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Fluid dynamics of gas exchange in high frequency oscillatory ventilation: *in vitro* investigations in idealised and anatomically realistic airway bifurcation models

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Abstract

The objective of this work is to develop understanding of the local fluid dynamic mechanisms that underpin gas exchange in High-Frequency Oscillatory Ventilation (HFOV). The flow field during HFOV was investigated experimentally using particle image velocimetry (PIV) in idealised and realistic models of a single bifurcation. Results show that inspiratory and expiratory fluid streams coexist in the airway at flow reversal, and mixing between them is enhanced by secondary flow and by vortices associated with shear layers. Unsteady flow separation and recirculation occurs in both geometries. The magnitude of secondary flow is greater in the realistic model than in the idealised model, and the structure of secondary flow is quite different. However, other flow structures are qualitatively similar.

Keywords

respiratory flow, mechanical ventilation, lung, particle image velocimetry

Introduction

The aim of this study is to develop an improved understanding of the fluid dynamics of High Frequency Oscillatory Ventilation (HFOV). HFOV is a shallow, fast mode of mechanical ventilation which can avoid the risk of lung injury associated with cyclic opening and closing of alveolar units in conventional ventilation.[16] HFOV has been successfully applied to neonates since 1981,[18] and more recently to adolescents[20] and adults.[2,7,19] HFOV results in better pulmonary outcomes than conventional mechanical ventilation for very low birth weight infants suffering with Respiratory Distress Syndrome.[27] In adults suffering from Acute Respiratory Distress Syndrome (ARDS), Derdak *et al.*[7] showed that HFOV is at least as effective as conventional ventilation strategy. However, HFOV remains a relatively new and incompletely understood mode of ventilation. An improved fundamental understanding of gas exchange in HFOV is essential in order to harness its potential to improve outcome in patients with ARDS.

Breathing frequencies from 60 to 2400 breaths per minute with tidal volumes of 35–150 ml have been used in HFOV,[16] whereas in normal breathing, the typical rate is 12 breaths per minute with a tidal volume of 500 ml. The anatomical dead space (the volume of the conducting airways) is approximately 150 ml. In normal breathing, therefore, 350 ml of the inspired 500 ml is transported to the alveoli for gas exchange. In HFOV, however, since the tidal volume is usually less than the anatomical dead space, gas exchange cannot be explained in terms of simple advective bulk transport to the alveoli.

More detailed knowledge of HFOV fluid dynamics is essential in order to develop HFOV systems and strategies that optimise gas exchange, and potentially improve outcome from Acute Lung Injury (ALI) and ARDS. It is apparent from previous research that Taylor dispersion and secondary flow, particularly at flow reversal, have a significant

impact on dispersion, and hence on the effectiveness of gas exchange in HFOV. Collectively, these can be classified as convective phenomena which act locally (within an individual bifurcation) to enhance axial mixing (that is, mixing between O_2-rich gas in upper regions of the airway and CO_2-rich gas in lower regions). The purpose of this study is to develop a better understanding of these fluid dynamic mechanisms and their interactions in HFOV.

This paper describes an investigation of the fluid dynamics of HFOV at the level of a single airway bifurcation. The goal is to identify and understand localised fluid dynamic mechanisms which augment bulk transport by promoting mixing in the axial direction. PIV and stereoscopic particle image velocimetry (SPIV) were used to measure the flow field quantitatively in *in vitro* models of a single bifurcation at flow conditions representative of HFOV in the upper airway. Models were scaled up to enhance the effective spatial and temporal resolution of the measurements. To identify and differentiate flow phenomena with the greatest possible clarity, experiments were carried out first in an idealised geometry. Measurements were also made in a realistic geometry, in order to investigate the effects of non-ideal geometry on flow structures.

Methods

The idealised and realistic geometry models are shown in Fig. 1. The idealised model consists of a single symmetric bifurcation with the geometry defined by Comer *et al.,*[6] based on the morphological work of Weibel.[34] The flow divider is rounded with a radius of 0.1 times the diameter of the daughter branch. The realistic model is based on images of a third-generation airway bifurcation from the inferior lobar bronchus of the left lung. The images were acquired from the female data set of the Visible Human Project.®[3]

Images were manipulated with image processing and surface modelling software (Mimics® and Rhinoceros® respectively) to generate a three-dimensional geometric description of a third-generation lung bifurcation. The parent and daughter branches have non-circular cross-sections, and their centrelines are not coplanar and have slight three-dimensional curvature. The right and left daughter branches divide at 24° and 20° to the parent branch centreline, respectively. Extension sections have been added to all inflow and outflow ports of the realistic geometry. In the extension sections, there is a smooth transition from the realistic anatomical cross-section of the airway branch to a circular cross-section, allowing circular-section tubes to be coupled to the physical model for experimental testing. Using the surface modelling software, continuity was enforced across surface boundaries to avoid any steps, sharp edges, or curvature discontinuities.

To improve the spatial and temporal resolution of the experiment, the geometry was scaled up by a factor of 5.6. Thus, the parent branch diameter scaled from 5.3 mm in the third-generation lung airway to 30 mm in the experimental model. To ensure validity of the scaling by complete dynamic similarity, the Reynolds number *Re* (dimensionless velocity) and Womersley number α (dimensionless frequency) must be preserved in the scale models. This was enforced with a 59/41% (by volume) glycerol/water mixture as the working fluid along with reductions in flow velocity and oscillation frequency. The frequency reduction enhances the effective temporal resolution of the experiment by a factor of 56.2. The composition of the glycerol-water mixture is also influenced by optical considerations, as discussed below.

Tests were carried out for two Reynolds numbers, Re = 740 and 1480, and two Womersley numbers, α = 4.3 and 7.0. The Womersley number $\alpha = D/2\ (2\pi f/\nu)^{1/2}$

characterises the unsteadiness of oscillatory flows, where $D = 2a$ is the airway diameter, f is the frequency of oscillation of the flow, and ν is the kinematic viscosity. For $\alpha << 1$ the flow is quasi-steady, but for high α unsteady effects are important. In terms of clinical parameters, frequency is proportional to α^2 in a given geometry. Throughout this article, the Reynolds number $Re = UD/\nu$ is based on the mean velocity in the parent branch at peak inspiration, denoted by U. At the level of the third-generation airway, $Re = 1480$ and $\alpha = 4.3$ corresponds to a tidal volume of 47 ml through the trachea at 377 breaths/min, representative of a HFOV condition. The test cases may also be characterised in terms of a dimensionless flow amplitude, L/a, where L is the mean displacement of a fluid particle over one cycle based on bulk transport, and a is the tube radius. The three dimensionless parameters are not independent, as $Re = \alpha^2 L/a$. The four test cases in this study are summarised in Table 1. All four test conditions fall within the flow regime dominated by the effects of convective acceleration, as defined by Jan *et al.*[14]

Rigid experimental models were made in Sylgard™ 184 (Dow Corning®) transparent silicone by a lost core technique described by Kelly *et al.*[15] The geometry of the experimental models was validated by comparing Computed Topography (CT) scans of the silicone models to the original CAD geometry models. Measurements were made at 118 cross-sections in the idealised and realistic models respectively, and it was found that the average cross-sectional area in the experimental models was 3% greater than in the original CAD geometry models.

The flow apparatus, shown in Fig. 2, supplied a controllable time-varying flow rate through the experimental model, while allowing unobstructed optical access for both two-component PIV and SPIV. A sinusoidal flow rate was applied to the experimental models

by a custom-built computer controlled piston pump, with a maximum stroke length of 1 m and a maximum stroke volume of 15.4 l. The pump was driven by a position-controlled stepper motor (Sanyo Denki Co. Ltd., 103-H8223-6540), which was controlled by a computer with a custom-designed LabVIEW™ 7 programme.

Straight tubes were attached at all inflow and outflow ports of the experimental models to minimise entrance and exit effects. Gerrard and Hughes[11] found experimentally that oscillatory flow is fully developed at a distance of $(0.08)a^2u_0/\nu$ from the tube entrance, where a is the tube radius, u_0 is the mean velocity based on peak flow, and ν is the kinematic viscosity. The entrance tubes used in this study were 30% longer than the required entrance length.

A flow disrupter and a flow straightener were attached to each straight tube section as shown in Fig. 2, at the opposite end from the bifurcation model, to condition the intake flow from the pump or reservoir. Each flow disrupter consists of a short tube full of glass spheres. Their function is to provide a uniform axial velocity profile. Each flow straightener consists of a bundle of parallel small-bore thin-walled tubes, intended to remove any radial and swirl velocity components. Flow rates through the daughter branches of the idealised geometry were measured in steady expiratory flow using PIV, and were matched to within 1% by adjusting the gate valves shown in Fig 2.

The refractive index of the working fluid and the Sylgard 184 silicone was measured at 20°C using an Abbe refractometer with sodium D light, and was found to be 1.411 for both materials. Chrome-coated hollow glass tracer particles of diameter 10 μm were added to the fluid mixture.

Instantaneous full field measurements were made using two-component PIV and three-component SPIV at different locations. The plane of measurement was illuminated with Nd:YAG lasers (New Wave Research™, Gemini PIV Laser). Images were captured using a 1000×1016 pixel CCD camera (TSI, PIVCAM™ 10-30) with 8-bit digitisation.

Measurement was triggered at the start of inspiration for each cycle by an electronic switch mounted on the frame of the piston pump. All PIV images were processed using 32 × 32 pixel interrogation windows, and each interrogation window overlapped neighbouring interrogation windows by 50%. A 32 × 32 pixel interrogation window corresponds to a 1.7 × 1.7 mm region of the measurement plane. SPIV images were processed using 16 × 16 pixel interrogation windows with 50% overlap. Each 16 × 16 pixel interrogation window measured 1.5 × 1.5 mm. The velocity field was sampled at a rate of 7.5 Hz for tests at $\alpha = 4.3$ and at 15 Hz for $\alpha = 7.0$. According to the Nyquist theorem, this would allow measurement of fluctuations up to 3.25 and 7.5 Hz, respectively, or up to 29 and 27 times the base cycle frequency. Particle images were processed by two-frame cross-correlation using Insight 3G™ 8.0.5 software.[1] Using the data of Raffel *et al.*,[25] measurement uncertainty for this PIV configuration was estimated as 0.007 m/s, or 1.6 % of the peak flow velocity.

The two-component PIV measurements were calibrated using dimensions measured in CT images of the experimental models. The SPIV measurements were calibrated by placing a three-dimensional calibration target in the fluid-filled model at the measurement location. A correction procedure was implemented for all SPIV measurements using Insight 3G™ 8.0.5 software[1] to reduce error due to misalignment between the calibration target and the measurement plane.

The locations of the measurement regions for the idealised and realistic geometries are illustrated in Fig. 1. In both experimental models, all two-component PIV measurements were acquired on a plane passing through the centres of the three inlet/outlet ports. This is defined as the *xz* plane in the Cartesian coordinate system shown in Fig. 1. This coordinate system is used throughout this article to indicate the relative orientation of measurement planes. For the idealised geometry, the *z*-axis corresponds exactly with the centreline of the parent branch. The outlines of the idealised and realistic models, shown in Fig. 1, are the intersections of these *xz* measurement planes with the silicone experimental models. Seven overlapping measurement regions, measuring 58 mm × 58 mm, were used in each geometry model to map the velocity field on the *xz* plane with two-component PIV. However, only those field measurements near the flow divider will be shown here. SPIV measurements were acquired on a number of *xy* planes (i.e. transverse sections) through the parent branch of each model, and results will be presented here for the measurement plane nearest the flow divider. PIV and SPIV measurements were captured at similar times, allowing PIV and SPIV results to be considered in association. All results are non-dimensionalised with respect to U, the mean velocity in the parent branch at peak inspiration.

The velocity fields presented are phase-averaged over 6 to 10 cycles. The cycle-to-cycle variation in the mean velocity on line A at peak flow was 2% of the average (line A is a line parallel to the *x* axis, passing through the centreline of the parent branch, as shown in Fig. 1).

Results

Detailed velocity measurements are presented in Figs. 3-6 for L/a = 15, Re = 740 and α = 7.0. Qualitatively similar flow structures were observed for all test cases, and these results may be considered representative. Fig. 3 and Fig. 4 illustrate the phase-averaged flow field in the idealised and realistic bifurcation geometries, at dimensionless times τ = 0.057, 0.115, 0.452 and 0.623. τ is dimensionless time t/T, where t = 0 is at the start of inspiration, and T is the period. In the two-component PIV visualisations (Fig. 3), vectors represent the magnitude and direction of velocity. In the SPIV measurements (Fig. 4), the colour map indicates the axial component (z-direction) of velocity (i.e. the component parallel to the nominal centreline of the parent branch) and the vectors indicate the magnitude and direction of the in-plane secondary flow.

Fig. 3 illustrates the two-component velocity field measurements in the idealised and realistic bifurcation models. In both models at the beginning of inspiration (τ = 0.057), the fluid at the walls reverses and begins to flow in the inspiration direction, while the faster moving flow in the core of the airway continues to flow in the expiration direction, as in classical Womersley flow.[35] As a region of inspiratory flow develops near the walls, the shear layer between the opposing flows moves outwards from the wall. Figure 4 illustrates the three-component SPIV measurements. The shear layers are also evident in the stereoscopic measurement on the parent branch cross-section. Throughout Fig. 3 and Fig. 4, the location of the shear layer is indicated approximately by the contour of zero z velocity. A secondary flow consisting of four longitudinal vortices is present in the parent branch of the idealised model during expiration, as illustrated in the stereoscopic measurement (Fig. 4). These vortices are associated with the curvature of the daughter branches as they merge into the parent branch. Similar Dean-type vortical structures have

been observed by other authors[9,14] during the expiration phase in idealised bifurcation models. In the stereoscopic view of the realistic model (Fig. 4) a single axial vortex is observed at the low y side of the cross-section, and is superimposed on the inspiratory and expiratory fluid streams.

At $\tau = 0.115$ in the idealised geometry PIV measurement, the shear layers are observed closer to the core of the parent and daughter branches. The secondary flow vortical structures, observed at $\tau = 0.057$, are still present in the parent branch, but the vortices have moved closer to the core of the branch. It is apparent from the stereoscopic measurement that these vortices span the counterflowing inspiratory and expiratory streams. The stereoscopic measurement (Fig. 4) also indicates that the axial flow in the cross-section is not axisymmetric at this location in the parent branch. At $\tau = 0.115$, inspiratory axial flow is observed first at the high and low x walls, but in the majority of the cross-section, flow is in the negative z (expiratory) direction.

In the realistic model at $\tau = 0.115$, the shear layers are still present in the parent and daughter branches, but their structure has changed from that observed at $\tau = 0.057$. There is now a single shear layer in both the parent and left daughter branches, while a separate shear layer is observed in the right daughter branch. The corresponding stereoscopic measurement indicates that the shear layer in the parent branch is closer to the core of the branch than was observed at $\tau = 0.057$. An axial vortex structure is still present.

At peak inspiration in the idealised model at $\tau = 0.250$, all flow is in the direction of inspiration, and the structure is essentially quasi-steady. (The quasi-steady flow structure has been described in previous studies[17] and is not illustrated here.) At $\tau = 0.452$, when the flow is decelerating from peak inspiration, flow separates upstream of the bifurcation,

and recirculation regions are created. However, the majority of the flow in the daughter branch is still in the inspiration direction. At τ = 0.623, shortly after the mean flow reverses from inspiration to expiration, the slower-moving fluid at the walls of the parent branch and at the outer walls of the daughter branches is the first to change to the expiration direction. The faster-moving fluid in the core of the parent branch and at the inside walls of the daughter branches is still moving in the inspiration direction. Shear layers exist between these opposing fluid streams. The stereoscopic measurement in the parent branch (Fig. 4) confirms that the two shear layers visible in the two-component PIV measurement (Fig. 3) are part of a single cylindrical shear layer structure. The structure of the shear layers in both models near flow reversal is visualised in Fig. 5 using instantaneous streamlines, which were computed using Tecplot™ 10 and initiated at points distributed throughout the bifurcation. In both geometry models, vortical structures are observed within the shear layers around both flow reversal times.

In the realistic model as the cycle reaches peak inspiration at τ = 0.25, the flow structure resembles quasi-steady flow. A small recirculation region forms on the inside wall of the bifurcation in the right branch. This recirculation region is caused by the local curvature at the flow divider, as shown in Fig. 3 at τ = 0.452. As the flow changes direction at the start of the expiration phase at τ = 0.623, the slower moving flow along the outer walls changes direction first, whereas the flow along the core of the parent and daughter branches is still moving in the inspiration direction. In the parent branch the shear layer is visible between the last of the inspiratory flow at the core of the branch and the expiratory flow at the perimeter.

The temporal variation of the mean magnitude of axial and transverse velocity at various section lines in the *xz* measurement plane is shown for both models in Fig. 6. The magnitude of transverse velocity is an indication of the level of secondary flow. At section 1W (in the parent branch of the idealised geometry), axial velocity dominates for most of the HFOV cycle, except at times of flow reversal. The magnitude of transverse flow increases just after peak expiration, but remains small compared to the axial velocity. Closer to the bifurcation, at section 2W, the transverse velocity is 2% of the axial velocity at inspiration, and increases to 10% during the expiration phase. In the left daughter branch, at section 3W and 4W, the peak transverse velocity occurs during the inspiration phase. Transverse velocity is higher than axial velocity during flow reversal at the start of the expiration phase.

In the parent branch of the realistic geometry, at section 1R, the transverse velocity is 25% of axial velocity at peak inspiration and expiration. Closer to the bifurcation, at section 2R, the transverse velocity is 30% of the axial velocity at peak flow. Transverse velocity is greater than axial velocity in the parent branch at section 2R during periods of flow reversal. The transverse velocity measured in the right and left daughter branches is significantly different. In the left daughter branch at section 4R transverse velocity reached 59% of the axial velocity at peak flow, whereas only minimal levels are recorded in the right daughter branch at section 3R.

Qualitatively similar flow structures are observed at all 4 conditions tested ($15 \leq L/a \leq 80$, Re = 740, 1480, and α = 4.3, 7.0). Shear layers are observed at all conditions. The duration of this counterflow increases as the flow becomes more strongly unsteady, from 14% of the cycle period at α = 4.3 to 22% at α = 7.0. At peak flow times, the velocity

profile in the parent branch is flatter for $\alpha = 7.0$. Similar levels of transverse velocity are measured at all stations in both bifurcation models for $\alpha = 4.3$ and 7.0. For the two Reynolds numbers tested, the shear layers are present in the bifurcation for the same duration, but thinner shear layers with higher velocity gradients are observed at the higher Reynolds number.

Fig. 7(a) shows the fluctuation of axial velocity over multiple cycles at Point A (Fig. 1) on the centreline of the parent branch in the idealised bifurcation model. At peak expiration (negative velocity), a high-frequency fluctuation can be observed. The amplitude of this velocity fluctuation is higher closer to flow divider. Fig. 7(b) shows the phase-averaged velocity over 10 cycles.

Discussion

The aim of this study is to enhance understanding of local convective flow mechanisms in a respiratory bifurcation during HFOV. Several previous authors have reported pointwise velocity measurements in *in vitro* airway models using Laser Doppler Velocimetry (LDV).[17,24] Particle Image Velocimetry (PIV), as a whole-field technique, can potentially give clearer insight into the spatial structure of the flow. Ramuzat and Riethmuller[26] adopted PIV to study steady and oscillatory flows in airway models, and reported steady streaming effects. Many previous studies of fluid dynamics in HFOV have been based on idealised models of the airways, neglecting realistic geometric features such as curvature, asymmetric branching, and non-circular cross-sections. However, the computational work of Nowak *et al.*[22] identified significantly different flow patterns in idealised and anatomically realistic airways under normal breathing

conditions. Große *et al.*[12] highlighted the influence of geometric asymmetry of the realistic lung on normal respiratory airflow.

In the present work, two-component and stereoscopic (three-component) PIV were used in both idealised and realistic bifurcation models. An idealised geometry model was used to facilitate clear identification of the generic fluid dynamic phenomena in the mixing of inspired and expired air. An anatomically realistic model was used to explore the additional effects of realistic geometric features such as local curvature, non-uniform non-circular cross-sections, and asymmetric branching angles.

The most important observed flow phenomena are sketched in Fig. 8 for four key points in the cycle. The main features of the flow are common to both the idealised and realistic models and to all four flow conditions modelled. Quasi-steady flow exists only at times of peak flow, as other authors have also observed.[17]

Counterflow at flow reversal

One of the significant findings of this study is that inspiratory and expiratory fluid streams coexist in the airways for significant periods of time of flow reversal, as a result of the unsteadiness of the HFOV cycle. At $\alpha = 7.0$, counterflow and shear layers are present in both models for longer than in the less strongly unsteady case, $\alpha = 4.3$. At flow reversals in both models, shear layers appear between the inspiratory and expiratory streams. In the realistic model, these shear layers persist for slightly longer in the wider left daughter branch than in the right daughter branch. Vortical structures are apparent as a result of roll-up of shear layers in both models, illustrated by instantaneous streamlines shown in Fig. 5 and in the schematic diagram of Fig. 8.

Secondary Flow

The axial curvature of the idealised bifurcation results in a velocity profile skewed towards the inside wall of the flow divider and the generation of secondary flow structures in the daughter branches during inspiration. Lieber *et al.*[17] and Peattie and Schwarz[24] observed similar flow structures in idealised bifurcation geometries. Eckmann and Grotberg[8] observed a skewed velocity profile in a curved tube. Secondary flow longitudinal vortical structures are observed throughout the expiration phase and at flow reversal. Jan *et al.*[14] visualised a similar flow structure during the decelerating phase of expiration for $\alpha = 21.3$ and $L/a = 20$. More recently Fresconi *et al.*[9] observed a similar secondary flow pattern in the parent branch during expiration in a similar idealised bifurcation model. Significant levels of secondary flow are observed in the daughter branches during inspiration and in the parent branch during expiration, as shown in Figure 7 at sections 1W and 2W.

In the realistic geometry, significant levels of secondary flow are observed during both inspiration and expiration in both parent and daughter branches. In the realistic model the parent and daughter branches have slight three-dimensional axial curvature, the daughter branches divide asymmetrically and the walls are non-parallel. The curvature and non-parallel walls of the parent branch result in similar levels of secondary flow in the parent branch during inspiration and expiration, as seen in Figure 7 at section 1R and section 2R. In inspiration, the skewed velocity profile and local wall curvature at the entrance of the right daughter branch cause a recirculation region. While the axial flow structure in the realistic bifurcation is similar to that observed in the idealised bifurcation, the non-

planar walls and slight three-dimensional axial curvature have a significant effect on the levels and structure of secondary flow.

Although axial and secondary flow structures in the idealised and realistic models are very different, the interaction between axial and secondary flow at flow reversals is similar in both models. In the parent branch of the realistic model, a single longitudinal vortex is observed, along with other more complex structure. This contrasts with the system of four vortices observed in the idealised geometry. In both models, the secondary flow vortices span the counterflowing inspiratory and expiratory flows at flow reversal.

Flow separation

Flow separation with recirculation is observed in both the idealised and realistic bifurcation models at different times in the HFOV cycle. Recirculation in the realistic geometry results from separation due to the local curvature of the inside wall of the right daughter branch during peak inspiration. In the idealised model, recirculation occurs on the outside of the bifurcation as the flow decelerates from peak inspiration, as shown in Fig. 4 for $\tau = 0.452$. Schroter and Sudlow[29] studied steady flow in an idealised bifurcation for $100 \leq Re \leq 4500$, and observed that the presence of recirculation regions was a function of the tube radius a of the parent branch and the radius of curvature R_c of the parent and daughter branch centrelines. They observed a region of recirculation in a model with $R_c \cong a$, but for $R_c = 4a$, no recirculation was observed. In the idealised geometry employed here, $R_c = (5.4)a$, and recirculation does not occur in the quasi-steady peak inspiration flow. Recirculation did not occur at any time in low-frequency oscillation experiments ($L/a = 634$, $Re = 740$, $\alpha = 1.08$; results not shown) in either geometry model. These observations are consistent with the findings of Schroter and

Sudlow[29] for steady flow. Thus, the transient flow separation and recirculation observed in the present work appears to be an unsteady effect, dependent on Womersley number.

Instability

There is evidence of instability in the flow, in the form of high-frequency velocity fluctuations which were detected in the parent branch at peak expiration in the idealised geometry, as shown in Figure 7(a). These fluctuations vary from cycle to cycle, and do not appear in the phase-averaged velocity shown in Fig. 7(b). This is consistent with the findings of Peattie and Schwarz,[24] who observed turbulent bursts in the parent branch of an idealised bifurcation at peak expiration.

Similar velocity fluctuations were observed in the realistic model on the parent branch centreline, at both peak inspiration and peak expiration. The magnitude of the fluctuations was highest near the flow divider in expiration, when a recirculation region forms between the fluid streams from the daughter branches. The temporal resolution of the PIV measurements in this study allows frequencies of 26-29 times the cycle frequency fluctuations as described to be resolved. As Peattie and Schwarz[24] detected fluctuations at 10-100 times the ventilation frequency, it is possible that some high-frequency fluctuations were beyond the resolution of the present experiments.

Significance for mass transport

The experimental results presented here describe the HFOV velocity field and flow structures in detail. However, concentration fields and mass transport effects have not been measured directly, and it is not possible to draw firm conclusions about respiratory mass transfer. Here we discuss the possible impact of some of the observed flow features

on mass transport and gas exchange in HFOV, in light of previous research on mass transfer.

It is known that lateral mixing by secondary flows can play an important role in mass transport. Eckmann and Grotberg,[8] Fujioka *et al.*,[10] Nishida *et al.*[21] and Tanaka *et al.*[30,31] have shown that effective diffusion is greater in curved and/or bifurcated tubes than in straight tubes for a range of conditions representative of HFOV, as a result of secondary flow. Tanaka *et al.*[31] also showed that significant transverse mixing occurred in an idealised model airway bifurcation at flow reversal, and found that a pause in the flow waveform around reversal allowed more time for transverse mixing, increasing the effective diffusivity by a factor of two. From these studies it is clear that the level of secondary flow has a significant effect on mass transport. Therefore, the enhanced levels of secondary flow due to axial curvature, asymmetric branching and local wall curvature, measured in the realistic model in this study, could have important implications for mass transport in the lung. The interaction of secondary flow with axially counterflowing streams may be a further mechanism to enhance axial mass transport. However, this hypothesis requires further investigation to test and quantify.

In the upper airways, Taylor-type dispersion is a dominant mechanism.[5,16] Taylor (or shear-augmented) dispersion occurs when a non-uniform velocity profile in the airway causes differential rates of axial convection for fluid at different transverse locations.[32,33] At peak flow, a flattened velocity profile is observed in the parent branch of the idealised and realistic bifurcation models, as shown in Fig. 3. However, in off-peak flow, the velocity profile in the parent branch is less uniform and hence more conducive to Taylor dispersion.

The transient flow separation observed in these experiments may act as a convective mechanism to enhance axial mixing in oscillatory flow. Fluid particles which get trapped in a recirculation region are detained in the airway while the main stream passes by, and released for mixing later, when the flow separation breaks down. Thus, recirculating particles are effectively transferred upstream relative to the bulk motion of the main axial fluid stream. Direct measurement or computation of the concentration field will be required in future work to test and quantify the overall effect of this phenomenon on mass transport, as for the other flow phenomena discussed above.

Clinical application

Acute Lung injury and Acute Respiratory Distress Syndrome (ALI/ARDS) are severe inflammatory diseases of the lung, and have a high morbidity and mortality. In fact, the mortality attributable to ALI in the United States is comparable to that seen with HIV infection, breast cancer, and asthma.[28] There is no cure for ALI/ARDS and the treatment is supportive. A key component of this supportive therapy is the mechanical support of respiration. Conventional mechanical ventilation strategies cause high lung stretch, and can damage lungs and contribute to mortality.[2] This has been a focus of intensive research effort in recent years, and has resulted in the development of 'low stretch' ventilatory strategies, which have been demonstrated to save lives.[2]

HFOV has been widely used in the clinical setting for the management of the critically ill neonate and child. HFOV is increasingly used in adults with severe acutely injured lungs, where it performs comparably to leading edge 'protective' ventilation strategies.[2,7] However, our understanding of HFOV has advanced relatively little since its introduction into clinical practice in the late 1970s[4]. A better understanding of the

mechanisms by which gas exchange occurs in HFOV offers the potential of developing more effective HFOV support therapies, by enhancing gas exchange while minimizing the potential for lung damage. Such novel HFOV strategies could significantly reduce death and disability from these devastating disease processes.

The research presented in this article offers new insights into the fluid dynamics of HFOV, which are a prerequisite for deep understanding of mass transport and gas exchange. In particular, the effect of complex realistic geometry on flow features has been investigated. In future work, these experimental results will be used to guide and validate high-resolution computational models, incorporating O_2 and CO_2 diffusion. Models of this type may be used to explore the design space for HFOV much more efficiently and safely than in pre-clinical and clinical trials, leading to opportunities for optimisation of HFOV tidal volume, frequency and waveform parameters.

Limitations

This study is based on *in vitro* models which have some inherent limitations. The models are rigid, and do not simulate dynamic deformations of the airway during the breathing cycle. However, this is a reasonable approximation for the upper conducting airway, in which deformation is restricted by the presence of cartilage in the walls.[23] A potentially more significant limitation of the present work is that models consist of a single tracheobronchial bifurcation. Previous studies of HFOV in the upper airway have shown that the flow structure in each bifurcation generation is influenced by the flow from higher and lower generations.[30] However, in a single-generation model, clearer insights into local fluid dynamics are possible without the added complexity of multi-generation effects. This fundamental approach was taken in the present work because a

detailed understanding of flow in a single generation is a prerequisite for understanding of HFOV mechanisms in the lung as a whole.

Conclusion

In summary, we have shown that around times of flow reversal, inspiratory and expiratory flows co-exist throughout the bifurcation for substantial periods of the oscillatory cycle, and the duration of this counterflow increases as the oscillation frequency increases. Shear layers form between these counterflowing streams, leading to the formation of vortices, which could enhance mixing between the streams. Secondary flows, due to airway curvature and the flow between parent and daughter branches, also mix the inspiratory and expiratory streams. Unsteady flow separation and recirculation occur in both geometric models, at or just after peak flow, although it appears to be highly geometry-dependent. Secondary flow velocity is significant in comparison with axial velocity at many times throughout the cycle, and greater than it in some cases near reversal. Secondary flow velocities are higher in the realistic geometry than in the idealised geometry, because of wall curvature. In the idealised geometry, the secondary flow in the parent branch resembles classical Dean vortices, but it is more complex in the realistic geometry. Axial velocity profiles are relatively flat at peak flow and most non-uniform at times of off-peak flow.

Understanding of the fluid dynamic mechanisms which underlie HFOV may ultimately contribute to the development of HFOV strategies to achieve more effective gas exchange, and ultimately increase survival in sufferers from ALI/ARDS. This work highlights the importance of transient counterflow, secondary flow and recirculation, and also shows that realistic geometry features have a significant effect.

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Figure 1. Geometrically accurate sectional views of the idealised and realistic experimental models, showing PIV and SPIV measurement locations.

Figure 2. Schematic diagram of the experimental flow apparatus.

Figure 3. Non-dimensionalised two-component PIV measurements in the idealised and realistic bifurcation models at L/a = 15, Re = 740, α = 7.0. Dotted lines represent contours of zero velocity.

Figure 4. Non-dimensionalised SPIV measurements in the idealised and realistic bifurcation models at L/a =15, Re = 740, α = 7.0. Dotted lines are contours of zero axial velocity.

Figure 5. Instantaneous streamlines at both flow reversals in the idealised and realistic models.

Figure 6. Magnitude of non-dimensionalised axial and transverse velocity averaged over various transverse lines (indicated at the top of the figure) in the measurement plane in the parent and daughter branches of (a) idealised and (b) realistic models.

Figure 7. Axial velocity measured at Point A (Fig. 1) on the centreline of the parent branch of the idealised model, (a) over multiple cycles and (b) phase-averaged over 10 cycles.

Figure 8. Sketch of the major flow structures observed in airway bifurcations throughout the cycle.

Table 1

Click here to download Table: Table1 Test Case.doc

Test Case	Re	α	L/a
1	740	4.3	40
2	740	7	15
3	1480	4.3	80
4	1480	7	30

Table 1. Summary of test conditions.

Figure 1

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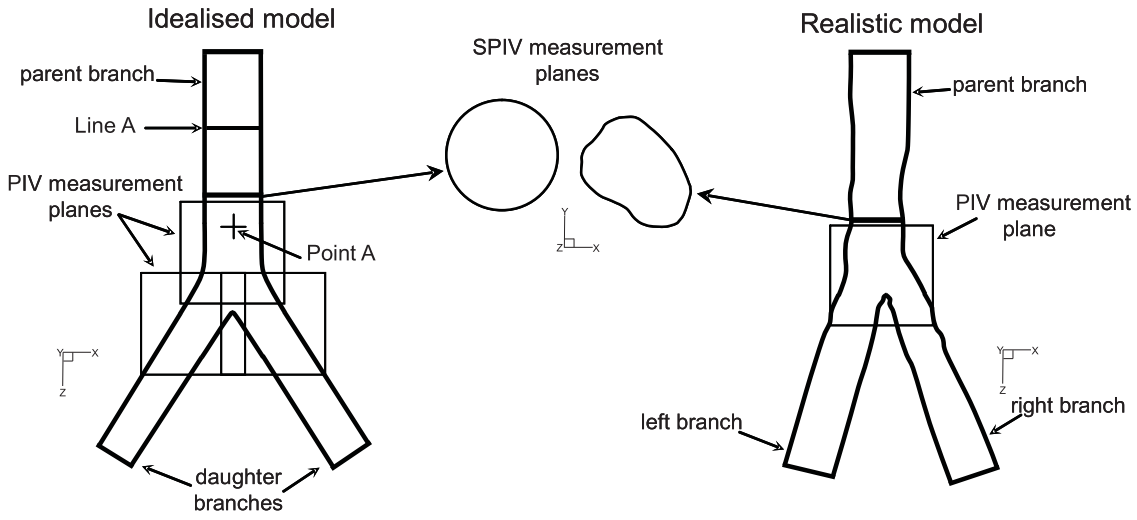

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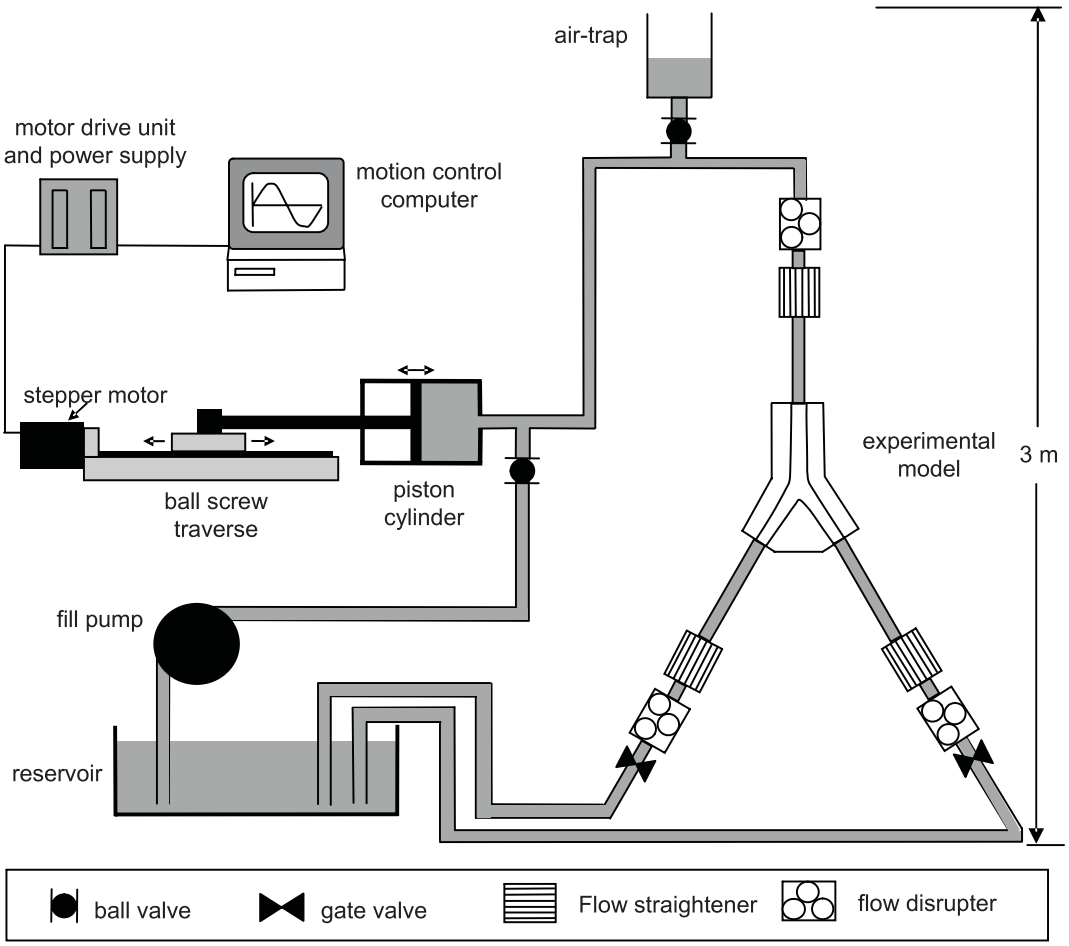

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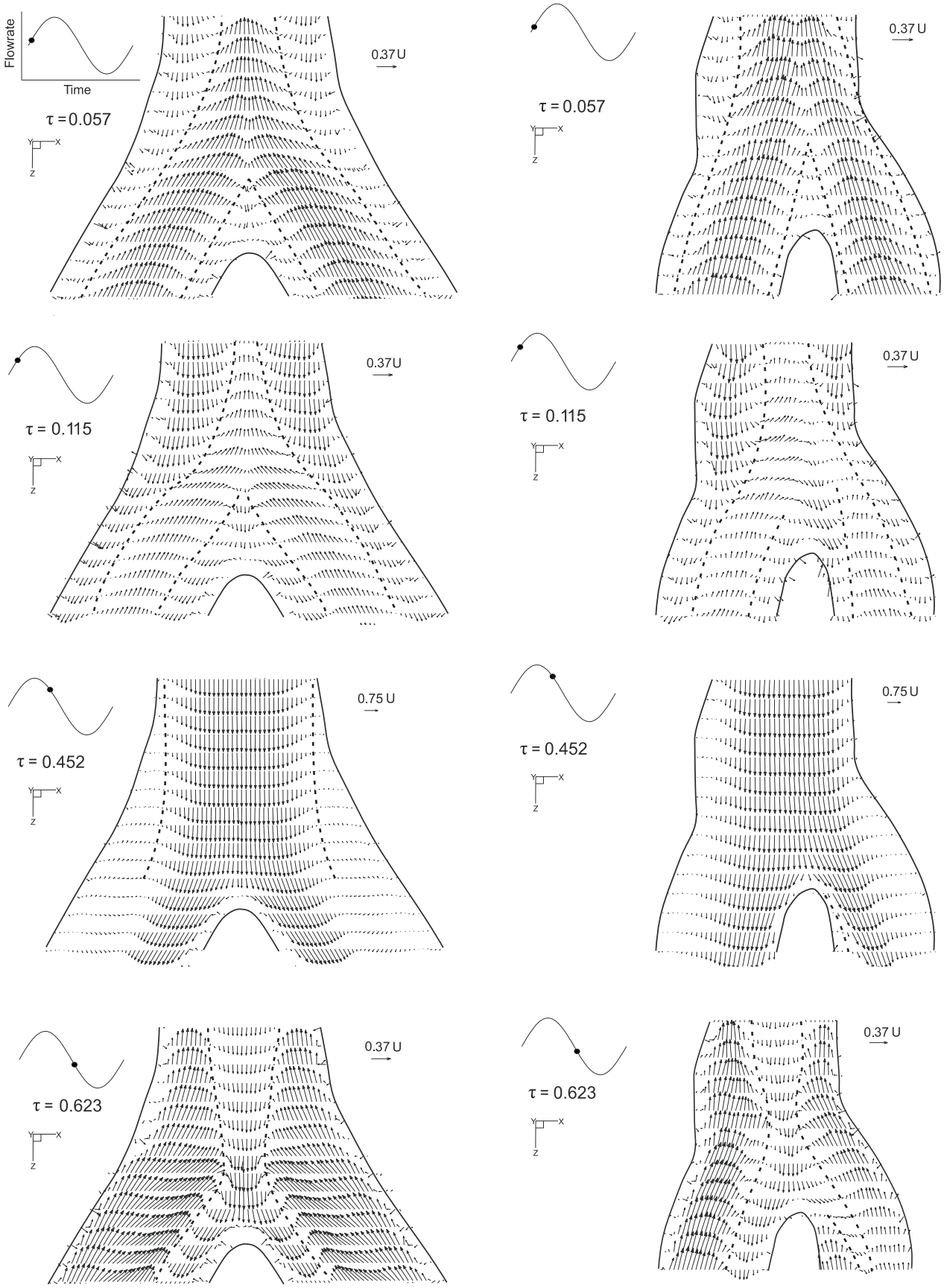

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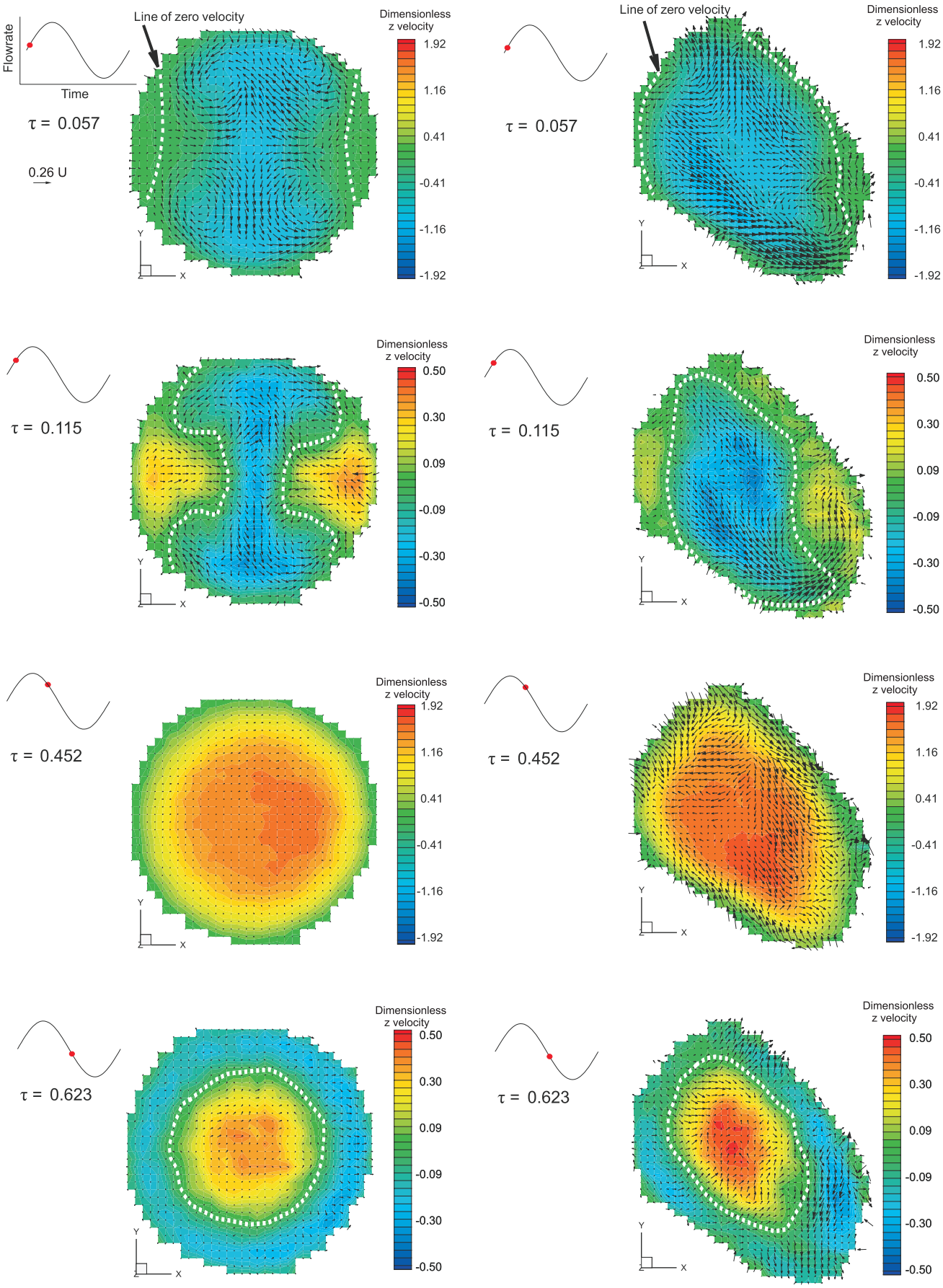

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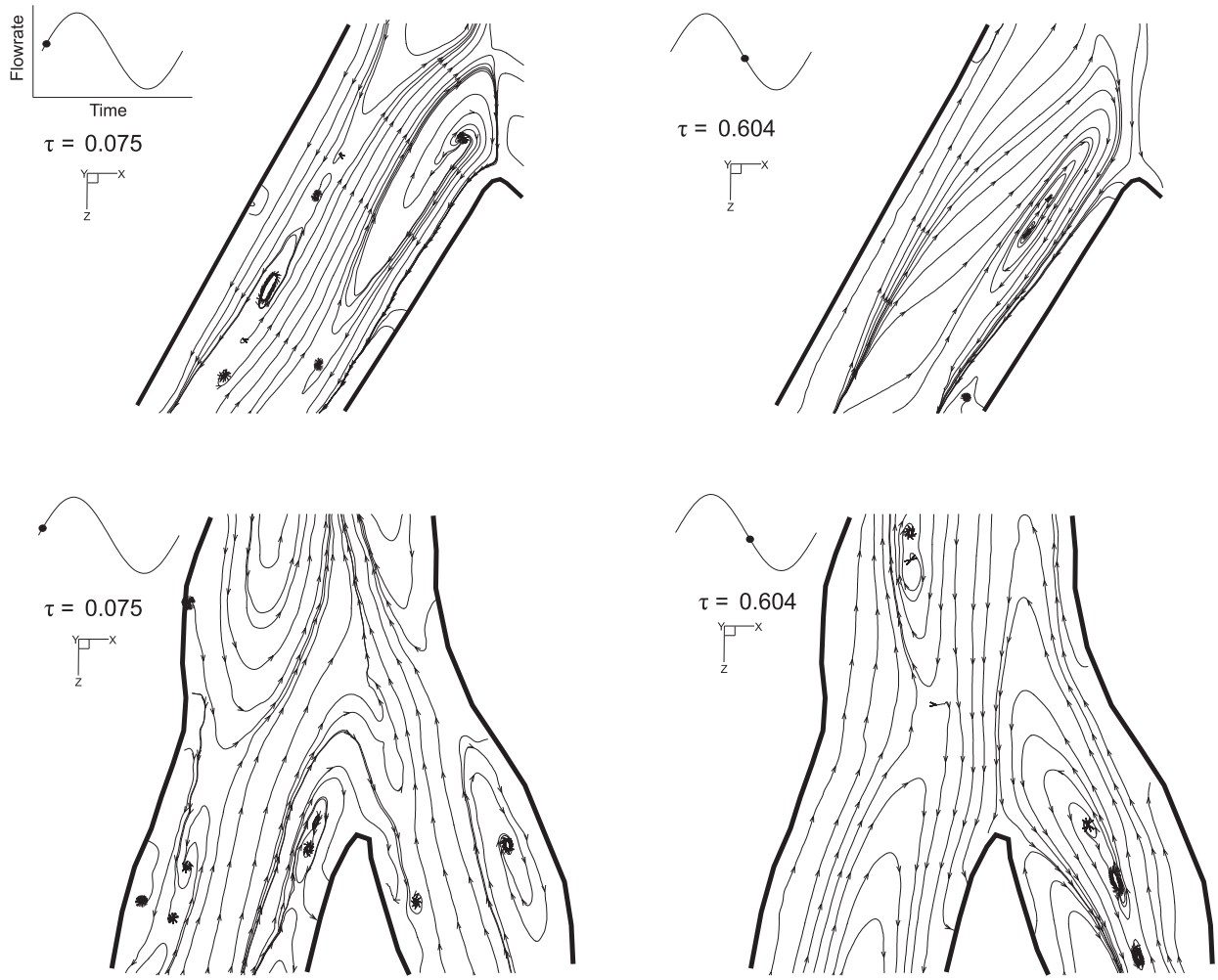

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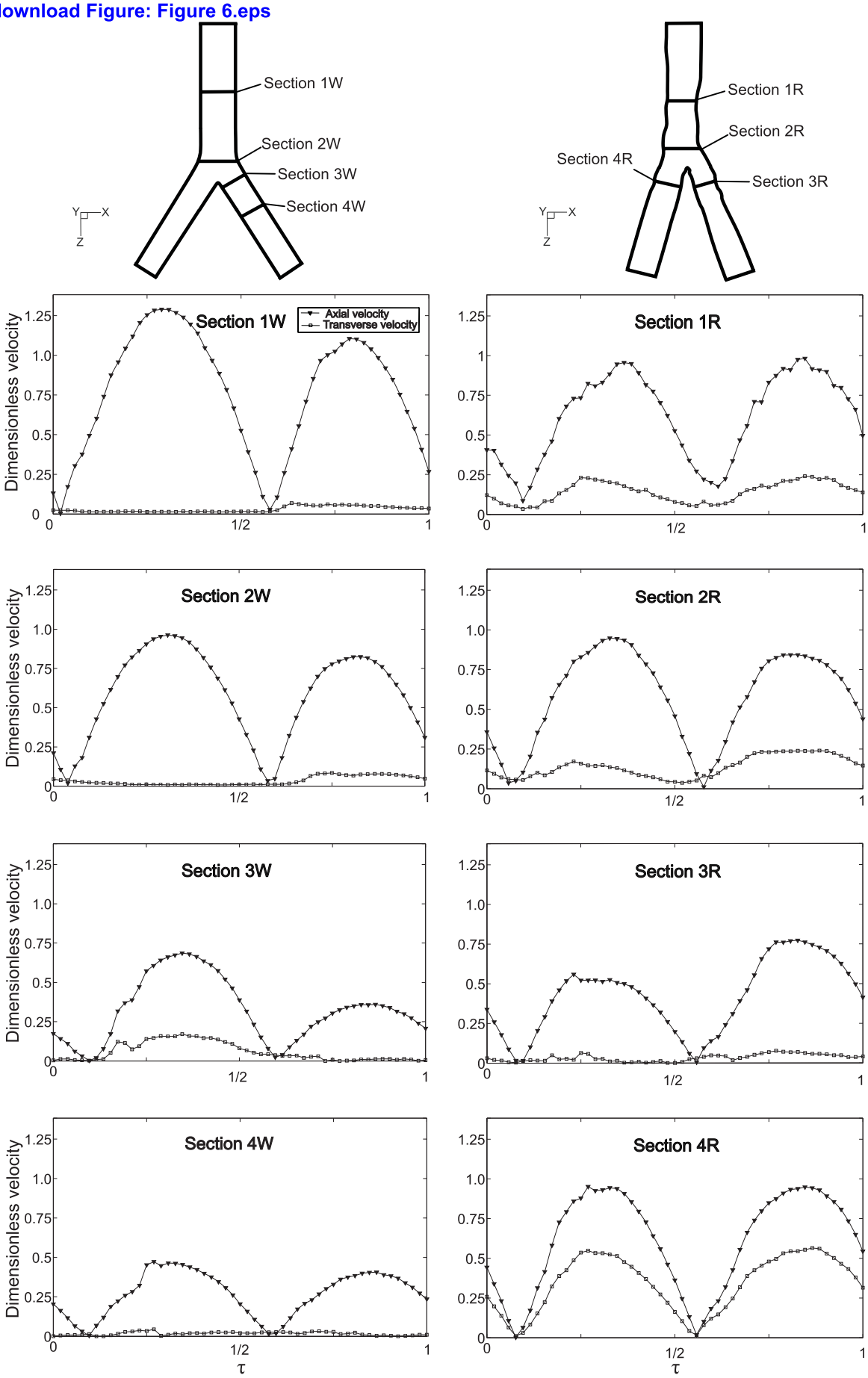

Figure 7a

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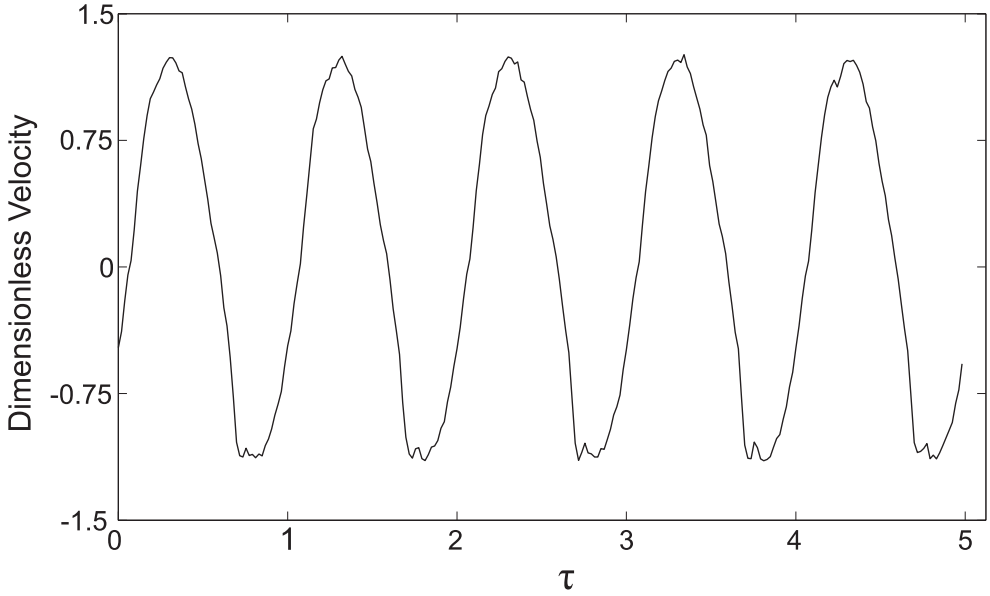

Figure 7b

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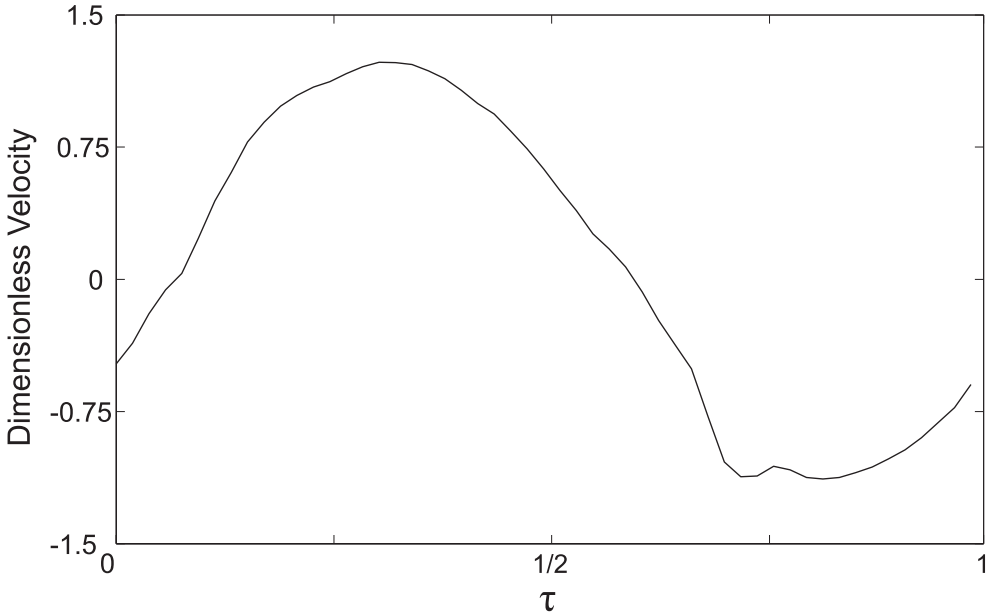

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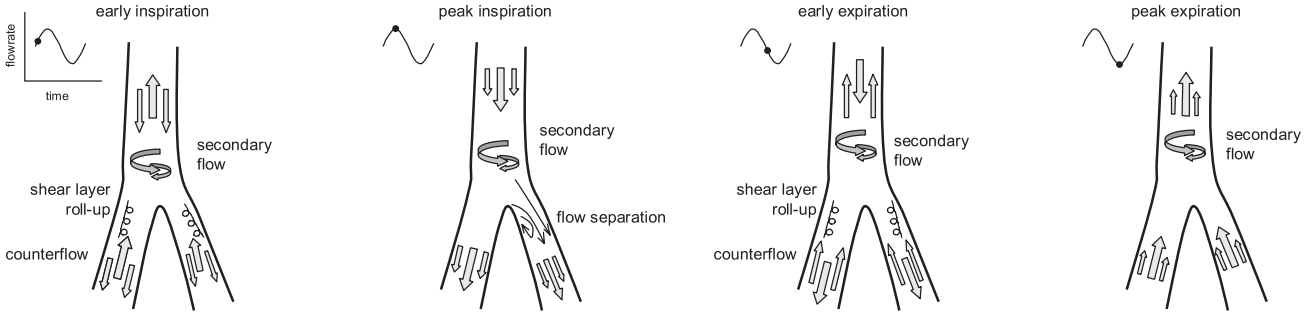
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