ENGINEERING TRANSACTIONS • *Engng. Trans.* • **64**, 4, 573–579, 2016
Polish Academy of Sciences • Institute of Fundamental Technological Research (IPPT PAN)
National Engineering School of Metz (ENIM) • Poznan University of Technology

Analysis of a Dynamic Response of the Cochlea Using Fluid-Structure Interaction Model

Konrad KAMIENIECKI[1)], Janusz PIECHNA[2)], Paweł BORKOWSKI[2)]

[1)] *Institute of Micromechanics and Photonics*
Warsaw University of Technology
Świętego Andrzeja Boboli 8, 02-525 Warszawa, Poland
e-mail: k.kamieniecki@mchtr.pw.edu.pl

[2)] *Institute of Aeronautics and Applied Mechanics*
Warsaw University of Technology
Nowowiejska 24, 00-665 Warszawa, Poland

A one-dimensional (1D) model of the cochlea of the inner ear has been built and validated against the previously built three-dimensional (3D) fluid-structure interaction (FSI) model of the cochlea. The 1D model has been used to assess the influence of the round window impedance on the pressure distribution in the cochlea. It was shown that high impedance, which enables compression reflection pressure wave at the round window, leads to the biggest pressure difference between the scala vestibule and the scala tympani in the cochlea, which may lead to a stronger excitation of the basilar membrane.

Key words: inner ear, cochlea, differential pressure.

1. Introduction

A lot of people in the world suffer from deafness. In order to be able to design better hearing aids it is important to understand well the macro-mechanics of the human hearing system.

1.1. Anatomy of human hearing system

The human hearing system, shown in Fig. 1, consists of three sections: the outer ear, the middle ear, and the inner ear. The inner ear can be divided into the semicircular canals and the cochlea. The semicircular canals are responsible for the sense of the body balance, while the cochlea is the organ responsible for hearing. The cochlea is a coiled bony tube filled with a liquid whose mechanical

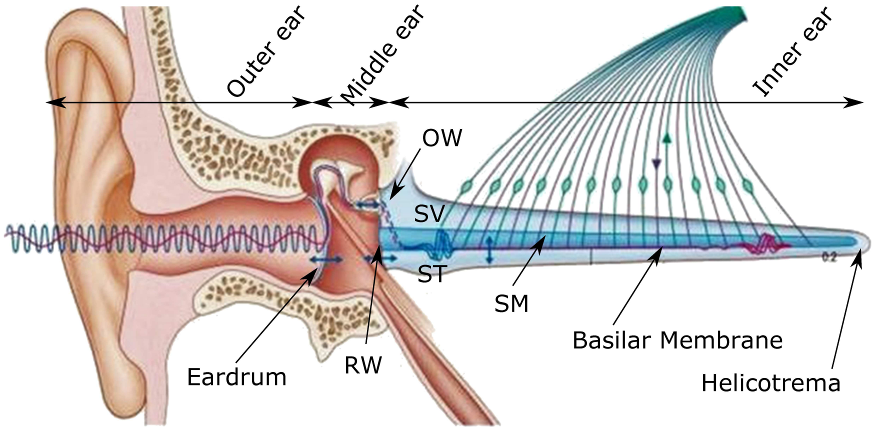

Fig. 1. Anatomy of the human hearing system. The cochlea is presented as uncoiled structure. The semicircular canals of the inner ear are not shown. OW – oval window, RW – round window, SV – scala vestibule, ST – scala tympani, SM – scala media. The picture is adopted from [3].

properties are similar to mechanical properties of water. This bony structure consists of three, parallel canals: the scala vestibule (SV), the scala tympani (ST) and the elastic scala media (SM) between them. The SM and the ST have a common, elastic wall called the basilar membrane (BM). The SV and the ST are connected at one end in a place called the helicotrema. The other end of the SV is closed by the oval window (OW). The ST ends with the round window (RW). In the literature [1, 2], the SM is very often modeled not as being built of few layers of cells, and the BM is placed directly between the SV and the ST.

The pressure wave in the air excites the eardrum, which transfers the vibrations to the auditory ossicles in the middle ear. The ossicles excite the OW in the cochlea. The vibrations of the OW introduce a pressure wave into the liquid. The pressure difference between the SV and the ST causes the BM to vibrate. Different sound frequencies excite different parts of the BM. High frequencies excite the part of the BM close to the OW, while low frequencies the part closer to the helicotrema.

Two waves can be distinguished in the cochlea: a pressure wave in the liquid, which travels with the speed of 1500 m/s and a wave in the BM, whose speed is one order of a magnitude lower than the one in the liquid.

1.2. State of the art model

Many experiments in hearing have been done in the past regarding frequency-position data [4], differential pressure in the cochlea [5], and deflection amplitude of the OW and the RW at specific frequencies [6, 7]. The experiments provide a lot of interesting information but due to a small size of the inner ear and difficult access location in the temporal bone, they are very difficult to carry out.

Based on the knowledge gained in the experimental work, numerical models are a good complement and help in better understanding of the inner ear macro-mechanics. Many numerical models of the inner ear have been built until now. First models were based on analogy modeling technique using mechanical and electrical components [8–11]. Also analytical models [12–14] and 3D finite element models [6, 15, 16] were calculated. The models were mostly calculated in a frequency domain that provides quasi-steady solution.

The authors of this paper developed a 3D FSI model in the cochlea, calculated in a time domain, which was presented at the IUTAM 2016 conference. This model was calculated using the ANSYS software. The obtained 3D FSI model led to interesting conclusions, but due to a complex numerical FSI scheme in 3D, it took several days to calculate the model.

As the next step in the FSI modeling of the cochlea we would like to propose a 1D FSI model that will be based on 1D Navier-Stokes equations for solving fluid dynamics and a dynamic equation of motion for solving the BM vibrations.

In this paper, we would like to present some preliminary results of the 1D modeling of the cochlea restrained to fluid dynamics coupled with the RW mechanics and possibilities for practical usage of this model. It was assumed that the BM is inflexible, and the acoustic wave in the cochlea does not cause any of its vibrations.

2. Methods

Schematic view of the 1D model of the cochlea is presented in Fig. 2. The SV and the ST are represented by two 1D straight pipes of a constant cross-section connected at the helicotrema. The pipes are filled with a liquid. The velocity and pressure profiles at the cross-sections are assumed to be uniform. The OW is represented by a velocity inlet boundary condition. The impedance boundary condition is applied to the RW. The impedance is defined as the ratio between pressure change and velocity change. By adjusting the impedance of the RW

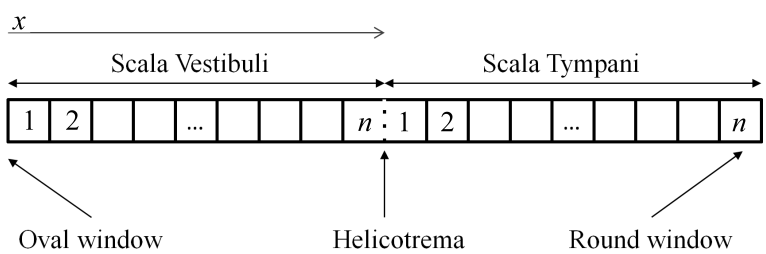

Fig. 2. Schematic view of 1D model divided into n discrete elements.

it is possible to obtain reflection or non – reflection condition for the pressure wave at the RW. The helicotrema is represented by an orifice between the two pipes, which reduces a flow of the fluid. The presence of this pipe narrowing has been taken into account by introducing a discharge coefficient.

The model was excited with harmonically changing velocity at the OW. The amplitude of the velocity is based on the experimental data [6]. Two cycles of the excitation wave were calculated in a time domain

The governing equations of the fluid flow are 1D Navier-Stokes equations (Eqs. (2.1) and (2.2)). The viscosity effects, as the transit time in the cochlea was relatively short, were not taken into account. The fluid equations were solved using the method of characteristics.

$$\frac{\partial\rho}{\partial t} + u\frac{\partial\rho}{\partial x} + \rho\frac{\partial u}{\partial x} = 0 \quad \text{– continuity equation,} \tag{2.1}$$

$$\rho\frac{\partial u}{\partial t} + \rho u\frac{\partial p}{\partial x} + \frac{\partial p}{\partial x} = 0 \quad \text{– momentum equation,} \tag{2.2}$$

where ρ – density, u – fluid velocity in x direction, p – pressure, t – time, and x – position along the cochlea canals.

3. Results

Firstly, a pressure – time history at a point in the SV, predicted by the new 1D model was compared with the results of 3D model. This comparison is shown in Fig. 3. In this case, the discharge coefficient of the helicotrema in the 1D model was adjusted to obtain similar results as in the 3D case. It can be seen that the curves are similar but the pressure amplitudes in the 1D model are bigger than in the 3D model. This may be due to the stiff BM in the 1D model and the fact that 3D effects at the OW are not taken into account in the 1D model.

After the model was validated, the influence of the RW impedance on the pressure distribution in the cochlea was studied. In this case, in order to eliminate too many input coefficients, the effect of orifice at the helicotrema was omitted. Thus, the fluid was able to flow through the helicotrema without any pressure losses. Figure 4 presents the pressure – time history along the SV and the ST at non – refection boundary condition at the RW ($\mathrm{RW}_{\mathrm{impedance}} = 1.3\mathrm{E}6$ $(\mathrm{Pa}\cdot\mathrm{s})/\mathrm{m}$). It can be seen that the wave propagates through the SV and passes to the ST. As there is no reflection of the pressure wave at the RW, the amplitude of the wave does not change at the RW. The maximum pressure difference between the SV and the ST in this case is approximately 25 Pa. If the RW is stiffer a compression reflection wave will occur, but if the RW is more elastic

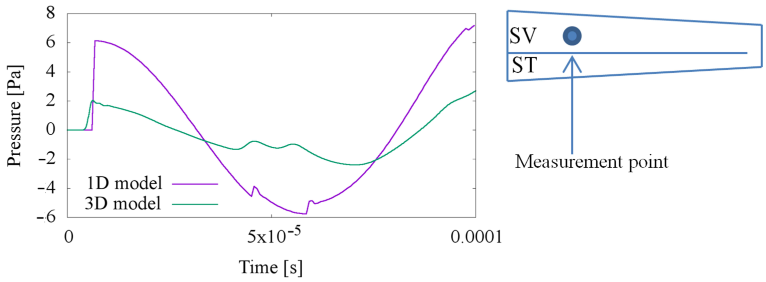

Fig. 3. Pressure – time history in 1D and in 3D model at a point in the SV offset by 10 mm from the BM base.

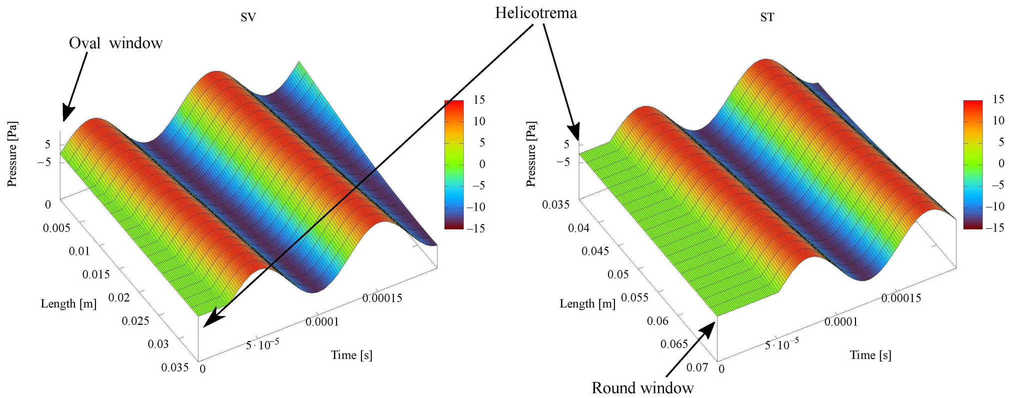

Fig. 4. Pressure wave in the SV and ST. The impedance of the RW is $R = 1.3\text{E}6$ $(\text{Pa}\cdot\text{s})/\text{m}$, the discharge coefficient of the helicotrema was chosen in such a way as not to give any pressure drop.

a decompression reflection wave will be present. A summary of results for the reflection and non – reflection boundary condition is presented in Table 1. As can be seen, with stiffening of the RW (higher value of the RW impedance) the pressure difference increases, thus it may lead to a stronger excitation of the BM.

Table 1. Summary of the results. Maximum pressure difference in reference to the RW impedance.

Impedance of the RW, R $[(\text{Pa}\cdot\text{s})/\text{m}]$	4.6E5 (reflection decompression wave)	1.3E6 (non-reflection)	1.3E7 (reflection – compression wave)
Maximum pressure difference between SV and ST, d_p [Pa]	20	25	40

4. Conclusions

The 1D model of the human cochlea has been built. Although the model contains many simplifications it is able to give results similar to 3D FSI model. The calculation time of one cycle of the excitation wave in 3D model takes around one week, while in 1D model it takes few seconds.

This 1D model may be used to explore the influence of the cochlea mechanical and geometrical parameters on its macro – mechanics. This may allow for a fast assessment of hearing loss and good results can be obtained using new hearing recovery methods before physical testing.

In the future, we plan to study a 1D fully coupled model with the BM represented by a shell structure.

Acknowledgment

This research was supported by the National Centre for Research and Developement PBS3/B7/25/2015.

References

1. Kwacz M., Marek P., Borkowski P., Mrówka M., *A three-dimensional finite element model of round window membrane vibration before and after stapedotomy surgery*, Biomechanics and Modeling in Mechanobiology, **12**(6): 1243–1261, 2013, doi: 10.1007/s10237-013-0479-y.

2. Gan R.Z., Reeves B.P., Wang X., *Modeling of sound transmission from ear canal to cochlea*, Annals of Biomedical Engineering, **35**(12): 2180–2195, 2007, doi:10.1007/s10439-007-9366-y.

3. Grey H., Williams P.L., Bannister L.H., *Grey's anatomy: the anatomical basis of medicine and surgery*, Williams P.L., (chairman of the editorial board), 38 th ed., Churchill Livingstone, New York 1995.

4. von Békésy G., Wever E.G., *Experiments in hearing*, McGraw-Hill, New York, 1960.

5. Nakajima H.H., Dong W., Olson E.S., Merchant S.N., Ravicz M.E., Rosowski J.J., *Differential intracochlear sound pressure measurements in normal human temporal bones*, JARO – Journal of the Association for Research in Otolaryngology, **10**(1): 23–36, 2009, doi: 10.1007/s10162-008-0150-y.

6. Kwacz M., Marek P., Borkowski P., Mrówka M., *A three-dimensional finite element model of round window membrane vibration before and after stapedotomy surgery*, Biomechanics and Modeling in Mechanobiology, **12**(6): 1243–1261, 2013, doi: 10.1007/s10237-013-0479-y

7. Kwacz M., Mrówka M., Wysocki J., *Differences in the perilymph fluid stimulation before and after experimental stapedotomy*, Acta of Bioengineering and Biomechanics, **14**(2): 67–73, 2012, doi: 10.5277/abb120209.

8. Mistrík P., Mullaley C., Mammano F., Ashmore J., *Three-dimensional current flow in a large-scale model of the cochlea and the mechanism of amplification of sound*, Journal of The Royal Society Interface, **6**(32): 279–91, 2009, doi: 10.1098/rsif.2008.0201.

9. Lyon R.F., Mead C., *Cochlear hydrodynamics demystified*, Caltech Computer Science Department Technical Report, Caltech-CS-TR-88-4, Caltech, Pasadena 1988.

10. Suesserman M.F., Spelman F.A., *Lumped-parameter model for in vivo cochlear stimulation*, IEEE Transactions on Biomedical Engineering, **40**(3): 237–245, 1993, doi: 10.1109/10.216407.

11. Vanpoucke F.J., Zarowski A.J., Peeters S.A., *Identification of the impedance model of an implanted cochlear prosthesis from intracochlear potential measurements*, IEEE Transactions on Biomedical Engineering, **51**(12): 2174–2183, 2004, doi: 10.1109/TBME.2004.836518.

12. de Boer E., van Bienema E., *Solving cochlear mechanics problems with higher-order differential equations*, Journal of the Acoustical Society of America, **72**(5): 1427–1434, 1982, doi: 10.1121/1.388675.

13. Diependaal R., *Nonlinear and active cochlear models: Analysis and solution methods*, PhD Thesis, TU-Delft, Netherlands, 1988.

14. Lim K.M., Steele C.R., *A three-dimensional nonlinear active cochlear model analyzed by the WKB-numeric method*, Hearing Research, **170**(1–2): 190–205, 2002, doi: 10.1016/S0378-5955(02)00491-4.

15. Kwacz M., Marek P., Borkowski P., Gambin W., *Effect of different stapes prostheses on the passive vibration of the basilar membrane*, Hearing Research, **310**: 13–26, 2014, doi: 0.1016/j.heares.2014.01.004.

16. Wada H., Metoki T., Kobayashi T., *Analysis of dynamic behavior of human middle ear using a finite-element method*, Journal of the Acoustical Society of America, **92**(6): 3157–3168, 1992, doi: 10.1121/1.404211.

Received November 3, 2016; accepted version November 12, 2016.