Two-dimensional Shear Elasticity Imaging Using External Mechanical Vibration

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Abstract—Compared to ultrasound radiation force, mechanical vibration is capable of delivering larger motion into deep tissues. However, the propagation direction of the shear waves generated by such method can be complicated and unpredictable, which make it difficult to measure shear wave speed ($c_s$). In this study, we propose using multiple vibration sources to produce a multi-directional shear wave field, and calculating a two-dimensional (2D) $c_s$ map based on cross-correlation after applying directional filtering to the field. The proposed method is tested using homogeneous and inclusion phantoms as well as in vivo liver in a healthy subject. Results show that the method has good penetration, boundary delineation, and consistent elasticity measurement.

Keywords—shear wave speed; mechanical vibration; directional filter

I. INTRODUCTION

Shear wave propagation speed ($c_s$) can be used to estimate tissue elasticity which is related to the state of tissue health [1]. Shear waves in tissues can be generated by ultrasound radiation force, mechanical vibration or intrinsic physiological movements such as breathing or cardiac motion.

Through an ultrasound push beam, ultrasound radiation force can produce shear waves remotely at specified locations within tissues, and the $c_s$ can be measured locally near the push axis [1-3]. However, shear waves generated by ultrasound radiation force are typically very weak (micrometers), making shear wave detection and $c_s$ measurement susceptible to noise. Therefore, such methods may become problematic for applications requiring deeper penetration.

Shear waves produced by external mechanical vibration sources can be of higher amplitude for more reliable motion detection in deeper tissues. However, shear waves produced with this manner often have complicated patterns with unknown propagation directions, making it difficult for traditional time-of-flight methods to measure $c_s$. Some actuators have been designed to better control the shear wave propagation direction [4], but their benefit is limited for deeper tissues. Physiological movements can also introduce shear waves without external sources. However, the waves can be even more complicated and unpredictable. Measurement of $c_s$ is biased higher than the true speed when $c_s$ is measured at an angle with its actual propagation direction, either within or out of the image plane.

A time reversal approach has been proposed for elasticity imaging based on correlations of a diffuse wave field [5]. Such a method can be applied to the shear waves generated both from external vibration [6] and from physiological movements [7]. However, a diffuse wave field is required in order to retrieve the Green’s function correctly [8], which may be difficult to achieve in some cases.

In this study, multiple vibration sources are used to produce a multi-directional shear wave field which is detected in 2D by an ultrafast imaging technique [9] called “flash imaging” over a period of several hundred milliseconds. Directional filters are used to separate waves moving in different directions. Out-of-plane and compressional waves are suppressed by imposing an upper limit of wave propagation speed. Then for each direction, a 2D $c_s$ map is calculated from the directionally filtered signals based on cross-correlation. The final $c_s$ map is constructed by compounding the $c_s$ maps calculated from all the directions. The proposed method does not require a diffuse wave field. Both phantom and in vivo results demonstrate the proposed method has good penetration, boundary delineation, and robust $c_s$ measurement.

II. METHOD

A. Vibration Sources

Multiple (4-10) miniature electromagnetic shakers are used to fill the region of interest (ROI) with multi-directional shear waves in several hundred milliseconds. They are placed randomly over the surface of the phantom or body where sufficient shear waves can be generated and delivered to the ROI. Each shaker is driven independently by random impulses to avoid repeated wave patterns or standing waves in the ROI that could compromise the performance of $c_s$ estimation using cross-correlation.

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B. Directional Filter

The multi-directional shear wave motion is detected in 2D (lateral x and axial z) over a short period of time using flash imaging. To isolate the shear waves propagating in different directions, the spatiotemporal \((x, z, t)\) motion signals are first converted to the spatiotemporal-frequency \((k_x, k_z, f_t)\) domain using a 3D Fourier transform. Then the directional filters are applied by multiplying the directional filter response with the signals in the k-f domain [10]. Fig. 1 shows the directional filter response in four directions. For each direction, shear waves propagating within an angle range near the primary direction can pass through whereas shear waves moving in directions further away from the primary direction are suppressed.

Low spatial frequency components corresponding to high speed waves (compressional and out-of-plane waves) can be suppressed in the k-f domain. Given a high speed limit \(c_h\) at a temporal frequency \(f_t\), the spatial frequency limit \(k_t = f_t c_h\) can be imposed within the directional filter. Spatial frequency components lower than \(k_t\) are removed as indicated by the dark round regions near the center of directional filter response in Fig. 1. Similarly, a low speed limit can be applied by removing spatial frequency components higher than \(k_s = f_t c_s\). A band-pass Butterworth filter is designed to apply smooth transitions for both \(k_t\) and \(k_s\).

After the directional filtering, the resulting shear wave signals in different directions are converted back to the spatiotemporal domain using the inverse 3D Fourier transform.

Figure 1. Directional filters at 0°, 45°, 90°, and 135° orientations

C. 2D \(c_s\) Estimation

Data filtered by each directional filter mainly contain waves propagating in that direction, which can be at an oblique angle to the lateral and axial coordinates \((x, z)\) of the ultrasound detection data grid. Then cross-correlation is applied in both axial and lateral directions to calculate the time delays between the temporal signals from two pixels with a certain distance apart. To improve the estimation accuracy, the temporal signals are upsampled 10 times and parabolic interpolation is applied to locate the sub-pixel correlation peaks [11]. If at one location the measured time delays per unit distance in the axial and lateral directions are \(\Delta t_x\) and \(\Delta t_z\), the combined time delay becomes \(\Delta t = \sqrt{\Delta t_x^2 + \Delta t_z^2}\) from which the 2D \(c_s\) can be calculated as \(1/\Delta t\).

To further improve the robustness of the measurement, the value of \(c_s\) at one pixel is determined not only by its own estimate, but also by the estimates of the surrounding pixels. Both inverse distance weighting (IDW) [12] and correlation coefficient weighting (CCW) are applied to calculate the average speed as described in Equation (1).

\[
\bar{c}_s = \frac{\sum c_{s,r} c_s^2 / r}{\sum c_{s,r}^2 / r}
\]

where \(c_{s,r}\) is the speed measured at a location with a CC of \(cc_c\) and a distance of \(r\) away from the center pixel.

D. Shear Wave Speed Image Compounding

After the 2D \(c_s\) maps in all the directions are calculated, they can be compounded into a final image. The cross-correlation coefficient can be used to weight maps from different directions, and a threshold can be applied to reject the unreliable results.

Alternatively, to further minimize the measurement bias caused by the remaining out-of-plane waves which would give artificially high speed estimates, the minimum measured speed from different directions can be used as the final speed after the coefficient threshold is applied.

III. EXPERIMENTS

A. Inclusion Phantom

The performance of the proposed method was evaluated using an elasticity QA phantom (Model 049, CIRS Inc., Norfolk, VA). As shown in Fig. 2(a), 10 miniature electromagnetic shakers made of small DC motors with off-centered weight were attached to the surface and the side walls of the phantom to produce multi-direction shear waves. The shakers were driven by 20 pulses randomly distributed over 500 ms with pulse widths each of 4 ms. A Verasonics ultrasound system (Verasonics Inc., Redmond, WA) with a linear array transducer L7-4 (Philips Healthcare, Andover, MA) was used to detect shear wave motion for 600 ms at a frame rate of 2 kHz with a pixel resolution of 0.3 mm.

The shear wave (particle velocity) signals were calculated from the IQ signals using 2D autocorrelation [13]. Eight directional filters were designed to separate the shear waves into 8 directions from 0° to 315° with 45° step. Four of the directional filters are shown as examples in Fig. 1. The allowed speed range was set from 0 to 6 m/s. For filtered shear wave
signals, 2D $c_s$ at each pixel was calculated using cross-correlation between two pixels ±1.2 mm apart, in axial and lateral directions respectively, and averaged over a 4 mm × 4 mm area using IDW and CCW. For the final $c_s$ map, the speed maps from all directions were CCW averaged, after a 0.95 CC threshold was applied.

**B. Homogeneous Phantoms**

Two homogeneous elasticity phantoms (CIRS Inc., Norfolk, VA) were used to further validate the proposed method. Five piston-like electromagnetic shakers as shown in Fig. 2(b) capable of generating high amplitude shear waves were attached to the surface and the side walls of the phantoms. A C5-2v curved array transducer (Verasonics Inc., Redmond, WA) was used for motion detection. Once triggered by the Verasonics, the shakers each activate sequentially once within a 200 ms time window. The generated shear wave motion was recorded using flash imaging for 500 ms at a frame rate of 1 kHz.

The same processing methods were used except the speed range was 0.5 – 3 m/s and the distance for cross-correlation was ±2 mm. To construct the final $c_s$ map, the minimum speed from all directions was used.

**C. In Vivo Liver**

Approved by the Institutional Review Board (IRB), the in vivo liver stiffness of a healthy volunteer was studied using the proposed method. To deliver more motion into the liver, 8 piston-like shakers were used on subject’s back. The shear wave motion was detected for 300 ms at a frame rate of 1 kHz. The speed range for directional filters was chosen to be 0.5 – 3 m/s, and the distance for cross-correlation was ±2 mm. To construct the final $c_s$ map, the speed maps from all 8 directions were CCW averaged. In cirrhotic liver, $c_s$ higher than 3 m/s may be underestimated by these speed limits. However, this underestimation will not be relevant for clinical application to separate fibrosis stages F0-F1 vs. F2-F4, because literature suggests that the cut point should be below 2 m/s [14]. In other words, a highly fibrotic liver with $c_s$ of 4 m/s may have a biased reading close to 3 m/s, but it will still be classified correctly as F2-F4 because the biased reading should still be above the clinical cut point.

**IV. RESULTS**

For the inclusion phantom experiment, the 2D $c_s$ maps calculated from the filtered shear wave signals in 8 directions are shown in Fig. 3. The final compounded image is shown in Fig. 4. The calculated Young’s modulus was 66.8 ± 8.0 kPa and 31.9 ± 5.4 kPa for the inclusion and the background, respectively, as compared to nominal values of 80 ± 8 kPa and 25 ± 4 kPa, respectively.

**Fig. 4.** Compounded $c_s$ image using the maps in 8 directions from Fig. 4.

![Figure 4](image)

For the in vivo liver experiment, the compounded $c_s$ map is superimposed over the B-mode image as shown in Fig. 6. The

**Fig. 5.** Compounded $c_s$ maps for homogeneous phantoms. (a) Phantom #1; (b) Phantom #2.

![Figure 5](image)
average $c_s$ in the liver region was 1.62 ± 0.09 m/s, which corresponds to a shear modulus of approximately 2.62 kPa.

Figure 6. Compounded $c_s$ image of in vivo liver using the proposed method.

V. DISCUSSION AND CONCLUSIONS

In this study, multiple vibration sources are used to introduce high amplitude multi-direction shear waves into deeper tissues. Directional filters are designed to isolate shear waves propagating in different directions and remove the out-of-plane waves. Phantom and in vivo experiments have shown this method has the potential of investigating the 2D elasticity distribution of deep tissues.

The bandwidth of the shear waves produced by mechanical vibration is generally lower than those generated by ultrasound radiation force, resulting in lower spatial resolution of the $c_s$ map. If the signal to noise ratio of the motion signals is sufficient, band-pass filtering or spatial gradient potentially can be applied to improve the bandwidth of the shear waves.

In this study, 8 directional filters were used to cover all the directions. In practical applications, the directional filter design can be more flexible. The center directions and angle ranges can be selected according to the prior knowledge of the shear wave directions.

The out-of-plane and compressional waves were suppressed by setting a speed limit in the directional filter, which is possible for most clinical applications, as the physiological value range is usually known. However, the proposed method could be used to study anisotropic media by measuring $c_s$ in various directions. Another advantage of using multiple vibration sources, to simultaneously generate as many shear waves coming from many directions as possible.

The $c_s$ image compounding method in this study assumes the medium is isotropic, meaning the $c_s$ is the same in all the directions. However, the proposed method could be used to study anisotropic media by measuring $c_s$ in various directions.

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