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Towards parametric modelling of human bronchial tree for computational fluid dynamics

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Abstract. This work discusses the development and application of a parametric CAD model of the human bronchial tree, for use in computational fluid dynamics simulations. The model, which represents the trachea, bronchi, and early airway bifurcations, is based on geometrical parameters derived from existing literature. It can be edited by easily varying parameters in an external spreadsheet, offering an efficient alternative to patient-specific models, which often require the use of time-consuming segmentation procedures. The developed model was utilized to run fluid dynamic simulations, including a scenario that represents respiratory system dysfunctions. These are typical of diseases such as acute respiratory distress syndrome, which can be also triggered by recently emerged COVID-19. The results of these simulations were critically analyzed: they turned out to be consistent with the stated objectives and methods, even in the context of the existing literature. The paper concludes by discussing the limitations and potential improvements of the research.

Keywords: bronchial tree, parametric modelling, CFD, ARDS.

1 Introduction

The respiratory system enables the body to exchange gases between the air and blood, and between the blood and the body's billions of cells. Most of the respiratory system helps to distribute air (conductive zone), while the alveoli and the alveolar ducts are responsible for actual gas exchange (respiratory zone). Computational modelling of the airflow inside the airways is of great importance in scientific research and clinics, due to its ability to provide detailed insights into the complex flow dynamics and transport phenomena within the respiratory system, in a fast and non-invasive way. Moreover, it can aid in the optimization of therapeutic strategies and the design of medical devices, thereby having significant implications for patient care and treatment outcomes [1].

In this work, we propose a framework for the airflow simulation inside the bronchial tree, leveraging on an easily adjustable parametric model of the trachea, bronchi, and early airway bifurcations, built upon geometrical data derived from the literature. This

parametric model is implemented to provide an alternative to patient-specific models (e.g., [2]), which necessitate the use of imaging techniques and time-consuming segmentation procedures. This can be realized without the need to acquire direct images from a patient, and can be easily adapted based on a few simple parameters.
Derived CAD models can be imported into the simulation environment, to run computational fluid dynamics (CFD) studies. With this regard, to realistically characterize not only the distribution of the airflow in the airways, but also the pressure field, a Matlab (MathWorks) code for non-zero pressure outlet boundary conditions was developed.
In this framework, a method for including in the model dysfunctions of the respiratory system, linked to the partial derecruitment of the alveoli, is proposed. This type of dysfunction is typical of diseases such as acute respiratory distress syndrome (ARDS) [3].
From the mid-900s until today, modelling and data acquisition techniques have evolved, from the first direct measurements to statistical analysis, up to modern imaging techniques and computational software. In 1971, Horsfield models [4] were developed, and they are still used nowadays as a reference to validate the most recent computational models of pulmonary morphology and activity. Horsfield models are based on data obtained from the measurement of a resin cast of a normal human bronchial tree. More recently, in [5], a hybrid 3D and 1D model of the airways was proposed. The 3D model, based on CT scan imaging, represents the upper airways down to the seventh generation, while the lower airways are modelled as 1D. The 1D model is used to define the outflow boundary conditions for the 3D model. Monjezi et al. [6] employed a similar approach, with a 3D model for the upper airways and a 1D model for the distal part. The acinar region is represented by lumped parameters. The impedance of the acinar and distal parts is calculated for use as outflow boundary conditions. Eventually, Elcner et al. [7] proposed a technique to calculate the pressure drop of air within the lung's airways, useful for estimating boundary conditions for a partial lung model.
The manuscript follows the subsequent structure: in the 'Material and Methods' section, we detail the creation of our model, while the 'Results' section presents the outcomes of our simulations, including the modelling of the pathological condition, and a comparison of our results with the Horsfield data. Finally, in the 'Discussion & Conclusions' section, we reflect on the utility of our model, its limitations, and potential areas for improvement.

2 Material and methods

2.1 Parametric CAD model generation

In the development of the 3D model of the bronchial tree, the work by [4] has been taken as the main reference: it takes into consideration the asymmetry of the airways inside each lung. This asymmetry is evident since the first bifurcation, being the left lung is significantly smaller due to the presence of the heart in the thoracic cavity. The Horsfield model provides information about the diameter, length and theoretical percentage of airflow for each duct of the airways, until the terminal respiratory bronchioles. The alveoli are not included in this model.

In this exploratory work, it was decided to implement a model down to the 4th airways generation, exploiting Inventor (Autodesk) as CAD software. A parametric approach has been adopted: the geometrical characteristics and the spatial disposition of the elements of the model are fully editable by changing the values of some key parameters.

The 3D model is defined starting from elementary units, consisting of a branch and a bifurcation. This elementary part is defined starting from circular sections (highlighted in green in Figure 1). The first two sections, starting from the top, are of the same diameter, and are associated with the same parameter. Consequently, this part of the model will always be cylindrical, regardless of the choice of parameters. The other two circular sections are each associated with a different parameter; this allows us to obtain an asymmetrical airway model, with airways of different sizes branching off from the same portion. Using the *loft* feature, the cylinder section is joined with the sections relative to the two bifurcations. The model is entirely solid, and the intersecting parts are joined with a Boolean operation. The axes of the circular sections of the bifurcation are inclined with respect to the axis of the cylindrical part. These inclination angles are defined by two parameters, so that the geometry of the bifurcation can be characterized. Each element can be rotated around the axis of the main duct, so the model can be developed in three-dimensional space, not just in the front plane (Figure 1).

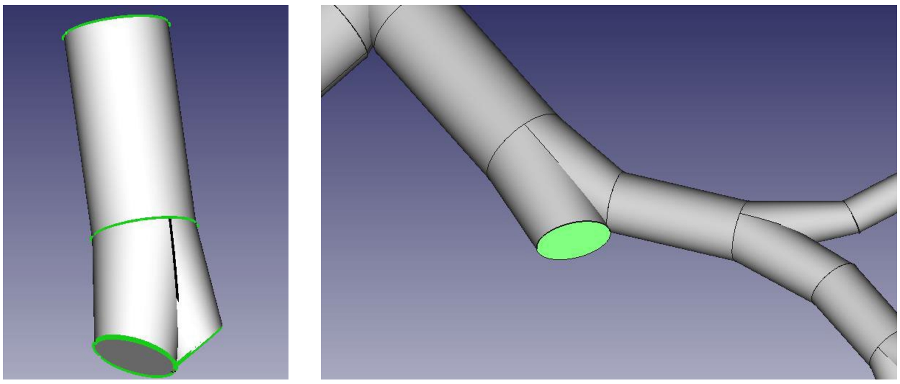

Fig. 1. On the left, basic unit of the parametric tree. On the right, tree assembly under construction

The parameters that determine the model are easily modifiable through an external calculation sheet linked to the Inventor file. The spreadsheet interface has been made as intuitive as possible (Figure 2).

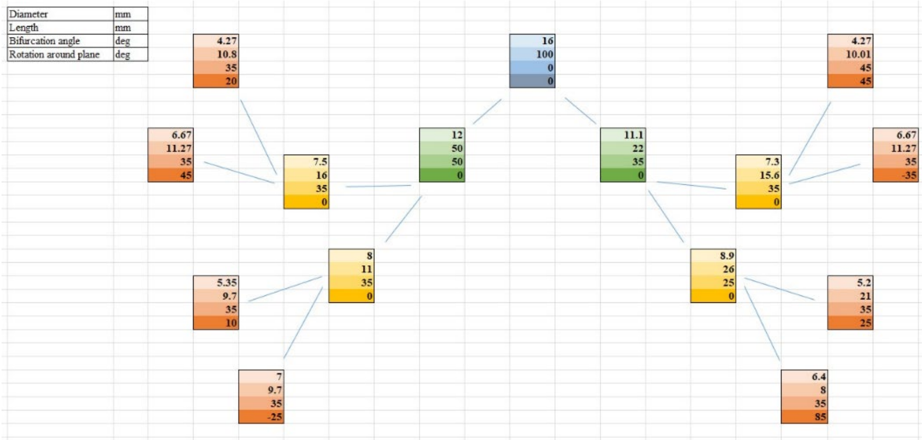

Fig. 2. External Excel spreadsheet to easily control the parameters of the model

The advantage of a parametric approach of this type makes it possible to locally vary the geometric characteristics of the model, to adapt it to simulate the airways of different subjects. In addition, it could be suitable for quickly representing a localized occlusion in a specific branch of the airway.

2.2 CFD simulations

The model of the airways that was generated is composed of multiple blocks joined together. When generating a mesh (tetrahedral), it is useful to match the cell vertices at the interfaces between two blocks. The part of the model that represents the airways with larger diameters was divided into elements with an average size of 2.5 mm. The same element size cannot be applied to the ducts with smaller diameters, so the mesh in these zones has been refined through the “sphere of influences” method. Reference systems have been placed in correspondence with the outlets of the model. Each reference system was used to define the centre of a sphere of influence, with 0.8 mm size elements.

The solver used for the analysis was Fluent (Ansys). For simplicity, steady-state simulations were performed. Walls were set as rigid, with no-slip shear condition. Air was simplified as an incompressible fluid [8]. The inlet boundary condition represents the maximum volume flow rate during light exercise, which was estimated at 0.8 l/s. Considering the inlet circular section with a diameter of 16 mm, the volume flow rate is equivalent to a fluid velocity of around 4 m/s, perpendicular to the inlet section.

Preliminary analyses were performed with a “traction-free” condition. This means that a zero-pressure boundary condition was imposed at each outlet of the models. In this case, the flow distribution and pressure inside the models are going to be entirely determined by the geometry of the models and the inlet boundary condition. The inlet flow rate is known, and the flow rate at each outlet of the model is calculated by the software.

Then, to refine the model, starting from the work by Park et al. [9], a Matlab code was written to calculate the resistance to flow of the lower generations airways. The Matlab code receives in input, for every outlet, the diameter and the outflow, calculated by the traction-free simulation. Knowing the values of resistance and flow at each outlet, non-zero pressure boundary conditions can be imposed. The Poiseuille formulation for airway resistance ($R_{Poiseuille}$) is used to estimate the total resistance of the airways downstream the branches of the 3D model:

$$R_{Poiseuille} = 8\mu L/\pi r^4,$$

where μ is the dynamic viscosity of the fluid, L is the length of considered pipe and r is its radius. To define r, a diameter reduction coefficient between consecutive generations equal to k=0.79 was considered [9]. The corresponding value of L was determined from a polynomial interpolation of experimental data derived from [4]. The polynomial equation was then used to get the values of length associated with the diameters.

By knowing the resistance, the total pressure drop between the outlets of the upper model and the alveoli can be estimated.

With the implemented model, the presence of a pathological partial derecruitment of the alveoli was simulated, too. We were interested in observing how the pressure range and the airflow vary in these conditions. In order to model a disease like ARDS, an assumption has to be made regarding the reduction of airflow that passes through the different branches of the model (increased resistance). It has been assumed that the lower part of the left lung is affected by a partial derecruitment and residual capacity is reduced to one-half. In Ansys Fluent, in correspondence with the left outlets, a “target mass flow rate” is imposed equal to half of the mass flow rate observed for the same model, but in normal conditions. At first, this method was applied to the traction-free model, with zero pressure boundary conditions, and then extended to the refined model.

3 Results

The resulting CAD model was exported in IGES format, to be employed in Ansys Fluent. After running the CFD simulations, pressure and velocity fields were analysed, both for the physiological and pathological cases. The resulting maps for the “non-zero” pressure boundary condition simulation are reported in Figure 3.

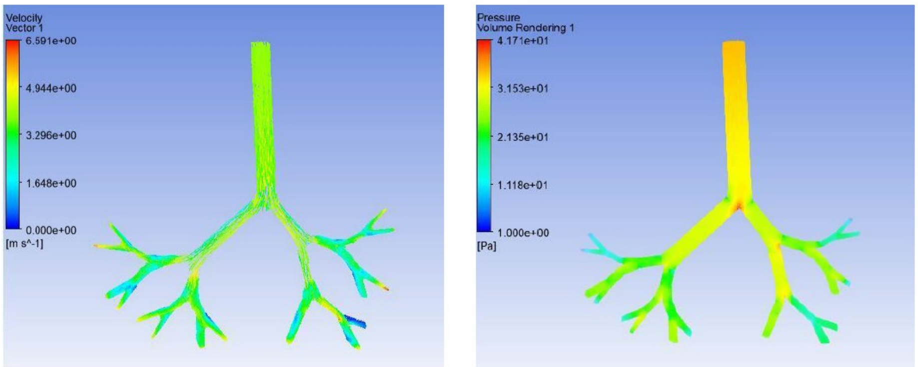

Fig. 3. On the left, velocity map for the simulation of the physiological condition. On the right, the corresponding pressure map

The percentage of the air that exits from each outlet is compared to the results from the traction-free simulation and to the theoretical values estimated by [4] (Table 1).

Table 1. Results comparison among the model with refined boundary conditions (BC), "traction-free" model (TF) and Horsfield work (H)

<table>
<tr><th></th><th>Flow rate [kg/s]</th><th>% Flow rate BC</th><th></th><th></th><th>% Flow rate TF</th><th></th><th></th><th>% Flow rate H</th><th></th><th></th></tr>
<tr><td>Outlet_1l (left)</td><td>3.90E - 05</td><td>4.0</td><td rowspan="4">20.2</td><td rowspan="8">48.9</td><td>2.5</td><td rowspan="4">19.1</td><td rowspan="8">49.4</td><td>3.8</td><td rowspan="4">18.9</td><td rowspan="8">45.0</td></tr>
<tr><td>Outlet_2l</td><td>3.60E - 05</td><td>2.5</td><td>0.2</td><td>1.5</td></tr>
<tr><td>Outlet_3l</td><td>8.60E - 05</td><td>8.9</td><td>12.6</td><td>9.9</td></tr>
<tr><td>Outlet_4l</td><td>4.70E - 05</td><td>4.8</td><td>3.8</td><td>3.8</td></tr>
<tr><td>Outlet_5l</td><td>6.90E - 05</td><td>7.1</td><td rowspan="4">28.7</td><td>5.9</td><td rowspan="4">30.2</td><td>5.2</td><td rowspan="4">26.1</td></tr>
<tr><td>Outlet_6l</td><td>3.50E - 05</td><td>3.6</td><td>3.1</td><td>2.0</td></tr>
<tr><td>Outlet_7l</td><td>1.13E - 04</td><td>11.6</td><td>16.0</td><td>13.7</td></tr>
<tr><td>Outlet_8l</td><td>7.10E - 05</td><td>6.3</td><td>5.2</td><td>5.2</td></tr>
<tr><td>Outlet_1r (right)</td><td>9.30E - 05</td><td>10.1</td><td rowspan="4">30.2</td><td rowspan="8">51.1</td><td>12.6</td><td rowspan="4">29.2</td><td rowspan="8">50.6</td><td>18.9</td><td rowspan="4">36.0</td><td rowspan="8">55.0</td></tr>
<tr><td>Outlet_2r</td><td>6.00E - 05</td><td>6.2</td><td>4.7</td><td>7.2</td></tr>
<tr><td>Outlet_3r</td><td>1.17E - 04</td><td>12.1</td><td>7.8</td><td>7.2</td></tr>
<tr><td>Outlet_4r</td><td>1.10E - 05</td><td>1.9</td><td>4.1</td><td>2.8</td></tr>
<tr><td>Outlet_5r</td><td>6.80E - 05</td><td>8.0</td><td rowspan="4">20.9</td><td>12.6</td><td rowspan="4">21.4</td><td>9.9</td><td rowspan="4">18.9</td></tr>
<tr><td>Outlet_6r</td><td>6.10E - 05</td><td>6.3</td><td>4.3</td><td>3.8</td></tr>
<tr><td>Outlet_7r</td><td>3.30E - 05</td><td>3.4</td><td>2.7</td><td>3.8</td></tr>
<tr><td>Outlet_8r</td><td>3.10E - 05</td><td>3.2</td><td>1.8</td><td>1.4</td></tr>
</table>

The simulation with non-zero pressure boundary conditions shows more consistent results regarding the distribution of the flow rate. However, even for the traction-free model a similarity between simulated flow distribution and the Horsfield data is noticeable. This fact underlines that the geometry itself strongly influences the airflow distribution inside the different regions of the lungs, as mentioned in the work of [5].

The values of pressure at the outlets are known, as they have been imposed as boundary conditions. At the inlet, a maximum pressure of 35.3 Pa is observed.

Coming to the modelling of the pathological case, results show an evident increase of pressure at the outlet sections (Figure 4), where the reduced functional capacity of the lung is modelled. In that area, the airflow is reduced, so a higher flow is diverted through the other branches of the model. The pressure rises to balance the reduced flow, which in normal conditions would be higher. Remembering the direct proportionality between pressure drop and flow in the airway, with the resistance as a constant of proportionality, a higher pressure is noticeable also in the parts of the model that are not modelled as directly affected by the reduced capacity.

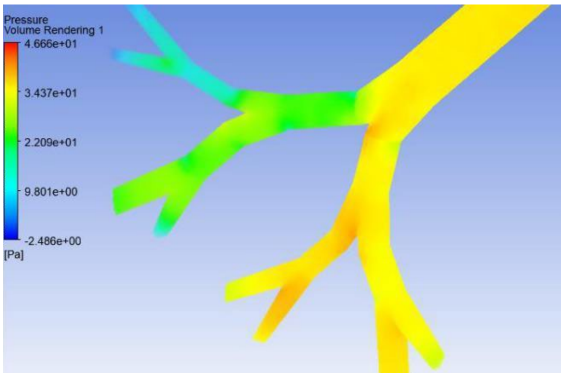

Fig. 4. Pressure map for the simulation of the pathological condition

4 Discussion & Conclusions

In this work, a parametric CAD model of the trachea, bronchi, and early airway bifurcations was implemented. The model was implemented to offer an alternative to patient-specific models, which require the use of imaging techniques and time-consuming segmentation procedures. Even if through segmentation accurate patient-specific models of the anatomy of interest can be obtained, and can also be used for dimensioning or virtual positioning of medical devices [10], this may not always be necessary. In this sense, a parametric CAD model can sometimes offer a more versatile and easily manageable alternative, also avoiding the extensive post-processing step needed to make a segmented model topology suitable for simulation.

It has been highlighted that a localized disease leads to a significant increase of pressure in the proximity of the diseased area, but also in that one not affected by the disease. Patients affected by ARDS, or serious infection caused by COVID-19, need to rely on mechanical ventilation. The parameters of mechanical ventilation need to be accurately selected to guarantee a sufficient gas exchange between alveoli and blood vessels, but at the same time, it is crucial to avoid ventilator-induced lung injuries [11]. These results show that the not-diseased part of the ventilated lung is potentially at risk from volutrauma due to an increase in lobular pressure over a normal functioning lung.

Concerning the adopted mathematical modelling, the air was treated as an incompressible fluid, as the pressures found in the lungs are not so high as to necessarily

take into account the compressibility of the air. Clearly, properly modelling air compressibility would increase the robustness of the method. Another possible significant improvement would be to consider boundary conditions in transient simulations, rather than stationary, in order to model the entire respiratory cycle.
In summary, this work has laid a solid foundation for the use of a preliminary parametric model in simulating human airways. With further refinement and validation, such models could become a useful tool in the study of respiratory disorders.

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