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Numerical simulation of flow structure and deposition of particles in Asthematic Airway Bifurcation

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Abstract: Symmetric airways bifurcation corresponding to generation 12-23 of Weible's model will be investigated through numerical simulation. Parent's airway segment is modeled as a smooth circular tube while the child segment is to be considered as asthmatic airway bifurcations. The effect of size of the lumen area and number of folds on particle deposition and pressure drop will be investigated. The present work extends to deal with asymmetric airway extracted from 2-3 branches of the model of Weible's in order to more appropriately model human air passage.

Keywords: Asymmetric bifurcation flow, human lung, lumen area.

Introduction

Theoretical models of aerosol deposition in the lung usually treat the tracheobronchial tree as a network of pipes. An essential building block of this approach is the mathematical description of particle motion in a pipe flow. The understanding of respiratory aerosol dynamics and deposition characteristics in human airways are very useful for the effect of dose analysis. Previous studies [1,2] are focused on healthy subjects, while airways of patients with respiratory diseases. The central airways of the human lung are a complex system of branching tubes. A statistical analysis of human bronchial morphometry data revolved distinct asymmetric features of the airway diameter, branching angle, and flow division could significantly affect the local distribution of particle deposition within airway bifurcations and ratios of air pressure, velocity profiles, in each branch should be balanced.

An appertain of their aerodynamic behavior is important to the understanding of their physiological role. In respiratory disease asthma is characterize by intermittent shortness of breath, possibly due to allergy to common allergies, cold air, viral infection or a particular strong event. Usually these types of disease developed in childhood. A few studies have investigated the effects of asthma on particle deposition [3,4,5,6]. By decreasing the diameters of normal airways an artificial model of asthmatic airways was constructed earlier. This study shows that under similar conditions in asthmatic airways deposition of particles is high than in normal airways [11,12,19] The earliest studies of airflow in the lung airways were the experimental work by [7,8,9,10]. A few velocity profiles and flow patterns (mass, and heat transfer) were presented for a double bifurcation model.

In other experimental studies the central airway up to the third generation of the bifurcation was used [13,14,15]. It was conducted that flow patterns were most likely dependent on the airway geometry. In the above studies, the respiratory flow was treated as a steady or quasi-steady condition based on the Womerslly parameter for normal breathing. Airflow patterns change as a result of an acute asthma episode where the combination of bronchospasm, mucus plugging, and mucosal edema build up leads to increased airways resistance as the diameters of the airways are reduced. Experimental studies with human subjects have shown that patients with asthma produce different particle deposition patterns [17,18].However most of the studies have considered airflow patterns affected by tumors, airway constructions and airway blockage associated with chronic obstructive pulmonary disease (COPD) in local sub-regions of the lung airways[20,21,22].

A double symmetric airway bifurcation, corresponding to generations12-23 of Weible's airway model (1963),the use of realistic models then leads to a lack of exact matching of comparative/compatibility studies as the data does not necessarily exist in the literature unlike the data for Weible's mode[16]l. In this paper we study the flow structure and particle deposition pattern for normal and constructed airway bifurcations (supposed to be asthmatic), are compared and analyzed the effect of various parameters by using available experimental data.

Numerical Approach

It is known that essential variations occur in the anatomy of the human lung airways bifurcations. Based on data from literature, a 2-D model of the lung bifurcation is considered. The geometry of the normal lung airways bifurcations was derived from the

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available literature. The lung bifurcation model and the essential geometrical data are illustrated in fig. (1). the inlet diameter of the first generation (at trachea) is $D = 3.50\,mm$ and near the first bifurcation diameter is $D = 2.90\,mm, 1.786\,mm$. The angle between the bifurcations (up and below) is 65^0. This bifurcation angle ($\theta = 65^0$) is same for the further bifurcations. Near the second (up and down side) bifurcation the diameters are $D = 2.23\,mm, \& 1.64\,mm$.

Governing equations

Flow of air in human lungs is considered to be incompressible and Newtonian, (including dust particles) consisting of the unsteady 2-D Navier-stokes equation in cylindrical system. For the convenience, the model is described using cylindrical coordinates (r, θ, z), where z axis is considered to be the axis of symmetry i.e. considered to be the central axis. Three component of velocity, u_r along the radius vector perpendicular to the axis, u_θ perpendicular to the axis and the radius vector, and u_z ,parallel to the axis of z. For the axi-symmetric case $u_\theta = 0$, therefore the remaining velocity component u_r, and u_z, are independent of θ due to symmetry.

The governing equation of continuity and equation of motion are given by:

$$\frac{1}{r}\frac{\partial (r u_r)}{\partial r} + \frac{\partial u_z}{\partial z} = 0 \qquad (1)$$

$$\rho\left(\frac{\partial u_r}{\partial t} + u_r\frac{\partial u_r}{\partial r} + u_z\frac{\partial u_z}{\partial z}\right) = -\frac{\partial p}{\partial r} + \mu\left(\frac{\partial^2 u_r}{\partial r^2} + \frac{\partial^2 u_r}{\partial z^2} + \frac{1}{r}\frac{\partial u_r}{\partial r} - \frac{u_r}{r^2}\right) \qquad (2)$$

$$\rho\left(\frac{\partial u_z}{\partial t} + u_r\frac{\partial u_z}{\partial r} + u_z\frac{\partial u_z}{\partial z}\right) = -\frac{\partial p}{\partial z} + \mu\left(\frac{\partial^2 u_z}{\partial r^2} + \frac{\partial^2 u_z}{\partial z^2} + \frac{1}{r}\frac{\partial u_z}{\partial r}\right) \qquad (3)$$

The following non dimensional variables were used

$$u_r = \frac{\overline{u_r}}{u_0};\, u_z = \frac{\overline{u_z}}{u_0};\; p = \frac{\bar{p}}{\rho u_0^2};\;\; z = \frac{\bar{z}}{R_0};\;\; r = \frac{\bar{r}}{R_0};\; t = \frac{\bar{t}}{R_0}u_0 \qquad (4)$$

Where p is the pressure, ρ the density, μ the viscosity of the fluid (like air) in lungs, and u_0 is the time-average velocity at inlet (nose/mouth). At the cross section of the bronchi the dimensionless parameters are used;

- The Reynolds number of the flow: $R_e = \dfrac{\rho u_0 R_0}{\mu} = \dfrac{u_0 R_0}{\nu}$ (4.1)

Where $\nu = \mu / \rho$, is the kinematics viscosity of the fluid (air).

- The Womersley number $\alpha = R\sqrt{\omega / \nu}$ (4.2)

$\omega = 2\pi f\,(= 2\pi / T)$, The angular velocity for the Poseuilic flow $\alpha^2 \le 1, \nu = 0.159\,cm^2 / \sec$.

Boundary conditions:

The air which we inhale passes through the trachea divides the lungs into two main branches (i) the left bronchi, and (ii) right bronchi. Since air is compressible Newtonian and viscous including dust particles during breathing mechanism (inspiration and expiration) for a very short time(for each inhale and exhale) can be zero because dust particles filtered by the lungs and only pure air circulates through the whole body. Usually in trachea flow of air is turbulent for the fourth and fifth generations. The velocities of the lung airways are always smaller than the speed of sound so the effects of compressibility are negligible.

Numerical Approach

The Reynolds number in generation 12-23 is typically in the range 250-1600, depending on breathing conditions. The airflow therefore considered as incompressible and laminar. The equations (1)-(3) with the proper boundary conditions stated above are

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solved by means of the Finite Element Method (FEM) using ANSYS Multiphysics Version 11.0. The 3-D cylindrical geometry of human upper airway lung bifurcation is modeled as 2-D geometry (fig. 1.1). The model was created in the software and was meshed with its mesh generator with FLUID141 element. The model is meshed automatically using ANSYS 'mesh tool'. The mesh is created using 'mapped meshing' instead of 'free meshing' (fig. 1.2). Mapped meshing is used when only quadrilateral (or triangular) elements are required. The meshing is specified as mapped when the command MSHKEY is set to 1. The grid had maximum 5680 elements with 563 corner nodes. Then boundary conditions are applied and good results.

Results and Discussion

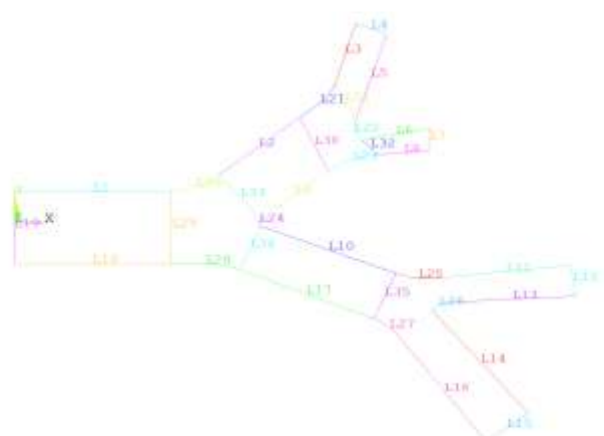
Numerical simulation of transient flow in the upper airway model for a cycle of breath is completed. We obtained value of velocity profile of asthmatic lung bifurcation at inlet and outlet position in the model. We found by numerically that asthmatic lung shows skew-parabolic behavior than normal lung inlet and outlet positions. The calculated velocity profiles are discussed over a range (250-1600) of Reynolds number. This range of Reynolds number is chosen to correspond to mouth air breathing rates of 0.272-2.261/s. There is a good comparison between ours and available data.

Velocity Profile

The axial velocity profiles were determined for Reynolds numbers 250 and 1600. These axial velocities were calculated at four different cross-sections, i.e., at inlet, near medial branch, in lateral branch and at outlet of lung bifurcation. Mean inlet velocities at the corresponding Reynolds number are used to scale the axial velocities. During inlet (fig.) the axial velocity profile is almost parabolic at the end of phase ($\mathrm{Re}_D = 250$). We also get similar velocity profile at the end of phase at ($\mathrm{Re}_D = 1600$). Axial velocity profile near the bifurcation (Fig.) during inspiration ($\mathrm{Re} \quad = 250{,}1600$) is almost parabolic similar to the profile at inlet. The velocity profile (fig.) near first bifurcation (L29, L20, L33, L24, L34, L28) is again parabolic ($\mathrm{Re}_D = 250$). But this parabolic nature is bandied below at the peak near second bifurcation (up side) ($\mathrm{Re}_D 1600$) ( ). Similar velocity profile (fig.) we get for outlet L4, L32, L15.

CONCLUSION

In this paper, we have numerically discussed velocity profile of asthmatic lung at inlet and outlet positions. The study was carried out in 2-D model of lung airway assuming the flow to be unsteady. Three-dimensional inspiratory flow in a three-generation asymmetric lung airway is numerically studied using finite element method. The method has validated against both by numerical simulation and available experimental data. The computations have been carried out for the Reynolds number 200 to 1600.



Fig(1.1). 2-D Lung bifurcation model

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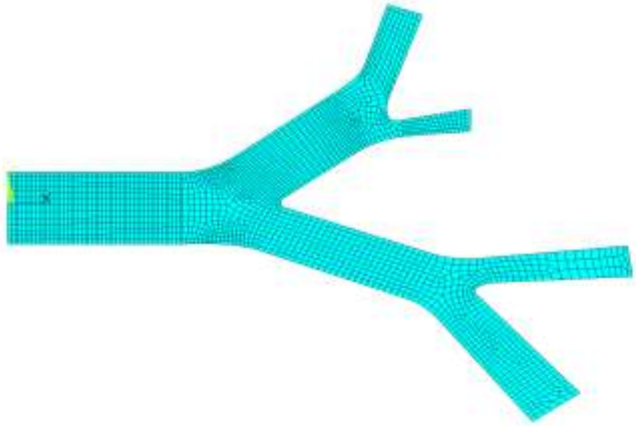

Fig.(1.2) Mesh generation of lung bifurcation model

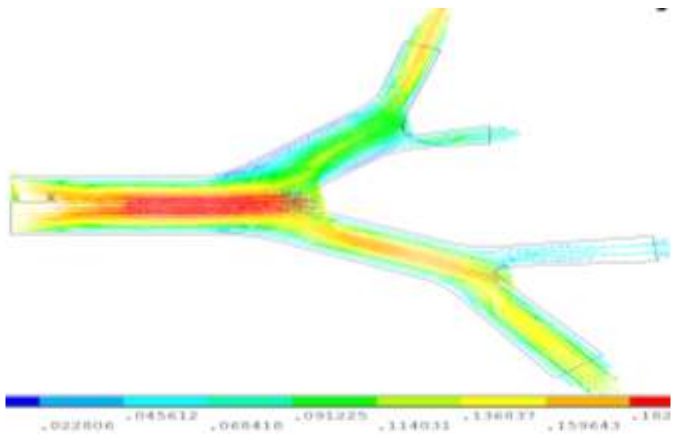

Fig (1.3) velocity profile of lung bifurcation

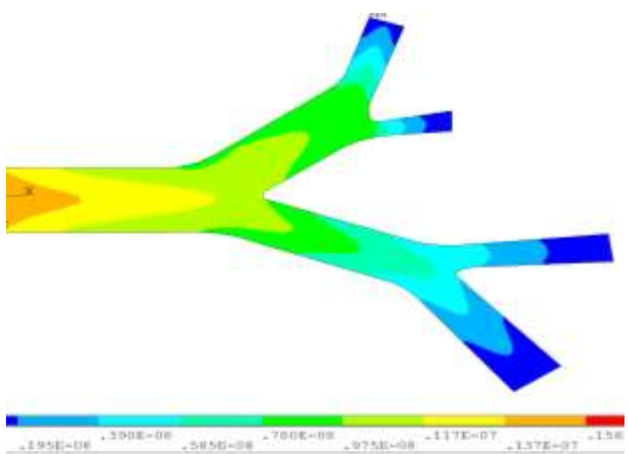

Fig (1.4) Color contours describing the velocity distribution over lung bifurcation

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