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Flow transport and gas mixing during invasive high frequency oscillatory ventilation

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ABSTRACT

A large Eddy simulation (LES) based computational fluid dynamics study was performed to investigate gas transport and mixing in patient specific human lung models during high frequency oscillatory ventilation. Different pressure-controlled waveforms (sinusoidal, exponential and square) and ventilator frequencies (15, 10 and 6 Hz) were used (tidal volume = 50 mL). The waveforms were created by solving the equation of motion subjected to constant lung wall compliance and flow resistance. Simulations were conducted with and without endotracheal tube to understand the effect of invasive management device. Variation of pressure-controlled waveform and frequency exhibits significant differences on counter flow pattern, which could lead to a significant impact on the gas mixing efficiency. Pendelluft-like flow was present for the sinusoidal waveform at all frequencies but occurred only at early inspiration for the square waveform at highest frequency. The square waveform was most efficient for gas mixing, resulting in the least wall shear stress on the lung epithelium layer thereby reducing the risk of barotrauma to both airways and the alveoli for patients undergoing therapy.

1. Introduction

A mechanical ventilator is commonly used in intensive care units to facilitate and support patient breathing during acute respiratory failure [1–3]. In many critical situations, as in acute lung injury and acute respiratory distress syndrome, usage of conventional mechanical ventilation (CMV) based therapy could lead to lung injury due to high tidal volumes of 6–10 mL/kg [1,4,5]. High frequency oscillatory ventilation (HFOV) is then considered a useful ventilation mode as it allows for adequate gas-exchange and prevents lung injury due to the usage of small tidal volumes of ~1–2 mL/kg [4], albeit at higher frequencies of 1–30 Hz [1,2,6]. Unlike CMV where the primary mechanism of flow transport is bulk convection, flow transport during HFOV is complicated and not fully understood [7,8]. Previous work [7–10] has identified several physical mechanisms that play a role in gas transport and mixing during HFOV: (a) bulk convection; (b) gas exchange between respiratory units or pendelluft; (c) counter flow; (d) longitudinal mixing; and (e) gas exchange by molecular diffusion. A detailed understanding of these mechanisms is essential for optimizing ventilation management strategies during HFOV and is the focus of this work.

Oscillatory flow and convective mixing through human lungs have been numerically investigated by Choi et al. [11] in a computed tomography (CT) based human airway model with HFOV conditions. Nagels and Cater [12] carried out a numerical simulation of HFOV in an idealized double bifurcation model and observed flow reversal near the wall due to residual motion of the fluid at different times of the ventilation cycle. Flow transport between main bronchus was by pendelluft mechanism, a feature also reported by Adler and Brucker [13]. Bauer and Brucker [14] studied, *in vitro*, airway reopening and observed that the collapsed airway can be recruited at a high Womersley number ($\alpha = a\sqrt{2\pi f/\nu}$, where a is airway radius, f is frequency and ν is kinematic viscosity). Secondary velocity was found to be highly dependent on geometry [15]. Heraty et al. [16] conducted *in vitro* investigation of gas exchange during HFOV in a single idealized and realistic bifurcation geometry and reported that bifurcated geometry features influence the flow leading to secondary flow structures. Hatcher et al. [17] studied mechanical performance of five clinically available neonate HFOVs and reported that ventilator performance varied widely for the models chosen even though the devices had similar operating parameters (the difference attributed to different waveform shapes). The presence of intubation during ventilation therapy causes a high speed-jet, which is released at the carina-trachea environment and induces shear stress at the airways walls. The wall shear stress (WSS) is seen to cause airways inflammation and epithelial erosion [18,19]. Green [18] studied WSS in an idealized model and found that the high flow rate at the peak expiration due

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to coughing caused a high wall shear stress, which subsequently leads to airways inflammation. Muller et al. [19] investigated experimentally the wall shear stress in a neonatal tracheal scaled model under high frequency jet ventilation (HFJV) by varying the location of the jets inside the endotracheal tube (ETT). High values of WSS were measured and it was observed that it doubled itself when the jet was moved closer to the wall from the center of the ETT.

The majority of *in vitro* and computational studies have looked at the effect of ventilation frequency or ventilation mode during HFOV. It is well-known that the HFOV waveform varies between clinically available models, and we hypothesize that this variation in the shape of ventilator waveform coupled with the frequency of the device plays an important role in gas transport and mixing. The current work is motivated by the need to study the effect of frequency (15, 10 and 6 Hz) and waveform shapes (sinusoidal, exponential, and square) for three commercially available high frequency ventilators.

2. Method

2.1. CT-scan based model reconstruction

Recent advances in medical imaging and availability of imaging processing algorithms allow extraction of patient specific lung models from CT scans [11,13,16]. A three dimensional human lung model (that of a 60-year-old female patient) was obtained by reconstructing CT scans. The scan parameters are listed in Table 1.

Table 1
Details of computed tomography (CT) scans.

Field of view	34.4 cm
Slice thickness	3 mm
Resolution (width × length)	512 mm × 512 mm
Pixel size	0.671 mm
Total images	260

3D reconstruction was performed using Mimics® software (Materialize Inc., Belgium) in which CT raw images (260 slices) were converted to appear in axial or transverse, sagittal and coronal planes (Fig. 1a). A region growing process was applied to eliminate all separated structures and a three dimensional geometry was created (Fig. 1b). Segmentation was performed subsequently to extract the final upper trachea-bronchial geometry that comprised seven consecutive generations from trachea (G0) to generation 7 (G7) with 53 outlets. Previous studies to predict flow phenomenon during HFOV have concluded that invasive devices like the endotracheal tube must be accounted for in order to capture the complicated flow physics [14]. An 8 mm dia. (interior) ETT [20] was thus chosen and added to the model to complete the lung geometry.

2.2. Numerical method and mesh generation

Transient three-dimensional high frequency oscillatory flow was solved using an implicit finite volume CFD solver (ANSYS® CFX13). Turbulence is modeled using a large Eddy simulation (LES)

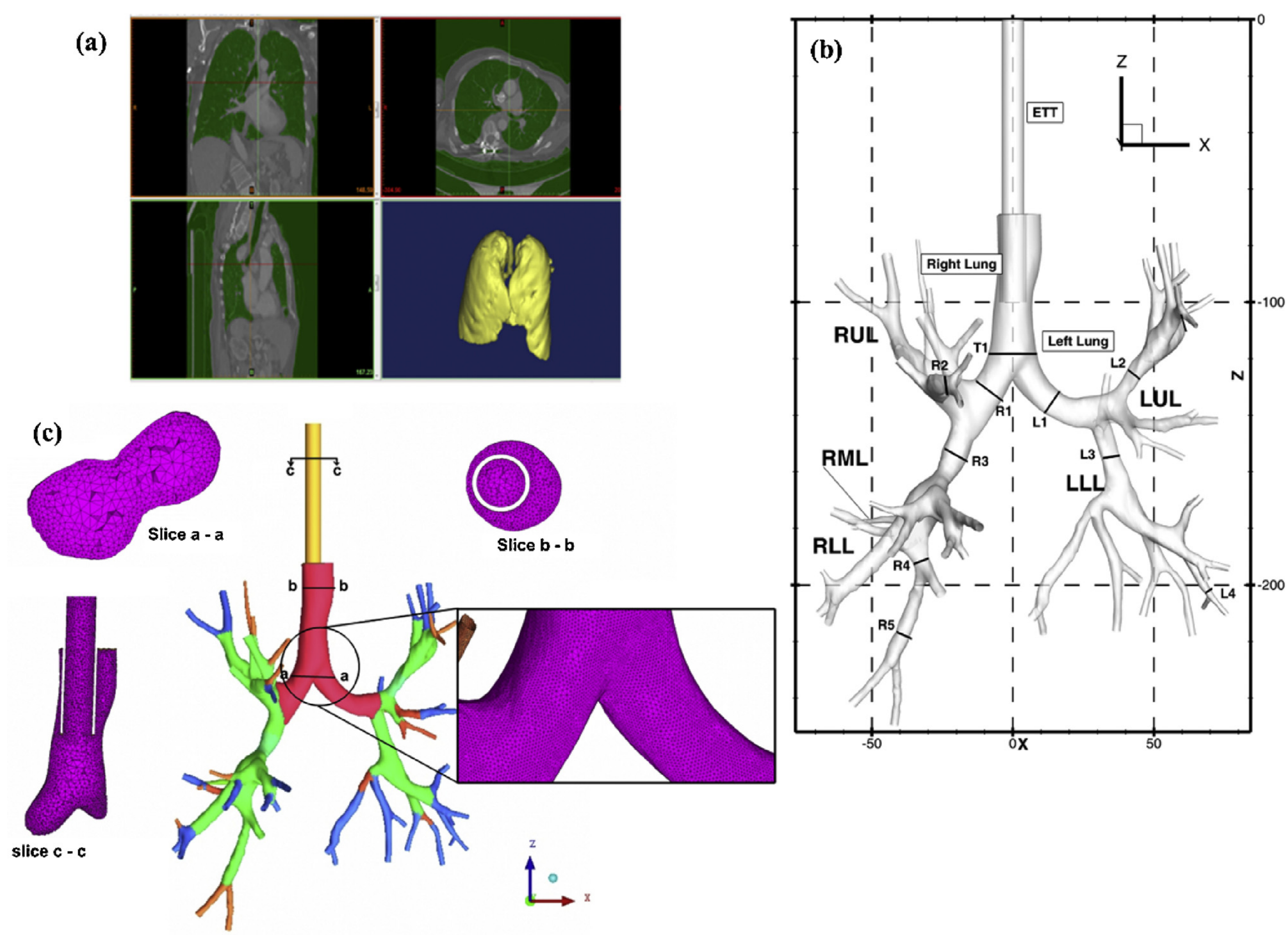

Fig. 1. (a) CT-scan images and the 3D reconstructed lung geometry; (b) final lung model with ETT; and (c) computational mesh. The expanded view displays the surface mesh. Slice a-a shows the volume mesh in a cut plane through the transition between parent and daughter branches in the first bifurcation. Slice b-b show the volume mesh in a cut plane through the trachea and the endotracheal tube. Slice c-c shows the volume mesh in a vertical cut plane through the endotracheal tube, trachea and main bronchi.