Article

Endotracheal Tubes Design: The Role of Tube Bending

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Abstract: Endotracheal tubes (ETT) passed inside the human trachea witness tube bending at different angles, affecting the local fluid flow dynamics. This induces a variable mechanical ventilation performance across patients' comfortability levels. Our understanding of the local fluid flow dynamics phenomena is thus crucial to enhance the maneuverability of ETT under operation. For the first time to our knowledge, we shed light on ETT through computational fluid dynamics (CFD) to investigate the bending effect of ETT on the local airflow in volume-controlled mechanical ventilation. We considered an ETT with 180° arc bend configuration, including Murphy's eye. We identified several flow phenomena associated with the bending, such as flow asymmetries, secondary flows, and vortex dynamics throughout the tube.

Keywords: endotracheal tube; mechanical ventilation; human health; human respiratory system; modeling and simulation; computational physics; fluid flow dynamics

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1. Introduction

Endotracheal tubes (ETT) are frequently used in the medical sector to provide mechanical ventilation to patients [1]. They are usually employed for long periods during anesthesia and surgical operations to give oxygen to the human lungs. They were also extensively used in intensive care units of hospitals during the last two years for COVID-19 patients. ETT is manufactured and delivered under different diameters and lengths. Their maneuverability is very complex and constitutes a susceptible procedure for their installation inside the human trachea. This is due to the bending of the tube that must fit well to the mouth-to-trachea internal topology.

ETT aims to cover a wide range of humans trachea tomographies. Farrow et al., 2012 [2] addressed the role of ETT size and its selection depending on the patient's physiology and conditions. Calderon et al., 2019 [3] investigated the mean flow and airway resistance in ETT of several diameters (6 to 10 mm) at average body temperature and constant pressure numerically. They showed that a single-sized ETT could facilitate endotracheal intubation (ET) without causing an increased airway resistance.

Recent research studies on ETT emerged during the last two years because of the COVID-19 pandemic [4–8]. Piazza et al., 2021 [9] shed light on long-term intubation and the high rate of tracheostomy in COVID-19 patients. Interestingly, they explained how this might determine an unprecedented increase in airway stenoses.

Thanks to advances in computer hardware and CFD methods and models, researchers can investigate the flow phenomena inside the human airways with considerable accuracy. For example, Kou et al., 2018 [10] studied the effect of the cough peak flow rate on airflow dynamics during a cough process in a CT (computerized tomography) scanned human respiratory airway model. Krenkel et al., 2010 [11] developed CFD simulations and experiments to investigate the impact of an ETT on the flow in a generic trachea (artificial lung ventilation). They shed light on models with bendings and connectors that induce more substantial secondary flows that can persist for long periods. Qi et al., 2014 [12] applied CFD simulations to study the airflow mechanisms in the trachea and main bronchi

for individuals with left pulmonary artery sling. Other scientists used numerical tools to investigate airflow in the human trachea at different conditions [13,14]. Mechanical ventilation (MV) has also been a recent topic of research by some investigators [15–20] in attempts to quantify the impact of different MV modes on the aerodynamics of the airways in specific patients.

Kingma et al., 2017 [21] conducted a recent comparison of four methods of endotracheal tube passage in simulated airways. They showed a need for improved techniques to enhance ETT design for better local airflow in the human airways (e.g., human trachea bifurcations) induced by artificial MV.

This study concerns CFD simulations to investigate the airflow dynamics in human trachea first bifurcation from an ETT with Murphy's eye [22–24]. We focus on the ETT vital role in inducing high local airflow circulations inside the trachea with symmetry and asymmetry features of the flow. This work shed light on the importance of appropriate manipulation of ETT during MV that physical practitioners and medical doctors usually manipulate.

2. Computational Methodology

We employed the three-dimensional (3D) transient compressible Navier–Stokes equations:

$$\frac{\partial \rho}{\partial t} + \nabla \cdot (\rho \mathbf{U}) = 0 \quad (1)$$

$$\frac{\partial (\rho \mathbf{U})}{\partial t} + \nabla \cdot (\rho \mathbf{U} \otimes \mathbf{U}) = -\nabla P + \nabla \cdot \tau \quad (2)$$

$$\frac{\partial (\rho E)}{\partial t} + \nabla \cdot [(\rho E + P) \mathbf{U}] = \nabla \cdot (\tau \cdot \mathbf{U}), \quad (3)$$

where ρ , $\otimes$, $\mathbf{U}$, P , E , τ represent the fluid density; tensorial product operator; velocity vector; pressure; internal energy and the shear stress tensor, respectively. The system of Equations (1)–(3) is discretized over the computational domain using the Finite Volume Method (FVM) [25,26] inside the open-source CFD code OpenFOAM® [27]. We model turbulence through the Reynolds-Averaged-Navier–Stokes in conjunction with the $k - \omega - SST$ model [28,29]. We conducted a mesh sensitivity analysis using three different grids. According to the Grid Convergence Index (GCI) by Celik et al. [30], we adopted a 3D non-uniform hexahedral mesh of about 1 million cells, including local mesh refinement. The trachea geometry is from a commercial provider of a scanned adult patient trachea, transformed into STL surface and treated to retrieve the zone of interest shown in Figure 1c. We applied an inhale–exhale breath with a transient flow rate profile (Figure 1b) imposed at the entry of the ETT extremity. This flow rate mimics a real volume-control (VC) mechanical ventilation signal. A VC ventilation cyclic mode is the most effective one and known to provide full mechanical ventilation. Applying mechanical ventilation with VC mode allows a better control such that for each inspiratory effort beyond a threshold, the ventilator will inject the initially predefined fixed tidal volume. The airflow temperature is imposed 22 °C at the entry and 33 °C at the ETT walls. An outlet pressure condition with atmospheric pressure value as a reference and an outlet zero Neumann temperature boundary condition were applied at the two outlets of the trachea (Figure 1). The trachea walls are assumed to be rigid walls with no-slip velocity conditions.

Figure 1. Endotracheal tube (ETT) 3D model of 8 mm internal diameter inserted into a human's trachea. (a) The computational model used in the simulations; (b) the inhale/exhale applied flow rate boundary condition that mimics a real volume-controlled (VC) mechanical ventilation (MV); (c) upper view of the ETT showing the trachea first bifurcation.

3. Results and Discussion

Figure 2 shows the local velocity streamlines in a cross-section post to the ETT extremity and before the bifurcation zone. At the end of the inhale period at $t = 0.6$ s (see Figure 2a), airflow asymmetry and small circulations near the trachea wall are observed. We have chosen those two times according to the flow rate cycle shown in Figure 1b. They represent the times at which the highest stresses can occur. The $t = 0.5$ s represents the maximum flow speed imposed during the inhalation period just before the exhalation period. The $t = 0.6$ s corresponds to the approximate maximum flow speed at the beginning of the exhalation period. At the beginning of the exhale period at $t = 0.6$ s (see Figure 2b), airflow asymmetry becomes less manifested with one big circulation (vortex) and the other three smaller vortices near the trachea wall. Note that strong vortices may induce damage to the trachea wall (e.g., inflammation, cracks), especially in patients with MV applied for long periods [9]. Figure 3 illustrates the local streamlines in a cross-section post to the ETT extremity and very close to the first bifurcation zone inside the trachea. At the end of the inhale period at $t = 0.5$ s (see Figure 3a), airflow symmetry and four medium to large vortices near the trachea wall are observed. At the beginning of the exhale period at $t = 0.6$ s (see Figure 3b), airflow symmetry is still prevailing, but with four small vortices at different positions compared to those observed at $t = 0.5$ s in Figure 3a.

Figure 2. Local velocity streamlines in a cross-section post to the ETT extremity and before the bifurcation zone. (a) Velocity streamlines at the end of the inhale period at $t = 0.5$ s; (b) velocity streamlines at the beginning of the exhale period at $t = 0.6$ s.

Figure 3. Local velocity streamlines in a cross-section post to the ETT extremity and very close to the bifurcation zone. (a) Velocity streamlines at the end of the inhale period at $t = 0.5$ s; (b) velocity streamlines at the beginning of the exhale period at $t = 0.6$ s.

Following the bifurcation zone, at the end of inhale period at $t = 0.5$ s, Figure 4a shows vital airflow circulations with two large vortices in each branch of the trachea. The local 3D streaklines and WSS are illustrated in Figure 5. Two large vortices at $t = 0.5$ s induce significant wall shear stress (WSS), as shown in Figure 5c. These two vortices per branch that were observed at the inhale period at $t = 0.5$ s become manifested, but only in the most significant branch addition alone at the beginning of the exhale period at $t = 0.6$ s (see Figure 4b).

Figure 4. Local velocity streamlines in a cross-section post to the ETT extremity and post to the bifurcation zone. (a) Velocity streamlines at the end of the inhale period at $t = 0.5$ s; (b) velocity streamlines at the beginning of the exhale period at $t = 0.6$ s.

The WSS values are in the same order of magnitude as those reported in the literature for an ETT (see Rhein et al., 2016 [15]). They are higher than the WSS values reported in the literature by Qi et al., 2014 [12] for a subject with a left pulmonary artery sling of the trachea (a numerical study without an ETT [12]). During the inhale phase, the ETT tube of Figure 1 induces more WSS values in the trachea zone following the first bifurcation (Figure 5c,d). In contrast, during the exhale period, it induces more airflow circulations in the trachea zone before the first bifurcation (Figure 5a,b). The flow direction reversibility

can explain the differences in the streaklines of Figure 5a,b, and thus its adaptation to different topologies with the flow direction. In Figure 5a, the flow goes from top to bottom (inhale), thus entering a bifurcation of two outlets, while in Figure 5b, the flow moves from bottom to top (exhale), thus entering Murphy's eye of the ETT.

Figure 5. Three dimensional velocity streaklines (a,b) colored by the axial velocity values U_z (m/s), and the local wall shear stress (c,d) values WSS (Pa) observed from a top view perspective. (a,c) At the end of the inhale period at $t = 0.5$ s; (b,d) at the beginning of the exhale period at $t = 0.6$ s. The small arrows in (c) indicate high local WSS values.

4. Conclusions and Perspectives

We shed light on the vital role of ETT bendings in affecting the local airflow in the human trachea during an MV under a VC mode. Employing CFD modeling and simulation, we showed complex local airflow phenomena inside the human trachea that can be observed easily within experiments. In this paper, we focused on an ETT with Murphy's eye and with an arc bending of 180° . Our computations revealed that the ETT tube with this bending angle during the inhale phase induces more WSS values in the trachea zone post to the first bifurcation. Furthermore, our numerical results showed that EET induces more airflow circulations in the trachea zone before the first bifurcation during the exhale period.

The present paper aimed only at raising the question of the bending tube issue through results that demonstrate the complex flow features such as the secondary vortices. At present, the medical practitioner's community are not aware that these effects exist and

can potentially cause harm to the patient. We aim at studying different configurations and provide a comparative study of different tubes and bendings in a future article.

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Abbreviations

The following abbreviations are used in this manuscript:

CFD	Computational Fluid Dynamics
CT	Computerized Tomography
ET	Endotracheal Intubation
ETT	Endotracheal Tube
FVM	Finite Volume Method
GCI	Grid Convergence Index
MV	Mechanical Ventilation
VC	Volume Controlled
WSS	Wall Shear Stress

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