Ultrasonic non destructive characterization of trabecular bone: estimation of the propagation velocity and attenuation

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Abstract. The non destructive characterization of porous structures with ultrasonic waves allows determining the propagation velocities and the attenuation for diagnosis of diseased bone (e.g., osteoporosis) by establishing correlations between ultrasonic parameters and their mineral density. Two compressional modes have been identified independently in bovine trabecular bone, a fast wave and a slow wave. The principal objective of this paper is to characterize the propagation velocity and ultrasonic attenuation as functions of frequency and porosity of bovine cancellous bone. The porosity of the used samples varies between 40 % and 75 %. A transmission technique is used. This method only requires the measurement of the specimen's thickness and recording of two pulses: one without and one with the specimen inserted between the transmitting and receiving transducers. From the two pulses, the attenuation can be determined using spectral analysis. The attenuation coefficient increases nonlinearly over the frequency from 200 to 700 kHz. The experimental results show a strong correlation between the bone density, the measured propagation velocity and the attenuation. The measurement of these velocities allows determining the bone elastic parameters. This study confirms the sensitivity of the ultrasonic propagation velocity to the change of bone porosity. The potential of ultrasound in bone tissue characterization seems to provide interesting results and would lead to predict bone pathology and particularly permit better diagnosis of bone fragility.

1 Introduction

The ultrasound field applied to the exploration of bone tissue and other biological organs is a promising alternative to the use of X-rays. Ultrasonic tissue characterization has become an essential method for clinical diagnosis of diseased bone (e.g., osteoporosis) [1]. The ultrasonic diagnostic methods of bone tissue concern both ultrasound image reconstruction and interpretation of acoustic parameters. Measurement of ultrasonic attenuation and velocity in cancellous bone are being applied to detect, identify and locate this disease. Bone with a low volume fraction of solid (less than 70%) is called cancellous or trabecular bone. Bone with above 70% solid is called cortical bone. The trabecular bone is on the acoustic level, a material much more complex than cortical bone. It is anisotropic, composite, porous and heterogeneous. Many theoretical approaches have been proposed for the analysis of the ultrasound interaction with the trabecular structure such as the poroelastic approaches (Biot [2, 3]), stratified [4] or scattering by heterogeneous medium models ([5, 6]). The propagation of elastic waves in porous media saturated by a fluid is described by the Biot theory. This theory has been developed and adapted to the description of the ultrasonic waves propagation in bone (trabecular bone) by several authors, in particular Hosokawa and Otani [7, 8], Hair and Langton [9], Fellah and al. [10]. Biot theory predicts the existence of two longitudinal waves (the slow wave and the fast wave) and a transverse wave. The "fast wave" corresponds to the wave in the solid and fluid moving in phase. The "slow wave" corresponds to their motion in phase opposition. These have been identified experimentally in the trabecular bovine bone [7].

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2 Biot theory applied to cancellous bone

The Biot theory was developed to predict the acoustical properties of fluid saturated porous rocks in the context of geophysical testing but has been used extensively to describe the wave motion in trabecular (cancellous) bone [10]. Biot model describes the acoustic wave propagation in the porous medium. It consists of a superposition of two coupled continuous media (fluid and solid). This is a poroelastic medium, i.e. a solid with connected cells in which a fluid circulates freely. The trabecular bone is considered as a porous biphasic medium, consisting of two parts: a solid phase (the skeleton) and a fluid phase (the bone marrow). This model is based on several assumptions (the skeleton is a viscoelastic medium, the fluid phase is continuous and the material is considered as homogeneous and isotropic).

A biphasic porous material submitted to a mechanical constraint is deformed elastically. The displacements u of the solid and U of the fluid obey to the propagation equations:

$$\bar{\rho}_{11}\frac{\partial^2 u}{\partial t^2} + \bar{\rho}_{12}\frac{\partial^2 U}{\partial t^2} = P\,grad(di\ddot{v}.u) + Q\,grad(div\,U) - N.rot(rot\,u)$$
$$\bar{\rho}_{12}\frac{\partial^2 u}{\partial t^2} + \bar{\rho}_{22}\frac{\partial^2 U}{\partial t^2} = Q\,grad(div.u) + R\,grad(div\,U) \qquad (1)$$

Biot and Willis [11] showed that the elasticity coefficients P, Q and R of the material which appear in the Eq. (1) of the movement are expressed through the relations:

$$P = \left(\frac{(1-\phi)(1-\phi-\frac{K_b}{K_S})K_S + \phi\frac{K_S}{K_f}K_b}{1-\phi-\frac{K_b}{K_S}+\phi\frac{K_S}{K_f}}\right) + \frac{4}{3}N, \qquad (1)$$

$$Q = \frac{(1-\phi-\frac{K_b}{k_s})\cdot\phi.K_s}{1-\phi-\frac{K_b}{K_s}+\phi\frac{K_S}{K_f}}, \quad R = \frac{\phi^2 K_s}{1-\phi-\frac{K_b}{K_s}+\phi\frac{K_s}{K_f}} \qquad (2)$$

where ϕ is the porosity and K_f, K_s, K_b are respectively the bulk modulus of the fluid, of the elastic solid and of the porous skeletal frame. N is the shear modulus of the composite as well as that of the skeletal frame. These parameters depend on the Young's modulus and Poisson's ratios of the solid and the skeleton, respectively, which are represented by the pairs (E_s, ν_s) and (E_b, ν_b).

$\bar{\rho}_{mn}$ are the "mass coefficients" which are related to the densities of solid ρ_s, fluid ρ_f and dynamic tortuosity $\alpha(\omega)$ by:

$$\bar{\rho}_{11} = (1-\phi)\rho_s - \bar{\rho}_{12}, \quad \bar{\rho}_{12} = -\phi\,\rho_f\,[\alpha(\omega)-1] \quad \text{and}$$
$$\bar{\rho}_{22} = \phi\,\rho_f - \bar{\rho}_{12}. \qquad (4)$$

The dynamic tortuosity $\alpha(\omega)$ is given by [12]:

$$\alpha(\omega) = \alpha_\infty\left[1 + \frac{\eta\phi}{j\omega\alpha_\infty\rho_f k_0}\sqrt{1 + \frac{4\alpha_\infty^2 k_0^2 \rho_f \omega}{\eta\Lambda^2\phi^2}}\right] \qquad (5)$$

where α_∞ is the geometric tortuosity and Λ the viscous characteristic length. This length is related to the size of the pore, the porosity ϕ and the viscosity η. It represents the scale where the phenomena of viscous dissipation occur.

The differential equations (1) are coupled via the coefficients $\bar{\rho}_{12}$ and Q, which represent respectively the inertial and the elastic coupling. The viscous coupling is integrated into the complex apparent densities ($\bar{\rho}_{11}$, $\bar{\rho}_{12}$, $\bar{\rho}_{22}$). The resolution of this equations system leads to two longitudinal velocities of propagation. One corresponds to the wave of the first kind (or fast wave),

the other to the wave of the second kind (or slow wave). For a harmonic plane wave of angular frequency ω propagating through the porous medium and at normal incidence, the following longitudinal velocities can be obtained:

$$V_{slow,fast}^2 = \frac{2(PR-Q^2)}{\Delta \pm \left(\Delta^2 - 4(PR-Q^2)(\bar{\rho}_{11}\bar{\rho}_{22} - \bar{\rho}_{12}^2)\right)^{0.5}} \qquad (6)$$

with $\Delta = P\bar{\rho}_{22} - R\bar{\rho}_{11} + 2Q\bar{\rho}_{12}$.

The signs $\pm$ in the denominator mean that v^2_{fast} will be obtained when "-" is selected, and v^2_{slow} will be obtained when "+" is selected.

The simulation parameters of the Biot model for a bovine trabecular bone are listed in Table (1). We consider two types of fluid saturating the porous material: water and bone marrow.

Table 1: Structural and acoustic parameters of bovine cancellous bone [7].

Young's modulus of solid bone : E_s	22 GPa
Poisson's ratio of solid bone: ν_s	0.32
Poisson's ratio of skeletal frame: ν_b	0.32
Solid density: ρ_s	1960 kg/m³
Compressibility modulus of the fluid : κ_f	
marrow	2.0 GPa
water	2.28 GPa
Fluid viscosity : η_f	
marrow	1.5 Pa.s
water	0.001 Pa.s
Fluid density: ρ_f	
marrow	930 kg/m³
water	10^3 kg/m³
Pore size: a	0.5 mm

Figure (1) illustrates the variation of the slow and fast longitudinal velocities depending on porosity through a porous bone saturated with marrow. In this figure we notice a substantial decrease in the velocity of the fast wave with porosity and a less pronounced increase in the velocity of the slow wave when the porosity increases. The velocity of the fast wave decreases from 4000 m /s, for porosity close to zero, to a velocity of 1400 m / s for a maximum porosity (ϕ equal to 1).

For the same porosity range, the speed of the slow wave increases from 0 to 1400 m/s. However, it remains close to 1400 m/s for almost the entire range of porosity considered. In Figures (2) and (3), we have respectively represented the velocity variations of the fast and of the low longitudinal wave for cancellous bone saturated with marrow and with water.

The fluid (water or marrow) has very little influence on the variation of the speed. This is explained by the fact that the velocity propagation in the marrow is
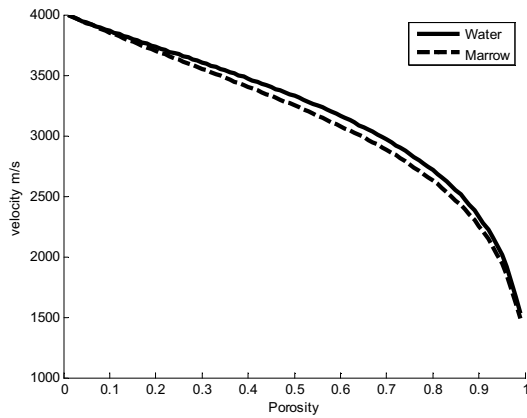
Fig 1. Variation of the fast longitudinal velocity versus porosity for a porous bone saturated with:
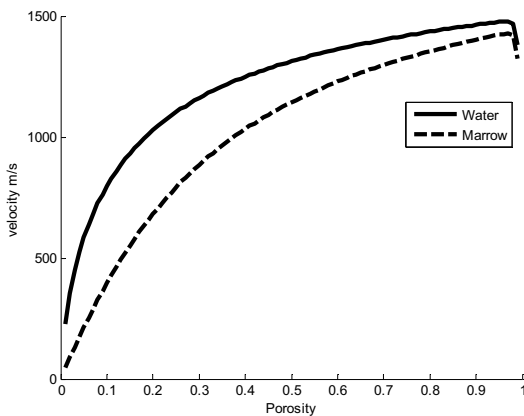---- marrow and __water.

Fig2. Variation of the slow longitudinal velocity versus porosity for a porous bone saturated with:
---- marrow and __water.

substantially equal to that of water. The densities of the two fluids are also similar. About 75% of red bone marrow is water; the rest being solid matter consisting of connective tissue, blood vessels and cells.

3. Experimental Setup

A number of measurements were performed on samples of bone from the proximal of bovine femurs. To observe the effect of changes in bone porosity, the specimens were taken in the range porosity 40% - 60%. The densities of these specimens were measured using Archimedes's principles. The thicknesses of the samples varied from 4 to 16 mm. Thickness between the two parallel planar surfaces were measured using a digital calliper. Each specimen was saturated with water followed by decompression to remove air bubbles. Ultrasonic measurements were performed in distilled water at room temperature (20°C). Samples were suspended by a clamp and placed half-way between the two transducers at normal incidence. The specimens were aligned parallel to the transducer faces by maximizing the

amplitude of the signal reflected from the planar front surface of the specimen.

We have used a pair of unfocused broadband transducers with a diameter of 19.68 mm and a nominal frequency of 500 kHz (Olympus V318-SU). We have used a transmission technique. Two signals were recorded, one without the bone specimen in the acoustic path (reference signal) and the other with it.

To measure the group velocity, arrival times of received broadband pulses were measured with and without the sample in the water path. Group velocity, V_g, was computed from:

$$V_g = \frac{V_w}{1 + \frac{V_w \Delta t}{d}} \qquad (7)$$

where Δt is the delay time between the traveling times of the reference signal and the signal with specimen, d is the sample thickness and V_w is the speed of sound in distilled water.

The attenuation $\alpha(f)$ was determined by using the same acquired signals. $\alpha(f)$ (Np/cm) at each frequency was calculated using the expression below:

$$\alpha(f) = \frac{1}{d} \ln\left(\frac{A_{ref}(f)}{A_S(f)}\right), \qquad (8)$$

where $A_{ref}(f)$ and $A_s(f)$ are the amplitude spectrum of the reference and sample signals respectively.

3. Results and discussion

A number of measurements were performed on samples of bovine bone. From heads of the femur bone samples of cylindrical or parallelepipedic shape of variable thickness and porosity were cut and used.

After having estimated the thickness and porosity of the bone sample, two recordings of two pulses were performed: one without the specimen inserted between the transmitting and receiving transducers and one with it.

The temporal signals acquired with one of the bone samples are shown in Figure (3). The transmitted waves, which appear on the temporal signal of this figure, exhibit the earlier arrival time due to the faster speed of sound in the specimen bone compared to that in water. Figure (4) shows the amplitude spectrum corresponding to the signals of Figure (3).

From the two temporal signals, the attenuation can be determined using Fourier analysis. Figure (5) shows the experimental attenuation coefficients of the fast wave for three specimens of porosity ø = 0.604, 0.598 and 0.457 in the direction perpendicular to the trabeculae as functions of the frequency. Attenuation was measured in the frequency range 200-650 kHz.

The attenuation increases with the frequency up to 650 kHz. It is more important for the low porosity in the frequency range 200-500 kHz. Its values depend strongly on the porosity of the samples and therefore depend on their internal structure.
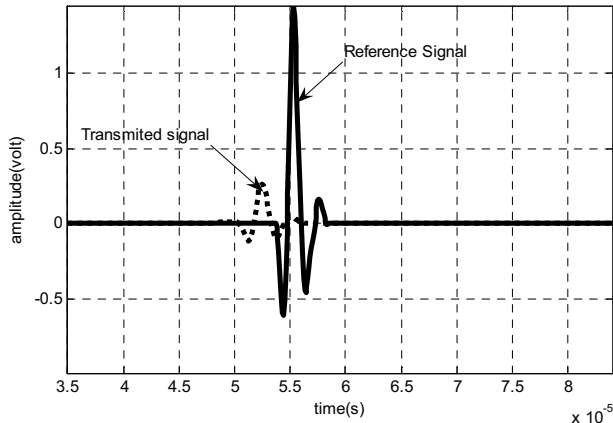
Fig 3. Reference signal through pure water and a typical transmitted signal through a bone specimen at a frequency of 500 kHz.
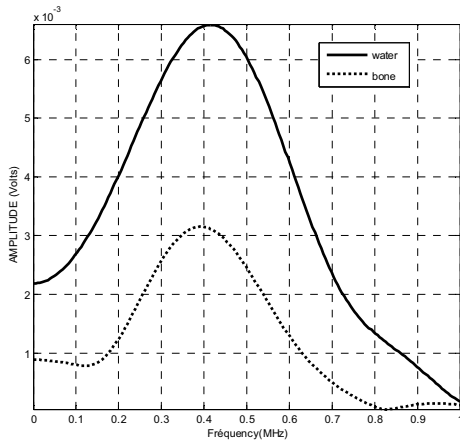

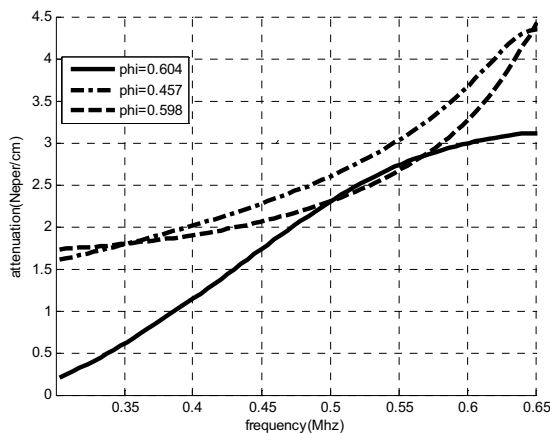**Fig 4.** Amplitude spectrums corresponding to the signals of Figure (3).

Fig 5. Attenuation of fast wave in cancellous bone at three porosities: ø =0.604, 0.598, and 0.457 as a function of frequency.

The ultrasonic velocity of the fast wave as function of porosity is illustrated in Figure (6). Error bars represent the standard deviation from several measurements. The speed of the fast wave varies from 2690 to 2310 m/s as the bone porosity increases. This variation is consistent with the theoretical curve achieved by using the Biot model. The velocity decreases with increasing porosity.
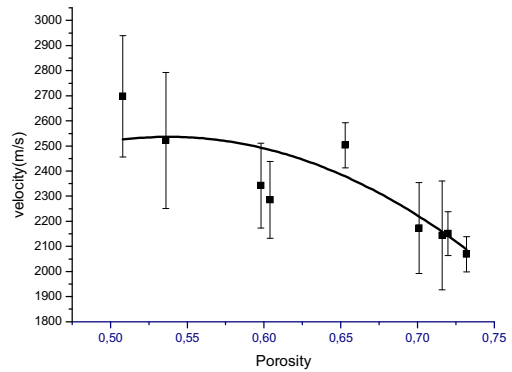

Fig 6. Variation of the fast longitudinal velocity versus porosity.

4. Conclusion

The present study shows good agreement between the measured and the calculated propagation speeds in bovine cancellous bone with the Biot theory. The experimental measurement shows a nonlinear dependence of the attenuation on the frequency and on the porosity. Indeed, according to the Biot theory the attenuation is essentially due to the viscosity of fluid while, the experimental measurement includes other mechanisms such as the reflection, scattering, absorption, mode conversion and diffraction. The attenuation increases with porosity in the region of interest. A cancellous bone shows also a decrease of the propagation velocity of the fast longitudinal wave with porosity as predicted by Biot model. These results suggest that the wave propagation is strongly influenced by the heterogeneity of the bone. Therefore a more elaborated theoretical model should be proposed to take into account all the physical phenomena that occur during the ultrasonic wave propagation through the trabecular bone.

References

1. P. Laugier, *Joint Bone Spine* **73**, 125 (2006).
2. M. Biot, *J. Acoust. Soc. Am.* **28**, 168 (1956)
3. M. Biot, *J. Acoust. Soc. Am* **28**, 182(1956).
4. M. Schoenberg, *Wave Motion* **6**, 303 (1984).
5. K.A. Wear, *J Acoust Soc Am.* **106**, 3659 (1999)
6. J. Jenson**,** F. Padilla**,** and P. Laugier*, Ultrasound Med. Biol.* **29 (3)**, 455 (2003).
7. A. Hosokawa and T. Otani, *J. Acoust. Soc. Am* **101**, 558.(1997).
8. A. Hosoakawa and T.Otani, *J. Acoust. Soc .Am* **103**, 2722 (1998).
9. T.J. Haire and C.M. Langton, *Bone* **24**, 291 (1999).
10. Z.E.A. Fellah , J.Y Chapelon, S.Berger, W. Lauriks, and C.Depollier ,*J Acoust Soc Am* 1**16**, 61. (2004).
11. Biot and G.Willis, *J.Appl.Mech* **24**, 594 (1957).
12. D.L. Johnson, J. Koplik, and R. Dashen. *Ultrasound Med. Biol.* **29 (3)**, 455 (2003).