

Article

Effect of Dielectric Properties of Cochlea on Electrode Insertion Guidance Based on Impedance Variation

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Abstract: The cochlear neuromodulator provides substantial auditory perception to those with impaired hearing. The accurate insertion of electrodes into the cochlea is an important factor, as misplaced may lead to further damage. The impedance measurement may be used as a marker of the electrode insertion guidance. It is feasible to investigate the impact of the dielectric properties of the cochlea tissue layers on the electrode insertion guidance using sophisticated bio-computational methods that are impractical or impossible to perform in cochlear implant (CI) patients. Although previous modeling approaches of the cochlea argued that the capacitive impact of the tissue layer can be neglected using the quasi-static (QS) approximation method, it is widely accepted that tissue acts as a frequency filter. Thus, the QS method may not always be appropriate due to short-duration pulses. This study aimed to investigate the impact of the frequency-dependent dielectric properties of the cochlea tissue layers on the impedance variation by following a systematic approach. The volume conductor model of the cochlea layers was developed, the dielectric properties of each tissue layer were attained, and the cochlea neuromodulator settings were applied to obtain the results based on both QS and transient solution (TS) methods. The results based on the QS and TS methods were compared to define to what extent these parameters affect the outcome. It was suggested that the capacitive impact of the cochlea layers should be considered after a certain frequency level.

Keywords: bio-impedance; capacitive effect; cochlea implant; computational modeling; dielectric properties; quasi-static simulation; transient simulation



Citation: Salkim, E. Effect of Dielectric Properties of Cochlea on Electrode Insertion Guidance Based on Impedance Variation. *Appl. Sci.* **2024**, *14*, 10408. <https://doi.org/10.3390/app142210408>

Academic Editor: Junseop Lee

Received: 21 August 2024

Revised: 16 October 2024

Accepted: 24 October 2024

Published: 12 November 2024



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1. Introduction

The global prevalence of hearing loss is a significant public health concern, affecting over 5% of individuals worldwide, with an anticipated doubling of this number by 2050 [1]. The cochlea function generates a sense of hearing by transforming the sound waves into mechanical vibrations of the hair cells and subsequently into electrical pulses to provide hearing sensation. Dysfunction of the cochlea may result in an impaired understanding of speech which is highly invalidating for professional, social, emotional, and cognitive well-being. Cochlear implant (CI) is a partially invasive neuromodulation technique that restores hearing loss through electrical stimulation of the hearing nerve from within the cochlea, as shown in Figure 1A. The electrode array (EA) of the CI is surgically inserted into the scala tympani (ST) of the cochlea to replace the function of the damaged hair cells. The clinical results have shown that the EA insertion process during the CI surgery is crucial for patient outcomes [2]. Thus, accurate electrode placement in the ST is a key parameter for hearing [3]. Misplacement may cause damage to intracochlear structures and compromise the remaining hearing. If the electrode touches the cochlea nearby layers, further hearing loss and insertion trauma may occur. It has been shown that electrode insertion trauma was observed in one in every three implantations [4]. Since the human cochlea is surrounded by bone, it is extremely challenging to assess the position of the EA within the cochlea during the insertion. Moreover, the interaction between the EA

and the cochlear structures is invisible due to the limitations of current electrode insertion guidance technologies. Thus, the EA insertion process primarily relies on the surgeon's tactile feedback and experience [5] although there are many visual inspections of the EA guidance techniques [3]. Computed tomography (CT) imaging is one of the methods to determine EA positioning but is not routinely performed due to the radiation risks [6]. Alternatively, the EA position can be assessed in real time by electrically evoked neural responses, electric field imaging, or impedance variations [7]. The results of the clinical studies have shown that the results for evoked neural responses in assessing EA positioning are highly variable [8]. Thus, this method may not be a reliable contributor to determining EA position guidance during insertion to the cochlea. Although the major error position of the electrodes in the ST could be registered using electric field imaging, it was not utilized to predict the positions of the electrodes in the ST [9]. It has been shown that bioelectrical impedance variations can be used as a biomarker for monitoring EA position in the ST [3,10–13] due to the higher conductivity of the ST compared to other layers. The hypothesis was that the measured electrode impedance to the ground should be higher when an electrode approaches the cochlea wall compared to when the electrode is in the middle of the scala [14]. The overview of experimental CI simulation and impedance recording procedures is highlighted in Figure 2A. Conventional clinical techniques may not be feasible to determine the EA positioning due to the associated risks. Alternatively, bio-computational modeling can be used to investigate the optimal position of the EA insertion [3,15].

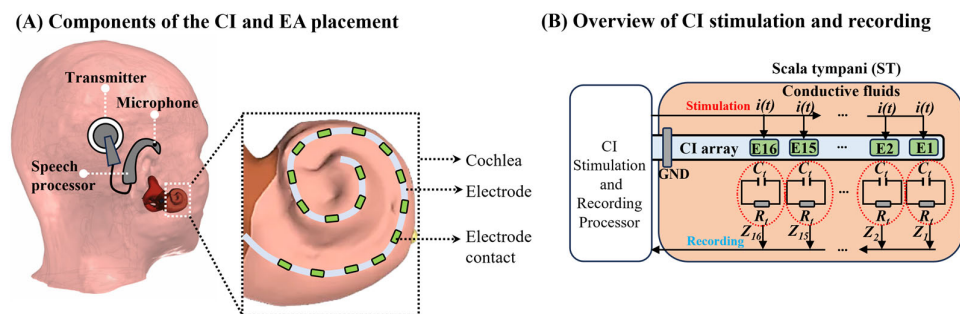
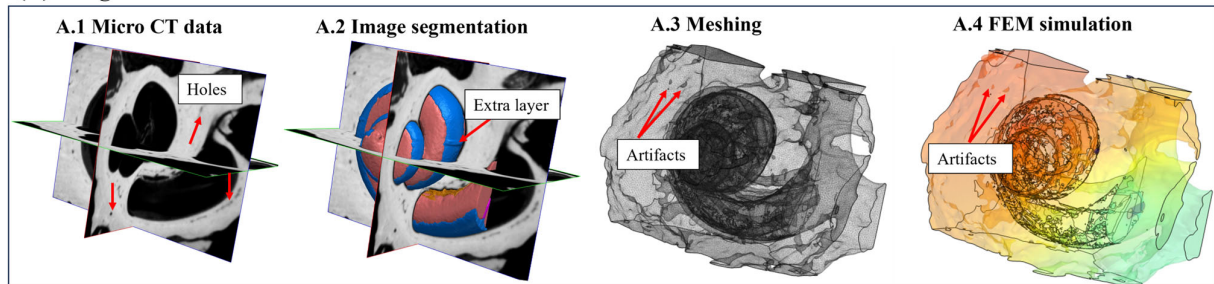


Figure 1. (A) shows the external components of the CI and insertion of the electrode array (EA) into the scala tympani. The EA and its contacts are highlighted. (B) shows an overview of experimental CI stimulation and recording procedures. $i(t)$ shows the applied current to the electrode contacts. The associated tissue capacitance and resistance are represented with C_t and R_t , respectively. GND: ground, Z : impedance, and E1: electrode contact 1.

Although the bio-computational modeling studies of CI have been carried out under the quasi-static solution (QS) by ignoring capacitive, inductive, and wave propagation effects [3,15,16], it has been indicated that the electrical impedance of body tissues depends on the alternating current signal frequency [17–19]. Also, the results based on the experiments showed that the capacitive component of impedance can affect the shape and amplitude of the voltage waveform for different applications [18]. In particular, the QS may not always be appropriate due to short-duration pulses ($\sim 100 \mu\text{s}$) [11–13]. Since it has been discussed that the dielectric parameters of the cochlea layers are frequency-dependent [20,21], more accurate results may be obtained using a transient solution (TS) by considering the capacitive effect of the time-dependent dielectric parameters. Thus, there is a need for a systematic study to show what extent these parameters affect the outcome. Using animal and human cadaveric models presents significant challenges as it is not possible to obtain a comprehensive systematic testing of dielectric parameters and their effects on the outcome. Alternatively, three-dimensional (3D) bio-computational models of the inner ear can be utilized to assist in investigating the impedance variation that influences CI outcomes [3,22–25]. Over the past decade, substantial research progress has been made to advance inner ear computational modeling. The first cochlea model was

developed based on a fixed grid. The grid size, and therefore the resolution of the model, was mainly limited due to computation. Semiautomatic and automatic segmentation image methods were generated to construct a cochlea volume conductor using finite element (FE) or boundary element methods [15]. The finite element modeling (FEM) allows the intricacies of the inner ear's mechanics to be reduced to simpler phenomena that can be verified with clinical results [26]. The impedance variation can be readily investigated using available commercial FEM software packages, such as COMSOL Multiphysics V5.2 (COMSOL, Ltd., Cambridge, UK), which includes many matured numerical toolsets for solving bioelectric phenomena problems [17,27–29].

(A) Image-based cochlea model



(B) Geometrical cochlea model

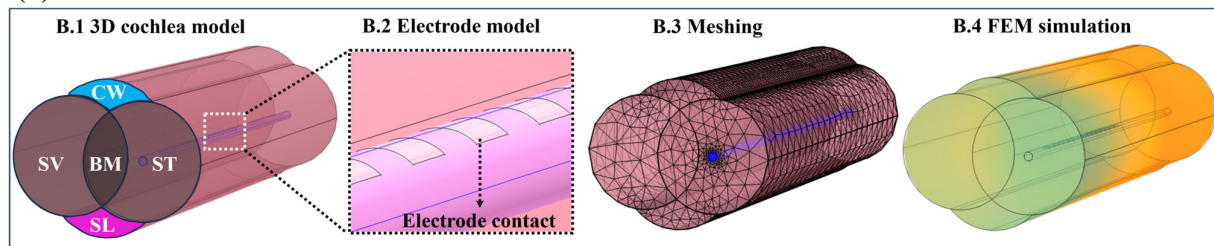


Figure 2. (A) shows the anatomical-based cochlea model. The holes are shown in the micro-CT data in (A.1). These holes cause extra layers during image processing as shown in (A.2). These small layers are observed in both meshing (A.3) and simulation (A.4) processes. (B) shows a multi-layered geometrical-based cochlea model without any extra layers and holes. The model is generated (B.1), the electrode model is merged with the cochlea model (B.2), and after applying the meshing process (B.3), the FEM results are calculated (B.4). The CI layers are scala tympani (ST), scala vestibular (SV), basilar membrane layer (BM), spiral ligament (SL), and cochlea wall (CW).

This paper develops an accurate human cochlea model that can be used to investigate the impact of the dielectric properties of tissue layers for the cochlea electrode insertion guidance based on the impedance variation as shown in Figure 2A. The models are implemented in FEM involving a volume conductor model representing various anatomical structures and the electrodes by their respective dielectric features and appropriate boundary conditions as shown in Figure 2. The impact of the dielectric properties on the outcome was defined by the following:

1. Impedance variation based on the EAs that are placed at the center and border of the ST by considering both capacitive and static (resistive) effects;
2. Variation in the electrical potential waveform to observe the capacitive effect;
3. Parametrizing the ST conductivity by keeping constant other layers' dielectric properties to examine the impact of this layer on the results;
4. Showing smooth electrical potential and current density variations over the outer wall of the ST based on QS and TS.

The paper is organized as follows. Section 2 presents the methods used to develop the model and the procedures to analyze the impact of the dielectric properties using different solution methods; Section 3 gives the results of these methods. The discussion and conclusion are given in Sections 4 and 5, respectively.

2. Methods

For all the subsequent simulations and operations, a computer with an Intel Core i7-8550U CPU at 1.99 GHz with 16 GB RAM (Intel, London, UK) was used.

2.1. Cochlea Model Consideration

It has been shown that the development of realistic computational models using image data is not efficient due to high computational costs [3], as shown in Table 1. The segmentation and meshing processes may lead to extra computation costs due to the low resolution of the image data [30]. Figure 2A shows the generation, segmentation, discretization, and simulation processes of the image-based cochlea model. It was shown that obtaining the results from this model was computationally expensive due to extra small layers, and noncontinuous parts in the image data as highlighted in Figure 2A. Thus, it has been shown that the computational cost can be significantly reduced with marginal added error by simplifying the model in certain applications [3,30,31]. This is relatively uncomplicated to set up modern software (e.g., COMSOL) if the relevant geometric parameters are available. These two models were compared based on the impedance variation for the full electrode array insertion and the recorded error was about 2% as shown in Table 1 [3]. The geometrical model was generated to investigate the goal of this study as the required steps are shown in Figure 2(B.1–B.4).

Table 1. Comparison of computational parameters of anatomical and geometrical cochlea models.

Parameters	Geometrical Model	Anatomical Model
Discretization time	<1 min	≈40 min
Number of elements	0.2 M	2.33 M
Computation time	<1 min	7 h 16 min
Mesh quality	0.65	0.6
Difference in results	≈2%	≈2%

The geometrical model was generated based on the statistical variation of the anatomical layers of the cochlea. The model was constructed based on the ST, scala vestibular (SV), basilar membrane layer (BM), spiral ligament (SL), and cochlea wall (CW) using smooth shapes in COMSOL. A relatively large bony layer was generated to complete the inner ear computational model. Spiral ligaments and nerves are distributed throughout the tunnel spiral in the modiolus called Rosenthal's canal. They were considered one layer as these layers have similar dielectric properties. The basilar membrane and the osseous spiral lamina layers were combined and modeled as one layer due to the discontinuity of the osseous spiral lamina. The thin membranes between the scala vestibuli and the vestibule were not modeled as their trajectory cannot be identified in the image processing methods.

2.2. Computational Model Validation

Since the target feature is impedance variation based on various electrode contacts, the impedance difference was calculated for each electrode contact for both anatomical and geometrical models based on a fully EA mid-scalar position. Although the impedance measurement was slightly higher using the geometrical model compared to the anatomical model, the impedance of electrodes at different insertion depths showed an approximately linear increasing change with electrode depth for both models [3]. It was shown that the impedance difference between the two models for different electrodes varies was

$\approx 2\%$. This may provide sufficient accuracy with far less computation cost by using the geometrical model.

2.3. Electrode Array (EA) Modeling

The electrode array (EA) was modeled using a commercially available neuromodulator setting [Advanced Bionics HiFocusTM Slim] electrode (Hannover, Germany)] with 16 platinum electrodes. These electrodes are supported by flexible silicone and are designed to face the inner cochlea wall as shown in Figure 2(B.2). Each electrode contact was designed using the actual dimensions of the SlimJ electrode system to obtain accurate results.

Two EA models were designed to analyze the impact of the dielectric properties of the anatomical layer on the outcome as shown in Figure 3. Initially, an EA model was designed to be placed at the center of the ST. The modeling steps in Figure 2B were followed and the results for this EA model were calculated for the defined frequency. Then, the EA was moved towards to the border of the ST to investigate the results between the center and the border of the ST based on impedance and electrical potential variations.

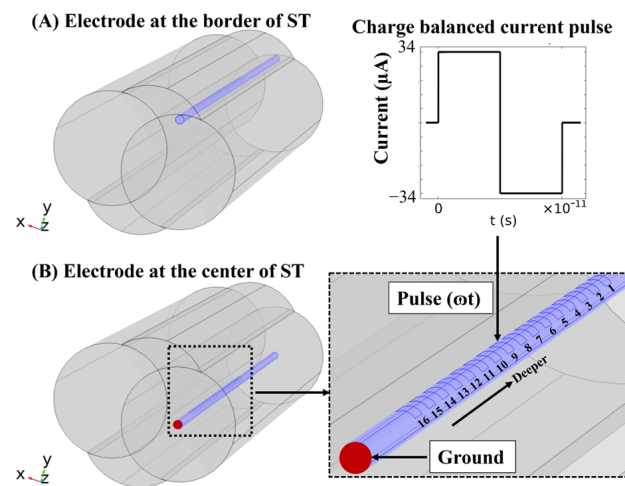


Figure 3. Electrode array (EA) modeling. (A) shows a border-placed EA model. (B) shows a center-placed EA model. The center-placed EA model is highlighted. The simulation waveform and current levels are highlighted. The simulation waveform was applied to each electrode contact. *Pulse (ωt)* shows a current pulse. Electrode contacts are numbered from bottom (16) to top (1).

2.4. Simulation Pulse Design

The stimulation waveform was designed based on the biphasic charge-balanced current pulse, as shown in Figure 3. To generate an anodic current pulse in COMSOL, the rectangular waveform was selected, and upper and low limits were defined based on the desired frequency. Then, the same procedures were applied to generate the cathodic current pulse. The time-based current pulse (e.g., *Pulse (ωt)*) was generated by combining the anodic and cathodic current pulses based on the defined expression. The stimulus current pulse based on different frequency ranges was applied on the electrode contacts for each simulation. The *boundary probe* was selected for each electrode contact to measure the surface electrical potential variation. The bottom of the electrode model was considered as a reference electrode by applying a voltage of 0 V as highlighted in Figure 3B. These procedures were repeated for both EA models and for each simulation.

2.5. Tissue Dielectric Properties

The response of the biological tissue to an applied electrical signal is defined by its dielectric properties, such as relative permittivity (ϵ_r) and conductivity (σ). These parameters were described for a wide frequency spectrum (from 10 Hz to 20 GHz) for biological tissues [21,32]. However, many computational studies analyzed the electrode array insertion based on the standard dielectric properties of the cochlea layers as shown in

Table 2. The tissue dielectric characteristics were defined based on current flows from a stimulating electrode to a return electrode for a specified frequency level [15]. Most of these values were obtained from guinea pig or gerbil animal models and some of these values were assumed to be representative of the human case. However, the dielectric properties of the bone varied between species. Thus, the value was updated to be like the human case [15].

Table 2. Standard conductivity values for cochlea layers.

Anatomical Layers	Conductivity (S/m)
Scalas	1.43
Basilar membrane	0.0125
Spiral ligament	1.67
Cochlea wall	0.3
Bony layer	0.0156

There are no available data that show the dielectric parameters of the cochlea layers based on the various frequency levels. Since it is always desired to place the EA at the center of the ST, extraction of the results based on the electrical parameters of this layer would greatly help in understanding current spread and electrode insertion guidance during cochlear electric stimulation. It has been shown that dielectric properties of cerebrospinal fluid (CSF) can be used for the ST as both layers are filled with the fluid (perilymph) [20,25]. The SV and ST were considered as one structure with the same dielectric properties of the CSF layer based on different frequency ranges as shown in Table 3. The remaining layers of dielectric properties in Table 3 were obtained from [21].

Table 3. Approximate values for the electrical properties of anatomical layers of the cochlea based on the frequency spectrum (10 Hz–100 MHz).

Anatomical Layers	Conductivity (σ)	Relative Permittivity (ϵ_r)
Scalas	1.789–2.46	16.6–68.4
Basilar membrane	2.02×10^{-1} – 9.78×10^{-1}	2.57×10^7 – 5.48×10^1
Spiral ligament	2.51×10^{-1} – 7.6×10^{-1}	1.99×10^7 – 4.56×10^1
Cochlea wall	4.08×10^{-2} – 3.12×10^{-1}	3×10^4 –6.2
Bony layer	2.04×10^{-2} – 1.56×10^{-1}	5.22×10^2 – 1.24×10^1

Also, the electrical parameters of the other layers of the cochlea were kept constant based on the defined frequency, and the conductivity of the ST was parametrized to examine the impact of the electrical parameters of the ST on the outcome as detailed in the result section. It was noted that frequency-dependent permittivity was found by averaging over frequency. It is noted that all tissue layers are assumed to be homogeneity and isotropic as there are no data in the literature on the anisotropy of the layer conductivities of the cochlea.

2.6. Finite Element Discretization and Simulation

After attaining the dielectric properties of each anatomical layer and neuromodulator settings, the cochlea model with an electrode array embedded and was meshed using tetrahedral finite elements for numerical solutions of partial differential equations in COM-SOL. The ST and the tissue layers near the ST were meshed finely to obtain sufficient accuracy while reducing excessive computation time. Mesh settings for the electrode model were adaptively adjusted to different sizes and growth rates in different models. Since the outermost layer (bone) was far from the region of interest, the discretization element size was selected to be larger (i.e., known as the normal tetrahedral setting) than the cochlea

layers. The meshing sizes were modified manually if the warnings were received from the software. The total number of the meshing elements varied between 0.1 million (M) to 0.2 M, depending on the model. Any surface that does not have a set value for potential or current has its current density set to zero to confine the current within the model.

To calculate the electric potential variation based on the QS, the cochlea was assumed to be electrically passive, with each anatomical structure having an isotropic specific conductivity, as shown in Table 2. The electrical potential was governed by Poisson's equation Equation (1). To measure the capacitive impact of the tissue on the induced electrical potential, the TS of Maxwell equation Equation (2) was used.

$$\nabla \cdot (\sigma \nabla V) = 0 \quad (1)$$

$$\nabla \cdot \left(\sigma \nabla V - \epsilon_0 \epsilon_r \nabla \frac{\partial V}{\partial t} \right) = 0 \quad (2)$$

where σ is the tissue conductivity, V is the measured electrical potential within the domains, ϵ_0 is the permittivity of free space, and ϵ_r is relative permittivity. ∇ is the Hamiltonian operator. After defining the solution method, an isotropic and homogeneous dielectric property of each tissue based on different frequency samples was assigned to each sub-domain of each cochlea model, as shown in Table 3. The wave propagation and inductive effects were assumed to be negligible. The charge-balanced on a fully differential rectangular waveform based on the desired frequency sample was applied to each electrode to obtain the results, as shown in Figure 4.

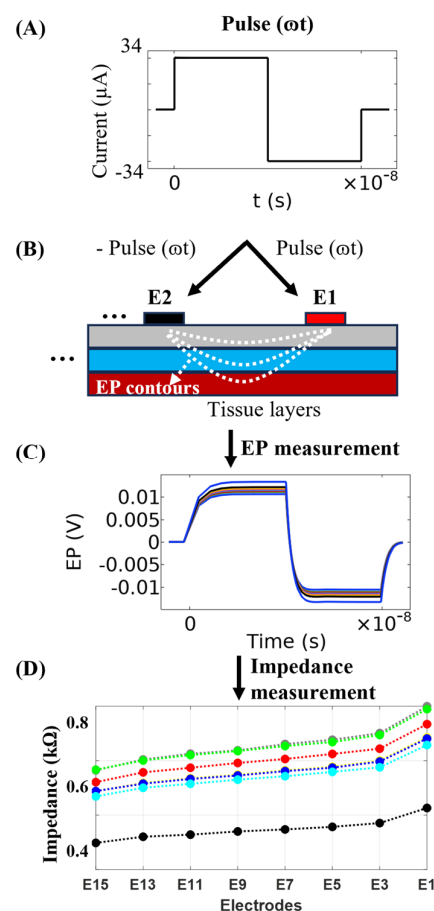


Figure 4. Simulation of the anatomical layers by inserting current via electrode contact. The applied current pulse is shown in (A). The electrical potential (EP) variation distributions within the anatomical layers are shown in (B). The recorded EP based on the specified frequency level is shown in (C). The extracted bio-impedance variation for each electrode is shown in (D).

It is aimed to simulate the whole nerve fibers in the cochlea to restore hearing loss. To achieve this, the induced current density and electrical potential are important parameters to define the smooth variation across the border of the ST. Thus, it was also aimed to measure the current density and electrical potential variations within the models based on different simulation frequency spectrums to examine the impact of the short and long simulation waveforms on the outcome. The current density was calculated using Equation (3).

$$J = \sigma \cdot E + \frac{\Delta D}{\Delta t} \quad (3)$$

where J is measured current density, E shows induced electrical field, and D indicates displacement.

To compare the impact of the longest (10 Hz) and shortest (100 MHz) current pulse waveforms on the electrical potential distribution at the border of the ST, the range was set to ± 1 mV for a fair comparison. The same settings of the streamline were defined in COMSOL to visualize the electrical potential distributions at the border of the ST.

2.7. Electrical Impedance Measurement Procedures

Figure 4 shows the general procedures of impedance measurements based on frequency-dependent various dielectric properties of the cochlea layers. The CI neuromodulator in this study was designed based on Advanced Bionics SlimJ EA and the electrical parameters of this CI EA were used, as shown in Figure 4A. The positive polarity of the current pulse was applied to the odd-numbered electrode contacts (E1, E3, ..., E15), and the negative polarity of the pulse was applied to the even-numbered ones (E2, E4, ..., E16), as shown in Figure 4B. The model was discretized and simulated based on each frequency by using associated tissue layers' dielectric parameters. As shown in Figure 4C, the electrical potential variation based on the simulation pulse was calculated using Equation (1) or (2) in the *Electrical Settings* section in COMSOL, depending on the solution method (e.g., QS, TS). The electrical potential variation for each frequency sample and current pulse were recorded in COMSOL for each EA contact and exported to MATLAB 2023 (a) to do post-processing. Since the applied current was constant, the impedance variation for each simulation was measured by following Equation (4) as the results were shown in Figure 4D. Thus, the impedance (Z) was measured by dividing the electrical potential ($V(t)$) measured by the current signal applied ($I(t)$).

$$Z = V(t) / I(t) \quad (4)$$

The current density was calculated using Equation (3) in the *Electrical Settings* and the data was obtained in the *post-processing section* in COMSOL by choosing current density variation in the result section in COMSOL.

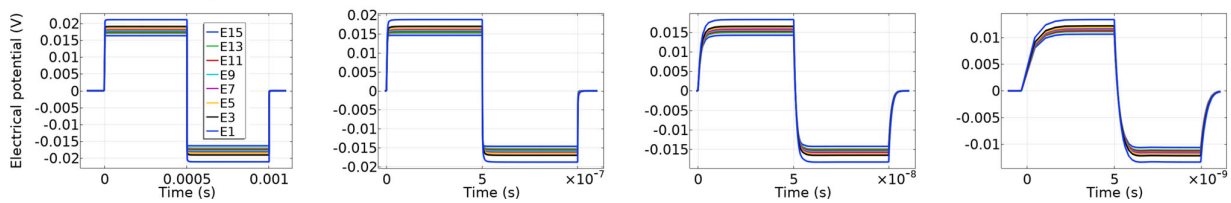
3. Results

The aim of this study was to analyze the impact of the dielectric properties on the measured impedance variation when the EA was inserted in the ST based on different EA insertion scenarios. The capacitive effect is measured based on electrical potential waveform using different frequency spectrums. The impedance variation is extracted from the electrical potential variation based on various dielectric parameters detailed as follows.

3.1. Capacitive Effect

The capacitive impact of the tissue layers based on different simulation times is defined by the induced electrical potential waveforms variation over time as selected samples (e.g., 1 kHz, 1 MHz, 10 MHz, 100 MHz) are shown in Figure 5. The electrical potential waveform versus simulation times is recorded for the EA model that was placed at the border (shown in Figure 5A) and the center (shown in Figure 5B) of the ST. The electrical potential waveform is obtained from all odd-numbered electrode contacts (E15, E13, ..., E1) for each frequency.

(A) EP variation for EA placed at the border of ST



(B) EP variation for EA placed at the center of ST

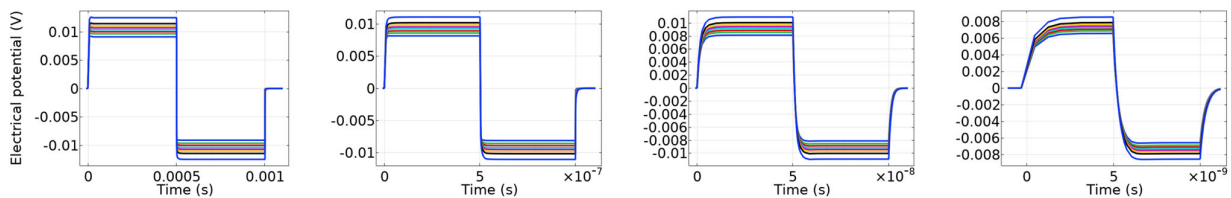


Figure 5. The capacitive effect of the tissue using various simulation frequency steps for odd-numbered electrode contacts. (A) shows the results for the EA model that is placed at the border of the ST. (B) shows the results for the EA model that is placed at the center of the ST.

The results show the effects of tissue dielectric properties in the model are clearly visible in the electrical potential waveform variation beyond a frequency sample (e.g., 10 MHz). There is no significant difference in the electrical potential waveform based on the frequency range from 10 Hz to 1 MHz. Although the electrical potential range is different for the center and border-placed EA models, the results for both models show the same capacitive effects. The slope of the electrical potential waveforms decreases when the simulation frequency increases. Therefore, considering the dielectric properties of the tissues, the calculated electrical potential waveforms show a smooth transition from the anodic to the cathodic phase after a certain frequency range. When the results for both the border and center-placed EA models are considered, the electrical potential variation is reduced using dielectric properties of the tissue layers at the higher frequency ranges (short simulation waveforms). This effect is present on all EA models and electrode contacts the edges of and it is more pronounced at the 100 MHz simulation compared to the other frequency levels.

Also, it is noted that the deeper electrode contacts have higher electrical potential waveform amplitudes (the deepest electrode contact is E1 as highlighted in blue in Figure 5).

3.2. Bioimpedance Measurement

Since it was aimed to derive a map of the impedance variation using dielectric properties of the cochlea tissue layers at various frequency values, the average impedance variation was extracted using (4). The impedance variation based on the frequency spectrum for the odd-numbered EA contacts is shown in Figure 6. The results for the border-placed and center-placed EA models are shown in Figure 6A,B, respectively.

It is clearly shown that there is a fluctuation in impedance variations for all EA contacts based on different frequency values for both EA models. In all cases, there is an inversely proportional relationship between impedance variation and simulation frequency for both EA models. When the simulation frequency increases, the impedance variation shows the opposite trend. For example, the impedance variation is significantly reduced when the dielectric properties of the cochlea tissue layers at 100 MHz are used. This is valid for both EA models. For example, the recorded impedance value is about 1.3 k Ω for 10 Hz simulation frequency, whereas this is about 0.8 k Ω for 100 MHz simulation frequency for the E1 contact, as shown in Figure 6A. However, the impedance results are relatively higher for the border-placed EA model compared to the center-placed one. Also, there is a direct relationship between insertion depth and recorded bio-impedance amplitude. The impedance variation is relatively higher at the deeper locations of the cochlea for both

EA models. The results for the deepest electrode contact (E1) are about 1.1 k Ω and this value is about 0.85 k Ω for the E15 based on 1 MHz simulation frequency, as shown in Figure 6A. As shown in Figure 6B, the same trend is observed with slightly lower variation. The impedance variations are 0.65 k Ω and 0.47 k Ω for the 1 MHz simulation frequency. This correlation is valid for all simulation frequency samples.

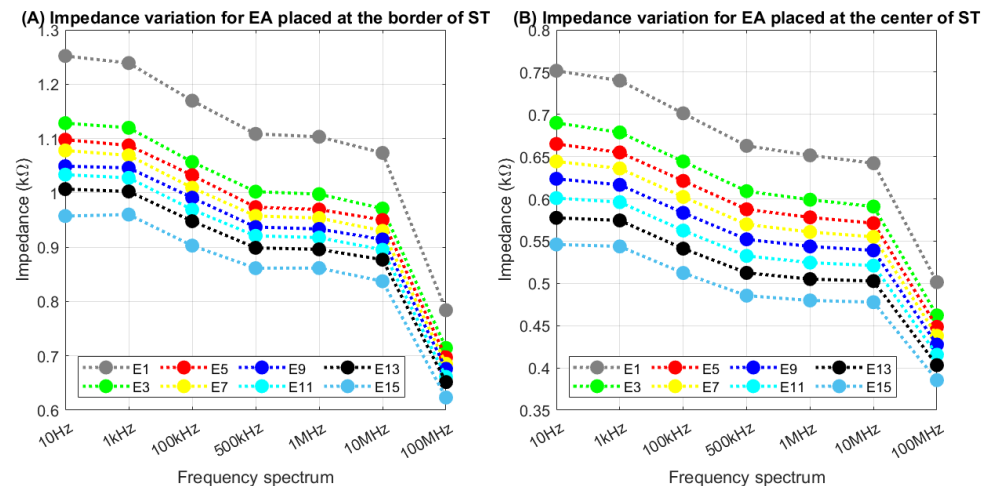


Figure 6. The impedance variation based on the frequency spectrum for the odd-numbered EA contacts. (A) shows the results for the EA model that is placed at the border of the ST. (B) shows the results for the EA model that is placed at the center of the ST.

Overall, the impedance increased with insertion depth and proximity to the inner wall of the cochlea. However, the lower impedance variation is recorded when the dielectric properties are used at the higher simulation frequency values for both EA models.

Since the most important anatomical layer is the ST for the electrode insertion, the second step of the study was to parametrize the conductivity of the ST to identify the impact of these dielectric parameters on the outcome by comparing impedance results for the QS and TS methods based on various dielectric properties of the cochlea layers. The impedance results based on various conductivity (σ) values of the ST are recorded using different simulation frequencies for TS and the obtained results are compared with QS, as shown in Figure 7, to examine the capacitive effect on the outcome. This variation is recorded based on the border-placed EA model. The trends between impedance measures and simulation types are similar. When the conductivity of the ST increased, the lower impedance was measured for both simulation scenarios. The highest impedance value is recorded at the lowest conductivity, and this is valid for all simulations. To highlight, the recorded impedance variation is about 2.1 k Ω and 0.75 k Ω for the lowest and highest conductivity values, respectively (based on 10 MHz simulation frequency). Although the amplitude of the impedance slightly shows different trends after a certain simulation frequency level (e.g., 500 kHz), the results for the QS and TS methods are slightly identical for the frequency levels from 10 Hz to 100 kHz, as shown in all subplots in Figure 7. The maximum impedance is measured based on the deepest electrode contact (E1) for all simulation scenarios. In addition to some fluctuations in the recorded impedance variations for the simulation frequency spectrum, the results are proportional to the electrode insertion depth for all parametrized conductivity values for both QS and TS methods. When the results for $\sigma = 3$ are compared for the QS and TS (based on 100 MHz) methods for E1, the result is significantly reduced.

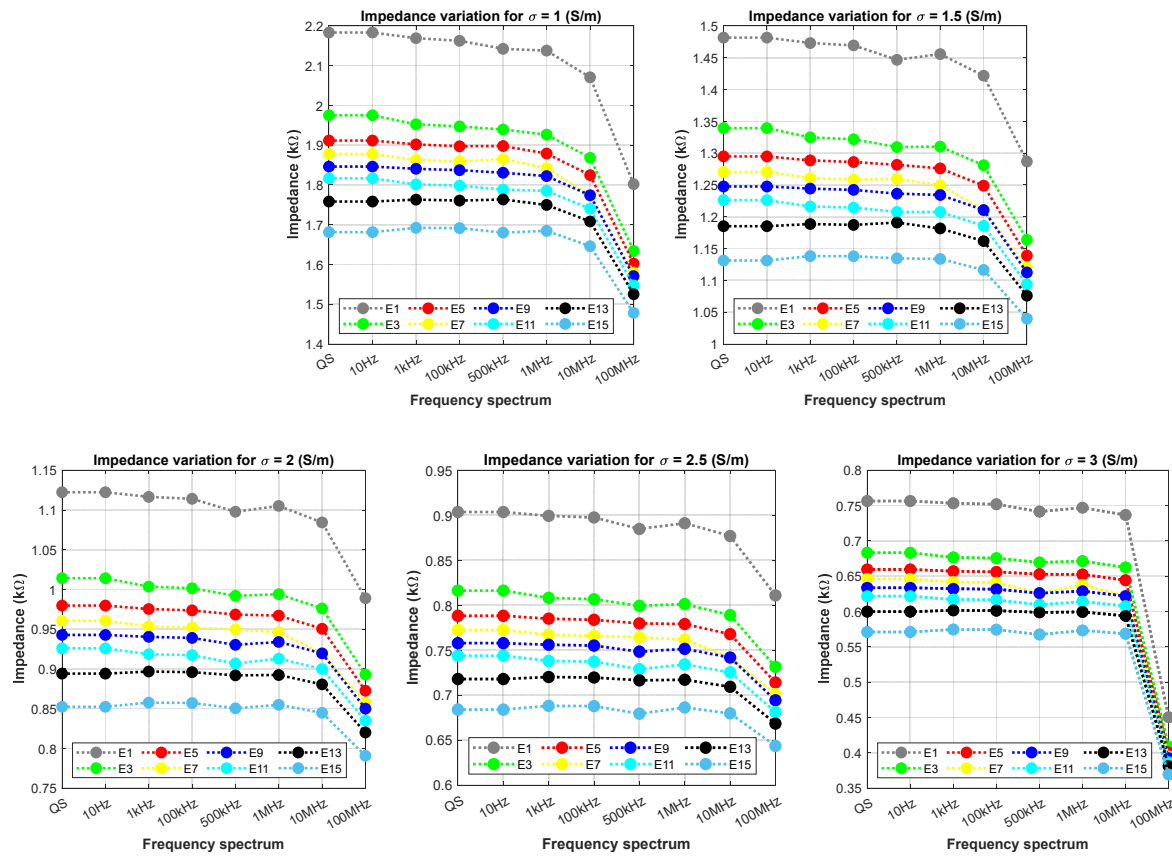


Figure 7. The impedance variation for the quasi-static and transient simulation based on various conductivities of the ST. The results for the quasi-static and transient simulations based on different frequency values are recorded for the odd-numbered electrode contacts (from E1 to E15).

3.3. Current Density and Electrical Potential Distributions

The electrical current density based on various simulation frequency spectrums using EA models is shown in Figure 8.

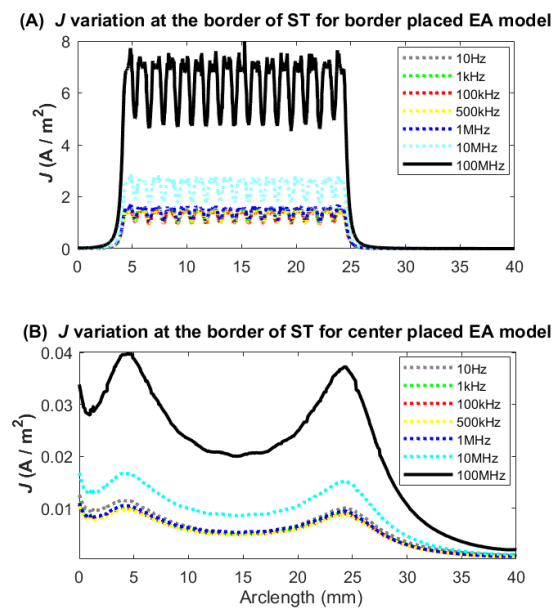


Figure 8. The electrical current density based on various simulation frequency samples for (A) border and (B) centered placed EA models.

The current density variation versus defined arclength at the border and at the center of the ST based on various frequency samples is shown in Figure 8A,B, respectively. There is a significant variation in current density based on EA models. The induced current density is higher when the border EA model is used compared to the center-placed EA model. The shorter simulation waveforms resulted in higher induced current density at the border of the ST. In particular, the current density for the 100 MHz simulation is relatively higher compared to the remaining for both EA models. This is more pronounced for the border-placed EA model.

The electrical potential distribution based on the QS and TS methods on the outer wall of the ST are shown in Figure 9A,B, respectively. The results are recorded for the TS methods based on the selected samples.

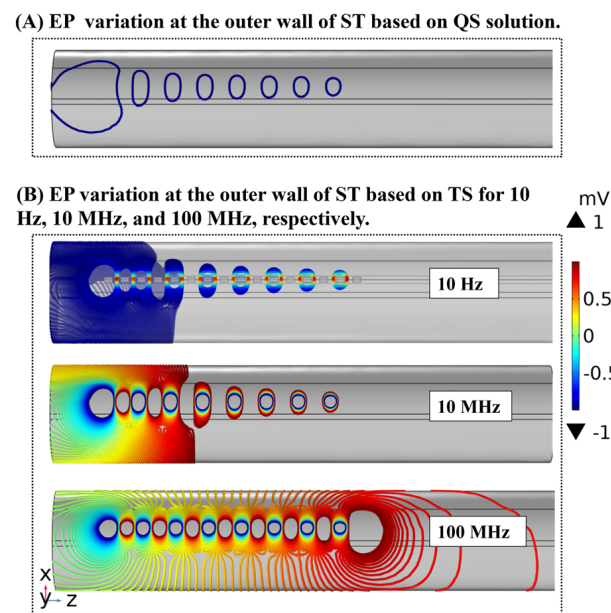


Figure 9. The electrical potential (EP) distributions at the border of the ST based on (A) 10 Hz (longest), and (B) 100 MHz (shortest) simulation frequency samples.

The electrical potential variation across different electrodes was recorded in both magnitude and contour lines at the outer wall of the ST for both QS and TS. Although the electrical potential variation has limited spread over the outer wall of the ST using the QS method, this is a relatively spread wide area of the outer wall. It is shown that the electric potential is finely distributed with relatively higher values over the outer wall of the ST when higher frequency values are used based on the TS method. Also, the higher induced electrical potential variation is measured in the vicinity of the electrode and the far region of the electrode when the higher frequency-based simulation is used.

4. Discussion

The development of neuromodulator therapy systems for neurological disorders has accelerated with improved technologies and an expanding understanding of the effect of electrical stimulation on the target tissue layer. The CI neuromodulator provides substantial auditory perception to those with severe or profoundly impaired hearing. However, visual inspection of the progress of electrode insertion is limited and mainly relies on the surgeon's tactile skills. It has been shown that the impedance variation can be used marker of the EA insertion guidance [3,9,11]. The available clinical measurement presently provides very limited information, and this may be due to variations in anatomical layers and dielectric properties [3,13,15,26]. It is not feasible to investigate these parameters using experimental methods due to complexity. Computational methods can be used to analyze the impact of such parameters on the electrode insertion guidance based on impedance

measurement. Most of the bioelectric models of the cochlea only considered the standard dielectric properties of the tissue layers [3,13,15]. However, it has been shown that the electrical impedance of body tissues depends on the applied current pulse frequency, thus, more accurate results may be obtained considering the capacitive effect of these layers [17–19]. Multi-layered cochlea model was developed to examine the impact of the various dielectric properties of the cochlea layers on the impedance measurement using simulation frequency samples. The results for the QS and TS methods were compared to analyze if the capacitive effect can be neglected. The results were simulated and recorded based on center and border EA models.

The results for the capacitive effect were examined based on electrical potential variation versus simulation frequency samples (e.g., 1 kHz, 1 MHz, 10 MHz, 100 MHz) using frequency-dependent electrical properties of the cochlea layers. The capacitive effect was clearly visible on the electrical potential waveforms after a certain frequency range for both EA models as shown in Figure 5. When the long current stimulus was used (at 500 μ s), the electrical potential was increased immediately and continued to grow until the anodic phase of the stimulus was completed. However, the waveform transitions were slower (rounded) due to attenuation of high frequencies. Since the magnitude of the impedance was increased when the EA was placed in the vicinity of the ST, the peak amplitudes were also proportionally changed for all frequency spectrum samples. In particular, the charging/discharging behavior was clearly observed at the start and end of the pulse due to tissue capacitance for the 100 MHz sample.

Figure 6 shows the impedance variation was proportional with both EA insertion depth and vicinity to the ST inner wall which is in agreement with both clinical and modeling studies [3,11]. However, when the simulation frequency was increased, the magnitude of the recorded impedance variation was reduced for all samples. This behavior can be explained by viewing the tissue as a filter. In other words, the measured impedance variation is inversely proportional to the tissue dielectric properties (e.g., conductivity) at higher frequency spectrums based on Equation (2) [33].

Since it is desired to place the EA at the center of the ST layer for the optimal simulation of cochlear implementation, the parametrization dielectric properties of this layer were investigated based on impedance variation. The results for both QS and TS methods were nearly identical apart from some fluctuations. Only a 20% difference in impedance magnitude was observed for the highest frequency sample compared to the remaining. The measured impedance variation dropped with frequency for all conductivity values based on the highest frequency sample, as shown in Figure 7 for all subplots. The reason there was no significant difference in impedance variation may be associated with the fluid in the ST as this layer is only filled with perilymph fluid. It has been shown that the impedance differences obtained due to complex and diverse internal tissues and anatomical compositions of the human body [34]. This was also proved by Figure 8.

The modeling results suggested that the outcome for the QS method and TS method were identical for simulation frequency samples from 10 Hz to 10 MHz. Thus, the electrode impedance insertion guidance can be used for this defined frequency range by neglecting the capacitive effect without inducing any errors in the results.

The induced electrical potential variation at the border of the ST (outer wall) significantly increased for the highest frequency sample, as shown in Figure 9. However, there was no distinguishable difference in the results for the QS and TS methods based on 10 Hz and 10 MHz samples. This may be proved that the electrical current penetrated the cochlea wall after a certain frequency value (e.g., 10 MHz). This may be used as a threshold frequency level to stimulate all nerve fibers in the vicinity of the ST for better hearing outcomes.

The electrical impedance of the biological tissue depends on the tissue composition, tissue anatomy, and frequency of the applied signal [21]. Thus, the current penetration and conduction paths significantly change with frequency. Since the applied current penetrates easily towards the inner structure of the cochlea layers based on the TS method (due to

higher frequency levels) compared to the QS method, the electrical potential may induce these layers. Thus, relatively lower impedance values were recorded at the higher frequency due to the higher frequency signal for the TS method, as shown in Figure 5. The applied current was more penetrated towards layers in the vicinity at a higher frequency and induced smooth electrical potential variations compared to the lower frequency simulation based on the TS and QS methods, as shown in Figure 9. The results for both QS and TS methods were identical until 10 MHz. This may be related to the dielectric properties of the cochlea wall as the permittivity of this layer is relatively higher compared to the ST. The current penetrates at higher frequency levels. These effects need to be considered when configuring stimulation protocols in CI patients.

This study has several limitations in terms of exploring the effect of the dielectric properties of the cochlea on the impedance variation. Electrode-tissue interface impedance has not been incorporated in simulation models. It has been shown that although changes in the impedance of the electrode-tissue interface directly affect the distribution of the electric field, the voltage distribution in the surrounding tissue remains unchanged [35,36]. However, the inclusion of nonlinear electrode properties in CI models may provide a more complete picture of electrode-tissue interface stability during long-term stimulation and can help with optimizing electrode design for high electrochemical stability. Second, since the dielectric properties of the ST layer are not available for all frequency levels based on the existing study, it was assumed the ST has similar dielectric properties to the CSF as both layers are composed of fluids. Also, small, discontinuous layers in the cochlea were not included in the cochlea FEM models. This may cause deviations in the electrical potential waveforms but may not affect the maximum measured impedance. Furthermore, the simplified geometrical model was used instead of the anatomical cochlea model to reduce computation cost which may lead to trivial error in the results. However, we believe the overall conclusions of the paper would not change.

Overall, although it has been discussed that the QS approximation can be used for various applications by ignoring the capacitive effect for the cochlea simulation [37], the results of this study showed that the capacitive effect for the cochlea simulation should be considered after a certain frequency threshold (e.g., 10 MHz) to obtain more accurate results and improve the credibility of the simulations. Thus, the findings of this study may guide the clinicians for accurate electrode insertion to the ST when the EA was inserted based on various simulation frequency ranges. Therefore, the outcome can be used to design for configuring stimulation protocols in CI patients. Since different cochlear characteristics, such as variations in tissue composition and fluid dynamics, may affect how capacitive impacts manifest in individual patients, these findings highlight the need for more personalized cochlear implant programming to account for individual anatomical differences, especially at higher frequencies. Also, clinicians may need to adjust the frequency of the simulation electrode insertion based on the cochlear structure and patient-specific anatomy to optimize stimulation of both low and high-frequency regions. The findings suggested that careful consideration must be given to prevent damaging the delicate cochlear structures, especially if capacitive effects significantly alter signal propagation at a certain frequency level.

5. Conclusions

Computational modeling can be employed to evaluate and quantify the specific contributions of the various aspects of the cochlea electrode insertion. In the current study, the impact of the various dielectric properties of the cochlea tissue layers on the electrical potential waveforms and measured impedance variation were investigated using a multi-layered 3D volume conductor model of the human cochlea. The computationally efficient geometrical model of the cochlea was used to reduce computation costs with trivial errors in the results. The electrical potential and impedance variation were recorded based on the QS and TS methods by attaining tissue dielectric features based on defined frequency samples to evaluate if the capacity effect of the tissue can be ignored or to what extent it can be neglected. The results showed that the capacitive impact of the tissue layers on the

electrical potential variation was distinguishable after a 10 MHz sample, but the maximum recorded impedance was not changed. However, there was a significant difference in both electrical potential waveform (capacitive effect) and extracted impedance variation for all EA models based on the 100 MHz frequency sample. Thus, the results suggested that the capacitive impact of the cochlea tissue layers should be considered after a certain simulation frequency.

Funding: This work was supported in part by the University College London (UCL) and in part by the UCL Library.

Data Availability Statement: Data is contained within the article.

Acknowledgments: The author would like to thank University College London (UCL) and Open Access Team for supporting the publication fee of this article.

Conflicts of Interest: The author declares no conflicts of interest.

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