STUDY THE EFFECTS OF VARIATIONS IN FOCAL PRESSURE AND SCANNING PATH IN ORDER TO REDUCE OVERHEATING OF THE POST TARGET BONE DURING ULTRASOUND BRAIN SURGERY

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INTRODUCTION

One of the most complicated tumor removal surgeries is in the case of brain tumors. Brain surgery requires precise and sophisticated instruments and is associated with healthy tissue damage. This can lead to quite complicated adverse effects after treatment. Presence of blood brain barrier makes the delivery of drugs very difficult [1]. Also conventional methods have so far failed to control tumor progression [2-4]. Because of the sensitivity of brain tissue and its central role minimally invasive and noninvasive procedures are desired. Compared to open surgeries such approaches offer the advantage of reducing (a) the surgery time (b) the tissue damage associated with surgery (c) transfusion requirements and their associated infection risks [5]. The result is shorter recovery time and hospital stay, a reduction in cost of health care and a generally superior therapeutic outcome [5].

In this respect high temperature thermal therapies are becoming increasingly acceptable. The applicable modalities in this method are: high intensity focused ultrasound, microwave, radiofrequency currents and laser. Between these ultrasound has the advantage of accurate energy focusing into the body [6], favorable range of energy penetration [7], the ability to shape power depositions [7] and the technical feasibility of constructing any size and shape [8]. However when it comes to practice the large discrepancy between the skull high acoustic velocity (about 3000 m s^{-1}) and the brain velocity (about 1500 m s^{-1}) severely distorts the beam shape [1] and prevents the effective focusing of beams.

For solving this problem two methods are proposed: 1- application of phased arrays to restore the focus through intact skull by applying phase shifts to each element [1] and 2- Transdural sonication after removing a piece of the skull bone [9,10]. The first method although promising and is a potential for near future is still in development. In the second method a craniectomy allows avoiding the absorption and distortion caused by the skull bone. However overheating of the post target bone or the normal brain tissue could still be problem [9,10]. This can be minimized by a suitable choice of frequency, intensity and scanning path. Therefore the effects of variations of these parameters should be studied. In this paper the transdural propagation of ultrasound from a transducer into a two dimensional MRI based brain model and the resulting temperature distribution is simulated applying finite element method (FEM) and solving Helmotz and Bioheat equations respectively. For an applicable value of frequency, the effects of variations in focal pressure and scanning path is studied on the temperature distribution and an appropriate choice of focal pressure and scanning path is discussed.

SIMULATION MEHODS

The simulations of this study are divided into three Categories: 1- A Two dimensional model was constructed from a Magnetic Resonance Image of brain. The model contains the skull, brain, cerebrospinal fluid and the sphenoid and frontal sinuses. Then a piece of the skull bone was removed to create an acoustic window to allow ultrasound to propagate inside the brain. 2- The pressure distribution inside the brain was calculated using Helmhotz equation by applying finite element method (FEM).3- The temperature evolution resulted by ultrasound absorption is calculated throughout the brain using Bioheat equation and applying FEM.

WAVE FIELD MODEL

The time harmonic ultrasound field is the solution of the Helmhotz equation in an inhomogeneous medium:

$$\nabla.\left(\frac{1}{\rho}\nabla p\right)+\frac{k^2}{\rho}p=0$$

Where ρ is density, c is the speed of sound and k is the wave number. In dissipative media the wave

number is defined as $k = 2\pi f / c + i\alpha$, where α is the absorption coefficient [11].

The Helmholtz equation was solved using (FEM) and by applying the acoustical parameters listed in table 1. In order to achieve a tolerable accuracy while using FEM to solve the Helmholtz equation, the density of the discretization points should be ten points per wavelength (i.e. $\lambda / h = 10$ where λ is the wavelength and h is the side length of a finite element). Also the density of discretization points per wavelength must increase with wave number increase to maintain accuracy on a fixed level [12].

In this paper the maximum element sizes of meshing were chosen to satisfy $\lambda / h = 12$.

Table 1: Acoustical parameters of the simulations [13,14]

Tissue	$c\left(ms^{-1}\right)$	$\rho\left(Kgm^{-3}\right)$
brain	1545	1030
bone	2652	1796
water	1500	1000
air	383	1.16

The attenuation coefficient of brain was chosen to be 7.5 $Npm^{-1}MHz^{-1}$[13, 15] and those for bone are listed in table 2.

Table 2: Attenuation of the bone corresponding to the value of frequency [14, 15].

Frequency $\left(MHz\right)$	0.3	0.6	0.8	1
Attenuation $\left(Npm^{-1}\right)$	20	52	86	124

THERMAL MODEL AND BIOHEAT EQUATION:

The temperature elevation resulted from the absorption of ultrasound in the model was computed using Bioheat equation [16]:

$$\rho C_T \frac{\partial T}{\partial t} = \nabla . k \nabla T - \omega_B C_B \left(T - T_A\right) + Q$$

Where T is the temperature, C_T is the heat capacity of the tissue, k is the tissue thermal conductivity, $\omega_B C_B$ denotes the blood perfusion term, Q denotes the heat source term and by neglecting the metabolic heat generation because of its small contributions, the only remaining part is the absorbed ultrasound energy in the tissue and is given by[17]:

$$Q = \frac{\alpha |p|^2}{\rho c}$$

Bioheat equation is solved using (FEM) and by applying the thermal parameters listed in table 3. The used mesh pattern is the one which was used to calculate the Helmhotz equation.

Table 3: Thermal parameters of the simulations [18-22]

tissue	$C_T\left(JKg^{-1}K^{-1}\right)$	$k\left(Wm^{-1}K^{-1}\right)$	$\omega_B\left(Kgm^{-3}s^{-1}\right)$
brain	3640	0.528	10
bone	1300	0.630	0
water	4180	0.615	0
air	1007	0.0263	0

SIMULATION PROCEDURE

A transducer (R=6cm, d= 8.9cm) was placed at the top of the brain so that to locate the focus at the center of the brain and 3.5 cm deep inside the tissue. An applicable value of frequency was chosen to study the effects of variations in focal pressure and scanning path. We have studied the frequencies of 0.3, 0.6, 0.8 and 1 MHz while keeping the focal pressure at 1.5 MPa. In all of the simulations we let the target to reach the toxic temperature of 56^{o}C. As shown in Ref.6 there is only one sec needed for the tissue to necroses at this temperature. The minimum applicable value of the frequency to be used in other parts of the simulations was chosen to be 800 KHz. Figure 1 shows the temperature distribution in case of using 800 KHz of frequency where the peak temperature of the target and bone are respectively 56^{o}C and 49^{o}C. In lower frequencies (300-600 KHz) the temperature of the bone exceeds the temperature of the target and results in the failure of the treatment.

By increasing the frequency the attenuation of the beam increases, therefore lower amount of heat is generated at the bone. However increasing the frequency is not always the best case as the overheating of the normal tissue between the target and the transducer may result [23].

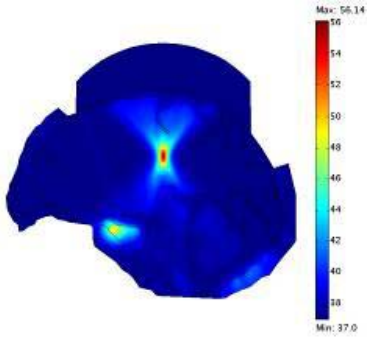

Figure1: Temperature distribution using the frequency of 800KHz and the focal pressure of 1.5 MPa when the target reaches the temperature of 56°C

EFFECTS OF PRESSURE VARIATIONS

In order to study the effect of variations in the intensity, the pressure distribution at the surface of the transducer was given so that to produce the focal pressures in the range of 1.8 - 5 MPa. The applied frequency was 800 KHz as determined suitable in the earlier part of the study. Like the previous part we let the target to reach 56°C.

VARIATIONS IN SCANNING PATH

The effect of variations in the scanning path was studied by rotating the transducer around the focal point in the range of (-20° to 20°). The applied frequency and the focal pressure are 800 KHz and 1.8 MPa respectively and fixed in all of the cases. Like the previous part we let the target to reach 56°C.

RESULTS ND DISCUSSION

Figures 2a-d show the temperature distribution corresponding to the value of the focal pressures of 1.8, 2, 3, 4 MPa respectively while applying the frequency of 800 KHz. They reveal that by increasing the focal pressure there is a significant reduction in the temperature of the post-target bone, pre-focal tissue and the tissue surrounding the target. The reason for this can be described as follows: The high blood perfusion in the brain (~10) ($Kgm^{-3}s^{-1}$) compared to its very low value in the bone (*~0*) ($Kgm^{-3}s^{-1}$) generates a strong heat sink in the target with respect to that of the bone. This heat sink grows linearly with temperature increase, so reduces the total heat source at the target. In turn this will decrease the rate of temperature increase. Considering the fact that the rate of temperature growth in the bone is constant and is not influenced by any heat sink, the temperature evolution in bone, approaches rapidly near the value of the target. By increasing the focal pressure the cooling effect of blood perfusion becomes negligible in comparison with the external heat source in the range of therapeutic levels (~56°C). In addition the decrease in the treatment time prevents the pre-focal tissue of gaining high energies and leads to the weakness of the effect of conduction and prevents the temperature of the surrounding normal tissue of being raised to higher values.

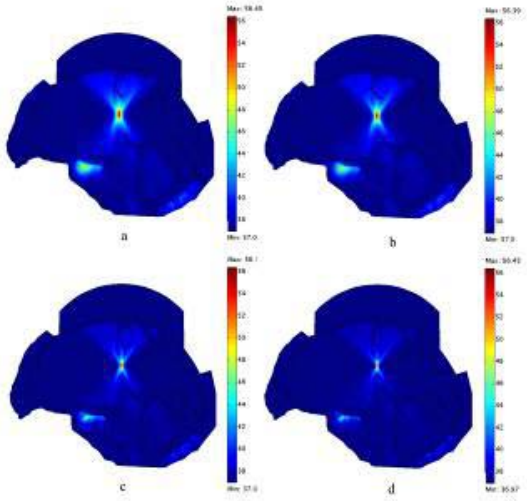

Figure2: Temperature distribution corresponding to the value of the focal pressure of: a- 1.8 b- 2 c- 3 and d- 4 MPa

Figure 3 shows the temperature difference between the target and the post-target bone with respect to the value of the focal pressure while the temperature of the target reaches 56°C.

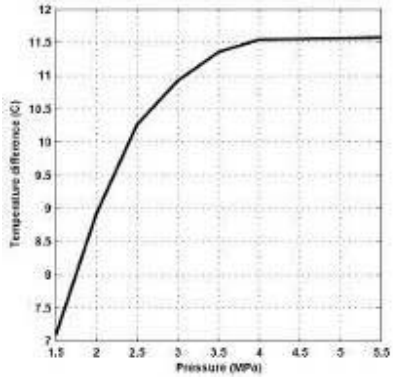

Figure2: Temperature difference between the target and the post-target bone with respect to the value of the focal pressure when the target reaches 56°C

It shows that the temperature difference grows by the increase in pressure. However the rate of growth reduces with pressure increase and nearly stops at a threshold of 4 MPa. As this is lower than the cavitation threshold at 800 KHz it can be set as the threshold of the maximum focal pressure in this frequency, scanning path and depth. The generated temperature at the post-target bone is 43.9^{o}C which is much lower than the case of applying 1.5 MPa of focal pressure.

Therefore not only increasing the frequency, but also increasing the focal pressure enhances a safer treatment by reducing the overheating of the post-target bone. Another advantage of using higher pressures is the reduction of the temperature of surrounding normal tissue. Figures 4a-d show the temperature distribution in case of the scanning angles of (-10, -20, 10, 20^{0}) respectively while applying the frequency of 800 KHz and the focal pressure of 1.8 MPa. They show that the highest temperature rise at the post-target bone happens when a part of the beams end up at the sinus interface. Choosing a suitable scanning path which moves the beams away from this region significantly decreases the overheating of the bone (i.e. as best seen in figure 4d). This is because the heat source depends on the square of the acoustic pressure and an small degree of constructive interference between the incident and reflected beams from the surface of the sinus air cavity leads to a very high acoustic heating.

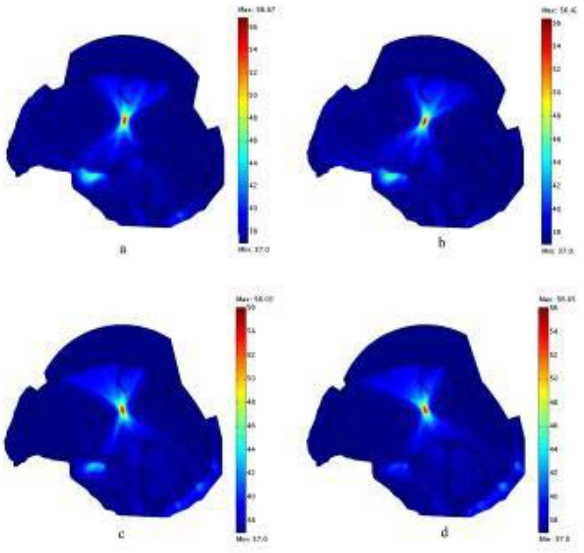

Figure4: Temperature distribution corresponding to the frequency of 800 KHz, focal pressure of 1.8 MPa and the scanning angles of: a(-10) b(-20) c(10) and d-(20) degrees

CONCLUSIONS

The effect of variations in focal pressure and scanning path was studied in this paper by using FEM simulations. It is shown that for a suitable value of frequency a threshold for the value of the focal pressure can be chosen so that the difference between the temperature of the target and the post-target bone is maximum. Also choosing a scanning path which moves the beams away from the sinus air cavities significantly reduces the temperature generated at the post-target bone.

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