Sādhanā (2020)45:229
https://doi.org/10.1007/s12046-020-01460-9

Fluid structure interaction studies of human airways

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MS received 20 April 2018; revised 21 May 2019; accepted 5 November 2019

Abstract. Study of flow characteristics of human airways using Fluid-Structure Interaction (FSI) analysis is very important in the context of prognosis, diagnostic and treatment of respiratory diseases. The present study is focused on effect of elasticity on the respiratory wall during inhalation. Airflow in Computed Tomography (CT) scan model with rigid and compliant airway walls is studied. FSI technique is used to simulate the airflow in the model. The comparison for the two different respiratory models (rigid and compliant) shows that FSI technique brings out more realistic results as compared to Computational Fluid Dynamics (CFD) analysis. It is found that respiratory wall elasticity affects the different flow parameters (pressure, wall shear stress, etc.) at different location of the model. Wall shear stress (WSS) and airway pressure were decreased due to flexibility effect of the airway wall. This will help medical practitioners to correlate the clinical assessment with this FSI results.

Keywords. Human airway model; compliant wall; fluid structure interaction (FSI).

1. Introduction

The airflow studies in the bifurcating respiratory are the most basic and enlightening problems. Because of the geometrical complexity of the human respiratory model, experimental studies of the internal flow dynamics is difficult. One of the challenging tasks for the numerical simulation is to get the realistic model from the CT scan. The development of computational potential with CT data makes it possible to create real model of respiratory tract for numerical/CFD simulation. A number of computational fluid dynamics studies have done the fluid flow characteristics considering rigid airway wall [1, 2]. However, very few studies have considered the airway wall elasticity in their simulation study [3].

Human respiratory tract are compliance which means ability of stretch and expand. This structural deformation occurs continuously during breathing and thereafter, dimensional changes occur during breathing. The changes in dimension can be determined by the mechanical properties of respiratory structure and composition of airway wall. The upper respiratory tract (trachea and first few generation of bronchus) have a higher number of cartilage rings and hence they are more rigid in comparison of the other part of respiratory tract. It is seen that elasticity of tracheal wall are not taken in most of the numerical simulation [1, 2].

Generally the airways are tethered to a very compliant and compressible tissue that undergoes large deformation during breathing. Hence the interaction between transmural pressure and airflow through airway is very important to consider in the study. Therefore, Fluid Structure Interaction (FSI) method must be employed for realistic results [3, 4].

Effects of wall elasticity on fluid flow characteristics was discussed by Koombua *et al* [4]. The result showed that it is very important to incorporate the airway wall elasticity into the computational model of the bifurcation when considering airway wall properties. FSI simulation is performed for two different models namely fluid and solid domain. It was computed that the airway wall elasticity affect the flow pressure and wall strain distribution but it has very less effect on airflow velocity and wall shear stress [3].

The objective of present study is to investigate the effect of airway wall flexibility on airflow characteristics, wall shear stress (WSS), and airway deformation in the airway during inhalation. Trachea to first generation CT model

This paper is a revised and expanded version of an article presented in "First International Conference on Mechanical Engineering" held at 'Jadavpur University', Kolkata, India during January 4–6, 2018 (INCOM-2018).

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Published online: 07 September 2020

with elasticity modulus of 2 kPa was considered. The material property of the airway wall was assumed to be homogeneous and isotropic having poison ratio of 0.45. Both rigid and flexible models are compared to see the effect of flexibility of airway wall on flow characteristics and stresses induced.

2. Computational domain

A three dimensional solid model from trachea to first generation is created using MIMICS software using CT scanned image. The hydraulic diameters at the inlet of trachea and outlets of bronchi were found to be 1.88 cm, 1.23 cm and 1.13 cm, respectively. Reynolds number based on average hydraulic diameter of trachea for realistic model was calculated as 2610 for an average inlet velocity of 2.03 m/s. Upper respiratory tract looks like a Y shaped bifurcated pipe (figure 1). Fluid and solid domain of the bifurcation were created separately. The fluid domain is the volume of the air flowing through model and solid domain is the airway wall surrounding the fluid domain. The thickness of the airway wall taken was 1 mm. The airway wall was assumed to be homogeneous and isotropic in nature.

This study mainly focused on CT based model from trachea to first generation.

3. Grid generation

A mesh generation option available in Ansys Workbench was used for the mesh generation. The meshing of both fluid domain and solid domain was done separately. Adaptation option is used for grid independence test. Refine and Coarsen options are selected for the enhancement of

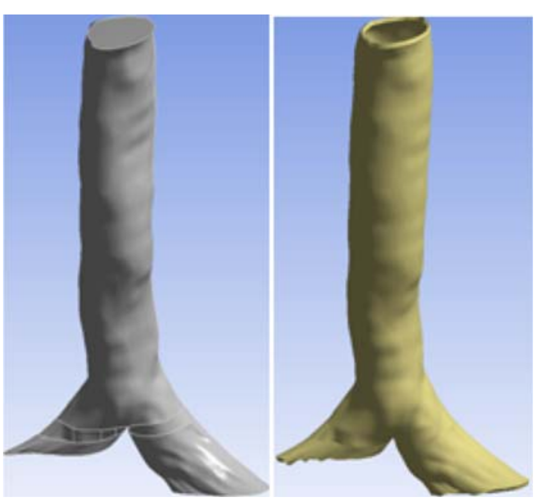

Figure 1. Geometry of fluid domain and solid domain of trachea to 1st generation model.

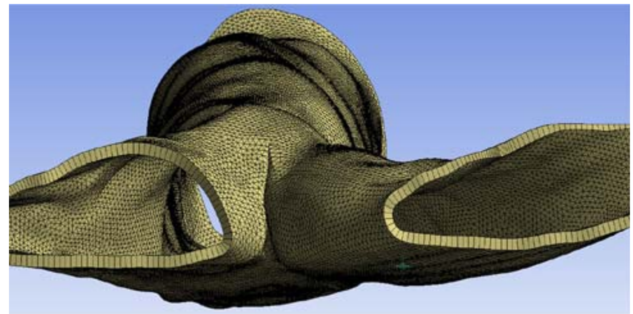

Figure 2. Grid distribution of solid domain of trachea to 1st generation Model.

mesh. Coarsen and refinement threshold are chosen in the curvature region. The fluid mesh has 15,46475 tetrahedral element with maximum skewness of 0.79. Solid mesh has 72890 triangular elements with maximum skewness of 0.54 (figure 2). A grid independence test is performed for both fluid domain as well as solid domain. Change in maximum velocity is used as convergence criteria for fluid domain and change in maximum displacement is used as convergence criteria for the solid domain. Convergence criteria equivalent to 10^{-4} was taken in this study. The computations were performed on a HP Z-800 workstation.

4. Governing equations

The airflow is assumed to be steady and incompressible. The basic equations governing the motion of fluid are as follows:

4.1 *Conservation of mass*

$$\frac{\partial(\rho)}{\partial t}+\frac{\partial(\rho u)}{\partial x}+\frac{\partial(\rho v)}{\partial y}+\frac{\partial(\rho w)}{\partial z}=0 \tag{1}$$

4.2 *Conservation of momentum*

$$\frac{\partial u_i}{\partial t}+\frac{\partial(u_i u_j)}{\partial x_j}=-\frac{1}{\rho}\frac{\partial p}{\partial x_i}+\frac{\mu}{\rho}\frac{\partial^2 u_i}{\partial x_j \partial x_j} \tag{2}$$

Since the flow is assumed steady therefore, time dependent terms are ignored.

Where, ρ is density of air, μ is viscosity, p is pressure under which flow is flowing, u_i(i = 1, 2, 3,…) velocity in x, y and z direction.

4.3 *Structural equations*

The basic equations governing the structural problem are as follows: [5].

4.4 *Equation of motion*

$$\begin{aligned}\frac{\partial \sigma_x}{\partial x}+\frac{\partial \tau_{xy}}{\partial y}+\frac{\partial \tau_{yz}}{\partial z}+F_x=\rho\frac{\partial^2 u}{\partial t^2}\\ \frac{\partial \tau_{xy}}{\partial x}+\frac{\partial \sigma_y}{\partial y}+\frac{\partial \tau_{yz}}{\partial z}+F_y=\rho\frac{\partial^2 v}{\partial t^2}\\ \frac{\partial \tau_{xz}}{\partial x}+\frac{\partial \tau_{yz}}{\partial y}+\frac{\partial \sigma_z}{\partial z}+F_z=\rho\frac{\partial^2 w}{\partial t^2}\end{aligned} \tag{3}$$

4.5 *Constitutive relation*

$$\begin{bmatrix}\sigma_x\\ \sigma_y\\ \sigma_z\\ \Gamma_{xy}\\ \Gamma_{yz}\\ \Gamma_{zx}\end{bmatrix}=C\begin{bmatrix}\varepsilon_x\\ \varepsilon_y\\ \varepsilon_z\\ \gamma_{xy}\\ \gamma_{yz}\\ \gamma_{zx}\end{bmatrix}=C\begin{bmatrix}\frac{\partial u}{\partial x}\\ \frac{\partial v}{\partial y}\\ \frac{\partial w}{\partial z}\\ \frac{\partial u}{\partial y}+\frac{\partial v}{\partial x}\\ \frac{\partial v}{\partial z}+\frac{\partial w}{\partial y}\\ \frac{\partial u}{\partial z}+\frac{\partial w}{\partial x}\end{bmatrix} \tag{4}$$

where, σ_x is normal stress in x direction, σ_y is normal stress in y direction, σ_z is normal stress in z direction, Γ_{xy} is shear stress in xy-plane, Γ_{yz} is shear stress in yz plane, Γ_{zx} is shear stress in zx plane, F_x is body force in x direction, F_y is body force in y direction, F_z is body force in z direction, U is the displacement in x direction, v is displacement in y direction, w is displacement in z direction, ρ is density, C is the elasticity matrix

5. Boundary conditions

The density of airway wall is 1024 kg/m^3 as reported by Xia *et al* [6]. There is very few research paper published about elastic properties of airway wall. Sera *et al* [7] have chosen Young modulus = 5.8 MPa. According to Koombua *et al* [4], Elasticity of modulus was 130.89 kPa and 74.07 kPa in circumferential and longitudinal direction respectively. Martino *et al* [8] have taken Poison ratio = 0.45. Corresponding to these parameters, very small deformation was found in the simulation.

In the present work, a range of Young's modulus from 200 kPa to 2 kPa was tested in the FSI simulation with the boundary condition of zero displacement at the outlet and inlet to provide tethering to the airway at Reynolds number = 3269. It is fond that Young's modulus of 2 kPa represents maximum deformation of human airway. Hence in the present work the value of Young's modulus taken is 2 kPa.

6. Fluid structure interaction algorithm

The effect of fluid flow pressure on a structure is significant when the structure is flexible in nature. In FSI, the governing equations are solved in an iterative manner. In the present work FSI algorithm is incorporated by using Ansys System Coupling. Fluent acquired a converged solution according to its own criterion of convergence and transfer the fluid force to Ansys Mechanical. Then the displacement value of structural member is obtained with help of solution provided by Ansys-Fluent [9] for the same time step. Now the difference exists compared to one way coupling. The calculated solution of Ansys-Mechanical solver is given back to the Ansys-Fluent solver to determine a new set of fluid forces according to nodal displacements of previous time step. This is said to be a coupling iteration and continues until the convergence criterion of data transfer is reached.

At the fluid–structure interface, the two meshes are conformed to each other meaning that the fluid mesh coincide with the solid mesh at the interface. Thus, the information about the mesh deformation, velocity and the fluid pressure is exchanged through this interface. The construction of FSI method is implemented by coupling a CFD solver with CSD (computational structure dynamics) solver. No slip boundary condition was defined at the fluid solid interface.

7. Validation

For the same boundary conditions, present results are compared (table 1) with the computational results reported by Koombua *et al* [4].

Table 1. Validation with Koombua *et al* [4].

	Present work	Koombua *et al*	Percent error
Velocity (m/s)	10.19	10.17	0.19%
Pressure (Pa)	49.81	54.423	9.26%
Total deformation	0.169×10^{-6}	0.177×10^{-6}	4.73%

8. Results and discussion

The effects of airway wall elasticity on distribution of airflow velocity, Wall Shear Stress, airway pressure were computed for both rigid and flexible model. Distribution of airway deformation Von-mises stress and normal stress and effect of airway wall elasticity on these parameters were discussed.

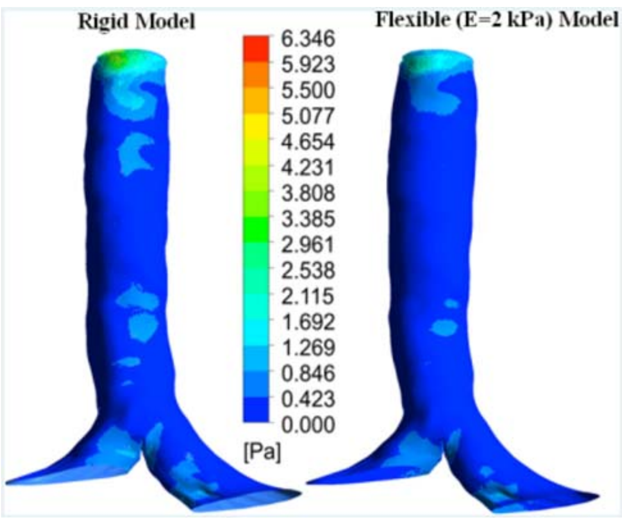

Figure 3. WSS distribution in rigid and flexible model.

8.1 *Wall shear stress distribution in rigid and flexible model*

The Wall Shear Stress distribution for a rigid model and elasticity modulus 2 kPa is shown in the figure 3. The maximum magnitude of wall shear stress in the rigid airway is about 6.346 Pa. Maximum magnitude of wall shear stress for flexible airway model was found to be 2.327 Pa. During the inspiration the high wall shear stress occurs close to inlet of trachea and near to the bifurcation as depicted in figure 3. The low wall shear stress is found at the most of the surface of trachea and first generation except the region close to inlet and near the bifurcation. The high shear stress in the region close to inlet is induced by the smooth bend, which consequently produces secondary vortices. The high wall shear stress behind the bifurcation is induced by the splitting of flow, which produces secondary vortices in the lower branches and increases the magnitude of the Wall Shear Stress. The high wall shear stress near the outlet is caused by the narrowing of the airway tract of lower branches, which leads to increase in the average velocity and thus increases the strain rate of the air velocity.

8.2 *Pressure and airway deformation*

The deformation of the respiratory wall indicates changes in wall shape due to the force of air on wall. Distribution of local pressure at two plane, first one lies in upper half of trachea and second one lies in the region close to bifurcation is shown in figure 4 given below. It can be seen that the pressure is maximum on interior right side of trachea on plane-1. This is due to the obstruction created by the cartilage ring to the flow. Hence form the figure it can be

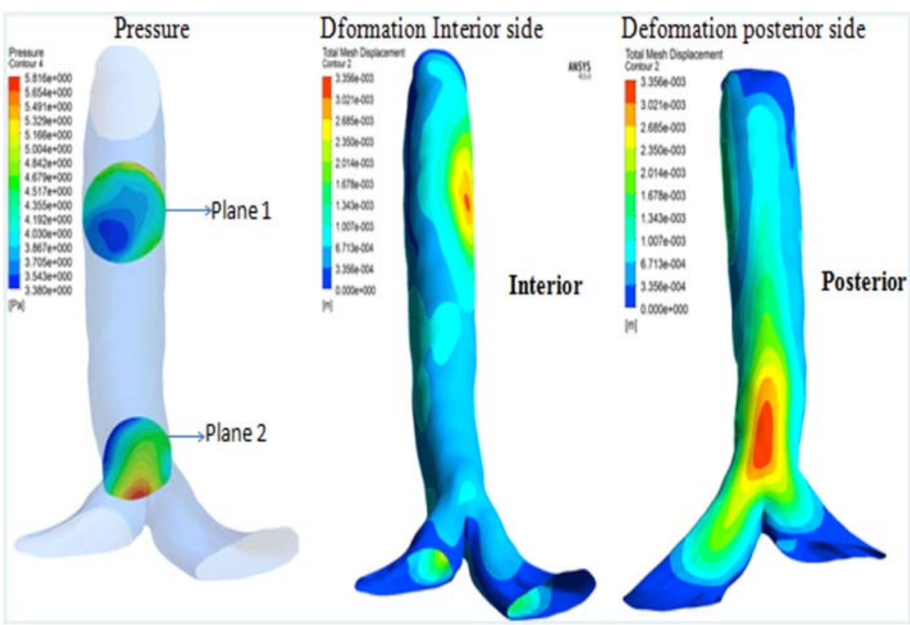

Figure 4. Distribution of Pressure and Deformation.

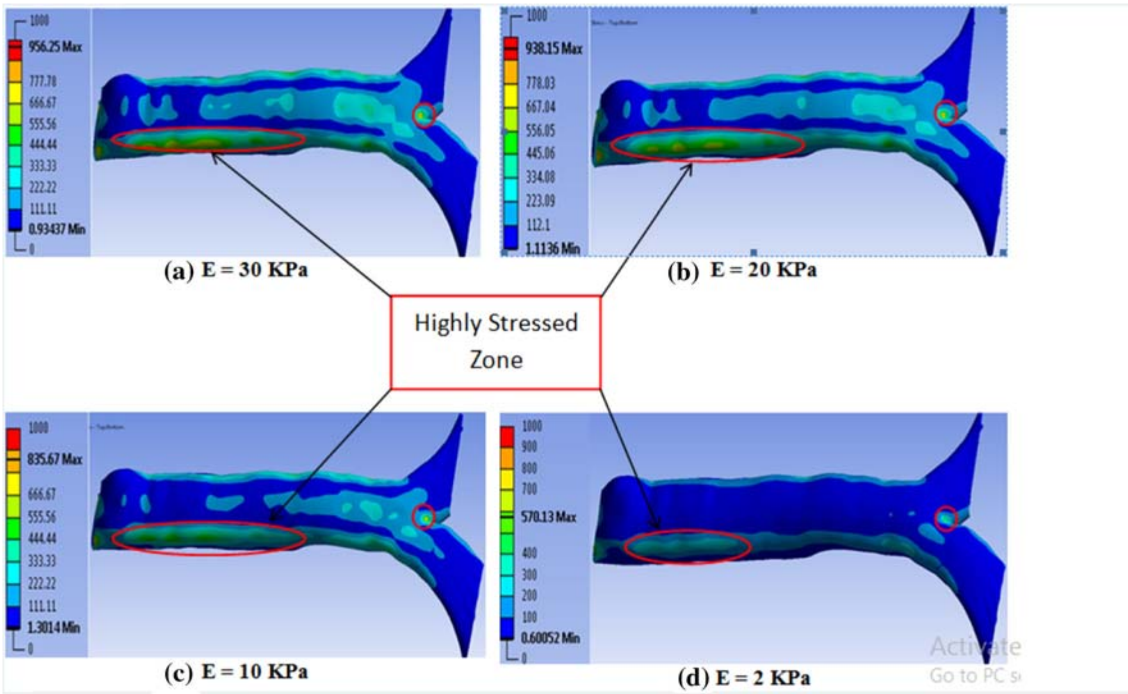

Figure 5. von Mises stress for different value of E (Modulus of Elasticity).

observed that the deformation is maximum in that region. On the plane-2, the pressure is maximum in the region of posterior side. Hence maximu deformation can be observed at posterior side of the airway wall.

8.3 *Distribution of vonmises stress and deformation*

The von Mises stress is the average stress in all direction. Distribution of von Mises stress is shown in the figure 5. The maximum value of Von-mises stress decreases as the value of modulus of elasticity decreases. It is observed from figure 5 that the magnitude of von Mises stress is maximum near the bifurcation region and at the left side of posterior wall. It means that the rigidity of the airway wallat this locationis more as compared to other regions of airway wall. Hence it may be possible that during the inhalation or exhalation, airway may collapse at the region where there is less stress induced. In other words the chance of failure of airway wall will be more where the magnitude of von Mises stress is less.

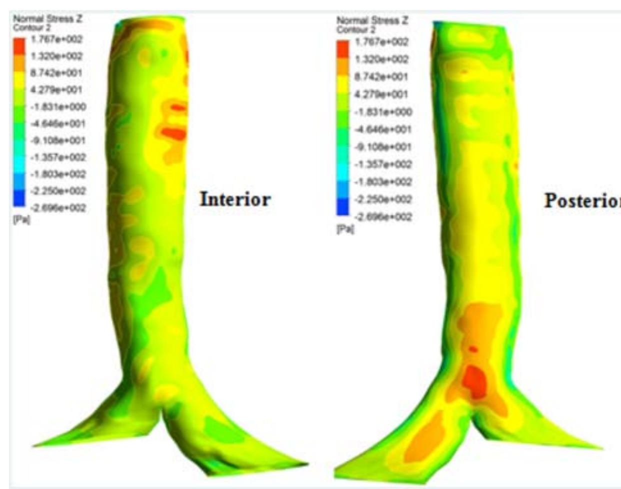

Figure 6. Distribution of normal stress in Z-direction.

8.4 *Distribution of normal stress*

In figure 6, distribution of Normal stress was shown for E = 2 kPa. It was noted that the normal stress was maximum in the region close to the bifurcation zone and at the region in upper half of trachea on interior left side of airway wall. These regions have a higher value of strain (deformation). In the previous section it was noted that in these region, the deformation was found to be maximum. As the value of deformation increases, the stress induced also increases. So in theses region high value of normal stress has been found.

9. Conclusions

The airway pressure, WSS, airway wall deformation and stresses induced in the airway wall of trachea to first generation CT model were analyzed in this study using finite element method with FSI algorithm. The main focus of this study was to investigate the effect of airway wall flexibility on the airflow characteristics and the stresses induced in the airway. It was noted that the wall shear stress and airway pressure were decreased due to the flexibility effect of airway wall as compared to the rigid airway wall. From the simulation result, it was noted that the deformation of the airway wall is maximum in the region close to the bifurcation and region in upper half of trachea on the interior left side. It was analyzed that the von Mises stress and longitudinal stress are maximum in the region where deformation was found to be maximum.

Acknowledgement

This paper is a revised and expanded version of an article entitled '*Analysis of Airflow Through Compliant Human Airways Using Fluid Structure Interaction*', Paper No. 218 presented in "First International Conference on Mechanical Engineering" held at Jadavpur University, Kolkata, India during January 4–6, 2018.

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