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Study of airflow in the trachea of idealized model of human tracheobronchial airways during breathing cycle

Jakub Elcner[1], Frantisek Lizal[1], Jan Jedelsky[1] and Miroslav Jicha[1]

[1]BUT Faculty of Mechanical Engineering, Energy Institute, 616 69 Technicka 2896/2, Brno, Czech Republic

Abstract. The article deals with a numerical simulation and its verification by experiments in the trachea of idealized geometry of tracheobronchial airways by using unsteady RANS method. The breathing cycle was simulated by sinusoidal function with period of 4 seconds and tidal volume of 0.5 litres of air, which corresponds to breathing during resting condition. Results were compared with experiments measured by laser-Doppler velocimeter in eight points of four cross sections in the trachea. Model consists of the mouth cavity, larynx and tracheobronchial tree down to fourth generation of branching.

1 Introduction

The breathing cycle can be divided into two separate sections, which differ in the flow direction through the airways. The first section, called inspiration, can be defined as a moment when the inhaled air flows into the respiratory tract through the mouth/nasal cavity, then continues to the larynx and over the trachea into the bronchial tree where is distributed to single alveolar sacs into the blood stream. After the exchange of vital gases air is exhaled out of the respiratory tract in reverse direction than during inhalation. Within these cycles, there are many phenomena which influence the character of air flow fields in given areas through the whole respiratory tract and affects many parameters in these areas such as pressure drop, velocity, volume flow etc. In 1963, Ewald R. Weibel published an idealized model of human lungs [1] where he defines a range of airways, in which inhaled air can be described by mechanisms of conduction. This range is between zeroth and seventeenth generation of branching and can be taken as a limitation of potential measurements or numerical simulations of airflow in the respiratory tract. We are able to achieve more accurate values in research of particle deposition, if we properly resolve the flow in the entire area and use the knowledge for prediction of dosing of drugs inhaled into the lung for the treatment of respiratory diseases. The aerosol deposition can be assessed using empirical equations. However, if we are interested in a detection of potential deposition hot spots, it is necessary to resolve flow patterns properly in the whole respiratory system [2].

Studies dealing with investigation of air flow during the respiration are usually focused on a narrow area of respiratory tract and resolve only a separate state of breathing (stationary inhalation/exhalation). The airflow in upper airways was investigated by Kelly et al. [3] or Chung and Kim [4]. Both teams performed an experimental study of flow patterns in a real geometry of nasal cavity using Particle Image Velocimetry (PIV) during inspiratory flow corresponding to 7.5 L/min. Computational Fluid Dynamics (CFD) simulations in oral cavity were performed by Wen et al. [5] and Zamankhan et al. [6] for flow rate of 15 L/min. More interesting part of respiratory tract, in terms of drug inhalation, is a region of mouth cavity and follow-up part of the upper respiratory tract (larynx, trachea, upper part of bronchial tree). Heenan et al. [7] performed a study on a simplified extrathoracic (oropharynx & trachea) geometry, which was focused on comparison of CFD using RANS method with PIV experiments for different air flows (15, 30 and 90 L/min). They concluded that the flow pattern does not differ significantly during the CFD calculation from the experiment when changing the Reynolds number. More recent study from Ball et al. [8], performed on identical geometry for lower flow rates (10, 15 and 30 L/min) found that k-omega models are not sufficiently accurate for investigating the airflow in larynx, but may be suitable for resolving airflow in the trachea and following areas. Fresconi et al. performed an experimental study [9] on a single bifurcation, where they studied phenomenona that arise from a combination of two air streams coming from daughter branches during stationary exhalation using PIV and Laser Induced Fluorescence (LIF). Calculations of a cyclic breathing driven by the sinusoidal function on a model of ten symmetrically branching generations were performed by Soni and Aliabadi [10]. A pressure boundary condition was prescribed on the outlet to the model and the inlets were specified by velocities. Many studies are based on geometries which begin on an interrupted trachea, before the entrance to the bronchial tree. However, in

[a] Corresponding author: yelcne00@stud.fme.vutbr.cz

reality, upper airways, where flow patterns are significantly altered, precede that region.. It has been proved by Lin et al. [11] that the laryngeal jet is one of the most significant phenomena that affect the airflow in respiratory tract.

The aforementioned studies, however, lack a detailed comparison of numerical simulations with experimental measurements. Correct and sufficient validation of these calculations allows us to obtain proper information required for further research of ventilation and aerosol deposition. This led us to conducting experiments compared with numerical calculations on two identical geometries (physical / digital) under identical boundary conditions.

2 Methodology

2.1 Model geometry

The model used for a comparison contains realistic geometry of mouth, throat and larynx which is connected to the idealized geometry of the tracheobronchial tree to the fourth generation of branching. The connection is located about 15 mm under the glottis and is done using a smooth transition. For the purposes of the experiment, the mouth cavity piece and single bifurcation pieces, were fabricated by a rapid prototyping method. Branches of bronchial tree were made from clear, thin-walled glass tubes to obtain better results during measurements using laser-Doppler velocimeter. A cylindrical tube of diameter 20 mm is connected to the model in the area of mouth cavity. This experimental model is identical to the geometry used for the numerical simulation. More information and description of the fabrication process of this model can be found in [12]. Model scheme with described boundary conditions and marked cross sections is on **figure 1**.

2.2 Numerical setup

2.2.1 Solver and mesh

Numerical simulations were performed using CD-Adapco StarCCM+ solver in version 8.02. A polyhedral mesh with a prismatic layer for better description of flow in the near wall region was made on the model geometry. The average size of a single volume is 0.7 mm in diameter and final mesh size contains 2.6 million of cells. The resolution of the mesh was chosen to achieve sufficient number of cells in the branch diameter and also to guarantee the value of low wall Y+ around 1 on a whole surface of the model.

2.2.2 Physics

One respiratory cycle was solved for the purposes of this article (inhalation and exhalation). The calculation was solved using unsteady RANS. The respiratory tract represents a net of continuously branching system with decreasing diameter of branches and wide range of Reynolds numbers ($300 < Re_{local} < 10^4$ according to [13]). For a proper modeling of laminar, turbulent and transient behavior of the velocity field a Menter hybrid SST model [14] with low Reynolds numbers modification was selected. This model showed good results in comparison with experimental and CFD data of a pulsating stream which was performed by Tan et al. [15].

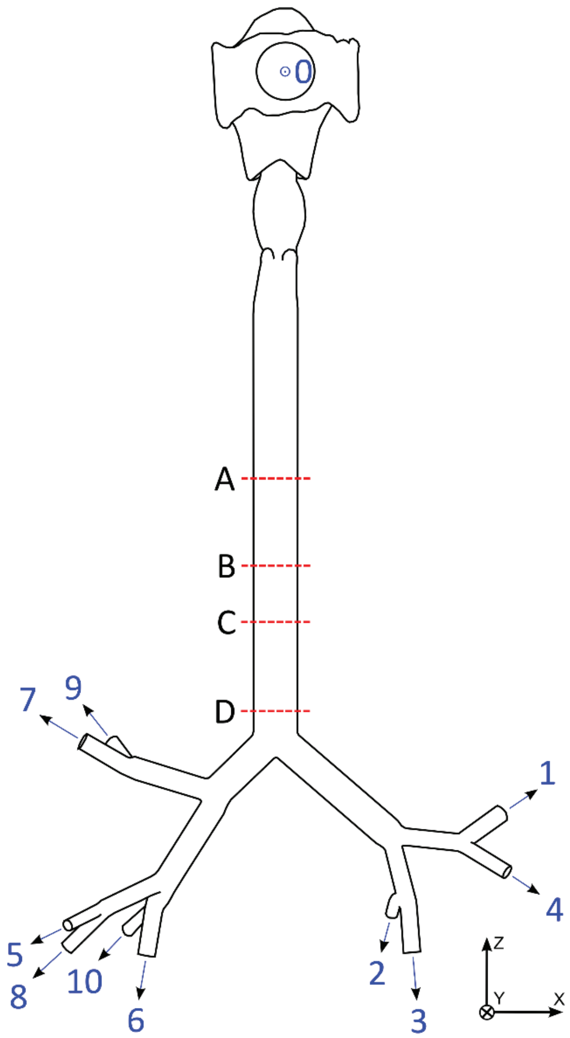

Figure 1. Model scheme

Different values of boundary conditions were set during inhalation and exhalation. This corresponds to the experiment, where the air flow rate was measured at the terminal branches of the bronchial tree of the model. The same flow rate was then set on the corresponding boundary in the numerical simulation. The list of flow rate values measured during inhalation and exhalation during experiment can be found in **table 1**. A zero pressure boundary condition was set on the inlet to the model (0). Velocity inlet boundary conditions with prescribed field function were set for simulation of cyclic behaviour of breathing on the terminal branches of boundary tree (1-10). This function can be defined using **equation 1**.

$$Q(t) = \frac{V_t \cdot \pi}{T} \cdot \sin\left(\frac{2\pi}{T} \cdot t\right). \qquad (1)$$

Where Q is flow rate ($m^3.s^{-1}$), V_t is tidal volume (m^3), T period (s) and t is time (s). One respiratory cycle takes 4 s and the time step for calculation was set to 0.001 s with respect to Courant number.

Table 1. Flow rates measured during experiment.

Boundary condition (-)	Branch diameter (mm)	Inspiratory volume (L)	Expiratory volume (L)
0	*20*	*0.5*	*0.5*
1	*6*	*0.06*	*0.05*
2	*4.8*	*0.04*	*0.08*
3	*6*	*0.06*	*0.04*
4	*4.8*	*0.04*	*0.02*
5	*4.8*	*0.04*	*0.04*
6	*6*	*0.06*	*0.06*
7	*6*	*0.07*	*0.03*
8	*4*	*0.04*	*0.04*
9	*4.8*	*0.05*	*0.08*
10	*4.8*	*0.04*	*0.05*

2.3 Experimental setup

Two-component laser-Doppler velocimeter (LDV) system Dantec Dynamics Flow explorer 9065x0341 (Dantec Dynamics A/S, Skovlunde, Denmark) was used for time-resolved point-wise measurement of velocity of individual particles flowing inside the airway model. Only one velocity component, corresponding to the local axial flow direction, was recorded. A built in, diode-pumped solid-state, laser generated a beam with 660 nm wave length. The beam was split into two parallel beams where the frequency of one of them was shifted by 80 MHz. A converging transmitting/receiving lens with 500 mm focal length was used to form an ellipsoidal probe volume with size app. 0.1×0.1×1 mm at the beam crossing point. The LDV worked in the back-scatter mode and used the Dantec BSA P80 flow and particle processor for signal processing. Following settings were used: photomultiplier sensitivity 700–1000 mW, signal gain 16–22 dB, record length: auto-adaptive with minimum and maximum record length set to 32 and 256 respectively, burst detector SNR level 1–4 dB. Particular settings of these values depended on the signal strength, position of the measured point and flow rates and were set for each regime and position individually. The cyclic flows were recorded always for two full cycles with velocity span 6–38 m/s and velocity center at 0 m/s (for all measurements). The stationary flows were recorded with velocity span 1,59–19,09 m/s. A minimum of 10,000 samples was acquired at each measured point.

3 Results

The comparison of measured and calculated values for four selected cross sections in the trachea is presented in **figures 2-5**. Measured values represent several consecutive cycles, which overlap in order to achieve sufficient data density. Results on the whole respiratory cycle gives us good agreement in both, inspiratory and expiratory parts of cycle. It is evident that the calculation (line) fits the measured values (point). Two time instants located at the peaks of the cycle (1000 ms and 3000 ms) were selected for the description of a character of inhaled and exhaled air.

3.1 Inhalation

During the inspiratory part of the breathing cycle, air flows through the mouth cavity, larynx and enters the trachea. The passage of air through the upper airways forms subject specific laryngeal jet. The volume of air flowing through the trachea is equal to the total amount of air prescribed on terminal boundary conditions. Maximal velocity in the trachea during inhalation is about 2.5 m/s and occurs in time about 1000 ms where peak of respiratory cycle can be found. In time approximately 400 ms, local maximum of velocity can be seen on all cross sections A-D, which is visible on both measured and calculated data. This fluctuation is related to the transition from laminar to turbulent flow. This phenomenon is also apparent in time about 2200 ms during exhalation which leads us to conclusion that the transition between laminar and turbulent phases occurs sooner during exhalation than during inhalation which may be caused by the collision of air flows from daughter branches and related mixing of air which supports turbulent behaviour. On scalar fields is also apparent that velocity profile in the peak of inspiration (at 1000 ms) is similar to parabolic profile but its maximum is shifted to front left quadrant of trachea cross section A and B. In cross section C and D is apparent movement of this maximum to the front right quadrant due to the highest air flow on the right side of the bronchial tree. We cannot see developed laminar or turbulent profile in whole length of the trachea as a result of its unsufficient length and unsteady behaviour of airflow.

3.2 Exhalation

Data comparison during exhalation shows slightly worse results than in inhalation. Cross sections A and B seems to have more flat profile than during inhalation and show us two local maxima located in the right rear and left front quadrants of trachea. Cross section A, which is located right above first bifurcation of bronchial tree, is the most interesting. Unusual shape of velocity profile which is caused by the collision of air flows from left and right part of the bronchial tree can be seen from the scalar field. Local minimums shows us negative values of axial velocities which is caused by collision of air flows and subsequent formation of vortices.

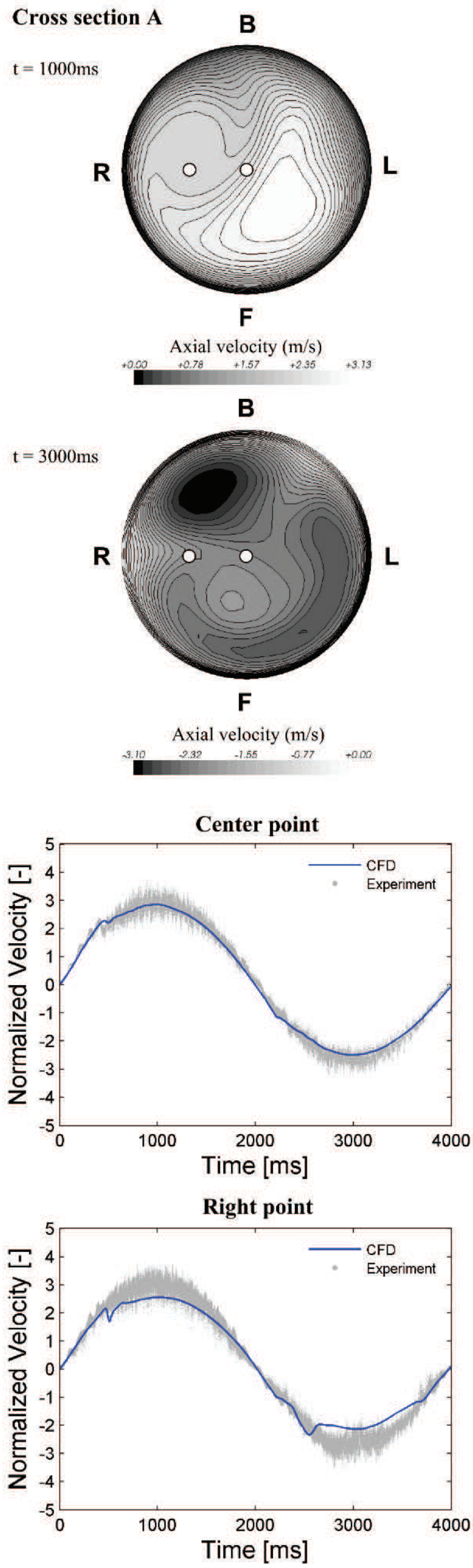

Figure 2. Cross section A analysis

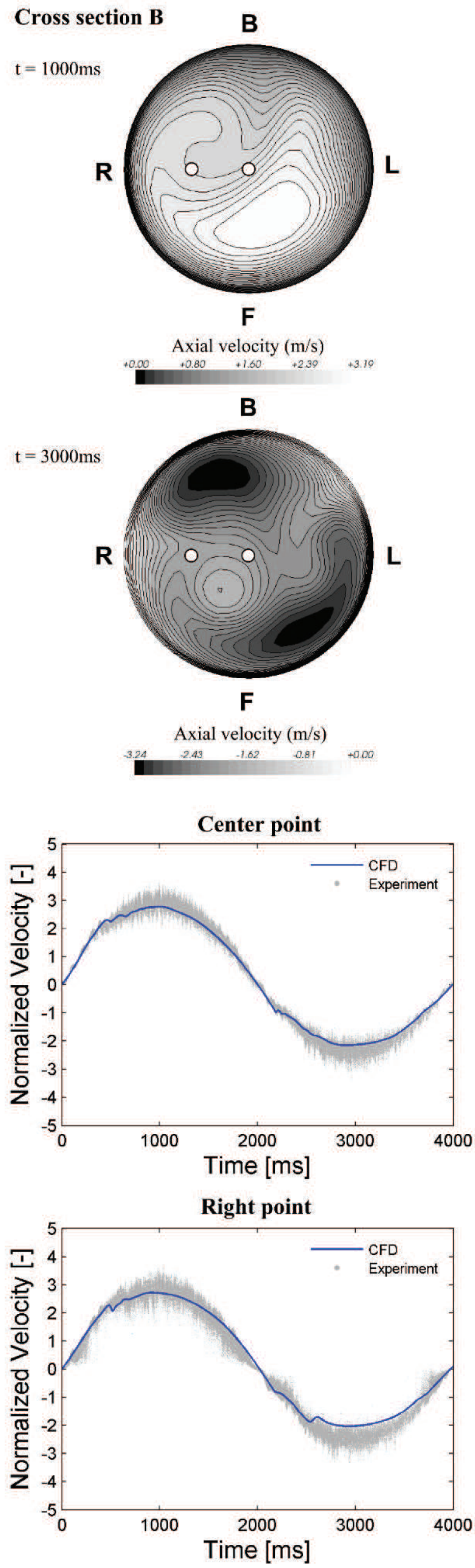

Figure 3. Cross section B analysis

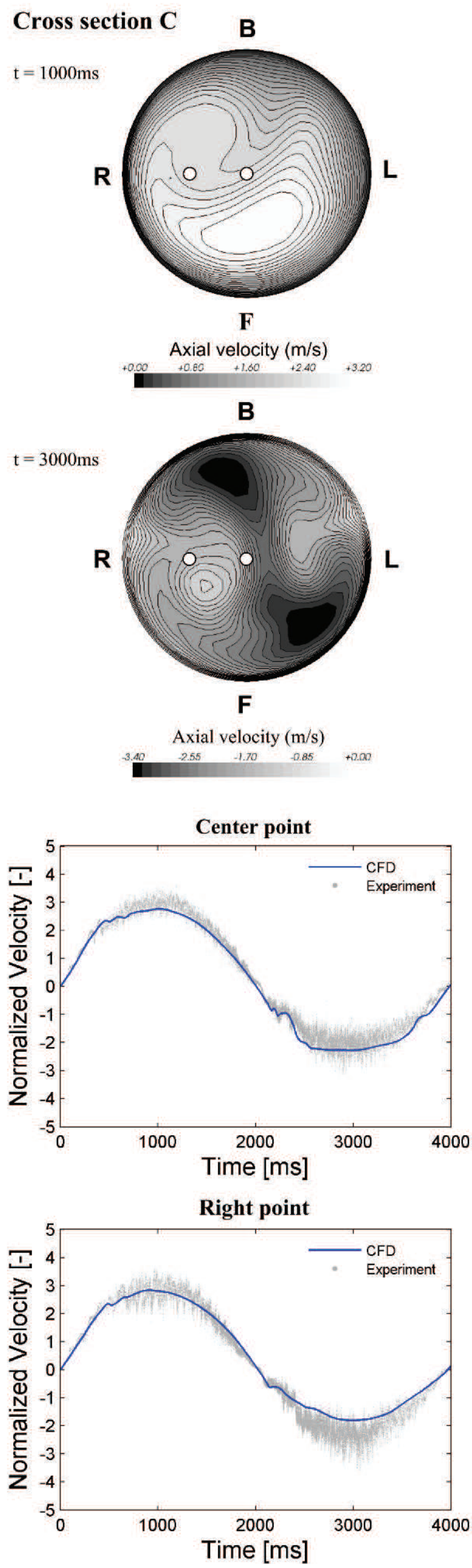

Figure 4. Cross section C analysis

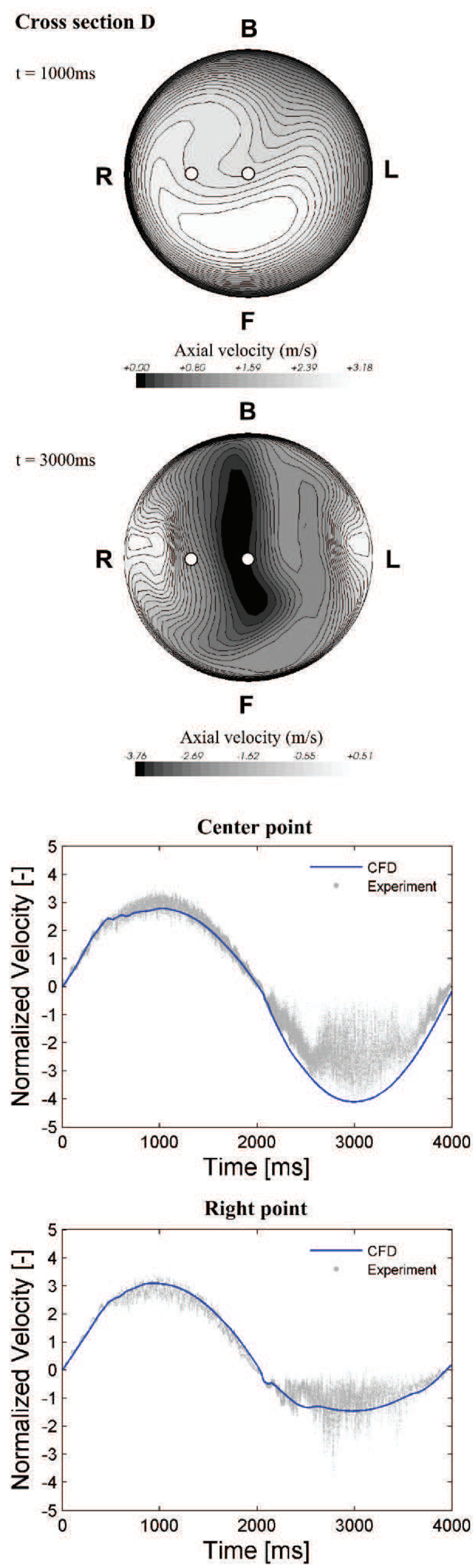

Figure 5. Cross section D analysis

4 Conclusion

The comparison of measurements performed using LDV and CFD calculations using unsteady RANS method showed us good agreement and it can be seen that RANS method, although by its nature unable to describe small vortices, can be used for prediction of the airflow in similar cases. Thanks to the validation with experiments we can try to perform calculations containing deposition of aerosols and expect similar quality of the results.

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