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Simulation of the Unsteady Oscillatory Flow in Three-dimensional Asymmetric Bifurcation Model of the Conducting Airway in the Human Lungs

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Abstract

A computational fluid dynamics (CFD) code based on finite volume method (FVM) was used to study the unsteady respiratory airflow dynamics within the human lung. The three-dimensional asymmetric single bifurcation model of the central airway based on the morphological data given by Horsfield et al [2] was used in the present study to simulate the oscillatory respiratory flow. Numerical simulation was made for the resting or normal breathing condition. Peak Reynolds number (Re) number of 1.75×10^{3} and Womersley number (α) of 2.73 were used, this corresponds to the tidal volume (VT) and the breathing frequency (f) of 0.5L and 0.8 Hz, respectively.

The simulation of respiratory flow within the single bifurcation was found to be sufficient to give a number of results which both qualitatively and quantitatively agree well with other published experimental works ([1], [3], [9], [10], [11], [12], [14] and [15]). The respiratory flow results were found strongly dependent on the convective effect and the viscous effect with some contribution of the unsteadiness effect. The secondary motions were observed at the outer walls of the branching curve of two daughter airways during inspiration and in left daughter airway and parent airway during expiration.

1. Introduction

Modeling of the airflow in the conducting airway in the human lungs is important to predict the particle deposition of both contaminate and the pharmaceuticals. In order to achieve good predictions of the deposition the airflow has to be modeled correctly.

Because of the complexity of the bronchial tree structure, the use of the small airway and highly asymmetric bifurcation structure, *in-vivo* studies are difficult for obtaining the global airways effect and the detail of the smaller airways downstream. While the experimental studies are crucial to obtain a somewhat satisfied detail by utilizing physiological scale-up model to study the airflow dynamics in the lungs. However those experimental studies are limited up to the 3rd generation of bifurcation and the smaller airways are in accessible to the most experimental technique even in the scale-up model ([1], [3], [11] and [12]).

Computational fluid dynamics (CFD) studies through the solutions of the Navier-Stroke equations seem to be a better way to access the smaller airways and easy to show a large number of data in three dimensional physiologically realistic airway geometries. Together with accurate computer modeling it should be possible to develop the insight to understand airflow pattern and the particle transport that can provide information which will avoid inappropriate clinical trials and waste effort.

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2. Mesh Model and Construction

The asymmetric single bifurcation model of the central airway based on the morphological data given by Horsfield et al [1] was used in the present study to simulate the respiratory airflow under resting condition.

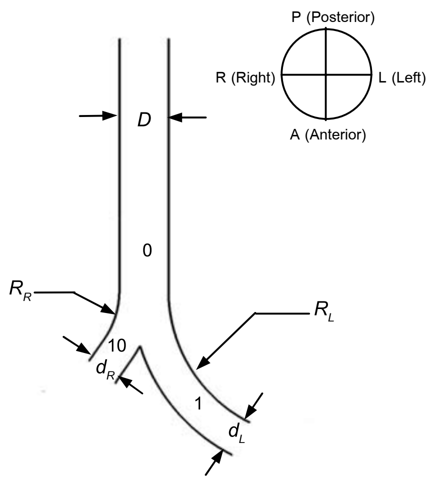


Figure 1 Model of single bifurcation of central airway (trachea and two main right and left bronchi).

Table 1 Model description of a single bifurcation model [1]

Branch No.	Diameter (mm)	Length (mm)	Branching angle	R/d	Axial curvature (R)
0 (parent branch)	D=16	100	-	-	-
1 (left daughter branch)	d_L=12	50	β=73°	4.5	54
10 (right daughter branch)	d_R=11.1	22	α=35°	3.0	33.5

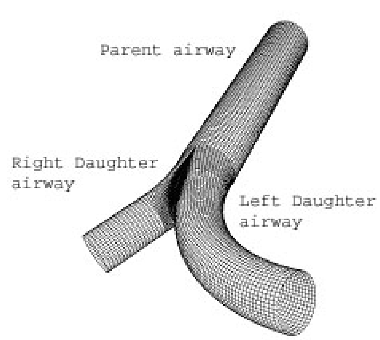


Figure 2 Three-dimensional mesh model of an asymmetric single bifurcation of the central airway

Due to the airway geometrical complexity, technique of multi block method was applied to ease the geometry and mesh generation. The airway mesh model consists of 84 blocks with structured hexahedral mesh cells of 159,872 node points. At the regions closed to the airway walls and the carinal ridge, the mesh cells were refined with well mesh grading ratio in order to capture high velocity gradients occurring near the airway walls and the carinal ridge.

The airway geometry dimensions and mesh model are given in Table 1, Figure 1 and Figure 2.

3. Respiratory Flow

The respiratory flow can be assumed to be incompressible laminar flow. The Navier-Stokes equations of the time-dependent incompressible flow are expressed by

Continuity equation:

$$\vec{\nabla}\cdot\mathbf{V} = 0 \qquad (1)$$

Momentum equations:

x-component $$\frac{\partial u}{\partial t} + (\mathbf{V}\cdot\vec{\nabla})u = -\frac{1}{\rho}\frac{\partial p}{\partial x} + \nu\Delta u \qquad (2\text{-a})$$

y-component $$\frac{\partial v}{\partial t} + (\mathbf{V}\cdot\vec{\nabla})v = -\frac{1}{\rho}\frac{\partial p}{\partial y} + \nu\Delta v \qquad (2\text{-b})$$

z-component $$\frac{\partial w}{\partial t} + (\mathbf{V}\cdot\vec{\nabla})w = -\frac{1}{\rho}\frac{\partial p}{\partial z} + \nu\Delta w \qquad (2\text{-c})$$

4. Numerical Boundary Condition and Solution

The Navier-Stokes equations are solved using a CFD code based FVM with the concurrent block Jacobi algorithm. The implicit scheme is used with the second order upwind scheme for the convective numerical flux and the central differencing for the diffusive numerical flux.

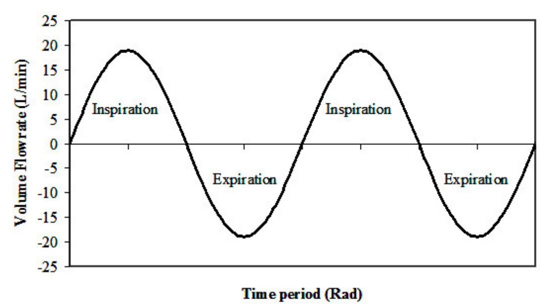


Figure 3 Respiratory cycle at resting condition

The airway walls can be assumed solid wall, hence no-slip boundary condition can be applied to the airway walls. To regulate oscillatory flow and control the flow rates in right and left daughter airways, the velocity boundary condition was applied on the inflow/outflow boundaries in the airway. The controlled flow rates in right and left daughter airways are 55% and 45% of the flow rate in the trachea, respectively. The uniformed velocity boundary conditions applied to the inflow/outflow boundary vary with time as a sinusoidal time function as shown in Figure 3.

Peak Reynolds number (Re) number of 1.75×10^{3} and Womersley number (α) of 2.73 were used, this corresponds to the tidal volume (VT) and the breathing frequency (f) of 0.5L and 0.8 Hz, respectively. Numerical respiratory flow parameters are given in Table 2

Table 2 Numerical parameters used in the study

Breathing condition	**Normal breathing**
VT (L)	0.5L
f (Hz)	0.2Hz
T (sec)	5sec
$\overline{U}_{T,\max}$ (mm/s)	1,562.5
Re	1.75×10^{3}
Time steps /cycle	100
Time increment, Δt	0.05sec
Dimensionless time at each time increment, $t^*=4\nu\Delta t/D^2$	0.01

5. Results and Discussions

5.1 The respiratory flow results

The results are presented as the velocity profiles normalized by the maximum average local velocity in each branch. The results are taken from 8 cross-sections station throughout the model. The oscillatory flow simulation results in a three-dimensional asymmetric bifurcation model at resting condition case are given in two separate parts, inspiration and expiration.

5.1.1 Inspiratory flow

An overview of the flow features for the inspiration phase under the resting condition is illustrated in Figure 4. During inspiration, uniform inlet was applied at the parent airway. The velocity profile at station P1 is still uniform and it develops downstream through a straight pipe passing station P2 and P3 of the parent airway with thicker boundary layer near the wall. When the flow has developed further downstream the peak velocity is found just next to the boundary layer near the wall as shown in Figure 4. The peak velocity near the wall is due to the diffusion of momentum over the cross-section while the flow tends to adjust itself during the oscillatory period. However this peak velocity is less noticeable further downstream and the velocity profile develops to the parabolic profile characteristic. When the flow from parent tube reaches the carinal ridge the flow has not yet fully developed, although the airway length is about 6.25 local diameters. This is because the time increment that allowed the velocity to change in magnitude is not large enough to allow the flow to adjust itself to be fully developed.

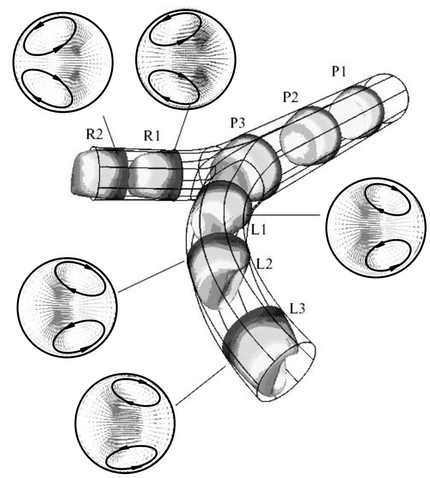
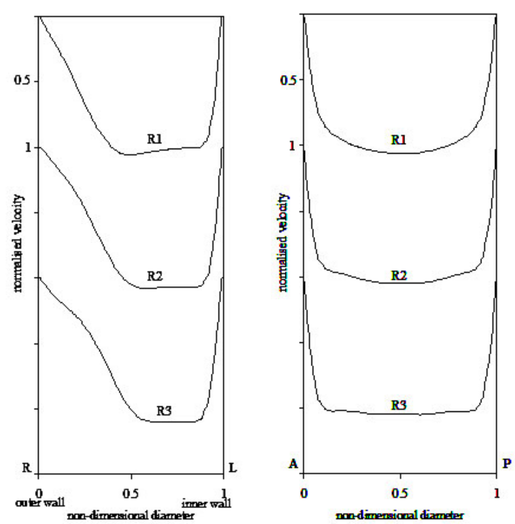


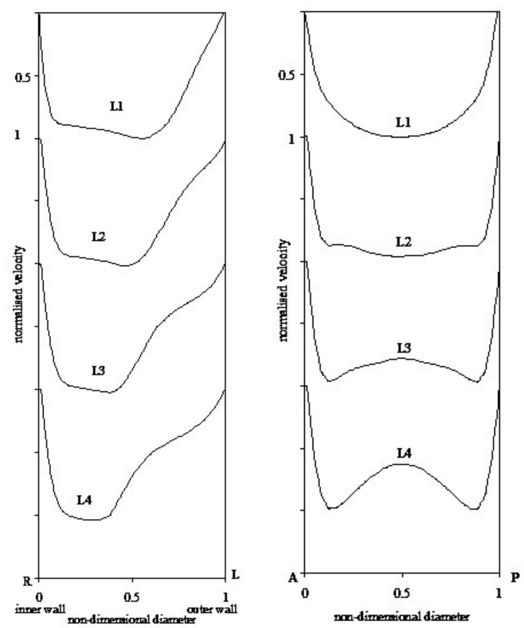
Figure 4 Inspiratory flow features at the maximum inspiration under the resting condition (Re = 1.75×10^{3}, α=2.37)



(a) (b)

Figure 5 Development of normalized axial velocity profiles in right daughter airway at peak inspiration

At the carina the incompletely fully developed flow from the parent airway is split into two daughter airways. When the flow enters two daughter branches, the axial velocities in bought daughter airways in the bifurcation plane (R-L) are initially skewed towards to the outer walls of the bifurcation (the inner walls of bend). The maximum velocity profiles are found next to the inner wall of the bifurcation as shown in Figure 5a and Figure 6a. While for the normal plane to the bifurcation (A-P), symmetric velocity profiles are found in both daughter airways as shown in Figure 5b and Figure 6b. The profiles are initially parabolic and then develop to the “M-shape” or “bi-peak” velocity profiles. The distortion of the axial velocity profiles in both right and left daughter airways are due to the secondary motions occurring when the flow undergoes the curved portions of the bifurcation.



(a) (b)

Figure 6 Development of normalized axial velocity profiles in left daughter airway at peak inspiration

5.1.2 Expiratory flow

In the expiration, the respiratory flow was regulated from the two daughter airways in which the uniform velocity profiles were imposed at the inflow boundaries. The expiratory flow feature is illustrated in Figure 7.

The right daughter is constructed with a short pipe continuing from the curved part at the bifurcation and the carinal ridge. When the flow moves through the straight pipe of right daughter airway at R2 and R1 velocity profile is still flat with a thicker boundary layer. The flow has not yet fully developed when it enters the bifurcation and the secondary effect was not found in right daughter airway during expiration phase.

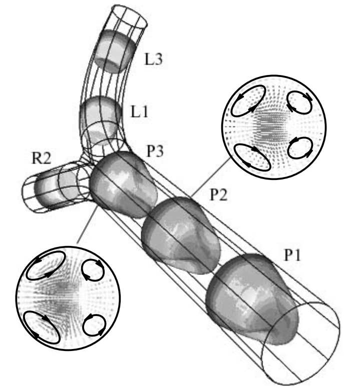


Figure 7 Expiratory flow features at the maximum Expiration under the resting condition (Re = 1.75×10^{3}, α=2.37)

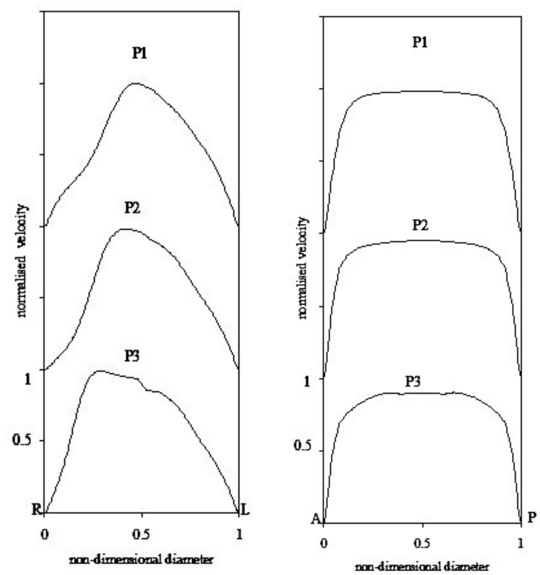


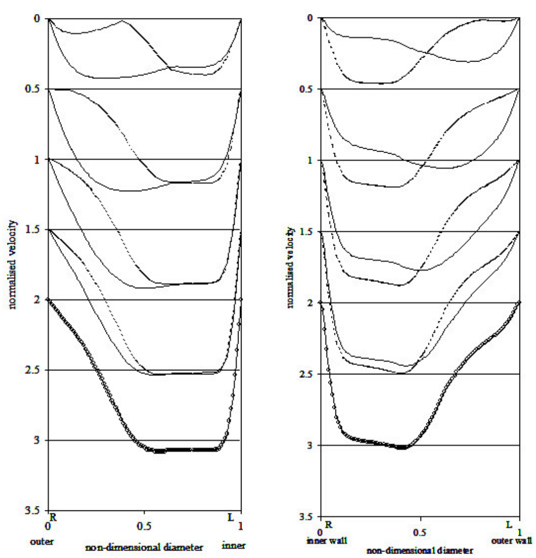
Figure 8 Development of normalized axial velocity profiles in parent airway at peak expiration

The left daughter airway was constructed with a long uniform curved pipe. A uniform inlet velocity in the bifurcation plane (R-L) develops downstream and pronounces skewed axial velocity profiles towards to the inner wall of the bifurcation (outer wall of the bend) within approximately 3 diameters from the inflow boundary before the flow reaches the bifurcation. The distortion of axial velocity profile in the left daughter airway during expiration is purely affected by the secondary flow caused by the curvature portion.

When two flow streams from the both daughter airways merge together in the bifurcation, two pairs of unequal vortices are formed downstream just above the carinal ridge in the parent airway as shown in Figure 7. In the plane of bifurcation (R-L) peak velocity profile is found near the right wall because the flow rate regulated from the right daughter airway is greater than that driven from the left daughter airway. While the axial velocity profile in the normal plane to the bifurcation (A-P) is symmetric characteristic with incomplete fully developed profile.

5.3 Discussions on the oscillatory flow effects

The numerical results of inspiratory flow and expiratory flow at peak flow rate under the resting condition were compared to the experimental results given by [1], [3], [9], [11], [14] and [15]. It was found that the numerical results are sufficient to give a number of results which both qualitatively and quantitatively agree well with the experimental works. The results show that velocity profiles during inspiration are strongly dependent on the airway geometry, meaning the convective and viscous effects play important role on the respiratory flow under the resting condition.



(a) (b)

—— Acceleration

- - - - Deceleration

Figure 9 Axial velocity profiles in the bifurcation plane (R-L) during inspiration cycle at (a) station R2 in right daughter airway and (b) station L3 in left daughter airway

It is noted that, both inspiratory and expiratory flows results are agreed with the previous experimental results when the flow reaches the peak flow during respiratory cycle. However, at the other flow rates during the same respiratory cycle (but between acceleration and deceleration) the velocity profiles obtained from acceleration and deceleration are different as shown in Figure 9. The flow separation at the near "zero" flow was observed when the flow changes direction at inspiration-to-expiration and expiration-to-inspiration during acceleration. The velocity profiles during acceleration are distorted at the low flow rate. The distortion of the velocity profiles at the beginning of the inspiration can be influenced by the flow separation effect at the reversal flow when the flow changes direction from expiratory to inspiratory flow.

6. Conclusions

It was found that the numerical results of the respiratory flow, both inspiratoy and expiratory flows, in an asymmetric single bifurcation of the central airway at peak flow rate are sufficient to give a number of results which both qualitatively and quantitatively agree well with the experimental results given by [1], [3], [9], [11], [14] and [15]. However, those experimental investigations used only the steady flow assumption and were not concerned the unsteadiness/oscillatory effect. The numerical results in our study suggested that the unsteadiness/oscillatory effect need to be taken into account and the steady flow cannot be assumed.

The results also showed that the respiratory flow under resting condition is strongly dependent on the convective and viscous effects. Hence, the realistic geometry of the airway is important to get the realistic results of the airflow patterns.

The secondary flow motions were observed in both daughter airways during inspiration and in the parent airway during expiration phase. The numerical flow information in the study is useful to track on the particle deposition during breathing cycle in order to investigate the efficacy of the pharmaceutical which most are in the form of aerosol particle.

7. Acknowledgement

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