IMechE Vicon Award Application 2019 **Shreya Singhal**
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Title: Developing a platform for the study of cochlear implants

Background

The objective of the project was to develop an 'artificial cochlea' platform with which to test cochlear implants (CIs). Due to the high conductivity of vestibular fluid and the limited spatial resolution of CIs (12-22 electrodes compared to the 3000 inner hair cells in the human cochlea), CI users often experience difficulty with music and speech perception in noise; therefore, optimising the spread of electrical stimulation at the electrode-auditory nerve interface is a key area of CI research (full literature review beyond scope of this report). However, this research is currently predominantly obtained using electronic feedback from the CIs (Electrical Field Imaging[1]), computational modelling[2][3] or psychometric studies from CI users (e.g. Electrical Compound Action Potentials[6]), which means research is dependent on cadaveric specimens or surgically implanted CI users, and does not experimentally demonstrate what auditory nerves actually see. Therefore, this project looks to develop an artificial cochlea which anatomically and electrically mimics the human cochlea, which, to the author's knowledge, has not been achieved.

Methodology

The 'artificial cochlea' used was a linear tube wired model[1], composed of clear methacrylate UV-cured resin with embedded silver-chloride electrodes and tube diameter anatomically matching the human cochlea[4]. The lumen was filled with NaCl solution and the CI (HiFocus® 1J Electrode, Advanced Bionics) inserted into the lumen. The plastic walls and saline filling of the tube were designed to mimic closely the electrical conductivity of the bony walls of the cochlea and vestibular fluid, respectively[5]. Confirmation of the hardware set-up was performed with impedance measurements and EFI (matched known in-vivo recordings). Essentially, the embedded tube electrodes act as a proxy for auditory nerve endings, their measurements upon implant stimulation therefore indicative of nerve stimulation in-vivo (Figure 1 shows conceptual set-up).

Results

Cathodic-leading charge-balanced biphasic stimuli were used for CI stimulation (using Bionic Ear Data Collection System, Advanced Bionics®). Figure 2 demonstrates an example of a typical tube electrode output waveform, showing the biphasic pulse detected. From this waveform, peak-to-peak amplitude and area under anodic pulse (charge density, AUC) were extracted, since it is unclear which the nerves respond to exactly. Linear regression indicates $1.3 * 10^{-3}\%$ deviation from perfect correlation, so peak-to-peak amplitude can predict AUC and vice-versa. Device stability was confirmed for stimulation amplitude and drift with time (graphs not shown).

3 key areas were investigated: stimulation mode (monopolar uses an extra-cochlear ground, bipolar and tripolar use other CI electrodes as intra-cochlear grounds), effect of distance of intra-cochlear ground (IG) from the stimulating electrode, and extent of IG (i.e. ratio of intra/extra-cochlear ground strength, named partial tripolar mode) - results demonstrated in Figures 4, 5 and 6 respectively.

[1]Designed and fabricated by Thomas Landry, EAR Lab, Biomedical Engineering, Dalhousie University

Discussion

Monopolar, bipolar and tripolar stimulation modes are intended to improve focusing of electrical stimulation (in that order). The results from Figure 4 demonstrate that this is indeed the case when we look at the normalised results. However, there are a couple of interesting phenomena worth noting. Firstly, in both the absolute and normalised cases, we see electrical shunting towards the basal end. It is proposed that this is due to device dimensions - at the apical end, the tube is essentially sealed, whereas the basal end is open to the saline bath, allowing a clear path for current shunting. This mirrors basal shunting in-vivo according to EFI data with monopolar stimulation. Secondly, although we see increased focusing according to stimulation mode in the normalised results, the absolute results indicate that the peak amplitude from tripolar stimulation is ~10x smaller than for monopolar - this is a significant problem for power consumption because peak amplitude will need to reach above a certain threshold for neuronal action potential firing and so a trade-off exists between stimulation focusing and device power consumption.

With regards to effect of distance between IG and stimulating electrodes (Fig 5), we see that they have almost identical focusing ability but at the cost of increased power consumption with reducing distance. We can therefore make preliminary recommendations supporting the use of tripolar stimulation modes with IG electrodes further separated (TP+2), for both improved focusing compared to monopolar or bipolar, and improved power consumption compared to placing closer IGs (TP+0).

Assessing the effect of strength of IG (Fig 6), the data largely obeys similar trends - higher strength IG improves focusing power with the cost of increased power consumption. The trend is confirmed using robust linear regression (graphs not shown), which shows linear proportionality between focusing ability and IG strength.[2]

Conclusions

The physical set-up and data obtained provide novel insight into the proposed stimulation modes for focusing electrical stimulation delivered by cochlear implants. It is the first time (to my knowledge) that data has confirmed the spread of stimulation between the electrode-auditory nerve interface directly and used this as a basis for comparison of stimulation modes. With these results, we can begin to provide truly results-driven recommendations to cochlear implant companies for optimal stimulation modes. Following this project, the device is being extended to more accurately reflect the anatomical and electrical human cochlea using denser electrode arrays and bioengineered materials.

References

[1] Vanpoucke, F. (2004). The Facial Nerve Canal: An Important Cochlear Conduction Path Revealed by Clarion Electrical Field Imaging. *Otology Neurotology*, 25(3), pp.282-289.

[2] Suesserman, M.F. (1993). Lumped-parameter model for in vivo cochlear stimulation. IEEE Trans. *Biomed. Eng.* 40, 237-245.

[3] Kral, A. (1998). Spatial resolution of cochlear implants: the electrical field and excitation of auditory afferents. *Hearing Research*, 121(1-2), pp.11-28.

[4] Thorne, M. (1999). Cochlear Fluid Space Dimensions for Six Species Derived From Reconstructions of Three-Dimensional Magnetic Resonance Images. *The Laryngoscope*, 109(10), pp.1661-1668.

[5] Binette, J. (2004). Tetrapolar Measurement of Electrical Conductivity and Thickness of Articular Cartilage. *Journal of Biomechanical Engineering*, 126(4), p.475.

[6] Tang, Q. (2011). Spatial channel interactions in cochlear implants. *Journal of Neural Engineering*, 8(4), p.046029.

[2] The data presented in this report is currently being prepared for submission for publishing.

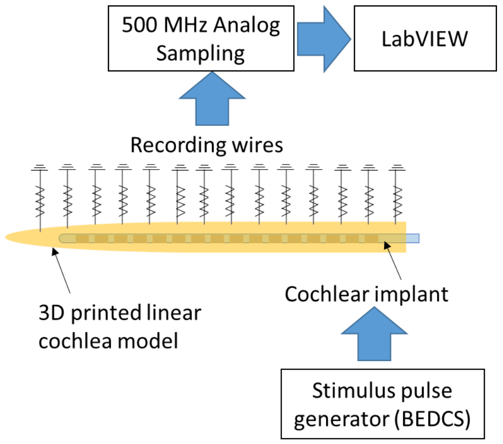

Figure 1: Diagram of measurement set-up

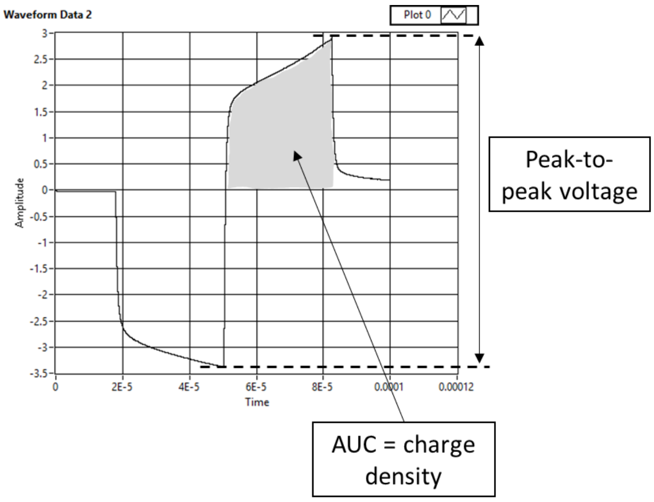

Figure 2: Example of recorded waveform

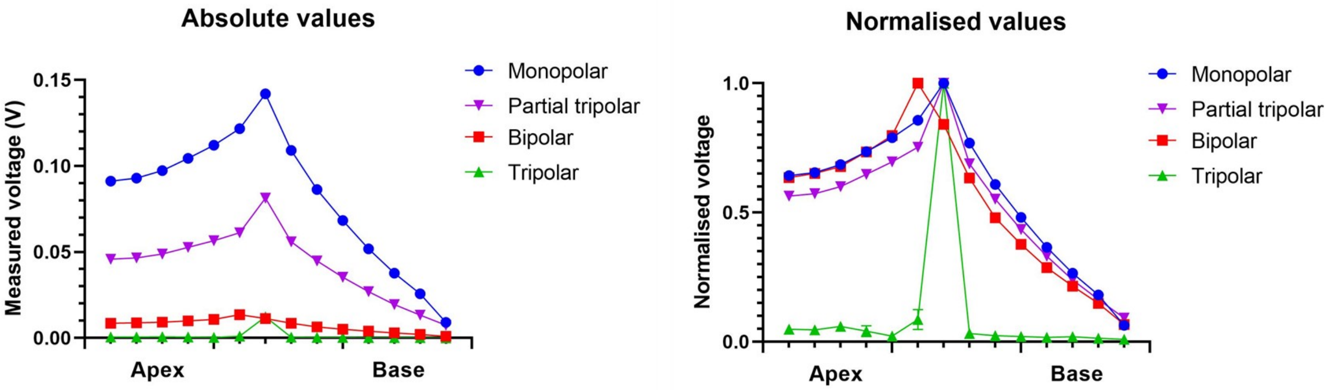

Figure 3: Peak to peak measurements for different stimulation modes

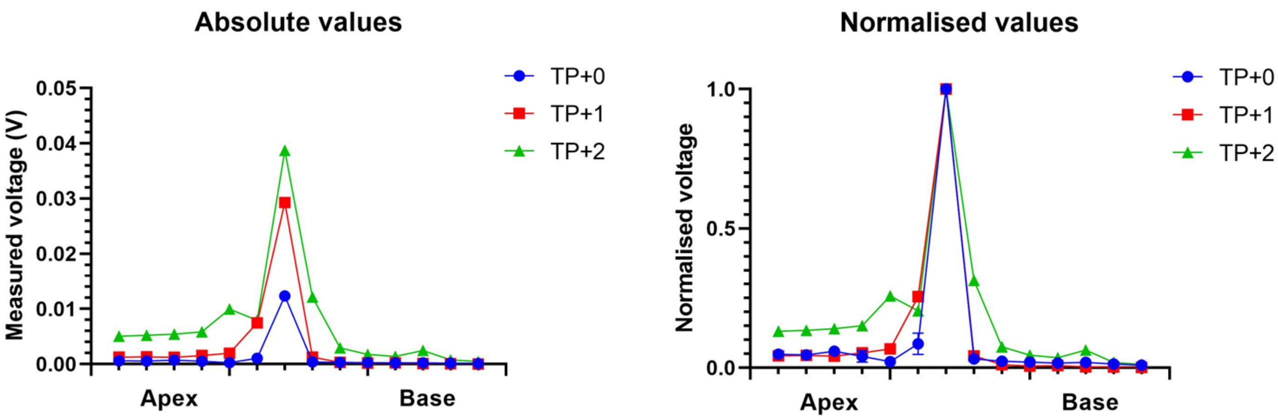

Figure 4: Peak to peak measurements for increased distance between intra-cochlear ground and stimulating electrode. NB, TP+x where x = number of electrodes between stimulating electrode and intra-cochlear ground electrodes.

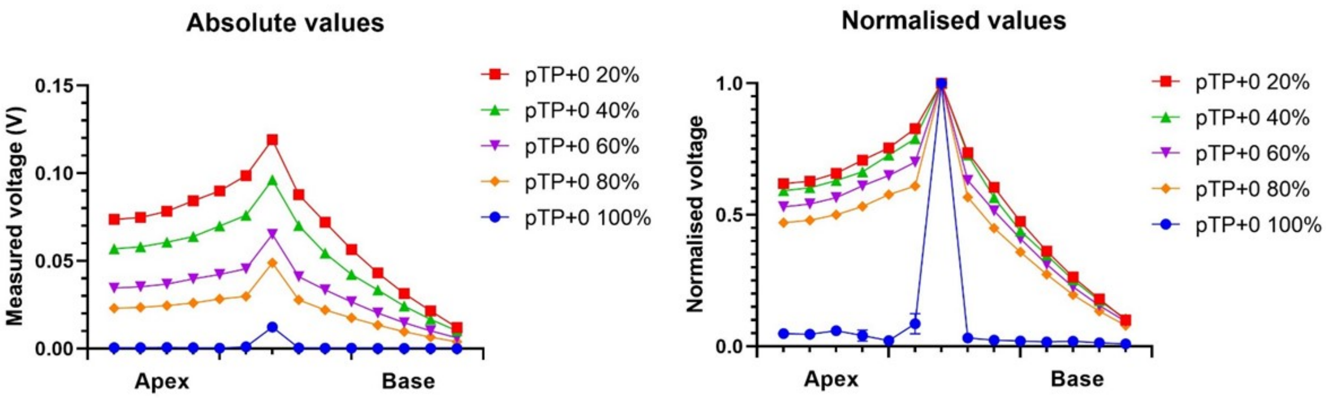

Figure 5: Peak to peak measurements for increasing strength of intra-cochlear ground (partial tripolar mode)