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Evaluation of rigid cochlear models for measuring cochlear implant electrode position

Ahmet Cakir, MRes¹, Robert F. Labadie, MD PhD², M. Geraldine Zuniga, MD², Benoit M. Dawant, PhD¹, and Jack H. Noble, PhD¹

¹Department of Electrical Engineering and Computer Science, Vanderbilt University, Nashville, TN 37232

²Department of Otolaryngology-Head and Neck Surgery, Vanderbilt University Medical Center, Nashville, TN 37232

Abstract

Objective—To investigate the accuracy of rigid cochlear models in measuring intra-cochlear positions of cochlear implant (CI) electrodes

Patients—93 adults who had undergone CI and pre- and post-operative CT imaging.

Main Outcome Measures—Seven rigid models of cochlear anatomy were constructed using micro-CTs of cochlear specimens. Using each of the 7 models, the position of each electrode in each of the 98 ears in our dataset was measured as its depth along the length of the cochlea, its distance to the basilar membrane, and its distance to the modiolus. Cochlear duct length was also measured using each model.

Results—Standard deviation across rigid cochlear models in measures of electrode depth, distance to basilar membrane, distance to modiolus, and length of the cochlear duct at two turns were 0.68, 0.11, 0.15, and 1.54 mm. Comparing the estimated position of the electrodes with respect to the basilar membrane, i.e., deciding whether an electrode was located within the scala tympani (ST) or the scala vestibuli (SV), there was not a unanimous agreement between the models for 19% of all the electrodes. With respect to the modiolus, each electrode was classified into one of the three groups depending on its modiolar distance: Close, medium, and far. Rigid models did not unanimously agree on modiolar distance for approximately 50% of the electrodes tested.

Conclusions—Inter-model variance of rigid cochlear models exists, demonstrating that measurements made using rigid cochlear models are limited in terms of accuracy due to non-rigid inter-subject variations in cochlear anatomy.

Keywords

Cochlear implant; scalar electrode position; cochlear model

Corresponding author: Ahmet Cakir, MRes. Department of Electrical Engineering and Computer Science, 2301 Vanderbilt Pl., Box 1679 Station B, Nashville, TN 37235-1679, Telephone: 615-322-8478, Fax: 625-343-5459, ahmet.cakir@vanderbilt.edu.

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Introduction

Cochlear implants (CIs) are considered standard treatment for patients who experience sensorineural hearing loss. Each CI has an electrode array with a certain number of electrodes, between 12 and 22, depending on the type and the manufacturer. The electrode array is surgically inserted into the cochlea. Although CIs have been very successful at restoring hearing, restoration to normal auditory performance is rare, and some CI recipients may experience poor outcomes. Electrode position is one of the factors that has been shown in several studies to affect hearing outcomes. However, the intra-cochlear based imaging modalities have been the only option for intra-operative imaging but are difficult to interpret and to implement in the traditional clinical workflow. Different groups have tried to estimate the shape of the cochlea and localize intra-cochlear electrode positions using post-implantation computed-tomography (CT) imaging, which remains the imaging modality best suited for post-operative CI imaging. Aschendorff et al.¹ used rotational tomography (RT) to manually determine whether electrodes were located within the scala tympani (ST) or the scala vestibuli (SV). In a later study², this imaging approach was used to show that hearing outcomes are correlated with the scalar positioning of the electrode array. Verbist et al.³ presented another CT imaging technique to evaluate electrode scalar position. Recently, Boyer et al.⁴ presented another technique using cone-beam CT to study the scalar positioning of electrode arrays. However, manually determining intra-cochlear electrode location using images alone is arduous. Further, while these methods may be accurate enough to detect correlations of gross differences in electrode positioning with hearing outcomes with a large enough dataset, they are prone to error. This is because it is not possible in CT to directly visualize fine scale intra-cochlear structures, such as the modiolar wall (the inner wall of the cochlea) or the basilar membrane and the bone that separate the ST from the SV, due to the lack of adequate resolution of the CT.

Magnetic resonance imaging (MRI) is another modality that has great potential for imaging the cochlea. It is the only modality that can directly visualize the ST and SV⁵. However, it is not possible to use MRI for post-implantation imaging because the electrodes within the cochlea corrupt the MRI signal in the region around the cochlea. In addition, only certain types of MRI are considered safe for cochlear implant recipients. Neri et al.⁶ developed a technique in which post-operative CT images were registered to 1.5 Tesla pre-operative MRI images in order to assess electrode location within the membranous labyrinth. Even though this method allows visualization of certain intra-cochlear structures, the ST and SV were non-differentiable on MRI images. It is important to note that with current MRI technology achieving the high signal quality and CT-like resolution necessary to accurately image ST and SV requires hours of scan time and the use of high field scanners not yet approved for clinical use, making use of MRI for large scale analysis of patient cochleae not yet a viable option. However, with future improvements to MRI technology it may become possible to fuse an MRI for pre-implantation imaging of the cochlea anatomy and a CT for post-implantation localization of the CI electrodes to accurately quantify intra-cochlear electrode position.

Another approach is to create a model of intra-cochlear anatomy that can be registered to a patient CT image to define the location of the basilar membrane, modiolus, and other intra-

cochlear structures, which are not directly visible in the images. This was the approach of Skinner et al.⁷, who created a rigid model of cochlear anatomy using a high-resolution multi-section histological image of a single cochlear specimen where the fine structures of the cochlea can be visualized. Using a spatially aligned CT volume, this model could be rigidly registered to new patient CT images to estimate the position of the electrodes relative to the model-estimated position of the basilar membrane and other fine intra-cochlear structures. This model has been used in several studies to investigate the relationship between electrode position and hearing outcomes. In Skinner et al.⁷ and Finley et al.⁸, it was found that scalar location of the electrode array is correlated with hearing outcomes. Later in Holden et al.⁹, it was found that in addition to scalar location, the distance of the electrode array to the modiolus is another factor that is predictive of outcomes. Depth of insertion into the cochlea is another factor that has been studied, but results are conflicting with some studies showing better outcomes are associated with deeper^{10,11} or shallower^{8,9} insertions. Thus, even with a model-based approach, some factors such as distance to the modiolus have been little studied, and others such as depth of insertion led to conflicting findings. One potential reason for this is that using a single model that is rigidly registered to patient images does not provide an accurate enough estimation of the position of intra-cochlear anatomy. Indeed, this approach does not permit to take into account normal non-rigid inter-subject anatomical variations and thus leads to error when localizing anatomic structures. For example, consider using a square model to approximate a circle. Even with optimal rotation and translation of the model, it never perfectly matches the circle in shape. Recent studies have shown that individual human cochleae differ not only in size but also in morphology^{12,13,14,15}. Studying 73 ears, Erixon et al.¹² found large variations in cochlea dimensions, concluding that each cochlea was uniquely shaped. Avci et al.¹³ also observed large variability across different measurements of cochlea, including height, width, length, and number of cochlear turns. Non-rigid variations, from the simplest variation of overall cochlea size to more complex variations such as changing proportions of specific areas (e.g. a deeper basal turn or a more horizontally positioned round window), cannot be captured by rotating or translating one geometric model to fit another and thus cannot be accounted for by rigid models.

A cochlear model that captures non-rigid variations in anatomy and that can be automatically and accurately aligned with new patient CT images has been proposed and developed by Noble et al.¹⁶, and the authors have begun to use it to assess electrode placement and its effect on outcomes^{17,18}. Consider the analogy of matching a square model to a circle above. With non-rigid, the square model can be warped to match the shape of the circle. Similarly, non-rigid models of the cochlea have the advantage of being able to be warped to more accurately match the shape of the target anatomy. When the non-rigid cochlea model was introduced by Noble et al.¹⁶, it was shown that it more accurately estimates the position of intra-cochlear structures than a rigid model; however, the effect that the anatomy localization method, i.e., rigid vs non-rigid, has on the accuracy of electrode position measurements has not been quantified. In this study, we propose an approach to quantify the accuracy of measurements of intra-cochlear electrode positions when using rigid cochlear models. In future work, we plan to design a study to quantify measurement

accuracy with non-rigid models which will require expanding our dataset as will be discussed in the conclusions section.

Another application of interest for which selection of an accurate cochlear model is important is image-guided CI programming (IGCIP)¹⁹. After CI surgery, each CI is programmed by an audiologist to attempt to optimize hearing performance. These programming sessions include deciding which electrodes will be activated, and assigning stimulation levels as well as sound frequency bands. With the intra-cochlear position of the electrodes unknown, audiologists have little quantitative information they can use to guide programming. Thus, they have to depend on the feedback that the patients provide in order to try to optimize the settings. However, the adjustment process often converges to sub-optimal settings, and it requires several programming sessions over the course of many years. In IGCIP, the goal is to accurately identify the intra-cochlear location of the CI electrodes and to use that information to assist the audiologist in selecting better patient-customized programming settings. Previous studies have shown that this approach leads to significant improvement in hearing outcomes^{20,21}. Since IGCIP depends on accurate localization of intra-cochlear electrode position for each patient, choosing an appropriate cochlear model is crucial.

Materials and Methods

We use a dataset of pre- and post-implantation CTs of 98 CI ears from 93 patients to quantify the accuracy of measurements of intra-cochlear electrode positions when using rigid cochlear models. Our approach is to (1) fit individual rigid models of cochlear anatomy we have created from 7 different cochleae to each of the CI ears; (2) evaluate intra-cochlear electrode position in each ear relative to each of the 7 models; and (3) quantify the errors due to non-rigid variations in cochlear anatomy as the differences in resulting measurement of electrode position across the 7 different models. Our approach is described in detail in the following sections.

Dataset

Our model construction dataset consists of 7 μ CTs of cochlea specimens in which the ST, SV, and modiolus were manually segmented to create 7 different rigid cochlea models. μ CTs rather than CTs must be used because intra-cochlear structures are not visible in CT images as discussed in the introduction section. Figure 1 shows a cropped μ CT image where ST, SV and modiolus are represented with blue, yellow and red contours, respectively. Our testing dataset consists of pre- and post-implantation CTs for 98 ears of 93 CI patients. CT images were obtained with various scanners and parameters and had voxel size ranging from approximately $0.2 \times 0.2 \times 0.3 \text{ mm}^3$ to $0.4 \times 0.4 \times 0.4 \text{ mm}^3$. As a pre-processing step, we have implemented the approach of Noble et al.¹⁶ to localize the ST, SV, and modiolus in the pre-implantation CT. We then localized the electrodes in the post-implantation CT^{22,23}, and rigidly registered the two CTs to bring the two datasets into alignment. Localizing the ST, SV, and modiolus in the pre-implantation CTs facilitates an automatic approach for fitting each of the 7 rigid models to the dataset as will be described below. Each pre-localization was visually inspected in the pre-implantation CT by an experienced Otolaryngologist,

although it is not necessary for our pre-localizations to be extremely accurate for the analyses we describe below.

Model fitting

The 7 rigid models represent the ST, SV, and modiolus as surfaces composed of a set of over 3000 points distributed uniformly over the surfaces of these structures. Each of these points corresponds to a specific anatomical site that was semi-automatically determined when the active-shape model of the cochlea was created¹⁶. Thus, the rigid models were created such that point correspondences exist not only across the models but also with the pre-localizations of the ST, SV, and modiolus in the testing dataset of 98 CI cases. Using these point correspondences, we automatically rigidly registered the 7 models to each patient using well known point-based registration techniques²⁴. All of the points that make up the ST, SV, and modiolus surfaces were used in the point-based registration. This method finds the translation and the rotation that will be applied to a set of points in order to spatially align them with another set of points while minimizing the sum of squared errors between the point sets. As it is an error minimizing method, it is more accurate than manual registration done based on visual observation. Point-based registration also has the advantage that it guarantees a solution that globally minimizes the surface error, as opposed to techniques such as image-based registration, where the image intensity-based error minimizing procedure could fall into a local minima and cause alignment errors. By using our point registration approach in which the models are registered to a reference in an unbiased way, we minimize the possibility that the differences we measure in relative electrode position across models are due to model alignment errors. Figure 2 and Figure 3 show an example result of this process.

Relative Electrode Position

In order to measure the relative position of the electrodes with respect to the basilar membrane, we first use the ST and SV to define a “basilar membrane curve,” or BMC for short, which is a curve that lies along the length of the cochlea on the basilar membrane, between the ST and the SV, and located midway between the modiolus and the lateral wall of the cochlea. The BMC is shown in white on Figure 4 where the ST and the SV are shown as blue and green 3D meshes, respectively. At each point c along the BMC, a vector, m , that points from c towards the mid-modiolar axis can be found. Next, a vector s , which is perpendicular to both m and the tangent of the BMC at c can be found. The mid-modiolar axis and the m and s vectors found along the length of the BMC for one case are shown in Figure 5 in red, light blue, and yellow, respectively. Using the BMC and these vectors, we can measure the intra-cochlear location of each electrode. For each electrode, first, the closest point p along the BMC is found. Then, the depth of insertion of the electrode can be found by measuring the geodesic distance, i.e., the distance along the BMC between p and the round window (RW). The geodesic distance is calculated as the sum of Euclidian distances of each line segment that defines the BMC between RW and p . Next, the modiolar distance of the electrode is found by measuring the distance to the electrode along the vector m associated with BMC point p . Finally, the scalar location of the electrode relative to the basilar membrane is found by measuring the position of the electrode along the s vector associated with BMC point p . Because p lies on the basilar membrane and s is defined to be

normal to the basilar membrane, electrodes that are found to be in the positive s direction relative to p are located in the ST, and those that are found to be in the negative s direction are located in the SV.

Cochlear Duct Length at 2 Turns

The cochlear duct length (CDL) is another quantity of interest for CI as its measurement has been proposed as an approach for pre-operative patient-specific selection of the electrode array. Measurement of CDL using a rigid model would not be useful because it would be measured identically for each patient. However, here in we report the standard deviation of the length of the cochlear duct at a depth of two turns across rigid models as an indicator for inter-subject variations in CDL.

Results

Standard deviation across rigid models in measures of electrode location relative to the basilar membrane was 0.11 mm. We also counted the number of times the rigid models were in unanimous agreement about the scala in which each electrode was located. Out of 1834 electrodes, the models were in agreement for 1348 electrodes being in ST and 133 electrodes being in SV. The models did not unanimously agree for 353 electrodes.

Standard deviation across rigid models in measures of distance to the modiolus was 0.15 mm. Electrodes found to sit more than 0.3 mm from the BMC towards the modiolus were considered to be close to the modiolus and labelled as group *I* electrodes. Those found to sit more than 0.3 mm from the BMC towards the lateral wall of the cochlea were considered far from the modiolus and labelled as group *III* electrodes. All other electrodes were considered as sitting a medium distance to the modiolus and labelled as group *II* electrodes. We found that out of 1834 electrodes, the rigid models unanimously agreed on 326 electrodes as belonging to group *I*, 169 as belonging to group *II*, and 426 as belonging to group *III*. The models did not agree on the remaining 913 electrodes.

Average standard deviation across rigid models in measurement of electrode depth within the cochlea was found to be 0.68 mm. We found the standard deviation in CDL across rigid models to be 1.54mm.

Conclusions

The primary aim of this study was to quantify the performance of rigid models in measuring intra-cochlear electrode position. Quantifying intra-cochlear position of CI electrodes is important for studying the relationship between electrode position and hearing outcomes. It is also important for image-guided cochlear implant programming techniques that rely on measurements of electrode positions to determine patient-customized programming settings. Rigid models are commonly used for measuring intra-cochlear electrode position, but the accuracy of electrode position measurements when using a rigid model has not been widely studied. Rigid models of anatomy are limiting because they are aligned by simple rotation and translation and assume identical size and proportions of the cochlea exist between

patients. This approach is effective if the assumptions are met but fails when there are substantial inter-patient differences.

In this study, we compared measurements of intra-cochlear electrode position on 93 subjects when using 7 different rigid models. Standard deviations of electrode position measures across rigid models are relatively high, leading to potentially different findings about the scalar position of electrodes and their proximity to the modiolus depending on which rigid model is used. The rigid models disagreed on scalar position for 19% of the electrodes in our dataset and on modiolar proximity for 50% of the electrodes. The error minimizing approach we used to align the rigid models ensures that we have minimized the likelihood that the disagreements we observe in measurements of electrode position are due to misalignment of the rigid models. These results motivate the use of non-rigid models of cochlear anatomy, such as that proposed by Noble et al.¹⁶, when measuring intra-cochlear electrode position as they have been shown to be more accurate than rigid ones. It was not possible in this study to similarly analyze variability in non-rigid models because it was not possible to create multiple unique non-rigid models to compare to each other. However, in future work, we plan to collect μ CTs of a large number of implanted cochlea specimens which will facilitate more direct measurement of errors in electrode positioning using both rigid and non-rigid models.

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Figure 1. Two different views of a cropped μ CT image where scala tympani (ST), scala vestibuli (SV) and modiolus are represented with light blue, yellow and red contours, respectively

Figure 2. Rigidly model fitting result. Scala tympani (ST - red) and modiolus (blue) are the surfaces in the pre-implantation CTs, and ST (green) and modiolus (yellow) belong to one of the rigid models

Figure 3.

Segmentation of anatomical structures using 7 different rigid models are represented with 7 different colors. Scala Tympani (ST) on the left, Scala Vestibuli (SV) in the middle, and modiolus on the right

Figure 4.
The basilar membrane curve (BMC) that separates scala tympani (ST) and scala vestibuli (SV) is shown in white. ST and SV are shown as blue and green 3D meshes, respectively

Figure 5. Scala tympani (ST) is shown as the blue 3D mesh. Mid-modiolar axis, $\mathbf{m}$ and $\mathbf{s}$ vectors are in red, light blue and yellow, respectively