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Experimental Analysis of Oscillatory Airflow in a Bronchiole Model with Stenosis

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Abstract : As the mechanism of gas transport and exchange in human respiratory ventilation, the complicated processes of mixing and diffusion in airways of human lungs are considered. However the mechanism has not been clarified enough. On the other hand, the analysis of detailed mechanism in the case of artificial ventilation like HFOV (High Frequency Oscillatory Ventilation) is strongly required for the development of clinical treatments on patients with respiration disorder. In HFOV, it is considered that pendelluft becomes one of the important factors of gas transport and exchange because of high frequency ventilation in comparison with natural breathing. As increase of the frequency, the different time constants of lung units generate phase lag of ventilation in airways of lungs. The phase lag of ventilation causes to generate pendelluft. The time constant is determined by compliance and flow resistance of lung unit. In order to investigate the effect of the different time constants induced by the difference of flow resistance in a part of respiratory bronchiole of human lungs, the experimental study has been carried out by using multi-bifurcated micro channels as a model of bronchiole. The flow resistance in the model channels was produced by a stenosis. The velocity distributions of ventilation flows in the channels with and without the stenosis have been measured by using μ-PIV technique. The results obtained show the frequency effects on the flow pattern in the bronchiole model channels and the appearance of pendelluft.

Keywords : Ventilation, Respiratory bronchiole, HFOV, Micro channel, μ-PIV

1. Introduction

Respiratory ventilation is the fundamental physiology of life-support. However, the mechanism of gas transport and exchange in airways of human lungs has not been clarified enough. Because the airways have the complex structure of bifurcations with 23 generations from trachea to alveoli, and respiratory ventilation in airways is accompanied by the complicated flow processes. These are bulk flow, Taylor's diffusion, molecular diffusion, streaming, pendelluft and so on. The conditions of gas transport and exchange in airways are affected by physical states of lung unit and by breathing condition which is in the resting, in the exercise or in the medical treatment by artificial ventilation like HFOV (High Frequency Oscillatory Ventilation). As the change of physical states of lungs, the decrease of compliance by lung disease and the increase of flow resistance in airways caused by some kind disease like stenosis by tumor are considered. The local change of physical states of lung unit brings about different time constants in the part of lungs, and it relates to the responses of ventilation to the change of respiration pressure. In the case of natural breathing in the resting or in the exercise at 0.2~0.8Hz, the difference of time constant is not important factor in consideration of gas transport and exchange mechanism in respiratory ventilation. However, with the increase of ventilation frequency, especially in the case of HFOV at 5~30Hz, the effect of the difference of time constant appears as phase lag of ventilation in airways, and it causes the generation of pendelluft.

Many studies on respiratory flow have been carried out by considering the human airway models. Haselton et al (1982) have shown flow visualization of steady streaming in oscillating flow through a bifurcating tube. They presented the results in the wide range of Reynolds and Womersley numbers using a Y-shaped bifurcation models with several fluids. Elad et al. (1998) developed a nonlinear lumped-parameter model to study the dependency of airflow distribution in asymmetric bronchial bifurcations on structural and physiological parameters. They derived the modified time-dependent expressions of resistance and compliance of each compartment, and presented that asymmetry in compliance of peripheral airways affected flow distribution in branch tubes and induced larger degrees of pendelluft. Ramuzat et al. (2002) carried out experimental investigation of oscillating flows in a 3D multiple-bifurcation lung model by using PIV. Martin et al.(2005) reported the flow structures in a U-shaped model channel with characteristic length 10mm by using μ-PIV technique in the range of Reynolds number 100~900. Kleinstreuer et al. (2003) described the development of highly efficient drug delivery systems for desired site deposition analysis of a four generation lung airway model with hemispherical tumors. They showed that the presence of a singular tumor in bronchial airways changed the flow distributions among different branches, and particle deposition on the tumor surface might be influenced by the local occlusion, as well as upstream flow and particle distributions as a function of Reynolds number in symmetrically bifurcating rigid lung airways with hemispherical tumors. Ultman et al. (1988) and High et al. (1991) carried out a study of pendelluft and mixing in a single bifurcation model during HFOV, and they estimated the mixing coefficients and the pendelluft volume fraction. However, the experimental investigation of detailed flow features in real scale model of bronchial tube has not been carried out.

The oscillatory flow phenomena in respiratory bronchiole have not been clarified enough not only in HFOV but also in natural breathing. Respiratory bronchiole is the end part of airways of human lungs, and it is being connected to alveolar duct. The bulk flow of aspiratory air reaches this part of lungs and the contact surface of inhalation air with physiologically residual air is formed in the part at phase finishing inspiration in the case of natural breathing. The Reynolds number and Womersley number of this part are very low. It means that the oscillatory flow in bronchiole is almost quasi-steady in the case of natural breathing. However, in HFOV, both Reynolds number and Womersley number become several times. And also compliance of alveolar duct units connected to respiratory bronchiole is not usually the same. Therefore it needs to clarify the effect of different time constants in respiratory bronchiole under the same condition of ventilation in human lungs, but there are some difficulties in the experimental analysis. The group of the authors performed the experimental investigations of oscillatory air flow in a single-bifurcated micro channel as a model of respiratory bronchiole terminated with different conditions of compliance by using μ-PIV technique, and they succeeded to obtain quantitative velocity profiles of pendelluft generated in the micro channel model for HFOV (Lee et al. (2006)). Flow resistance is the other factor to determine time constant. The inhomogeneity of flow resistance in bronchiole branches induces phase lag of ventilation in the branches, and it causes pendelluft generation.

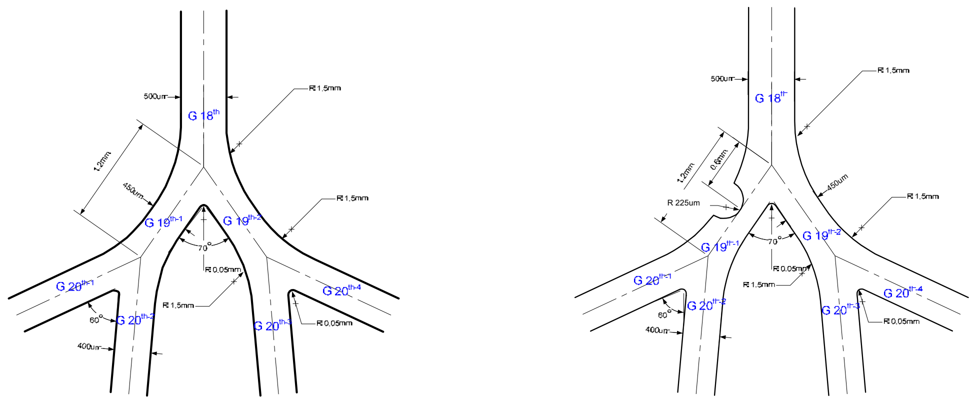
In this report, the effect of inhomogeneous flow resistance in bronchi branches on the respiration flow was experimentally investigated by using multi-bifurcated micro channels as a model of respiratory bronchiole. The geometry of the channels are corresponding to the 18th~20th generations of airways of human lungs by Weibel's lung classification scheme (Weibel, 1963). The difference in flow resistance was produced by a semicircular stenosis formed in the one side branch of the 19th generation. On the other hand, each branch of the 20th generation was terminated with the same condition of compliance. The velocity profiles in the models with and without stenosis were obtained by μ-PIV measurements for different breathing conditions. As a result a flow pattern of pendelluft has been observed.

2. Experimental Method and Setup

2.1 Micro Channel Model

The oscillatory flows in the end part of airways of human lungs were analyzed in this investigation. For the total analysis of mechanism of gas transport and exchange by mixing and diffusion in this part, the 3D flow structure is a factor to be considered, and also the effect of tube wall elasticity. In that case, an elastic tube model of bronchiole with circular cross-section is suitable to use for the experimental analysis. The subjects of this investigation were the frequency dependent stream-wise phenomena in oscillatory ventilation flows through micro scale bronchiole. The phenomena were investigated by the measurements of 2D velocity distributions in multi-bifurcated micro channels with rectangular cross-section as a model of bronchiole under the assumption that the effects of 3D flow structure and tube wall elasticity are ignorable.

Figure 1 shows the configurations of multi-bifurcated micro channel models corresponding to the 18th~20th generations of airways with and without a semicircular stenosis in one side branch of the 19th generation. When stenosis by tumors is considered, its shape is usually not simple. In this experiment, the simplest shape of stenosis was considered as an element producing flow resistance. The test channels were made by precision machining of aluminum plate with 500μm thickness, and they were preprocessed with black Alumilite coating in order to prevent optical scattering. The processed aluminum plates were put between the transparent plates of glass and Plexiglas of 1cm thickness, and then the micro channels were formed with the following dimensions. The depth of channel was 500μm, and the channel widths of the 18th~20th generations were 500μm, 450μm and 400μm, respectively. The radius of the stenosis was 225μm, and it corresponded to the half width of the channel. The length of the branch of the 19th generation was 1.2mm along the center line. The angles of the 1st and 2nd junctions were 70 and 60 degrees, respectively. The each branch of the 20th generation in both channels was terminated with closed flexible tube with the same length, and it means that the each branch of the 20th generation was terminated with the same condition of compliance.



(a) micro channel (b) micro channel with semicircle stenosis

Fig. 1 Multi-bifurcated micro channel models of bronchiole.

2.2 Experimental Setup

Figure 2 shows schematic of the experimental setup consisted of ventilation system, test channels, high power macro lens and image capturing system for μ-PIV.

The ventilation of natural breathing modes in the resting and in the exercise was produced by the reciprocal motion of a small volume syringe (250μl) driving by a stepping-motor. In HFOV mode, the part of syringe was replaced by HFOV driver. The pressure sensor was set in the upstream side of micro channel, and the signal of the sensor was applied for the phase locked measurements of velocity fields at different phases.

The μ-PIV system for 2D velocity field measurements in the micro channels was consisted of double-pulsed Nd:YAG laser, long-distance high power lens (×500), CMOS camera, timing hub, delay generator, oscilloscope and computer for image capturing. The test section was illuminated by expanded laser beam from the laser light source with 532nm wavelength, 50mJ/pulse output and 5ns duration from the back side of the micro channels at angle of 60 degrees to prevent damage of the camera. The camera has the resolution of 1280(H)×1024(V) pixels in space and 10 bit in depth. The frame rate of the camera is 1,000, and it is not enough for direct time series measurement in HFOV. The phase locked measurements were carried out by using this camera. The oil-mist generated by Laskin atomizer nozzles was used as seeding particles for μ -PIV measurements. The oil-mist size is less than 1μm, and its material is a kind of vegetable oil. The seeding particles were mixed at the upstream of the test channels. The PIV images were captured at different phases of ventilation by using pulse generation system controlled by phase of sinusoidal pressure change at the inlet of the channels.

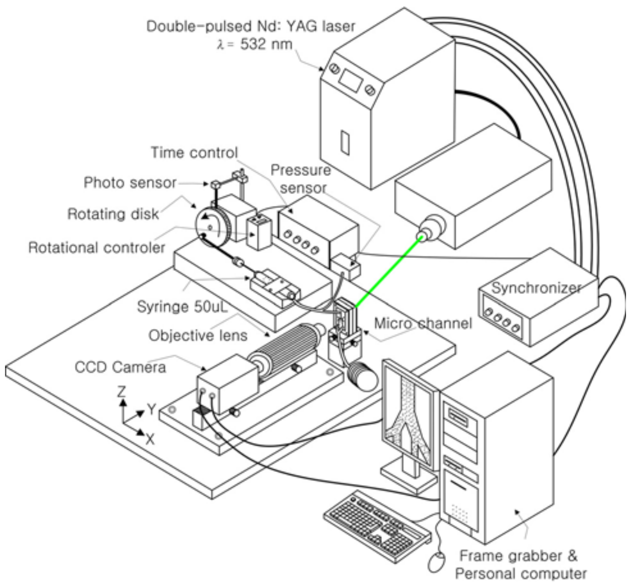


Fig.2. Schematic of experimental setup.

3. Experimental Results and Discussions

3.1 Experimental Conditions

In the analysis of oscillatory flow in channels, the fundamental parameters are Reynolds number of oscillating flow (Re_δ) in which the characteristic length is the boundary layer thickness ($\delta \propto \sqrt{\nu/\omega}$), and Womersley number (α). The parameters are defined as follows.

$$\mathrm{Re}_\delta = \frac{2u'}{\sqrt{\nu\omega}}, \qquad (1) \qquad\qquad \alpha = \frac{D_h}{2}\sqrt{\frac{\omega}{\nu}}, \qquad (2)$$

where ν is kinematics viscosity, ω and u' are angular frequency ($\omega = 2\pi f$) and velocity amplitude of the oscillating flow, and D_h is hydraulic diameter of micro channel. Both of Reynolds number and Womersley number depend on the oscillating frequency, and in particular, the latter is important parameter in order to consider the effect of frequency on the velocity profiles of oscillating flow in channels. The frequency of oscillating flow in HFOV is much higher than the frequency in natural breathing, and Womersley number increase in HFOV.

In order to investigate the frequency effects, the experiments for different ventilation frequencies have been carried out at 0.2Hz and 0.8Hz as natural breathing in the resting and in the exercise, and in the range of 10Hz~20Hz for HFOV. The flow rate in the micro-channel model in the case of natural breathing was almost the same with it in human breathing at the 18th generation estimated by Weibel. The experimental conditions in HFOV were determined as follows. The stroke volume was 30 ml, and MAP (Mean Airway Pressure) was kept at 6kPa. The experimental conditions in this investigation are shown in table 1.

Table 1. Experimental conditions of ventilation.

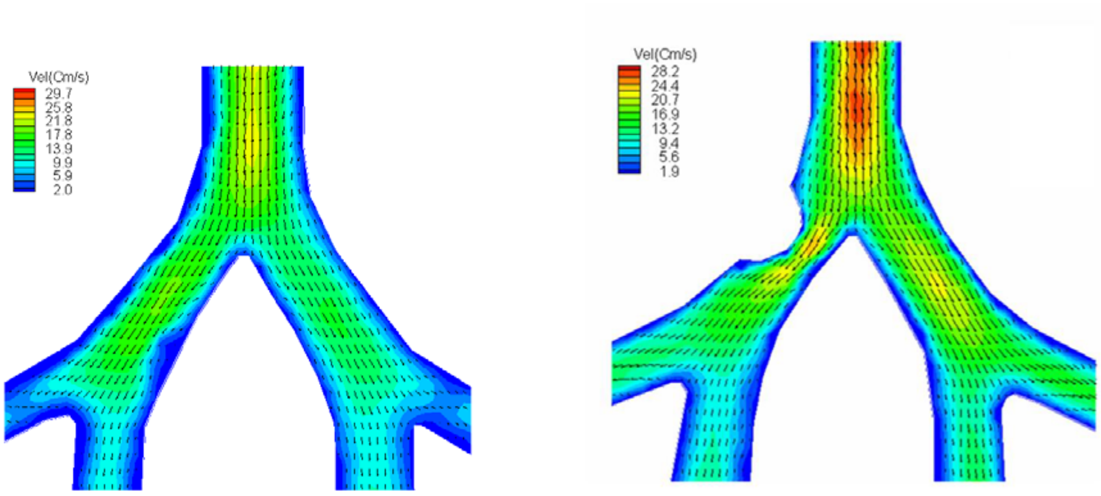
Breathing Conditions	Frequency (Hz)	Tidal volume ($m\ell$)	Re_δ	α
in the resting	0.2	62.5~250	7.75~33.23	0.072
in the exercise	0.8	62.5~250	39.81~120.5	0.144
in HFOV	10~20	30	28.36~45.7	0.51~0.72

3.2 Experimental Results

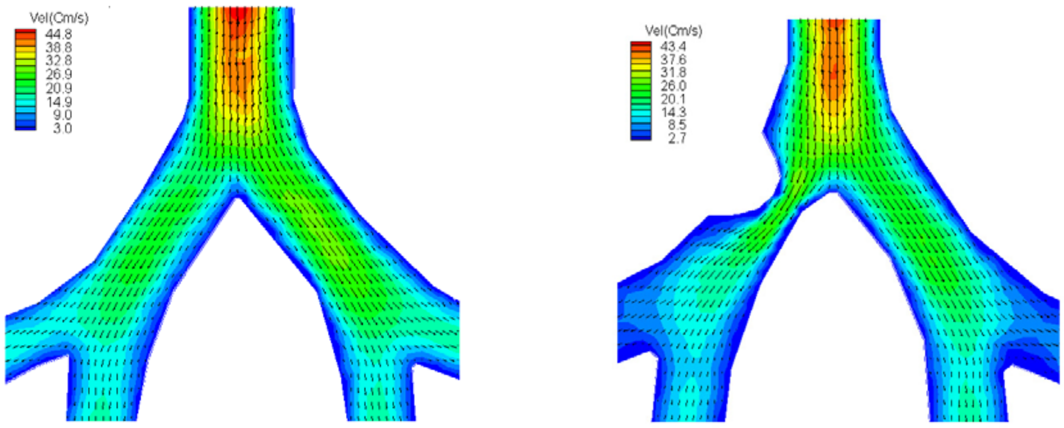
The 2D velocity distributions of oscillating flows in multi-bifurcated bronchiole models with and without stenosis were obtained by phase-locked μ-PIV measurements for different ventilation conditions. The plane of measurements was fixed at the middle of the depth direction. The phase control of measurements was based on the sinusoidal variation of ventilation pressure at the inlet of the channel. The origin of phase angle in the measurements was fixed at the state of the maximum inlet pressure. The error in phase angle was estimated as about $\pi/10$, because the amplitude of pressure variation was very small, and the signal included some noise. The averaged velocity vector fields were reconstructed with the PIV measurement of 500 times at fixed phase of ventilation.

The velocity vector maps of inspiration flows at the state of the maximum inlet pressure are shown in fig.3 for 0.2Hz, 0.8Hz, 10Hz and 15Hz. The size of observation area was 2.8×2.3mm. The interrogation window size for velocity reconstruction was 16×16 pixels, and it was corresponding to the spatial resolution of 2μm. In order to minimize the error in the velocity reconstruction, the laser pulse interval was optimized for particle displacements in PIV images and the sub-pixel processing was applied. However, the results included some errors. The major source of error was image noises caused by glass wall contamination with tracer particles.

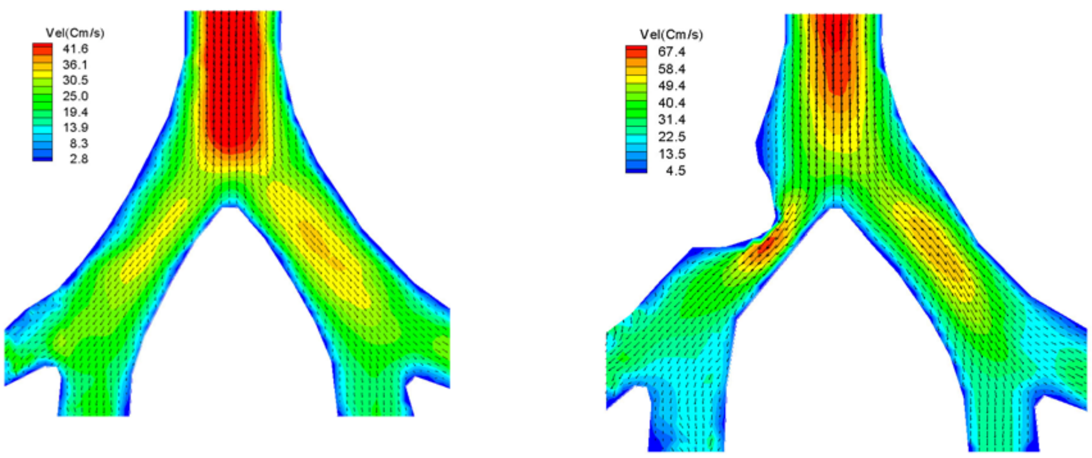
As shown in fig.3, the flow patterns in the channel without stenosis are almost symmetry, and it verifies that the end of each branch of the 20th generation was terminated with the same physical state. The peak velocities of the flow in the 18th generation channel in both cases of natural breathing and HFOV increase with the increase of frequency, because the tidal volume was kept at the same in each ventilation condition. In the channel with stenosis, the flow rate difference between the branches of the 19th generation is observed. The flow separation around the stenosis is not observed in all cases, but the inhomogeneous inspiration flow toward the junction of the 20th generation appears in the downstream side of the stenosis. The tendency of the inhomogeneity becomes remarkable with the increase of flow velocity at the part of stenosis.



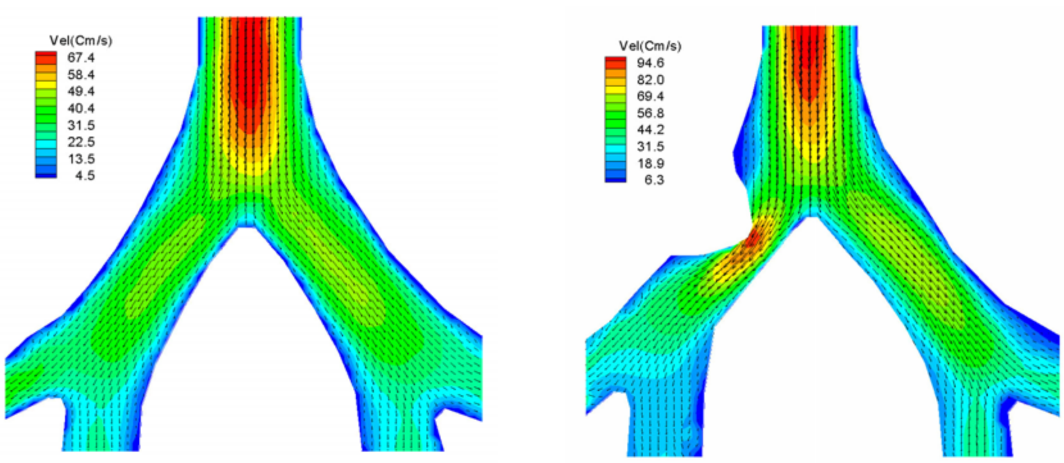
(a) natural breathing in the rest (at 0.2Hz)



(b) natural breathing in the exercise (at 0.8Hz)



(c) HFOV mode (at 10Hz)



(d) HFOV mode (at 15Hz)

Fig. 3 Velocity vector maps of inspiration flow in model channels.

From the flow patterns shown in fig.3, it is not able to discuss about the effect of time constant difference, because the velocity vector maps were obtained at fixed phase of ventilation. The effect of time constant difference was verified at phase of critical ventilation condition. Figure 4 shows the velocity vector map of the flow in the channel with stenosis at phase of switching from the expiration to the inspiration in HFOV at 10Hz. The figure shows that the flow of the residual expiration in the left branch is joining with the starting flow of the inspiration in the right branch. The phase lag of ventilation in the left branch was brought by the difference of time constant of each branch units. The increase of time constant of the left branch unit was caused by the increase of flow resistance produced by the stenosis. This type of flow caused by phase lag of ventilation is known as pendelluft.

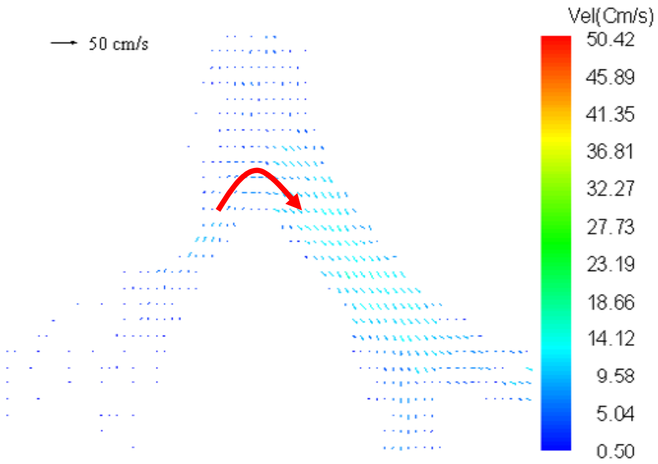
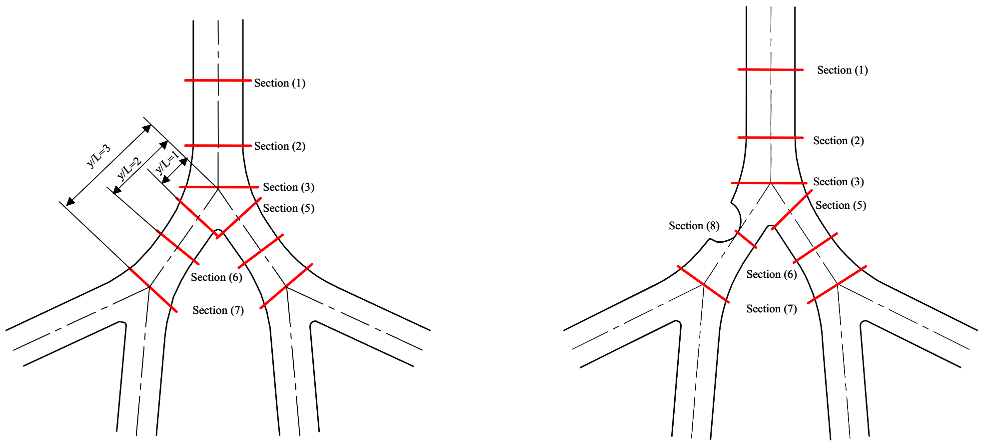


Fig.4 Velocity vector maps of flow at state of switching from expiration to inspiration.

Figure 5 shows the measurement sections of the velocity profiles. Figure 6 shows velocity profiles at sections 2, 3, 5 and 6 in both channels at the states of the maximum and the minimum inlet pressures for different ventilation conditions. The transverse axis in fig.6 is non-dimensional distance from the left side wall of the channel, where D is the width at each section shown in fig.5. The values of D at the sections 2 and 6 are equal to the channel widths of the 18th and the 19th generations, but the values at the sections 3, 5 and 7 are larger than the channel width. The error of about 5% in the determination of both ends of the transverse axis was included, because it was difficult to determine the exact positions of the channel walls from the PIV images in which the wall boundaries were recorded as blurred lines by the effect of optical arrangement. The vertical axis indicates the non-dimensional velocity, where U_0 is the amplitude of ventilation velocity at the inlet of the channel in each condition. The positive side in the figures shows the expiration velocity profiles, and the negative side is the inspiration.

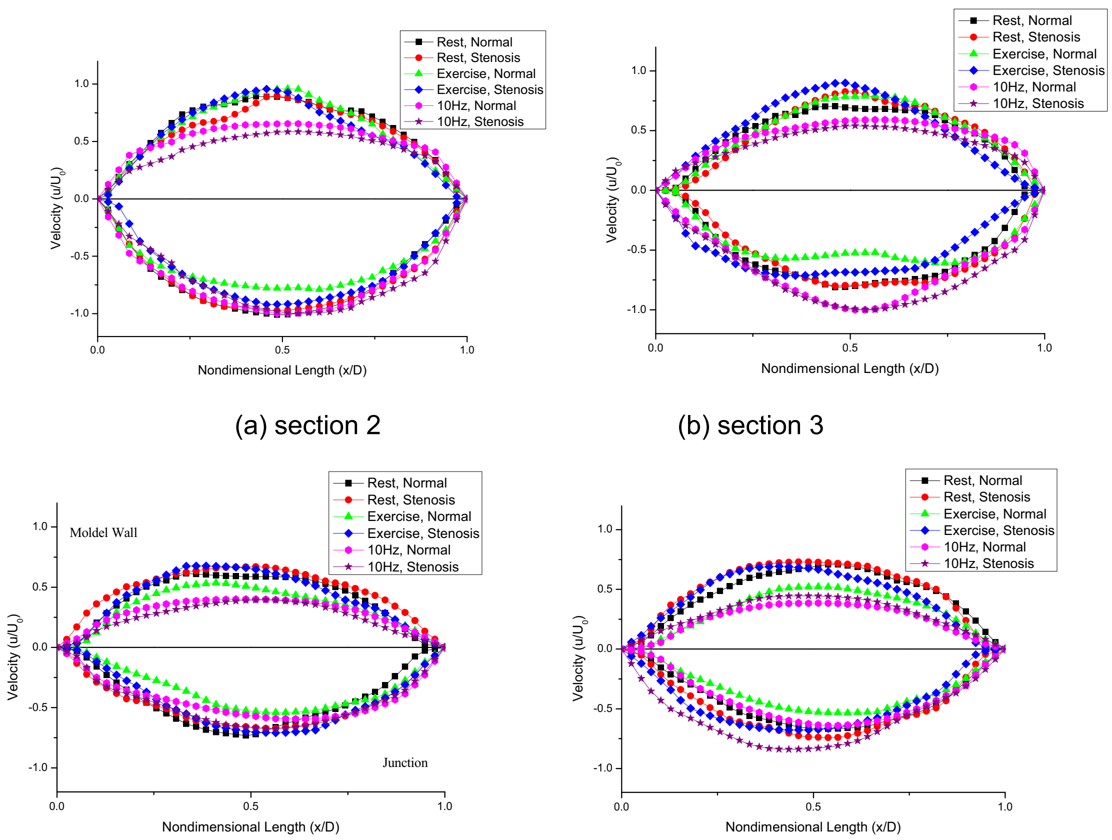
From the velocity profiles shown in fig.6, it is estimated that the flow rate of expiration at the condition of the minimum inlet pressure is less than the flow rate of inspiration at the condition of the maximum inlet pressure. And the difference increases with the increase of frequency. The cause of the difference is considered as the phase of the minimum inlet pressure dose not coincide with the phase of the maximum expiration, and the difference of phase increases in HFOV. In all cases shown in fig.6, it seems that the velocity profiles in natural breathing are affected by the channel geometry and the ventilation conditions, but the almost smooth profiles are observed in HFOV.

The velocity profiles at section 2 in the 18th generation branch are shown in fig. 6 (a). The effects of ventilation frequency and channel stenosis are not clear in the inspiration velocity profiles at this section. The velocity profiles of expiration in cases of low frequency show that the joined flows from the both branches of the 19th generation is not enough developed at this section. The velocity profiles at section 3 which is close to the corner of 1st junction with distance 200μm include some disturbances as shown in fig.6 (b). It is assumed that the effects of the junction and the 3D flow structure are appeared in the velocity profiles. The velocity distributions of inspiration and expiration at the inlet of the right branch of the 19th generation (at section 5) show the different tendencies in cases of natural breathing. The positions of peak velocity of inspiration shift toward outside wall, and toward inside wall in expiration. The velocity distributions at the middle part of the right branch of the 19th generation (at section6) are almost symmetry as shown in fig.6 (d). In all cases shown in fig6, the effect of flow resistance difference dose not clearly appear in the velocity profiles, but the variations with the ventilation frequency are clearly observed.



(a) bronchiole model (b) bronchiole model with semicircle stenosis

Fig.5 Measurement sections of velocity profiles.



(a) section 2 (b) section 3

(c) section 5 (d) section 6

Fig.6 Velocity profiles at sections 2, 3, 5 and 6.

The velocity profiles at section 7 in both branches of the 19th generations are shown in fig.7. The effect of stenosis on the velocity profiles is not clear in the cases of low frequency ventilation, but the effects of the 2nd junction similar to the results shown in fig.6 (b) are observed. On the other hand, the effect of the stenosis in HFOV appears clearly as shown in this figure. The asymmetric velocity distribution at the downstream side of stenosis causes the different flow rates in the branches of the 20th generation as shown in fig.3 (d). It is assumed that the difference of flow rate induces phase lag of ventilation in branches of the 20th generation. However, it was not clarified in this investigation.

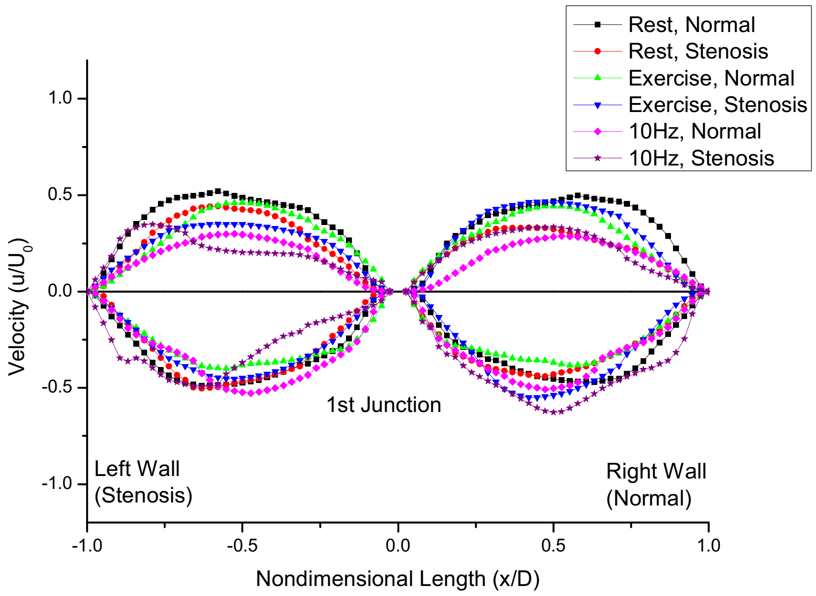


Fig. 7 Velocity profiles at section 7.

4. Conclusion

The effect of flow resistance produced by channel stenosis on the oscillatory airflow simulating the ventilation flow in the part of respiratory bronchiole of human lungs has been experimentally investigated by using the multi-bifurcated micro channels as the model of bronchiole for different ventilation conditions.

The results of 2D velocity measurements of the flows in the model channels with μ-PIV show the changes of flow patterns in the whole field of the channels with the increase of ventilation frequency and with the existence of channel stenosis. As the effects of the increase of frequency, the velocity profiles different from those in Poiseille type flow and phase lag of the ventilation flow from the variation of ventilation pressure were observed. And as the effects of stenosis, it is considered that the time constant difference between branch units with and without stenosis was generated. It was verified by the generation of pendelluft at phase of switching from expiration to inspiration.

The detailed estimations of the effects of ventilation frequency and flow resistance are required for the quantitative analysis of gas transfer and exchange in the end part of airways of human lungs, but it has not been accomplished in the investigation because the experimental difficulties still remain in the measurements of oscillating air flow in micro scale channels.

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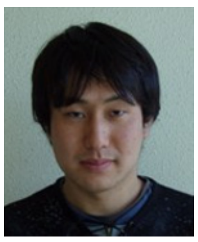

Yusuke Iwasaki: He received his Master degree in Mechanical Engineering in 2008 from Saitama University. The subject of his Master Degree Thesis was the experimental analysis of pendelluft generated in a micro channel as a model of bronchiole with different terminating conditions of compliance.