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ELASTICITY ESTIMATES FROM IMAGES OF CRAWLING WAVES GENERATED BY MINIATURE SURFACE SOURCES

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Abstract

We present a surface-based approach to generate shear wave interference patterns, called crawling waves (CrW), within a medium and derive local estimates of biomechanical properties of tissue. In previous experiments, elongated bars operating as vibration sources were used to generate CrW propagation in samples. In the current study, however, a pair of miniature circular vibration sources is applied to the overlying skin to generate the CrW within the medium. The shape and position of the miniature sources make this configuration more applicable for *in vivo* implementation. A modified ultrasound imaging system is used to display the CrW propagation. A shear speed mapping algorithm is developed using a detailed analysis of the CrW. The proposed setup is applied to several biomaterials including a homogeneous phantom, an inhomogeneous phantom, and *ex vivo* human liver. The data are analyzed using the mapping algorithm to reveal the biomechanical properties of the biomaterials.

Keywords

Crawling waves; Sonoelastography; Elasticity imaging; Local shear speed estimation; Shear waves

INTRODUCTION

Biomechanical properties of biological soft tissues can serve as biomarkers of health or pathology (Hoyt et al. 2008b; Parker et al. 2011). Some diseases such as breast and prostate cancer exhibit localized differences in stiffness where cancerous tumors are formed in softer healthy tissue. Other diseases such as liver fibrosis and liver cirrhosis exhibit increased overall stiffness of the organ itself. Manual palpation has been commonly used by physicians to detect the presence of hard lesions within soft tissue. However, the limited

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accessibility of palpation and low elastic contrast between the lesion and the background tissue typically encountered in clinical situations limit this qualitative approach.

There have been research studies conducted for over two decades in the area of elastographic imaging methods. Elastography, which refers to the branch of noninvasive elasticity imaging, can be classified in terms of the imaging modality used to scan soft tissue and display the resulting elasticity maps. The commonly used modalities are ultrasound and magnetic resonance imaging (MRI) (Parker et al. 2011). In ultrasound elastography, the methods can be divided with respect to the applied external excitation: vibration sonoelastography (Krouskop et al. 1987; Lerner and Parker 1987; Lerner et al. 1988; Yamakoshi et al. 1990), compression Elastography (Ophir et al. 1991), vibrational transient Elastography (Catheline et al. 1999; Sandrin et al. 2003), and acoustic radiation force techniques (Fatemi and Greenleaf 1998; Sarvazyan et al. 1998; Nightingale et al. 1999; Nightingale et al. 2001; Konofagou and Hynynen 2003; Chen et al. 2007; McAleavey and Menon 2007; Hah et al. 2012). Extensive reviews of the different techniques are available (Sarvazyan et al. 1995; Gao et al. 1996; Ophir et al. 1999; Greenleaf et al. 2003; Parker et al. 2005; Garra 2007; Parker et al. 2011). Sonoelastography is an ultrasound based imaging technique which utilizes continuous harmonic vibrations and power Doppler scanning. It is a readily available, noninvasive, real-time imaging tool used in all experiments presented in this paper.

Wu et al. (2004) used sonoelastography to image a slowly propagating shear wave interference, crawling waves (CrW), generated by a pair of mechanical vibration sources. The sources, in the shape of elongated bars, were placed in close contact at opposite sides of the medium and oscillated along a direction parallel to the sides as illustrated in Fig. 1(a) (the pair of sources labeled I), thereby generating shear wave from each side of the medium. The ultrasound probe is placed at the same cross-section with the vibration sources and displays the shear wave interference. By applying vibrations with a slight offset between the vibration frequencies, the displayed interference would appear on the screen as slowly propagating stripes, and thus the term “crawling waves”. The data are further analyzed to reveal the elastic properties of the medium. This method provided reliable *ex vivo* results (Castaneda et al. 2009; Barry et al. 2012), but is not well suited for *in vivo* use, mainly because of the location and size of the sources. Hoyt et al. (2008c) further suggested placing the bars and the ultrasound probe on the top surface in order to generate the CrW as shown in Fig. 1(a) (sources labeled II). Here, the bars vibrate in the direction normal to the surface. This setup was employed on *in vivo* skeletal muscle and a previously developed algorithm (Hoyt et al. 2008a) was used to map the viscoelasticity of the muscle tissue. However, the algorithm used is valid only for an ROI in which the CrW propagation pattern can be approximated as a planar propagation. Moreover, the size of the bars renders this setup unsuitable for *in vivo* use such as liver scanning.

In this paper, a revised source configuration is introduced where a pair of circular indenters with a small surface area is used as the vibration sources instead of the elongated bars as illustrated in Fig. 1(b). The location and dimensions of the sources make this elasticity mapping method a more applicable approach for *in vivo* scanning. The theory behind this configuration is described including detailed explanations of: the sonoelastography imaging mode that is used to detect and display the wave propagation within the medium; the vibration source configuration used to generate shear wave propagation using both one and two vibration sources; and the algorithm for mapping the elastic properties within the medium. Detailed analysis of the displacement field is presented and consequently, the estimation algorithm is developed which is valid for the entire scanned ROI. Experiments and results for a homogeneous phantom, an inhomogeneous phantom, and an *ex vivo* human liver are described to examine the performance of the suggested method.

THEORY

Elasticity theory

The stiffness of soft tissues can be quantitatively expressed in terms of shear modulus (Pascal units). In a linear, isotropic, and elastic medium, the relationship between the shear modulus G , shear speed c_s , mass density ρ , angular frequency of the shear wave ω , and shear wavenumber k_s , is

$$c_s = \sqrt{\frac{G}{\rho}} = \frac{\omega}{k_s}. \quad (1)$$

It is acceptable to represent stiffness using only the shear speed value because the mass density is nearly 1 gram/cm³ in most soft tissues. A quantitative elasticity map is advantageous over the qualitative result, especially in the cases of liver fibrosis and liver cirrhosis, where the average stiffness of the liver in an absolute scale is desired. All the results in this paper are presented in the form of shear speed maps.

Sonoelastography imaging

Sonoelastography is an ultrasound based imaging technique in which external continuous harmonic vibrations excite the tissue to generate shear waves within the desired organ. The tissue scatterers oscillate in a sinusoidal fashion around their equilibrium position as the harmonic wave propagates. Regions of high stiffness will respond with lower displacement, as compared with relatively soft tissues (Lerner et al. 1990). Doppler scanning is utilized to detect displacements of vibrating scatterers. The power spectrum spread of the backscattered echo from a vibrating scatterer is linearly proportional to the displacement amplitude of the vibration (Huang et al. 1990), and is given by

$$\sigma = \frac{\sqrt{2}\omega_0\omega}{c} \cdot u, \quad (2)$$

where σ is the spread (standard deviation) of the Doppler power spectrum, u is the displacement amplitude, ω_0 is the angular frequency of the insonifying ultrasound wave, ω is the angular frequency of the vibration source, and c is the compressional speed of sound. Sonoelastography employs a real-time computation of spectral variance to represent the intensity of the local particle displacement. The variance of the power spectrum, derived by Kasai et al. (1985), is computed with

$$\sigma^2 = \frac{2}{T_{PRF}} \left(1 - \frac{|R(T_{PRF})|}{R(0)} \right), \quad (3)$$

where T_{PRF} is the time interval between the successive ultrasound Doppler pulses and R is the autocorrelation of the backscattered signals of a given Doppler packet. Local variance, corresponding to the local stiffness, is estimated using eqn (3) and displayed on the ultrasound screen in real-time (Taylor et al. 2000). This variance image is called a *sonoelastogram*.

Vibration field from a circular load vibrating normal to the surface

To analyze the proposed double-source vibration setup shown in Fig. 1(b), a more fundamental configuration is examined first where only a single source is employed. The geometry and coordinate system of the setup are illustrated in Fig. 2(a), and initially we consider only the left source (source 1) (the subscript 1 corresponding to source 1 is eliminated here for convenience, e.g. $\omega_1 = \omega$). This source is positioned on top of the surface

and vibrates with angular frequency ω in the direction normal to the surface (z axis), where A_s is the vibration amplitude, d is the lateral location of the source, r is the distance from the source and θ is the angle from the z axis. A rectilinear ultrasound transducer is placed on the same surface and scans the medium. Since the conventional Doppler system processes the axial component of the particle motions, only the z component of the displacement is of interest. According to Miller and Pursey (1954), the closed-form solution of the displacement field in the axial direction, valid in the far field, is

$$u_z = A_s \frac{ia^2 \cos \theta}{2c_{44} k_c r} \cdot \left[\frac{2\mu^3 \sin^2 \theta \sqrt{\mu^2 \sin^2 \theta - 1}}{F_0(\mu \sin \theta)} \cdot \exp(-ik_s r) + \frac{i \cos \theta (\mu^2 - 2 \sin^2 \theta)}{F_0(\sin \theta)} \cdot \exp(-ik_c r) \right] \exp(i\omega t), \quad (4)$$

where $i = \sqrt{-1}$, a is the radius of the circular load, c_{44} is the bulk modulus, k_c is the wavenumber of the compressional wave, k_s is the wavenumber of the shear wave, ω is the angular frequency of the shear wave, μ is the k_s / k_c ratio, and $F_0(\zeta)$ is

$$F_0(\zeta) = (2\zeta^2 - \mu^2)^2 - 4\zeta^2 \sqrt{(\zeta^2 - 1)} \sqrt{(\zeta^2 - \mu^2)}. \quad (5)$$

$F_0(\zeta)$ is a complex function and thus can be expressed in the phasor form as

$$F_0(\zeta) = |F_0(\zeta)| \exp(i\gamma), \quad (6)$$

where

$$\gamma = \tan^{-1} \left\{ \frac{\text{Im}[F_0(\zeta)]}{\text{Re}[F_0(\zeta)]} \right\}. \quad (7)$$

The axial displacement field provided in eqn (4) consists of shear and compressional propagation components. The compressional propagation, however, can be neglected because the wavelength of the compressional wave is larger than human organs. This implies that the phase of the compressional wave is approximately constant over the scanned ROI. We add an exponential attenuation term α_s for shear waves to account for loss in biological tissues, and express the function F_0 in its phasor form, then the displacement field can be expressed as

$$u_z = \frac{A}{r} \exp(-\alpha_s r) \cdot \exp \left[i \left(\omega t - k_s r - \gamma + \frac{\pi}{2} \right) \right], \quad (8)$$

where A is a real number and given by

$$A = A_s \frac{a^2 \cos \theta}{2c_{44} k_c} \cdot \frac{2\mu^3 \sin^2 \theta \sqrt{\mu^2 \sin^2 \theta - 1}}{|F_0(\mu \sin \theta)|}. \quad (9)$$

Eqns (8) and (9) describe a spherical wave with axial symmetry (Blackstock 2000), characterized by a geometric loss, where the field is inversely proportional to the distance r . A simulation using eqn (8) is performed for an ROI of 50 by 40 mm² with the following parameters: $a = 0.5$ cm, $\rho = 1$ gram/cm³, $c_c = 1540$ m/s, $c_s = 3$ m/s, $f = 300$ Hz, $k_c = 2\pi f / c_c$, $k_s = 2\pi f / c_s$, $c_{44} = \rho \cdot c_c^2$, and $\alpha_s = 0.25$ Np/cm. The farfield distance in this case is $a^2 / \lambda_s = a^2 f / c_s = 0.25$ cm. The normalized magnitude and the phase from the simulation are illustrated in Fig. 2(b) and Fig. 2(c), respectively.

In the sonoelastography mode, where the square of the vibration amplitude is displayed, the displacement generated by a single vibration source is

$$|u_z|^2 = u_z \cdot u_z^* = \frac{A^2}{r^2} \exp(-2\alpha_s r). \quad (10)$$

As can be seen in eqn (10), the phase information that contains the shear speed is not displayed in sonoelastography when a single source configuration is employed. In order to overcome this problem and increase signal-to-noise ratio, the CrW technique is employed, where two sources are used to generate a slowly propagating interference of two shear waves. CrW excitation using a pair of normally vibrating sources is presented in the following section.

Shear wave interference patterns (CrW)

In the setup for generating CrW propagation with a pair of circular vibration sources, the ultrasound transducer is positioned between the vibration sources and the center of the probe is defined as coordinate (0,0) as illustrated in Fig. 2(a). Two circular loads, vibrating normally to the surface, generate shear wave interference within the medium, where the subscripts 1 and 2 correspond to the left and right sources, respectively. The axial displacement of the interfering shear waves, u_T , is

$$\begin{aligned} u_T = u_{z1} + u_{z2} = & \frac{A_1}{r_1} \exp(-\alpha_s r_1) \cdot \exp[i(\omega_1 t \\ & - k_{s1} r_1 - \gamma_1 + \frac{\pi}{2} + \phi_1 + \frac{A_2}{r_2} \exp(-\alpha_s r_2) \cdot \exp[i(\omega_2 t \\ & - k_{s2} r_2 - \gamma_2 + \frac{\pi}{2} + \phi_2)]] \end{aligned} \quad (11)$$

where ϕ_1 and ϕ_2 are arbitrary phase terms. The squared amplitude of the displacement is

$$|u_T|^2 = u_T \cdot u_T^* = (u_{z1} + u_{z2}) \cdot (u_{z1} + u_{z2})^*. \quad (12)$$

By substituting eqn (11) into eqn (12), the interference pattern as displayed in sonoelastogram is given by

$$\begin{aligned} |u_T|^2 = & \frac{A_1^2}{r_1^2} \exp(-2\alpha_s r_1) \\ & + \frac{A_2^2}{r_2^2} \exp(-2\alpha_s r_2) \\ & + 2 \frac{A_1 A_2}{r_1 r_2} \exp[-\alpha_s(r_1 \\ & + r_2) \cdot \cos[(\omega_1 \\ & - \omega_2)t \\ & + k_{s2} r_2 - k_{s1} r_1 + (\gamma_2 - \gamma_1) + (\phi_1 - \phi_2)]] \end{aligned} \quad (13)$$

It is noted that the phase term containing the shear speed is preserved. The following notations are used in order to further simplify eqn (13):

$$\omega_1 - \omega_2 = \Delta\omega \quad \gamma_2 - \gamma_1 = \beta \quad \phi_1 - \phi_2 = \phi_0, \quad (14)$$

where $\Delta\omega$ is the difference between the angular vibration frequencies, ϕ_0 is a constant phase term, and β is a function of the coordinate system and the location of the vibration sources. The vibration sources are adjusted such that the difference between the frequencies is very small, *i.e.* $\Delta\omega \ll \omega$. Accordingly, since $\Delta k_s = \Delta\omega / c_s \ll k_s$, then the shear wavenumbers of the propagating shear waves can be denoted as $k_{s1} \approx k_{s2} \approx k_s$. Applying the notations of eqn (14) and the condition $\Delta\omega \ll \omega$ to eqn (13), the interference is

$$|u_T|^2 = \frac{A_1^2}{r_1^2} \exp(-2\alpha_s r_1) + \frac{A_2^2}{r_2^2} \exp(-2\alpha_s r_2) + 2 \frac{A_1 A_2}{r_1 r_2} \exp[-\alpha_s(r_1 + r_2)] \cdot \cos[\Delta\omega t + k_s(r_2 - r_1) + \beta(r_1, r_2) + \phi_0]. \quad (15)$$

The shear speed parameter k_s appears in the argument of the cosine term of eqn (15), and therefore, can be estimated using an appropriate algorithm. A simulation using eqn (15) is performed for an ROI of 50 by 40 mm² with the following parameters: $a = 0.5$ cm, $\rho = 1$ gram/cm³, $c_c = 1540$ m/s, $c_s = 3$ m/s, $f_1 = 300$ Hz, $f_2 = 300.4$ Hz, $k_c = 2\pi f / c_c$, $k_s = 2\pi f / c_s$, $c_{44} = \rho \cdot c_c^2$, $\alpha_s = 0.2$ Np/cm, $d_1 = 2$ cm, and $d_2 = 2$ cm. The normalized squared amplitude from the simulation is illustrated in Fig. 2(d). As can be seen, the CrW patterns are displayed in sonoelastography as curved stripes. The distance between the stripes and their shape correspond to mechanical properties of the medium at frequency 300 Hz. The argument of the cosine term in eqn (15) is further analyzed in the following section to develop the estimation algorithm that maps the local shear speed values in the ROI.

Local shear speed estimation algorithm for elasticity mapping

The goal of the estimation algorithm is to estimate the local shear speed values within the scanned ROI. Substituting eqn (1) into eqn (15), the shear wave interference can be expressed by

$$|u_T|^2 = B + 2 \frac{A_1 A_2}{r_1 r_2} \exp[-\alpha_s(r_1 + r_2)] \cdot \cos[\Delta\omega t + \varphi(c_s, r)] \quad \varphi(c_s, r) = \frac{\omega}{c_s}(r_2 - r_1) + \beta(r_1, r_2) + \phi_0, \quad (16)$$

where $\varphi(c_s, r)$ denotes the spatial phase term of the interference and B is the baseline term. The rectilinear ultrasound probe, used in this study, provides the data in Cartesian coordinates. Applying appropriate coordinate transformations (polar to Cartesian) to the spatial variables r_1 and r_2 , the spatial phase $\varphi(c_s, r)$ is

$$\varphi(c_s, x, z) = \frac{\omega}{c_s} \left(\sqrt{(x - d_2)^2 + z^2} - \sqrt{(x - d_1)^2 + z^2} \right) + \beta(x, z) + \phi_0. \quad (17)$$

By computing the gradient of $\varphi(c_s, x, z)$ in x direction, the local shear speed estimate $\hat{c}_s(x, z)$ can be extracted with

$$\hat{c}_s(x, z) = \omega \left(\frac{\partial \varphi}{\partial x} - \frac{\partial \beta}{\partial x} \right)^{-1} \left(\frac{x - d_2}{\sqrt{(x - d_2)^2 + z^2}} - \frac{x - d_1}{\sqrt{(x - d_1)^2 + z^2}} \right), \quad (18)$$

where all parameters on the right hand side are known from the geometry and theory or measured from the data.

The data provided by the ultrasound scanner are in the form of I/Q data samples. A three-dimensional (3-D) matrix of sequential frames (axial $\times$ lateral $\times$ time) is generated from the

I/Q data and illustrated in Fig. 3(a). Since the frame rate used in sonoelastography is comparable to the conventional Doppler ultrasound (~ 10 Hz), the time axis of the frame sequence is known as slow-time axis. The spatial phase is computed from the slow-time signal for every coordinate (x, z) , thereby generating a phase map for the entire scanned ROI. An example of the slow-time signal for a location is shown in Fig. 3(b). In order to compute the gradient of $\phi(c_s, x, z)$, the phase map is unwrapped using one-dimensional algorithm of MATLAB computational software (The Mathworks Inc., Natick, MA, USA) applied in both lateral and axial dimensions. The local shear speed is then calculated for each coordinate using eqn (18) to generate the shear speed map $\hat{c}_s(x, z)$. The two-dimensional (2-D) phase map and its unwrapped version are shown in Fig. 3(c) and Fig. 3(d), respectively. As a practical matter some spatial averaging is used to minimize the effects of noise. A 2-D median filter of 2 by 2 mm² is applied for each CrW frame, and a 4 by 4 mm² median filter is applied to the final shear speed map. This algorithm is applied to the experiments presented in the following section.

EXPERIMENTS

Five experimental data sets have been acquired and analyzed in order to validate the performance of the proposed experimental setup illustrated in Fig. 1(b), and the estimation algorithm. Four experiments were performed on tissue-mimicking phantoms and one experiment was performed on *ex vivo* human liver under informed consent and IRB approved protocol.

Two phantoms were prepared: a homogeneous 17% gelatin phantom, and one inclusion phantom with 17% gelatin within the inclusion and 9.3% gelatin within the background. The percentages are with respect to the weights of the gelatin and water. The process for homogeneous phantom preparation is analogous to the phantom preparation procedure used in *ex vivo* prostate and liver experiments (Castaneda et al. 2009; Barry et al. 2012). The homogeneous phantom consists of 2 liters degassed water, 409 g porcine gelatin (300 Bloom Pork Gelatin; Gelatin Innovations Inc., Schiller Park, IL, USA), 19.66 g salt (Sodium Chloride, BDH, PA, USA), and 3.27 g agar (Difco Agar Technical Solidifying Agent, BD, Sparks, MD, USA). The components were mixed together and heated to a temperature of 55°C. The molten mixture was then poured into a cube-shaped mold for hardening. The inclusion phantom preparation consists of two main steps: inclusion fabrication and embedding the inclusion into the background phantom. In the process of inclusion fabrication, once the mixture had been heated to 55°C, the solution was poured into a cylindrical mold (approximately 10 mm in diameter) and placed in a refrigerator (approximately 5°C) for hardening. Once solidified, the inclusion was placed into a cube-shaped box. The water-gelatin mixture of the background material (2400 ml degassed water, 246 g gelatin, 3.6 g agar, 21.6 g salt) was heated to 55°C and then cooled to 32°C before poured into the box. Since the inclusion and the background are 17% and 9.3% gelatin, respectively, the inclusion is stiffer than the background.

The prepared phantoms were scanned in three different cross-sections. A single cross-section in the 17% gelatin phantom was scanned using the setup shown in Fig. 1(b). The cylinder phantom was scanned in two cross-sections as outlined in Fig. 4. A homogeneous section (9.3% gelatin) in the cylinder phantom, dotted outline in Fig. 4, was scanned using setups presented in Fig. 1(a) (sources labeled I) and Fig. 1(b) for comparison. The vibration frequencies of the sources were 250 Hz and 250.4 Hz. Finally, a cross-section that includes the cylinder, dashed outline in Fig. 4, was scanned using the proposed setup with vibration frequencies of 220 Hz and 220.4 Hz. In all phantom experiments the temperature of the phantoms was approximately 19°C (room temperature). In addition to the phantom experiments, a whole human liver was scanned using the same setup (Fig. 1(b)). The liver

was provided for the experiment after hepatectomy, under approved consent. The sources, vibrating at 90 Hz and 90.45 Hz, excited the liver through a gelatin layer that operated as a standoff pad. Note that different vibration frequencies were used in liver and phantom experiments. The choice of frequency range depends on the shear wave attenuation, α_s , and the sample size of the biomaterial. The farfield distance in all experiments (according to a^2 / λ_s) starts at a distance that is less than 1 cm.

In each experiment, a dual channel signal generator (AFG3022B, Tektronix, Beaverton, Oregon, USA) produces two sinusoidal signals with slight difference between the frequencies, $\Delta\omega$. In order to generate a slowly propagating shear wave interference pattern (CrW), the frequency difference is set in the range of 0.3–0.5 Hz. Each output signal from the function generator passes through a digital power amplifier (Model LP-2020A+, Lepai, Bukang, China) and then supplied to a miniature vibration source (Linear Actuator, model NCM02-05-005-4JB, H2W, Santa Clara, CA, USA). The vibration sources are placed on top of the biomaterial and vibrate in a direction that is normal to the surface of the medium (z axis in accordance with Fig. 2(a)). A linear array ultrasound transducer (M12L, GE Healthcare, Milwaukee, WI, USA) is positioned between the vibration sources and scans the medium. The sonoelastography imaging mode displays the CrW on the ultrasound display (Logiq 9, GE Healthcare, Milwaukee, WI, USA). A DICOM (Digital Imaging and Communications in Medicine) file is saved for each experiment and transferred to a personal computer (PC) for processing.

The estimated average shear speed of the 9.3% and 17% homogeneous cross-sections is 3.66 ± 0.3 m/s and 6.63 ± 0.5 m/s, respectively, calculated over an ROI of 35 by 30 mm², centered at 30 mm depth. This verifies that the 17% phantom is stiffer than the 9.3% phantom. A single frame of each CrW movie and the corresponding shear speed maps are illustrated in Fig. 5. The bright and dark stripes (green and black color schemes, respectively, in color Doppler) indicate regions of high and low displacement amplitudes, respectively. The distance between the stripes is proportional to the shear speed. Since both phantoms were scanned with the same vibration frequencies, it is evident from Fig. 5, that the 17% phantom has higher shear speed than the 9.3% phantom (less stripes indicate higher shear speed). The shear speed value obtained for the 9.3% homogeneous section using the elongated bars is 3.5 ± 0.1 m/s over an ROI of 35 by 30 mm². This value correlates with the estimated shear speed (3.66 m/s) for the same cross-section obtained with the circular sources setup. In addition, mechanical measurement using a stress relaxation test, as described by Zhang et al. (2007), was previously applied to a 9.3% phantom and showed shear speed value of 3.6 m/s, which also correlates with the estimated shear speed in this study.

For the inclusion experiment, the B-scan, a CrW movie frame, the unwrapped phase map, and the shear speed map are illustrated in Fig. 6. The average shear speed within the inclusion and the background is 5.34 ± 0.35 m/s and 3.7 ± 0.4 m/s, respectively. The shear speed value of the background correlates with the estimated shear speed from the experiment performed on the homogeneous cross-section of the same phantom. However, the estimated shear speed within the 17% inclusion and 17% homogeneous phantom are different. This may stem from the general rule that in order to detect a structure in a medium, the wavelength should be larger than the dimension of the target structure. This condition is not satisfied since the inclusion diameter is approximately 10 mm and the wavelength is about 30 mm (assuming that $c_s = 6.5$ m/s and given that the vibration frequency is 220 Hz). Although the estimated shear speed within the inclusion may be underestimated, the inclusion can be revealed on the shear speed map as illustrated in Fig. 6(d).

The liver was recovered from a previously healthy patient who died of a ruptured cerebral aneurysm. All liver biochemical tests were normal and liver biopsy showed less than 5% steatosis (fat concentration). Consent for research was obtained from the patient's family prior to recovery. The liver was not used for transplant because of logistical issues resulting in a prolonged cold ischemia time. After hepatectomy, the liver was kept in a plastic bag filled with a preservative solution and refrigerated, for approximately 48 hours before scanning. Liver images including the B-scan, a CrW movie frame, the unwrapped phase map, and the computed shear speed map are shown in Fig. 7. The estimated average shear speed within the liver is 3 ± 0.1 m/s over an ROI of 35 by 30 mm², centered at 30 mm depth. This value correlates with the estimated shear speed obtained by (Barry et al. 2012) in an *ex vivo* human liver. *In vivo* studies (Sandrin et al. 2003; Muller et al. 2009), however, show lower shear speed values than the *ex vivo* liver studies. This difference is due to the different temperatures of the livers during the scanning. Lower temperature results in an increased stiffness. Therefore, shear speed estimations obtained in *ex vivo* experiments (at room temperature of approximately 19°C) are greater than the estimations obtained *in vivo* (at body temperature of approximately 37°C).

CONCLUSIONS

The configuration of two miniature vibration sources together with the ultrasound transducer can serve as an *in vivo* elasticity imaging technique for larger organs such as the liver. A detailed analysis of the displacement field generated by the vibration sources and the developed algorithm allows mapping of the elastic properties within the scanned biomaterial. This technique demonstrated its capabilities in identifying an inclusion on the order of 10 mm in diameter. It was also shown that the proposed double-source setup can produce CrW within *ex vivo* human liver, and elasticity was mapped using the developed algorithm. Some limitations of the current study include the need for testing in more complex models including the effects of additional layers and multiple inhomogeneities. However, the results demonstrated that this approach can be further developed to become a hand-held ultrasound-based device for elasticity mapping within soft tissues.

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Fig. 1. Vibration setups used in previous (a) and current (b) studies. Previously, elongated bars were used to generate the crawling waves (CrW) by vibrating the sides (sources labeled I) or the top (sources labeled II) of the medium. In this study, the sources with a small circular surface area vibrate on top of the medium surface to generate the CrW propagation as shown in (b).

Fig. 2.

Schematic and simulations of the vibration setups. Two mechanical vibration sources shown in (a) generate shear wave interference within the medium via continuous harmonic vibrations in the direction normal to the surface of the medium. The sources and the ultrasound transducer define the imaging plane. The parameters ω , r , θ , and d , represent respectively the angular frequency of the vibrations, the distance from the source, the angle from the normal to the surface, and the lateral location of the source. The normalized magnitude in (b) and the phase in (c) of the axial displacement field generated by a single source (source 1) are illustrated using simulation and eqn (4). A single frame of the simulated crawling waves movie (both sources vibrating) is shown in (d). This image is the

normalized square amplitude of the particle displacement. The shape and location of the stripes correspond to the mechanical properties of the medium.

Fig. 3.

3-D matrix of the crawling waves (CrW) is illustrated in (a). The lateral, axial and slow-time axes correspond to the depth, width and time of the CrW movie, respectively. A set of data samples through the slow-time axis for a specific coordinate is called slow-time signal (dashed line). A slow-time signal for a location is shown in (b). The number of samples of a slow-time signal equals the number of frames of the movie sequence. In order to recover the phase ϕ , the slow-time signal should contain at least one sinusoidal cycle. The phase map in (c) and the unwrapped phase map in (d) were computed from the simulated CrW.

Fig. 4.
The cylinder phantom was scanned in two different cross-sections: a homogeneous section (dotted line) and a section that includes both the cylinder and the background (dashed line).

Fig. 5.

Homogeneous phantoms experiments. Crawling waves movie frames of the 9.3% phantom in (a) and 17% phantom in (b) are illustrated. The shear speed maps of the 9.3% and 17% phantoms are given in (c) and (d), respectively. In both experiments the vibration frequency and the vibration frequency offset are 250 Hz and 0.4 Hz, respectively.

Fig. 6.

Inclusion phantom experiment. The inclusion is 17% gelatin and background is 9.3% gelatin. The vibration frequency and the vibration frequency offset are 220 Hz and 0.4 Hz, respectively. (a) is the B-scan of the phantom, (b) is a single frame of the crawling waves movie, (c) is the unwrapped phase map, and (d) is the computed shear speed map.

Fig. 7.
Ex vivo human liver. The vibration frequency and vibration frequency offset are 90 Hz and 0.45 Hz, respectively. (a) is the B-scan of the liver, (b) is a single frame of the crawling waves movie, (c) is the unwrapped phase map, and (d) is the shear speed map.