Ultrasonics xxx (2013) xxx–xxx

Contents lists available at ScienceDirect

Ultrasonics

journal homepage: www.elsevier.com/locate/ultras

On ultrasound waves guided by bones with coupled soft tissues: A mechanism study and *in vitro* calibration

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A R T I C L E I N F O

Article history:
Available online xxxx

Keywords:
Bone assessment
Coupling effect
Quantitative ultrasound (QUS)
Soft tissue
Ultrasound waves

A B S T R A C T

The influence of soft tissues coupled with cortical bones on precision of quantitative ultrasound (QUS) has been an issue in the clinical bone assessment in conjunction with the use of ultrasound. In this study, the effect arising from soft tissues on propagation characteristics of guided ultrasound waves in bones was investigated using tubular Sawbones phantoms covered with a layer of mimicked soft tissue of different thicknesses and elastic moduli, and an *in vitro* porcine femur in terms of the axial transmission measurement. Results revealed that presence of the soft tissues can exert significant influence on propagation of ultrasound waves in bones, leading to reduced propagation velocity and attenuated wave magnitude compared with the counterparts in a free bone in the absence of soft tissues. However such an effect is not phenomenally dependent on the variations in thickness and elastic modulus of the coupled soft tissues, making it possible to compensate for the coupling effect regardless of the difference in properties of the soft tissues. Based on an *in vitro* calibration, this study proposed quantitative compensation for the effect of soft tissues on ultrasound waves in bones, facilitating development of high-precision QUS.

1. Introduction

The increasing needs for monitoring the bone health status, for example diagnosis of osteoporosis, have entailed a number of quantitative bone assessment techniques, typified by quantitative ultrasound (QUS), X-ray computed tomography (CT) and magnetic resonant imaging (MRI) [1–3]. In particular, QUS has been deemed as a most promising candidate for quantitative bone evaluation, due to its competitive nature of non-radiation, ease of manipulation and cost-effectiveness [1]. With the application of various measurement configurations, the ultrasonic waves can be injected into the bone structure and captured after they propagate either axially along the bone axis (i.e., axial transmission (AT)), or circumferentially across the bone cross-section (i.e., transverse transmission), or in the bone thickness in a reflection manner (i.e., pulse echo or backscattering) [1,4–12]. The bone properties can be evaluated in different respects with applications of different techniques, among which the AT technique remains most competitive, because it is capable to reflect not only the material properties of the bone, but the bone geometrical features [13–16]. With such a fascination, the AT-based QUS has gained a good reputation as promising for osteoporosis evaluation [15–18].

However, the prevalence of such a technique has been considerably undermined by the fact that the soft tissue covering the bone introduces unwanted disturbances and severe alterations to the propagation of ultrasonic waves in the bone, significantly preventing the AT-based QUS technique toward a clinical application of high precision and accuracy [17,19–25]. With such a concern, considerable efforts have been directed to developing novel methods to remove the influence of soft tissues. In this regard, Moilanen et al. [26] invented an axial transmission device with receiver shifted at a constant step during the measurement. With such an operation, a distance–time diagram was obtained, from which the wave propagation velocities can be determined without the interference from the overlying soft tissues. Bossy et al. [17] developed a bidirectional transmission technique using a probe consisting of two groups of emitters with a single group of receivers in between. The generated ultrasound waves travel along the bone in opposite directions. By taking into account time delays of waves propagating in opposite directions, influence arising from unequal thicknesses of the coupled soft tissues and the probe inclination can be compensated for. However, previous efforts considered the soft tissue as an addition layer to the bone that only provide extra wave propagation routines. The coupling effect on wave propagation in real bone structures has not been explored but is of great importance.

Our previous results demonstrated that a coupling layer (fluid or mimicked soft tissue) can significantly alter the wave

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0041-624X/$ - see front matter © 2013 Elsevier B.V. All rights reserved.
http://dx.doi.org/10.1016/j.ultras.2013.08.002

propagation characteristics in solid wave guides (i.e., metal or bone-mimicking plate) [19–21,27,28]. However, the coupling effect on wave propagation in real bone structures which is much different from plates has not been explored but of great significance. With such a concern, in this study, a series of tubular Sawbones samples covered with a layer of artificial silicon rubber (ASR) (serving as mimicked soft tissue and considered as *Tissue Equivalent Materials* (TEM)) varied in thickness and elastic modulus was ultrasonically interrogated at multiple frequencies, as well as an *in vitro* porcine femur with soft tissue but marrow removed. The propagation characteristics of the first arrival signal (FAS) and second arrival signal (SAS) in the soft tissue–bone mimicking phantoms and *in vitro* porcine femur were analyzed. This study further contributes to the understanding of the soft tissue coupling effect on the propagation of ultrasonic guided waves, paving the way for development of high-precision QUS techniques for clinical bone assessment.

2. Ultrasound waves in a coupled cylindrical medium

Ultrasonic wave propagation in soft tissue–bone-coupled (SBC) media can be simplified to wave propagation in a fluid–solid bilayer (FSB) for a first level approximation, by regarding the soft tissue as fluid [17,23,26,29]. Here, the analytical description of wave propagation in the coupled media, in particular in the tubular structure, is recalled, treating the bone as a sort of tubular structure.

First, considering a homogeneous, isotropic and elastic medium, the equation of particulate motion in the medium can be expressed as [30]

$$\mu \nabla^2 u + (\lambda + \mu)\nabla(\nabla \cdot u) = \rho \frac{\partial^2 u}{\partial t^2}, \tag{1}$$

where u, ρ, λ and μ are the displacement field, density and the two Lamé constants of the material, respectively. In a FSB as illustrated in Fig. 1, the displacement (u) in either the solid or fluid part can be decomposed as, according to the Helmholtz decomposition [31],

$$u = -\nabla \Phi + \nabla \times \Psi, \tag{2}$$

where Φ is the scalar potential, and Ψ the vector potential. Therein, the displacement in a solid cylinder can be decomposed by its corresponding scalar potential (Φ^S) and vector potential (Ψ^S) as [23]

$$\Phi^S = [A_1 J_n(\alpha r) + A_2 Y_n(\alpha r)] \cdot \cos(n\theta) \cdot e^{i(k_z z - \omega t)}, \tag{3a}$$

$$\Psi_r^S = [B_1 J_{n+1}(\beta r) + B_2 Y_{n+1}(\beta r)] \cdot \cos(n\theta) \cdot e^{i(k_z z - \omega t)}, \tag{3b}$$

$$\Psi_\theta^S = -[B_1 J_{n+1}(\beta r) + B_2 Y_{n+1}(\beta r)] \cdot \cos(n\theta) \cdot e^{i(k_z z - \omega t)}, \tag{3c}$$

$$\Psi_z^S = [C_1 J_{n+1}(\beta r) C_2 Y_{n+1}(\beta r)] \cdot \sin(n\theta) \cdot e^{i(k_z z - \omega t)}, \tag{3d}$$

As fluid is unable to sustain shear stresses, the vector potential of displacement in fluid remains zero. As a result, the displacement in fluid can only be express by the scalar potential, namely [19]

$$\Phi^F = [D_1 J_n(\alpha^F r) + D_2 Y_n(\alpha^F r)] \cdot \cos(n\theta) \cdot e^{i(k_z z - \omega t)} \tag{4}$$

In Eqs. (3) and (4), $\alpha^2 = \omega^2/C_L^2 - k^2$, $\beta^2 = \omega^2/C_T^2 - k^2$, $\alpha^{F^2} = \omega^2/C_F^2 - k^2$. J_n and Y_n are Bessel functions of the order n. ω, k, C_L, C_T, C_F are the angular frequency, wavenumber, longitudinal wave velocity in solid, transverse wave velocity in solid and longitudinal wave velocity in fluid, respectively. Note that k_Z is the wavenumber in dimension Z, while k is the wavenumber of dimensionless.

At the interface of the fluid and solid, only normal components of the displacement and stress are continuous, while the continuity of the shear components never holds. The boundary conditions are [32]

$$\begin{aligned} &\sigma_{rr} = \sigma_{r\theta} = \sigma_{rz} = 0, \quad \text{at } r = a \\ &u_r = u_r^F,\ \sigma_{rr} = \sigma_{rr}^F,\ \sigma_{r\theta} = \sigma_{rz} = 0, \quad \text{at } r = a + h_s \\ &\sigma_r^F = 0 \quad \text{at } r = a + h_s + h_F \end{aligned} \tag{5}$$

where σ_{rr}, $\sigma_{r\theta}$, σ_{rz} are the three stress components in the cylindrical coordinate. u_r, σ_{rr} and $\sigma_{r\theta}$ are the radial (or normal) component of displacement and stress, circumferential component of stress in the solid, respectively. u_r^F, σ_r^F and σ_{rz} are the radial (or normal) component of displacement and stress, circumferential component of stress in the fluid, respectively. a, h_S and h_F are the inner radius, thickness of the solid cylinder and thickness of the fluid layer, respectively, as indicated in Fig. 1. Combining the boundary conditions (i.e., Eq. (5)) together with the governing wave equations (i.e., Eqs. (3) and (4)), it yields the characteristic equation of ultrasonic wave propagating in the FSB, i.e., the determinant of the coefficient matrix consisting of $A_1, A_2, \ldots, D_2$ in Eqs. (3) and (4), (more details can be referred to elsewhere [19])

$$|M = 0|. \tag{6}$$

Based on Eq. (6), Fig. 2 plots the dispersion curves of cylindrical Lamb waves in a cortical bone cylinder (inner radius: 4 mm; wall thickness: 3 mm; material properties are shown in Table 1) in the absence and presence of a layer of fluid (thickness: 1 mm, as listed Table 1), to find that the features of the guided modes in the bone cylinder coated with a layer of fluid behave much

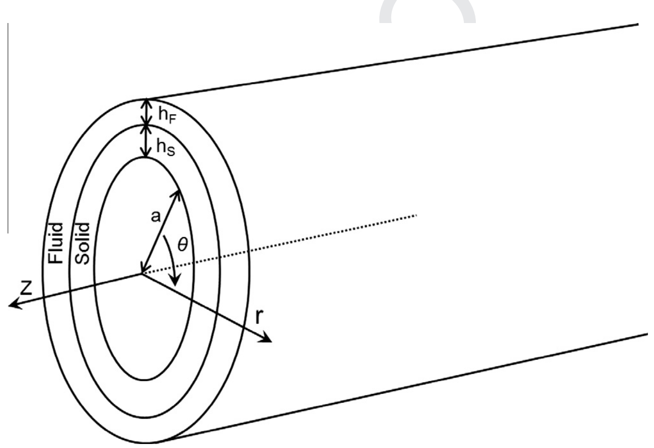

Fig. 1. A hollow cylinder covered with a layer of fluid of an infinite extent in z-direction and a finite thickness in the cylindrical coordinates (a: inner radius of the cylinder, h_S: thickness of the cylinder, h_F: thickness of the fluid layer).

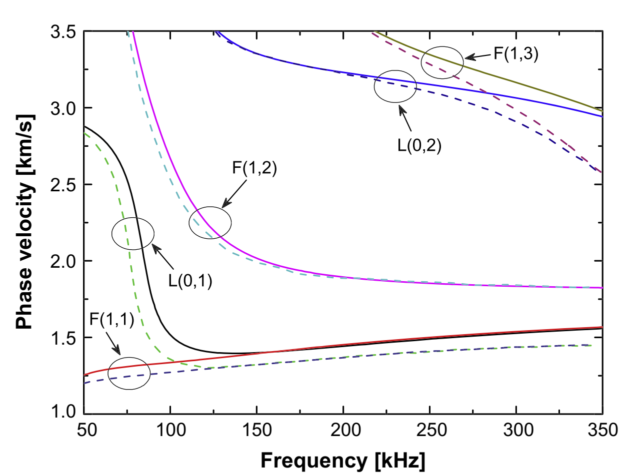

Fig. 2. Dispersion curve of cylindrical Lamb waves in (a) a free bone tube (solid lines) and (b) a bone tube covered with a layer of fluid (thickness:1 mm) (dash lines).

Please cite this article in press as: J. Chen, Z. Su, On ultrasound waves guided by bones with coupled soft tissues: A mechanism study and *in vitro* calibration, Ultrasonics (2013), http://dx.doi.org/10.1016/j.ultras.2013.08.002