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● *Original Contribution*

CARDIAC SHEAR WAVE ELASTOGRAPHY USING A CLINICAL ULTRASOUND SYSTEM

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Abstract—The propagation velocity of shear waves relates to tissue stiffness. We prove that a regular clinical cardiac ultrasound system can determine shear wave velocity with a conventional unmodified tissue Doppler imaging (TDI) application. The investigation was performed on five tissue phantoms with different stiffness using a research platform capable of inducing and tracking shear waves and a clinical cardiac system (Philips iE33, achieving frame rates of 400–700 Hz in TDI by tuning the normal system settings). We also tested the technique *in vivo* on a normal individual and on typical pathologies modifying the consistency of the left ventricular wall. The research platform scanner was used as reference. Shear wave velocities measured with TDI on the clinical cardiac system were very close to those measured by the research platform scanner. The mean difference between the clinical and research systems was 0.18 ± 0.22 m/s, and the limits of agreement, from −0.27 to +0.63 m/s. *In vivo*, the velocity of the wave induced by aortic valve closure in the interventricular septum increased in patients with expected increased wall stiffness. (E-mail: m.strachinaru@erasmusmc.nl)

Key Words: **Shear waves, Elastography, Stiffness, High-frame-rate tissue Doppler.**

INTRODUCTION

Many rapidly occurring mechanical phenomena have been described in the heart, such as electromechanical activation, blood flow noise and shear waves generated in the heart walls by closure of the valves (Cikes et al. 2014; Kanai 2005). The shear waves could potentially be used to estimate non-invasively the stiffness of the myocardium (Brekke et al. 2014; Couade et al. 2011), with huge possible implications for the diagnosis and treatment of multiple pathologies characterized by deterioration of the diastolic properties of the left ventricle. To track these rapidly occurring mechanical waves, high-frame-rate (>200 frames/s) imaging is mandatory. The frame rate in conventional ultrasound imaging is limited by the finite velocity of sound in human tissue (around 1540 m/s) and imaging depth (15 cm in an apical view), as well as the reconstruction of one image frame from many transmit–receive events. This leads to a conventional recording time of about 30 ms per frame, yielding a frame rate of around 30 Hz. Yet, modern clinical scanners achieve frame rates well over 50 Hz in gray-scale 2-D imaging by multiline acquisition, in which multiple lines are reconstructed simultaneously from a single transmit–receive event (Tong et al. 2012). Recent approaches to increase the frame rate even further include plane wave imaging or diverging waves, as well as high-level multiline transmit beamforming (Cikes et al. 2014) and selective field-of-view imaging (Brekke et al. 2014; Kanai 2005), which reach frame rates between 500 and 12,000 Hz, depending on the technology and depth of the tissue imaged. Moreover, recent advances in full-channel capture systems indicate that high frame rates can be achieved with relatively high contrast, signal-to-noise ratio (SNR) and sector size (Papadacci et al. 2014), albeit with resolution similar to that of

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conventional or multiline acquisition (MLA) beamforming techniques. However, to date, none of these technologies have been implemented in a clinical cardiac ultrasound system (Couade et al. 2011; Konofagou et al. 2011; Lee et al. 2012; Song et al. 2013). On the other hand, current clinical tissue Doppler imaging (TDI) applications use frame rates up to 200 Hz, by multiline acquisition and reduced resolution (Cikes et al. 2014; Sutherland et al. 1999). The use of a clinical scanner to track waves in the human heart has already been described (Pislaru et al. 2014) at frame rates of 350–450 Hz. In this study, we reached higher frame rates in TDI using a clinical cardiac ultrasound system by carefully tuning the imaging parameters, and hypothesized that this fast TDI modality could allow the detection and quantification of shear waves after valve closure. Similar to any other measurement method, both accuracy and precision of the measurement are important in clinical practice. However, such measurement *in vivo* is very difficult because there is no ground truth method for cardiac shear wave tracking, preventing estimation of accuracy, and because every heartbeat is different, preventing estimation of single-shot precision. The core aim of our study therefore was assessment of the accuracy and precision of the clinical TDI method to track shear waves. To obtain a reliable ground truth, we used a phantom setup where the propagation velocity of shear waves is constant and verified by using a high frame rate research scanner.

METHODS

Shear parameters

The *propagation velocity of shear waves* in an isotropic, homogeneous, elastic bulk material is related to the shear modulus μ and density ρ (Shiina et al. 2015) by

$$V_s = \sqrt{(\mu/\rho)} \quad (1)$$

The simplifying conditions (isotropic, homogeneous, elastic, bulk material) will not be met in cardiac tissue, and therefore, we refrained from converting the measured shear wave velocity to shear modulus or Young's modulus E ($E \approx 3\ \mu$ in soft biological tissue [Couade et al. 2011]) in the *in vivo* pilot data. However, we presume a monotonic relation between shear wave velocity and tissue stiffness. In the phantom experiment described below, the conditions are well met and this relation is used to convert Young's modulus into an expected shear wave propagation velocity.

Materials

The investigation was carried out on five different tissue ultrasound phantoms (CIRS, Norfolk, VA, USA). The physical properties of these phantoms are summarized in Table 1. Baseline calibration was performed on the multipurpose 40 GSE model, and further testing for different tissue stiffness was performed on the Model 039 phantom set.

We used two ultrasound scanners. The first was a research platform (R) inducing a shear wave through an acoustic radiation force push pulse and tracking it (as reference). It was a Verasonics Vantage system with extended burst option (Verasonics, Kirkland, WA, USA), equipped with a linear array L7-4 probe (Philips, Bothell, WA, USA). Recorded raw channel radiofrequency data and reconstructed ultrasound images were stored for off-line analysis. The second scanner was a normal clinical cardiac ultrasound system (C). This was a Philips iE33 system (Philips Medical, Best, Netherlands) with an S5-1 probe. A two-heartbeat TDI video was recorded and stored in DICOM format for off-line analysis. Philips QLab 9 post-processing software was used for the data analysis.

Setup

In preparation for the measurements, the probes were placed on the upper surface of the phantom, using clinical ultrasound gel as a contact medium. The probes were carefully aligned with their 2-D sectors in-line and thus oriented perpendicular to the direction of propagation of the shear wave. The leads of both an external cardiotachometer (CWE CT-1000) and the clinical scanner were attached to one of the researchers (M.S.) to

Table 1. Tissue phantom properties

	Model 40 GSE	Model 039			
Phatom properties	Calibration phantom	Phantom 1	Phantom 2	Phantom 3	Phantom 4
Expected shear wave velocity	2.80 m/s	1.01 m/s	1.57 m/s	2.43 m/s	3.56 m/s
Young's modulus E	25 kPa	2.7 kPa	11 kPa	20 kPa	48 kPa
Density			1030 kg/m^3		
Poisson ratio			0.5		
Attenuation			0.5 dB/cm/MHz		
Velocity of sound			1540 m/s		