Hydrodynamics of Otoacoustic Emissions

Taha Jaffer[1,2] and Hans Kunov[1,2]
1. Institute of Biomaterial and Biomedical Engineering, University of Toronto, Toronto, Ontario, Canada. M5S 3G9.
2. Department of Electrical and Computer Engineering, University of Toronto, Toronto, Ontario, Canada. M5S 3G4.

1. Introduction

The discovery of Otoacoustic Emissions (OAE's) by David Kemp in 1978 has greatly aided the study of the mammalian auditory system. OAE's are sounds that are emitted from the cochlea and have been related to hearing function. OAE diagnosis is objective and non-invasive, however, the diagnostic value of OAE's can only be appreciated if the underlying physiology of the cochlea is properly understood.

Classical cochlear models suffer from an inability to realistically simulate OAE's because of the simplified treatment of the hydrodynamics at the interface between the oval window, stapes, and cochlear fluid. In the classical model, the cochlea is considered a tapered fluid-filled canal, split into two chambers by an elastic partition. There are two boundary conditions in this model. The first relates to a hole at the apex of the partition, known as the helicotrema, whose function is to null the trans-chamber pressure. The second describes the fluid displacement by the stapes at the base of the canal in response to sound entering the ear. However, when the sound is removed, the continued movement of the cochlear partition causes volume displacement in the cochlear canal, forcing fluid flow, and creating OAE's. The purpose of this present work is to mathematically describe this added hydrodynamic mechanism.

2. Classical Cochlear Hydrodynamics

The fluids within the cochlea are considered mechanically similar, acting in a linear, lossless, and incompressible fashion. The cochlear fluid flows in one direction, parallel to the Cochlear Partition (CP). Following the treatment given by Dallos (1973), the cochlear hydrodynamics are described by the following relation,

$$2\frac{\rho b(x)}{S(x)}\frac{\partial^2 z}{\partial t^2}=\frac{\partial^2 P}{\partial x^2} \qquad [1]$$

where $z(x,t)$ is the vertical displacement of the CP, $P(x,t)$ is the fluid pressure difference between the scala tympani and vestibuli, $b(x)$ is the width of the CP, and $S(x)$ is the cross sectional area of the canals.

One-dimensional cochlear models are constrained by two boundary conditions. The first describes the pressure variation at the helicotrema, a tiny opening at the apex of the CP. The second boundary condition describes the fluid displacement by the stapes at the base of the canal in response to sound entering the ear. Depending on the model of the middle ear used, this condition can vary. Generally, the boundary condition is taken to be

$$\frac{\partial P}{\partial x}(0,t)=G_{ME}(t)P_s(t) \qquad [2]$$

where $G_{ME}(t)$ describes the middle ear response and $P_s(t)$ describes the sound stimulus pressure.

The basal boundary condition [2] does not allow for reverse transmission after the presentation of sound stimuli. If the applied pressure in the ear canal is zero, that is, $P_S(t)=0$, then the fluid velocity at the oval window is also zero. Thus, the original boundary condition cannot reproduce this behavior and is incomplete and needs to be revised.

3. The Revised Basal Boundary Condition

The purpose of this derivation is to describe the boundary condition at the oval window with reference to the fluid flow in the entire cochlea. The CP is not perfectly compliant, it does not follow the applied sound pressure exactly because it has inertia and viscoelasticity. When the applied pressure is removed, the motion of the CP should continue, causing fluid volume displacements in the cochlear canal, thus, driving the oval and round windows. As the partition comes to rest after stimulation, only then will the reverse transmission cease. To incorporate the dynamics suggested, the original boundary condition at the base of the CP needs to be revised.

The actual fluid pressures seen in the base of the scala vestibuli is sum of the driving stapes and the pressure induced by the fluid flow in the scala vestibuli caused by CP motion. Using Newton's Second Law of Motion, and assuming that the cochlear fluids behave incompressibly, the fluid flow at the base of the CP can be described by,

$$\frac{\partial P}{\partial x}(0,t)=G_{ME}(t)P_s(t)-\frac{\partial P}{\partial x}(+0,t) \qquad [3]$$

where $+0$ is a point on the CP marginally apical to the base. It remains to determine the contribution of the CP motion to the fluid flow seen at its base. The classical model ignores this added contribution.

4. Basal Effect of CP Motion

The effect of CP movement without the sound stimulation can be deduced from [1]. Consider integrating [1] along x, from +0 to L_c, which amounts to summing the volume displacements induced by every portion of the CP. The result is

$$\frac{\partial P}{\partial x}(L_c,t)-\frac{\partial P}{\partial x}(+0,t)=2\delta_{cp}(t) \qquad [4]$$

where

$$\delta_{cp}(t)=\int_{+0}^{L_c}\frac{\rho b(x)}{S(x)}\frac{\partial^2 z}{\partial t^2}dx \qquad [5]$$

With the knowledge that fluid flow through the helicotrema is negligible, the revised basal boundary condition can be obtained by substituting [4] into [3], and introducing a reverse inefficiency parameter $G_{rv}(x,t)$. The result is the revised CP basal boundary condition,

$$\frac{\partial P}{\partial x}(0,t) = G_{ME}(t)P_s(t) + G_{rv}(x,t)\ddot{\xi}_{cp}(t) \qquad [6]$$

Note, the derivation reduces to the well-known classical basal boundary condition [2], when the flows induced by the movement of the CP are not taken into account. In addition, after the stimulus pressure is turned off, the movement of the partition still influences the overall dynamics and sends acoustic energy back out into the ear canal to be recorded by the microphone as an OAE.

5. Simulation of TEOAE's

To test the efficacy of this revised basal boundary condition [6] a preliminary simulation was constructed. The simulations were carried out using a finite difference approach in a MATLAB computing environment.

Following Viergever (1980), cochlear parameters of mass, viscoelasticity, and stiffness were selected to best represent a traveling wave on the CP. A simple lossy lever model of the middle ear was employed based on a description given by de Boer (1980). The reverse transmission inefficiency was taken to be a constant and estimated based on the tone-burst response from the model cochlea. The simulation was run for 1 kHz, 1.5 kHz, and 3 kHz tone bursts of duration 50ms with amplitudes 44 dB (SPL), 44 dB (SPL), and 40 dB (SPL). After the presentation of the stimuli all stimulation was turned off and the resulting oval window movement computed. The results are shown in Figure 2 and corresponding live observations are shown in Figure 1. The results demonstrate that the responses obtained in simulation closely resemble that recorded in patients both in frequency content and time evolution as seen in Geisler (1998). The "echo" like behavior is apparent.

6. Summary

In this paper the basal boundary condition for the classical one-dimensional cochlear model was reconsidered. A revision of the basal boundary condition was derives by arguing that the movement of the CP introduces additional fluid volume displacements to the stapedial pumping of the oval window. Simulations incorporating this additional dynamic were successful in producing output very similar to current Transiently Evoked OAE's using tonal bursts. Future work will explore the applicability of this boundary condition to other classes of OAE's including the Spontaneous, Stimulus Frequency, and Distortion Product families.

8. Acknowledgments

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7. References

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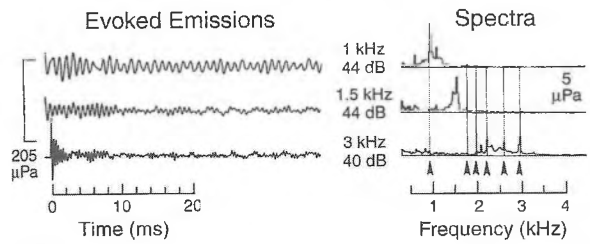

Figure 1. Time evolution and spectra of typical Transiently Evoked OAE from a normal human ear. Adapted from Geisler (1998), page 161, with permission from Elsevier.

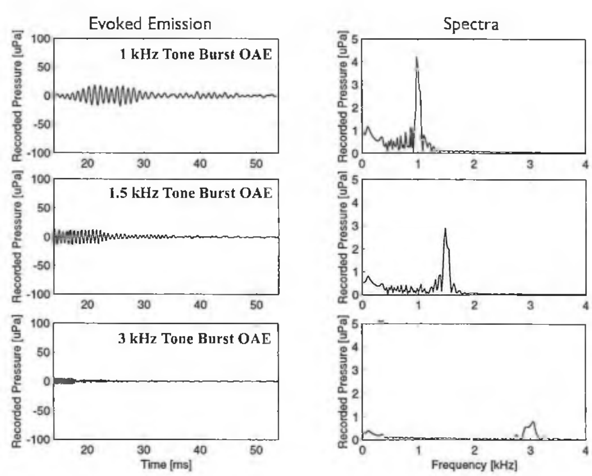

Figure 2. Simulated time evolution and spectra of Transiently Evoked OAE in a model of the human ear with the revised basal boundary condition. The similarity between Figure 1 and Figure 2 are striking.