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AXIAL TRANSMISSION METHOD FOR LONG BONE FRACTURE EVALUATION BY ULTRASONIC GUIDED WAVES: SIMULATION, PHANTOM AND *IN VITRO* EXPERIMENTS

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Abstract

Mode conversion occurs when the ultrasonic guided waves encounter fractures. The aim of this study was to investigate the feasibility of fracture assessment in long cortical bone using guided-mode conversion. Mode conversion behavior between the fundamental modes S₀ and A₀ was analyzed. The expressions proposed for modal velocity were used to identify the original and converted modes. Simulations and phantom experiments were performed using 1.0-mm-thick steel plates with a notch width of 0.5 mm and notch depths of 0.2, 0.4, 0.6 and 0.8 mm. Furthermore, *in vitro* experiments were carried out on nine ovine tibias with 1.0-mm-wide partial transverse gap break and cortical thickness varying from 2.10 to 3.88 mm. The study confirmed that mode conversion gradually becomes observable as fracture depth increases. Energy percentages of the converted modes correlated strongly with fracture depth, as illustrated by the frequency-sweeping experiments on steel phantoms (100–1100 kHz, $r^2 = 0.97$, $p < 0.0069$) and the fixed-frequency experiments on nine ovine tibias (250 kHz, $r^2 = 0.97$, $p < 0.0056$). The approaches described, including mode excitation, velocity expressions and energy percentage criteria, may also contribute to ultrasonic monitoring of long bone fracture healing.

Keywords

Long cortical bone; Fracture severity assessment; Ultrasonic guided waves; Mode conversion; Mode velocity; Energy percentages

INTRODUCTION

Bone fracture is a medical condition in which bone discontinuity is created by stresses higher than the bone can bear. Fracture healing is a proliferative process, and full recovery can take 3–5 years (Martin et al. 1998; Phillips et al. 2005). To minimize the negative consequences of delayed fracture consolidation, relatively inexpensive and effective diagnostic and monitoring tools are essential (Calori et al. 2007; Dahabreh et al. 2009; Kanakaris 2007). Conventional radiography continues to be the most common means of

bone fracture evaluation in clinical practice (Davis et al. 2004). Recently, increased attention has been paid to bone quantitative ultrasound, potentially one of the best alternatives to radiologic methods with the advantages of low cost, portability and lack of ionizing radiation exposure (Lasaygues et al. 2005; Laugier and Haïat 2011; Li et al. 2013; Protopappas et al. 2008; Rose et al. 2004; Ta et al. 2008; Tatarinov and Sarvazyan 2003). As an elastic wave, ultrasound is intrinsically suitable for evaluation of biomechanical properties and for monitoring of fracture healing. The axial transmission (AT) technique, first used in the late 1950s to evaluate fracture healing, is the classic method used to study long cortical bones (Abendschein et al. 1972; Gerlanc et al. 1975; Siegel et al. 1958). In the AT technique, typically, a pair of transducers are positioned on the same side of the long bone, in contact with the skin, to measure the transmitted signal along the axial direction (Chen et al. 2012; Kilappa et al. 2013; Tatarinov et al. 2011). Experimental and theoretical analysis indicates that at least two distinct contributions, the first arriving signal (FAS) and a slower energetic component, can be measured by the AT technique (Bossy et al. 2002; Laugier and Haïat 2011; Nicholson et al. 2002; Tatarinov et al. 2011).

Recent studies have revealed that the nature of the FAS depends on the ratio of longitudinal wavelength to cortical bone thickness. When cortical thickness is greater than the longitudinal wavelength of the bone, the FAS corresponds to a lateral wave propagating along the surface between the soft tissue and cortical layer at the longitudinal velocity of the bone. In cases of lower thickness or larger wavelength, however, FAS velocity decreases to that of the symmetric guided mode S₀ (Bossy et al. 2002; Nicholson et al. 2002). Results from simulation, phantom, animal and clinical studies have revealed that the velocity and signal loss of the FAS can be used to monitor fracture healing (Barbieri et al. 2011; Dodd et al. 2007; Gerlanc et al. 1975; Protopappas et al. 2005; Saulgozis et al. 1996). Some FAS-based clinical devices for long bone measurement have already been developed, for example, the Soundscan 2000, which operates at 250 kHz (Myriad Ultrasound Systems, Rehovot, Israel); the Omnisense, which operates at 1.25 MHz (Sunlight Medical, Tel-Aviv Israel); the wearable ultrasound system USBone, which uses 1-MHz transducers to monitor and accelerate fracture healing (Computer Science Department, University of Ioannina, Ioannina, Greece); and a prototype that measures the FAS using bidirectional axial transmission at 1 MHz (Laboratoire d'Imagerie Paramétrique, Paris, France) (Bossy et al. 2004; Muller et al. 2005; Njeh et al. 1997; Protopappas et al. 2005).

Ultrasonic guided waves are well-established and widely used means of cost-effective damage identification and structural health monitoring in the field of nondestructive ultrasonic testing and evaluation (Rose et al. 2004). Driven by recent advances and technical breakthroughs in sensor technology and signal processing, several groups of researchers have shifted the innovative emphasis toward ultrasonic guided waves in long bones (Laugier and Haïat 2011; Le et al. 2010; Lefebvre et al. 2002; Moilanen 2008; Protopappas et al. 2008; Ta et al. 2006; Tatarinov et al. 2011). Compared with the traditional FAS, guided modes transmit throughout the entire cortical layer, with significant sensitivity to the endosteal region and cortical thickness (Moilanen 2008). It has been found that guided modes, propagating in the tubular human tibia and radius, can be identified and separated by suitable signal processing algorithms (Moilanen et al. 2006; Ta et al. 2006). Multiple parameters of the guided modes, such as attenuation (Ta et al. 2009), velocities (Chen et al.

2012; Minonzio et al. 2010; Moilanen et al. 2006) and energy (Protopappas et al. 2006, 2007; Xu et al. 2012), have been extracted to estimate bone properties. In particular, guided wave-based cortical thickness determinations have been carried out on specimens of bovine tibias ($r^2 = 0.79$, $p < 10^{-5}$ [Ta et al. 2006]), human radiuses ($r^2 = 0.81$, $p < 0.001$ [Moilanen et al. 2007]; $r^2 = 0.6$, $p < 10^{-5}$ [Sasso et al. 2009]) and human proximal tibias ($r^2 = 0.84$, $p < 0.01$ [Tatarinov et al. 2011]). Use of a bone scanner system (Bone UltraSonic Scanner) that combines guided waves (0.1 MHz) with the FAS (1 MHz) in the diagnosis of osteoporosis has been reported (Sarvazyan et al. 2009). However, the use of ultrasonic guided waves to evaluate fractured long cortical bones, for which results are limited to several simulation studies, has progressed very slowly (Laugier and Haïat 2011; Moilanen 2008; Protopappas et al. 2008). Mode conversion has been observed in 2-D (Protopappas et al. 2006) and 3-D (Protopappas et al. 2007) simulations of fractured and healing long bones. It has been found that the modal energy of transmitted signals is gradually reconstructed during consolidation of the callus, which provides a means of monitoring fracture healing (Protopappas et al. 2006). However, because of an immature understanding of mode conversions in fractured long cortical bones (Laugier and Haïat 2011; Protopappas et al. 2008) and the challenges of guided mode identification and separation (Moilanen 2008; Sasso et al. 2009; Song et al. 2011; Xu et al. 2010), a quantitative relationship between guided mode conversion and indicators of fracture status, such as geometric and biomechanical properties, has not been studied in diaphysis assessment. Recently, a preliminary analysis of guided mode interaction with a diaphyseal transverse osteotomy was performed by 2-D finite-difference time-domain simulation, and reflection and transmission coefficients of the guided modes in the fractured long bone were determined quantitatively (Xu et al. 2014a). It was found that mode conversion characteristics of the fractured long bone have great potential in the assessment of fracture status, even in the early diagnosis of minimal gap breakage (gap width < 0.5 mm).

The purpose of this study was to investigate guided mode conversion in fractured long bones and to propose sensitive parameters for long bone fracture evaluation. We first offer a comparative analysis between numerical simulations and experimental measurements of plate phantoms with notch depth variation to analyze the fundamental principle of mode conversion occurring in fractured long cortical bones. A mode velocity relationship and energy criteria verified by frequency-sweeping measurements from 100 to 1100 kHz are proposed. *In vitro* experiments using 250-kHz excitation carried out in diaphyseal regions of nine ovine tibias with partial transverse fractures are described. Finally, correlations between original and converted energy and fracture percentages are discussed.

METHODS

In the experiments, the AT technique was adapted to measure guided modes in steel plate phantoms and ovine tibias. The emitter and receiver were placed on opposite sides of the fracture, and ultrasound signals were recorded at axially shifted distances by the receiver scanning (Protopappas et al. 2008) or transducer (Chen et al. 2012) array.

In this section, we present a fundamental description of guided mode conversion, excitation and measurement. Next, the numerical simulation and experimental setup are explained,

followed by details on the plate phantoms and tibia specimens. Finally, signal processing procedures, including velocity measurement and energy percentage (EP) calculation, are introduced.

Mode conversion of guided waves

Guided waves are mechanical vibrations that can be supported by the long cortical bone (Laugier and Haïat 2011; Lefebvre et al. 2002). Numerous guided modes can exist in a long cortical bone with different properties such as velocity, attenuation and energy. However, the common ground is that the velocities of all the guided modes can be theoretically determined by the dispersion functions, which correlate to waveguide geometry and acoustical properties, including thickness, density, Young's modulus, longitudinal velocity and shear velocity (Laugier and Haïat 2011; Lefebvre et al. 2002; Moilanen 2008; Rose et al. 2004; Tatarinov et al. 2011). The dispersive velocities are usually formulated as functions of the frequency-thickness product ($f \cdot d$). In ultrasonic studies of long bones, thickness refers to the thickness of the diaphysis cortex. Within the relatively low $f \cdot d$, most of the wave energy propagates in two fundamental zero-order guided modes, symmetric (S0) and anti-symmetric (A0) (Moilanen 2008; Rose et al. 2004).

When guided waves encounter defects, the incident guided modes usually convert to other modes, a process known as mode conversion (Alleyne and Cawley 1992; Protopappas et al. 2006, 2007, 2008). As illustrated in Figure 1, while the forward propagating S0 and A0 come across the notch, mode conversion occurs, so that the received signals generally contain the originally transmitted S0 and A0, the A0-converted mode, S0_{A0}, and the S0-converted mode, A0_{S0}. Assuming it is a narrowband excitation with a small dispersion, the group velocities of A0 and S0 can be regarded as two constants, V_{S0} and V_{A0} . The distance from the center of the left transmitter to the notch is x_1 , the distance from the notch to the right receiver center is x_2 and the total propagation distance is $x_1 + x_2$. Thus, we calculate the propagation delays and velocities of S0_{A0} and A0_{S0} as follows (Xu et al. 2014b):

Group delay:

$$T(A0_{S0}) = x_1/V_{S0} + x_2/V_{A0} \quad (1a)$$

$$T(S0_{A0}) = x_1/V_{A0} + x_2/V_{S0} \quad (1b)$$

Velocity:

$$V(A0_{S0}) = \frac{x_1 + x_2}{x_1/V_{S0} + x_2/V_{A0}} \quad (1c)$$

$$V(S0_{A0}) = \frac{x_1 + x_2}{x_1/V_{A0} + x_2/V_{S0}} \quad (1d)$$

Because at low θ , V_{S0} is larger than V_{A0} , the packets of $S0_{A0}$ and $A0_{S0}$ propagate in the middle between $S0$ and $A0$. If x_1 is very small, then $V(S0_{A0})$ and $V(A0_{S0})$ are close to $V(S0)$ and $V(A0)$, respectively. However, if x_2 is very small, then $V(S0_{A0})$ and $V(A0_{S0})$ are close to $V(A0)$ and $V(S0)$, respectively. The difference may be used to evaluate uncentered fractures. In particular, if $x_1 = x_2$, we obtain (Xu et al. 2014b).

$$V(S0_{A0}) = V(A0_{S0}) = 2V_{A0} * V_{S0} / (V_{A0} + V_{S0}) \quad (2)$$

In eqn (2), modes $S0_{A0}$ and $A0_{S0}$ merge as a mixed wave packet ($S0_{A0} + A0_{S0}$) propagating between the original $S0$ and $A0$ modes.

Mode excitation and measurement

The incident angle and waveform are two critical factors for mode excitation and measurement. According to the source influence theory, normal contact transducers have wide phase velocity spectra, resulting in difficulty in mode identification and velocity calculation (Luo et al. 2004). Compared with the contact transducer, angle beamed transducers have narrower phase velocity spectra, which is helpful for simplicity of the signal. Thus, a couple of 45° Perspex wedges (Olympus, Waltham, MA, USA) were used.

Because the pulse signal has a very broad frequency band, multiple guided modes can be excited, causing difficulties in signal processing. Windowed tone bursts (four-cycle Gaussian-windowed sinusoids), rather than simple pulses, were adopted in the simulations and experiments, ensuring excitation of relatively non-overlapping wave packets with narrowband and smooth spectra (Alleyne and Cawley 1992).

Simulations

The simulations were carried out with self-developed software using the 2-D finite-difference time-domain method (Liu et al. 2013).

The model is illustrated in Figure 1. Mode excitation and measurement were adjusted according to experimental conditions. The steel material used in the simulations had a density (ρ) of 7.932 g/cm³, longitudinal wave velocity (V_l) of 5960 m/s and shear wave velocity (V_t) of 3200 m/s (Xu et al. 2012). Plate thickness (d) was 1.0 mm, and notch width (w) was fixed at 0.5 mm. Notch depth (h) varied from 0.1 to 0.9 mm with a 0.1-mm interval, to imitate degree of diaphyseal fracture from 10% to 90%. For consistency in comparisons, the distance between the receiver and transmitter was kept at a constant 100 mm, and transverse notches were artificially arranged exactly in the middle of emitter and receiver.

To analyze the robustness of the mode conversion parameters, a frequency-sweeping test was used in the simulations (Xu et al. 2014b). According to the theoretical dispersion curves

of the steel plates, at $f \cdot d < 1100 \text{ kHz} \cdot \text{mm}$, guided signal energy is confined in the form of S0 and A0 (Alleyne and Cawley 1992). Consequently, in the simulation, the center frequencies of the excitations varied from 100 to 1100 kHz, with a 100-kHz interval to cover the whole $f \cdot d$ range, where only S0 and A0 modes exist.

Experimental apparatus

All measurements were performed at room temperature (20°C) in air. Ultrasonic gel (Echo Jelly, Aloka Medical Equipment, Shanghai, China) was used as coupling agent to ensure good contact between the transducers and the cortical surface. A couple of broadband piezoelectric transducers (Olympus) with a central frequency of 1 MHz and diameter of 10 mm were used in the measurements, one acting as the transmitter and the other as the receiver. Figure 2 is a schematic diagram of the experimental setup. To adjust the propagation distances of guided waves in the axial direction of long bones or bone phantoms, transducer positions were controlled with UltraPAC step motors (Physical Acoustics, Princeton, Junction, NJ, USA). The power amplifier (AG 1021, T&C Power Conversion, Rochester, NY, USA) delivered the Gaussian windowed tone burst from the arbitrary waveform generator (33220 A, Agilent, Englewood, CO, USA) to the emitter. Simultaneously, excitations were also used as synchronized signals to trigger the oscilloscope. The pulser/receiver unit (5900 PR, Olympus) was used to filter and amplify the received signals. In all measurements, signals were averaged 256 times and digitized to 8 bits (HP54642A, Agilent, Santa Clara, CA, USA) at a sampling rate of 20 MHz. Meanwhile, the same measurements were repeated three times under the same experimental conditions.

Phantoms

Steel plates were used as bone phantoms in this study. Although steel has higher density and faster longitudinal wave and shear wave velocities, it is appropriate for revealing the fundamental mechanism of mode conversion.

One intact and four notched 1.0-mm-thick steel plates were prepared with a notch width of 0.5 mm and notch depths of 0.2, 0.4, 0.6 and 0.8 mm, respectively, which were made by a computer numerically controlled milling machine. The plates were 600 mm long and 200 mm wide to suppress the effects of ultrasonic reflection from the edges.

Similar to the simulation, the distance between receiver and transmitter was kept at a constant 100 mm with the notch axially centered. However, as the time resolution of the 100-kHz frequency is too low for mode identification and separation, the incident frequency was swept from 200 to 1100 kHz.

Ovine tibias

Nine fresh ovine tibias were used in the *in vitro* experiments (Table 1). The experimental protocol received approval from Fudan University's Institutional Animal Care and Use Committee. The diaphyseal length of these specimens ranged from 17 to 20 mm. Mean outer diameter varied from 12.15 to 20.31 mm, and mean cortical thickness varied from 2.10 to 3.88 mm. Notch widths, determined by the saw blade, were all 1.0 mm. Density (ρ),

longitudinal velocity (V_l) and shear velocity (V_t) of the ovine tibias were $\rho = 1.5 \text{ g/cm}^3$, $V_l = 4063 \text{ m/s}$ and $V_t = 1846 \text{ m/s}$, respectively (Protopappas et al. 2007).

The epiphysis regions were extracted, and the overlying soft tissue was removed. For each sample, the emitter and receiver were fixed with the transverse notch exactly in the middle of the diaphyseal pathway. Notch breaks were sawed step by step to imitate different degrees of fracture. After the experiments, each sample was sawed off at three different places to measure average cortical thickness.

Higher excitation frequency denotes a shorter duration and better temporal resolution, which is beneficial for mode identification and separation. However, in the case of some thick cortical samples, high-frequency pulses usually excite the unwanted high-order guided modes and complicate the multimode separation. To achieve a compromise between temporal resolution and mode suppression, 250-kHz four-cycle Gaussian windowed excitation was used in measurements of ovine tibias.

Data processing

To separate the individual components of the guided signals in the long bone, several algorithms are available, such as time-frequency representation (Tatarinov et al. 2011), multiridge-based time-frequency separation (Xu et al. 2010), 2-D Fourier transform (Moilanen et al. 2006), singular value decomposition (Minonzio et al. 2010), dispersion compensation (Xu et al. 2012) and joint approximate diagonalization of eigenmatrices (Song et al. 2011). In the study, taking advantage of the narrowband excitation and angle beamed transducer, modal packets in the received signals are not strongly dispersive and overlapped and can thus be identified and obtained by simply cutting windows (Alleyne and Cawley 1992; Luo et al. 2004; Xu et al. 2014b). However, because the frequency-sweeping measurements were performed in simulations and experiments, lengths of the mode packets were not fixable, but varied with excitation duration. Consequently, the cutting window lengths were adjusted to comply with excitation durations. In particular, according to eqn (2), because the notches were arranged exactly in the middle of the transmitter and receiver, converted modes $S0_{A0}$ and $A0_{S0}$ were detected with the same arrival time and processed together as a whole converted packet. Three individual components, that is, original modes $S0$ and $A0$ and the converted packets of $S0_{A0}$ and $A0_{S0}$ ($S0_{A0} + A0_{S0}$), were thus separated from each guided signal by the fixed-length cutting windows (Xu et al. 2014b).

Modal velocities were then determined from the known source-receiver distance and the arrival time of the mode packet maximum amplitude. Mode energy was calculated as the sum of signal amplitude squared (Cohen 1989). Normalized EPs were used to evaluate mode conversion and further predict fracture status.

RESULTS

Simulation study

Figure 3a presents the simulated waveforms of the steel models under 500-kHz excitation, with notch depths ranging from 10% to 90% of the thickness. As expected, mode conversion did not occur in the intact model, and there were only two original wave packets, $S0$ and $A0$,

at the testing frequency. As the increasing fracture depth aggravated the mode interaction effects, the converted packet, $S0_{A0} + A0_{S0}$, gradually emerged and enlarged in the middle of the original modes S0 and A0. To quantify mode conversion phenomena, original packets of S0 and A0 and converted packets were extracted, and the corresponding EPs were calculated. As illustrated in Figure 3b, mode conversion and energy transfer are obvious from the EP curve variation. When notch depth reached 60% of plate thickness, the converted $S0_{A0} + A0_{S0}$ gained most of the energy from the original S0 and A0 through mode conversion, and its EP value increased to 40%. In contrast, the energy of the original modes S0 and A0 decreased monotonically. At notch depths > 80% of the thickness, the EPs of original S0 and A0 were <20%.

Figure 4 illustrates the results for the intact model (0%) and fractured models (10%–90%) with the incident frequency sweeping from 100 to 1100 kHz. Significant consistency in velocity was observed between the simulated results and theoretical curves. According to Xu et al. (2014b), the theoretical velocity can be calculated with eqn (2). Because of the relatively poor time resolution at lower frequencies, the maximum velocity errors were taken at 100 kHz ($\Delta d = 100 \text{ kHz} \cdot \text{mm}$), where the root-mean-square-error (RMSE) of S0, A0 and $S0_{A0} + A0_{S0}$ were 0.10, 0.45 and 0.53 km/s, respectively. The frequency-sweeping (100 to 1100 kHz) EPs of S0, A0 and $S0_{A0} + A0_{S0}$ are plotted versus fracture percentage in Figure 4b. In strong agreement with the results in Figure 3b, it was found that as notch depth increased, the EP curve of the converted modes increased monotonically, whereas the curves of the original modes decreased gradually.

Phantom experiment

Mode conversions were further analyzed in the steel phantoms. The signals excited by a 500-kHz four-cycle Gaussian envelope pulse are given as examples (Fig. 5a). It can be seen from the waveform of the intact plate (0% fracture percentage, bottom of Fig. 5a) that the converted packet did not appear and there were only two wave packets, S0 and A0. However, as fracture percentage increased, the relative magnitudes of the converted $S0_{A0}$ and $A0_{S0}$ gradually increases, whereas the magnitudes of the original modes gradually decreased.

Similar to the simulations, the frequency-sweeping tests (200–1100 kHz) were carried out on steel phantoms. In Figure 5b, the velocities of S0, A0 and $S0_{A0} + A0_{S0}$ measured in frequency-sweeping tests were generally in good agreement with theoretical results. Because of a decrease in time resolution, maximum errors occurred at 200 kHz ($\Delta d = 200 \text{ kHz} \cdot \text{mm}$). The RMSEs of S0, A0 and the converted packet $S0_{A0} + A0_{S0}$ were 0.29, 0.32 and 0.49 km/s, respectively. The frequency-sweeping EP results of the phantoms are marked by symbols in Figure 5c, in which fracture percentage varies from 20% to 80%. Obvious mode conversions were observed in the phantom experiments. The EPs of the converted modes increased monotonically with increasing fracture percentage (Fig. 5c). The slope of the plot of converted EP versus fracture percentage was about 0.3, different from that for the simulation in Figure 4c, which can be attributed to the coupling effects between the transducers and phantoms.

In vitro experiment

The raw signals measured in specimen 1 are given as examples in Figure 6a, where the percentiles denote the degree of fracture. Deepening fracture was found to significantly weaken the transmitted energy. Compared with the signal for the intact tibia (specimen 1, bottom of Fig. 6a), the maximum amplitude of the signals for the fractured tibias gradually decreased to half of the original magnitude (80% fracture percentage, top of Fig. 6a). The $f \cdot d$ values varied from 400 to 1000 kHz•mm depending on the thickness of the sample. Good consistency in velocity was obtained between measurements (symbols) and theoretical predictions (lines). The EPs of ovine tibias are plotted in Figure 6c. Similar to the phantom results (Figs. 4b and 5c), fracture deepening gradually aggravated mode conversion and increased the converted EPs. As the fracture percentage of the tibias reached 80% of cortical thickness, the converted components grew to be the most energetic, with an EP value of 45.3%.

Comparative analysis

Results for the plate phantoms and ovine tibias consistently indicate that EPs of the converted modes are highly related to the variation in notch depth. In Figure 7a are the fitted EP curves of the original and converted modes measured in the five notched phantoms with $f \cdot d$ varying from 200 to 1100 kHz•mm. The slope of the energy conversion in the plate phantoms was 0.3, and the correlation between converted EP and fracture percentage was $r^2 = 0.97$, $p < 0.0069$. In Figure 7b are the fitted EP curves of the original and converted modes measured in the *in vitro* experiments on nine ovine tibias. The correlation between converted EP and fracture percentage was $r^2 = 0.97$, $p < 0.0056$. The increasing slope of the converted energy was approximately 0.6.

DISCUSSION

Although X-ray imaging has been widely accepted for the clinical diagnosis of bone fracture, there is significant interest in the use of ultrasound methods for monitoring fracture healing and preventing the delayed union of bone fractures, especially before the occurrence of the radiographic evidence (Barbieri et al. 2011; Moed et al. 1995; Protopappas et al. 2008). This study represents a systematic and quantitative investigation of the use of ultrasonic guided waves to evaluate bone fractures. Although only transverse notches were considered in the numerical simulations and experiments on plate phantoms and *in vitro* ovine tibias, the simplified model yielded interesting findings, including the observation of converted modes and energy variation with notch depth. These findings can also be used in investigations of other behaviors and the underlying mechanisms, for instance, transmission through oblique fractures (Dodd et al. 2008) and signal energy recovery during fracture healing (Barbieri et al. 2011; Protopappas et al. 2006, 2007). In addition, the establishment of energy criteria for the converted modes illustrates the potential of quantitative prediction of cortical fracture status on the basis of ultrasound guided mode conversion.

According to traditional FAS analysis of the soft tissue-covered long bone, wideband and even high-frequency excitation can also be used to measure the apparent ultrasound velocity and further assess bone quality (Barbieri et al. 2011; Dodd et al. 2007, 2008; Gerlanc et al.

1975; Protopappas et al. 2006). Theoretically, in thick cortical bone or at high frequency (thickness $\gg$ longitudinal wavelength), the FAS corresponds to the non-dispersive lateral wave or head wave, propagating at longitudinal velocity along the soft tissue-bone surface, whereas at thin cortical thickness or low frequency (thickness $\ll$ wavelength), guided wave theory can successfully be employed to interpret the FAS and propagation of slower energetic waves (Bossy et al. 2002; Laugier and Haïat 2011; Moilanen 2008; Nicholson et al. 2002; Tatarinov and Sarvazyan 2003). Hence, low-frequency excitation can guarantee that the wavelength is much larger than the cortical thickness and satisfy the excitation conditions of low-order guided modes. Furthermore, in the aim of suppressing broadband dispersion of the guided modes, we adopted four-cycle Gaussian envelope sinusoids as excitations (Alleyne and Cawley 1992). Wave packets of the original and converted components of fundamental modes S0 and A0 can thus be correctly distinguished and extracted for later quantitative analysis (Xu et al. 2014b).

As illustrated in Figure 3, the EPs of the converted modes are highly related to notch depth variation. The simulations illustrate mainly two trends in ultrasonic transmission caused by notch aggravation. First, because of ultrasound scattering and acoustic impedance mismatch at the fracture site, fracture deepening enhances the mode conversion and transmitted energy loss. Second, at fracture percentages $< 70\%$, enhancement of mode conversion is able to overwhelm the transmitted loss and enlarge the absolute amplitude of the converted modes, but at fracture percentages $> 70\%$, the converted amplitudes begin to decrease. Therefore, the proportional parameter EP is more suitable for describing the degree of mode conversion than the absolute amplitude or attenuation.

On the basis of the frequency-sweeping method, mode conversion phenomena, including velocities and energy percentages, were further investigated in simulations and experiments over a wide $f \cdot d$ range (100–1100 kHz•mm) (Figs. 4 and 5). Figure 7 illustrates the linear relationship between EP and fracture percentage that was observed in the studies of both plate phantoms and ovine tibias. Compared with the results at higher $f \cdot d$, larger RMSEs of the velocity determination are found at low $f \cdot d$ (Figs. 4a and 5b). The main reason is that at low frequency, the relatively long duration of mode packets impairs time resolution, and the relatively high dispersion of mode A0 also affects the velocity measurements. The phenomenon denotes that sensitive frequencies are important for evaluation of fractured bone with ultrasonic guided waves, and selection of the excitation frequency is requires further study. In the ovine tibia experiments, a fixed excitation of 250 kHz was selected. Hence, the results for ovine tibias (Fig. 6c) featured smaller EP error bars compared with the frequency-sweeping results (Fig. 5c).

As outlined in eqns (1) and (2), the converted mode velocities are affected by mainly two factors, propagation distance and notch position. For consistency in the comparison, the transverse gap breaks were all artificially made in the middle shaft. All these samples were examined axially with the fractured gap exactly in the middle of the transducer couple. Clinically, in some fractures, which are located close to the epiphysis rather than in the mid-diaphysis, the fracture-centered arrangement of the transducer couple may actually cause difficulties in measurement. However, if the fracture gaps are not in the middle of the transducers, the converted modes, propagating through the identical pathway, but in opposite

axial directions, may have totally different characteristics (Xu et al. 2014b). Taking the converted $S0_{A0}$ as an example, according to eqn (1d), after propagating over a long distance x_2 , the A0-converted $S0_{A0}$ tends to split from the original A0 as an individual packet, whereas over a short distance x_2 , $S0_{A0}$ may still be close to or even overlap with the original A0. Similar trends in converted $A0_{S0}$ can also be expected. The underlying differences in the converted $S0_{A0}$ and $A0_{S0}$ between the two opposite-direction measurements may also be helpful in locating the fracture site and further assessing fracture status.

Notched plates were used as phantoms in this study. The main limitation of the model is that the notch width is fixed, which is different from the situation in a clinical cortical fracture. Furthermore, during fracture healing, recovery of the mechanical properties of connective tissue at the fracture site, for example, from the cartilage to the woven bone, could also influence the energy distribution of the guided modes. Theoretically speaking, the mode conversion is able to reveal fracture geometry (Alleyne and Cawley 1992) and may also provide other clinical information, which needs to be analyzed in future work.

Soft tissue was removed before measurements, and coupling effects were not discussed. Basically, because of the dominant out-of-plane displacements, the mode A0 easily leaks energy to the surrounding soft tissue, whereas the S0 mode has mostly in-plane displacement and, thus, is less attenuated than A0 (Chen et al. 2012). The presence of soft tissues has a significant effect on mode amplitude, but mode conversion phenomena still can be observed (Xu et al. 2014a).

According to Snell's law and source influence theory, the specific incident and receiving angles can be used to selectively excite and receive certain guided modes and also narrow the phase velocity spectra. Previous studies (Moilanen et al. 2006; Nicholson et al. 2002) illustrated that A0 mode is stronger than S0 mode by using contact transducers, that is, parallel to the interface. In this study, to receive close magnitudes of modes S0 and A0 (Figs. 3a, 5a and 6a), a couple of Perspex wedges with the incident angle of 45° were used. A theoretical explanation and some examples of selective mode excitation and incident angle can be found in Li et al. (2001), Luo et al. (2004) and Ta et al. (2009).

CONCLUSIONS

This study found that the mode conversion theory of guided waves can be used to interpret ultrasound propagation in long cortical bones with partial diaphyseal fractures. Results of numerical simulations and phantom experiments illustrated that EPs of the converted modes monotonically increase with increasing notch depth. The utility of EP was further confirmed in *in vitro* experiments on nine ovine tibias with partial transverse gap breaks. The finding and interpretation of mode conversion at the fracture sites and the establishment of energy criteria may indeed contribute to ultrasonic assessment of long cortical bone fractures. Future work should focus on *in vivo* verification of mode conversion in fractured long bones, particularly clinical trials assessing ultrasonic monitoring of long bone fracture healing.

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Fig. 1.

Schematic representation of the mode conversion that occurred at the fracture site, where T and R represent the transmitter and receiver. w and h are the width and depth of the transverse notch, respectively. d is cortical thickness. The distance from the center of the left wedge to the notch is x_1 , and the distance from the notch to the right wedge center is x_2 . The original guided modes (S0 and A0) are excited by the transmitter. The mode conversion of the guided waves occurs as the waves encounter the notch in the cortical bone, leading to the newly converted modes ($S0_{A0}$ and $A0_{S0}$), which can be detected by the receiver together with the original modes S0 and A0.

Fig. 2.
Experimental setup.

Fig. 3. Simulated results for plate models under 500-kHz excitation with fracture percentage ranging from 10% to 90%. (a) Comparison of guided signals. (b) Energy percentages of the original guided modes S0 and A0 and the converted wave packet ($S0_{A0} + A0_{S0}$). The simulated energy percentages of S0, A0 and $S0_{A0} + A0_{S0}$ are denoted by *circles*, *diamonds* and *triangles*, respectively.

Fig. 4. Simulated results for the intact model (0%) and fractured models (10%–90%) with incident frequency sweeping from 100 to 1100 kHz. (a) Modal velocities versus $f \cdot d$, where the lines are the theoretical dispersion curves. (b) Modal energy percentage versus fracture percentage. The simulated velocities and energy percentages of S0, A0 and S0A0 + A0S0 are denoted by *circles*, *diamonds* and *triangles*, respectively.

Fig. 5.

Experimental results for the steel phantoms with respect to fracture percentages from 0% to 80%. (a) Example of guided signals measured on the intact phantom (0%) and fractured phantoms (20%–80%), under 500-kHz excitation. Frequency-sweeping results from 200 to 1100 kHz. (b) Modal velocity versus $f \cdot d$, where the lines are the theoretical dispersion curves. (c) Modal energy percentage versus fracture percentage. The measured velocities and energy percentages of S0, A0 and S0A0 + A0S0 are denoted by *circles*, *diamonds* and *triangles*, respectively.

Fig. 6.

Results of experiments on nine ovine tibias in which the fracture percentage ranged from 0% to 80%, under 250-kHz four-cycle Gaussian excitation. (a) Typical signals measured in specimen 1. (b) Modal velocity versus $f \cdot d$, where the lines are the theoretical dispersion curves. (c) Modal energy percentage versus fracture percentage. The measured velocities and energy percentages of S0, A0 and S0A0 + A0S0 are denoted by *circles*, *diamonds* and *triangles*, respectively.

Fig. 7. Energy percentage of the original and converted modes as a function of increasing fracture depth. (a) Results for the plate phantoms. (b) Results for the nine ovine tibias. *Circles* and *triangles* denote the energy percentages of the original and converted modes, respectively.

Table 1

Cortical thickness and mean diameter of ovine tibias used in this study

	Bone								
	1	2	3	4	5	6	7	8	9
Thickness (mm)	2.10 ± 0.07	2.44 ± 0.04	2.48 ± 0.06	2.64 ± 0.06	2.76 ± 0.03	3.01 ± 0.10	3.32 ± 0.07	3.40 ± 0.17	3.88 ± 0.11
Diameter (mm)	12.15 ± 0.56	15.27 ± 0.26	15.83 ± 0.51	18.07 ± 0.14	15.09 ± 0.15	18.48 ± 0.19	14.17 ± 0.17	16.73 ± 1.22	20.31 ± 1.75