Piloting the Recording of Electrode Voltages (REVS) using Surface Electrodes as a test to Identify Cochlear Implant Electrode Migration, Extra-Cochlear Electrodes and Basal Electrodes Causing Discomfort

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Abstract

Objective: To determine if Electrode Voltage (EV) measurements are potentially suitable as a test for detecting extra-cochlear electrodes in cochlear implants (CIs).

Methods: EV measurements were made using surface electrodes in live mode in 17 adult cochlear implant (CI) users. Repeatability, the effects of stimulation level, CI active electrode position, (active) recording electrode position and stimulation mode (for Nucleus devices) were investigated.

Results/Discussion: Recordings made in monopolar mode showed good repeatability when the active recording electrode was placed on the ipsilateral earlobe; voltages increased linearly with stimulation level as expected. EVs for basal electrodes differed greatly between partially inserted/migrated devices, fully inserted devices with all electrodes activated, and those with deactivated basal electrodes [$\chi^2(2)=10.2$, $p<0.05$ for the most basal electrode].

EVs for Nucleus devices were small for electrodes on the array when compared to those for monopolar return electrodes, except for the participant with extra-cochlear electrodes. We argue that fibrosis around the electrode array facilitated current flow across the round window in this case.

Conclusion: The test appears to be a viable approach to detect electrode migration and extra-cochlear electrodes in adult CI users and may also be sensitive to discomfort caused by current leakage from the basal end of the cochlea.

Keywords: cochlear implant, electrode, voltage, integrity test, partial insertion, migration, extra-cochlear, fibrosis

Introduction

Cochlear implantation is recognised as a highly successful intervention for severe and profound deafness but outcomes vary considerably (Holden et al., 2013) and the intervention carries a small risk of complications. A proportion of this variability can be attributed to the position of the electrode array. Holden et al. found a significant difference in speech perception outcomes for electrodes located in the scala tympani compared with those in the scala vestibuli. Other authors have found that less successful outcomes may result from partial insertion, especially if associated with ossification (Yan et al., 2019). Deterioration in performance may result from migration of the electrode array out of the cochlea (Vaid et al., 2011, Connell et al., 2008). Limited migration has been reported to occur more commonly than was previously suspected (van der Marel et al., 2012), having been found in 10 cases out of a series of 35 individuals, who had had more than one post-operative CT scan. In the majority of cases the second CT scan was not arranged due to concerns about performance with the device. Nine out of the ten instances of migration occurred in 19 patients with lateral wall electrode arrays. Signficant migration potentially leading to re-implantation is relatively rare, accounting for 3.62% of adverse events recorded in the relevant FDA database in 2010 (Causon et al., 2013). Care should be taken to attend to such cases in a timely manner if fibrosis or ossification is developing, as this can make re-implantation more difficult and a more shallow insertion may result (Manrique-Huarte et al., 2016).

Even with a full insertion of the electrode array, stimulation of basal cochlear implant (CI) electrodes can cause pain or discomfort at comfortable listening levels or a lack of auditory sensation in some

recipients (Broomfield et al., 2000, Stoddart and Cooper, 1999). In such cases one or more electrodes require deactivation. At present, there is no easy way to identify basal electrodes which may be at risk of causing discomfort (without actually doing so). Similarly, identification of extra-cochlear electrodes by direct observation at the time of surgery is inaccurate (Holder et al., 2018) and standard clinical procedures such as electrical impedance measurements may give readings within the normal range for electrodes outside the cochlea (Dietz et al., 2016). One study, in which CT scans were performed for recipients with a range of different devices, found that 13% of devices had at least one extra-cochlear electrode (Holder et al., 2018). A fast, reliable test of migration and extra-cochlear position which is easy to perform within an audiology clinic would therefore be very helpful.

Electrode voltage measurements (EVs) are well established as a test of CI device function, as they are used as part of the standard Nucleus integrity testing battery. They are measured using EEG. Typically, surface electrodes are attached to the scalp: on the mastoids, the nape of the neck, the forehead, a combination of these or the earlobes, and EVs are measured by the manufacturer's bespoke equipment or alternatively an evoked potentials system. Voltages are typically averaged over a number of recordings, facilitated by an external trigger to the recording system from the manufacturer's software, which also stimulates the cochlear implant. The voltages recorded are heavily dependent on the position of the implant's return electrode, as shown in figure 1: when the return electrode is outside the cochlea, known as monopolar mode, voltages are typically large and similar for different active electrodes (Mens and Mulder, 2002). If the return electrode is within the cochlea, voltages are considerably smaller, suggesting that current flow is mostly contained within the cochlea. In Common Ground (CG) mode, where the active electrode is on the array and all the remaining electrodes on the array are connected together to act as the return electrode, voltages vary in both amplitude and polarity (Hughes et al., 2004, Kileny et al., 1995, Cullington and Clarke, 1997). In other intra-cochlear modes, where a single electrode acts as the return electrode (pseudomonopolar and bipolar modes), the polarity of the voltage recorded is determined by whether the return electrode is located in a more apical or basal position than the active electrode (Hughes et al., 2004). EVs show these same patterns in both straight and curved electrode arrays (Cullington and Clarke, 1997) (Pijl et al., 2008).

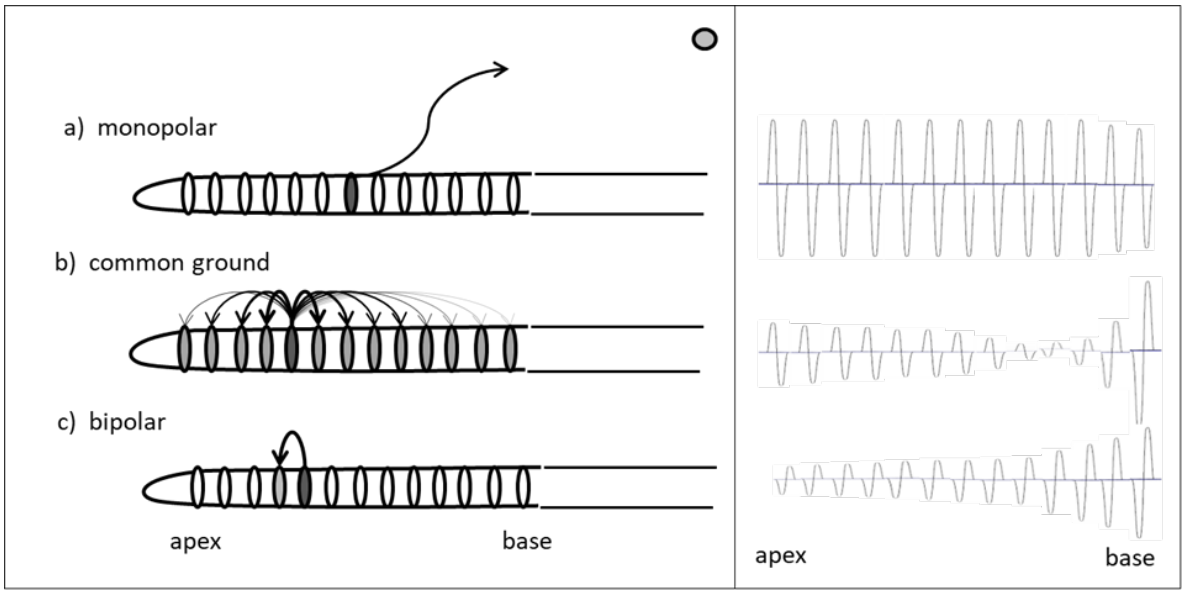


Figure 1: Current flow for different positions of the return electrode in cochlear implants. The left pane shows the direction of current flow in different stimulation modes for a hypothetical electrode array, where the active electrode is shaded dark grey, the return electrode(s) is shaded light grey and other electrodes have no shading. The right pane shows typical EVs associated with activation of each electrode along the array in turn in different stimulation modes, where the active recording electrode is on the ipsilateral mastoid or earlobe (not to scale).

EV measurements in all stimulation modes are effective at identifying open circuits on individual electrodes; short circuits and partial short circuits are evident in common ground mode and bipolar modes, where the return electrode(s) is on the array (Hughes et al., 2004).

EVs also show deviations from the typical patterns in cases of abnormal anatomy (Rotteveel et al., 2008, Mens and Mulder, 2002) and have been found to be different in cases of partial insertion (Monin et al., 2006, Kileny et al., 1995, Mahoney and Proctor, 1994, Shallop, 1993). EVs may be more variable for children than adults (Mahoney and Proctor, 1994). EVs are normally similar between electrodes in monopolar mode (Mens and Mulder, 2002, Mens et al., 1994), but one study found changes of both amplitude and polarity in a minority of paediatric cases (Monin et al., 2006).

Previous EV research has suggested that the current path which allows surface voltages to be measured, at least for intra-cochlear modes, is likely to be through the cochlear fluid along the scala tympani towards the basal end (Hughes, 2013). Mens et al. (Mens et al., 1995) found that when bipolar surface potentials were measured, the measurements reflected the equivalent dipole in the head for otosclerosis patients only. In a later volume conduction modelling study (Mens et al., 1999), they found that the uniform increase of potential with distance between stimulating electrodes found in most implanted subjects was reproduced rather well using a single-cavity cochlea (with a low impedance exit to the ipsilateral skin), in a highly resistive bone compartment, surrounded by the brain. The return of current in non-otosclerosis patients was effectively directed away from the most basal electrode (which was used as the active electrode), irrespective of which intra-cochlear electrode was used as the return electrode. This suggests that the voltage which can be measured on the skin is associated with current which flows in or out of the cochlea via a single current path, rather than in multiple directions, as would be expected if significant current flow from CI electrodes occurred across the bony walls.

More recent volume conduction modelling work for non-otosclerosis cases has focused on reproducing trans-impedance measurements made in monopolar stimulation mode. Detailed models suggest that the majority of the current exits the cochlea via its bony walls (Nogueira et al., 2016, Tran et al., 2015), but 20-30% may exit the cochlea via the basal end. Figure 5 from Tran et al., 2015 suggests that the voltage on the surface is likely to be highest near to the reference electrode. The voltage associated with the current field around the active electrode may not be sufficiently large to be measurable on the skin.

The path which current takes as it leaves the cochlea is of interest for this study, as electrodes which are on the array but outside the cochlea are not in direct contact with the cochlear fluids and current flowing towards the return electrode(s) may take a different path from that for intra-cochlear electrodes, resulting in altered EVs. The voltage measured will be dependent on both the direction of current flow and the conductivity of the material through which the current is flowing.

This article describes work that tests the hypothesis that EVs for partially inserted or migrated arrays would differ from those for fully inserted electrode arrays, assuming normal anatomy and device function in all cases. We also develop an empirically-based model of EVs in different stimulation modes, which provides insights into the likely paths taken by current from fully inserted and partially inserted arrays, based on the data for participants with Nucleus devices.

Materials and Methods

Participants

All participants were adult cochlear implant users. Those in experiment 1 had Advanced Bionics (AB) or MED-EL (M) devices; those who participated in experiment 2 had Nucleus (N) devices. The inclusion criteria for the study were normal device function, as evidenced by normal impedances (except for open circuits related to migration) and no suspicion or diagnosis of a device issue; also normal morphology of the cochlea.

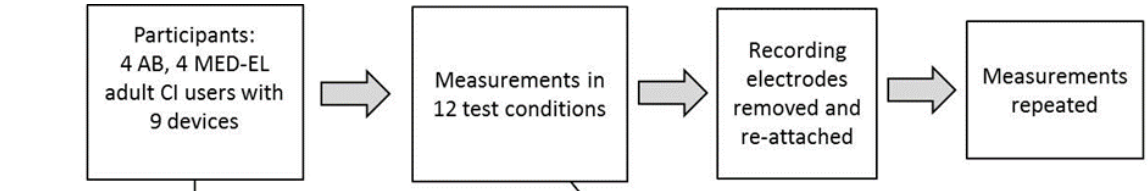
17 participants were recruited with 18 devices; 1 participant was subsequently excluded due to an open circuit and deactivated apical electrodes. Participants 1M6, 1M8L (experiment 1) and 2N5 (experiment 2) had partial insertions or migration. All other participants had full insertions at the time of surgery, confirmed by a post-operative X-ray. Four of these (2N6, 2N8, 1AB3, 1M5) had one basal electrode deactivated in their everyday map and one had four basal electrodes deactivated in their everyday map (1AB2).

Ethical review

Ethical review was performed by the University of Southampton Research Governance Office and by the National Health Service's National Research Ethics Service (Reference numbers 16/WA/0209 and 17/SC/0360).

EVs were recorded using surface electrodes in all cases. No averaging was performed, as the devices were activated in live mode whilst the test was undertaken and triggering is not available in live mode. Hence each measurement was for a single biphasic stimulation pulse on an individual electrode. Peak-to-peak voltages were recorded and were labelled as positive if the first phase was positive, or negative if the first phase was negative.

Experiment 1



Participant	Electrode array	Insertion	basal electrodes deactivated
1AB1	AB Mid Scala	full	0
1AB2	AB Mid Scala	full	4
1AB3	AB Mid Scala	full	1
1M4	MED-EL Flex 28	full	0
1M5	MED-EL Flex 28	full	1
1M6	MED-EL standard	2 outside	4
1AB7	AB Mid Scala	full	0
1M8L	MED-EL standard	1 outside	3
1M8R	MED-EL Flex 28	full	0

Stimulation levels (charge units)	50, 65 and 80 (AB); 6, 8 and 10 (MED-EL)
Active recording electrode positions	Ipsilateral earlobe and mastoid
High pass filter settings	0.1 and 100 Hz

Figure 2: Experimental details for EV Measurements for Experiment 1

Experiment 1 assessed the repeatability of recordings, the effect of stimulation level on EVs, and compared EVs measured with the active recording electrode placed on the mastoid or earlobe. Details are summarised in figure 2.

Surface electrodes were positioned on the ipsilateral mastoid, ipsilateral earlobe, forehead (Fz) and the contralateral mastoid. The active recording electrode was either the ipsilateral mastoid or earlobe; the reference recording electrode was the forehead and the ground electrode was the contralateral mastoid. The skin was wiped with an alcohol wipe prior to attaching the electrode, but only limited skin preparation was undertaken prior to recordings being made.

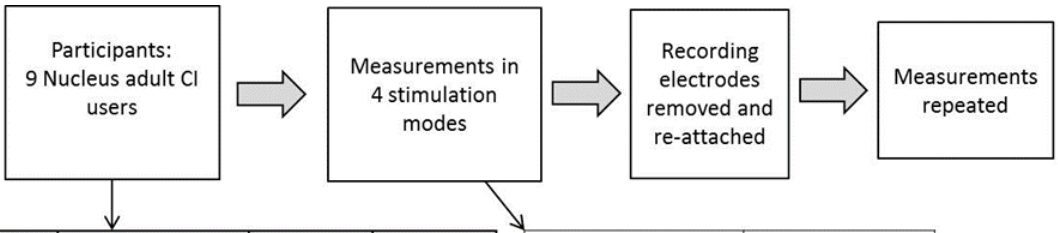
A new map was made for the duration of the test and was activated in live mode whilst the recordings were undertaken. All electrodes were activated, as there were none which showed open or short circuits on impedance telemetry. All devices were programmed in monopolar mode with the implant's reference electrode located on the body of the receiver-stimulator package. Stimulation levels, both threshold and comfort levels, were set to the same level for all electrodes, except for one which was set to a higher level and used as a marker, to ensure that the electrodes could be identified. The voltage measured for the marker electrode was reduced in proportion to the stimulation level in order that all electrodes could be compared with an equivalent stimulation level. The coding strategy used was HD-CIS for MED-EL recipients or HiRes-S for Advanced Bionics (AB) recipients. These strategies stimulate one electrode at a time and the channel order is sequential from apex to base. The pulse width was set manually to 100 µs per phase. The map was activated in live mode but no sound stimuli was used, beyond what naturally occurred in the room, as the strategies used produce constant stimulation at threshold level.

The stimulation levels used for recording were 6, 8 and 10 charge units for MED-EL devices (=60, 80 and 100 µA) and 50, 65 and 80 charge units for AB recipients (=39, 51, 62 µA); current levels for AB recipients were limited by participants' comfort levels. A Biologic Auditory Evoked Potentials system was used for recording. The high pass filter used for recording was either 0.1 Hz or 100 Hz. The recording window was 5.33 ms, the low pass filter was 15 kHz and the gain was 100. A single recording was made for each combination of stimulation level, high pass filter setting and position of

the active recording electrode. The electrodes were then removed and the patient given a break if desired. Following this the electrodes were re-applied and a further set of recordings were made.

The participant's usual map was reviewed and any deactivated electrodes were noted.

Experiment 2



Participant	Electrode array	Insertion	Basal electrodes deactivated (number of open circuit electrodes)
2N1	Contour advance	full	0 (0)
2N2	Contour advance	full	0 (0)
2N3	Contour advance	full	0 (0)
2N5	Slim straight	6 outside	11 (2)
2N6	Contour advance	full	1 (0)
2N7	Contour advance	full	0 (0)
2N8	Contour advance	full	1 (0)
2N9	Slim straight	full	0 (0)

Stimulation mode	Common ground, monopolar 1, monopolar 2 and variable bipolar (reference = 20)
Active recording electrode position	Ipsilateral earlobe
High pass filter setting	0.1 Hz

Figure 3: Experimental details for experiment 2

Experiment 2 assessed the effect of stimulation mode on EVs measured with surface electrodes and the repeatability of these measurements; details are shown in figure 3.

Nine adult participants were recruited with Nucleus devices: all devices had been fully inserted but one device had migrated to some extent (confirmed by a CT scan). EVs were recorded for this device on two occasions, separated by three months, during which further migration had occurred, as evidenced by altered impedance telemetry. On the first occasion, electrodes 4 and 5 had open circuits, whereas on the second occasion, electrodes 5 and 6 showed open circuits; electrodes 1 to 6 were considered to be extra-cochlear at this point. The EVs from the second session are included in the analysis.

Surface electrodes were positioned on the ipsilateral earlobe (active), forehead (Fz) (reference) and the contralateral mastoid (ground). Skin preparation was limited, as in experiment 1.

Recordings were made in monopolar modes MP1 and MP2, also in common ground mode (CG) and in variable bipolar mode (VBP). In this mode, the return electrode is the same regardless of which active electrode is used: in this case the return electrode was always E20, towards the apical end of the array. The same recording system and parameters were used as in experiment 1 but the high pass filter for recording was always set to 0.1 Hz. The active electrodes were 1 to 22 in all modes except VBP mode, where recordings were made for electrodes 1 to 10. Electrodes which were open circuit at the time of testing were not included and were assigned a voltage of 0 μV for the purpose of analysis. No short circuits were recorded for any device.

The processing strategy used was ACE for monopolar modes and SPEAK for CG and BP modes. The stimulation rate was set to 250 Hz for all maps for consistency. There is a clear gap between stimulation pulses at this rate. Neither the SPEAK or ACE strategy provides constant stimulation at threshold level, so a sound stimulus was constructed for the purpose of the experiment. This was comprised of a series of frequency modulated tones, with base frequencies at the centre frequencies of the filters for a 12-channel map and modulated by 5%. The level of each tone was based on an estimate of the effect of the microphone on stimulation levels for different frequencies. The modulation rate varied between tones so as to minimise any reduction in level during the sound presentation. The sound was presented at 70 dB(A) via Sony supra-aural headphones and was looped for continuous presentation. All participants used behind-the-ear processors (CP910 or CP810).

The pulse width was set to 50 μs on all active channels except one, which was set to 100 μs (either E8 or E15). This channel was used as the marker, to enable individual electrodes to be identified. The stimulation order for Nucleus devices is sequential from base to apex; the recording for the marker electrode was identified by its wider pulse shape and the recordings corresponding to the remaining electrodes were labelled sequentially, consistent with the stimulation order. The stimulation levels were set to 80 current units for threshold levels and 82 units for comfort levels for monopolar recordings (or the highest comfortable level available) and the highest available comfortable level for CG and VBP recordings. The highest comfortable stimulation levels were used for CG and VBP recordings as EV amplitudes can be small in these modes, with the intention of minimising the number of EVs in the noise floor. The dynamic range was always set to the minimum value of two units. After the measurements were made, all voltages were normalised to the equivalent of an 80 current unit stimulation level, using the relevant equation from the manufacturer (=76 μA). The number of maxima available is limited to eight for SPEAK maps and so this limited the number of electrodes which could be measured at any one time. The number of maxima was set equal to the number of active electrodes in each map. The electrode array was tested in three sections, E22 to E15, E15 to E8 and E8 to E1 in all modes except VBP, where E10 to E6 were tested and then E5 to E1. It was expected that eight biphasic pulses would be present between marker stimuli in each recording, or five in VBP mode. If there was a gap in the recordings, indicating that one electrode had been missed, the recording was repeated if there was no repeat of the electrode within the trace. Recordings were made in each stimulation mode, then the recording electrodes were removed and the participant was given a break if required and the recordings were repeated.

The participant's usual map was reviewed and any deactivated electrodes were noted.

Results

Experiment 1

The repeatability of recordings made with the active recording electrode on the earlobe and mastoid are shown in figure 4. In each case the reference recording electrode was the forehead (Fz) and the ground electrode was the contralateral mastoid.

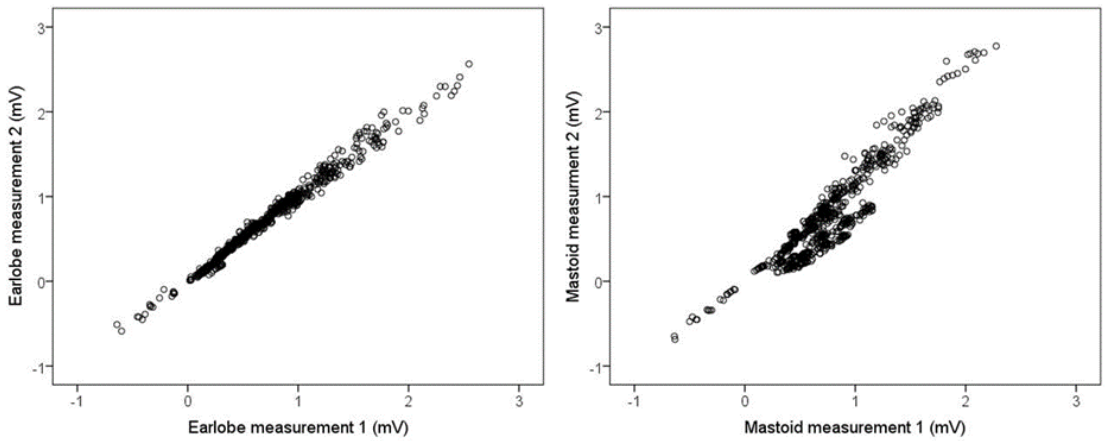


Figure 4: Electrode Voltages for repeated measurements on the mastoid and earlobe

A 2-tailed non-parametric correlation was performed, Spearman's ρ, and a significant correlation was found between repeated measurements on both the earlobe and mastoid (ρ=0.995, p<0.001 for the earlobe and ρ=0.889, p<0.001 for the mastoid).

The greater variability for the mastoid recordings was investigated. Figure 5 shows the recordings for participant 1AB2 at the highest stimulation level tested of 80 charge units. All traces show a decrease in the voltages at the basal end of the array compared to the apical end. Following removal and replacement of the recording electrodes, the voltages were much lower across the electrode array. However, it is noticeable that the voltages have not changed relative to their original value: there is a drop of approximately 350 µV for all electrodes. This suggests that there are two components to these EVs, one which is common to all electrodes and is sensitive to the position of the recording electrode on the mastoid and another which affects only the basal electrodes and is insensitive to the position of the recording electrode on the mastoid.

This effect was not observed for earlobe recordings for the same participant, also shown in figure 5. There was a drop in the voltages towards the basal end of the electrode array but very little difference in the voltages for repeated measurements.

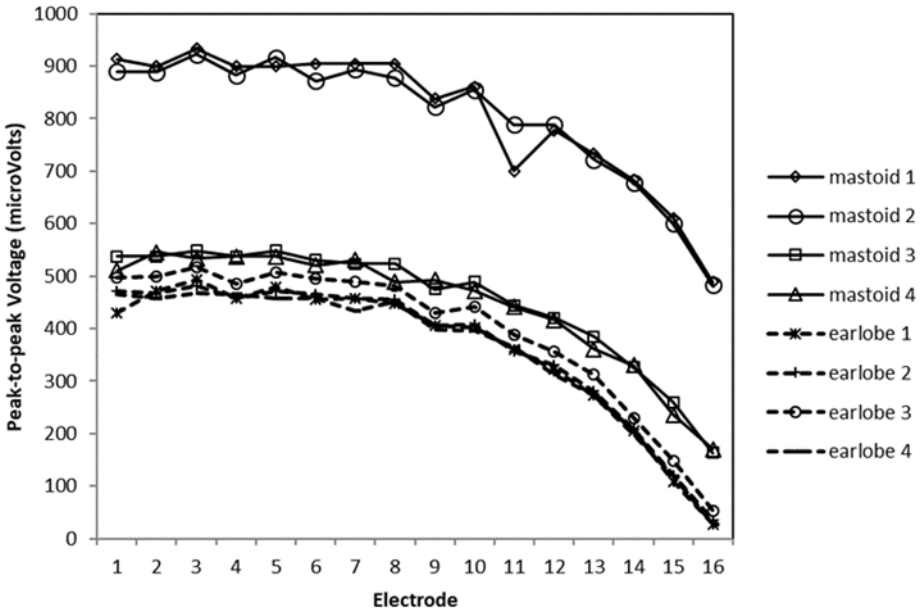


Figure 5: Electrode voltages for participant 1AB2 at 80 charge units with the active recording electrode on the mastoid and earlobe: dotted lines show earlobe recordings and solid lines show mastoid recordings.

Measurements for different stimulation levels showed expected increases with level across the electrode array when the active recording electrode was placed on the earlobe, as shown in figure 6.

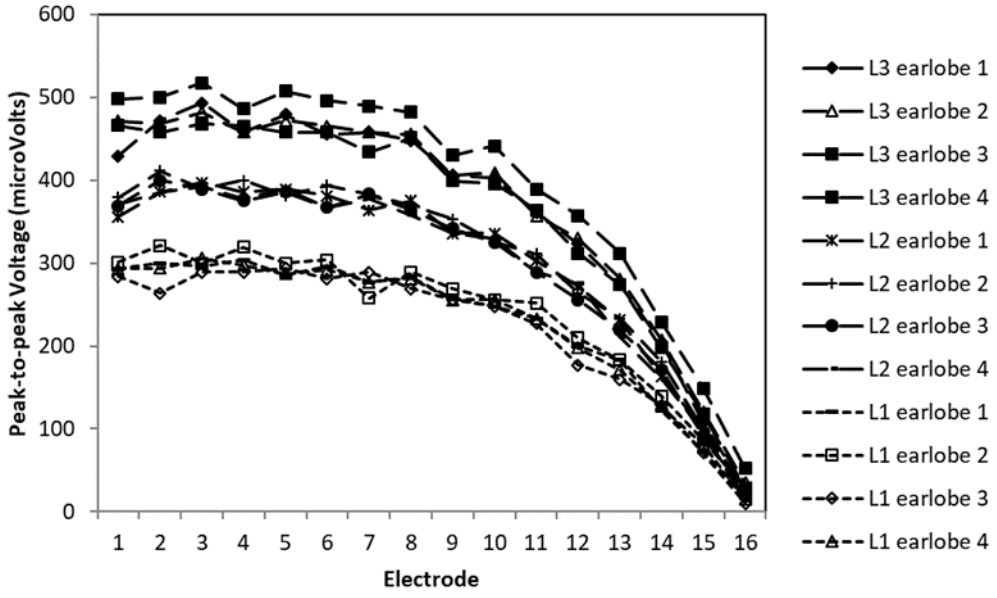


Figure 6: EVs for participant 1AB2 in different test conditions and at stimulation levels of 50, 65 and 80 charge units.

The average EV and standard error for the earlobe and mastoid recordings were reviewed for the level 3 measurements for individual participants. Standard errors for individual electrodes for the four level 3 measurements were calculated for earlobe and mastoid recordings and were averaged

across the array and expressed as a function of the mean EV for all electrodes for each participant. These are shown in figure 7. For earlobe measurements, the standard error was less than 5% of the mean EV in all but one case (participant AB1). This participant also had the smallest absolute voltages and some traces showed marked baseline drift (earlobe level 3 average 213 μV; average for the group 959 μV). For mastoid measurements, the standard error was less than 5% of the mean EV in only three out of nine cases.

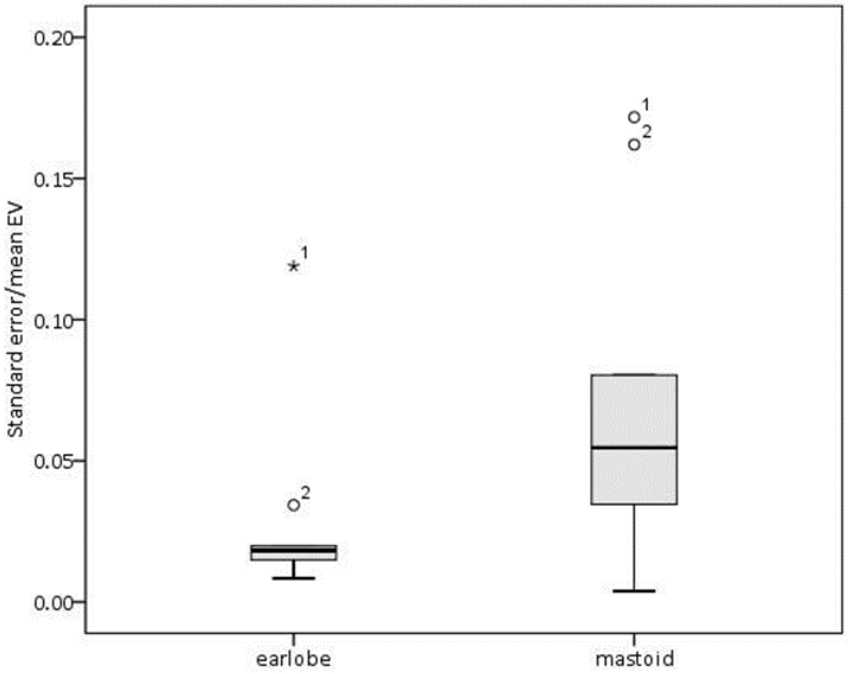


Figure 7: Standard errors for four level 3 measurements averaged across the electrode array for individual participants, expressed as a fraction of the mean value for earlobe and mastoid recordings.

The remaining analysis was performed with the earlobe recordings only.

EVs were averaged across electrodes for each condition. Average EVs were found to be normally distributed for each condition (Shapiro-Wilk $p>0.05$). Repeated measures ANOVA was performed to investigate the effects of stimulation level, filter setting and first or second measurement.

Mauchly's test of sphericity gave a significant result ($p<0.001$), so the Greenhouse-Geisser correction was applied. There was no significant effect of filter setting [$F(1,8)=0.677$, $p>0.05$] and no significant effect of repeated recording, as expected [$F(1,8)=0.177$, $p>0.05$] but there was a significant effect of stimulation level [$F(2,16)=20.7$, $p<0.01$]. This can be seen in figure 8. The changes of EV amplitude are consistent with the changes in stimulation level used. There were no significant interactions between any variables ($p>0.05$).

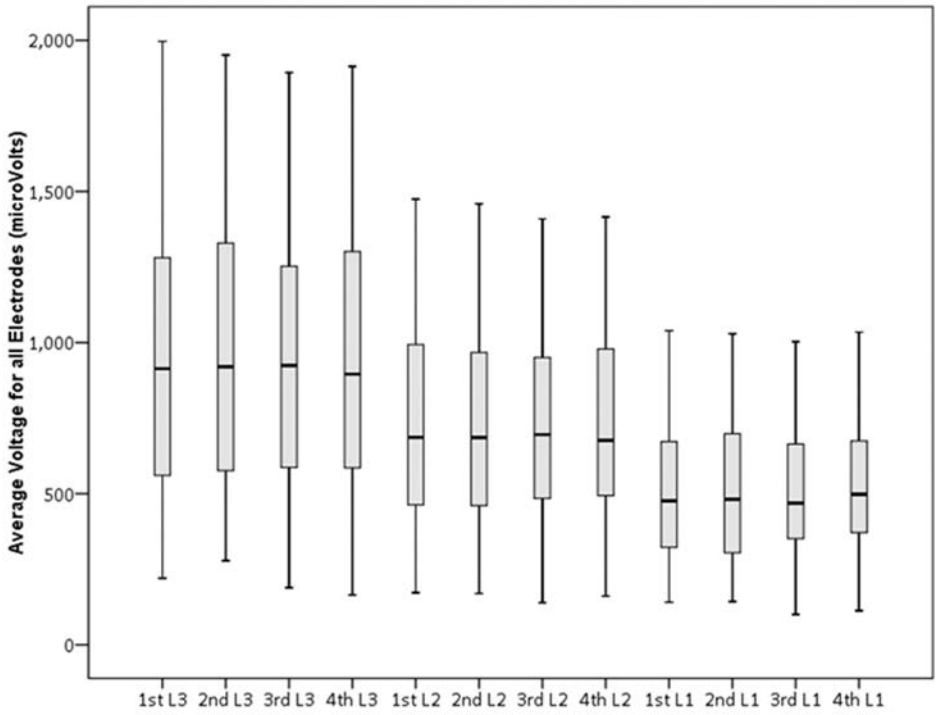


Figure 8: Boxplots showing the voltages averaged across electrodes in different test conditions. The heavy line shows the median value, the boxes represent the 25th and 75th centiles and the whiskers represent the range of data. L represents the stimulation level, which was either 1,2 or 3.

Experiment 2

First a check of the repeatability of measurements was performed. Measurements were repeated at the same level and pulse width (two recordings) for electrode 15 in MP1, MP2 and CG modes. The standard error for repeated measurements was calculated for each mode of stimulation for each participant and in each case was found to be less than 5% of the mean EV for that electrode. However, in two cases, the comparison was not possible in CG mode, due to the voltages being within the noise floor.

EV measurements in different stimulation modes are shown in figure 9.

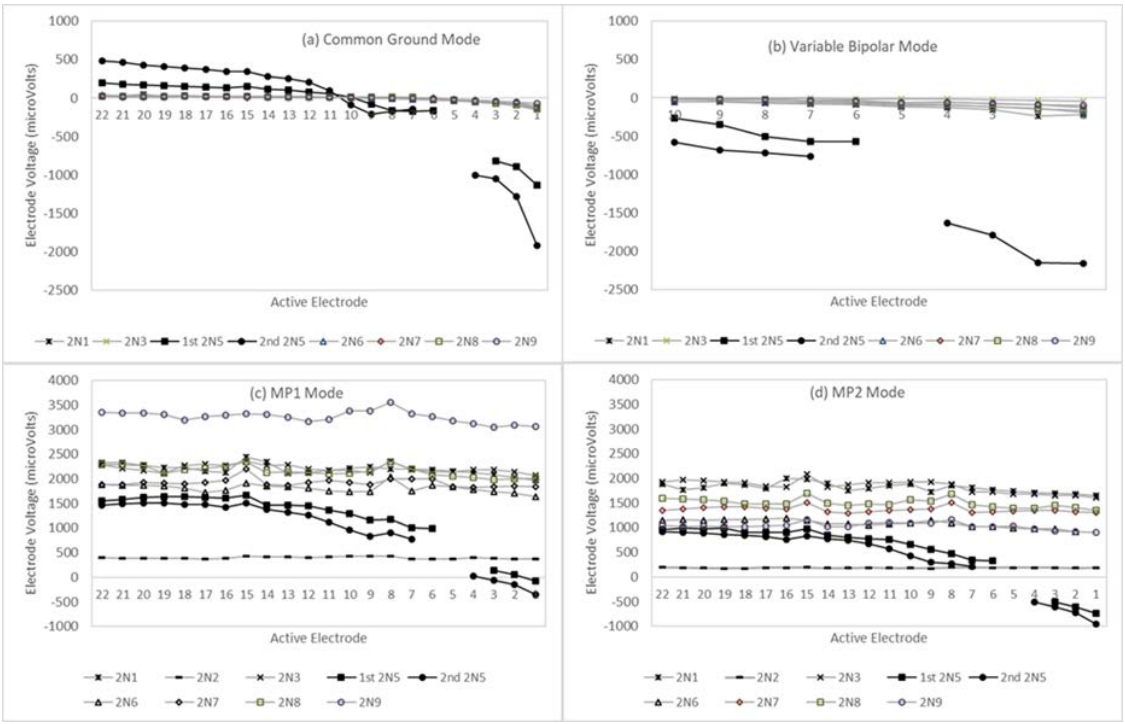


Figure 9: EVs in different stimulation modes for participants with Nucleus devices. EVs are shown for all 8 participants in MP1 and MP2 modes; in CG and VBP modes, the results for 7 participants are shown, as the voltages for P2 were too small to measure. EVs for participant 2N5 with electrode migration are shown for two time intervals. The first measurement was made 3-months post-activation and the second one six months post-activation. At 3-months post-activation, electrodes 1 to 5 were extra-cochlear; at 6-months post-activation, electrodes 1 to 6 were extra-cochlear; electrodes 7 to 22 were intra-cochlear. Note the differences in scale for different stimulation modes.

EVs were compared for electrode 10, as this electrode was tested in all modes and is in the middle of the electrode array, so was intra-cochlear for participant 2N5. Friedman's test was used, as the data were not normally distributed [$\chi^2(3)=21.0$, $p<0.001$]. Wilcoxon signed ranks test was used to compare EVs for electrode 10 between different stimulation modes. This gave the same significant result for all comparisons [$Z=-2.37$, $p<0.05$].

Eleven electrodes were assigned a voltage of 0 in CG mode around the null point, due to the voltages being within the noise floor.

EVs were very different for the migration case 2N5 when compared with those for participants with full insertions. In monopolar modes, EV amplitudes dropped substantially at the basal end for the migration case, becoming negative for the extra-cochlear electrodes, especially in MP2 mode. CG and VBP EVs are small for participants with full insertions, relative to the MP1 and MP2 EVs. The amplitudes of the CG and VBP EVs are an order of magnitude larger for the migration case than for those with full insertions for most electrodes.

In order to investigate the effect of active electrode position (intra- or extra-cochlear) on monopolar EVs in more detail, the results for experiments 1 and 2 were combined for those with repeatable recordings (N=16). There were three participants with extra-cochlear electrodes; all of these also

had deactivated intra-cochlear basal electrodes. A further five participants with full insertions had deactivated basal electrodes. These participants were considered as a separate group from the participants with full insertions and all electrodes activated. In order to compare results for implants from different manufacturers with different numbers of electrodes, six electrodes representing different portions of the array were identified from across the electrode array, as shown in table 1.

Table 1 Labelling of Electrodes for Different Devices

Electrode	Advanced Bionics	MED-EL	Nucleus
Apical 1 (A1)	2	2	21
Apical 2 (A2)	5	4	17
Medial 1 (M1)	8	6	13
Medial 2 (M2)	10	8	9
Basal 1 (B1)	13	10	5
Basal 2 (B2)	16	12	1

EVs for these electrodes were compared with the most apical electrode in each case (E1 for AB and MED-EL, E22 for Nucleus), as shown in figure 10.

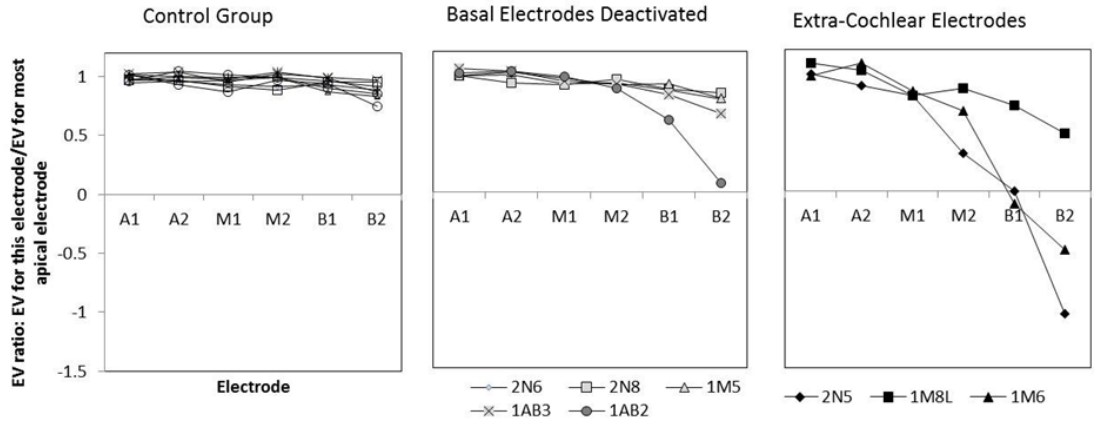


Figure 10: Monopolar EVs relative to the EV for the most apical electrode for all participants for both experiments for two apical, two medial and two basal electrodes. Within the basal electrodes deactivated group, participants with one electrode deactivated have markers with light shading and the participant with four electrodes deactivated has dark shading.

Within the control group, EVs were compared for the six selected electrodes using repeated measures ANOVA, as these were normally distributed. A significant effect of electrode was found $F(5,7)=7.87$, $p<0.001$. Pairwise comparisons with Bonferroni corrections showed that the EVs for apical electrode A2 were significantly different for those for basal electrode B2.

The non-parametric Kruskal-Wallis test was used to compare individual electrodes between the three groups (fully inserted, all electrodes activated; fully inserted, basal electrode(s) deactivated; partially inserted/migrated). Bonferroni corrections for multiple comparisons were applied where appropriate. A significant effect was found for basal electrodes B1 [$\chi^2(2)=10.6$, $p<0.05$] and B2 [$\chi^2(2)=10.2$, $p<0.05$]. There was no significant effect of group for apical electrodes A1 [$\chi^2(2)=2.84$,

p>0.05]or A2 [$\chi^2(2)$=0.759, or middle electrodes p>0.05], M1 [$\chi^2(2)$=7.01, p>0.05] or M2 [$\chi^2(2)$=8.95, p>0.05]. Pairwise comparisons between groups for the basal electrodes, using the Mann-Whitney test, gave significant differences between the control group and the basal electrodes deactivated group for basal electrodes B1 [Z=-2.49, p<0.05] and B2 [Z=-2.34, p<0.05]. Similarly, for the extra-cochlear electrodes group, the Mann-Whitney test gave significant differences between this group and the control group for basal electrodes B1 [Z=-2.45, p<0.05] and B2 [Z=-2.45, p<0.05].

As a significant difference was observed between EVs for basal electrodes between the full insertion, basal electrodes deactivated and extra-cochlear electrodes group, further analysis was undertaken to investigate EVs for activated and deactivated electrodes. EVs for the last active electrode were compared with those for the first deactivated electrode, both relative to the most apical electrode, as shown in figure 11 for the eight individuals with at least one deactivated basal electrode. A paired samples t-test [t(7)=3.56, p<0.01] found that the average EV for the first deactivated electrode was significantly lower than that for the last active electrode. This is unlikely to be due to deactivation, as EVs change smoothly across the electrode array for these participants, with no discontinuity at the point of deactivation. The drop in EV suggests that there is a relationship between the active electrode voltage (relative to that of the most apical electrode) and the comfort of stimulation on the electrode.

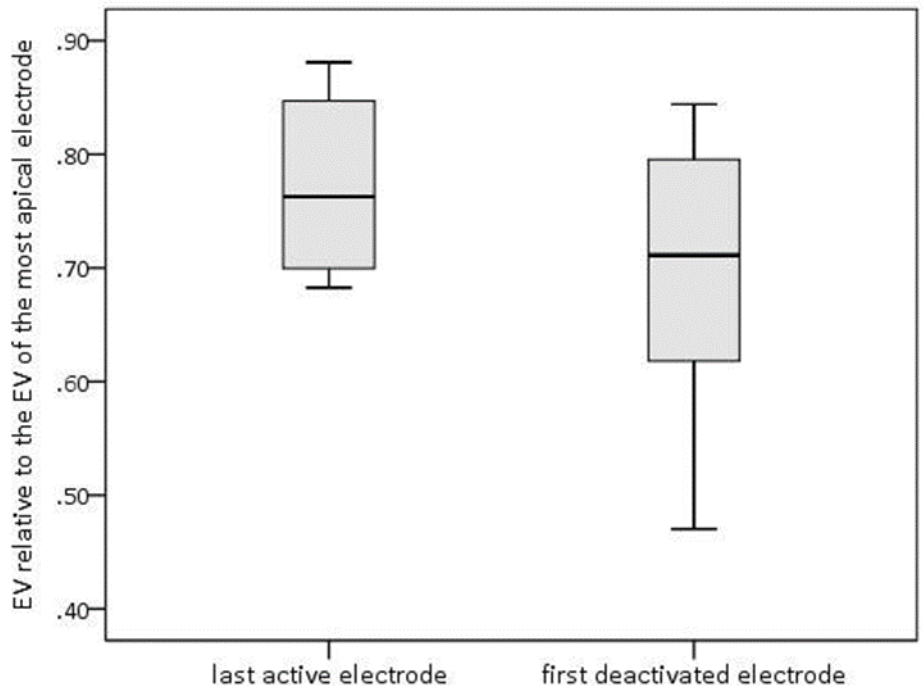


Figure 11: Electrode voltages for the last active electrode and first deactivated electrode relative to that for the most apical electrode.

Discussion

We aimed to evaluate the application of EV measurements, a test normally undertaken to determine device hardware faults, for the detection of extra-cochlear electrodes and associated electrode migration. A voltage change of >20% between adjacent electrodes is considered abnormal when EV

measurements are made for the purpose of testing a CI device (Goehring et al., 2014). In order for these comparisons to be made, repeated measurements on the same electrode should give only small changes of amplitude if these comparisons are to be meaningful. We considered that the standard error of the measurements should be <5% of the mean voltage.

EV measurements without averaging met this criterion value in monopolar mode for 16 out of 17 participants, when the active recording electrode was placed on the ipsilateral earlobe and the reference recording electrode on the forehead. This suggests that this method of recording is suitable for use in clinic, with the proviso that the repeatability of the measurements is checked by making at least two measurements with the same recording conditions. If the voltages are not repeatable, a higher stimulation level may improve the repeatability of the measurements. Each recording takes less than five seconds, so repeating the measurements or making further recordings using different stimulation levels is not at all onerous. On the contrary, this method of recording voltages is convenient, as it does not require bespoke equipment and can be performed using equipment which is commonly used for evoked potential measurements in audiology clinics.

EV measurements were less repeatable on the mastoid, suggesting that the voltages change significantly in this area over short distances. The mastoid is also larger than the earlobe, so the placement of the recording electrode is less likely to be consistent.

Common ground recordings were successful for some participants without averaging but not for all, due to the voltages being very small and within the noise floor in some cases, especially around the null point. As such it would be better to use averaging for common ground recordings, to reduce noise and facilitate measurements in a greater proportion of adult cochlear implant users for a larger number of electrodes.

As reported in previous studies (Hughes et al., 2004), EVs differed between stimulation modes. Intra-cochlear modes gave small EVs for electrode arrays which were fully inserted; VBP recordings gave negative leading traces, similar to measurements reported in the literature for bipolar+1 recordings (Hughes, 2013), whilst monopolar traces were positive leading for all electrodes for fully inserted devices. The profiles of CG EVs across electrode arrays for fully inserted devices followed a distinct pattern, which is typical of CG traces reported in the literature (Garnham et al., 2001, Hughes, 2013).

Model of EVs in Different Stimulation Modes

In order to assist with the interpretation of recordings in different stimulation modes, a simple empirically-based mathematical model was developed. As surface potentials are a far-field recording, the measured voltage is likely to be dominated by differences in the current path between the electrodes and the skin, rather than by the interface between each of the electrodes and the immediately adjacent tissue. If current flows out from the cochlea via both the basal end and the bony walls, as modelling based on trans-impedance measurements suggests, there will be at least two current paths contributing to the measured voltages for each electrode pair; wheras if current flows out via the basal end only, as Mens et al.'s model suggests for bipolar stimulation, then the voltage measured at the surface in bipolar modes can be written as:

$$V_m(a) = V_{be}(a) + V_{be}(r) \qquad [1]$$

Where $V_m(a)$ is the measured voltage and $V_{be}(a)$ and $V_{be}(r)$ are potentials at the surface associated with stimulation of the current path from the active and return electrodes via the basal end of the cochlea to the skin respectively.

If this holds true, it will also be possible to predict the voltages in CG mode from voltages measured in bipolar modes, because current flow will be via the same current path and the measured voltages in CG mode will be as follows:

$$V_m(a) = \frac{1}{Nr}\sum_{i=1}^{Nr} V_{be}(a) + V_{be}(r_i) \qquad [2]$$

Where V_m is now the measured CG voltage, Nr is the number of return electrodes (21 for Nucleus devices) and $V_{be}(r_i)$ are the potentials at the surface associated with stimulation of the return electrodes in bipolar modes. We hypothesise that $V_{be}(a)$ and $V_{be}(r_i)$ reflect a natural distribution of voltages in the cochlea for this current path and are equal for a given electrode but of opposite charge. The mean $V_{be}(r)$ for E20 was estimated for the five participants in experiment 2 with full insertions and a full set of recordings in each stimulation mode, by extrapolating the mean VBP voltages in the apical direction. This was based on the assumption that the voltages will diminish in magnitude towards the apical end, as shown in figure 12, in keeping with recordings made in other studies in bipolar stimulation modes e.g. (Cullington and Clarke, 1997). This value was then added to the extended VBP voltage measurements to estimate the mean voltage at the surface for each active electrode $V_{be}(a)$ and from there the mean $V_m(a)$ in CG mode was predicted (also shown in figure 12). The correlation between the anticipated and measured values was very strong (Spearman's ρ = 0.992, p<0.01).

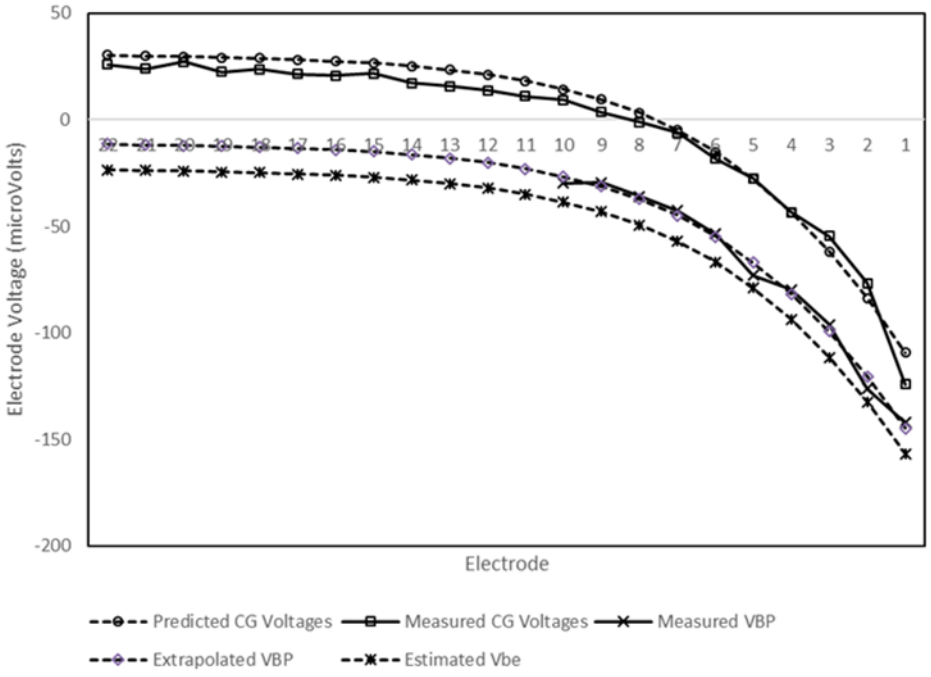


Figure 12: Measured and Predicted EVs in CG and VBP modes for the control group, and predicted voltages associated with stimulation of active electrodes on the array, with current exiting the cochlea at the basal end (Vbe)

The monopolar voltages were also considered. As the return electrodes are distant from the electrode array, there will be separate paths for the return electrodes to the skin. In MP2 mode the electrode is positioned directly underneath the skin, so we expect there to be a single current path associated with that electrode and the measured voltage to be:

$$V_m(a) = V_{be}(a) + V_2(r) \qquad [3]$$

Where $V_2(r)$ is the voltage at the surface associated with stimulation of the return electrode; this should be the same for each $V_{be}(a)$. The estimated $V_{be}(a)$ from the extended VBP recordings was subtracted from $V_m(a)$ for the monopolar recordings, giving an estimate of the mean value of $V_2(r)$ for MP2 recordings. The same procedure was applied for MP1 recordings, although in this case the return electrode is not in direct contact with the skin so there may be more than one current path from the return electrode to the skin. The estimated $V_2(r)$ values are shown in figure 13.

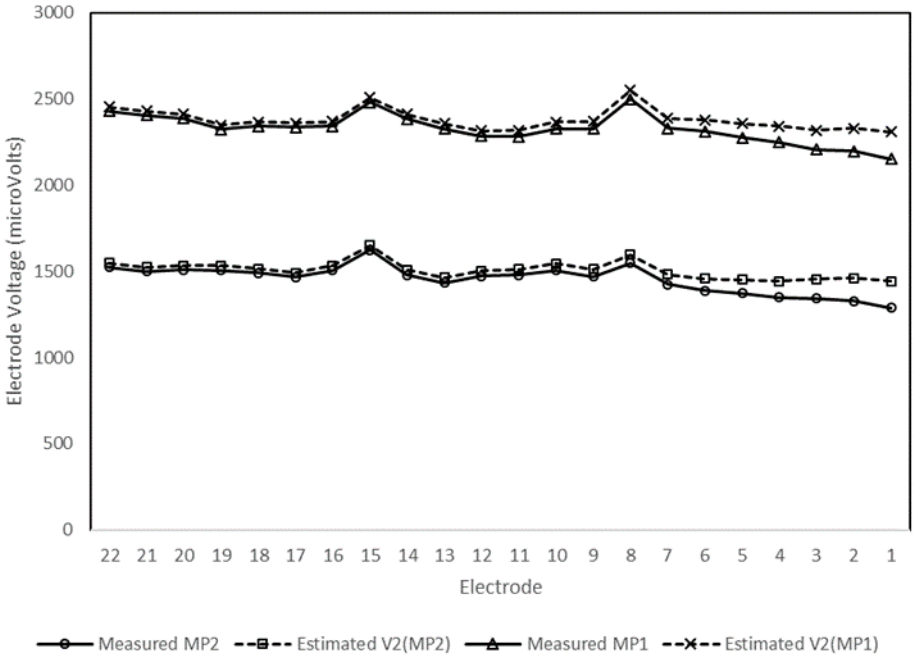


Figure 13: Measured MP EVs and predicted voltages associated with stimulation of CI return electrodes (V2(MP1) and V2(MP2)) for the control group

$V_2(r)$ should have the same value for each active electrode in each monopolar mode; here the estimated value of $V_2(r)$ is similar across the electrode array (except for E8 and E15, where the stimuli had wider pulse widths), suggesting that the same current path from the active electrode via the basal end to the skin was used in monopolar stimulation as in CG and VBP stimulation. The fact that $V_2(r)$ is so much larger than $V_{be}(a)$ suggests that only a limited amount of current exited the cochlea or alternatively that the current path from the cochlea to the recording electrode was much more difficult than that from the return electrode to the skin, for those within the control group.

For the migration case, VBP recordings were only available for four intra-cochlear electrodes, meaning that $V_{be}(E20)$ could not be easily estimated from the VBP measurements. Instead, the variation in $V_{be}(a)$ across the electrode array was taken from the MP2 recordings and then $V_{be}(a)$ for each electrode was estimated by adding an offset in keeping with the VBP recordings. This is shown

in figure 14. Finally, $V_m(a)$ in CG mode was predicted and compared with the measured values, also shown in figure 14. The agreement between the predicted and measured values was very good (Spearman's ρ=0.991, $p<0.01$), suggesting that there was only one current path between the active electrode and the skin in the migration case also. The voltages were very much larger in CG and VBP modes when compared to those for the control group, suggesting that more current was able to exit the cochlea in the migration case. V_{be} for the basal electrodes was larger in magnitude than $V_2(r)$, so there is no suggestion of a difficult current path from the cochlea to the recording electrode on the surface.

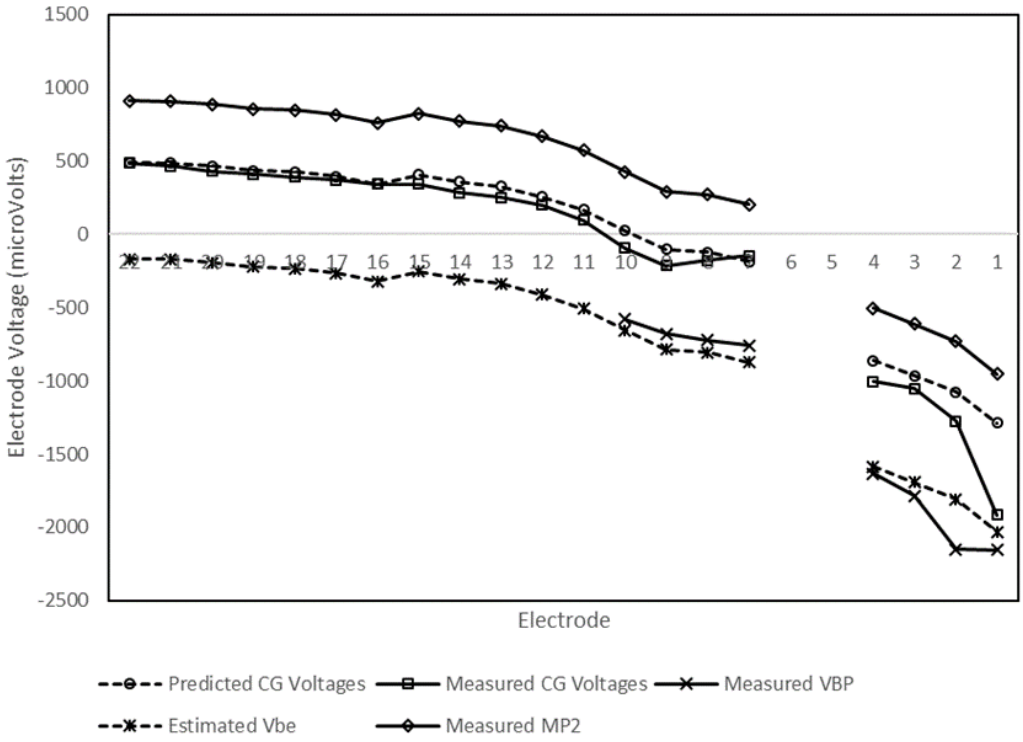


Figure 14: Measured EVs in CG, VBP and MP2 modes for the migration case, and predicted voltages in CG mode and those associated with stimulation of active electrodes on the array, with current exiting the cochlea at the basal end (Vbe)

Temporal bone studies have shown that fibrotic tissue frequently develops around CI electrode arrays, but varies considerably in thickness and morphology between individuals (Kamakura and Nadol, 2016). We considered the possibility that fibrosis might have facilitated current flow across the round window for both intra-cochlear and extra-cochlear electrodes. The implant which had migrated was explanted and the patient received a new device. Extensive tissue growth around the electrode array was observed at the time of revision surgery. On removal of the array a cuff of tissue was found which was firmly adhered to the array. This tissue was fixed then removed from the array and prepared for histological analyses. A case report detailing the hearing performance and electrical measures along with the distribution of connective tissue and cell types within and along the fibrotic mass has been accepted for publication(Hough et al., in press).

It is likely that current flowed through the fibrotic tissue for the extra-cochlear electrodes in the migration case, which may account for the normal impedance measurements. The EV measurements suggest that the direction of current flow in MP modes was back into the cochlea rather than along the electrode wire towards the receiver/stimulator (as negative leading pulses are associated with current flow in the apical direction, as observed for measurements made in variable bipolar mode). They indicate that the current path for extra-cochlear electrodes was similar to that for intra-cochlear electrodes but their presence increased the amount of current flow along that path; this was evidenced by large EVs in intra-cochlear modes, even when apical electrodes were stimulated.

Looking back to figure 2 and the finding that there were two contributions to the MP EVs for participant 1AB2, the likely explanation for this is that $V_2(r)$ varied across the mastoid but not on the earlobe. The earlobe is well placed to pick up voltages associated with stimulation of electrodes both on the array and on the receiver/stimulator package.

Another interesting finding was that EVs differed for intra-cochlear basal electrodes which were activated or deactivated, although this was not associated with a discontinuity in the voltages along the electrode array at the point of deactivation. The deactivated intra-cochlear electrodes had all been deactivated prior to the study, typically because of discomfort or poor loudness growth. A possible explanation for the discomfort or poor loudness growth which resulted in deactivation is that current was able to flow out of the cochlea into the middle ear for the basal electrodes, causing discomfort. The EVs for the basal electrodes suggest that the limit of comfort for basal electrodes may be represented by a drop of approximately 30% in the EV when compared with the EV for the most apical electrode. Further investigation of this effect would be beneficial, including comparing voltages for basal electrodes with different apical electrodes, as the most apical electrode did not always have the highest EV. This suggestion is consistent with the observation that the number of deactivated electrodes was approximately double the number of extra-cochlear electrodes for the three patients with extra-cochlear electrodes in this study.

Conclusions

Electrode voltage measurements for CI electrodes were made without averaging using standard evoked potentials equipment and were found to be mostly repeatable for recordings made in monopolar mode with adult CI users with the active recording electrode on the earlobe and the reference recording electrode on the forehead.

EVs measured in monopolar modes differed greatly for basal electrodes for fully inserted devices with all electrodes activated compared with those with deactivated basal electrodes or extra-cochlear electrodes. In two cases with extra-cochlear electrodes, basal EVs had reversed polarity compared to EVs for apical electrodes.

The polarity reversals suggest that the voltage associated with current leakage from the cochlea via the basal end was larger than that recorded from the return electrode. In one case, the device was explanted as a result of migration and it was found that a large amount of fibrotic tissue had developed around the electrode array. We hypothesise that this facilitated current flow into and out of the cochlea. If a migration case does not result in the development of fibrosis around the extruded electrodes, electrode impedances may show an open circuit and EVs may have very low

amplitudes on these electrodes. A single electrical test for electrode migration may not be possible but a combination of measures could be effective for this purpose.

Our results suggest that recordings of EV made in monopolar mode with adult CI users are sensitive to the position of the basal electrodes and may be suitable as a test for electrode migration, particularly when combined with impedance or trans-impedance measurements. The results also yield novel insights into the nature of current flow from CI electrodes, suggesting that the basal end is the main source of current leakage from the cochlea for adults with normal anatomy. Further investigation is required to develop the test for routine use in CI clinics.

Declaration of Interest Statement

The authors report no conflict of interest.

References

BROOMFIELD, S., MAWMAN, D., WOOLFORD, T. J., O'DRISCOLL, M., LUFF, D. & RAMSDEN, R. T. 2000. Non-auditory stimulation in adult cochlear implant users. *Cochlear implants international,* 1**,** 55-66.

CAUSON, A., VERSCHUUR, C. & NEWMAN, T. A. 2013. Trends in Cochlear Implant Complications: Implications for Improving Long-Term Outcomes. *Otology & Neurotology,* 34**,** 259-265.

CONNELL, S. S., BALKANY, T. J., HODGES, A. V., TELISCHI, F. F., ANGELI, S. I. & ESHRAGHI, A. A. 2008. Electrode migration after cochlear implantation. *Otology & Neurotology,* 29**,** 156-159.

CULLINGTON, H. E. & CLARKE, G. P. 1997. Integrity testing of cochlear implants in the awake child. *British Journal of Audiology,* 31**,** 247-256.

DIETZ, A., WENNSTROM, M., LEHTIMAKI, A., LOPPONEN, H. & VALTONEN, H. 2016. Electrode migration after cochlear implant surgery: more common than expected? *European Archives of Oto-Rhino-Laryngology,* 273**,** 1411-1418.

GARNHAM, J., MARSDEN, J. & MASON, S. M. 2001. Profiles of AEVs for intra- and post-operative integrity test measurements in young children with the Nucleus mini 22 cochlear implant. *British Journal of Audiology,* 35**,** 31-42.

GOEHRING, J. L., NEFF, D. L., BAUDHUIN, J. L. & HUGHES, M. L. 2014. Pitch ranking, electrode discrimination, and physiological spread of excitation using current steering in cochlear implants. *Journal of the Acoustical Society of America,* 136**,** 3159-3171.

HOLDEN, L. K., FINLEY, C. C., FIRSZT, J. B., HOLDEN, T. A., BRENNER, C., POTTS, L. G., GOTTER, B. D., VANDERHOOF, S. S., MISPAGEL, K., HEYDEBRAND, G. & SKINNER, M. W. 2013. Factors Affecting Open-Set Word Recognition in Adults With Cochlear Implants. *Ear and Hearing,* 34**,** 342-360.

HOLDER, J. T., KESSLER, D. M., NOBLE, J. H., GIFFORD, R. H. & LABADIE, R. F. 2018. Prevalence of Extracochlear Electrodes: Computerized Tomography Scans, Cochlear Implant Maps, and Operative Reports. *Otology & Neurotology,* 39**,** E325-E331.

Hough, Kate , Sanderson, Alan Peter , Grasmeder, Mary , Mitchell, Tim , Verschuur, Carl and Newman, Tracey (2020) Inflammation at the tissue-electrode interface in a case of rapid deterioration in hearing performance leading to explant after cochlear implantation. *Otology & Neurotology* **.** (In Press) HUGHES, M. L. 2013. *Objective Measures in Cochlear Implants*, Plural.

HUGHES, M. L., BROWN, C. J. & ABBAS, P. J. 2004. Sensitivity and specificity of averaged electrode voltage measures in cochlear implant recipients. *Ear and Hearing,* 25**,** 431-446.

KAMAKURA, T. & NADOL, J. B. 2016. Correlation between word recognition score and intracochlear new bone and fibrous tissue after cochlear implantation in the human. *Hearing Research,* 339**,** 132-141.

KILENY, P. R., MEITELES, L. Z., ZWOLAN, T. A. & TELIAN, S. A. 1995. COCHLEAR IMPLANT DEVICE FAILURE - DIAGNOSIS AND MANAGEMENT. *American Journal of Otology,* 16**,** 164-171.

MAHONEY, M. J. & PROCTOR, L. A. R. 1994. THE USE OF AVERAGED ELECTRODE VOLTAGES TO ASSESS THE FUNCTION OF NUCLEUS INTERNAL COCHLEAR IMPLANT DEVICES IN CHILDREN. *Ear and Hearing,* 15**,** 177-183.

MANRIQUE-HUARTE, R., HUARTE, A. & MANRIQUE, M. J. 2016. Surgical findings and auditory performance after cochlear implant revision surgery. *European Archives of Oto-Rhino-Laryngology,* 273**,** 621-629.

MENS, L. H. M., BROKX, J. P. L. & VAN DEN BROEK, P. 1995. Averaged electrode voltages: Management of electrode failures in children, fluctuating threshold and comfort levels, and otosclerosis. *Annals of Otology Rhinology and Laryngology,* 104**,** 169-172.

MENS, L. H. M., HUISKAMP, G., OOSTENDORP, T. & VAN DEN BROEK, P. 1999. Modelling surface potentials from intracochlear electrical stimulation. *Scandinavian Audiology,* 28**,** 249-255.

MENS, L. H. M. & MULDER, J. J. S. 2002. Averaged electrode voltages in users of the Clarion cochlear implant device. *Annals of Otology Rhinology and Laryngology,* 111**,** 370-375.

MENS, L. H. M., OOSTENDORP, T. & VANDENBROEK, P. 1994. IDENTIFYING ELECTRODE FAILURES WITH COCHLEAR IMPLANT GENERATED SURFACE-POTENTIALS. *Ear and Hearing,* 15**,** 330-338.

MONIN, D. L., KAZAHAYA, K. & FRANCK, K. H. 2006. Routine use of the crystal device integrity testing system in pediatric patients. *Journal of the American Academy of Audiology,* 17**,** 722-732.

NOGUEIRA, W., SCHURZIG, D., BUCHNER, A., PENNINGER, R. T. & WURFEL, W. 2016. Validation of a Cochlear Implant Patient-Specific Model of the Voltage Distribution in a Clinical Setting. *Frontiers in Bioengineering and Biotechnology,* 4**,** 16.

PIJL, S., WESTERBERG, B. D., GUSTIN, C. & FONG, L. 2008. Objective Findings with Malpositioning of a Nucleus 24RE(CA) Cochlear Implant. *Journal of the American Academy of Audiology,* 19**,** 435-442.

ROTTEVEEL, L. J. C., BEYNON, A. J., MENS, L. H. M., SNIK, A. F. M., MULDER, J. J. & MYLANUS, E. A. M. 2008. Cochlear implantation in 3 patients with osteogenesis imperfecta: Imaging, surgery and programming issues. *Audiology and Neuro-Otology,* 13**,** 73-85.

SHALLOP, J. K. 1993. OBJECTIVE ELECTROPHYSIOLOGICAL MEASURES FROM COCHLEAR IMPLANT PATIENTS. *Ear and Hearing,* 14**,** 58-63.

STODDART, R. L. & COOPER, H. R. 1999. Electrode complications in 100 adults with multichannel cochlear implants. *Journal of Laryngology and Otology,* 113**,** 18-20.

TRAN, P., SUE, A., WONG, P., LI, Q. & CARTER, P. 2015. Development of HEATHER for Cochlear Implant Stimulation Using a New Modeling Workflow. *Ieee Transactions on Biomedical Engineering,* 62**,** 728-735.

VAID, N., ROLAND, J. T. & VAID, S. 2011. Extracochlear electrode extrusion. *Cochlear implants international,* 12**,** 177-80.

VAN DER MAREL, K. S., VERBIST, B. M., BRIAIRE, J. J., JOEMAI, R. M. S. & FRIJNS, J. H. M. 2012. Electrode Migration in Cochlear Implant Patients: Not an Exception. *Audiology and Neuro-Otology,* 17**,** 275-281.

YAN, T., ZONG, F. R., MA, X. J., XU, X. B., CHEN, W. L., SONG, Z. Y., HAN, X., WANG, X. J. & ZHANG, H. B. 2019. Cochlear implantation in patients with ossified cochleas. *American Journal of Otolaryngology,* 40**,** 183-186.