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Numerical study of a 5-generation Weibel lung compared to experiments

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ABSTRACT

For a better understanding of the human breathing and the improvement of artificial respiration, it is necessary to understand pressure drops and volume flow in the human lung. The importance of this topic is shown by the fact that 50% of long-term ventilated patients die within one year after artificial respiration due to baro- or volu-traumata (Raymondos et al. 2017) To research this topic, a 5-generation lung model with transient numerical simulations and experiments is created.
Simulating the whole lung with its 23 generations is nearly impossible, since it would imply to implement the 2^{23} branches of the bronchial tree. To reduce the model of the lung, it is necessary to know pressure drop and volume flow in a single branch. Therefore, the impact of the pressure losses on one bifurcation to another must be known.
In order to do experimental research, a 5-generation lung model based on the human lung model of Weibel (1963) was designed. The geometry was scaled up with Reynolds similarity. The model is closed up after the last bifurcation with an elastic component on each branch, which follows that only the inhaled air can be exhaled. Within the test rig the volume in- and outflow can be measured as well as the pressure in every branch of the lung.
Numerical simulations with the model were done in parallel to the experimental studies. Since the pressure drop velocity component is at the speed of sound, the pressure will work in opposite direction than the volume flow, which makes it challenging to realize the zero crossing of the volume flow at the inlet.
In order to solve this problem, a closed box is placed at every opening of the 5th generation. Within these boxes, the pressure decreases and only the incoming air can be exhaled. Therefore, no opening is necessary, only an inlet will be used for in- and outflow.
The results of the numerical simulation were validated by the experimental results. As a conclusion the extent of the numerical model can be reduced to a single branch which decreases the calculation time of the simulation. Furthermore, both methods give deeper insight on pressure distribution in the human lung. This leads to an improvement of artificial respiration, and therefore decreases the cases of damaged lung tissue due to baro- or volu-traumata.

1. INTRODUCTION

The function of the respiratory system, consisting of lung and airways, is the exchange of oxygen and carbon dioxide in the human body. If this function cannot be assured independently by the human body, because of respiratory disorder or during surgery with anesthetics, the body has to be supported by artificial respiration (Clauss et al. 2009) During artificial respiration, the lung tissue can be damaged in consequence of the application of too high pressure or volume. A German clinical study from Raymondos et al. (2017) shows that 50% of long-term-ventilated patients with ARDS (acute respiratory distress syndrome) die within one year after artificial respiration.
Consequently, it is important to gain a better understanding of volume and pressure in the human lung, so that the cases of volu- or baro- traumata due to artificial respiration can be reduced. To research this topic, numerical simulations and experiments are used.
As shown by other works, such as Finlay & Martin (2008), Comer et al. (2001) and Kolanjiyil et al. (2017), which are mainly focusing on particle transport in the lung, it is possible to use only one branch of the bronchial tree for a numerical simulation of the human lung. To be able to do so it is necessary to completely understand the flow and pressure conditions in the whole lung. To validate the numerical simulations, experiments are used. The base of the experimental and the numerical lung model is given by the human lung model of Weibel (1963).

2. MATERIAL AND METHODS

The bronchial tree of the human lung as described by E. R. Weibel consists of 23 generations with a total of 2^{23} branches. Starting from the trachea the branches are split up into two equal, smaller branches at every bifurcation of each generation. The upper airways, which will be of main interest in the following, can be approached as stiff tubes since they are made of ring cartilage. In the experiment as well as in the numerical simulation, only inspiration is studied. That is because inspiration is the critical part of the breath cycle during artificial respiration. The here applicated volume flows and pressures may lead to volu- or baro-traumata and thereby tissue damage in the lung. The expiration is a passive process and hence controlled neither during relaxed natural breathing nor during artificial respiration. Therefore, expiration is not critical for tissue damage during artificial respiration.

2.1 Numerical Model

Since it is impossible to simulate the whole lung with its 2^{23} branches, only the branches of generation 0 to 5 are used for the numerical model. Furthermore, the highest volume flows and pressure drops are expected in the first generations. The values for diameter and length of each branch are given by the lung model of E. R. Weibel and are shown in Tab. 1. The geometry of the numerical lung model is pictured in Fig. 1.

Tab. 1 Dimensions of the numerical lung model

Generation	Diameter [m]	Length [m]
0	0.0180	0.1200
1	0.0120	0.0476
2	0.0083	0.0190
3	0.0056	0.0076
4	0.0045	0.0127
5	0.0035	0.0107

To prevent overlapping of the branches, each branch of generation 4 and 5 is rotated by 5 degree to the branch in the generation before.

For the mesh, every branch and bifurcation was meshed individually and the tubes were filled with two C-grids. This leads to a total of 2,309,274 elements and a y+ of one at the highest volume flow.

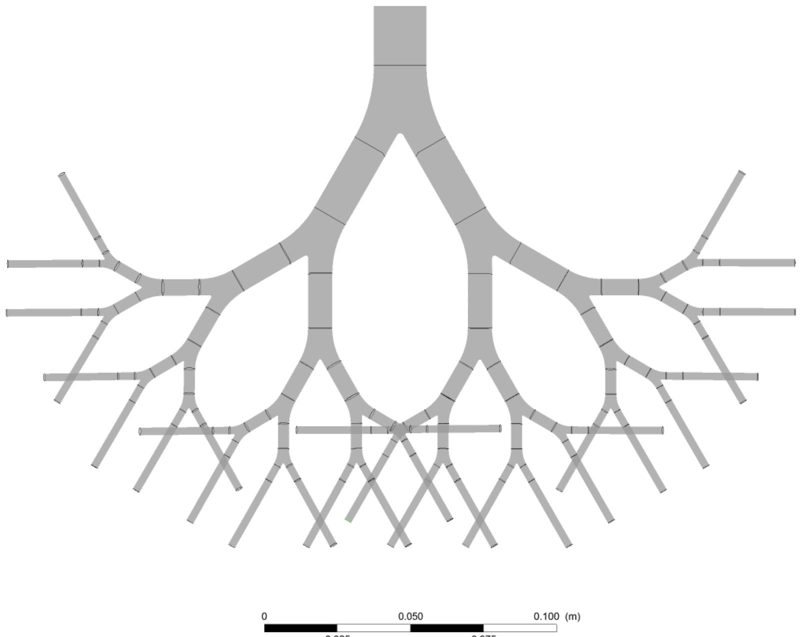

Fig. 1 Geometry of the numerical lung model

For the simulation, the program ANSYS CFX is used. As turbulence model, the shear stress transport model (SST) with gamma theta transition model is chosen. That is because, depending on the volume inflow, turbulent flow conditions can be expected in the first generations, but always laminar flow conditions in the higher generations.

As a reference point for the value of the volume flow in the simulation, relaxed human breath cycles of volunteers were recorded and averaged. This leads to a maximum volume flow of 46 l/min during inspiration. Consequently, the volume flows 30, 40, and 50 l/min were simulated.

2.2 Experimental model

The basis of the experimental model was also the geometry of E. R. Weibel. To improve handling and measurements especially of the smaller generations, the geometry was scaled up by the factor 0.72^{-1}. For that Reynolds similarity was used at the point of the highest volume inflow during inspiration.
Reynolds similarity means that the Reynolds numbers Re

$$Re = \frac{u \cdot d}{\nu} \tag{1}$$

in the numerical model and the experimental model are identical at the chosen point of the breath cycle despite the scale up. As a consequence the flow conditions of the two models are the same. Therefor the velocity u of the fluid has to be divided by the factor 0.72^{-1} by which the diameter d of the branch is multiplied. ν is the kinematic viscosity of the fluid and stays constant through the scale up. From the continuity equation

$$Q = u \cdot A = const., \tag{2}$$

where Q is the maximum volume flow of the fluid and A the cross-section area of the considered branch, it follows that the volume flow Q in the experiment has to be scaled up by the factor 0.72^{-1}.
Finally the pressure p

$$p = \frac{1}{2} u^2 \tag{3}$$

from the numerical simulation, has to be divided by the factor 0.72^{-2} so that it can be compared to the experimental results. An overview of the scaling factors is given in Tab. 2.
The resulting diameters and lengths for the experimental model are shown in Tab 3.

Tab. 2 Scaling factors from numerical model to experimental model

parameter	experimental model	numerical model
diameter	$0.72^{-1} \cdot d_{model}$	d_{model}
length	$0.72^{-1} \cdot l_{model}$	l_{model}
velocity	$0.72 \cdot u_{model}$	u_{model}
volume flow	$0.72^{-1} \cdot Q_{model}$	Q_{model}
pressure	$0.72^{2} \cdot p_{model}$	p_{model}

Tab. 3 Dimensions of the experimental lung model

Generation	Diameter [m]	Length [m]
0	0.0250	0.1667
1	0.0169	0.0611
2	0.0115	0.0264
3	0.0078	0.0106
4	0.0063	0.0176
5	0.0049	0.0149

All parts of the lung model were printed by a 3D printer. The experimental lung model is pictured in Fig. 2.

The complete experimental set-up consists of the lung model, sensors for pressure and volume and a fan. The pressure sensors are connected to each branch of every generation of the lung model. Within the model, in the small rings around the branches, the pressure is averaged over four points around the branch. The fan is connected to the trachea, which can be seen in the top of Fig. 2. The volume flow sensor is placed in between the trachea and the tube of the fan.
The volume flow in the experiment has to be scaled up according to Tab. 2. The resulting values compared to the numerical values are shown in Tab. 4.

Tab. 4 Volume flow for simulation and experiment

volume flow simulation [l/min]	volume flow experiment [l/min]
30	42
40	56
50	69

For the comparison of the numerical data with the experimental data, the scale up from one to another has to be taken into account, too.

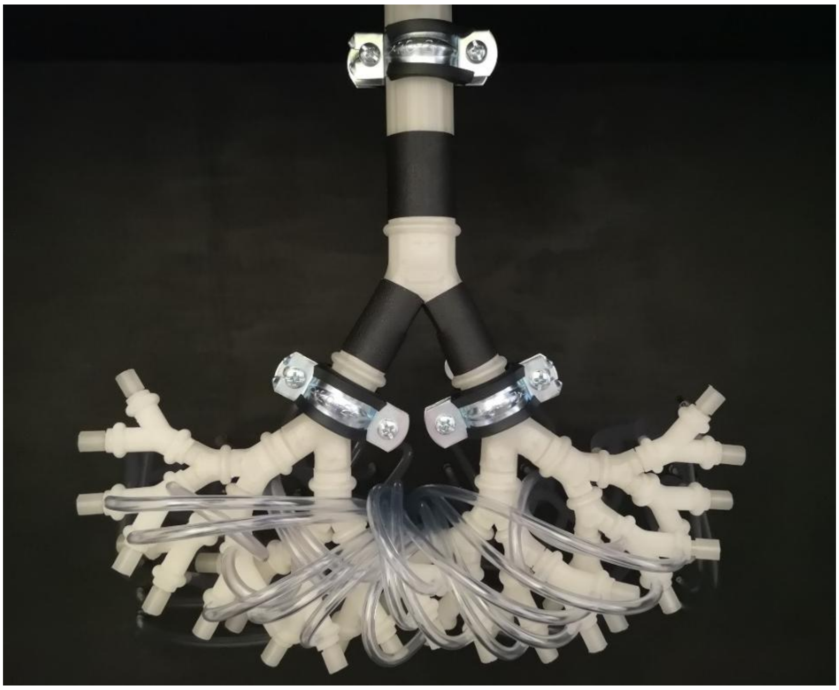

Fig. 2 Experimental lung model

3. RESULTS

In the following, the results of the numerical simulations are scaled according to Tab. 2 for the comparison with the experimental data.
The pressure in simulation and experiment from generation 0 to 5 for the different volume flows is shown in Fig. 3.

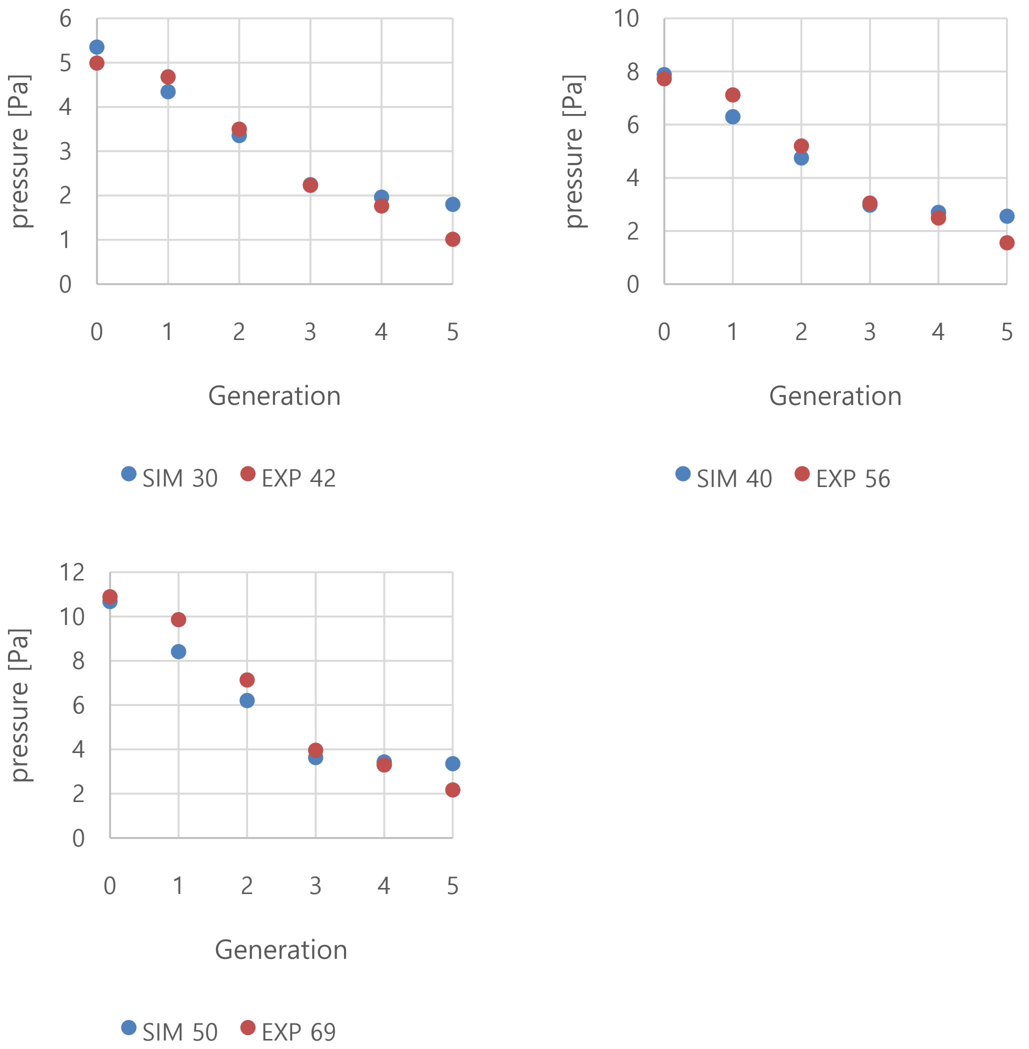

Fig. 3 Pressure in generation 0 to 5 for the volume flow combinations: simulation 30 l/min experiment 42 l/min, simulation 40 l/min experiment 56 l/min, simulation 50 l/min experiment 69 l/min

The velocity distribution for generation 5 is shown in Fig. 4. Because the two sides of the bronchial tree are axially symmetric, only the results for the left half of the lung model is shown here. The branches are counted from left to the middle of the lung model.

Figure 4 Velocity distribution in generation 5 for the volume flow combinations: simulation 30 l/min experiment 42 l/min, simulation 40 l/min experiment 56 l/min, simulation 50 l/min experiment 69 l/min

The velocity distribution in the whole numerical lung model is pictured in Fig. 5.

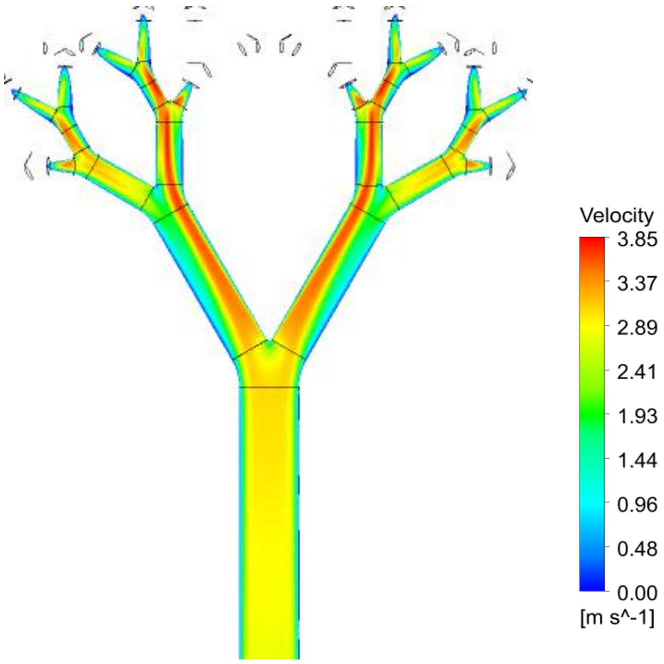

Fig. 5 Velocity distribution in the numerical lung model

4. DISCUSSION

In general, the results of the experiments and the simulations shown in Fig. 3 and 4 align well. So the simulation can be validated by the experiments.
It becomes clear that the pressure and the pressure loss depend on the incoming volume flow and the generation. With increasing volume flow pressure and pressure loss increase especially in the generations 0 to 3, but also in the higher generations. Here the pressure and pressure loss are lower than in the lower generations. From this, it follows that the main pressure loss is caused by generations 0 to 3 while the highest pressure loss is from generation 2 to 3. Generation 3 is a special case, because it has the lowest length to diameter ratio, which leads to the conclusion that pressure loss is also influenced by the length of the tubes. The difference of the pressure values in generation 5 between simulation and experiment that can be observed for every volume flow could be explained by the difference in length of the end parts in the experimental and the numerical model after the 5^{th} generation.
The velocity distribution over generation 5 shown in Fig. 4 is inhomogeneous over the branches. In Fig. 5, it can be seen that this phenomenon is not limited to the 5^{th} generation but can also be observed in the rest of the model from the second bifurcation on. That is due to the fact, that there is no fully developed flow within the branches, except for generation 0, the trachea. Here the tube is long enough to reach the point of a fully developed flow. Thus, the airflow is split up evenly at the first bifurcation. Due to the inertia of the flow it is attached to the inner walls of the 1^{st} generation. Because of the short length of this generation, no fully developed flow is reached until the next bifurcation. Here the flow does not hit the middle of the

bifurcation but, when following the flow on the right hand side of the lung model in Fig. 5, the inner wall of the left of the branches that emerge from this bifurcation. Consequently, the flow is not split up evenly and the main flow follows the path of the left branch. From there on this procedure can be observed in every generation, which leads to different velocities respectively volume flows and pressures in the branches of one generation.

By comparing the numerical with the experimental results, it stands out that the pattern of pressure gradient and velocity distribution differ less for different volume flows for the simulation than for the experiment. This leads to the assumption that there may be factors influencing the experiments that are not regarded by the simulation, yet. Such a factor could be the roughness of the surface in the experimental model while the surface in the numerical model is assumed as smooth.

5. CONCLUSIONS

By the numerical and experimental investigation of a 5-generation lung model, a better insight in pressure and pressure loss as well as velocity distribution, and by that also volume flow and pressure distribution, in the human lung was given.
Pressure and pressure loss in the lung depend on incoming volume flow, generation of the bronchial tree and on the length of the tubes. This is shown by the fact that the highest pressure loss is observed from generation 2 to 3 where generation 3 is the generation with the lowest length to diameter ratio. Furthermore, it became clear that the main pressure loss is caused by generations 0 to 3.
Because of the good alignment of experimental and numerical results, the simulation can be validated by the experiments, even though there may be a factor influencing the experiment that has not been taken into account in the simulation, yet. However, the reduction of the model is difficult because of the inhomogeneity in velocity respectively volume flow and pressure in the branches of one generation. Which is shown in detail for generation 5. A reduction to half of the model is possible because of the axially symmetry of the used Weibel model for the human lung. A further reduction could be done depending on the aim of the investigations done with the model. For tissue damage due to too high volume flow or pressure for example, the branches with the highest pressure and volume flow would be used, and for investigating the less aerated lung parts during artificial respiration, branches with low volume flow and pressure. Thereby an improvement of artificial respiration can be reached.

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