IDENTIFICATION OF AIRWAY CHARACTERISTICS USING THE INPUT IMPEDANCE

Tung[1], V. X., Al-Jumaily*[1], Cheong[2], S.H, Ro[2], S.H

[1]Biomedical Engineering Centre (BioMEC)
Auckland University of Technology, Auckland, New Zealand

[2]School of Mechanical Engineering
Kumoh National Institute of Technology, Gumi-Si, Gyeongbuk, South Korea

*ahmed.al-jumaily@aut.ac.nz

Abstract

In an attempt to determine the correlations between the input acoustic impedance and the variations of the physical characteristics of the terminal elements, a five-lobe branched tube-network is mathematically developed and experimentally simulated using a lung simulator. The model takes into account some realistic conditions such as varying cross-sectional areas, flexible wall properties and branching. The effects of airway constrictions expressed by lobe stiffness variations on the impedance are determined for a range of frequencies up to 256 Hz. It is concluded that the developed model is capable of non-invasively predicting various physiological changes in the airway passages.

INTRODUCTION

Early research on the modelling of the respiratory system had focused on the flow resistance and pressure drop characteristics across the airway passages but did not include airway narrowing into the models [1,2]. Al-Jumaily and Mithraratne [3] developed a theoretical acoustic model of the respiratory system based on the Weibel's symmetric model [9]. Al-Jumaily and Al-Saffar [4] extended the model developed by Fredberg [5] by including the effect of the wall inertia to study the dynamic response of the respiratory system measured at the trachea. Further Al-Jumaily and Al-Fakhiri [6,7] developed a mathematical model to study the influence of elastance variation on the respiratory system dynamics. They used the acoustical approach to determine the impedance at the throat using impedance recursion formulas for both symmetric and asymmetric structure to consider the response of the lung system. They cited that "the overall normalised input impedance frequency spectrum could be used to give a reasonable signature for identifying such abnormality" [7]. Further, Al-Jumaily and Du [8] modelled and simulated the airways for identifying and detecting obstruction. The results demonstrated that the input impedance resonant frequencies can locate the obstruction with its degree of severity in any of the airway branches.

To understand how the input impedance measured at the throat of the respiratory system relates to the mechanical characteristics of the lung, a 5-lobe model is proposed and analysed. The model accounts for the effects of airways with varying cross-sectional area and flexible wall properties in the bronchial tree within the lungs. The model is then validated using a lung simulator.

MODEL DEVELOPMENT

A human bronchial tree is a system of wet, elastic, and bifurcated tubes. The relationship between the impedance of the proximal end Z_0 and terminal end Z_L of a branch is frequency dependent and may be written in the following recursion form [1]

$$Z_0 = \frac{\bar{z}_s Z_L \cosh(\gamma L\omega) + \bar{z}_s^2 \sinh(\gamma L\omega)}{\bar{z}_s \cosh(\gamma L\omega) + Z_L \sinh(\gamma L\omega)} \qquad (1)$$

for $H < 0$, and

$$Z_0 = \frac{\bar{z}_s Z_L \cos(\gamma L\omega) + j\bar{z}_s^2 \sin(\gamma L\omega)}{\bar{z}_s \cos(\gamma L\omega) + jZ_L \sin(\gamma L\omega)} \qquad (2)$$

for $H > 0$, where

$$H = \left(\frac{\omega_1^2 - \omega^2}{\omega_0^2 - \omega^2}\right) \qquad (3)$$

and the characteristic impedance $\bar{z}_s = \frac{1}{A_0}\sqrt{\frac{M_B \rho_0}{|H|}}$. In the above equations, the tube wall natural frequency ω_0 and ω_1 without and with air-filled, respectively can be expressed as:

$$\omega_0^2 = \frac{K_w}{M_w} \qquad \text{And} \qquad \omega_1^2 = \frac{K_w + M_B}{M_w} \qquad (4)$$

where M_w is the tube wall linear density (kg/m), K_w is tube wall Elastance, M_B is fluid bulk modulus, ρ_0 is air density; and A_0 is original wall area. Values for these frequencies and for different generations are available in reference [1].

Figure 1a shows a parent link and daughters in the simulation model. The impedance at the lower end of the parent link of order $K+1$, $Z_{L,\,k+1}$ is given by the parallel combination of the impedances at the upper ends of the daughter links of order K. Therefore for a given bifurcation this may be written as:

$$Z_{L,K+1} = \frac{1}{\frac{1}{Z_{0,K}} + \frac{1}{Z_{0,RC}}} \tag{5}$$

where RC is the recursion index of the daughter link.

An actual human lung consists of five lobes, three to the right and two to the left. Figure 1b is proposed to fulfil this requirement with right lobes R1, R2, R3, left lobes L1, L2 and conducting airway passages from 1 to 9. The basic principle of obtaining the impedance at the inlet of the parent branch 1 is based on determining the input impedance in terms of the output impedance for each segment using equation 1.

During uninterrupted mechanical ventilation, respiratory gas is moved in and out of the lobes. The lobe mechanics deals with the pressures acting on the lobe and the concomitant changes in pulmonary gas volume produced. The static and dynamic behaviour of the lobe is mathematically described by a simple mass-spring damper system which can be expressed in terms of the lobe volume V and a net pressure P (difference between atmospheric and plural pressure) as follows

$$V(i\omega) = \frac{1}{\sqrt{\left(1 - \frac{\omega^2}{\omega_n^2}\right)^2 + \left(2\zeta \frac{\omega}{\omega_n}\right)^2}} P(i\omega) \tag{6}$$

The impedance transfer function of each lobe (pressure/volume) can be written as

$$Z = \frac{M(i\omega)^2 + B(i\omega) + K}{A^2(i\omega)} \tag{7}$$

From the specifications of each lobe, Table 1, one can determine the impedance from Eq. (7) and then substitute in Eq.(1) to determine the input impedance for each of the branches 4 and 6-9. Eq. (5) is then used to determine the equivalent impedance at each bifurcation point. Using these impedances, again Eq.(1) is used to determine the input impedance of branches 3, 5 and 2, respectively. The equivalent impedance at branch 1 bifurcation point can be determined by using Eq.(5) for impedances of branch 2 and 3. Finally Eq.(1) is used to determine the overall impedance of the system.

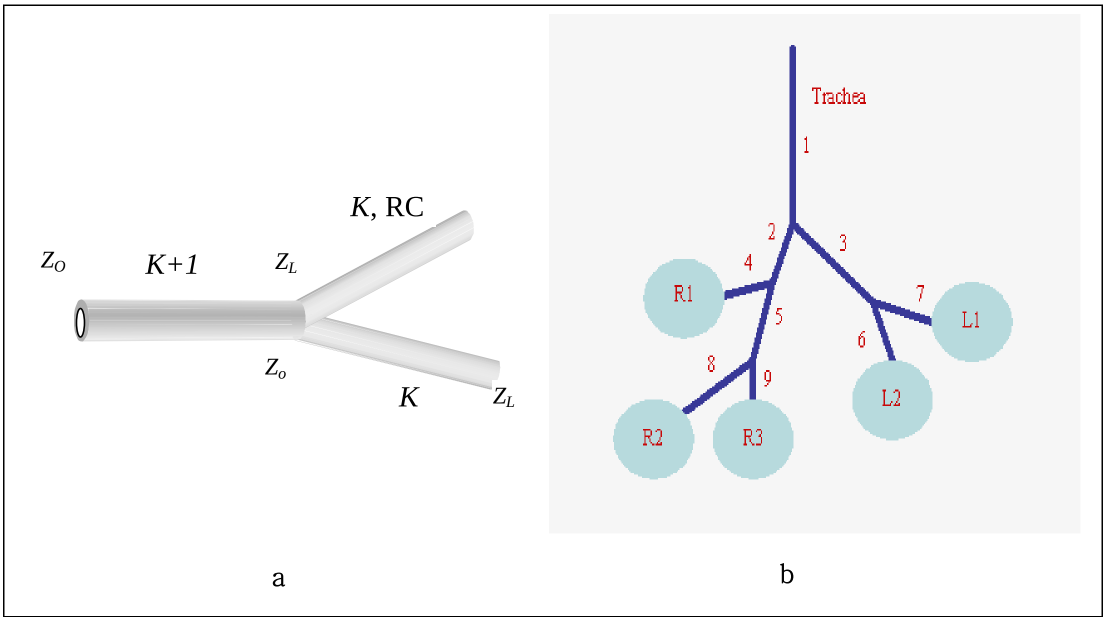

Fig. 1- Model: (a) branching element, (b) 5-lobe

Table 1: Geometric and Mechanical Properties of the Simulated Lung Model

Airway No.	Length (m)	Radius (m)	Elastance (kN/m^2)
1	0.12	0.010	6000
2	0.10	0.008	6000
3	0.10	0.006	6000
4	0.08	0.005	6000
5	0.08	0.004	6000
6	0.08	0.003	6000
7	0.08	0.002	6000
8	0.07	0.002	6000
9	0.07	0.002	6000

EXPERIMENTAL VALIDATION

Figure 2 shows a photo and a schematic diagram of the lung simulator used in this investigation. The basic principle of operation of the lung simulator is to issue a breath cycle command from the computer using LabView™ to the control valve, which allows air to enter the pneumatic drive in accordance with certain modes. Once the actuator starts moving, it will drive the piston inside the cylinder. The cylinder, which resembles the lung, is filled with water to replicate the lung environment. The schematic diagram Fig. 2b represents the mechanical system together with the actuator, interface electronics, low-level position controllers and the control computer. The real-time controller of the entire system has been implemented in LabView 6.1 installed on a personal computer. The computer control with the Data Acquisition generate the analogue output signals and the volume reference (*Vref*). The pneumatic system is then driven by the output signals and its motion dynamically changes the volume inside the pneumatic cylinder. The piston position is measured using an encoder as a feedback to the system. The change in volume of the cylinder creates a negative pressure inside the chamber, which is the driving force to generate the pressure change and airflow in the tested-lung. This pressure change is measured by a pressure sensor. The flow is measured by TSI flow- meter (model 4043) as pressure drop over the tested-lung. The accuracy of the TSI flow-meter was recorded as 2% of the reading or 0.05 Std l/min. It also monitors temperature and static pressure. In-house designed of electrical cards and pressure transducers were used to measure air velocity and pressure drop, respectively, at the terminal locations inside the tested-lung.

The output signals from the tested-lung, together with the signals representing the changes of the airways pressure are recorded via the LabView™ program and stored on the computer hard disk. All the data is given in time domain and are used later to determine the frequency domain analysis. To detect changes in the input impedance spectrum due to changes in lobe characteristics, each lobe was set to be obstructed by adding another rubber lobe around it, which would double the stiffness of the lobe. Obstruction was introduced by running experiments with one, two, three, four or five lobes being stiffened.

RESULTS

For the five-lobe mathematical model, the results were obtained by implementing the values of the terminal impedances calculated using Table l. For the experimental investigation, the results of measured impedance obtained from the pressure and volume flow rate were transformed into the frequency domain. Measured impedances were then used to calculate the input impedance of the entire system using the above model. The theoretical model is validated by comparing the results to the experimental investigations. The Input Impedances were calculated for the simulated lung and presented up to 256 Hz (at 512 Hz sampling rate).

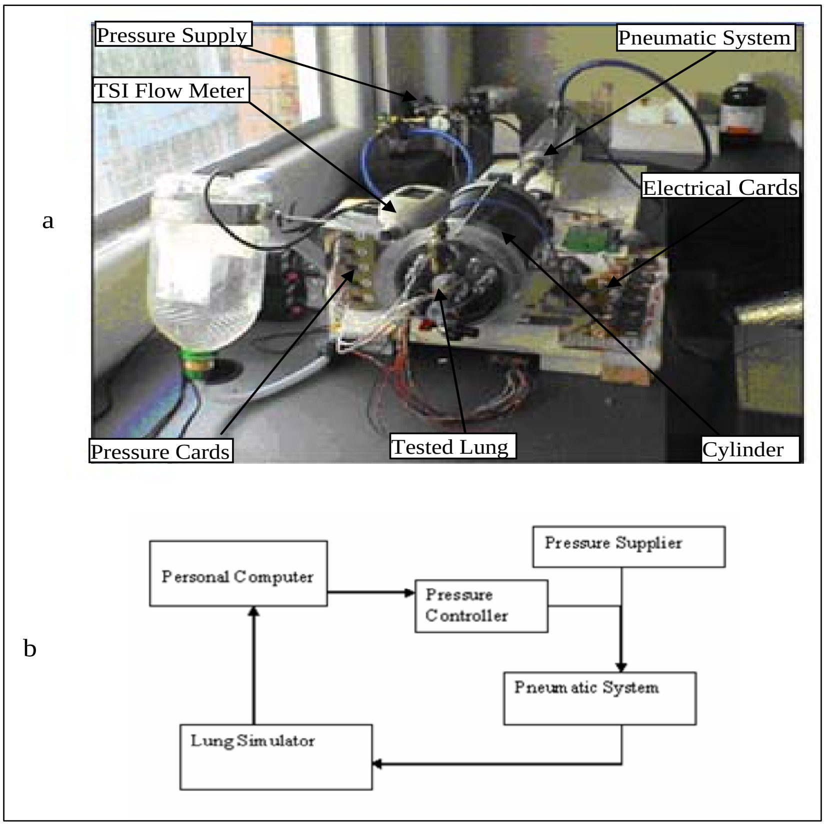

Figure 2- Lung Simulator: (a) setup, (b) schematic Diagram

DISCUSSION

This is a twofold discussion. First validation of the proposed theoretical model using the lung simulator will be discussed. Second the use of the model to determine the effect of changing the lobe characteristics on the input impedance will be elaborated on.

Figure 3 shows a typical frequency spectrum of the input impedance results used in the present discussion. In Table 2 Both theoretical and experimental results are presented for comparison. Several dominant resonance frequencies of the input impedance are observed during the process. The first resonant frequency falls around 4 Hz while the second falls between 7-8 Hz. Producing several spectra similar to Figure 3 for both theoretical and experimental simulation can lead to a table similar to Table 2. Six identical resonant frequencies were found within the range of interest from 0 to 10 Hz. This table gives good agreement between the theoretical and experimental models. It is obviously clear that changing the breath cycle frequency has insignificant contribution to the resonant frequency. This is attributed to the fact that the breath cycle is a forcing function and should not have any contribution to the system characteristic frequencies.

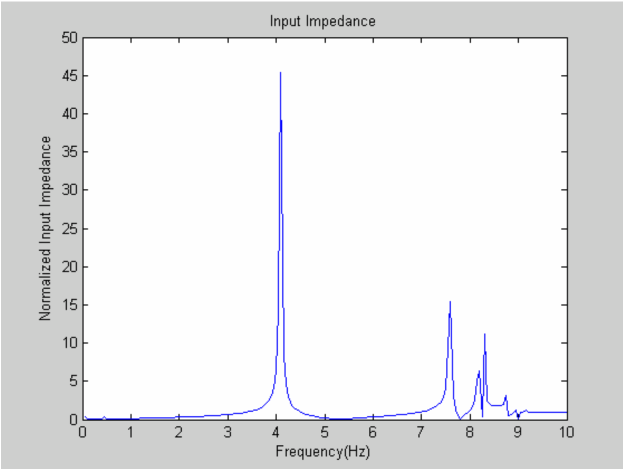

Figure 3: Typical spectrum of the simulated Input Impedance in normal conditions

Inspired by the accuracy of the model as compared with the experiments, the research was expanded to investigate the effect of various lobe stiffness values on the input impedance. Several frequency spectrum curves were generated for various values of lobe stiffness and for different lobes. For brevity a summary of one of the results is given in Table 3. The result indicates that for one-lobe stiffness changes, the first natural frequency shifts from above 4 Hz to around 3 Hz while the other frequencies remain almost unchanged. However, simultaneous changes of more than one lobe stiffness would result in changes in other frequencies. Therefore, more precise computations are essential to specify the specifics of each lobe variations. In conclusions, the change in stiffness parameter has an impact in the results of resonant frequencies.

Table 2 Predicted and measured resonances (Hz)

Predicted	Measured
4	4.1
7.5	7. 7
8.3	8.8
9	9.5

Table 3 Resonances (Hz) for one-lobe stiffening

Normal	R1-	R2-	R3-	L1-	L2-
4	3	2.9	2.4	2	0.2
7.5	7.2	6.8	6.8	6.5	6.3
8.4	8.2	8	8.3	7.5	7.5
9	8.7	8.8		8.8	8.5

CONCLUSIONS

This research is a feasibility study to investigate how the input impedance determined at the inlet of a five-lobe model correlates with changes in lobe characteristics. Both experimental simulation and mathematical modeling confirm strong correlations which could be implemented in a future diagnostics technique. It is possible to link the variation in any of the terminal branch to a particular frequency. The results indicate that stiffness variations in terminal branches can be predicted from the trend of variation of the associated resonant frequency. Also, it is indicated that considering more lobe constrictions in the model stimulates more resonant frequencies within the same frequency range.

ACKNOWLEDGEMENT

This work was supported by the Korea Research Foundation and The Korean Federation of Science and Technology Societies Grant funded by Korea Government (MOEHRD, Basic Research Promotion Fund).

REFERENCES

[1] Dekker, E., “The transition between laminar and turbulent flow in the trachea”, J. Applied Physiology, 16: p. 1060-1064, (1961).

[2] Affin, M.Y. and P. Kesic, “irway resistance: a fluid mechanics approach” Journal of Applied Physiology, Vol. 36, No. 3, pp.354-361,(1974).

[3] Al-Jumaily, A. M., Mithraratne, P., “Simulation of respiratory system for identifying airway occlusion”, J. of Nonlinear Sci. & Num. Simulation, 2(1):21-28, (1974.).

[4] Al-Jumaily, A.M., Al-Saffar, A.M. “Dynamic Response Characteristics of Upper Respiratory System”, J. of the Acoust. Society of America, **106**: 4, pt 2, 2172, (1999).

[5] Fredberg, J.J. and J.A. Moore, “The distributed response of complex branching duct networks. Journal of Acoustic Society of America, 63:p. 954 – 961, (1978).

[6] Al-Jumaily, A. M., Y. Al-Fakhri. “Occlusion identification in respiratory system using asymmetric model”, Comm. in Nonlinear Science & Numerical Simulation 8(1): 37-47, (2003).

[7] Al-Jumaily, A. M. and Y. Al-Fakhri. “Asymmetrically respiratory system simulation to identify airway occlusion”, ASME, International Mechanical Engineering Congress and Exposition, New York, (2001).

[8] Al-Jumaily, A. M. and Y. Du. “Obstruction identification in a compliant tube with application to airway passages”, J. of Vibration and Control 8(5): 643-657, (2002).

[9] Weibel, E.R., “Morphometry of the Human Lung”, N.Y.: academic Press, (1963).