufficient to restore speech understanding in most patients, the quality of the 'sound' leaves something to be desired and CIs are notoriously bad at conveying pitch information accurately. Furthermore, complications can pop up, such as unintended co-stimulation of non-auditory nerve fibres, chiefly in the facial nerve, which runs close to the cochlea, stimulation of which can trigger painful muscle spasms in one side of the patient's face. Therefore, despite the CI's success, there is an ongoing need for further research on how to improve and optimise its design and performance.

Computational modelling

Though the above summary of how CIs function applies to all modern clinically available devices in broad strokes, there are many details in which commercial implants and clinical practices differ. Each manufacturer has made different design decisions and each of them offers multiple devices intended for different clinical circumstances. This means that there is a variety of different CIs available, with different electrode array designs (e.g.,

differing in length, thickness, number of electrode contacts, target placement within the scala tympani) and hardware capabilities (e.g., available stimulus parameters, telemetry options, the ability to stimulate multiple contacts simultaneously in a controlled manner). In addition, the external and internal hardware of modern CIs are versatile enough to allow for more experimental speech coding and stimulation strategies than the ones used in standard clinical settings.

All of this means there is a great need for research into the precise functioning and optimisation of CI-enabled hearing. This is difficult, since the human cochlea itself is small, only roughly 1 cm in maximum diameter, with all of the relevant anatomical structures much smaller than that (for example an individual auditory nerve fibre is only a few micrometres thick) and the whole structure is completely embedded in bone, making it very difficult to interact with it in an ethically responsible way. Experiments with human subjects are therefore mainly limited to psychophysical testing and objective measures obtained through the CI's telemetry capabilities. Animal testing opens up more options, but it has the drawback that there are many anatomical and (neuro)physiological differences between animals and humans, which complicates comparison of animal data to human data and still requires experimentation on the small, difficult to access, inner ear.

However, the underlying physics of electrical neural stimulation can be well understood in terms of classical electromagnetism. The cochlear system is too complex to meaningfully describe CI functioning analytically, but it is an ideal candidate for computational modelling. The electrical fields induced by the CI can be simulated using a volume conduction model that solves the relevant Maxwell equations for a geometrical approximation of the human cochlea and an implanted electrode array. These calculated fields can subsequently be used to simulate responses of auditory nerve fibres, modelled as active electrical networks that react to the CI's stimuli.

The work of this thesis was to develop and use such a computational model to offer insight into the working mechanisms of CI stimulation and to enable virtual experiments that would be unfeasible or outright impossible in real life. The model that was used for this thesis was not built from scratch but was an updated and expanded version of the one developed at Leiden University Medical Centre by Johan H.M. Frijns and Jeroen J. Briaire. In essence, this thesis forms a continuation of their PhD work.

The general goals of each of the modelling studies presented in this thesis were

(I) To improve the accuracy of the Leiden computational CI-model to gain a better understanding of CI-induced hearing.

(II) Where possible, validate the model using available data from electrophysiological or psychophysical experiments.

(III) To use the model to offer predictions that may help to improve the function or design of CIs in the future.

Overview of this thesis

Chapter 2 gives a historical overview of computational modelling of cochlear implants, including the work that directly preceded this thesis (as well as the work presented in chapters 3 through 6).

Chapter 3 describes how the Leiden computational CI model was used to study place-pitch percepts resulting from CI stimuli. For this purpose, the model was updated to include more realistically curved auditory nerve fibre trajectories, based on histological data.

Chapter 4 concerns the modelling of facial nerve stimulation and otosclerosis. A representation of the facial nerve was added to the model, after which the effects of otosclerosis were simulated by reducing the electrical conductivity of temporal bone so that the implications for the stimulation thresholds of the auditory and facial nerves could be examined.

Chapter 5 presents a model analysis of so-called dual electrode stimulation strategies, which involve stimulating two electrode contacts, either simultaneously or in quick succession, in order to create intermediate place-pitch percepts.

Chapter 6 details a study of different current focussing strategies in the model. Current focussing strategies use simultaneous opposite polarity stimulation on multiple electrode contacts in an attempt to restrict spread of neural excitation. For this study, nerve fibre trajectories in the model were updated to include the spatial distribution of cell bodies of the auditory neurons in the cochlear modiolus.

Chapter 7 examines the effect of stimulus polarity in the model and tests the hypothesis that polarity sensitivity to CI stimuli may be used as an indicator of auditory neural health. For this study, the nerve fibre model was updated to one based on available human neurophysiological data.

Finally, chapter 8 presents a general discussion of the work of this thesis and directions for future research.