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Effect of Upper Airway on Tracheobronchial Fluid Dynamics

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ABSTRACT

The upper airways play a significant role in the tracheal flow dynamics. Despite many previous studies, however, the effect of upper airways on the ventilation distribution in distal airways has remained a challenge. The aim of this study is to experimentally and computationally investigate the dynamic behaviour in the intra-tracheal flow induced by the upper respiratory tract and to assess its influence on the subsequent tributaries. Patient-specific images from two different modalities (MRI of the upper airways and CT of the lower airways) were segmented and combined. An experimental phantom of patient-specific airways (including the oral cavity, larynx, trachea, down to generations 6-8) was generated using 3D printing. The flow velocities in this phantom model were measured by the flow sensitized phase contrast MRI technique and compared to the computational fluid dynamics simulations. Both experimental and computational results show a good agreement in the time-averaged velocity fields as well as fluctuating velocity. The flows in the proximal trachea were complex and unsteady under both lower- and higher-flow rate conditions. CFD simulations were also performed with an airways model without the upper airways. Although the flow near the carina remained unstable only when the inflow rate was high, the influence of upper airways caused notable changes in distal flow distributions when the two airways models were compared with and without the upper airways. The results suggest that the influence of the upper airways should be included in the respiratory flow assessment as the upper airways extensively affect the flows in distal airways and consequent ventilation distribution in the lungs.

Keywords: Upper airway – trachea – MRI – CFD – flow – ventilation

1. INTRODUCTION

Recent progress in medical imaging and image-based modelling technique enables accurate flow assessments in patient-specific airways geometries [1-3]. Computational fluid dynamics (CFD) is a valid tool to assess the flow velocity and pressure in respiratory airways [4-10]. Patient-specific CFD modelling studies of central airway (from the trachea down to several generations) flows can describe the tracheal flow characteristics such as velocity profile, wall shear stress and pressure in detail [11-17]. In the meantime, the importance of upper airways geometry has been often overlooked in past lung modelling studies, and this is partly because the upper airway is not routinely included in the scope of conventional lung imaging protocol in the radiological assessment of respiratory disease due to additional radiation exposure to patients [18,19].

Instead, the flows in the human upper airways were modelled separately without lower airways in multiple studies [20-25]. Their modelling results revealed the complex and unstable flow characteristics in the upper airways while the influence on the consecutive flows in the trachea was still less well understood. Recently, experimental and computational studies included the upper airway geometries to elucidate the tracheobronchial flow dynamics [26-29]. Phuong et al. measured the flow velocities in a realistic replica of the human airway track using particle image velocimetry (PIV) under constant breathing conditions and compared to the Reynolds Averaged Navier-Stokes (RANS) CFD models [26]. Calmet et al. Conducted a large-scale CFD simulation with the finest meshes (350 million elements) to highlight the unsteady flow characteristics during a rapid inhalation [27]. Banko et al. demonstrated the time-varying flow velocity field in an anatomically accurate experimental model using phase contrast magnetic resonance velocimetry [28]. Lambert et al. performed a

Large Eddy Simulation (LES) of the flows in a computed tomography (CT) based human airway model to show the left-right lung asymmetry of particle ventilation [29]. In those studies, the geometries of upper respiratory tracks (including the oral cavity, oropharynx and larynx) generated complex flow structures such as laryngeal jets, localised vortices and secondary flows followed by significant pressure drops and altered wall shear stress distributions in the tracheas. Their flow characteristics were different from the less skewed cross-sectional velocity profiles and simple flow patterns found in the lower airway only models [30-32]. Choi et al. demonstrated the formation of a laryngeal jet at the glottis in their LES model. The laryngeal fundamentally affected the turbulent flow characteristics in the trachea. They consequently showed that the formation of laryngeal was prevented by removing glottis constriction and the tracheal flow was laminar without the upper airway track [33]. The turbulent flow behaviour induced by the upper airway was gradually attenuated as the flow moves towards the distal branches. However, unsteady flow fluctuation was still discernible not only in the trachea but also in the primary and secondary bronchi [34]. While the local flow dynamics caused by the upper airway is well described in literature, the potential influence of upper airways affecting ventilation distribution in the lung has not been fully discussed yet.

This paper aims to investigate the tracheobronchial flow alterations caused by the upper airway geometry utilising a patient-specific CFD model. The combined airway (CA) models of upper and lower airway geometries are obtained from magnetic resonance imaging (MRI) and CT images from the same patient. The pulmonary flow in a combined airway model is validated using the phase contrast velocimetry (PCV-MRI) measurements [35-38]. 3D and 2D/1D unsteady MRI flow measurements in the patient-specific airways model are compared with CFD simulations. CFD simulation results are also compared with and without the upper airways to illustrate their difference in tracheal flow characteristics and bronchial flow distribution. The effect of complex upper airway structure and the tracheal flow dynamics on the ventilation distribution in distal airways is discussed.

2. METHODS

2.1. Model geometry

A forty-nine-year-old female patient's imaging data was recruited in this study. The upper airway was segmented from the MRI images using ScanIP (Simpleware, Exeter, UK) while the lower airway model was reconstructed from the CT images using Mimics (Materialise, Leuven, Belgium). The segmented airway geometries were mutually registered and combined into a complete central airway model (Fig. 1). As a consequence, the resultant combined airway (CA) geometry consisted of the oral cavity, pharynx, larynx, trachea, primary bronchi, main bronchus, and up to the seventh generation of following local airway branches. It was noticeable that there was a significant constriction at the laryngeal airway. Based on the combined geometry of the airway model, a phantom model was 3D printed using the stereolithography technique (Materialise, Leuven, Belgium) and an MR-compatible material (TuskXC2700T / Tusk2700W, Tusk Somos®, Elgin, IL). In addition to the combined model of the upper and lower airways, we developed a lower airway (LA) geometry model to compare the computed flows with the CA model. The inlet boundary plane of the LA geometry was defined with extra care to obtain similar flow conditions in the trachea to those in the CA flow model.

2.2. MRI Flow Measurements

We used two different inflow rates (lower-flow, Q_{low} = 3.5 mL/s and higher-flow, Q_{high} = 20.0 mL/s) in the study. Those flow rates are equivalent to 55 mL/s and 314 mL/s of the air flow rates to represent a quiet and a fast breathing, respectively. Corresponding Reynolds numbers ($Re = UD/v$) of the lower-flow (LF) model and the higher-flow (HF) model were Re=350 and 2000, respectively. The flow velocities in the patient specific 3D printing model were measured by the flow sensitized phase contrast MRI technique [39]. 3D flow MRI measurements were performed on a 3 T MRI scanner (Philips, Ingenia, Netherlands) with a multi-channel cardiac coil during a constant flow of water (Q_{low}). The lung model was immersed in a water container. The oral cavity inlet was connected to a water reservoir and the flow was controlled by a constant height difference between the reservoir and a small diameter outlet in the container. Copper sulphate at the concentration of 15 mmol/L was added to the water to increase magnetic resonance signal. A 3D fast field echo sequence with flow encoding the gradients along the three axes was used with the following parameters: field of view of 200 mm × 160 mm × 250 mm, resolution of 0.39 mm × 0.39 mm × 1 mm, partial echo, echo time/repetition time of 3.9/7.8 ms, flip angle of 5˚. The acquisition was repeated twice with two different maximum velocity encoding values of 30 and 10 cm/s corresponding to an acquisition time of 12 and 15 minutes, respectively.

In addition to 3D flow measurements, 2D/1D velocity profiles measurements were performed on a 1.5 T scanner (GE, HDx, USA) with a birdcage head coil during a constant flow of water. A 2D Cartesian encoded spoiled gradient echo sequence with flow encoding gradient perpendicular to the slice orientation was implemented with the following parameters: field of view of 250 mm × 187.5 mm, resolution of 0.98 mm × 1.95 mm, slice thickness of 6 mm, flip angle of 60°, echo time/repetition time of 7.9/15 ms. The acquisition time for each 2D slice was 2.9 s. Multiple slices from the trachea entrance to the trachea carina were acquired (see locations S1 to S5 in Fig. 1 and Table 1) for a full acquisition time of ~15 s. To obtain 1D successive velocity profiles, the sequence was repeated with phase encoding gradients off. The time resolution of two successive 1D velocity profiles was 30 ms. The experiments were repeated with different maximum velocity encoding gradients values of 20, 30 and 63 cm/s.

2.3. CFD Modelling and Data Analysis

The segmented model geometries were imported into the Ansys ICEM CFD (Ansys, Abingdon, UK) for pre-processing. A total of 4.8 and 11.3 million hybrid volume meshes consisted of tetrahedrons and prism layers were generated for the LA and CA geometry models, respectively (Fig. 2). In order to achieve accurate flow assessments in the near wall region, seven prism layers were generated within the viscous sublayer while the rest of the space was filled by tetrahedral meshes. Here, the refinement levels of meshes were determined from a grid independence study. We applied two different inlet flow rates (Q_{low} and Q_{high}) in the CFD models as those in the experiments. To provide the equivalent outlet boundary conditions with the experimental phantom, the CFD outlet boundaries were assumed to be uniformly constant (P_{outlet} = 0 Pa). The density and dynamic viscosity of water in the CFD model were 997.56 kg/m^3 and 8.887×10^{-4} Pa·s, respectively. No-slip and rigid boundary conditions were imposed on the airway wall.

The finite volume based CFD code, Ansys CFX (Ansys, Abingdon, UK) was used to solve the Navier-Stokes equations on a fine grid. Even though the inlet boundary flows in this study were set for constant inhalations, the unsteadiness of flows could be developed in the physiological geometry of the airways. Thus, we assumed the transient, incompressible flow conditions to solve the governing equations. The pressure and velocity in the governing equations were solved in a finite volume domain and the PISO algorithm for the coupling. The equations were discretised as the advection fluxes were solved using a high-resolution scheme which is essentially second-order accurate and bounded. An implicit second-order accurate time differencing scheme is used for the transient flows. Since the low-Reynolds number nature of the flows in most of branches, the laminar flow model was used for the lower-flow conditions (Re = 300). However, the localised turbulence due to the presence of a strong jet was observed in the larynx region under the higher-flow condition, we applied the LES model for the higher-flow conditions (Re = 2000). In this model the subgrid-scale eddies were modelled with the wall-adapting local eddy-viscosity model (WALE) while the large-scale eddies were resolved with the transport equation [40]. The flow was computed at every millisecond for 21 seconds and the model solutions were assumed to be converged when the root mean square (RMS) residual of the dimensionless mass and momentum were less than 10^{-5}. The data in last 3 seconds were used to compare with the experimental measurements to avoid an influence of initial conditions. All the computations were carried out on the high performance computing cluster (Lenovo NeXtScale nx360 M5 servers with 2 x Intel Xeon E5-2630 v3 2.4 GHz Haswell 8-core processors; 16 cores per node; 203 nodes; 3488 cores; 64 GB DDR4 memory per node / 4 GB per core) at the University of Warick.

Once the computations of flow properties were completed, the CFD datasets were manually registered and compared to the MRI measurements. For the combined airway geometry with lower-flow (CA-LF) model, the velocity fields on the cross-sectional planes (S1-S8 in Fig. 1 and Table 1) in the CFD models were integrated at every computational time step and averaged for the data acquiring period (3 s). Those time-averaged cross-sectional velocity fields were compared to those in the experimental measurements. Moreover, the local flow rates at twelve different locations in the trachea and main bronchi were compared between the CFD and MRI models.

Since the flow in the coupled airway geometry with higher-flow (CA-HF) model could be highly unsteady depending on its location, the variation of the flow velocity profiles as well as the flow distribution on the cross-sectional planes (S1-S4 in Table 1) in the trachea was illustrated. Here, the velocity profiles over the cross-sectional planes in the CA-HF model were obtained by line averaging of the normal flow velocities from the anterior to posterior direction.

In addition to the flow fields, the spectral energy of the dynamic flow was analysed to elucidate the development of the turbulent characteristics in the tracheal flow in the CA-HF CFD model. The Kolmogorov's energy spectrum of turbulence (E) in the inertial subrange where the energy density depends only on the scale (k) and energy dissipation rate (ε) becomes

$$E(k) = C \cdot \varepsilon^{\frac{2}{3}} \cdot k^{-\frac{5}{3}}, \tag{1}$$

where C is a constant and k is the wave number defined as $k=2\pi Fr$ and Fr is the frequency. The characteristics of power spectra of the kinetic energy in the flows at three different locations (centre points on the cross-sectional planes, S2-S4) in the trachea are presented in this study.

To clarify the effect of the upper airway on the tracheal flow, the flows in the lower-airway-only geometry were simulated with lower-flow (LA-LF) and higher-flow (LA-HF) CFD models. The temporal flow distributions on the coronal plane in the LA geometry models were compared to those in the CA models. In addition to the temporal flow characteristics in the trachea, we compared the dynamic flow split ratios of the tracheal flows into the right main bronchus in the LA and CA models. Furthermore, we visualised the flows in the full airway networks in the models to distinguish the influence of upper airway on the distal airway flow distribution.

3. RESULTS

The tracheal flows in the CA-LF models of the MRI and CFD are almost steady. Figure 3 illustrating the time-averaged flow velocity fields on the cross-sectional planes reveals a great resemblance between the MRI and CFD. The flows in the upper airway tract are deflected by the posterior wall of the curved airway at the oropharynx and consecutively encounter a laryngeal constriction. The narrowing of air pathway before trachea increases the flow velocity and decreases the pressure. Consequently, the increased flow velocity by the laryngeal constriction provokes the formation of the jet stream in the trachea (S8). The cross-sectional flow field in the proximal trachea shows skewed distribution to have a fast flow near the anterior wall (S1). This distribution gradually develops as the flow moves downstream and becomes nearly axisymmetric at the distal trachea (S3-4) due to the low *Re* nature of the flow in the LF model. The tracheal flow splits into two at the carina of the trachea and goes furthers down to the left and right main bronchi (S5-7). In addition to the tracheal flow distribution, shown in Figure 4 and Table 2 are the flow rates in the trachea and main bronchi in the CA-LF model. Again, the local flow data reveal an excellent agreement between the MRI measurement and the CFD analysis. It is notable that more flow goes into the right main bronchus then the left in both MRI and CFD models.

Tracheal flow distributions on the cross-sectional planes in the CA-HF models are demonstrated in Figure 5(a). There are differences in the boundary shapes of the flow fields on the proximal cross-sections, S1 and S2, between the MRI and CFD because the minimal flow velocities (< 0.5 cm/s for S1 and < 0.1 cm/s for S2) are shown in white colour in the MRI, while those are dark blue in the CFD. Despite those differences, the MRI and CFD flow velocity contours reveal substantial similarity. The flow distribution is skewed to the anterior wall to form a strong jet in the proximal trachea (S1-2) and disperse to the larger area in the downstream trachea as it is previously shown in the CA-LF model.

Figure 5(b) illustrates the transition of temporal flow velocity profiles in the CA-HF models. The velocity profiles reveal the flows in the upstream trachea (S1-S2) are highly unsteady with significant variations and the temporal flow distributions on these planes are frequently biased. However, the dynamic characteristics of the flows are gradually attenuated as the flows are going down to the distal trachea (S3-4). In summary, table 3 shows the changes in the mean flow velocity and standard deviation of the velocity profiles along the trachea. Both mean velocity and standard deviations from the MRI and CFD models decrease as the flows move to the downstream trachea. The variations of the temporal velocity profiles in the CFD model are smaller than those in the MRI measurements.

The root-mean-square (RMS) of the tracheal flow velocity fluctuations in the LA-LF and LA-HF models were less 0.002% and 0.004% of the mean flow velocity, respectively.

The root-mean-square (RMS) of the tracheal flow velocity fluctuations in the LA-LF and LA-HF models were less 0.002% and 0.004% of the mean flow velocity, respectively. As the tracheal flows in the CA models are unstable unlike LA models, we computed the spectral energy of the dynamic flows at three different locations along the trachea (S2-4) in the CA CFD models. Figure 6 demonstrates the spectral kinetic energy distribution of the tracheal flow. The slope of energy dissipation at the proximal trachea (S2) reveals the existence of the inertial subrange of turbulent flow in the CA-HF model while the flow in the CA-LF model remains virtually laminar. The kinetic energy of the turbulent flow is dissipated in the downstream flows (S3-4). Consequently, the turbulent flow activity is weakened in the distal trachea (S4) even though this does not imply that the flow is consistent and stable at the distal trachea in the CA models.

To shed light on the effect of the upper airway on the flows in the trachea and main bronchi, shown in Figure 7 is the temporal flow distributions on the coronal planes in the CFD models. The LA geometry models demonstrate even and stable flow distributions in the trachea. Contradistinctively, the tracheal flows in the CA geometry models are more complex and unstable compared to the LA models. The skewed flow distribution induced by the laryngeal jet in the CA-LF model is widely disseminated and disappears near the carina of the trachea. Meanwhile, the CA-HF model shows irregular and inconsistent flow distribution in the entire trachea. The tracheal flows are split by the carina and enter the right or left main bronchi. Thus, the dynamic flow characteristics in the distal trachea directly affect the flows in the right and left main bronchi and even possibly interfere with the flows in subsequent branches.

Figure 8 depicts the ratio of the tracheal flow into the right main bronchus in the CFD models. The tracheal flows in all models are biased towards the right main bronchus (> 0.5) as it is found earlier in the MRI flow measurement with the CA-LF model (Fig. 4). The biased flow ratios are more significant (1.5-2 %) in the HF models compared to the LF models. It is notable that the flow ratios in the LA geometry models are stable regardless of the flow condition. Even though the flow ratio in the CA-LF model changes slowly, it shows a close resemblance to the LA-LF model. Whereas, the flow in the CA-HF model constantly fluctuates and the average flow ratio is slightly (0.3-2.3 %) lower than the LA-HF model.

In addition to the dynamic flow behaviours in the main bronchus, the time-averaged flow distributions over the entire airway networks in the CFD models are illustrated in Figure 9. The boxes in this figure depict the airway elements from the trachea to the terminal branches in the model. The width of each box demonstrates the flow rate which confirms that the flow in a parent branch is equivalent to the summation of two descending branch flows. The colour of the boxes shows the ratio of the local airway flow to the tracheal flow. The number of terminal branches increases from left to the right. We marked only multiples of five in the plots. This figure illustrates the biased flows in the trachea to more likely enter the right main bronchus. It also shows the influence of the upper airway on distal airway flows. There are similarities in the flow distribution between the LF models (Fig. 9 (a) and (b)), and HF models (Fig. 9 (c) and (d)) which are geometrically identical despite their difference in flow scales due to the boundary conditions. However, the effect of upper airway increases the flows into the RUL and LLL in the CA models. It subsequently affects the distal airway flows. Figure 10 clearly illustrates the flow rates at the bronchial elements in the LA and CA models. The data demonstrates the similarity in the proximal airway flows (Gen 3-4) between

LA and CA models gradually diminishes as the flow propagates to the middle (Gen 5-6) and the distal airways (Gen 7-10).

4. DISCUSSION

The measurement uncertainty in standard MR imaging is usually defined by the image signal to noise ratio. In the flow imaging, however, the quantification of uncertainty is more complex since the velocity is encoded in the phase of the signal and not its magnitude. Few experimental parameters can influence and minimized phase measurement errors such as signal magnitude but not only. The error is also directly proportional to the pixel velocity and inversely proportional to the choice of maximum velocity encoding value. In the 3D measurement of the lower flow model, for example, we chose to repeat the experiment twice with two different maximum velocity encoding value: 30 cm/s to avoid phase wrapping in the upper airways and 10 cm/s to improve velocity to noise ratio in the regions with low velocities. In-vitro experiments like the one we are reporting in this work offer the possibility to repeat several times the measurement and optimize signal and velocity to noise ratio through careful choice of sequence parameters. The resulting velocity errors are therefore usually small and can be neglected compared to hardware related source of error. The latter consists of eddy currents, Maxwell terms and gradient field inhomogeneities [41] that are strongly dependent on scanner model hardware due to the difference in gradient coil design, maximum gradient strength and slew rate. These phase measurement errors are dominant, but they are hard to assess and requires dedicated experiment to correct for them [42]. As an example of the order of velocity uncertainty to be expected in our measurement, Giese et al.

[42] quote an RMSE of ~2.5 % in a system similar to our (3T Philips Achieva system). The phase error will, however, be dependent on object geometry with minimum inaccuracies in the iso-centre of the magnet (corresponding to the centre of the trachea) and increasing errors at the periphery of the lung model.

A curved and irregularly shaped upper airway tract has been known to bring radical alternations in the tracheobronchial flow dynamics. The narrow larynx and resulting jet formation contributed to the biased flow distribution on the cross-sectional planes while the curved airway structure was most likely responsible for the secondary flow generation [43]. The current CFD models and MRI measurements enabled us to confirm the appearance of complex tracheal flow characteristics induced by the upper airway such as the laryngeal jet, biased cross-sectional airway flow distribution, secondary flow, and dynamic velocity profiles which have been illustrated in previous modelling studies [24, 26, 28, 34]. Those complex flow behaviours are highly dependent on the airway geometry as well as the flow rate. Lin et al. [44] showed the laryngeal jet biased towards posterior wall of their realistic subject-specific upper respiratory tract model. By contrast, it was biased towards to anterior wall in other studies [45, 46] as we found in both CFD and experimental analyses of our CA models. The disturbed flow patterns in the upstream trachea appeared in both LF and HF models, but the behaviours of the downstream tracheal flows were different from each other. The laryngeal jets in the CA-HF models were strong and highly unstable compared to the weak and steady jets in the CA-LF models and it affected to the subsequent flows in the main bronchi.

In an earlier study, Luo and Liu [47] extended and modified the trachea geometry to have a biased tracheal flow velocity profile to simulate the effect of a laryngeal jet. In their conclusions, the flow ratio to the left and right lobes was insensitive to the *Re*. We partly agree with those conclusions. As it was previously discussed, the flows at downstream trachea in the CA-HF models were highly unstable and subsequent flow ratios to the right lobes were variable. Moreover, the time-averaged flow ratios in the high *Re* (CA-HF) and low *Re* (CA-LF) models were different. These results illustrated the tracheal flow split ratio could be sensitive to the *Re*. We presume that some of the key flow characteristics such as secondary flows and vortices at the tracheal inlet could hardly be included in their model with the modified geometries. If so, the simplified inlet flow boundary assumptions could result in the different flow behaviours compared to the models with realistic upper airway geometry in the current study.

We showed that the tracheal flow ratios to the main bronchi are different between the LA-HF and CA-HF models. Relatively, the difference was minimal for the LF models as the effect of upper airway provoking flow perturbations almost diminished at the downstream trachea. However, even though they were small in the upstream airways (G3-4), the flow rate discordances between LA and CA models increased rapidly as the flow moved towards the tributary. Consequently, the flow rates in those models became significant in the distal branches (G7-10). These results demonstrated a nature of cumulative error reproduction in an assessment of ventilation distribution within an airway network system due to its branching structure. Thus, minor alterations at the airway boundary of a model could result in notable distortions in tributary flows regardless of breathing conditions.

Complex flow patterns and contorted air pathways incited by an upper airway tract could extend the travel distance and the residence time of the inspired particles. Accordingly, the particles could have more chance to impact the airway wall. Multiple studies have shown the relationship between the *Re* and particle deposition [48, 49]. Luo and Lin demonstrated the significantly increased amount of particle deposition in the turbulent flow models compared to the laminar flow models [47]. Lambert et al. found that the particle-laden turbulent laryngeal jet caused a disproportion of particle split to go into the primary bronchi and succeeding branches [29]. The particle transport and deposition models were not included in the current study. Nonetheless, we confirmed the turbulent flow characteristics in the upstream trachea and unsteady flow behaviours in the main bronchi of the CA-HF model. In addition, as it was aforementioned, the upper airway still affected the flow distribution in distal airways despite the inspiratory flow rate was low. These results allow us to deduce that the upper airway directly makes an impact on the particle deposition on the trachea and main bronchi if the flow rate is high, while it extensively affects the particle distribution in small airways when the flow rate is low.

We assumed the airway walls in the CFD models were rigid. It was a fair assumption to model the surface of the experimental phantom but different from the physiological airway wall. In spite of the difference, the interaction of airflow and compliant airway wall was considered negligible so that the rigid wall assumption has been widely accepted in previous CFD studies to model the airway flows. Furthermore, we decided that the small-scale peripheral airways and acini models were beyond the scope of the current study. As a consequence, the pressures at the outlet boundaries were simplified to be uniformly constant without the influence from the peripheral airway flows and compliant acinar dynamics. Again, those boundary assumptions were similar to the experimental conditions for the MRI flow measurements but could limitedly represent the variant flow conditions in the patient-specific airway geometry. One of the options to improve the outlet boundary assumptions in the current models could be using a full-scale conducting airway model [31, 50]. It could allow assessing the upper airway effects on the flows in an extended range of branches and possibly on a whole lung ventilation. Further studies need to be performed to clarify those.

5. CONCLUSIONS

The effects of upper airways on tracheal flow dynamics in a patient-specific airway model are assessed using MRI flow measurement and CFD analysis. The experimental MRI measurements and CFD simulation results show good agreement with each other. The upper airway morphology and the laryngeal constriction enhance the turbulent kinetic energy in the upstream tracheal flows. The impact is weakened in the downstream branches but the flows in the distal trachea are still complex and unsteady when the flow rate is high. The small difference in the main bronchial flows between the LA and CA models increased in the distal bronchi. The results suggest that the influence of upper airways on the flows in the trachea and tributaries may significantly affect the ventilation distribution of a lung. We conclude that flow in the upper airways needs to be borne in mind when performing CFD analyses of ventilation distribution as well as particle deposition in the airways of the lungs.

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TABLES

Table 1. Locations of cross-sectional planes.

Cross section ID	Location
S0	Larynx opening
S1	After the larynx opening
S2	Beginning of trachea
S3	Middle of trachea
S4	A few centimetres above carina
S5	Carina bifurcation
S6	Right main bronchus
S7	Left main bronchus
S8	Sagittal middle cut of trachea

* Locations of the tracheal cross sections (S1-4) in the higher-flow (HF) model are 5 *mm* closer to the larynx than those in the lower-flow (LF) model.

Table 2. Flow rates at different locations in the MRI and CFD models.

Location	MRI	CFD
Trachea after uvula	3.16	3.46
Trachea before carina	3.52	3.5
Right main bronchus	1.82	1.94
Left main bronchus	1.68	1.55
Left lower lobe bronchus	0.68	0.59
Truncus intermedius	0.82	0.94
Left upper lobe	0.69	0.64
Left lower lobe	0.94	0.82
Right upper lobe	1.0	0.89
Right middle lobe	0.2	0.16
Right lower lobe 1	0.8	0.77
Right lower lobe 2	0.04	0.03

(Unit: *ml/s*)

Table 3. Changes in the mean and standard deviation of the velocity profile along the trachea

Cross section ID	**Distance from S1 (cm)**	**MRI (cm/s)**	**CFD (cm/s)**
S1	0	21.6 ± 6.1	23.8 ± 4.8
S2	2.2	15.4 ± 3.6	15.5 ± 2.3
S3	4.5	12.4 ± 1.8	11.4 ± 1.6
S4	6.9	11.3 ± 1.3	10.4 ± 1.3

FIGURE LEGENDS

Figure 1. A schematic illustration of the CFD analysis and the MRI flow measurement. Cross-sectional planes to measure flow velocities are indicated in the lower left, MR image based flow measure picture (S1-S8). Larynx is located above the cross-sectional plane, S1.

Figure 2. CFD meshes for the CA model (left and middle: surface meshes, right: interior volume meshes appearing on cross-sectional planes at proximal and distal trachea).

Figure 3. Flow velocity fields on the cross-sectional planes (S1 - S8) in the LF model. Anatomical directions, anterior (A) and posterior (P), are indicated on the first contour plot (S1-MRI).

Figure 4. Comparison of flow rates at different locations (2 trachea, 2 main bronchi, 8 lobar/distal bronchi) in the MRI and CFD models.

Figure 5. Tracheal flow characteristics in the HF model. (a) Time-averaged velocity field, (b) temporal velocity profiles (black solid lines), time averaged velocity profile (red solid lines) and standard deviations (red dashed lines) on the cross-sectional plan.

Figure 6. Kinetic energy spectra of tracheal flows at three different locations (centre of the S2-4) in the HF model. The straight dashed line indicates the Kolmogorov's law (k -5/3). (a) CA-LF model, (b) CA-HF model.

Figure 7. Temporal flow velocity fields on the coronal planes in the CFD models at two different instants of time (t1 and t2). (a) LA model, (b) CA model.

Figure 8. Ratio of the tracheal flow into the right main bronchus ($Q_{right}/Q_{trachea}$).

Figure 9. Time-averaged flow distribution in the LA and HF CFD models. (a) LA-LF, (b) CA-LF, (c) LA-HF, (d) CA-HF model. Terminal branch numbers of the boxes in the flow distribution plots increase from the left to the right as the circled numbers indicate. The corresponding branches, as well as a full set of terminal branch numbers, are shown in the right-hand side tree picture. The colour of each box indicates the partitioning to the tracheal flow.

Figure 10. Comparison of the time averaged flow rates in the proximal (Generation 3-4), middle (5-6) and distal (7-10) bronchi between the LA and CA models. (a) LF model, (b) HF model.