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Investigation on the Respiratory Airflow in Human Airway by PIV

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Abstract: The creation of the accurate transparent flow passage is essential to analyze the flow inward a geometrically complex flow passage like human airway by PIV. We established the procedure to create a transparent box containing a model of the human airway for PIV measurements. A flow passage includes the whole human upper airway, nasal cavities, larynx, trachea, and 2 generations of bronchi. The phase averaged mean and RMS velocity distributions in sagittal and coronal planes are obtained for 7 phases in a respiratory period by tomographic PIV. Some physiologic conjectures are obtained. The main stream went through the backside of larynx and trachea in inspiration and the frontal side in expiration.

Keywords: Tomographic PIV, Biomedical flow, Human airway, CT, Physical Model

1. Introduction

The Knowledge of airflow characteristics in total airway is essential to understand the physiological and pathological aspects of nasal breathing. Several studies have utilized physical models of the nasal cavities to understand the patterns of the airflow (Scherer et al. 1989, Hess et al. 1992, Hopkins et al. 2000). Though the physical models become more sophisticated and PIV experiments on these models become popular recently (Kim and Son 2002, Doorly et al. 2006, Taylor et al. 2006), most investigators deal with models of airway only in part under constant flow condition. There have been some articles that have investigated the airflow of the mouth breathing both experimentally and numerically (Heeman et al. 2003, Liu et al. 2003, Johnstone et al. 2004.) The investigation on the airflow of nose breathing in a whole airway is very rare due to its geometric complexity. In this article, we investigate airflow in a whole airway including nasal cavities, larynx, trachea, and two generations of bronchi under periodic flow condition.

Creating the accurate transparent flow passage is essential to analyzing the flow inward a complex flow passage by PIV. Kim and Son (2002) improved the procedure to produce the better cavity model with high-resolution CT scan data and the surface rendering. Thin sliced CT data and meticulous refinement of model surface under the ENT doctor's advice provided more sophisticated nasal cavity models. We applied this procedure to create models for abnormal nasal cavities with adenoid vegetation and asymmetry (Kim and Son 2004). Now, we extend this procedure to create a whole upper airway. This model is believed to be one of most anatomically correct one. The models of a nasal cavity and a whole airway are compared with the other models around world, which confirms the adequacy of our models. We design and construct the periodic pumping system to generate periodic flow simulating the physiological data of human respiration.

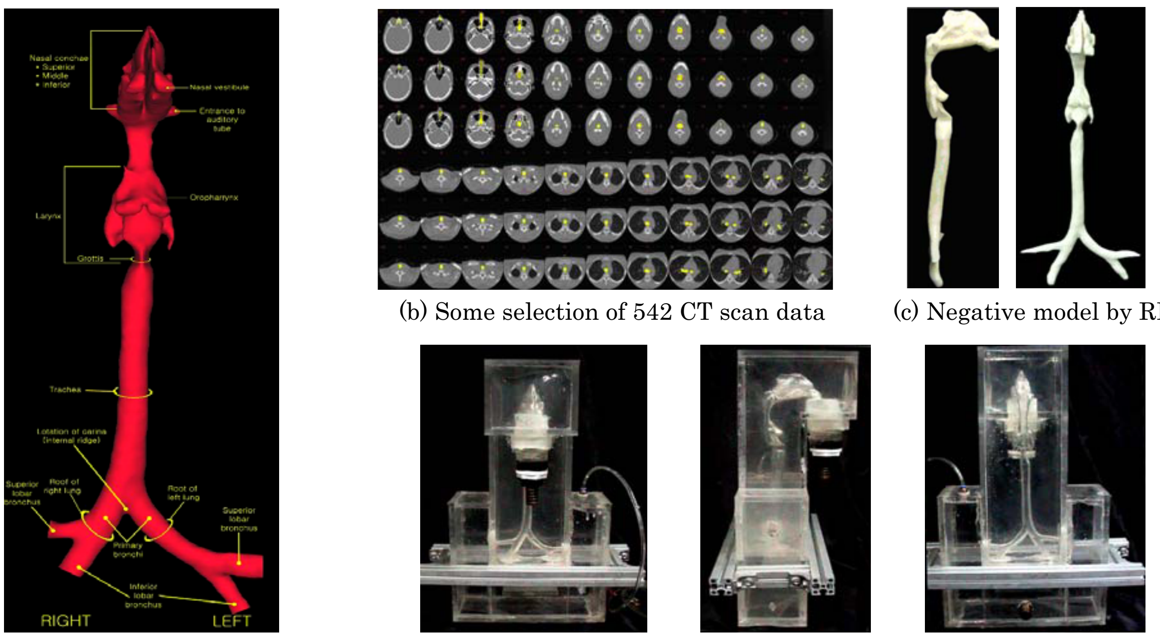
The phase averaged mean and RMS velocity distributions in sagittal and coronal planes were obtained for 7 phases in a respiratory cycle. The tomographic PIV technique is introduced to investigate complex flow structure in nasal cavity. Sixty sagittal sections with 1 mm in thickness are investigated by PIV measurements. The CBC PIV algorithm (Hart 2000) with window offset (64*64 to 32*32) is used for vector searching in PIV analysis (Kim 2001). The three dimensional reconstruction of velocity fields gives insights on flow characteristics of each part. Some physiologic conjectures are obtained. The main stream went through the backside of larynx and trachea in inspiration and the frontal side in expiration. There exist vortical motions in inspiration, but no prominent one in expiration.

Since we published many results on nasal airflows, presentations of results are focused on the airflow near larynx and trachea in this article. The paradigm established in this paper can be applied to many kinds of otorhinolaryngological and airway diseases and is believed to contribute to the diagnosis and treatment including surgical operation of airway diseases.

2. Experimental methodology

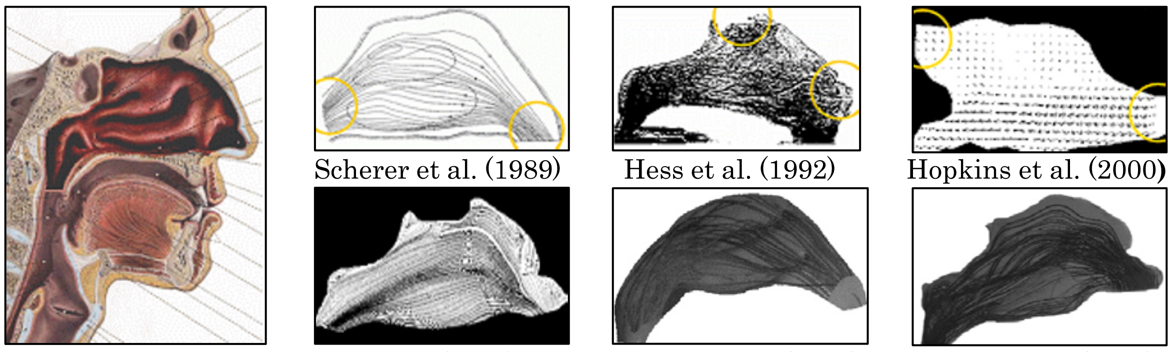
2.1 Creation of flow passage and working fluids

Creating an accurate transparent flow passage is essential to analyze the flow inward a complex flow passage by PIV. The key to producing a geometrically complex flow passage suitable for PIV is the recent availability of a rapid prototyping machine and water-soluble material for a negative model. Rapid prototyping is a well-accepted method for quickly generating replicate prototypes from computer files including CT scan data. A human airway is composed of nasal cavities, larynx, trachea, and 1-4 generation of bronchus, 5-(17-23) generations of bronchi, as shown in Fig.1a. The procedure of creating flow passages is summarized as follows; At first, a solid computer model for building the replicate model is created from coronal CT scan data (Somatom plus 4, Siemens Co., 0.6mm scan rate) of Korean adults (Fig.1b). Then, a replicate prototype of the nasal cavity, made of water-soluble cornstarch, is created by a RP machine (Z Co. MA. USA). This prototype is suspended in a rectangular Plexiglas box. Then, clear silicone is poured around the prototype carefully. After the silicone has been cured in an oven, the cornstarch prototype is removed with cold water. Finally, a rectangular box containing the form of the nasal cavity can be made. To remove the difference in the index of reflectance, the mixture of water and glycerin (52:48 in this case) is used as a working fluid. Due to the large amount of CT scan data (542 scan, 0.6mm increment) and the careful surface rendering, more sophisticated airway model can be made and used in this article. Irrelevant parts, such as sphenoid sinus, are removed to reduce the optical noise. While a 2X model of nasal cavity was made by the aforementioned procedure for the nasal airflow study, an actual size of upper airway is created in this article. For this model, the constant flow-rate assumption is less persuasive from the dynamic similarity condition (Chung et al. 2006). Fig.2 and 3 show existing nasal cavity and upper airway models. Comparing with anatomical figures, the adequacy of our model is prominent. For the upper airway model, it is believed to be one of most accurate model. (Private discussion in World Congress on Biomechanics at Munich, 2006)



(a) Human airway anatomy (b) Some selection of 542 CT scan data (c) Negative model by RP (d) Flow passage after washing out by water

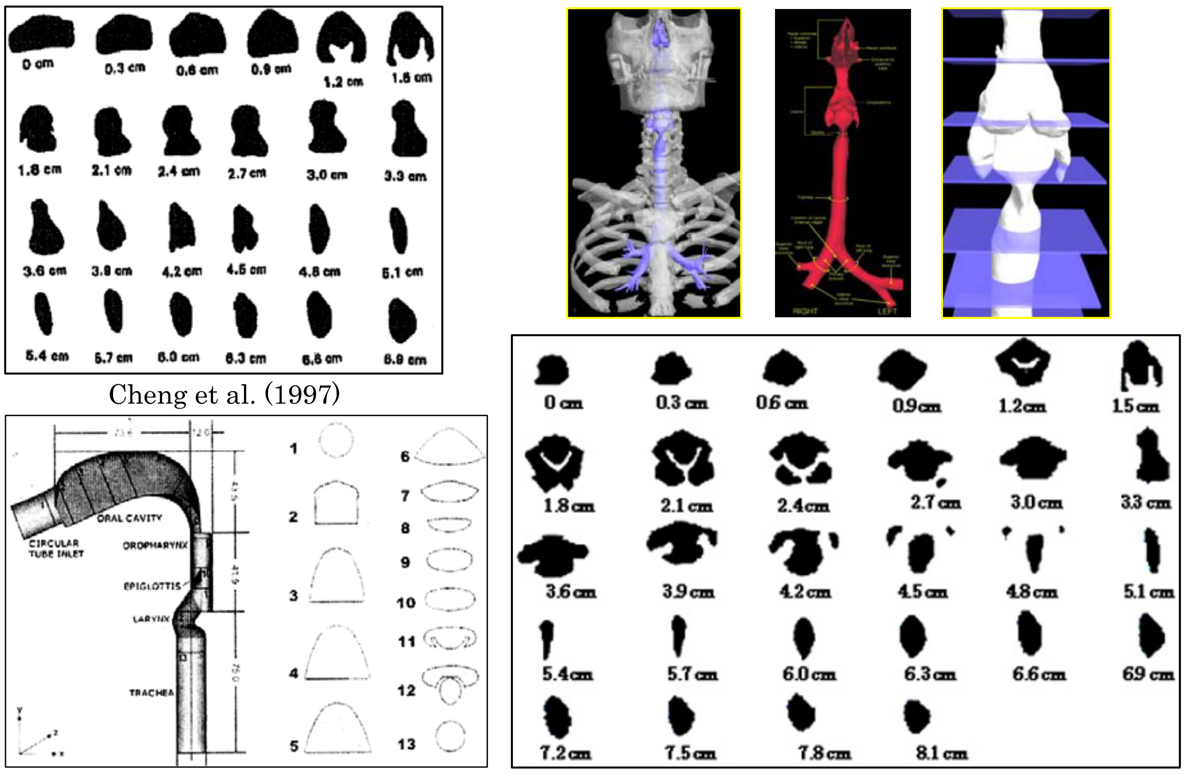
Fig. 1. Human airway anatomy and the flow passage



Scherer et al. (1989) Hess et al. (1992) Hopkins et al. (2000)

Atlas of Anatomy Kim (2002) Kleven et al. (2006) Dooley et al. (2006)

Fig. 2. Comparison of existing nasal cavity models



Cheng et al. (1997)

Johnstone et al. (2004) Our model (Male, 26years old, 65 Kg, 168cm, 2006)

Fig. 3. Comparison of existing human upper airway models

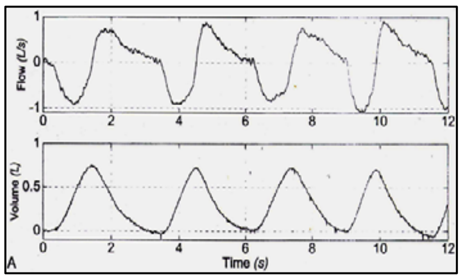
2.2 Piston Pump and Cam

As shown in Fig.4, the main body of the pump is a cylinder with a diameter of 180 mm. The movement of the piston was controlled by the specially made cam, which simulated normal respiration at rest. In designing the cam, the respiratory cycle was divided into 17 unequal portions and the distance of movement was calculated at each portion using the volume curve (from Samsung Medical Center). (Kim et al. 2005) The pump was connected to the model at the chamber that encloses exits of 4 bronchi. The nostril part of the model was connected to the reservoir from underneath. The respiratory cycle was divided into four phases: inspiratory acceleration and deceleration, and expiratory acceleration and deceleration. The PIV measurements were performed at 7 phases a period: the first one was at the change from expiration to inspiration, the second was near the maximum flow rate of inspiration, the fourth was at the change from inspiration to expiration, and the fifth was at the maximum expiration.

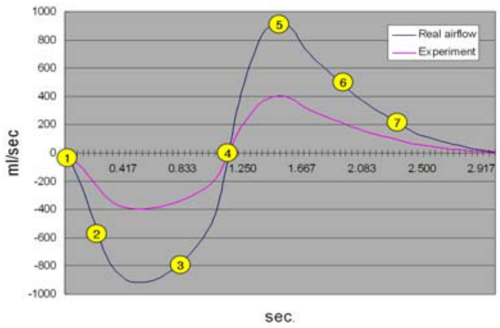
2.3 Experimental set-up

Since the flow passage is too long in shape, the measurement is conducted in three parts; nasal cavity, larynx and trachea part, and bronchiole part. The measurements were performed in 34 sagittal planes of the airway model at 1-mm intervals by the controlled transverse units. Each measurement was repeated 30 times. The time interval between two consecutive images was about 300 μs, depending on the flow rate. Data sampling for phase averaging is controlled by the image acquisition and trigger system (Fig.4e). When marks on a cam pass the light sensor, the 15 volt signal initiates the synchronizer unit which activates the Nd:Yag laser and a CCD camera simultaneously. A schema of the experiment depicts in Fig.5. We calculated the phase averaged velocity and root mean square (RMS) value from 30 vector fields in each time phases (128 by 80 vectors in each section). We try to find the optimal number and found out that results became consistent if the number is bigger than 30. Since Reynolds number is small except for the region near nasal valve and vocal sector, the averaging over 30 realizations is believe to be enough for average velocity fields. Coronal reconstructions for the mean velocity and RMS value were performed at 7 phases to show the results three-dimensionally from two-dimensional results. Also, the results were displayed as phase averaged velocity vector, distribution of RMS, and streamlines at 34 sections in the sagittal plane so as to include the nasal cavity, larynx and trachea. Since we used a mixture of water and glycerin rather than air, the dynamic similarity must be satisfied to have same Reynolds number and Womersley number. The typical value of period of rest respiratory is 3-5 seconds. Among these values, we choose 3 seconds. With a measured density and kinematic viscosity (6.5*10-6 m2/s) of the mixture, as shown in Tab.1, a period in experiments is calculated as 7 seconds to have same Womersley number. Reynolds number, based on a diameter of the external nares, varies from 0 to 800 as respiratory flow rate varies from 0 to 923 ml per second. Polyvinyl spherical particles with 30 m in diameter are used as tracer.

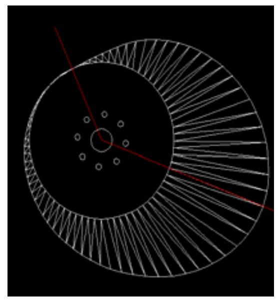
$$\text{Reynolds Number} : Re = \frac{VL}{\nu}, \quad \text{Womersley Number} : Wo = L\sqrt{\frac{\omega}{\nu}}$$



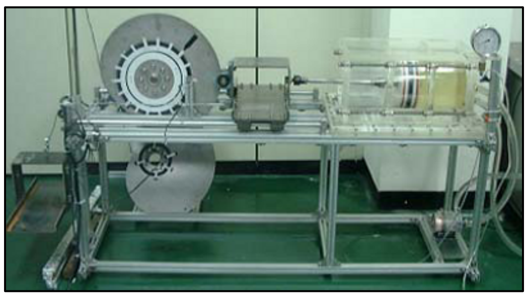
(a) Physiological data of Flow rate and tidal volume for rest respiration



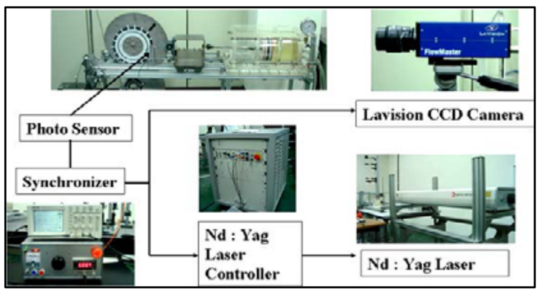
(b) Selected flow rate positions in a period



(c) CAD drawing of Cam



(d) Periodic pumping system



(e) Image acquisition and trigger system

Fig. 4. A periodic pumping system and an image acquisition system for phase averaging

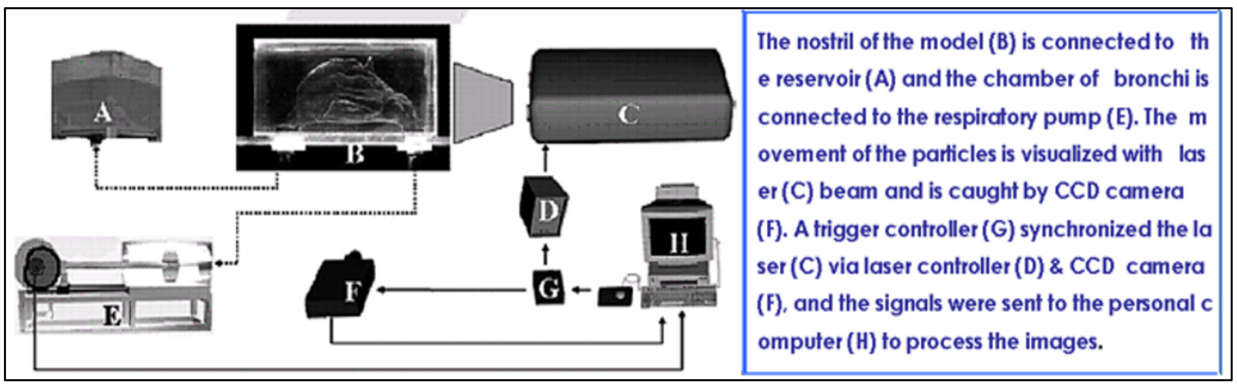


Fig. 5. A schema of the experiment

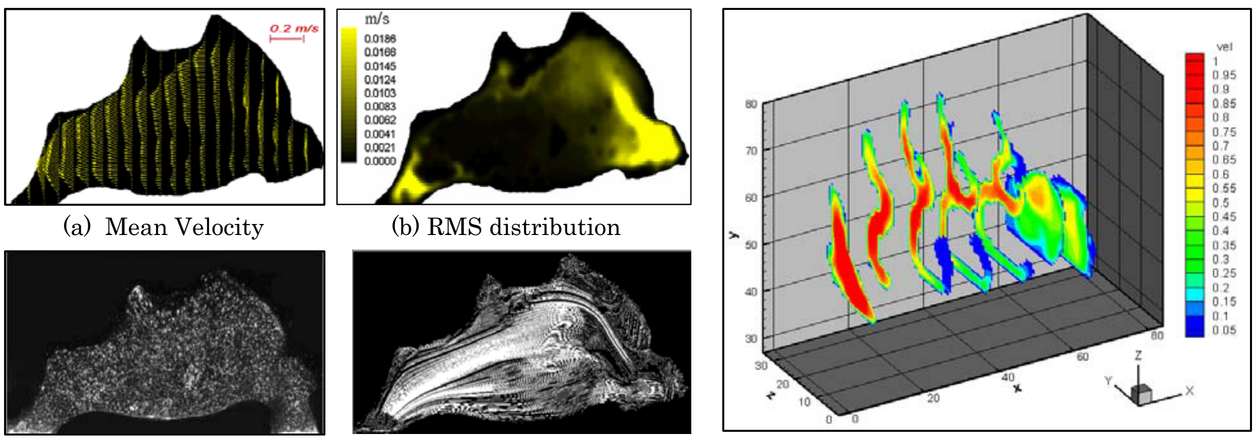
Table 1 Physiological and experimental condition

	Size	Flow rate	period
Human	1	0 ~ 923 ml/sec	3 sec.
Model	1	0 ~ 400 ml/sec	7 sec.

3. Results and discussion

Typical PIV results in nasal cavity (near maximum inspirational velocity phase) are given in Fig. 6 and the variation of average velocity in coronal sections of nasal cavity depict in Fig.7. Larger RMS values, equivalent to larger heat and mass transfer, can be seen at the level of the anterior part of the middle airway, and the flow rate was highest at the middle airway in nasal cavity (Kim et al. 2002). Velocity levels denote amplitude of average and RMS velocities. The physiological flow in nasal cavity at rest has been believed to be a predominantly quasi-steady flow. The partial evidence of this fact can be also found in our results. There exists the similarity in velocity profiles for most of the time during a period except for the moment of changing phase, as shown in Fig. 7. The velocity field for this short moment showed a complex flow pattern, which affects the sedimentation of dust or pollutant particles (Comer et al. 2001, Chung et al. 2006, Chung & Kim 2008). This localized complexity can be thought as the non-linear interaction of the periodic flow and the complex geometry of nasal cavity. Therefore the airflow analysis under periodic flow condition is indispensable for this sort of problem, like drug delivery or pollutant air inhalation. The investigation by dynamic PIV is recommended for the precise look at this. (Lee et al. 2008, Kim et al. 2008)

The phase averaged velocity distributions in a flow distributions in coronal planes provide better visualization for the direction of main stream, where colored passage near larynx and trachea are obtained from a 30 velocity vector set by PIV analysis, as shown in Fig.8 (cross-sectional plane) and Fig.9 (coronal section). Due to its geometrical shape, the magnitude of axial mean velocity and RMS increase in near and downstream of the vocal section. Can the inspired fresh air always reach to the alveoli in one cycle of respiration? If not, how can we prevent the mixing of fresh and exhausted air? We have one conjecture to explain this. The main stream went through the backside of larynx and trachea in inspiration and the frontal side in expiration as shown in Fig. 10. Since fresh air cannot always reach to alveoli in one cycle of respiration, this prevents mixing of fresh and exhausted air inside human upper airway during successive respirations. There exist vortical motions in inspiration, but no prominent one in expiration.



(a) Mean Velocity (b) RMS distribution

(c) Digital image (d) Mean Stream (e) Coronal Velocity distribution

Fig. 6. Results for nasal airflow in the middle stage of inspiration

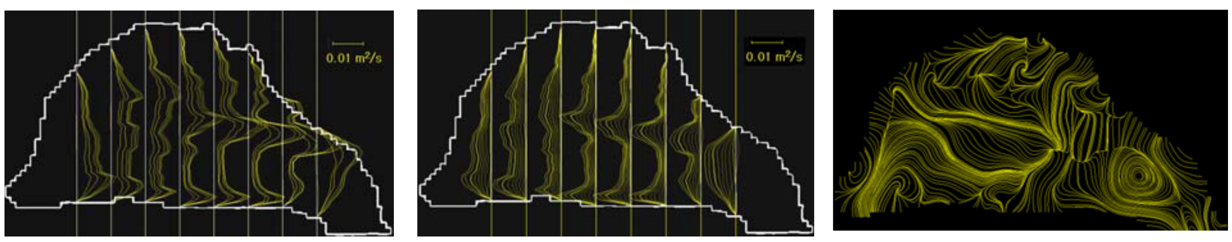


Fig. 7. Variation of flow profile (flow-rate/unit length) in coronal sections for inspiration (left), expiration (middle) and changing phase (right)

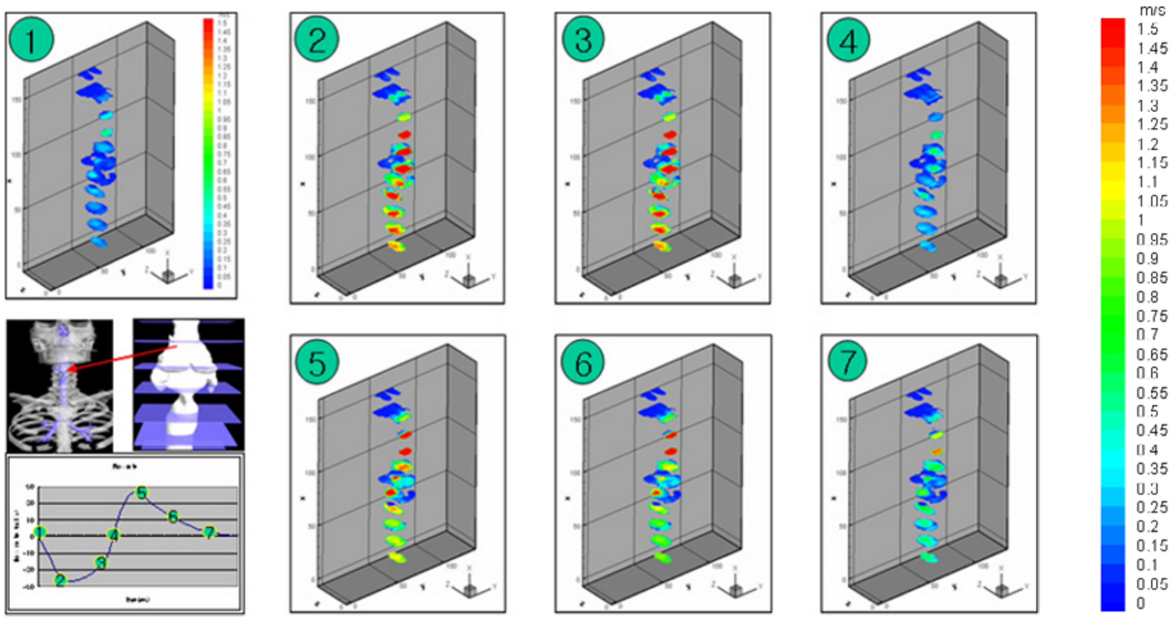
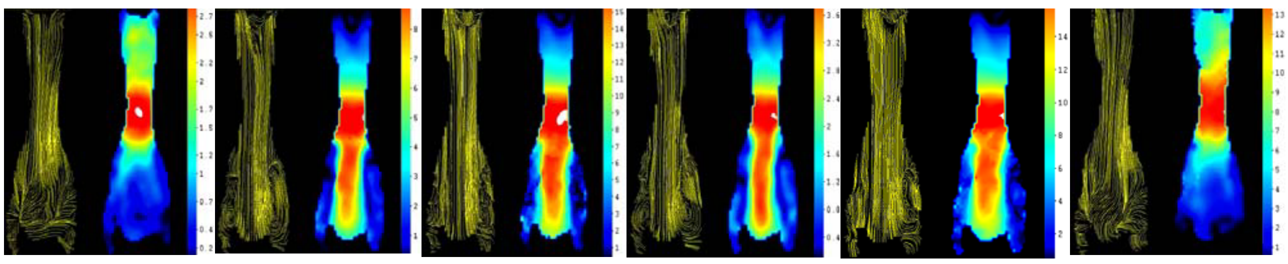
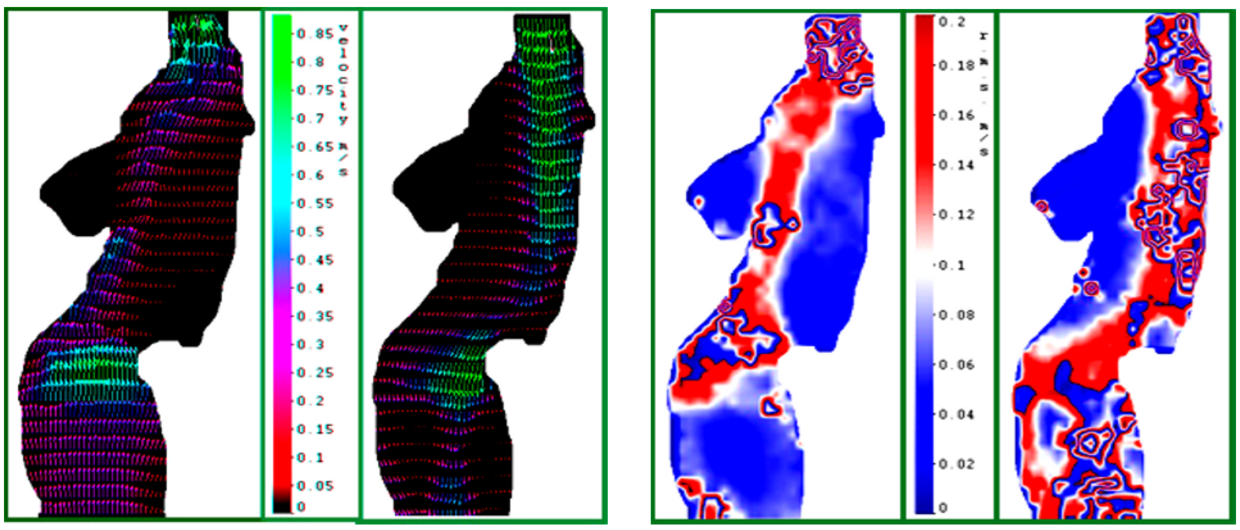


Fig. 8. PIV results for flow-rate on cross-section in larynx, pharynx, and trachea



(a) phase 1 (b) phase 2 (c) phase 3 (d) phase 4 (e) phase 5 (f) phase 6

Fig. 9. PIV results for flow-rate on coronal plane in pharynx, larynx, and trachea

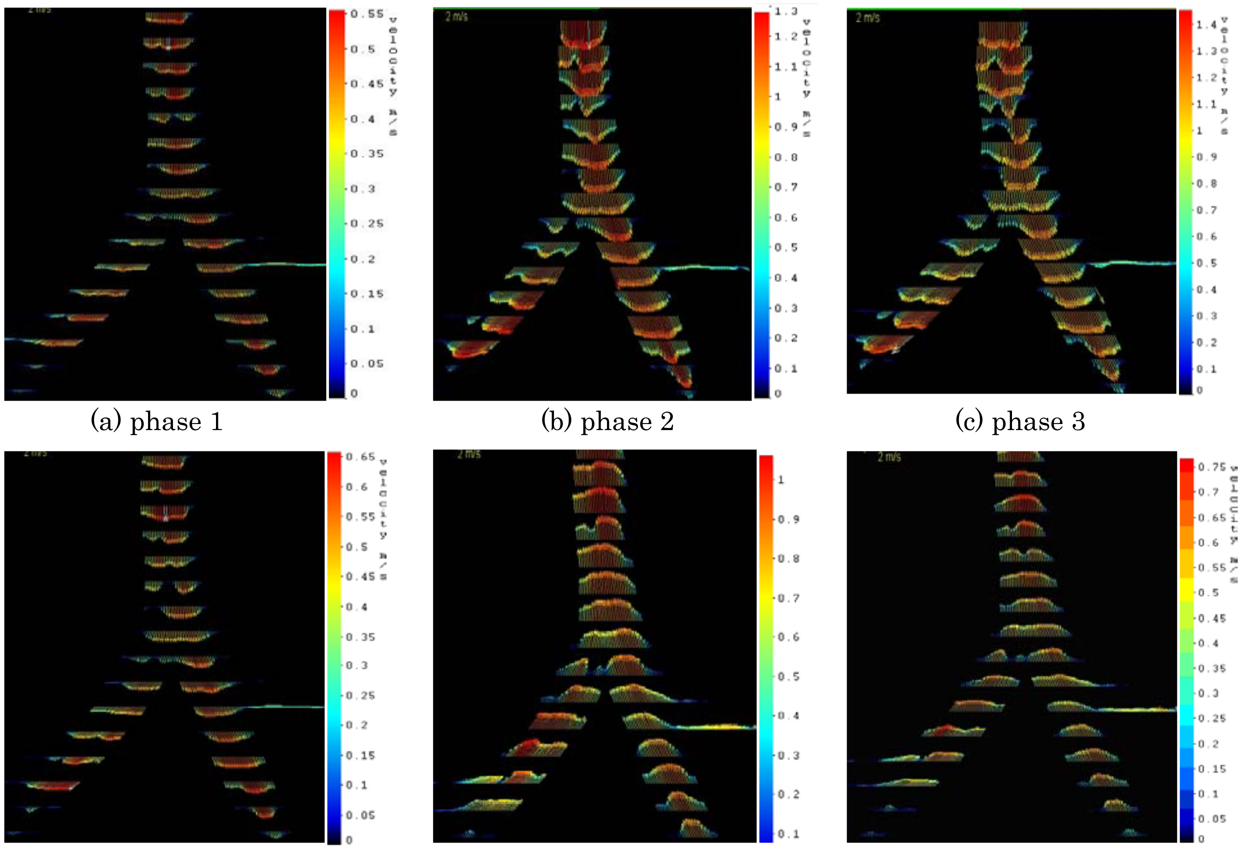


(a) Average velocity distribution for expiration (left) and inspiration (right)

(b) RMS distribution for expiration (left) and inspiration (right)

Fig. 10. PIV results for airflow in Larynx

This can be seen in the movie files of images and sequential velocity distributions of coronal plane. Average velocity distributions in cross-sections of bronchi are obtained from a 30 velocity vector set by PIV analysis, as shown in Fig.11. Since the shape of bronchi is asymmetric due to the existence of a heart, there exist some difficulties in obtaining 2 dimensional images in bronchi area. This may cause inaccurate results in this area. (Lee et al. 2006)



(a) phase 1 (b) phase 2 (c) phase 3

(d) phase 4 (e) phase 5 (f) phase 6

Fig. 11. PIV results for airflow in bronchi

4. Conclusion

We have shown the typical application of PIV measurements to biomedical problem in the area of the otolaryngology. The procedure to create the sophisticated upper airway models and the periodic pumping system are established, that results in obtaining the reliable PIV results for the realistic flow condition. We offer the first quantitative results for the full airway model under physiological flow condition. We have some biophysical conjectures. The main stream went through the backside of larynx and trachea in inspiration and the frontal side in expiration during respiration cycles. These results can be fundamental data for ENT doctors and the CFD investigator of these problems. The paradigm established in this work can be easily applicable to the other biomedical flow problem like blood flow in the circulatory system.

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