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MODELING OF TEMPERATURE FIELDS INDUCED IN TISSUES BY PHASED ARRAYS

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The influence of regular or random distribution of elements on a spherical shell of a two–dimension ultrasound phased array on acoustics and temperature fields in biological tissues is investigated. The occurrence of secondary maxima of the intensity in the acoustic field is analyzed, when the focal region is moved along the axis and in the transverse direction to it. Numerical algorithms are developed for the modeling of temperature distributions in biological tissues created by ultrasound phased arrays with different parameters. The developed algorithms allow the simulation of 3-D temperature fields of any spatial structure. In order to solve the problem, it is divided into two coupled parts. The Rayleigh-Sommerfeld integral is used for simulation of the acoustic field of the array. Heat transport and temperature rise are modeled using the Pennes bio-heat transfer equation with the relaxation term. The Pennes equation takes into account diffusion and cooling due to the microvasculatures (perfusion). Parameters of real phased array systems and biological tissues are used. It is shown that a random distribution of elements on the surface of the array eliminates the appearance of grating lobes and overheating of undesirable regions of tissues as compared with regularly distributed elements.

The application of focused ultrasound in surgery and high-temperature hyperthermia of tumors, where it is necessary to realize strictly controlled and properly located heating of biological tissues, attracts increasing attention. One of mechanisms of the effect of ultrasound on the tissues is the thermal mechanism applied in two modes [1]. The first of them is called hyperthermia. In this case the tissue is heated by ultrasound to a temperature of 42 - 45°C using moderate intensities of 1 - 10 W/cm^2 for a 30 - 60 minutes exposure. The second regime, in which a shorter exposure (approximately one second or less) and higher intensities of 500 - 1500 W/cm^2 are used, is called ultrasound surgery. Focused ultrasound radiators with high gains are employed to obtain such high intensity in a focal region. This regime enables to increase rapidly the temperature in the small focal region up to 60 - 90°C at which tissue ablation takes place. Alongside focusing radiators with a fixed focal length, designed on the base of concave piezoceramic plates [1], the use of phased arrays [2-6] has become perspective in many medical applications. They allow the creation of one focus or multiple simultaneous foci and steer them electronically at the necessary distances, thereby allowing an increasing the volume of irradiated tissue and reduce essentially the duration of medical procedure [2, 5-6]. However, the steering of the focal region along the axis of the array or in the perpendicular direction to it may result in the appearance of the secondary maxima of the intensity in the acoustic field created by the array. The physical nature of these maxima is caused by the presence of discrete elements in the array. If the steering of a single focus or multiple foci becomes significant (for example, more than 30 – 40 mm along the axis or 4 - 5 mm in the perpendicular direction to it), these secondary maxima can induce unsafe influence on healthy tissues outside the given site of irradiation. Some authors have investigated ways allowing of reducing this effect [4]. It is shown [3, 6] that the use of a random distribution of elements on the surface of two-dimensional array can be an effective method of suppression of secondary maxima of the intensity.

The aim of our work is to develop a method of numerical modeling of arbitrary three-dimensional temperature distributions without any restriction on spatial symmetry of the field, and, with the use of this method, to investigate the influence of various methods of distribution of the elements on the surface of the spherical array on the distributions of thermal sources and temperatures in biological tissue. For theoretical modeling of heating process, the problem is divided into two parts.

In the first one, the acoustic field and the corresponding field of thermal sources are modeled. For this purpose, a phased array with parameters typical for real two-dimensional arrays is considered: having the radius of curvature F = 12 cm, radius of the shelf r_0 = 5.5 cm, operating frequency f = 1.5 MHz and diameter of circular elements of 5 mm. The maximum center to center distance of the edge elements is 10 cm, the number of elements is 256. Two ways of the distribution of elements on the surface of the array are compared: a regular distribution in square patterns (which is the most popular way described in the literature) and a random distribution [6].

The three-dimensional field of the pressure of the circular element, taking into account its radial symmetry, is obtained by the numerical decision of the Rayleigh-Sommerfeld integral [2, 3]

$$p\,(\vec{r}\,) = \frac{i\; r_0\, c_0\, k}{2p} \int_S \frac{u(\vec{r}\,')\exp\left(-(i\,k + m\,)\,\left|\vec{r}-\vec{r}\,'\right|\,\right)}{\left|\vec{r}-\vec{r}\,'\right|}\, d\,S\,, \tag{1}$$

where p is the acoustic pressure in the beam, $ñ_0$ = 1500 m/s is the velocity of sound, r_0 = 1000 kg/ì^3 is the density of tissue, k is the wavenumber ($2p\,/\,l$, where l is the wavelength), $\vec{r}$ is the coordinate vector of a point (x, y, z), in which the pressure is calculated, $\vec{r}\,'$ is the coordinate vector of microelements with the sizes smaller than l , into which each element of the array was divided, z is the coordinate along the axis, x, y are the cross-section coordinates, m is the attenuation coefficient including absorption and scattering, a_0 = 5 m^{-1} is the absorption coefficient for the stated f, u is the normal to the surface of the array component of the oscillatory velocity . The phases of signals on the elements are determined from the condition of arrival of all the signals in the same point of the focus at the same time. The amplitude of the oscillatory velocity u_0 is identical for all the elements. With the use of the obtained phases and the angle coordinates of all the elements, the acoustic field of the whole array is calculated by summation of the fields of all 256 elements in the stated volume of interest. With the obtained values of the pressure, the intensity field is calculated $I\left(x,\, y,\, z\right) = \left|\,p\left(x,\, y,\, z\right)\right|^{2} \Big/ 2\; r_0\, c_0$, which is used for the reconstruction of the field of the thermal sources q_v .

$$q_v\left(x,\, y,\, z\right) = 2\;\, a_{\;0}\;\; I\,(x,\, y,\, z)\,.$$

In this model, the effects of acoustic nonlinearity and cavitation, which are characteristic for high-temperature hyperthermia and surgery, are not taken into account [7-8].

The time and spatial temperature distributions are computed using the Pennes bio-heat transfer equation [9]:

$$\frac{¶\,T}{¶\,t} = k\;\;\Delta T\; - \;\frac{T - T_0}{t}\; + \frac{q_v}{c_v}\,, \tag{2}$$

where $T = T\left(\vec{r}, t\right)$ - the temperature of tissue, T_0=36.6°C - the ambient temperature of tissue. Modeling of the process of heating is performed for a medium with parameters, characteristic for the real biological tissue: k = 1.399×10^{-7} m^2/s - the thermal diffusivity, t =250 s - the time of perfusion, $ñ_v$ = $ñr_0$ = 4.31×10^6 W·s/(m^3·°C) - the specific heat of tissue. The equation (2) takes into account the processes of cooling caused by diffusion of heat and intensive blood flow in vessels, located both in the heated volume and outside it (process of perfusion). The equation allows calculation of the temperature field in a tissue for arbitrary spatial distribution of the thermal sources q_v. The numerical finite-difference algorithm of integration of the equation (2) involves the use of the method of splitting procedure [8].

A model is considered, when the wave propagates in a tissue only. The modeling of heating process is performed for both regular and random distributions of elements on the surface of the array. The value of the intensity on the array surface in both the cases was selected to provide the maximum value I_{max} = 650 W/ñm^2 in the focus. In Fig. 1, the contours of the intensity distribution in the yz - plane at x = 0 are presented for the cases of regular (à) and random (b) distributions of elements on the array surface. The focal region here is shifted both along the axis at 2 cm from the geometrical focus of the

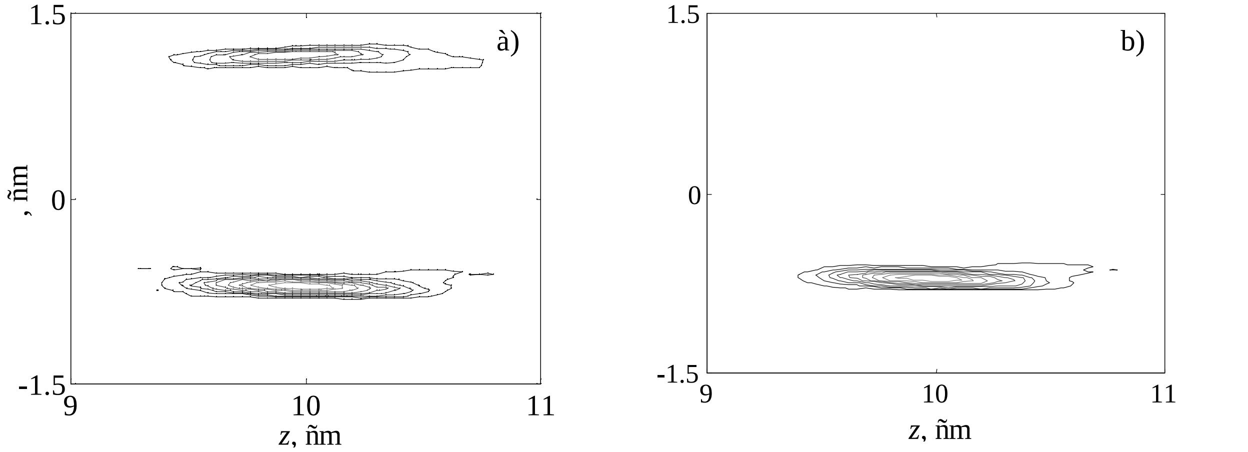

Fig.1. Intensity distributions in the yz - plane for the case of steering the focus to the point (x = 0 cm, y = - 0.7 cm, z = 10 cm) for regular (à) and random (b) distributions of elements of the array.

array, and off the axis at the distance y = - 0.7 cm. The maximum value of the intensity in the focus is equal to I_{max}. In Figs. 1à and 1b, the external contour corresponds to the level of the intensity of 0.1 I_{max}. The increment between the neighboring levels of the intensity is also 0.1 I_{max}. It is seen from Fig. 1a, that the shift of the focal region off the axis results in occurrence of the strongly expressed secondary maximum, the intensity in which exceeds 50% I_{max} in the main focus. In Fig. 1b, the spatial distribution of the intensity in the same *yz* - plane is shown for the random distribution of the elements on the array

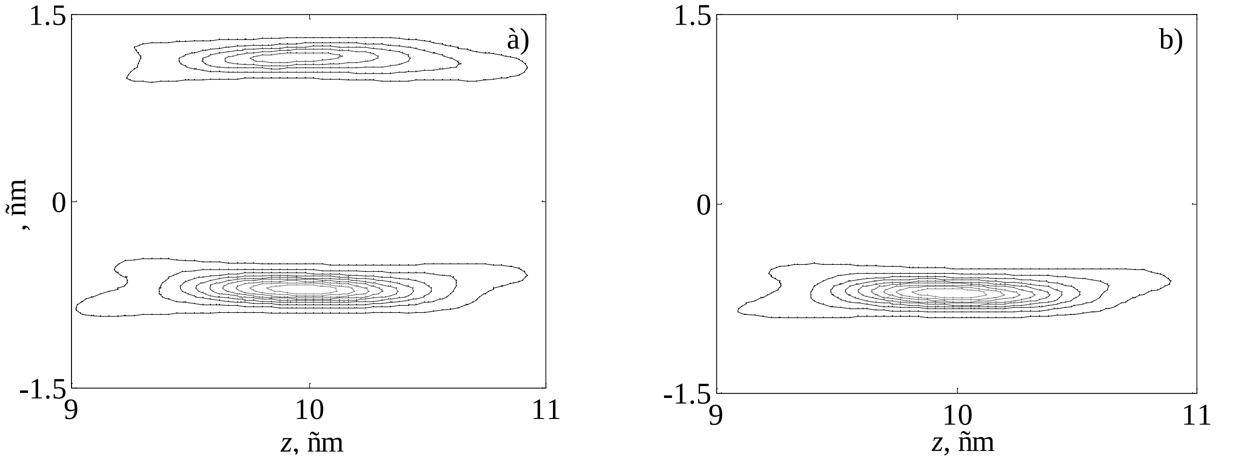

Fig. 2. Spatial distribution of temperature in the *yz* - plane for the case of steering the focus to the point (x = 0 cm, y = - 0.7 cm, z = 10 cm) for regular (à) and random (b) distributions of elements of the array; duration of heating t = 5 s.

surface. It is seen, that the secondary maximum of the intensity is absent, and only clearly outlined desired heated area of the tissue is observed. In Fig. 2, the spatial distributions of temperature in the *yz* - plane are presented at x = 0 cm for the cases of regular (à) and random (b) distributions of the elements on the surface of the array, for the correspondent intensity distributions presented in Fig. 1. The temperature fields were obtained for the exposure t = 5 s. Maximum temperature in the main focus $Ò_{max}$ = 90°C. The external contour in Figs. 2a and 2b corresponds to 10% of the maximum elevation of temperature $\Delta Ò_{max}$ = $Ò_{max}$ - 36.6 °C. The increment between the next levels of the temperature is equal to 0.1 $\Delta Ò_{max}$. It is seen, that diffusion smoothes the distribution of temperature, both along the array axis, and in the perpendicular direction, for both modes of the distribution of the elements on the surface of the array. From Fig. 2à, it is seen that, as well as in the case with the regular distribution of the intensity (Fig. 1a), the strongly expressed secondary maximum, in which the maximum value of the temperature exceeds 50%$\Delta \grave{O}_{max}$ that corresponds to $\grave{O}_{max} > 63°C$. Such essential overheating, at the mentioned above value of the exposure, can result in thermal necrosis of the tissue [1]. In Fig. 2b, for the case of the random distribution of elements on the array surface, the well expressed and clearly outlined focal region is seen in the given site of heating; the sites of overheating outside this area being absent.

It is known that phased arrays can be used successfully for synthesizing multiple simultaneous foci [2, 5-6]. The problem of occurrence of secondary maxima of the intensity is a vital question in this case too. The numerical algorithms developed in the present work enable evaluations of temperature fields in tissues for the case of creation and scanning of multiple foci.

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