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INVESTIGATION OF AIR FLOW IN IDEALIZED MODEL OF HUMAN RESPIRATORY TRACT

Jakub Elcner*, František Lízal*, Jan Jedelský*, Miroslav Jícha*

Validation of a numerical simulation using experimental data is a necessary prerequisite for verification of proper use of numerical method. This article deals with a comparison of velocities as predicted by an idealized model of human upper airways during stationary inspiration for three different breathing regimes. For the purpose of this study, a model which includes a realistic geometry of the mouth cavity and glottis coupled with an idealized geometry of the trachea and bronchial tree up to the fourth generation of branching was made. Calculations were compared with experimental data acquired by Phase-Doppler Particle Anemometry (P/DPA) on the identical geometry. Velocity data were compared at three points in the trachea. Specific air flow characteristics are documented and discussed based on results of the numerical simulation of the velocity field.

Keywords: *air flow, Human respiratory tract, CFD*

1. Introduction

Breathing is a complex phenomenon which represents an air flow through a system of channels called the respiratory tract. Inspiration is initiated by diaphragm contraction and resulting in the volume change of air paths in bronchial tree thus pressure difference. Due to this pressure difference, air flows through the mouth cavity, larynx, trachea and bronchial tree into the alveolar region, where blood is enriched with oxygen and gets rid of carbon dioxide. After that, deoxidised air is conducted during expiration, through the same path as it comes in and is exhaled to atmosphere. Along this path, the character of air flow is constantly affected by physiological and anatomical factors. Most of the factors which affect the flow characteristics can be observed in trachea. The first phenomenon, visible at the beginning of the trachea, was studied by Lin [1] and it is called the laryngeal jet. The laryngeal jet is caused by constriction of air path in the larynx, where velocity and turbulent kinetic energy grows and the shape of the velocity profile is strongly affected. Second phenomenon is caused by the first bifurcation at the end of the trachea. This bifurcation changes the shape of the velocity profile, during inspiration, in distance of several diameters of the trachea before the bifurcation itself. The phenomena mentioned above can be investigated by *In vivo*, *In vitro* or *In silico* methods. *In vivo* methods are not commonly used nowadays for ethical reasons and for its demanding measurements. *In vitro* methods, for example High Resolution Computer Tomography (HRCT), Laser Doppler Velocimetry (LDV) or Phase Doppler Particle Anemometry (P/DPA), are commonly used, but the complicated geometry of the bronchial tree makes the measurement in the highest generations of branching very difficult and it does not give us complete idea of flow in lungs very often. If

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* Ing. J. Elcner, Ing. F. Lízal, Ph.D., doc. Ing. J. Jedelský, Ph.D., prof. Ing. M. Jícha, CSc., Brno University of Technology, Faculty of Mechanical Engineering, Energy Institute, Technicka 2896/2, Brno

we want to know the character of flow fields in these parts, the best way is to employ a CFD calculation validated by *In vitro* measurement, which completes the whole image of air flow in the human respiratory tract.

2. Model

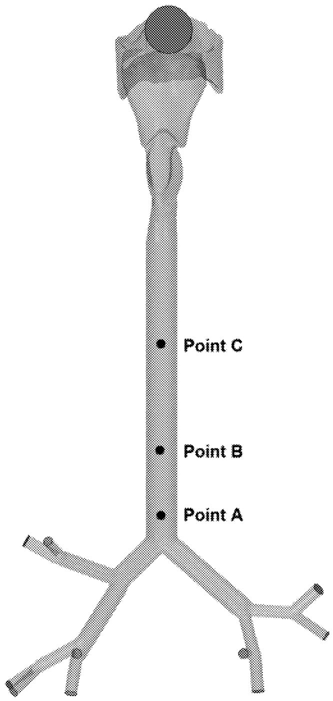

Fig.1: Idealized model

The main reason to design an idealized model (Fig. 1) lies in obtaining a more accurate image of flow characteristics inside the trachea and early generations of the bronchial tree. Experiments employing the model are performed using P/DPA technology which accuracy is dependent on the material used for the model fabrication as well as the shape of the geometry which surrounds the measuring point. Therefore, a combination of realistic and idealized geometry was chosen. The upper part of the model (mouth cavity and larynx) and particular bifurcations, where a realistic geometry is essential for authentic airflow development were rapid prototyped method which allowed us to create an arbitrary shape of the airway with respect to a characteristic wall curvature which can be found on real human airways. Rest of the model (trachea and branches of bronchial tree) was an assembly of thin-walled glass tubes. Using of thin-walled tubes with known inner diameter and length of tube allow us more specific assignment of measured points and it follows definition of these points in numerical model which leads to more accurate results. The length and diameter of particular branches were determined to resemble the realistic model formerly used during measurements at the Department of Thermodynamics and Environmental Engineering of Brno University of Technology [2]. Angles of branching were also determined using realistic model by measuring the angles which are held by daughter branches of a single bifurcation. More information about the model development as used for this study can be found in [3].

3. Experiment

The experiment was performed using P/DPA technique. 1D PDA (Dantec Dynamics) with Ar-Ion+ Laser ILT 5500A-00 (max. power 300 mW) was used for measurement of time-resolved flow velocity at multiple points of the model for three cyclic breathing patterns. The PDA focal length of both the transmitter and receiver was 310 mm; 1^{st} order light refraction and scattering angle 45° was used. Tracer particles of di-2-ethylhexyl Sebacate (DEHS) with 3μm in diameter were produced by a condensation monodisperse aerosol generator (CMAG TSI 3475); they were mixed with air in a static mixer and led to the model. The cyclic breathing patterns with sinusoidal course of flow are defined by tidal volume V_{t} and breathing period T (Tab. 1). The measurement was done at three locations within trachea, marked as A, B and C in Fig. 1. A velocity component in direction perpendicular to the cross-section of the trachea (expected direction of air flow in the model) was measured. The particle velocity is considered to match with air flow velocity due to low Stokes number (less than 0.02). More information about experimental setting can be found in [4].

Activity	Q (l/min)	V_t/T (l/s)	Re
Resting condition	15	0.5/4	1404
Deep breathing	30	1/4	2809
Light activity	60	1.5/3	5618

Tab.1: Investigated breathing regimes

4. Calculation

The calculations were performed with the use of CD-Adapco StarCCM+ software. Orientation of the model fits the lungs position in the human body. Tracheal axis was parallel to the Z axis of the global coordinate system and the input into the model lies in the plane formed by the X and Z axes. Computational mesh consists of polyhedral cells and contains approximately 2,000,000 cells. A prismatic layer was developed on the model wall for better description of the flow in the boundary layer. Unsteady RANS solver with k-omega turbulence model (SST Menter) was applied to the calculation. The velocity inlet condition prescribed by time-dependent equation (1), which simulates the breathing cycles corresponding to the regimes mentioned in Tab. 1, was imposed at the input to the model and the pressure outlet condition was set at the end of the branches.

$$Q(t) = \frac{V_\mathrm{t}\,\pi}{T}\,\sin\!\left(\frac{2\pi}{T}\,t\right) \qquad (\mathrm{m^3\,s^{-1}}) \tag{1}$$

where Q is flow rate, V_t is tidal volume in m^3.

The calculation was done in three points which corresponds to the experiments. Cross section areas, perpendicular to axis of trachea leading through the points A, B and C were also monitored.

5. Results and discussion

5.1. Breathing cycles

Comparison of measured and calculated values for three different points in the trachea and three different breathing regimes according to Tab. 1 is shown in Fig. 2–4. Graphs show the change of velocity during inspiration and expiration, at points A, B and C (see Fig. 1). These points cover most interesting parts of the trachea. Point A is situated above the carinal ridge of the first bifurcation, where turbulent behaviour due to mixing of air streams from daughter branches in first generation of the bronchial tree is expected. Point B can be found in the second third of the trachea, which is the area where we expect a developed profile of the air flow during inspiration and expiration. Point C is situated after constriction in glottis which is one of the mayor reasons for creation of laryngeal jet [1] and this point should monitored changes of velocity which this phenomenon affects.

Velocities measured by P/DPA system and velocities in Z axis of the local coordinate system as calculated by means of CFD are compared. Because of the reverse orientation of coordinate systems used during calculation and measurement, the data needs to be fitted together, this was performed in MATLAB. Measurement data were not statistically evaluated. A comparison made at the three points show slightly greater velocity of the calculated values than that of the measured data for all three breathing regimes. As you can see on Fig. 2–4, shape of breathing cycle has sinusoidal profile within each breathing cycle which is given by equation (1). Good agreement between experiment and calculation is seen for deep

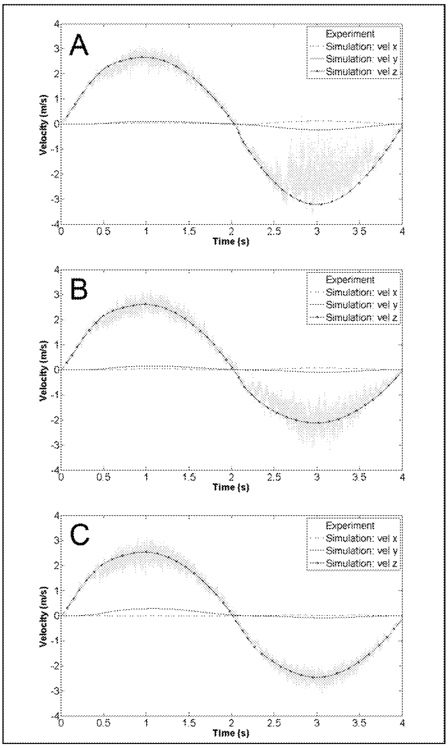

Fig.2: Comparison of measured and calculated velocities for resting condition

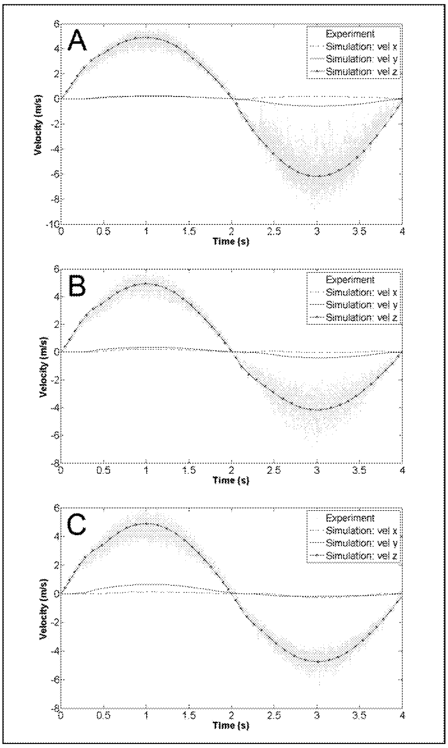

Fig.3: Comparison of measured and calculated velocities for deep breathing

breathing and light activity during inspiration at time approx. 0.25 s, where similar velocity change is apparent for both measured and calculated values. This change can be found at all measuring points which means that it has not influenced by surrounding geometry changes (laryngeal jet for point C or carina ridge of bifurcation for point A). The comparison also shows greater values of velocities in tangential direction during expiration which is caused by air mixing in first bifurcation and higher turbulence level in this area. Tangential velocities is further investigated in figures 6, 8 and 10 and discussed in following paragraph. This finding is significant for future research and it cannot be discovered without numerical simulation because P/DPA system gives us only one-dimensional information on velocity in the investigated area.

5.2. Scalar velocity fields

In order to good agreement of numerical simulation with data from experiments we can fulfil the information about flow in selected cross-sectional area of trachea. Fig. 5, 7 and 9 shows the scalar area of velocity in direction parallel to axis of trachea for points A, B and C. Positive value of velocity represent air, which flows from bronchial tree to the mouth cavity. Figures are divided in two columns. First column gives us information about flow fields in trachea during maximal inspiration (maximal inspiration during breathing cycle corresponds to point in time at 1 s for resting condition and deep breathing, and for 0.75 s for light activity). Figures show that the velocity profile during inspiration is strongly affected by shape of glottis which defines the direction of the axis of air flow. In our model,

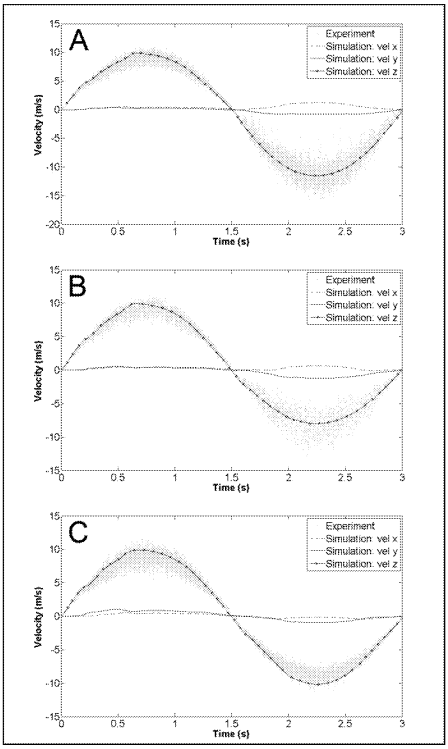

Fig.4: Comparison of measured and calculated velocities for light activity

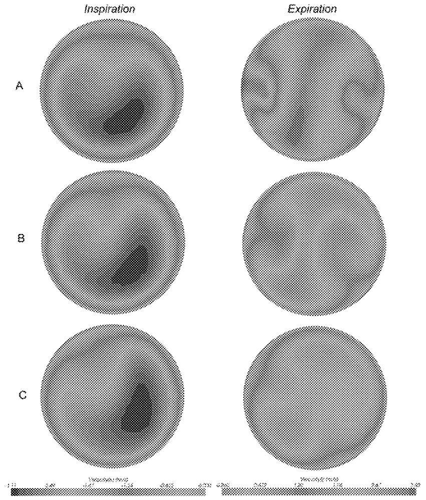

Fig.5: Scalar velocity fields during inspiration and expiration for resting condition

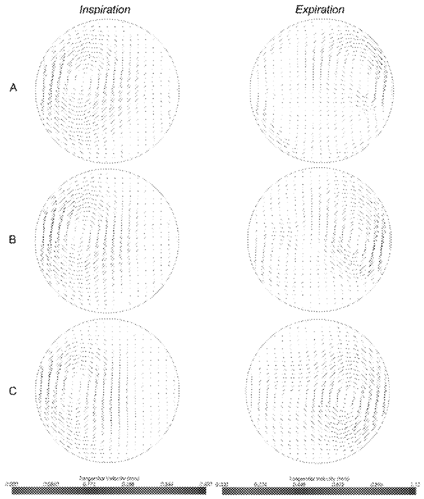

Fig.6: Tangential velocity vectors during inspiration and expiration for resting condition

the airflow tends to the left anterior area of the model and the air flow profile suit to this side. Second column represents the airflow distribution during expiration (point in time 3 s for resting condition and deep breathing and 2.25 s for light activity). Velocity profile during expiration is formed by first bifurcation, where air streams from left and right parts of the

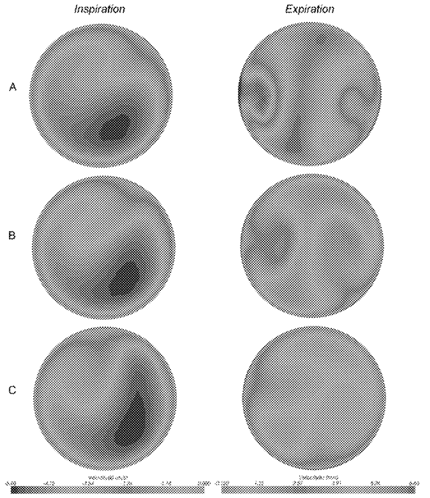

Fig.7: Scalar velocity fields during inspiration and expiration for deep breathing

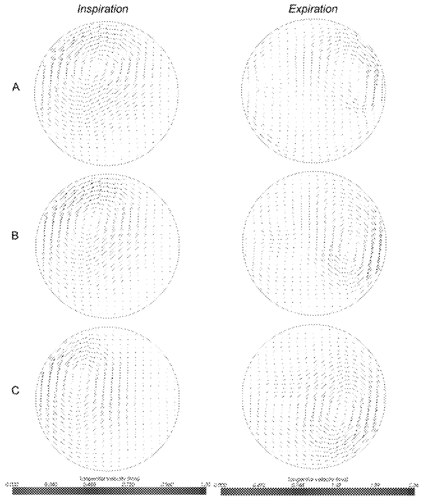

Fig.8: Tangential velocity vectors during inspiration and expiration for deep breathing

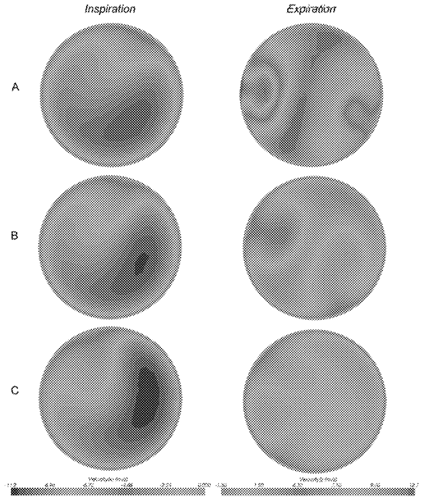

Fig.9: Scalar velocity fields during inspiration and expiration for light activity

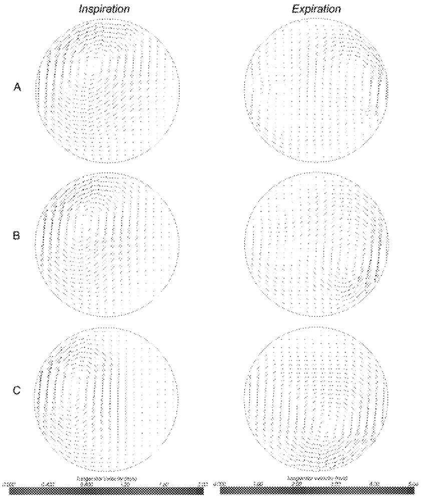

Fig.10: Tangential velocity vectors during inspiration and expiration for light activity

model are mixed. This air mixing leads to different shape of velocity profile, than the profile during inspiration.

5.3. Vector velocity fields

Figures 6, 8 and 10 shows velocity vectors in direction tangential to axis of trachea. Comparison of breathing regimes shows that the higher values of tangential velocities are achieved during expiration. Paths of velocity vectors during inspiration form the single vortex with core near the centre of trachea. Maximal values of tangential velocities can be found on the right side of the model where laryngeal jet raises the pressure resistance by impact of the air on the model wall. During the expiration we can observe production of two vortexes. First is visible on the left anterior and second on the right side of the model. These vortexes are formed by mixing of airstreams from branches of bronchial tree. However, values of tangential velocities achieved much lower values than velocities in normal direction which are parallel to main air stream.

6. Conclusion

Good agreement of numerical simulation and experimental data in the axis of the trachea allows us to fulfil the 1D velocity measurement in human upper airways, obtained by experiments provided by P/DPA, with 3D data of numerical calculations and give us more complex knowledge about flow characteristics in the investigated area. Results of the comparison were discussed formerly in the article. The knowledge gained from this article will help us with future deposition research, where proper setting of physics and model of turbulence is necessary for accurate calculations. Velocity fields in trachea will also help us to determine probably hotspot in first bifurcation of bronchial tree, where air from trachea is divided to two separate streams and then flows to left or right lung.

Acknowledgments

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