Modeling the heat effect of ultrasound in the irradiated tissues

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The temperature of the body tissues during ultrasound irradiation depend on the tissue and treatment properties. The newly developed computational model can determine the temperature changes in the skin-muscle-tumor-blood system between different tissue and treatment parameters. The model can calculate with the respiration (315-318K) and denaturing (e.g. HIFU) hyperthermia too. The basic parameters are the geometry, (m); specific heat, (J/gK); density, (g/cm^3); heat conductivity, (J/msK); sound velocity (m/s); attenuation coefficient (Np/cm MHz) and the perfusion rate of blood in the different tissue layers (ml/g$_{tissue}$/min). The effective treatment parameters are the emitted electric output, (J/s); the mechanical efficiency (%); the number (pieces); and the surface area of the transducers (m^2); and the area (m^2) and temperature (K) of the cooling surfaces. The model calculates many of intermediate data from these. Some of these is the sound path in the different tissues, (m); the blood quantity in the tissue layers, (m); the geometry of the irradiated tissue bodies, (m, cm^2, cm^3); the acoustic hardness, (Pa s/m); sound reflection and sound transmission occurring at the interfaces, (Np); heat exchanger wall thickness of the irradiated bodies, (m); heat dissipation and heat exchanger surface areas, (m^2); flow rate of blood in the tissues located in the path of ultrasound, (ml/tissue mass in g/min); and the sound attenuation effect in the different tissues, (Np). The basic and intermediate data gives the generated heat (K/s) decreased by the heat energy transported (J/s) to the surrounding tissues by blood and heat conductivity, and the actual temperature (K) of the irradiated tissue. The algorithm contains the pre-cooling possibility of the tissues before and during the irradiation with a specific area and temperature cooling surfaces. The model considers that single rectangular transducers are focused to a target zone, which equal to the surface of one transducer.
The actually modeled situation is that the tumor must be reaching a 316K temperature; however the temperatures of other two layers must be under than 313K. The heat generation was examined in the function of thickness and perfusion rate of blood in the different tissue layers. During the solution, we varied the pieces and ultrasound power of transducers and the surface and temperature of the cooling surfaces. The conclusion is that the aims were solvable between every situation. The future is that the model can handle the ray interferences and the angle of incidences.

Keywords: *Ultrasound hyperthermia, tissue layers, losses, irradiation*

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URL: the poster accessible in PDF format at: www.lorincza.hu/UIAposter2007

1. Introduction

According to Mátai (2002) the hyperthermia can inhibit the tumor proliferation and in many cases the size of the tumors decreasing by the treatment. The perfusion rate of blood in the tumors is inadequate. This reasoned that same dose of irradiation gives higher temperatures in the tumor than in the surrounding tissues. The surrounding tissues supplied the tumors with blood. The inside of the tumor is necrotized and the intermediate zone is anoxic because the oxygen absence. During the irradiation the aeration intensity of the cells are increasing because the temperature increasing. The oxygen demands of the cells are in direct ratio with the temperature. The chance of survive of the irradiated cells are decreased when the bloods supply can't adapt to the increased cell activity at higher temperatures. Base on these, the cancerous cells will be necrotize because the oxygen absence during ultrasound treatment and the surrounding tissues remains intact. This is the theory of the respiration type hypertharmia. The method is effective, when the temperature of the tumor is between 43-45°C. Between 37-42°C, the cell proliferation in the tumors increasing. The necessary treatment period usually from half to one hour on 43°C. From 43-46°C, the irradiation period is halved with every centigrade, however the temperatures above 46°C to endangered the surrounding tissues. Base on these, the optimal treatment period is one hour between 42-43,5°C (315-316,5K). In the case of denaturing hyperthermia, the temperature of treatments are over on 43,5°C and the target zones are considerably close and the effect based on the cell protein denaturing. Mise et. al., (1990) treated cancer cells with ultrasound in water baths with 42°C and concluded that the proliferation of the cells increased to 30min but after decreased. Lai et. al., (2002) treated prostates with self-tuning ultrasound hyperthermia system with 43-45°C. The advantage of the method is that the control of the ultrasound intensity based on the actual temperatures, and not on the knowledge of heat physical parameters of the tissues. According to Hurwitz et. al., (2001) the most effective temperature and period is 43-45°C and 30-60min to the prostate cancer therapy. Fosmire et. al., (1993) reported that the application of ultrasound hyperthermia is most reasoned when the patient's has other diseases too (e.g. heart disease). Wojcik et. al., (1995) examined the tissue ablation effect caused by focused ultrasound with finite element modeling technique. The calculation capable modeling the nonlinear acoustic phenomena and solves the electromechanic and bioheat equations in 2D/3D inhomogeneous elastic and acoustic media. Fry et. al., (1954) showed the thermal effect of ultrasound on soft tissues and these results were applied by Heimburger (1985), Sanghvi et. al., (1984), Purnell et. al. (1964) and Lizzi et. al., (1994), and Foster et. al. (1990) for treating deep breast tumors, lung tumors, in eye therapy, and in treating prostate hypertrophy, respectively. Based on the observation of Fry (1993), high intensity focused ultrasound (HIFU) has two types of effect. One of them is the lesion or cut that spreads in a regular way in the focus by way of heat coagulation. The other effect is that vapors are generated in the focus and these vapors rapidly expand towards the ultrasound source. Lizzi (1987) proved that due to vaporization a gas body can be formed, or even cavitation may occur in the tissue range located in the focal point of ultrasound. According to Lizzi (1993) this can be a factor limiting the intensity of focused coagulate ultrasound applied in hyperthermia. Properties of the tissues are changed by the initial focused ultrasound beam, e.g., the absorption, scattering, focal point are changed. Damianou, et. al., (1995) applied linear models for studying the effect of heat generated in the tissues by ultrasound. Non-linear models based on e.g. the KZK equations (Aanonsen 1983) were applied by Christopher and Parker (1991)**.** These models calculate the propagation properties of shock waves in body tissues.

1.1. Aims with the model

Our goal was make a simple, interactive, easy to use computational ultrasound hyperthermia model to the fields of research and development, demonstration and education. The model handles now three tissue layers. The treatment parameters can variables 0-∞. The model calculates and graphically displayed the heat effect of ultrasound between different treatment and tissue parameters. The model calculates with heat and acoustical losses too. The model can calculate with respiratory and denaturing hyperthermia.
The actually modeled situation, that the tumor must be reach the 316K temperatures, however the other two tissue layers must be under than 313K.

2. Matherials and Methods

2.1. Bases

The model considers that single rectangular transducers irradiated to the target zone. The modelled situation is that the tumour must be reach a 316K temperature, however the other two layers must be under than 313K. The heat generation was examined in the function of thickness and perfusion rate of blood in different tissue layers. The model calculates the heat effect of ultrasound and calculates with the sound attenuation, reflection and main losses. The model can handle from 0 to ∞ the piece of transducers in pc, the tissue thickness in m, the active and cooling surfaces in m^2 and the intensity of transducers in W. The Fig. 1. shows the theoretically model scheme with one transducer. The model determines the heat effect generated by the ultrasound, due definite number of single rectangular transducers, on a specific volume element, while the sound passes through the tissue layers (skin, muscle, tumour).

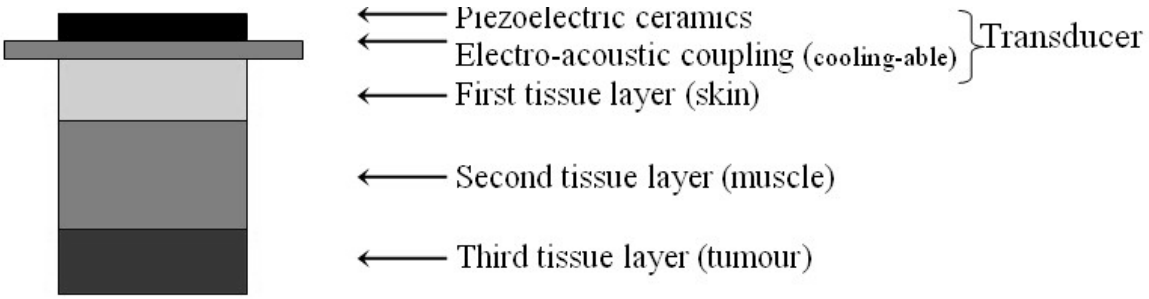

Figure 1. The scheme of the examined layers

The heat generation in the tissues is determined after the correction with losses. These the heat energy removed by blood, cooled transducer and surrounding tissues. There is a pre cooling possibility before and during of irradiation. This gives a flexible treatment position between large-scale intensities. The type of the tissue layer is represented in the model by its acoustic and thermal parameters. The most important tissue parameter the sound velocity, specific heat, density, attenuation coefficient, heat conductivity and the perfusion rate of blood in the tissues. The generated heat depends on the material properties, determined by the geometry of irradiated tissue bodies, the irradiation parameters (intensity, frequency, wave shape, mode, the extent of focusing) and the occurred losses. The Fig 2. shows the operator window of the model.

HIPERTERMIA MODEL	
OPERATOR WINDOW	
EMITTED ELECTRIC INTENSITY/TRANSDUCER, W	4,8
WIDENESS OF THE TRANSDUCERS (m)	0,01
LENGTH OF THE TRANSDUCERS (m)	0,01
TRANSDUCERS TARGETING THE TUMOUR (Pieces)	7
COOLING SURFACE/TRANSDUCER (m^2)	0,0005
TERMPERATURE OF THE COOLING SURFACE (K)	283
DEPTH OF THE SKIN (m)	0,02
DEPTH OF THE MUSCLE (m)	0,05
DEPTH OF THE TUMOUR (m)	0,01
ELECTRO-ACOUSTICAL EFFICIENCY OF TR. (η, %)	50
CALC. ACOUSTICAL SURFACE INTENSITY (W/cm^2)	2,4
PERFUSION RATE OF BLOOD, SKIN ($ml/g_{tissue}/min$)	0,2
PERFUSION RATE OF BLOOD, MUSCLE ($ml/g_{tissue}/min$)	0,027
PERFUSION RATE OF BLOOD, TUMOUR ($ml/g_{tissue}/min$)	0,25

Figure 2. Operator window of the model

2.2. Tissue geometry

The incidence of the beam is perpendicular to the tissue layers. The beam transmission surface is

$$S=ab \ [cm^2] \quad (1)$$

where *a* the width and *b* the length of transducer in cm. The treated tissue volume is

$$V=Sh \ [cm^3] \quad (2)$$

where S is the surface and h is the tissue thickness in cm. The treated tissue mass is:

$$M=V\rho \ [g] \quad (3)$$

where ρ, g/cm^3 the density of tissue.

From the multiplication of treated tissue mass and specific heat C [J/gK], the energy needed for increasing the temperature of the subject tissue by 1K can be calculated in J/K.

2.3. Perfusion rate of blood

The knowledge of perfusion rate of blood in the treated tissue volume:

$$g_{blood}/g_{actuallytissuemass}/min \ (B_2) \quad (4)$$

based on the:

$$ml/g_{tissue}/min \ (B_1) \quad (5)$$

connection. The calculation of B_2 requires the knowledge of blood and tissue densities and the treated tissue volume. When using the perfusion rate of blood/minute value, the model assumes that this is the quantity of blood that is present in the subject tissue at any time. Of course, the realistic value may differ from this quantity and this blood quantity changes further during the treatment as a result of the increased perfusion rate and heart rate. Due to

this, the perfusion rate of blood is considered by the model as an input parameter, that is it can be adjusted for the individual tissues on the operator window. The length of the path of the sound beam in the blood flowing through the tissues is also a factor of decisive importance. This means that the height of blood in the irradiated tissue volume determined by the irradiation surface S, cm^2 and the thickness of the specific tissue shall be determined. This corresponds to those assumptions that the vascular system of the tissues is so dense and the walls of the blood vessels are so thin, that the heat exchange between the tissues and blood can be considered instantaneous. As a consequence of this, it is enough to determine the quantity of blood in the treated tissue volume having a specific surface, and through this to determine the height of the blood body in the tissue, hb, cm with the

$$hb=(B_2/\rho b)/S \quad (6)$$

formulae in order to calculate the heat generated in the blood present in the tissue by the irradiated acoustic energy, and to correct the generated heat J/s by heat energy factors representing the quantity of heat transported to, and from, the treated tissues by the perfusion of blood.

2.4. Intensity connections

In order to determine the different intensity changes, the sound intensity is calculated in the following units of measure: W/cm^2, dB and Np. From the emitted total electric intensity I_e, W=J/s, the initial acoustic intensity I_i can be calculated by using the initial acoustic intensity, W and the electro-acoustic coefficient (η, %) of efficiency according to the equation:

$$I_i=I_e\eta. \quad (7)$$

The acoustic intensity per surface unit in W/cm^2 can be calculated from formulae:

$$I_s=I_i/S. \quad (8)$$

Traditional way of recalculating this variable in dB unit of measure is shown by the equation

$$dB=10\log_{10}(I_S/I_0) \qquad (9)$$

where

$$I_0=10^{-16}\ [W/cm^2]. \qquad (10)$$

Recalculation of INp from this equation is described by

$$INp=IdB/8{,}686. \qquad (11)$$

Of course, the units of measure can also be recalculated backward using the inverse forms of the appropriate equations. Attenuation caused by the specific tissue volume can be calculated by

$$I=I_0e^{-2\alpha h} \qquad (12)$$

where INp is the actual Np intensity and I_0 is the initial intensity expressed in Np, α is the attenuation coefficient (Np/cmMHz), and h is the length of the path of the sound beam in the given medium expressed in cm. From here the model calculates the W/cm^2 intensities for each media backwards through IdB, then it determines the reflections R at the medium interfaces

and through this, the transmitted intensity T. If it is assumed that the incidence is perpendicular to the interface between the media, the rate of reflection is

$$R=(Z_2-Z_1)/(Z_1+Z_2), \quad (13)$$

which is a function of the difference between the acoustic hardness (PaS/m) of the media

$$Z=c\rho, \quad (14)$$

where c, m/s is the sound velocity, ρ, g/cm^3 is the density, Z_1 and Z_2 are the acoustic hardness of the resultant and receiving media, respectively. From these results the level of relative sound intensity transmitted through the interface can be calculated from

$$T=1-R, \quad (15)$$

provided that 1=100%, that is in the absolute acoustic intensity, the acoustic intensity going through the interface between the media in W/cm^2. The heat energy generated in the examined tissue (ter Haar, 1988) volume can be calculated by the equation

$$dT/dt=2\alpha I/C\rho, \text{ [K/s]}, \quad (16)$$

provided that both the numerator and the denominator of the formula have been corrected by the distance covered and by the treated tissue volume, respectively. By solving this equation to sec K can be arrived at, namely the result show how much the temperature increases in the system in a given period of time where C is the specific heat J/gK of the examined tissue. The reason why the Rayleigh-Sonnerfeld (Goodman, 1968) equation, the equation that handles xyz coordinates was not used in the model is that our goal was to calculate the spatial average temperature of the tissue volume to be treated and the model assumes that the position of the tissue volume to be treated is known, because this is the location where the transducers are focused to. If the calculation was carried out on the basis of the xyz coordinates, the application of the bioheat transfer equation (BHTE) (Pennes, 1948 and Sapareto and Dewey, 1984) would be necessary, although this equation does not deal with the presence of blood vessels and the changes in the perfusion rate of blood that occur during the treatment. The equation systems applied in the model are equivalent with the BHTE system without space coordinates and they take into consideration the input (feed energy) and output (losses) sides of the energy balance. The acoustic intensity reaching the tumor can be calculated as a sum of the residue intensities of all the transducers focused to the tumor, without considering the interferences. Generated heat is determined form this intensity by time units. Basic spatial scheme of the hyperthermia superstructure is shown in figure 3.

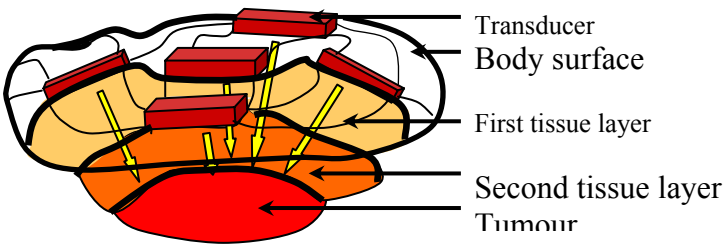

Figure 3. Theoretical buildup of ultrasound hyperthermia tissue model.

2.5. Losses

When determining the heat generated in the treated tissue volume during the ultrasound irradiation, two factors shall be known; these are the level of heat energy generated by the acoustic energy of the continuous input radiation and the level of heat loss. Three main sources of energy loss are handled by the applied model. These are parts of the energy removed by blood (I_b), by the cooled transducer surface (I_s) and by the heat transported to the neighboring tissues (I_t) by heat conduction. Determination of the latter two sources of loss (I_s and I_t) are based on determining the heat flow I, J/s along a planar heat exchanger wall surface by using the equation

$$I=\lambda/w[Sq(t_1-t_2)] \qquad (17)$$

where λ, J/msK is the heat conductivity of the given tissue, w, m is the wall thickness of the heat exchanger, Sq, m^2 is the size of the heat transfer surface, and t_1, K is the actual tissue temperature or the average temperature of the internal tissue volume. t_2 is the external temperature prevailing on the wall of the heat exchanger that is usually considered as 310K, that is 37°C body temperature by the model when calculating the heat conducted towards the neighboring tissues. For a rectangle, calculation of the w wall thickness of the heat exchanger is done by formulae

$$w=[(a/2)+(b/2)+(\sqrt{\ }a2+b2)]/3. \qquad (18)$$

When determining the q, J/s heat flow towards the neighboring tissues, the Sq, m^2 heat exchange surface means the skirt surface of the irradiated tissue volume given by the formulae

$$Sq=[(2ha)+(2hb)]/10000, \qquad (19)$$

except in the case of the last tumor layer where the heat exchanger surface located opposite to the beam entry surface shall be added to the calculated surface. In the case of the J/s heat energy removed by the cooled transducer the wall thickness and the heat exchanger surface are equal to the depth of the treated tissue expressed in m and to the surface of the transducer or the S surface to the treated tissue volume, respectively. In case of the top layer, the cooling surface can be adjusted and a cooling surface different from the transducer surface can be applied if necessary. In case of the I_c heat flow removed by the cooled transducer the t_1 temperatures of the individual layers are equal to the respective temperature of the contamination interface of the layer located above the subject layer in the point of time just preceding the point of time to which the calculation is made. The energy removed by blood I_b, J/s can be calculated from the difference of the introduced and removed energies according to the formulae

$$Ib=[(t_1B_2C_b)-(310B_2C_b)], \qquad (20)$$

where C_b, J/gK is the specific heat of blood and t_1 is the average internal temperature of the currently treated tissue mass.

2.6. Main equation

Based on the above, the heat generated by the ultrasound in the individual tissue layers can be calculated from the next equation:

$$dT/dt=(((2\alpha_t h_t ((I_{input}S)-(I_t+I_{cooling})))/C_t\rho_t V_t)+(2\alpha_b h_b ((I_{input}S)-I_b)/C_b\rho_b V_b)) \text{ [K/s]} \qquad (21)$$

where *b* the blood, *t* the tissue and *c* the cooling surface, and the result given in K/s. In the equations I_t, W/cm^2 is the value of acoustic intensity at the tissue interface. This value shall be multiplied with the irradiated surface (S, cm^2) to calculate the total intensity arriving into a volume element. The symbol of I_t [J/s] is the heat flow towards the surrounding tissues, $I_{cooling}$ [J/s] towards the cooled transducer surface and the tissue interface located above the subject tissue layer, and I_b [J/s] the heat removed by the blood. The basic expression that the treatments are carried out by sine waves at a frequency of 1 MHz. The frequency can be adjusted in the model and this results a modification of the attenuation coefficient α, Np/cmMHz.

2.7. Basic data

The basic data are shown in table 1. Parameters affecting the heat effect of treatment are: the emitted electric intensity, [W]; width and length of the transducer(s), [m]; number of transducers focused to the tumor [pc]; cooling surface of a transducer, [m^2]; temperature of the cooling surface, [K]; thickness of skin, muscle and tumor layers, respectively, [m]; coefficient of acoustic efficiency of the transducer, [%].

Table 1. Basic data

	Attenuation coefficient, α, [Np/cmMHz]	Sound velocity, c, [m/s]	Specific heat, C, [J/gK]	Heat conductivity, λ, [J/msK]	Density, ρ, [g/cm^3]	Perfusion rate of blood, B_1, [ml/g$_{tissue}$/min]
Ceramics (c)		2220			7,6	
Electro-acoustic coupling (co)		6320			2,7	
Skin (s)	0,24	1720	3,66	0,37	1,01	0,2
Muscle (m)	0,15	1566	3,639	0,55	1,04	0,027
Tumour (t)	0,085	1549	3,9	0,545	1,04	0,25
Blood (b)	0,019	1566	3,89		1,06	

3. Results and discussion

The transducers were placed on the skin surface and these radiated in the direction of tumor. There were four modeled situations (Table 2.)

Table 2. Coded body situations

	Perfusion rate of blood, B_1 [ml/g$_{tissue}$/min]		Thickness of tissue, cm	
	Code: 1	Code: 2	Code: A	Code: B
Skin	0,5	1	1	1
Muscle	0,5	1	2	4
Tumour	0,25	0,5	1	1

During the modeled ultrasound hyperthermia treatments, the electro-acoustic efficiency was always 50%. The size of the transducers and the cooling surfaces were 10X10 and 25X20mm respectively. The irradiation frequency was always 1MHz. The minimal cooling, maximal tumor and surrounding tissue temperature were: 273, 316,5 and 313K respectively. The initial tissue temperature was 313K without cooling. When the temperature of the skin and muscle (surrounding tissues) were higher than 313K, we applied a pre-cooling step during 600 seconds and when the 273K cooling temperature wasn't enough to drain the heat energy from the tissues, then we increased the cooling surfaces. The treatments started always at the 600. second. To the reach of our goal, we adjusted the electric power, transducer number, cooling surfaces and temperatures (input data).

The basics: the temperature of the tumor must be reaching the 316K in the 4200. second (600+3600) in every experiments. During the treatments, between the different experimental

situations, the time period necessary to the 315K-tumor temperature and the 315-316K period were examined. We named the 315-316K like „effective temperature”. The temperatures of healthy tissues (skin and muscle) and the relationships between this and the applied transducer number and electric powers were determined.

Table 3. Calculated thermal data in the case of 1/A. situation

	Parameters	***Situation code: 1/A***					
		Without cooling					*With cooling*
Adjusted values	Electric power, W	1,014	1,263	1,667	2,49	4,838	4,838
	Number of transducers, pc	10	8	6	4	2	2
	Cooling temperature, K	310	310	310	310	310	273
	Cooling surface, cm^2	5	5	5	5	5	10
Calculated values	Skin, K	310,6	310,7	311	311,4	312,8	298,2
	Muscle, K	310,7	310,9	311,2	311,8	313,4	313
	Tumour, K	316	316	316	316	316	316
	Time period 315K, s	1580	1580	1580	1580	1600	1620
	315-316K, s	2620	2620	2620	2620	2600	>3600
	The reach of the 316K, s	4200	4200	4200	4200	4200	4200

Table 4. Calculated thermal data in the case of 2/A. situation

	Parameters	***Situation code: 2/A***					
		Without cooling					*With cooling*
Adjusted values	Electric power, W	1,85	2,31	3,06	4,54	8,854	8,854
	Number of transducers, pc	10	8	6	4	2	2
	Cooling temperature, K	310	310	310	310	310	273
	Cooling surface, cm^2	5	5	5	5	5	10
Calculated values	Skin, K	311,1	311,3	311,8	312,6	315,1	301
	Muscle, K	310,7	310,8	311,1	311,6	313,2	313
	Tumour, K	316	316	316	316	316	316
	Time period 315K, s	1580	1580	1600	1580	1600	1580
	315-316K, s	2620	2620	2600	2620	2600	>3600
	The reach of the 316K, s	4200	4200	4200	4200	4200	4200

Table 5. Calculated thermal data in the case of 1/B. situation

	Parameters	***Situation code: 1/B.***					
		Without cooling					*With cooling*
Adjusted values	Electric power, W	1,021	1,273	1,689	2,512	4,9	4,9
	Number of transducers, pc	10	8	6	4	2	2
	Cooling temperature, K	310	310	310	310	310	273
	Cooling surface, cm^2	5	5	5	5	5	5
Calculated values	Skin, K	310,4	310,5	310,7	311,1	312,1	306
	Muscle, K	310,7	310,8	311,1	311,6	313,2	312,9
	Tumour, K	316	316	316	316	316	316
	Time period 315K, s	1720	1720	1720	1720	1720	1740
	315-316K, s	2480	2480	2480	2480	2480	>3600
	The reach of the 316K, s	4200	4200	4200	4200	4200	4200

Table 6. Calculated thermal data in the case of 2/B. situation

	Parameters	***Situation code: 2/B.***					
		Without cooling					*With cooling*
Adjusted values	Electric power, W	1,862	2,32	3,082	4,59	8,96	8,96
	Number of transducers, pc	10	8	6	4	2	2
	Cooling temperature, K	310	310	310	310	310	273
	Cooling surface, cm^2	5	5	5	5	5	5
Calculated values	Skin, K	310,8	311	311,3	3119	*313,8*	307,8
	Muscle, K	310,6	310,7	311	311,5	313,2	312,9
	Tumour, K	316	316	316	316	316	316
	Time period 315K, s	1740	1720	1740	1720	1720	1740
	315-316K, s	2460	2480	2460	2480	2480	2720
	The reach of the 316K, s	4200	4200	4200	4200	4200	4200

The 2-5. tables shows that the 316K tumor temperature without surface cooling available with minimum four transducers. In the case of two transducers there was necessary to applying two $10cm^2$ cooling surfaces with 273K temperatures. The applied transducer number decreased from 10 to 2. Our main goal (316K tumor temperature at the 4200. second) was accessible with proper treatment parameters without theoretical damage of surrounding tissues. In the cooling situations the time necessary to reach the 315-316K increased, and because the heat drain, the temperature gradient decreased. Lower ultrasound intensity was necessary to achieve a same temperature and heat gradient during lower perfusion rate of blood.

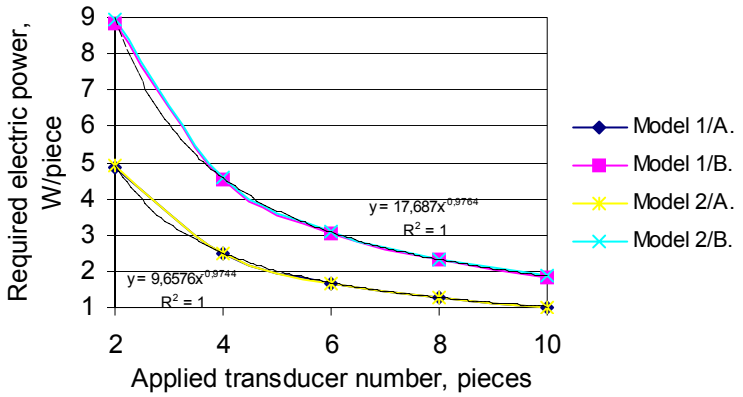

Figure 4. Relationship between piece and electric power requirement of transducers

It is shown on the Fig. 4. that with increasing of transducer numbers, the power necessity to attain the basic aim decreasing. The connection between the transducer numbers and power necessity is power function-like (Figure 4.). The power necessity was influenced due the tissue thickness in the highest degree. The reason is maybe that the sound absorption of the blood is same in the different tissues. When the tissues were two times thicker, the power necessity was near two times higher. The temperature gradients of treated skin and muscle see on Fig. 5. and 6. It is clearly seen that in the case of thicker layers (situation "B") - mainly at lower transducer numbers - the temperature rises in the surrounding tissues. This heat effect increased with lower perfusion rates too (situation 1.) because the blood can't carry the heat energy from the irradiated tissues.

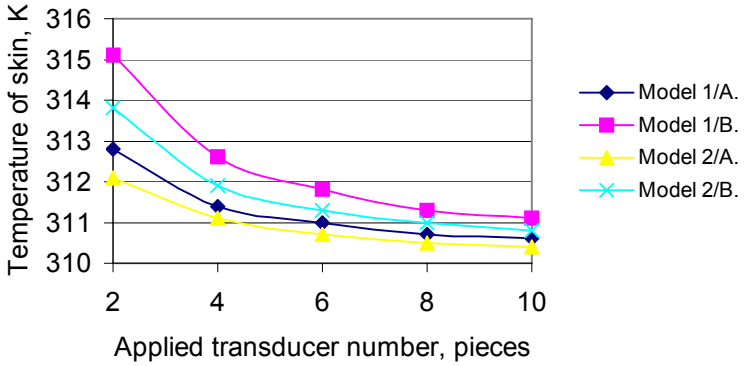

Figure 5. Relationship between the averaged skin temperature and transducer numbers during the ultrasound hyperthermia

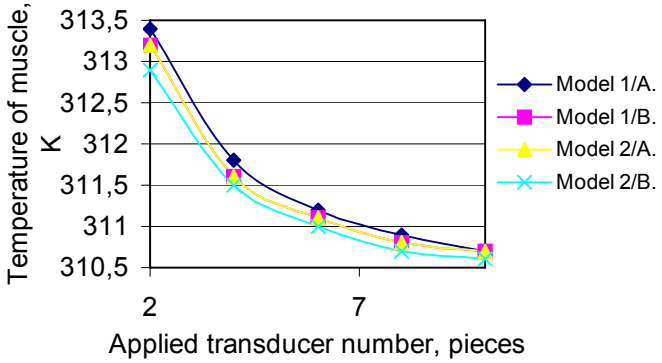

Figure 6. Relationship between the averaged muscle temperature and applied transducer numbers during the ultrasound hyperthermia

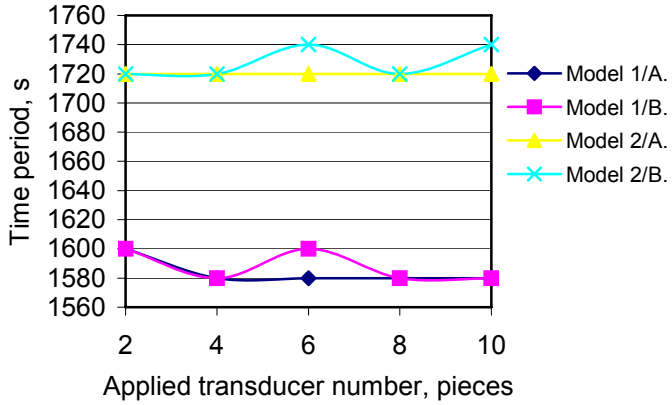

Figure 7. Relationship between the time period to 315K tumor temperature and the applied transducers

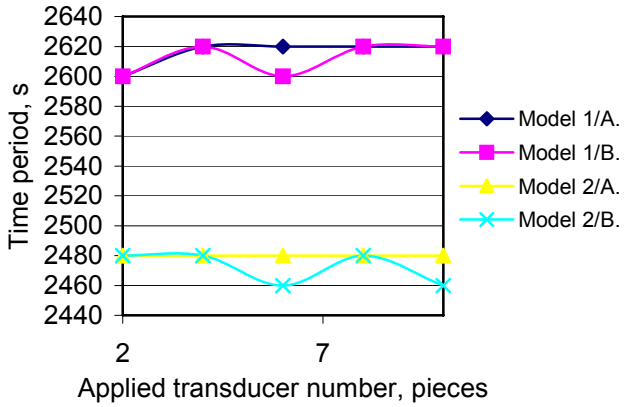

Figure 8. Relationship between the time period of 315-316K temperature zone and the applied transducers

The Fig. 7. and 8. shows that the achieving time to 315K and the length of 315-316K period based on the perfusion rate of blood in the tissues. This is interesting, because in the case of Fig 4., the effect of the blood was negligible.

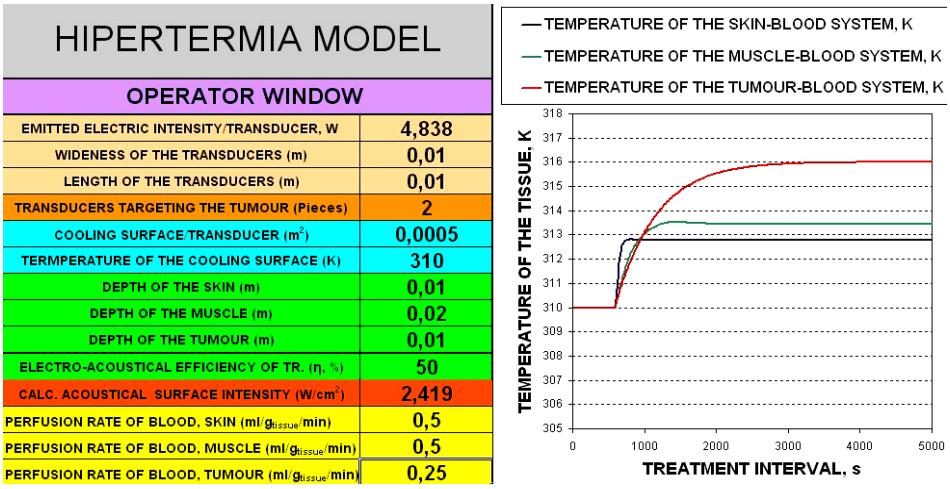

Figure 9. Denature temperature in the muscle during irradiation (situation: 1/A.)

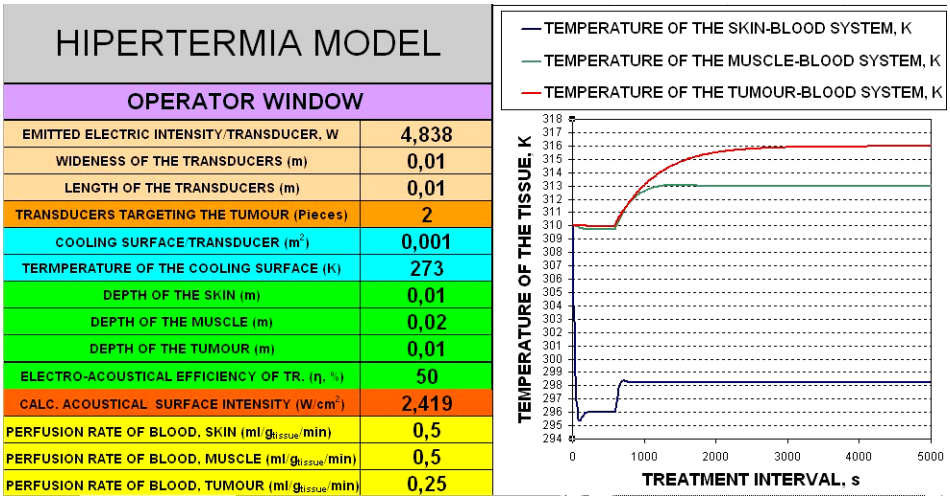

Figure 10. Decreased temperature with cooling

The Fig. 9. shows the control and indicator panel of model, where shown that during the irradiation in the muscle tissue generated more than 313K temperature. In this situation cooling surfaces were situated to the skin with 310K temperatures and because these, the skin temperature don't raised above the denature value. The tumor temperature always attained the 316K in the 4200. second and the temperature gradient were always saturation-likes. The Fig. 10. shows that with the increasing of cooling surfaces to 10cm^2 and decreasing of the cooling temperature to 273K, the heat energy can drain from the deeper muscle layers too, avoidable the heat denature danger of healthy tissues. During skin surface cooling, the tumor temperature decreased negligible. The treatments can be successful without surrounding tissue lesions. On the Fig 11 and 12. seen a same situation just here the skin and muscle temperature increased over the denature level without cooling. With use of cooling mode, avoidable the heat denature in the healthy tissues.

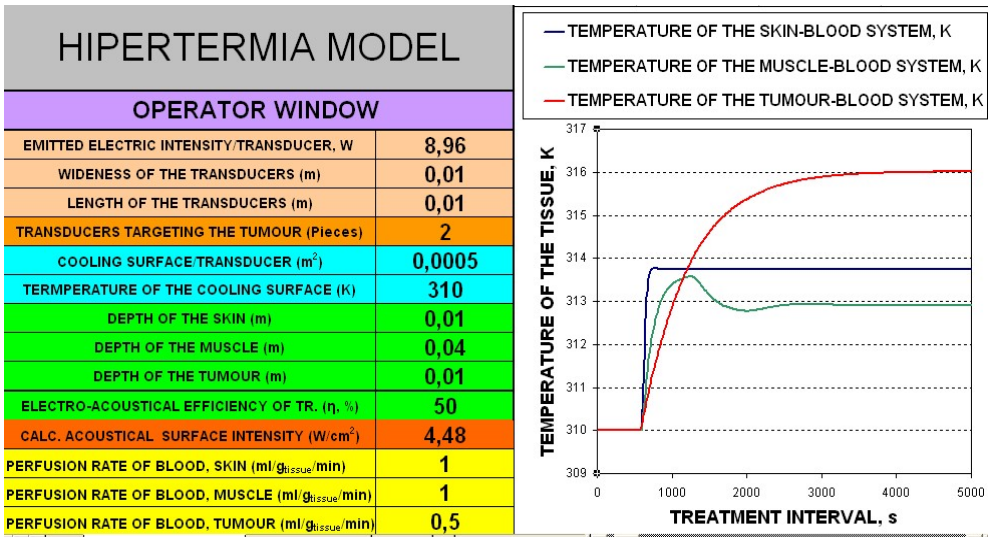

Figure 11. Denaturation temperature in the muscle and skin tissue (2/B. situation)

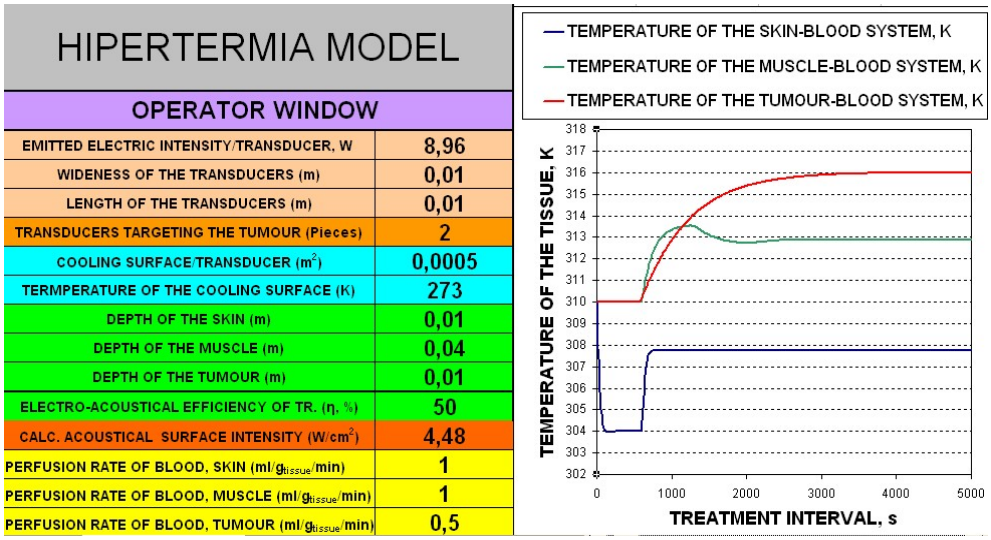

Figure 12. Eliminated over temperatures at 2/B. situation with cooling

4. Conclusions

- With use of the new computational hyperthermia model, determine-able the temperature gradient of the irradiated tissues during ultrasound treatment.
- The results are logical.
- The model takes in view many biophysical parameters and gives new information.
- The data in the model can be modifying.
- The ultrasound hyperthermia model is an effective tool in the field of research, study and presentations.

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