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Identification of airway characteristics using the input impedance

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Abstract

To understand how the input impedance determined at the throat correlates with changes in the dynamic characteristics of the airways, a simplified 5-lobe model is developed and simulated. The model takes into account some realistic conditions such as varying cross-sectional areas, flexible wall properties and branching. The lobe terminal impedances are implemented in the model to predict the input impedance at the throat. The effects of airway constrictions and wall elastance variations on this impedance are determined for a range of frequencies. It is concluded that the developed model is capable of predicting various physiological changes in the airway passages.

Keywords: Impedance; Airway; Respiratory system; Trachea; Lung

1. Introduction

Early research on the modelling of the respiratory system 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 physiological scheme [4]. Incorporating the effect of the wall inertia, they extended Fredberg's model [1] to study the dynamic response of the respiratory system by monitoring the input impedance at the trachea. Also Al-Jumaily and Al-Fakhiri [5, 6] developed a mathematical model to study the influence of elastance variation on the respiratory system dynamics. To study the effect of the wall properties on the lung response, they incorporated the acoustical approach to determine the impedance at the throat using impedance recursion formulas for both symmetric and asymmetric structures. They cited that "the overall normalised input impedance frequency spectrum could be used to give a reasonable signature for identifying such abnormality" [6]. Further, Al-Jumaily and Du [7] modelled and simulated the airways for identifying and detecting obstructions. The results demonstrated that the input impedance resonant frequencies determined at the throat can identify the location and severity of airway obstruction in any of the airway branches.

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

2. Model Development

An actual human lung consists of five lobes, three to the right and two to the left, Fig. 1(a). Fig. 1(b) is proposed to fulfill this requirement with right lobes R1, R2 and R3 and left lobes L1 and L2. The conducting airway passages leading to these lobes are generations 28-35 labeled according to Horsfield's scheme [8] with generation 35 as the trachea. During uninterrupted mechanical ventilation, the respiratory

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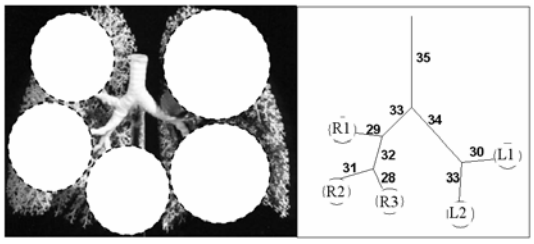

Fig. 1. Lung: (a) image with prescribed lobes, (b) 5-lobe model.

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 dynamic behavior of a lobe can be described by a second order mass-spring system as

$$F(t) = Kx(t) + B\dot{x}(t) + M\ddot{x}(t) \tag{1}$$

where *M, B* and *K* as the equivalent mass, damping and stiffness of the lobe, respectively; *F* is the applied force to produce a displacement *x*. The dots and double dots indicate first and second derivative with respect to time, respectively. Since the actual lobe, Fig. 1, has a volume variation rather than a displacement variation, multiply Eq. (1) by an equivalent piston cross sectional area *A* to get

$$A^2P = KV + B\frac{dV}{dt} + M\frac{d^2V}{dt^2} \tag{2}$$

with *P* as the pleural pressure. Thus, the impedance transfer function of each lobe (pressure/volume flow rate) can be written as

$$Z = \frac{MD^2 + BD + K}{A^2 D} \tag{3}$$

where ω is the frequency of excitation and *D* is *d/dt* . *Z* will be taken as the termination impedance of each airway passage; later this will be transferred to the frequency domain by replacing D by *iω,* where *i=sqrt (-1)*. However, consideration of airway asymmetry is important, especially, when analyzing flow to different regions of the lung. Thus, in this paper an asymmetrical model is considered to describe the model of the bronchial tree, which includes some of the asymmetry of the real structure and yet permits physiological calculation to be made at the same time. The impedance at a network compliant branch, Fig. 2, may be defined as the ratio of the complex amplitude of pressure and volume flow rate.

Fig. 2. Compliant branch.

$$Z_0 = \frac{P_0}{A_0 \cdot V_0}$$
$$Z_L = \frac{P_L}{A_L \cdot V_L} \tag{4}$$

where subscripts *0* and *L* refer to the proximal end and distal end of an airway passage, respectively. For any branch, the relationship between the impedance of the proximal end Z_0 and terminal end Z_L may be written as [3]:

$$Z_0 = \frac{\bar{z}_s Z_L \cosh(\gamma L\omega) + \bar{z}_s^2 \sinh(\gamma L\omega)}{\bar{z}_s \cosh(\gamma L\omega) + Z_L \sinh(\gamma L\omega)} \tag{5}$$

for $H<0$, and

$$Z_0 = \frac{\bar{z}_s Z_L \cos(\gamma L\omega) + j\bar{z}_s^2 \sin(\gamma L\omega)}{\bar{z}_s \cos(\gamma L\omega) + jZ_L \sin(\gamma L\omega)} \tag{6}$$

for $H>0$. The index H is defined by $H = (\omega_1^2 - \omega^2)/((\omega_0^2 - \omega^2)$, the characteristic impedance $\bar{z}_s = (1/A_0)\sqrt{M_B\rho_0/|H|}$ and $\gamma = \sqrt{\frac{\rho_0}{M_B}H}$. In the above equations, the passage wall natural frequency ω_0 and ω_1, without and with air-filled, respectively, can be expressed as:

$$\omega_0^2 = \frac{K_w}{M_w} \quad \text{and} \quad \omega_1^2 = \frac{K_w + M_B}{M_w} \tag{7}$$

where M_w and K_w is the branch wall linear density (kg/m) and elastance, respectively; ρ_0 and M_B is the air density and bulk modulus, respectively; A_0 is the original wall area. The equivalent impedance Z_E at each bifurcation junction, Fig. 1(b), is determined from the impedance of the daughter branches Z_{B1} and Z_{B2} at the junction. Namely,

Table 1. 5-lobe bronchia specifications.

Gen No.	$K_w 10^{-2}$ (kN/m^2)	$M_w 10^{-3}$ (kg/m)	D mm	L mm	f_o (Hz)	f_1 (Hz)
28	39	0. 520	2.3	7.60	436	2668
29	39	0.700	2.8	9.00	376	2300
30	39	1.100	3.5	10.7	300	1834
31	47	1.600	4.5	12.7	273	1525
32	47	3.000	5.6	7.60	199	1114
33	47	15.10	8.3	19.0	89	496
34	57	22.60	12.2	47.6	80	407
35	73	30.20	18.0	120.	78	354

$$Z_E = 1/(\frac{1}{Z_{B1}} + \frac{1}{Z_{B2}}) \quad (8)$$

In the present work five lobes and the connecting bronchia with the specifications given in Table 1 are considered [5]. The impedance of each lobe is deduced from previous work as 32680 Pa.s/m^3 [3]. Using this impedance as the terminal impedance (for Eq. (3)), the input impedances of branches 28-33 are determined by using Eqs. (4) through (7), then the bifurcation impedance at each junction is determined from Eq. (8). Continuous recursive computations lead to the overall input impedance at the throat.

3. Results and discussion

In order to facilitate correlation and comparison of the results, all the results of this work are expressed in terms of normalized dimensionless input impedance at the throat. Therefore, the calculations of the input impedance Z_0 in Eqs. (5) and (6) are normalized by dividing by the specific impedance, z_s of the trachea (branch 35). The normalized impedance at the throat is computed for the human respiratory system along each of the bronchia. A typical output of this process is given in Fig. 3. To explain the variation in the spectrum due to any changes in the system, this work focuses on identifiable characteristics which could be used for diagnostic purposes such as the overall minimum or maximum of the normalize input impedance [3, 5, 6].

The present 5-lobe model is simulated by using Matlab^{TM} and several frequency spectra of the input impedance are generated for two practical scenarios. The first is by assuming rigid walls where the airways are treated as non-elastic (the value of the index $H = 1$ for Eqs. (5) and (6)), while the second is assuming compliant walls where the elastance and inertance of the airways are considered. For each scenario two cases of termination are considered. The first case is assuming a complete collapse of the lobe where the terminal impedance is set to infinity, while the second is assuming healthy lobe where the compliant termination with appropriate impedance value of 32680 Pa.s/m^3 [3].

Table 2. Normalised input impedance for rigid and compliant passages with occlusion at the indicated branches.

Occlus. at	First Minimum (Hz) NII$\times 10^3$				First Maximum (Hz) NII$\times 10^3$			
Branches	Rig.	Com	Rig	Com	Rig	Com	Rig	Com
Normal	517	65	33.2	90.1	918	75	4.9	5.2
28	534	65	26.8	47.8	978	75	7.8	8.1
29	537	65	29.4	51.8	988	75	7.5	8.5
30	537	65	33.8	57.0	992	75	6.7	8.3
31	551	66	41.9	44.2	1031	75	6.2	11.2
32	549	66	33.6	50.0	1009	75	6.1	9.4
33	586	67	107.	109	1080	75	2.7	6.9
34	632	69	302	358	1103	76	1.4	1.4

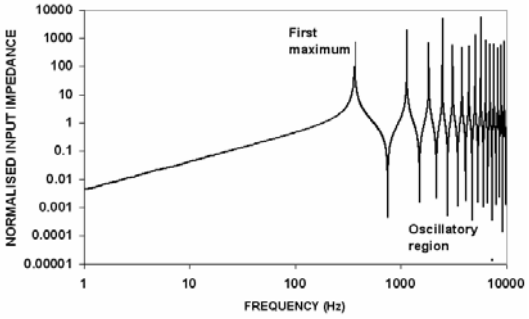

Fig. 3. Typical normalized input impedance vs. frequency with identifiable first maximum.

Table 2 presents the results for the rigid and compliant wall models with compliant termination (healthy lobes). Also, an occlusion is introduced at different levels of the bronchia as indicated in column 1 of the table. An occlusion at generation 29, 31, 28, 30 or 33 indicates the collapse of lobe R1, R2, R3, L1, or L2, respectively. For diagnostic purposes, the identifiable parameters used in this work are the First Minimum (FM) and the First Maximum (FX) of the normalized input impedance. The following observations are made:

(i) First Minimum: With rigid walls the frequency

and the impedance value of the FM increases as the occlusion progresses towards the trachea and it becomes maximum when there is a complete occlusion of the respiratory system (the trachea with infinite terminal impedance). This is attributed to the fact that as the occlusion progresses towards the trachea, shorter passages are obtained and higher frequencies and amplitudes are expected. In contrast for compliant branches the frequency of the first minimum does not change significantly, but the value of the impedance increases as the occlusion progresses towards the trachea.

(ii) First Maximum: With rigid walls the frequency of the FX increases as the occlusion progresses towards the trachea, becoming a maximum when there is a complete occlusion of the respiratory system. This is attributed to the same fact of shorter passages mentioned above. However, this frequency does not experience significant changes when compliant branches are used. The impedance value of FX for both rigid and compliant branches does not give a recognizable trend of variation for identification purposes.

Table 2 also indicates that the frequencies where FM and FX occur for the compliant-wall model always have lower values compared to those of the rigid-wall model, because elastic walls tend to damp out resonances by absorbing energy from the wall vibrations.

To study the effect of wall property variations on the normalized overall impedance determined at the throat, elastance variation is introduced in the bronchia, generation 34, 33 and 32. It is indicated that there is no significant change in the overall maximum; however, distinguishing changes are observed in the overall minimum. Fig. 4 indicates that the amplitude of the overall minimum impedance decreases as the wall-elastance increases; however, this reduction is more pronounced when the variation is in bronchus 33.

In conclusion, regarding all limitations of the mathematical model, this study suggests that the approach adapted in this work is a valuable tool for assessing an obstruction in a patient suffering from an asthmatic attack or collapse of one of the lobes. Also, the present model is much simpler than previous models and easier to validate experimentally [9]. Changing the elastance of the lobes did not show systematic changes as those observed when changing the elastance of the bronchia.

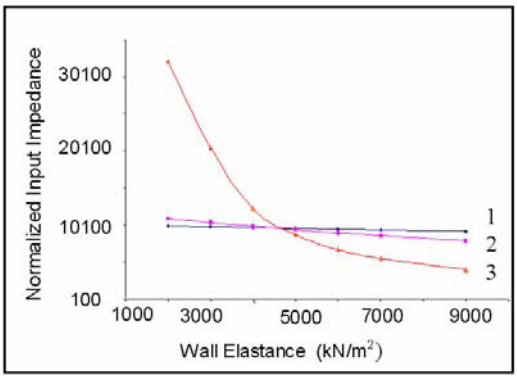

Fig. 4. Overall normalised impedance vs. wall elastance variation in bronchus: 1 Gen. 34, 2 Gen. 33 and 3 Gen. 32.

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