FACILITY FOR THE ELECTROMAGNETIC HYPERTHERMIA BASED ON THE PHASED ARRAY

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Hyperthermia is a promising approach to improve of the chemo- and the radiotherapy efficiency by means of increasing tumor's temperature. Hyperthermia is an additional method to conventional treatments of oncological disease wherein tumor temperature is increased up to 40…43°C. The phased array of applicators for electromagnetic hyperthermia was suggested earlier. The heating is provided by absorption of electromagnetic energy focused in tumor by varying phases and amplitudes of each of dipoles. Operating frequency plays the significant role in specific absorption rate (SAR) distribution forming. The phased array antenna is under consideration. Principles of choosing of operating frequency are discussed. Numerical estimates of heat localization depending on the radiation area are presented. Also simulation results with voxel model are considered.

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INTRODUCTION

Hyperthermia is a method of cancer treatment by means of increasing tumor temperature up to 40…44°C. Hyperthermia is usually applied in oncology as an adjunct to the traditional methods. Under specific condition established treatment methods such as radiotherapy and chemotherapy are less effective. Hyperthermia makes tumor cells be more sensitive to the radiation and to the anticancer drugs. Many researches have shown that high temperature can damage and even kill tumor cells. The effect on surviving fraction depends both on the magnitude of the temperature and on the duration of the expose. Experimental studies show that tumor cell heating alone for 60 minutes at 43°C is damaging and that the period of exposure decreases by a factor of two for each increasing degree in temperature above approximately 43°C [1, 2]. So, 60-minute hyperthermia treatment at 43°C can be replaced by 15-minutes treatment at 45°C. The main mechanism for cell death is probably protein denaturation, observed at temperatures above 40°C, which leads to alternations in multimolecular structures such as cytoskeleton and membranes, and changes in enzyme complex for DNA synthesis and repair [3]. Also heat enhances the cytotoxicity of X-rays. Increased cytotoxicity observed over and above what would be expected on the basis of additivity of the two treatments, and it is maximized when these are given simultaneously. It decays with time when the treatments are separated by more than one or two hours [4].

The thermal therapy combined with the radiation (thermoradiotherapy, TRT) has been applying in N.N. Blokhin Russian Oncological Research Center (RORC) since 1980-th. More than 1000 patients have been treated up to now. Data presented in [5] demonstrate better efficiency of TRT than the radiation therapy applied alone. The complete tumor regression rates higher after combined treatment.

Also high temperature damages the healthy tissues and one of the most important problems of the hyperthermia technique is to prevent the overheating of these tissues. Various mechanisms for cell killing by heat have been proposed. But non-invasive heating of deep seated tumors is a difficult technique challenge. In this case the electromagnetic field has an advantage over other methods of physical expose (for instance, ultrasound) to create higher level temperature in a given volume of tissue. Radio frequency (RF) fields have a greater penetration rate than optic or infrared waves. Using the single applicator to heat tumors located deeper than 2…3 cm is not effective method. The most evident approach to reach deep-seated tumors is using an array of applicator situated around the patient body [6]. It should be a phased array to steer maximum of absorption power in patient body. Optimized distribution is reached by varying of the amplitudes and the phases of waves produced by each applicator. As a result of the constructive interference of E-field in the focus region, multichannel phased array of applicator can provide deeper penetration level of RF energy without overheating of skin and superficial healthy tissues.

1. THE PHASED ARRAY

Suggested phased array consists of eight copper dipoles, attached on the inner side of the dielectric cylinder, and surrounds a patient body. Top view of this structure is shown in Fig. 1,a. The aperture radius is up to 60 cm which can be applied in more cases. Dipoles are fed independently permitting to control phases and amplitudes of waves produced by each dipole. Space between the dipoles and the patient body is filled by deionized water (conductivity $\sigma \approx 0.001$). Thus applicators are squeezed from the inner side by lossy medium with high permittivity (deionized water $\varepsilon \approx 80$), and from the outer side by medium with low permittivity (air ε=1). The conducting elements of antenna are isolated from lossy medium by thin layer of an insulator (thickness $h \approx 1$ mm*)*. Because of energy density of electrical fields (1/2 *ED*) inside the dielectric tank is higher by a factor of ε (the relative dielectric constant of medium) than in the air outside, energy is mainly concentrated inside the array. Fig. 1,b,c care presented to demonstrate SAR distributions, which produced in a muscle tissue cylinder (conductivity at 150 MHz σ=

0.73 S/m and relative permittivity ε = 72.2) with diameter of 200 mm. Space between the muscle cylinder and the phased array is filled with deionized water (see Fig. 1,b) and with air (see Fig. 1,c). These distributions have similar behavior, but SAR value, produced in the phased array filled with deionized water, is substantially greater than without water because in the second case one side of each dipole has a contact with water. The RF field components in water are in ε times intensive than in air. SAR ratio is about 200. Thus deionized water not only cools body surface and superficial tissues but is also a matching medium.

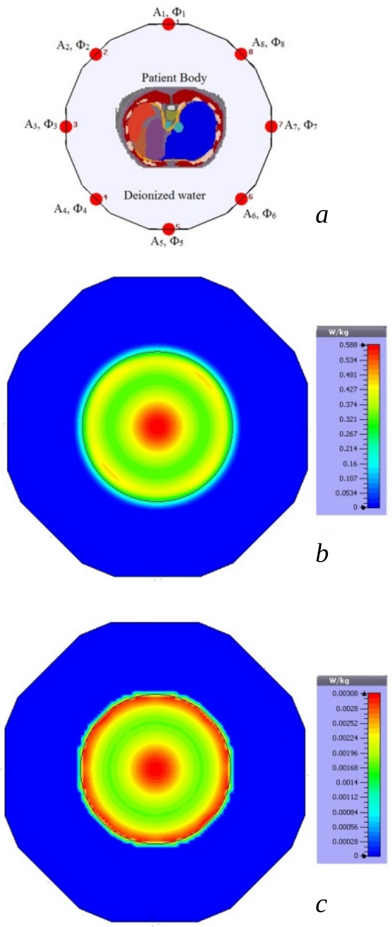

Fig. 1. Top view of phased array surrounding patient body (a); SAR distributions are produced in the phased array filled with deionized water and (c) with air space between dipoles and patient body (b)

Detailed scheme of dipoles including sizes is depicted in Fig. 2. Half of dipole is shown because of symmetry. In this picture two horizontal segments of antenna (signed as L_h) form two-conducting transform line that guides RF power to the vertical section (L_v). Vertical sections are radiating elements and horizontal segments contribution to the E-field canceled everywhere except proximity to the copper elements. Moreover vertical parts of antenna, insulation layer and lossy medium with high permittivity can be considered as a microstrip transmission line.

Electric fields lines are parallel to the axis of dipoles inside the phased array. As it known heat absorption is proportional to tissue conductivity (σ, which value for fat are significant lower than for muscle or tumor tissue). Thus overheating of skin and superficial tissues is reduced in comparison with using of capacitive applicators.

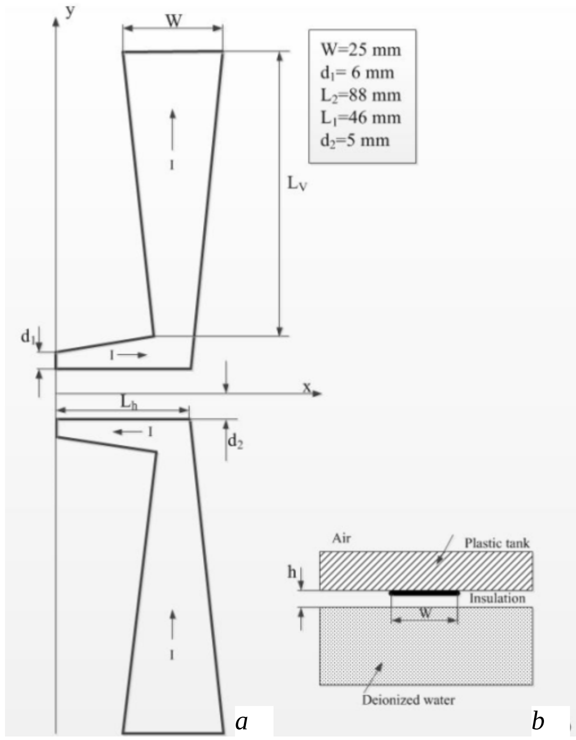

Fig. 2. Detailed scheme of dipole with dimensions (a); Cross-section view of radiating part of dipole (b)

E_z is the only component which able to control by shifting the amplitudes and phases of eight dipoles.

E-field generated by each of the dipoles is given by:

$$E_j = A_j E_{j0}(x,y)\exp\left[-i\left(\omega t - \Phi_j\right)\right], \qquad (1)$$

where E_{j0} – is the complex E-field for $A_j = 1, \Phi_j = 0$, and $A_k = 0$ for $j \neq k$. A_j – is a scaling factor of amplitude, Φ_j – wave phase, j and k – are numbers of dipoles. It is able to move peak of interference pattern and to focus it into the tumor site with the variation of these two parameters (phase and amplitude). The measure of the rate at which energy is absorbed by the body when exposed to a radio frequency (RF) electromagnetic field is a Specific Absorption Rate (SAR). It is defined as the power absorbed per mass of tissue and has units of watts per kilogram (W/kg):

$$SAR = \frac{\sigma \vec{E}^2}{\rho}, \qquad (2)$$

where σ – is the electrical conductivity of the tissue (S/m), ρ – is the density of the tissue (kg/m^3), E – is the root mean square electric field.

2. SIMULATION RESULTS

To demonstrate that the phased array can produce a maximum SAR distribution inside patient body, radiation simulations were performed. The simulations are performed with CST Microwave Studio. In previous report [7] SAR focusing ability was presented. Simulations were performed with limbs. The phase array can be used for different regions of patient body, but different regions require determining the optimal frequency range. For this purpose neck model was used and it is shown in Fig. 3. The model is a muscle cylinder of diameter 150 mm. Trachea filled by air, cartilage, spinal chord, bone and tumor are included in this model. The tumor is situated as more as deeper but not in the center of model to demonstrate focusing ability of absorbed power. Simulations were performed for two operating frequencies: 150 and 434 MHz. The simulations are performed also with CST Microwave Studio. Simulation results are presented in Fig. 4. Cross section SAR distributions with the coherent feeding of antennas are depicted in Figs 4,a,c, the optimized SAR distributions with relative phases are given in Figs 4,b,d.

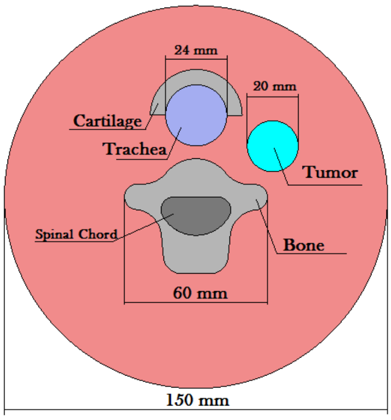

Fig. 3. Neck model used for simulation

Optimized SAR distributions are achieved with varying both the phases and the amplitudes of produced waves. The phases were chosen by the principle that the peak SAR shifts away from the dipole which has phase delay. The relative values of the phases and the amplitudes are noted opposite to each dipole. In this case not only phases varying are necessary due to presence of the bone tissue (ε=14.41; σ=0.07 at 150 MHz) and air filled trachea(ε=1; σ=0) in the center of the model. For coherent feeding at 150 MHz the peak SAR is situated in the muscle and tumor tissues around the low conductive and higher dense bone tissue (see Fig. 4,a). The penetration depth of electromagnetic fields at 150 MHz is greater than it necessary because the great volume of muscle tissue in the pattern of the optimized SAR absorbs too much power (see Fig. 4,b). As known the higher wave frequency, the higher tissue conductivity and the lower penetration depth, so simulation with operating frequency of 434 MHz was performed. When antennas fed coherently the peak SAR is situated, as in case of operating frequency of 150 MHz, in the muscle tissue (see Fig. 4,c), but the value of the peak SAR at 434 MHz is higher than at 150 MHz. The optimized SAR distribution is reached by means of varying both the phases and the amplitudes which are noted opposite the each antenna, and it is presented in Fig. 4,d. The good focusing of absorbed power in the tumor tissue is depicted in Fig. 4,d. Also the majority part of the healthy tissues is staying at normal temperature range. Thus the tumor localization size is 20…30 mm without overheating of the healthy tissues at 434 MHz for the neck model. For example, 40…50 mm localization size for limbs and 10…20 mm for the breast were reached during simulations (not presented in this paper).

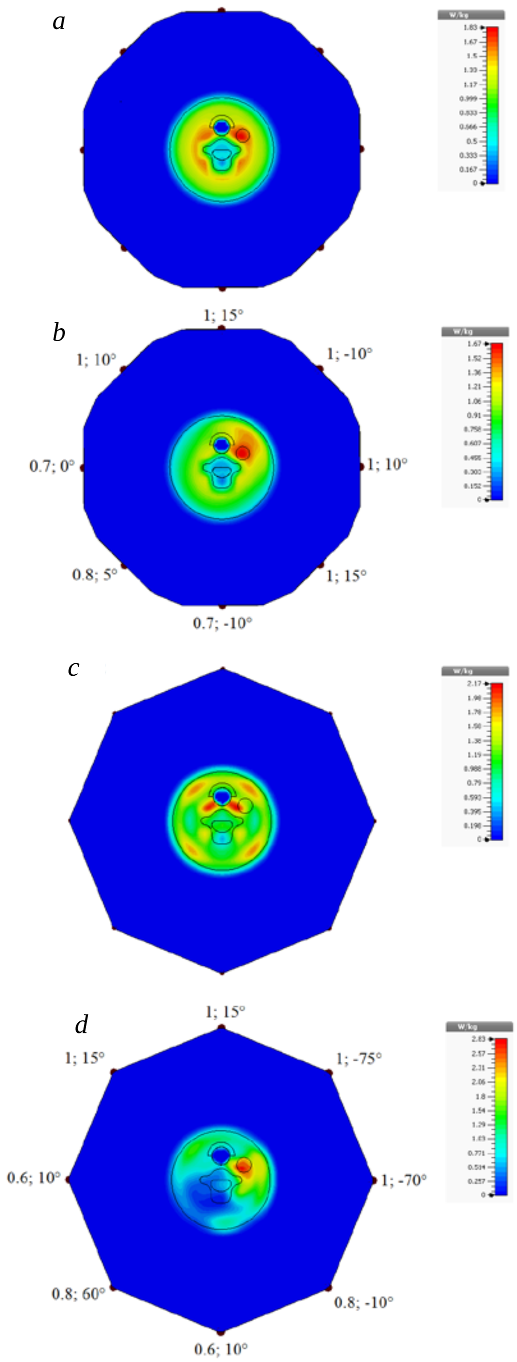

Fig. 4. SAR distribution with the coherent feeding at 150 MHz (a); optimized SAR distribution at 150 MHz (b); SAR distribution with the coherent feeding at 434 MHz (c); optimized SAR distribution at 434 MHz (d)

3. SIMULATION WITH VOXEL MODEL

A voxel (volumetric pixel) is a volume element, representing a value on a regular grid in three dimensional space. This data point can consist of a single piece of data, such as opacity, or multiple pieces of data. The value of a voxel may represent various properties. In CT scans, the values are Hounsfield units, giving the opacity of material to X-rays. Different types of value are acquired from MRI or ultrasound. After scanning the model is reconstructed. The voxel model of human body including 86 different tissues is presented in Fig. 5,d. Such parameters like dielectric and thermal properties, densities can be assigned to the each tissue [8]. Thereby the most accurate simulation can be performed by using voxel models. To demonstrate that the phased array can produce a maximum SAR distribution inside patient body in desirable region, radiation simulations with different operating frequencies were performed. The material properties of water shell, enclosure and antennas are also included in simulation. The blood flow and the thermal conductivity aren't factored in the simulation, thus the simulated pattern of SAR distribution have local maximum within vessels. The real SAR distribution hasn't these locals.

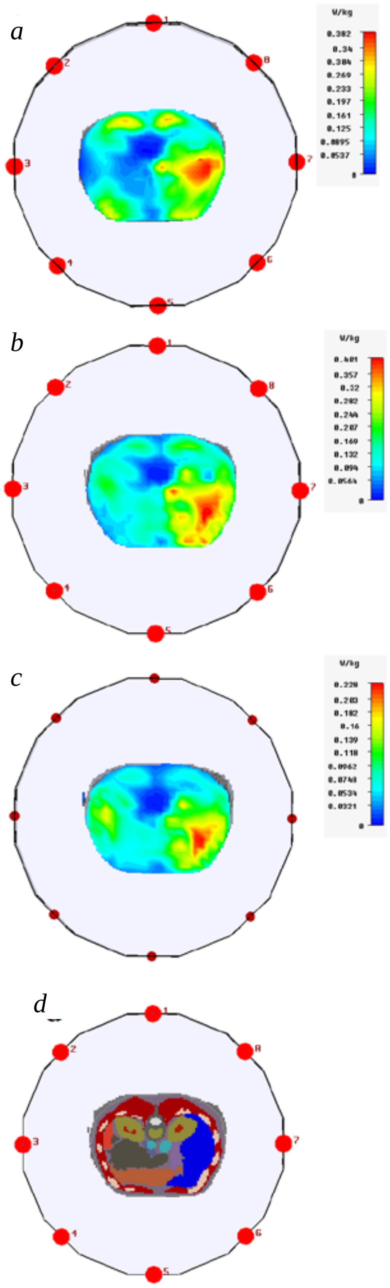

Fig. 5. Cross-section pattern of the SAR distribution with different operation frequencies (a) 150 MHz; (b) 100 MHz; (c) 80 MHz and with input phases of 50°, 50°, 50°, 50°, 0°, -30°, -40°, -10° applied to channels 1,2…8 respectively; (d) cross-section of the voxel model

There are three operating frequencies were used for SAR simulation: a) 150 MHz; b) 100 MHz; c) 80 MHz. Unfortunately, this model doesn't include the tumor tissue. Phases of each dipole were changed for estimating focusing ability in any region of the body. The liver was selected as a focus region (marked as blue on Fig. 5,d). Also we tried to minimize absorption rate in other soft tissues. Cross-section SAR distributions with relative phases are depicted on Fig. 5. The simulated focus is steered to the liver with input phases of 50°, 50°, 50°, 50°, 0°, -30°, -40°, -10° applied to channels 1,2…8 respectively. Also the goal was to prevent occurring hot spots in other regions and to maximize the SAR value in relative to the healthy tissue SAR. The phases were chosen by the principle that the peak SAR shifts away from the dipole which has phase delay, i.e. the peak SAR will be shifted away from the dipole #3 due to the phase delay on this dipole will be 50°. Amplitudes were not varied.

In the case of optimized SAR distribution the maximum SAR value was observed with the operating frequency of 100 MHz, and the minimum SAR value – of 80 MHz (see Fig. 5,b,c). This means that SAR distribution with lower operating frequency is more uniformly and the peak is specialized weakly. It can be concluded that the operating frequency of 100 MHz is the optimal for the voxel model with such transverse size. For models with greater transverse sizes the using of the lower operating frequencies is desirable.

The SAR distribution with operating frequency of 100 MHz, depicted on Fig. 5,b, has a peak in a liver site, and it is more uniform in the area outside the peak in comparison with other operating frequencies. Local peaks in other soft tissues (muscle) aren't appeared. It seems that the operating frequency 100 MHz is more preferable for the model of these transverse sizes. However it is hard to confirm that using of this operating frequency will be equally effective for the other transverse model sizes. It argues that hyperthermia planning procedure is necessary for each patient.

CONCLUSIONS

Heating of deep-seated tumors can be realized by means of focusing of radiofrequency energy inside the patient body. The phased array is suggested to increase the temperature level in the tumor side and on the other hand to prevent overheating of the healthy tissues. Dipoles geometry was discussed. Differences in the dielectric properties of tumor and healthy tissues play significant role in power absorption of RF fields and the voxel models, based on the real data like computer tomography or magnetic resonance imaging, provide the more accurate simulation of the hyperthermia planning procedure. It is shown that production of desirable SAR distribution inside the patient body by using the phased array is possible. By proper selection of phases and amplitudes of the waves radiated from each dipole it is possible to concentrate resulting EM-field in the tumor site. The phases are choosing by the principle at which the peak SAR shifts away from the dipole that has a phase delay. SAR distributions simulation results are presented and it promising performances are demonstrated. But the issues deals with the most effective operating frequency are still under discussion by reason of the different models with different cross sectional dimensions comply with relevant penetration depth of the EM-field. For the used voxel model of abdomen the operating frequency of 100 MHz is optimal. It is supposed that frequency range 60…120 MHz can overspread essential dimensional ranges. It is shown that the heating localization size can be not higher than 20…30 mm without overheating of the healthy tissues at 434 MHz for the neck model. The phased array prototype construction is the further step in our work.

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УСТАНОВКА ДЛЯ ЭЛЕКТРОМАГНИТНОЙ ГИПЕРТЕРМИИ, ОСНОВАННАЯ НА ФАЗИРОВАННОМ МАССИВЕ ИЗЛУЧАТЕЛЕЙ

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Гипертермия является многообещающим способом для увеличения эффективности химио- и лучевой терапии посредством повышения температуры опухоли. Гипертермия – дополнительный к традиционным метод лечения онкологических заболеваний, при котором происходит повышение температуры опухоли до 43…44°С. Фазированный массив излучателей для электромагнитной гипертермии был предложен ранее. Нагрев обеспечивается поглощением электромагнитной энергии, сфокусированной в опухоли при изменении фаз и амплитуд волн, генерируемых каждым из излучателей. Частота излучения играет значительную роль в формировании распределения удельного коэффициента поглощения (УКП). Рассматривается фазированный массив излучателей. Обсуждаются принципы выбора рабочей частоты. Представлены численные оценки локализации нагрева в зависимости от области облучения. Также рассмотрены результаты моделирования с использованием воксельных моделей.

УСТАНОВКА ДЛЯ ЕЛЕКТРОМАГНІТНОЇ ГІПЕРТЕРМІЇ, ЩО ҐРУНТУЄТЬСЯ НА ФАЗОВАНОМУ МАСИВІ ВИПРОМІНЮВАЧІВ

А.М. Фадєєв, С.М. Полозов, Е.А. Перельштейн, С.М. Іванов

Гіпертермія є багатообіцяючим способом для збільшення ефективності хіміо- і променевої терапії за допомогою підвищення температури пухлини. Гіпертермія – додатковий до традиційних метод лікування онкологічних захворювань, при якому відбувається підвищення температури пухлини до 43...44°С. Фазований масив випромінювачів для електромагнітної гіпертермії був запропонований раніше. Нагрівання забезпечується поглинанням електромагнітної енергії, сфокусованої в пухлині при зміні фаз і амплітуд хвиль, що генеруються кожним з випромінювачів. Частота випромінювання відіграє значну роль у формуванні розподілу питомого коефіцієнта поглинання (ПКП). Розглядається фазований масив випромінювачів. Обговорюються принципи вибору робочої частоти. Представлено чисельні оцінки локалізації нагрівання залежно від області опромінення. Також розглянуто результати моделювання з використанням воксельних моделей.