Computational modeling methods for simulating obstructive human lung diseases

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Aims

Provide a computational model for the human lung that:

- Allows the processing of the patient specific lung geometry.
- Simulates the effects of bronchoconstriction.
- Allows the visualization of the airflow for different kinds of geometry alterations.

Methods

The developed geometry processing approaches, allows the simulation of the obstructive diseases in the 3D lung structure and study their impact on airflow in the lung, by using a 3D CFD simulation. The surface of the given object is iteratively contracted in the direction of the inward normal based on the Laplacian mesh processing scheme presented in [1]. A lung 3D model is described by a mesh $G = (\mathbf{V}, \mathbf{E})$ with vertices $\mathbf{V}$ and edges $\mathbf{E}$. Using cotangent weighting the Laplacian coordinates approximate the curvature-flow inward normal. Thus,solving iteratively the discrete Laplacian equation $\mathbf{LV} = 0$ we can achieve mesh contraction or converge to and 1D shape . Let $\mathbf{L}$ be the Laplacian operator and $\mathbf{V}'$ the vertices final position, then we have:

$$\mathbf{L}_{i,j} = \begin{cases} \omega_{i,j} = cot\ a_{ij} + cot\ b_{ij}, & (i,j) \in E \\ \sum_{i,k \in E}^{k} -\omega_{ik}, & i = j \\ 0, & \text{otherwise} \end{cases} \quad (1)$$

where a,b are opposite angles for edge (i, j).Since $\mathbf{L}$ is singular, further constrains need to be used in order to ensure a unique solution for $\mathbf{v}'$. Thus we focus on solving the equations:

$$\begin{bmatrix} \mathbf{W}_L \mathbf{L} \\ \mathbf{W}_H \end{bmatrix} \mathbf{V}' = \begin{bmatrix} \mathbf{0} \\ \mathbf{W}_H \mathbf{V} \end{bmatrix} \quad (2)$$

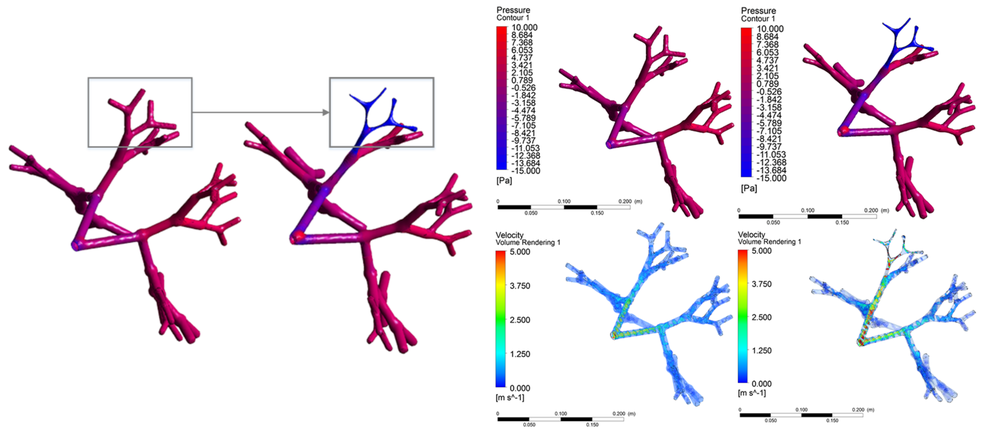
where $\mathbf{W}_L$ and $\mathbf{W}_H$ are the diagonal matrices. However, the solution of (2) does not immediately converge to an 1D shape so that an iterative scheme is employed. In order to simulate narrowing we employ stopping criteria related to the difference between the updated and the initial position.

1. If I is a unitary matrix, k a double constant and A the average face area of the model initialize W_L and W_H in the following manner:
$$\mathbf{W}_L = k \cdot \sqrt{A} \cdot \mathbf{I} \qquad \mathbf{W}_H = \mathbf{I} \quad (3)$$
2. Solve (2) for $\mathbf{V}'$.
3. Update $\mathbf{W}_L$ and $\mathbf{W}_H$. t denotes the iteration index and A_i^t the one-ring area of face i at iteration t.
$$\mathbf{W}_L^{t+1} = s_L \cdot \mathbf{W}_L^t \qquad W_{H,i}^{t+1} = W_{H,i}^0 \cdot \sqrt{A_i^0 / A_i^t} \quad (4)$$
4. Recompute $\mathbf{L}$.
5. Repeat steps 2 to 4 for a given number of iterations.

The resulting mesh is used for computational fluid dynamics simulation using FLUENT,ANSYS Inc. The volume inside the surface mesh is assumed to be the fluid domain consisting only of air. The flow is assumed to be incompressible and the fluid to be Newtonian. The Reynolds Averaged Navier-Stokes equation is solved allowing the simulation of turbulence. The solution scheme uses the Semi-Implicit Method for Pressure Linked Equations algorithm (SIMPLE) [2]. We used a synthetic mesh with a similar to lung bronchial tree branching pattern consisting of connected tubular meshes. The flow is assumed to be laminar while the inlet velocity is assumed to be 5 m/s and the pressure at the inlet 100000 Pa ($\approx 1Atm$).

Results

The mesh contraction process allows the deformation of the airways under investigation, simulating bronchoconstriction.Finally, the results of the CFD simulation are presented below.By inspecting the figure we can conclude that air velocity increases in the contracted airway branches as compared to the non-contracted branch while the air pressure drops in the contracted version of the model. Specifically for a 40% diameter reduction in a terminal airway, we observe a pressure drop of 48% with relevance to the lowest observed pressure value.



(a) (b)

Figure: (a)Narrowing 4^{th} and 5^{th} generation airways. (b)Pressure and velocity distributions for initial and narrowed geometry

Conclusion

The aforementioned approaches are essential for creating patient specific 3D models, corresponding to different levels of airway narrowing related to different levels of inflammation, from existing 3D models that have been constructed from available CT/MRI scans.

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