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Airflow velocities in the airways during expiration on different end-expiratory lung volumes: Computational study

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Abstract— **We used our computational model of the respiratory system which features non-linear variation of airway dimensions and airway-generation-based structure to show airflow velocities (cm/sec) during natural slow expiration on different end-expiratory lung volumes. Expiratory airflow rates at the mouth can be easily measured using a flow meter. However, because there is no practical non-invasive method that is currently available to measure airflow velocity in the airways, the airflow velocities in airway generations 0 ~ 16 were studied using the computational model. An airflow velocity is given by an airflow rate (ml/sec) ÷ a cross sectional area (cm^2). The cross sectional areas vary depending on inflation and deflation of a lung during respiration, and thus, knowing expiratory airflow rates at the mouth does not go far along the way to find out airflow velocities in the airways. In this study, we first predicted variation of expiratory airflow rates on six different end-expiratory lung volumes using a concept of a time constant, a product of lung compliance and airway resistance, and computational simulation. Then airflow velocities during expiration on the six end-expiratory lung volumes were computed and compared at the conducting airways, airway generations 0 ~ 16.**

Keywords— **Respiratory system, computational modeling, simulation, time constant, airflow velocity**

I. Introduction

There are many aspects of the lung that determine airflow rates at the mouth during tidal breathing but they can be summarized as one of three parameters of the lung mechanics – airway resistance, lung capacitance, and inertnace of air. Since inertance of air is negligible during quiet slow breathing with normal air, airway resistance and lung capacitance together characterize airflow rates (ml/s) in terms of lung mechanics. Provided that one exhales in a usual and comfortable manner, a time constant [10] [11] [12], a product of airway resistance and lung compliance, determines strength and duration of expiratory airflow rates. Even if the amount of ventilated air during respiration is the same, inspiratory airflow is actively controlled by spontaneous efforts and hence unpredictable but expiratory airflow should be always similar as long as the parameters of lung mechanics are identical. In other words, expiratory airflow rates can be characterized by the parameters of lung mechanics [6]. Since both airway resistance and lung compliance depend upon airway dimensions, the parameters of lung mechanics are easily changed by lung volumes. With increased lung volume, airway resistance and lung capacitance decreases and so does the time constant. A smaller time constant allows a shorter time to complete expiration, therefore it results in the larger maximum value of airflow rates, and vice versa.

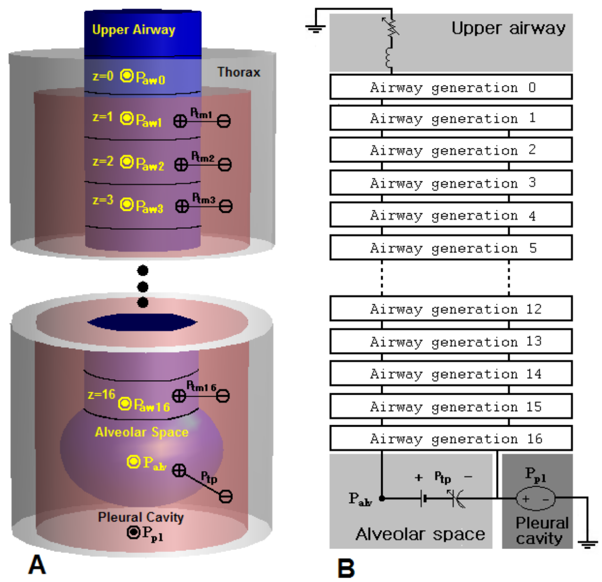

Fig. 1. (a) The conceptual model based on Weibel's morphometry of the lung [22]. Five regions are defined to describe the lung and airways. (i)=alveolar space, (ii)=compliant conducting airways, (iii)=pleural cavity, (iv)=thorax outside the lung, and (v)=outside thorax. Conducting airways, (ii), consist of the upper airway and airway generations from 0 to 16. (b) Overall circuit analogue converted from (a). Each airway generation is represented by compartments.

However, since velocity of airflow depends on both airflow rates and cross sectional areas (cm^2) (airflow rate ÷ cross sectional area), airflow rates measured at the mouth are not a direct indicator for airflow velocities (cm/sec) in the airways. Because there is no practical non-invasive method that is currently available to measure an airflow velocity, airflow velocities in the conducting airways (airway generations 0 ~ 16) are studied using the computational model. In this paper, we first briefly introduce the computational model and then present characteristics of expiratory airflow velocities in the conducting airways.

II. Methodology

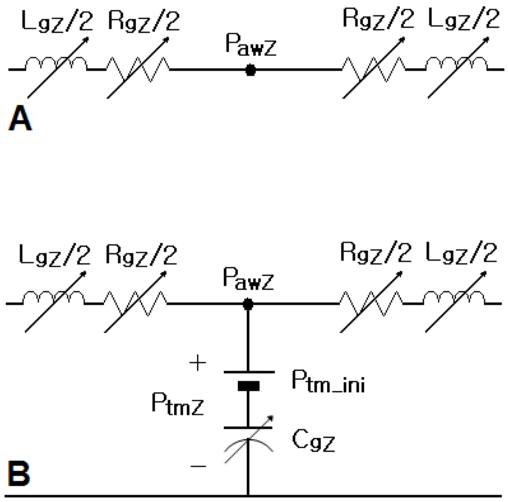

Fig. 2. (a) Electrical circuit analogue for airway generation 0. (b) Electrical circuit analogue for airway generations 1 ~ 16.

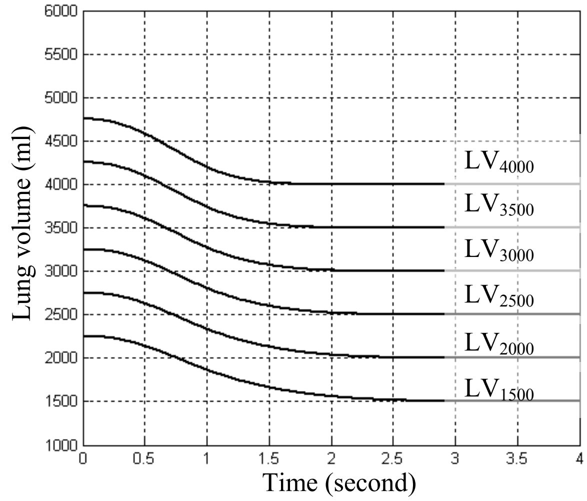

Fig. 3. Variation of the lung volume on the six different lung volumes. LV_{XXXX} represents a lung volume after expiration, which is the same as end-expiratory lung volume is XXXX ml. Expiratory volume of air is 750 ml. Due to difference of the time constants, expiration curves of each case differ from one another.

The computational model was developed based on the previous studies on physiological and anatomical characteristics of the bronchial airways and the details were described in [16][17].

For anatomical details of the airways, Weibel's morphometry of the lung [19], which suggested the bronchial airways as a 24-times branched symmetric dichotomous structure, delineated airway dimensions. We simplified each airway generation from 0 to 16 (conducting zone) as a big tube of which cross sectional area equals the total airway cross sectional area. Meanwhile airway generation 17 ~ 23 (respiratory zone) was considered as a lump and this zone is defined as alveolar space [3]. He defined the trachea as airway generation 0 and the number of airway generation increases as the airways are branched. Fig. 1 (a) illustrates our concept for simplification. Fig. 1 (b) shows the overall model that is equivalent to the conceptual model. Each airway generation is represented by a compartment of which electrical circuit analogues are shown in fig. 2. Airways in airway generations 4 to 16, which are the bronchioles and terminal bronchioles, are not supported by cartilages and are very compliant. Reinforcing Weibel's morphometry of the lung that delineated airway dimensions, Lambert et al. [8] presented the physiological study on variations of airway cross sectional areas during respiration. They suggested a tube law that relates transmural pressure and cross sectional areas.

This computational model featured non-linear variation of airway dimensions and airway-generation-based structure. Non-linear variation of airway dimensions was carefully reflected based on previous physiological studies [2] [5] [8] [13] [15], therefore, airway resistance, lung compliance, as well as inertance of air in the model changes during respiration in accordance with known physiological knowledge. To compute airflow velocities, accurate estimation of cross sectional areas of the airways was necessary and this was realized by the non-linear elements in the electrical circuit analogue. The geometrical structure of the airways was simplified minimally so that the model allows identifying airflow velocities in each airway generation, which are significantly different from each other.

Lung compliance consists of alveolar compliance and airway compliance and most of lung compliance is accounted for by alveolar compliance. Since the airway walls become stiffer as the lung is inflated, increase of the lung volume reduces lung compliance. Airway resistance is determined by both airway dimensions and airflow rate though the airways. A smaller airway diameter and faster airflow creates greater airway resistance during breathing.

The *TLC* (total lung volume), *FRC* (functional residual capacity), and *RV* (residual volume) of the lung model were defined as 6000 ml, 2700 ml, 1000 ml, respectively. *FRC* is 45 % of *TLC* and tidal volume is 750 ml. The six different end-expiratory lung volumes were chosen as 1500 ml (LV_{1500}), 2000 ml (LV_{2000}), 2500 ml (LV_{2500}), 3000 ml (LV_{3000}), 3500 ml (LV_{3500}), and 4000 ml (LV_{4000}) and those volumes represented 45% ± ~20% of *TLC*. Fig. 3 is natural slow expiration of 750 ml on the six different end-expiratory lung volumes. Since a larger lung volume induces smaller airway resistance and smaller lung compliance, a time constant decreases and a lung is deflated more rapidly.

Based on those six cases, airflow rates at the mouth and airflow velocities in the conducting airways were investigated. Fig. 4 shows a diagram that show how airflow velocities were calculated by simulation program. The codes were written on MATLAB (MathWorks, Natick, MA).

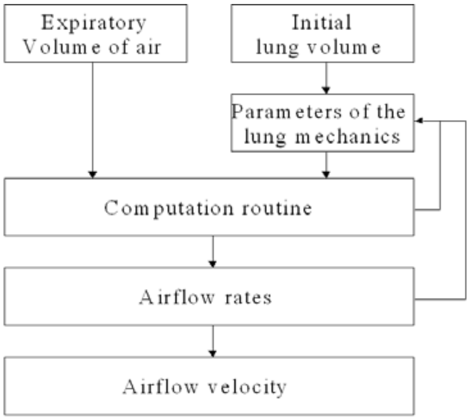

Fig. 4. A diagram for the structure of the simulation codes. An expiratory volume of air and an end-expiratory lung volume are the initially set parameters. Parameters of lung mechanics are continuously updated based on both the lung volume and the airflow rates.

III. RESULT

If a longer time is needed for expiring the same amount of air, it is intuitive to understand that smaller maximum value of an airflow rate should result, as graphs in fig. 5 illustrates. LV_{1500}, which took the longest time for expiration, proves the smallest maximum value of airflow rates during natural slow expiration while LV_{4000} shows a larger lung volume creates a greater airflow rate. Because the maximum airflow rate of LV_{4000} is about 40% larger than that of LV_{1500}, one may erroneously consider that airflow velocity in the conducting airways would be also the largest with LV_{4000}.

However, our simulation suggests that airflow velocities in airway generations 4~16 cannot be presumed by airflow rates at the mouth and airflow velocities only in airway generations 0~3 are in proportion to airflow rates at the mouth. In Fig. 6, the maximum airflow velocities at all the conducting airway generations are drawn as a line – there are six lines that correspond to each of the six different end-expiratory lung volumes. A line for LV_{1500} is set as a benchmark (shown as 100%) and lines for other lung volumes are expressed as the percentage difference being compared to LV_{1500}. This figure indicates that the maximum airflow velocities in airway generations 4~16 actually get smaller as the lung is inflated. Therefore, it can be conjectured that, in these airway generations, airway walls are so compliant that enlargement of the cross sectional areas have more significant influence on airflow velocities than increase of airflow rates do. In our computation, there is only ~4 % difference of cross sectional areas between airway generation 1 of LV_{4000} and LV_{1500} while a cross sectional area at airway generation 13 of LV_{4000} is as greater as ~67 % than that of LV_{1500}. This colossal difference in cross sectional areas explains why larger airflow rates at the mouth do not guarantee larger airflow velocity in all the airways.

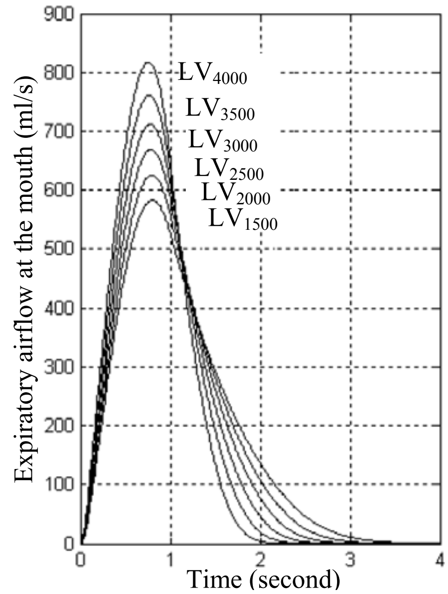

Fig. 5. Expiratory airflow rates for the six different lung volumes. Since the time constant of LV_{1500} is the largest, it takes the longest time for expiration and the maximum value of an expiratory airflow rate is the smallest. On the contrary, the maximum airflow in LV_{4000} is the largest due to its smallest time constant.

Additional interesting finding is that, among all airway generations, airway generation 3 holds the greatest airflow velocity among all airway generations with no exception. This may be an artifact of the study of Weibel [19]. In Weibel's morphometry of the lung, the total cross sectional areas of airway generation 3 is considered as smaller than any other airway generation, which is not necessarily true in the living organ.

IV. DISCUSSION & CONCLUSION

In the viewpoint of some caregivers who are concerned about airway clearance, larger expiratory airflow rates are frequently considered most desirable because it is believed that larger airflow rates create stronger outward shear forces to mucus. However, it should be stressed that shear forces to mucus is not estimated with an airflow rate but with an airflow velocity as many researchers pointed out [1] [4] [7] [9] [18]. Although interaction between airflow and mucus in the airways is one of the physics unresolved by modern technology [14], airflow velocity is the most prominent factor to estimate shear forces for mucus movement.

Although our simulation did not cover the entire range of a lung volume variation, the simulation results clearly demonstrate that, airflow rates at the mouth is not always a good reference for characterizing airflow velocities in the bronchial airways.

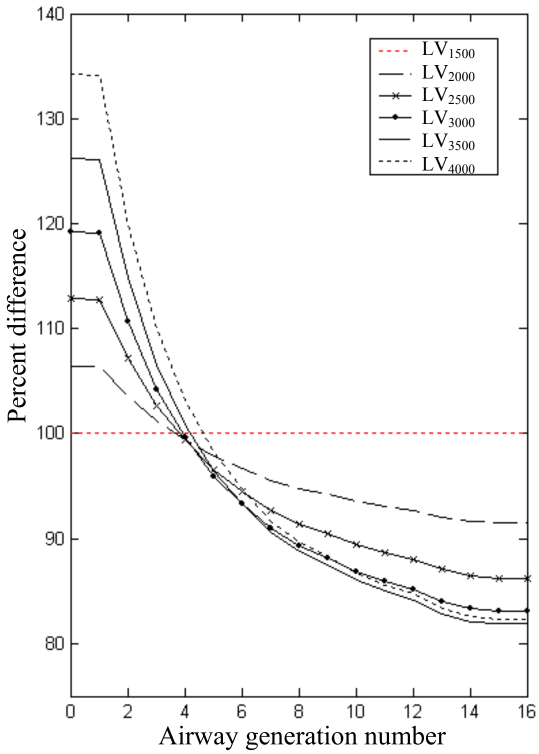

Fig. 6. The maximum airflow velocities at all the conducting airway generations are compared to each other. There are six lines that correspond to each of the six different end-expiratory lung volumes. A line for LV_{1500} is set as a benchmark (100%) and lines for other lung volumes are expressed as the percentage difference being compared to LV_{1500}. Our simulation suggests that airflow velocities in airway generations 4~16 cannot be presumed by airflow rates at the mouth although airflow velocities in airway generations 0~3 are in proportion to airflow rates at the mouth.

We hope that our suggestion in this study could be useful in improving airway clearance techniques as well as opening a new field of advanced studies on the respiratory system.

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