FINITE ELEMENT MODELLING OF HUMAN EAR CONSIDERING HYPERELASTIC AND VISCOELASTIC PROPERTIES OF SOFT TISSUES

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Soft tissues of the human ear usually exhibit viscoelastic and hyperelastic properties in the physiological function range. To accurately reflect these material properties, a three-dimensional (3D) finite element (FE) model of human ear consisting of the ear canal, middle ear, and cochlear is first constructed, and then the hyperelastic and viscoelastic properties of soft tissues in middle ear are taken into consideration to study the harmonic response to pure tone stimulus. The displacement response of the tympanic membrane umbo and the stapes footplate, the basilar membrane (BM) responses and the fluid pressure distribution in the cochlea are derived, and the FE model-derived results are compared with the published experimental data in the literatures. The results show that the FE results match well with the published data. An application of present model is to predict the dynamic responses of human ear to high intensity sound. This study provides an accurate dynamic model of human ear considering of hyperelastic and viscoelastic properties of soft tissues to analyse the characteristics of human ear using FE method.

Keywords: human ear, nonlinearity, soft tissues, FE modelling

1. Introduction

The human ear system consists of the pinna, external ear canal, middle ear and cochlea. Its function is to transfer the sound waves collected by the pinna to the cochlear via the middle ear, and then convert the vibration to the nerve impulses in the inner ear. Since the human ear system has complex geometry, non-homogeneous material properties and ultrastructural characteristics, the finite element (FE) method has been becoming a powerful tool to study the dynamic characteristics of human ear[1].

At present, many FE models of human ear have been reported. Many representative scholars, including Koike[2], Lee[3] and Nie[4] and so on, constructed the three-dimensional (3D) models of middle ear, and their cochlea models were simplified as springs and dashpots. Moreover, some scholars, like Gan[5], Zhang[6] and Yao[7], etc, modelled the cochlea as fluid-filled spiral or uncoiled structure in their studies. These FE models of human ear have been used to study the dynamic performance of human ear. However, in above models, biological tissues of the human ear like the tympanic (TM), suspensory ligaments and muscle tendons were assumed as linear elastic materials, or only one type of material properties (hyperelasticity or viscoelasticity) was considered. These assumptions didn't agree with the published experimental data[8-10], which experiments have indicated that soft tissues in middle ear have hyperelastic and viscoelastic material behaviours. So it’s necessary to establish a FE model of human ear considering hyperelastic and viscoelastic properties of soft tissues for studying the dynamic response of human ear.
In this paper, a FE model of human ear consisting of the ear canal, middle ear, and cochlear is firstly established. Both of hyperelastic and viscoelastic material properties of soft tissues are considered in this model. The validity of the model is conducted through comparing the dynamic responses and experimental data in the literature, and the calculated results show a good agreement with measurements. The present results and methods in this study will contribute to improving the accuracy of modelling human ear.

2. Methods

2.1 Finite element model of human ear

A human temporal bone extracted from the right ear of a 45-year-old man is firstly scanned using a micro-computer tomography scanner (GE Healthcare, eXplore Locus SP). A total of 882 serial images with a resolution of 43.5×43.5 μm/pixel are obtained. Then the software Simpleware (Simpleware Ltd, UK) is adopted to construct the geometric model of external ear canal and middle ear. And this geometric model is imported into the FE pre-processing software Hypermesh (Altair Engineering, Troy, MI) to be meshed. The air in the external ear canal is meshed by 22366 four-node tetrahedral elements. And a total of 48043 four-node tetrahedral elements are created to mesh three ossicular bones, the incudo-malleolar joint (IM-joint), the incudo-stapedial joint (IS-joint) and seven ligaments/tendons. In current FE model, the cochlea is simplified as an uncoiled, two-chambered and fluid-filled duct consisting of the oval window (OW), round window (RW), scala vestibuli, scala tympani, the BM and the supporting structure, which has been commonly accepted for the study of cochlear mechanics using FE model. The interfaces between the BM and the cochlea fluid are defined as fluid-structure interaction surfaces, and the surfaces of the cochlea fluid are set as the rigid walls (Neumann condition) which mean the normal pressure gradient of the cochlea fluid is zero. Meanwhile, boundary conditions for middle ear are set by fixing the displacements of the ligaments and tendons, tympanic membrane annulus (TMA) and the stapedial annular ligament (SAL) to the middle ear cavity bony wall.

Figure 1 shows the FE model of human ear consisting of the external ear canal, middle ear and the cochlea. In order to see the internal structure of the cochlea model, the fluid in the cochlea are displayed as partially transparent. And this model has been used to evaluate the performances of implantable hearing devices in our previous studies[11, 12].

![Figure 1: The FE model of human ear consisting of the external ear canal, middle ear and cochlea.](image-url)
2.2 Modelling of hyperelastic and viscoelastic materials

Recently, several experiments [8, 13] have shown that the TM and the soft tissues in the middle ear are hyperelastic. As for these hyperelastic properties, the Ogden model is suitable for describing the nonlinear hyperelastic behaviours of soft tissues in the middle ear [13]. According to the Ogden model, the strain energy function can be expressed as

\[ U = \frac{2\mu}{\alpha_i} \left( \lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3 \right) \]  

(1)

Where \( U \) is the strain energy, \( \lambda_1, \lambda_2, \lambda_3 \) are the three principle stretch ratios, \( \mu_i \) and \( \alpha_i \) are the constant hyperelastic parameters. Assuming that soft tissues are incompressible, the stress-strain relationship for uniaxial tension can be express as

\[ \sigma = \frac{2\mu_i}{\alpha_i} \left[ \lambda^{(\alpha_i-1)} - \lambda^{-0.5(\alpha_i+1)} \right] \]

(2)

Where \( \sigma \) is the stress, and \( \lambda \) is the stretch ratio.

Taking the derivative of Eq. (6) with the respect to \( \lambda \), the elastic modulus can be expressed as the function of \( \lambda \),

\[ E(\lambda) = \frac{d\sigma}{d\lambda} = \frac{2\mu_i}{\alpha_i} \left[ (\alpha_i-1)\lambda^{(\alpha_i-2)} + (0.5\alpha_i + 1)\lambda^{-0.5(\alpha_i+2)} \right] \]

(3)

The hyperelastic material parameters, derived from Cheng’s works [13], were listed in Table 1. Note that for the TM, the SAL and the incudo-stapedial joint, the values of \( \alpha_i \) are obtained from Cheng’s experiments while the values of \( \mu_i \) are derived using the cross-calibration method.

Table 1: Hyperelastic properties for soft tissues in the middle ear

<table>
<thead>
<tr>
<th>Component</th>
<th>( \mu_i ) (MPa)</th>
<th>( \alpha_i )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tympanic membrane</td>
<td>3.5</td>
<td>26.76</td>
</tr>
<tr>
<td>Superior mallear ligament</td>
<td>0.05</td>
<td>17.40</td>
</tr>
<tr>
<td>Lateral mallear ligament</td>
<td>0.05</td>
<td>17.40</td>
</tr>
<tr>
<td>Posterior incudal ligament</td>
<td>0.05</td>
<td>17.40</td>
</tr>
<tr>
<td>Anterior incudal ligament</td>
<td>0.078</td>
<td>13.69</td>
</tr>
<tr>
<td>Posterior mallear ligament</td>
<td>0.05</td>
<td>17.40</td>
</tr>
<tr>
<td>Superior incudal ligament</td>
<td>0.05</td>
<td>17.40</td>
</tr>
<tr>
<td>Stapedial annular ligament</td>
<td>0.01</td>
<td>23.52</td>
</tr>
<tr>
<td>Stapedial annular ligament</td>
<td>0.1</td>
<td>5.75</td>
</tr>
<tr>
<td>Incudo-stapedial joint</td>
<td>2.0</td>
<td>9.18</td>
</tr>
<tr>
<td>Incudo-malleolar joint</td>
<td>20</td>
<td>9.18</td>
</tr>
</tbody>
</table>

Meanwhile, the mechanical properties of soft tissues also show viscoelastic behaviours [8, 14]. In this study, seven soft tissues including the TMA, the TM-Pars tensa, the TM-pars flaccida, the IM-joint, the IS-joint, SAL and round window membrane (RWM) are modelled as viscoelastic properties. As for these viscoelastic properties, the first-order Wiechert model is appropriate to describe the viscoelastic behaviours of soft tissues in the middle ear [14]. The first-order Wiechert model can be expressed in time domain as

\[ E(t) = E_0 + E_1 e^{-\frac{t}{\tau_1}} \]

(4)

The Eq. (4) also can be written in terms of the frequency using Fourier transform as

\[ E(\omega) = E_0 + E_1 \frac{\omega^2 \tau_1^2}{1 + \omega^2 \tau_1^2} + iE_1 \frac{\omega \tau_1}{1 + \omega^2 \tau_1^2} \]

(5)

Where \( E_0, E_1, \) and \( \tau_1 \) are the viscoelastic parameters.
Note that under the condition of small strains, $E_0$ for hyperelastic materials can be estimated as\[15\]

$$E_0 = 3\mu_l$$  \hspace{1cm} (6)

Therefore, the Young’s modulus for the TM-Pars tensa, the TM-pars flaccida, the IS-joint, the IM-joint, SAL can be derived from Eq. (6). And the viscoelastic parameters of seven soft tissues are listed in Table 2. The parameters of $E_i$ are derived from the Zhang and Gan’s work\[14\].

<table>
<thead>
<tr>
<th>Table 2: Viscoelastic properties for soft tissues in the middle ear</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_0$ (MPa)</td>
</tr>
<tr>
<td>-----------</td>
</tr>
<tr>
<td>$E_i$ (MPa)</td>
</tr>
<tr>
<td>$\tau_i$ ($\mu s$)</td>
</tr>
</tbody>
</table>

3. Results

3.1 Middle ear functions

Figure 2 shows the FE model-derived middle ear function consisting of displacement responses of the TM umbo and the stapes footplate (FP) in comparison with the published data\[5, 16\]. The input sound pressure is 90 dB SPL at the TM. The figure indicates that the results derived from present FE model are closed to the experimental data\[16\] and the representative simulation data\[5\]. Although at higher frequencies above 4000 Hz, the displacement magnitude of FP is somewhat lower than experimental data, but they all have the same trend in the frequency range of 200–8000Hz. It is found that the displacement magnitudes of TM and FP are stable at frequencies below 1000 Hz, and they decrease as the frequency increases at frequencies above 1000 Hz. Meanwhile, the phase curve indicate that the phase degrees of the TM and FP decrease as the frequency increases. and the phase degree of FP is lower than that of the TM umbo.

![Figure 2: The FE model-derived displacement of the TM umbo and the FP in comparison with published data\[5, 16\]. (a) Magnitude; (b) Phase.](image)

3.2 Cochlea responses

In present FE model, the cochlea has been simplified as two-chambered and fluid-filled duct. The fluids in the scala vestibuli and scala tympanic are separated by the BM, and they meet at the Helicotrema at the apex of the cochlea. Figure 3 shows the BM displacements which have been normalized by the FP displacement at six locations corresponding to best frequencies of 400, 800, 1000, 2000, 4000 and 8000 Hz. It can be found clearly that every displacement-frequency curve shows a narrow peak, and the best frequency of BM decreases from the base to apex along the BM length. The BM movement is very sensitive to the sound frequency.
Figure 3: The displacement ratio of the BM to the FP ($d_{BM}/d_{FP}$).

Figure 4 shows the FE model-derived pressure responses of the scala vestibuli and scala tympani at six locations close to BM best frequencies of 400, 800, 1000, 2000, 4000 and 8000 Hz. The Fig. 4(a) indicates that the scala vestibuli pressure is relatively higher in the middle cochlea, and the maximum values occur at the frequency range of 800–1200 Hz. At lower frequencies below 400 Hz and higher frequencies above 5000 Hz, the scala vestibuli pressures hover around 1 Pa when a uniform pressure of 90 dB SPL is applied at the TM. These results are consistent with the report by Gan et al.[5]. Meanwhile, the pressure responses of scala tympanic are displayed in the Fig. 4(b). The results show that the pressure magnitudes of scala tympanic reach the maximum values at the frequency range of 800–1200 Hz and remain stable around the value of 1 Pa, which is similar to the pressure responses of scala vestibuli (seen in Fig. 4(a)).

Figure 4: Pressure responses of scala vestibuli and scala tympanic at six locations close to BM best frequencies of 400, 800, 1000, 2000, 4000 and 8000 Hz. (a) Scala vestibuli pressure; (b) Scala tympanic pressure.

The differential pressure between the scala vestibuli and the scala tympanic is the energy source that drives the movement of BM[17]. The transfer function, which is defined as the ratio of the differential pressure between the scala vestibuli ($P_{SV}$) and the scala tympanic ($P_{ST}$) to the input sound pressure at the TM ($P_{EC}$), is displayed in Fig. 5. In the current transfer function calculation, we extract the fluid pressure of scala vestibuli and scala tympanic near the oval window and round window, and the extracting location is close to that of Nakajima et al’s experiment[17]. The comparison between FE model-derived results and the Nakajima et al’s data[17] shows that the FE results match well with
the experimental data. In the frequency range below 1000 Hz, the transfer function magnitude increases as the frequency increases. And in the frequency range above 1000 Hz, the magnitude decreases as the frequency increases.

![Figure 5: The transfer function (|P_{SF} - P_{ST} / P_{EC}|) in comparison with published data[17].](image)

4. Discussion and conclusion

In this study, a 3D FE model of human ear is improved by considering the hyperelastic and viscoelastic properties of soft tissues. The Ogden model is adopted to describe the hyperelastic behaviours of the TM and the ligaments in the middle ear. Meanwhile, the first-order Wiechert model is used to describe the viscoelastic properties of the TM, the ossicular joints, the SAL and the RW. The validation of this nonlinear model is conducted by comparing the FE model-derived middle ear functions and cochlea responses with the experimental data. And the results show a good agreement with measurements. Taking into account of hyperelastic and viscoelastic properties helps to develop the accurate dynamic model of human ear.

This study is the first time to develop a FE model of human ear considering the hyperelastic and viscoelastic properties of soft tissues as far as we know. Compared with the previous linear model, this nonlinear model can be used to derive the dynamic responses of human ear exposed to high intensity sound, blast overpressure waves or middle ear pressure. The nonlinear mechanism of human ear system can be further studied in the future.

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