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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 s 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 (±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.

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 produces 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.

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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 s 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.

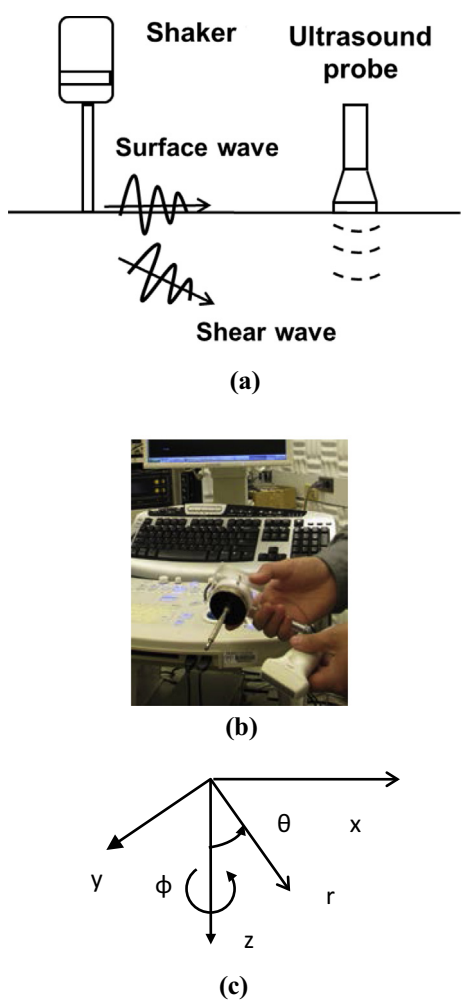

Fig. 1. (a) In surface wave elastography (SWE), a electromagnetic shaker is used to generate a 0.1 s harmonic vibration at 100 Hz on the surface of tissue. The resulting surface wave propagation on the tissue and shear wave propagation inside the tissue are detected using an ultrasound probe at 5 MHz; (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.

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]

$$(\lambda+2\mu)\nabla\nabla\cdot\mathbf{u}+\mu\nabla^2\mathbf{u}-\rho\frac{\partial^2\mathbf{u}}{\partial t^2}=0, \tag{1}$$

where $\mathbf{u}$ 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 $r \leqslant a$. The displacement fields are derived in the r and z directions at any location in and on the surface of the medium as

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

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 $\eta=k_2/k_1=\sqrt{\{2(1-\sigma)/(1-2\sigma)\}}$, $k_1=\omega\sqrt{\rho/(\lambda+2\mu)}$, $k_2=\omega\sqrt{\rho/\mu}$, $\xi=k_1 a$, 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 Eq. (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

$$c_s=\frac{1}{1.05}\sqrt{\frac{\mu}{\rho}}. \tag{3}$$