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Effect of airflow and material models on tissue displacement for surgical planning of pharyngeal airways in pediatric down syndrome patients

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ABSTRACT

Pharyngeal narrowing in obstructive sleep apnea (OSA) results from flow-induced displacement of soft tissue. The objective of this study is to evaluate the effect of airflow parameters and material model on soft tissue displacement for planning surgical treatment in pediatric patients with OSA and Down syndrome (DS). Anatomically accurate, three-dimensional geometries of the pharynx and supporting tissue were reconstructed for one pediatric OSA patient with DS using magnetic resonance images. Six millimeters of adenoid tissue was virtually removed based on recommendations from the surgeon, to replicate the actual adenoidectomy. Computational simulations of flow-induced obstruction of the pharynx during inspiration were performed using patient-specific values of tissue elasticity for pre and post-operative airways. Sensitivity of tissue displacement to selection of turbulence model, variation in inspiratory airflow, nasal airway resistance and choice of non-linear material model was evaluated. The displacement was less sensitive to selection of turbulence model (10% difference) and more sensitive to airflow rate (20% difference) and nasal resistance (30% difference). The sensitivity analysis indicated that selection of Neo-Hookean, Yeoh, Mooney-Rivlin or Gent models would result in identical tissue displacements (less than 1% difference) for the same flow conditions. Change in pharyngeal airway resistance between the rigid and collapsible models was nearly twice for the pre-operative case as compared to the post-operative scenario. The tissue strain at the site of obstruction in the velopharyngeal airway was lowered by approximately 84% following surgery. Inclusion of tissue elasticity resulted in better agreement with the actual surgical outcome compared to a rigid wall assumption, thereby emphasizing the importance of pharyngeal compliance for guiding treatment in pediatric OSA patients.

1. Introduction

Airway narrowing is likely to occur when forces promoting airway collapse exceed the forces in soft tissue tending to dilate the airway (Bilston and Gandevia, 2014). Narrow airways or extremely compliant tissue supporting the pharynx enhance the probability of sleep disorders such as snoring and obstructive sleep apnea (OSA) (Suratt et al., 1984). Moreover, negative pressures generated during inspiratory airflow have been hypothesized to result in upper airway collapse (Roberts et al., 1985). P_{crit} is the critical value of transmural pressure resulting in complete pharyngeal obstruction (Patil et al., 2004). The number of events per hour when airflow reduces or ceases to flow in the pharynx is referred to as the Apnea-Hypopnea index (AHI) (Ruehland et al., 2009). High AHI or positive values of P_{crit} indicate greater OSA severity. Pharyngeal collapse in pediatric patients with Down syndrome (DS) results from hypotonia, obesity, midface hypoplasia, relative macroglossia, large tonsils and adenoids (Shott et al., 2006). Collapsibility of upper airway structures is more pronounced in DS patients with OSA during sleep, due to decreased muscle tone as well as supine position (Fleck et al., 2013). Commonly used first line treatment

Abbreviations: OSA, Obstructive sleep apnea; AHI, Apnea Hyponea Index; DS, Down syndrome; CPAP, Continuous Positive Airway Pressure; CFD, Computational fluid dynamics; FSI, Fluid-structure interaction; CSM, Computational structural mechanics; IRB, Institutional Review Board; MR, Magnetic resonance; GV, Gray value; RANS, Reynolds Averaged Navier Stokes; SST, Shear Stress Transport; LES, Large eddy simulations; DNS, Direct numerical simulations

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of OSA in children include non-surgical options such as continuous positive airway pressure (CPAP) (Waters et al., 1995) and mandibular advancement or surgical procedures such as tonsillectomy, adenoidectomy and glossectomy (Ayappa and Rapoport, 2003). The aforementioned surgical treatment options have only had moderate success rates (Shott et al., 2006). An enhanced understanding of the complex interaction between pharyngeal airflow and supporting tissue would be necessary to guide treatment and improve surgical outcomes.

Computational fluid dynamics (CFD) assumes a rigid pharynx and has been employed previously to calculate airflow patterns and variations in luminal pressure distribution in patients with OSA (Mihaescu et al., 2008; Mylavarapu et al., 2009; Xu et al., 2006) and normal subjects (Wootton et al., 2014). CFD studies to obtain dynamic pressure and shear stress distributions on the pharyngeal wall in pre and post-operative airways have been performed previously (Luo et al., 2014; Mihaescu et al., 2011). Studies involving CFD have also described the use of 'virtual' surgery to assess the efficacy of a surgical procedure prior to application (Mylavarapu et al., 2013). Fluid-structure interaction (FSI) simulations have been performed to assess the effect of airflow on the displacement of soft tissue to simulate snoring (Pirnar et al., 2015) and velopharyngeal narrowing (Zhu et al., 2012). Furthermore, FSI studies have also been employed to calculate airflow patterns in the pharynx following nasal surgery (Wang et al., 2012) and mandibular advancement (Zhao et al., 2013). Besides, the aforementioned studies both limited the analysis to the soft-palate or the mucosal layer supporting the pharynx and employed elasticity values that were not subject-specific. Three-dimensional (3D) computational structural mechanics (CSM) simulations of pharyngeal narrowing in human (Carrigy et al., 2016) and animal models (Xu et al., 2009) with subject-specific mechanical properties have been performed for static loads and do not consider the effects of airflow.

The present study describes a computational methodology to simulate narrowing of the pharynx in a pediatric DS patient with OSA using patient-specific 3D geometries of airway tissue and corresponding stiffness values, in response to airflow. We describe the sensitivity of tissue displacement to flow parameters, selection of material models and discuss the importance of this analysis to guide surgical procedures. Additionally, we illustrate the significance of including tissue elasticity to obtain an improved correlation between the simulated surgery and actual treatment outcome.

2. Materials and methods

2.1. Patient clinical history, sleep and imaging studies

Institutional Review Board (IRB) approval was obtained for the diagnostic sleep and imaging studies described in this section. The patient analyzed in the study is a 3 year old male with DS and moderate OSA (AHI: 6.8, P_{crit}: −790 Pa) (Subramaniam et al., 2016a, b). Electroencephalogram, electrooculogram, electromyogram, electrocardiogram, desaturation index and AHI were recorded using computerized overnight polysomnography. A light anesthesia was provided using dexmedetomidine to reduce muscle tone and simulate natural sleep with minimal respiratory depression (Nelson et al., 2003). Magnetic resonance (MR) images of the patient were obtained using a 1.5-THDxt MRI scanner (software version 16; General Electric), following adequate depth of sedation. The patient's head and neck were placed in a vascular coil and head motion was minimized by securing the patient's head using adhesive tape. MR scans were acquired in the supine position. Specific imaging parameters included acquisition matrix 256 by 256; echotrain length 64; slice interval 0.8 mm and slice thickness 1.6 mm (Subramaniam et al., 2016a). CPAP was administered using a face mask approved by IRB. A normalized inspiratory flow signal was obtained using a modified BiPAP device (Phillips Respironics Inc.) as described previously (Subramaniam et al., 2016b). Imaging was performed during the expiratory phase of the breathing cycle at three pressures: zero mask pressure and two values of CPAP that were based on airway caliber and collapsibility. An end inspiratory respiratory triggering was employed and the acquisition time was approximately 5 min, depending on the efficiency of respiratory triggering (Fleck et al., 2013). The linear portion of the curve relating negative pressure to the normalized flow was extrapolated to zero flow to compute P_{crit} (Patil et al., 2004). A statistical analysis was performed to evaluate the association between airway collapsibility and measures of OSA severity for a group of 31 pediatric OSA patients with DS (Prasad et al., 2016). Linear regression plots indicated a strong correlation between apnea-hypopnea index and fractional collapse of the airway. The pre to post-operative change in AHI was thereby employed to estimate the effectiveness of the adenoidectomy recommended for the patient described in this study.

2.2. Estimation of tissue stiffness

Patient-specific stiffness of soft tissue was estimated iteratively using a previously developed computational method involving finite element simulations of airway collapse (Subramaniam et al., 2016b) and is briefly described in this section. Outline of the tissue domain was identified manually using two-dimensional axial scans of the patient's airway. An in-house MATLAB (Mathworks Inc., Natick, MA) code (Subramaniam et al., 2016a) was employed to automatically identify the airway outline. The profile of the airway corresponding to a CPAP value of 350 Pa was used as the starting point of the simulation. The geometric model was meshed using GAMBIT v2.4 (ANSYS Inc., Canonsburg, PA) and exported in the Gmsh format (Geuzaine and Remacle, 2009) using the mesh conversion utility available in the Elmer finite element software (Järvinen et al., 2008). Grid independent solutions without excessive mesh distortion was obtained using an optimum element size of 0.5 mm. Non-linear behavior of the airway tissue was modeled using a nearly incompressible Neo-Hookean hyperelastic material (Subramaniam et al., 2016b) described using the following strain energy function (W),

$$W = \frac{G_0}{2}(I_1 - 3) + \frac{2}{K_0}(J_{el} - 1)^2 \tag{1}$$

where, G_0 is the shear modulus, K_0 is the bulk modulus, I_1 is the first invariant of the right Cauchy-Green strain tensor and J_{el} is the elastic volume strain. The negative transmural static pressure difference $\delta P_{\text{collapse}} = P_{\text{crit}}$ − CPAP was applied as a linearly increasing load on the airway wall. Self-contact of the airway wall was modeled as a sliding contact and the posterior wall adjacent to the spine was constrained in all directions. Symmetry boundary conditions were applied in the axial direction to constrain the motion to two-dimensions. Gravitational forces were equilibrated and not considered in the governing equations since the images were acquired in supine position. Finite element simulations of airway collapse were performed using the FEBio v2.5 (Maas et al., 2012) implicit finite element solver. The shear modulus was iteratively changed from an initial guess value of 2420 Pa obtained using MR elastography (Brown et al., 2015) until the airway collapsed completely and the difference between the slope of the experimental (linear regression co-efficient: 0.96) and numerical (linear regression co-efficient: 0.99) pressure-flow curves (Fig. 1a) was minimized. Accordingly, the value of shear modulus obtained using the iterative method was equivalent to 5800 Pa (~140% higher than the starting value). For the patient analyzed, the difference between the slopes was 4%. As can be seen in Fig. 1b, a U-shaped collapse pattern was observed for this patient. A detailed description of the slope matching procedure and airway collapse patterns in pediatric OSA patients with DS is described elsewhere (Subramaniam et al., 2016b).

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