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A noninvasive ultrasound elastography technique for measuring surface waves on the lung

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Abstract

The purpose of this work was to demonstrate an ultrasound based surface wave elastography (SWE) technique for generating and detecting surface waves on the lung. The motivation was to develop a noninvasive technique for assessing superficial lung tissue disease including interstitial lung disease (ILD). ILD comprises a number of lung disorders in which the lung tissue is stiffened and damaged due to fibrosis of the lung tissue. Currently, chest radiographs and computed tomography (CT) are the most common clinical methods for evaluating lung disease, but they are associated with radiation and cannot measure lung mechanical properties. The novelty of SWE is to develop a noninvasive and nonionizing technique to measure the elastic properties of superficial lung tissue. We propose to generate waves on the lung surface through wave propagation from a local harmonic vibration excitation on the chest through an intercostal space. The resulting surface wave propagation on the lung is detected using an ultrasound probe through the intercostal space. To demonstrate that surface waves can be generated on the lung, an *ex vivo* muscle-lung model was developed to evaluate lung surface wave generation and detection. In this model, swine muscle was laid atop a swine lung. A vibration excitation of 0.1 second 100 Hz wave was generated on the muscle surface and the surface waves on the lung were detected using a linear array ultrasound probe at 5 MHz. To test its feasibility for patient use, SWE was used to measure both lungs of an ILD patient through eight intercostal spaces. The mean wave speed was 1.71 ± 0.20 m/s ($\pm$ SD) at the functional residual capacity, while the mean wave speed was 2.36 ± 0.33 m/s at the total lung capacity. These studies support the feasibility of SWE for noninvasive measurement of elastic properties of lung and demonstrate potential for assessment of ILD.

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1. Introduction

Lung disease is the third leading killer in the United States. Lung disease death rates are increasing, while death rates due to other major causes of death, such as heart disease and cancer, are declining [1]. Many lung diseases such as interstitial lung disease (ILD), chronic obstructive pulmonary disease, and acute respiratory distress syndrome are associated with dramatic changes in mechanical properties of the lung. ILD is an umbrella term for various lung disorders that are associated with dramatic changes in lung stiffness due to fibrosis of the lung parenchyma. ILD can lead to other complications including pulmonary hypertension and respiratory failure. The incidence rate of ILD has increased from 27 to 34 per 100,000 person-years recently [2]. Diagnosis of lung fibrosis can be difficult, especially early in the disease course, because the symptoms are nonspecific (most commonly shortness of breath and a dry cough) [3]. Current diagnostic tools include medical history and physical examination, chest radiography, computed tomography (CT), pulmonary function tests (PFTs), and lung biopsy. The findings of physical examinations are usually nonspecific. ILD may be first suspected after an abnormal chest radiograph, but in most cases, the radiographs also provide nonspecific diagnosis. High-resolution CT (HRCT) is the clinical standard for diagnosing lung fibrosis [4, 5]. HRCT is a special type of CT acquisition technique that uses 0.5-1 mm thick slices to produce high detail images. However, HRCT substantially increases radiation exposure for patients. In addition, the potential for frequent HRCT use is limited by its expense. Lung fibrosis results in stiffened lung tissue. However, CT provides imaging of the lung but not measurement of lung elastic properties.

We propose a novel and noninvasive surface wave elastography (SWE) technique to measure the elastic properties of superficial lung tissue. SWE uses a handheld electromagnetic shaker to generate a local harmonic vibration of 100 Hz on the chest and produces surface wave propagating on the lung. The surface wave propagation on the lung is detected using a 5 MHz ultrasound probe through an intercostal space. The purpose of this work was to demonstrate the SWE technique for generating and detecting surface waves on the lung. The paper is organized as follows: Section 2 is a brief description of the principles of SWE; Section 3 describes an *ex vivo* muscle-lung model developed to evaluate lung surface wave generation and detection. *In vivo* testing of both lungs of an ILD patient through eight intercostal spaces is presented; Section 4 is discussion of the SWE technique; and Section 5 provides conclusions from these results.

2. Methods

In the ultrasound-based SWE technique (**Fig. 1**), a handheld shaker is made using an electromagnetic shaker (Model: FG-142, Labworks Inc., Costa Mesa, CA 92626). The handheld shaker applies a local excitation on the skin through an indenter with 3 mm diameter. The excitation is a 0.1 second harmonic vibration (for example, 10 cycles of 100 Hz signal). The resulting propagation of the tissue wave motion (typically a few micrometers) is detected using an ultrasound probe. Due to the vibration excitation on the tissue surface, the surface wave propagation along the tissue surface is used to examine the surface tissue, while the shear wave propagation inside the subcutaneous tissue is used to

examine the deep tissue. The measurements of wave speed and wave attenuation enable calculation of viscoelastic properties. In our previous study on skin viscoelasticity of scleroderma patients, we found that both elasticity and viscosity were higher in scleroderma patients than those of healthy subjects [6]. In order to measure surface waves on the lung, the indenter of the handheld shaker is placed on the skin of an intercostal space. The ultrasound probe is positioned about 5 mm away from the indenter in the same intercostal space to enable measurement of the generated surface wave propagation on the lung.

Surface wave propagation can be analyzed as wave propagation in a semi-infinite linear viscoelastic medium under harmonic excitation by a uniformly distributed stress on a circular surface area. The equation of wave propagation in an isotropic and linear viscoelastic medium is [7, 8]

$$(\lambda+2\mu) \nabla \nabla \cdot \vec{u} - \mu \nabla \times \nabla \times \vec{u} = \rho \frac{\partial^2 \vec{u}}{\partial t^2}, \quad (1)$$

where $\vec{u}$ is the displacement, ρ is the density of the medium, and λ and μ are, respectively, the Lamb coefficients of the medium.

The problem can be solved in the cylindrical polar co-ordinate system (**Fig. 1c**). Consider a harmonic force excitation with a uniform stress on the surface of the medium in the circular region of $r = a$. The displacement fields are derived in the r and z directions at any location in and on the surface of the medium as

$$\begin{aligned} u_z &= \frac{a}{\mu} \int_0^\infty \frac{\sqrt{(\xi^2-1)} J_1(\xi k_1 a)}{F_0(\xi)} \left\{ 2\xi^2 e^{-k_1 z} \sqrt{(\xi^2-\eta^2)} + (\eta^2 - 2\xi^2) e^{-k_1 z} \sqrt{(\xi^2-1)} \right\} J_0(\xi k_1 r) d\xi \\ u_r &= \frac{a}{\mu} \int_0^\infty \frac{\xi J_1(\xi k_1 a)}{F_0(\xi)} \left\{ 2 \sqrt{(\xi^2-1)} \sqrt{(\xi^2-\eta^2)} e^{-k_1 z} \sqrt{(\xi^2-\eta^2)} + (\eta^2 - 2\xi^2) e^{-k_1 z} \sqrt{(\xi^2-1)} \right\} J_1(\xi k_1 r) d\xi, \end{aligned} \quad (2)$$

where a is the radius of the distributed stress, and ξ is the integration parameter in the wave number domain, which has been normalized with respect to k_1 . The divisor function of the integration functions is $F_0(\xi) = (2\xi^2 - \eta^2)^2 - 4\xi^2 \sqrt{(\xi^2-1)} \sqrt{(\xi^2-\eta^2)}$, where $\eta = k_2/k_1 = \sqrt{2(1-\sigma)/(1+2\sigma)}$, $k_1 = \omega \sqrt{\rho/(\lambda+2\mu)}$, $k_2 = \omega \sqrt{\rho/\mu}$, σ being Poisson's ratio for the medium, k_1 and k_2 denote the wave numbers for compression and shear wave propagation, respectively, and J_0 and J_1 refer to Bessel functions of the first kind.

The wave motion can be solved using equation (2). However, our method is to measure the wave speed for estimating the viscoelasticity of tissue. The wave speed is only a function of local tissue material properties and does not depend on the detection and excitation. The surface wave speed can be solved by $F_0(\xi) = (2\xi^2 - \eta^2)^2 - 4\xi^2 \sqrt{(\xi^2-1)} \sqrt{\xi^2 - \eta^2} = 0$.

We have found that the surface wave is 5% slower than the shear wave for soft tissues. The surface wave speed can be related to the shear modulus of tissue as [8]

$$c_s = \sqrt{\mu / \rho / 1.05}. \quad (3)$$

The viscosity of tissue can be analyzed by the decay of wave amplitude with distance. Alternatively, viscoelasticity may be expressed using the Kelvin-Voigt model

$\mu(t) = \mu_1 + \mu_2 \dot{E} \dot{Z} / \dot{E} \dot{Z}(t)$ where μ_1 is shear elasticity and μ_2 is shear viscosity. Equation (3) can be modified as [8, 9]

$$c_s = \frac{1}{1.05} \sqrt{\frac{2(\mu_1^2 + \omega^2 \mu_2^2)}{\rho(\mu_1 + \sqrt{\mu_1^2 + \omega^2 \mu_2^2})}}, \quad (4)$$

where ρ is mass density, and ω is the angular frequency.

The wave speed is measured by the phase gradient method using $c_s(\omega) = \omega r / \phi$, where r is the distance between two detection locations, ϕ is the phase change over that distance, and ω is the angular frequency. The estimation of wave speed can be improved by measuring the phase change over multiple locations using a regression model $\Delta \hat{\phi} = -\alpha \Delta r + \beta$, where $\Delta \hat{\phi}$ denotes the regression value of multiple ϕ measurements, α and β are regression parameters, and $c_s(\omega) = \omega / \alpha$. A good measurement of wave speed can be assured by the sum of squares of regression residuals (R^2) being 0.8. By using wave speed measurements at several frequencies, μ_1 and μ_2 can be estimated from **Eq. 4** by using a nonlinear least-squares fitting technique [9]. Young's modulus E can be obtained from the shear elasticity μ measurement by $E = 3\mu$ [10]. The shear wave in tissue is 5% faster than the surface wave, and **Eq. 4** is applied to the shear wave without the factor of 1.05 [8].

3. Experiments and Results

Implementation of SWE into a commercially available ultrasound system requires only software modification. A function generator (Model: FG33120A, Hewlett Packard, Palo Alto, CA 94301) sends a digital pulse to the ultrasound system (**Fig. 2**), which initiates collection of ultrasound data. At the same time, a synchronized signal of vibration excitation is amplified and sent to a handheld shaker, which applies a local vibration to the skin. The propagation of the resulting waves in the skin and underlying tissue are then measured by the ultrasound probe. SWE has been implemented using a Sonix RP ultrasound system (Ultrasonix Medical Corp; Richmond, BC, Canada) for applications in multiple tissues, including lung [11], skin [9], abdominal muscles [12], and tendons [13]. A newer implementation of SWE uses a Verasonics ultrasound system (Verasonics, Inc; Kirkland, WA 98053). The Verasonics system can collect high frame rate imaging data up to a few thousand frames per second by using plane wave transmit pulses. This allows us to collect

wave imaging data in a few hundred Hertz for the whole imaging area. The Verasonics system collects data from all tracking beams in one test, whereas, the Ultrasonix system collects data only from a few tracking beams depending on the depth of imaging and frame rate.

Ex vivo swine lungs were used to develop the technique for measuring lung tissue viscoelasticity [11, 14, 15]. Ultrasound measurements were also validated on *ex vivo* swine lungs using an optical device (laser vibrometer system, Polytec-PI, Inc., Auburn, MA 01501). The effect of pulmonary pressure on surface wave speed was investigated using *ex vivo* swine lungs. A swine lung was pressurized through a device inserted in the trachea. The lung was inflated or deflated using a syringe pump (Model 210, KD Scientific Inc., Holliston, MA 01746). The pressure inside the lung was measured by a pressure transducer (PX319, Omegadyne Inc., Sunbury, OH 43074). Inspiration and expiration were simulated by injecting air into or withdrawing air from the lung [16]. Our results indicated that wave speed increased with the pulmonary pressure consistent with pulmonary mechanics and physiology.

In order to demonstrate SWE for generating wave propagation on the lung, an *ex vivo* muscle-lung model was developed to evaluate lung surface wave generation and detection. In this model, swine muscle was laid atop a swine lung (**Fig. 3a**). A vibration excitation of 0.1 second 100 Hz wave was generated on the muscle surface. The surface wave propagation on the lung was detected using a linear array ultrasound probe L7-4 with a central frequency of 5 MHz (Verasonics, Inc, Kirkland, WA 98053). Under ultrasound image guidance, a region of interest (ROI) on the lung surface was selected to analyze lung surface waves. For example, the yellow area in **Fig. 3b** shows a 1×8 mm portion of the lung surface. This ROI was under a 20 mm thick muscle layer. The excitation indenter was placed on the muscle about 25 mm lateral to the ROI. The maximum tissue velocity in the ROI was about 2 mm/second for the 100 Hz excitation. **Fig. 3c** clearly shows wave propagation in the ROI. The wave speed at each pixel in the ROI was analyzed by the phase gradient method. The wave speed was 2.25 ± 0.38 m/second in the ROI, which was consistent with our previous results. This muscle-lung model demonstrates that the surface wave propagation on the lung can be generated by excitation on the muscle.

Human lung studies were approved by the Mayo Clinic Institutional Review Board (IRB). Detection of *in vivo* human lung surface motion is guided by ultrasound imaging. In **Fig. 4a**, eight locations over a length of 7 mm on the lung surface were selected for analyzing surface wave propagation on the lung. The tissue motion at a given location on the lung surface can be analyzed by cross-correlation analysis of the ultrasound tracking beam [17]. The tissue motion is measured at these locations in response to a harmonic vibration on the chest wall. A high frame rate (1,000-2,000 frames/second) is used to detect tissue motion in response to the vibration excitation between 100 and 200 Hz. **Fig. 4b** shows the surface motion at the first location for a patient, which was in response to a 0.2-second excitation at 100 Hz. The linear array ultrasound probe L7-4 was used. The tissue velocity was less than 0.5 mm/second, but the response signal was clear by the ultrasound tracking method. Surface wave speed is measured by determining the change in wave phase at the remaining locations relative to the first location (**Fig. 4c**) using eight ultrasound tracking beams [8, 11, 18]. The

surface wave speed was 2.29 ± 0.33 m/second with a 95% confidence interval in the format of mean $\pm$ standard error. The wave speed on the lung surface is determined by analyzing ultrasound data directly from the lung. Therefore, the wave speed measurement, consequently, is local and independent of the source of excitation.

Table 1 shows the wave speed measurements for both lungs of the patient through eight intercostal spaces. The patient was examined in a sitting position. The anterior upper lobes of both lungs were tested through the second intercostal space in the midclavicular line. The lower lateral areas were tested through the fourth intercostal space in the lateral midaxillary line. The posterior lungs were tested at the infrascapular and suprascapular levels. The patient was tested at two lung volumes. Functional residual capacity (FRC) is at the end of a normal tidal expiration and total lung capacity (TLC) is at the end of a maximal inspiration.

The mean wave speed is 1.71 ± 0.20 m/s in the format of mean $\pm$ standard deviation (SD) at FRC. The mean wave speed is 2.36 ± 0.33 m/s ($\pm$ SD) at TLC. The wave speed is greater at TLC than at FRC, which is consistent with lung physiology. The patient's measurements highlight several concepts. First, wave speed values are relatively higher in the patient with ILD compared with a healthy volunteer. For example, wave speeds in the second space were 2.95 vs 2.4 m/second for the patient and a healthy volunteer, respectively, at the same place for the TLC volume [11]. Second, wave speeds were consistent over multiple areas in keeping with diffuse disease. Third, some local variation in wave speed may provide additional information on focal disease and regional tissue heterogeneity. For the patient, the wave speed at the right midclavicular line was higher than that of the left side (1.84 vs 1.46 m/second, respectively). The reason for this variation is uncertain, but she had undergone a surgical lung biopsy on the right side a few years ago. Fourth, more changes of wave speed were apparent in the upper lobes than in the lower lobes during a maximal inspiration. The lower lobes were more affected by ILD, which may reduce expansion of the lower lobes with maximum inspiration.

The *ex vivo* muscle-lung model demonstrates that surface wave propagation can be generated on the muscle surface. Measurements of *in vivo* human lungs for a patient through multiple intercostal spaces suggest that SWE is feasible for noninvasive assessment of lung mechanical properties.

4. Discussion

SWE is a noninvasive and clinically feasible method for measuring lung elastic properties. SWE uses a handheld shaker to generate a harmonic mechanical vibration on the chest and an ultrasound probe to measure the resulting surface wave propagation on the lung through an intercostal space. Because the surface wave is generated by mechanical vibration on the chest, high intensity ultrasound is not used for generating waves. Diagnostic ultrasound is only used for wave detection. The SWE technique is a safe method for testing lung and screening patients.

SWE uses the surface wave to evaluate superficial lung tissue. Most elastography methods are based on shear wave propagation in an infinite medium. However, these techniques are

not technically suitable for measuring tissue surfaces, and most ILDs typically affect the lung surface. SWE is capable of measuring both superficial and internal tissues. Some ultrasound elastography methods use ultrasound radiation force to generate the shear waves inside the tissue. These techniques include acoustic radiation force impulse [19], supersonic shear imaging [20], and the bending wave method in vessels [21]. Ultrasound radiation force should not be applied to lung tissue because it may cause lung tissue injury due to the air inside the lung tissue.

SWE may be used to assess multiple lung disorders such as airless or “wet” lungs. Wet lungs, so termed because the alveoli are filled with fluid, can be imaged by ultrasonography [22]. In “wet” lungs, ultrasound may penetrate deeper in the diseased tissue than healthy lung tissue. Therefore, deep “wet” lung tissue may be imaged using ultrasound. Lung ultrasonography is an emerging technique for evaluating lung disorders such as pneumothorax [23], peripheral lung lesions [24], and lung consolidations [25]. SWE provides both ultrasonography imaging of lung structures and measurements of tissue elastic properties, thereby, offering an additional tool for assessing lung disease.

5. Conclusion

In this paper, an ultrasound based surface wave elastography (SWE) technique is proposed for measuring surface wave propagation on the lung. An *ex vivo* muscle-lung model demonstrates that the surface wave propagation on the lung can be generated by a mechanical vibration on the surface of an overlaid muscle layer. SWE was used to test both lungs of an ILD patient through eight intercostal spaces. The mean wave speed was 1.71 ± 0.20 m/s ($\pm$ SD) at the functional residual capacity, while the mean wave speed was 2.36 ± 0.33 m/s at the total lung capacity. These studies support the feasibility of SWE for noninvasive and safe testing of human lungs. SWE may be useful for frequent follow-up of ILD patients to assess disease progression and treatment.

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References

- [1]. Lung Disease Data: 2008. American Lung Association; New York: 2008.
- [2]. Kornum JB, et al. The incidence of interstitial lung disease 1995-2005: a Danish nationwide population-based study. *BMC Pulm Med.* 2008; 8:24. [PubMed: 18983653]
- [3]. Sperandio M, et al. Transthoracic ultrasound in the evaluation of pulmonary fibrosis: our experience. *Ultrasound Med Biol.* May.2009 35:723–9. [PubMed: 19111972]
- [4]. Mathieson JR, et al. Chronic diffuse infiltrative lung disease: comparison of diagnostic accuracy of CT and chest radiography. *Radiology.* Apr.1989 171:111–6. [PubMed: 2928513]
- [5]. Verschakelen JA. The role of high-resolution computed tomography in the work-up of interstitial lung disease. *Curr Opin Pulm Med.* Sep.2010 16:503–10. [PubMed: 20644479]

- [6]. Zhang X, et al. Quantitative assessment of scleroderma by surface wave technique. *Med Eng Phys.* Jan.2011 33:31–7. [PubMed: 20888282]
- [7]. Miller GF, Pursey H. The field and radiation impedance of mechanical radiations on the free surface of a semi-infinite isotropic solids. *Proceedings of the Royal Society of London.* 1954:521–541. Series A, Mathematical and Physical Sciences.
- [8]. Zhang X, Greenleaf JF. Estimation of tissue's elasticity with surface wave speed. *J Acoust Soc Am.* Nov.2007 122:2522–5. [PubMed: 18189542]
- [9]. Zhang X, Osborn TG, Pittelkow MR, Qiang, Kinnick RR, Greenleaf JF. Quantitative assessment of scleroderma by surface wave technique. *Medical Engineering & Physics.* 2011; 33:31–37. [PubMed: 20888282]
- [10]. Sarvazyan A, et al. An Overview of Elastography - an Emerging Branch of Medical Imaging. *Curr Med Imaging Rev.* Nov.2011 7:255–282. [PubMed: 22308105]
- [11]. Zhang X, Qiang B, Hubmayr RD, Urban MW, Kinnick R, Greenleaf JF. Noninvasive ultrasound image guided surface wave method for measuring the wave speed and estimating the elasticity of lungs: A feasibility study. *Ultrasonics.* 2011; 51:289–295. [PubMed: 20971489]
- [12]. Zielinski M, Zhang X. The Biomechanical Effects of Flaccid Paralysis Induced by Botulinum Toxin a after Damage Control Laparotomy: A Randomized Clinical Trial. *Mayo Clinic IRB Human study.* 2012
- [13]. Wang Y, et al. A non-invasive technique for estimating carpal tunnel pressure by measuring shear wave speed in tendon: a feasibility study. *J Biomech.* Nov 15.2012 45:2927–30. [PubMed: 23031416]
- [14]. Zhang, X.; Kinnick, RR.; Greenleaf, JF. Viscoelasticity of lung tissue with surface wave method; presented at the 2008 IEEE International Ultrasonics Symposium; 2008.
- [15]. Zhang, X.; Qiang, B.; Urban, MW.; Kinnick, R.; Hubmayr, RD.; Greenleaf, JF. Quantitative surface wave method for measuring local viscoelasticity of lungs; presented at the 2009 IEEE International Ultrasonics Symposium; 2009.
- [16]. Venegas JG, et al. A comprehensive equation for the pulmonary pressure-volume curve. *J Appl Physiol.* Jan.1998 84:389–95. [PubMed: 9451661]
- [17]. Hasegawa H, Kanai H. Improving accuracy in estimation of artery-wall displacement by referring to center frequency of RF echo. *IEEE Trans Ultrason Ferroelectr Freq Control.* Jan.2006 53:52–63. [PubMed: 16471432]
- [18]. Zhang X, et al. Comparison of the surface wave method and the indentation method for measuring the elasticity of gelatin phantoms of different concentrations. *Ultrasonics.* 2011; 51:157–164. [PubMed: 20800256]
- [19]. Nightingale K, et al. Shear-wave generation using acoustic radiation force: in vivo and ex vivo results. *Ultrasound Med Biol.* Dec.2003 29:1715–23. [PubMed: 14698339]
- [20]. Bercoff J, et al. Supersonic shear imaging: a new technique for soft tissue elasticity mapping. *IEEE Trans Ultrason Ferroelectr Freq Control.* Apr.2004 51:396–409. [PubMed: 15139541]
- [21]. Zhang X, et al. Noninvasive method for estimation of complex elastic modulus of arterial vessels. *IEEE Trans Ultrason Ferroelectr Freq Control.* Apr.2005 52:642–52. [PubMed: 16060513]
- [22]. Volpicelli G. Lung sonography. *J Ultrasound Med.* Jan.2013 32:165–71.
- [23]. Noble VE. Think ultrasound when evaluating for pneumothorax. *J Ultrasound Med.* 2012; 31:501–504. [PubMed: 22368142]
- [24]. Lie CH, et al. New image characteristics in endobronchial ultrasonography for differentiating peripheral pulmonary lesions. *Ultrasound Med Biol.* Mar.2009 35:376–81. [PubMed: 19110372]
- [25]. Barillari A, Franco FD, Colonna F. Chest ultrasound helps to diagnose pulmonary consolidations in pediatric patients. *J Medical Ultrasound.* 2011; 19:27–31.

Figure 1.

(a) In surface wave elastography (SWE), a electromagnetic shaker is used to generate a 0.1 second harmonic vibration at 100 Hz on the surface of tissue. The resulting surface wave propagation on the tissue and shear wave propagation inside the tissue are detected using an ultrasound probe at 5MHz; (b) Photos of a handheld shaker and an ultrasound probe for a SWE system; (c) The cylindrical polar coordinate system is used to analyze wave propagation in the tissue in response to a mechanical excitation on the tissue. The tissue surface is on the plan of x and y coordinates.

Figure 2.

Implementation of SWE into a commercially available ultrasound system requires only software with no modification of the system hardware. A function generator (FG33120A) sends a digital pulse to the ultrasound system, which initiates collection of ultrasound data. At the same time, a synchronized signal of 0.1 second of 100 Hz wave excitation is amplified and sent to a handheld shaker, which applies a local 0.1 second vibration of 100 Hz on the tissue. The propagation of the vibration wave on the tissue and inside the tissue is measured using the ultrasound probe at 5 MHz.

Figure 3.

(a) An *ex vivo* muscle-lung model was used to evaluate lung surface wave generation and detection. In this model, swine muscle was laid atop a swine lung. A mechanical vibration of 0.1 second 100 Hz was generated on the muscle surface. The surface wave propagation on the lung was detected using a linear array ultrasound transducer at 5 MHz; (b) A region of interest (ROI) on the lung surface was selected to analyze lung surface waves. This ROI was under a 20-mm thick muscle layer; (c) Surface wave imaging in the ROI in response to a 100 Hz excitation on the muscle surface. This muscle-lung model demonstrates that waves on the lung surface can be generated by mechanical excitation on the muscle.

Figure 4.

(a) In *in vivo* human lung testing, eight locations over a length of 7 mm on the lung surface were selected to measure the tissue surface motion using ultrasound tracking beams; (b) Lung surface motion at the first location was in response to a 0.2 second 100-Hz vibration excitation; (c) The wave phase delay of the remaining locations, relative to the first location, is used to measure the surface wave speed at 100 Hz.

Table 1

Measurements of surface wave speed (m/s) at 100 Hz through eight intercostal spaces for a patient. The lung was tested at the functional residual capacity (FRC) and the total lung capacity (TLC). Both lungs were tested. The anterior upper lungs were tested through the second intercostal space in the midclavicular line. The lower lateral lungs were tested through the fourth intercostal space in the lateral midaxillary line. The posterior lungs were tested at the infrascapular and suprascapular levels.

Lung volume	Anterior upper area through the second space		Lateral lower area through the fourth space		Posterior lower area at the infrascapular level		Posterior upper area at the suprascapular level	
	Left side	Right side	Left side	Right side	Left side	Right side	Left side	Right side
At FRC	1.46 ± 0.11	1.84 ± 0.15	1.55 ± 0.17	1.98 ± 0.23	1.99 ± 0.19	1.64 ± 0.17	1.60 ± 0.14	1.64 ± 0.21
At TLC	2.95 ± 0.54	2.72 ± 0.30	2.29 ± 0.15	2.20 ± 0.37	2.12 ± 0.25	2.35 ± 0.14	1.94 ± 0.25	2.29 ± 0.33