The Biomechanics of the Human Tongue

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SUMMARY

The human tongue is composed mainly of skeletal-muscle tissue, and has a complex architecture. Its anatomy is characterised by interweaving, yet distinct muscle groups. It is a significant contributor to the phenomenon of Obstructive Sleep Apnoea Syndrome (OSAS). A realistic model of the tongue and computational simulations are important in areas such as linguistics and speech therapy. The aim of this work is to report on the construction of a geometric and constitutive model of the human tongue, and to demonstrate its use in computational simulations for OSAS research. The geometry of the tongue and each muscle group of the tongue, including muscle fibre orientations, are captured from the Visible Human Project (VHP) dataset. The muscle model is based on the Hill three-element model that represents the constituent parts of muscle fibres and is a linearised version of a recent model due to Martins et al. The mechanics of the model are limited to quasi-static, small-strain, linear-elastic behaviour. The main focus of this work is on the material directionality and muscle activation. The transversely isotropic behaviour of the muscle tissue is accounted for, as well as the influence of muscle activation. The behaviour of the model is illustrated in a number of benchmark tests and for the case of a subject in the supine position. Copyright © 2010 John Wiley & Sons, Ltd.

KEY WORDS: Obstructive Sleep Apnea; Human Tongue; Hill Model; Finite Element Method

1. INTRODUCTION

This paper deals with the development of a finite element model of the human tongue for studies of Obstructive Sleep Apnoea Syndrome (OSAS). OSAS is a pathological condition affecting about 2% of women, and 4% of men, according to USA survey figures [1, 2]. Patients with OSAS experience various respiratory problems, an increase in the risk of heart disease, a significant decrease in productivity and an increase in motor-vehicle accidents. OSAS may be defined as the partial or complete closing of any part of the human upper airway (HUA) occurring during sleep.

The human tongue is an important contributor to the phenomenon of OSAS as it is directly involved in the collapse of the HUA in OSAS patients. It is composed mostly of skeletal-type muscle, and has a complex architecture characterised by interweaving muscle groups, each having strongly directional properties.

Computational models of the human upper airway (HUA), of varying complexity, have been developed in the past. Most of these models form the required geometry computationally, from available MRI or CT scans. Computational simulations are non-invasive, less expensive than physical testing and can be used to focus on specific factors which may influence OSAS in a patient.

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and may also reveal potential solutions to the problem. Sophisticated computational fluid dynamics (CFD), finite element (FE) models, fluid structure interaction (FSI), and neuromuscular models of the HUA are currently being developed to improve the understanding of OSAS, as well as other phenomena. Some works on modelling of the muscles of the tongue, important in OSA, have been developed by Malhotra et al. [3, 4, 5, 6]. These studies focus on muscle activation, and muscle control involved in OSA. Other FE models by Gerard et al. [7], Wilhelms-Tricarico [8, 9], Vogt et al. [10] and Fujita et al. [11, 12] were developed to improve the understanding of the role of the tongue in linguistics. Some recent work in FSI in the HUA was carried out by Chouly et al. [13, 14, 15] and Huang and Malhotra et al. [16, 17, 18, 19].

The muscle model is based on the Hill three-element model [20] that represents the constituent parts of muscle fibres and is a linearised version of a recent model due to Martins et al. [21, 22]. A complex model of the soft palate and uvula, making use of the Hill muscle model, was developed by Kojic et al. [23]. In this case, two- and three-dimensional models of the geometry are constructed, and the effects of passive and active muscle fibres are taken into account. Similar studies on active muscle behaviour effects on other areas of the body have also been carried out. Some of these include cardiac muscle studies by Humphrey et al. [24, 25, 26], and pelvic floor muscles by d’Auglignac and Martins et al. [27, 21, 22]. Research into muscle behaviour has been carried out by many in the past, including Joyce et al. [28, 29], Blemker et al. [30, 31, 32], and Aigner et al. [33]. Important work on biological materials by Holzapfel et al. [34], and also on transversely isotropic materials in general by Weiss et al. [35], proved to be valuable stepping stones in muscle related studies.

Neuromuscular effects are significant in the mechanism of OSA as well. There has been research conducted into neuromuscular models as well, e.g. those by Huang [36] and Saboisky [37]. In the case of in-vitro studies, surface electrode arrays are used to record EMG signals, which give an indication of muscle activity, e.g. the genioglossus muscle in O’Connor [38].

This paper describes the development of a realistic muscle material model, together with computational simulations of the behaviour of the tongue. This computational model focuses on the behaviour of the human tongue, taking into account its attachment points in the mouth. The mechanics of the model is limited to static, small-deformation, anisotropic, linear-elastic behaviour. The body position of the patient during an apneic episode will be considered for the simulation i.e. for the subject lying on their back or in the standing position. The ultimate objectives of this work are to examine how activation of specific muscle groups, or combinations thereof, under gravitational loading may influence the constriction of the human upper airway (HUA) and hence OSAS, by way of the tongue.

The structure of the rest of this paper is as follows. Details on the anatomy of the human upper airway (HUA), and the human tongue and skeletal muscle tissue are covered in Section 2. The capturing of the geometry of the human tongue is discussed in Section 3. Each muscle group of the tongue is visually identified, and its geometry and muscle fibre orientations captured. The governing equations, including the constitutive relations, and FE approximations are developed in Sections 4 and 5 respectively. The results of the finite element analysis of the human tongue under OSA type loading conditions are presented and discussed in Section 7. The focus of these simulations is on the behaviour of the tongue with activation of various muscle groups, or combinations thereof, with and without gravitational loading. These results obtained are analysed and compared to those available in the literature. The conclusions drawn from this work and recommendations for future work based on these conclusions are presented in Section 8.

## 2. ANATOMY OF THE HUMAN TONGUE

For the purposes of this work, it is necessary to have a good understanding of the anatomy of the HUA, and more specifically, the human tongue. The human upper airway (HUA) consists of the nasal passages, nasopharynx, velopharynx, oropharynx, hypopharynx, larynx, tongue, soft palate, uvula and the tonsils (Figure 1). This is the region of the airway in OSAS patients where the major obstruction occurs.
The human tongue is a complex muscular organ used mainly for digestion and speech, and also has a major effect on the process of breathing. The tongue is an important contributor to the phenomenon of OSAS as it is directly involved in the collapse of the HUA in OSAS patients. It is mainly made up of interweaving, but distinct groups of muscle fibres, illustrated in Figures 2 and 3.

There are intrinsic and extrinsic muscle groups, the former contained totally within the tongue, and the latter attached to a point outside the body of the tongue, and inserted into it. Intrinsic muscles generally alter the shape of the tongue, while extrinsic muscles move the tongue in various directions.

It is noted that the tongue is symmetric about the midsagittal plane with a thin medial septum of connective tissue dividing it into lateral halves. This septum runs along the entire length of the tongue, from the tip to the body of the hyoid bone. There are two jaws, namely, the maxillary (upper) and the mandibular (lower). The tongue has muscle groups originating in both of these bones. The hyoid bone located in the base of the tongue, is the only bone in the human body not articulated with another bone. It is believed that the hyoid bone plays an important role in speech in humans. Some muscles of the tongue are attached directly to it.
One of the earliest modern studies on the morphology of the human tongue, by Abd-El-Malek [43], examines its complexity by physical dissection, identifying muscle groups and their arrangement within the tongue.

Another detailed study on the tongue conducted by Takemoto [44] reveals the interweaving patterns of the fibres of the different muscle groups of the tongue. This complex myoarchitecture is what gives the tongue its high maneuverability and strength. A schematic cross-section adapted from the work of Takemoto, highlighting muscle fibre orientations of different muscles is illustrated in Figure 4. It can be seen from this figure that specific muscle groups are nearly orthogonal to each other at this cross-section, i.e. the muscle groups are mostly vertical, longitudinal, or transverse in direction.

2.1. Muscle groups of the tongue

The tongue is composed of interweaving muscle bundles or muscle groups. Each muscle group has been identified and a brief description of its structure, and muscular action is provided in Table I. The hyoid bone, located in the base of the tongue is also examined [43, 44].
Table I. Muscle groups of the tongue. See Figures 2 to 4 for illustration of these muscles [45].

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Origin</th>
<th>Main action</th>
</tr>
</thead>
<tbody>
<tr>
<td>Genioglossus (GG)</td>
<td>superior part of mental spine of mandible</td>
<td>depresses tongue; its posterior part pulls tongue anteriorly for protrusion</td>
</tr>
<tr>
<td>Geniohyoid (GH)</td>
<td>inner surface of mandible</td>
<td>moves tongue and hyoid anteriorly</td>
</tr>
<tr>
<td>Hyoglossus (HG)</td>
<td>body and greater horn of hyoid bone</td>
<td>depresses and retracts tongue</td>
</tr>
<tr>
<td>Inferior Longitudinal (IL)</td>
<td>root of tongue and body of hyoid bone</td>
<td>curls tip of tongue inferiorly and shortens tongue</td>
</tr>
<tr>
<td>Superior Longitudinal (SL)</td>
<td>submucous fibrous layer and median fibrous septum</td>
<td>curls tip and sides of tongue superiorly and shortens tongue</td>
</tr>
<tr>
<td>Transversalis (T)</td>
<td>median fibrous septum</td>
<td>narrows and elongates the tongue</td>
</tr>
<tr>
<td>Verticalis (V)</td>
<td>superior surface of borders of tongue</td>
<td>flattens and broadens the tongue</td>
</tr>
<tr>
<td>Mylohyoid (MH)</td>
<td>inner sides of mandible</td>
<td>elevates the hyoid and base of tongue</td>
</tr>
<tr>
<td>Stylohyoid (SH)</td>
<td>styloid process of temporal bone</td>
<td>elevates and pulls tongue posteriorly</td>
</tr>
<tr>
<td>Styloglossus (SG)</td>
<td>styloid process and stylohyoid ligament</td>
<td>retracts tongue and draws it up to create a trough for swallowing</td>
</tr>
<tr>
<td>Palatoglossus (PG)</td>
<td>palatine aponeurosis of soft palate</td>
<td>elevates posterior part of tongue</td>
</tr>
<tr>
<td>Digastric (DG)</td>
<td>inner surface of mandible and mastoid process of temporal bone</td>
<td>elevates hyoid and depresses mandible for opening of mouth</td>
</tr>
<tr>
<td>Hyoid Bone</td>
<td>located at posterior base of tongue</td>
<td>support structure for tongue; significant role in speech, swallowing and breathing</td>
</tr>
</tbody>
</table>

2.2. Skeletal Muscle

The human tongue is made up mainly of skeletal muscle tissue. Skeletal muscles are voluntary muscles, as they contract and relax consciously. Some of the main functions of skeletal muscle tissue are to induce motion, provide stability, and to move substances within the body [46]. Muscles of the tongue and HUA also play an important role in speech and breathing.

Skeletal muscles exhibit a hierarchical structure when observed at different levels of magnification. This structure is illustrated in Figure 5. A muscle, at the largest scale, is made up of many bundles of muscle fascicles. These are composed of long cylindrical cells, namely, muscle fibres. Muscle fibres consist of many force-producing cells, known as sarcomeres. Sarcomeres are the basic contractile part of the muscle tissue [47, 20, 48, 22, 49]. Myofibrils consist of numerous amounts of sarcomeres arranged in series and parallel to each other. A group of myofibrils arranged in parallel, make up a muscle fibre. This repetitive nature of the structure of the muscle tissue suggests that, in terms of its mechanical behaviour, the muscle is ultimately a scaled up version of a sarcomere.

The muscle fascicles of the human tongue are arranged into identifiable muscle groups. In some regions of the tongue, one specific muscle group may be dominant, but there are regions where there is more than one muscle, and the fibres of each muscle group are interdigitated. This gives the tongue a highly complex structure and maneuverability, illustrated in Figure 6 [40].

2.2.1. Muscle stimulation and muscle tone A twitch is a single stimulation-contraction-relaxation sequence in a muscle fibre. Wave summation occurs when muscle stimulation signals arrive at the
Figure 5. A breakdown of muscle fibre structure. (a) muscle fibre, (b) myofibril, (c) muscle filament, (d) sarcolemma, (e) sarcoplasm

Figure 6. Muscle structure of the tongue: transversalis (T), verticalis (V) and longitudinal (L) muscles.

Muscle before it has fully relaxed from a previous stimuli. This causes an increase in the contractile state of the muscle. At a certain stimulation rate, the muscle can partially relax between successive stimuli. This is called incomplete tetanus. At a certain increased rate, there is no time for relaxation between successive stimuli, and complete tetanus occurs. This process is illustrated in Figure 7. This allows for a wide variation of the force produced in the muscle.

Figure 7. Myogram showing an example of (a) complete tetanus, and (b) incomplete tetanus, of muscle tissue.

Muscle tone is the result of the involuntary activation of a small number of motor units, causing a sustained firmness in the relaxed muscle. A small percentage of muscle fibres are contracted, while the rest are in a relaxed state, providing overall firmness without contracting the muscle. Hypotonia is the condition of decreased muscle tone, which plays a role in some OSAS cases [46].
2.2.2. Isotonic and isometric contractions  An isotonic contraction occurs when a muscle undergoes a change in length under an applied load. Isotonic contractions can be either concentric or eccentric. For concentric contractions, the muscle tension exceeds the resistance and the muscle shortens. For eccentric contractions, the muscle tension developed is less than the resisting force, and the muscle elongates.

An isometric contraction occurs when the muscle does not or cannot change length, but the load in the muscle increases. An example of this is holding a weighted object in a fixed position. The load causes stretching, and the muscle counteracts this by contracting and an increase in tension is experienced. Although there is no movement, energy is still expended in maintaining the increased tensile force in the muscle. Most movements of the body use a combination of isotonic and isometric contractions [46].

3. GEOMETRICAL MODELLING OF THE HUMAN TONGUE

The geometrical data for the human tongue was acquired from the open-source Visible Human Project (VHP) dataset [50]. The Visible Human Project (VHP) is an ongoing effort to create highly detailed three-dimensional anatomical models of the human body. The original dataset includes photographs of a male and female cadaver that were sliced and photographed at 1mm and \(\frac{1}{3}\)mm intervals respectively. A single image of the VHP dataset is shown in Figure 8.

![Figure 8. VHP female dataset image sample taken midway through the head [50]. The blue material surrounding the subject is an immersive gel used for suspending and holding the cadaver in place.](image)

The female dataset was selected for use in this project due to its higher level of detail. In addition, the female dataset has less post-mortem deformation in the tongue than the male. The medical image processing software package, Mimics [51], was used to capture the geometry of the tongue, its individual muscle groups, and muscle fibre architecture. Mimics is a commercial image processing software package used for design and modelling [51].

3.1. Image processing in Mimics

The VHP data was imported into Mimics as a series of images. The voxels generated by this process are \(\frac{1}{3} \times \frac{1}{3} \times \frac{1}{3}\) mm, for the female dataset. A set of 200 sequential images were used ranging from just below the nasal opening, to just below the digastric muscle and epiglottis, and measure 66mm axially. This range of data is enough to capture the entire oral cavity, illustrated in Figure 9.

Various image processing tools are available in Mimics. Thresholding and region growing tools were the best for identifying regions of interest in the voxel dataset. Thresholding allows one to select different tissues from a range of grayscale values in the voxel dataset. Segmentation allows separation of different muscle groups of the tongue, as well as separation of the tongue itself from surrounding tissue. For example, segmentation of the uvula and epiglottis can be seen in Figure 10.
Region growing involves the selection of initial seed points within the voxel set. Neighbouring voxels of the initial seed are selected according to a specified thresholding procedure. The region can hence be “grown” according to the number of neighbouring voxels desired.

The final tongue geometry extracted using Mimics is illustrated in Figure 11 together with the mandible bone. This three-dimensional model was smoothed and processed to reduce irregularities without losing significant geometrical detail.
3.2. Capturing the myoarchitecture

The muscle fibre architecture of the tongue is highly influential on the overall mechanical behaviour. There is usually a clear distinction in the muscle fibre arrangement in the dataset. MedCAD, a module of Mimics, has a tool called “cylinder” which can be used to map vectors in geometry. It has been successfully used in the past for veins and arteries [52], and in this case, for muscle fibre directions. Individual muscle groups are illustrated in Figure 12.

![Figure 12. Examples of tongue muscles extracted using Mimics.](image)

The trajectory is aligned along the muscle fibre directions and control points allow one to manipulate the trajectory in three dimensions. The process of fibre extraction was carried out for each muscle group of the tongue. Using colour, grain and striations (see Figure 10) as visual indications of directionality of muscle fibres and also knowledge of the anatomy, the individual muscle groups were identified and segmented using advanced image and geometry processing tools in Mimics. In most instances, there is a clear distinction in the muscle fibre arrangement in the dataset. The number of elements of the mesh containing fibres belonging to each of the 9 muscle groups extracted are summarized in Table II.

<table>
<thead>
<tr>
<th>number</th>
<th>muscle name</th>
<th>number of elements containing fibres</th>
</tr>
</thead>
<tbody>
<tr>
<td>01</td>
<td>digastric</td>
<td>171</td>
</tr>
<tr>
<td>02</td>
<td>genioglossus</td>
<td>139</td>
</tr>
<tr>
<td>03</td>
<td>geniohyoid</td>
<td>261</td>
</tr>
<tr>
<td>04</td>
<td>hyoglossus</td>
<td>207</td>
</tr>
<tr>
<td>05</td>
<td>inferior longitudinal</td>
<td>219</td>
</tr>
<tr>
<td>06</td>
<td>mylohyoid</td>
<td>253</td>
</tr>
<tr>
<td>07</td>
<td>superior longitudinal</td>
<td>312</td>
</tr>
<tr>
<td>08</td>
<td>transverse</td>
<td>669</td>
</tr>
<tr>
<td>09</td>
<td>verticalis</td>
<td>754</td>
</tr>
</tbody>
</table>

The fibre data for each muscle group, obtained using Mimics, was processed to determine its location within the finite element mesh. An element in the mesh is assigned a single fibre vector if the fibre data lies within the same geometrical region as the element. This process is carried out for each fibre group. The combined fibre and geometry data is displayed using Paraview [53] in Figures 13(a) and 13(b).
4. GOVERNING EQUATIONS

For the purposes of this research, a linearised, small strain formulation of a muscle model is developed, based on Martins’ model [21, 22], which in turn draws on the model developed by Hill [20] for muscle fibres. It is referred to henceforth as the LMM (linearised muscle model).

The classical Hill’s muscle model consists of three components, namely the contractile (CE), series (SE), and parallel (PE) elements, each representing a specific component of a muscle sarcomere (see Figure 14).

![Figure 14. Schematic view of Hill-type muscle fibre. (a) muscle fibre, (b) myofibril, (c) muscle filament](image)

### 4.1. Linearised muscle model

The classical one-dimensional Hill type model (Figure 14) is used as a basis for this work. The three-dimensional model is consistent with the one-dimensional model used by Hill [20], Huxley [48] and Zajac [54]. The one-dimensional longitudinal muscle tension (in the direction of fibres) is the sum of the stresses in the SE and PE i.e.

\[ T = T_p + T_s . \]  

The tension in the CE is equal to the tension in the SE, i.e.

\[ T_c = T_s . \]
The total stress in the LMM is the sum of the ground substance, and the muscle fibre stress in each muscle group present, i.e.

\[ \sigma = \sigma_m + \sum_i \sigma_f^{(i)}, \]

where \( i \) = number of muscle groups present.

Here, \( \sigma_m \) is given by

\[ \sigma_m = C_m \varepsilon, \]

where \( C_m \) is the constitutive tensor of the ground material, which is typically isotropic and \( \varepsilon \) is the strain in the material.

Generally, for any stretch \( \lambda \), one can write

\[ \lambda = 1 + \varepsilon, \]

where the strain, namely, \( \varepsilon << 1 \).

For small strains, \( \lambda \approx 1 \), and the stress in the fibre, \( \sigma_f^{(i)} \), is given by

\[ \sigma_f^{(i)} = \lambda_f^{(i)} T^{(i)} m^{(i)} \otimes m^{(i)} \approx T^{(i)} m^{(i)} \otimes m^{(i)}. \]

where \( T^{(i)} \) is the total tensile force in the fibre of each muscle group present.

The strain in a muscle fibre, \( \varepsilon_f \), is given by

\[ \varepsilon_f = m \cdot \varepsilon m. \]

The multiplicative split of the stretches has an advantage over the additive split method, commonly used in Biomechanics, in that it does not require information on the partition of the initial fibre length between the CE and SE.

Using (5), the multiplicative split equation (8) can now be rewritten as

\[ \lambda_f = 1 + \varepsilon_s + \varepsilon_c + \text{higher order terms}, \]

where, for small strains, the higher order terms are assumed to be insignificant. Rearranging the terms in (10), and considering (5), we obtain a formulation for SE and CE strain in terms of fibre strain:

\[ \lambda_f - 1 \simeq \varepsilon_s + \varepsilon_c, \]

hence

\[ \varepsilon_f \simeq \varepsilon_s + \varepsilon_c. \]

The stress in the PE is given by

\[ T_p(\varepsilon_f) = T_0 f_p(\varepsilon_f). \]

Linearization of \( f_p \) gives

\[ f_p(\varepsilon_f) = \begin{cases} m_p \varepsilon_f, & \varepsilon_f > 0 \\ 0, & \text{otherwise} \end{cases} \]

The resulting linearised stress-strain relation for the PE can be seen in Figure 15(a). Here \( m_p = 2aA \) is the slope of the curve for \( \varepsilon_f > 0 \), and \( T_0, a \) and \( A \) are material parameters.
The stress in the SE is given by

\[ T_s(\varepsilon_f, \varepsilon_c) = T_0 f_s(\varepsilon_f, \varepsilon_c) . \]  

(15)

Using Taylor expansion, and noting that for small displacement \( x \), we have \( e^x \simeq 1 + x \). Applying this approximation and the approximation in (12), it can be shown that

\[ f_s(\varepsilon_f, \varepsilon_c) = \begin{cases} 
  m_s (\varepsilon_f - \varepsilon_c), & \varepsilon_s \geq 0 \\
  0, & \text{otherwise} 
\end{cases} . \]  

(16)

The resulting linearised stress-strain relation for the SE can be seen in Figure 15(b). Here \( m_s = 10 \) is the gradient of the straight line.

The stress in the CE is given by

\[ T_c(\varepsilon_c, \dot{\varepsilon}_c, \alpha) = T_0 f_c^l(\varepsilon_c) f_c^v(\dot{\varepsilon}_c) \alpha(t) , \]  

(17)

where \( \alpha(t) \) is the activation function in (20).

Using (5) and (12) gives the relationships for \( f_c^l(\varepsilon_c) \) and \( f_c^v(\dot{\varepsilon}_c) \) as

\[ f_c^l(\varepsilon_c) = \begin{cases} 
  1, & -0.5 \leq \varepsilon_c \leq 0.5 \\
  0, & \text{otherwise} 
\end{cases} , \]  

(18)

and

\[ f_c^v(\dot{\varepsilon}_c) = \begin{cases} 
  0, & \dot{\varepsilon}_c < -5 \\
  1, & -5 \leq \dot{\varepsilon}_c \leq 3 \\
  1.6, & \dot{\varepsilon}_c > 3 . \]  

(19)

The resulting linearised stress vs. strain and stress vs. strain-rate relations for the CE can be seen in Figure 16. The blue line in each of the Figures 16(a) and 16(b) illustrate the linearised functions used for the LMM, superimposed on the original non-linear functions, shown in green.

4.2. Muscle activation function

The time-dependent muscle activation function illustrated in Figure 17 is given by the solution to the first-order differential equation used by Pandy and Zajac et al. [55, 54],

\[ \dot{\alpha}(t, u) = \frac{1}{\tau_r} (1 - \alpha(t)) u(t) + \frac{1}{\tau_f} (\alpha_{min} - \alpha(t))(1 - u(t)) , \]  

(20)

where \( \tau_r \) and \( \tau_f \) are rise and fall time constants for activation and deactivation of the muscle, respectively, \( \alpha_{min} \) is the minimum value of the activation, and \( u(t) \) is the neural excitation as a
function of time given by

\[ u(t) = \begin{cases} 1, & 0 < t \leq 1 \\ 0, & \text{otherwise.} \end{cases} \]  \hspace{1cm} (21)\]

The backward Euler method is an implicit method which uses the current and previous states of the system to find the solution at the current state. Using this method, the activation rate in Equation (20), can be approximated by

\[ \dot{\alpha} = \frac{\alpha^n - \alpha^{n-1}}{\Delta t}, \]  \hspace{1cm} (22)\]

where \( \Delta t \) is the time-step size, \( \alpha^n \) is the activation at the current time-step, \( \alpha^{n-1} \) is the activation at the previous time-step. Applying the backward Euler method to the activation function in (20), using the neural input function in (21), and solving for \( \alpha^n \), gives the solution as

\[ \alpha^n(t) = \begin{cases} \frac{\alpha^{n-1} \tau_r \tau_f + \Delta t \tau_f}{\tau_r \tau_f + \Delta t \tau_f}, & u = 1 \\ \frac{\alpha^{n-1} \tau_r \tau_f + \Delta t \alpha_{\text{min}} \tau_r}{\tau_r \tau_f + \Delta t \tau_r}, & u = 0 \end{cases}. \]  \hspace{1cm} (23)\]

This implicit method allows one to determine the activation level at the current increment in time based on the activation level at the previous time increment. It is in this way that the activation function is handled computationally.
4.3. LMM parameters

The parameters of the constitutive equation were obtained from multiaxial testing data, from Humphrey and Yin \cite{26}, and are given as

\[ c = 3.87 \text{gf/cm}^2, \quad b = 23.46, \quad A = 8.568 \times 10^{-4} \text{gf/cm}^2, \quad a = 12.43. \]

In addition, the activation stress constant is chosen from Martins et al. \cite{21}, namely

\[ T_0 = 6280 \text{gf/cm}^2, \quad (24) \]

It can be shown that the shear modulus is related to the Young’s modulus as follows,

\[ G = \frac{E}{2(1 + \nu)}, \quad (25) \]

Using the parameters \( b \) and \( c \), a relationship between Young’s modulus, and the Poisson’s ratio, is given as

\[ E = 2(1 + \nu)bc. \quad (26) \]

Thus, for a selected Poisson’s ratio \( \nu = 0.45 \), Young’s modulus is given as \( E \approx 26000 \text{Pa} \).

5. FINITE ELEMENT APPROXIMATIONS

The weak form of the equilibrium equation of the LMM boundary value problem is given as

\[ \int_{\Omega} \sigma(u, \varepsilon_c) : \varepsilon(v) d\Omega = \int_{\Omega} f \cdot v \ d\Omega, \quad (27) \]

Expanding (3) using the relationships in (4) and (6) gives

\[ \sigma = C_m \varepsilon + T_0 \sum_i [\beta \varepsilon_f^{(i)} + \gamma \varepsilon_c^{n(i)}] m^{(i)} \otimes m^{(i)}, \quad (28) \]

where \( n \) is the current solution increment, and the introduced parameters are defined as

\[ \beta = m_p + m_s, \quad \gamma = -m_s. \quad (29) \]

\[ \gamma = -m_s. \quad (30) \]

From the constraint stated in the Hill three-element model in (2), the contractile stress must be equal to the series stress, i.e.

\[ \sigma_c = \sigma_s. \quad (31) \]

Expanding the contractile and series element stress functions using (19) and (16) for each fibre (i), gives

\[ f_c^l \left(m_c^\nu \varepsilon_c + e_c^\nu\right) \alpha^n = m_s (\varepsilon_f - \varepsilon_c^n). \quad (32) \]

Using the backward-Euler method, the contractile strain rate \( \dot{\varepsilon}_c \), can be approximated by

\[ \dot{\varepsilon}_c = \frac{\varepsilon_c^n - \varepsilon_c^{n-1}}{\Delta t}, \quad (33) \]

where \( \varepsilon_c^n \) and \( \varepsilon_c^{n-1} \), are the current and prior contractile strains.

Combining Equations (32) and (33), the current contractile strain in a fibre is then given by

\[ \varepsilon_c^n = n_\phi \varepsilon_f + \phi (\varepsilon_c^{n-1} + \kappa \Delta t), \quad (34) \]
where, assuming $f_c^l \neq 0$, $\alpha^n \neq 0$, $m_c^v \neq 0$, parameters are formed by rearrangement of terms as

$$\eta = \frac{m_c^v \Delta t}{m_c^v f_c^l \alpha^n}, \quad \phi = (1 + \eta)^{-1}, \quad \kappa = -\frac{e_c^n}{m_c^v}.$$  \hspace{1cm} (35)

Substitution of $e_c^n$ using equation (34) into equation (28) gives

$$\sigma = C_m \varepsilon + T_0 \sum_i \left[ (\beta + \gamma \eta \phi) \varepsilon_f^{(i)} + \gamma \phi (\varepsilon_c^{(n-1)(i)} + \kappa \Delta t) \right] m^{(i)} \otimes m^{(i)}.$$  \hspace{1cm} (36)

Equation (36) together with the weak equilibrium equation in (27) is to be solved using the finite element method. Applying (27) to (36) to the linear system of equations,

$$Kd^n = F,$$  \hspace{1cm} (37)

where $K$, $d^n$, and $F$, are the stiffness matrix, displacement vector, and force vector, respectively. This gives the equilibrium equation for the LMM as

$$\begin{cases}
K_m = \int_{\Omega} B^T C_m B \, d\Omega + \sum_i \int_{\Omega} T_0 (\beta + \gamma \eta \phi) B^T m^{(i)} \otimes m^{(i)} \otimes m^{(i)} B \, d\Omega \\
K_f \sum_i \int_{\Omega} T_0 \gamma \phi (\varepsilon_c^{(n-1)(i)} + \kappa \Delta t) B^T m^{(i)} \otimes m^{(i)} \, d\Omega
\end{cases}$$

$$= \int_{\Omega} N^T b \, d\Omega - \int_{\Omega} T_0 \gamma \phi (\varepsilon_c^{(n-1)(i)} + \kappa \Delta t) B^T m^{(i)} \otimes m^{(i)} \, d\Omega$$

$$= \left\{ \begin{array}{ll}
\begin{array}{ll}
K_m \sum_i \int_{\Omega} T_0 (\beta + \gamma \eta \phi) B^T m^{(i)} \otimes m^{(i)} \otimes m^{(i)} B \, d\Omega \\
F_b
\end{array}
\end{array} \right.$$

where $K_m$ is the ground substance stiffness matrix, $K_f$ is the fibre stiffness matrix, $F_b$ is the body force and $F_f$ is the fibre force.

6. EXAMPLE: CONCENTRIC CONTRACTION

To illustrate the functioning of the model, we demonstrate concentric contraction with an example similar to that presented in [21]. A $10 \times 10 \times 1$ mm$^3$ block, depicted in Figure 18(a), was discretised into $14 \times 14 \times 2$ hexahedral elements and had muscle fibres aligned in the X-direction. A planar constrained was applied to the surfaces with a Z-normal and on the -X orientated face, which was also pinned at its central node. A tensile load of a total of 4.9gf, which was linearly ramped over a period of 1s and subsequently held constant, was applied to the +X surface. After 1s had elapsed, the passive muscle was activated by setting $u = 1$, and remained active for a duration of 1s, after which the stimulus was removed.

Figure 18(b) demonstrates that with a 1% activation force, the displacement of the loaded face was qualitatively comparable to that shown in [21] for a higher contractile force. The displacement of the measured point mimicked the activation pattern governing the contractile force generated in the muscle both during contraction and relaxation. In the first second, the passive muscle underwent extension due to the traction force. As the muscle was excited, the contractile force exceeded the applied load and the material entered a contractile state after 0.2s of activation. At 1.8s, maximal excitation was achieved and the maximal displacement of $-2.75$ mm was obtained. As the material was linear in nature, it was more compliant at large deformations than models used in the literature, requiring a lower contractile force to produce the motion than that used by [21]. The material returned to the tensile state as the muscle was relaxed.

7. FINITE ELEMENT ANALYSIS (FEA) RESULTS

The results for the FEA of the model of the human tongue are presented in this section. It is widely believed that the OSA event is highly influenced by gravity, low airway pressure during
inspiration, and reduced muscle tone, amongst other effects. Some of these effects are tested using the tongue geometry and linearized muscle model (LMM) presented in this paper. The LMM is tested in levels of increasing complexity, starting from the most basic isotropic (without fibres) case under gravitational load. The case with all muscle fibre groups being passive is also tested under the same gravitational loading condition. The activation of individual muscle groups on the movement of the tongue are also tested, firstly without gravitational loading, then under gravitational loading.

The activation function used for these tests is the same as that introduced in Figure 17, and is based on the work of Martins et al. [21, 22]. In addition, the activation stress parameter is given as $T_0 = 68.20 \text{gf/cm}^2$, i.e. 1% of the activation level specified in Martins’ work. The behaviour of the model can be qualitatively compared to the work on pelvic floor muscles presented by Martins. Although airway pressure is a significant contributor to OSA, it is not modeled in these simulations as the focus of this work is on muscle fibre directionality under gravitational loading and how this relates to OSAS.

One of the main objectives of these simulations is to examine the effects of activation of various muscle fibre groups of the tongue on OSA, or more specifically, on the prevention of constriction of the airway. These simulations would also give insight into the role of each muscle group in the kinematics of the tongue. This is done by monitoring a point on the back of the tongue model under various loading conditions and muscle activity, and comparing the results.

7.1. Smoothed mesh of the tongue

A mesh of the right sagittal half of the tongue, generated from the geometrical data of the VHP and smoothed and processed in Gambit, is illustrated in Figure 19. The tongue and loading conditions are assumed to be symmetrical about the mid-sagittal plane. This mesh of the tongue consists of 4800 hexahedral elements. A node near the back of the tongue indicated as node A in Figure 19, was selected for monitoring.

This node was selected due to its location near the unconstrained posterior part of the tongue, where maximum displacement of tongue material is expected to occur. The mid-plane of the tongue geometry is fixed in the $X$-direction due to symmetry. The attachment points of the tongue to the mandible and hyoid bones are fully fixed for all degrees of freedom.

7.2. Isotropic tongue displacement results

In this test we consider an isotropic formulation (i.e. without the effect of muscle fibres), and under gravitational loading. There are two test cases, one simulating a person in the standing position.
with gravity in the positive \( Z \)-direction, and the other with the person lying on their back (supine position) with gravity in the negative \( Y \)-direction. This is used as a basis for the behaviour of the tongue under gravitational loading for more complex tests involving muscle activation.

The displacement contours for the isotropic tongue with gravity in the \( Z \)-direction with the person in the standing position for the LMM UEL element are displayed in Figure 20 (displacement plot scaled to 1000 \%). From the illustration, it can be seen that the entire body of the tongue (except fixed nodes,) move in the \( Z \)-direction, i.e. downward. It should be noted that in all displacement contour plots, the values at only one node per element are rendered during post-processing.

The displacement contours for the isotropic tongue with gravity in the negative \( Y \)-direction for the LMM UEL element type are displayed in Figure 21 (displacement plot scaled to 1000 \%). From the illustration, it can be seen that the entire body of the tongue (except fixed nodes,) move in the negative \( Y \)-direction. This loading direction with the person in the supine position is selected for muscle activation tests to follow.

### 7.3. Active tongue displacement results

The effect of muscle activation is tested on the mesh of the tongue both with and without gravitational loading by activating individual muscle fibre groups. In these simulations, the tongue was first loaded with gravitational loading, ramped up from zero to maximum value at \( t = 1 \) s. Once the tongue is fully loaded, specific muscles or combinations thereof are activated.
It was found that activating the genioglossus (GG), hyoglossus (HG) and verticalis (V) muscles had the greatest effect on pulling the body of the tongue away from the back of the HUA, and would thus reduce constriction in an OSAS event. Activating this set of muscles pulls the tip of the tongue, as well as the back of the tongue forward. The displacement contour for the active GG, HG and V muscles, without gravitational loading, is illustrated in Figure 22.

The displacement contour for the active GG, HG and V muscles under gravitational loading is illustrated in Figure 23. When compared to the isotropic case, depicted in Figure 21, it can be seen that activating this set of muscles reduces the displacement at the back of the tongue caused by gravitational loading.

The displacement results of node A (illustrated in Figure 19) for the activation of each muscle group, or combinations thereof, without and with gravitational loading (g), is summarised in Table III.

7.4. Discussion
The results of the finite element simulations of the tongue are examined in detail and presented here.
7.4.1. Passive and active contributions  In all the simulations presented using gravitational loading, the gravitational load was applied by ramping up from zero to its maximum value over a period of one second, starting at the beginning of the simulation, and maintaining this force until the end of the simulation. Once the gravitational load reached its peak value at $t = 1 \text{s}$, specific muscles were activated for a period of one second, and deactivated at $t = 2 \text{s}$ (except in the passive and isotropic cases, where no muscles are activated).

The $Y$-displacement history of node A for the case with all muscles being passive under gravitational loading is shown in Figure 24(a). The $Y$-displacement history of node A for the case with the GG, HG and V muscles being activated simultaneously under gravitational loading is shown in Figure 24(b), and this corresponds to the displacement contour in Figure 23.

From these two figures, the difference between the active and passive behaviour is clear. The contribution of the activation load to the displacement behaviour can be superimposed on the contribution of the gravitational load in the passive case. In the passive case, only the gravitational
load applies, and this results in a displacement in the negative $Y$-direction. In the active case, the three muscles being activated all contribute to pulling the body of the tongue towards the front of the mouth, resulting in a positive displacement in the $Y$-direction for the duration of muscle activation. When the muscle is deactivated in this case, the tongue eventually returns to the position it held in the passive case (see Figure 24(b)).

7.4.2. Muscle activation without gravitational loading

The results for the active tongue muscles without gravitational loading are displayed in Table III. From the results generated for activation of specific muscle groups, it is seen that some muscles move the tongue away from the back of the mouth (anteriorly), while others move it towards the back of the mouth (posteriorly). Those that move the tongue anteriorly (in the positive $Y$-direction) when activated, reduce the potential for an OSA event occurrence. The muscle which moves the tongue furthest forward, is the verticalis muscle, as it has the largest positive displacement in the $Y$-direction. From highest to lowest positive displacement in the $Y$-direction, the HG, GG, GH, IL, and DG muscles also move the tongue forward, but less so than the V muscle. The muscle which moves the tongue posteriorly the most, is the T muscle. The SL and MH muscles also move the tongue posteriorly, the SL more so than the MH. The reason for the V and T muscles having the greatest effect on Node A is due to the fixed displacement constraint on the hyoid bone. All the other muscles of the tongue interact with the hyoid bone, and are hence more constrained by it. The V and T muscles are near the surface of the tongue, and have more freedom and hence a greater effect on the tongue movement.

Some of the muscles which move the tongue anteriorly were activated simultaneously, and the resulting displacement is shown in the last row in Table III. For this case, the GG, HG and V muscles were selected and activated. The combined activation moves the tongue even further forward than if they were individually activated, as the muscles work together. The displacement history of node A in the $Y$-direction for the case with GG, HG and V muscles being activated simultaneously is shown in Figure 25(b), and this corresponds to the displacement contour presented in Figure 22.

Another test was done with all muscles being activated simultaneously. This result is shown as the second last entry in Table III. The displacement history of node A in the $Y$-direction for the case with all muscles being active is shown in Figure 25(a).

In these figures, it can be clearly seen that activation of all muscles simultaneously has a mixed effect on results, leads to a smaller positive displacement in the $Y$-direction, and hence is not as effective as activating selected muscles as in the case with GG, HG and V activation. In this case, the muscles are working against each other.

7.4.3. Muscle activation with gravitational loading

The results for the active tongue muscles under gravitational loading are also given in Table III. In these simulations, the tongue was first loaded...
with gravitational loading, ramped up from zero to maximum value at $t = 1$ s. Once the tongue is fully loaded, specific muscles, or combinations thereof, are activated. It was found that only the V muscle had a positive displacement in the $Y$-direction. The DG, GH, GG, HG and the IL muscles also reduce the negative displacement caused by the gravitational load, but are not powerful enough to cause a positive displacement in the $Y$-direction when combined with the gravitational load. The T, SL and MH muscles combine with the force of gravity to pull the tongue further back.

A test was done combining the activation of the GG, HG and V muscles under gravitational loading. The $Y$-displacement history of node A for the case with GG, HG and V muscle being activated simultaneously under gravitational loading is shown in Figure 26(b), and this corresponds to the displacement contour in Figure 23.

An additional case with all muscles being active under gravitational loading was also evaluated. The $Y$-displacement history of node A for the case with all muscles being active under gravitational loading is shown in Figure 26(a). The muscle behaviour in the cases with gravitational loading was found to be similar as the cases without gravitational loading, i.e. the same muscles had similar effects either with or without gravitational loading.
results of the action of a combination of muscles is not a linear combination of the individual muscle actions.

For the case with all muscles being activated simultaneously, the maximum value of displacement in the $Y$-direction after application of the gravitational load is $-551.77 \times 10^{-3}$ mm, whereas in the case with only the GG, HG and V being activated, the maximum value of displacement in the $Y$-direction after application of the gravitational load is $+379.17 \times 10^{-3}$ mm. By activating only a selected group of muscles, the gravitational load can be overcome, but activating all the muscles simultaneously leads to a much smaller effect. Again, this is due to the action of component muscles canceling each other out. Ultimately, a select group of muscles must work simultaneously to most effectively prevent the posterior motion of the tongue and thus airway constriction.

8. CONCLUDING REMARKS

A full finite element model, with defined muscle groups and muscle fibre orientations taken into account, has been developed. The geometry of the human tongue is based on the assumption of symmetry. This is a reasonable assumption taking into account the direction of loading and the anatomy of the tongue.

The high level of detail of the geometry extracted was deemed to be accurate enough for the purposes of this work. The voxel size of $\frac{1}{3} \times \frac{1}{3} \times \frac{1}{3}$ mm means that features as small as this could be captured. This highly detailed geometry was used to generate a mesh of 4800 elements. This mesh size seemed reasonable for the size of the domain used. The entire body of the tongue could fit into a space of roughly $100 \times 100 \times 100$ mm. The muscle fibre data of each muscle group within the tongue was successfully combined with the mesh.

This linearised muscle model is a suitable step in the development and analysis of the model of the tongue. The muscle model is based on the works of Humphrey [56], Hill [20], and the recent work of Martins [21, 22]. With this model, we were able to simulate the behaviour of the tongue with each muscle group individually activated, and also with combinations of active muscle groups. We were also able to examine how different muscles contribute or counteract the OSA event when they are activated.

These simulations allow insight into muscle behaviour which cannot be easily gained through physical testing. We have observed that activation of specific muscle groups of the tongue simultaneously leads to a reduction of airway constriction in this simulated OSAS case. These are, in decreasing effect, the V, HG, GG, GH, IL, and DG muscle groups. It has also been noted that the activation parameter $T_0$ affects stiffness passively, and also the magnitude of the activation. This $T_0$ parameter affects the muscle model similarly to muscle tone in the tongue, which is also a contributing factor in OSAS.

The results are validated at relevant points in development, increasing in complexity at each point. Isotropic uniaxial testing, Cook membrane tests, passive fibre directionality testing, active fibre testing, and body force testing were all done on varying mesh sizes. The isotropic tests were compared to standard element types available in Abaqus, and relevant literature. Passive and active test results are compared to those results found in Martins’ work. Results for the isotropic validation tests were, in most cases, 100% accurate when compared to the results of Abaqus standard C3D8 elements under similar loading. The results for the isometric and isotonic tests are comparable to the results found in Martins’ work. The displacement solution for these tests display similar results. The differences in the numerical results are due to the limitations imposed on the model, i.e. linear elasticity, no selective reduced integration, and small strain assumptions. Even with these limitations, both the results and behaviour prove to be similar.

Gravitational forces were applied in the standing and supine positions in successive tests. The effects of individually activated muscle groups on the displacement of the tongue in the gravitational field were examined. It was found that even when only passive, the tongue displaced less in the gravitational field when any muscle fibres were present. These fibres provide an additional stiffness to the material, even without active contraction. With active contraction, the stress experienced by the material increases.
Using this model and simulations, it has been possible to examine the behaviour of the human tongue in a simplified simulated case of OSAS, where gravity is assumed to be the main contributor. From the results it is observed that activation of specific muscles in the tongue simultaneously leads to a reduction in constriction of the airway in this OSA case. Future models will incorporate airway pressure, which is another significant contributor to OSAS.

REFERENCES


