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Electrical Potential Distribution within the Inner Ear: A preliminary study for vestibular prosthesis design*

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***Abstract*— Rotational cues in patients that suffer from bilateral vestibular loss can be delivered by vestibular prosthesis. Even though great efforts towards the development of a vestibular implant have been made, many parameters have still to be optimized. Numerical simulations of the neural activation during electrical stimulation can give important indications about the optimal electrode insertion site, stimulation waveform and electrode configuration, in terms of the highest selectivity. The first step of this type of numerical simulation requires the digital reconstruction of the human inner ear and the calculation of the spatial electrical potential distribution by means of finite-element methods.**

I. INTRODUCTION

The vestibular system provides linear and angular information about head motion and orientation. The semicircular canals (SCCs) are the part of the vestibular system that is sensitive to angular accelarations. These clues are essential for mantaining vertical postural (with the vestibulo-spinal reflex) and to keep eyes fixed on a target (with the vestibulo-ocular reflex) [1].

A cranial trauma, injury, a too high dosage of gentamicin (an aminoglycoside antibiotic), or other diseases, can cause damage to the vestibular functions. The hair cells, other parts of the inner ear, the neural pathways, or the portion of the brain devoted to process vestibular information, can be malfunctioning. The person experiences blurried vision, dizziness, and inhability to orient and walk properly. This makes falling more likely, especially in elderly people [2]. Pharmacological treatements are not completely effective.

The success of cochlear prosthesis, which can restore hearing in profoudly deaf people, drew to the concept of a vestibular prosthesis. The ideal vestibular prosthesis will process external inputs, provided by gyroscopes fixed to the skull, and selectively stimulate the ampullary nerves to make them spike as they would in a normal-functioning vestibular system [3], [4]. Even though promising trials have been performed in humans [5], which have assessed its feasibility, there are still many issues to be addressed. One of the most urgent concerns is about elicitation of non-target nerves, due to current spread and unwanted electrode movements [6].

Therefore, there are many parameters that have to be carefully chosen and optimized. For example, electrode configuration (size, orientation, multipolar versus monopolar, choice of the reference point) and especially electrode positioning is particularly crucial to elicit different target nerves and achieve the highest prosthesis selectivity. Also the stimulus waveform (amplitude, duration, frequency) plays an important role in fiber recruitment [7].

To facilitate design of electrodes and stimulus protocols for a vestibular prosthesis, parts of the inner ear have been assembled in a 3D digital reconstruction. This provides the first step of modeling vestibular nerve excitation by electrical currents for prosthetic applications.

II. MODEL DESCRIPTION

A. Model Source

The model is elaborated from the free software '3D Virtual Model of Human Temporal Bone', created by Eaton-Peabody Laboratory of Auditory Physiology, of Massachussets Eye and Ear Infirmary (used with permission).

The anatomical surfaces shown in the *Temporal Bone* software were created from histological section from a 14-year old male. The views were digitazed and imported in Amira v3.1 (Mercury Computer Systems/TGS, San Diego, CA) for surface rendering. The software displays several morphological elements. Among others, there is the temporal bone, the ossicles of the middle ear, the cochlea, the bony SCCs, the sensory epithelia of cochlear and vestibular labyrinth, the auditory, the vestibular (the ampullary nerves and the otolith nerves), and major blood vessels (Figure 1a).

In this preliminary work, the considered surfaces are those that are close to the electrode insertion site. Therefore, the bony semicircular canals, the vestibular, the cochlear and the facial nerves are imported (Figure 1b). It is assumed that the temporal bone embedds each part. A saline layer is surrounding the temporal bone.

B. Finite Element Model

Finite element analysis is performed to estimate the electrical potential through the inner ear for a specific electrode configuration. The surfaces of the elements of interest are exported and go through a reverse engineering process, to smooth the parts where the surfaces are more rough and therefore hardly meshable. This makes sense because usually the average is performed between the model anatomies acquired from different animals [8].

Since dielectric relaxation times are much shorter than the typical duration of stimulus pulses, *quasi-static conditions* are assumed. Therefore, tissue reactances are not required

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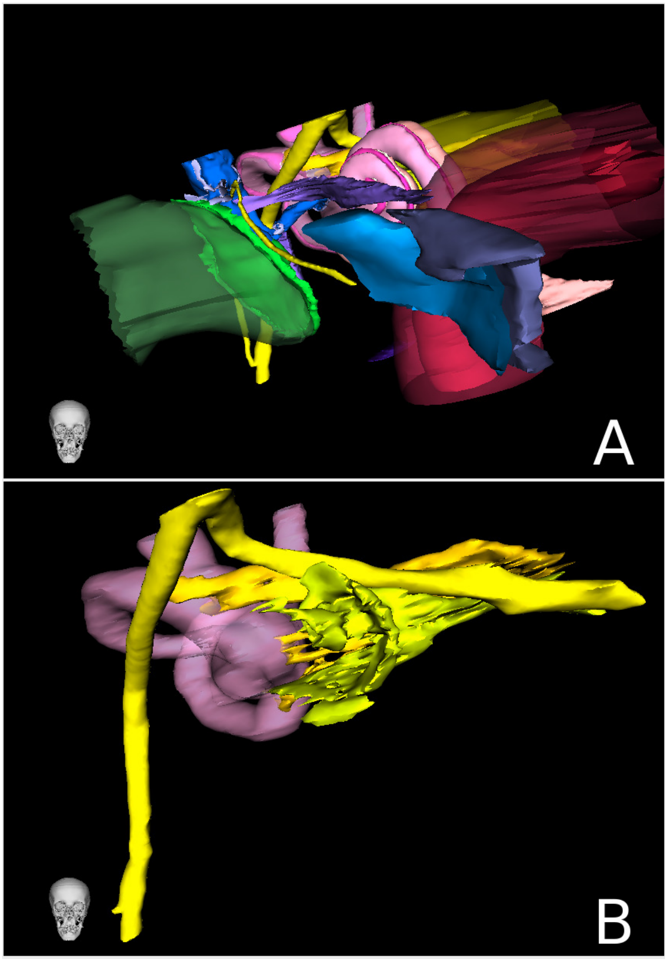

Fig. 1. Screenshots of the '3D Virtual Model of Human Temporal Bone': (a) all elements, except the temporal bone, are shown; (b) bony semicircular canals and the vestibular, cochlear and facial nerve are shown.

and conductivities are sufficient to fully describe the problem. These assumptions imply that, if the electric source changes in time, the transients are negligible and the potential at each mesh vertex reaches almost instantaneously its new equilibrium value.

To represent every electrode configuration is sufficient to perform a series of simulations where a single source is injecting a unitary current. The arranged effects of activated electrodes will be obtained with a linear combination of their voltage distributions, thanks to the superposition principle for linear systems. The linear equation that links the injected current and the potential will be described below.

Moreover, the voltage distribution can be interpolated within the mesh nodes, to determine the potential at any point of the computational domain and linearly rescaled, according to the stimulus temporal profile. It is worth highlighting that even if the finite-element simulation is in a steady-state configuration, the potential values will be tuned accordingly to the biphasic current pulses. The fiber recruitement will be shown in a separate paper.

More specifically, the *volume conductor* problem has to be solved, in order to find the solution of bioelectric fields generated from a given setting of current sources, located in conducting media. The problem consists in calculating the potential field for a specific source configuration, and it is expressed by Poisson's equation:

$$\nabla^2 \Phi = \frac{\nabla \cdot J}{\sigma} \tag{1}$$

which relates the second spatial derivatives (the laplacian) of the potential Φ to the divergence of J, the current density; σ is the domain electrical conductivity [9]. Because of the complexity of the surface geometries, this equation can not be solved analitycally, but by means of numerical implementations. When the domains are delimited by non-trivial boundaries and a non-staggered mesh is used (in this case tetrahedrical), finite element methods (FEM) are a powerful tool to accomplish the numerical solution.

A single electrode is positioned at 0.5 mm from the posterior ampullary nerve. This nerve is chosen according to the work by Wall C. et al. [10], which is the first demonstration of the feasibility of using multiphasic pulse trains applied to one branch of the peripheral vestibular nerve, through a thin layer of bone, to produce reflexive eye movements in human subjects.

This particular electrode configuration is the first in a series of new simulations, where each single ampullary nerve will be elicited in different sites (for example, along the nerve itself or near the corresponding ampulla). In this work, the distance between the electrode and the posterior ampullary nerve has been chosen because it was considered a reasonable value. Papers about functional electrical stimulation of human vestibular nerves [5], [11] do not provide information about the distance.

The electrode is a sphere of radius $100 \mu m$, representing the active part of the stimulating electrode described in [10]. The injected current density is 1 A$/m^3$.

The bone and the semicircular canals have isotropic conductivity, while the nerve tissue has anisotropic conductivity that differs in the radial and longitudinal directions. The axial conductivity was assigned along the main nerve direction (Table I).

TABLE I
TISSUE CONDUCTIVITIES [8]

Tissue Domain	Conductivity (Ω^{-1}/m)
Bone	0.0139
Nerve (longitudinal)	0.3333
Nerve (transverse)	0.0143
Saline layer	2.0
Electrode	1e6

Each mesh is performed with COMSOL Multiphysics (COMSOL Inc, Burlington, MA) for finite element analysis using COMSOL AC/DC Electric Currents Module, in Stationary mode. Meshing has required a particular effort, due to the different sizes of surfaces and volumes of the element of interest (Figure 2). Each simulation is solved iteratively on a 64-bit multicore processor using the conjugate gradients

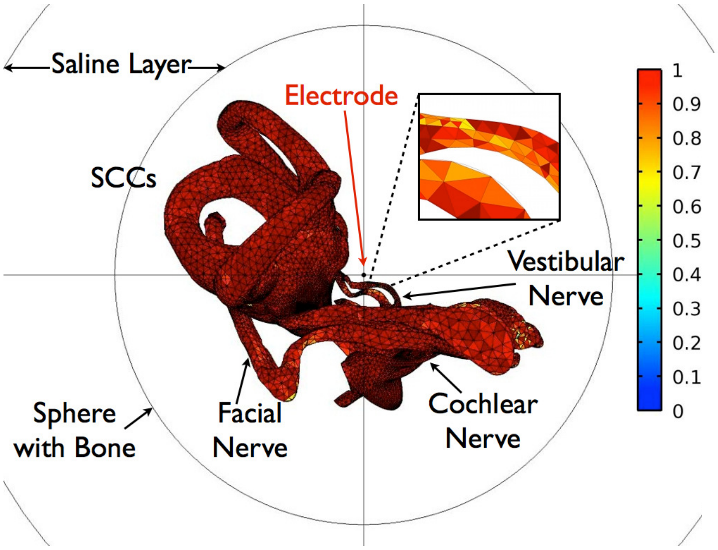

Fig. 2. Mesh quality of semicircular canals, vestibular, facial and cochlear nerves. The colorbar refers to the aspect ratio of tetrahedral elements.

method. The resulting solution, defining the potential at each mesh vertex for a unit stimulus current, will be sampled at points in space corresponding to the nodes of Ranvier along the axons in each nerve branch and will drive the extracellular stimulation.

III. RESULTS

Figure 3 displays two different views of normalized potential field normalized for semicircular canals, the vestibular, cochlear and facial nerves. The highest voltage values, in the domain of interest, are near the posterior ampullary nerve, which is the target nerve.

The fiber recruitment is driven by not only the extracellular potential amplitude at a given point, but also by the second spatial derivative along the nerves, the so-called *activation function* [12]. It is used to predict the sites of action potential initiation sites and estimate the electrical stimulation efficacy.

Moreover, nerve activation depends also on the fiber geometry and neural modeling. The last two deeply affect the activation threshold. Therefore, cochlear and facial nerves may be elicitated because of their different neuro-physiology with respect to the vestibular nerve, even though they are more distant from the electrode.

In clinical trials, it was observed that minimal displacements of the electrode resulted in drastic changes in the amplitude of the response or in the occurrence of facial movements [5]. Therefore, the neural selectivity (percentage of activated fibers of the target nerve versus the other ampullary nerves and versus the cochlear and facial nerves), together with the eye movement responses, will be the main feedback of the insertion site.

Figure 4 displays the normalized potential field relative to each singular nerve and conveys information about the possible activation site of each nerve.

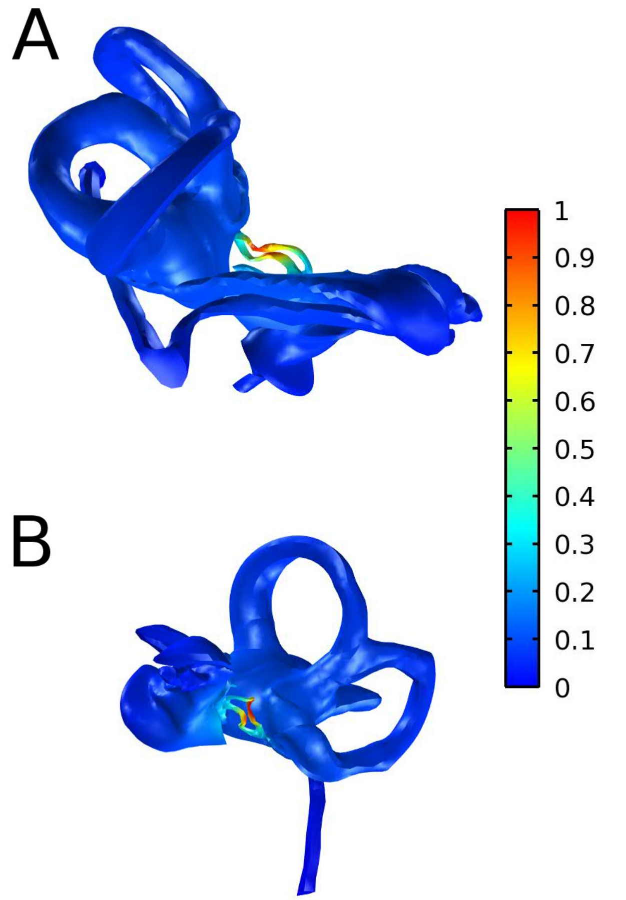

Fig. 3. Normalized voltage spatial distribution: (A) XY view; (B) YZ view.

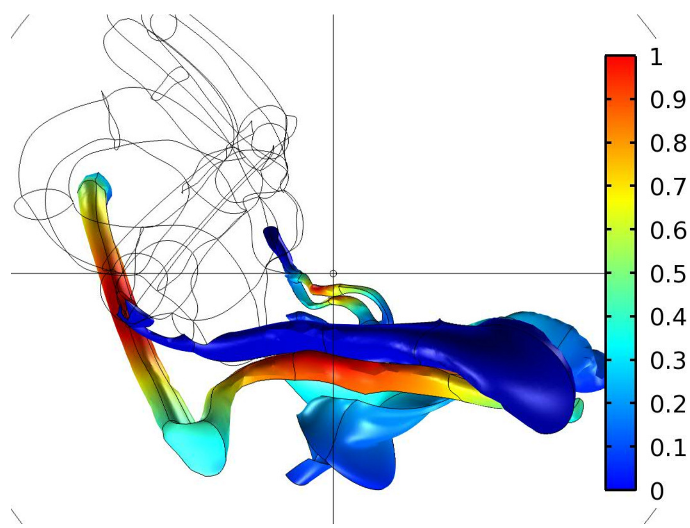

Fig. 4. Normalized voltage spatial distribution relative to each nerve.

IV. CONCLUSIONS

The aim of this work is to calculate the spatial distribution of the electrical potential, due to an external electrode near an ampullary nerve. The values of the potential field will be used to drive extracellular stimulation at each node of Ranvier of vestibular, facial and cochlear fibers, after a proper modeling of their geometry and physiology.

Future work will include positioning the electrode in different sites, such as near the other ampullary nerves [5], [11] and near the ampullae [13]. In each location different types of electrodes will be inserted (monopolar versus multipolar), with several types of stimulus waveforms, to check which configuration promotes the highest selectivity.

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