ULTRASONIC TOMOGRAPHY OF LONG BONE: RESOLUTION AND QUANTIFICATION

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Abstract

Ultrasonic propagation in bone suffers from the severe mismatch between its acoustic properties and those of the surrounding soft tissues, in particular at skeletal sites like vertebrae or thighbones. LIP and LMA united to work out an Ultrasonic Tomography. The algorithm combines Ultrasonic Reflection Tomography (URT) and Ultrasonic Transmission Tomography (UTT) and relies on *a priori* information of the analysed medium. URT gives the shape of the body and UTT gives the velocity map. For the Time-Of-Flights detection in UTT, we use a wavelet analysis tool adapted to ultrasound signals that allows the transmitted signals to be cleaned and filtered and the useful information to be separated from the unwanted noise. All these treatments were implemented to examine an academic target and real human thighbones.

Introduction

The objective of this study is to show the feasibility of long bone imaging using ultrasonic tomography. Ultrasonic propagation in bones suffers from the severe mismatch between the acoustic properties of this biological solid and those of the surrounding soft medium, (soft tissues *in vivo* or water *in vitro*). This propagation depends on the skeletal site we are analysing, which is more or less heterogeneous and anisotropic. Because this study is a new approach to ultrasonic bone examinations, the skeletal site that we have been investigating corresponds to the cortical zone of the femur (the diaphysis) where the biological structure was considered weakly heterogeneous beside the wavelength and isotropic to the perpendicular plan of the vertical fibres.

Bone imaging is a non-linear inverse-scattering problem. Our group showed [1] that preliminary *in vitro* quantitative images of sound velocities in a human femur cross section could be reconstructed by combining ultrasonic reflection tomography (URT) and ultrasonic transmission tomography (UTT). URT gave the shape of the body and UTT gave the velocity map. Our first results were very promising.

In this paper we present a novel image-processing tool to extract the external and internal boundaries and a new processing, based on wavelet analysis of transmitted ultrasonic signals, which allowed us to detect time-of-flight more precisely from.

A brief review of the ultrasonic tomography that we developed with correction algorithm of the wavepaths and compensation procedures, as well as the first results of our analyses on models and specimens of long bone using our new iterative quantitative protocol are outlined below.

Iterative Combined Quantitative Ultrasonic Tomography

Ultrasonic Tomography is based on a linearization of the inverse acoustic scattering problem. It allows perturbations (theoretically small) of a reference medium to be visualized.

For soft tissues, the reference medium is *in vivo* the biological mean-medium and *in vitro* the water. The approximation of weak scattering is generally used such as Born approximation. This leads to the Inverse Born Approximation method, whose practical solution results in regular angular scanning with broadband pulses, allowing one to cover slice-by-slice the spatial frequency spectrum of the imaged object. This leads to "Reconstruction-From-Projections" algorithms like those used for X-ray Computed Tomography.

The acoustic impedance of the cortical of long bones is more highly contrasted compared to the surrounding soft tissues (or water), and so, perturbs ultrasonic propagation (refraction, attenuation and diffraction). Therefore, bone imaging is a non-linear inverse problem with no single solutions. However, for some hypotheses and *a priori* knowledge of the analyzed bones, we showed [1, 2] that one solution is possible.

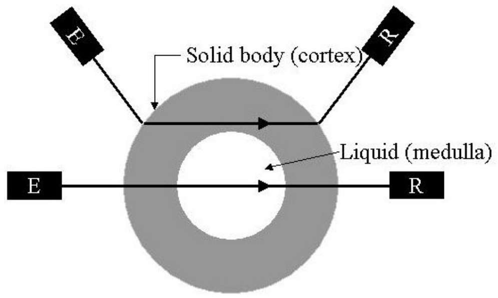


Figure N°1: Correction of the refraction and two-zones compensation procedure [2]

Firstly, we adapted our *in vitro* acquisition protocols to take into account the refraction phenomena due to the high contrast and, then, we adapted the inversion algorithms developed for soft tissues to these hard mediums.

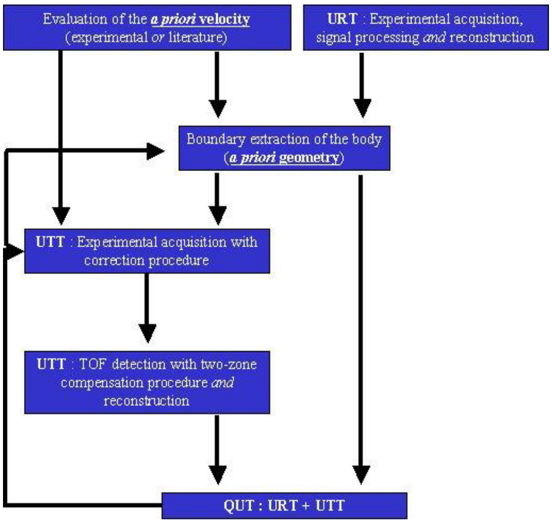


Figure n°2: Iterative algorithm of Quantitative Ultrasonic Tomography

Secondly, because the wavelength we used (typically 3-4 mm) was greater than the porosity of the shaft zone of the femur, it could be considered weakly heterogeneous and the ultrasonic propagation was less disturbed and was in straight lines inside the body.

Our algorithm combines two tomographies, Ultrasonic Reflection Tomography (URT) and Ultrasonic Transmission Tomography (UTT). URT is used to build the external contour of the object precisely, and the program we developed is particularly effective

Long human bones are effectively irregular hollow tubes (medullar canal) (figure n°1) and should support the propagation of more complex waves similar to elastic volume waves. The dimensioning of the internal boundary by tomography is impossible if the sound velocity model inside the shell is uncorrected next to the background model. In our algorithm, a mean value of velocity may be introduced in order to correct the relative error. Then, URT provides the realistic shape of the bodies [3], and, because the corrections of the wave refraction throughout the shell are based on the knowledge of these dimensions (Snell-Descartes laws), URT is used to improve the experimental protocol of transmitted signal acquisitions. This is then connected to the UTT, which is especially adapted to mapping local velocities inside the area defined.

A velocity ground level value is then necessary to initialize both tomographies. After combining URT and UTT, a mean velocity can be measured on the final image and used to improve the dimensioning and the quantification, ensuring, the complete iterative Quantitative Ultrasonic Tomography (figure n°2) in which the behavior and the convergence are good.

Results

Optimization of Time-Of-Flight detection using wavelet analysis

The experimental ultrasonic acquisition protocol enabled us to digitize correctly the transmitted signals (recorded by the receiver) according to the compensation procedure throughout the object. The transmitted signals were composed of several packages of waves, which had followed various pathways within the object [4][5] (figure n°3). To use transmission tomography, transmitted signals were processed from each view angles, and projections were constructed by Time-Of-Flight (TOF) detection. Our first process for assessing TOF was based on detecting the arrival of the signal corresponding to the wave propagating at, or close to, the longitudinal velocity in the diaphysis. However, the subjective thresholds were very sensitive to the signal-to-noise (SNR) ratio. The number of bad TOF increased because these thresholds were not adapted to all the values (typically 15 728 640 in the results presented here) that we must automatically treat. Indeed, the choice of the lower limit of the threshold was fixed for each view angles in relation to the noise. If we extracted the useful signal of this noise, and if we standardized all the packages with a criterion, which does not perturb their temporal position, the errors decreased.

Our idea was to exploit the spectral information on the transmitted signals. For this, we have selected wavelet analysis, which allowed simultaneous joint studies in the time and frequency domain.

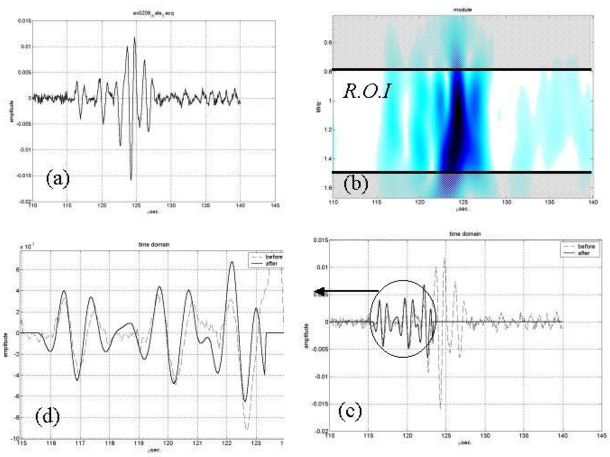


Figure n°3: Example of wavelet decomposition with R.O.I., and synthesis of 1 transmitted signal in cortical bone

Two aspects were considered in particular: the significant spectral bandwidth of the emitted signal; i.e. the transmitted signal without the object (figure n°3) and the axial resolution of the transducer, brought back to the support of this emitted signal. Then, we defined a Region-Of-Interest (ROI) consisting of this time and frequency bandwidth for the wavelet analysis, filtering and synthesis (sum of the wavelet coefficients).

In figure n°3, we can see the R.O.I. that was used to synthesize the wavelet coefficients (b), [110 – 140 µs] for time and [0.8 - 1.6] MHz for spectral. The results are given in (c) and (d). We can see that the SNR is higher than before treatment, and that the signal is filtered for a low and high frequencies. Finally, the algorithm improved the resolution and the quantification on the signal. The same threshold used for all treated signals, would not engender as much false TOF-detection as when it was not treated, or treated only in time.

Analyzing the frequencies provides the most successful method and the TOF-detection less sensitive to the noise. The advantage of our algorithm is that the noise reduction, the thresholding or the TOF-detection may be made simultaneously on the image of the wavelet coefficients.

Test target and femur imaging

We applied this *i*CQUT to an test and an anatomical object.

The academic target is a non-circular PVC tube (longitudinal wave velocity = 2700 m/s) (25 x 31 mm for external dimensions and a 11 mm internal diameter). 90 projections were acquired following 128 transverse displacements with a space sampling of 330 µm (figure n°4). Geometrically, the URT of the object is faithful to real dimensions. Quantitatively, the *a priori* velocity introduced was 2700 m/s and the mean velocity given on the QUT was 2742 m/s.

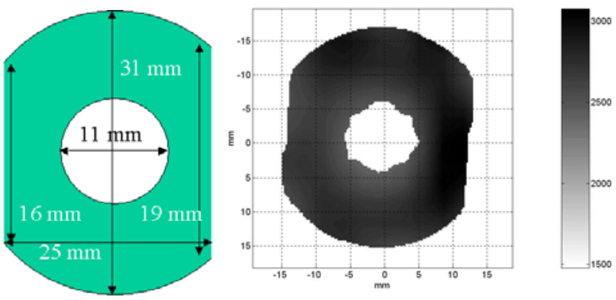


Figure n°4: Quantitative Ultrasonic Tomography of a non-circular PVC tube

The human sample was a 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 [1, 2].

By introducing a more realistic diameter, obtained from each view angle by URT, the results were much better (figure n°5). The fluid in the internal shape was reconstructed with a correct velocity value (≈ 1500 $m.s^{-1}$) 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 $m.s^{-1}$.

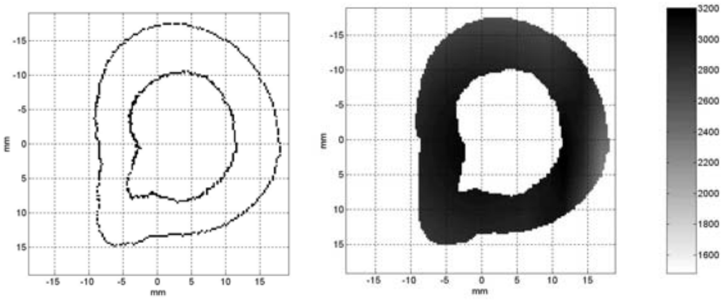


Figure n°5: Quantitative Ultrasonic Tomography of a human femur

Comparison X-ray / Ultrasonic Reflection Tomography

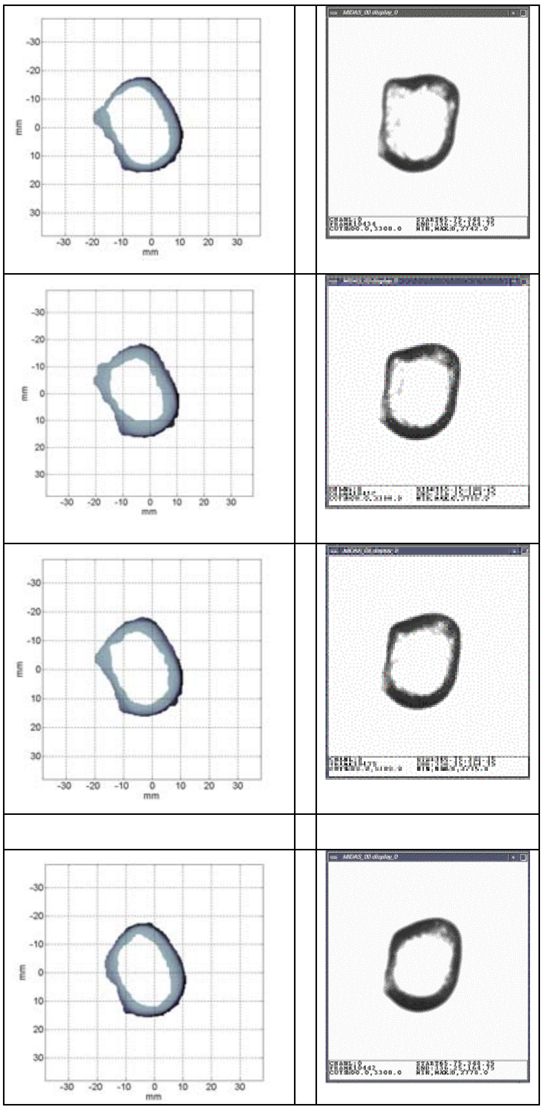
We now present ultrasonic reflection tomographic images obtained with the 2D-scanner and a comparison with X-ray tomographic images obtained with the conventional x-ray scanner GE – ND 12000.

The specimen is a femur of a woman about 30 ± 5 mm in external diameter and 20 ± 5 mm in internal diameter. Figure 6(a) represents four 300 x 300 reconstruction of the femur. The operating frequency is 1 MHz. For comparison, figure 6(b) represents the tomographic images of the same slices acquired with the conventional x-ray tomograph.

The thickness of the acoustic beam is 2 mm and the slice thickness with x-ray is of 1 mm. The step between two slices is 2 mm. The resolution of the x-ray tomograph is 0.25 x 0.25 mm.

We can see that the external shape of the acoustic and the x-ray tomographies are coherent. The internal boundaries are differents but it's important to note that the resolution of the images depends on the dynamics of colors chosen by the operator. In this case, the resolution of the x-ray tomography is not the same as ultrasound images. We cannot compare the image with a quantitative criteria but only qualitative criteria; such as the comparison of external and internal boundaries.

Using iCQUT should allow us to improve the interpretation of the images.



(a) *(b)*

Figures n°6 : Comparison of images obtained from ultrasonic reflection tomograph (a) and conventional x-ray tomograph (b)

Conclusion

An implementation scheme for bone ultrasonic imaging by Ultrasonic Tomography is presented. Because of strong contrast, the problem with bone is more complicated so one must change the usual framework of the tomography (linear inversion of the data with a linear approximation of the direct problem – Born approximation). It is necessary to study the non-linear inverse problem and an iterative method is inescapable (minimization of the distance between measurement and simulation of the forward problem). This approach depends on the configuration of departure (role of the *a priori* knowledge).

In this article we present the progress of our strategy based on Born Iterative method with iteration on the campaigns of acquisition. We assimilate the bone to a weak contrast object (the local fluctuations of acoustic characteristics of a long bone, on a cutting, are weak) immersed in a quasi-homogeneous middle (water). An iterative transmission tomography was implemented with correction of the refraction in the interface bone / water. The *a priori* information about the velocity is realistic. The method gives good results.

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[1] Ouedraogo E, Lasaygues P, Lefebvre J.P., Talmant M, Gindre M., and Laugier P. (2001), Multi-step compensation technique for ultrasound tomography of bone, Acoustical Imaging, 26

[2] Ouedraogo E., Lasaygues P., Lefebvre JP., Gindre M., Talmant M. and Laugier P., (2002) Contrast and velocity ultrasonic tomography of long bones, Ultrasonic Imaging 24,135 – 146.

[3] Lasaygues P., Lefebvre JP., (2001) Cancellous and cortical bone imaging by reflected tomography, Ultrasonic Imaging 23, pp. 55-68

[4] Mc Cartney, R.N., Jeffcott L.B., Mc Carthy, R.N., Transverse path of ultrasound waves in thick-walled cylinders, Med. & Biol. Eng. & Compt. 33, pp. 551-557, (1995)

[5] Mc Cartney R.N., Jeffcott L.B., Combined 2.25 MHz ultrasound velocity and bone mineral density measurements in the equine metacarpus and their in vivo applications, Med. & Biol. Eng. & Comput. 25, pp. 620-626, (1987)