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● *Original Contribution*

ULTRASOUND IMAGING OF LONG BONE FRACTURES AND HEALING WITH THE SPLIT-STEP FOURIER IMAGING METHOD

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Abstract—We applied the split-step Fourier imaging method to back-propagate the ultrasound zero-offset wavefields acquired on the bone surface to the sources of scatterers, which are the reflecting interfaces. The method required, as an input, an estimated slowness (reciprocal of half the velocity) model to map the time-dependent sonogram to the depth image, which provides the geometric properties of the interfaces. The slowness was approximated by a depth-dependent term and a first-order spatially varying perturbation. Simulated data sets were used to validate the method. The reconstructed images show proper mapping of the interfaces and the fracture, and a reasonable cortical thickness measurement with 8.3% error. The images also illustrate clearly the bone fracture healing process of a 1-mm-wide 45° inclined crack with different in-filled tissue velocities for various healing stages. Reconstruction of a fractured bone plate using data from an *in vitro* experiment is also presented. This study suggests that the proposed imaging method has good potential in quantification of bone fractures and monitoring of the fracture healing process. (E-mail: lawrence.le@ualberta.ca) © 2013 World Federation for Ultrasound in Medicine & Biology.

Key Words: **Ultrasound imaging, Cortical bone, Image reconstruction, Bone fracture, Bone healing.**

INTRODUCTION

It has been estimated that more than 2 million osteoporosis-related fractures occurred in the United States in 2005. This figure is projected to increase by 50% by 2025 (Burge et al. 2007). Fracture accounts for about 24% of injury-related costs and is associated with losses amounting to billions of dollars in the United States (Finkelstein et al. 2006). Therefore, fracture diagnosis and subsequent healing care are identified as major health priorities. Although conventional radiography is still the most common modality used to assess fracture healing, ultrasound assessment is emerging as a promising diagnostic tool (Atkinson and Lennon 2003; McManus et al. 2008).

Over the years, researchers have tried to incorporate ultrasound into fracture diagnosis procedures, especially in children. Not only is ultrasound free of ionizing radiation, but it is also an important complement to radiography, because radiography-based diagnosis involves radiation and does not provide complete information. Ultrasound can detect occult fractures, which cannot be identified by radiography (Legome and Pancu 2004). Controlled experiments have been carried out to detect fractures in children using both ultrasound and radiography (Moritz et al. 2008). Results indicated a sensitivity of 92.9% for ultrasound and 93.2% for radiography, and ultrasound was superior to radiography in detecting clavicle fracture. Although different sensitivity values were obtained by several groups (Ackermann et al. 2009; Weinberg et al. 2010), they have unanimously come to the conclusion that ultrasound is a valuable and safe alternative to radiography in the diagnosis of bone fractures. It was even proposed that for children with trauma, ultrasound is the imaging method of choice. In the presence of compound fracture, radiography should be used to scan the region of interest pre-defined by ultrasound (Ackermann et al. 2009; Hubner et al. 2000; Moritz et al. 2008), reducing the radiation exposure of pediatric patients.

Ultrasonography has been applied at the beginning of the fracture healing process (6–9 wk) when radiography failed to detect the non-ossified callus formed in

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the first stage of healing (Craig et al. 1999; Maffulli and Thornton 1995; Moed et al. 1998). The success rate in predicting the status of union was 97% in a study involving 47 patients with tibial fractures (Moed et al. 1998). To quantify the healing process, a vibrational technique known as computerized sonometry (Morshed et al. 2008) was used to measure the velocity of ultrasound traveling from a transmitter to a receiver across a fracture. The measurement was then compared with a baseline measurement on intact bone to determine the stage of the healing process. Apart from velocity, the use of fracture transmission loss to assess bone strength and monitor fracture healing has also been studied (Dodd et al. 2007, 2008; Gheduzzi et al. 2009). When measurements were made at multiple stations deployed collinearly from the transmitter in an axial transmission configuration, the data indicated various wave types such as bulk waves and Rayleigh guided waves (Le et al. 2010; Ta et al. 2006). Developed for non-destructive evaluation (NDE) of materials and structural defects, guided waves are elastic waves propagating in plates, rods and shells and depend on the wavelength/thickness ratio (Prosser et al. 1999). Recently, guided-wave modes for intact and healing bones were studied by simulation and *ex vivo* measurements (Protopappas et al. 2006, 2007).

Conventional B-mode medical ultrasound is used mainly in soft tissue imaging. For hard tissue imaging, B-mode images of bone tissues lack image quality and spatial resolution partly because of the limited penetrability of ultrasound through bone. Research in ultrasonic bone imaging is challenging because wave propagation is complicated in bone tissues. Ultrasonic image reconstruction methods such as quantitative ultrasonic tomography (Lasaygues 2006; Lasaygues et al. 2005, 2007) have been developed to image cross-sectional bone structures.

The split-step Fourier imaging (SSFI) method, also known as the split-step Fourier migration method in geophysics, was developed by Stoffa et al. (1990) and has been successfully applied to image earth structures for decades (Fehler and Huang 2002; Stoffa et al. 1990) using zero-offset data. The method extrapolates the wavefields in depth and back-propagates them in the Fourier domain to reconstruct the images of the reflecting surfaces. The reconstruction method approximates the material slowness by a depth-dependent reference slowness and a first-order spatially varying slowness perturbation, and, therefore, is valid for moderately lateral slowness variations.

In this article, we describe our investigation of the use of the SSFI method to image bone structures using pulse-echo ultrasound data. We present a new application of the method that has never been reported in the field of bone imaging. The validity of the method in imaging bone structures and monitoring bone healing was examined using simulated data for intact and fractured bone models. Moreover, we describe an image reconstruction of a fractured bovine bone sample.

METHODS

Split-step Fourier imaging method

In our study, we used a pulse-echo model in which the same transducer sends the pulse and receives the echoes, also known as a zero-offset case. We considered acoustic (compressional) waves because we assumed normal incidence where mode conversion did not occur, and we mimicked soft tissue using water in which shear waves do not exist. In actual clinical applications, the shear wave velocity of soft tissue is very small (<100 m/s) around 2 MHz (Madsen et al. 1983).

The application of the SSFI method to the zero-offset case is based on the exploding reflector model (Loewenthal et al. 1976). The acoustic waves emanate from a source and travel through the medium at a material velocity v. After being reflected at an interface, some of the waves travel back along the initial path to the receiver located at the same position as the source. Alternatively, according to Huygens' principle, we can consider the reflected waves to be generated by a series of diffracting source points along the reflecting surface at time zero and to travel one way to the recording surface at half the velocity, that is, $v/2$. Therefore, the downward ray path from the source to the reflector is identical to the upward ray path from the reflector to the receiver.

For a two-dimensional (x, z) isotropic medium, the Helmholtz wave equation for the acoustic wavefield $u(x,z,t)$ is

$$\nabla^2 U(x,z,\omega)+\omega^2 s^2(x,z)U(x,z,\omega)=0 \tag{1}$$

where $s(x,z)$ is the spatially varying slowness, which is the inverse of half the velocity, $2/v(x,z)$, ω $(=2\pi f)$ is the angular frequency, and $U(x,z,\omega)$ is the Fourier transform of the wavefields $u(x,z,t)$. We approximate the slowness with the background slowness, $s_0(z)$, and the first-order perturbation, $\Delta s(x,z)$, that is,

$$s(x,z)\simeq s_0(z)+\Delta s(x,z) \tag{2}$$

where $|\Delta s(x,z)| \ll s_0(z)$ and all higher-order terms are ignored. The reference or background slowness, $s_0(z)$, depends only on the depth level, z, and can be obtained via $s_0(z)=2/\bar{v}(z)$, where $\bar{v}(z)$ is the average of $v(x,z)$ with respect to x, that is, $\bar{v}(z)=\sum_{i=1}^{m} v(x_i,z)/m$, and m is the number of x-position of the transducer. The perturbation term, $\Delta s(x,z)$, takes the moderate variation of the lateral slowness into account. With (2) in (1), eqn (1) can be rewritten as the inhomogeneous wave equation