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**IMAGING THE SHAPE OF LONG BONES USING ULTRASOUND
TOMOGRAPHY**

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In this paper, we investigate ultrasound measurement of the cortical thickness of child long bones. Due to the acoustic impedance mismatch between surrounding tissues and bone, the higher the frequency, the smaller the portion of energy going through and the lesser the resolution of the shaft. To estimate the cortical thickness, we propose ultrasonic tomography, with frequencies lower than those of conventional echography and especially adapted to the strong contrast medium. The ultrasonic tomographic imaging techniques produce cross-sectional images of the spatial distribution of some physical parameters of an object from scattered ultrasonic measures. The measurements are made for more or less dense sets of transmitter and receiver positions and of frequencies of the interrogating wave. We solved this inverse scattering problem by using a Born approximation, which leads to a simple and attractive linear relation between the object function and the scattered field, particularly in the far-field. This approach makes it possible, in principle, to reconstruct the object almost in real time for a large set of scattering data. We investigated wide-band acoustical tomography experimentally.

1. Introduction

Some bone pathologies like osteoporosis are directly related to bone remodeling dysfunctions leading to low bone mass, alterations in material properties or micro architecture, and ultimately impaired bone strength. X-ray densitometry techniques represent the gold standard for assessing skeletal status and provide accurate and precise measurement of bone mineral density (BMD) [i]. Although BMD accounts for most of the variability of bone strength, several other bone properties also influence bone strength, such as material properties and micro architecture. Recent development of high resolution MRI and CT imaging techniques permits an estimate of several structural parameters of cancellous bone and cortical bone. Because ultrasound involve compressive and shear stress, quantitative ultrasound (QUS) has been introduced as an alternative to X-ray densitometry to probe multiple bone properties that contribute altogether to bone strength. QUS has been applied to the measurement of peripheral skeletal sites and many studies report the ability of QUS to prospectively predict fracture risk. However, interaction mechanisms of ultrasound with the structure of bone are not fully understood yet and the complexity of the phenomena has hindered the technical evolution of QUS and its widespread use in clinical settings. Several QUS measuring techniques are currently available, including transverse transmission to measure the heel bone (calcaneus) [ii] and axial transmission which can be applied to multiple peripheral skeletal sites (radius, metacarpal,

ulna, etc.) [iii]. The basic principle of this technique relies on ultrasound transmission and estimates of the average value of the ultrasonic parameters (attenuation, sound velocity) for the wave pathway through the skeletal site. Other approaches exploit the reflection of ultrasound from cortical bone (ultrasound critical reflectometry) [iv] or the signal backscattered from cancellous bone micro architecture [v] in order to provide a direct assessment of material properties or micro architecture features. Ultrasound tomography represents an alternative to reach one or more significant parameters like acoustic impedance, ultrasonic velocity and attenuation. In a previous study, our group has shown that preliminary *in vitro* quantitative images of sound velocity in a human femur cross section could be reconstructed by combining ultrasonic reflection tomography (URT) and ultrasonic transmission tomography (UTT). The dramatic impedance mismatch between soft tissue and bone causes refraction, an effect which must be taken into account and compensated before the straight-ray reconstruction by the backprojections technique can be applied to UTT data acquisition. We have proposed to use URT first to reconstruct the external and internal boundary of cortical bone, and to modify the acquisition scheme for (UTT) according to the shape of the object in order to compensate for refraction.

2. Quantitative Ultrasonic Tomography (QUT): distorted Born iterative method

Ultrasonic Tomography is brought about by a linearization of the inverse acoustic scattering problem. For soft biological tissues, the reference medium is water, *in vitro*, and the mean-medium (the soft tissue *in vivo*, the water tank *in vitro*). In this case, the approximations of weak scattering are generally used, for example the approximation of Born or Rytov. On the other hand, if the medium much is more contrasted than the surrounding reference medium, the problem is non-linear and there is in general no single solution. This is the case for bones. Our idea is to adapt the methodology and algorithms of inversion developed for soft tissues to bone (or solid tissue). This approach, analogous to the Distorted Born Iterative Method, relies on two parts: the hypothesis relative to the bone structure, and thus essentially to the choice of the site studied, and the hypothesis relative to the propagation of a wave in this kind of heterogeneous media. We use the geometrical theory of diffraction, and more simply, a mode of propagation in straight beams.

Hypothesis and assumptions: Distorted Born

The overall architecture of bone is divided into cancellous bone, and cortical bone. Cortical bone forms a compact shell around highly porous cancellous bone, which is formed by an interconnecting latticework of bone filled with marrow. In general, cancellous bone is found in the axial skeleton (spine), small bones of the peripheral skeleton (e.g., calcaneus) and distal parts (epiphysis) of long bones such as radius and femur, while the diaphysis of long bones is composed primarily of cortical bone (e.g., radius, femur, tibia). This

study is focused on cortical bone. One admits that the shaft zone is weakly heterogeneous, one can suitably apply, inside the medium, a linearization of the propagation and a "fluid" modeling. Cortical bone is generally assimilated to a transverse isotropic medium, with a principal axis of symmetry oriented along the longitudinal axis of the diaphysis. This assumption was confirmed by several independent measurements of compression and shear wave velocities in various cortical bone specimens and in various directions of isotropy [vi, vii]. The correction of the refraction of the waves on the water/bone and bone/water interfaces allows us to consider the scatterer as a weakly heterogeneous medium, and to be coherent with the hypothesis of the Born approximation.

Ultrasonic Transmission Tomography (UTT), based on the measurement of the relative fluctuation of time-of-flight (TOF), provides the value of the sound velocity along the propagation in the object. The relative fluctuation of velocity will be calculated by considering the TOF along a straight line of the acoustic wave in the cortical zone. TOF is directly related to the weak disturbance of the refraction index. Consequently, the velocity map may be obtained by inverting the TOF data.

Getting the shape of the body via Ultrasonic Reflected Tomography

Ultrasonic Reflected Tomography gives a qualitative image of the shape of the body. At this level, the algorithm developed for the low-contrast media can be applied without restriction related to the material. After the URT is built, the external boundary of the object is determined. For interior boundary, the problem is more complicated because one must determine the "apparent" internal border, which does not correspond to the real limit. Indeed, the basic assumption is that of the media slightly diffracting and the induced parameters of reconstruction are those of water. There is thus an aberration of imaging in the internal border. Lasaygues *et al* [4] showed that URT with signal processing [viii] improves the resolution by eliminating interferences (speckle) and thus makes it possible to distinguish various boundaries from the objects.

Ultrasonic tomography is then ideally suited to reconstruct the external boundaries of high-contrast cylindrical bodies. For the inner cavity, the real position is obtained by modifying the distance between the external boundary and the "apparent" internal boundary previously measured. This correction of distance uses the *a priori* wave velocity in the medium (c_{bone}). The first phase of the algorithm consists in representing the URT image in the cartesian plane mm / degrees (Fig. 1). The second phase determines the position of the external border (z_{ext}) and then of the internal border (z_{intl}). For modeling of the forward URT problem, the host-middle is water (c_{water}). As a consequence the new internal border is determined relative to this reference velocity:

$$z_{int2} = \frac{c_{bone}}{c_{water}} \left[z_{intl} - z_{ext} \right] \quad (1)$$

The results are highly accurate. This improved URT algorithm is based on the same initial information as that used for UTT.

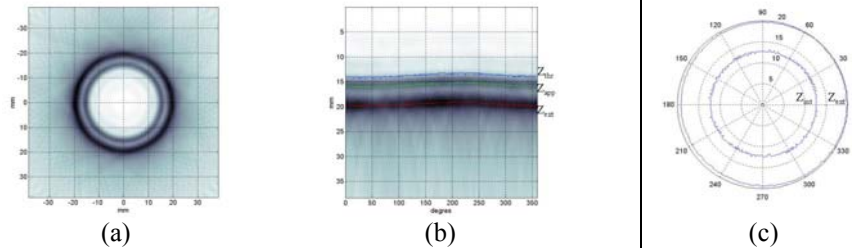


Fig. 1. Determination of boundaries of a Plexiglass cylindrical tube ($\phi_{ext} = 20$ mm; $\phi_{int} = 12$ mm, initial mean-velocity = 2700 m/s); a -URT, b - polar map, c - polar representation.

Correction of refraction phenomena for Ultrasonic Transmission Tomography

UTT is perturbed when the media are strongly contrasted. Basic assumptions on the tomography are also not respected and several phenomena appear. To preserve the same hypotheses and the same approximations as when weak scattering tomography is transposed to the long-bone tomography, the solution is to compensate refraction of the waves on the water/bone and bone/water interface, and thereby to preserve a straight-line propagation inside materials [ix, x].

The compensation procedure may be achieved based on *a priori* knowledge of the geometrical shape and acoustical properties of the object [xi]. Using Snell's laws one can compute the appropriate position and orientation of the transducers for straight parallel ray propagation in the object. For the femur that we compare to a tube, we use the refraction correction everywhere except in the hollow part (medullar canal), considered as a multi-layer medium with parallel and planar interfaces (Fig. 2). Then we define an original construction procedure to acquire acoustic sequence, called "two-zone correction procedure".

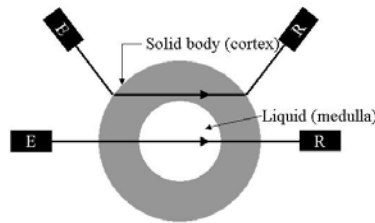


Fig. 2. Diagram of ultrasonic pathways for two-zone correction procedure for time-of-flight detection (zone 1: cortex, zone 2: cortex – medulla – cortex)

Iterative Quantitative Ultrasonic Tomography algorithm

Quantitative Ultrasonic Tomography is the combination of the images obtained with URT and UTT. URT gives the size and the boundaries of the shape, and UTT quantifies each pixel between the boundaries by speed-of-sound

values. When *a priori* information is poorly known, the experimental protocol becomes an iterative algorithm. Each QUT gives the geometrical and acoustical mean parameters of the next iteration. The beginning is done using initially known basic information. The complete iterative algorithm presented in the general diagram operates as follows. We carry out the reflected tomography of the object and initialize *a priori* velocity with values (literature or acoustic measurements). Then we determine external boundary and, for each view angle, we calculate the apparent acoustic diameter (i.e. *a priori* geometry). With *a priori* velocity, we determine the internal boundary of the object (tube case). By using this *a priori* information, we carry out acquisitions according to the algorithm of correction of refraction and we determine the TOF according to the two-zones and build projections for each view angle and the corresponding transmission tomography. Ultimately, we build the QUT.

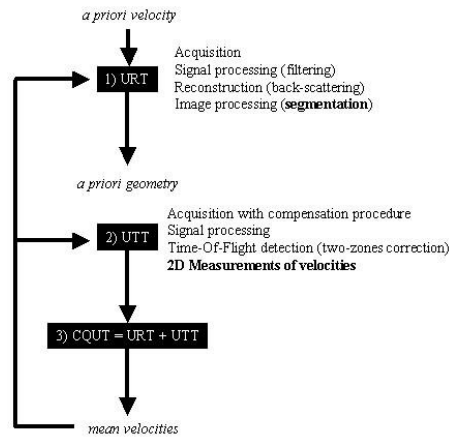


Fig. 3. General algorithm for Quantitative Ultrasonic Tomography (QUT)

2. Measurement system

The general architecture of the mechanical system is that of a first-generation X-ray tomograph: a main symmetric arm holds two transverse arms allowing the parallel translation of two transducers (1 MHz). Angular scanning is allowed by the rotation of either the main arm or the object holder. The transducers can also be positioned and oriented with high precision, allowing linear and sectorial scanning. Two 1 MHz ultrasonic transducers were connected to a computer via a digital oscilloscope. The axial resolution was 6 mm (4 μ s) and the lateral resolution was 4 mm.

3. Results

Human femur

The sample was a human femoral specimen about 32 ± 5 mm in external diameter and 16 ± 2 mm in internal diameter. For the *a priori* velocities, we chose

3400 m.s⁻¹ in bone and 1478 m.s⁻¹ in water. For the diameter, we initially compared the femur to a circular tube but the result was not satisfactory. By introducing a more realistic diameter, obtained for each view angle by URT, the results are much better. The fluid in the internal shape was reconstructed with a correct velocity value ($\approx 1500 \text{ m.s}^{-1}$) and the dimension of this cavity was 15 - 17 mm. The greatest external shape (30 - 34 mm) was exact and the mean velocity estimated in the cortical shell was 3150 m.s⁻¹.

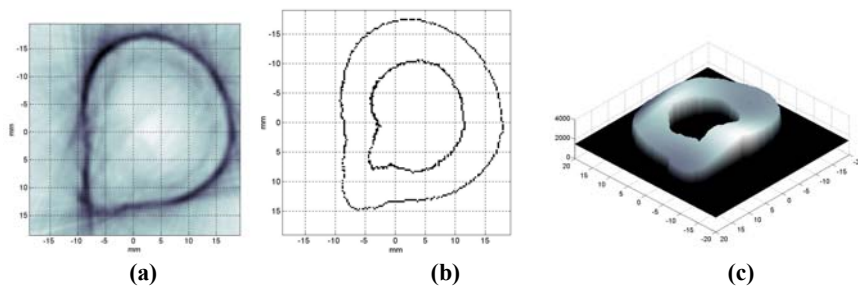


Fig. 4. 1 MHz human femur UT: a –basic URT b – Acoustic boundaries plot; c – QUT = combined URT and UTT with two-zone correction;

3D representation

Here, we compare two 3D US reflected tomogram of a human female thighbone. The first one is obtained on a non-menopausal woman. The second is obtained on a menopausal and illness (osteoporosis) woman. We notice that the cortical thickness decreased at the illness woman's. This modification of the thickness is a criterion of diagnosis of this pathology today.

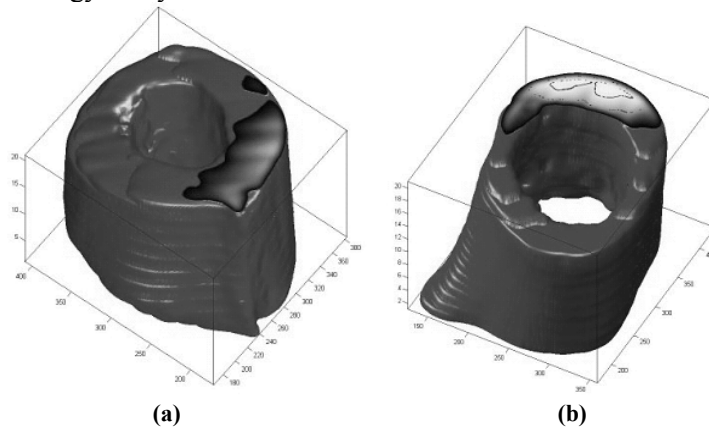


Fig. 5. Human femur (in vitro) without marrow (a) un-illness (b) illness (osteoporosis disease, 1 MHz)

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IMPROVEMENT OF THE PERFORMANCE OF HIGH-INTENSITY LINEAR AND SPHERICAL PHASED ARRAYS FOR APPLICATION IN SURGERY

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Disadvantages of powerful phased arrays designed for noninvasive surgery include the unwanted presence of grating lobes and other secondary intensity maxima in acoustic fields. Methods for improvement the quality of the acoustic fields generated by the arrays are discussed and the efficacy of these approaches is illustrated by the results of measurements and computer modeling. For linear arrays, it was shown that optimization of number of active elements with the values of aperture, focal distance, shift of the focus and other focusing conditions permits to reduce the level of secondary maxima. For two-dimensional spherical phased array, it was shown that randomization of elements distributions on the surface of the array improved significantly the performance of the arrays by suppressing grating lobes in the acoustic field. The other necessary condition is