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Modelling the Generation Of the Cochlear Microphonic

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***Abstract*— The cochlear microphonic (CM) is one of the electrical signals generated by the human ear in response to sound stimulus. Difficulty in recording this signal and inadequate understanding of its origin have restricted its use for human auditory research. Modelling can help to improve our understanding of this signal. In this paper, an electromechanical model for the generation of the cochlear microphonic is proposed. The results of the model can also explain discrepancies between the basilar membrane and CM tuning curves.**

I. INTRODUCTION

The cochlear microphonic (CM) is a by-product of cochlear activities in response to a sound stimulus and is mainly generated by the outer hair cells (OHCs) [1]. Malfunction of the OHCs causes audiometric (sensitivity) hearing loss [2], therefore, the CM may provide a valuable tool for objective audiometric tests, especially for non-cooperative patients such as newborn babies [3]. This information can supplement the information that is available from Otoacoustic Emissions (OAEs) [4]. There is much information available about individual parts of the human hearing system including the OHCs and electrical properties of the organ of Corti. However in order to improve understanding of the hearing process, modelling techniques can be used.

Despite the passing of more than eighty years from its discovery, the CM is rarely interpreted or used as indicator of cochlear performance [5]. Modelling can help to extend our knowledge of the CM and its interpretation. Even though modelling mechanical parts of the cochlea has been of great interest, modelling electrical parts of the cochlea has not received much attention. In most of the recent cochlear models the electrical parts of the model are ignored or observed as black boxes which provide local positive feedback to the mechanical parts [6]. In some models, electrical parts are modelled separately without considering mechanical connections [7] or electrical activities of the model are not addressed [8].

The next two paragraphs briefly outline the process by which the CM is generated:

Basilar membrane displacement resulting from sound stimulus deflects stereocilia on hair cells and activates the cells. Hair cells convert mechanical to electrochemical activity through the *transduction process*. Deflection of the stereocilia opens and closes pores known as *mechanoelectrical transduction* (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, comprising a *transduction current*. Since these channels are embedded in the electrical network of biological resistances and capacitances of the organ of Corti, changing their currents also produces changes in extracellular current flow and creates the CM [5].

The *transduction current* causes depolarization and hyperpolarization of the OHCs' membrane. Because of the motor protein of the OHC (the somatic motor which relies on prestin molecule) the OHC length changes, creating an active force on the basilar membrane. This process is referred to as OHC electromotility. There is also evidence of force resulting from hair bundle motility [9].

Based on anatomical and physiological observation of the cochlea a complete electromechanical model for the cochlea is proposed in this paper. In the proposed model physical connections are considered and recent measured parameter values are used [10]. The results of the model are then used to investigate the difference in broadness between tuning curves of the basilar membrane and the CM.

The rest of paper is arranged as follows: a detailed model of the cochlea is described in Section II. The methodology used to analyse the proposed model is outlined in Section III. The results are shown and explained in Section IV. Finally, V is the conclusion.

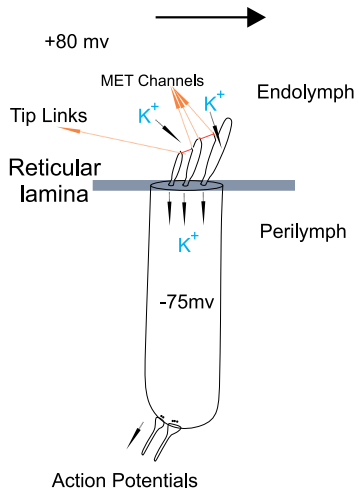

Fig. 1. The operation of hair cells. Deflection of the stereocilia opens and closes the MET channels which leads to ion flow, driven by the voltage difference between the endolymph and the intracellular potential.

II. MODELLING

A. Mechanical Components of the Model

The mammalian cochlear function initiates at the stapes which directs energy from the eardrum to the oval window.

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The movement of the oval window causes waves to travel through the fluid from the base toward the apex. The cochlear fluid is practically incompressible, so the wave is propagated by movement of the fluid and the basilar membrane. The round window moves in the opposite direction to make room for the incompressible fluid. Different stiffnesses along the cochlea cause natural or passive tuning of the basilar membrane.

Vibrations of the basilar membrane are transformed to a shearing movement between the reticular lamina and tectorial membrane causing the stereocilia of the OHCs to deflect resulting in activating the cochlear amplifier which makes the basilar membrane sharply tuned [6, 11]. High frequency stimulus cause vibration near the base and low frequency stimuli cause vibration near the apex due to the cochlea's tonotopic property; in other words, as the wave propagates down the cochlea, the stiffness decreases and the wave comes at a point identified as the *best place* for that input frequency. At this point the membrane will vibrate with maximum amplitude. Beyond that point the basilar membrane becomes less stiff and highly damped so the wave energy dissipates rapidly.

For modelling cochlear macromechanics and micromechanics, the well-known model of [12], and the later version [13] have been used.

B. Electrical Components of the model

Deflecting the stereocilia of the OHCs changes the MET channel currents. Since these channels are embedded in the electrical network of the organ of Corti, changing their currents also produces changes in extracellular current flow and creates the CM [5].

The MET currents affect the membrane potentials of the OHC. These potential changes cause a change of the OHC length which provides an active force on the basilar membrane and tectorial membrane. In other words, these potential changes activate the cochlear amplifier.

The simple and widely used model of the generation of the CM is the *battery and variable resistance model* by Davis [14]. In this model, the resting potentials of each radial section the organ of Corti have been modelled by two batteries. The primary battery is in the hair cells, the accessory battery is in the stria vascularis, and the MET channels are modelled by variable electrical resistors [9]. Accordingly, the current through the hair cells is modulated by changing electrical resistances resulting from cilia deflection. These result in electrical potential changes which comprise the CM.

This simple model was extended later by adding longitudinal resistor and biological capacitors [7, 15–18]. Linearization of the variable resistors results in the current sources included in the model which is shown in Fig. 2.

III. METHODOLOGY

Fig. 3 shows one section of the micromechanical model. Each section of the micromechanical model consists of two parts representing the mass, stiffness and damping of the basilar membrane and OHC mechanical load respectively.

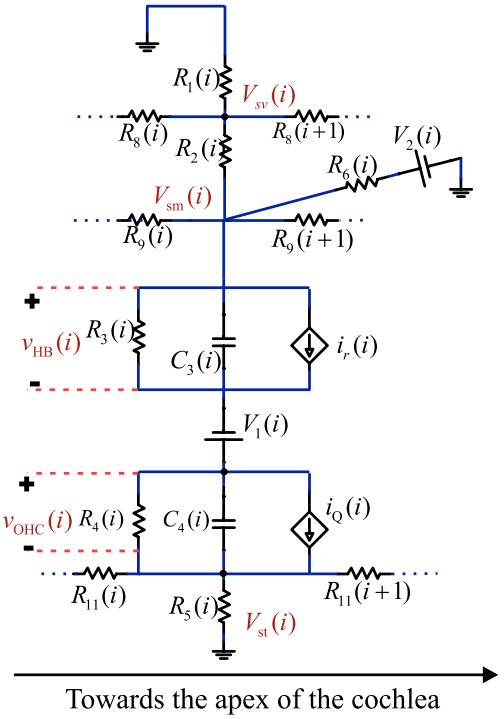

Fig. 2. Electrical lumped model of a radial section of the organ of Corti. R_8, R_9 and R_{11} represent the resistances along the scalae vestibuli, media and tympani respectively. R_1, R_5 and R_6 represent resistance between each of these scalae and the surrounding spiral ligament which is considered to be ground (0 V) in this model. R_3 and C_3 represent the apical resistance and capacitance. R_4 and C_4 represent basolateral resistance and capacitance.

The force f_{OHC} is induced by OHC electromotility and pressure (P) is induced by the cochlear fluid.

By applying Newton's second law to each section of micromechanical model, the following equations are derived:

$$f_{\text{OHC}} = M\ddot{\xi}_{\text{o}} + R\dot{\xi}_{\text{o}} + K\xi_{\text{o}} \tag{1}$$

$$m\ddot{\xi}_b + r\dot{\xi}_b + k\xi_b = -\frac{P}{w\Delta x} \tag{2}$$

$$\xi_b = \xi_{\text{r}} + \xi_o \tag{3}$$

where w is the width of the basilar membrane and Δx is the length of a cochlear section (see [13, 19] for more details of the other parameters).

For describing the electrical network of the organ of Corti, nodal equations can be written for the circuit in Fig. 2 and equations related to v_{OHC} and v_{HB} can be extracted. $i_{\text{Q}} = \dot{\xi}_{\text{o}}/T$ (where T is the piezoelectric transformer ratio.) and i_{r} can be written as a nonlinear antisymmetric saturating function of displacement and velocity of the reticular lamina [13].

To summarise, displacement and velocity of both reticular lamina and OHC together with OHC and hair bundle voltages in each section of the organ of Corti form a state vector as follows:

$$x_i = [\xi_{\text{r}}, \dot{\xi}_{\text{r}}, \xi_{\text{o}}, \dot{\xi}_{\text{o}}, v_{\text{OHC}}, v_{\text{HB}}]^{\text{T}} \tag{4}$$

i indexes the discrete section section of the cochlea. The complete system can be represented as one n-dimensional

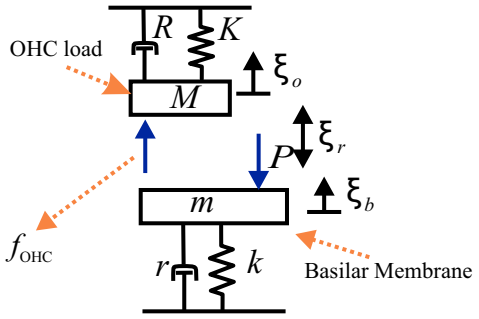

Fig. 3. ith part of micromechanical model of [13]. ξ_b, ξ_r and ξ_o represent the displacement of the basilar membrane, reticular lamina and OHC in ith section respectively. P is the pressure difference between scala vestibuli and scala tympani and f_{ohc} is the active force that is generated by the OHC and by which the electrical parts of the model can be linked to the mechanical part of the model. M, R and K are mass, resistance and stiffness of the OHC load impedance. m, r and k are mass, resistance and stiffness of the basilar membrane

first order vector differential equation:

$$\dot{x} = f(t, x, u) \tag{5}$$

$$y = Cx \tag{6}$$

where u is input to the cochlea and the C matrix selects the desired state variables. For time domain analysis, these ordinary differential equations can be solved numerically using any conventional method such as the ode45 solver in MATLAB.

By considering a linearized version of the MET channel current (i_r) [13], the system can also be analysed in the frequency domain.

IV. RESULTS

A. Responses to Stimulus (Frequency responses)

Fig. 4 represents the outputs of the linearized model for the basilar membrane velocity for different stimulus frequencies. As can be seen from this figure, high frequency stimulus causes vibration near the base and low frequency stimulus causes vibration near the apex. In addition, the tuning curves of the basilar membrane velocity are sharply tuned, both of which agree with experimental recordings [20].

B. Cochlear Microphonic

The CM is usually recorded from the round window for clinical purposes using a transtympanic membrane electrode [21] or glass micropipet electrodes in the scala media [22] for research purposes, and can be observed as the potential of the endolymphatic space above the basal hair cells (V_{sm} in Fig. 2)[7].

Fig. 5 illustrates the amplitudes of these potential variables as a function of the cochlear length. The observable CM as shown in Fig. 5 (c) exhibits much broader tuning than the basilar membrane velocity. This also agrees with experimental results [22]. Despite the very similar amplitudes of v_{OHC} and v_{HB}, they have a phase differences of π near their peaks and nearly cancel each other (see Fig. 6). According to the model, a significant contributing factor to the broadness of the CM tuning curves is the near π phase difference between v_{OHC} and v_{HB} near their peaks.

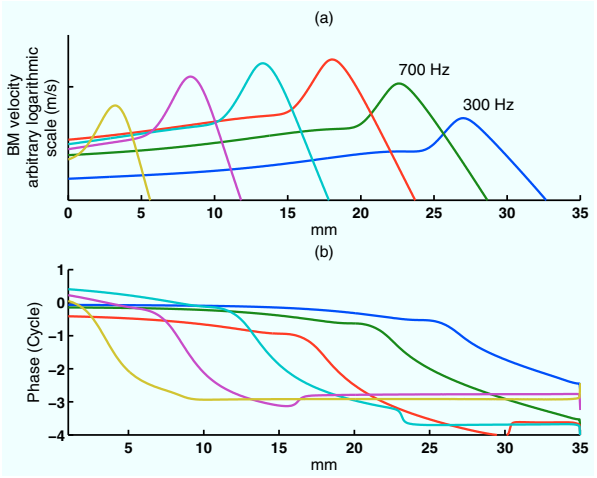

Fig. 4. Basilar membrane (BM) velocity, as a function of the cochlear length. (a) shows magnitude and (b) shows phase of the basilar membrane velocity for four different stimulus frequencies: 12000, 5900, 2900, 1400, 700 and 300 Hz.

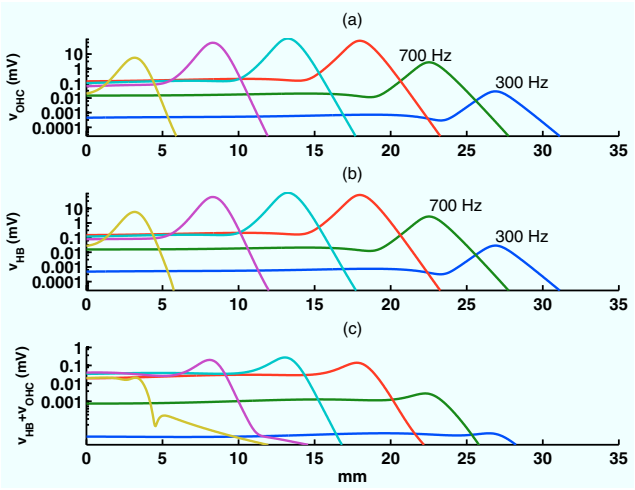

Fig. 5. Amplitudes of these potential variables as a function of the cochlear length. (a), (b) and (c) show amplitudes of v_{OHC}, v_{HB} and $v_{OHC} + v_{HB}$ frequencies: 12000, 5900, 2900, 1400, 700 and 300 Hz.

C. Responses to Stimulus (Time domain responses)

Time domain analysis has been performed on the model including nonlinearity (equation (5)). Fig. 7 shows results of a time domain analysis at three different locations along the cochlea for the voltage of the scala media.

Fig. 8 depicts v_{OHC} and v_{HB} at one location. These responses are in agreement with the frequency analysis showing that v_{OHC} and v_{HB} have nearly π phase difference.

V. CONCLUSION

In this paper an electromechanical model for the CM is proposed. The model integrates in simplified form aspects of

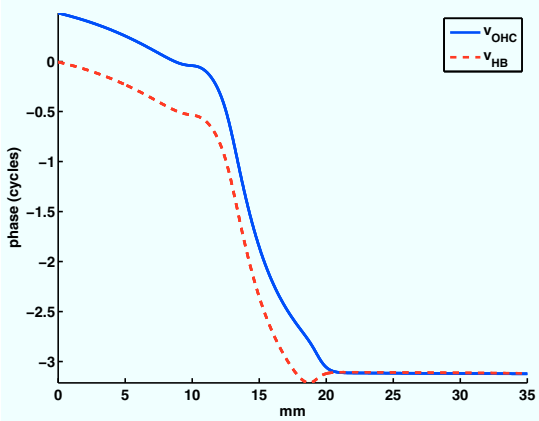

Fig. 6. Phase of v_{OHC} and v_{HB}. For clarity, only curves for the characteristic frequency of 2900 Hz have been shown.

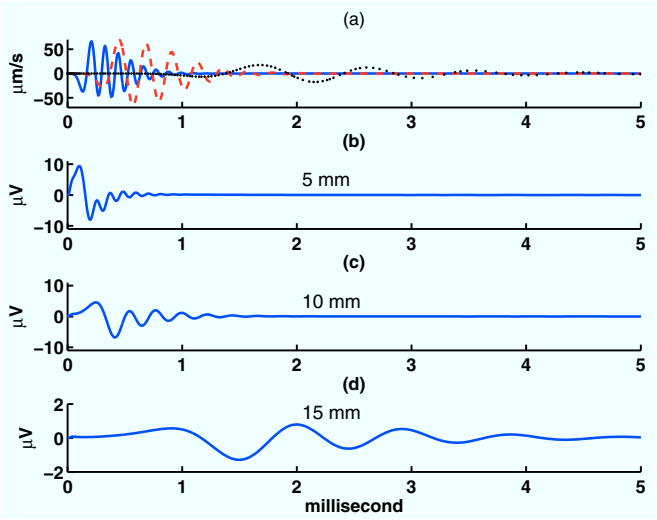

Fig. 7. (a) shows the basilar membrane velocities at locations 5 mm (solid line), 10 mm (dashed line) and 15 mm (dotted line) from the stapes. (b), (c) and (d) show voltage of the scala media (V_{sm}) at locations 5, 10 and 15 mm from the stapes respectively, in response to a step input.

cochlea function that have not previously been considered together. Outcomes of the model agree with experimental observations. By assessing the OHCs potentials, the results of the model also show a near π phase difference between v_{OHC} and v_{HB} which can be considered as a contributing factor to the difference in sharpness between the tuning curves of basilar membrane motion and the CM.

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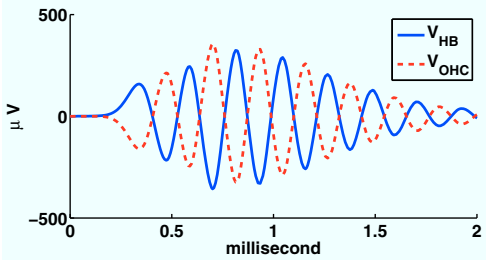

Fig. 8. v_{OHC} and v_{HB} at location 10 mm from the stapes, in response to a step input.

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