Original Article

The Effect of Reference Electrode Position in Cochlear Implants

Angel Ramos-Miguel, Angel Ramos-Macías, Jaime Viera Artiles, María Teresa Perez Zaballos

Departamento Ciencias Quirúrgicas, Universidad de Las Palmas of Gran Canaria, Gran Canaria, Spain (AR Miguel, MTPZ)
Departamento de Sistemas Inteligentes, Universidad de Las Palmas of Gran Canaria, Gran Canaria, Spain (AR Macías)
University Hospital Insular De Gran Canaria, Gran Canaria, Spain (JVA)

OBJECTIVE: This study aims to evaluate the increase in power of cochlear implants (CI) as the reference electrode (RE) position changes. Patients in whom it is necessary to use the full power of the device to achieve the desired stimulation levels, this strategy will ensure that stimulation capability is maximized.

MATERIALS and METHODS: The variability in RE placement in the temporal bone has a measurable effect on the electrical current, impedance, and power consumption, and if the electrode position has a functional effect on the stimulation intensity. The following three approaches were used: 1) classical circuit analysis, 2) 2-dimensional numerical simulations, and 3) real temporal bone measurements using a purpose-made CI.

RESULTS: The three approaches demonstrate a significant decrease in the current intensity and electrical resistance for distances that are closer to the intra-cochlear electrode. The results also demonstrate that to maintain a constant current, shorter distances require 33% less power.

CONCLUSION: Reference electrode position during surgery can make significant differences in CI power consumption and threshold intensity, which allows a more powerful stimulation in complicated patients (i.e., those with otosclerosis). This study presents an attractive perspective to surgeons, as it shows a way to decrease consumption that might result in a longer battery life or more power to be devoted to coding strategy performance.

KEYWORDS: Cochlear implant impedance, cochlear implant, power consumption, reference electrode, threshold intensity

INTRODUCTION

The reference electrode (RE) position in cochlear implants (CI) has been surprisingly understudied. There are no articles investigating a correlation between battery life and position of RE. Most centers do not even use this electrode, with just the body of the implant being used. From a physical point of view, RE position has a bigger influence regarding power consumption than intracochlear electrode position.

Despite using different electrode designs and coding strategies across CI companies, the variability in auditory abilities of recipients appears to be more similar across devices than for the same device across individuals. This suggests that at an individual level, significant recipient-dependent factors limit the overall auditory ability. Aspects, such as ossifications, are known to have a significant effect. Unfortunately, to improve patient performance in such cases split electrodes have to be used, and they require exquisite optimization of parameters to obtain good results [1]. Goehring et al. [2] discovered that 12.4% of implantations intraoperatively demonstrate high impedances. In 8.2% of cases, impedances postoperatively remain high.

There is extensive literature wherein many variables concerning intracochlear electrodes are evaluated. In contrast, very few studies have explored the effects of position, number, or shape of RE, without which CI would not properly function [3]. A field that has been explored in more depth is the comparison between mono- and bipolar stimulation on the still controversial hypothesis that more restricted current fields can provide better speech perception. However, there are problems with bipolar stimulation: 1) Bipolar electrodes require higher currents for supra-threshold stimulation because of a current shunt from electrode to electrode and 2) high potentials and high current densities form around each of the two electrodes and supra-threshold excitation is possible near both the source and sink electrodes [4-9]. Therefore, power consumption is still a major drawback in this stimulation mode, thereby making monopolar the preferred choice of most audiologists. Therefore, it is important to evaluate whether RE position in monopolar stimulation can decrease power consumption and improve electrical field sharpness.

In monopolar stimulation, the current is injected through a contact in the scala tympani and returned to a far-field electrode contact. Stickney [10] studied the current distribution in multiple electrode contacts along the scala tympani using this mode. The result demonstrated that electrical potential field patterns caused by monopolar stimulation are broad in nature and that potential relatively and slowly drop off as one moves away from the stimulating contact. High impedances in the tissues can cause losses in the electrical impulse quality, thereby requiring high power for the patient to efficiently detect this signal. It can even lead to the deactivation of some electrodes because their power consumption is so high that they hinder the global stimulation strategy. When impedance is too high, particularly at high current levels, then it is possible that the implant voltage will not be sufficient to generate the programmed electric current level. In this case, an electrode is said to be "out of compliance." This leads to the situation where perceptual loudness does not increase with increases in the current level. To increase loudness, the width of stimulation pulses can be increased. In some cases, impedance is too high that this approach is not feasible because of too large pulse duration required. Hence, high impedance results in a decrease in battery life but also in a wider spread of excitation because of the large current intensity required.

However, decreasing impedance is feasible and can solve the issue of battery consumption. How this might be done depends on the source of impedance. It could be achieved by reducing the space between RE and signal electrode. RE can be the extracochlear electrode of some types of implants, which can be placed in different parts of the mastoid, but also the body of the implant, used as ground electrode as well. By reducing the amount of tissue between the ground and intracochlear electrodes, impedance could be lowered, and the pulse intensity required could be obtained with less power [11, 12].

In CI surgery, not much attention is given to the specific position of RE. The surgical technique historically suggests positioning RE as far as possible from the electronic part of the implant to avoid interferences during stimulation. This part is currently isolated from this type of interference by Faraday cages. Therefore, RE can be currently positioned closer to the stimulation regions [13, 14].

This study aims to examine the variability in extracochlear electrode position as a contributing factor to power consumption in CI. The hypothesis is that closer the RE is placed from the stimulation region, lesser the power consumption will be required to stimulate auditory neurons. This simple change can be of great advantage regarding the battery life for patients with ossifications of the cochlea. Because they require high currents to perceive electrical stimulation, a closer distance between the intra- and extracochlear electrodes may enable the generation of larger current intensities without the need to upturn the pulse duration to increase auditory sensation.

MATERIALS and METHODS
Ethics committee approval was not required for this study because it did not involve human subjects.

Classical Circuit Analysis
The first method comprised a theoretical characterization of the physical system using a simplified electrical model.

In this approach, three electrical formulas were used. The first is Ohm's Law, which was postulated by Geor Simon Ohm in 1826. It states that a current I through a conductor between two points is directly proportional to the potential difference V across the two points, with resistance R being the constant of proportionality between them [15]:

$$V = I \cdot R \qquad (1)$$

In this case, V is the potential difference produced by CI between the signal and RE, R is the impedance caused by the tissue between the electrodes, and I is the electrical current passing through it. With respect to I, it becomes clear that to raise this value, V needs to be increased or R needs to be lowered. Thus, the only way to do this without supplying extra energy is by reducing the resistance between RE and intracochlear electrode.

The second equation is the electrical power (P) formula (eq. 2). It represents the rate of energy that is converted from the electrical energy of the moving charges to some other form, e.g., heat, mechanical energy, or energy stored in electric or magnetic fields. Power is measured in Watts (W) and is used as a measure of the efficiency of a device. It is directly proportional to the square of the electrical current and resistance of a circuit [15].

$$P = V \cdot I = R \cdot I^2 \qquad (2)$$

Thus, for a fixed I, the higher the R, the more the power required to be supplied by the circuit to keep I constant. Electrical hearing threshold occurs when the current I delivered to the nerve fibers reaches a minimum value. This value can be taken as the fixed value of I. Then, it becomes clear that the further away RE is from the active one, i.e., the higher the resistance between them, the more power will be required to reach that threshold.

The last formula used is the mathematical definition of the electrical resistivity of a material, which is the resistance opposed by a material to an electrical current flow across it. This formula gives the characteristic R of each biological tissue between the RE and intra-cochlear electrode.

$$R = \rho \frac{l}{S} \qquad (3)$$

Where l is the length of the conductor, S is its cross-section area, and ρ is the electrical resistivity coefficient of the material, which is measured in Ohms per meter (Ω·m). In this study, the values of ρ for endolymph (ρ=0.75 Ω·m), bone (ρ=20 Ω·m), and muscle (ρ=3.5 Ω·m) were obtained from the database that was created by IT'IS Foundation. (IT'IS Database Low-Frequency Conductivity)

A simple circuit analog of the real-life system was designed to study the power required to keep an electrical current fixed between two electrodes, while the bone thickness varied from 50 to 100 mm in increments of 10 mm. The circuit design is illustrated in Figure 1. It comprises the RE and intracochlear electrode; a voltage supply of 5 V and three resistors, which are equivalent to a combination of 1 mm of endolymph; 3 mm of muscle; and 50–100 mm of the bone.
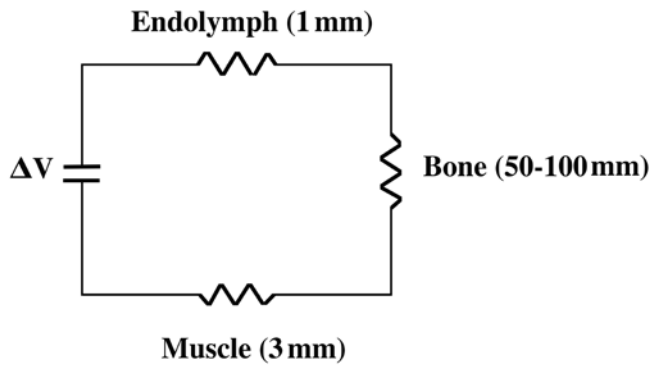
Figure 1. Circuit analog of the interelectrode region. The different tissues located in this space act as three resistors in series, and thus, can be grouped as a single one that is the sum of all its components

2-D Numerical Simulations

The second technique used was numerical simulation. The advantage of using this approach is that the physical phenomenon can be recreated in a completely controlled environment where all the variables and geometries can be defined, measured, and modified to match the real-life system.

The simulation software used was Mathematica (Wolfram Research Inc.; Mathematica, Version 10.0, 2014 Champaign, IL, USA). A 2-dimensional finite element simulation of electromagnetic phenomena for low-frequency currents was build. This computational software is widely used in science and computer engineering because it enables fast modeling thanks to the libraries that it contains.

A useful approach for calculating electric potentials (voltage) is to relate that potential to the charge density. Because electric charge is the source of electric field, the electric field at any point in space can be mathematically related to the charges present. For multiple point charges, a vector sum of each of their electrical fields is required. For a continuous distribution of charge, vector sums are better handled through calculus. This is done through the divergence relationship [16]:

$$\nabla \cdot \vec{E} = -\frac{\rho}{\varepsilon_0} \qquad (4)$$

Where E is theelectric field vector, ε_0 is the permittivity, and ρ is the charge density.

Because the electric field is also the derivative of the potential, we obtain Poisson's equation:

$$\nabla \cdot \vec{E} = \nabla \cdot \nabla V = -\frac{\rho}{\varepsilon_0} \qquad (5)$$

In a charge-free region of space or when the frequency is low, the governing equation for the electromagnetic field is the static equation, which is Laplace's equation. This is a differential equation with a smooth solution, without singularities. It is used to define the potential energy field caused by a given charge distribution:

$$\nabla \cdot \nabla V = 0 \qquad (6)$$

The model's task will then be to determine the solution to this equation for a biological model, which involves various materials and geometries. This equation gives the second derivative of V; however, because it is V itself that is needed, integration is required. However, while integrating, constants appear that cannot be numerically determined unless the boundary or initial conditions of the system are known. These are the values at the borders of the system's space or the initial value of V before it starts moving through space and time [16].

Two types of conditions were used: Dirichlet and null Neumann. The first specifies the conditions that the electrical potential requires to take along the boundary. The second sets the value of its derivative along this boundary. Dirichlet conditions of 0 and 5 V were imposed on the RE and intracochlear electrode, respectively. The null Neumann condition was applied to the edges that limit the geometry of the system [17].

Another important step in designing the computer model is the geometry, which is illustrated in Figure 2. The model represents three electrode positions. They all share the same elements; however, the distance between the electrodes decreases by a factor of 2 between one case and the next.

As in the previous approach, each element was assigned the same electrical resistivity. Moreover, air was used in this case, with $\rho=1.3\times10^{12}$ $\Omega \cdot$m [18].

Temporal Bone Measurements

A device was designed to stimulate and monitor electrical impulses that were delivered to a temporal bone while measuring impedance, current, and voltage.

The purpose-built implant comprises one intracochlear and six extracochlear electrodes. Because only the effects of relative position between the RE and intracochlear electrode are of interest in this study, CI was simplified and comprised a single intracochlear electrode. In contrast, to assess extracochlear electrode currents along different regions at the same time, six of them where located along the temporal bone. They all have spherical ends with a mean diameter of 0.9 mm. All extracochlear electrodes can be set to be RE, one at a time, thus providing six different measurement positions. The remaining electrodes act as probes throughout a given stimulation trial.

The system comprises a microcontroller (Arduino UNO; Somerville, MA, USA) that is connected to a computer via USB. Figure 3 shows a real image of the device. It was programmed to transmit electrical signals, sample each of the probes, and store data. It also calculates impedances, currents, and power that are generated during stimulation. The board is connected to a breadboard with the electrodes and probes to be inserted into the temporal bone.

The circuit is based on a number of voltage dividers. The schematics are depicted in Figure 4. The two resistors in series form a voltage divider circuit. One end of the resistor pair is fed 5 V, while the other end is connected to the ground. Five volts that the Arduino provides are divided between the two resistors depending on their resistance. The resistor, which holds the greater resistance, gets more of the
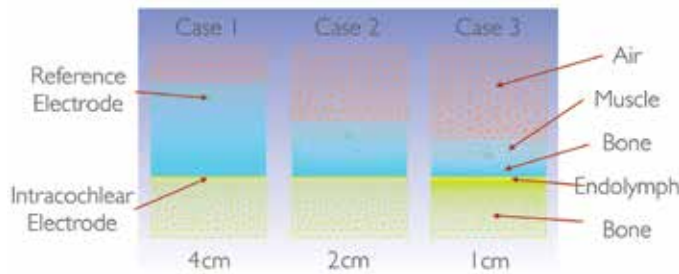
Figure 2. Geometries used in the numerical model. The interelectrode bone layer thickness was modified to study its effect on impedance, and thus, electrical consumption
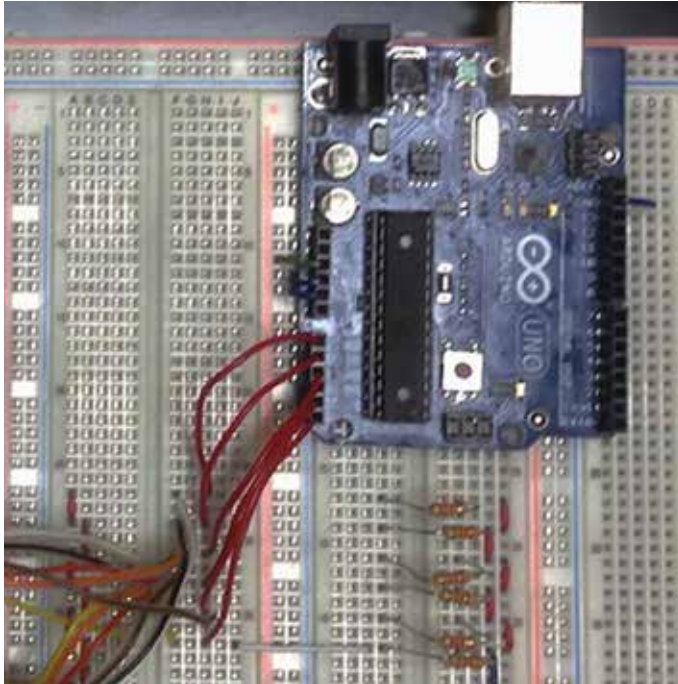

Figure 3. In-house implant. The microcontroller board is in charge of delivering the stimulus and sampling the probes during stimulation trials. The intra- and extracochlear electrodes are visible in the left hand corner

Table 1. Electrical resistance and current intensity as a function of bone thickness

Bone thickness (cm)	Impedance (kOhm)	Current (mA)
5	2.53	1.98
6	3.03	1.65
7	3.53	1.42
8	4.03	1.24
9	4.53	1.10
10	5.03	0.99

voltage, according to Ohm's law formula. The voltage that falls across a component is directly proportional to the amount of resistance it contains [15]:

$$R_{body} = \frac{R_{reference} - V_{in}}{V_{out}} - R_{reference} \qquad (7)$$

Where R_{body} is the impedance of the average of the tissues of the region of interest and is proportional to the RE impedance, $R_{reference}$; in-

Table 2. Voltage and power values as a function of bone thickness for fixed electrical current

Bone thickness (cm)	Impedance (kOhm)	Voltage (V)	Relative power (P)
5	2.53	2.53	100%
6	3.03	3.03	120%
7	3.53	3.53	140%
8	4.03	4.03	159%
9	4.53	4.53	179%
10	5.03	5.03	199%

put voltage, V_{in}; and output voltage, V_{out}. Using this principle, a model can be set up to determine the resistance on the basis of the voltage division. With this data, all electrical characteristics of the circuit can be defined, i.e., impedances, currents, and voltages.

The software to control the implant was purposely designed by the research group.

The electrode implantation was done on a temporal bone with the cochlea exposed to have access to a wider range of measurement points. The intracochlear electrode was inserted at 6 mm from the cochleostomy and then fixed. To facilitate electrical conduction, a steady uniform flow of saline solution was applied over the entire temporal bone throughout the test.

Measurements were taken at three different points across the temporal bone: the mastoid region, standard position, and root of the zygomatic process of the temporal bone. The specific locations are indicated with black squares in Figure 5. The implant processor returned and average for each point based on 2000 measurements, to minimize the effect of noise on the signals.

The statistical analysis was performed using IBM SPSS Statistics for Macintosh, V. 21.0. (IBM Corporation; New York, USA).

RESULTS

Classical Circuit Analysis
An increase in bone thickness from 5 to 10 cm, with the cross-section maintained constant at 2 cm^2, causes R in the circuit model to linearly increase. Because the resistors that represent the endolymph, muscle, and bone are in series, they can be associated as a single resistance that will be called "body" for the rest of the analysis. The increase in R that is associated with bone thickness can be found in Table 1, with increments of 0.5 kΩ/cm.

Thus, for a fixed potential of 5 V, the current through the circuit will linearly decrease as the bone thickness increases. Eq. (2) has been used to generate the theoretical values of current as a function of thickness in Table 1, which are inversely proportional.

Following Eq. (1), V must be increased to maintain a constant current of 1 mA through the body as the bone thickness increases.

In the case of power, using Eq. (3), it is observed that as the bone thickness increases, so should P increase to maintain a constant cur-

rent (Table 2). Here the power values have been normalized with re-spect to the closest point (5 cm) to illustrate how for a bone thickness of 10 cm, the power has to be doubled to obtain the same current.

2D Numerical Model

The finite element method was applied to each of the generated geometries to solve the electrostatic problem, which is defined by the potential between the RE and intracochlear electrode. After performing the simulation, the electric field lines were evaluated. Figure 6
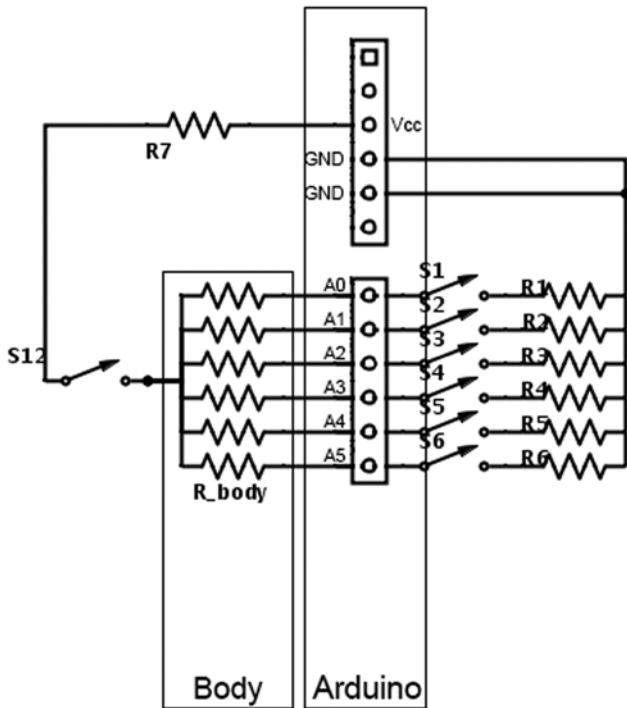
Figure 4. Schematics of the implant circuit. The reference/probe electrodes were attached to the analog ports of the microcontroller board. Switchers S1–S6 define which probe to select as RE. S12 disconnects or reconnects the intracochlear electrode. Resistors R1–R7 all have a value of 1.5 kΩ
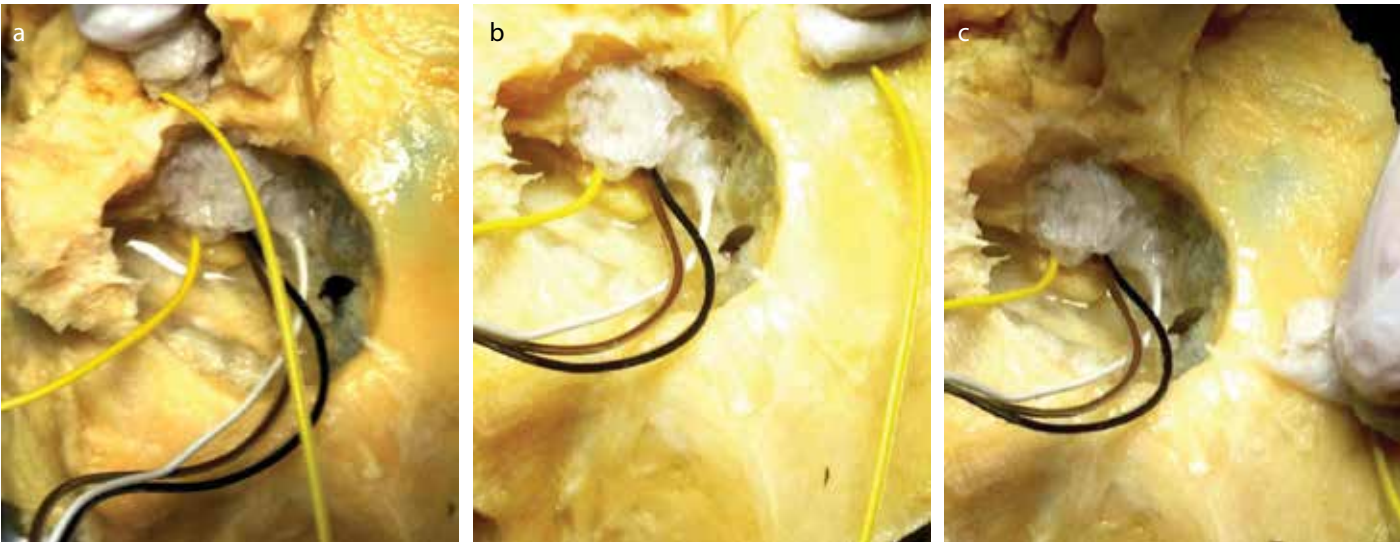
illustrates the results. It can be observed how as the electrodes get closer together, the field lines are closer to the intracochlear electrode, which means that the current generated becomes higher.

The next measurement taken was the behavior of the current at the Y axis, which is defined as the distance perpendicular to the longitudinal axis of the electrode circumference. The zero point was established in the intracochlear electrode, and the maximum point was 6-mm away from it. Figure 7 shows how in the case of the shortest distance (bone thickness, 1 cm), the value of the current is always higher than in the case of maximum distance measured (bone thickness, 4 cm). The maximum difference is observed in the region near the electrode, which is the target zone for an effective stimulation of the auditory nerve.

Finally, the evolution of the flow at a distance of 2 mm from the electrode at the X axis was studied. This was defined as the longitudinal axis of the electrode circumference. As observed in Figure 8, moving along the X axis also shows how the current generated is higher in the cases where the interelectrode distance is shortest.

However, the current density j generated for a fixed potential increases as the electrodes approach. Table 3 shows the normalized current density for each case. It can be observed how in the case where the electrodes are closest together, the current density is 33% higher than in the case where they are the furthest apart.

Table 3. Current density as a function of bone thickness. The variables show an inverse relationship that is consistent with current findings from the previous method used in this study

Case	Current Density
A	100.00%
B	115.16%
C	133.01%

Case A: the furthest away from the cochlea; Case B: mid-distance from the cochlea; and Case C: the closest to the cochlea

Figure 5 a-c. Real images showing the three RE positions studied: (a) at the root of the zygomatic process of the temporal bone; (b) in the standard temporal bone position; and (c) far away from the cochlear in the mastoid region
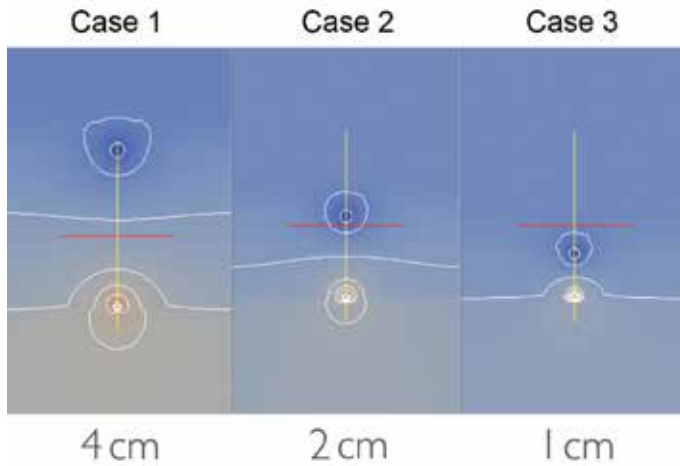
Figure 6. Electric field lines as a function of bone thickness. As RE is placed closer to the intracochlear one, the field lines show higher intensity closer to the region where the auditory fibers are located. The cross represents the origin of the coordinate system (0.0), and the white lines represent electric field contours of equal value
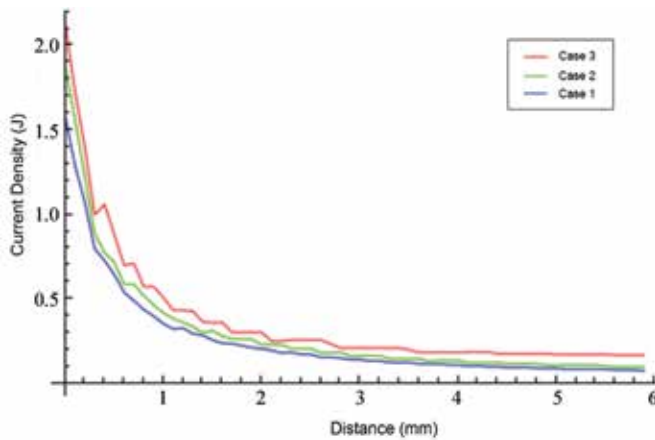

Figure 7. Current as a function of interelectrode distance at the Y axis (perpendicular to the electrode longitudinal circular axis). Each line represents a bone thickness (Case 1 being the one with the highest value and Case 3 with the lowest). All of them follow exponentially decaying profiles, although the values are higher for the case where bone thickness is lowest (Case 3)
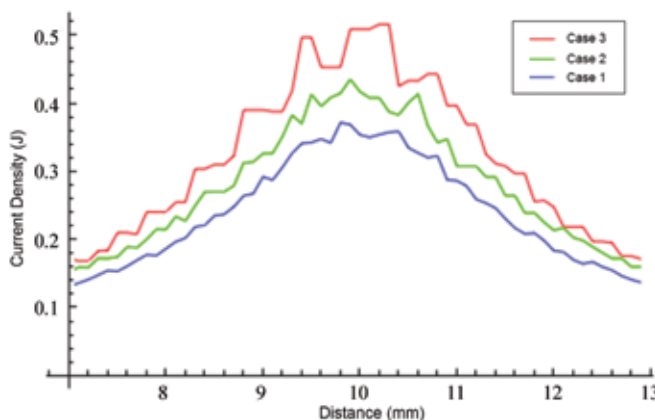

Figure 8. Current as a function of interelectrode distance at the X axis (along the electrode longitudinal circular axis). Each line represents the same bone thickness as in the previous figure. The three cases follow a similar profile, although the lowest thickness one is the one showing the highest current values

Table 4. Resistance, voltage, current, and power consumption of each RE position

Case	Impedance (kOhm)	Voltage (V)	Current (mA)	Relative Power (P)
A	45.88	4.38	0.0955	61.22%
B	30.80	4.20	0.1364	83.87%
C	24.14	4.05	0.1682	100.00%

Case A: the furthest away from the cochlea; Case B: mid-distance from the cochlea; and Case C: the closest to the cochlea

Temporal Bone Measurements

Impedances, currents, and potentials that were produced on the temporal bone in the three positions described in the previous section were measured (Table 4). It can be observed that as the RE approaches the intracochlear one, the generated current increases. The power when the electrodes are closest (C) is 38.78% greater than in cases when they are farthest apart (A). This causes a current that is 176.12% stronger with a lower potential difference.

DISCUSSION

The RE position defines many aspects of the electrical behavior of CI once implanted. As shown in the Results section, current intensity, impedance, and power that are necessary to stimulate the auditory nerve are all dependent on it. In this study, the effect of distance between the RE and intracochlear electrode on these parameters has been evaluated. To this end, three independent and complementary methods were used. First, a theoretical model based on the analysis of classical circuits was designed. Then, a numerical model was built using the finite element analysis. Finally, measurements were taken from a temporal bone.

The resulting data were used to validate the consistency of the three models because they all demonstrated the same physical behavior. These results were particularly good because one can then use them as complementary techniques to provide a wide range of direct and indirect measurements.

Studies on other areas have demonstrated the influence that a correctly positioned RE can have on the efficiency of electrical stimulation. Butson and McIntyre [19] demonstrated that the inclusion of either electrode or tissue capacitance reduces the volume of tissue activated in a manner dependent on the capacitance magnitude and the stimulation parameters (amplitude and pulse width) in deep brain stimulation.

On the basis of the available measurements and modeling results, the scala tympani is usually considered to be a preferential current pathway that acts like a leaky transmission line. Therefore, most studies assumed that current thresholds exponentially decay along the length of the scala tympani. Briaire and Frijns [20] suggests that although the relatively well-conducting scala tympani turns out to be the main current pathway, the exponential decay of current is only a good description of the far-field behavior. In the vicinity of the electrodes, higher current densities are found, which are best described by a spherical spread of the current.

No studies have been published where the position, shape, and impedance of RE are investigated in detail. Yet there is evidence that a

change in the impedance of the RE interface causes substantial current variations. The essential results that have been observed from the work of Black et al. [21] and Spelman et al. [22] and the present study is that the position of the RE lines affects impedance and currents generated and therefore, the power consumption of the device when stimulating the auditory nerve.

Bone thickness was the adjustable variable of choice in both the theoretical and numerical models in this study because its constant resistivity (ρ) is larger than the other tissues; therefore, it will have the strongest effect on conductivity. A real-life example is observed in patients with ossifications. In these cases, both current threshold and electrical impedance are higher than average. This occurs because the resistivity (ρ) increases. As shown in Eq. (5), for constant size and length, but increased resistivity coefficient, the resistance of the material increases. The increase in ρ is because of the increased bone density that is caused by ossification. Therefore, we postulate that placing RE closer to the edge of the surgical mastoidectomy may result in a decrease in IC power consumption, particularly in the case where the patient presents ossified regions of the cochlea, because in such cases, a small decrease in this value can mean the difference in deciding whether to keep a certain electrode activated or to disconnect it because of its high power consumption [23].

Peer-review: Externally peer-reviewed.

Author Contributions: Concept - A.R.Mi., M.T.P.Z., J.V.A.; Design - A.R.Mi., M.T.P.Z., J.V.A.; Supervision - M.T.P.Z., A.R.Ma.; Resources - A.R.Mi., M.T.P.Z., J.V.A.; Materials - A.R.Mi., M.T.P.Z., J.V.A.; Data Collection and/or Processing - A.R.Mi., M.T.P.Z., J.V.A.; Analysis and/or Interpretation - A.R.Mi., M.T.P.Z.; Literature Search - A.R.Mi., M.T.P.Z.; Writing Manuscript - A.R.Mi., M.T.P.Z.; Critical Review - A.R.Ma.

Conflict of Interest: No conflict of interest was declared by the authors.

Financial Disclosure: The authors declared that this study has received no financial support.

REFERENCES

1. Firszt JB, Holden LK, Skinner MW, Tobey EA, Peterson A, Gaggl W, et al. Recognition of speech presented at soft to loud levels by adult cochlear implant recipients of three cochlear implant systems. Ear Hear 2004; 25: 375-8. [CrossRef]

2. Goehring JL, Hughes ML, Baudhuin JL, Lusk RP. How well do cochlear implant intraoperative impedance measures predict postoperative electrode function? Otol Neurotol 2013; 34: 239-44. [CrossRef]

3. Black RC, Clark GM, Patrick JF. Current distribution measurements within the human cochlea. IEEE Trans Biomed Eng 1981; 10: 721-5. [CrossRef]

4. Choi CTM. Shape optimization of cochlear implant electrode array using genetic algorithms. In Proceedings of the 23rd Annual International Conference of the IEEE; 2001; Istambul, Turkey: Engineering in Medicine and Biology Society. p. 1445-8. [CrossRef]

5. Cohen LT, Xu J, Xu SA, Clark GM. Improved and simplifed methods for specifying positions of the electrode bands of a cochlear implant array. Otol Neurotol 1996; 17: 859-65.

6. Wolfgang G, Franz P, Hamzavi J, Plenk Jr H, Baumgartner W, Czerny C. Intacochlear position of cochlear implant electrodes. Acta Otolaryngol 1999; 119: 229-33. [CrossRef]

7. Hochmair-Desoyer IJ, Hochmair ES, Burian K. Design and fabrication of multi wire scala tympani electrodes. Ann N Y Acad Sci 1983; 405: 173-82. [CrossRef]

8. Bierer JA, Middlebrooks JC. Auditory cortical images of cochlear-implant stimuli: dependence on electrode configuration. J Neurophysiol 2002; 87: 478-92.

9. Henry BA, McKay CM, McDermott HJ, Clark GM. The relationship between speech perception and electrode discrimination in cochlear implantees. J Acoust Soc Am 2000; 108: 1269-80. [CrossRef]

10. Stickney GX, Loizou PC, Mishra LN, Assman PF, Shannon RV, Opie JM. Effects of electrode design and configuration on channel interactions. Hear Res 2006; 211: 33-5. [CrossRef]

11. Loizou PC. Introduction to cochlear implants. IEEE Eng Med Biol Mag 1999; 18: 32-42. [CrossRef]

12. Zierhofer CM, Hochmair-Desoyer IJ, Hochmair ES. Electronic design of a cochlear implant for multichannel high-rate pulsatile stimulation strategies. IEEE T Rehabil Eng 1995; 3: 112-6. [CrossRef]

13. Cohen NL, Roland Jr JT, Fishman A. Surgical technique for the Nucleus® Contour™ cochlear implant. Ear Hear 2002; 23: 59S-66S. [CrossRef]

14. Friedland DR, Runge-Samuelson C. Soft cochlear implantation: rationale for the surgical approach. Trends Amplif 2009; 13: 124-38. [CrossRef]

15. Bobrow LS. Fundamentals of electrical engineering: Oxford University Press; 1996.

16. Chari MVK, Salon SJ. Numerical methods in electromagnetism: Academic Press; 2000.

17. Strikwerda J. Finite difference schemes and partial differential equations: SIAM; 2007.

18. IT'IS Database Low-Frequency Conductivity. [Online]. [cited 2014 september. Available from: http://www.itis.ethz.ch/itis-for-health/tissue-properties/downloads/.

19. Butson CR, McIntyre CC. Tissue and electrode capacitance reduce neural activation volumes during deep brain stimulation. Clin Neurophysiol 2005; 116: 2490-500. [CrossRef]

20. Briaire JJ, Frijns JHM. Field patterns in a 3D tapered spiral model of the electrically stimulated cochlea. Hear Res 2000; 148: 18-30. [CrossRef]

21. Black RC, Graeme MC. Differential electrical excitation of the auditory nerve. J Acoust Soc Am 1980; 67: 868-74. [CrossRef]

22. Spelman FA, Pfingst BE, Clopton BM, Jolly CN, Rodenhiser KL. Effects of electrical current configuration on potential fields in the electrically stimulated cochlea: field models and measurements. Ann Otol Rhinol Laryngol Suppl 1995; 166: 131-6.

23. Shallop JK. Objective measurements and the audiological management of cochlear implant patients. Adv Otorhinolaryngol 1997; 53 : 85-111. [CrossRef]