BIOCYBERNETICS AND BIOMEDICAL ENGINEERING 34 (2014) 206–219

Available online at www.sciencedirect.com

ScienceDirect

journal homepage: www.elsevier.com/locate/bbe

Original Research Article

An integrated electromechanical model for the cochlear microphonic

Mohammad Ayat *, *Paul D. Teal*, *Mark McGuinness*

Victoria University of Wellington, Wellington, New Zealand

ARTICLE INFO

Article history:
Received 4 February 2014
Received in revised form
9 June 2014
Accepted 21 June 2014
Available online 5 July 2014

Keywords:
Cochlear microphonic
Electrophysiology
Outer hair cells
Cochlear modelling

ABSTRACT

The cochlear microphonic (CM) is an electrical signal generated inside the cochlea in response to sound. This electrical signal reflects mechanical activity in the cochlea and the excitation processes involved in its generation. However, the difficulty of obtaining this signal and the simplicity of obtaining other signals such as otoacoustic emissions have discouraged the use of the cochlear microphonic as a tool for studying cochlear functions. In this article, a model of the cochlea is presented which integrates both mechanical and electrical aspects, enabling the interaction between them to be investigated. The resulting model is then used to observe the effect of the cochlear amplifier on the CM. The results indicate that while the cochlear amplifier significantly amplifies the basilar membrane displacement, the effect on the CM is less significant. Both of these outcomes are consistent with previous physiological findings.

Moreover, the close match between mechanical and electrical predictions of the model and experimental measurements validates the model, and suggests that further investigations using the model into various pathologies and anomalies are warranted.

1. Introduction

The cochlear microphonic (CM) is an important product of electrical activities in the cochlea and can be employed in clinical practice and auditory research [1]. However, despite the importance of the CM, difficulty in obtaining this very small signal and uncertainty in its interpretation have meant that it is rarely used as an indication of cochlear performance, even after more than eighty years since its discovery [2,3].

The cochlea includes electrical and mechanical parts which interact bidirectionally to convert vibrations to neural stimulations.

For investigating the properties and effects of the electrical activities of the cochlea, it can be considered as an electrical network of biological resistances, capacitances, voltage and current sources or simply an electrical model. The *Battery and variable resistance* model by Davis [4] is an initial attempt to model the electrical network properties and the distribution of potentials in the cochlea. Strelioff [5] suggested a network model of resistors and batteries to simulate the generation and distribution of the cochlear potentials. In [6–8], different cochlear models have been proposed which integrates the electrical, mechanical and acoustical elements of the cochlea. In a modified version of the model of Neely and Kim [9], simplified electrical components without any longitudinal

* *Corresponding author* at: Victoria University of Wellington, P.O. Box 600, Wellington 6140, New Zealand.
E-mail address: mohamad.ayat@gmail.com (M. Ayat).
http://dx.doi.org/10.1016/j.bbe.2014.06.001
0208-5216/

coupling also has been considered [10]. However assessing electrical activities inside the cochlea is not the main focus of these models. In this paper, an integrated detailed model of the electrical and mechanical properties of the cochlea is presented including nonlinearity and longitudinal electrical coupling.

2. Physiological background

The basilar membrane (BM) displacement deflects the stereocilia and activates the hair cells. Hair cells convert mechanical to electrochemical activity in a process which is called *mechanoelectrical transduction* (*MET*) [11]. Deflection of the stereocilia opens and closes pores known as MET channels. Due to the voltage difference between the endolymph and the intracellular potential, the opening of the MET channels causes an inflow of ions (K^+), comprising a *transduction current*. Accordingly, the auditory neurons are stimulated by the inner hair cells (IHCs) and mechanical force is generated by the outer hair cells. The K^+ then is actively pumped back to the scala media via the spiral ligament and stria vascularis [12]. Therefore, electrical activities in mammalian cochleae are a significant part of the auditory sensing process.

3. Modelling approach

In the following, an integrated detailed model of the electrical and mechanical properties of the cochlea is presented. This integration allows bidirectional interaction between the mechanical and electrical aspects of the model. None of the components and configurations of the proposed model are original but these configurations have never been integrated for investigating the cochlear microphonic. The model of this paper is based primevally on that of Liu and Neely [10]. We proceed now to briefly describe the model, and then highlight the modifications we have made in order to model the CM.

For the integrity of our current work we repeat some of the equations and definitions from Liu and Neely [13,10].

3.1. *The model of Liu and Neely [10]*

Liu and Neely's model is summarised in Fig. 1. Parameters $\{M_d, R_d, K_d\}$ represent the sound source (which is here assumed to be the diaphragm of an earphone). The dynamics of the diaphragm are described by:

$$M_d\dot{v}_d = f(t) - K_d x_d - R_d v_d + P_d A_d \tag{1}$$

where $\dot{v}_d$, v_d and x_d[1] denote the acceleration, velocity and displacement of the earphone diaphragm, respectively. P_d is the pressure in the enclosed space of the ear canal, and A_d is the area of the earphone diaphragm. $f(t)$ is the stimulus force on the earphone diaphragm.

The earphone diaphragm and the eardrum are coupled to each other via the enclosed air filled ear canal between them. This air filled area can be modelled by an acoustic compliance K_c, resulting in:

$$P_d = K_c(x_d A_d - x_m A_e) \tag{2}$$

where x_m is the displacement of the malleus (equal to the displacement of the eardrum. See Fig. 1), and A_e is the area of the eardrum.

3.1.1. *Model of the middle ear*

The eardrum–malleus–incus system is modelled by parameters $\{M_m, R_m, K_m\}$. The malleus–incus lever ratio is $g \leq 1$. The joint between the incus and the stapes is modelled by parameters $\{R_i, K_i\}$. The stapes and its surrounding structures are modelled by parameters $\{M_s, R_s, K_s\}$. P_d and $P(0)$[2] are coupled to each other via the following equations:

$$M_m\dot{v}_m = -K_m x_m - R_m v_m + g f_i + P_d A_e \tag{3a}$$

$$(M_s + M_r)\dot{v}_s = -(K_s + K_r)x_s - (R_s + R_r)v_s - f_i - P(0)A_s \tag{3b}$$

$$f_i = K_i(x_s - g x_m) + R_i(v_s - g v_m) \tag{3c}$$

A_s is the effective area of the stapes footplate; x_s and v_s denote the displacement and the velocity of the stapes. Parameters $\{M_r, R_r, K_r\}$ represent the round window (see part (B) of Fig. 1).

3.1.2. *Cochlear macromechanics*

The Navier–Stokes equation governs the motion of an incompressible Newtonian fluid [14, Chapter 2]. The fluid inside the cochlea chambers is usually assumed to be incompressible and inviscid and gravity effects are ignored. Accordingly, the Navier–Stokes equation in Cartesian coordinate, is simplified to [15,16, Chapter 20]:

$$\rho\frac{\partial u}{\partial t} + \nabla P = 0 \tag{4}$$

where vector P is pressure, u is the fluid velocity and ρ is the density of the fluid. The incompressibility assumption also indicates that the divergence of the fluid velocity is zero [17, Chapter 5].

$$\nabla \cdot u = 0 \tag{5}$$

Eqs. (5) and (4) can be simplified to one dimension. By neglecting dependency on the y direction (see Fig. 1), Eq. (5) can be simplified to

$$\partial_x u_x + \partial_z u_z = 0 \tag{6}$$

where x denotes the longitudinal direction from base to apex, and z denotes the vertical direction. u_x and u_z are components of the fluid velocity u in these directions.

Assuming that the fluid velocity linearly changes from a maximum value at $z = 0$ to zero at $z = H$ and the height of channel (H) is assumed to be constant along and across the width of the BM (see Fig. 1(C)), u_z can be written as [17, Chapter 5]:

$$u_z = \frac{\dot{\xi}_r(x)(H - z)}{H}$$

[1] Henceforth, when x denotes displacement, v and $\dot{v}$ will denote velocity and acceleration, respectively.

[2] $P(0)$ is the pressure at the stapes which is equal to the cochlea fluid pressure at the oval window.