Investigation of Gas Redistribution in Doubly Bifurcated Respiratory Channel of Human Lung*

Mahtab Uddin AHMMED**, Hiroyuki HIRAHARA***, Tomonori YAMAMOTO** and Katsuya IWAZAKI**

**Graduate School of Science and Engineering, Saitama University,
***Division of Human Support and Production Sciences, Saitama University
Shimo-Okubo-255, Sakura, Saitama, 338-8570, Japan
E-mail: hhira@mail.saitama-u.ac.jp

Abstract
An oscillatory flow in human lung was investigated with a real scale model channel experimentally. One objective of the present study is to clarify the gas exchange mechanism in the peripheral channel, e.g. from the 18-th to 20-th generations, which are the leading region of gas exchanges. The second object is the inspection of the flow convection under the HFOV (high frequency oscillatory ventilation) condition. The experiment was carried out with a real scale two-dimensional model of airway, instead of using a similarity model, in order to take into account the practical condition of HFOV. The velocity profiles were obtained from ensemble mean measurements and the air trajectories were demonstrated with the ensemble mean velocities in the test section. It was confirmed that the air trajectories of expiration were deviated from those of inspiration in the intermediate zone of our channel. It was predicted that a convective flow mixing might be took place depending on the geometric configuration and the asymmetric compliance ratios of the bronchioles.

Key words: Human Lung, Respiratory Region, HFOV, Particle Image Velocimetry, Pendelluft, Alveoli, Air Trajectory

1. Introduction
Respiration flows in mammal’s lung seems to be a simple reciprocal motion, whereas the molar fraction of O₂ and CO₂ in the lung is sustained suitable values within a severe condition through the respiration even a small tidal volume. The lung is morphologically consisted of complex ramifications, which are successively bifurcated from trachea to the alveolar zone. Since in-vivo inspection of pulmonary flows is seemed to be impossible due to the complexities, it should be conducted by a suitable modeling, analogical analysis or numerical calculations. In particular, the studies related to gas mixing in the alveolar pulmonary airways need not only numerical approach but also experimental one. The important physical parameters in the respiration are tidal volume and frequency. The former is most important to consider the gas exchange and its ratio to the dead volume of lung determines the penetration distance of fresh air. The latter also influences on the gas exchange rate and the combined effect of those may determine the active region in the gas exchange area. In contrast to the normal breathing condition, the pulmonary condition may sometimes be different. When one has a disease in lung, the treatment should be changed depending on the patient’s condition. For example, when an infant is suffering from respiratory distress syndrome (ARDS), high frequency oscillatory ventilation (HFOV) is
mostly effective as an artificial ventilation support \(^{(1,2)}\). Recently, HFOV has been applied to adult treatment and it was gotten good results \(^{(3,4)}\). HFOV has a feature as minimal pressure fluctuations with a small tidal volume, which can avoid the risk of lung injury associated with cyclic opening and closing of alveolar elements, especially, for very low birth weight infants. The effectiveness of HFOV has been proved in the clinical fields, however, the treatment condition was determined empirically. Then, a fundamental understanding of gas exchange should be required in order to improve the clinical treatment and its application. For the usual treatment, the reciprocating piston of HFOV system is driven with small displacement volume and at relatively higher cycle rather than a normal one, as 10 to 20 Hz. As the effectiveness of HFOV has been recognized, many studies of respiratory flow in HFOV have been carried out both experimentally and theoretically. In the previous research, e.g. Chang considered five gas transport mechanisms during ventilation in HFOV \(^{(5)}\), and seven potential mechanism were described by Duval et.al \(^{(1)}\). The important mechanisms of those might be, direct alveolar ventilation, convective mixing due to inhomogeneous time constants, convective transport for asymmetric inspiratory and expiratory velocity profiles, longitudinal dispersion due to turbulent eddies, and molecular diffusion by laminar flow and radial mixing.

On the other hand, macroscopic analysis has been carried out with a lumped-parameter model. Otis et al. presented the mechanical behavior of the lungs by the lumped-parameter model of RC circuit analogy \(^{(6)}\). He showed that the breathing frequency is an important parameter in distribution of ventilation when time-constants of the separate pathways are different. The RC model was expanded to RLC model introducing the inertia effect for high frequency by Ultman et.al.\(^{(7)}\) and High et.al. \(^{(8)}\) They studied the pendelluft flows during ventilation at tidal volumes of 5 to 15 mL and frequencies, \( f = 6 \) to \( 26 \) Hz. They concluded that the asymmetry in compliance and inertance produced greater pendelluft than an asymmetry in resistance. They also inferred, Pendelluft does not occur in symmetric airway bifurcations. The RLC models are linear model whereas there can be many sources of nonlinearity in the respiratory system such as the relationship between pressure and volume. The investigation was developed by considering stability due to the nonlinearity effect in RLC model by Feng and Poon \(^{(9)}\). They conducted stability analysis to show the pendelluft phenomenon in a symmetric airway bifurcation. A nonlinear lumped-parameter model for asymmetric bronchial bifurcations has been calculated as further extension of investigation by Elad et.al. \(^{(10)}\). The time-dependent expressions for both of airway resistance and compliance and pressure loss for the mixing were derived in their paper. They also showed that asymmetric compliance of peripheral airways might affect the flow distribution in daughter tubes and induce a larger amount of pendelluft.

In order to study the influence due to the anatomical configuration, the inspiration process for symmetric multi-generations bronchial tree was studied numerically for G0 to G3 by Andrade et.al. \(^{(11)}\). They concluded that flow distributions were significantly heterogeneous due to the inertia effect at \( Re = 4800 \) and quite uniform at low \( Re = 150 \). The effective diffusion is greater in curved and bifurcated tubes than in straight tubes for secondary flow during HFOV conditions \(^{(12-14)}\). Thereafter, particle deposition in the distal pulmonary regions of human airways was investigated \(^{(15-17)}\). In their papers, they calculated the deposition fraction and particle loss due to deposition in each generation on inhalation. The respiratory flows along realistic and idealized model channel were investigated simultaneously. The amount of secondary flow was higher in realistic geometry than in the idealized geometry with quite different structure \(^{(18)}\). The flow structure in each generation might be influenced by the flow from higher and lower generations \(^{(19)}\).

A lot of experimental and modeling works have been studied concerning the flow mechanisms in pulmonary region of human lung. CFD simulation is alternative strong tool to validate the study with experimental or numerical work \(^{(20-22)}\). Nagels and Cater
investigated the suitability of using LES within trachea and bronchi and they showed the occurrence of pendelluft numerically \(^{(20)}\). Ma et al. simulated the air flow in upper airways and alveolated airways successfully \(^{(21)}\). The dissimilarity in particle trajectories was found from experiment and CFD simulations due to different wall conditions for impaction. Fujioka et al. showed the influence of channel curvature on the secondary swirling flow \(^{(14)}\). Furthermore, Inagaki et al. investigated the oscillatory flow in realistic model human airways and they concluded the expiration flow strongly depend on the airway geometry \(^{(22)}\).

As described above, pendelluft flow as bulk flow, convective mixing due to the different time constants of the branches, and asymmetric flow at inlet and outlet flows may be important in the investigation of HFOV. In the present study, the fluid dynamical mixing in multi-bifurcated bronchial tubes was investigated with phase averaged velocity measurement experimentally. Especially, the flow in intermediate channel between the branches was discussed in detail. The ensemble mean velocity vectors were obtained with PIV technique and Lagrangian path lines of virtual particles were calculated from these data. Thus, the air trajectories were reconstructed and their deviations for mutual flow were discussed. The phase delay obtained from experimental data and RLC circuit analogy was also compared and the asymmetric flow pattern was discussed in terms of out-of-phase.

2. Anatomical Background and Oscillatory Flow Characteristics

According to the respiratory flow characteristics, the lung is mainly divided into two regions, that is, air transport and gas diffusion regions. The airways from trachea, G0, to the terminal bronchioles, G16, are bifurcated repeatedly and gas is transported without any gas exchange. Here, G represents generation. The volume of this region is called as the anatomical dead space. The respiratory zone from G17 to G23 with respiratory bronchioles, alveolar ducts and alveolar sacs plays a role of gas exchange. This region contributes in gas exchange between O\(_2\) and CO\(_2\). A numerous tubes from the terminal bronchiole to the most distal alveolus contain about 2.5 to 3 liters of air whereas the standard adult lung contains about 5 liters of air. Active gas mixing and exchange will be expected in the alveolar respiratory region to maintain the effective molecular diffusion. Therefore, the air flow from respiratory inlet region, G17 to G20 is very interesting to distribute a fresh air to distal alveolus. In this reason, we will inspect the fluid dynamical mixing in G18, G19 and G20 with a two-dimensional model in real scale as shown in Fig.1. The model channel was designed to investigate the convective flow between the branches without any three-dimensional influence such as a secondary swirling flow \(^{(14)}\).

On contrary to the normal breathing, when we use HFOV for the treatment, an oscillatory flow is induced in the deep region of lung with high frequency. In general, the
oscillatory flow is characterized with Stokes layer thickness, \( \delta = \sqrt{\nu/\omega} \), where \( \nu \) and \( \omega \) (=2\( \pi f \)) are kinematic viscosity and angular frequency of oscillatory flow, respectively. And also, the flow is characterized with three fundamental dimensionless parameters, i.e., Reynolds number, \( Re = UD/\nu \), Womersley number, \( Wo = (D/2)/\delta \), and Peclet number, \( Pe = Sc \cdot Re = UD/\alpha \). Where, \( U \), \( D \), \( \alpha \), and \( Sc \) are representative velocity, channel width, molecular diffusivity and Schmidt number, respectively. \( Wo \) and \( Pe \) represent the ratio of channel half width to Stokes layer thickness and that of convective speed to molecular diffusion speed. Figure 2 shows the typical values of \( Re \), \( Wo \), and \( Pe \) in human adult lung under the HFOV conditions when a tidal volume, \( TV = 150 \text{mL} \) and \( f = 10 \), 15, and 20Hz. Here, the value of tidal volume was determined from Fesseler et.al. \(^{(23)}\). On the other hands, the dead space from G0 to G17 is about 220mL \(^{(24)}\). The above tidal volume is less than the dead space of G0 to G17, so the fresh air does not reach to the respiratory region directly. Therefore, the investigation on the air flow near G18 as the starting region of gas exchange is very interesting subject. As shown in this figure, \( Re \) decreases from 20000 at the trachea to 10 at terminal bronchus and is less than 0.8 in the alveolar region in the HFOV driving condition. At a lower value of \( Wo \) less than 10, the flow shows almost laminar velocity profile without any remarkable phase delay between the center and the near wall of the channel. The value of \( Pe \) is of order of \( 10^6 \) to \( 10^7 \) in G19. It means molecular diffusion plays an important role as well as convection. In the present paper, we will discuss the enhancement of gas mixing in the oscillatory flows between the branches due to the convective effect.

3. Experimental Techniques and Setup

As shown in Fig.1, a doubly-bifurcated bronchial model channel in real scale was used in the present experiment. According to the respiratory morphologies, a doubly-bifurcated symmetric channel was manufactured by a machining. Each branch of channel was fabricated axially symmetric. The surface roughness of finish of the machining is less than 1 \( \mu \text{m} \), which is better than a laser processing. The channel is made of aluminum and consists of three generations, G18, G19, and G20. The surface is black anodized aluminum in order to reduce the light scattering. The channel widths of G18, G19 and G20 are 500, 450 and 400\( \mu \text{m} \), respectively, where channel depth is 500\( \mu \text{m} \). The length of G19 is 1.2mm along the centre line and those of G18 and G20 are long enough to reach up to fully
developed flow. The angles of upper junction and lower junctions are 70 and 60 degrees, respectively.

HFOV provides a minimal pressure fluctuation with small tidal volume and high frequency for the patients with lung's disease. The HFOV frequency is about 50 times larger than the normal breathing, and the tidal volume is much smaller than dead space volume of human lung. The usual pressure amplitude is 500 Pa at the trachea. Due to the pressure loss by the wall resistance of the bifurcated tubes, the pressure amplitude decreases towards the deep region. According to the results in a preliminary experiment, the flow rate in G18 was matched with a practical adult condition at $P_0=30$ Pa at inlet of the test channel. In this condition, the representative maximum flow speed is about 0.15 m/s at G18 and the corresponding tidal volume at trachea is 150 mL. The present experiments were carried out under this pressure condition, and the driving frequency was 10 Hz.

Figure 3 shows the schematics of the experimental setup. The fresh air is supplied by a linear driver to the channel via buffer tank. A micro-manometer (KYOWA EIC, PDV-50A) was mounted between the test channel and buffer tank. The signal from micro-manometer was transferred to PC and a timing unit for phase locked image acquisition in PIV. The closed compliant tubes (TYGON R-3603), which represent the alveoli, are connected to the terminal of G20’s branches. We adjust the compliance by changing the tube length. Table 1 shows the compliance of TYGON tubes in preliminary test. The test was carried out under a static pressure condition. After a tube was filled by a specific volume of air with a syringe, the static pressure was measured and the compliance was calculated from the charged volume and the measured pressure. The result was summarized in Table 1.

### Table 1 Compliance of tubes.

<table>
<thead>
<tr>
<th>Tube length</th>
<th>Compliance $[\mu L/\text{Pa}]$</th>
<th>Compliance per length $[\mu L/\text{Pa cm}]$</th>
<th>Averaged $[L/\text{Pa cm}]$</th>
</tr>
</thead>
<tbody>
<tr>
<td>40 cm</td>
<td>$3.42 \times 10^{-2}$</td>
<td>$8.55 \times 10^{-4}$</td>
<td>$8.11 \times 10^{-10}$</td>
</tr>
<tr>
<td>100 cm</td>
<td>$7.68 \times 10^{-2}$</td>
<td>$7.68 \times 10^{-4}$</td>
<td></td>
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The experiment was conducted for two different compliance combinations, which are $[C_1; C_2; C_3; C_4] = [1:1:1:1]$ and $[10:10:1:1]$. Here, the average compliance at G20 is calculated as

$$C_{20} = C_L \times \frac{V_{20}}{V_L} = 0.2 \times \frac{(5850 - 720)}{5850}$$

$$= 2.0 \times 10^{-7} [\text{L/cmH}_2\text{O}]$$

Here, the average compliance, $C_L$ was estimated from the previous reports $^{25,26}$. The total lung volume for standard adult, $V_L$ is 5850 mL and a dead space volume is 720 mL. A unit compliance in the present experiment is $C_0 = 2.4C_{20}$, which was determined to represent the suitable flow rate in HFOV driving as described above. The flow velocity in the channel was controlled by adjusting the piston displacement when the pressure amplitude was 30 Pa at the inlet of the channel. In the present experiment, the stroke volume for piston displacement was 5 and 4.5 mL for $[C_1; C_2; C_3; C_4] = [1:1:1:1]$ and $[10:10:1:1]$, respectively. Under these conditions, the representative peak flow rate was 25 mm$^3$/s, so the tidal volume in G18 was about 0.4 μL.

Figure 4 shows the arrangement of PIV measurement. A CMOS camera (IDT,
XS-5, 1280H×1024V pixels in resolution, 10 bits in intensity depth) was used for the image acquisition. For the micro-PIV observation, a macro lens with long working distance (Keyence, VQ-Z50A) was used. The focal depth is 44 μm, which is small enough to obtain the two-dimensional flow image along the axis of the micro channel. The micro channel was aligned as normal to the camera axis, which was carefully adjusted with a micro tilting stage. A double-pulsed Nd-YAG laser (NewWave, SoloIII, 532nm in wavelength, 50ml/pulse), a timing hub, and a computer for image capturing were employed. The test section was illuminated from the backside at the angle of 30 degrees inclination from horizontal plane. The angle was determined to record the maximum scattering intensity of forward scattering from tracer. Sub-micron oil-mist generated by Laskin nozzle was used as a tracer. The nominal diameter of the tracer is 1 μm and the diameter of the airy disc is 40 μm. The photographic image is much larger than the real dimension of particle. Also, the particle diameter satisfy sufficiently the limiting condition of traceability for the flow fluctuation, because the time response is less than $10^{-5}$ sec for 1 μm diameter (27). The interrogation size in the correlation calculation was 64×64 pixels. The correlation calculation is based on FFT algorithm. The tracer was introduced into the buffer tank before the test. After filling it, the PIV measurement was started. 100 pairs of phase locked images were obtained for each delay, thus the ensemble velocity averaging was executed with these data. The phase locked data were acquired by shifting the trigger signal delayed by $1/16T$ step. As shown in Fig.2, $Re$ number is less than 10, so the flow is presumed quasi-steady. Also, the scale of turbulence and velocity fluctuation is estimated as $0.95D$ and $0.21U$ for $Re=5$ from the turbulent theory (28). These values imply that the flow will not include the higher order turbulence practically. In the present paper, we are aiming to discuss the flow deviation from the reversible flow. According to the above estimation, we decided the image acquisition period as $1/16T$.

4. Results and Discussions

4.1 Ensemble Mean Velocity Maps

The PIV experiments were carried out at 10 Hz in HFOV driving, so the flow period, $T$ was 100ms. The raw image is shown in Fig.5 as an example. The ensemble mean velocity vectors are shown in Fig.6 for homogeneous compliance condition and Fig.7 for inhomogeneous one.

In the homogeneous alveoli, there is no phase delay in velocity profile between the center and near wall regions due to the small $W_m$, which is 0.511 at G18. On the other hands, in the inhomogeneous case, a fast flow is observed in the left channels. Here, the uncertainty of velocity was $\overline{v} \pm 1.8\%$ at most. $\overline{v}$ represents the local time mean velocity. And it was not observed local vortical flow in the channels. Comparing these figures, the flows were almost laminar and no vortical structure was observed in both conditions. Where, as stated by Chang (5), even though the channel is symmetric, a deviated convective transport was observed in the channel. Especially in G19L and G19R, the flow was faster in the outside than in the inside at the inlet, and vice versa at the outlet. It is considered that this deviation is aroused due to the inertia force. These convective transport deviations will be discussed in §4.4 to §4.6.
Alternative result for inhomogeneous compliance of $[C_1: C_2: C_3: C_4] = [10:10:1:1]$ is

(a) $t=0/16T$

(b) $t=4/16T$

(c) $t=8/16T$

(d) $t=12/16T$

Fig. 6 Ensemble velocity maps at $f = 10$ Hz and $[C_1: C_2: C_3: C_4] = [1:1:1:1]$

Fig. 7 Ensemble velocity maps at $f = 10$ Hz and $[C_1: C_2: C_3: C_4] = [10:10:1:1]$

Alternative result for inhomogeneous compliance of $[C_1: C_2: C_3: C_4]= [10:10:1:1]$ is
shown in Fig.7. The flow shows the asymmetric pattern at this condition. At \( T_{1/6} \), the maximum flow rate in inspiration period, a fast flow was observed along G18 \( \rightarrow \) G19L \( \rightarrow \) G20LR. Inversely, at \( T_{3/6} \), the maximum flow rate in expiration period, a fast return flow was observed along G20LR \( \rightarrow \) G19L \( \rightarrow \) G18. Between the inspiration and the expiration, a pendelluft as bulk flow was observed as depicted in Fig.7 (a) and (c). An incoming flow at G19R in Fig.7(a) and an outgoing flow at G19L in Fig.7(c) were clearly observed. In this condition the uncertainty of measured velocity was \( \pm 4.2\% \) in G19L and \( \pm 1.9\% \) in other channels. The increase of error in G19L may be arisen due to low density of tracer particles. In fact, the tracer density decreased with time in fast flow. The problem had not been fixed in the present experiment.

4.2 Velocity Profile and Flow rate for Homogeneous Alveoli

In this section, we will discuss a deviation of the velocity profiles at each branch in inspiration and expiration. Ensemble mean velocity profiles at G18, G19L and G19R are shown in Fig. 8 (a), (b), and (c) at 16 phases for \( f=10 \) Hz and \([C_1:C_2:C_3:C_4]=[1:1:1:1]\). The velocity profiles were interpolated with ensemble velocity maps at the sections A, B and C, which were shown in Fig.6(a) with red lines. The velocity was normalized with the maximum velocity in G18 and span-wise distance with the corresponding width of the channel, respectively. As shown in the velocity profiles of the parent channel G18, the velocity profile is almost parabolic, i.e. there is no phase delay between the flows in the center and near the wall because of low \( W_o \) number.

Comparing the velocity magnitude at \( x/x_{max}=0.2 \) and 0.8 in Fig.8 (b), the inlet flow near the left wall is faster than that near the right side in G19L, and vice versa in G19R. The
main flow through G18 collides on the first junction and stagnates. Therefore, it becomes decelerated near the junction and accelerated in far side wall. This slight flow deviation is induced by the difference between the inertia and viscous forces. Such deviation from the symmetric profile is small but the air trajectory is influenced even the small $Re$ number as shown in the next section.

Figure 8(d) shows the flow rate variation in G18, G19L and G19R in homogeneous compliance condition. Here, the flow rate was calculated from the velocity profile data in Fig. 8(a), (b), and (c). Since it is very difficult to take a three dimensional effect into consideration for the different channel width, the flow rate was expressed for a unit depth. We can see a short phase delay in the flow rate. The delay was well recognized from the figure as the zero flow rate such as the beginning of inspiration or expiration. Unfortunately, it was very difficult to confirm the conservation of mass flow at the moment from the flow rate curves even though a lot of experimental run were conducted. The error may be aroused by a de-correlation by the background noise due to the defocused particles or the slight deviation of the observation plane from the channel center axis, while the periodic flow does not show the considerable phase delay.

4.3 Velocity Profile and Flow rate for Inhomogeneous Alveoli

In this section, we will discuss about inhomogeneous distribution of alveoli. We set the asymmetric compliance ratio of $[C_1; C_2; C_3; C_4]=[10:10:1:1]$ as asymmetric distribution. A large compliance ratio was set to compare the flow phenomena with the symmetric one. Figure 9 (a), (b), and (c) show the time variation of velocity profiles of section B, C and D, respectively. The velocity profiles of section B and C, G19L and G19R shows a typical pattern in asymmetric case. Comparing Fig. 9 (a) and (b), the velocity in left channel, G19L is much faster than that in G19R. Also, the velocity deviation described in the previous section was much emphasized in this case. As shown in Fig.9 (a), the flow speed in $x/x_{\text{max}} < 0.5$ is faster than that in $x/x_{\text{max}} > 0.5$. The similar tendency is shown in Fig. 9 (c). The peak velocity in G19L was about 5 times larger than that of G19R whereas the compliance ratio was 10. The velocity curves in G19R were more flat for small convection.

Fig. 9 (a), (b), and (c) are time variation of velocity profiles. (d) shows flow rate comparison during the respiration at $f=10$ Hz and $[C_1; C_2; C_3; C_4]=[10:10:1:1]$. 
These profiles show the existence of the asymmetric bulk flow in the inspiration and expiration. Fig. 9 (d) shows the flow rate at sections A, B, C and D. The flow rate curves show the phase delay due to the pendelluft as shown in Fig. 7 (a) and (c). In the present inhomogeneous compliance, a bulk flow due to pendellift was 13.4 mm$^2$/s in G19L at $t=0.05S$ and 12.1 mm$^2$/s in G19R at $t=0.1S$. According to the lumped parameter analysis, the phase delay was calculated as 0.93 rad (53 degrees). The same order of phase delay was shown in the experimental result.

### 4.4 Air Trajectory Demonstration

The air flow in lung is so complicated that a consideration for the air streaming based on the ensemble mean flow is valuable to understand the air mixing mechanism. If we know the air trajectory during the respiration, we can discuss the fundamental mechanism of flow mixing of the first order of the respiratory motion by using the experimental results. Therefore, an air trajectory analysis was conducted with the ensemble vector maps obtained for 16 phases. In order to reconstruct the air trajectories from PIV data, we have to expand the experimental data into time axis. Afterward we can follow the particle location within a suitable time resolution. For this sake, we prepared seven vector maps which were interpolated with a cubic spline interpolation function between each phase data as shown in Fig. 10. The virtual air particles were tracked by using these data as well as
with a cubic spline interpolation. The calculated trajectory was shown in Fig. 11. Fig. 11(a) and (b) shows the air trajectories during one cycle of respiration for homogeneous and inhomogeneous compliances condition, respectively. Red lines show the inspiration phase and blue lines the expiration phase. The virtual particles were distributed initially on the line at $y=1.0$ in Fig. 11(a) and $y=2.0$ in Fig. 11(b), respectively. In the case of homogeneous compliance, the particles show almost laminar motion during the inspiration and expiration period as shown in Fig. 11(a). However, the particles draw curves near the turning point, especially close to the branches. Then, some particles returned into G18, and others remained in G19L and G19R. On the other hand, in the case of inhomogeneous compliance, almost air entered into G19L. The trajectories show the laminar behavior as well as for homogeneous case, whereas the path lines were not same in inspiration and expiration as seen in G19L. The slow stream in G19R draws curve and some particles shows the complicated locus. From these consideration of trajectory tracking, the air will take place a dynamic mixing during the repeated periodic motion.

4.5 Air Trajectory Deviation for Homogeneous Compliance

In this section, we will discuss the peculiar path lines near the second branches. The instantaneous response of corresponding pressure and flow during oscillatory flow in mother tube, G18 shows the features of unsteady Poiseuille flow. The flow in daughter branches showed different characteristics as phase delay and convective deviation. The air trajectories were demonstrated in Fig. 12 for homogeneous compliance condition, $[C_1; C_2; C_3; C_4] = [1:1:1:1]$. Three curves were drawn from the different initial position. In these figures, the trajectories from $t=4/16T$ to $16/16T$ are shown. The trajectories show the regular inlet and outlet motion but U-shape curves near the bifurcation. Although the test channel is symmetric, the geometry is not symmetric about the center axis of G19L and G19R. Then the inlet and outlet velocity profiles are not symmetric completely, so the air path draws the peculiar curved shape. Especially, as shown in this demonstration, when the air turns in the vicinity of the head of the branch, the air may turn with the larger radius of curvature. It will be concluded that the air path is deviated from the ordinal flow pattern due to the convective effect that is aroused by the inertance. Here, let us consider the lumped parameter method, which is used in the analysis for bulk flow in network flow system. The differential equation is fundamentally expressed for the any section as the following.

$$P_0 \sin \omega t = R_k Q_k + L_k \frac{dQ_k}{dt} + \frac{1}{C_k} \int Q_k \, dt$$

(2)

Where, $P_0$, $Q_k$, $R_k$, $L_k$, and $C_k$ are pressure amplitude, flow rate, resistance, inertance and compliance of the k-th channel, respectively. This equation treats a bulk flow from the
macroscopic view, and gives us information about the contribution of wall resistance, capacitance and flexibility of tube and alveoli. Especially, the inertia is related with the acceleration terms in Navier-Stokes equation. Also we can estimate the phase delay and the starting time of pendelluft as a bulk flow as described in the previous section. Now, we note the effect of the inertia in this equation. Supposing that $Q_k$ is fluctuated sinusoidally, $Q_k$ becomes zero at $\omega t=0$ and $\pi$. At this moment, the pressure change by resistance is zero. On the other hand, the second term in Eq.(2) is not zero because the time derivative of $Q_k$ has the maximum. From this inference, the flow is not at rest at the moment and will generate a fluctuation depending on the local pressure distribution due to the time derivative of inertia difference. Based on this reason, it may conclude that the trajectory deviation shown in the Fig.12 was aroused by the change of inertia in the flow field.

4.6 Air Trajectory Deviation for Inhomogeneous Compliance

Fig.13 shows the air trajectories for inhomogeneous compliance condition, $[C_1; C_2; C_3; C_4] = [10:10:1:1]$. As well as in Fig.12, three curves were drawn from the different initial position. In this condition, the flow in G19L is faster than that in G19R. In Fig.13(a), the particles enters and return through G19L channel. It was found that the inlet and outlet path was different and its deviation was larger near the head of the bifurcation. On the contrary to this motion, the flow in G19R is slow. The trajectories show the complicated lines without tracking the reversible lines through inspiration and expiration. In this figure, the particle (1) in Fig.13(b) shows to be faltering. This particle entered downwards and returned upwards at $t=8/16T$ with short delay. Then it was pull back at the end of expiration ($t=16/16T$). The behavior indicates the influence of pendelluft. In this condition, the particle motion was not laminar. The complexity of the particle path may be generated by the mutual effect of pendelluft and inertia. As discussed here and previous section, the air trajectory was deviated from the reversible motion. The deviation was induced neat the zero velocity phase. Although the mixing ratio of old and fresh air has not been deduced from the experimental results, the important mechanism of redistribution during the oscillatory flow was demonstrated practically. The estimation of gas mixing should be carried out as based on these physical processes.

5. Conclusions

A velocity measurement in a model channel of human lung was carried out by using PIV technique under the HFOV driving condition. The doubly-bifurcated model channel was designed in real scale from G18 to G20 to investigate the gas mixing mechanisms. The experiment was carried out for two alveolar compliances, $[C_1; C_2; C_3; C_4] =$
[1:1:1:1] for homogeneous compliance and [C_1; C_2; C_3; C_4] = [10:10:1:1] for inhomogeneous one. The experimental velocity profile and flow rate were evaluated with ensemble averaging. Also, the air trajectories were demonstrated with ensemble averaged velocity data in order to discuss the relation between the air motion and respiration phase. For homogeneous case, comparing the velocity profile during the respiration period, in the parent tube G18, the flow shows the almost parabolic, but in daughter tubes, the velocity profile shows a slight deviation. It is found that the deviation was induced by the difference between the change of inertia and viscous forces even for homogeneous compliance. Such deviation from the symmetric profile is small but the air trajectory is influenced even the small Re number. The mixing process was demonstrated with virtual trajectory tracking. For inhomogeneous case, the velocity vector maps show the asymmetric flow and the occurrence of pendelluft. The demonstrated trajectories show the faltering motion. The complexity of the particle path may be generated by the mutual effect of pendelluft and inertia variation.

References


