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Experimental Investigation of Oscillatory Air Flow in a Bronchial Tube Model with HFOV Mode

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Abstract : Mechanical ventilations for artificial respiration have been developed to improve the medical treatment of the patients showing respiratory disorder. In various types of ventilation, High Frequency Oscillatory Ventilation (HFOV) is one of the most effective techniques of medical care for pulmonary disease patients, especially, infants or premature infants. HFOV is a ventilation technique with high breathing rate in comparison with the normal breathing rate. Some successful studies have focused on the effect of treatment using HFOV. However, the mechanism of gas exchange in bronchial tube under the medical treatment by HFOV has not been clarified. In this study, the oscillatory flow in a micro-channel model of bronchial with single bifurcation in HFOV mode has been investigated experimentally with micro Particle Image Velocimetry (micro PIV) technique. The phase averaged velocity profiles changing with the driving frequency of HFOV have been investigated.

Keywords : Respiration, Bronchial tube, Oscillatory flow, Micro channel, Micro PIV.

1. Introduction

Analysis of flow in airways of human lugs is important factors in order to clarify the fundamental mechanism of pulmonary respiration. Many studies on respiratory flow have been carried out with models of the human airway. Schroter et al. (1968) experimentally investigated secondary flow in a bifurcation model. Lieber et al. (1998) carried out a study on oscillatory flows in a bifurcation airway model. Sarangapani et al. (1999) simulated aerosol bolus dispersion in a bifurcation model of human airways. Fresconi et al. (2003) presented experimental results on expiration flow in a symmetric bifurcation by using PIV and LIF techniques. And, Kim et al. (2004) carried out an investigation on airflow in pathological nasal airway by PIV. However, the analysis on gas transfer and exchange mechanism related to flow pattern in the airways, especially at the respiratory zones from bronchioles to alveoli, has not been clarified in detail. In addition, fluid dynamic analysis of artificial respiration methods applying as clinical procedures for patients showing respiratory disorders is an important problem for improvement of practical treatments.

In various types of artificial ventilation methods, High Frequency Oscillatory Ventilation (HFOV) is known as an effective clinical procedure for respiratory disorders (Bohn D. J. et al., 1980;

Drazen J. M. et al., 1984). In HFOV operation, the efficiency of gas exchange shows maximum in the ventilation frequency range of 10~20Hz. This is much higher than the normal breathing rate for adults, which is in the range of 0.2~0.8Hz. However, the mechanism of improvement of gas exchange by high frequency ventilation and the optimum frequency range in HFOV operation have not been clarified.

The velocity profile of oscillatory flow in the airways is not simple because of the complicated bifurcated structure of bronchial tree. And the increase of ventilation frequency must generate more complicated flow features in the bronchus. In order to analyze gas exchange mechanism in the airways, 3D flow effect, turbulence and streaming effect in the oscillatory flow must be important factors in both normal breathing and artificial respiration cases.

In this report, as the first step of the analysis of the frequency effect on gas transfer and exchange at the respiration zones of the airways, the method of experimental analysis for high frequency oscillatory air flow in a simple model of bronchiole has been established by using micro Particle Image Velocimetry (micro PIV) technique in HFOV operation. A simple micro channel model with single bifurcation was used as a model of an 18th~19th generation of bronchiole, and oscillatory velocity profiles for different frequencies in the tube have been measured.

2. Experimental Setup and Method

2.1 HFOV

Over the past 40 years, mechanical ventilation for artificial respiration has been developed for clinical treatment. The advantage of mechanical ventilation is that it is easy to control the oxygenation and to maintain lung volume. In addition, it can reduce the patient lord for breathing. There are several ventilation techniques for clinical use, e.g., CMV (Control Mode Ventilation), AMV (Assist Mode Ventilation), IMV (Intermittent Mandatory Ventilation), CPAP (Continuous Positive Airway Pressure), PSV (Pressure Support Ventilation), HFOV (High Frequency Oscillation Ventilation), etc. Above all, it is known that HFOV has an advantage in rapid improvement for critical care. On the other hand, this technique is not enough effective in the case of spontaneous breathing.

In HFOV operation, the sinusoidal pressure change over the mean airway pressure (MAP) is produced by harmonic piston motion with the ratio I/E = 1, where I/E is maximum and minimum peak pressures ratio at inhalation and exhalation phases. The ventilation effect depends on the amplitude of pressure, the frequency of the piston motion and the MAP. The rate of ventilation in HFOV operation is much higher than the normal human breathing rate, however the tidal volume V_T in the case of HFOV is smaller than the anatomical dead space ventilation volume (~150 $m\ell$) associated with normal breathing.

2.2 Outline of Respiration

As shown in Fig. 1, the lungs have a complicated bronchial tree system, and it consists of trachea, bronchi, bronchioles and alveoli. Especially, the terminal zones of the lungs indicate very complicated micro scale structure with multi-bifurcations and alveolar sacs. The physiological gas exchange occurs in the respiratory zone from respiratory bronchioles to alveoli. Inhaled air by inspiration fills trachea, bronchi, bronchioles and alveolar ducts. However, the inhaled air does not directly reach to the inside of alveoli, so the contact surface between the inhaled air and physiologically residual air generates in the respiratory zone. In order to transport O_2 rich air into alveoli, some kinds of flow mechanism such as molecular diffusion, gas mixing and streaming must exist between inhaled air and residual air. Table 1 shows respiratory flow parameters for each respiratory condition. Under normal respiration condition, the tidal ventilation volume for an adult is about 500 $m\ell$. The fraction of inhaled air, which fills the part of lung system unrelated to physiological gas exchange, is about

150 $m\ell$, and it is called the dead space ventilation volume.

The terminal airways structure, where the inhaled air contacts with the residual air, has complicated bronchial tree with bifurcated micro-channels in which the complicated gas exchange mechanism must occur not only by molecular diffusion but also by fluid motion effects generated by oscillatory flow. The fundamental parameters of this oscillatory flow are Reynolds number (Re_δ) and Womersley number (α). Reynolds number is different at each zone of the airway. In the terminal zone of the lungs, Re_δ is very small (usually less than 0.01 in the alveolar region), and in the case of normal respiration, α is also very small because the ventilation frequency is very low (about 0.2 Hz). This means that the oscillatory flow in the bronchioles can be assumed to be a quasi-steady laminar flow in the case of normal breathing.

When HFOV is applied as artificial ventilation, the characteristics of oscillatory flow in the bronchioles change markedly. In typical HFOV operation, the tidal volume is less than for natural respiration, and the frequency is in the range of 5 to 20 Hz. This is 25 to 100 times of the normal breathing. Womersley number increases as the frequency increases. The increase of Womersley number introduces the effect of unsteadiness in oscillatory flow in the bronchioles. Thus gas exchange in the lungs system will take place by a different mechanism.

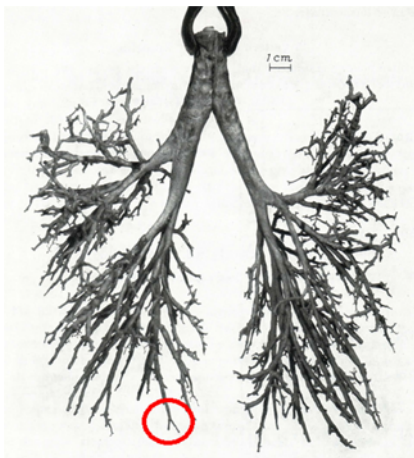


Fig. 1. Bronchial tree (Weibel, E. R., 1963) and bronchiole indicated by a circle.

Table 1. Respiratory flow parameters for different respiratory conditions.

Parameter	Normal breathing	Exercise case	HFOV operation
Tidal volume; V_T ($m\ell$)	500	~3300	*70~150
Breathing frequency; f (Hz)	0.2	0.8	5～20

* The tidal volume V_T ($m\ell$) is smaller (i.e., 1~3 $m\ell$/kg (weight)) than the anatomical dead space (150 $m\ell$).

2.3 Experimental Setup

The shape and geometry of the model shown in Fig. 2 was determined from the anatomical features of a terminal airway bifurcation in human lungs based on Weibel's model (1963). The test channel is made of aluminum plate, and it has dimensions corresponding to 18th ~19th generation of bronchioles.

The dimensions of parent channel are 500 μm in width and 500 μm in depth, and the daughter channels are 450 μm in width and 500 μm in depth. The length of parent and daughter channels is 20 mm, which is sufficiently long to eliminate entrance effects. The branching angle between the daughter tubes is 60 degrees, which is typical of most airway bifurcations. The radii of rounding at the branching point of the daughter tubes and the connecting point of side walls between the parent tube and daughter tubes are 50 μm and 2 mm, respectively. The aluminum plate was preprocessed

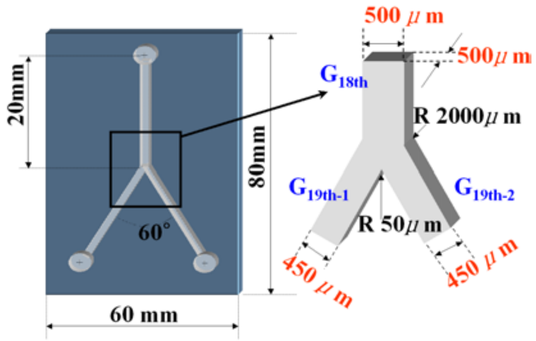


Fig. 2. Schematic of a bifurcated micro channel.

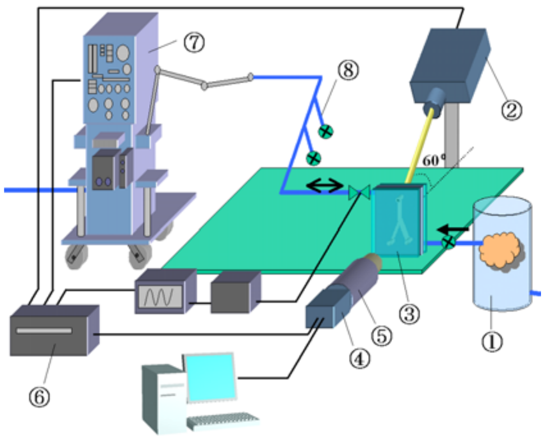


Fig. 3. Experimental setup.

with black coating in order to reduce the reflection. The micro channel was hold by two pieces of transparent glass with 10 mm thickness from both sides.

The experimental setup is shown in Fig. 3. Seeding particle selection is difficult in micro PIV applications to micro-scale gas flows. However, in this experiment, velocity profiles of oscillatory airflow were successfully measured by applying a conventional seeding technique using incense smoke as a source of tracer particles. After charging smoke particles into the micro channel from the smoke generation chamber (1), the chamber was isolated by a valve. The test section (3) was illuminated by laser beam from a double-pulsed Nd-YAG laser (2). The wavelength of the light source was 532 nm and the output was 50 mJ/pulse. The forward scattering was used to capture PIV images by rear side illumination at an angle of 60 degrees. The images were captured by using a CCD camera (4) with 1360(H) ×1035(V) [pixels] and a micro lens (5). Data acquisition was synchronized with the HFOV driving signal from a digital pulse generator (6), and its control signal was pressure wave at the outlet of HFOV driver. The HFOV driver (7) used in this experiment was operated under various ventilation conditions. The parameters of HFOV operation were frequency, mean airway pressure, and pressure amplitude. The output of the HFOV driver was connected to the micro-channel though a plastic tube with an inner diameter of 4 mm. The flow rate was controlled by two bypass valves (8).

3. Results and Discussion

3.1 Experimental Conditions

The HFOV driver used in this experiment can control the MAP in the range of 0 to 3 kPa, and a stroke volume in the range of 0 to 30 $m\ell$. The ventilation frequency and the driving pressure amplitude can be controlled in the range from 10 to 25 Hz and from 1 to 5 kPa, respectively. In practical treatment, the MAP in HFOV operation is usually controlled at 0.5 kPa, but it depends on the condition of the patient. In this experiment, the MAP was fixed at 2 kPa, which is almost the highest condition for practical cases. The driving frequency was in the range of 10 to 20 HZ, and the pressure amplitude was kept at 1.6 kPa at the outlet of the HFOV controller. The pressure amplitude at the inlet of the micro channel was 0.5 kPa

Phase averaged velocity profiles in the bronchial model were measured at the phases of maximum and minimum pressure, which correspond to the influx and efflux phases of respiration. The timing of the pulsed-laser illumination for PIV measurements was controlled at each phase of pressure change at the outlet of HFOV driver by using trigger signals from an external pulse generator. The CCD camera was synchronized with the pulsed laser oscillation, and the pulse separation for PIV measurements was chosen on the condition that particle displacements were less than 10 pixels.

In order to investigate the oscillatory flow in the micro channel model, the following dimensionless parameters must be considered. The Stokes layer thickness δ corresponding to the oscillatory boundary layer thickness is given by

$$\delta = \sqrt{\frac{2\nu}{\omega}} \qquad (1)$$

where ν is kinematic viscosity and ω is angular frequency (ω = 2πf). Reynolds number of oscillatory flow Re_δ, Womersley Number (Shear Wave Number) α , and Strouhal number *Str* for oscillatory flows are defined as,

$$\mathrm{Re}_\delta = \frac{2u'}{\sqrt{\nu\omega}} \quad (2), \qquad \alpha = \frac{D}{2}\sqrt{\frac{\omega}{\nu}} \quad (3), \qquad \text{and} \qquad Str = \frac{\alpha^2}{\mathrm{Re}} \quad (4),$$

where u' is velocity amplitude of the oscillatory flow, which is expressed as $u = u'\cos\omega t$. *D* is the characteristic length of the micro channel. Knudsen number *Kn* is defined as

$$Kn = \frac{\lambda}{D} \qquad (5)$$

where λ is the mean free path.

The values of these dimensionless parameters are shown in Table 2. Knudsen number is small enough to apply non-slip condition of the flow in the model. Stokes layer thickness is larger than the half-width of the channel, and also Reynolds number is small enough. It means that the flow in the micro channel model can be fundamentally assumed as a quasi-steady laminar flow. However, these values were estimated for straight channel. In the case of micro channel flow with a bifurcation, experimental investigation is needed to clarify the flow condition affected by bifurcating and joining.

Table 2. Physical parameters of oscillatory flow in present study.

Parameters	Values
Stokes layer thickness, δ [mm]	0.49 ~ 0.69
Peak Reynolds number, Re_δ	2.75 ~ 7.78
Womersley number, α	0.51 ~ 0.72
Strouhal number, *Str*	0.0334 ~ 0.1885
Knudsen number, *Kn*	1.30E-04~1.40E-04

3.2 Results Obtained

The velocity field measurements of oscillatory flows in the bronchial tube model were carried out by using a micro PIV technique. The velocity fields were reconstructed by conventional cross-correlation evaluation from PIV images. The image size was 1360×1035 pixels, and it corresponds to a measurement window size of 1.56×1.2 mm. The interrogation spot size was 16×16 pixels. Erroneous vectors caused by optical noise, such as halation spots, were removed by a simple filter based on threshold of absolute value of velocity.

Figure 4 shows a raw image of flow in the micro channel model at the influx phase. Figure 5 shows a phase averaged velocity vector map at the influx phase, φ = π/2, and at the efflux phase, φ = 3π/2 of oscillatory flow at frequency f = 10Hz. The phase-averaged velocity fields were calculated from 200 frames of instantaneous velocity field.

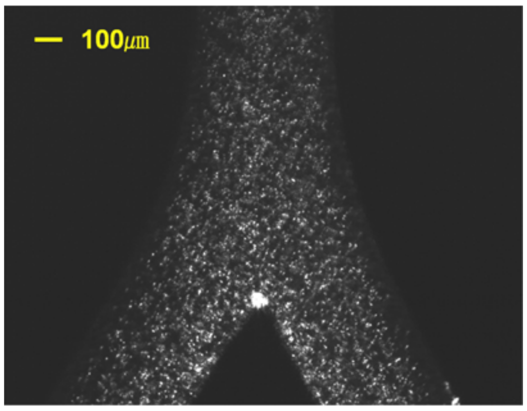
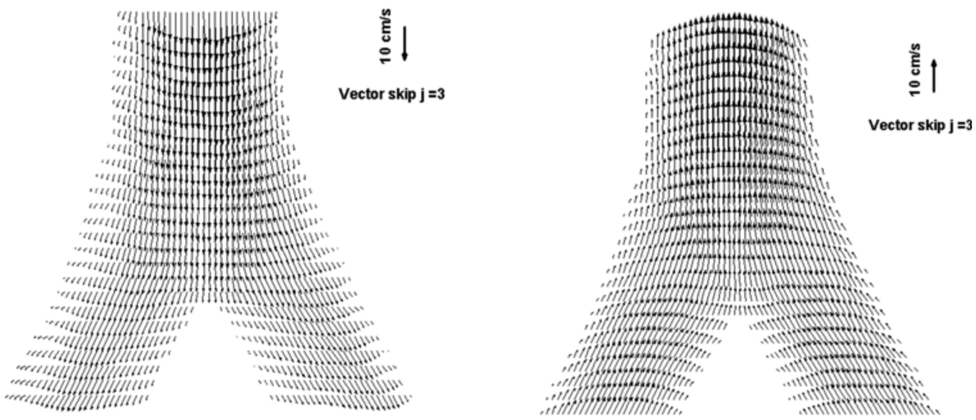
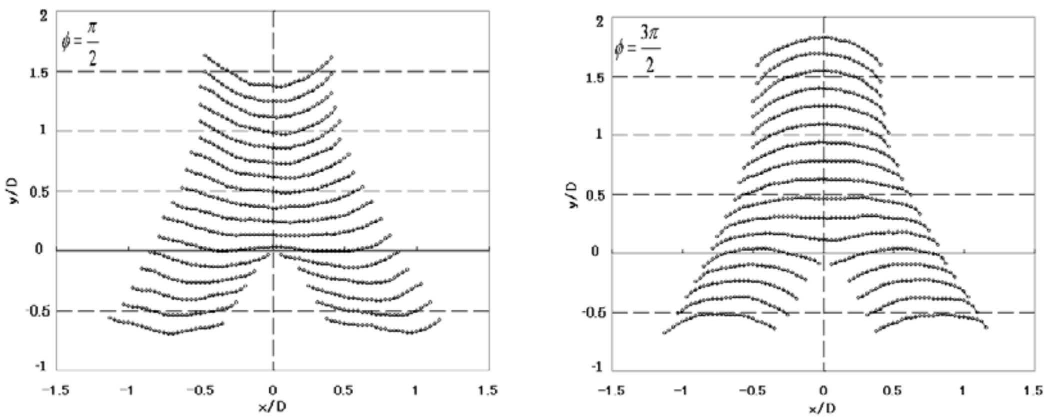


Fig. 4. Raw image of flow in the micro channel at influx phase.



(a) influx phase (b) efflux phase

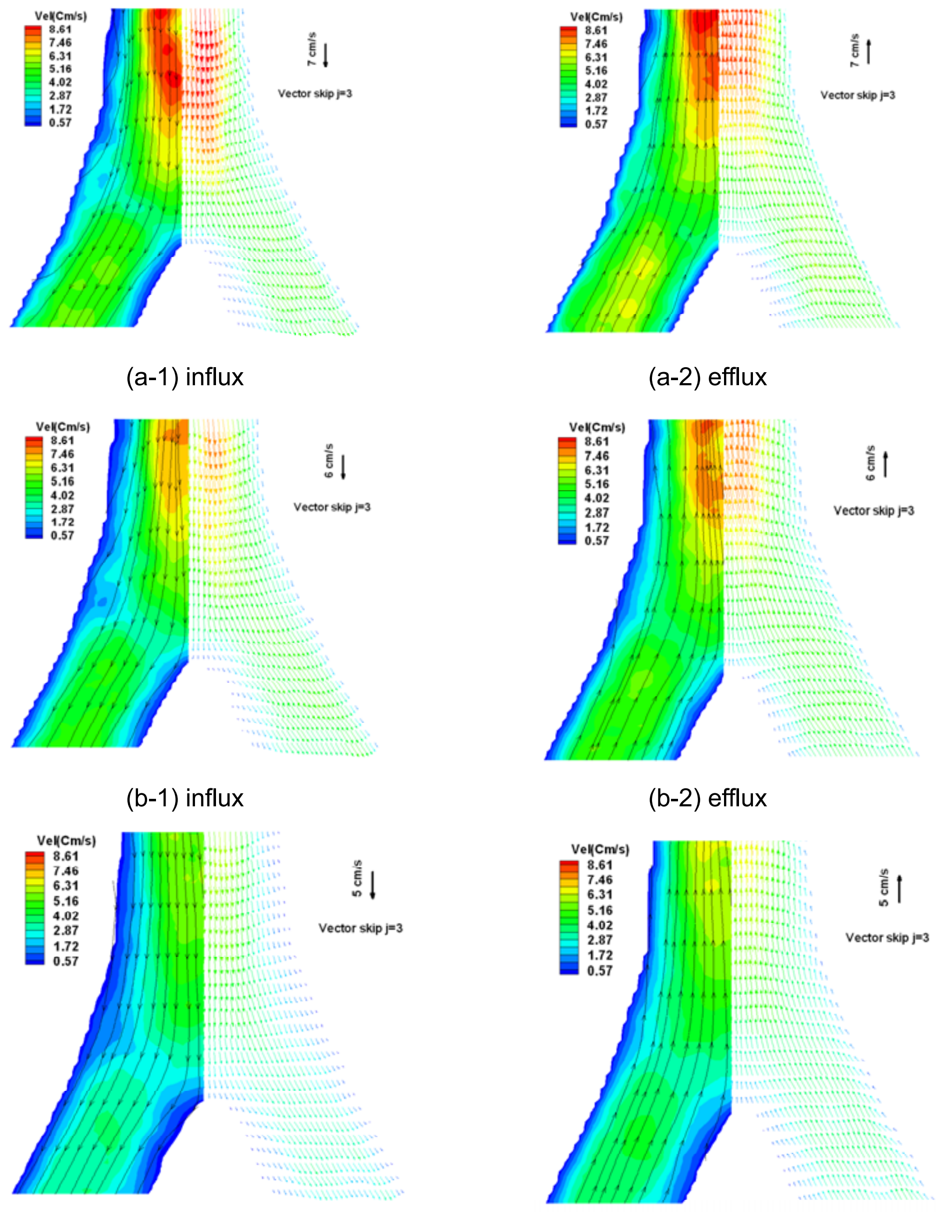
Fig. 5. Phase averaged velocity vectors at influx and efflux under HFOV driving conditions.



(a) influx phase (b) efflux phase

Fig. 6. Oscillatory velocity profiles in the bifurcating micro channel (α = 0.51).

Figure 6 shows velocity profiles at each phase. The abscissa is the dimensionless distance x/D, where x is the distance from the centerline of the micro channel, and D is the width of the parent channel. The ordinate is the dimensionless distance y/D, where y is the distance from the branching point of the daughter channels. The velocity profiles for $\alpha = 0.51$ are shown at the phases of $\varphi = \pi/2$ and at $\varphi = 3\pi/2$. These figures show the peak of the velocity profile in the parent tube appears along the centerline in the axial direction during the influx phase, however, in the daughter branches the peak positions are shifted toward the outside wall. During the efflux phase, the flows from the both daughter branches join and develop quickly in the parent tube.



(a) f = 12Hz (a-1) influx (a-2) efflux

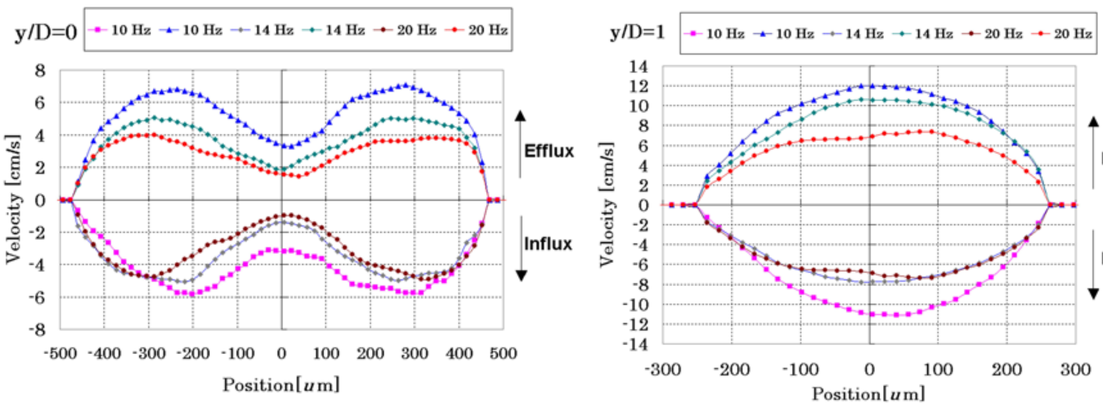
(b) f = 16Hz (b-1) influx (b-2) efflux

(c) f = 20Hz (c-1) influx (c-2) efflux

Fig. 7. Streamlines and velocity vectors at peak influx and efflux.

Figure 7 shows streamline and axial velocity contours (left side), and a velocity vector map (right side) for frequencies of 12, 16 and 20 Hz. These figures show the velocity gradually decreases as the frequency increases.

The variations in velocity profiles for different driving frequencies at the influx and the efflux phases are shown in Fig. 8, where (a) and (b) show the velocity profiles at the branching (y/D = 0) and in the parent channel (y/D = 1), respectively. The HFOV driving frequencies were 10, 14 and 20 Hz. The results obtained show that the oscillatory velocity amplitude decreases as the frequency increases, and the velocity profiles change with frequency, and the velocity profile at y/D=1 in the case of low frequency shows like as Poiseuille flow distribution, but the velocity profile changes from it as the frequency increases.



(a) velocity profiles at y/D = 0 (junction) (b) velocity profiles at y/D = 1 (parent channel)

Fig. 8. Variation of velocity profiles with frequency.

In order to explain the frequency dependence of the velocity profiles in each phase, detailed time series measurements of the oscillatory flow are required. Additionally, more precise measurements with better spatial and temporal resolution must be carried out to clarify the complicated phenomena in bronchial associated with oscillatory ventilation flow of respiration, such as secondary flow, steady streaming flow (Haselton et al., 1982), and annular effect (Uchida et al., 1956). And also it is necessary to change the experimental model to the near model in actual shape with multi-bifurcations.

4. Conclusion

In this study, experimental method of oscillatory air flow in a micro channel with single bifurcation as a model of bronchiole by using micro PIV technique has been established, and fundamental characteristics of oscillatory flow in the model in HFOV operation were investigated. Phase averaged oscillatory flow patterns and velocity profiles in the micro channel for different HFOV driving frequencies were measured, and the following results have been obtained. At the influx phase, the velocity distribution in the flow in the daughter branches is not symmetric about the center axis of the branches. At the efflux phase, the confluence from the daughter branches develops quickly to the profile of developed laminar flow in the parent tube, but the velocity profile changes as the driving frequency increases. These results indicate the possibility of examining the change of gas exchange mechanism with ventilation frequency. In order to clarify more detailed oscillatory flow conditions in the terminal zone of airways, measurements with improved spatial and temporal resolutions are required.

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