The influence of the mechanical behaviour of the middle ear ligaments: 

a finite element analysis

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The influence of the mechanical behaviour of the middle ear ligaments: a finite element analysis

Abstract: The interest in Computer Modelling, mainly by using the Finite Element Method (FEM), of biomechanical systems has been increasing, in particular, to analyze the mechanical behaviour of the human ear. In this work, a finite element model of the middle ear was developed to study the dynamic structural response to harmonic vibrations for distinct sound pressure levels applied on the eardrum. The model includes different ligaments and muscle tendons with elastic and hyperelastic behaviour for these supportive structures. Additionally, the non-linear behaviour of the ligaments and muscle tendons was investigated, as they are the connection between ossicles by contact formulation. Harmonic responses of the umbo and stapes footplate displacements, between 100 Hz and 10 kHz were obtained and compared with previous published works. The stress state of ligaments (superior, lateral and anterior of malleus and superior and posterior of incus) was analyzed with the focus on balance of the supportive structures of the middle ear, since ligaments make the link between the ossicular chain and the walls of the tympanic cavity. The results obtained in this work highlight the importance of using hyperelastic models to simulate the mechanical behaviour for the ligaments and tendons.

Keywords: Computer Modelling; Simulation; Finite Element Method; Biomechanics; Middle ear; Elastic; Hyperelastic.
1. Introduction

Some studies have been developed and published concerning the mechanical behaviour of the human middle ear, [1-4]. However, in order to obtain more realistic results, some improvements in the numerical simulation models are needed. To overcome the current limitations, a finite element model of the middle ear is described in the present work. Hence, this study starts with the construction of 3D solid models of the ossicles and eardrum from the imaging data of a normal ear. The discretization of these components is made using tetrahedral solid elements for the ossicles and hexahedral elements for the eardrum. The finite elements model also includes the ligaments, muscles and respective tendons and the cochlear fluid. For the ossicles and eardrum the mechanical properties used are from the current literature [1,2,4]. Additionally, the connection between the ossicles is carried out using contact formulation, which can be interpreted as a simulation of the capsular ligaments [5]. The numerical simulation was carried out with the commercial software ABAQUS [6].

As the most important audible frequency range is located between 125 Hz and 8 kHz (normally used in an audiogram), the dynamic study in this work focuses on the range between 100 Hz and 10 kHz. In addition, in terms of the umbo and stapes footplate displacements, a dynamic study based on a structural response to harmonic vibrations is presented and the results are compared with published works [1, 4, 7-10]. Normally, the works based on the Finite Element Method (FEM) use a linear elastic material model for the supportive structures of the middle ear [1-3, 11]. However, as has been recently shown by Cheng and Gan, the behaviour of ligaments and muscle tendons is non-linear [12]. From the mechanical point of view, this non-linearity can be treated with one hyperelastic model [13] for the anterior maleolar ligament, stapedius and tensor tympani tendons. In the present work, the hyperelastic constitutive model of Yeoh [14] is used for all ligaments and muscle tendons. Harmonic responses of the state of ligament stresses, with elastic and hyperelastic behaviour, are obtained and compared for a load of 20 Pa applied on the eardrum.

2. Methods

2.1. Geometric and finite element mesh
The geometry of the eardrum and ossicles (malleus, incus and stapes) was constructed from Computerized Tomography (CT) images from the right ear of a 65 year old healthy subject. The slices obtained were 0.5 mm thick. Many computational algorithms have been developed for semi or fully automatic medical imaging processing and analysis; particularly, for image segmentation, image registration and 3D shape reconstruction [15-20]. However, in this work, such algorithms could not be used due to the small size of the live structures involved and also because of the low resolution of the original images. Thus, each original 3D slice was manually segmented and the extracted contours were then interpolated and the corresponding surfaces and volumes were built up. Afterwards, those volumes (solid models) were fed into the ABAQUS software [6] and the related finite element meshes were obtained. The finite element meshing of the middle ear is then carried out, including the ligaments (superior, lateral and anterior of the malleus, superior and posterior of the incus and annular ligament of the stapes), two muscle tendons (tensor tympani and the stapedius) and simulation of the cochlear fluid, Figure 1.

The finite elements used for the ossicles (a total of 67,287) were four-node tetrahedral elements (C3D4 in ABAQUS) and for the eardrum (11,165 elements) 8-node hexahedral (C3D8 in ABAQUS) elements were used. The *pars tensa* was divided into 3 layers; the inner and outer ones being considered isotropic and the middle one (fibrous) orthotropic, with radial and circumferential fibres. Linear elements (T3D2 in ABAQUS) were used to simulate the ligaments (including the annular ligament) and the muscle tendons. The cochlear fluid was modelled using fluid elements (F3D3 in ABAQUS), assuming isochoric conditions. Comparing the geometry of the eardrum with the ossicles, one can conclude that the eardrum geometry is simpler. As a consequence the discretization of the eardrum was based on hexahedral elements, but for the ossicles the established meshes were based on tetrahedral elements.

The boundary conditions applied to the finite element model include the tympanic annulus, the connection between the stapes footplate and the cochlea and the connection of the suspensory ligaments and muscle tendons to the temporal bone. The tympanic annulus is firmly attached to the *pars tensa*, being less bent in its posterior-superior part. The attachment of the stapes footplate to the cochlea, in the oval window, is made by one-dimensional linear elements, which simulates the stapes annular ligament. The free extremities of these elements are fixed, while the others are connected to the stapes nodes. In the present finite element model,
the superior, lateral and the anterior suspensor ligaments of the malleus and the posterior and superior suspensor ligaments of the incus were studied. The five suspensory ligaments are all fixed in their free extremities: the superior ligament of the malleus and the superior ligament of the incus simulate the tegmen tympani; the posterior ligament of the incus simulates the fossa incudis; the anterior ligament of the malleus simulates the anterior wall of the tympanic cavity; and the lateral ligament of the malleus attaches the neck malleus with the tympanic annulus. The tensor tympanic muscle is fixed to the handle of the malleus, in a lateral direction and the stapedius muscle in the posterior crus of stapes.

The connection between ossicles, malleus/incus and incus/stapes, simulating incudomallear and incudostapedial joints, respectively, is done using contact formulation. Hence, the basic Coulomb friction model available in ABAQUS program [6] is used with a friction rate equal to 0.7 [5].

The friction coefficient is assumed as a function of the slip rate, contact pressure, temperature, and field variables:

\[
\mu = \mu(\dot{\gamma}_{eq}, p, \bar{\theta}, \bar{f}^\alpha),
\]

where \(\dot{\gamma}_{eq}\) is the equivalent slip rate, \(p\) is the contact pressure, \(\bar{\theta} = (1/2)(\theta_A + \theta_B)\) is the average temperature at the contact point and \(\bar{f}^\alpha = (1/2)(f_A^\alpha + f_B^\alpha)\) is the average predefined field variable \(\alpha\) at the contact point; \(\theta_A, \theta_B, f_A^\alpha\) and \(f_B^\alpha\) are the temperature and predefined field variables at points \(A\) and \(B\) on the surfaces; Point \(A\) is a node on the slave surface and point \(B\) corresponds to the nearest point on the opposing master surface.

In the present model, it is assumed that the friction coefficient is only dependent on \(\dot{\gamma}_{eq}\).

The load applied, to the eardrum, simulates a uniform Sound Pressure Level (SPL) of 80, 105 and 120 dBSPL that can be defined by:

\[
dBSPL = 20 \times \log_{10}(p/p_0)
\]

where \(p_0 = 20\mu Pa\) is the reference sound pressure corresponding to the audibility threshold [21].

Thus, distributed normal loads of 0.20 Pa (80 dBSPL) and 3.56 Pa (105 dB SPL) of sound pressure were applied in order to analyze the harmonic response of the umbo and footplate stapes (in terms of
displacements). Also, a pressure of 20 Pa (120 dBSPL) was applied to study the maximum principal stress on the ligaments.

2.2. Material Properties

Some authors have assumed viscoelastic properties for the entire middle ear system [1, 2, 4, 10]. In the present work, the eardrum and the ossicles were also assumed to have viscoelastic behaviour. Their elastic properties are summarized in Table 1, assuming an isotropic behaviour. The Poisson’s ration is assumed to be equal to 0.3 for all viscoelastic materials.

The mechanical behaviour of many materials obeys hyperelastic models [22-25]. In several publications on biomechanics, including in textbooks, the different body ligaments are presented as having hyperelastic behaviour [26-29]. Considering the Yeoh model [14], this work also uses hyperelastic non-linear behaviour for the ligaments and muscle tendons, and the present results are compared with the elastic model, Table 2.

The density for all components was adopted to be equal to 1.0E+3 Kg/m$^3$.

Assuming the Yeoh constitutive model, the strain-energy function can be written as:

\[
\psi = c_1 (I_1 - 3) + c_2 (I_1 - 3)^2 + c_3 (I_1 - 3)^3,
\]

where $I_1$ is the first right Cauchy-Green tensor invariant [22] and $c_1$, $c_2$, and $c_3$ are the material constants [30], included in Table 2, for all ligaments and muscle tendons.

Wang et al. obtained experimental curves (stress versus stretch) for the eardrum, anterior maleolar, stapedius and tympani tendons [13] and they assumed that all ligaments have the same mechanical properties as the anterior maleolar. Based on these experimental results, the material constants, for the Yeoh model, were established in the present work, Table 2. All ligaments have the same properties as the anterior maleolar ligament. In this context, we assumed a geometrically nonlinear analysis [31].

The deformation gradient tensor $\mathbf{F}$ is defined as:
\[
\mathbf{F} = \begin{pmatrix}
\frac{\partial x_1}{\partial X_1} & \frac{\partial x_1}{\partial X_2} & \frac{\partial x_1}{\partial X_3} \\
\frac{\partial x_2}{\partial X_1} & \frac{\partial x_2}{\partial X_2} & \frac{\partial x_2}{\partial X_3} \\
\frac{\partial x_3}{\partial X_1} & \frac{\partial x_3}{\partial X_2} & \frac{\partial x_3}{\partial X_3}
\end{pmatrix},
\]

where \((X_1, X_2, X_3)\) represents the position of the particle in the reference configuration and \((x_1, x_2, x_3)\) its position in the current configuration. The particular form of \(\mathbf{F}\) in the case of hyperelastic materials subjected to a uniaxial tension is:

\[
\mathbf{F} = \begin{pmatrix}
\lambda & 0 & 0 \\
0 & \frac{1}{\sqrt{\lambda}} & 0 \\
0 & 0 & \frac{1}{\sqrt{\lambda}}
\end{pmatrix},
\]

Assuming that the stress is applied along the \(x_1\) direction, taking \(\lambda_1 = \lambda\) and noting that the incompressibility condition, described by:

\[
J = \prod_{i=1}^{3} \lambda_i = \det(\mathbf{F}) = 1,
\]

requires that \(\lambda_2 = \lambda_3 = 1/\sqrt{\lambda}\), the right Cauchy-Green tensor \([\mathbf{C}]\) can then be obtained as:

\[
[\mathbf{C}] = [\mathbf{F}]^T [\mathbf{F}] = \begin{pmatrix}
\lambda^2 & 0 & 0 \\
0 & \lambda^{-1} & 0 \\
0 & 0 & \lambda^{-1}
\end{pmatrix},
\]

\(I_1\) being the first right Cauchy-Green tensor invariant:

\[
I_1 = tr(\mathbf{C}).
\]

For this particular case, \(I_1\) has the form:

\[
I_1 = \lambda^2 + 2/\lambda.
\]

According to Holzapfel [22], in the case of a uniaxial tension, the Cauchy stress, \(\mathbf{T}\), as a function of the strain invariants, is:
\[ T = 2 \left( \lambda^2 - 1/\lambda \right) \left( \frac{\partial \Psi}{\partial I_1} \right). \] (10)

Using this invariant form and the strain-energy function \( \Psi \) we obtain:

\[ T_{\text{exch}} = 2 \left( \lambda^2 - 1/\lambda \right) \left( c_1 + 2c_2 (I_1 - 3) + 3c_3 (I_1 - 3)^3 \right). \] (11)

The points of inserting the ligaments and muscle tendons were identified based on anatomy books [32-34].

2.3. Dynamic analysis

To obtain the structural dynamic response (x) of the middle ear system, the solution of the dynamic equilibrium equation [35]:

\[ M \ddot{x}(t) + C \dot{x}(t) + Kx(t) = F(t), \] (12)

is needed. In this equation, we have:

\[ M = \int_{\Omega} \rho N^T N d\Omega, \quad C = \int_{\Omega} \mu N^T N d\Omega, \quad K = \int_{\Omega} B^T D B d\Omega, \] (13)

where \( M, C \) and \( K \) represent the mass, the damping and the stiffness matrices, respectively, \( F \) being the load as a harmonic function given by:

\[ F = \int_{\Omega} N^T b dV + \int_{S} N^T T dS. \] (14)

The nodal displacement vector of any element is represented as:

\[ \mathbf{d} = \{d_1, d_2, \ldots, d_n\}^T, \] (15)

where \( n \) is the number of degrees of freedom of the element.

Assume the displacement shape functions \( N(x,y,z) \) that relate generic displacements to nodal displacements \( \mathbf{u} \) as follows:

\[ \mathbf{u} = N(x,y,z) \mathbf{d}. \] (16)

Using Rayleigh proportional damping, \( C \), is a linear combination of the mass and the stiffness matrices:

\[ C = \alpha M + \beta K, \] (17)

where \( \alpha \) and \( \beta \) are the damping parameters and their units are \( \text{s}^{-1} \) and \( \text{s} \), respectively.

In this work, we used \( \alpha = 0 \) \( \text{s}^{-1} \) and \( \beta = 0.0001 \) \( \text{s} \) [1, 2].
Therefore, the displacement response is expressed as the linear combination of the natural modes of the system. Considering undamped free vibration, i.e. without damping ($C = 0$) and external load ($F = 0$), the result is a matrix equation of motion given as:

$$\mathbf{M}\ddot{\mathbf{X}}(t) + \mathbf{K}\mathbf{X}(t) = 0.$$ \hspace{1cm} (18)

In this work, the dynamic response [36, 37] is obtained by using ABAQUS [6], as the software predicts the linear response of a structure subjected to a continuous harmonic excitation, and the analysis of the response obtained provides the steady-state amplitude and phase of the response of a system due to harmonic excitation at a given frequency. Thus, this procedure has the advantage of being able to obtain the response of a user-specified range of frequencies.

As already pointed out, the common frequencies used by audiograms (the more important audible range) are between 125 Hz to 8 kHz. Consequently the present work used the frequency sweep from 100 Hz to 10 kHz.

3. Results

3.1. Structural response to harmonic vibrations

Considering a SPL of 80 dBSPL and of 105 dBSPL (0.20 and 3.56 Pa respectively), applied on the eardrum and the two constitutive models (elastic and hyperelastic) for the ligaments and muscle tendons, umbo and stapes footplate displacements were obtained from a frequency range between 100 Hz to 10 kHz. Comparing the two models, one can conclude that for the umbo and stapes footplate displacements, Figures 2 and 3 respectively, there are no significant differences.

To confirm the validity of the present model, the results were compared with other works published [1, 4, 10], in particularly, with the results from Lee et al. that had compared their results with experimental data obtained from human temporal bones [8, 9].

For the motion of the umbo, the results obtained were between those obtained by other authors and near the experimental results of Huber et al [9], Figure 2.
In Figure 3, the displacements of the stapes footplate were compared with the ones indicated in [1, 4, 10].
For all frequencies, the values obtained were close to those indicated by Prendergast et al. [1] and Lee et al. [10] but for higher frequencies the results of Gan were somewhat lower than ours.
Considering a sound pressure level of 105 dBSPL, applied on the eardrum, the umbo and stapes footplate displacements, Figures 4 and 5, respectively, were obtained and compared with the corresponding values indicated by Kurokawa et al. [7], who had studied six temporal bones, of 61 to 74-year-old healthy (male) subjects, using a Laser Doppler Vibrometer. As we can see from those figures, the results are close to the values obtained by Kurokawa et al.

3.2. **Maximum Principal Stress in the suspensory ligaments**

The stress state, measured as the maximum principal stress, for the ligaments (superior, lateral and anterior of malleus and superior and posterior of incus) are shown in Figures 6 and 7, using elastic and hyperelastic constitutive models, respectively, for a sound pressure level of 120 dB SPL applied on the eardrum. For the model in which the ligaments were considered to have an elastic behaviour, the ligament with the highest stress was the anterior ligament of the malleus. This value of stress was obtained for low frequencies (with a peak value of 370 Pa near a frequency of 700 Hz).

The two superior ligaments presented low values. For middle frequencies, the ligament with the highest stress was the incus posterior (200 Pa for 1 kHz).

In the model in which the ligaments were assumed to be hyperelastic, the ligaments that reached the highest stress were the two superior ligaments (450 Pa for a frequency of 500 Hz). Other ligaments also presented high values of stress near this frequency, but with lower values. For lower frequencies (< 300 Hz), the ligament with the highest stress was the malleus anterior ligament (near 50 Pa).

All ligaments (with elastic and hyperelastic behaviour) have a peak of maximum stress near the first eigenvalue, Figure 8. In this Figure, the first ten natural frequencies are plotted for both models.

4. **Discussion and Conclusions**
A biomechanical dynamic study of the middle ear was conducted, and its harmonic response and the maximum principal stress of the suspensory ligaments were analysed. Concerning the suspensory ligaments and muscle tendons, two constitutive models were prepared, one with elastic and the other with hyperelastic behaviour. The structural response to harmonic vibrations was analyzed for the human audible frequency range. Harmonic responses of the umbo and stapes footplate displacements were obtained from 100 Hz to 10 kHz, for a SPL of 80 and 105 dBSPL applied on the eardrum. The results were compared with the ones indicated in previous works and were very similar. This comparison showed that the displacements obtained for the umbo and stapes footplate are close to the results found in the literature. Additionally, one can gather that the material model used (linear elastic model or hyperelastic model) has a great influence on the equilibrium in the ossicular chain.

However, if we focus on the analysis of the maximum principal stress state on the suspensory ligaments and muscle tendons, it is possible to identify significant differences. For the elastic model the highest value for the maximum principal stress was obtained on the malleus anterior ligament. On the other hand, the hyperelastic model gave the highest value for the two superior ligaments (of malleus and incus). Both these peak values of maximum principal stress were near the first natural frequency obtained.

One of the limitations of this work is that there are no comparative studies between elastic and hyperelastic models, including ligament stresses, in the literature. From a structural point of view this is essential, and perhaps this can be very important for clinicians, because the ligaments acts as a support throughout the ossicular chain, anchoring ossicles to the walls of the middle ear cavity. From the results obtained was possible to verify that the two superior ligaments present more stress.

A key conclusion of this work is that the use of hyperelastic material models for the computer simulation of middle ear ligaments and tendons is of extreme importance.

5. **Acknowledgments**

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6. References


10. Lee, CF, Chen, PR, Lee, WJ, Chen, JH, Liu, TC, Computer aided three-dimensional reconstruction and modeling of middle ear biomechanics by high-resolution computed tomography and finite


FIGURE CAPTIONS

Figure 1: Finite element model of the middle ear.

Figure 2: Umbo displacements for a sound pressure level of 80 dBSPL.

Figure 3: Stapes footplate displacements for a sound pressure level of 80 dBSPL.

Figure 4: Umbo displacements, for a sound pressure level of 105 dBSPL.

Figure 5: Stapes footplate displacements, for a sound pressure level of 105 dBSPL.

Figure 6: Maximum principal stress (Pa) of malleus and incus suspensory ligaments, with elastic behaviour, for 120 dBSPL.

Figure 7: Maximum principal stress (Pa) of malleus and incus suspensory ligaments, with hyperelastic behaviour, for 120 dBSPL.

Figure 8: The first 10 natural frequencies for the two models (elastic and hyperelastic).
TABLE CAPTION

Table 1: Material properties for the eardrum and ossicles.

Table 2: Elastic and hyperelastic properties for the ligaments and muscle tendons.
FIGURES

Figure 1

Figure 2

- Elastic
- Hyperelastic
- Chia et al.
- Nishiara et al.
- Huber et al.
- Prendergast et al.
Figure 6

Figure 7

Figure 8
### Table 1

<table>
<thead>
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<th>YOUNG’S MODULUS (N/m²)</th>
<th>DENSITY (Kg/m³)</th>
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