PLANNING SYSTEM FOR THE THERMORADIOTHERAPY PROCEDURE

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Hyperthermia is a method of treatment of oncological diseases by means of increasing tumor's temperature up to 40…43°C. Hyperthermia isn't applied alone but usually like adjunct to the established treatment methods such as radiation and chemotherapy. High temperature of tumor cells makes its more sensitive to the radiation and to the anticancer drugs. Creating of high level temperature distribution in the tumor site by the phased array of applicators was proposed. Varying phases and amplitudes of each dipole provides focusing electromagnetic energy in the deep situated tumors. It is very important to save healthy tissues from overheating. Prediction and control of temperature distribution inside the patient body is a significant part of the treatment process. The planning system for the thermoradiotherapy procedure is under development. Numerical simulation methods of the E-field and temperatures distributions produced by the phased array are described. These simulation techniques use a finite-element method. Resulting distributions of E-field and temperature are also presented.

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INTRODUCTION

Hyperthermia is an adjunctive method of treatment of oncological diseases by means of increasing tumor's temperature up to 40…43°C. Under this conditions tumor cells can be damaged and, moreover, efficiency of radiation and chemotherapy can be improved. But it is very important to save healthy tissues from heat destined for the tumor.

Treatments by means of radio frequency (RF) radiation are difficult challenge because RF energy is rapidly absorbed by human tissue. Therefore, treatment of tumors situated deeper than a few centimeters with a single applicator is not effective. Alternative decision is a using of several applicators positioned around the patient body in a configuration which allows obtaining constructive interference of the RF patterns. This conception was presented earlier in [1]. Suggested device consists of eight dipoles evenly spaced around a 60 cm annular space. Each dipole is powered and controlled by its own amplifier and phase shifter consequently. All dipole applicators are driven by the same frequency but, thus, each can have separate amplitude and phase values. This ability permits to steer maximum of the absorption power in the tumor site.

The operating frequency of dipoles can vary in the range of 80 to 150 MHz. Amplitude and phase variations together with frequency variation present a wide range degree of freedom in selecting of the treatment mode. Amplitude and phase values are made by intuition. The main rule of focusing strategy is that the maximum of the absorption power is shifted form the delayed dipole applicator. But it is very hard, and hence longstanding, to find the optimal values of amplitudes and phases guided by only this rule, especially for real time treatment. Moreover human body consists of huge amount of different tissues with its own dielectric properties and densities and each patient has its own anatomic peculiarity. That's why development of planning system for the hyperthermia procedure is very important for saving healthy tissues surrounding tumor from overheating.

1. ANALYTICAL METHOD

The problem of temperature prediction consists of two tasks: first, predict the rate of heat deposition in the various tissues; and second, predict the resultant temperatures. The latter problem is described by a role of blood circulation and thermal conductivity of tissues in the temperature regulation. But the main goal is to present the electric field distribution in the studied volume, which later will be used for temperature calculation with the same element grid. Since the power deposition per unit volume of tissue is given by (without blood flow and thermal conductivity):

$$SAR = \frac{\sigma |\boldsymbol{E}|^2}{2\rho}, \tag{1}$$

where $\boldsymbol{E}$ is the rms electric field, σ is a tissue conductivity and ρ is tissue density, the main goal is to complete the electric field distribution.

In this paper the general Galerkin formulation for the electric field is used.

In general case we should to solve Maxwell's equations:

$$\nabla \boldsymbol{E} = i\omega\mu \boldsymbol{H}\ , \tag{2}$$

$$\nabla \boldsymbol{H} = -i\omega\varepsilon_c \boldsymbol{E}\ , \tag{3}$$

where $\boldsymbol{E}$ and $\boldsymbol{H}$ are the time-variant complex amplitude of the electric and magnetic field correspondingly; μ is the magnetic permeability; $\varepsilon_c = \varepsilon + i\sigma/\mu$ is the complex permittivity; σ is the electric conductivity; ε is the permittivity; ω is the radiant frequency. As known μ is constant in tissue, but ε and σ vary with both frequency and tissue type. Therefore after isolating $\boldsymbol{E}$ the result is:

$$\nabla(\nabla \boldsymbol{E}) - k^2 \boldsymbol{E} = 0, \tag{4}$$

where $k^2 = \omega^2\mu\varepsilon + i\omega\mu\sigma$ is the complex wavenumber.

Since the object of study is the human body, it was decided that the Finite Element Method [2] was to be implemented. The principal idea of the method is the subdivision of the whole domain into a composite of simpler parts. Several advantages prompted the use of this method of which the ease of representation of complex geometry and greater flexibility in regard to the use of discontinuous coefficients in the equation, as compared to Finite Difference schemes, play the major role, with the latter being of even greater consequence than the former due to the nature of the problem.

Approximate solution is based on the weak form of (4):

$$\langle[\nabla(\nabla \boldsymbol{E})]\varphi_i\rangle - \langle k^2 \boldsymbol{E}\varphi_i\rangle = 0\,, \tag{5}$$

where $\langle\ \rangle$ indicates integration over the volume and φ_i is any scalar function. In this implementation of the method, however, the same weighing functions that are utilized for the field expansion on the elements of the mesh – so-called form function – are used. The first term in (5) may be integrated by parts:

$$\begin{aligned}\langle[\nabla(\nabla \boldsymbol{E})]\varphi_i\rangle &= \langle\nabla(\nabla\varphi_i \boldsymbol{E})\rangle - \langle\nabla\varphi_i(\nabla \boldsymbol{E})\rangle = \\ &= \oint \boldsymbol{n}(\varphi_i \nabla \boldsymbol{E})\,ds - \langle\nabla\varphi_i(\nabla \boldsymbol{E})\rangle.\end{aligned} \tag{6}$$

And thus the weak form is presented by:

$$\langle(\nabla \boldsymbol{E})\varphi_i\rangle - \langle k^2 \boldsymbol{E}\varphi_i\rangle = i\omega\mu\oint(\boldsymbol{H}\times\boldsymbol{n})\varphi_i\,ds\,. \tag{7}$$

Note that $\boldsymbol{H}$ was reintroduced purposefully for applying boundary conditions where $\boldsymbol{E}$ is not specified and computing $\boldsymbol{n}\times\boldsymbol{H}$ on the boundaries where $\boldsymbol{E}$ is specified. The unknown $\boldsymbol{E}$ should be expanded to complete the numerical discretization as follow:

$$\boldsymbol{E} = \sum_{i=1}^{N} \boldsymbol{E}_i\varphi_i\,, \tag{8}$$

where φ_i is the real-valued weighting functions. This Galerkin approximation is then implemented on conventional linear, hexahedral finite elements, where φ_i and $\boldsymbol{E}$ are continuous with piecewise continuous first derivatives.

The linear hexahedral element was chosen as an initial approach due to the fact that this type of element allows for non-constant gradients of the unknown quantity, hence making the mesh more sensitive to change.

2. SIMULATION RESULTS

In this study following model was used for simulations. Cylinder dielectric tank filled by deionized water is surrounded by just a pair of dipoles. Two dipoles are enough to demonstrate the ability of interference. Conductivity of deionized water is very low ($6\cdot10^{-7}$ S/m), thus it is expected that the absorbed power will be close to zero. What's why cylinder filled with conductive water (0.5 S/m) is situated in the center of the tank. Dipoles separate from water by thin dielectric layer (conductivity 0.02 S/m). The overall view of studying model is presented in Fig. 1.

This model is performed by CST Microwave Studio. Moreover, simulation results will be compared with CST results. In these simulations thermal conductivity was ignored, thus it is can be expected abrupt change of the absorbed power level in deionized water and dielectric tank interface and deionized and conductive water interface. Here conductive water is an equivalent of human tissue, for instance, muscle which has similar dielectric properties such as conductivity and permittivity.

Electric field distribution with equal phases along the line connected two dipoles is presented in Fig. 2. This line is sighed on the top of the figure. And corresponding SAR distribution along the line is depicted in Fig. 3. Cylinder with conductive water starts at 90 and ends at 190 mm.

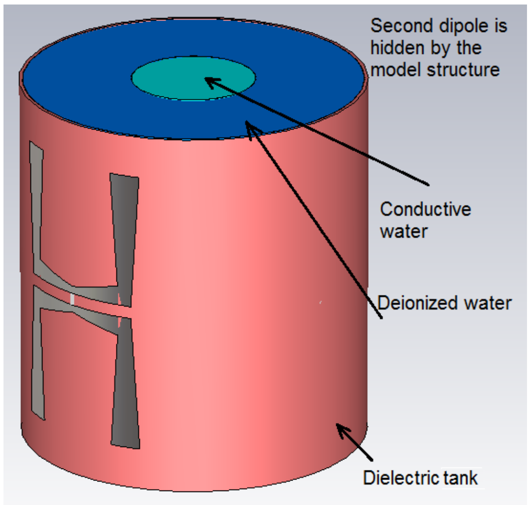

Fig. 1. Overall view of studying model

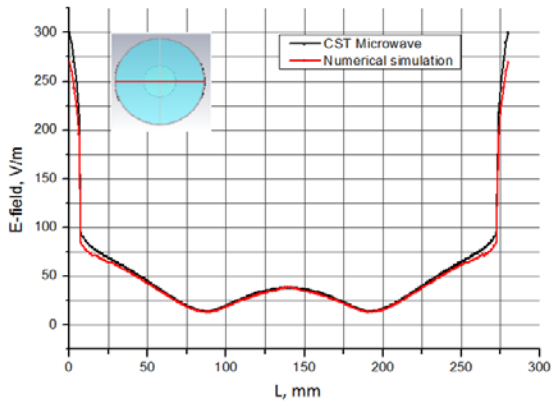

Fig. 2. Electric field distribution along the line

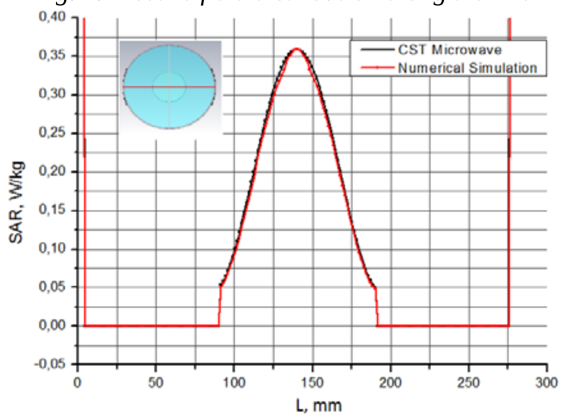

Fig. 3. SAR distribution along the line

Here black line is for CST simulation and red line is for numerical simulation with using Galerkin approximation.

Electric field has maximums values near the dipoles and they are situated in the dielectric tank. SAR distribution has the same maximums. Because of non-zero conductivity of dielectric tank it can be damaged, but in these calculations thermal conductivity is ignored. Thus, in this case the power absorption will be decreased.

Maximum of absorbed power in the center is a result of the summation of the absorbed power from each of two dipoles. Zero absorbed power in other region caused by low conductivity of deionized water. In the presence of thermal conductivity in simulating SAR distribution would be smoothed and high absorbed power values near the dipole region would be decreased. These figures show ability of deep heating.

SAR distribution with phase delay at the left dipole is shown in Fig. 4. Phase shift is 45 degrees.

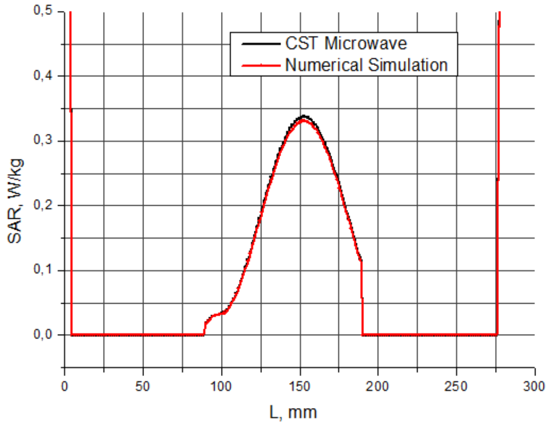

Fig. 4. SAR distribution along the line with the phase delay at the left dipole

According to the rule described in the first section the maximum of absorbed power should be shifter from delayed dipole, i.e. from center toward the right dipole. This behavior is depicted in the figure.

CONCLUSIONS

Hyperthermia is a promising adjunctive method of the treatment of oncological disease. Heating tumors by RF fields can be effective in case of multiple applicators using, where desirable absorbed power distribution is reached by amplitudes and phases varying of each dipole. Thus, it is very important to choose the most suitable values for the most effective treatment and for saving healthy tissues from overheating.

For these purposes planning system is under developing. Numerical method, which permits to produce electric field distribution, is presented. Simulation results were compared with CST Microwave Studioresults. It is shown that deep heating of biological tissues is possible and also it is possible to move peak of absorbed power.

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СИСТЕМА ПЛАНИРОВАНИЯ ДЛЯ ПРОЦЕДУРЫ ТЕРМОРАДИОТЕРАПИИ

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Гипертермия – это метод лечения онкологических заболеваний путем повышения температуры опухоли до 40…43°С. Гипертермия обычно не применяется отдельно, но используется как дополнение к установленным методам лечения, таким как лучевая и химиотерапия. Повышенная температура опухолевых клеток делает их более восприимчивыми к облучению и химиопрепаратам. Для создания повышенного уровня температур в опухоли предложено использовать независимо фазируемый массив излучателей. Варьирование фаз и амплитуд волн от каждого из излучателей обеспечивает фокусировку электромагнитной энергии в глубоко расположенной опухоли. При этом очень важно предотвратить перегрев здоровых тканей. Прогнозирование и контроль распределения температуры внутри облучаемого тела является значимой частью процесса лечения. Разрабатывается система планирования для процедуры терморадиотерапии. Описываются методы математического моделирования распределения электрических полей и температуры внутри тела, производимые фазированным массивом. Эти техники основаны на методе конечных элементов. Также представлены результирующие распределения электрических полей и температур.

СИСТЕМА ПЛАНУВАННЯ ДЛЯ ПРОЦЕДУРИ ТЕРМОРАДІОТЕРАПІЇ

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Гіпертермія – це метод лікування онкологічних захворювань шляхом підвищення температури пухлини до 40…43°С. Гіпертермія зазвичай не застосовується окремо, але використовується як доповнення до встановлених методів лікування, таких як променева і хіміотерапія. Підвищена температура пухлинних клітин робить їх сприйнятливішими до опромінення і хіміопрепаратів. Для створення підвищеного рівня температур у пухлині запропоновано використовувати незалежно фазований масив випромінювачів. Варіювання фаз і амплітуд хвиль від кожного з випромінювачів забезпечує фокусування електромагнітної енергії в глибоко розташованій пухлині. При цьому дуже важливо запобігти перегріванню здорових тканин. Прогнозування і контроль розподілу температури всередині опромінюваного тіла є значущою частиною процесу лікування. Розробляється система планування для процедури терморадіотерапії. Описуються методи математичного моделювання розподілу електричних полів і температури всередині тіла, вироблені фазованим масивом. Ці техніки базуються на методі кінцевих елементів. Також представлено результуючі розподіли електричних полів і температур.