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Simulation of upper airway occlusion without and with mandibular advancement in obstructive sleep apnea using fluid-structure interaction

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ABSTRACT

Obstructive Sleep Apnea (OSA) is a common sleep disorder characterized by repetitive collapse of the upper airway (UA). One treatment option is a mandibular advancement splint (MAS) which protrudes the lower jaw, stabilizing the airway. However not all patients respond to MAS therapy and individual effects are not well understood. Simulations of airway behavior may represent a non-invasive means to understand OSA and individual treatment responses. Our aims were (1) to analyze UA occlusion and flow dynamics in OSA using the fluid structure interaction (FSI) method, and (2) to observe changes with MAS. Magnetic resonance imaging (MRI) scans were obtained at baseline and with MAS in a known treatment responder. Computational models of the patients' UA geometry were reconstructed for both conditions. The FSI model demonstrated full collapse of the UA (maximum 5.83 mm) pre-treatment (without MAS). The UA collapse was located at the oropharynx with low oropharyngeal pressure (−51.18 Pa to −39.08 Pa) induced by velopharyngeal jet flow (maximum 10.0 m/s). By comparison, simulation results from the UA with MAS, showed smaller deformation (maximum 2.03 mm), matching the known clinical response. Our FSI modeling method was validated by physical experiment on a 1:1 flexible UA model fabricated using 3D steriolithography. This is the first study of airflow dynamics in a deformable UA structure and inspiratory flow. These results expand on previous UA models using computational fluid dynamics (CFD), and lay a platform for application of computational models to study biomechanical properties of the UA in the pathogenesis and treatment of OSA.

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1. Introduction

Obstructive sleep apnea (OSA) is a rapidly increasing sleep disorder affecting at least 17% of adults and 2% of children (Young et al., 1993, Ali et al., 1993, Young et al., 2005). It is characterized by repetitive episodes of complete (apnea) or partial (hypopnea) collapse of the upper airway during sleep (Force, 1999). A range of treatment modalities is available to treat the disorder, including weight loss, continuous positive airway pressure (CPAP), and various types of surgery (eg. uvulopalatoharyngoplasty, maxillomandibular advancement) (Riley et al., 1989, Croft and Golding-Wood, 1990, Sher et al., 1996, Barnes et al., 2004). Mandibular Advancement Splints (MAS) are an alternative approach for OSA treatment. MAS devices hold the lower jaw in a protruded position and help to prevent the upper airway (UA) from occluding (Cistulli et al., 2004). MAS have a high acceptance as they are comparatively easy to use, but have similar health benefits compared to CPAP, which is attributable to better treatment adherence (Gotsopoulos et al., 2002, Ng et al., 2003, Phillips et al., 2013). Although 60–70% of patients benefit clinically from MAS, a complete treatment to OSA (apnea-hypopnea index (AHI) < 5/h after treatment) is achieved in 35–40% of patient (Chan et al., 2010). The mechanisms of UA collapse in OSA, and why some patients respond to MAS treatment while others do not, are not well understood. A key unresolved issue is the factors that influence treatment outcome, and whether it is feasible to predict treatment response before implementation of treatment. Computational modeling shows promise as a method to understand treatment responses in OSA.

Computational fluid dynamics (CFD) has recently been introduced in OSA research by modeling the UA flow. Compared with idealized

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UA models used in early attempts (Heenan et al., 2003, Martonen et al., 2002), anatomically-accurate upper airway models have been proven to be advantageous in capturing important individual flow characteristics (Collins et al., 2007). These patient-specific UA models are reconstructed from imaging scans such as Computerized Tomography (CT) and magnetic resonance imaging (MRI), and knowledge of the response of the particular intervention (De Backer et al., 2007, Huynh et al., 2009, Zhao et al., 2013). However, rigid wall boundary assumption limits these CFD models, by not taking into account the interaction between UA flow and UA wall movement, which occurs in OSA patients (Xu et al., 2006, Jeong et al., 2007, Wang et al., 2009, Van Holsbeke et al., 2011).

There have been a limited number of studies (particularly with experimental validation) utilizing FSI in OSA research, with one of the first being an idealized 2-D tongue model in 2007 (Chouly et al., 2008), which later developed into a 3-D tongue model in 2011 (Rasani et al., 2011). Early studies also used FSI to model soft palate movement within a realistic UA geometry (Sun et al., 2007). A comparison of uvula displacement from pre- and post-surgical cases using FSI simulations was able to indicate the effect of nasal surgery on OSA (Wang et al., 2012). Although promising, the previous realistic UA FSI models have not been experimentally validated.

In terms of MAS therapy, previous studies have used CFD simulation to model the treatment effects on UA flow parameters which related to pressure drop and flow resistance (Vos et al., 2007, Van Holsbeke et al., 2011). These parameters were found to accurately reflect clinical response with MAS in the patient sample (Zhao et al., 2013). We now extend these CFD models by developing patient specific FSI models to incorporate UA collapse. The entire UA structure is simulated to be fully coupled with the flow field. We aimed to simulate upper airway occlusion in OSA and the effect of MAS therapy in a patient responsive to this form of therapy. Additionaly, our method will be validated experimentally by a physical model.

2. Material and methods

2.1. Reconstruction of geometry and meshing

A male adult OSA patient (52 years, BMI 29.4 kg/m^2) with severe OSA (AHI 41.5 events/hour on overnight polysomnograhy) who was a complete responder to treatment (post-treatment AHI < 5/h or no OSA) was selected from participants in a previous MAS research study using upper airway Magnetic Resonance Imaging (MRI) (Chan et al., 2010). Written informed consent was obtained for use of the patient data and all procedures have been described previously (Chan et al., 2010). The MRI protocol has previously been described in detail (Zhao 2013, Chan 2010). Briefly, T1 weighted UA scans were obtained with the patient in standardised head position during quiet nasal breathing. Scans of the UA were obtained without and with MAS, which represented the pre- and post- MAS treatment conditions. These two scans were imported into a commercial medical image post-processor (Amira 5, Visage Imaging, United States) to segment the UA volume (Fig. 1). The UA space was segmented on axial image slices from the level of the hard palate to vocal cord using automated thresholding of pixel values corresponding to air (Chan 2010, Schwab RJ et al. 2003, Am J Respir Crit Care Med 168:522-530). UA surface models were reconstructed for the two UA models and a vertex-averaged smoothing algorithm was applied. The boundary vertices were recalculated according to the average position of their adjacent vertices, while the UA geometrical charactristics were preserved. Table 1 lists the major dimensional changes of the UA geometry by the MAS. Significant enlargement of the restricted cross-sectional area near the softpalate was found in the UA (with MAS).

UA surface models were developed into 3-D UA volumes for the fluid domains and 2 mm thick airway wall models for the solid domains. The 2 mm wall thickness was selected as representative of the minimum thickness of the surrounding soft tissue which was between the posterior UA wall and spinal cord. According to our previous work (Zhao et al., 2013), refined unstructured hybrid volume meshs with 1.3 million and 1.6 million elements were generated respectively for fluid regions in each condition. The UA solid wall mesh used structured hexahedral elements (Fig. 2). An additional grid convergence study was perfomed to determine the most efficient mesh for the wall. Four meshes with increasing grid points were included in this study. The result show 20,000 hexahedral computational grids (solid 186, ANSYS, United States) were adequate for the pre-treatment UA structure. For the post-treatment model, a 27,000 hexahedral element mesh was sufficient (Fig. 3).

2.2. Governing equations

The estimated range of Reynolds numbers (Re) indicated the UA flow to be laminar or transitional. The standard k–ω shear stress transport (k–ω SST) model was proven to be appropriate to simulate this complex flow (Zhao et al., 2013). The main governing equations of the fluid domain can be expressed as:

$$\frac{\partial u_i}{\partial x_i}=0 \tag{1}$$

$$\rho u_i\frac{\partial u_j}{\partial x_i}=-\frac{\partial P}{\partial x_i}+\frac{\partial}{\partial x_i}\left[\mu\left(\frac{\partial u_i}{\partial x_j}+\frac{\partial u_j}{\partial x_i}\right)-\rho\overline{u'_i u'_j}\right] \tag{2}$$

where u is the fluid velocity, ρ is the fluid density, P is the fluid pressure, μ is the dynamic viscosity, i and j represent the Cartesian coordinates. The fluid and solid domains are coupled through the stress tensor. The Cauchy stress tensor in an isotropic Newtonian fluid field is

$$\sigma_{ij}^f=-p\delta_{ij}+2\mu\left(\varepsilon_{ij}-\frac{1}{3}\varepsilon_{kk}\delta_{ij}\right), \tag{3}$$

where $\sigma_{ij}^f$is the fluid stress tensor,$\varepsilon_{ij}$is the rate of the strain tensor which is given by:

$$\varepsilon_{ij}=\frac{1}{2}\left(\frac{\partial u_i}{\partial x_j}+\frac{\partial u_j}{\partial x_i}\right) \tag{4}$$

In the structural field, the governing equation of linear elasity are:

$$\frac{\partial\sigma_{ij}^s}{\partial x_j}+F_i=\rho\frac{\partial^2 D_i}{\partial t^2} \tag{5}$$

where F_i is the body forces, $D_i$is the structural displacement, $\sigma_{ij}^s$is the solid strss tensor, t is time.

The Cauchy stress tensor in the structural field is defined as:

$$\sigma_{ij}^s=\frac{Ev}{(1+v)(1-2v)}\varepsilon_{kk}\delta_{ij}+\frac{E}{(1+v)}\varepsilon_{ij} \tag{6}$$

where $E$is the Young's modulus, v is the Poisson's ratio, $\varepsilon_{ij}$is the strain tensor Eq. (4).

Table 1
Summary of upper airway geometrical changes with MAS. Data is shown as percent change from baseline with MAS.

Patient	AHI	Volume(m^3)	Restricted area (m^2)
Without MAS	41.5	12.5×10^{-5}	2.04×10^{-5}
With MAS	2.1	16.3×10^{-5}	3.30×10^{-5}
% change	94.9%	30.4%	61.8%

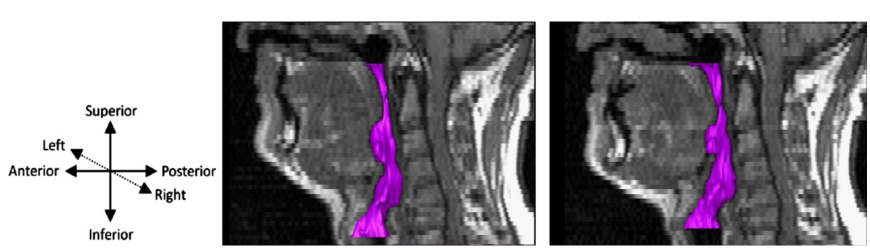

Fig. 1. The 3-D reconstruction (purple) of UA geometries before and after MAS treatment, imposed on a 2-d MRI slice. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

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