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Experimental and Computational Studies of Sound Transmission in a Branching Airway Network Embedded in a Compliant Viscoelastic Medium

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Abstract

Breath sounds are often used to aid in the diagnosis of pulmonary disease. Mechanical and numerical models could be used to enhance our understanding of relevant sound transmission phenomena. Sound transmission in an airway mimicking phantom was investigated using a mechanical model with a branching airway network embedded in a compliant viscoelastic medium. The Horsfield self-consistent model for the bronchial tree was adopted to topologically couple the individual airway segments into the branching airway network. The acoustics of the bifurcating airway segments were measured by microphones and calculated analytically. Airway phantom surface motion was measured using scanning laser Doppler vibrometry. Finite element simulations of sound transmission in the airway phantom were performed. Good agreement was achieved between experiments and simulations. The validated computational approach can provide insight into sound transmission simulations in real lungs.

1. Introduction

1.1. Background

Alterations in the structure and function of the pulmonary system that occur in disease or injury often give rise to measurable changes in lung sound production and transmission. Lung sounds are known to contain spatial information that can be accessed using simultaneous acoustic measurements at multiple locations. It has been shown that lung consolidation, pneumothorax, and airway obstruction, to name a few conditions, alter the production and/or transmission of sound spectrally, temporally and regionally. If properly

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quantified, these detected changes might provide additional information about the severity and location of the trauma or pathology [1-4].

A better understanding of sound transmission and how it is altered by injury and disease might improve interpretation of lung sound measurements. This includes new lung imaging modalities that are based on an array measurement of the acoustic field on the chest wall surface [5-7] or on 3-dimensional measurement of the lungs and chest wall acoustic field using magnetic resonance elastography [8-11]. With this in mind, the authors are developing a comprehensive computer simulation model of pulmonary acoustics, known as the Audible Human Project™. A long-term goal of the project is to develop a computational acoustic model that would accurately simulate generation, transmission and noninvasive measurement of sound and vibration within the pulmonary system and chest wall. As a first step, it is necessary to carry out acoustic studies on airway and lung mimicking phantoms to validate the computational approach.

Simulations of sound transmission in lung mimicking phantoms and the human chest have been carried out by several groups using analytical calculations or numerical simulations. Wodicka et al. [12] developed a theoretical model of sound transmission from within the respiratory tract to the chest wall due to the motion of the walls of the large airways. The vocal tract, trachea, and the first five bronchial generations were represented over the 100 to 600 Hz frequency range by an equivalent acoustic circuit. The radiation of sound through the surrounding lung parenchyma was represented as a cylindrical wave in a homogeneous mixture of air bubbles in water. The predicted magnitude of acceleration over the extra-thoracic trachea and at three locations on the posterior chest wall in the same vertical plane was consistent with experimental observations. The acoustic circuit model has limitations in predicting location-dependent motion on the lung or torso surface. In Vovk et al. [13], a theoretical 2D axisymmetric layered model for the torso region was used. The model included annular regions for the parenchyma, rib cage region, soft tissue and skin. Breath sound transmission from trachea to the body's surface was calculated and the predicted mechanical impedance of the chest surface had similar frequency dependence with experiments. In Royston et al. [14], a theoretical model similar to Vovk et al. [13] was used to predict sound transmission from trachea to chest surface under normal and pneumothorax conditions. The predicted sound energy loss caused by pneumothorax was confirmed by experiment and finite element simulations. That 2D axisymmetric layered model simplified the airway tree as a circular region for the trachea. That model has limitations in predicting location-dependent motion on the lung or torso surface. In Acikgoz et al. [15] experimental studies on sound transmission in a lung-chest phantom that mimics pneumothorax were compared with boundary element simulations. These studies were helpful in quantifying the effect of pneumothorax on the resulting acoustic field measured at the surface of the phantom that mimicked the chest. Wochner et al. [16] used a two-fold symmetric FE model to study mouse lung response to underwater sound. A simplified 3D model of the lung (with trachea and main stem bronchi) and rib cage was constructed. Simulations were compared to published results for mice and used in part to calculate a preliminary acoustic damage threshold curve for humans as a function of frequency. Peng et al. [17] conducted wide frequency band percussion studies on human and pig chest. Computational models of the upper torso were validated by comparison with experiment measurements.

From the above studies, it is seen that the airway geometry was significantly simplified to enable theoretical modeling and/or phantom construction. In the present study a more complex phantom model of sound transmission in the lungs is analyzed experimentally and computationally, with the goal of validating the computational simulation approach, which could be extended to even more complex and realistic airway structures.

1.2. Objectives

In Sec. 2, a self-consistent model proposed by Horsfield et al. [18, 19] is used to describe the bifurcating human airway tree. The acoustics of an airway segment with a bifurcation is also reviewed in Sec. 2. In Sec. 3, the acoustics of the branching airway network phantom are studied by measuring the acoustic pressure at network terminal segments and comparing measurements with analytical solutions and finite element simulations. Sound transmission in the solid (soft-tissue) portion of the phantom is covered in Sec. 4. The phantom top surface velocity is measured by scanning laser Doppler vibrometry and compared with finite element simulations. The validity of the simulations is assessed by comparing the phantom top surface velocity amplitude distribution patterns at several discrete frequencies and the surface velocity amplitude of some selected points over the entire frequency range of interest. Discussion is provided in Sec. 5 followed by an overall conclusion in Sec. 6.

2. THEORY

2.1. Horsfield Model of Human Airway Tree

The model of sound propagation in the sub-glottal region consists of two fundamental parts. First, it requires a mathematical description of the acoustic properties of individual airway segments, the description of which must be sufficiently sophisticated to include all of the phenomena of the gas and airway wall considered to be significant. Second, the dynamical description of individual airway segments must be integrated into a complex asymmetrical airway tree encompassing approximately 10^7 terminal branches in the case of human subjects. This task is accomplished by using the Horsfield self-consistent model of asymmetric dichotomy for the bronchial tree. Horsfield et al. [18,19] proposed an approximate but comprehensive model based on detailed lung castings for both human and canine subjects. The Horsfield model of the human lung categorizes the airway tree into 35 different segment sizes, starting with $n = 35$, the trachea, and ending with $n = 1$, the terminal bronchiole, which itself terminates into two alveoli. In addition to specifying the average of each segment's length $l^{(n)}$, diameter $d^{(n)}$, wall thickness $h^{(n)}$, and area fraction of cartilage $c^{(n)}$, the Horsfield model specifies the degree of asymmetry at each airway bifurcation through a recursion index $\alpha^{(n)}$. An airway of order n bifurcates into two airways of order $n-1$ and $n-1-\alpha^{(n)}$. The Horsfield model is self-consistent in the sense that, for a particular airway order n , the bifurcated airway types (or daughter airways) are the same throughout the lung. A partial diagram of the lung airway tree based on this model down to $n = 23$ is shown in Fig. 1, as adapted from Fredberg [20].

These bronchial models of Horsfield, as modified by Habib et al. [21,22] to account for non-rigid airway walls and terminal respiratory tissues, are based on lung airway casts and consist of acoustically one-dimensional tube-like branches that split successively into

smaller and smaller components with a self-consistent asymmetrical dichotomy terminating with the alveoli. This modeling approach ought to be valid up to at least 5 kHz; at higher frequencies pressure variations across the large airway cross sections are expected, necessitating a more complex three-dimensional acoustic description.

2.2. Mathematical and Diagrammatic Description of the Subglottal Model

It is of interest to be able to calculate the acoustic impedance (ratio of acoustic pressure to acoustic particle velocity as a function of frequency) at the input to the trachea as a function of the airway geometry and any geometrical changes it experiences. Additionally as a means of estimating sound transmission in the lung, it is also of interest to be able to explicitly calculate the acoustic pressure distribution within the airways. The self-consistent nature of the Horsfield model renders the first of these tasks, the input impedance calculation, relatively simple in spite of approximately 6 million airways present in the model. The first step is to calculate the input acoustic impedance at a terminal bronchiole, $n = 1$. Then, “march up” the recursion ladder from $n = 2$, then 3 and so on until calculating the input impedance for $n = 35$, the trachea. Referring to Fig. 2, for the n^{th} order airway segment of length $l^{(n)}$, the input acoustic impedance $Z_{in}^{(n)}[\omega]$ (taken at the end closer to the trachea) is given by [20]

$$Z_{in}^{(n)}[\omega] = \frac{Z_T^n[\omega] + Z_0^n[\omega] \tanh[\gamma_0^{(n)}[\omega] l^{(n)}]}{1 + (Z_T^n[\omega] / Z_0^n[\omega]) \tanh[\gamma_0^{(n)}[\omega] l^{(n)}]}, n=1, \dots, 35 \quad (1)$$

The terms $Z_0^{(n)}[\omega]$ and $\gamma_0^{(n)}[\omega]$ are the characteristic impedance and propagation coefficient of the n^{th} airway segment, respectively, and are computed using the following equations [21-23].

$$Z_0^{(n)}[\omega] = \sqrt{Z^{(n)}[\omega] / Y^{(n)}[\omega]} \quad (2)$$

$$\gamma_0^{(n)}[\omega] = \sqrt{Z^{(n)}[\omega] Y^{(n)}[\omega]} \quad (3)$$

$$Z^{(n)}[\omega] = \frac{j\omega\rho_g}{A^{(n)}(1 - F_\nu^{(n)}[\omega])} \quad (4)$$

$$Y^{(n)}[\omega] = \frac{j\omega A^{(n)}}{\rho_g c_g^2} (1 + 0.402 F_t^{(n)}[\omega]) + \frac{1}{Z_w^{(n)}[\omega] l^{(n)}} \quad (5)$$

$$F_\nu^{(n)}[\omega] = \frac{2}{a^{(n)} \sqrt{-j\omega\rho_g/\eta_g}} \frac{J_1[a^{(n)} \sqrt{-j\omega\rho_g/\eta_g}]}{J_0[a^{(n)} \sqrt{-j\omega\rho_g/\eta_g}]} \quad (6)$$

$$F_t^{(n)}[\omega] = \frac{2}{a^{(n)} \sqrt{-j\omega C_g/\kappa_g}} \frac{J_1 \left[a^{(n)} \sqrt{-j\omega C_g/\kappa_g} \right]}{J_0 \left[a^{(n)} \sqrt{-j\omega C_g/\kappa_g} \right]} \quad (7)$$

$$Z_w^{(n)}[\omega]^{-1} = \frac{c^{(n)}}{Z_{w,c}^{(n)}[\omega]} + \frac{1-c^{(n)}}{Z_{w,s}^{(n)}[\omega]} \quad (8)$$

Here ω is the frequency, ρ_g and η_g are the air density and viscosity. C_g and κ_g are the air specific heat and thermal conductivity. $F_v^{(n)}$ and $F_t^{(n)}$ accounts for the sound attenuation by air viscosity and thermal dissipation. $Z_w^{(n)}[\omega]$ is the effective volumetric wall impedance of the n^{th} order airway segment that is composed of two distinct material types, cartilage and soft tissue, their fractional proportions denoted by $c^{(n)}$ and $1-c^{(n)}$, respectively. Cartilage rings are present in the trachea and larger airway orders in decreasing proportion from $n = 35$ down to $n = 17$ for human subjects. For, $n = 1, \dots, 16$ only soft tissue is present ($c^{(n)} = 0$).

Both the impedance of the cartilage component $Z_{w,c}^{(n)}[\omega]$ and the soft tissue component $Z_{w,s}^{(n)}[\omega]$ can be computed by invoking a series resistance, inertance and compliance (R_w - I_w - C_w) representation of each wall compartment as

$$Z_{w, cors}^{(n)}[\omega] = R_w^{(n)} + j \left(\omega I_w^{(n)} - 1/\omega C_w^{(n)} \right) \quad (9)$$

where

$$R_{w, cors}^{(n)} = \frac{4h^{(n)} E_{I, cors}}{\pi (d^{(n)})^3 l^{(n)} \omega} \quad (10)$$

$$I_{w, cors}^{(n)} = \frac{h^{(n)} \rho_{cors}}{\pi d^{(n)} l^{(n)}} \quad (11)$$

$$C_{w, cors}^{(n)} = \frac{\pi (d^{(n)})^3 l^{(n)}}{4h^{(n)} E_{R, cors}} \quad (12)$$

Here, ρ_{cors} denotes the density, $E_{R, cors}$ and $E_{I, cors}$ denote the real and imaginary part of the Young's modulus of the cartilage or soft tissue, respectively.

The term $Z_T^{(n)}[\omega]$ denotes the acoustic impedance at the far end of each segment, which is given by

$$Z_T^{(n)}[\omega] = \begin{cases} \frac{N_T}{j\omega C_g + 1/[R_t + j(\omega I_t - 1/[\omega C_t])]} & n=1 \\ \frac{1}{1/Z_T^{(n-1)}[\omega] + 1/Z_T^{(n-1-\Delta[n])}[\omega]} & n=2, \dots, 35 \end{cases} \quad (13)$$

Here, N_T denotes the total number of terminal ($n=1$) bronchiole segments, which are effectively in parallel with respect to the termination into the soft tissue of the parenchymae. For the Horsfield human model, this can be calculated using the following recursion formula, taking $N_T^{(1)}=1$ and $N_T=N_T^{(35)}$,

$$N_T^{(n)} = N_T^{(n-1)} + N_T^{(n-1-\Delta[n])} \quad (14)$$

The result is $N_T \approx 2.35$ million in this case. The term C_g denotes the alveolar gas compression compliance based on the Dubois [24] six-element terminal airway model. Also based on the Dubois model, R_t , I_t and C_t denote the terminal tissue resistance, inertia and compliance. (The lumped airway wall inertance and resistance of the Dubois six-element model are replaced here by the more comprehensive airway tree model.)

Calculating acoustic pressure (e.g. complex values as a function of frequency referenced to the input acoustic pressure at the trachea) throughout the airway tree is not as simple as the impedance calculation. For pressure calculations, the starting point is at the trachea, $n = 35$, respectively, and then proceeds down the tree. Pressure values are not self-consistent in that all branch segments of the same order n where $n < 35$ will not have the same pressure values as they all are different lengths from the trachea. I.e., there is a different sequence of orders between them and the trachea, which results in different pressure values. While the acoustic pressure levels will not be consistent among airways of the same order “ n ”, the ratio of acoustic pressure $P_{rat}^{(n)}[\omega]$ at the far end $P_T^{(n)}[\omega]$ to that at the near end $P_{in}^{(n)}[\omega]$ (closer to the trachea) is consistent among airway segments of a particular order “ n ” and is given by:

$$P_{rat}^{(n)}[\omega] = \frac{P_T^{(n)}[\omega]}{P_{in}^{(n)}[\omega]} = \frac{Z_T^{(n)}[\omega] Z_0^{(n)}[\omega]}{Z_{in}^{(n)}[\omega] \left(Z_T^{(n)}[\omega] \sinh \left[\gamma_0^{(n)}[\omega] l^{(n)} \right] + Z_0^{(n)}[\omega] \cosh \left[\gamma_0^{(n)}[\omega] l^{(n)} \right] \right)} \quad (15)$$

Consequently, given the acoustic pressure at the input to the bronchial tree (top of trachea), the above formula is used to calculate the acoustic pressure at the base of the trachea, which is the same as the pressure at the top of the two main-stem bronchi. This process can be repeated to obtain the acoustic pressure at the top or base of any airway segment. The acoustic pressure along a particular segment can also be computed based on the pressure values at either end of the segment. Taking $x = 0$ as the end of a segment nearer to the trachea and $x=l$ as the far end, gives

$$P^{(n)}[x, \omega] = \frac{Z_0^{(n)}[\omega]}{\sinh \left(\gamma_0^{(n)}[\omega] l \right)} \cdot \left(\frac{P_{in}^{(n)}[\omega]}{Z_{in}^{(n)}[\omega]} \cosh \left(\gamma_0^{(n)}[\omega] (x-l) \right) - \frac{P_T^{(n)}[\omega]}{Z_T^{(n)}[\omega]} \cosh \left(\gamma_0^{(n)}[\omega] x \right) \right) \quad (16)$$

3. Theoretical, Experimental and Computational Studies of Sound Transmission in Airway Phantom

3.1. Method

Theoretical, experimental and computational studies were conducted on an airway phantom to gain a better understanding of (i) sound propagation through an airway tree, and (ii) sound coupling into soft biological tissues. The phantom with an airway tree was built using tissue-mimicking silicone gel Ecoflex 00-10 (Smooth-On Inc, Easton, PA) and is shown in Fig. 3. The airway tree in the phantom was created by pouring the Ecoflex 00-10 into the mold with plastic rods. The phantom dimensions are $26 \times 15 \times 6.5$ cm. The Horsfield model was adopted to categorize this airway tree into 9 different segment sizes. The schematic diagram of the airway tree is shown in Fig. 4. Segment 4 and segment 6 are the same in length, diameter and wall thickness, but their daughter segments are not the same so different Horsfield order numbers are assigned to them. Air properties and structural parameters of the airway tree are listed in Table 1 and Table 2. Here, fractional cartilage contents $c^{(n)}$ are all zero so they are not listed in Table 2.

The phantom has a density of 965 kg/m^3 . Shear viscosity of Ecoflex 00-10 was taken from magnetic resonance elastography measurements on the same material in Yasar et al. [25]. The frequency-dependent complex-valued shear modulus of the material was obtained by employing geometric focusing of the shear wave to a simple axisymmetric cylindrical phantom and fitting the measured shear wave motion to analytical solutions. The estimated shear modulus were fit to several common models of linear viscoelasticity (Fig. 5.), including those with fractional derivative operators, and the best model was identified over the entire frequency span considered (200 Hz to 7.75kHz). The best fit model was a fractional Voigt model with complex-valued shear modulus $\mu = [13.3 \times 10^3 + 1956(j\omega)^{0.34}] \text{ Pa}$ [25].

The shear wave speed and attenuation in the material are governed by the shear viscoelasticity while the compression wave speed and attenuation are governed by both the bulk and shear viscoelasticity. For linear elastic material, the bulk elasticity and shear elasticity are simply related by a real and constant Poisson's ratio. For linear viscoelastic material, the attenuations for compression and shear wave can differ, depending on the material, because an isotropic material has two independent stiffness values and two independent loss tangents [26]. Experiments were also conducted to measure the compression wave speed in the material. Fig. 6 shows the schematic diagram of the compression wave speed measurement setup.

A periodic chirp signal (100-4000 Hz) was generated from a dynamic signal analyzer (SignalCalc ACE, Data Physics, San Jose, CA) and was fed into a power amplifier (P 3500S, Yamaha, Buena Park, CA) that was connected to an electromagnetic shaker (ET-132, Lab-Works Inc., Mesa Costa, CA). An impedance head (288D01, PCB Piezotronics, Depew, NY) was mounted on the shaker and was placed in contact with the phantom with the intent of measuring the speed of sound through the 6.4 cm thickness of the sample. A small circular area on the front side of the material center was driven and its

acceleration was measured by the impedance head. Out-of-plane velocity of the circular area on the back side of material was measured by a laser Doppler vibrometer (LDV) (PDV-100, Polytec, Irvine, CA). The measurement was repeated at two other positions (3 cm to the left and right of the first position) of the material. The acceleration and velocity measurements were recorded by the same signal analyzer with a sampling frequency of 102.4 kHz. The input signal $x(t)$ is the left side acceleration measured by the impedance head. The output signal $y(t)$ is the right side acceleration which is taking derivative of the velocity measured by the LDV with respect to time. The compression speed can be calculated from the phase angle θ_{xy} of the cross-spectral density function between $x(t)$ and $y(t)$. The transit time τ is related to the slope of the θ_{xy} - frequency curve by [27]

$$\tau = -\frac{1}{2\pi} \frac{d\theta_{xy}}{df} \quad (17)$$

For linear viscoelastic material, the complex-valued compression wave number k_p is given by

$$k_p = k_{pR} + jk_{pI} = \frac{\omega}{c_p} = \frac{\omega}{\sqrt{(K + 4\mu/3)/\rho}} \quad (18)$$

Here, K is the complex-valued bulk modulus and ρ is the material density. The real part of the compression wave number k_{pR} is related to the phase speed by

$$k_{pR} = \frac{\omega}{c_{ph}} \quad (19)$$

The compression wave attenuation is governed by the imaginary part of the wave number k_{pI} . In our current experiment setup, the planar compression wave created by the impedance head connected with the shaker will become spherical wave as it travels in the sample. The measurement was not used to estimate k_{pI} as the attenuation is caused by both geometrical damping due to the spherical wave and the material viscoelasticity. To determine k_{pI} , the finite element simulation described below was performed. Here, the complex-valued compression wave speed and shear wave speed were chosen as the input material properties for the simulation. Simulation was run for k_{pI} from -50 to 0 with an increment of 5 to find the best fit k_{pI} value. From Eq. (18), the complex-valued bulk modulus can be estimated and thus the Young modulus is

$$E = E_R + jE_I = \frac{9K\mu}{3K + \mu} \quad (20)$$

In the sound transmission experiment, the airway tree acoustic pressure at each terminal segment was measured by a 1/4 inch microphone (377A01, PCB Piezotronic, Depew, NY) (Mic 2). The reference pressure was measured by inserting a 1/2 inch microphone (4176, Brüel & Kjær Sound & Vibration, Denmark) (Mic 1) into segment 10 from the phantom top surface. Periodic chirp sound wave from 100 to 1600 Hz (The major frequency content of

the breadth sound is in this range.) was introduced into the airway through a hard plastic hose inserted into the inlet by 2.5 cm. Metal pins were inserted at all terminal segments by 1 cm to create the fixed boundary conditions. The measured pressure was recorded by a dynamic signal analyzer (SignalCalc ACE, Data Physics, San Jose, CA). The frequency response function (FRF) of the two microphone measurements can then be acquired.

The FRF of the two microphone measurements can also be calculated analytically. The first step is to calculate the input acoustic impedance at a terminal segment, $n = 1$. Then, “march up” the recursion ladder until calculating the input impedance for $n = 9$. As the FRF between the reference pressure and the pressure at the airway terminal segment remains the same for different input pressure values at the inlet, so assuming the pressure at the inlet to be 1 Pa, Eq. (15) is used to calculate the acoustic pressure at the base of segment 9, which is the same as the pressure at the top of segment 8. This process can be repeated to obtain the acoustic pressure at the top or base of any segment. For example, referring to Fig. 4, we have the path 9-8-4-3. The acoustic pressure at the end of the $n = 3$ path is:

$$P_T^{(3)}[\omega] = P_{in}^{(9)}[\omega] P_{rat}^{(9)}[\omega] P_{rat}^{(8)}[\omega] P_{rat}^{(4)}[\omega] P_{rat}^{(3)}[\omega] \quad (21)$$

As the microphone can be approximated as a fixed boundary condition, the impedance $Z_T^{(n)}$ at the terminal segment approaches infinity, so Eqs (1) and (15) simplify to

$$Z_{in}^{(n)}[\omega] = \frac{Z_0^n[\omega]}{\tanh[\gamma_0^{(n)}[\omega] l^{(n)}]} \quad (22)$$

$$P_{rat}^{(n)}[\omega] = 1/\cosh[\gamma_0^{(n)}[\omega] l^{(n)}] \quad (23)$$

Take location G as an example, the FRF is

$$FRF[\omega] = \frac{P_T^{(3)}[\omega]}{P_{ref}[\omega]} = \frac{P_T^{(3)}[\omega]}{P^{(9)}[x_0, \omega]} \quad (24)$$

where x_0 is the location where Mic 1 was inserted into segment 9, the reference pressure P_{ref} can be calculated from Eq. (16).

The acoustics of the airway tree is also simulated using the acoustic-solid interaction module in Comsol 4.3a (Comsol Inc, Burlington, MA). The phantom was modeled as a linear viscoelastic solid. Since the phantom is symmetric, only the left half was simulated by creating a plane of symmetry in Comsol. A planar compression wave of 1 Pa is applied to the inlet. The places where microphones or metal pins were inserted were specified as fixed (sound hard) boundary conditions in Comsol 4.3a. Fig. 7 shows the 3D geometry of the left half of the phantom.

3.2. Result

Fig. 8 shows the θ_{xy} -frequency curve of three points for the compression wave speed measurement and it is seen that θ_{xy} can be regarded as a linear function of frequency above 1000 Hz. The phase angle below 1000 Hz was affected by artifacts as the motion of the sample material caused by the shaker had a larger influence on the θ_{xy} -frequency curve at low frequencies. By fitting each θ_{xy} -frequency curve with a linear function, the transit time at each point was estimated from Eq. (17) and the average compression wave phase speed is 120.98m/s.

It was also found from the simulation that k_{pI} greatly affects the phantom top surface velocity amplitude while it has a negligible effect on the phantom airway acoustic pressure. When k_{pI} was properly selected, the amplitude of the FRF between the phantom top surface velocity and the reference pressure were close to experiment for all frequencies. Under each k_{pI} value, the error between the simulation and experiment is defined as

$$error = \sum_{i=1}^m \sum_{j=1}^n |u_{i,j}^s - u_{i,j}^e| \quad (25)$$

where u is the amplitude of the FRF between the phantom top surface velocity and the reference pressure. The surface velocity, itself, was not chosen to calculate the error as the acoustic pressure being sent into the phantom inlet is frequency-dependent. Here, m is the number of scanning points and n is the number of frequencies. Superscript s and e denote simulation and experiment, respectively. For each k_{pI} the error was calculated and $k_{pI} = -20$ resulted in the smallest error. Then the complex-valued Young's modulus of phantom was calculated from Eq. (20).

The FRF at different terminal locations is plotted in Fig. 9. In each subplot, the red and blue curves represent experimental results measured on the left and right side of the phantom. The Comsol simulation matches the experiment reasonably well. The analytical calculation matches the experiment reasonably well above 600 Hz, but shows some discrepancy below the first peak in the experiment. A possible reason for this discrepancy is that the airway walls are assumed to be annular while in our experiment the airway tree was embedded in the phantom and so the airway wall thickness varies circumferentially.

4. Experimental and Computational Studies of Structure-Acoustic Coupling in Airway Phantom

As the walls of the phantom airway tree are viscoelastic, they will have radial motion when sound propagates through the airway tree; the sound energy is coupled into the solid part of the phantom and then travels within that solid. Sound transmission can also be detected by measuring the phantom top surface velocity. As shown in Fig. 10, bands of retro-reflective tape were adhered on the phantom top surface to enhance laser reflectivity. A total of 1,126 scanning points were defined on the retro-reflective tape. The same B&K microphone was used to measure the reference pressure. Steel plugs were inserted at all terminal segments to a depth of 1 cm to create fixed end conditions for these airways. The input sound wave was broad band with spectral content between 100 Hz and 1600 Hz. The surface velocity on top

of the phantom was measured by a scanning laser Doppler vibrometer (SLDV) (PSV-400, Polytec Inc, Irvine, CA) and the FRF of the phantom surface velocity with respect to the reference pressure was recorded. Sound transmission in the phantom was also simulated in the acoustic-solid interaction module in Comsol 4.3a. The FRF amplitudes at 400Hz, 700Hz, 1000Hz, 1300Hz and 1600Hz are shown in Figs. 11-15. Since there is no exact analytical solution of the phantom surface velocity, only the experimental and computational results are displayed. In each figure, the left half is from the simulation and the right half is from experiment as the phantom was geometrically symmetric. The color bar is in dB with the units of m/s/Pa reference to 1 Pa input pressure. In the experiment, a small area on the phantom top surface was shaded by the fixture of the microphone and thus could not be scanned by the SLDV. It is seen that the FRF amplitude distribution between simulation and experiment generally matches well at these five frequencies.

Figs. 10-15 show a good match of the FRF amplitude distribution at five frequencies, while the FE model can be further validated by checking the FRF amplitude over the whole frequency range. Six points on the phantom top surface (shown in Fig. 16) are selected and their FRF amplitudes are shown in Fig. 17. There is a generally good match between the experiment and simulation. The agreement between experiment and simulation of point 1, 3, 4 and 5 are better than that of point 2 and 6 as these four points are over or very close to the phantom airway and their FRF from simulations are less affected by the deviation from phantom material properties. To evaluate the discrepancy between the experiment and simulation, the percentage error relative to the measured FRF (phantom top surface velocity with respect to reference pressure) amplitude is defined as

$$error = \sqrt{\frac{\sum_{i=1}^n [u_i^s - u_i^e]^2}{\sum_{i=1}^n [u_i^e]^2}} \times 100 \quad (26)$$

where n is the number of frequencies, u denotes the FRF amplitude, the superscript s and e denote simulation and experiment, respectively. The percentage error of each point is listed in Table 4. Points 2 and 6 have larger percentage error than the other four points. The percentage error is acceptable as the predicted FRF curves have an overall good match with measured curves and the predicted resonant peaks are very close to the measured peaks. The environment noise could be one of the causes of discrepancy. The reflective tape on the phantom top surface could also be a possible cause.

5 Discussion

Sound transmission in a lung was studied using a mechanical phantom comprised of a branching airway network embedded in a compliant viscoelastic medium. A corresponding computational model was validated by comparing simulations with measured airway acoustic pressures and phantom surface velocity over a wide frequency range. In Acikgoz et al. [15], sound transmission through a lung phantom with an airway was studied. A polyurethane foam phantom was used to mimic the lung parenchyma. The porous foam matched some aspects of the lung parenchyma better than ecoflex. Compression wave speed

in the foam was ~30 m/s. Although this speed is closer to compression wave speed in the lung than in ecoflex, the latter has mechanical properties that better mimic the larger lung airway walls, which will help to more accurately simulate sound coupling from the airways to their walls. In addition, ecoflex properties are relatively stable and well documented (Royston et al. [28], Yasar et al. [25], Mansy et al. [29]), which thus provides for a reliable test bed to validate the FE-based computer simulation. Hence, ecoflex was used in the current study.

It is seen from Fig. 17 that the finite element simulation – with material property values based on measurements of shear wave viscoelasticity, compression wave speed and attenuation – has a good match with the experiment. Even though the compression wave attenuation in the frequency range of a few kHz can be more accurately measured by Hopkinson bar type experiments [30,31] (by making the sample into cylindrical bars and measuring stress wave propagation in the bar), the available properties from previous studies suffice here. They are provided to the model and the main objective of this study is to validate a computational acoustic model of sound transmission in an airway phantom.

In the Horsfield model of the human or dog airway tree, segments with the same length, diameter and wall thickness are assigned with a same order number n . In this study, the Horsfield model was adopted to number the individual airway segments in the phantom and couple their acoustics into the whole branching network. Although with same length, diameter and wall thickness, segment 4 and segment 6 have different order numbers as their daughter segments are different. In this way, the self-consistency of the Horsfield model is preserved (for a particular airway order n , the bifurcated airway types – daughter airways – are the same throughout the lung).

It is noted that actual lung structure is far more complex than the phantom studies here. The lung parenchyma is comprised of soft biological tissue and vasculature, as well as millions of microscopic air sacs (alveoli) that are connected through a complex branching airway structure. However, for small-amplitude mechanical wave motion in the parenchyma whose wavelengths are much larger than the microscopic heterogeneous features of the lung, macroscopic homogenized representations of the lung's physical properties have been proposed. Based on this homogenous view, two different models for wave propagation have been put forth. One is sometimes referred to as the “effective medium” or “bubble swarm” theory. It has been prominently used in the literature for modeling lung acoustics since the 1980's [32,12]. More recently, there has been an interest in applying Biot's theory of poroviscoelasticity to the lung [33,34]. Application of Biot theory leads to a more complex theoretical model that predicts a frequency dependent wave speed that, in previous studies of the authors, is a better match to experimentally measured wave speeds [34].

It is expected that the lung surface velocity amplitudes will have different distribution patterns at different frequencies. In a human lung, the airway acoustic pressure could not be measured as easily as in the phantom. Moreover, millions of airway segments will make the FE simulation computationally prohibitive. It is reported that for modeling the air flow in the human airways, the real airway tree can be simplified to an airway tree with the first few bifurcations by specifying the flow pressure and velocity at the terminal segments of the

simplified airway tree [35, 36]. From the current study the acoustic impedance at any segment in the airway tree could be calculated. This allows the real airway tree to be simplified to a tree with the first few bifurcations by specifying the acoustic impedance at the terminal segments of the simplified airway tree. An FE model could be used to calculate and visualize vibroacoustic pressure and motion inside the lung and its airways caused by the acoustic input. Lung structural and mechanical property changes can also be simulated and their influence on sound transmission in the lung could aid in the interpretation of wave images that are acquired using, for example, magnetic resonance elastography [10, 11, 37], to reconstruct a quantitative map of variation in mechanical properties that can correlate with injury, the progression of disease and/or the response to therapy.

In summary, the present study investigated sound transmission in a simplified model of the airways both experimentally and computationally. Although the geometrical structure of the airway phantom was simpler than actual airways, it is more complex than previous models (Acikgoz et al. [15]). The proposed computational simulation approach was validated, and can now be extended to more complex and realistic airway structures.

6. Conclusion

Sound transmission in the airway tree and the surrounding lung parenchymal tissue was studied using an airway-mimicking phantom with a built-in bifurcating airway tree embedded in a compliant viscoelastic medium. The Horsfield self-consistent model for the bronchial tree was adopted to topologically couple the individual airway segments into the branching airway network. Finite element simulations of phantom airway acoustics agreed well with experimental measurements and analytical solutions in the frequency range of 100 Hz to 1600 Hz except for some discrepancies at low frequencies. Simulations of sound transmission in the phantom also have good agreement with measurements made using scanning laser Doppler vibrometry (SLDV). The FE model is validated by comparing the predicted phantom top surface velocity amplitudes to SLDV measurements. The phantom used in the current study served as a platform to validate the computational approach which can now be applied in the simulation of sound transmission in much more complex airway structures, such as in the human lung. By using this computational approach in the human respiratory system to simulate a variety of pulmonary diseases, we hope to enhance understanding of vibroacoustic changes in health and disease, and potentially lead to improved tools for the diagnosis of pulmonary diseases.

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Fig. 1.
Diagram of partial airway model of Horsfield (from 35th to 23rd order), adapted from Fredberg et al. [19].

Fig. 2.
Airway acoustic model with one bifurcation.

Fig. 3.
Airway phantom with branching network inside.

Fig. 4.
Schematic diagram of the airway tree in the phantom.

Fig. 5.

Schematic representation of a Maxwell, SLS, Generalized Maxwell, Voigt, fractional order spring-pot, and fractional order Voigt model (Taken from [24], with permission).

Fig. 6.
Schematic diagram of compression wave speed measurement.

Fig. 7.
3D geometry of the left half of the phantom.

Fig. 8.
Phase angle θ_{xy} -frequency curve.

Fig. 9.

FRF of Mic 2/Mic 1 (Pa/Pa), fixed boundary condition at all terminal segments, (a) Location A&B (b) Location C&D (c) Location E&F and (d) Location G&H. For locations refer to Fig. 4. Key: — simulation, —○— theoretical prediction, — — and - - - experiment.

Fig. 10.
Experimental setup of sound coupling into phantom measurement.

Fig. 11.
FRF of phantom top surface at 400 Hz. (Left: Simulation, Right: Experiment, color bar:
m/s/Pa in dB scale)

Fig. 12.
FRF of phantom top surface at 700 Hz. (Left: Simulation, Right: Experiment, color bar:
m/s/Pa in dB scale)

Fig. 13. FRF of phantom top surface at 1000 Hz. (Left: Simulation, Right: Experiment, color bar: m/s/Pa in dB scale)

Fig. 14. FRF of phantom top surface at 1300 Hz. (Left: Simulation, Right: Experiment, color bar: m/s/Pa in dB scale)

Fig. 15. FRF of phantom top surface at 1600 Hz. (Left: Simulation, Right: Experiment, color bar: m/s/Pa in dB scale)

Fig. 16.
Selected points on phantom top surface

Fig. 17.

FRF of selected points on phantom top surface. (a) Point 1 (b) Point 2 (c) Point 3 (d) Point 4 (e) Point 5 and (f) Point 6. Key: — experiment, — simulation.

Table 1

Material properties of the phantom

air density	ρ_g (kg/m ³)	1.23
air specific heat	C_g (cal/kg/K)	240
air viscosity	η_g (Pa s)	1.82×10^{-5}
sound speed in air	c_g (m/s)	343
air thermal conductivity	κ_g (cal/m/s/K)	6.4×10^{-3}

Table 2

Structural parameters of the airway tree (n , Horsfield airway order; $l^{(n)}$ and $d^{(n)}$, airway length and diameter, respectively; $h^{(n)}$, wall thickness, $i^{(n)}$, recursion index)

n	$l^{(n)}$ (cm)	$d^{(n)}$ (cm)	$h^{(n)}$ (cm)	$i^{(n)}$
1	2.73	0.6	2.2	0
2	4.3	1.3	1.85	0
3	7.7	0.6	2.2	0
4	3.12	0.6	2.2	2
5	3.7	0.6	2.2	0
6	3.12	0.6	2.2	4
7	4	1.3	1.85	4
8	3.7	1.3	1.85	3
9	6.85	2.2	1.4	0

Table 3

Coordinates of selected points on phantom top surface (unit: mm)

Point #	x	y	z
1	40.35	25.00	-38.32
2	84.99	25.00	-38.17
3	56.29	25.00	-0.41
4	129.13	25.00	-0.19
5	93.95	25.00	37.83
6	111.61	25.00	37.89

Table 4

Percentage error relative to the measured FRF amplitude

Point #	error %
1	3.85
2	6.76
3	4.09
4	3.73
5	4.88
6	6.11