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Modeling of the fluid structure interaction of a human trachea under different ventilation conditions☆

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ABSTRACT

This work is focused on the analysis of the response of the tracheal wall to different ventilation conditions. Thus, a finite element model of a human trachea is developed and used to analyze its deformability under normal breathing and mechanical ventilation. The geometry of the trachea is obtained from computed tomography (CT) images of a healthy man. A fluid structure interaction approach is used to analyze the deformation of the wall when the fluid (in this case, air) moves inside the trachea. A structured hexahedral-based grid for the tracheal walls and an unstructured tetrahedral-based mesh with coincident nodes for the fluid are used to perform the simulations with the finite element-based commercial software code (ADINA R & D Inc.). The tracheal wall is modeled as a fiber reinforced hyperelastic solid material in which the anisotropy due to the orientation of the fibers is introduced. Deformation of the tracheal walls is analyzed under different conditions. Normal breathing is performed assuming a sinus shape of the pressure at the inlet and air speed at the outlet based on real data which represent the inspiration and the expiration processes respectively. Mechanical ventilation is simulated as smooth square shape velocity airflow considering positive values of pressure using data from a mechanical ventilation machine. Deformations of the tracheal cartilage rings and of the muscle membrane, as well as the maximum principal stresses in the wall, are analyzed. The results show that, although the deformation and stresses are quite small for both conditions, forced ventilation does not exactly imitate the physiological response of the trachea, since with always positive pressure values the trachea does not collapse during mechanical breathing.

1. Introduction

The upper human airway is the primary duct for inspiration in the breathing process. Air entering the mouth, passes through the pharynx and flows into the trachea via the glottal region. The trachea is able to adapt itself to regulate the pressure during the different ventilation conditions such as breathing, sneezing or coughing. The main components that constitute the trachea are the cartilaginous rings, and the muscular membrane that runs longitudinally and posteriorly to the trachea. The main role of the tracheal cartilaginous structures is to maintain the windpipe open despite the inter-thoracic pressure during the respiratory movements. Smooth muscle contraction and transmural pressure generate bending and tensile stresses in the cartilage and collapse it to regulate the air flow and modulate the diameter of the airway. Although a better understanding of how this process is performed and how the implantation of a prostheses affects the response of the trachea is not only important but challenging, few studies have analyzed the behaviour of the trachea under different ventilation conditions. This is especially relevant for instance in patients that have to be mechanically ventilated or after implantation of a tracheal prostheses. Therefore, understanding of the breathing process or forced ventilation in healthy trachea, and how the airflow and stresses in the airway wall are distributed, are essential in order to design more convenient prostheses or mechanical ventilation techniques. Most of the developed numerical studies in the respiratory system till now have analyzed the airflow pattern using rigid walls and approximated airways geometries [4,18,25]. Cebral and Summers [4] developed a CFD model using a 4-generation geometry. They studied the central tracheal and bronchial airways by using a virtual bronchoscopy reconstruction method. Nowak et al. [25] demonstrated a four-subunit CFD simulation method for the human tracheobronchial tree. They economized on computational effort by segmenting the first 12 generations into four generation “tranches”. Ma and Lutchen [18] developed a 6-generation model in which the flow pattern was predicted under turbulent conditions using zero-pressure outlet conditions. Only few studies are based on an accurate airway geometry coming from computed tomography (CT) or magnetic resonance (MRI) [16,23,37,38]. Moreover, these studies

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(both using simplified or real geometries) do not take airway deformation into account [1,3,13,14,18,25,36]. In particular, Zhang and Kleinstreuer [38] analyzed targeted aerosol drug deposition in a rigid triple bifurcation tracheobronchial airway model. Nithiarasu et al. [23] performed a steady state simulation of a human tracheal model while Liu et al. [16] analyzed the unsteady flow patterns in a patient-specific human healthy trachea during inspiration under a turbulent condition. One interesting analysis that analyzed the collapsibility of the trachea under different pressure conditions was developed by Costantino et al. [6] although they did not consider the fluid itself. Including this aspect, FSI studies in lower airway geometries have been only done in the lower cartilage-free generations of the lung and were restricted to simplified models/geometries [5,9–11,34]. The work developed by Wall and Rabczuk [33] is the first analyzing the behaviour of the trachea using a FSI analysis but using simplified constitutive models for the structural behaviour of the trachea. More recently, Koombua and Pidaparti [27] study the flow characteristics and stress distributions in the airways during inhalation. Using a FSI approach, they tested an isotropic and an orthotropic material model to represent airway flexibility. Regarding the constitutive behaviour of the tracheal walls, there is a large dispersion of the mechanical properties of the different tissues that compose the trachea, and only few studies have analyzed their mechanical behaviour for humans [28,29,31,35]. In most of these works, the isolated tracheal cartilage was considered to be a linearly elastic isotropic material. Moreover, many of the previous works studying the human tracheal smooth muscle dealt with its plasticity, stiffness and extensibility and the influence of the temperature on force–velocity relationships [8]. In a previous work [19], we showed smooth muscle deflections of a human healthy trachea undergoing simplified normal breathing and coughing conditions comparing the results with the introduction of a tracheal prosthesis. In this work, we presented a realistic constitutive model of the tracheal wall using a FSI approach to compute deformations and stresses of the trachea under different ventilation conditions. In particular, we put our attention in the muscular membrane deformation, which is critical for the physiological function of the human trachea. The final aim of this work is to contribute to a better understanding of the response of the human trachea during natural breathing and forced ventilation. Forced mechanical ventilation is in fact responsible for clinical complications as local inflammatory phenomena or postintubation stenosis [24,30]. This study is focused on the solid tracheal behaviour and its main goal is to achieve a basic view in the analysis of its response that has been only made very approximately or just clinically. Our interest is less directed toward local flow patterns (which are crucial for local particle deposition) than toward overall stress and strain characterization of the tracheal muscle (which are determinant factors for stenting technique). In this sense, this is the first step to create a tool for analyzing the healthy and pathological tracheas under different conditions in order to assess variables that cannot be measured *in vivo*, as pressure drop or wall stresses.

2. Materials and models

2.1. Solid model of the trachea

The finite element model of the human trachea was made based on a CT performed to a 70 year old healthy man. The DICOM files coming from the scan provide a clear picture of the black internal cavity (filled with air). A non automatic segmentation of the CT scan was accomplished to determine the real geometry of the trachea and to distinguish between the muscle membrane and the cartilage rings. To identify the tracheal tissues, the cartilage rings could be isolated through their higher density. With the help of MIMICS©, the different material densities could be distinguished through different tones of the gray scale (i.e. high density corresponds to white colour while low density corresponds to dark colour). For the thickness, very slight variations along the tracheal axis were detected, and therefore we chose to simplify it as constant. An IGES file of the segmented geometry was created to construct the associated computational grid. A full hexahedral mesh of around 50,000 elements was made using PATRAN© (see Fig. 1), where the different tissues are shown in different colours. In order to avoid significant variations between segmented geometry and finite element model a grid independence study was performed. Finally the average solid-grid element size was around 1 mm. To determine the properties of the different tissues of the trachea, different experimental tests were conducted. The goal of this paper is not the experimental characterization of the trachea, so, only a summary of the data acquisition is included. For further explanations see [19,32]. Human tracheas were obtained at the time of autopsy from two subjects (aged 79–82 years). Samples of cartilage and muscle were dissected from the tracheas for histology and mechanical analysis procedures. The histology revealed that in the cartilage rings, the collagen fibers run randomly. Therefore an isotropic material was used to define its behaviour. On the contrary, the muscular membrane presented two orthogonal families of smooth muscle cells, one running mainly longitudinally and the other transversely. Then, for this tissue a constitutive model that takes into account the anisotropy coming from the different fiber orientation was used. The specimens were mounted on the Instron MicroTester 5548 to perform tensile tests. For cartilage, since there is no preferential orientation, a Neo-Hookean model, with strain energy density function (SEDF) defined as $\Psi = C_1(\bar{I}_1 - 3)$, was used to fit the experimental results ($C_1 = 0.57$ MPa). Regarding the smooth muscle,

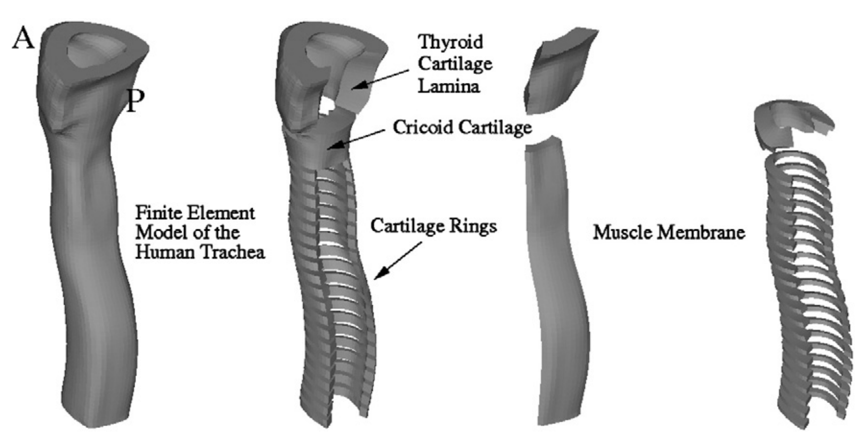

Fig. 1. Finite element mesh of the trachea. The cartilaginous parts of the trachea are plotted separately from the muscular membrane. (A denotes anterior part and P posterior part).