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Impedance Measures during *in vitro* Cochlear Implantation predict Array Positioning

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Abstract

Objective—Improper electrode placement during cochlear implant (CI) insertion can adversely affect speech perception outcomes. However, the intraoperative methods to determine positioning are limited. Because measures of electrode impedance can be made quickly, the goal of this study was to assess the relationship between CI impedance and proximity to adjacent structures.

Methods—An Advanced Bionics CI array was inserted into a clear, plastic cochlea one electrode contact at a time in a saline bath (9 trials). At each insertion depth, response to biphasic current pulses were used to calculate access resistance (R_a), polarization resistance (R_p), and polarization capacitance (C_p). These measures were correlated to actual proximity as assessed by microscopy using linear regression models. **Results**: Impedance increased with insertion depth and proximity to the inner wall. Specifically, R_a increased, C_p decreased, and R_p slightly increased. Incorporating all impedance measures afforded a prediction model ($r = 0.88$) while optimizing for sub-mm positioning afforded a model with 78.3% specificity.

Conclusion—Impedance *in vitro* greatly changes with electrode insertion depth and proximity to adjacent structures in a predictable manner.

Significance—Assessing proximity of the CI to adjacent structures is a significant first step in qualifying the electrode-neural interface. This information should aid in CI fitting, which should help maximize hearing and speech outcomes with a CI. Additionally, knowledge of the relationship between impedance and positioning could have utility in other tissue implants in the brain, retina, or spinal cord.

Index Terms

impedance measurement; cochlear implants; implantable biomedical devices

I. Introduction

COCHLEAR implantation (CI) is a major advance in auditory rehabilitation. Although speech perception outcomes with CIs have improved on average over time, these outcomes are still highly variable [1]. Biographical factors, audiological factors, and surgical approach can only account for 25% of the variance in speech outcomes [2], and up to ~40% when incorporating the magnitude of tone-evoked responses of the cochlea just before implantation [3]. A substantial portion of the remaining variance is likely due to positioning of the CI relative to the auditory nerve as a result of implantation [4].

An ideally placed CI electrode is completely contained within the scala tympani (ST), coiled along the curvature of the cochlea and facing inwards towards the modiolus, a bony structure which contains the auditory nerve's spiral ganglion cell bodies. Recent improvements in post-insertion imaging have strengthened the intuitive relationship between major placement errors, i.e., electrodes that traverse from ST (correct placement) into scala vestibuli (adjacent structure), and poor speech perception outcomes [5–8].

Yet even for completely-within-ST insertions, speech outcomes are still highly variable [9] and may be influenced by differences in the proximity to and integrity of the electrode-neural interface [10]. For instance when individual CI electrode contacts are farther from the modiolus the stimulus current level needed to activate adjacent neurons is greater [11], which can increase the spread of excitation and the risk of channel-channel interactions known to inhibit speech discrimination [12]. As such, it would be useful to assess the proximity of each electrode contact to the modiolar wall in order to optimize stimulation parameters to obtain the best possible speech outcomes [5].

Correct placement by surgeons has always been the goal during implantation [13], but visual inspection of the implant is limited to the site of insertion and tactile perception is limited in its ability to predict overall cochlear positioning [14]. Post-operative CT imaging is the best way to determine positioning, but is not routinely performed in adults and because of the radiation risk it will never be routinely performed in children. However, intra-operative device testing is almost universally employed [15]. Three measures which can be rapidly assessed from the implant at the time of implantation include electrically-evoked neural responses, electric field imaging (EFI), and impedances [16].

The clinical utility of evoked responses in assessing electrode positioning is highly variable. Mittmann et al. found a correlation between evoked response thresholds and completely-within-scala positioning [17], while Miller et al. found no correlation of response thresholds to either intrascalar position or outcomes [18]. In a study of 2,365 CI insertions in human adults, the relationship between electrode-to-modiolus distance and the minimum current level to reach response threshold was significant, but very weak within a given array type

($r=0.12$) [19]. As such, evoked responses are not a reliable contributor to determining electrode geometry relative to adjacent structures.

With regard to EFI, Vanpoucke et al. thoroughly modelled the equivalent circuit of a CI within the ST [20]. The model was able to detect major placement errors, including tip-rollover of the implant and ossification of the cochlea, but was not utilized to predict the relative medio-to-lateral orientation within a scala [21]. A final metric, impedance, is typically used to assess CI integrity after implantation yet unique recent approaches to impedance measurement have gleaned additional information about the electrode's surroundings and proximity to adjacent tissue.

One experimental approach, utilized by Tan et al., analyzed impedance of electrode contacts at two stages during CI insertion: before and after stylet removal, a surgical step in some CI models which tightly wraps the array around the modiolus of the cochlea [22]. They used fluoroscopy to confirm that stylet removal successfully caused the array to coil inwards and increases were seen in impedance at nearly every CI contact, consistent with models which describe the electrode- electrolyte interface and electrolyte-tissue interface as major contributors to resistance [20]. This same approach was also employed by Pile et al. who similarly found the change in impedance across electrodes increased after stylet removal, and could vary by surgical insertion technique [23]. Tykocinski et al. followed post-insertion impedance in CI subjects over time, further characterizing the electrode-electrolyte interface to track inflammation and fibrosis [24].

The common approach utilized by these three groups to compare *changes* in impedance is paramount, because impedance on its own is not correlated with CI positioning [25]. This is because differences in baseline impedances between CI electrode contacts can routinely vary by kOhm [26], completely obscuring changes in the aforementioned studies which could be within hundreds of Ohms. A technique for characterizing the electrode-saline interface of an implant before introducing biologic tissue has been utilized in other applications [27] but has not been utilized in CI arrays, and was the foundation for this research.

The goal of this study was to determine the relationship between CI impedance and proximity to adjacent structures in a saline environment. Our approach was to assess impedance across all contacts of a CI throughout sub-steps of insertion into a plastic cochlea submerged in saline. Our hypothesis was that impedance characterization of the electrode-saline interface before CI insertion would allow changes in impedance during insertion to sensitively infer electrode proximity to adjacent structures.

II. Methods

A. Cochlear Implant and Current Pulse Stimuli

An Advanced Bionics (AB) Hi-Focus 1j CI electrode array was used for all experiments (Valencia, CA, USA). The AB 1j electrode array has 16 individual platinum contacts (E1 to E16) which curl inwards, towards the modiolus. Contacts are spaced apart by 1.1 mm, leading to an overall active array length of 17 mm. At the base of the implant array there is a full- circumference ground electrode (Fig. 1, top), but the processor case can also be used as

a ground in a clinical setting. Electrode contacts and the ground are individually shielded within a flexible silicon carrier and connected internally to the processor via platinum-iridium wires. Each electrode is driven by a separate current source, and the processor has a built-in amplifier with an analog-to-digital converter which can sample at 56 kHz with 9 bit resolution (Fig. 1, center). Sampled data is sent via telemetry from the implant's magnet to the processor and interpreted with Advance Bionics Bionic Ear Data Collection System (BEDCS) software. Stimuli were biphasic pulses with an amplitude of 34 uA lasting 179.6 usec per phase, separated by 100 ms to minimize any interference with charging between subsequent pulses. At a 56 kHz sample rate, 33 samples per recording epoch were taken in monopolar recording mode, that is to say the recording electrode and stimulating electrode were the same, and potentials were recorded relative to the ring electrode ground.

B. Calculation of Access Resistance and Polarization Impedance

The approach used here is described by Tykocinski et al., who model the CI electrode-electrolyte interface as a resistor and capacitor in parallel, and the bulk tissue resistance as a resistor in series [24]. The implant generates a current pulse which passes from an electrode contact through the surrounding medium and returns on the ring ground (Fig. 1, bottom). The major sources of impedance are the bulk resistance through the cochlear tissue (R_a) and the impedance at the electrode-electrolyte interface (Z_p), which are in series. Polarization Impedance (Z_p) is composed of both resistive (R_p , Faradaic Resistance) and capacitive (C_p , double layer capacitance) elements in parallel.

The response waveform (Fig. 2, center) to a long stimulus pulse (Fig. 2, top) of an implant in saline demonstrates the two sources of voltage increase consistent with this model: an immediate voltage increase from the frequency-independent resistive elements between the contact and the ground (access voltage, V_a) and a slowly-rising limb demonstrating a charge accumulation at the electrode-electrolyte interface (polarization voltage, V_p) as in (1).

$$V_{\text{tot}}(t) = V_a(t) + V_p(t) \quad (1)$$

Calculation of access resistance is simply the access voltage at the first sampled time point in V_a divided by the current pulse amplitude (2).

$$R_a = \frac{V_a}{i} \quad (2)$$

The total polarization impedance is the voltage growth after the first time point until the end of the first phase divided by the current used (3).

$$Z_p(t)_{180\text{us}} = \frac{V_p(t)_{180\text{us}}}{i} \quad (3)$$

Unlike the R_a , Z_p changes as a function of time. In our model, the relationship between R_p , C_p , and Z_p are described as a standard RC circuit:

$$V_p(t) = i * Z_p(t) \quad (4)$$

$$Z_p(t) = R_p[1 - \exp\left(\frac{-t}{R_p * C_p}\right)] \quad (5)$$

Using Matlab (MathWorks, Natick, MA) to fit the $Z_p(t)$ segment of the measured response (4) to the model function (5), it is possible to approximate magnitudes of R_p and C_p (Fig. 2, bottom).

C. Recording Impedance in the Plastic Cochlea during CI Insertion

The AB CI was placed in a saline bath alongside a clear, 3D printed plastic cochlea. The cochlea was printed to mimic the approximate size of the human ST [28, 29], but was slightly wider due to constraints in fabrication. Just inside the RW, the plastic ST is roughly twice the diameter of a human ST (as determined by studies of human cochlear histology) but by 4 mm of insertion depth our plastic cochlea ST size is within 0.35 mm of real cochleae [30]. At the deepest level the CI was inserted (17 mm), the plastic lateral wall is 1.8 mm from the modiolus, slightly larger than the human ST width at this depth, which ranges from 1.25 to 1.6 mm (-SD to +SD) [30]. Post-operative CT imaging of CI subjects receiving this exact AB array had contacts which ranged from 0.4 to 1.75 mm to the modiolus [19]. As such our plastic cochlea is much wider than a human cochlea at the base, but differences from 4 mm of insertion to full insertion depth were usually within 0.2 mm and were thus considered negligible.

With the CI still in saline and completely outside the plastic cochlea, baseline measurements of electrode impedances were sequentially made between each contact and the ground ring (E1 to ring, E2 to ring, ... E16 to ring). Next, the implant array was inserted 1 electrode contact into the plastic cochlea, such that E1 was just inside the cochlea while E2 to E16 were still outside. Impedances were again assessed for all 16 contacts. For the rest of the implantation, each time after the CI was inserted 1 electrode contact deeper, impedances were again assessed across all contacts (Fig. 3). Thus there were 17 recording locations (saline + 16 CI insertion depths) during which all 16 electrode contacts were individually assessed, totaling 272 impedance recordings per CI insertion. Recordings at each contact were normalized by subtracting the impedance component values in saline. Nine complete CI insertions were performed.

D. Imaging to determine CI Positioning

The 3D printed cochlea was completely transparent, and allowed for microscopy at all stages of CI insertion. A Zeiss (Oberkochen, DE) Axioskop microscope with was used with accompanying Canon (Tokyo, JP) EOS digital camera and software to acquire a high

resolution photograph at each CI insertion depth (Fig. 4a). In this way, each time an impedance measurement was made for a given electrode contact, it would also be possible to determine the associated distance of that contact to the inner wall (Fig. 4b).

E. Impedance Modeling to Predict CI Positioning

The goal of the study was to determine if CI positioning of each electrode contact could be predicted solely from impedance measures. Specifically, to determine if a model could be developed which input the overall CI insertion depth, specific electrode number, and recorded impedance response to predict the distance of that contact to the modiolus (6).

$$\text{PredictedDistanceToModiolus} = f(V(t))|_{\text{EL,Depth}} = f(R_a, R_p, C_p)|_{\text{EL,Depth}} \quad (6)$$

III. Results

A. Impedance and Insertion Depth

When solely observing the voltage response waveforms of the deepest/apical electrode (E1) during several stages of CI insertion (Fig. 5, top), it is clear that there is a direct relationship between insertion depth and recorded voltage amplitude (Fig. 5, bottom). Deeper locations within the plastic cochlea have a more resistive path to ground, necessitating an increase in the required CI's voltage to allow the fixed current pulse level. Consistent with this observation, in a fully-inserted CI (Fig. 6, top), waveform amplitudes across multiple EL contacts demonstrate that deeper electrodes have higher waveform amplitudes (Fig. 6, bottom).

Trends in R_a , Z_p , Z_{tot} (total impedance), R_p , and C_p , across all contacts for 4 different insertion depths are depicted in Fig. 7. Each column represents a CI insertion depth while each row demonstrates microscopy of the CI when the data were collected (Fig. 7, top row), trends in R_a , Z_p , and Z_t at each electrode contact (Fig. 7, center row), and characterization of polarization components R_p and C_p (Fig. 7, bottom row). Column one shows the array with only a single contact inserted into the ST. At this depth, there have been no appreciable changes in any impedance measures compared to those in saline. Column 2 demonstrates 9 contacts inserted. At this depth, the total impedance Z_{tot} is driven mostly by R_a , and those electrode contacts which were visually inspected to be closer to the modiolus (black arrow) had much higher rates of impedance growths than the mid-scalar electrodes closer to the RW. With 12 electrode contacts inserted (Column 3), electrodes 4-7 were closer to the modiolus and again R_a growths (black arrow) were much higher than the linear growth pattern observed on electrodes both deeper and shallower. The final column represents a full insertion, during which all electrodes are more lateral and far from the modiolus, demonstrating a fairly linear growth in R_a as a function of depth with a small increase in Z_p . No appreciable changes in either R_p or C_p occurred during any insertion.

B. Insertion Depth and Proximity to the Modiolar Wall

The first 100 degrees of rotation of the cochlea has a wider radius of curvature than the remaining two turns running towards the apex [29]. As such, the CI array changes its medio-

lateral positioning throughout insertion. By analyzing the proximity of any contact by cochlear position (1=base, 16=apex), microscopy revealed that this particular CI array had a reproducible range of distances to the modiolus (Fig. 8). At the base of the cochlea, the distance from electrodes to the modiolus was large. As the electrode approached the basal turn of the cochlea, roughly around position 8, the array was consistently close to this inner edge. Throughout the rest of the insertion, the array remained roughly 1.5 mm away from the modiolar wall.

C. Impedance and Proximity to the Modiolar Wall

When analyzing changes in impedance (relative to saline) as a function of cochlear position, R_a shows a positive trend increasing with depth (Fig. 9, top left), R_p does not drastically change with depth (Fig. 9, center left), and C_p decreases with depth (Fig. 9, bottom left). Incorporating microscopy into the dataset, individual measurements in the left column of Fig. 9 were ranked and colored by proximity to the modiolus - where a red circle indicates a closer position to the modiolus, a blue circle indicates a far position from the modiolus, and a black circle was a midscalar position. At a given cochlear position, the range of impedances for R_a could be partially explained by the electrode proximity (Fig. 9, top right). However, trends of proximity were not so obvious with R_p (Fig. 9, center right) or C_p (Fig. 9, bottom right).

D. Impedance Model to Predict Modiolar Distance of Electrode Contacts

The trends between impedance measures and proximity which were introduced visually (Fig. 8) were fully realized in a linear regression model. The independent variables of this model are designed to be the metrics available to the surgeon at the time of surgery – a given CI insertion depth (d) and an array contact (EL), with the associated calculated impedance measures (R_a , R_p , and C_p). The dependent variable to be predicted is the distance from contact to the modiolus, in mm. The linear regression optimization algorithm introduced these variables in a stepwise fashion, creating 5 models and evaluating whether each variable significantly increased the adjusted r^2 (Table I, models [a] through [e]).

The majority of the observed variance (59%) in electrode-to-modiolus distance can be explained solely with R_a , while incorporating insertion depth and electrode number added ~9% each to the adjusted r^2 . While still significant, C_p and R_p contributed just 2.5% collectively to the explained variance. Ultimately, a Pearson correlation of 0.885 could be achieved with the Table I model [e], herein called Model 1.

E. Impedance Model to Predict Sub-millimeter Proximity to Modiolus

During the experiments performed, 267 individual electrode contacts were found to be microscopically within 1 mm of the modiolus. Using these sub-millimeter cases as inputs to the model in Model 1, the predicted distances were greater than 1 mm in nearly 77% of recordings. Such underestimation of proximity is not ideal. As such, a new model was created utilizing training data solely from electrode contacts which were closer to the modiolus (Table II). In this model, the output metric was a boolean – whether the contact was within 1 mm of the modiolus – rather than an estimated distance as was the case in Model 1.

A linear regression model introduced these same variables in a stepwise fashion but here the best model resulted with a Pearson correlation of 0.717 with Table II model [e], herein called Model 2. The adjusted r^2 of this model is slightly lower than that of Model 1, but the rate of false negatives when detecting sub-mm proximity dropped from 77% to 22.1%. In addition to better detection of sub-mm positioning, Model 2 also reduced the rate of false-positives (array contacts being flagged <1 mm despite actually being farther from the modiolus) from 43% to 25%.

IV. Discussion

Qualifying the electrode-neural interface in cochlear implant recipients is crucial for understanding the variance in speech perception outcomes. A key component of this interface is the positioning, or geometry, of the electrode array relative to cochlear anatomy. This report introduces impedance collection throughout many stages of CI insertion, and expands upon existing impedance models for approximating CI positioning *in situ*.

First the plastic ST model was used to determine the relationship between derived CI impedance measures and proximity to adjacent structures at 16 discrete steps during CI insertion. We found that the total impedance increased with both insertion depth and proximity to adjacent structures – consistent with Tan et al. and Pile et al. who demonstrate an increase in overall (R_a) impedance after stylet removal [22, 23]. In our Model 1, R_a accounted for the majority (59%) of the adjusted r^2 , while the addition of electrode number and insertion depth brought this correlation to 0.76. Polarization components R_p and C_p added a small but significant amount to this correlation.

Because of the profound interest by some implant manufacturers to place the implant either as close to the modiolus as possible to minimize current spread, or intentionally as far from the modiolus as possible to avoid modiolar insertion trauma, we also wanted to determine if impedance could detect when any given contact was grossly lying along the modiolar wall or along the lateral wall. One manufacturer, for example, produces pre-curved arrays which have a median modiolar distance of 0.4 mm *in vivo* whereas their lateral wall arrays have a median modiolar distance of 1.2 mm [19]. In fact, their perimodiolar arrays with electrode-modiolar distances greater than 1 mm in this study were considered outliers. With this cutoff in mind, Model 2 was created, wherein the metric to be optimized was a Boolean – whether any contact was within 1 mm of the modiolus – even at the expense of knowing the precise modiolar distance as in Model 1.

A comparison of Model 1 and Model 2 (Table III) demonstrates that Model 1 is better for determining precise positioning anywhere within the ST, but Model 2 is much more sensitive and specific to detecting which electrodes are very close proximity to the modiolus. As such, each model provides discrete but complimentary information regarding the position of the electrode relative to the modiolus, which may be clinically valuable in assessing the electrode-neural interface. In addition, implementation of an approach like Model 2 throughout CI insertion may also be advantageous if a surgeon wanted to receive intraoperative feedback on whether the cochlear implant is approaching adjacent tissue structures [31, 32]. The speed of the measurements being less than 1 ms per electrode, and

the ability to concurrently use the processor case as the ground electrode, make the runtime feedback a feasible goal. However, the role of impedance as a metric for trauma is not currently known, as penetration damage from the array tip, for example, could occur despite recording contacts being appropriately distanced from tissue.

Currently, the use of intraoperative impedance testing only occurs only after full CI insertion, and is used to confirm CI processor function and identify malfunctioning array contacts [15]. Electrodes which are shorting have abnormally low impedances, whereas electrodes with an open or disconnected contact have abnormally high impedances. Functioning devices are manufactured to satisfy a range of acceptable impedance tolerances, but slight differences between contacts are not uncommon. As such, a single post-insertion impedance scan is not sufficient to determine which electrodes are closest to the modiolus [25]; differences can be due to factors both within the CI or at the CI interface rather than reflecting the environment around the device. To minimize this confounder, it is absolutely paramount that the impedance of each individual electrode contact is characterized in saline before CI insertion; the trends noted in this report were not robust when analyzing impedance without first doing so. This may become even more important when utilizing this approach in living tissue, when slight discrepancies in individual electrode contact impedances can lead to drastic differences in recordings to biphasic current stimulation [33]. In the surgical setting, an approach to obtain pre-insertion impedance measures could include briefly flooding the surgical field and immersing the array with saline, while a more elegant approach may be to use the first non- artifact impedance measure from each contact throughout insertion as its own baseline.

The vast majority of CI current *in vivo* is confined to the ST, with a ST conductance roughly 100x greater than transversal current pathways towards the modiolus [20]. When current is limited to the ST, the return path to ground becomes longer and the cochlear conductive space becomes narrower as the electrode advances deeper. This may explain why total impedance increases with both insertion depth and proximity to the modiolar walls [22, 23], consistent with our results in this study where the plastic model afforded no “leaky” transverse channels whatsoever. In addition to current not returning through the modiolus, our plastic model also negates the possibility of current favoring a path out of the ST apex and through the facial canal rather than towards the RW [34]. This characteristic also likely explains why the polarization resistance and capacitance of the surface monolayer was negative compared to saline; that is to say patterns of inductance were more easily obtained down the ST than in an open body of saline. Recordings *in vivo* also necessarily introduce bone, which will increase the R_a portion of the necessary return path depending on temporal bone thickness [35]. The roles of polarization impedance, specifically R_p and C_p , played a minimal role in assessing positioning in the plastic model but may be much more integral in cadaveric or living cochlear models because of the current pathways outside the ST, comprising a variety of tissue types and interfaces between them. In fact, polarization impedance comprised nearly 2/3 of the total impedance across CI contacts within the first week after CI implantation [24].

Knowledge of array positioning has been previously used to optimize speech coding strategies during rehabilitation for individual subjects, by reprogramming and deactivating

contacts to limit spread of excitation [36]. As such, attempts to assess positioning are clinically worthwhile, particularly for those where imaging is unavailable. Future studies will thus be directed at characterizing the relationship between impedance measures and CI positioning in temporal bone and cadaveric models.

V. Conclusions

Components of electrode impedance change during CI insertion due to proximity to local sources. Building a template of CI electrode insertions with intraoperative insertion depth and impedance measures can accurately predict the positioning of the CI array during insertion in a plastic ST model. Of the metrics obtained, access Resistance (R_a) is the best at inferring positioning of electrode contacts to the wall of the cochlear model, while polarization resistance and capacitance significantly contributed to a smaller degree. Impedances normalized to saline aided in the creation of two regression models to predict 1) overall electrode positioning in terms of proximity to the modiolus and 2) predict whether contacts were grossly located along the modiolar or lateral wall.

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Fig. 1. (top) Current passes from each cochlear implant (CI) electrode contact, E1 for example, through the surrounding medium and returns at the ring ground. (center) The processor contains a current source and ADC while the array contains contacts which interface with saline. (bottom) Major contributors to impedance for a single cochlear implant contact include the bulk resistance of the medium (access resistance, R_a), and the polarization impedance of the electrode-electrolyte interface (Z_p), which can be modeled as a parallel circuit with polarization resistance (R_p) and capacitance (C_p).

Fig. 2. Stimulus, Recording, and Analysis of Impedance. (top) The CI generates a biphasic square pulse. The measured response (middle) includes an immediate jump in voltage (V_a) and a polarizing growth (Z_p). V_a is used to calculate R_a while Z_p is split into R_p and C_p (bottom) by modelling the circuit as a resistor and capacitor in parallel.

Fig. 3. Relationship between Position in Cochlea and Insertion Depth. Each row represents a depth the CI is inserted into the model cochlea, and each boxed number is an electrode contact. Each column is the position of the cochlea. E1 is the deepest electrode throughout insertion.

Fig. 4.

From a micrograph of the CI array in a plastic cochlea model (a), it is possible to assess the distance (b, solid bar) from each contact to the modiolar wall (b, dashed line).

Fig. 5. Relationship between E1 response magnitude and insertion depth during CI insertion. (top) Microscopy of mid-CI insertion, just before CI insertion (1 EL contact into the cochlea) and during insertion (8 EL contacts inserted). (bottom) The voltage required to allow the current pulse increases with E1 insertion depth. Grossly, total impedance of E1 grows with insertion depth.

Fig. 6. Relationship between response magnitude and electrode contact in a fully-inserted CI. (top) demonstrates a fully inserted CI and corresponding electrode contact numbers. (bottom) demonstrates response magnitude is greatest for the deepest electrode contacts.

Fig. 7.

Components of impedance across all CI electrode contacts at 4 stages of CI insertion normalized by saline. (Column 1) with 1 EL inserted (Row 1), R_a , Z_p , and Z_t are the same as in saline (Row 2). Row 3 demonstrates splitting of Z_t into R_p and C_p . At 9 EL inserted (Column 2), exponential growth in Z_t is dominated by R_a , and these apical electrodes are close to the modioloar wall (arrow). At 12 EL inserted (Column 3), electrode contacts 4-8 are close to the modioloar wall and demonstrate exponential growth in R_a . At full CI insertion, (Column 4), no electrodes are significantly close to the inner modiolar wall and Z_t is roughly linear.

Fig. 8. Patterns of Electrode Proximity during CI insertion. Array contacts passing the base of the cochlea (position 1) are further in proximity than those passing the basal turn (position 8). Toward the apex (position 8), the array bows out to roughly 1.5 mm.

Fig. 9.

Changes in access resistance, polarization resistance, and polarization capacitance as a function of cochlear position (left column). Incorporating microscopy data (right column), demonstrates proximity highly affects R_a at a given insertion depth. R_a , R_p , and C_p are normalized (values from saline subtracted out)

TABLE I

Models for Estimating Proximity of All Electrode Contacts to Modiolus at Any Stage of CI Insertion

	R	R ²	Adj. R ²	Std. Error of Estimate	R ²	F	Sig.
<i>[a]</i>	0.767	0.588	0.588	1.392	0.588	3282	< 1E-14
<i>[b]</i>	0.820	0.673	0.672	1.241	0.085	596.6	< 1E-14
<i>[c]</i>	0.871	0.758	0.758	1.066	0.086	814.9	< 1E-14
<i>[d]</i>	0.874	0.764	0.764	1.053	0.006	57.9	< 1E-14
<i>[e]</i>	0.885	0.783	0.783	1.010	0.019	200.8	< 1E-14

[a] Predictors: Ra,*[b]* Predictors: Ra, Insertion Depth*[c]* Predictors: Ra, Insertion Depth, EL*[d]* Predictors: Ra, Insertion Depth, EL, Cp*[e]* Predictors: Ra, Insertion Depth, EL, Cp, Rp

TABLE II
Models for Estimating Proximity of Array to Modiolus within 1 mm during CI Insertion

	R	R²	Adj. R²	Std. Error of Estimate	R²	F	Sig. F
<i>[a]</i>	0.662	0.379	0.438	0.693	0.439	782.9	< 1E-14
<i>[b]</i>	0.687	0.485	0.484	0.664	0.047	90.5	< 1E-14
<i>[c]</i>	0.703	0.495	0.493	0.658	0.009	18.0	< 1E-14
<i>[d]</i>	0.710	0.504	0.502	0.652	0.010	19.2	< 1E-14
<i>[e]</i>	0.717	0.514	0.512	0.646	0.010	20.3	< 1E-14
<i>[a]</i> Predictors: Ra							
<i>[b]</i> Predictors: Ra, EL							
<i>[c]</i> Predictors: Ra, EL, Cp							
<i>[d]</i> Predictors: Ra, EL, Cp, Rp							
<i>[e]</i> Predictors: Ra, EL, Cp, Rp, Insertion Depth							

TABLE III

Comparison of Models for Array Positioning

$\text{PredictedDistanceToModiolus} = \beta_0 + \beta_1 \cdot \text{Depth} + \beta_2 \cdot \text{EL} + \beta_3 \cdot \text{R}_a + \beta_4 \cdot \text{R}_p + \beta_5 \cdot \text{C}_p$			
	<u>Model 1</u>	<u>Model 2</u>	
Utility	Accurate positioning of any electrode	Detecting any electrode within 1 mm of modiolus	
β_0	5.054	2.310	
β_1	-0.241	-0.090	
β_2	0.170	0.134	
β_3	-0.176	-0.063	
β_4	0.236	0.074	
β_5	-275.383	-112.601	
Pearson R	0.885	0.717	
Sensitivity to < 1 mm	22.1 %	78.3 %	
Specificity to > 1mm	56.8 %	75.0 %	