ISSN 1413-3555
Rev Bras Fisioter, São Carlos, v. 12, n. 3, p. 204-14, may/june 2008
Revista Brasileira de Fisioterapia©

Didactic software for modeling heating patterns in tissues irradiated by therapeutic ultrasound

Software didático para modelagem do padrão de aquecimento dos tecidos irradiados por ultra-som fisioterapêutico

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Abstract

Introduction: Ultrasound is a resource commonly used in Physical Therapy. However, its inadequate application may produce insufficient heating or cause damage to biological tissues. Therefore, the knowledge on the optimum parameters for achieving the appropriate temperature, within safe limits, is necessary. Heat generation depends on equipment parameters and the physical properties of tissues. This study presented a software that simulates the energy and temperature variation in tissues over time, thus allowing users to view the heating patterns in tissues as a function of these parameters. **Methods:** The software was implemented based on the bioheat transfer equation for four layers (skin, fat, muscle and bone), in which the user can change the thickness and thermal or acoustic properties of these tissues. The intensity, frequency and time of application can also be chosen. Graphs showing the percentage energy absorption in relation to depth and the respective temperature variation per millimeter of tissue are presented. **Results:** Simulations were produced to give examples of situations of interest for therapy, by varying the time of application, thickness and ultrasound frequency. Differences in heating patterns are seen, especially at the interfaces. **Conclusions:** The software made it possible to study the heating of biological tissues by ultrasound and can be used both for teaching purposes and for planning heating doses for continuous waves. In the future, the software will be adapted, in order to estimate which dose should be regulated in the apparatus to maintain the desired temperature for the time chosen. **Software available in:** http://www.peb.ufrj.br/lus.htm.

Key words: physical therapy; education; ultrasound; thermal field.

Resumo

Introdução: O ultra-som é um recurso bastante utilizado em Fisioterapia. Entretanto, a aplicação inadequada pode promover aquecimento insuficiente ou causar danos aos tecidos biológicos. Por isso, é importante que se conheçam os parâmetros ótimos para atingir a temperatura adequada, dentro dos limites seguros. A geração de calor é função dos parâmetros do equipamento e das propriedades físicas dos tecidos. Este trabalho apresentou um software que simula a variação da energia e da temperatura nos tecidos ao longo do tempo, permitindo ao usuário visualizar o padrão de aquecimento nos tecidos em função dos parâmetros. **Materiais e métodos:** O software foi implementado com base na equação biotérmica, supondo quatro camadas (pele, gordura, músculo e osso), das quais o usuário pode alterar espessura e propriedades acústicas e térmicas. Pode-se também escolher intensidade, freqüência e tempo de aplicação. São apresentados gráficos com o percentual de energia absorvida ao longo da profundidade e a respectiva variação de temperatura por cada milímetro de tecido. **Resultados:** Foram realizadas simulações exemplificando situações de interesse para a terapia, variando tempo de aplicação, espessura e freqüência do ultra-som. Podem ser observadas as diferenças do padrão de aquecimento, em especial nas fronteiras das interfaces. **Conclusões:** O software permitiu o estudo do aquecimento de tecidos biológicos por ultra-som e pode ser usado tanto para fins didáticos como para planejamento de doses de aquecimento, para ondas contínuas. Numa próxima etapa, pretende-se adequá-lo para estimar qual dose deve ser regulada no aparelho, para manter a temperatura desejada pelo tempo escolhido. **Software disponível em:** http://www.peb.ufrj.br/lus.htm.

Palavras-chave: fisioterapia; ensino; ultra-som; campo térmico.

Received: 04/07/2007 – **Revised:** 03/12/2007 – **Accepted:** 25/04/2008

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Parts of this work has been previously published at the Proceedings of the Brazilian Congress of Biomedical Engineering 2006 (see reference 12) and an abstract has been presented at the III International Congress of Physical Therapy, 2007 (see reference 13).

Introduction

In Physical Therapy, thermotherapy treatments are usually classified as deep or superficial, and in both cases heat transfer to the tissues may occur in three different ways: by conduction, convection or radiation[1]. In the case of deep thermotherapy treatments, in which tools (ultrasound, microwave, and short waves) are capable of heating internal tissues with little influence on superficial tissues, the main form of heat transfer is radiation (diathermy)[2].

The physiological alterations produced by therapeutic ultrasound (TUS) on biological tissues are traditionally grouped into two categories: thermal and mechanical (non-thermal) effects. Both occur on the body, but their rate and magnitude depend on the supply cycle and on the output intensity[3].

Since its development in the 1950s, TUS has been used to promote heating in the treatment of musculoskeletal disorders. When tissues reach a temperature range of 40 to 45°C for approximately five minutes, the following biological effects are triggered increased metabolism, pain relief, reduced joint stiffness and increased blood flow[4]. Temperatures above this threshold may cause cell damage, whereas lower temperatures are unlikely to promote the desired effects.

Ultrasound heating results from the interaction between the acoustic and thermal properties of the tissues and the irradiated wave. In order to use TUS safely and efficiently, the physical therapist must know the number, sequence, and thickness of the tissue layers to be irradiated, besides the following physical properties of each tissue: acoustic impedance, absorption coefficient, specific heat, and thermal conductivity. As for the source of irradiation (TUS equipment), the professional must define wave frequency (1 or 3MHz), mode of emission (continuous or pulsed), intensity, and length of treatment.

In clinical practice, the physical therapist usually selects the dosage (intensity and length of treatment) based on the information contained in the equipment's instruction manuals, which simply recommend dosage tables according to each pathology with no adequate reference to the scientific literature[5]. Electrotherapy textbooks only provide general knowledge of ultrasound and its effects on biological tissues[3,6], leaving the physical therapist to learn the method empirically. It should be noted at this point that the latest studies have shown a systematic procedural error in TUS application that has a deeper cause, namely the lack of theoretical development and of controlled experiments to evaluate and even to quantify the efficiency of ultrasound in physical therapy[7-10].

Several mathematical models have been implemented to aid the understanding of energy and temperature behavior in continuous ultrasound beams[11,12]. Nevertheless, the application of these methods frequently requires advanced mathematical knowledge and complex software that is not accessible to health professionals and comes with pre-established, hard to manipulate and often unalterable physical parameters.

Given this scenario, this paper puts forward an educational software that simulates the ultrasound heating pattern of tissue, based on a simplified mathematical model of the biothermal equation (Bioheat Transfer, BHT)[13]. This equation describes the behavior of temperature in tissues over time, at different depths. The simulator contains adjustable acoustic as well as thermal parameters, and allows the operator to visualize the evolution of energy and temperature distribution to the tissues (skin, fat, muscle, and bone) during the ultrasound propagation. The purpose of this software is to provide physical therapists with a tool to educate them about the basic physical phenomena involved in ultrasound heating, thus improving their ability to evaluate the dosages suggested for each case.

Materials and methods

This section consists of two parts. In the first part, the model used to create the software is briefly described for the benefit of readers interested in the mathematical formulation of the problem. In the second part, the simulator interface is introduced, showing the four layers of tissue in different types of graphs and the user-defined parameters.

Mathematical modeling

The implemented model was obtained from the equation that describes the temperature variation of a body according to the amount of heat transferred[14], and its detailed description can be seen in the reference section.

The model stems from the principle that the amount of energy responsible for the heating is related to the ultrasound intensity $I(x)$ supplied to the tissue along its depth x (mm), which is described by equation (1).

$$I(x) = I_o . e^{-2.\alpha . f . x} \quad (1)$$

where I_o is the initial intensity, given in $W.cm^{-2}$, e is the neperian base, α is the coefficient of tissue attenuation ($Np.mm^{-1}$), and f is the ultrasound frequency (MHz). The amount of energy $\Delta I_{(x)}$ ($W.cm^{-2}$) that remains in a Δx fraction of the layer of tissue after TUS application is obtained by the difference between the intensity that goes into and the intensity that comes out of that layer, according to equation (2)[15].

$$\Delta I_{(x)} = I_o\left(e^{-2.\alpha.f.(x+\Delta x)} - e^{-2.\alpha.f.x}\right) \quad (2)$$

Thus, we have a one-dimensional, simplified version of the biothermal equation (BHT), commonly used to simulate ultrasound heating, which may be seen in equation (3). For the sake of simplicity, it does not take into account the loss by blood perfusion, and the temperature variation per space unit is considered equal to the temperature variation per time unit, which may be applied to thin mediums (measured in millimeters)[16,17].

$$\theta_f = \frac{Io.(e^{-2.\alpha.f.(x+\Delta x)} - e^{-2.\alpha.f.x}).t}{\rho.c\Delta x^2 + k.t} + \theta_o \quad (3)$$

where, θ_f is the final temperature, after t seconds of heating. The ρ parameter is the tissue density (g.cm^{-3}), c is the specific heat (J.g^{-1}.ºC^{-1}), k is the coefficient of thermal conductivity (W.cm^{-1}.ºC^{-1}) and θ_o is the initial temperature of the human body.

The equations were used to simulate TUS application to an area of the body with four tissue layers, throughout its depth (with 1mm increments) and time (with 1s increments), assuming a fixed (head) transducer is used. It is important to note that, given the way equation (3) was implemented, one can simulate thicker tissues, and the software itself internally subdivides each layer in consecutive 1mm thick slices before applying the equation in question. This procedure is transparent to the user.

The simulation software

When the physical therapy ultrasound is applied, its beam reaches up to four layers of tissue: skin, hypoderm (fat), muscle, and bone[18]. The software was developed with the use of LabVIEW® 7.1 (National Instruments®), implementing these layers (according to tissue properties, as in Table 1[19,20]), and simulating the passage of a flat wave according to the equations previously described. The software allows the user to define the following parameters of each layer: thickness (x), attenuation coefficient (α), acoustic impedance (Z), density (ρ), specific heat (c), and thermal conductivity (k). Thus, in order to simulate other tissues (dense connective tissue, such as tendons), organs (liver, spleen), or even prosthetic material such as metal (titanium), the user simply needs to have the properties requested by the software in relation to the simulated layer.

The software allows the user to define the values selected in the equipment in order to carry out the therapy: transducer frequency (f), initial applied intensity (I_o), and time of application (t). The initial temperature of the tissue (θ_o) is standardized at 36ºC, the average human body temperature, but it can also be altered in the simulation, according to user needs. The software displays a screen with 12 graphs distributed on a 3x4 matrix (Figure 1). In the first row are the graphics for intensity variation by tissue depth $I(x)$ (as in equation (1)), which in addition to representing the percentage loss of ultrasound intensity during its passage (continuous line) also show the amount of energy (I_f) that arrives at each interface. The dashed line represents the percentage loss of intensity, after suffering one reflection. The Et(%) parameter indicates the total percentage of energy that remained in the tissue layer, after the passage of the ultrasound (including one reflection).

In the second row of graphs, the gray scale (in the software the scale is in color) represents the variation of the amount of energy $\Delta I(x)$ which remains in each Δx of 1mm thick tissue, after the passage of the ultrasound (according to equation (2)). The colors vary from black (minimum) to white (maximum), covering different hues of blue, depending on the intensity of energy deposited in each millimeter of each layer.

In the third row of graphs, there is a representation in gray (the software shows it in color scale) of the final θ_f temperature in each millimeter of x thickness, after the irradiation for t seconds (as shown in equation (3)). The color code varies from blue, yellow, and orange to red, according to rise in temperature. A column analysis yields graphs that represent the tissues through which the wave will pass: skin, fat, muscle, and bone. In the upper part of the screen, there are control buttons that allow the user to alter the absorption coefficient, thickness, and the acoustic impedance of each tissue (Figure 1). The percentage of reflected intensity (R1, R2 and R3) and transmitted intensity (T1, T2 and T3) in the interface between mediums is automatically calculated, based on the acoustic impedance difference among them, and can be seen on the top margin of each graph in the first row.

Table 1. Thermal and Acoustic Tissue Properties.

Tissues	α (Np.mm^{-1}.MHz^{-1})	Z (x10^6 kg.m^{-2}.s^{-1})	ρ (g.cm^{-3})	c (J.g^{-1}.ºC^{-1})	k (W.cm^{-1}.ºC^{-1})
Skin	0.024	1.87	1.20	3.59	0.0023
Fat	0.007	1.37	0.95	2.67	0.0019
Muscle	0.011	1.65-1.74	1.04	3.64	0.0055
Bone	0.150	3.75-7.38	1.38-1.80	1.25	0.0230

The loss of intensity along the tissue is not constant; it decreases exponentially (as seen in equation (1)). Hence, the four intermediate graphs represent the distribution of percentage energy $\Delta I(x)$ absorbed by each layer having a Δx thickness, along the tissues. This energy is used in equation (3) to calculate the final temperature of each 1mm Δx slice in the tissue and to show the temperature distribution over time (four bottom graphs in Figure 1).

Results

Below are examples of four simulations of interest to physical therapy professionals, showing the potentiality of the software to communicate the influence of the physical properties of the tissue and of the ultrasound wave.

First simulation: influence of bone density variation on tissue heating

According to the literature[18], acoustic impedance depends on the density and speed of the wave in the medium (Z=$\rho.c$). Because bone tissue varies in density from 1.38 to 1.81g.cm^{-3}) and in acoustic impedance from 3.75 to 7.40kg.m^{-1}s^{-2}, it is important to know the influence of the variations of these properties on tissue heating. To do so, two simulations were conducted, using the four consecutive tissue layers, the skin being 4mm thick, and the other tissues, 10 mm thick. They were all irradiated for 180 seconds, with an intensity of 2.0W.cm^{-2} at a frequency of 1MHz. In the first simulation (Figure 2), bone density (ρ) was 1.38g.cm^{-3} and acoustic impedance (Z) was 3.75kg.m^{-1}s^{-2}.

In the second simulation (Figure 3), bone density was 1.80g.cm^{-3} and impedance 7.4kg.m^{-1}s^{-2}. The influence of density and, consequently, acoustic impedance on the final temperature can be clearly seen. When these values were minimal, bone temperature reached a limit value of 43.25ºC; when values were maximal, the temperature rose to 40.63ºC. That shows the influence of density on acoustic impedance and consequently on the reflection of the energy (15.12% for minimal values, and 40.37% for maximal values) in the muscle-bone interface.

It should be noted that fat and muscle heating curves practically overlap when values are minimal (Figure 4), and muscle heating is higher in relation to fat heating due to reflection when values are maximal (Figure 5).

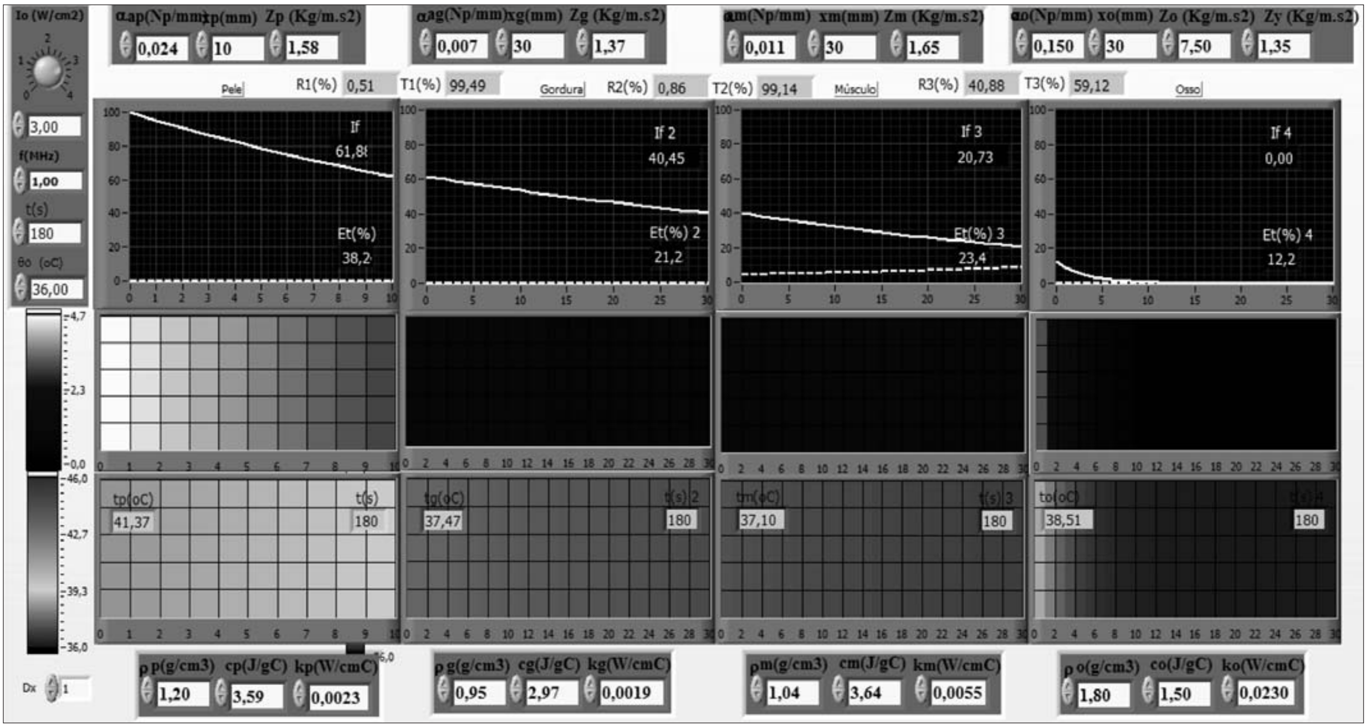

Figure 1. Software screen with the simulation of four tissue layers with TUS of 1 MHz and 3 W/cm2, during 180 seconds. The superior graphs show the intensity curve in each tissue layer. The middle graphs represent the energy distribution in each tissue layer, due to the ultra-sound absorption (in the software, the scale is in color). The lower graphs demonstrate the temperature maps generated due to the energy applied in each tissue layer (in the software, the scale is in color). The parameters for each layer are: α_i (absorption coefficient); x_i (thickness); Z_i (impedance); ρ_i (density); c_i (specific heat); k_i (thermal conductivity); θ_0 (oC) is the initial temperature and t(s) tissue exposition time to ultra sound.

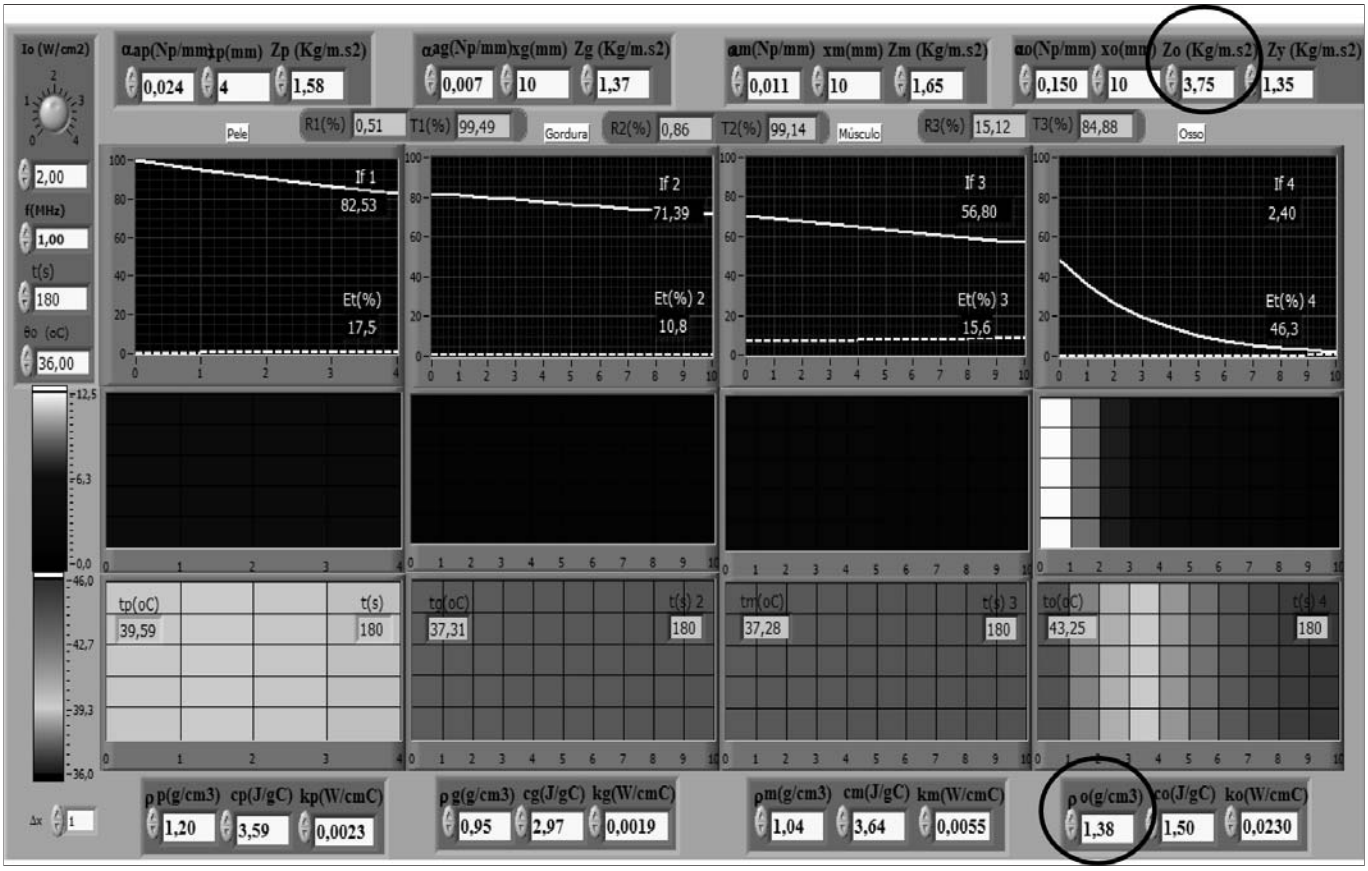

Figure 2. Software simulations with minimum values for bone density and acoustic impedance

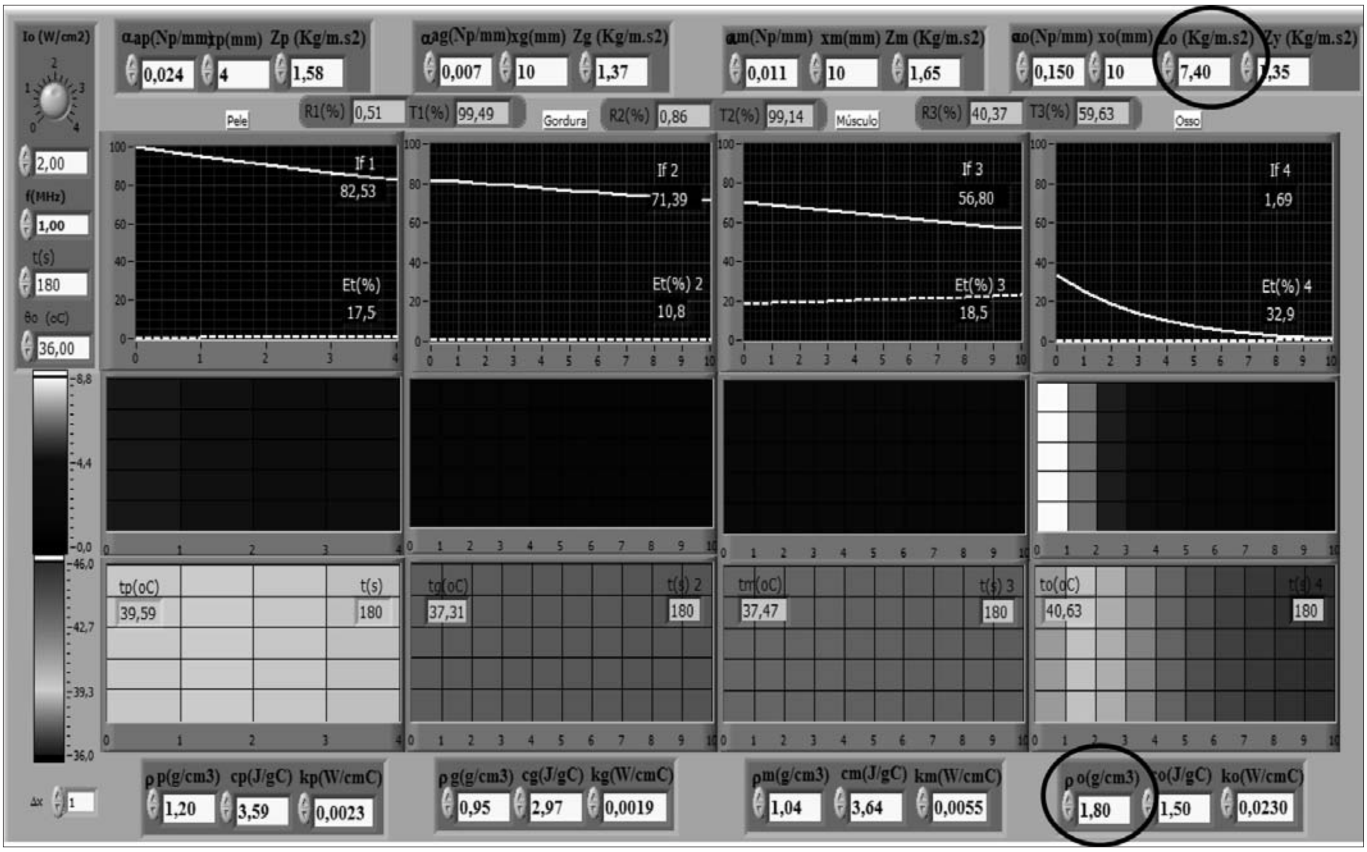

Figure 3. Software simulations with maximum values for bone density and acoustic impedance

Second simulation: influence of muscle thickness on heating

In this case, heat was applied for 180 seconds to skin 4mm, fat and bone 10mm, and muscle 30mm. The remaining values were kept according to the previous simulation, with maximal bone density and acoustic impedance values. It is common for the physical therapist to assume that the tissue that absorbs the highest amount of energy will heat the most. However in this simulation we found that, although the muscle absorbed 41.3% of energy and fat absorbed 10.8%, the final temperature in the outermost layer of these tissues was similar (Figure 6).

A temperature vs. time curve graph was used to verify the thermal behavior of these tissues (Figure 7). It is relevant to point out that the time of ultrasound application determines which of the tissues will heat more in relation to the other. When heating lasted 60 seconds, the final temperature of the bone layer (37.66ºC) was greater than that of the skin (37.27ºC). However, when the time of application was 160 seconds, the initial part of the skin layer heated more (39.22ºC) than that of the bone (38.84ºC). This phenomenon can also occur between fat and muscle, although this difference may be less significant.

Third simulation: influence of fat thickness on the heating of the other tissues

In order to compare the influence of fat thickness on the energy absorbed by the other tissues, its thickness was tripled in relation to the previous simulation, and the remaining parameters were kept constant. It is often claimed that the thickness of fat has little influence on the heating of other tissues because of its low absorption coefficient. The percentages of energy located in the muscle and in the bone had a significant reduction, whereas fat thickness increased to 30 mm (Figure 8). In the first case (Figure 6), the muscle and the bone absorbed 41.3% and 21.2% of the total energy respectively. However, when the fat layer was thicker (Figure 8), these percentages fell to 31.2% and 16% respectively. Nonetheless, the reduction of the final temperature in the most superficial layer of the muscle was 0.32ºC (from 37.29ºC to 36.97ºC), and in the bone, 0.73ºC (from 38.98ºC to 38.25ºC), which characterizes a light to moderate reduction.

Fourth simulation: influence of frequency on heating

It is commonly stated, in the field of Physical Therapy, that the deeper the treatment, the lower the frequency of the

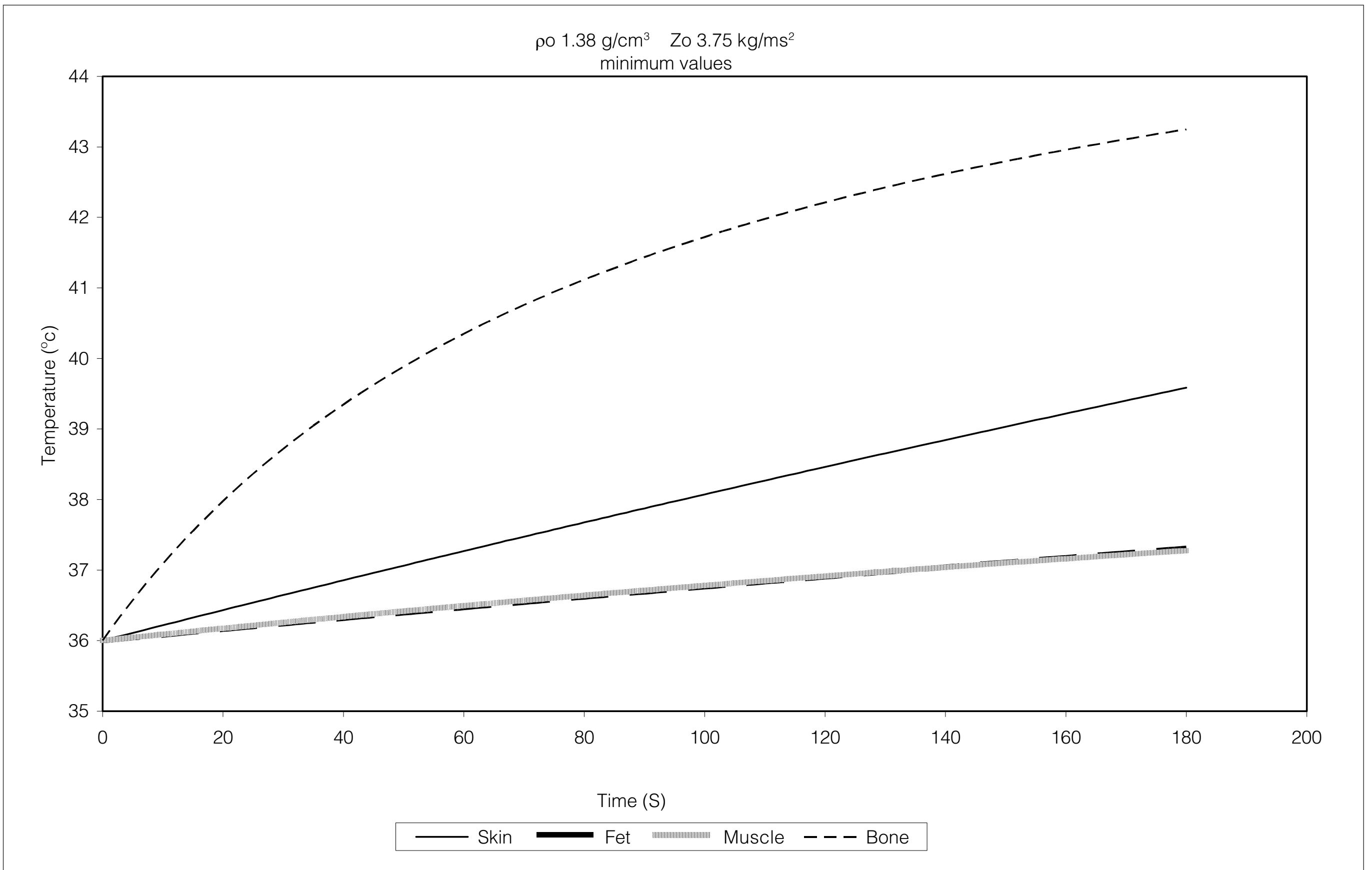

Figure 4. Graphs on the temperature time variation in the first millimeter of the skin, fat, muscle and bone layers for the minimum values for bone density and acoustic impedance.

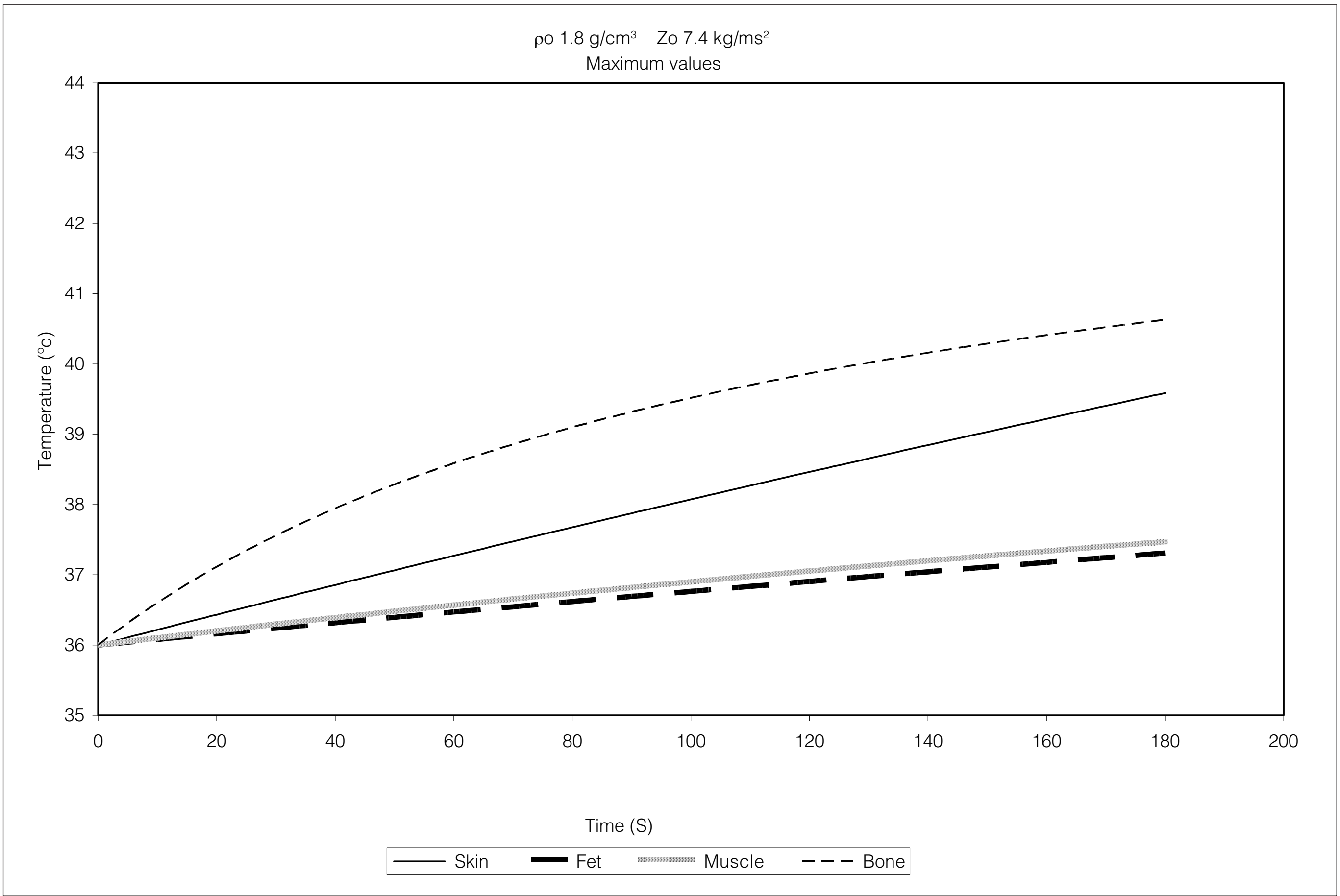

Figure 5. Graphs on the temperature time variation in the first millimeter of the skin, fat, muscle and bone layers for the maximum values for bone density and acoustic impedance.

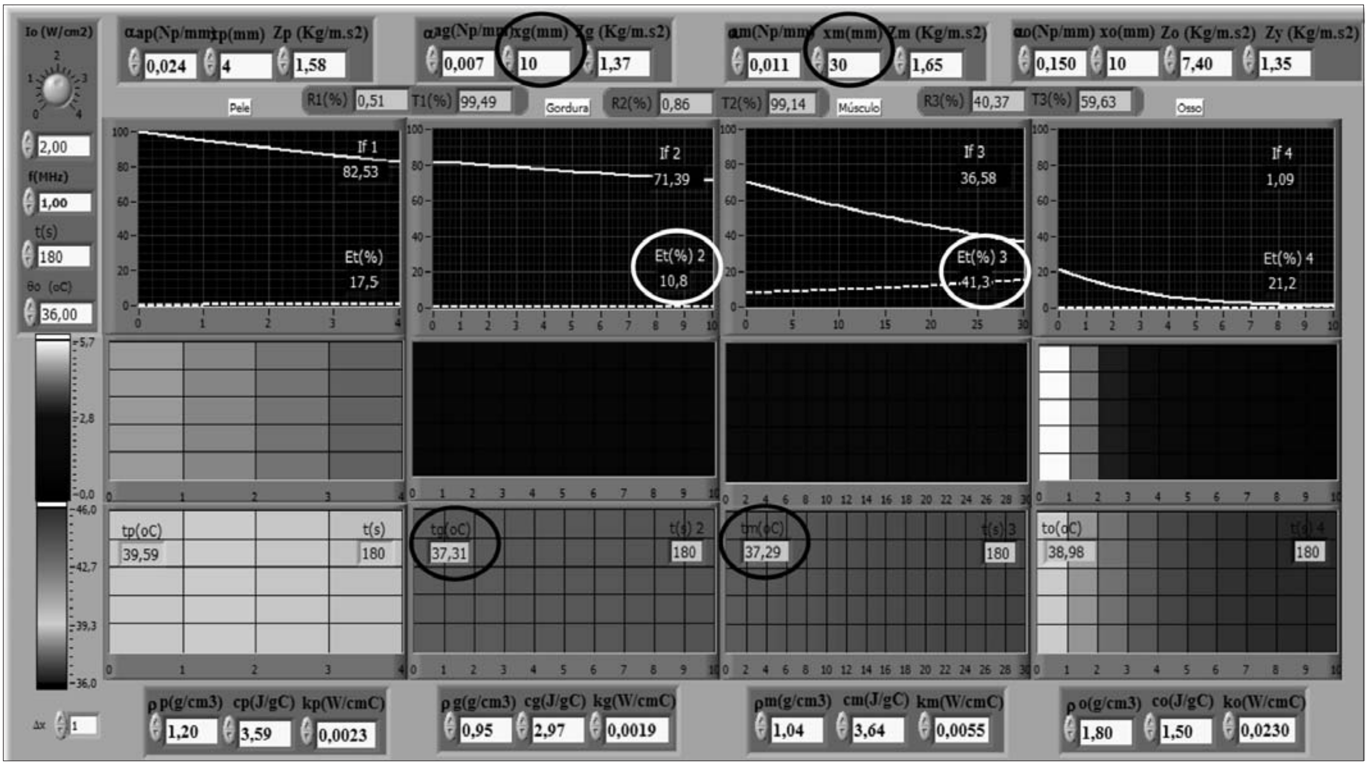

Figure 6. Software simulations with skin (4mm), fat (10mm), bone (10mm) and muscle (30 mm) thickness, during 180 seconds and 2.0 W/cm^2.

transducer (1MHz, in this case). Therefore, if a superficial treatment is required, the transducer should be used at higher frequencies (e.g. 3 MHz)[1]. For this simulation, the following parameters were used: 4mm for skin and fat, 30mm for muscle and 10mm for bone during 180 seconds at 3MHz. There was increased ultrasound energy absorption by the skin, and therefore greater heating (the final temperature was 46ºC) (Figure 9) when compared to the results obtained with a 1MHz frequency (Figure 8). That occurs because the coefficient of absorption increases with frequency[2] and with the amount of energy deposited in each centimeter of the most superficial tissues, possibly resulting in serious injury.

Discussion

We developed a software capable of simulating a biological environment with four layers of tissue. Its main purpose is to create conditions for physical therapists to develop their knowledge of the physical phenomena involved in ultrasound propagation and heating of biological tissue. This will allow these professionals to choose more suitable dosages and educate them about the evolution of the suggested treatment.

The implemented mathematical model takes into account the heat loss by conduction to the adjacent tissues but it still does not consider cooling due to blood perfusion. Therefore, the final temperature values may be overestimated, which does not invalidate the qualitative results obtained. Another limitation of the model is that the biothermal equation (BHT) has not been solved analytically, however we proposed an approximate solution by infinitesimal increments. In addition to that, the model does not predict the mobility of the transducer, as this form of application is difficult to be modeled given the various head shapes and movement speeds. However, these results with the fixed head allow the physical therapist to predict with certainty how long the ultrasound can be applied to the same treatment area to reach the desired temperature level.

Four simulations were carried out, exemplifying common situations of interest to the physical therapist. In the first one, we analyzed the effect of the variation of the reflection coefficient on the muscle-bone interface in relation to the variation of its impedance variation. When the impedance doubled, the

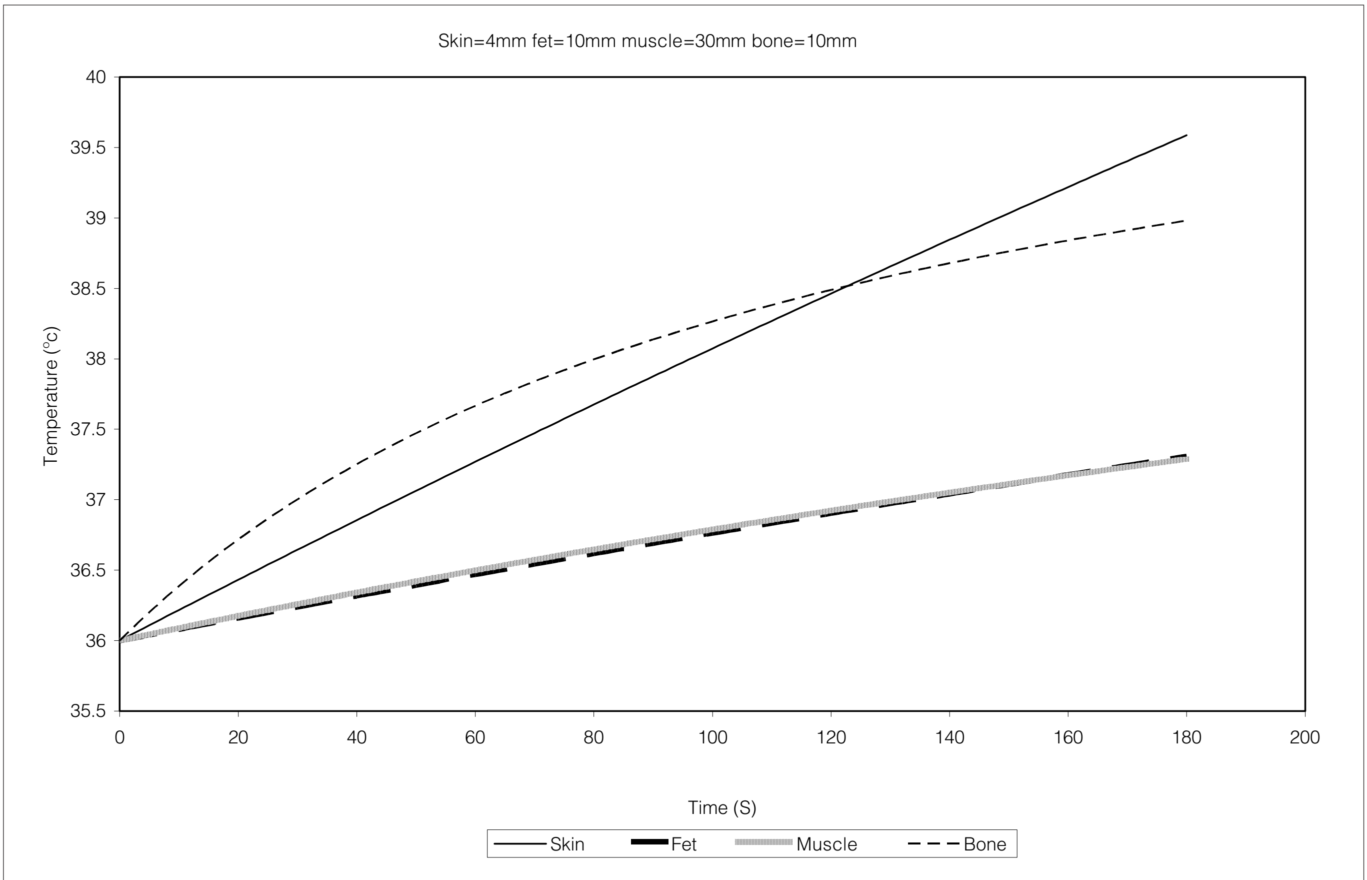

Figure 7. Temperature graphs at the first millimeter in the skin (4mm), fat (10mm), bone (10mm) and muscle (30 mm) thickness, during 180 seconds and 2.0 W/cm^2.

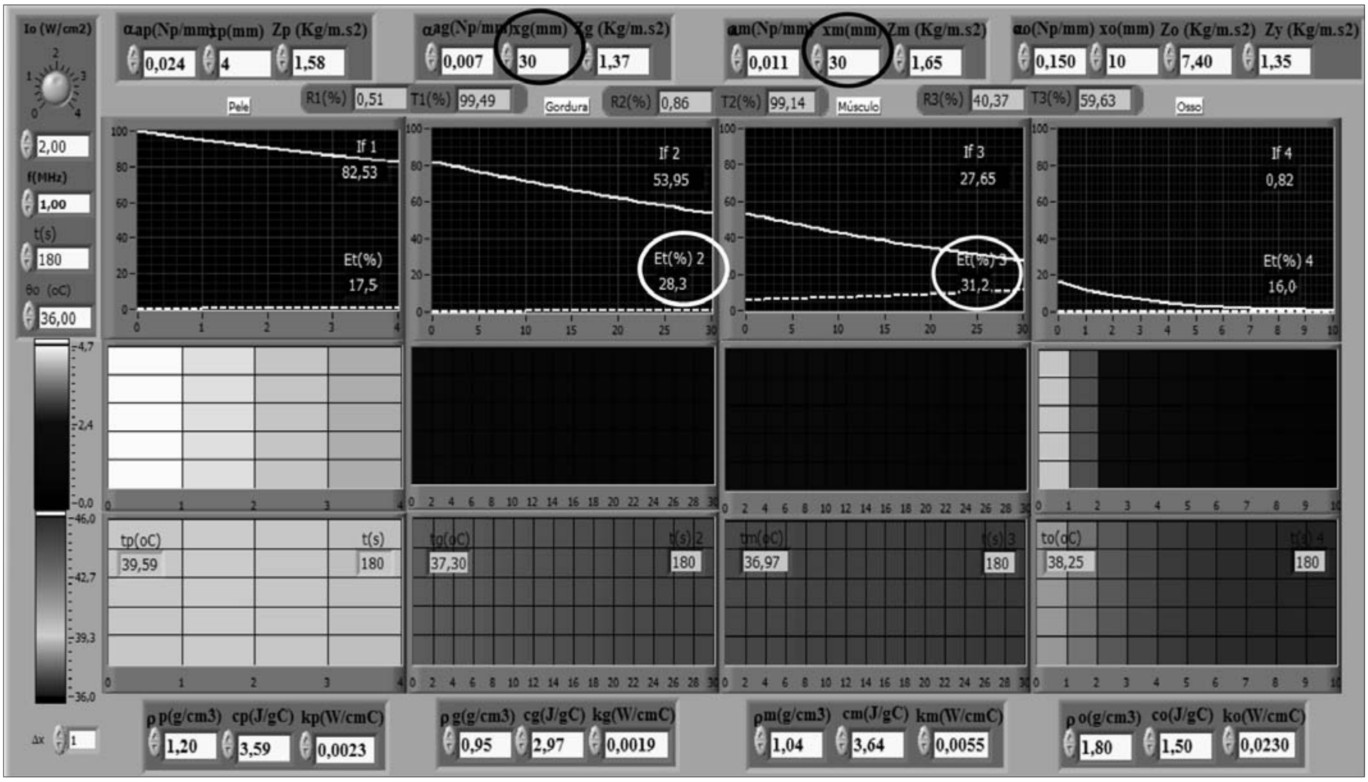

Figure 8. Software simulations with skin (4mm), fat (30mm), bone (10mm) and muscle (30 mm) thickness, during 180 seconds at a frequency of 1MHz.

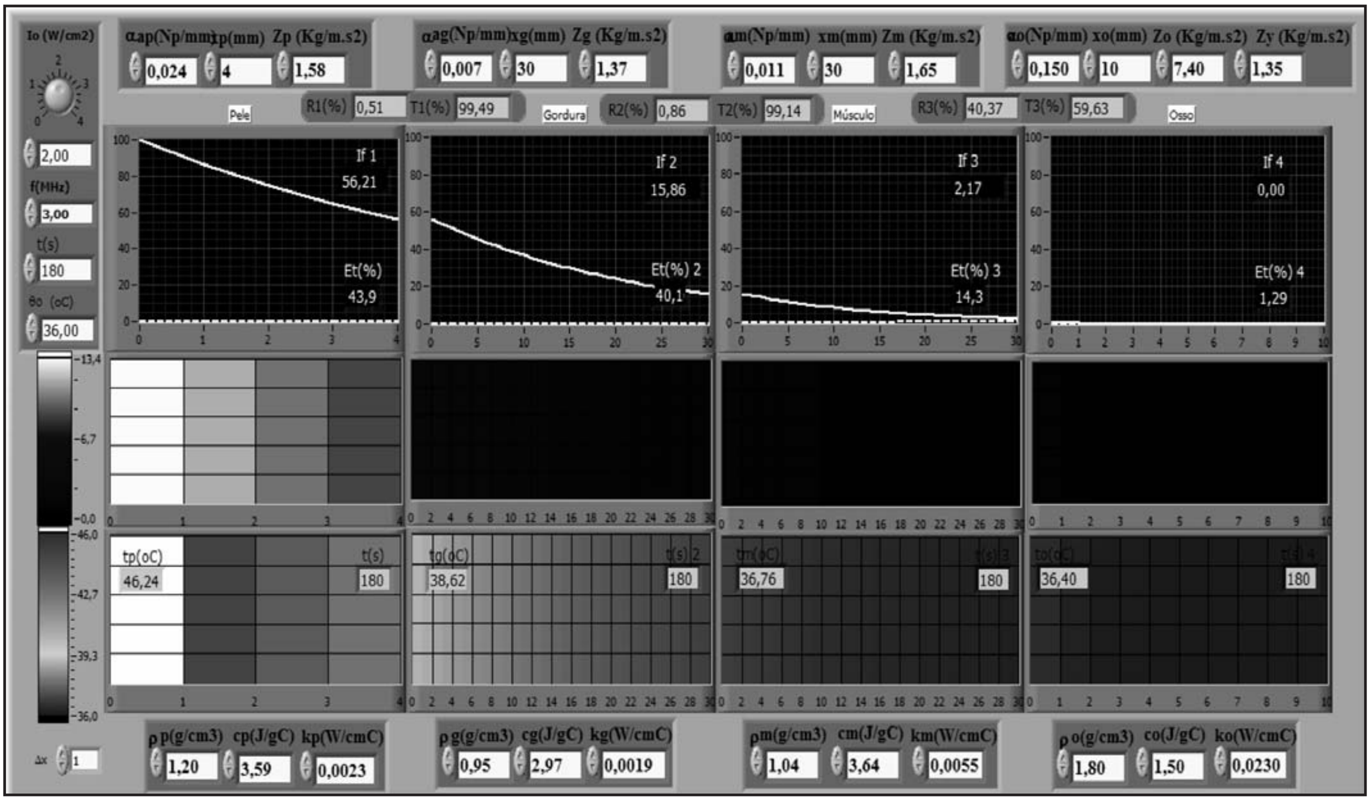

Figure 9. Software simulations with skin (4mm), fat (30mm), bone (10mm) and muscle (30 mm) thickness, during 180 seconds at a frequency of 3MHz.

reflection coefficient tripled, causing a reduction of approximately 2ºC in the superficial layer of the bone tissue.

In the second simulation, the tissue that absorbed the highest amount of energy did not necessarily heat the most. That is due to the fact that temperature variation in the medium does not depend solely on the absorption coefficient, but also on the amount of heat loss (thermal conductivity) and on the amount of energy it needs to elevate its temperature by 1ºC (specific heat). This simulation also showed that the tissue that heats the most is not always the same, but that this depends on the time of application. This shows that the heating produced by ultrasound propagation in the biological tissues is a complex phenomenon, and that the rules commonly found in instruction manuals to aid the professionals may be oversimplified and devoid of effective practical value.

In the third simulation, we analyzed the influence of the thickness of the fat layer on heating. When fat thickness was tripled, the temperature variation in the muscle was almost irrelevant, but in the bone it was moderate, which indicates that the thickness of the fat layer should also be evaluated, contrary to what is commonly claimed in this field.

In the last simulation, we studied the effects of change of frequency and found that, at 1MHz, the final temperature was higher in the skin and bone tissues, and that at 3MHz it was higher in the skin and fat tissues. According to the literature[4], there is significant heating in the most superficial tissues (skin and fat), however no attention is given to the fact that besides being more superficial, this heating is also more intense than it would be if the application was made using a lower frequency, with the same intensity. Because the attenuation is greater at higher frequencies, the percentage drop in energy in relation to depth is more significant. That causes the energy delta to be greater in the initial centimeters of the tissues, which in turn promotes a more pronounced temperature increase that may surpass 46ºC and cause protein denaturation. In short, for the same intensity and length of exposure, the heating is greater at 3MHz than at 1MHz.

In conclusion, the proposed simulation software was useful for the study of intensity and temperature distribution in biological tissues and may be useful for educational purposes such as regular Biophysics and Electrothermotherapy courses aimed at physical therapists. Though simple, the implemented model was capable of reproducing situations of interest to the professional who works with TUS.

The next steps are to develop the equation to include the losses by perfusion and to provide the simulator with the ability to calculate the necessary dosage (intensity and frequency) to produce a specific heating temperature at a given depth for a fixed period of time. This feature would allow the professional to prepare treatment protocols for testing.

We expect to produce an executable version of the software using the resources available in LabVIEW®. In future, it should be available to potential users online by accessing the Ultrasound Laboratory website of the Biomedical Engineering Program at COPPE (UFRJ).

Acknowledgements

To Brazilian Grant Agencies CNPq and Capes for the financial support.

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