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Computational fluid-structure interaction simulation of airflow in the human upper airway

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ABSTRACT

Obstructive sleep apnoea syndrome (OSAS) is a breathing disorder in sleep developed as a consequence of upper airway anatomical characteristics and sleep-related muscle relaxation. Fluid-structure interaction (FSI) simulation was adopted to explain the mechanism of pharyngeal collapse and snoring. The focus was put on the velopharyngeal region where the greatest level of upper airway compliance was estimated to occur. The velopharyngeal tissue was considered in a way that ensures proper boundary conditions, at the regions where the tissue adheres to the bone structures. The soft palate with uvula was not cut out from the surrounding tissue and considered as an isolated structure. Both, soft palate flutter as well as airway narrowing have been obtained by 3D FSI simulations which can be considered as a step forward to explain snoring and eventual occlusion. It was found out that during the inspiratory phase of breathing, at given elastic properties of the tissue and without taking gravity into consideration, velopharyngeal narrowing due to negative suction pressure occurs. Furthermore, soft palate flutter as the main attribute of snoring was predicted during the expiratory phase of breathing. The evaluated flutter frequency of 17.8 Hz is in close correlation with the frequency of explosive peaks of sound that are produced in palatal snoring in inspiratory phase, as reported in literature.

1. Introduction

Obstructive sleep apnoea syndrome (OSAS) is a highly prevalent sleep-related breathing disorder (Peppard et al., 2013) characterised by recurrent episodes of complete or partial pharyngeal collapse during sleep, causing cessations of breathing (apnoeas) or reductions in airflow (hypopnoeas), despite ongoing respiratory effort. The two most common complaints of OSAS are excessive daytime sleepiness attributable to fragmented sleep and loud snoring (American Academy of Sleep Medicine, 2014; Kavčič et al., 2013).

Given the "Venturi tube" shape of the pharynx, the Bernoulli pressure principle plays a major role in snoring, based on CT images analysis (Fajdiga, 2005). If greater pharyngeal narrowing was the unique factor, computational fluid dynamics (CFD) would have been sufficient tool to investigate the OSAS phenomenon. Extensive CFD studies of airflow dynamics in the human upper airway using patient-specific modelling have been conducted (Jeong et al., 2007; Žun et al., 2008; Mihaescu et al., 2008; Cheng et al., 2014), some also concerning methodology validation (Xu et al., 2006; Mylavarapu et al., 2009; Mihaescu et al., 2011; Zhao et al., 2013a).

On the other hand, searching for the first principles of snoring, an attempt was made to idealise uvula flutter by solving the mechanical problem with a flexible plate or a thin membrane model in an unbounded fluid flow which is fixed at one edge and exposed to the transmural pressure (Huang, 1995; Connell and Yue, 2007 for example). According to Patil et al. (2007), upper airway patency depends on mechanical loads which impact on the pharyngeal wall and the compensatory neuromuscular response, indicating a complex interaction between the two. The upper airway is compliant and under certain conditions is prone to collapse, particularly in the velopharynx. The assumption of a rigid wall boundary in the CFD model or a thin membrane flapping instability in an unbounded flow is therefore inappropriate or insufficient and anatomically accurate 3D fluid-structure interaction (FSI) analysis can be recognised as a promising method to investigate pharyngeal collapse in sleep apnoea. Several contributions employing and discussing fluid-structure interaction to simulate airflow through upper airway models have been published so far, but none to the best of our knowledge obtained soft palate flutter. For instance, Sun et al. (2007) built finite element (FE) models of the upper airway and soft palate to analyse

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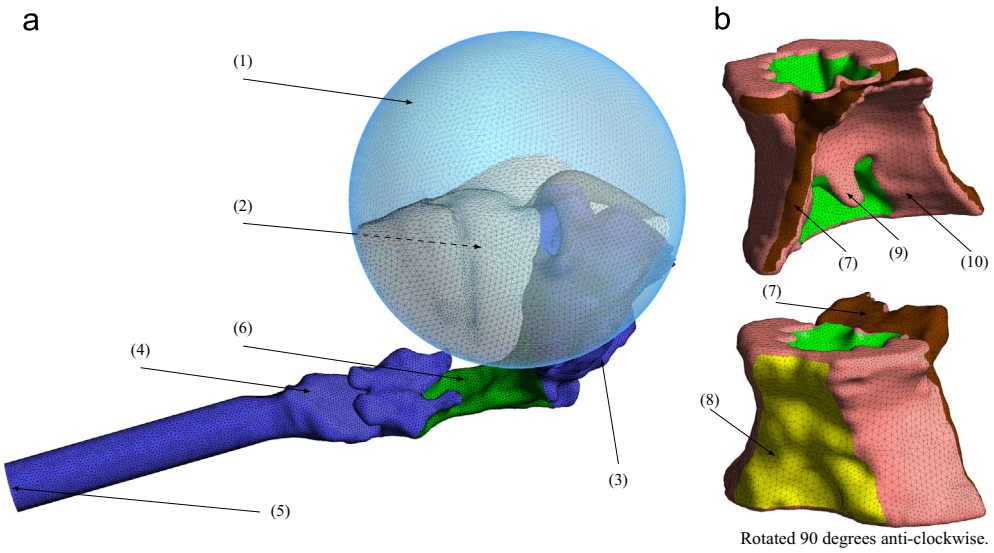

Fig. 1. Meshed 3D models of the (**a**) upper airway and (**b**) velopharyngeal soft tissue with identified boundary regions. (1) ambient, (2) face, (3) nasal cavity, (4) larynx (wall), (5) trachea, (6) wall of the velopharyngeal space (interface region), (7) bone structures, (8) soft tissue area connected to spine bone, (9) uvula, (10) tissue. (For interpretation of the references to colour in this figure, the reader is referred to the web version of this article.)

differences in the airflow field and tissue movement in apnoeic and non-apnoeic subjects. They pointed out that a higher velocity gradient and pressure difference in the narrowed velopharynx of the OSAS patient induce additional airway constriction and a decrease in static pressure that might lead to obstruction, resulting in snoring and apnoea. Kim et al. (2010) investigated airflow in the human upper airway via uni-directional FSI simulation. A pressure field from a large-eddy simulation was applied as the load in the structural analysis and the largest wall deformation was predicted to occur in the larynx. In the work by Zhu et al. (2012), passive movement of the human soft palate was studied assuming a laminar airflow regime. They pointed out the soft palate displacement is caused predominantly by the force exerted by static pressure, since it is much larger than shear force in all three directions. In the simulation they conducted, the soft palate movement was insignificant during the inspiratory phase and more profound in the expiratory phase when the soft palate moved towards the posterior pharyngeal wall. Huang et al. (2013) developed a 3D finite element model of the upper airway considering corresponding head and neck soft tissues. The simulation pointed out significant posterior direction movement during the inspiratory phase and much less profound in the expiratory phase, contrary to Zhu et al. (2012) results.

Powell et al. (2011) observed a strong correlation between airflow improvements predicted by CFD and polysomnography findings in patients before and after maxillomandibular advancement (MMA) surgery. Mandibular advancement splint (MAS) treatment also enlarges the pharyngeal volume and stiffens surrounding tissue and its success was assessed by Zhao et al. (2013a, 2013b) using CFD and FSI simulations, respectively. Qualitatively, their predictions correlated well with the known patients' clinical response. Zhao et al. (2013b) found the highest negative pressure to be in the velopharynx, but oropharyngeal collapse was predicted to occur, since the pharyngeal tissue was modelled as a homogeneous tube with a constant wall thickness. Although FSI simulation provided additional information about the pharyngeal deformation pattern, it should be acknowledged that the results were affected by the tissue geometry (i.e. soft palate had no free movement and high curvature in the velopharynx prevented its occlusion).

In distinction to the most of the data available from the literature, the velopharyngeal tissue geometric model has been reconstructed in our FSI model in a way, where the soft palate with uvula is not cut out from the surrounding tissue and considered as an isolated structure. This ensures anatomically correct boundary conditions at the regions where the tissue adheres to the bone structures. Maximum soft palate posterior movement of 1 mm was identified in our case during inspiratory phase in the narrowest velopharyngeal cross-section. Airway narrowing during the inspiratory and widening during the expiratory phase of breathing are consistent with our in vivo observations. They are in contrast to Zhu et al. (2012) and in accordance with Huang et al. (2013) results. Besides narrowing, soft palate flutter was identified which can be considered as a step forward towards numerical simulation of coupled phenomena like snoring and repeated occlusion.

2. Materials and methods

2.1. Segmentation of CT image data and 3D model reconstruction

Raw 3D upper airway and soft tissue (surrounding the velopharynx) models were reconstructed from CT images of a patient using the open source medical image segmentation tool ITK-SNAP developed by Yushkevich et al. (2006). The CT image in-plane voxel size was 0.52 mm with a slice thickness of 0.8 mm. The procedure of CT imaging is described thoroughly in Koren et al. (2008). The threshold value of −300 Hounsfield units (HU) was chosen in order to perform an automatic segmentation of the upper airway. The isolated region was visually inspected and manually repaired if necessary. Soft tissue was segmented by hand. Models were afterwards imported in MeshLab (Visual Computing Lab ISTI-CNR), where irregular geometric features (e.g. holes, spikes) were removed and the surface was smoothed using the built-in algorithm. The meshed models with identified boundary regions are shown in Fig. 1. Fig. 1a represents the geometrical model of the airflow region, whereas Fig. 1b represents soft tissue region (coloured in ochre) with identified bone structures (coloured in yellow and brown) and FSI interface (coloured in green), respectively.

In the final step, the unstructured tetrahedral meshes with 740 k and 155 k computational cells were generated in ANSYS ICEM CFD (Release 15.0, 2013) in order to spatially discretise fluid and structural domain, respectively. A cross-sectional view of the meshed regions in the mid-sagittal plane is shown in Fig. 2. Six prism layers were generated adjacent to the airway wall to meet the turbulence model's near-wall spacing requirement. Maximum y^+ value was 0.8. A SOLID 185 element was used for the modelling of the soft tissue structure.