The Measurement of Broadband Ultrasonic Attenuation in Cancellous Bone—A Review of the Science and Technology

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(Invited Paper)

***Abstract*—The measurement of broadband ultrasonic attenuation (BUA) in cancellous bone at the calcaneus was first described in 1984. The assessment of osteoporosis by BUA has recently been recognized by Universities UK, within its EurekaUK book, as being one of the "100 discoveries and developments in UK Universities that have changed the world" over the past 50 years, covering the whole academic spectrum from the arts and humanities to science and technology. Indeed, BUA technique has been clinically validated and is utilized worldwide, with at least seven commercial systems providing calcaneal BUA measurement. However, a fundamental understanding of the dependence of BUA upon the material and structural properties of cancellous bone is still lacking. This review aims to provide a science- and technology-orientated perspective on the application of BUA to the medical disease of osteoporosis.**

I. Cancellous Bone and Osteoporosis

By definition, cancellous bone has a minimum porosity of 30%. It consists of a complex open-celled porous framework of rod- and plate-like trabeculae perfused with bone marrow, and serves primarily as a biomechanical "shock-absorber" and a focus of high metabolic activity. It is found near the joint surfaces of long bones and within irregular bones such as the spinal vertebrae and calcaneus. The density of our skeleton increases from birth, reaching a maximum between the third and fourth decades, and thereafter gradually decreasing with advancing age. Osteoporosis has been defined as "a decrease in bone mass and architectural deterioration of bone tissue, leading to enhanced bone fragility and consequent increase in fracture risk" [1].

Osteoporosis results from a negative remodelling imbalance creating a loss of bone tissue and structural integrity. The predominant factor for osteoporosis is the female menopause, with additional "secondary" causes including rheumatoid arthritis, renal osteo-dystrophy, and steroidal therapy [2]. It is clinically manifested in the form of fractures, predominantly of the wrist, spine, and hip, typically occurring in female subjects from the ages of 50, 60, and 70 years, respectively. The current annual cost of osteoporosis to the UK Department of Health has been estimated to be £1 billion [3], and is steadily rising due to the increasing number of elderly subjects in the population. Associated with this, 25% of subjects suffering a hip fracture die within 12 months, and 50% of those remaining never regain full independent living [4]. Osteoporosis is often called the "silent disease" because bone loss occurs quietly without symptoms until the subject suffers a fracture. One in three women and one in five men over 50 will suffer a fracture due to osteoporosis; this increases to one in two women and one in three men over 60. It has been projected that there will be a four-fold increase in hip fracture incidence by 2050 [5].

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II. Broadband Ultrasonic Assessment of the Calcaneus

The measurement of broadband ultrasound attenuation (BUA) through the cancellous bone of the human calcaneus for the assessment of osteoporosis has recently been recognized by Universities UK as being one of the "100 discoveries and developments in UK Universities that have changed the world." Indeed, BUA has been clinically validated in terms of prediction of hip fracture risk [6], and is utilized worldwide, with at least seven commercial systems providing BUA measurement. However, after 23 years since BUA was first described [7], a fundamental understanding of the dependence of BUA upon the material and structural properties of cancellous bone is still lacking.

The calcaneus in the heel is the most popular measurement site used for quantitative ultrasound for several reasons. The calcaneus is approximately 90% cancellous bone with a thin cortical shell (Fig. 1) and is easily accessible; it is further considered to reflect the mechanical environment experienced by the weight-bearing osteoporotic anatomical sites of the proximal femur and spine. The calcaneus has been reported to be the optimal bone mineral density (BMD) measurement site, in terms of clinical sensitivity and utility, for routine screening of perimenopausal women to predict the risk of any type of osteoporotic fracture [8], [9]. The posterior aspect of the calcaneus is measured in

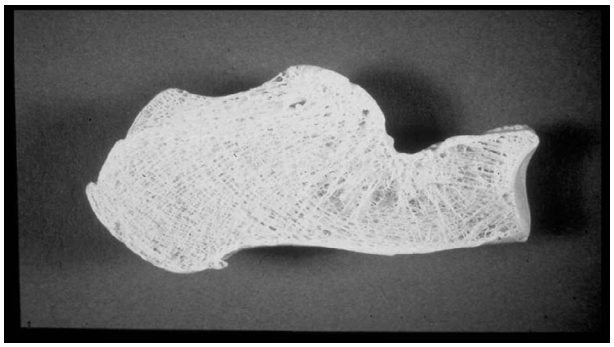

Fig. 1. Cross section of the human calcaneus illustrating the thin cortical shell and high proportion of cancellous bone.

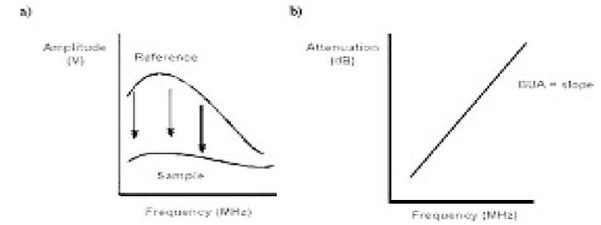

Fig. 2. Diagrammatic representation of BUA measurement, where (a) describes measurement of frequency spectra through a reference material (usually water) and the test sample. Attenuation is plotted in (b) against frequency, the regression slope being the BUA parameter.

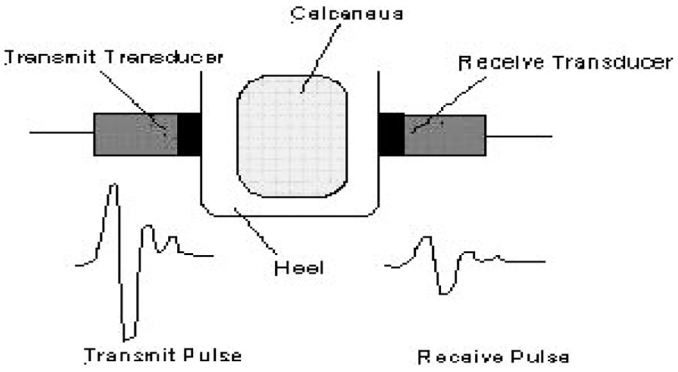

Fig. 3. The transmission technique illustrating the low-frequency filtering nature of cancellous bone to a propagating ultrasound signal.

the medio-lateral direction where the medio-lateral surfaces are approximately flat and parallel.

The frequency range of 0.1–1 MHz is the most useful for bone characterization by frequency-dependent ultrasound attenuation; below 0.1 MHz, attenuation is relatively insensitive to frequency, and above 1 MHz signal-to-noise becomes a significantly limiting factor.

The intensity of a plane wave propagating in a direction x decreases with distance as

$$I_x = I_o.e^{-\mu(f).x},$$

where I_o and I_x are the intensities incident and at a distance x (cm), respectively, and $\mu(f)$ is the frequency-dependent intensity attenuation coefficient (dB cm^{-1}). There are a number of attenuation processes involved in the propagation of ultrasound through bone, including absorption, scattering, reflection [10], diffraction [11], [12], mode conversion, and phase cancellation [13]; although attenuation may be readily measured experimentally, it is extremely difficult to predict or to transpose an attenuation value into material and structural parameters.

Attenuation is typically reported in decibels (dB), a logarithmic scale defined in terms of intensity, or, more generally, the measured signal voltage amplitude (A):

$$10.\log(I_1/I_2) \text{ for intensity (W m}^{-2}) \text{ or}$$
$$20.\log(A_1/A_2) \text{ for amplitude (volts).}$$

The total attenuation (μ) is approximately linearly proportional to frequency (f), given as $\mu(f) = \alpha.f$ where α is the slope of attenuation against frequency (dB MHz^{-1}cm^{-1}). In clinical practice, this has become known as broadband ultrasound attenuation (BUA). BUA is measured by recording the amplitude spectrum of an ultrasound pulse through a reference material $A_{\text{ref}}(f)$, chosen to be degassed water, and through the bone to be studied $A_{\text{bone}}(f)$, illustrated in Fig. 2. The attenuation (dB) at each frequency (f) is calculated from the amplitude through water and through sample, and plotted as a function of frequency between 0.2 and 0.6 MHz. The slope of this plot is defined as the BUA index, with units of dB MHz^{-1}. Dividing this by the sample width provides a volumetric parameter with units dB MHz^{-1} cm^{-1}. The specific frequency range of 0.2 to 0.6 MHz was chosen since it provides the greatest sensitivity to osteoporosis, i.e., greatest slope difference between healthy and osteoporotic subjects.

III. BUA Instrumentation

Until recently, due to the highly attenuating nature of cancellous bone, a transmission technique was adopted, whereby two transducers are utilized, one acting as transmitter, the other as receiver, being coaxially aligned (Fig. 3).

In the first system to report the measurement of calcaneal BUA [Fig. 4(a)], the transmitter was a spike generator and the receiver a computer-interfaced spectrum analyzer [7]. The first commercial calcaneal BUA system, the Osteosonics UBA1001 (Osteosonics, Doncaster, UK) [Fig. 4(b)], utilized a swept-frequency toneburst generator, the receiver being an RF-to-dc converter, a sample-and-hold circuit, and an analog-to-digital converter (ADC), both generator and receiver being computer-interfaced [14]. The first calcaneal BUA rectilinear scanner, the Walker Sonix UBA575 (Walker Sonics Inc., Worcester, MA), also utilized this measurement approach [Fig. 4(c)]. The first gel-coupled "dry" BUA system utilized a spike generator and ADC, the received time-domain digitized data being converted into a frequency spectrum via a fast Fourier transform [Fig. 4(d)–(e)]. This was subsequently commercialized as the McCue CUBA-

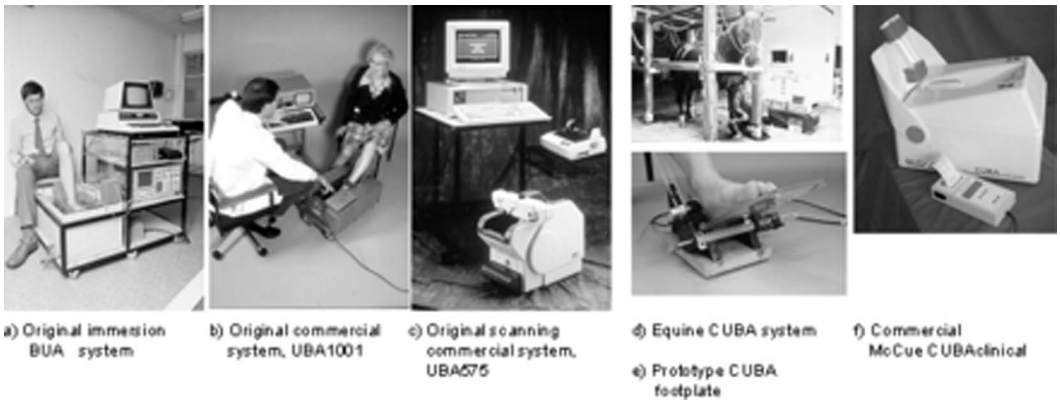

Fig. 4. Development of BUA systems for the assessment of osteoporosis: (a) Original immersion BUA system; (b) Original commercial system, UBA1001; (c) Original scanning commercial system, UBA575; (d) Equine CUBA system; (e) Prototype CUBA footplate; and (f) Commercial McCue CUBAClinical system.

Clinical (McCue Plc, Winchester, UK) [Fig. 4(f)] [15]. Most recently, a 2D detector array has been utilized [16] that has the potential to provide imaging of the calcaneus and hence improved positioning.

There are currently available a number of commercial calcaneal BUA systems, illustrated in Fig. 5. These instruments have significant differences between them, such as coupling method and scanner design, and including, for example, transducer size, shape, center frequency, and bandwidth, among other aspects. Thus the readings obtained on different systems vary significantly [17]. Instead of describing the individual systems, the following sections will describe the general features. The individual systems have been described by Njeh *et al.* [18].

Although pulse-echo attenuation coefficient techniques have been utilized for decades for the characterization of soft tissues [19], they have only recently been applied to bone through analysis of backscatter from the trabecular structure of the calcaneus [20], [21]. By applying a time-window to the received data, echoes originating from a particular depth from within the calcaneus may be analyzed by calculating BUA.

It is very important to apply a stringent quality analysis (QA) program when using BUA in bone status assessment, mainly because changes due to osteoporosis or its treatment are relatively small. Therefore, measurements of bone status changes have to be very precise because procedural errors, malfunctioning equipment, or erroneous data analysis may cause substantial interference and hence mis-diagnosis, even if the data are erroneous by only a few percentage points. Measuring the same subject on different systems is not recommended due to basic inherent machine differences, lack of an absolute ultrasound bone phantom, or lack of a universally accepted cross-calibration procedure, resulting in BUA variation between systems.

The degree of complexity of the QA program will depend on whether it is for an individual site or for a multicenter clinical trial. Most manufacturers provide system-specific phantoms. However, these system-specific phantoms are not anthropomorphic and their daily changes may not reflect what might happen *in vivo*. Non-manufacturer-produced quantitative ultrasound (QUS) phantoms are the Vancouver phantoms (University of British Columbia, Vancouver, BC, Canada), Leeds phantoms (Leeds Test Objects Ltd., North Yorkshire, UK), and CIRS phantom (CIRS, Inc., Norfolk, VA) [22], [23]. The Leeds and Vancouver phantoms have BUA values that fall within the biological range. Ultrasound measurement of these phantoms are highly influenced by temperature variations [23]; thus, if used in the QA program, the temperature of the phantom must also be monitored. An alternative QA approach incorporates an electronic circuit to simulate the "low-pass filter roll-off attenuation" behavior of ultrasound propagation through the calcaneus. Two commercial systems, the McCue CUBAClinical and the DMS UBIS (DMS, France) have adopted this approach [24].

IV. Sources of Error

There are a number of potential sources of error that are likely to affect BUA, including diffraction, interface losses, and phase cancellation. Historically, these have often been neglected in the context of clinical measurements, and existing evidence does suggest that their impact is limited. However, investigation of these effects is likely to be an important part of the search for improved accuracy and precision in clinical measurements. Diffraction errors associated with the immersion method are considered to be negligible since they incorporate fixed transducer separation and the heel velocity is close to that of water [11]. With a contact method, diffraction errors may potentially be significant because the transducer separation changes to accommodate different heel or sample thicknesses. Numerical diffraction corrections may be applied, however [25]. Theoretical work suggests that diffraction errors in BUA will be of the order of 0.6 and 10 dB MHz^{-1} for immersion and contact measurements at the heel, respectively [26]. Interface losses are generally assumed to be frequency independent, being negligible in cancellous bone samples ($\approx$ 0.5 dB) but appreciably higher (up to 20 dB) when

Fig. 5. Commercial devices incorporating BUA for the assessment of osteoporosis: (a) Hologic Sahara (US), (b) GE Achilles (US), (c) Med-Tec QUS2 (US), (d) Aloha AOS 100 (JP), (e) Osteometer DTU-one (DK), (f) Medlink Osteospace (FR), and (g) DMS UBIS-5000 (FR).

overlying cortical surfaces are present [27], [28]. In the human calcaneus, the frequency dependence of the interface losses has yet to be studied.

Another potential source of BUA measurement error is phase cancellation. It occurs in phase-sensitive receivers as used in single element piezoelectric transducers. It has recently been demonstrated *in vivo* that phase-sensitive BUA measurements are approximately 15 dB MHz^{-1} higher than corresponding phase-insensitive measurements [29]. The curved medial and lateral cortical surfaces of the calcaneus have been shown to create a phase cancellation artifact of the order of 3–5 dB $MHz^{-1}cm^{-1}$ [30]. The cortical plate itself may also serve as a sound wave modulator and introduce an additional BUA artifact, the magnitude of which will depend upon the thickness of the cortex, having been demonstrated by both experiment and model simulation [31]. The potential for scanning confocal ultrasound to define both surface topology and thickness has recently been described [32], with the potential to reduce errors associated with the cortex.

Calcaneal edema has been demonstrated to reduce BUA [33]. Foot positioning is probably the major cause of clinical measurement imprecision for BUA [34].

A summary of the dependence of ultrasound propagation, intrinsic BUA measurement, and clinical system BUA estimation upon bone shape, structure, visco-elasticity, bone thickness, and soft tissue thickness is shown in Table I.

V. In Vitro Experimental Findings

The relationship between BUA and bone density has been studied extensively *in vitro*. Early studies [35], [36] found high correlations ($r = 0.83$–0.85) between BUA and apparent density in samples of cadaveric calcanei. Even higher correlations ($r = 0.97$) have been reported in human vertebra samples measured in the anterior-posterior direction [37]. Others have reported similar strong and positive relationships between density and BUA [38]–[41]; however, in the more dense bovine cancellous bone, the relationship is much weaker or even completely absent, and both positive and negative regression slopes have been reported [39], [42]–[44]. These findings may be explained by the observation in natural tissue samples and phantoms of a near-parabolic relationship between BUA and porosity (and inversely related density), with BUA rising to a maximum at a porosity of approximately 70% [44]–[46]. The relative role of absorption and scattering in determining ultrasound attenuation provides a qualitative explanation for this nonlinear behavior, for example, that absorption primarily determines attenuation in low density cancellous bone with scattering becoming important only in dense cancellous bone samples. It has been further suggested that the parabolic symmetry could be related to the scattering cross-sectional area between bone and marrow, being similar for low porosity (few pores) and high porosity (few trabeculae) [44].

The ability of BUA to determine the mechanical properties of Young's modulus and strength in cancellous bone has also been studied [38], [47], [48]. QUS can add predictive power beyond that of density for mechanical properties estimation [49]–[54]. BUA has also been shown to be an independent predictor of Young's modulus when the correlation was adjusted for trabecular density both *in vitro* and *in vivo* [54]. The ability of BUA to predict bone mechanical properties is diminished when measuring the bone *in vivo* rather than as cancellous cubes or when predicting the strength of a bone at a remote location. BUA of the heel correlates moderately with the strength of the calcaneus itself ($r = 0.79$) [52] and the proximal femur ($r = 0.57$–0.71) [55], [56]. Femoral BMD is, however, a significantly better predictor of femur strength (0.77–0.94) than heel BUA [55]; similarly, calcaneal BUA is not as good as lumbar spine BMD in predicting vertebral strength [57], [58]. However, in contrast to these results, Lochmüller *et al.* [59] found that calcaneal QUS correlates with failure load of the proximal femur in a manner similar to femoral neck BMD. Caution should be expressed, however, since mechanical and ultrasound measurements of whole bones are compounded by many error sources and hence tend to be unreliable.

Since calcaneal BUA was first described in 1984, it has been proposed that it provides information on bone structure in addition to density. Reports in the scientific literature that describe a poor association between BUA and BMD have attributed this to an additional dependence of BUA upon structure. However, a poor association could be also due to many other factors including measurement errors and anatomical discordance [60]. Evidence of the structural dependence of QUS has come mainly from anisotropic, histomorphometric, and fractal analysis studies. Due to the difficulty of obtaining meaningful parameters characterizing trabecular structure, studies have often been qualitative in nature. Noting that density is by definition isotropic, BUA anisotropy has been reported in

TABLE I

SUMMARY OF THE DEPENDENCE OF ULTRASOUND PROPAGATION, INTRINSIC BUA MEASUREMENT, AND CLINICAL SYSTEM BUA ESTIMATION UPON BONE SHAPE, STRUCTURE, VISCO-ELASTICITY, BONE THICKNESS, AND SOFT TISSUE THICKNESS. THE NUMBER OF TICKS REPRESENTS THE RELATIVE MAGNITUDE OF DEPENDENCE.

	Bone shape	Structure	Visco-elasticity	Bone thickness	Soft tissue thickness	
Propagation	Refraction and phase cancellation	Scattering	Absorption			
External factors						Temperature
Intrinsic measurement (Gold standard and artifact free)	√√	√√√	√√	√√√		√
System estimation				√√	√	

cancellous bone cubes from the equine metacarpus [61], the bovine femur [62] and radius [63], and human vertebra [37] and femur [55], implying that structure affects BUA independent of density. Further evidence came from demineralization and crushing of bovine cancellous bone samples [64]. A few studies have reported relationships between BUA and quantitative microstructural measurements [61], [65], [66]. Again, there is a discrepancy in findings that may be attributed to the natural tissue studied, for example, whether it is relatively high density bovine or relatively low density human cancellous bone. Interestingly, in a bone-mimicking material spanning a range of densities and with two distinct pore sizes, BUA was higher for the larger pore size, indicating the sensitivity of BUA to a structural factor (pore size) independent of density [45]. While it is clear that structural effects on BUA can be demonstrated, it is by no means certain that these effects are significant in clinical measurements. The strong correlations between density and BUA measured mediolaterally in human cancellous bone *in vitro* implies only very limited room for structural factors to play a role [38], [67]. This, along with the other evidence cited above, suggests caution with regard to claims that clinical heel measurements are useful indicators of cancellous structure.

What may, however, be concluded is that we currently have a poor understanding of exactly how the ultrasound propagation is manifested in terms of attenuation and hence BUA, along with other measurement parameters such as velocity. Although there appear upon initial consideration to be well-defined fundamental relationships available describing velocity, both simply as distance traveled divided by time taken, and as the square root of modulus of elasticity divided by density, these are complicated. For example, the actual propagation distance within cancellous bone will be greater than this due to structural tortuosity [68]. Until we have a better understanding of propagation, the plethora of reported inter-parameter correlations between BUA, velocity, BMD, and so forth are somewhat founded upon a statistical rather than a fundamental scientific basis. There is, however, interesting evidence, both scientific [29] and clinical [69] that has contributed to an improved understanding of these relationships: for example, the study of primary parathyroidism where a greater cortical than trabecular bone loss was manifested in differential changes in BMD, BUA, and velocity, both compared to control subjects and following treatment [69].

VI. THEORETICAL MODELLING

BUA describes a measure of the frequency dependence of total attenuation, being a combination of absorption and scattering in cancellous bone, that may itself be characterized in terms of density and structure, noting that these relationships have not to date been elucidated. Conflicting evidence on the relative role has been given as to whether absorption [20] or scattering [70] is the predominant attenuation mechanism in cancellous bone. Allied to the inadequacy of experimental studies to determine these relationships, a number of theoretical approaches have therefore been considered, including those of Biot and Schoenburg, noting, however, that modeling the frequency dependence of ultrasound attenuation has proved extremely difficult.

Scattering is caused by sudden spatial changes in elastic properties, the magnitude being dependent on the relative size of inhomogeneities and the ultrasound wavelength [71]. Multiple scattering may also be considered, being a combination of the original and previously scattered waves. An alternative approach is to consider cancellous bone to be a mixture of two components, bone and marrow, with a bone volume fraction δ, velocity c may be expressed as

$$c = ((\delta\rho_1 + (1-\delta)\rho_2) \cdot (\delta k_1 + (1-\delta)k_2))^{0.5}$$

where ρ_1 and ρ_2 are the densities, and k_1 and k_2 are the bulk moduli for bone and marrow, respectively.

This fundamentally simple approach has been shown to be reliable for a number of complex media such as suspensions, although for porous media such as cancellous bone, it is limited by not taking into account mode conversion. For attenuation, the simple mixture theory predicts trends with ultrasound frequency, but does not provide accurate quantitative data. Chernov's theory combines scattering

and simple mixture theory via velocity fluctuations and scatterer size [72] and has been shown to provide reliable quantitative data for both velocity and attenuation in cancellous bone. An alternative approach is Schoenberg's theory that assumes an idealized microstructure of periodic stratified layers and predicts two longitudinal waves for all propagation angles but an angle-dependent anisotropy [73]–[75]. A scattering model based upon velocity fluctuations in a binary mixture (marrow fat and cortical matrix) has also been considered to estimate ultrasonic attenuation in cancellous bone [76], [77]. The model predicted nonlinear trends very similar to those previously observed experimentally and also demonstrated that attenuation was dependent on scatterer size in addition to porosity. This further supports the argument that attenuation is influenced by structure. Potential limitations in this approach include the failure to include absorption in the model.

A simple multi-echo simulation has also been considered, the primary source of ultrasound attenuation being phase cancellation [78].

The Biot theory [79] 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 [80]–[83]. It allows for an arbitrary microstructure. The different motions of the solid elastic framework (bone) and the interspersed fluid (marrow), induced by the ultrasonic wave, are considered separately. It includes energy loss due to viscous friction between solid (bone) and fluid (marrow). The theory gives rise to three elastic parameters that are dependent on the structural and elastic properties of the porous media. The elastic parameters P, Q, and R were defined using the assumption that porosity is constant for small strains, and linked the elastic coefficients to measurable physical constants:

$$P = \frac{\beta\left(\frac{K_s}{K_f} - 1\right)K^* + \beta^2 K_s + (1-2\beta)K_s - K^*)}{1 - \beta - \frac{K^*}{K_s} + \beta\frac{K_s}{K_f}} + 4\frac{\mu^*}{3},$$

$$Q = \frac{\left(1 - \beta - \frac{K^*}{K_s}\right)\beta K_s}{1 - \beta - \frac{K^*}{K_s} + \beta\frac{K_s}{K_f}},$$

$$R = \frac{K_s\beta^2}{1 - \beta - \frac{K^*}{K_s} + \beta\frac{K_s}{K_f}},$$

where K_s is the intrinsic bulk modulus of the solid material, K^* is the bulk modulus of the frame, K_f is the bulk modulus of the fluid, μ^* is the shear modulus of the frame; β is the porosity (volume fraction of the fluid phase). The other three main parameters are the mass coefficients that describe the effects of viscous and inertial drag, taking into account the fact that the relative fluid flow through the pores can be nonuniform:

$$\rho_{11} + \rho_{12} = \rho_1$$
$$\rho_{22} + \rho_{12} = \rho_2$$
$$\rho_{12} = -(\alpha(\omega) - 1)\beta.\rho_f$$
$$\rho_1 = (1-\beta)\rho_s$$
$$\rho_2 = \beta.\rho_f$$

where ρ_s and ρ_f are the densities of the solid and the fluid phase, respectively. These parameters are complex terms taking into account the theory of dynamic tortuosity and permeability. ρ_{11} is the effective density of the solid moving through the liquid, ρ_{22} is the effective density of the fluid moving through the solid, ρ_{12} is the inertial drag that the solid exerts on the fluid, and $\alpha(\omega)$ is the Johnson-Koplik-Dashen (JKD) dynamic tortuosity [84]. The above parameters can be defined without using the JKD formulation of tortuosity; in this case the tortuosity is a purely geometric variable (sometimes referred to as the sinuosity) [68]. The JKD tortuosity is the formulation most commonly used and is adequate for most situations.

The Biot theory predicts three modes of propagation for an ultrasonic wave in a porous media. Two dilatational waves (longitudinal waves), termed waves of the first kind and waves of the second kind, alternatively named fast and slow waves; and one rotational (shear wave). The usual explanation given for the existence of the separate fast and slow waves is that the fast wave represents the fluid and solid vibrating in phase and the slow wave corresponds to vibration in anti-phase (half a wavelength separation). The wave equations for these waves are:

$$V^2_{\text{fast/slow}} = \frac{\Delta \pm \sqrt{\Delta^2 - 4}\,(\rho_{11}\,\rho_{22} - \rho_{12}^2)\,PR - Q^2}{2\,(\rho_1\,\rho_2 - \rho_{12}^2)} \tag{1}$$

where

$$\Delta = P\rho_{22} - 2Q\rho_{12} + R\rho_{11}, \tag{2}$$

$$V^2_{(\text{shear})} = \left(\mu\left(\rho_{11} - \frac{\rho_{12}^2}{\rho_{22}}\right)\right). \tag{3}$$

Eq. (1) has two complex roots, corresponding to the fast and the slow waves, and (2) has one; the real part of the root (qr), provides the wave speed as ω/qr (m s^{-1}), where ω is the angular frequency of the wave. The imaginary part of the root (qi), provides the attenuation.

The greatest difficulty in the application of the Biot theory to cancellous bone is the large number of physical parameters that have to be measured or estimated. A copious amount of information exists in the literature about the experimental determination of the parameters necessary for the Biot theory, applying both general experimental methods and those more specifically applicable to cancellous bone. Many of the parameters required by the Biot theory are unknown and may only be estimated. The intrinsic ultrasonic (velocity and attenuation) and physical parameters (density (ρ_s), Young's modulus (E_s), bulk modulus (K_s) and Poisson's ratio (υ_s)) for cancellous bone tissue are assumed to be those for solid bone

material. Even with this assumption, some difficulty can arise, however, in the experimental measurement of these parameters. Once the intrinsic material properties have been measured or calculated, it is possible to calculate the parameter values for the trabecular framework. Poisson's ratio is either assumed to have a specific value (typically 0.5) or experimentally measured, with inherent difficulty. The Young's modulus for the cancellous bone frame can be determined using three methods: by calculation, by compressive testing at a low strain rate, or by ultrasound measurement. Since bone marrow is mainly composed of fat with very little blood and tissue fluid, the physical parameters for fat are normally used for the pore fluid. Water may be substituted as the pore fluid, particularly if the theoretical results are to be compared to experiments performed *in vitro* where the marrow is often completely removed and replaced with water. Permeability relates the rate of fluid or gas flow through a material to the sample thickness, the cross-sectional area, and the pressure causing the flow, and is defined by the relationship known as Darcy's law. One of the most elusive parameters in the Biot theory is the tortuosity ((() or the sinuosity. This is defined as the ratio of the length of true path of flow for a fluid to the shortest distance between the inflow and the outflow. It is important to realize that this definition is kinematic, not geometric, and must be experimentally measured, the most common method being electrical resistivity. The final parameter to be measured is the pore size parameter ((), a measure of the intrinsic dynamically interconnected pore sizes. The pore size parameter can be estimated by measuring the mean trabecular plate separation using standard histomorphometric techniques. For cancellous bone, the pore size parameter is assumed to be half of the mean trabecular plate separation.

Finally, the Biot theory essentially assumes isotropic behavior, whereas most elastic and structural parameters of cancellous bone are anisotropic. Hence, for a true representation of ultrasound propagation through cancellous bone, consideration of sample orientation should be given.

VII. Summary

In summary, although the measurement of broadband ultrasonic attenuation in cancellous bone was first described 23 years ago, a fundamental understanding of the propagation dependence upon material and structural parameters is still lacking. There still remains, therefore, the attractive potential that an improved understanding that has been scientifically validated will enhance the clinical utility of this technique. Allied to this is the potential that pulse-echo backscattered measurements may be performed at the clinically relevant anatomical fracture site of the proximal femur.

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