RELATIONSHIP BETWEEN THERMAL AND ULTRASOUND FIELDS IN BREAST TISSUE IN VIVO

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The study shows the direct relationship between the temperature field and the parallel changes that are taking place in backscattered ultrasonic signals from breast tissue in vivo when heated to temperature of approximately 42°C. The non-uniform temperature field inside the tissue being heated was determined by a numerical model using FEM. It is shown that the spatial distribution in some areas of intensities of the backscattered signals coincides with the temperature distribution field predicted by the numerical model. The result indicates the possibility of indirect measurement of the temperature rise in breast tissue in vivo by measuring the intensity variations of the ultrasound echo.

INTRODUCTION

The special methods of Quantitative Ultrasound (QUS), which is a useful tool for soft tissue differentiations, cf. [13], were adapted to the detection of temperature variations inside specimens, cf. [5, 9, 2]. In recent decades the links between temperature variations and different parameters which measure the variations in RF (Radio Frequency) signals backscattered from the internal region of samples were studied. All experiments carried out to evaluate the possibility of the temperature field visualization inside the soft tissue, with the help of backscattered ultrasound signals, have, until now, been performed only on soft tissue phantoms and samples of soft tissue *in vitro*. The paper is the first one where the feasibility of temperature registration with the changes in intensities of backscattered signals by data from the *in vivo* experiment are studied. The problem of temperature distribution inside breast tissues is crucial in the field of breast thermography. Numerical models applied to modeling the temperature fields inside the breast tissue with different lesions in natural body temperature conditions were proposed in [1, 6, 11, 10, 14]. A basic hypothesis about existence of differences in heat production in healthy and diseased tissues was posed. The main goal of research in the above mentioned papers was the identification of lesions inside the breast by measuring the temperature distribution on the skin surface.

Our considerations are radically different. The skin surface was heated *in vivo* and we registered the ultrasound echoes from the tissue before and after the heating. Moreover, we observed the reaction of internal regions of the breast to skin surface heating. The numerical modeling of this heat transfer process together with special ultrasound characteristics yielded the possibility of differentiate the lesion region from healthy tissue regions. The paper is organized as follows: First, the temperature field inside the breast was calculated with the help of FEM (Finite Element Methods). Similar modelling of the temperature field inside the tissues was performed by us in [3, 4, 8] for different physical boundary value problems. Secondly, the mean intensities of backscattered signals in nine regions, before and after heating, were calculated. Furthermore, the local signal phases were also calculated for two different thermal data sets arising from different circumstances. Variations caused by the temperature rise, tied to the level of signal intensities, as well as in-phase distributions, are discussed. The links between the temperature field predicted by the numerical analysis and spatial variations of intensities are discussed. Section I introduces the experimental materials and methods and Section II presents the results. In Section III, conclusions are drawn about the contributions of this study.

1. MATERIAL AND METHODS

An ultrasound imaging system (SonixTOUCH, ULTRASONIX, BritishColumbia, Canada), equipped with a 128-elements linear transducer (L14-5/38), was used for acquisition of the ultrasonic radio frequency (RF) echoes from the patient breast, operating at 10 MHz center frequency. The patient breast had been heated 6 minutes through the skin by a thermal sack containing the warm water of temperature c/a 45°C. The FR signals, from which were made ultrasound B-mode images, were collected before and after heating. The USG were taken by a medical doctor from regions of the breast in which the well-recognized cyst is visible, see Fig. 1. All signal processing performed below was done with use of Matlab software (Mathworks Inc., Natick, Massachusets, USA).

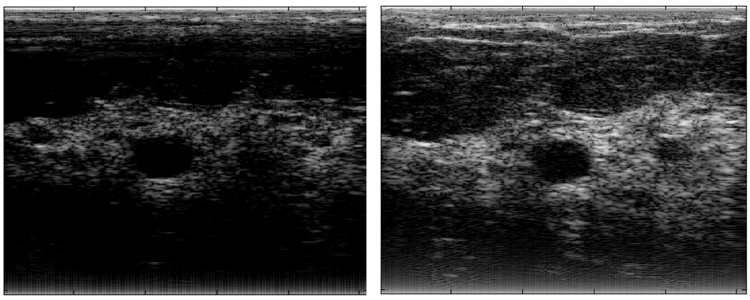

Fig. 1. Two B-mode images left: normal temperature, right: 6 minutes after heating.

2. BIOHEAT TRANSPORT IN HEATED BREAST

The breast tissue content can be reconstituted from the B-mode interpretation. The area of the breast considered includes the skin layer of 2 mm, unevenly distributed fatty tissue of 2 mm - 9 mm, breast tissue (gland breast tissue), and the cyst filled with fluid of about 5 mm x 7 mm. The cyst is located at a distance of 19 mm from the surface of the breast. The mathematical model of the heat transfer in the breast during the heating experiment was based on the bioheat equation proposed by Pennes, see [12], with spatially inhomogeneous coefficients. The simplified geometry of the 2D problem is presented in Fig. 2.

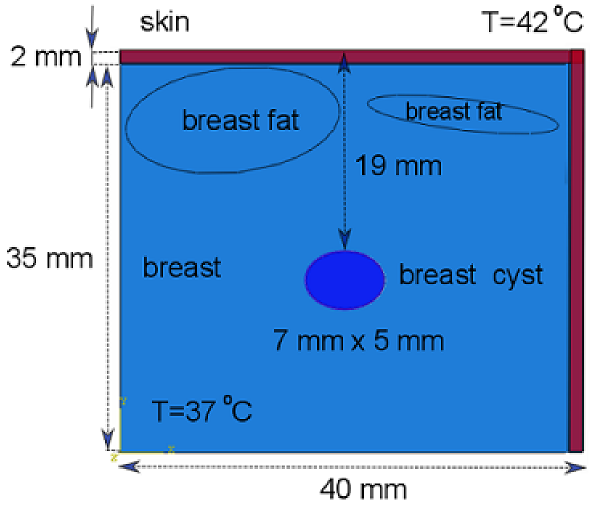

Fig. 2. 2D simplified geometrical model used in FEM calculations.

General form of the bioheat transfer equation in an inhomogeneous thermally anisotropic medium, occupying domain V in 3D real space, may be written as

$$\rho(\mathbf{x})C(\mathbf{x})\frac{\partial T(\mathbf{x},t)}{\partial t} = \nabla \cdot K(\mathbf{x}) \cdot \nabla T(\mathbf{x},t) + Q_p(\mathbf{x},t) + Q_{\text{int}}(\mathbf{x},t) + Q_{\text{ext}}(\mathbf{x},t), \qquad \text{for} \qquad \mathbf{x} \in V, \quad (1)$$

where T, t, ∇, ρ, K, Q_p, Q_{int}, Q_{ext} denote, respectively, temperature, time variable, gradient vector, density, specific heat, thermal conductivity of a medium in the case of a 2nd order tensor, perfusion, internal heat generation and external heat sources. On the basis of the experiment described above, the initial boundary value problem of the bioheat Eq. (1) was stated. When external heat sources were applied, thermal transport was caused by a temperature of 42^{o}C applied to the skin boundary, the right upper edge of the rectangle in Fig 2. The rest of the boundary preserves constant body temperature of 37^{o}C, which was the initial temperature at $t = 0$ for the whole domain V. The medium under consideration consists of four materials occupying the domain $V = V_s \cup V_f \cup V_g \cup V_c$, V_s – skin tissue, V_f – fatty tissue, V_g – breast gland tissue and V_c – cyst tissue, respectively, cf. Fig. 2. The initial boundary value problem for Pennes equation (1) was solved by FEM. Different physical properties of three components of breast tissue, namely fat tissue, skin tissue and gland breast tissue, necessary for the numerical calculation, are given in Tab. 1 and Tab. 2, [7, 15, 16]. The cyst is assumed to be filled with fluid and its physical properties are the same as properties of water.

3. RESULTS

In Fig. 1 it is easy to recognize that the B-mode image after heating is much lighter, which implies that the amplitude of the RF signals has higher values with increasing temperature. To describe these changes, the intensity of the ultrasound field were determined in 9 rectangular regions marked in Fig. 3. The cyst is located in the center of the 8 surrounding areas.

Tab. 1. Physical material parameters of three types of breast tissue

Tissue	Density $[kg/m^3]$	Conductivity $[W/m \cdot K]$	Specific heat $[J/kg \cdot K]$
Breast gland	1060	0.42	3540
Breast fat	911	0.21	2348
Breast cyst	1000	0.6	4200
Skin tissue	1109	0.37	3391

Tab. 2. Biofunctional parameters of three types of breast tissue

Tissue	Blood perfusion rate $[ml/s]$	Metabolic heat generation rate $[W/m^3]$
Breast gland	0.00018	700
Breast fat	0.00009	450
Breast cyst	0	0
Skin tissue	0.0001	368

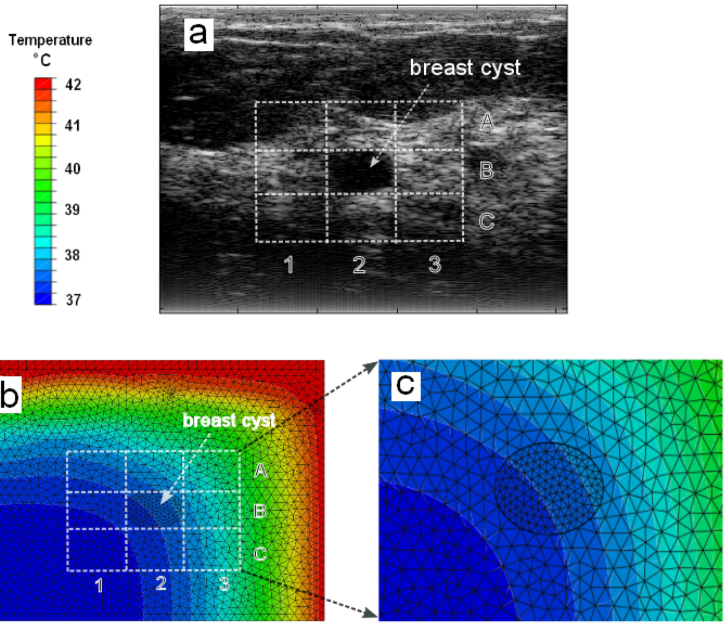

Fig. 3. (a) selected areas on the ultrasound image, (b) the same areas highlighted in the FEM model, (c) bigger FEM image with better visibility of the cyst.

Acoustic intensity is defined as the average rate of energy transmission per unit area perpendicular to the direction of propagation of the wave. Its relation with amplitude A can be written as

$$I = A/2\rho \cdot c,$$

where ρ and c denotes the equilibrium density and the speed of sound, respectively. The physical properties of a tissue change with temperature. Fortunately, in the range 37^{o}C – 42^{o}C, the changes in the density and speed of sound are negligible. They are of the order of 0.2‰ – 1.3‰ [8]. Thus, the ratio of intensities in the heated areas (lower index – hot), denoted by r to the intensity in the body temperature (lower index – cold) are equal to the ratio of the amplitude squares:

$$r = \frac{I_{hot}}{I_{cold}} = \frac{{A_{hot}}^2}{{A_{cold}}^2}.$$

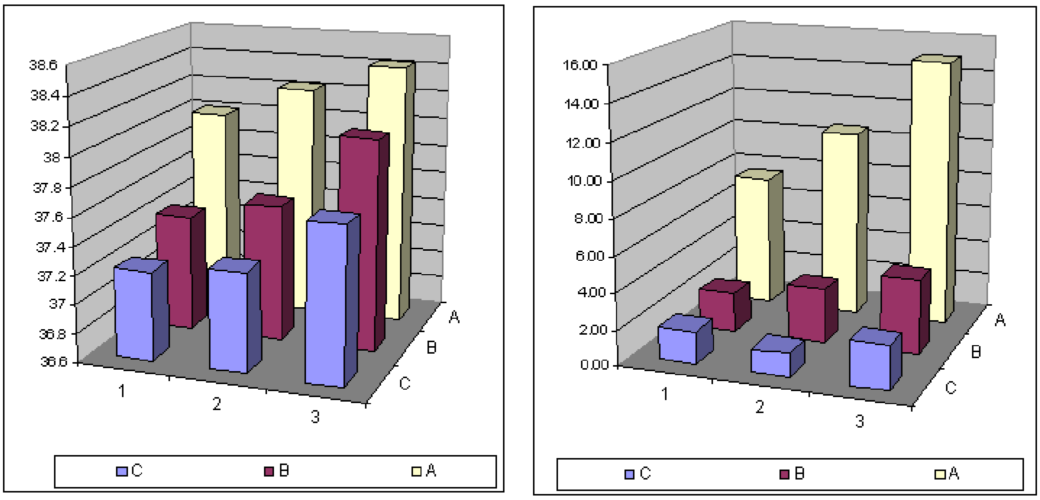

Fig. 4. The comparison of temperature level, left, after 6 minutes of heating calculated by FEM and r value of RF signals intensities for 9 regions surrounding the cyst, right.

4. CONCLUSIONS

Two methods of imaging living breast tissue being heated in vivo were considered. The first, based on FEM calculations, helped to create a 2D map of temperature field which was obtained by the solution to the initial-boundary value problem of the bioheat transport equation. The second method, based on analysis of the backscattered RF signals obtained during the experiment, helped to create a map of the acoustic intensity variations. The changes in intensities were calculated as the ratio of the intensity of the ultrasound signal in the heated tissue to its intensity in the body temperature. The comparison was done on a grid of nine contiguous rectangles forming a large rectangle with the central rectangle containing the cyst. The calculated mean values of the temperature and the relative intensities in these areas are shown in Fig. 3. Both approaches lead to qualitatively similar result, see Fig. 3. The 'maps' are qualitatively consistent: an increase in temperature in an area correlates with an increase in the intensity ratio (except for an area 2C). This unique region 2C is located under the cyst, see Fig. 2. The cyst has different acoustical characteristics and less value of a scattering coefficient from those of normal breast tissue. That particular inconsistency in maps possibly arises from the effect of reflection or refraction of the acoustic wave at the boundary between the breast tissue and the cyst. Accuracy of the calculated fields of the temperature distribution and variations in the backscattered signal intensities show that the temperature rise can be indirectly measured by the ultrasonic signal collected from the breast *in vivo*. Besides, the marked improvement in the B-mode image quality makes it advisable to consider a thermal process in order to improve ultrasound diagnosis of breast tissue *in vivo*.

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