Experimental investigation of the bronchial velocity distribution
using scanning stereo PIV

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Abstract   A thorough analysis of the unsteady flow field in a realistic shaped, transparent model of the
human lung is presented in this experimental study. The model consists of the bronchial tree up to the third
generation of bifurcation. Spatial focus is set on the first bifurcation between the right main bronchus and the
subsequent lobe bronchi, whereas the temporal focus concentrates on the transition from inspiration to
expiration. Due to the highly three-dimensional and asymmetric character of the flow field time resolved 3D-
3C measurements are performed using quasi-volumetric stereo scanning particle-image velocimetry (PIV)
and a refractive-index matched fluid. The spatial expansion of the investigated volume is 20x18x18mm³ and
covers the total bronchial cross section. The temporal resolution is in relation to the ventilation period
smaller than 1 degree for both measured Womersley numbers of α₁ = 3.35 and α₂ = 4.11. The investigated
Reynolds number of Re₀ = 3,000 based on the hydraulic diameter and the bulk velocity at maximum
inspiration represent human breathing under light activity condition. The quality of the presented advanced
PIV method is validated by the means of a detailed cross correlation analysis. The results evidence the
scanning stereo particle image velocimetry to be a feasible quasi-volumetric acquisition technique for flow
field measurements that is capable to capture the 3D-3C velocity fields in intrinsic and complex geometries.
The temporal and spatial developments of the flow field for three different temporal states in six successive
planes are presented. The analyses determine the location and the size of vortical structures and the temporal
and spatial extension of regions with different velocity magnitude and direction. Some results are drawn to
verify the quality of the presented PIV technique. Results are that at the beginning of the expiration phase the
outflow from the lower lobe bronchus lags the outflow from the left lobe bronchus and that the magnitude of
the secondary flow structures increase with higher Womersley numbers.

1. Introduction

The human lung consists of a repeatedly bifurcation network of tubes with progressively decreasing
diameters. It is necessary to understand the highly unsteady and non-linear flow field within these
tubes of the human respiratory organ to develop aerosol drug delivery systems and to improve the
efficiency and usability of artificial respiration systems.

Studies show that in Europe approximately 100,000 patients a year are artificially ventilated due to
the Respiratory Distress Syndrome (RDS) (The Acute Respiratory Distress Syndrome Network
2000). According to the predictions of the World Health Organization (WHO), this value will be
increase by 100% by the year 2020. Unfortunately, many details of the lung function are still
unknown. Therefore, a meaningful control mechanism for lung machines could not be derived, yet.
That is, the mortality rate of patients with Acute Lung Injury (ALI) is still in the range of 40 - 50%.
For this reason, general clinical studies, e.g., from Dreyfuss & Saumon (1998) and Zambon &
Vincent (2008) investigated different possible mechanisms of ventilation and related them to the
mortality rate. This vital need for newly optimized respirators and novel ventilation patterns that
can only be achieved by a deeper understanding of the complete flow field in the human lung.
Numerous investigations in a single plane of the human lung have been performed in the last years.
Martonen et al. (2001) report a secondary flow pattern in form of vortices which occur in daughter
airways downstream of the carina. Experiments were conducted at \( Re = 200 \) – \( 2,000 \) showing that the shapes and intensities of the secondary flows are a function of the inlet conditions. Furthermore, Grotberg (2001) found the secondary flow amplitude to decrease for \( Re = 200 \) – \( 1,200 \) if the Womersley number and the flow amplitude are reduced. Razumat & Riethmueller (2002) investigated the oscillatory flow within a three-dimensional model of three successive generations of lung bifurcations. Their results show a quasi-steadiness of the flow when the frequency of the oscillations decreases. Moreover, flow structures can repeat themselves from one bifurcation to the other under certain conditions despite the presence of strong secondary motions.

Recent experimental studies, e.g., from Große et al. (2007, 2008) and Eitel et al. (2010), focused on the analysis of steady and unsteady flow in a realistic model of the first six bifurcations of the upper human airways. Große et al. (2007, 2008) analyzed the flow field at steady inspiration and expiration for two different Reynolds numbers and detected a minor role of the Reynolds number on the extension of the flow phenomena. On the other hand, the Womersley number plays a major role at unsteady flow. The size of the evolving secondary flow structures strongly depends on the instantaneous Reynolds number and Womersley number. Eitel et al. (2010) compared numerical and experimental data for the pulsatile flow field of the human lung. The experiments for the first six generations were conducted using particle-image velocimetry (PIV) in a 1:1 realistic lung cast. Subsequently, the secondary flow structures were analyzed on the basis of numerical solutions. Therefore, time-resolved and three-dimensional measurements are required.

Such experiments can be based on the scanning particle-image velocimetry technique which has been applied to various flow problems. First investigations using scanning particle-image velocimetry were performed by Brücker (1995, 1996) where a mirror mounted on a stepper motor generated nine scanning light sheets in an instantaneous flow behind a circular cylinder at a Reynolds number \( Re=300 \). Hori & Sakakibara (2004) conducted stereo scanning PIV measurements with one scanning mirror for the three-dimensional distribution of three-component velocity in a turbulent round jet at \( Re=1,000 \) in 50 velocity planes.

Burgmann et al. (2008a) and Burgmann & Schröder (2008b) used stereo scanning PIV to investigate the vortex induced unsteadiness of a separation bubble on an airfoil. The number of scanning light-sheet planes depended on the chord length based Reynolds number which varied in the range of \( Re = 20,000 \) - \( 60,000 \). For those measurements a high-speed laser system consisting of a row of infrared laser diodes mounted in a single rack was used. Kallweit et al. (2008) applied stereo scanning PIV to investigate unsteady and relative slow biomedical fluid flows. They extend the stereo scanning PIV by using a polygon mirror. The system was limited by the pixel resolution of the cameras and the number of polygon planes of the scanner.

Considering the existing knowledge on the scanning PIV technique in the context of low Reynolds number three-dimensional flow this technique will be used to investigate the flow field in the upper human airway system.

The paper is organized as follows. Firstly, the experimental method and the equipment are
described. Secondly, 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 Große et al. (2007, 2008) and Eitel et al. (2010). The scanning stereo PIV experiments were implemented in a realistic, three-dimensional lung model that extends to the 3rd generation of the bronchial system. The manufacturing process of the model encompasses radiological scanning of a human proband, digital geometry reconstruction, segmentation, surface generation and iterative surface smoothing, geometry post-processing, corn starch rapid prototyping, surface coating, inlet insert attachment, transparent silicon casting, and hot water wash-out. The extensions of the model’s exits to the outer dimensions of the cast have been considered in geometry post-processing to avoid manual channel drilling. The material of the hollow lung model consists of transparent silicon to allow perfect optical access. A detailed description of the manufacturing process can be found in Große et al (2007, 2008).

The model can either be embedded 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 breathing cycle, are technically realized by a linear actuating piston. The ventilation mode of the oscillatory flow can be parameterized as sinusoidal or user-defined at different frequencies and amplitudes. Figure 1 shows a schematic illustration of the experimental setup for oscillatory flow.

Figure 1. Schematic of the experimental setup for oscillating flow, geometry of the inlet cross section, and two photographs of the model.

The inflow conditions are critical since the velocity profile in the trachea has a strong influence on
the flow field in the first lung bifurcation (Yang et al. 2006). To allow a comparison with numerical results, a defined inflow condition is necessary. Hence, the fluid has to enter the lung model through an anatomically shaped trachea with a dorsal indentation to uniquely define the inflow characteristics. For that purpose a pipe section with constant cross-section matching that in the upper part of the trachea with a length of \( L = 500 \text{ mm} \) was manufactured to ensure a fully developed velocity profile. The hydraulic diameter \( D \) of the pipe is approximately 17.9 mm such that \( L/D \) is 27. The Reynolds number based on the hydraulic diameter and the bulk velocity at maximum inspiration is approximately \( \text{Re}_d = 3,000 \). The total length of the human trachea is 10 - 12 cm measured from the laryngeal region (cartilage cricoidea) to the first bifurcation (bifurcation tracheae) at an average hydraulic diameter of 15 - 27 mm. Consequently, the real flow in the human body cannot be considered fully developed.

The flow exits the model through realistically formed endings. The outlet pressure in the experiment can be assumed nearly constant for all branches of the 3th generation exiting in the tank since there is hardly any disturbance of the hydrostatic pressure distribution in the tank.

A water/glycerine mixture with a refractive index of \( n = 1.44 \) being identical to that of the silicone block of the hollow model was used. The refractive-index matched fluid ensures optical access without distortion. The dynamic viscosity of the mixture with 60.7 mass percent glycerine is \( \eta = 10.3 \times 10^{-3} \text{ Pa·s} \) and the density \( \rho = 1.153 \times 10^3 \text{ kg/m}^3 \) for a temperature \( T = 23.8^\circ\text{C} \) which was held constant during the experiments.

The experiments are performed at two Womersley numbers \( \alpha_1 = 3.35 \) and \( \alpha_2 = 4.11 \). That is, typical human breathing cycles under normal conditions can be simulated with this setup.

An electrolytic unit upstream the above mentioned inlet pipe generates hydrogen bubbles which serve as tracer particles. The number of bubbles in plain water/glycerine is rather low due to missing electrolytes in the fluid. Thus, a small amount of citron acid is added to fluid to enhance particle generation. The voltage applied to the electrolytic unit, the distance between the metal cathode and anode of the electrolytic unit, and the mean flow velocity through the unit determine the size of the tracer particles. The size of the tracer particles is in the range of is 1 - 5 \( \mu \text{m} \). A homogeneous distribution of the particles is achieved by starting the electrolytic unit at least 15 min prior to the measurement series. The lifespan of the bubbles ranges from 10 to 30 min which is large compared to the measuring time.

The predesign of the flow configuration is based on typical human respiration pattern. A natural respiration cycle is nearly sinusoidal with an approximate time period of \( T_{V,1} = 5 \text{ s} \) and an approximate tidal volume of \( V_T = 0.5 \text{ l} \). Respiration rates during activity easily increase to \( T_{V,2} = 2.5 \text{ s} \) resulting in a maximum volume flux of \( \dot{V}_1 = 0.2 \text{ l/s} \) and \( \dot{V}_2 = 0.4 \text{ l/s} \), respectively. The tracheal cross sectional area of the 1:1 lung model illustrated in Fig. 1 is \( A_T = 2.79 \text{ cm}^2 \) yielding a maximum bulk velocity at inspiration for an average volume flux \( u_R = \dot{V}/A_T = 1.08 \text{ m/s} \).

All experiments of this study were carried out using the stereo scanning PIV system shown in Fig. 2. A continuous wave Ar+ laser Innova 90C-A6 (Coherent GmbH) operating at a wavelength of \( \lambda = \)
514.5 nm and a maximum light power of \( P = 2.4 \) W is used as light source. The Beer-Lambert absorption law \( \frac{I}{I_0} = \exp(-\mu \cdot x) \) describes the attenuation of the emitted light \( I_0 \) by the water/glycerine mixture as a function of the absorption coefficient \( \mu \) of the fluid and the penetration depth \( x \) in the tank. Assuming that the absorption coefficient is proportional to the wavelength of the laser light, the intensity \( I \) of the light in the measurement volume decreases at increasing wavelength \( \lambda \) such that lower wavelengths are favorable for illumination.

A combination of three lenses (I: convex lens; II: cylindrical lens; III: convex lens) is used to generate the light sheet. A polygon mirror linked to a high precision DC motor (maxon motor GmbH) deflects the incident light sheet. The motor is equipped with an MR encoder and uses the positioning control unit Epos 70/10 with velocity feedback regulation to control the rotation speed. The polygon mirror consists of 15 aluminium coated facets (Alflex B) with quartz protection. The polygon width is \( h = 30 \) mm resulting in a scan angle of \( \beta = 34^\circ \). A plano-convex lens in the light path between polygon mirror and tank rectifies the fanning light sheet to parallel measurement planes.

A glass fiber cable is positioned in the upper right corner of the scanning area \((y \geq 0, z \geq 0)\) and outside of the investigated zone. It is connected to a high speed photo sensor which converts the bypassing light impulse into an electric signal such that the light sheet related to the origin is exactly positioned during scanner rotation. The Gaussian distributed electrical signal is conditioned to a standard TTL signal with an increased supply current up to \( I = 20 \) mA to support the trigger prerequisites of both SA3 high-speed cameras (Photron, Inc.). The conditioning steps include amplification, offset shift, and digitalization. The high current analog to digital conversion is performed by a comparator/ transistor circuit which controls simultaneously both ‘TrigIn’ interfaces of both cameras.

The tank possesses a quadratic base area \((x-y\) plane\) with a lateral length of \( L = 300 \) mm and consists of transparent PMMA (Plexiglas®) walls. Obscure black foils protect for hazardous laser radiation and increase optical contrast during measurements by generating a dark image background. To ensure undisturbed outflow conditions, i.e., to minimize the influence of the walls, the lung is placed in the center of the container.

The light speed of the water/-glycerine mixture and air differs. To ensure likewise path lengths from the object plane \((x\)-axis\) to the image plane \((lens)\) and as such to keep the image perpendicular to the optical axis, two prisms at similar inclination as object to image plane and material with a light speed analogous to the test fluid are used for the optical paths to the stereoscopically arranged cameras. The prisms are glycerine filled containers with adjustable inclination angle and light protection. The lenses are two AT-X 100 Pro D (Tokina Industrial, Inc) prime lenses with an aperture of \( A = 5.6 \). The lenses and the cameras are mounted on a Scheimpflug adapter with an angle of \( \varphi = 9^\circ \) between the two optical axes to obtain a clear image in the \( x \)-direction. Bellows between lenses and cameras enclose both components to light and dust and allow fine adjustment of the Scheimpflug angle. Both cameras are synchronized and trigger a high-level signal during
recording via ‘CamOut1/2’. The signals are used to synchronize the cameras with the motion of the linearly actuated piston and to control the camera synchronization. Each camera is connected via high speed Ethernet with a standard PC.

In Fig. 2 ‘A’ shows a long-time exposure of the scanning light sheets over a translucent calibration grid that is positioned normal to the y-direction. The length of one unit of the grid is 5 mm. Hence, a homogenous distributed scan distance of approximately 30 mm is generated by the rotating polygon mirror. The left and the right boundaries are stronger illuminated since the refraction near the framework of the plano-convex lens is inhomogeneous. Therefore, only the measurement volume illuminated by the inner part of the light sheet will be analyzed.

![Diagram of the optical arrangement](image.png)

Figure 2. Top view of the complete optical arrangement. The image A shows the expanded volume caused by the high frequency rotation rate of the polygon mirror in the sagittal plane.

The SA3-120K high-speed cameras are capable to record full frame (1024x1024 px) images at a maximum frame rate of $f_{\text{frame}} = 2 \text{ kHz}$ with a total frame number $n_{\text{Total}}$ depending on the size of the camera memory. A calibration grid with an equidistant grid spacing of 5 mm has been placed in the x-z-plane and images have been captured for $-10 \text{ mm} < y < 10 \text{ mm}$ in $\Delta y = 1 \text{ mm}$ discretization steps. The ratio between real and virtual units is $20 \text{ px} = 1 \text{ mm}$. The lens foci are adjusted to be perfectly on the center plane ($y = 0 \text{ mm}$). The size of the aperture is a compromise between depth of field and light intensity. Both cameras are positioned in the positive x-direction to take advantage of
the increased light intensity of the forward scattered light. The ‘TrigIn’ signal, resulting from the laser light passing the fixed trigger point, is delayed \( t_{\text{Delay}} \) and used to initialize the recording of the data. The delay synchronizes the exposure with the light sheet entering from the bronchus border. The subsequent frames \( n_{\text{Frame}} = 8 \) with a frame time of \( t_{\text{Frame}} = 1/f_{\text{Frame}} = 1/10 \) kHz scan through the bronchus and capture the complete flow field in \( l_{\text{Scan}} = 1.2 \) mm steps. The total light sheet thickness \( l_{\text{Scan}} \) reads

\[
l_{\text{Scan}} = l_{\text{Stop}} + u_{\text{Motor}} \cdot t_{\text{Shutter}}
\]

where \( l_{\text{Stop}} = 0.6 \) mm represents the light sheet thickness for a stopped scanner and \( u_{\text{Motor}} = \omega \cdot D_{\text{PM}}/2 \) is the rotation speed of the scanner motor with \( D_{\text{PM}} \) as the diameter of the polygon mirror and \( t_{\text{Shutter}} = 1 / f_{\text{Shutter}} \) as the shutter time of the camera. The setup has been designed following the relation \( l_{\text{Scan}} = 2 \cdot u_{\text{Motor}} \cdot t_{\text{Shutter}} \) that guarantees a seamless image acquisition without separation and overlap. After the last frame of the first sequence is stored the camera returns to ready status to record the next sequence. To obtain an optimal pulse distance \( t_{\text{PD}} \) for a cross correlation, the camera and motor parameters are \( u_{\text{Motor}} = 1,000 \) 1/min and 382x352 px at \( f_{\text{Frame}} = 10 \) kHz. Image post-processing steps include double image sortation, perspective mapping based on the calibration images, and a dewarp algorithm. In the following analysis are due to low image quality only the frames from 2 to 7 taking into account.

![Figure 3. Schematic of the camera signal processing and principle image acquisition for cross correlation.](image)

3. Results and discussion

Figure 4 and 5 depict the results of the stereo scanning PIV measurements in the first bifurcation between the right main bronchus and the subsequent lobe bronchi for Womersley numbers of \( \alpha_1 = 3.35 \) and \( \alpha_2 = 4.11 \), respectively. The peak Reynolds number based on the hydraulic diameter \( D \) of the trachea on the maximum bulk velocity during inspiration is \( \text{Re}_D = 3,000 \). The analysis focuses on the transition from inspiration to expiration for a sinusoidal form of ventilation. The vectors indicate the in-plane velocity components \( u \) and \( v \), whereas the contours represent the out-of-plane component \( w \) of the velocity field. The solid lines are simplified illustrations for the inner bronchial outline extracted from the geometrical data set.
Figure 4. Velocity distributions in six selective planes at the first bifurcation between right main bronchus and the subsequent lobe bronchi. Φ1, Φ2, and Φ3 denote different temporal points during the transmission from inspiration to expiration for a Womersley number of $\alpha = 3.35$. The vectors indicate the in-plane velocity $(u, v)$ and the contours represent the out-of-plane component ($w$).
Figure 5. Velocity distributions in six selective planes at the first bifurcation between right main bronchus and the subsequent lobe bronchi. \( \Phi_1, \Phi_2, \) and \( \Phi_3 \) denote different temporal points during the transmission from inspiration to expiration for a Womersley number of \( \alpha_2 = 4.11 \). The vectors indicate the in-plane velocity \( (u, v) \) and the contours represent the out-of-plane component \( (w) \).
For reasons of clarity and comprehensibility, only every second measurement plane is displayed in Fig. 4 and 5, respectively. Hence, the lateral spacing of the vector field equals 2.4 mm while the light sheet thickness was 1.2 mm during these experiments. Φ1, Φ2, and Φ3 denote different temporal states described by the phase angle during the transition with respect to maximum inspiration (Φ = 0°). The comparison of the in-plane and out-of-plane velocity distributions of different planes of the same temporal state show the three-dimensional spatial extension of the flow structures, e.g. vortices, high-speed regions, and shear layers.

The contours of plane 4 (α1) and planes 4 & 5 (α2) indicate at the end of inspiration (Φ1) a three-dimensional vortex structure in the left sub-bronchus that’s vortex vector is parallel to the x-y-plane. The spatial extension in the x-y plane of this vortex is similar to the diameter of the related bronchus.

Planes 2 - 5 (α1) and planes 3 & 4 (α2) at temporal state Φ3 show a remaining high-speed region that occurs during inspiration. The interaction between the outflow of the left sub-bronchus and the high-speed region of the inflow generates a second vortex (plane: 2 & 3 for α1, plane: 2 - 4 for α2) with a vortex vector parallel to the z-direction. The second vortex is located at the inner wall below the bifurcation. The spatial extension in the x-y plane and y-z plane is approximately half the diameter of the lobe bronchus.

Concerning the temporal results that have obtained with a corresponding discretization accuracy of ΔΦ = 0.58° for α1 and ΔΦ = 0.86° for α2 it can be clearly shown that the outflow from the lower lobe bronchus lags the outflow of the left lobe bronchus. The shear layer, resulting by the opposite flow directions, moves with increasing phase angle Φ from the point of bifurcation to the right outer wall until no inflow remains.

Due to the slight difference of the regarded Womersley numbers α1 = 3.35 (Fig. 4) and α2 = 4.11 (Fig. 5) a small variance in the presented results was obtained. However, the comparison evidences that the spatial extension in the x-y plane of the second vortex (Φ3) is reduced for a higher Womersley number. Figure 5 (plane 6, Φ3) shows also in comparison to Fig. 4 (plane 6, Φ3) a more pronounced w component between the lower lobe bronchus and the main bronchus. The latter indicates a frequency dependent increase of the secondary flow during expiration in accordance to the numerical simulation performed by Eitel et. al. (2010).

In order to increase the amount of analyzed data different properties of the experimental setup have to be redesigned. Using an advanced high-speed camera either the spatial resolution of the time-resolved data will increase or the out-of-plane error will be reduced due to the higher frame rate. Future modifications will include an optimized electrolysis unit to increase the amount of tracer particles and keep it constant during the ventilation cycle. In addition, to render the light sheet positions more precisely the balance of the scanner has to be optimized.
4. Conclusion and outlook

Time-resolved and three-dimensional velocity measurements and a thorough analysis of an oscillating flow in a realistic shaped model of the human lung were presented. The model and the experimental setup has been described in detail in Sec. 2 and from the presented results it can be concluded that the stereo scanning particle image velocimetry is a feasible quasi-volumetric acquisition technique for flow field measurements for complex and intrinsic geometries. The temporal accuracy is sufficiently high to capture the flow structures and their development during normal breathing conditions, whereas the spatial extension in scan direction is capable to capture the complete flow field of bronchial cross sections.

Results were presented for two Womersley numbers of $\alpha_1 = 3.35$ and $\alpha_2 = 4.11$ at a constant peak Reynolds number of $Re_D = 3,000$. The data were focused on six selective planes through the first bifurcation of the right main bronchus and three temporal states during the transition from inspiration to expiration. Analyses show the occurrences and development of different vortical structures and determine their location and size. The comparison between different Womersley numbers evidence a frequency dependent increase of secondary flow structures.

Future research will include different investigation regions and phase angles of the ventilation cycle to gain more knowledge about mass flux distribution, mixing, particle deposition, and wall shear stresses. Additionally, the experimental setup will be optimized to increase the quality of correlation to enable the opportunity for statistical analyses.

Acknowledgement

The support of the ‘Intelligent Laser Applications GmbH’ and the funding of the ‘Deutsche Forschungsgemeinschaft’ are gratefully acknowledged.
5. References


