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● *Original Contribution*

# IMPROVEMENT OF SHEAR WAVE MOTION DETECTION USING HARMONIC IMAGING IN HEALTHY HUMAN LIVER

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**Abstract—Quantification of liver elasticity is a major application of shear wave elasticity imaging (SWEI) to non-invasive assessment of liver fibrosis stages. SWEI measurements can be highly affected by ultrasound image quality. Ultrasound harmonic imaging has exhibited a significant improvement in ultrasound image quality as well as for SWEI measurements. This was previously illustrated in cardiac SWEI. The purpose of this study was to evaluate liver shear wave particle displacement detection and shear wave velocity (SWV) measurements with fundamental and filter-based harmonic ultrasound imaging. In a cohort of 17 patients with no history of liver disease, a 2.9-fold increase in maximum shear wave displacement, a 0.11 m/s decrease in the overall interquartile range and median SWV and a 17.6% increase in the success rate of SWV measurements were obtained when filter-based harmonic imaging was used instead of fundamental imaging. (E-mail: amadorcarrascal.carolina@mayo.edu) © 2016 World Federation for Ultrasound in Medicine & Biology.**

*Key Words:* **Liver elasticity, Filter-based harmonic imaging, Shear wave elasticity imaging, *In vivo*.**

## INTRODUCTION

A major application of shear wave elasticity imaging (SWEI) is the non-invasive quantification of liver elasticity, which has revealed that higher shear wave velocities correlate with progressive liver fibrosis (Bavu et al. 2011; Chen et al. 2013; Fierbinteanu-Braticevici et al. 2009; Gheonea et al. 2010; Kirk et al. 2009; Yin et al. 2007). The most common methods used to quantify absolute liver elasticity with ultrasound imaging include transient elastography (TE) (Sandrin et al. 2002), acoustic radiation force impulse (ARFI) imaging (Nightingale et al. 2002), shear wave elasticity imaging (SWEI) (Sarvazyan et al. 1998) and supersonic shear wave imaging (SSI) (Bercoff et al. 2004). SWEI methods are attractive, as shear waves are generated inside the tissue of interest; shear wave propagation is then monitored in space and time by pulse–echo ultrasound, and the tissue elasticity is estimated from the shear wave propagation velocity.

The ability to successfully detect shear wave propagation can be impaired by ultrasound image quality, which is susceptible to mechanisms such as phase aberration, ultrasound attenuation and clutter. In most ultrasound imaging systems the focusing and steering time delays, which ensure that signals on all channels are in phase at the focal area, are calculated based on the speed of sound (O'Donnell and Flax 1988). The speed of sound actually varies from 1,470 m/s in fat to 1,665 m/s in collagen (Goss et al. 1978). Focusing delays are usually inaccurate when tissue layers with varying acoustic velocity lie between the transducer and the organ of interest (Kremkau and Taylor 1986). The ultrasound beams become defocused because of the inhomogeneities in acoustic velocity. This effect is commonly known as phase aberration.

In addition to phase aberration, ultrasound attenuation is another mechanism that can affect ultrasound image quality; it results from the interaction of ultrasound waves with tissue through absorption and scattering mechanisms (Kremkau and Taylor 1986). Ultrasound attenuation is usually modeled to be proportional to frequency; therefore, by lowering the beam frequency, ultrasound attenuation effects are reduced, but as a consequence, image resolution is compromised.

Ultrasound clutter is a common ultrasound artifact that appears as a diffuse haze as a consequence of sound

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reverberation in tissue layers, off-axis scattering from tissue structures, beam distortion, previously transmitted pulses signal and random electronic or acoustic noise (Kremkau and Taylor 1986). The effects of inhomogeneities in acoustic velocities, ultrasound attenuation and clutter have been studied extensively for conventional ultrasound B-mode imaging (Foster and Hunt 1979; Lediju et al. 2008; Lyons and Parker 1988; Smith et al. 1988; Trahey and Smith 1988), and several methods have been proposed (Montaldo et al. 2009; Ng et al. 1994; Tranquart et al. 1999) to ameliorate the effects.

To address these issues of poor ultrasound image quality, tissue harmonic imaging (THI) is widely used and significantly improves ultrasound image quality. THI takes advantage of the non-linear propagation of ultrasound in tissue, which creates ultrasound signal frequency components that are not present in the incident beam (Thomas and Rubin 1998). It is believed that because harmonic signal passes through attenuating and/or aberrating layers only once, THI suffers less from phase aberration and/or attenuation; however, it has been found with simulations that the second harmonic experiences aberration to that experienced by the fundamental component (Varslot et al. 2007) and that the major contribution of THI emanates from reverberation suppression (Pinton et al. 2011), allowing for better spatial and contrast resolution (Shapiro et al. 1998; Ward et al. 1997).

Recently, it was found that THI can significantly improve SWEI measurements. Use of SWEI in myocardial tissue has been significantly improved thanks to conventional pulse-inversion harmonic imaging (PIHI) shear wave motion detection. Song et al. (2013), in a study of six healthy patients, reported that the left ventricular myocardium stiffness could be measured transthoracically with a 93.3% success rate using a phased array transducer. Conventional PIHI requires a pair of phase-inverted pulses to reconstruct an image, so the effective pulse repetition frequency (PRF) used to track shear wave propagation is decreased by a factor of 2, which could be a limitation when tracking faster shear waves as in the case of stiffer tissues. On the other hand, Song et al. (2014) reported the implementation of filter-based harmonic imaging (FHI) for abdominal applications using a curved array transducer in tissue-mimicking phantoms. Nonetheless, the advantages of THI for shear wave detection have not been significantly explored for *in vivo* abdominal applications.

The purpose of this study was to evaluate shear wave displacement detection with fundamental imaging and FHI in *in vivo* healthy livers. Shear wave displacement and shear wave velocity estimation are evaluated using fundamental imaging and FHI to track shear waves resulting from focused acoustic radiation force shear wave generation at two depths below the liver capsule, in addition to imaging the liver from two intercostal spaces.

## METHODS

### *Patients*

Healthy individuals with no history of liver disease were included in the study. The experiment protocol was approved by the Mayo Clinic institutional review board and compliant with HIPAA. Written informed consent was obtained before scanning. Seventeen patients (10 female, 7 male) with a body mass index (BMI) of 28 ± 6 kg/m$^2$ (mean ± standard deviation, n = 17) and aged 43 ± 12 y (mean ± standard deviation, n = 17) were recruited for the study.

Images and measurements were obtained by an experienced sonographer with the ultrasound probe positioned at the seventh and eighth intercostal spaces during breath holds. Under the guidance of B-mode imaging, the sonographer located the tissue of interest and regions of interest (ROIs) 1.5 and 2.5 cm below the liver capsule. Five independent acquisitions were performed. The total number of acquired data sets per subject was 20 (2 intercostal spaces, 2 ROIs, 5 acquisitions).

### *Experimental design*

A Vantage ultrasound system (Verasonics, Kirkland, WA, USA) equipped with a curved array transducer (C5-2 v, Verasonics) was used. The detection beams were wide beams with an *f*/9.9 focal configuration transmitted with a frequency of 2 MHz, which produced a second harmonic signal around 4 MHz. During receive, the radiofrequency (RF) signals from two steering angles (−1°, +1°) were coherently compounded (Montaldo et al. 2009), giving an effective pulse repetition frequency (PRF) of 2.77 kHz. The pulse sequence was the same for both fundamental imaging and FHI, and a finite-impulse response (FIR) filter was used to extract the fundamental (named *fundamental tracking* hereinafter) and second harmonic (named *harmonic tracking* hereinafter) components of the RF signal separately; therefore, the resulting RF signals are from the same tissue. The Verasonics system converts the RF signals to in-phase/quadrature (IQ) signals, which are saved for later analysis.

### *Data analysis*

Axial particle displacement ($U_z$), which will be referred to as shear wave displacement, was then calculated from both fundamental and harmonic components using an autocorrelation method (Pinton et al. 2006). The shear wave displacement data within the focal zone were averaged along the axial direction (focal depth: ±2.5 mm) to create spatiotemporal maps. Then, shear wave maximum displacement (Max $U_z$) and shear wave