THE POROVISCOELASTIC MODEL OF THE LUNGS UNDER LOW-FREQUENCY PREDICTED BY BIOT'S THEORY

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1 Introduction

The conventional method for chest physiotherapy involves a technique called clapping, generating vibrations on the chest surface [1]. This action influences the properties of bronchial mucus, making it more fluid and easier to expel due to its viscoelastic, shear-thinning, and thixotropic characteristics [2]. Recent research suggests that acoustic airway clearance devices (ACDs) may enhance this process significantly, being potentially more effective than traditional methods and supplying autonomy for patients [2]. Consequently, High-Frequency Chest Compression (HFCC) therapy, facilitated by such devices, has emerged as the prevalent approach for managing excessive mucus buildup [1,2]. However, there is inconsistency in these kinds of devices' operating conditions such as frequency range [2]. Therefore, a realistic numerical model is necessary to develop a representative model of the thorax to test the frequency ranges [3].

The numerical model of the human organs, especially the lungs, the unique poroviscoelastic material, needs to be determined precisely for an accurate model. Key assumptions within Biot's framework are tailored to the pulmonary environment [4,5]. Firstly, it posits infinitesimal transformations between reference and current states of lung tissue deformation, enabling the application of continuum mechanics to observable macroscopic values [5]. These values are derived as volume averages of corresponding microscopic quantities, reflecting the homogenized nature of lung tissue.

A numerically developed [3] and validated [6] CT-FEM is used in this study at 28 Hz as a resonance frequency and 0.138 m/s^{2}N peak in frequency response function (FRF). The Biot's theory offers a crucial lens through which to understand the propagation of waves within the intricate structure of the lungs, where the solid framework of the lung tissue interacts with the fluid saturation of air [3]. Biot's theory abstracts away from the microscopic complexities, allowing us to apply continuum mechanics to discernible macroscopic quantities, illuminating the dynamics of airflow and tissue deformation in different transpulmonary pressures [5].

In this study, it is aimed to investigate the poroviscoelastic behavior of the lungs considering its accelerance at the low-frequencies by using the Biot's theory at 5-100 Hz, by a realistic 3D FEM of the human thorax. Therefore, we illustrate the accelerance behaviour of the thorax response at the low-frequency range, which is supported by an experimental study. Thus, this study contributes to the application of the Biot's theory not only to offer a theoretical foundation for understanding wave propagation in the lungs but also to provide insights critical for studying respiratory function and dysfunction, guiding advancements in pulmonary medicine and respiratory therapy.

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2 Method

The human thorax model has been used by using CT scans in order to have accurate geometries [3,6]. The HFCC effect was applied as determined in our previous numerical study. After solving the frequency domain, internal organs, especially the lungs were checked. Many points were selected on the lungs to take the average of the lungs for the acceleration, as shown in the figure. To homogenize the heterogeneous media, fully saturated material features of the lungs, the Biot's theory is used to calculate the complex fast compression waves (c_{pf}) and slow compression waves (c_{ps}) as well as shear wave speeds (c_s). The physical properties of the lungs have been calculated by using the Biot's theory by using the following Eqn. 1 and in Eqn. 2 [5].

$$\mu u_{i,jj} + \left(K + \frac{\mu}{3}\right) u_{j,ij} - (\alpha - \beta) P_{,i} + F_e = -\omega^2 (\rho - \beta \rho_f) u_i \quad (1)$$

$$\beta p_{,ii} + \frac{\varphi^2}{R} \rho_f \omega^2 P + \rho_f j \omega r_g = -\rho_f \omega^2 (\alpha - \beta) u_{i,i} \quad (2)$$

The first equation (1) describes the dynamic displacement (u), dynamic air pressure (p_a), coupling parameters (δ, ε, ζ), external forces (F), angular velocity (ω), and gas volume rate (α). In the absence of external excitation, it simplifies equation (3), illustrating shear behaviour. Furthermore, Eqn (2) represents the dynamic pressure (p), and the gas volume rate (α), and the effects of external excitation and the other parameters were considered and calculated as it is described in our previous study [5].

3 Results of the Acceleration Amplitude of the Lungs and Discussion

As a result of this study, 3D FEM of the human thorax is used, and the acoustic harmonic excitation is investigated by a 28 mm radius cylindrical shape under 146 dB_{SPL} onto the back-chest surface to represent the HFCC therapy [6]. The acceleration amplitude data is read from 5 k different homogeneously distributed points as illustrated in Fig. 1a and Fig. 1b for 10 cm H_2O and 20 cm H_2O, respectively in the frequency range of 5-100 Hz. The average acceleration was found as 0.15336 m/s^2 at 46 Hz and 0.15416 m/s^2 at 47 Hz for 20 cm H_2O and 10 cm H_2O, respectively with having similar acceleration behaviour.

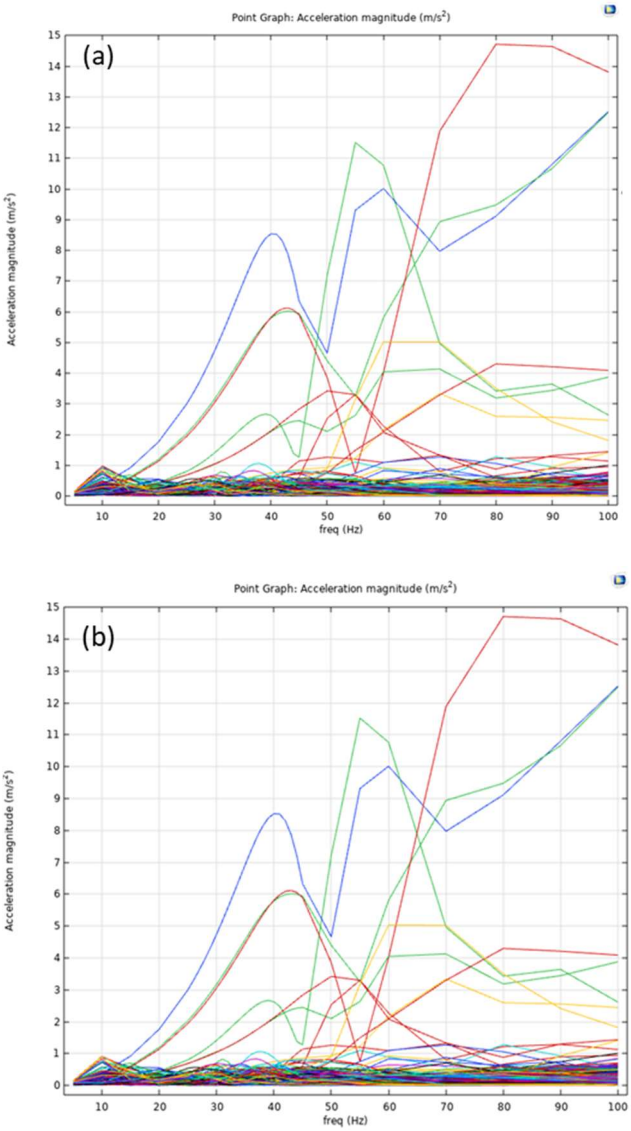

Figure 1: Acceleration magnitude at 5k points on the lungs according to different transpulmonary pressures: (a) 10 cm H_2O and (b) 20 cm H_2O

This would be explained by an experimental study as they investigated that wave with velocities similar to lung shear waves were detected at transpulmonary pressures above 15 cm H_2O in the non-edematous lung and above 25 cm H_2O transpulmonary pressures in the edematous lung [7]. Furthermore, they investigate a cut-off frequency of 40 Hz at 25 cm H_2O transpulmonary pressures and lung density of 0.2 g/cm^3 [7], which is close to both lungs' behaviour investigated in this study.

4 Conclusion

This study clarifies the internal effects of HFCC on the lungs under an acoustic ACD by numerical simulations. This paper revealed the acceleration of the lungs at different transpulmonary pressures using previously validated CT-FEM of the human thorax. Similar average values are obtained at these different transpulmonary pressures, which could be explained by another experimental study.

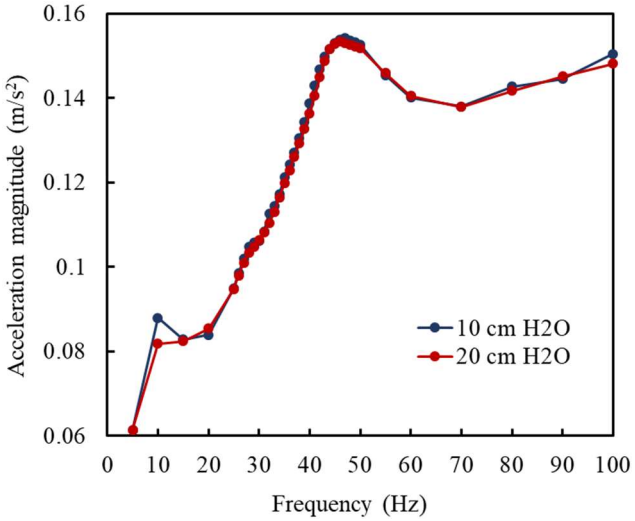

Figure 2: Average acceleration amplitude of the lungs at different transpulmonary pressures: 10 cm H_2O and 20 cm H_2O.

In a future study, a set of new numerical simulations is planned to obtain both the kinetic energy density and strain energy density of the lungs as they illustrate the material features that are the most important for respiratory care. Therefore, the objective of future work will be to identify the effectiveness of the ACDs considering the material features of the lungs to optimize mucus transport in the lungs.

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