A SIMULATION OF USING ULTRASOUND ENERGY DEPOSITION FOR HYPERTHERMIA

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ABSTRACT

A simulation of using ultrasound energy for hyperthermia treatment is presented . The biological medium considered is composed of layers of human skin , fat , muscle and bone. The temperature distribution in each layer is obtained numerically. The calculations are based on solving the bioheat transfer equation by the finite difference time domain (FD) method. The results showed that an adequate temperature elevation can be achieved inside the muscle layer by ultrasound energy deposition .

Keywords: Ultrasound, Hyperthermia, Bioheat transfer equation.

INTRODUCTION

Ultrasound is widely used for imaging and diagnosis process. Recently, the deposition of ultrasound for hyperthermia takes more considerations. Hyperthermia means a specific elevation of temperature at a certain location for a period of time. Several papers [1]-[3] have described methods and introduced applicators for focused ultrasound hyperthermia. One layer of human tissue was considered in those papers. In this paper a theoretical method is presented to introduce temperature elevation in a multilayered human tissues medium for hyperthermia treatment. Blood flow is not considered due to its nonuniform distribution during the heat treatment [4]. The finite difference method is chosen because of its applicability to simple and complex configurations to solve the bioheat transfer equation. In the first part of the present paper, the theoretical model and the bioheat transfer equation solution by the FD technique are described. The results obtained are then presented and discussed. The last section is devoted to the conclusion.

THEORETICAL MODEL

For the simulation, the ultrasound energy is produced by an ultrasound transducer placed on the top surface of the biological medium consisting of skin , fat , muscle and bone The transient bioheat transfer equation is solved to estimate the temperature elevation in each tissue due to the ultrasound energy deposition. In rectangular coordinates and for the case of y-symmetry it is written as [4]:

$$\rho\, c_t \frac{\partial T}{\partial t} = k\left(\frac{\partial^2 T}{\partial x^2} + \frac{\partial^2 T}{\partial z^2}\right) + Q \qquad (1)$$

where ρ is the tissue density, c_t is the specific heat of tissue, k is the thermal conductivity of the tissue and T is the temperature. Q is the absorbed acoustic power and it is given as [4] :

$$Q = 2\alpha f I_o e^{-2\alpha z f} \cos\left(\frac{\pi x}{2x_o}\right) \qquad (2)$$

where, I_o is the intensity at the top surface, α is the ultrasound attenuation coefficient inside the tissue, $2x_o$ is the ultrasound transducer width along the x axis and f is the frequency. The proposed model is selected to be a two dimensional (x z) model to reduce the computation time. The finite difference form of the time and

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space derivatives for the bioheat transfer equation is given as

$$\rho.c_t \frac{T^{p-1}_{m.n} - T^{p}_{m.n}}{\Delta t} = k[\frac{T^{p}_{m-1.n} + T^{p}_{m-1.n} - 2T^{p}_{m.n}}{(\Delta x)^2}$$

$$+ \frac{T^{p}_{m.n-1} + T^{p}_{m.n-1} - 2T^{p}_{m.n}}{(\Delta z)^2}] + \qquad (3)$$

$$(2\alpha.f.I_O.e^{-2.\alpha.f.(m.\Delta z)} \cos(\pi(n\Delta x) / 2x_o)$$

where, m and n are used to designate the z and x locations of nodal points, respectively. The term p *is* used to denote the time dependence of T , thus t = p Δ t. Therefore, the nodal temperature at the new (p + 1) time is given as

$$T^{p-1}_{m,n} = F_o(T^{p}_{m-1,n} + T^{p}_{m-1,n} + T^{p}_{m,n-1} + T^{p}_{m,n-1})$$

$$+ (1 - 4F_o)T^{p}_{m,n} +$$

$$(\Delta t / \rho.c_t) .$$

$$(2\alpha.f.I_O.e^{-2.\alpha.f.(m.\Delta.z)} \cos(\pi(n\Delta x) / 2x_o) \qquad (4)$$

where $F_o = k\Delta t / \rho.c_t.(\Delta x)^2$. The spacing of elements in the finite difference grid is $\Delta x = \Delta z$ = 0.2 mm in the x z plane and the time increment Δt = 0.06 s. These values are selected to satisfy the stability condition $F_o \leq 1/4$ [5]. In addition, the constant temperature boundary condition is considered during the calculations [5-7]. In order to avoid the heating effect on the upper tissue surface, a water bolus may be placed between the transducer and the upper tissue surface. Figure 1 shows the configuration of the transducer with a water bolus and the human tissues. Skin , fat , muscle and bone constants are listed in Table I [8, 9].

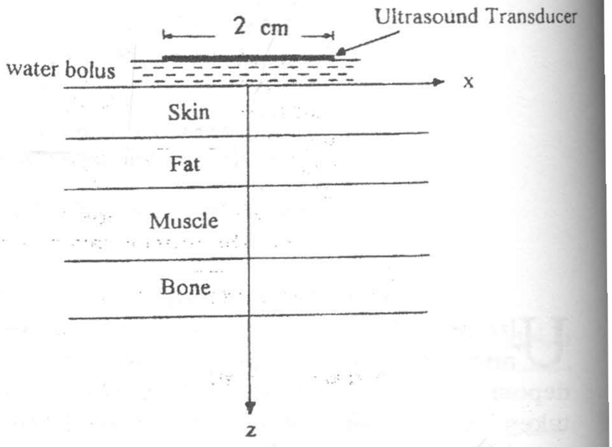

Figure 1 Human tissues with transducer and a water bolus

Table 1 Skin, fat muscle and bone parameters

	ρ (kg/m³)	Cₜ (J/Kg/K)	K(W/m/K)	α(Nₚ/m/MHz)
Skin	1050	3610	0.48	5.5
Fat	1200	1005	0.192	0.006
Muscle	1060	3720	0.5	6.0
Bone	1350	1020	0.22	0.0055

RESULTS AND DISCUSSION

We start with the case of a semi-infinite human skin layer in order to show the temperature elevation as a function of exposure time at two different ultrasound intensities and an operating frequency of 5 MHz. Figure 2 shows that temperature becomes constant after 15 seconds for I_o = 20, 30 W/cm². Therefore, it is considered as the maximum exposure time. Then the skin layer is replaced by three human tissues skin (2mm depth), fat (2 cm depth) and muscle (5 cm depth) . The temperature distribution in each layer is calculated and plotted in Figure 3 for I_o = 20 W/cm². It is found that the effect of energy deposition is neglected inside the muscle layer despite of increasing the exposure time. Therefore, the operating frequency must be reduced and 1MHz is used as an operating frequency.

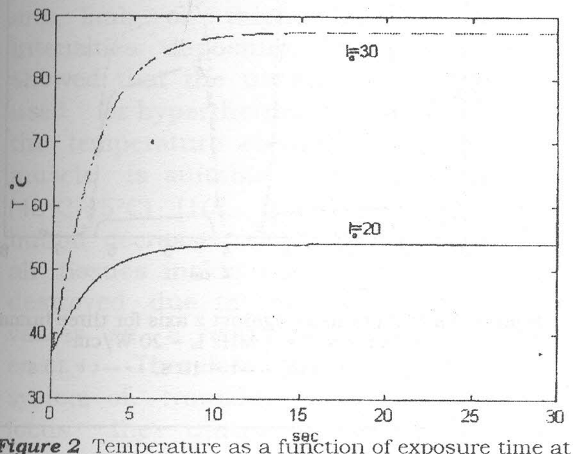

Figure 2 Temperature as a function of exposure time at different intensities f = 5 MHz

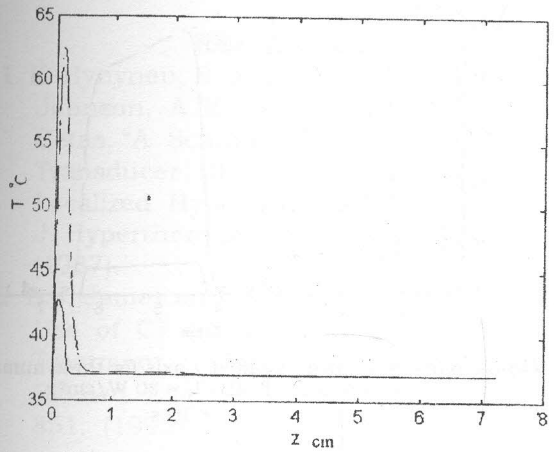

Figure 3 Temperature against z axis for three human a tissues at f = 5 MHz I_o = 20 W/cm² ____ t = 1 s _ _ _ t = 10 s

The temperature distributions against x and z for different times are plotted in Figure 4. In Figure 4-a , it is observed that for skin and muscle tissues the temperature values increase as exposure time increases at I_o = 20 W/cm². Also, for a certain time, the temperature distribution increases to a maximum value then decreases as the depth increases. Since the fat has a lower thermal conductivity than that of muscle, it is found that the maximum values of temperature inside the muscle layer are in the therapeutic temperature range. While inside the fat layer the temperature is constant and close to 37 °C . It is observed that the peak values for higher intensity needs less exposure time. Thus, in Figure 4-b, the exposure times are t = 1 s , 10 s for I_o = 30 W/cm². Figure 4-c shows the temperature profile against x axis for 1 s, 10 s and 15 s exposure time and I_o = 20 W/cm² while Figure 4-d is plotted for I_o = 30 W/cm2 at 1 s and 10 s . It is seen that the temperature peak values exist close to the feeding point at x = 0 then the temperature values decrease as x increases. As a practical model of arm, the three human tissues are replaced by seven tissues skin (2mm), fat (1 cm), muscle (2 cm), bone (1 cm), muscle (2 cm), fat (1 cm) and skin (2mm).

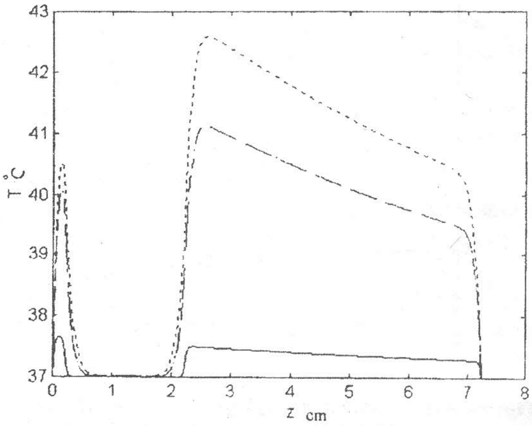

Figure 4-a Temperature against z axis for three human tissues at f = 1 MHz I_o = 20 W/cm² ____ t =1 s, _ _ t = 10 s, _ _ _ t = 15 s

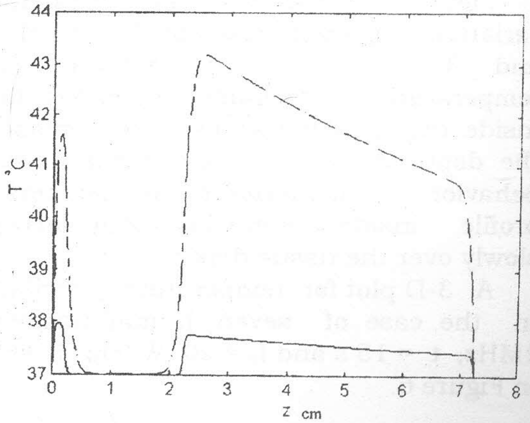

Figure 4-b Temperature against z axis for three human tissues at f = 1 MHz I_o = 30 W/cm² ____ t =1 s _ _ t = 10 s

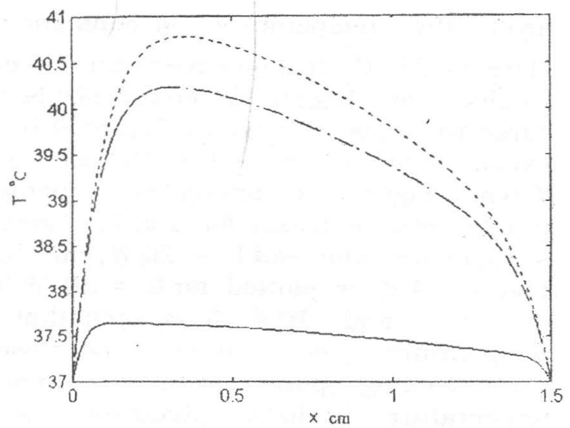

Figure 4-c Temperature against x axis for three human tissues at f = 1 MHz I_o = 20 W/cm² ____t = 1 s,— — t = 10 s - - - - t = 15 s

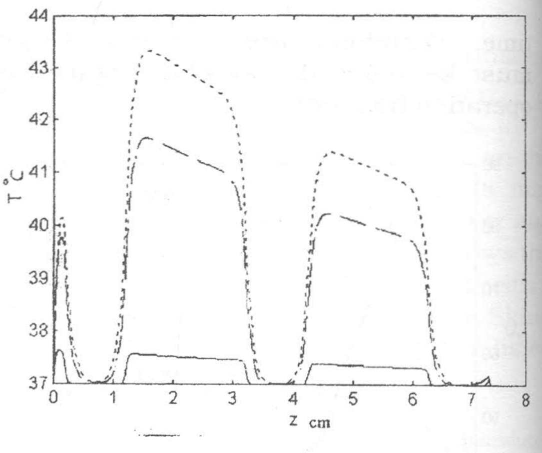

Figure 5-a Temperature against z axis for three human tissues at f = 1 MHz I_o = 20 W/cm² : ______ t =1s, — —t = 10 s, - - - -t = 15 s

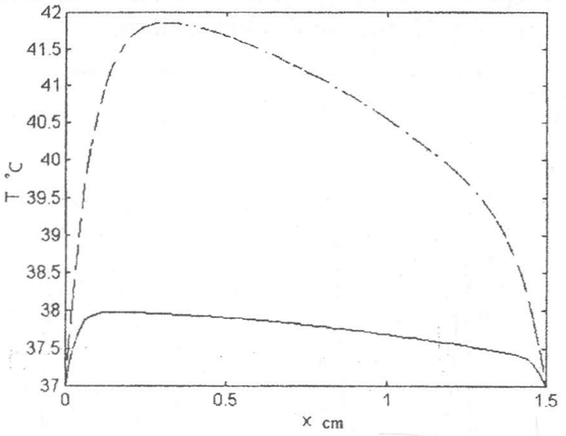

Figure 4-d Temperature against x axis for three human tissues at f = 1 MHz I_o = 30 W/cm² : ____ t =1s, — —t = 10

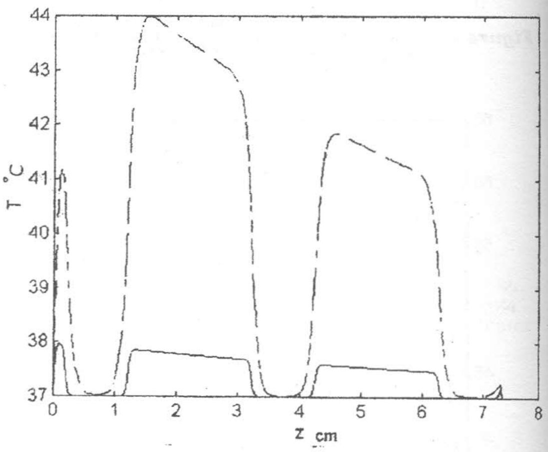

Figure 5-b Temperature against z axis for three human tissues at f = 1 MHz I_o = 30 W/cm² : ______ t =1s, — — t = 10 s

Figure 5 shows the temperature variation in each tissue for I_o = 20 W/cm² and 30 W/cm². It is observed that the temperature maximum values are found inside the muscle tissues and decreases as the depth increases due to the attenuation behavior. Furthermore, the temperature profile inside the muscle tissue decreases slowly over the tissue depth.

A 3-D plot for temperature distribution in the case of seven human tissues, f = 1MHz, t = 15 s and I_o = 20 W/cm² is shown in Figure 6.

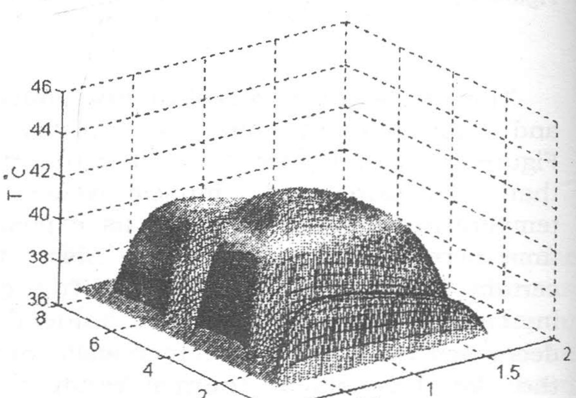

Figure 6 3-D Temperature distribution for seven human tissues at f = 1 MHz I_o = 20 t = 15 s

CONCLUSION

A practical application of solving the bioheat transfer equation is to get temperature distribution in the human arm, limb or thigh due to ultrasound intensities deposition. The results obtained showed that the ultrasound energy can be used for hyperthermia treatment. Although the temperature elevation range exits in muscle is suitable for tumor treatment (40°C-45°C) [10], the temperature distribution decreases slowly. This means that all tissues inside the muscle layer will be destroyed due to the high temperature values as a result of the ultrasound energy. Therefore, we suggest that a system of transducers is required to focus the ultrasound energy at the tumor location in order to use ultrasound successfully for hyperthermia.

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محاكاة لاستخدام الموجات فوق الصوتية فى العلاج برفع درجة الحرارة

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ملخص البحث

تستخدم الموجات فوق الصوتية عادة فى الفحص والكشف عن الأورام. حديثا اتجه التفكير لاستخدام هذه الموجات فى العلاج ايضا عن طريق رفع درجة حرارة الوسط الذى يحوى الأورام. يتطلب العلاج برفع درجة الحرارة الى امتصاص طاقة محددة لفترة زمنية عند موضع معين داخل الوسط.

فى هذا البحث اعتبرنا الوسط البيولوجى مكون من طبقات متعددة من الجلد والدهن والعضل والعظام.

وقد تم ايجاد التوزيع الحرارة نتيجة لامتصاص موجات فوق صوتية ذات شدة مختلفة داخل هذه الطبقات بحل المعادلة البيوحرارية باستخدام طريقة الفروق المحددة فى حيز الزمن.

وقد اثبتت النتائج ان الموجات فوق الصوتية يمكن ان تستخدم بكفاءة عن طريق تجميع الموجات فوق الصوتية وتركيزها لرفع درجة حرارة الوسط الى القيم المطلوبة لعلاج الأورام.