Basic flow regimes in a human airway model measured by stereo scanning TR-PIV

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Abstract The detailed analysis of the flow in the human lung is of high relevance for the optimization of mechanical ventilation. In this paper, the spatial and temporal development of the flow field in a realistic shaped model of the human lung is investigated for different pulsatile flow regimes. The model is based on the physiological data of a middle-aged and healthy male. The lung model consists of the tracheobronchial tree up to the 4 – 7th generation and is made from transparent silicone. The full optical access and the use of a refractive index matched test fluid offers the possibility to use Stereo-Scanning PIV to perform time-resolved three-dimensional measurements for all velocity components that cover the complete bronchial volume. The results of the analysis of the flow field will help to optimize ventilation strategies for artificial respiration. The investigated regions cover the trachea, the primary bronchi, and the right secondary bronchi, whereas all measurements are performed for a complete sinusoidal ventilation cycle. Three parameter combinations of the ventilation, composed of Reynolds and Womersley numbers, defined as viscous (Re = 10; \(\alpha = 1.5\)), unsteady (Re = 40; \(\alpha = 5\)), and convective (Re = 150; \(\alpha = 1.5\)) were chosen with respect to the dominating fluid mechanical effects. The flow conditions are characterized by pressure measurements. In addition, steady measurements using a multi-plane PIV technique were analyzed for laminar (Re = 150) and turbulent (Re = 2650) flow for inspiration and expiration.

The following conclusions can be drawn from the results. The critical Reynolds number depends on the Womersley number for the onset of turbulence. The velocity profiles for the steady and the pulsatile convective flow domain are similar, defining the flow at \(\alpha = 1.5\) to be quasi-steady. The maximum velocity is shifted to the lower wall for inspiration and towards the upper walls of the bronchi for expiration. For inspiratory flow, a U-shaped high-speed velocity profile develops only inside the left primary bronchus, whereas both primary bronchi contain one vortex-pair. During expiration, one vortex-pair from each main bronchus merges to a two-vortex pair system inside the trachea. In addition, the vorticity inside the left primary bronchus is more than two times higher than in the right one. An undersupply for the right upper lobe is noticed for low ventilatory frequencies, whereas the unsteady flow domain leads to a more homogeneous ventilation. Furthermore, temporal analysis shows a variable phase lag for unsteady flow that results in two different internal frequencies for inspiration and expiration. Additionally, a premature outflow of the upper right lung lobe is observable.

1. Introduction

The focus of the current work lies on patients who suffer from the Acute Lung Injury (ALI) and the Acute Respiratory Distress Syndrome (ARDS). The early respiratory failure is characterized by a massive inflammation that results in a severe pulmonary dysfunction with an edema of the lung (Müller, 2000). Studies number the incidence of ARDS between 1.5 and 75 per 100,000 per year and a mortality rate of approximately 40 to 60 % (Dreyfuss and Saumon, 1998; Ware and Matthay, 2000). Therapies of ALI and ARDS consist in ventilation of the injured lung. Recent studies show that the mortality rate decreases significantly when the patients are ventilated with a lower tidal volume, e.g., 6 to 8 ml per kilogram body weight, have a significant reduced mortality (The ARDS Network, 2000). This shows the relevance of understanding the complex flow field in the human airways.

Numerous investigations of generic lung models have been performed during the last decade. Martonen et al. (2001a, 2001b, 2001c) reported a secondary flow pattern in form of vortices that occur in daughter airways downstream of the carina. Experimental and numerical investigations conducted at Re = 200 – 2,000 show that the shape and intensity of the secondary flow structures depend on the inlet conditions, whereas the measurements were performed using hot-wire anemometry. Grotberg (2001) found the secondary flow amplitude to decrease for Re = 200 – 1,200 if the Womersley number and the tidal
volume are reduced. Furthermore, Ramuzat and Riethmueller (2002) investigated via standard PIV the oscillatory flow within a three-dimensional model for three successive generations of lung bifurcations. Their results show a quasi-steadiness of the flow when the frequency of the oscillation decreases. Moreover, flow structures can repeat themselves from one bifurcation to the other under certain conditions, despite the presence of strong secondary motions. Despite numerous experimental and numerical investigations of the flow field inside the lung, there is still a considerable amount of uncertainty concerning the complex flow field inside the human lung due to the intricacy of its geometry. Therefore, results of the flow distribution for realistic models are rare. Most investigations are based on generic geometries.

Große et al. (2007) analyzed the two-dimensional flow field inside the first bifurcation for steady and unsteady flow at a range of Womersley numbers $3.3 \leq \alpha \leq 5.8$ for three Reynolds numbers $Re = 1050, 1400,$ and $2100$ using standard PIV. They found that the flow behavior is independent of the Reynolds number for steady inspiratory flow. For expiration, the unsteady and steady flows are rather similar. Eitel et al. (2010) compared 2D/2C PIV results with numerical results based on the Lattice Boltzmann method for a maximum Reynolds number of $Re = 1600$ and two Womersley numbers of $\alpha = 3.64$ and $\alpha = 5.15$. The authors showed that the inspiratory flow is highly influenced by secondary flow structures. In contrast, at expiration the streamwise flow distribution is more homogeneous with a significant higher level of vorticity. Soodt et al. (2011) performed time-resolved 3D-3C measurements in a realistic three-generation lung model for the right primary bronchus and the subsequent secondary bronchi using the Stereo-Scanning PIV technique. The flow field and the development of the vortical structures was analyzed in detail for two physiological Womersley numbers of $\alpha = 3.4$ and $\alpha = 4.2$ during the transition from inspiration to expiration. Based on the broadening of the frequency-dependent inspiratory velocity profiles and the fact that an increased mass flux into the right superior bronchus is evident, it is concluded that higher frequencies homogenize the inhaled air and could therefore be beneficial for the oxygenation of patients during mechanical ventilation.

All recent investigations mentioned above were carried out using either a generic geometry, a complex geometry without a three-dimensional PIV technique, or a three-dimensional PIV in a low generation lung model, the latter only covering parts of a full ventilation cycle. Considering the existing knowledge on the scanning PIV technique and the necessity of the application on such complex flows, this technique will be used to investigate the flow field in a realistic model up to the 7th generation of a healthy human lung.

This paper is organized as follows. First, the experimental method and the equipment are described. Second, the results of the stereo scanning PIV measurements are presented and discussed. Finally, these findings are summarized and conclusions are drawn.

2. Experimental setup

The current study enhances the investigations of Soodt et al. (2011). The PIV experiments were implemented using a realistic, three-dimensional lung model that extends to the 4 – 7th bifurcation generation of the bronchial system. The manufacturing process of the model encompasses radiographical scanning of a human proband, digital geometry reconstruction, segmentation, surface generation and iterative surface smoothing (German Cancer Research Center, Heidelberg, Germany), geometry post-processing, rapid prototyping (Institute of Applied Medical Engineering (CVE), Aachen, Germany), surface coating, silicone casting, and kernel removal. The extensions of the model’s exits to the outer dimensions of the cast have been incorporated in the digital data set during geometry post-processing to reduce manual channel drilling to a minimum. The material of the hollow lung model consists of transparent silicone to guarantee perfect optical access.

The model can be embedded either in a bidirectional circuit for steady inspiration and expiration or in an open-circuit flow facility for oscillating flow. Measurements at oscillating flow, which were designed to simulate a sinusoidal breathing cycle, are technically realized by a linear actuating piston. Figure 1 shows two photographs of the Rapid-Prototype kernel (a) and the hollow silicone lung model (b) as well as a schematic illustration of the experimental setup for steady and oscillatory flow (c).
The inflow conditions are critical since the inflow velocity profiles have a strong influence on the flow field in the first generations of the model. Therefore, an extended trachea with a length of \( L = 500 \text{ mm} \) is used to ensure a smooth transition from the circular cross-section of the pipe to the tracheal cross-section. The number of bifurcating generations is set to a maximum of seven to allow realistic inflow conditions for the expiratory phase. Since there is hardly any disturbance of the hydrostatic pressure distribution in the tank, the outlet pressure in the experiment can be assumed nearly constant for all branches exiting in the tank.

A water/glycerine mixture with a refractive index \( (n = 1.41) \) identical to that of the silicone block of the hollow model, both experimentally measured by an Abbe refractometer, was used. The refractive-index matched fluid ensures optical access without distortion. The dynamic viscosity of the mixture with 64.6 mass percent glycerine is \( \eta = 12.0 \cdot 10^{-3} \text{ Pa·s} \) and the density \( \rho = 1.167 \cdot 10^3 \text{ kg/m}^3 \) for a temperature \( T = 25.0 \text{ °C} \) that was held constant by means of a heating unit with integrated control unit inside the circuit during all experiments. The parameters were calculated using the model of Cheng (2008).

In these investigations, small air bubbles are used for flow visualization, because solid tracer particles deposit on the inner surfaces of the airways inside the model and decrease the image quality considerably. The air bubbles are created by means of a micro-bubble generator (Ishikawa, 2009) that is placed inside a second circuit and connected to the bottom of the measurement tank. A feeding pump supplied by pressurized air generates the velocity through a Venturi nozzle. This Venturi nozzle creates shear forces that cause the bubbles to burst into micro-bubbles. The median size of the generated air bubbles is about \( d_B = 11.9 \mu \text{m} \). The velocity of the buoyancy is \( v_B = 0.01 \text{ mm/s} \) and the response time \( t_B = 0.002 \text{ ns} \). Both values are sufficiently small to guarantee satisfactory particle behavior to measure the flow field.

The PIV measurements were performed using a continuous-wave \( \text{Ar}^+ \) laser (Innova 90C-A6, Coherent, Dieburg, Germany) operating in multi-line mode. The point-shaped laser beam is spanned to a light-sheet using a combination of three lenses (I: convex lens, II: cylindrical lens, III: convex lens), see Figure 2, and directed on a rotating polygon mirror (Kugler GmbH, Salem, Germany), which has a diameter of 55 mm and 10 facets. The light is reflected off the facets of the polygon mirror towards the measurement region of the lung model. Between the polygon mirror and lung model, a cylindrical lens emends the fanned light plane into parallel planes.

The polygon mirror is driven by a high-precision DC motor that is equipped with a gear, which reduces the rotating speed, leading to a more stable concentricity. Nevertheless, due to the clearance inside the gear and velocity fluctuations, a trigger signal for each facet is essential. A trigger unit consisting of a
lens, a photo sensor, and a comparator circuit initializes the cameras for the acquisition. Two SA-3 (Photron, West Wycombe, United Kingdom) high-speed cameras are used for the recording of the flow. Both cameras are positioned at an angle of $50^\circ$ to the measured volume and fulfill the Scheimpflug-criterion at an angle of $15^\circ$. Each camera receives the high-speed trigger signal from the sequential facet (Scanner) and records a set of frames after a general signal of the pump is once set (Figure 2 b). This pump signal is provided by the linear actuator, which can be set to any phase angle of the breathing cycle. Therefore, the double-image pulse distance ($t_{\text{facet}}$) is the time the mirror needs to turn to the sequential facet. A delay time ($t_{\text{delay}}$) between the scanner signal and the first recorded image adjusts the positions of the planes inside the measurement volume. A second pump signal stops the acquisition.

Figure 2 a) Schematic sketch (top view) of the PIV setup. b) State diagram of trigger signals and frame acquisition.

The first step in post-processing is the calibration to reconstruct the stereoscopic three-dimensional vector field. A translucent calibration grid was positioned in the same plane (x-y plane) as the particle images were captured; however, it has been displaced in z-direction. The recorded picture sequences of the breathing cycle were sorted depending on the position of the light-sheet and were then analyzed with VidPIV 4.7 (Intelligent Laser Applications GmbH, Jülich, Germany). The final window size was 24 x 24 px$^2$ with an overlap of 50%. This represented a spatial resolution of 0.408 x 0.408 mm$^2$. The last step merges the 2D vectors of each camera into a 3D vector field that was finally exported and further post-processed by means of MATLAB (MathWorks, Natick, USA). In the first step, average velocities were calculated over all recorded cycles per position and parameter. In the second step, averaging over $N$ phase angles was performed. The amount depends on the discretion size of the parameter but at least a temporal resolution of $\varphi < 6^\circ$ is achieved. Moreover, the absolute velocities $u_{\text{abs}}$ (by means of the arithmetic mean for three velocity components $u,v,w$, and the normalized velocities $\bar{u} = u_{\text{abs}}/\bar{u}$, the latter referred to the peak bulk velocity $\bar{u} = Re \cdot \eta/(\rho \cdot d_{\text{hyd}})$ at maximum inspiration) were calculated for a better comparability between the different parameters. In addition, the vorticity $\omega$ and the relative helicity $H_r = \bar{v} \cdot \bar{a}/(\|\bar{v}\| \cdot \|\bar{a}\|)$ were determined. While the vorticity is direction-dependent, the helicity can be used to analyze vortices in complex geometries because the scalar product is calculated for the helicity, i.e., the flow direction can change due to bifurcations.
3. Results and discussion

Steady and time-dependent pulsatile flow distributions have been investigated for a set of parameters, represented in Figure 3. Steady measurements were performed for two Reynolds numbers Re = 150 and Re = 2650, for inspiration and expiration using 2D time-resolved PIV technique. Thirteen planes with a vertical spacing of 1 mm were recorded in test-section TS*. Pulsatile measurements with sinusoidal ventilation cycle were investigated for three Reynolds numbers, i.e., Re = 10 and Re = 150 at a Womersley number α = 1.5 and Re = 40 at α = 5. Stereo-Scanning PIV was used and each parameter was recorded for three different test sections, i.e., TS 0, TS 1 and TS 2. For the pulsatile measurements, between five and eight planes were recorded, depending on the geometry of the lung and the parameters of the flow. The spatial resolution in z-direction is 1.4 mm and the angular shift has values between ϕ = 0.1° and ϕ = 1.9°. The test-sections capture the end of the trachea including the bifurcation into the right and left primary bronchi (TS 0), the right primary bronchus with the bifurcation to the three secondary bronchi (TS 1) and the bottom right secondary bronchus (TS 2).

![Figure 3](image-url)  
**Figure 3** The locations of the test-sections inside the lung in conjunction with the parameters for the 2D time-resolved PIV measurements at steady inspiration and expiration (TS*) and for pulsatile measurements (TS 1) are shown. The arrow indicates the scan direction for the pulsatile measurements.

The calculation of the Reynolds number is based on the hydraulic diameter of the trachea \( d_{\text{hyd}} = 15.7 \) mm and on the maximum bulk velocity \( \bar{u} \), the latter of which is represented for each parameter in the tables (Figure 3).

The parameters for the Reynolds and Womersley numbers were chosen with regard to Jan et al. (1989) and presented graphically in Figure 4. The abscissae shows the squared Womersley number \( W_0^2 = \alpha^2 \) and the ordinate the dimensionless stroke length \( \frac{L}{a} \). The boundaries of the different zones are indicated with solid lines and separate the unsteady, viscous, and convective flow domains from each other. In addition, boundaries found by previous pressure measurements show the transition from laminar over transitional (dashed lines) to turbulent (dash-dotted) flow for low and high Womersley numbers. The transition from convective to transitional flow is given at a critical Reynolds number of approximately \( Re_{\text{crit}} = 400 \) and \( Re_{\text{crit}} = 500 \) for low and high Womersley numbers, respectively. It can be seen, that each of the three chosen parameters represents one specific flow domain, i.e., the flow domain unsteady (‘U’) is represented by the parameters \( Re = 40 \) and \( \alpha = 5 \), the flow domain viscous (‘V’) by \( Re = 10 \) and \( \alpha = 1.5 \), and the flow domain convective (‘C’) by \( Re = 150 \) and \( \alpha = 1.5 \). The change from transitional to
turbulent flow was found at critical Reynolds numbers of approximately $Re_{crit} = 2000$ and $Re_{crit} = 4500$ for low and high Womersley numbers. The turbulent flow domain was not determined since the maximum camera frame speed determines the limit of the Scanning-PIV. The approximate region of physiological breathing is indicated by the small-dotted ellipse. This region was found by using the similarity theory for the Reynolds and Womersley number between water/glycerine mixture and air.

Figure 4 Flow regimes based on the concept by Jan et al. (1989). Measured flow parameters representing the flow domains viscous (‘V’), unsteady (‘U’) and convective (‘C’).

3.1 Steady flow

For the steady measurements, the 2D time-resolved multi-plane PIV technique was used. The focus lies on the velocity distribution and the helicity.

Figure 5 shows the high-speed velocity distributions in three-dimensional view for values of $u^* = 1$ by contours for $Re = 150$ (a, c) and $Re = 2650$ (b, d) at inspiration (a, b) and expiration (c, d).

Figure 5 3D view of the high-speed velocity distributions by contours for $Re = 150$ (a, c) and $Re = 2650$ (b, d) at inspiration (a, b) and expiration (c, d). The slice through the volume, showing the normalized velocity $u^*$, reinforces the velocity distribution inside the primary bronchi.
The slice through the volume, showing the normalized velocity \( u^* \), reinforces the illustration of the velocity distribution inside the primary bronchi. For all four parameters, the tracheal high-speed distribution resembles the cross-sectional shape and requires most of the cross-sectional area, whereas the distribution inside the right primary bronchus is round-shaped with minor spatial extension.

In the left bronchus, the high-speed distribution differs strongly for inspiration and expiration. For inspiration, the distribution shifts towards the outer bend of the bronchi and deforms to a U-shaped pattern at increasing Reynolds numbers. In addition, a non-uniform stagnation region at the first bifurcation can be identified. At expiration, this region is homogeneously distributed over the bronchial cross-section, i.e., elliptical, due to the merging of the fluids out of the sub-branches. Moreover, only for \( \text{Re} = 150 \) a part of the expiratory fluid flows out of the upper secondary bronchus. That results in an undersupply of air in that lobe, when taking into account the reduced inspiratory flow of that bronchus.

The helicity is calculated to identify vortices inside the fluid. Figure 6 shows in two views the helicity contours for inspiration (a) and expiration (b) at \( \text{Re} = 150 \) for values of the relative helicity \( |H_r| > 0.35 \) and \( |H_r| > 0.6 \) and five slices through the volume with the helicity as contour. The streamlines indicate the flow direction.

Both results of the primary and the secondary flow pattern are in fundamental agreement with the findings given by Dean (1928a, 1928b), where the Dean number for the right and the left main bronchus are \( \text{De}_R \approx 80 \) and \( \text{De}_L \approx 70 \), respectively.

![Image of helicity contours](image-url)
3.2 Pulsatile flow

This section presents the results for the pulsatile measurements for which the Stereo-Scanning PIV technique was used, where the focus is on the spatial and temporal velocity distribution.

Figure 7 compares the velocity distribution of the convective ‘C’ (Re = 150) and the unsteady ‘U’ (Re = 40) flow domain at maximum inspiration in test-section TS 1. The contours show the averaged normalized velocity $u^*$ and the vectors evidence the in-plane components in each recorded plane (scan-direction in negative z-direction). Every second vector in the y-direction is left out to give a clear picture about the velocity profiles. The convective flow profile inside the right primary bronchus is shifted to the outer wall of the bend, i.e., the wall closest to the carina. A very small part of flow enters the upper secondary bronchus while the main part continues downwards into the lower secondary bronchus. For ‘U’ the flow is more broadly distributed over the cross-section. The main part still flows straight downwards into the lower secondary bronchus, but more than the half of the velocity magnitude is reached at the bifurcation and upwards into the upper secondary bronchus, as it can be seen in the center plane.

![Figure 7 Maximum inspiratory flow for ‘C’ (top) and ‘U’ (bottom) in TS 1. The contours show the normalized velocity distribution and the in-plane vectors indicate the flow direction.](image)

Figure 8 Normalized averaged velocity profiles for ‘C’ (left) and ‘U’ (right) for ‘cut 1’ (see Fig. 9) at 8 instants during one sinusoidal breathing cycle, whereas the abscissae represents the normalized distance $x^*$ over the hydraulic diameter $d_{hyd} = 15$ mm. Inspiration has negative values, expiration positive.
Figure 8 shows the velocity profiles (‘C’ and ‘U’) for this center plane (‘C’: plane ‘5’, ‘U’: plane ‘4’) during one ventilation cycle. The convective flow profiles are parabolically shaped and temporally asymmetrical, i.e., the velocity profile at maximum inspiration is shifted to the left wall with the peak at \( x^* = 0.44 \), whereas the peak at maximum expiration is at \( x^* = 0.56 \). The inspiratory profile during acceleration is symmetric along the center axis. The unsteady velocity profiles are temporally highly symmetrical between inspiration and expiration. The maximum inspiratory and expiratory profiles are more broad than for ‘C’ and shifted to the inner wall of the bend with the maximum at \( x^* = 0.7 \). The velocity profiles for the deceleration of the fluid at \( 0.5 \cdot u^*_{\text{insp, max}} (I \rightarrow E) \) and \( 0.5 \cdot u^*_{\text{exp, max}} (E \rightarrow I) \) have more parabolic shapes that are shifted to the inner wall of the bend. Conversely, the velocity profiles for the acceleration at \( 0.5 \cdot u^*_{\text{insp, max}} (E \rightarrow I) \) and \( 0.5 \cdot u^*_{\text{exp, max}} (I \rightarrow E) \) possess an M-shape with a distinct peak near the inner wall of the bend.

In Figure 9 (top), the flow domains viscous (left) and unsteady (right) are compared for the center plane and for maximum expiration. The results evidence a changed off-plane velocity contour to an M-shape structure due to the higher ventilatory frequency. Figure 9 (bottom) plots the averaged absolute velocity for the three in-plane sections (‘1’ – ‘3’) and the associated flow domains viscous and unsteady. Concerning the fact that all measurements are triggered with the inspiratory peak velocity of the linearly moving piston a variable phase lag is evident. Moreover, a sectional (‘3’ to ‘1’, ‘2’) phase shift during the in-/expiratory transition for ‘U’ can be observed. The temporal velocity distributions of sections ‘1’ – ‘3’ show that the low frequency parameter (‘V’) is in phase to the applied ventilation and can be therefore characterized as quasi-steady. In contrast, unsteady (‘U’) ventilation is strongly delayed.

**Figure 9** Top: Maximum expiration for ‘V’ (left) and ‘U’ (right). Bottom: Absolute velocity for the sections ‘1’, ‘2’, and ‘3’ and ‘V’ and ‘U’, where the first peak represents maximum inspiration and the second the maximum expiration.
4. Conclusion and outlook

Time-resolved PIV measurements inside a realistic human lung model for steady and pulsatile ventilation were presented, the latter for three parameters, each parameter covering one flow regime defined by an order-of-magnitude estimation of the relevant fluid-mechanical equations (Jan et. al., 1989). Steady measurements were performed for Re = 150 and Re = 2650 for inspiration and expiration. Pulsatile measurements using a stereo scanning technique for Re = 10 and Re = 150 at α = 1.5 represent the flow domains viscous (‘V’) and convective (‘C’), respectively, whereas Re = 40 at α = 5 describes the unsteady flow domain (‘U’).

The zones ‘V’ and ‘C’ are evidenced to be quasi-steady for this realistic lung geometry. The pulsatile convective flow profile resembles the steady measurements with slightly different shifted peaks for the flow distribution. The maximum velocity is shifted to the lower wall for inspiration and towards the upper walls of the bronchi for expiration. The velocity distribution of the viscous flow domain is almost parabolic shaped, but slightly influenced by the sub-branches.

For steady inspiratory flow, a U-shaped high-speed velocity profile develops due to high centrifugal forces that occur because of the curvature inside the left primary bronchus. This profile is more pronounced for higher Reynolds numbers. Conversely, the elliptical profile inside the right primary bronchus is independent of the Reynolds number.

An undersupply of the right upper lobe is noticed for convective flows (‘C’). The inspiratory flow is pressed against the lower wall of the bronchi, which leads to a higher inflow into the lower lung. The solution for this undersupply consists of homogeneous ventilation over the unsteady flow domain. The high-frequency broadens the velocity profile that results in an M-shaped form. The homogeneous ventilation leads to an enhanced ventilation of the lateral bronchi. Moreover, for patients suffering from ARDS, the dorsal and lower lung areas possess a reduced function, which increases the importance of greater ventilation inside the functioning upper lung areas to ensure a sufficient supply of oxygen and a reduced flux inside the lower lung areas for no further lung damage.

During inspiration, in both the right and left primary bronchus, one vortex-pair develops which begins approximately half a hydraulic diameter above the carina. At expiration, one vortex-pair develops inside each primary bronchus and two vortex-pairs form inside the trachea. The dominance of the left or right-rotating vortex of the vortex-pair varies.

Comparing the spatial velocity distribution at maximum expiration for viscous and unsteady flow a changed off-plane velocity contour to an M-shape structure is evident. This is the result of the frequency-dependent cross-sectional velocity distribution.

Temporal analysis of the averaged absolute velocity for three in-plane sections inside the first right bifurcation, denoted as TS 1, and the associated flow domains viscous and unsteady show a variable phase lag for the unsteady flow. Hence, two different internal frequencies for inspiration and expiration are evident inside that multi-bifurcation system for unsteady flow. Therefore, the temporal peak velocity for expiration is reduced compared to inspiration. Additionally, a premature outflow of the upper right lung lobe is observable in accordance with Soodt et al. (2011).

The Scanning-PIV measurements were limited to a Reynolds number of approximately Re = 300. Higher applicable Reynolds numbers would enable investigations for physiological breathing and turbulent flows. Additional experiments should be performed with more and different parameters to verify the investigated flow structures of the different flow domains and also to achieve further understanding about the transition between the flow domains. Furthermore, the discrete mass flow has to be calculated for each recorded plane and then summed up to generate the volumetric mass flow to determine the exact temporal changes.

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