Finite element analysis of active Eustachian tube function

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Ghadiali, Samir N., Julie Banks, and J. Douglas Swarts. Finite element analysis of active Eustachian tube function. J Appl Physiol 97: 648–654, 2004. First published March 26, 2004; 10.1152/japplphysiol.01250.2003.—The inability to open the collapsible Eustachian tube (ET) has been related to the development of chronic otitis media. Although ET dysfunction may be due to anatomic and/or mechanical abnormalities, the precise mechanisms by which these structural properties alter ET opening phenomena have not been investigated. Previous investigations could only speculate on how these structural properties influence the tissue deformation processes responsible for ET opening. We have, therefore, developed a computational technique that can quantify these structure-function relationships. Cross-sectional histological images were obtained from eight normal adult human subjects, who had no history of middle ear disease. A midcartilaginous image from each subject was used to create two-dimensional finite element models of the soft tissue structures of the ET. ET opening phenomena were simulated by applying muscle forces on soft tissue surfaces in the appropriate direction and were quantified by calculating the resistance to flow ($R_v$) in the opened lumen. A sensitivity analysis was conducted to determine the relative importance of muscle forces and soft-tissue elastic properties. Muscle contraction resulted in a medial-superior rotation of the medial lamina, stretching deformation in the Ostmans’s fatty tissue, and lumen dilation. Variability in baseline $R_v$ values correlated with tissue size, whereas the functional relationship between $R_v$ and a given mechanical parameter was consistent in all subjects. ET opening was found to be highly sensitive to the applied muscle forces and relatively insensitive to cartilage elastic properties. These computational models have, therefore, identified how different tissue elements alter ET opening phenomena, which elements should be targeted for treatment, and the optimal mechanical properties of these tissue constructs.

Young’s modulus; biomechanics; elasticity; respiratory airway; compliance; fluid-structure interactions; mathematical modeling

THE DEVELOPMENT OF PERSISTENT OTITIS MEDIA (OM) has long been associated with an inability to open the collapsible Eustachian tube (ET), i.e., ET dysfunction. The ET, which connects the nasopharynx (NP) with the middle ear (ME), has three physiological functions: 1) protection of the ME from opportunistic pathogens of the NP, 2) clearance of ME fluids, and 3) ventilation of the ME to ambient pressures (4). Although the ET normally exists in a “closed” configuration to protect the ME, the clearance and ventilation functions require periodic openings of the ET. In healthy individuals, these openings occur during swallowing, where muscle contraction deforms the surrounding soft tissue resulting in an increase in the cross-sectional lumen area and a reduction in the resistance to airflow (3). In the absence of these openings, the balance between gas adsorption by the ME mucosa and resupply via the ET is disrupted (10). The inability to ventilate the ME via the ET results in negative ME pressures (relative to atmospheric), fluid transudation into the ME, and infection/inflammation of the ME mucosa, i.e., OM. Although an inability to open the ET is the primary etiology of OM (4), standard treatment therapies (e.g., tympanostomy tubes and antibiotics) do not address the underlying mechanical and/or anatomic abnormalities responsible for ET dysfunction. The development of novel treatment therapies for persistent OM will therefore require an understanding of how these structural properties influence ET opening phenomena.

The anatomic structure of the ET is highly complex in that the lumen is surrounded by several muscular, cartilaginous, and fat tissue elements (see Fig. 1) and is bounded by fluid-coated mucosal tissue. Several investigators have demonstrated that paralysis of the tensor veli palatini muscle (TVPM), the primary muscle associated with ET function, results in negative ME pressures (7), fluid accumulation in the ME (1), and a significant decrease in the compliance or elastic properties of the ET (14). Although it is well established that contraction of the TVPM is required for normal ET function, several other mechanical properties may also play a role in opening the collapsed ET. For example, previous investigations (15, 29) have suggested that the elastic and viscoelastic properties of the cartilage and/or fat tissue may be important determinants of ET function. Specifically, hypercompliant tissue properties may impair the active opening of the ET by diminishing the forces transmitted to the lumen, whereas very large muscle forces may be required to open rigid or inelastic ETs. The opening of the ET may also depend on the surface tension, adhesive, and inflammatory properties of the ET mucosa (5, 11, 13). Although these tissue mechanical properties have been implicated in ET dysfunction, the relative importance of the various tissue elements as well as a quantitative understanding of how variations in mechanical properties affect opening phenomena have not been investigated.

In addition to the mechanical environment, several investigators (20, 24, 30) have utilized histological imaging techniques to identify the morphometric relationships that may contribute to ET dysfunction. The volume of cartilage and/or muscle tissue, curvature orientation of the lumen, height-to-width ratio of the ET cartilage, and insertion angle of the TVPM have all been suggested to be important determinants of ET function. Several three-dimensional (3D) anatomic features, including variations in muscle insertion angles and

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cartilage size along the length of the ET, may also influence ET function (25, 27). Although these investigations have identified the anatomic properties that may contribute to ET dysfunction, a major limitation of these observational studies is that they can only speculate on how the measured morphometric relationships influence the tissue deformation processes responsible for active ET opening. As a result, these studies cannot be used to obtain a detailed understanding of ET opening phenomena or to quantify the relative importance of the various anatomic properties.

Although insufficient ET opening during swallowing may be due to both mechanical and/or anatomic abnormalities, previous investigations of ET function were performed independently in that the interaction between the ET’s mechanical and anatomic environment was not considered. As a result, the precise mechanisms by which these structural properties alter the function of the ET have not been investigated. The goal of the present study was to develop a two-dimensional (2D) finite element mathematical model of active ET opening phenomena, which can be used to investigate these structure-function relationships. Although the 3D structure may influence ET function, this study focuses on developing 2D modeling techniques that capture essential anatomic and mechanical features in a single cross section. These initial models are based on the mechanical and anatomic properties measured in normal adult subjects. Tissue deformation and lumen opening during swallowing are simulated by applying muscle forces on the appropriate soft tissue elements, and a flow resistance parameter is calculated to quantify the degree of lumen opening. A parameter variation analysis is conducted to determine the relative importance of various tissue mechanical properties, including muscle force magnitude and soft-tissue elastic properties. As a result, these models elucidate how the mechanical properties influence ET opening phenomena and identify which tissue elements are the most important determinates of ET function. The development of these computational models also elucidates how modifications in the anatomic and mechanical structure of the ET influence its function and, therefore, may be useful in the development of novel structure-based treatment therapies for OM.

**Methods and Materials.**

**Protocol.** Data were obtained from eight adult human temporal bones, which included the ET cartilage, TVPM, levator veli palatini muscle (LVPM), and Ostmann’s fatty tissue (OFT). Subjects had no documented history of ME disease or ET dysfunction, an age range of 39 ± 23 yr with no individual <18 yr in age, and equal numbers of female and male subjects. All specimens were processed histologically according to the technique developed by Sando et al. (23). Briefly, each specimen was fixed in 10% formalin, decalcified in 5% trichloroacetic acid, dehydrated in graded concentrations of ethanol, and embedded in celluloid. Serial 30-μm histology sections were cut vertically in the plane perpendicular to the long axis of the ET. Every 20th section was stained with hematoxylin and eosin for light microscopy. One cross-sectional image was selected from each subject for image analysis and construction of the 2D finite element models. The selection of this cross-sectional image was based on morphological features of the TVPM. The TVPM is a flat, ribbon-like muscle that attaches to the cranial base and lateral portions of the cartilage and OFT, descends inferiorly, and ends in a tendon that winds around the pterygoid hamulus (16) (see Fig. 1). In the proximal region of the ET (i.e., near the NP), the TVPM is primarily attached to the cranial base (not shown in Fig. 1). In contrast, the TVPM is only attached to the cartilage and OFT in distal regions. For this study, the most distal cross section in which the TVPM is not attached to the cranial base and is attached to the cartilage and OFT was selected. The definition of this midcartilaginous section is consistent with the observations of Takahashi and colleagues (28) with respect to the location of ET dysfunction. This definition also ensures that the selected cross sections are comparable between all subjects. All histological specimens were generously provided by the Elizabeth McCullough Knowles Otopathology Laboratory at the University of Pittsburgh.

**Histological image processing.** The selected cross-sectional image from each subject was visualized with a Diaphot inverted microscope (Nikon Instruments) with a charge-coupled device video camera attachment (Carl Ziiess). Digital images were acquired at an average resolution of 0.039 ± 0.011 mm/pixel with the Metamorph image analysis package (Universal Imaging). High-quality contours of the cartilage and OFT as well as the medial lumen surface were obtained as shown in Fig. 2A. These contours were generated in a piecewise fashion such that important surface locations could be identified. Specifically, the medial-superior attachment of the cartilage to the OFT as well as the anterior and inferior portions of the OFT and cartilage (D and E in Fig. 2A) were recorded. Finally, the insertion angle of the TVPM was measured at two locations, the superior OFT-cartilage junction (θs) and the inferior insertion with the OFT (θb).

**Finite element modeling.** The geometric contour information from each subject was imported into the ADINA finite element program (Watertown, MA) for mesh generation and tissue deformation analysis. First, the cartilage and OFT regions in Fig. 2A were subdivided into small six-noded triangles or elements (see Fig. 2B) using free-form meshing techniques. The size and quality of these elements is directly related to the accuracy of the finite element technique. Note that the cartilage and OFT regions were meshed with two different element types to account for differences in mechanical properties (see below). These triangular elements were then used to solve the complex but well-established equations that govern tissue deformation and stress generation in 2D (2, 12). Specifically, standard 3D deformation equations were simplified with a 2D plain-strain analysis such that all deformations occur in the cross-sectional or y-z coordinate plane (Cartesian coordinates). Solution of these 2D plain-strain equations requires specification of several modeling parameters, including boundary conditions, applied loads, and tissue mechanical properties.
Fixed boundary conditions (i.e., no deformations) were specified on the medial-superior surface of the cartilage (section A in Fig. 2A) to approximate attachment to the nondeformable cranial base, whereas free boundary conditions (i.e., unrestrained deformations) were specified at all other non-load-bearing surfaces. Muscle contraction was simulated by applying forces on the load-bearing surfaces (sections B to E), and insertion angles of TVPM [superior (θS) and inferior (θI)].

Baseline force magnitudes used in this study, FLVPD and FLVPE, are the total force generated by the TVPM in the y-z plane on sections D and E, respectively. FLVP, total force generated by the TVPM in the y-z plane.

Application of the appropriate boundary conditions, muscle loads, and tissue mechanical properties results in tissue deformation and an opening of the ET lumen (see Fig. 4). This open lumen area was analyzed with a one-dimensional fluid-flow model, similar to the one utilized by Dai et al. (9), to calculate a viscous flow resistance parameter

\[ R_v = \frac{\Delta P}{Q} = \frac{\mu L}{A^2} \]  

where \( L \) is the assumed length of the ET (3 cm), \( \mu \) is the viscosity of air, \( A \) is the cross-sectional area, \( \Delta P \) is the change in pressure, \( Q \) is the volumetric flow rate, and \( \Gamma_v \) is a generalized hydraulic-geometric shape factor, which can be calculated from any arbitrary shape (see Ref. 15 for details). \( R_v \) therefore, depends on both the cross-sectional area and the shape of the ET lumen and can be used to quantify the degree of opening, where a large \( R_v \) indicates minimal ET opening.

Calculation of \( R_v \) via Eq. 2 requires a reasonably uniform cross-sectional lumen shape and area along the length of the ET. Previous histological measurements (25, 26) have demonstrated that axial variations in the lumen’s cross-sectional shape and area, within the deformable cartilaginous region, are minimal, and therefore the use of Eq. 2 is justified. Note that \( R_v \) is consistent with the standard definition of resistance, i.e., the ratio of pressure to volumetric flow rate, and can therefore be compared with active flow resistance measurements obtained experimentally during the force-response test (3, 6).

Human experiments and model validation. Although a majority of the mechanical parameters utilized in the finite element models are

\[ G = \frac{E}{4(1 + v)} \]  

\[ \kappa = \frac{E}{3(1 - 2v)} \]  

Fig. 2. A: histological midcartilaginous image used to determine cartilage and OFT contours, location of cranial base (A), location of load-bearing surfaces (B to E), and insertion angles of TVPM [superior (θS) and inferior (θI)]. B: finite element representation of cartilage and OFT soft-tissue elements, fixed boundary conditions, and location and directions of applied muscle loads. FLVPD and FLVPE, magnitudes of the forces applied by the LVPM in the y-z plane on sections D and E, respectively; FTVP, total force generated by the TVPM in the y-z plane.
based on accepted literature values, the baseline magnitude of OFT tissue elastic modulus ($E_{\text{OFT}}$) requires validation. This validation was accomplished by simulating the dynamics of a recently developed oscillatory force-response test (15). In this test, the ET is passively opened via inflation of the ME. Once opened, an oscillatory airflow rate is imposed, and the resulting oscillations in pressure are recorded. These pressure-flow rate relationships can be analyzed with a mathematical model of flow in a collapsible tube to determine mechanical properties of the surrounding tissue (15) including the overall compliance defined as $dA/dP$, where $A$ is the cross-sectional lumen area, $P$ is the applied pressure, and $dA/dP$ represents the average slope of the area-pressure curve. This oscillatory force-response protocol was conducted in a separate group of healthy adult volunteers with no history of ME disease (age range of 33 ± 13 yr with no individual <21 yr old; 3 women and 5 men). Note that all subjects were examined by a qualified physician to document normal ME status before undergoing the voluntary myringotomy required to perform the oscillatory force-response protocol. This myringotomy (<1 mm) was performed by a qualified surgeon, and all myringotomies healed within 1 wk with no complications. Approval for obtaining these measurements, including the voluntary myringotomy, was obtained from the Human Rights Committee and Institutional Review Board at Children’s Hospital of Pittsburgh. Analysis of this data resulted in an average compliance of $7.02 ± 6.31 \times 10^{-7} \text{cm}^2/\text{mmHg}$.

An analogous compliance parameter was obtained by modifying each histological subject’s finite element model to account for experimental conditions. Specifically, unrestrained deformation boundary conditions and zero muscle forces were specified on surfaces B to E to emulate passive conditions. Loading was accomplished by applying a range of normal pressures, corresponding to experimental values (200–400 mmHg) (15), on internal lumen surfaces. This loading resulted in an opened lumen cross-sectional area, which was calculated as a function of the applied pressure. Note that the current finite element models do not account for time-dependent viscoelastic effects, and thus the area calculation is independent of the rate of pressure application. For baseline mechanical properties of cartilage elastic modulus ($E_{\text{cart}}$) = 3.4 MPa and $E_{\text{OFT}}$ = 0.5 MPa, analysis of all eight finite element models yielded an average area-pressure slope of $dA/dP = 7.71 \times 10^{-7} ± 6.53 \times 10^{-7} \text{cm}^2/\text{mmHg}$. This value is in excellent agreement with the experimental measurements and therefore validates the magnitude of $E_{\text{OFT}}$ under baseline conditions.

**RESULTS**

The finite element models generated from each histological subject are shown in Fig. 3. Free-form meshing of the cartilage region resulted in an average of 1,017 ± 220 triangular elements per model, whereas meshing of the OFT region resulted in an average of 515 ± 77 elements per model. The average number of nodes used to create each model was 4,752 ± 615. In addition to the morphometric variability demonstrated in Fig. 3, histological measurements of the TVPM insertion angles resulted in $\theta_2 = 55.7 ± 18.2^\circ$ and $\theta_1 = 25.1 ± 15.0^\circ$. For all subjects, surfaces A to E could be accurately identified, and therefore all models presented in Fig. 3 were analyzed for tissue deformation and ET lumen opening.

The tissue deformation and lumen opening that occurred in a representative subject (subject e37) is shown in Fig. 4. These results utilize baseline mechanical properties of $E_{\text{cart}} = 3.4$ MPa, $E_{\text{OFT}} = 0.5$ MPa, $F_{\text{LVPD}} = 5$ N, $F_{\text{LVPE}} = 10$ N, and three specific $F_{\text{TVP}}$ loading conditions (60, 120, and 180 N), which correspond to the range of muscle forces measured during human jaw movement (8). The outlines in Fig. 4 represent the undeformed configuration, whereas the contour plots represent the degree of stretching within the soft tissue as measured by the effective strain (21)

$$\epsilon_{\text{eff}} = \sqrt{\frac{2}{3}(\epsilon_y^2 + 2\epsilon_{yz}^2 + \epsilon_z^2)}$$

$\epsilon_{\text{eff}}$ accounts for relative elongation in the $y$ and $z$ directions ($\epsilon_y$ and $\epsilon_z$) and in-plane shearing ($\epsilon_{yz}$) but does not account for rigid body rotations. Figure 4 demonstrates that, at nominal $F_{\text{TVP}}$ magnitudes, lumen opening is relatively small compared with the opening observed at higher $F_{\text{TVP}}$ values. Figure 4 also demonstrates that a majority of tissue stretching occurs along the lateral wall of the OFT and that the magnitude of this stretching increases with the applied force. Finally, minimal stretching or strains develop in the medial lamina of the cartilage, indicating that deformation in this tissue is primarily a rotation about the fixed cranial base.

The finite element models shown in Fig. 3 were used to investigate how specific tissue mechanical properties influence active ET opening phenomena. Specifically, several simulations were performed by either varying $F_{\text{TVP}}$, $E_{\text{OFT}}$, or $E_{\text{cart}}$ independently from the baseline conditions of $E_{\text{cart}} = 3.4$ MPa, $E_{\text{OFT}} = 0.5$ MPa, $F_{\text{LVPD}} = 5$ N, $F_{\text{LVPE}} = 10$ N, and $F_{\text{TVP}} = 50$ N. Figure 5A demonstrates that increasing the magnitude of the TVPM forces results in a significant decrease in $R_v$ due to a large opening in the cross-sectional lumen area (see Fig. 4). In contrast, Fig. 5, B and C, demonstrates that increasing the Young’s modulus of the OFT and the cartilage results in a stiffer ET, which is more difficult to open, and thus an increase in $R_v$.

**DISCUSSION**

An impaired ability to open the ET and increase the cross-sectional lumen area during swallowing has been related to the development of OM. Although several anatomic and mechanical soft tissue properties may be responsible for this debility, previous investigations could only speculate on how these structural properties influence ET opening phenomena. The goal of this study was to overcome this limitation by developing finite element models of the tissue deformation processes
responsible for ET opening. These models were constructed using histological data from eight normal adult subjects, and model parameters were validated by comparing theoretical and experimental cross-sectional area-pressure relationships. These 2D finite element models were then used to simulate active ET opening by applying muscle forces to the appropriate soft-tissue surfaces. Opening of the ET lumen was quantified with a flow resistance parameter, and a parameter variation study was conducted to determine the influence of various tissue mechanical properties on opening phenomena.

The models developed in this study (Fig. 3) demonstrate significant variability with respect to tissue size and minimal variability with respect to tissue morphology, except for a large OFT height-to-width ratio in subject e38. For equivalent mechanical properties, simulation of ET opening in these models resulted in significant variability in $R_v$ (see Fig. 5A; $R_v$ at $F_{TVP} = 60$ N). This variability in $R_v$ was correlated to the variability in tissue size by performing a least-squared regression between the closed lumen length in the cross-sectional $y$-$z$ plane and the magnitude of $R_v^{-1}$ calculated under baseline conditions.

Fig. 4. Simulation of tissue deformation and ET opening during swallowing in subject e37 using a total applied TVPM force of 60 N (A), 120 N (B), and 180 N (C). Contour plots represent the effective strain or stretching within the soft-tissue elements, whereas the outlines represent the undeformed configuration.

Fig. 5. Variation in resistance to airflow ($R_v$) in the actively opened ET lumen as a function of $F_{TVP}$ (A), OFT tissue elastic modulus (B), and cartilage elastic modulus (C).
conditions. This analysis, which resulted in a statistically significant correlation ($r^2 = 0.63, P < 0.02$), indicates that for constant tissue mechanical properties, ETs with long lumens are easier to open (smaller $R_v$) and that the magnitude of $R_v$ is strongly influenced by tissue size. Although the magnitude of $R_v$ varies significantly between subjects, Fig. 5 indicates that all models exhibit similar trends with respect to $R_v$, as a function of $F_{TVP}$, $E_{OFT}$, and $E_{cart}$. These similarities may be due to the consistent tissue morphologies demonstrated in Fig. 3. However, future studies, which model tissue deformation in OM patients with varying morphologies, are required to ascertain the precise relationship between tissue morphology, tissue mechanics, and ET function. In particular, a detailed understanding of how tissue mechanical properties influence opening phenomena under different anatomic conditions may be useful in identifying which tissue-based treatments may be most effective in patient populations with different morphological properties (e.g., cleft palate, chromosomal aberrations, and chronic OM patients).

Unlike previous histological measurement techniques, the present mathematical models are able to simulate and quantify the tissue deformation processes responsible for ET opening as a function of the applied muscle load. For example, Fig. 4A indicates that inferior aspects of the cartilage’s medial lamina undergo a medial-superior rotation during muscle contraction. This rigid body rotation, which is consistent with in vivo endoscopic observations (22), is not accompanied by significant tissue stretching or deformation, as indicated by the near-zero strains in the medial lamina. In contrast, Fig. 4, B and C, indicates that large tissue strains (~50–75%) develop in the OFT region at $F_{TVP}$ greater than ~120 N. Although these large $F_{TVP}$ values lead to a significant reduction in $R_v$ (see Fig. 5A), the large tissue strains could also result in tissue remodeling, which typically involves tissue fibrosis and hyperplasia (17), and an increase in $E_{OFT}$, which would impair ET opening (see Fig. 5B). In addition, the eight healthy adult subjects described in Human experiments exhibited an average in vivo active resistance of 1.9 ± 0.64 mmH2O·mL⁻¹·min⁻¹ [measured via a standard force response test (3)]. Therefore, a modest increase in $F_{TVP}$ to ~60 N (see Fig. 5A) may be sufficient to relieve ET dysfunction without creating excessive tissue strain.

Figure 5 demonstrates that several tissue mechanical properties can alter ET opening behavior. For example, a reduction in $R_v$, which indicates greater ET opening, can be accomplished by either increasing the applied muscle forces (Fig. 5A) or decreasing the Young’s modulus of the OFT and cartilage tissue (Fig. 5, B and C). However, equivalent changes in these mechanical properties do not produce equivalent changes in $R_v$. Therefore, the following sensitivity parameters were calculated to evaluate the relative importance of $F_{TVP}$, $E_{OFT}$, and $E_{cart}$ where $\Delta_{TVP}$, $\Delta_{OFT}$, and $\Delta_{cart}$ were calculated from data in Figs. 5 and represent the amount of change in $R_v$ for an order of magnitude change in the mechanical parameters. The results of this analysis ($\Delta_{TVP} = 440\times, \Delta_{OFT} = 48\times, \Delta_{cart} = 9.4\times$) indicate that ET opening phenomena are highly sensitive to the applied muscle forces, whereas $\Delta_{OFT} > \Delta_{cart}$ indicates that lumen opening is more sensitive to OFT properties than to cartilage properties. Although muscle forces appear to be the most important factor, it is important to recall that excessive $F_{TVP}$ could also result in large tissue strains as shown in Fig. 4.

In addition to analyzing data from OM-prone populations, future studies will include several model enhancements that could increase the utility of these computational models. First, although the 2D approximations made in the present models may be appropriate due to the relatively uniform lumen cross-sectional area reported in the literature (25, 26), these models cannot account for variations in cartilage or OFT morphology along the length of the ET and the complex 3D structure of the TVPM (see Fig. 1). Because these 3D morphological features could significantly influence the accuracy of our predictions, future studies should focus on developing and analyzing sophisticated 3D finite element models of the ET soft-tissue structure. It is important to note that, although 3D models may be more accurate, the present 2D models may be more clinically relevant since they would be easier to construct using nonhistological imaging data with lower resolution (i.e., MRI).

The present modeling techniques also do not account for the adhesive and/or surface-tension forces within the ET mucosa. As a result, these models may only be appropriate when the adhesive properties are minimal. For example, during the force-response test of active ET function, muscle forces are applied to the soft tissues after the lumen has been partially opened via air inhalation (6). In contrast, the lumen is completely collapsed before the application of muscle forces during normal swallowing events. Under these physiological conditions, the adhesive forces within the mucosa, which would be elevated during inflammation, serve to keep the ET closed until the stress at the lumen surface exceeds a yield value. As a result, any changes that diminish the stress transmitted to the lumen, such as a reduction in $E_{OFT}$, could hinder ET opening. The incorporation of mucosal adhesion dynamics into the present tissue deformation models would therefore result in a more accurate description of ET opening phenomena under inflammatory conditions. Finally, the time-dependent or hysteresis pressure-flow behavior observed in experiments (15) can be accounted for by incorporating viscoelastic material properties, whereas the quantity of air/fluid transported during ET openings can be ascertained by incorporating fluid-structure modeling techniques.

In conclusion, we have developed 2D finite element models of the tissue deformation processes that govern ET opening. These modeling techniques were able to account for both anatomic variability in normal subjects and the interaction between the anatomic and mechanical environment. These models, which were validated by comparing theoretical and experimental data, were used to simulate the muscle-assisted opening of the ET during swallowing. Flow-resistance measurements were used to quantify this opening, and results indicate that ET opening is highly sensitive to the applied muscle forces and relatively insensitive to cartilage elastic properties. These models have therefore identified how differ-

\[ \Delta_{TVP} = \frac{R_v \text{ at } F_{TVP} = 20 \text{ N}}{R_v \text{ at } F_{TVP} = 200 \text{ N}} \]

\[ \Delta_{OFT} = \frac{R_v \text{ at } E_{OFT} = 5 \text{ MPa}}{R_v \text{ at } E_{OFT} = 0.5 \text{ MPa}} \]

\[ \Delta_{cart} = \frac{R_v \text{ at } E_{cart} = 34 \text{ MPa}}{R_v \text{ at } E_{cart} = 3.4 \text{ MPa}} \]
ent tissue elements alter ET opening phenomena, which elements should be targeted for treatment, and the optimal mechanical properties of these tissue constructs. Further development of these computational models will help elucidate how modifications in the anatomic and mechanical structure of the ET influence its function and, therefore, may be useful in the evaluation of novel structure-based treatment therapies for OM.

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