Modeling ECAP in Cochlear Implants Using the FEM and Equivalent Circuits

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Evoked compound action potential (ECAP) has been used clinically to determine whether auditory nerves are responding to electrical stimulation in cochlear implant systems. In this paper, a novel scheme is proposed to model ECAP and its measurement using a generalized Schwarz-Eikhof-Frijns nerve model and equivalent circuits based on the finite-element (FE) method. A 3-D FE model of a cochlear implant is used in this scheme. The ECAP modeling result is validated with clinical ECAP measurement results.

Index Terms— **Coupling circuits, electrical stimulation, equivalent circuits, evoked compound action potential (ECAP), finite element method (FEM), generalized Schwarz-Eikhof-Frijns (GSEF) model.**

I. Introduction

COCHLEAR implants have been an effective treatment for hearing impaired patients. Evoked compound action potential (ECAP) has been used clinically to determine whether the auditory nerves are responding to cochlear implant stimulation. In addition, ECAP has been used clinically as an objective mean to estimate stimulation parameters such as the threshold stimulation level and the most comfortable stimulation level, which is the stimulation dynamic range of each channel. Currently, it can take up to 1–2 hours to measure the ECAP clinically for the 12–22 channels in the available cochlear implants in the market. An alternative is to create a cochlear implant model to study the means of speeding up the ECAP measurement process. Currently there is no easy way to model ECAP in a cochlear implant system. This paper presents a novel scheme to model ECAP using the finite-element (FE) method [1], [2], the generalized Schwarz–Eikhof–Frijns (GSEF) nerve fiber model [3], and equivalent circuit models on a 3-D cochlea model. The simulation results are validated by the ECAP measured in cochlear implant users.

II. Method

FE models of cochlear implants have been developed to study electrical stimulation [1], [2]. In Fig. 1, a 3-D FE model of a cochlear implant is used to study ECAP. This FE model is created by rotating a 2-D cross section of a human cochlea (Fig. 2) spirally [1], [2]. The Poisson equation (1) is solved to compute the electrical potential distribution in the human cochlea (inner ear) and the auditory nerve fibers. The conductivity of the medium and other parameters are given in [2]–[4]

$$-\nabla^2 \phi = f \tag{1}$$

where ϕ is the electrical potential and f is the source function [3]–[5]. First, a sufficiently large voltage or current input is
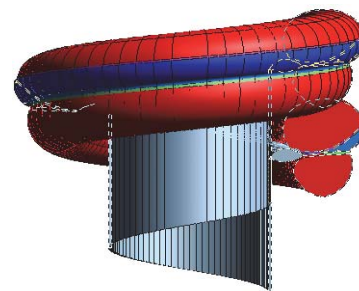
Fig. 1.   3-D FE model of a cochlear implant.

imposed on the stimulating electrodes on the electrode array in the FE model. Next, the electrical potential distribution is computed to determine whether the extracellular potential on the nerve fibers is sufficiently large to excite (or evoke) the nerve fibers [6]. Fig. 3 shows how ECAP is evoked and measured. The nerve fibers are shown in blue in the center of Fig. 3. Stimulating electrodes (in blue) on the electrode arrays are used to excite (or evoke) the nerve fibers (upper center). Once the nerve fibers are excited, their action potentials are generated and coupled electrically to the sensing electrode (purple) on the electrode array to become ECAP. Typical action potential generated by a nerve fiber is shown in Fig. 4 [3], [6]. This paper follows the GSEF model specifications [3] as listed below and nerve fibers are put into the 3-D cochlea model (Figs. 1, 2 and 5). A nerve fiber is like a cable with 16 nodes of Ranvier and separated by internode myelin sheaths. The first node of the nerve fiber has a length of 10 μm, the other nodes have a length of 1 μm and the central axon has a diameter of 3 μm. Nodes 4 and 5 (gap 4) are separated by the soma which has a length of 20 μm and a diameter of 14.286 μm (including the internode myelin sheath). The gap distance between the other nodes are listed in the following. The distance of gaps 1, 2, 3 and 5 are 150 μm, while gaps 6, 7 and 8 are 200 μm, 250 μm, and 300 μm, respectively with a diameter of 4.286 μm (including the internode myelin sheath). The nodes of Ranvier are modeled as floating equipotential surfaces (the potential is computed using the FE model) while the electrodes are modeled as imposed equipotential surfaces.
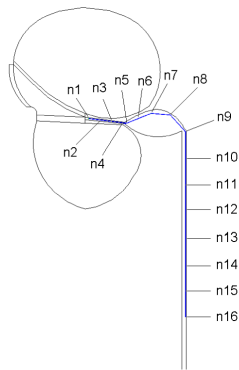
Fig. 2. Cross section of a human cochlea. The nerve fiber with its 16 nodes is illustrated in blue.
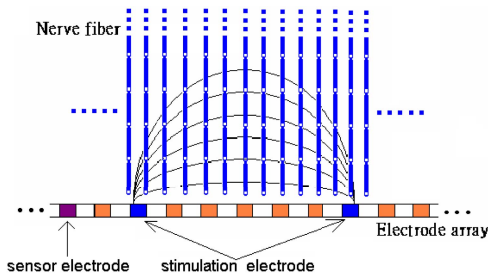

Fig. 3. Illustration of how ECAP is evoked and measured. The stimulating electrodes excite the nerve fibers in the upper centerpart which generate action potential (Fig. 4). The action potential is coupled to the sensing electrode on the bottom left.
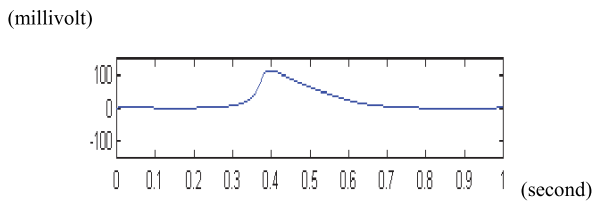

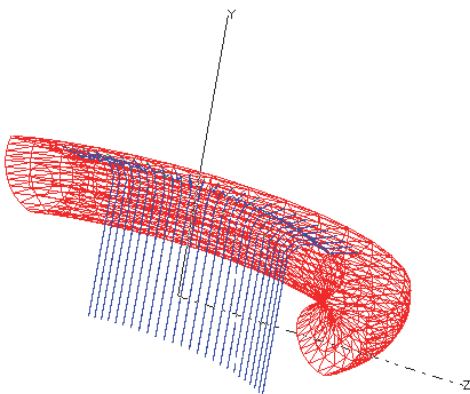Fig. 4. Action potential generated by a nerve fiber.

Fig. 5. Auditory nerve fibers' position relative to scala-tympani (the lower chamber of the cochlea), where cochlear implant electrodes are located.

Fig. 5 shows a half turn 3-D cochlear implant model, where 40 GSEF models were put into the first half turn of the cochlea with an interval of 4.5°. These 40 GSEF models represent the typical 10 000 auditory nerve fibers distributed over the cochlea uniformly over the same space.
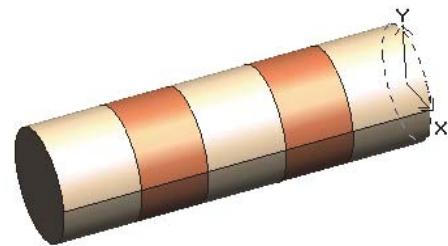

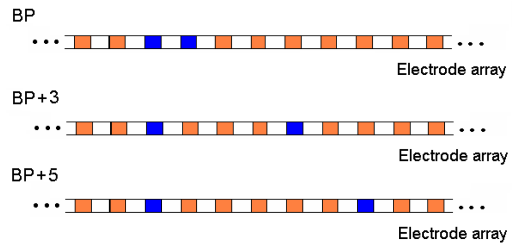Fig. 6. Segment of the ring electrode contacts (dark brown).

Fig. 7. BP, BP+3, and BP+5 configurations. The electrode contacts in blue show stimulating electrode pairs.
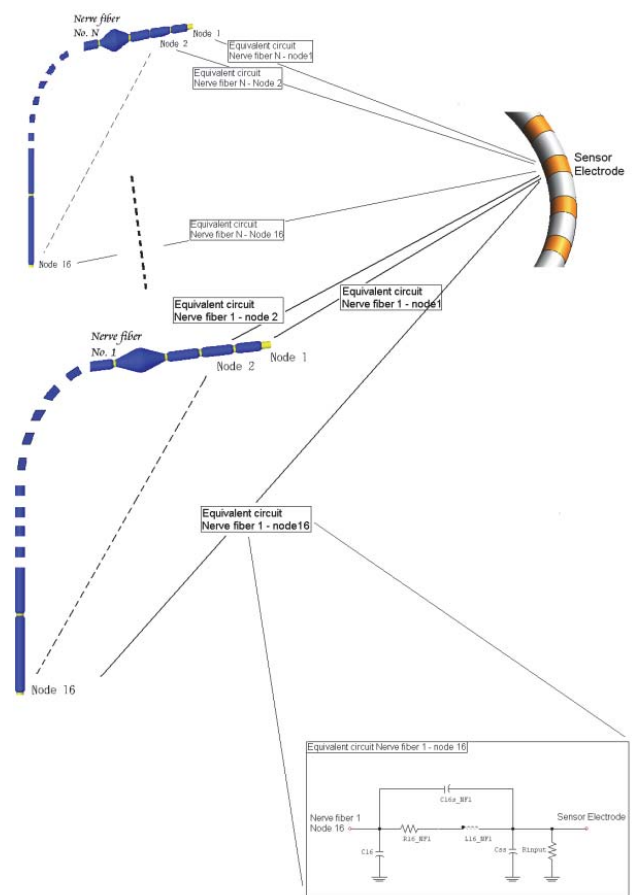

Fig. 8. Equivalent circuit [8] between auditory nerve fibers and a sensing electrode.

This simulation system uses the Nucleus 24-CI24M electrodes [7]. A full electrode array is composed of 22 platinum electrode contacts and is embedded in a silicon carrier Fig. 6 shows a segment of the ring electrode contacts. The electrode
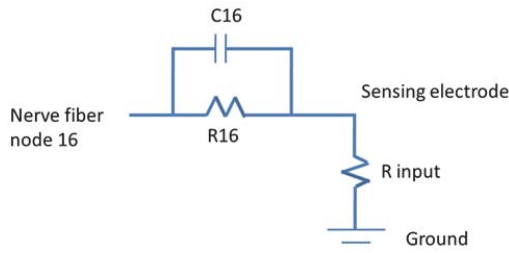
Fig. 9. Equivalent circuit between node 16 of one nerve fiber, the sensing electrode, and the system ground. The equivalent parallel resistor and capacitor circuits are computed between nerve fibers and the sensing electrode based on the 3-D FE model in Fig. 1. An equivalent resistor between the sensing electrode and the system ground is computed using the same 3-D FE model.
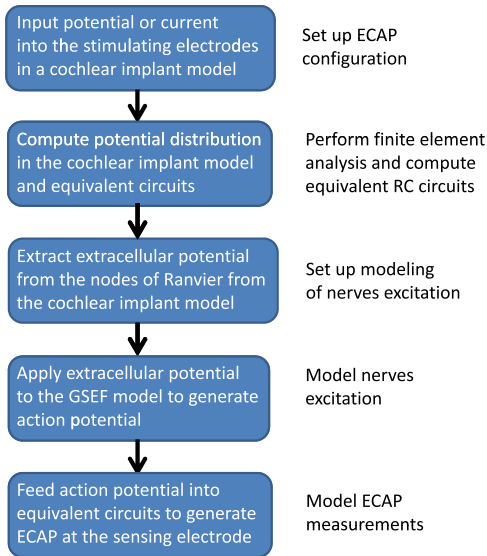

Fig. 10. Flowchart showing how to compute ECAP using a 3-D FE model, GSEF models, and equivalent circuits.
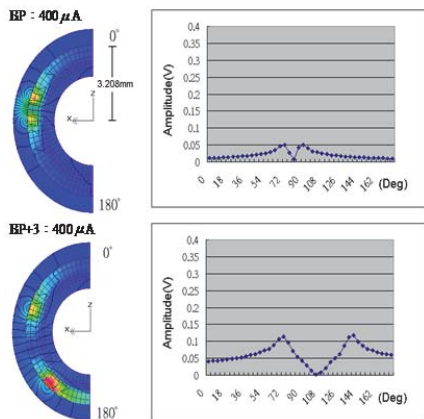

Fig. 11. Potential distribution (orange/red are peaks) of a half turn cochlea (left) and the voltage plot versus angle (right) for BP and BP+3 configurations with 400 μA input current.
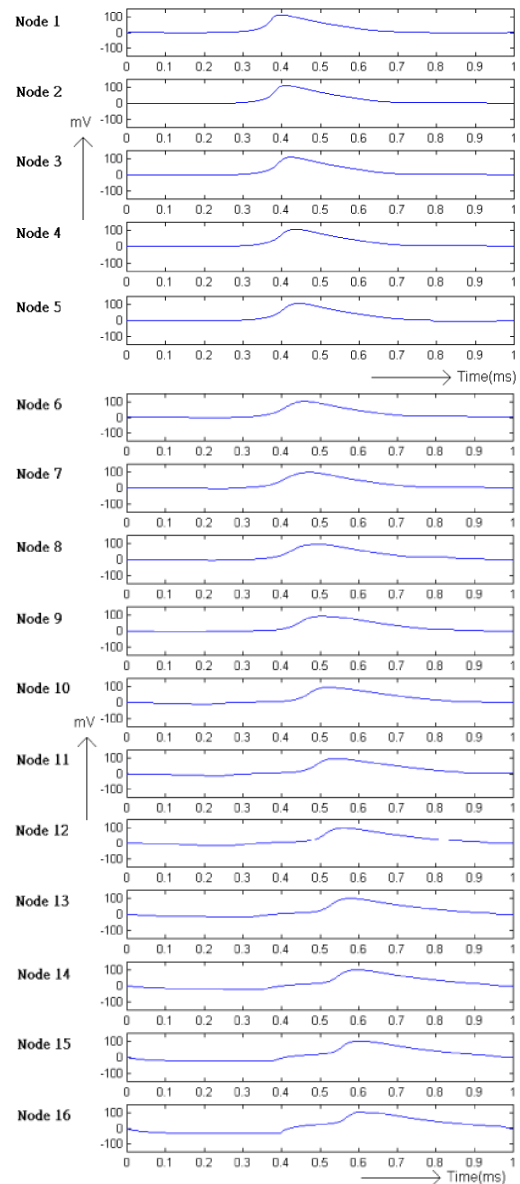

Fig. 12. Action potential for nodes 1–16 generated by the GSEF model with electric potential extracted from the cochlear implant as extracellular potential.

contact is 0.3 mm in length and 0.5 mm in radius and is separated by 0.45 mm of silicon between the electrodes.

Typically, two neighboring electrodes are used as the stimulating electrode pair in the bipolar (BP) mode (Fig. 7), where one is positive and the other negative. When the two stimulating electrodes are separated by three electrodes, it is

called BP+3. Similarly, if two stimulating electrodes are separated by five electrodes, it is called BP+5.

An equivalent circuit between auditory nerve fibers and a sensing electrode is shown in Fig. 8. The coupling between the nodes of the nerve fibers and the sensing electrode can be modeled by an equivalent parallel resistance, inductor and capacitor circuit (RLC circuit in the lower bottom right of Fig. 8) [8]. The parameters of the equivalent RLC circuits are determined from the FE simulation. Likewise, the coupling between the sensing electrode and ground can be modeled by an equivalent resistor circuit from the same 3-D FE model. With an RLC circuit for each node and the sensor, 16 equivalent circuits are required for the 16 nodes of the nerve fibers, i.e., n1–n16. Since there are 40 auditory nerve fibers or GSEF models in our CI model, a total of 640 equivalent circuits are required to compute the ECAP for a particular ECAP configuration (i.e., for a given stimulating electrode pair). Since the actual
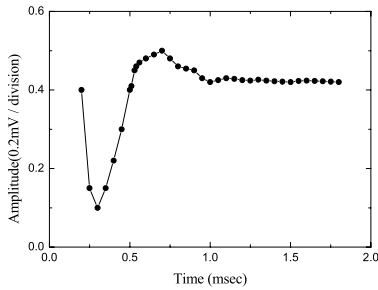
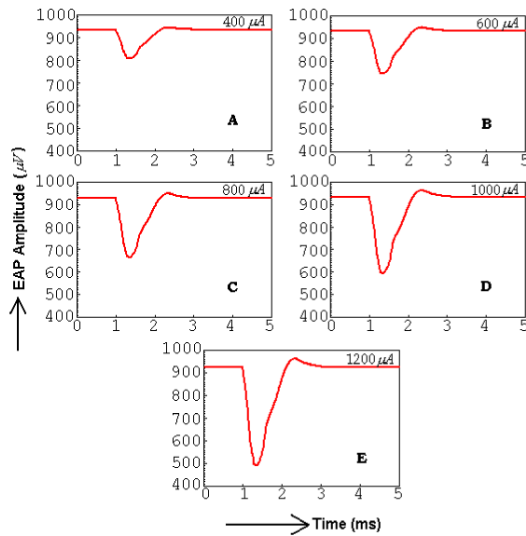Fig. 13.   Typical ECAP measured [7] from a CI patient in time domain.

Fig. 14.   ECAP generated from the FE model, GSEF models and equivalent circuits in time domain.
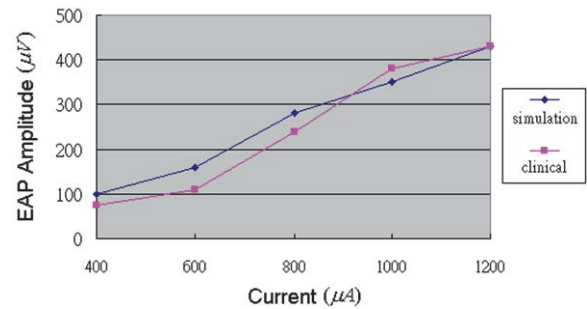

Fig. 15.   ECAP generated from simulation versus clinically measured ECAP is plotted against the current.

with various BP+5 input currents. The ECAP generated by computer models is consistent with the pattern of ECAP as measured by clinical experiments [7]. The pattern generated by BP and BP+3 inputs also show a similar pattern (not shown) and are consistent with clinically measured results. Fig. 15 compares the clinically measured ECAP with the ECAP generated by computer models with various current inputs for BP+5. Again, the ECAP generated by computer models matches the pattern of the ECAP measured clinically in time and with various input currents. The results for BP and BP+3 are also consistent with the clinical ECAP results, but cannot be included due to space limitation.

IV. CONCLUSION

The measurement of ECAP is modeled using a 3-D FE model, GSEF models and equivalent circuits. The ECAP model is validated by clinical ECAP measurement results. This makes the use of computer modeling approach to study ECAP measurement feasible.

ACKNOWLEDGMENT

stimulation frequency for the input current is <500 Hz, the inductance is negligible. The RLC equivalent circuit can be simplified into an *RC* circuit (Fig. 9). In addition, the 640 equivalent circuits can be reduced to 80–160, depending on the area being activated. A flowchart showing a step-by-step process to compute ECAP is shown in Fig. 10.

III. RESULTS AND DISCUSSION

With a 3-D cochlear implant model, equivalent circuits [8] are computed based on the steps as shown in Fig. 10. A typical auditory nerve will generate a spike such as the one shown in Fig. 4 when computed by the GSEF model. From steps 1 and 2 of the flowchart as shown in Fig. 10, a typical potential distribution is obtained, such as the BP and BP+3 stimulation configurations with 400 μA as input current which are shown in Fig. 11. The peaks are in red and orange while zero is in blue. The activated region for BP is much narrower than BP+3 and BP+3 is narrower than BP+5 (not shown). Then the extracellular potential from nodes 1 to 16 of the auditory nerve fibers are extracted and imported to the GSEF model to generate intracellular action potential. Action potentials from nodes 1 to 16 generated by a GSEF model are shown in Fig. 12. The action potential is then used as the input to the corresponding equivalent circuit to generate an ECAP at the sensing electrode. Fig. 13 shows the clinically measured ECAP from a patient whereas Fig. 14 shows the ECAP generated from computer models

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