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Simulation of Respiratory Flows

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Abstract

Simulations were conducted on the respiratory tract of a juvenile rhesus monkey, to determine if turbulence modelling is required in such simulations. Simulations were conducted using two different breathing rates and both laminar and turbulent flow models. The analysis was complicated by the complexity of the geometry and the nature of the flow. Given the pulsatile nature of the flow, a lower Reynolds number for transition was applied, based on experimental work in the literature. We have found that at higher breathing rates there is a need for a turbulence model.

Introduction

Respiratory research has been ongoing for quite sometime, which is unsurprising given the importance of lung function. Despite this interest however, there is relatively little information available in terms of local flow and pressure fields within the lungs. While lung function may be readily assessed through the use of spirometry, this only provides global data.

The ability to resolve localised flow and pressure fields within in the lungs may be beneficial in terms of improved understanding of lung flow behaviour and the effects of lung diseases on these flows. There is also the potential to improve the delivery of aerosolised medications through this understanding and the possibility of optimising their delivery through the control of factors such as particle size distribution.

Simulation offers a means to study localised flow patterns in the lungs, and has been used by a number of research groups, from both engineering [3], [9], [10] and medical backgrounds [5], [4]. Many of these works utilise simplified geometries and/or unidirectional flow, which while offering the advantage of computational efficiency, may not be physiologically realistic.

Another consideration is the expansion and contraction of the lungs during breathing. In our previous work [8] we demonstrated that lung motion influenced both flow and particle deposition in the lungs. The current work will expand on the previous work through consideration of the use of a turbulence model.

Studies in the literature vary in their approaches to turbulence, with some employing Reynolds Averaged Navier Stokes (RANS)/Large Eddy Simulation (LES) models [7], [6] and others considering the flow as purely laminar [3], [8], [10], some of which noted that turbulence may need to be considered in parts of the extrathoracic airway [10].

One of the difficulties in assessing the need for turbulence modelling is the tortuous nature of the geometry and the pulsatile nature of the flow. As such, simple determinations of Reynolds number (*Re*), treating the system as a pipe flow, may be inappropriate, as in reality the transition to turbulent flow may occur at much lower values of *Re* [11].

In their work Winter and Nerem [11] experimentally determined that transition could occur at *Re* values between 400-500 in a pipe subject to pulsatile flow, and went on to suggest that transition could occur at lower values in physiological flows.

The Reynolds number (*Re*) is given by;

$$Re = \frac{\rho u d}{\mu}, \tag{1}$$

where ρ is the fluid density, μ is the fluid viscosity, u is the flow velocity and, d is the characteristic length.

In this work we will consider the flow in the airways and lungs of a juvenile Rhesus monkey, at two different breathing rates (tidal breathing and breathing under exertion). Simulations will be conducted using a laminar flow model and then an LES model.

Methodology

This work was performed using a geometry developed from x-ray computed tomography (CT) scans of a juvenile Rhesus monkey, originally developed by Asgharian et al. [1] and shown in 1 The mesh used was an unstructured hexahedral mesh, of approximately 17 million cells. The inlet is a plane set in front of the entrance to the mouth an nose, with the ends of each airway branch in the lungs initially set as walls.

Simulations were performed using an open source computational fluid dynamics (CFD) package (OpenFOAM), with mesh motion prescribed by a custom library. The function of the library is described in Mead-Hunter et al. [8].

In order to determine the appropriate inlet velocity over time, initial simulations for the two breathing rates were conducted with a prescribed mesh motion that ensured that the appropriate tidal volume was inhaled with each breath. This step was taken to account for airways of the lung, with could not be sufficiently resolved from the CT-scans, and is discussed in Mead-Hunter et al. [8]. This required an expansion factor of 2.18, which is physiologically unrealistic, and so only used to generate data for the inlet velocity.

The inlet velocities were extracted from these initial simulations and used in new simulations using a time varying mapped fixed value boundary condition. The airway extremities previously specified as walls were reset as uniform total pressure boundary

Figure 1: The rhesus monkey airway and lung geometry

conditions, thereby allowing the appropriate volumetric flow in and out of the lungs, when a new more realistic expansion factor of 1.4 was used.

Only the breathing rates were altered between the two cases, and not the volume of the breath. The tidal breathing rate was set as 37 breaths per minute [2], with the breathing rate under exertion set as 50 breaths per minute, which is the upper recorded breathing rate.

Simulations were conducted at each breathing rate, with a laminar flow model and then an otherwise identical simulation with an LES model. The sub-grid scale model employed utilised a dynamic Smagorinsky approach.

Results & Discussion

In order to streamline the analysis of results, a number of planes were extracted from the completed simulation. These were taken, through the upper regions of the nose, nasal passages, below the pharynx, around the midpoint of the trachea, near the carina of the trachea and through the upper airways of the lung (see Figure 2). Comparisons between simulations were conducted based on the point in the breathing cycle were the velocity was highest at each plane. As the mesh motion was identical for both the tidal and exertion cases, the plane through the upper lung is consistent, so that comparisons may be made.

Reynolds numbers were evaluated for the highest velocity at each plane. In planes with multiple airway branches, the branch with the largest velocity was chosen. As an example one of the sample planes (P3) is shown in Figure 3. This was extracted from the laminar flow simulation for the exertion case, and illustrates the velocity field.

The Reynolds numbers are highest in the planes taken through the upper nasal passages and the pharynx, and lowest in the planes taken through the upper lungs. These may be seen in the plot shown in FIgure X. This is as expected given the reduction in velocity that occurs as the flow moves towards the extremeties of the lung, due to flow being controlled by the expansion of the lungs. If we were to assume simple pipe flow then the Reynolds numbers for all cases are below 2300, indicating laminar flow, with no transition. However, given nature of the flow this assumption is overly simplistic, as evidenced by the experimental work of Winter and Nerem **??**. If as their work suggests, we take the beginning of transitional flow to be between *Re* values of 400 and 500, then the presence of transitional flow in the pharynx and trachea needs to be considered for the exertion case. Furthermore, given this lower threshold for transition there is a possibility that turbulence may develop at lower values of *Re*, also. In the exertion case the value of *Re* through P3, reaches 1300, which may in this case of pulsatile

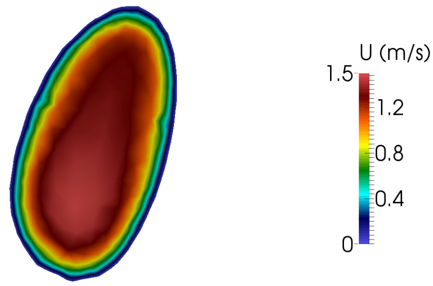

Figure 3: Velocity field through P3, for the exertion case, with a laminar flow solver.

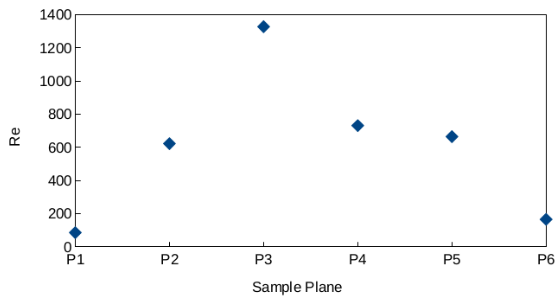

Figure 4: Reynolds numbers through the six sample planes, for the exertion case, with LES model.

flow be indicative of turbulence. At the very least it suggests that the use of a turbulence model should be considered.

For the tidal breathing case, the Reynolds numbers are lower, however if we consider the lower threshold to transition, then an *Re* value of 664 at P3, would indicate, transitional flow. Suggesting that a laminar flow solver may not itself be sufficient tor resolve the flow in the region around the pharynx. It does however, appear that the flow has returned to a laminar state by P4 (about 2/3 down the trachea).

Given the potential issues in identifying the need for turbulence modelling from consideration of Reynolds number alone, the vorticity was also considered. The vorticity for the turbulent, exertion case is shown in Figure 5, with the regions of high vorticity indicated in colour.

As indicated in Figure 5 the regions of maximum vorticity occur in the region in between and around P2 and P3. The additional vorticity resolved using the LES model, combined with the suggested turbulent flow in the exertion case, indicates the need to implement a turbulence model in the exertion case. While this may not be necessary for flow in the lungs themselves (i.e. a regions below P5), it will be necessary in order to resolve the flow in the upper respiratory tract.

This need is less evident in the tidal breathing case, where vorticity is 2 orders of magnitude lower. However, given the presence of transitional flow through the pharynx (P3), it may need to considered if the flow in this region needs to be resolved. For the purposes of simulating flow in the lungs, it may however be unnecessary, as any effect of transitional flow will likely be diminished by the time the flow reaches the lower part of the

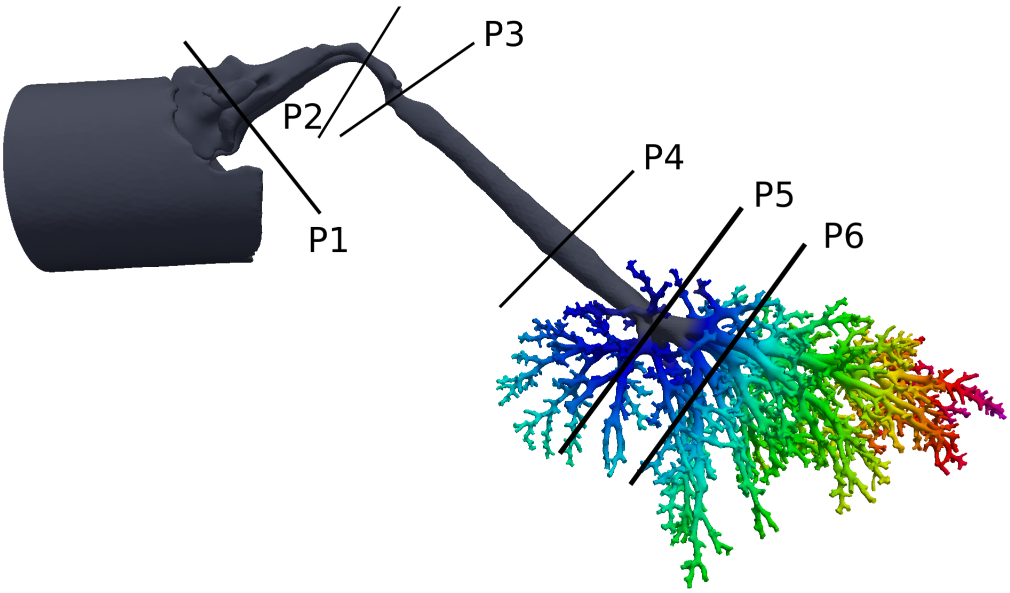

Figure 2: The geometry showing the locations of the sample planes. The geometry is coloured by expansion, with the violet regions indicated maximum expansion and the grey areas, minimum/no expansion.

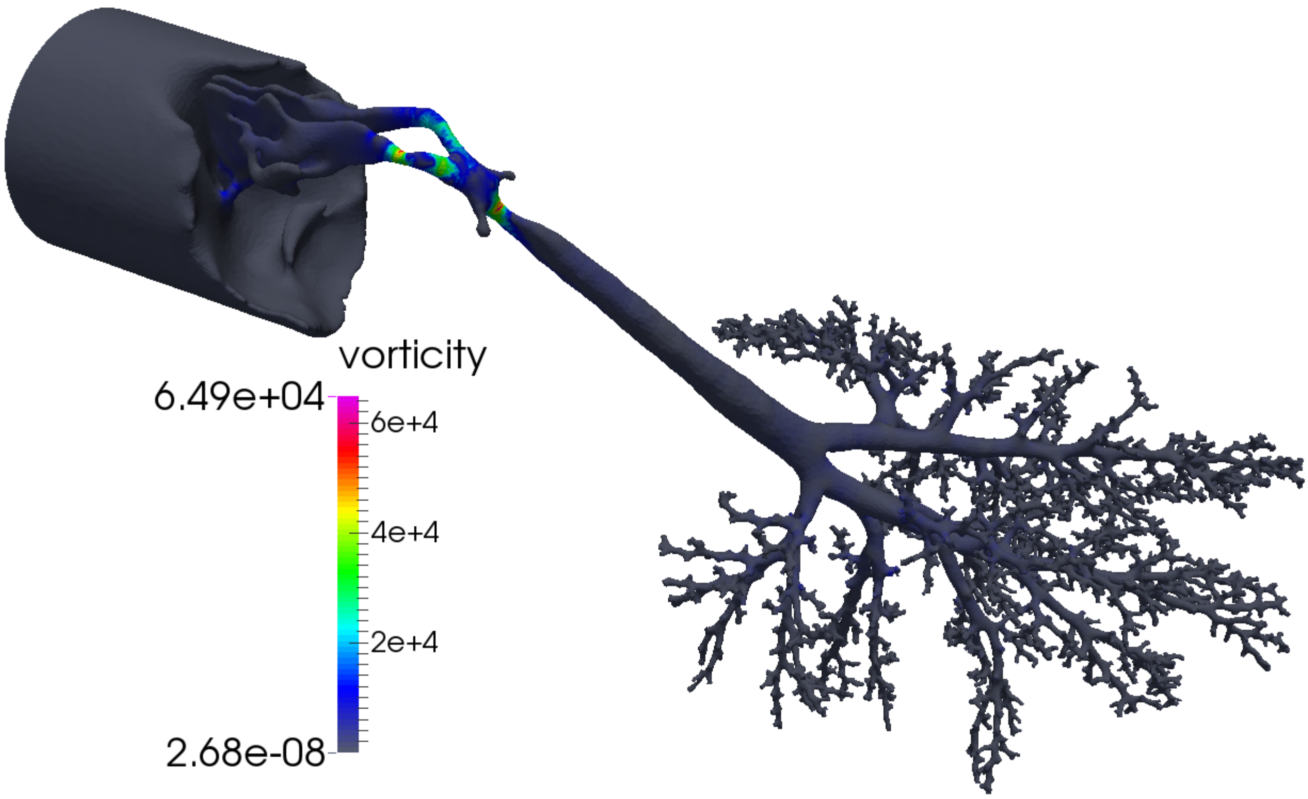

Figure 5: Image of the lung geometry, indicating areas of vorticity. The colour scheme ranges from violet for highest vorticity, to grey for the lowest.

trachea (P4).

If however, particle deposition was to be considered, as in the case of the delivery of aerosolised medications, the influence of transitional flow through the pharynx may not be able to be ignored. This would need to be investigated further.

Conclusions

Simulations were conducted using both laminar and turbulent flow models on otherwise identical cases. Given the pulsatile nature of the flow, and the structure of the geometry it is likely that transitional and turbulent flow occur at lower Reynolds numbers than in a simple pipe flow scenario. Based on the experimental work of Winter and Nerem [11] it appears that transitional flow appears, in some regions, in both the tidal breathing and exertion cases. This suggests that turbulence modelling will be necessary in the case of elevated breathing rates, as in our exertion case. Even if this is just to ensure the flow in the upper airway is fully resolved. In the tidal breathing case only one plane indicated transitional flow, which does not persist to the next sample plane. Suggesting that any effects of transitional flow may be isolated around the pharynx. It may therefore still be appropriate to use a laminar flow solver in this case. However, if other factors, such as particle deposition, were to be considered, the influence of any transitional flow in this region would need to be considered.

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