Prediction of the skull overheating during High Intensity Focused Ultrasound transcranial brain therapy

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Abstract—**Ultrasound brain therapy is currently limited by the strong phase and amplitude aberrations induced by the heterogeneities of the skull. However the development of aberration correction techniques has made it possible to correct the beam distortion induced by the skull and to produce a sharp focus in the brain. Moreover, using the density of the skull bone that can be obtained with high-resolution CT scans, the corrections needed to produced this sharp focus can be calculated using ultrasound propagation models. We propose here a model for computing the temperature elevation in the skull during a High Intensity Focused Ultrasound (HIFU) transcranial therapy. Based on CT scans, the wave propagation through the skull is computed with a 3D finite differences wave propagation software. The acoustic simulation is combined with a 3D thermal diffusion code and the temperature elevation inside the skull is computed. Finally, the simulation is experimentally validated by measuring the temperature elevation in several locations of the skull.**

HIFU; skull bone; brain; therapy; ultrasound; CT scan

I. INTRODUCTION

High Intensity Focused Ultrasound (HIFU) brain therapy remains very limited because of the strong aberrations induced by the skull. Indeed, a large discrepancy between the skull high acoustic velocity (about 3000 $m.s^{-1}$) and the brain velocity (about 1540 $m.s^{-1}$) combined with a severe attenuation of ultrasound in the bone, strongly distort the beam shape [1]. However, in the last decade, several ultrasonic techniques have been developed to achieve minimally invasive therapy of brain tumors [2][3]. Thomas and Fink [2] proposed in 1996 to use a time reversal mirror with amplitude compensation to correct the skulls aberrations. In this technique a hydrophone is implanted in the brain, and the signals relating this hydrophone to the transducers of the HIFU array are recorded. Then, one has to emit the time reversed signals with amplitude compensation in order to correct both phase and amplitude aberrations induced by the skull.

Recently, several studies have shown the feasibility of a non-invasive procedure based on high resolution CT scans [4] [5]. The acoustic properties, such as the sound velocity, the density, and the ultrasound attenuation can be modelled thanks to the CT scan of the entire skull bone. Using a finite differences numerical simulation of the complete wave equation, it is then possible to compute in 3D the propagation of an ultrasonic wave generated by a virtual source located inside the brain [5]. The distortion of the ultrasonic wave through the skull is computed, and one can predict the phase correction in order to correct for these aberrations.

However, the heating of the skull bone during a HIFU treatment is a major concern in ultrasound brain therapy, and recent studies have shown that skin burns can be induced [6]. Because of the high and inhomogeneous absorption of the skull, the temperature elevation in the skull bone should be predicted in order to avoid such skin burns. In this paper, we propose to use the CT data to model the acoustic properties of the skull bone as well as its thermal properties. The acoustic propagation software is coupled with a 3D thermal diffusion code to compute the temperature elevation in the skull bone during HIFU treatments. Finally, the temperature elevation in the skull bone is compared to thermocouples measurements performed during an experiment.

II. SKULL MODELLING

A. Acoustic model

The acoustic properties (density, sound velocity and attenuation coefficient) of a human skull are deduced from high-resolution CT images (0.42 mm in-plane resolution, slices 0.7mm thick, Siemens, CT VA1). The models have been developed in a previous work [7]. An example of a velocity map computed from the raw CT data is given in Fig.1.

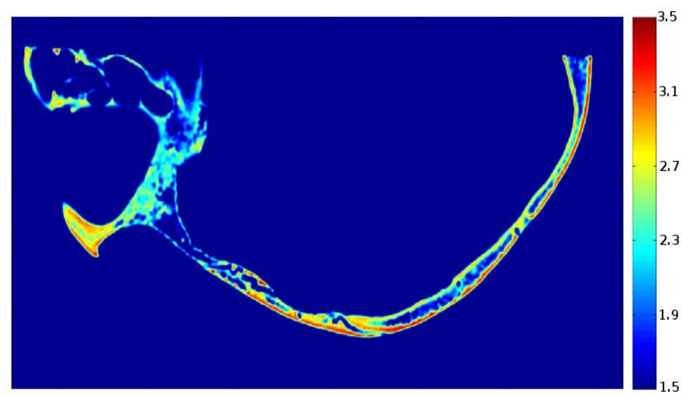

Figure 1. Slice of the velocity map ($mm.\mu s^{-1}$).

The ultrasonic wave propagation is computed using a finite differences code (ACEL) of the full 3D wave equation in an inhomogeneous medium [7]:

$$\left(1+\tau(\vec{r})\frac{\partial}{\partial t}\right)\rho(\vec{r})\,\mathrm{div}\left(\frac{\overrightarrow{\mathrm{grad}}\,p(\vec{r},t)}{\rho(\vec{r})}\right)-\frac{1}{c^2(\vec{r})}\frac{\partial^2 p(\vec{r},t)}{\partial t^2}=0 \qquad (1)$$

where p is the acoustic pressure, c the sound velocity, ρ the density and τ the relaxation time of the ultrasound attenuation.

B. Thermal model

The temperature elevation was computed using a 3D finite differences code based on Penne's bio-heat equation [8]:

$$\rho C_t\frac{\partial T}{\partial t}=\nabla.k\nabla T+Q_0+Q \qquad (2)$$

where T is the temperature, C_t the heat capacity, k the thermal conductivity, Q_0 a blood perfusion term. Q is the heat source term that is given by:

$$Q=\frac{\alpha|p|^2}{\rho c} \qquad (3)$$

where α is the ultrasound absorption coefficient.

The thermal properties of brain and compact bone given in table 1. were found in [9]. The heat capacity and thermal conductivity of the skull were computed according to the local density of the bone found on the CT scan.

	Heat capacity (J/g/K)	Thermal conductivity (W.m/K)
Water	4.18	0.54
Brain	3.7	0.52
Compact bone	1.84	1.30

Table 1. Thermal properties of tissues.

The absorption of the skull bone is much more difficult to determine than the attenuation coefficient used in the acoustic model. The relative contribution of absorption and scattering to ultrasound attenuation is indeed completely different for compact and porous bone. In this study a simple model for the absorption was determined experimentally as a function of the bone porosity:

$$\alpha=\alpha_{min}+(\alpha_{max}-\alpha_{min})(1-\Phi) \qquad (4)$$

where Φ is the bone porosity, α_{min} = 0.2 dB/mm and α_{max} = 5 dB/mm.

III. EXPERIMENTAL SETUP

Experiments were performed with a high power ultrasonic array previously designed for transcranial ultrasound therapy [10]. It is composed of 200 high power transducers randomly distributed on a spherical holder. Each transducer works at 1 MHz central frequency, is linked to its 50 electrical matching and could generate a monochromatic wave with output acoustic intensity of 20 $W.cm^{-2}$ during 5 s (the electrical to acoustical efficiency reached 50%). The transducers are connected to a 200 channel electronic driving system. Each electronic channel is fully programmable and has its own high power emission electronic board that can generate 16 W.

Figure 2. 200-Element high power ultrasonic array.

The skull was mounted in a PVC frame and attached to the phased array. The PVC frame had markers that were visible in the CT images, so that the CT and the experimental coordinates registered. The skulls aberrations were corrected using a 0.4 mm PVDF bilaminar calibrated hydrophone (Golden Lipstick, SEA, Soquel, CA) located at the geometrical focus of the ultrasonic array. A signal is emitted by each transducer one at a time and is recorded by the hydrophone. Then the signals are time-reversed and the amplitudes were adjusted to the same level.

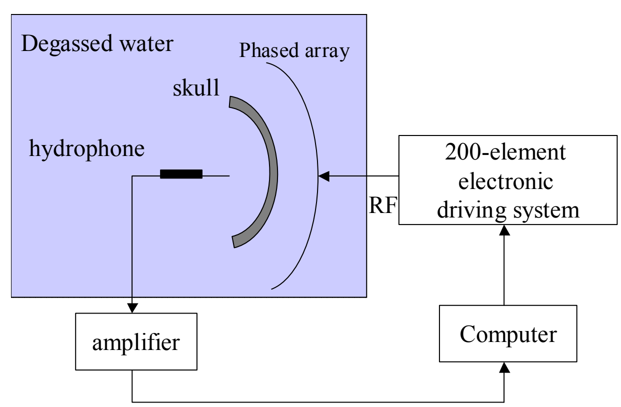

Figure 3. Experimental setup

In order to validate the models, 3 thermocouple sensors were attached on the outer surface of the skull with adhesive tape. 3 other sensors were placed on the inner surface, and 2 thermocouples were inserted in the diploë (see Fig. 4). The sensors were connected to an 8-channel temperature acquisition board.

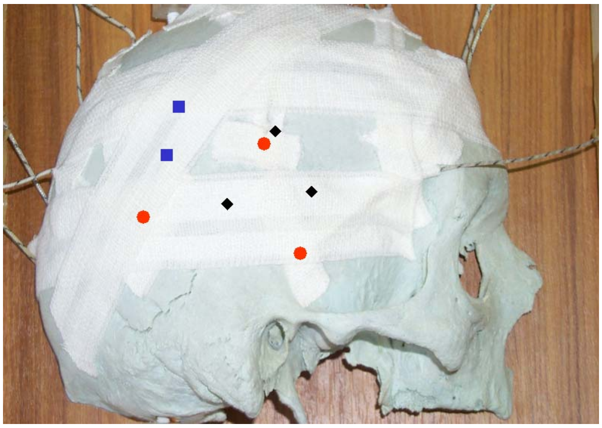

Figure 4. Location of the 8 thermocouples sensors. (on the outer surface ●, on the inner surface ◆, inside of the skull ■).

IV. Results

A. Simulation results

The propagation of the ultrasonic wave was computed through the skull bone. A 1 MHz continuous wave was emitted by the 200 transducers with the phase individually corrected. The amplitude of the pressure field inside the skull bone was calculated in continuous-wave mode. The temperature elevation was then computed using the thermal diffusion code. The heat source term was calculated locally as a function of the local amplitude of the pressure field and the absorption (see Eq. 3).

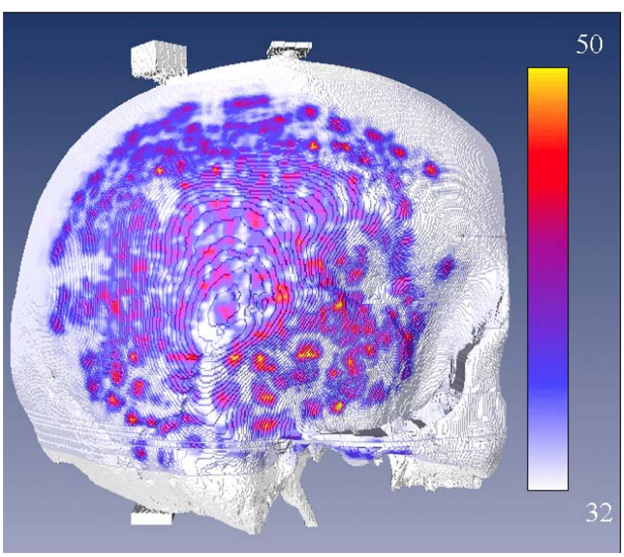

Figure 5: Temperature distribution on the surface skull after 5s sonication.

A 5s sonication was simulated and the temperature distribution was recorded every 1s. The initial temperature was 31°C in order to compare with experimental results performed at the same temperature. Fig. 5 and Fig. 6 show the temperature distribution after a 5s sonication. The distribution was found to be highly inhomogeneous on the surface skull. Temperature elevation can reach up to 17 °C at several locations on the outer surface of the skull facing the transducers. High temperature elevations are observed on the outer surface, but one should notice that the inner surface is particularly preserved from overheating (Fig. 6). In the inner table, the temperature elevation was found to be between 1°C and 4°C.

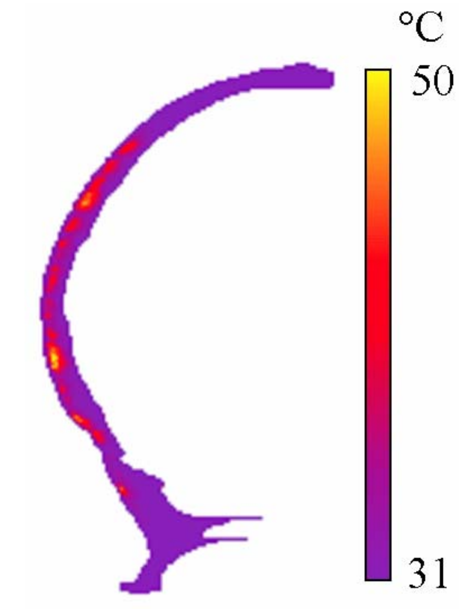

Figure 6. Temperature distribution in a slice of the skull bone, after a 5s sonication.

B. Experimental Validation

The hydrophone was removed in the experimental setup, and a 5s sonication was performed with a focal intensity of 1200 W.cm^{-2}. During the sonication, the temperature was recorded by the thermocouples with a repetition time of 0.2 s.

The experimental temperature is compared to the simulation. As shown in Fig. 7 the simulation is in good agreement with the experiment. The average error was found to be less than 1.5°C for the 8 different locations, on the outer surface, on the inner surface or in the diploë.

Finally, a simulation was performed with *in vivo* conditions. The initial temperature was set to 37 °C and the blood perfusion to 0.02 mL/min/mLof tissue. An external cooling system was set on the outer surface of the skull bone by maintaining external water at a constant temperature of 10°C. A 5s sonication was computed with an intensity at focus of 1200 W.cm^{-2}. The temperature at focus was simulated and reached 65 °C. The temperature in the skull was computed during the whole sonication. The mean temperature elevation in the skull bone was found to be 2.6°C and the standard deviation was 5.3°C. The temperature distribution is shown on Fig. 8 for different sonication times.

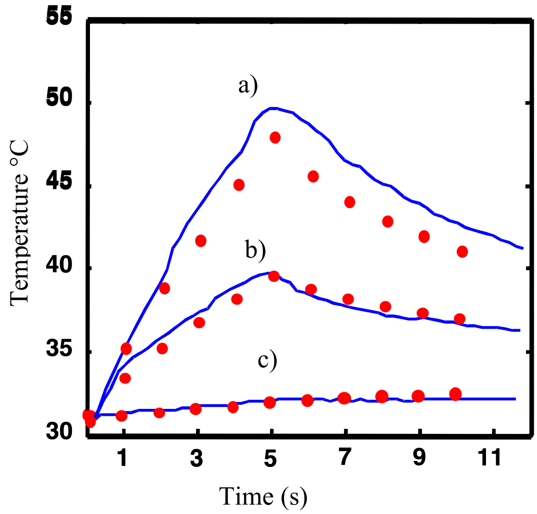

Figure 7. Temperature measured (- lines) and simulated (●) at 3 different locations located a) on the outer surface b) inside of the skull and c) on the inner surface.

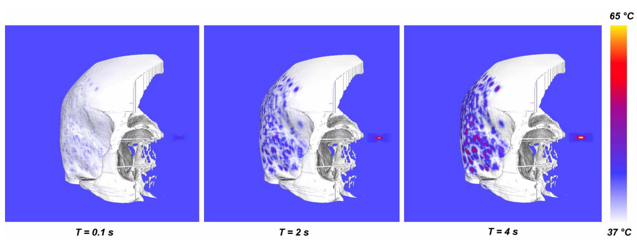

Figure 8. Temperature distribution on the outer skull surface achieved after a) 0.1s b) 2s c) 4s sonication.

V. CONCLUSION

A non-invasive procedure was achieved for HIFU brain therapy. High-resolution CT scans were used for the correction of the skulls aberrations, and secondly for the prediction of the temperature elevation during a HIFU treatment. The acoustic wave propagation software was coupled to the bio-heat equation and the temperature elevation in the skull bone was computed. The models were experimentally validated by thermocouples measurements. It shows that the temperature elevation is highly inhomogeneous inside the skull. These results show also that an external cooling system must be used to prevent skin burns.

VI. REFERENCES

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