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Functional Simulation of the Cochlea for Implant Optimization

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***Abstract*—Cochlear implantation is a surgical technique which aims to restore hearing in people with deep hearing loss. However, outcomes of the surgery still exhibit a large variability between patients. Among the factors that contribute to variability the most important are morphological differences in anatomical structures between patients and incorrect implant placements. In order to address these issues, it would be desirable to have a functional model of the cochlea which incorporates inter-patients variability and simulate electrode placement. To this end, we present a finite element model which captures the interaction between the cochlear partition, modeled as an elastic solid with finite deformation, and the perilymph fluid, modeled as a compressible, viscous fluid. Numerical results show that the membrane responds to changes in the stimulation frequencies.**

***Index Terms*—FEM, cochlear partition, fluid-structure interaction, perilymph, Elmer**

I. INTRODUCTION

HEARING impairment or loss is among the most common reasons for disability. Worldwide, 27% of men and 24% of women above the age of 45 suffer from hearing loss of 26dB and more [1], [2], meaning that they can only hear sounds with intensity higher than 26dB. The cochlear implant (CI) is a surgically placed device that converts sounds to electrical signals, bypassing the hair cells and directly stimulating the spiral ganglion cells. Extreme care has to be taken when inserting the CI's electrode array into the cochlea to obtain the best possible improvement in hearing while not damaging residual hearing capabilities [3].

The HEAR-EU European project[1] aims at reducing the inter-patient variability in the outcomes of surgical hearing restoration by improving the design and operative protocol. In this context we propose that the availability of a good functional model of the cochlea can improve implant design, insertion planning and selection of the best treatment strategy for each patient. To this end, we present in this paper an initial model of the interaction between fluid and structure in the human cochlea.

In the following sections we discuss briefly the physiology of hearing and the approximation used in our model. Then we present the general derivation of the equation of motion in both fluid and solid. Finally we present the results achieved by running our model on simple input signals.

A. Physiology of hearing

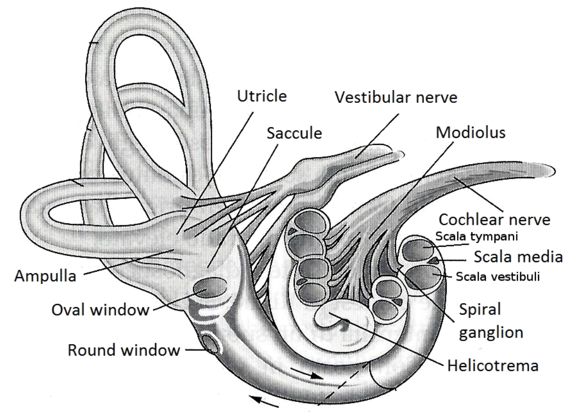

Figure 1. Anatomical structure of the cochlea. Pressure waves that originate from the oval window propagate through the scala vestibuli and scala tympani and cause vibration of the structures within the scala media. Adapted from [4]

The perception of sound is a staged process wherein vibrations from the environment are transduced into nerve impulses to be interpreted by the brain. These impulses encode frequency (tone), and intensity information. The transduction of pressure waves, which we interpret as sound, begins at the tympanic membrane. Here these waves induce a vibration of the membrane, and this movement is transferred to the cochlea via a chain of three minute bones called ossicles: malleus, incus and stapes. The final bone (stapes) interfaces with a membrane (oval window) on the cochlea, and induces pressure waves in the fluid within this structure. The cochlea is a spiral shaped structure (See Fig. 1) with a rectified length of about 30 mm wrapped around a conical bone, the modiolus. Its section has a diameter of approximately 9 mm and is internally organized in three tubular sections filled with liquid. The upper and lower cavities are connected by the helicotrema while the middle structure contains the basilar membrane and the organ of Corti, the sensory organ of hearing. When the stapes transmits vibrations to the oval

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[1]http://www.hear-eu.eu

window, pressure waves propagate through the tympanic and vestibular scales and cause the basilar membrane to oscillate in a complex form. These movements are transmitted to the inner hair cells (IHCs) which transduce them as electrical signals and then send these signals to the spiral ganglion cells. The complex oscillatory pattern of the basilar membrane is caused by the significant change of its mechanical properties along its length. At the apex the membrane is large and elastic while at its base it is thin and stiff. This change allows the basilar membrane to separate the frequency of a propagating sound, acting as a mechanical spectrum analyzer and it is essential to hearing. Low frequencies are keener to resonate on the membrane in the apex of the cochlea, while higher frequencies on the base. This is known as tonotopic mapping of the frequencies in the cochlea. Due to this mapping, different groups of hair cells respond to different frequencies, activating a glutamate-driven chemical synapses with the spiral ganglion cells [4], [5].

B. Related work

Edom et al. [6], [7] present a computational model for dynamical phenomena in the cochlea, where they analyze viscous effects and cochlear responses to non-harmonic stimuli. Finite volume (FV) simulations have also been reported [8]. Because of the coupling with elastic structural equations in fluid-structure interaction (FSI), a finite element approach (FEM) such as ours is more natural and thus preferred over the finite volume (FV) method.

II. METHODOLOGY

In this section we detail the mathematical formulation of the dynamics of the fluid and the elasticity of the solid body. We also discuss the cochlear modeling and the coupling strategies for FSI.

A. Geometry and mesh model

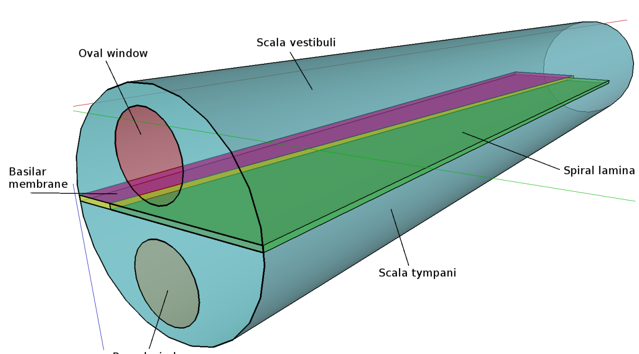

Figure 2. Simplified 3D model of the cochlea. Pressure signal is applied to the oval window and propagates through the fluid domain. The fluid domain is composed by scala tympani and scala vestibuli, while the solid domain is made up by spiral lamina and basilar membrane. The scala media is not included in this simplified model.

As described in Section I, the sound is perceived by the cochlea as a vibration of the basilar membrane that separates the scala vestibuli from the scala timpani. In our model we make several approximations. First, the coiling of the cochlea is a secondary effect [9] and we neglect it. Then we only include the spiral lamina and the basilar membrane in our representation of scala media. The uncoiled and simplified cochlear anatomy is depicted in Fig. 2. There you can note the oval (red) and round (dark green) window elastic membranes -where pressure waves are generated and absorbed- and the fluid and the solid compartments. The former is composed by scala tympani and scala vestibuli while the latter by spiral lamina and basilar membrane. This is modeled taking into account the approximate real geometry which is shorter at the base and larger at the apex. However, basilar membrane is currently modeled as an isotropic solid with a constant elastic module. Actually, the variable stiffness along its length is important to cause the peak vibration for tones of different frequency in a different location of the cochlea. Therefore the propagation of pressure waves in our geometry would not follow exactly the Greenwood model [5]. Finally, the active cochlear amplification mechanism was not considered in this stage of modeling [10]. Nevertheless we expect to capture the essential qualities of the interaction of pressure waves and the solid membrane.

The geometrical model described so far was then meshed using SALOME[2] and Netgen[3] algorithm and it is shown in Fig. 3 with the round window and the basilar membrane colored in red. The resulting mesh had 21753 nodes. The volume elements corresponding to the fluid domain were not shown to prevent cluttering of the image.

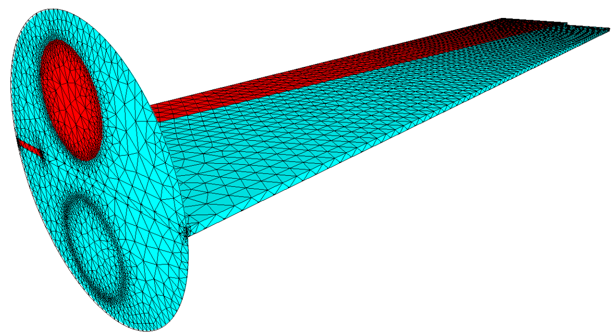

Figure 3. Mesh of the simplified 3D model. Here volumetric elements of the mesh which represent fluid are not shown because they would clutter the illustration. Basilar membrane and oval window are depicted in red.

The system is stimulated by applying a sinusoidal function of time with frequency f as a pressure boundary condition in the oval window region. The amplitude of the input pressure signal was chosen as 1 Pa. The walls are modeled as rigid.

B. Fluid dynamics

In solid and liquid materials viscous fluid flow is governed by Navier-Stokes equations, which can be derived from the

[2]http://www.salome-platform.org/
[3]http://www.hpfem.jku.at/netgen/

Table I
MODEL PARAMETERS APPLIED TO THE SOLID AND FLUID DOMAIN. PARAMETERS WERE SELECTED BASED ON [8]

Symbol	Description	Value	Domain
μ	Dynamic viscosity	1.2 mPa·s	Fluid
λ	Volume viscosity	3.09 mPa·s	Fluid
ρ_F	Density	1000 kg/m^3	Fluid
ρ_S	Density	1000 kg/m^3	Solid
E	Young's module	1000 Pa	Solid
γ	Poisson ratio	0.3	Solid

basic principles of conservation of mass, momentum and energy [11]. We briefly summarize our analysis from the Cauchy momentum equation:

$$\rho\left(\frac{\partial}{\partial t}\mathbf{u}+(\nabla\cdot\mathbf{u})\mathbf{u}\right)=\nabla\cdot\sigma+\mathbf{f} \qquad (1)$$

where u models the velocity of the fluid, σ is the stress tensor and $\mathbf{f}$ the resulting of all other forces such as gravity. If we decompose the stress tensor σ into its hydro-static (π) and deviatoric ($\mathbb{T}$) components we obtain the general form of the Navier-Stokes equations:

$$\rho\left(\frac{\partial}{\partial t}\mathbf{u}+(\nabla\cdot\mathbf{u})\mathbf{u}\right)=-\nabla\pi+\nabla\cdot\mathbb{T}+\mathbf{f} \qquad (2)$$

To complete the equations we need to precise further the constitutive law for the stress tensor, the continuity equation and the state equation. We consider a Newtonian isotropic fluid for which the stress tensor depends linearly on the strain such as: $\mathbb{T}_{ij}=\mu\left(\frac{\partial u_i}{\partial x_j}+\frac{\partial u_j}{\partial x_i}-\delta_{ij}\frac{2}{3}\frac{\partial u_k}{\partial x_k}\right)$ and $\pi=p-\lambda\frac{\partial u_k}{\partial x_k}$ where μ is the dynamic viscosity coefficient, p is the thermodynamic pressure and λ the volume viscosity. If we consider the continuity of mass for a compressible fluid in absence of sources:

$$\frac{\partial\rho}{\partial t}+\nabla\cdot(\rho\mathbf{u})=0 \qquad (3)$$

and substitute the expression of the stress tensor in Eq. 2 we obtain the Navier Stokes formulation for compressible Newtonian fluid:

$$\rho\left(\frac{\partial\mathbf{u}}{\partial t}+\mathbf{u}\cdot\nabla\mathbf{u}\right)=-\nabla p+\nabla\cdot\left(\mu(\nabla\mathbf{u}+(\nabla\mathbf{u})^T)\right)+\nabla\left((\lambda-\frac{2\mu}{3})\nabla\cdot\mathbf{u}\right)+\rho\mathbf{g} \qquad (4)$$

where $\mathbf{g}$ represents the gravity force.

These are the equations solved on the fluid domain.

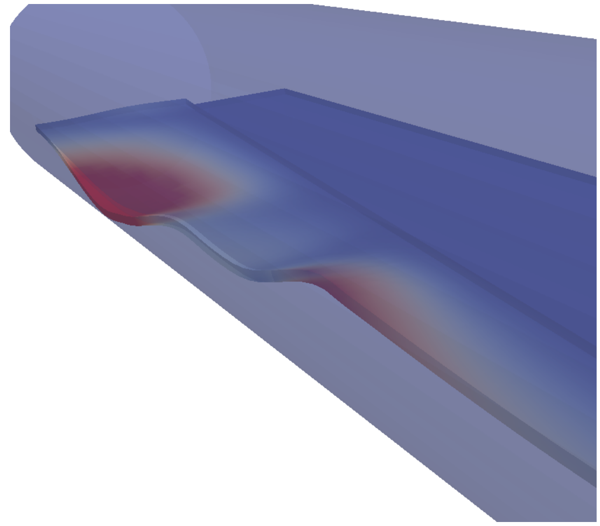

Figure 4. Detail of a traveling wave in a 3D simulation of the entire cochlear domain. Displacement of the basilar membrane is exaggerated for the purpose of visualization.

C. Elastic solid

In the solid domain, we are modeling biological soft tissues with possibly large deformation. To this end, we use the finite deformation formulation [12]

$$\begin{cases}\rho_0\ddot{u}-\nabla\cdot S & =\rho_0 b_0\\ S & =F\bar{S}(C)\\ F & =I+\nabla u\\ C & =F^TF\end{cases}$$

where ρ_0 gives the density when the body is in the reference position, the tensor field S is referred to as the first Piola-Kirchhoff stress, and $b_0=b_0(x(p,t),t)$ gives the body force measured per unit mass. The response function $\bar{S}(C)$ characterizes the second Piola-Kirchhoff stress as a function of the right Cauchy-Green tensor C. Here we use:

$$\bar{S}=\lambda tr(E)I+2\mu E$$

where E is the strain tensor:

$$E=\frac{1}{2}(C-I).$$

III. RESULTS

We implemented our FEM model using the Elmer open-source computational tool [13]. The iteration scheme to solve fluid-structure iteration problems is the following: the Navier-Stokes equations for the fluid domain are solved first and under the assumption of a constant geometry. Then the surface forces acting on the solids are calculated and utilized to calculate stress and displacement of the elastic structure. Using the resulting displacement velocities as fixed boundary conditions the fluid domain is solved again. The procedure continues until the solution has converged.

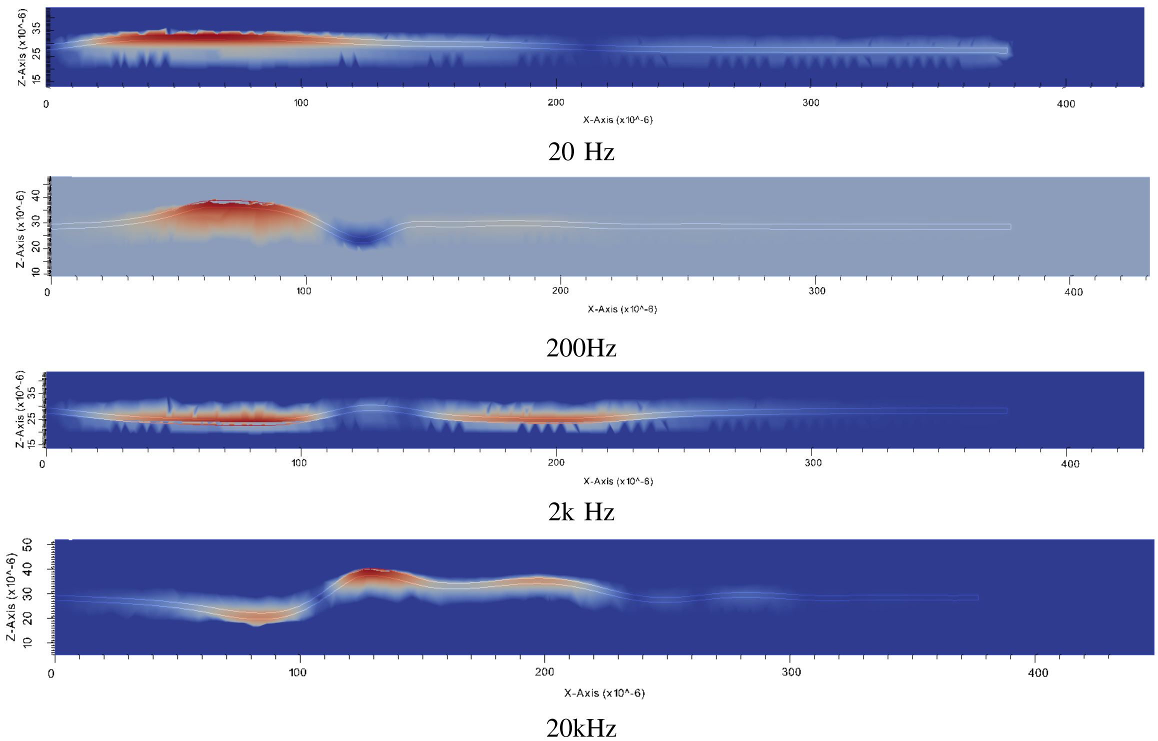

Figure 5. Configuration of the cochlear membrane corresponding to the maximum absolute displacements along z axis. The position of the maximum and the configuration of the membrane depend on the input frequency. Here we present the results for 20, 200, 2k and 20k Hz.

We selected four frequencies inside the human hearing spectrum (20, 200, 2k and 20kHz) and simulated the response of our model to those stimuli. A separate simulation with a timestep size of 100 μs was run for each stimuli. Each simulation run for 100 time-steps and took approximately 30 minutes on an HP Z620 workstation with 10 cores and 8 GB of RAM. Fig. 4 shows an illustrative detail of the traveling waves found in the 3D simulation of the cochlear domain. Displacement is exaggerated to ease visualization. In Fig. 5 we present the response of the basilar membrane to the four simulated input frequencies. For each frequency we selected the deformed configuration which presented the maximum displacement along the z axis. In real cochlea, low frequency tones should have the peak at the cochlear apex, while high frequency tones at cochlear base, close to the oval window. Even if the model is not detailed enough to reproduce exactly the same behavior observed in real cochlea, these preliminary results support in our opinion the idea that FEM formulation of the interaction between pressure waves in the perilymph and cochlear partition is a feasible and affordable way to study functional mechanics of he cochlea. Further improvements will focus on the quality of the model of the cochlea and will use more physiological stimulation patterns.

IV. Conclusions

The proposed model is able to simulate the interactions between the most important cochlear structures, coupling the fluid with the internal structures of the ear. Acoustical waves in the lymph and traveling waves in basilar membrane were observed, supporting the use of FEM modeling in the study of cochlear dynamics.

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