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Pressure losses and volume flows in a 5 generation human lung model

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Abstract: *To understand the human breathing and get better mechanical ventilation systems it is important to understand pressure drops and volume flow. To investigate a 5-generation lung model with instationary numerical simulations is created.*

The whole lung with 23 generations is nearly impossible to simulate, because it would imply to implement the 223 branches of the bronchial tree. Knowledge of pressure drop and volume flow in a single branch aids in reducing the extend of the whole model. For this, the impact from the pressure losses on one furcation to another must be known.

The geometry of the simulation is based on the lung model of weibel and is fully parametrized. Depending on the volume flow the flow conditions will change, there for a SST with gamma-theta turbulence model is used.

The zero crossing of the volume flow at the inlet is difficult to realize, because velocity of the pressure is sonic speed. Consequently, the pressure will work in opposite direction then the volume flow. To challenge this problem, at every opening in the 5th generation a closed box will be placed. The pressure decreases in this box and only the incoming air can be exhaled, so no opening is necessary, only an inlet will be used for in- and outflow. The outflow is only pressure driven, like the natural human breathing.

The results were validated with an analytical solution.

With the results of these numerical simulations the whole bronchial tree can be reduced to a model of only one branch.

1 INTRODUCTION

In case of a failure of the human breathing it is necessary to support the human body. In situations of emergency, operations or other disease a mechanical respirator is mandatory. A 2017 German clinical study shows that 50 % of long term-ventilated patients will die in the first year after the artificial respiration. Reason for this can be baro- or volu- traumata [1]. In the first case, the mechanical respiration will realized to high pressure and in the second case to high volume of air in the lung. In both cases, the ventilation air will destroy the lung tissue. To reduce this damage of the tissue the airflow in the lung must be complete understand. For this computer fluid dynamics is an important tool.

With this CFD many papers will be published focused on particle transport in the lung, such as Finlay & Martin, Zhang, Kleinstreuer et al. [2, 3, 4]. These papers often use a lung model with only one branch of the bronchial tree. To use only one branch, it is important to understand the flow conditions in the whole lung. Important parameters to reduce the whole tree are volume flow and pressure loss in every generation and branch. To calculate the pressure loss, the discharge coefficient is necessary. This coefficient is depending of geometry, volume flow and viscosity. In case of a constant viscosity of air in the lung, the three parameter can be reduced to geometry and volume flow. E. Weibel gives the geometry 1963 by plastification human lungs of dead people [5]. This is the base model of the numerical solutions with CFD.

The volume flow will be investigating in this work with numerical simulations. With the results the analytic solution of the discharged coefficient will be adapted for the lung and then it will be possible to reduce the whole bronchial tree to one branch.

2 MATERIAL AND METHODS

In the lung, the trachea is subdivided in two new branches for 23 generations, so in the end it will be 2^{23} branches. This lung model was released 1963 by E.R. Weibel. The first 16 generations are only for the air management and at the other 7 generations, there are the alveoli's for the gas exchange. The upper airways are build out of ring cartilage and can adopted as stiff tube. The diameter of the branches in one generation is equal, every generation has different diameter and length of the branches. So a fully parametrized model of a bifurcation was implemented with the reduction of the diameter. The included angle is defined with 60 degree for all bifurcations, to reduce the effect of rotation. The branches are rotate 5 degree to the next generation, this is necessary that the outlets at the end of the 5 generations do not overlapped (Figure 1).

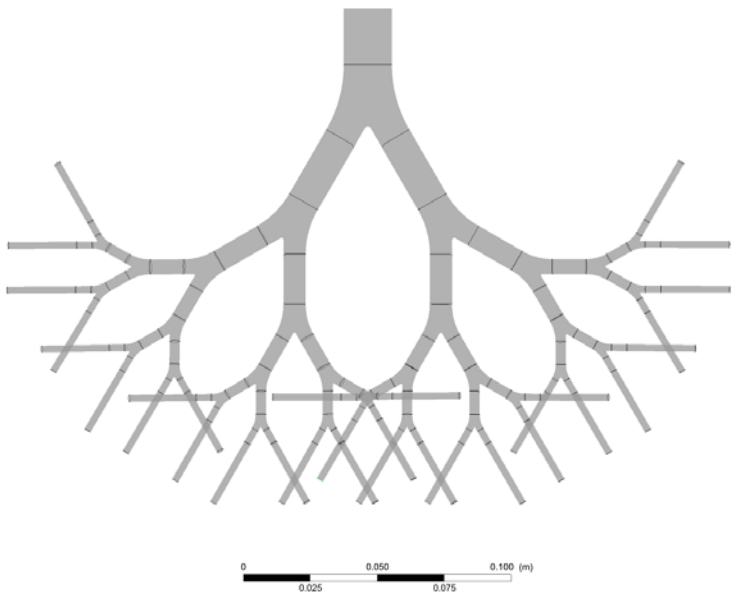

Figure 1: Geometry of the lung model.

The dimensions of the 5 generations are shown in table 1. As inlet is used a stiff tube with the length of $10d_0$ and at every outlet is a stiff tube with the length of 10 d_5.

Generation	Diameter [m]	Length [m]
0	0.0180	0.1200
1	0.0122	0.0476
2	0.0830	0.0190
3	0.0560	0.0760
4	0.0450	0.0127
5	0.0350	0.0107

Table 1: Dimension of the 5 generation lung model.

The volume flow for the transient simulations at the inlet is a sinus function with amplitude of 45 l/min. This is lean to an experimental investigation of the human breathing; the result of the experiment is shown in figure 2 [6]. The mass flows are calculated from the volume flows. The duration of breathing cycle is 7 s.

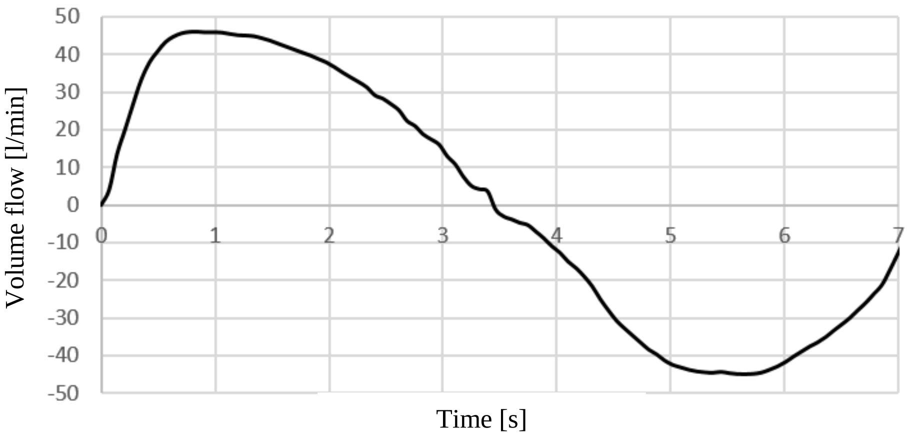

Figure 2: Experimental data of a volume flow within a human breath cylce .

Every branch was meshed alone; the same was with the tubes between the bifurcations. The pipe is filled with two C-grids, also the pipes, so the cell size is the same at the interfaces. The whole mesh, buildup of the bifurcations and the pipes has 2309274 elements. With the highest volume flow amount a y+ of one. The cross-section of the mesh and the mesh of a bifurcation with two pipes will be shown in figure 3.

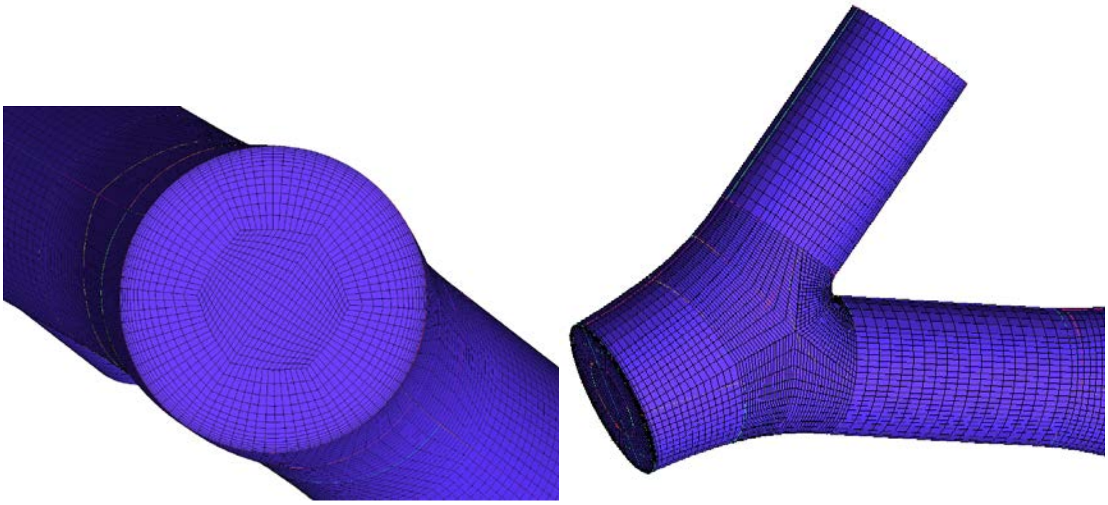

Figure 3: Mesh of the lung model.

Depending of the airflow and the generation are different flow conditions expected. With a high airflow in the first two generations turbulent conditions and with a lower airflow or in higher generations it will be laminar conditions. The curves of the Reynolds numbers are shown in figure 4. The critical Reynolds number, when laminar flow conditions changes to turbulent is normally 2300, but in the case of the lung, the pipes are not long enough to develop a full develop pipe flow. So the transition regime increases and turbulent conditions are possible at lower Reynolds numbers.

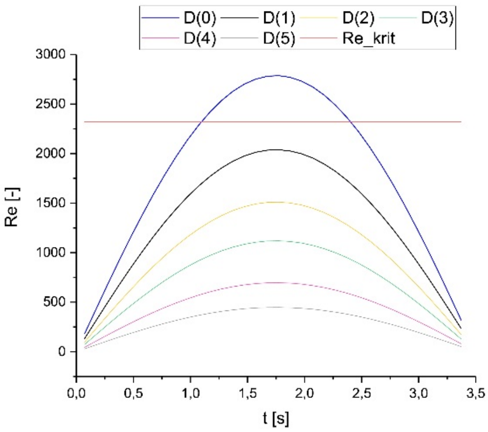

Figure 4: Reynolds number in the different diameters of the 5 generations.

So the turbulence model must represent both parts. For the simulations the shear stress transport model (SST) with gamma theta transition model. The time step was set as 0.025 s and the complete simulation time was 35 s. This are five breathing cycles. The first four are only to engage the flow situation in the lung.

As simulation program ANSYS CFX is used, because the industry of mechanical respirators use the same program to laying up there products. The calculation are made on a high performance computer cluster on 5 nodes, every node has 12 cores.

One important time step of the numerical simulation is when the volume flow cross the zero point. This is the moment when the inspiration change to expiration or the other way round. Because the volume flow is driven by the pressure gradient between two points, and the pressure velocity is sonic speed. So the pressure gradient changes to the opposite direction as the volume flow. In this moment the pressure works against the volume flow. This time step is hard to calculate for the cfd solver.

3 RESULTS

As results the pressure losses, volume flows and the discharge coefficient are evaluated. The whole time for the calculation of the transient simulation was 17 days for the whole 35 s.

Only at the first bifurcation, the volume flow is divided in two equal parts. In the deeper generations there is a difference up to 12 %, these is shown in the figures 5.

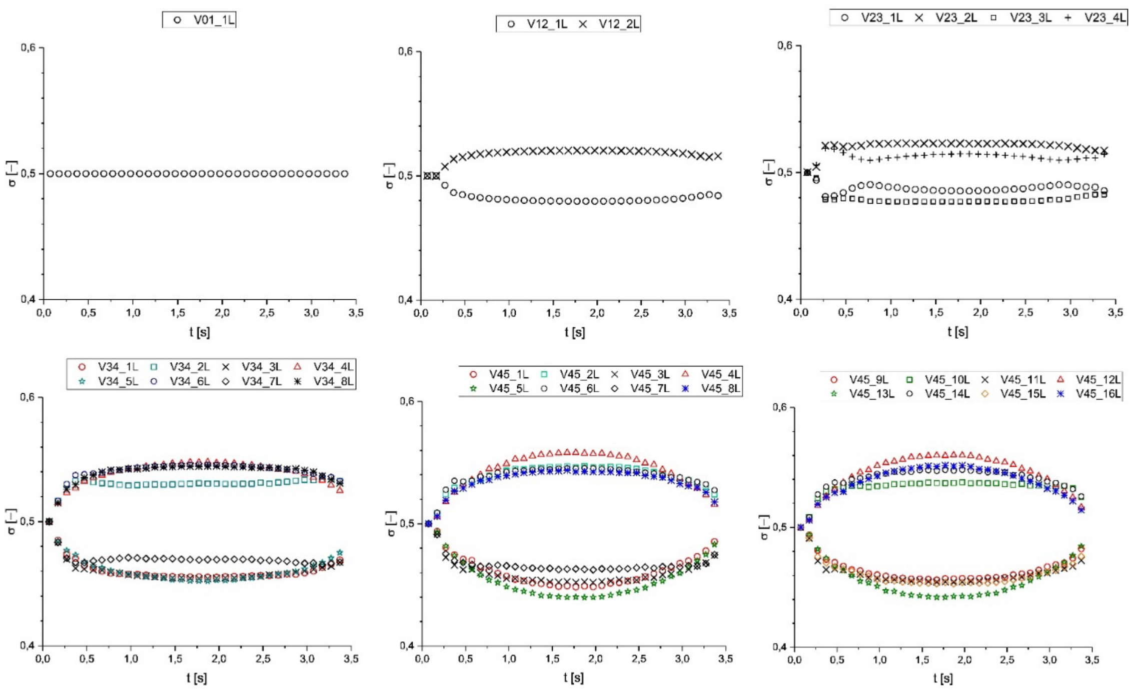

Figure 5: Subdivison of the volumeflow in the different generations of the human lung.

With the pressure at every in- and outlet of the bifurcation the discharge coefficients ζ can be calculated with the formula:

$$\zeta = \frac{2\Delta p_v}{\rho u^2} \tag{1}$$

With Δp_v as pressure loss at the bifurcation, the density of air ρ and the velocity in the bifurcation u.

The discharge coefficient varied in every generation and with every volume flow. With the results of the volume flow, the discharge coefficient is not constant. The results are shown in figure 6.

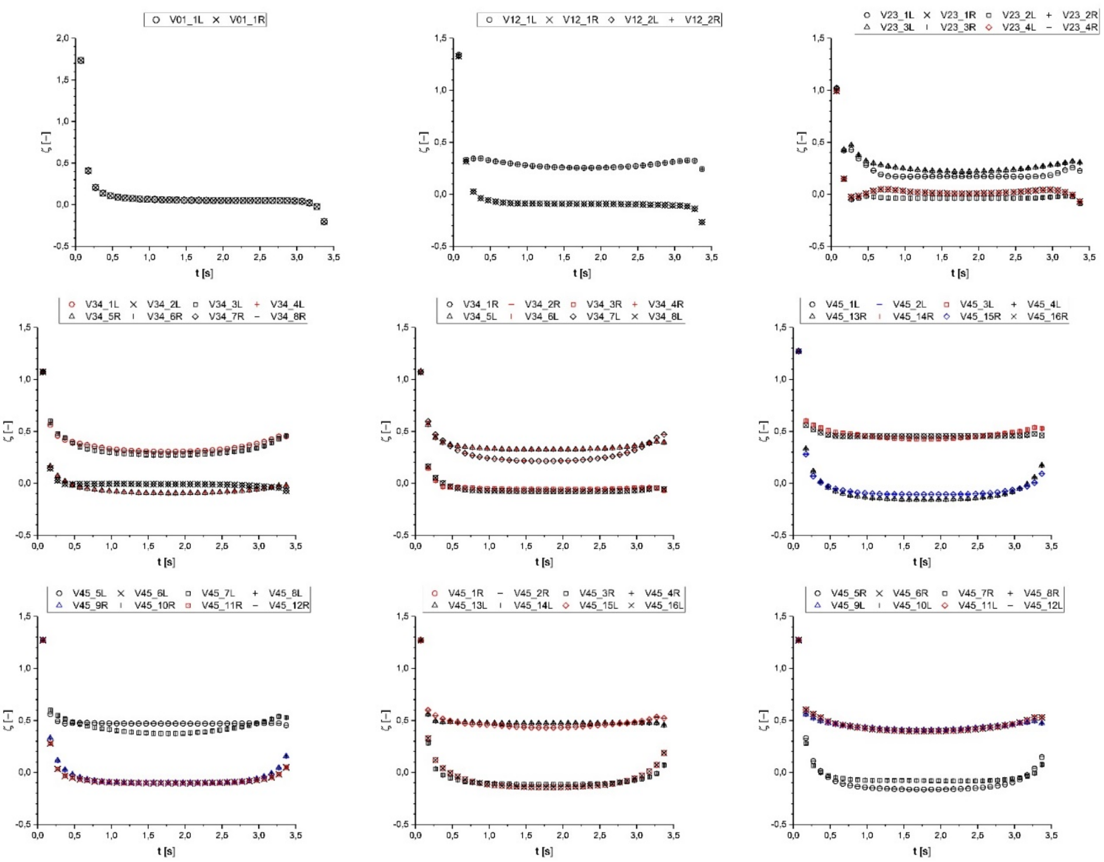

Figure 6: Discharge coefficient in the different lung generations.

4 DISCUSSION

For numerical simulations of mechanical respirators, it is important to know the discharge coefficients of the inhalation for every bifurcation in the lung. The discharge coefficient is not constant. The numerical simulations have shown, that the generation at the volume flow is important for this coefficient. In the case of volume flow, the flow conditions are the main influence on the distribution of the volume flow. Only on the first bifurcation between the zero and first is a uniform partition of the volume flow. At the end of the five generations there is no uniform partition of the volume flow shown. In case of this, the velocity in unequal in the branches and this is the main influence to the discharge coefficient. The numerical results shown, that the volume flow do not split in two symmetrical flows, only on the first bifurcation. In the upper region of the lung, the volume flow is symmetrical in both main parts of the lung. The first 16 branches and the branches 17 to 32 are mirror-inverted, that can be seen in figure 7. By reason of the flow conditions in the lung there are different volume flows at the end of the branches, depending on the turbulences inside the lung.

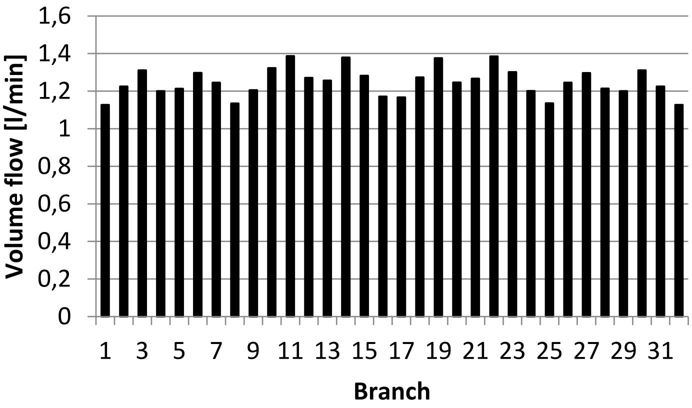

Figure 7: Volume flow in the branches of the 5th generation.

Known of this symmetrical distribution is the first step to simplify the lung model. The next step is to understand the distribution of the discharge coefficient to reduce the whole bronchial tree to only one branch. This is possible, if the pressure losses on every bifurcation can be preset for the reduced branch. With the formula 1, the discharge coefficient can be calculated for the stationary, full-developed pipe flow. In case of the lung, this formula must be adapted to the form:

$$\zeta = \frac{2\Delta p_v}{\rho u^2} \cdot \frac{1m^2}{A_1} \cdot \frac{A_1^2}{\dot{V}^2(t)} \tag{2}$$

In this case A is the area of the inlet in the bifurcation based on the diameter and $\dot{V}$ is the volume flow at the inlet of the bifurcation.

In figure 8 are the differences between the normed discharged coefficient and the numerical solution is shown. This analytic solution is through the velocity volume flow depending. The volume flow itself is time depending.

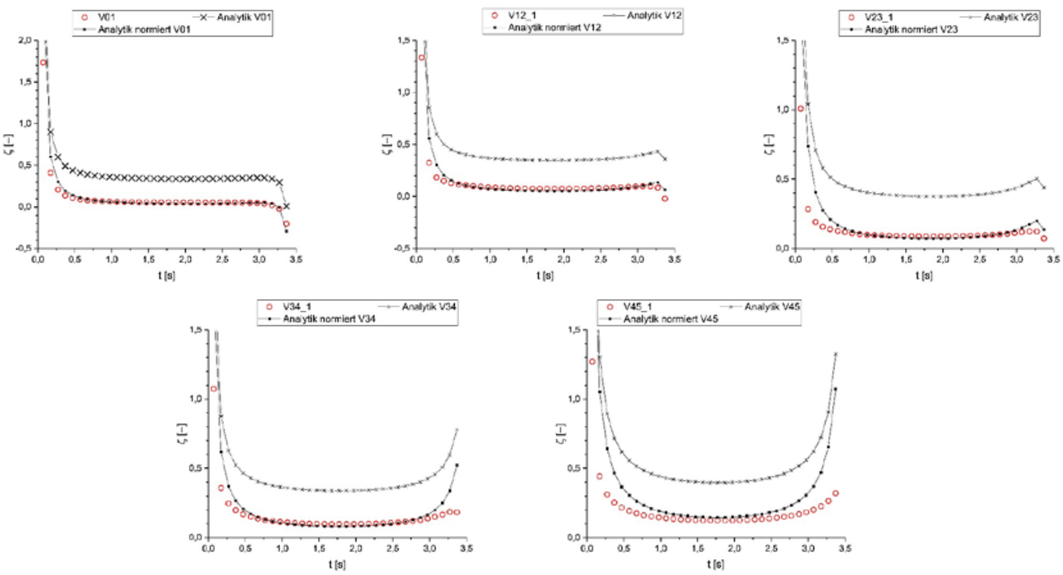

Figure 8: Analytic, adapted analytic and simulated discharge coefficient.

With the analytic solution it is possible to set at every bifurcation a time depending function for the pressure loss of the reduced branch. Then it is possible to reduce the whole bronchial tree to one branch. For this case it is important to understand which part of the lung will be investigate. Because the volume flow is different in every branch, in case of the particle tracking, a branch with high volume flow is more important than in a case of research the oxygen transport in the deeper lung generations. To understand the biomechanics of the lung branches with high volume flow as well as low volume flow is important, because the destroying of the lung tissue is possible in all parts of the lung.

With the reduced model, the simulation time can be reduced of 80 %. This means, the numerical solution of a five-generation lung model is possible in 3 days.

REFERENCES

[1] Raymondos, Konstantinos, et al.. Outcome of acute respirators distress syndriome in university and non-university hospitals in Germany. *Critical Care*, 21:122, 2017.

[2] W.H. Finley, A.R. Martin, Recent advances in predictive understanding of respiratory tract deposition. *Journal of Aerosol Medicine and Pulmonary Drug delivery*, 21(2), 189-206, 2008.

[3] J. K. Comer, C. Kleinstreuer, Z. Zhang. Flow structures and particle deposition patterns in double-bifurcation in airway models.*Journal of Fluid Mechanics*, 435, 25-54, 2001

[4] A. V. Kolanjiyil, C. Kleinstreuer. Computational analysis of aerosol-dynamics in a human wholelung airway model. *Journal of Aerosol Science,* 114, 301-306, 2017

[5] E.R. Weibel. Morphometry of the human lung. Springer-Verlag, Berlin, 1963

[6] B. Schöneberger, M. Semel, A. Delgado. Prädiktion von Drücken und Volumenströmen in einem 5-Generationen Lungenmodell auf Basis von Analytik und CFD-Simulationen. Deutsche Gesellschaft für Laser-Anemometrie – German Association for Laser Anemometry GALA e.V., 2017