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CFD MODELLING OF AIRFLOW IN HUMAN RESPIRATORY SYSTEM

Vivek Kumar Srivastav[1], Anuj Jain[1], Akshoy Ranjan Paul[1], Shrey Joshi[2]

[1]Department of Applied Mechanics, Motilal Nehru National Institute of Technology, Allahabad, India.
[2]Depatment of Mechanical Engg, Motilal Nehru National Institute of Technology, Allahabad, India

ABSTRACT

The information regarding the regional and local depositions of toxic/therapeutic particles in the human airways are useful for understanding the causes/cure of certain respiratory diseases. The aim of this study is to locate the region where particles/aerosol deposits. In this paper the airflow in human airway was numerically solved using a three dimensional model in FLUENT 6.3. The dimensions of the human airway was adopted from Luo et al.(2004). The computational work was performed for the inspiration velocity 2.03 m/s corresponding to a heavy breathing condition. The velocity distribution (velocity vector and velocity contour) and wall shear stress were obtained and used as a tool to predict the regional deposition of aerosol/particles in human airway. In this study, bifurcation region was focused to predict the deposition pattern of aerosol/particles in this region. This information is useful in targeted drug delivery for treatment of specific diseases and to diagnose the diseases at bronchial bifurcation

Keywords: Computational Fluid dynamics, Bifurcated Human Airways, Wall Shear Stress

1. INTRODUCTION

A human airway is similar to a Y-shaped bifurcated duct. For health point of view, it is important to understand the complicated airflow fields and local deposition of toxic/therapeutic particles in human lung airways [1]. During the heavy breathing turbulence is formed in upper respiratory tract [2]. The particles are deposited through the nose causes bronchial constriction or in medical word ‘asthma’ and large deposition of pollutant cause tissue injuries and sometimes, it is called lung cancer [3]. Generally, turbulent flows are recognised by eddy size, and particles (aerosol/toxic/drugs) flow/deposition is influenced by formation of large eddies.

Most of the researchers have computationally simulated particle deposition in human airways; however few of them have performed experimentally [13, 8, 17]. Previous studies in this filed show that the flow in bifurcation is most important part for diagnosis of respiratory diseases [2, 4, 5, 6]. The airflow inside the human airway is provided important information for clinical purpose (regional ventilation) and for the inhalation treatment [7] As the particles are entered with the air, so mechanics of airflow is very important to improve the drug delivery efficiency, medical therapy and health impact of inhaled toxic particle [8, 9].

Li et al. have predicted that most of the particles are deposited at the carinal ridge, however few particles at the outer wall due to secondary flow. They have found that most of the particles are deposited near the bifurcation (in the neighbourhood of trachea) due to cartilaginous ring structure. They have also observed that the deposition fraction in the left lung is higher than the right lung[8]. Zhao et al. have mentioned that the secondary swirls and axial velocity are due to high shear stress, which creates inner wall injury [10]. Luo et al. have used the LES modelling for airflow in single bifurcated human airways. Their results shows that LES modelling captured the eddy formation and flow separation at the corners, which is very useful for predicting the aerosol/particle deposition in the human respiratory system [2]. Xia et al. have discussed the wall shear stress of rigid and compliant realistic human airways. It was pointed that the maximum wall shear stress near the outlet due to narrow of the airways [16]. Luo et al. have discussed the particle deposition in the obstructed airways based on Weibel Model. They have found that the effects of particle deposition occurred at the downstream of the obstruction [18].

A large number of studies have focused on effect of wall shear stresses on the human airways. A relatively small group of these studies have focused on local (regional) effect of wall shear stress. Previous studies only focused on the medical aspect of wall shear stress near the bifurcation region but unable to predict the exact position where wall shear stress is maximum.

Although these previous studies, prediction of aerosol/particle is still not clear on the bifurcation. The current CFD simulation model focuses on prediction of regional deposition (along the radial and perpendicular cross section) of particle/aerosol and influence of bifurcation region. This study is provided valuable information of airflow patterns in the respiratory tract of different cross sections (radial and perpendicular cross sections). Flow field information is played very important role in correctly predicting aerosol/dust particle deposition in human respiratory system. Findings will help in targeted drug aerosol delivery and the health impact region of inhaled toxic particles.

2. AIRWAY GEOMETRY

The human airways model was adopted from Luo et al. [2]. The model contains only trachea and two bronchus; in addition model is considered as rigid and regular (fig.1). The bifurcation region is considered from cross section-3 up to cross section-6. Right bronchus is the right and left bronchus is the left of the patient.

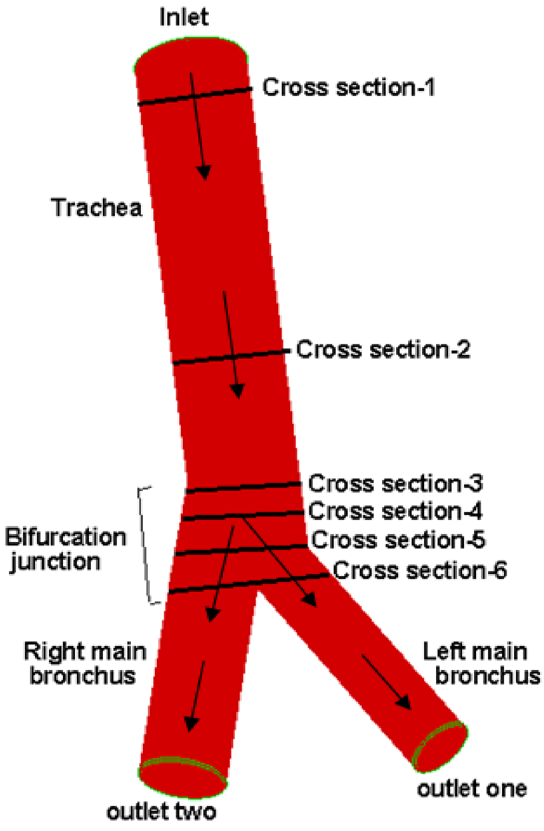

Fig 1. Schematic view of 3D human airways

3. BOUNDARY CONDITIONS

The boundary conditions used in this study are: inlet velocity, pressure outlet and fixed wall with no slip condition. The velocity inlet and pressure outlet was applied at the inlet and outlet section of the trachea and bronchus respectively. In addition, no-slip condition and no reflection of wall elasticity was assumed in this analysis. The properties of air at normal conditions was considered in this study i.e., density is 1.19 kg/m^3, the viscosity is 1.82×10^{-5} kg/m-s. In this study, an inspiratory inlet velocity of 2.03 m/s, which corresponds to high breathing rate, was adopted from Luo et al. [2]. For the inlet velocity, turbulence intensity= 5% and turbulence length scale = 0.0217 m have also taken from Luo et al. [2].

4. NUMERICAL SIMULATION

The Fluent 6.3 was used in this work for numerical simulation of airflow in realistic human airway. The SIMPLE algorithm was used for the pressure-velocity coupling. The different terms of the transport equation were discretized by using second order upwind numerical scheme. The realizable $k-\varepsilon$ turbulence model was selected to capture the airflow in the internal flow of the realistic human airway [2, 11, 12]. "Enhanced wall treatment" option in the Fluent solver was used to determine the flow near the wall. Mesh boundary layer was also created on the wall of trachea and the airway. The residual value of the governing equation was 10^{-4}, adopted from Luo et al [2]. The computation was performed on an IBM workstation with 8 GB RAM and Intel Xeon processor. A typical run time for the CFD simulation was approximately 4 hours. The results were found to converge after 1120 iterations with under-relaxation factors: pressure = 0.3, density = 0.5, body forces = 0.5, momentum = 0.4, turbulent kinetic energy = 0.1, turbulent dissipation rate = 0.1 and turbulent viscosity = 0.2.

5. RESULTS AND DISCUSSION

5.1 CFD Validation

The present result was validated with the CFD results reported by Luo et al. [2]. In the 3D analysis, the maximum velocity obtained was 2.66 m/s against the value 2.6 m/s observed by Luo et al. showing a deviation of 2.3 %.

5.2 Velocity Distribution

Fig.2 shows the velocity contour and velocity vector for inspiration flow at the cross sections-3 to 6. The mainstream velocities are located at the centre of trachea and bifurcation. At cross-section-6, high velocity is found at the top and bottom corner of right bronchus (Right Lung) and side-corner of left bronchus (Left Lung) while the region of low velocity is in wide area near the inner wall of left bronchus (Left Lung). This pattern of velocity contour suggests a larger deposition of particles/aerosol at the inner wall of the left bronchus (Left Lung) as compared to right bronchus (Right Lung).

Velocity vector

Cross section-3

Cross section-4

Cross section -5

Cross section -6

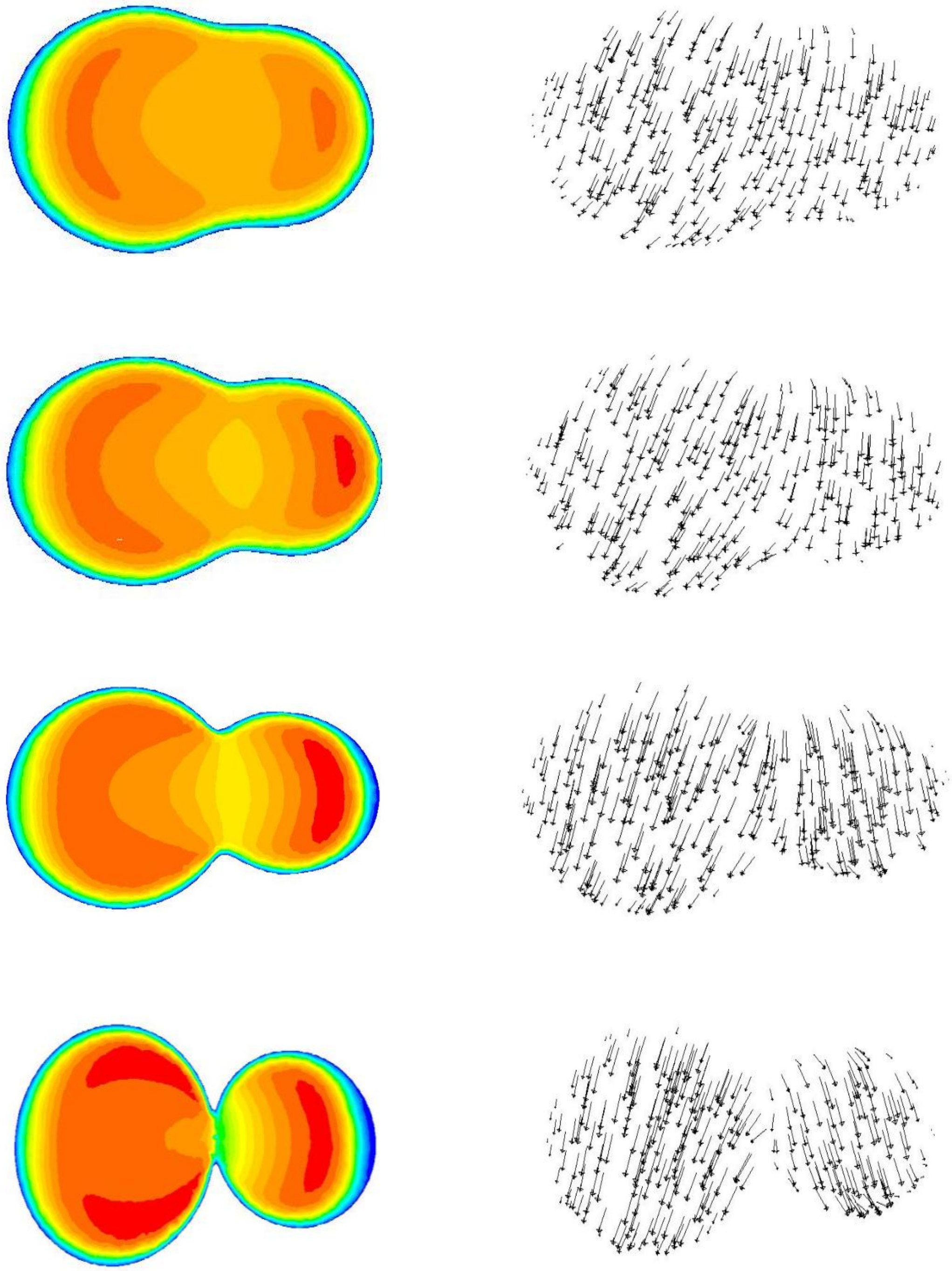

Fig 2. Velocity contour (m/s) and vector along the radial cross sections
Velocity contour

The fig.2 shows that the velocity vectors are turned in the downwards direction. The maximum velocities are obtained near the inner wall at the cross section-5, 6. Counter-rotating vortices are developed on the bifurcation (cross section-5 & 6), while co-rotating vortices are developed above the bifurcation (cross section-3, 4).

Fig.3 shows the velocity distribution along the perpendicular cross section. The magnitude of maximum velocity obtained was 2.66 m/s. Non uniform velocity contour is obtained at the bifurcation (fig.3), which indicates the flow separation at the curved surface of the bifurcation. The velocity vector and contour shows that the vortices are formed near the bifurcation due to curve boundaries. Stagnation point is observed at the corner (fig.3). It is found that the corners of the model introduced more flow disturbance because of the existence of corner vortices. However, CFD simulation of 3D model reveals that the flow separation zones occurring towards the curved portion of the bifurcation. The separation region is quite large, comparable in size to that observed in the bifurcation. Flow separation occurred at the bifurcation points, which leads zero velocity towards the wall of the left and right bronchus.

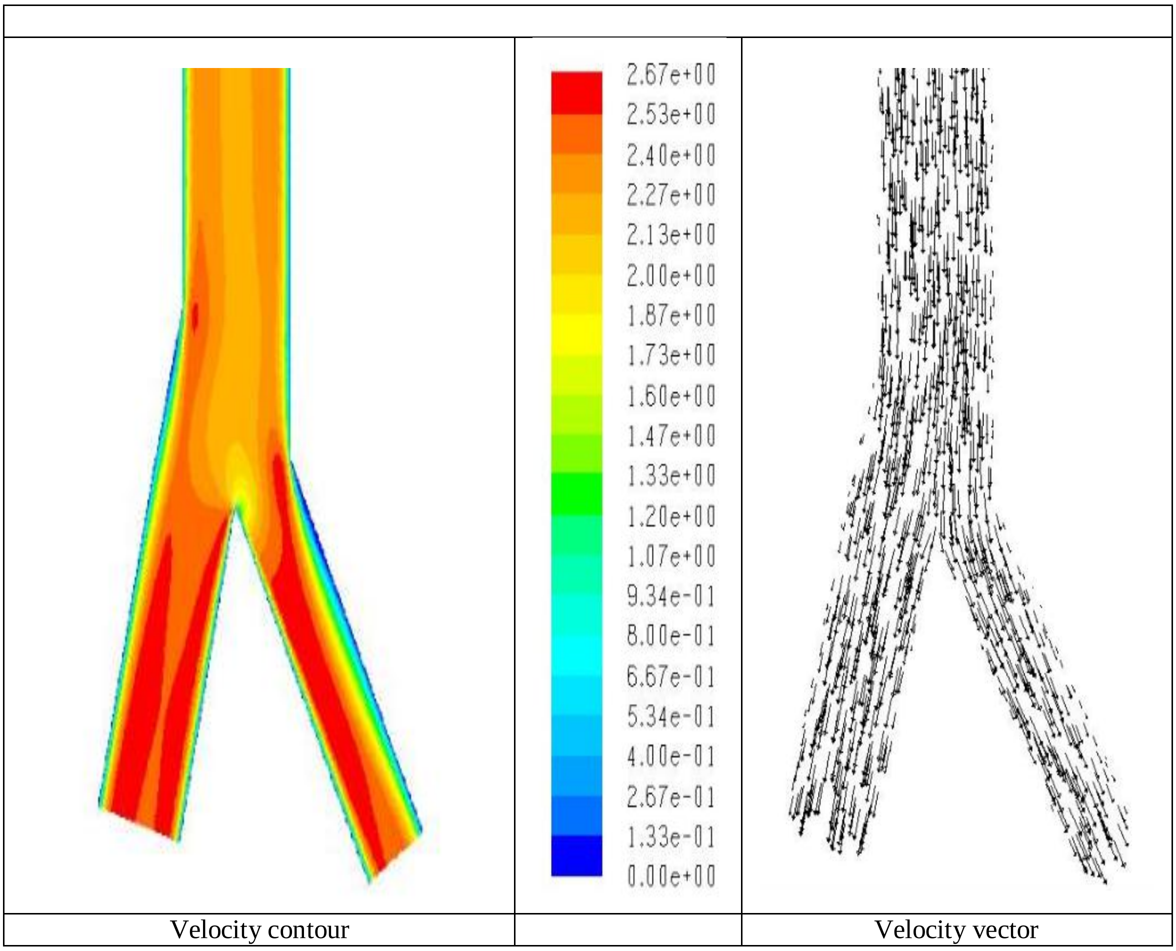

Fig 3 . Velocity contour (m/s) and vector of the perpendicular cross sections

5.3. Wall Shear Stress

The effect of wall shear stress on the walls of the airways plays a vital role for the respiratory diseases [13]. Shifting in the flow velocity maxima indicates that higher shear stress along the inner wall of the airways (cross section-6). The variation of local wall shear stress at different cross-sections has plotted in fig.4. The wall shear stress distribution at the cross-section-1 and 2 is more or less uniform whereas large variations are observed at the cross section-6 (fig.4). However as the air flow towards the bifurcation, at cross section-2 (around middle of trachea) low wall shear stress is noticed. The average wall shear stresses for the cross section-1, 2 and 6 are 0.179 N/m^2, 0.0433 N/m^2 and 0.092 N/m^2 respectively. It is found that the peak absolute (local) wall shear stress at cross section-6 (fig.4) is 0.44 N/m^2 occurring near the inner wall of the bifurcation.

The wall shear stress distribution at cross section-1 near the entry is uniform because of uniform velocity applied at the trachea. Maximum velocity gradient obtained at the inner wall of the bifurcation for cross-section-6 (fig.2 & fig.4) is responsible for increase in the wall shear stress at this location.

The high magnitude of wall shear stress suggests maximum particle deposition on the inner wall of the bifurcation (cross section-6 in fig.4). The high value of shear stress on the inner wall of the trachea and bifurcation indicates possibility of inner wall injury.

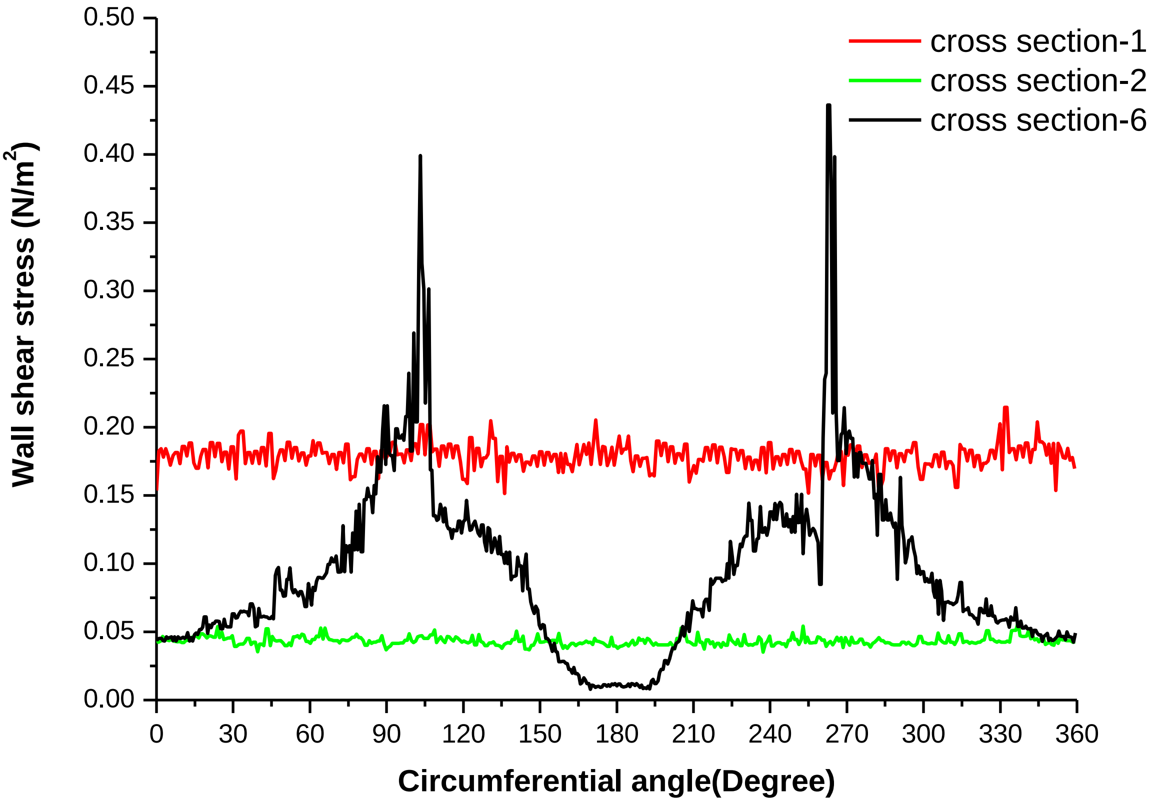

Fig 4. Wall shear stress vs circumferential angle

6. CONCLUSION

The airflow in human airways was numerically solved using the realizable k–ε turbulence model. CFD simulation of airway model shows the flow separation occurring near the walls of both bronchi. The realizable $k–\varepsilon$ turbulence model is used to capture the flow physics of the human airways during the heavy breathing. Thus it is concluded that: (1) the maximum velocity is reduced near the inner wall of left bronchus which is close to the bifurcation, this leads a maximum deposition at the inner wall of the left bronchus (Left Lung) rather than right bronchus (Right Lung). (2) Maximum wall shear stress at the bifurcation is due to the high velocity or high velocity gradient near the bifurcation. The computation of wall shear stress also indicates the maximum deposition of aerosol and its probable locations. This leads a further understanding of effective drug delivery, which holds both the medical and engineering aspects.

Thus, findings (predicting probable locations of toxic/therapeutic particles deposition and peak value of wall shear stress at bifurcation region) provide useful information for targeted drug delivery and deposition of toxic particles.

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8. MAILING ADDRESS

Vivek Kumar Srivastav
Department of Applied Mechanics
Motilal Nehru National Institute of Technology,
Allahabad-211004, India
Phone: +91-532-2271212
E-mail: vivekapril@gmail.com