A PROTOTYPE OF A PHASED ARRAY FOR DEEP THERMORADIOTHERAPY

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Abstract

It is proven that hyperthermia increases radiation and chemotherapy efficiency. In oncology, the generation of a higher temperature at a tumor-involved region of the body is called hyperthermia. The thermoradiotherapy is widely and effective uses. A phased array of eight dipoles for the hyperthermia treatment of deep-seated tumors is proposed earlier. The power and phase coherently delivered to the radiating elements can be varied, so that the electromagnetic field is increased at the tumor location and decreased in the normal tissues. The prototype of the phased array of two dipoles and the RF power scheme are presented and results of experiments are discussed. Measured and simulated temperature distributions along the line connecting two dipoles are discussed in this paper.

INTRODUCTION

Hyperthermia is an efficient adjuvant for the common modalities such as surgery, radiation and chemotherapy. Many researches have shown that hyperthermia temperature can damage and kill tumor cells, thus reduces tumor size. However the main advantage is that hyperthermia is a promising approach to increase efficiency of chemotherapy or radiation therapy. Under hyperthermia temperature some tumor cells become more sensitive to the radiation and anticancer drugs. The effect on surviving fraction depends both on the temperature increase and on the duration of the expose [1, 2]. Treatment requires that temperatures within tumor remain above 43 °C during 30-60 min, while maximum temperature in normal tissues have to be lower than 42°C.

In previous papers the phased array for deep hyperthermia was suggested [3, 4]. This phased array consists of eight dipole antennas arranged on an inner side of a cylindrical dielectric tank. Dipoles are surrounding the patient body and the amplitudes and phases of each antenna are under control of the operator as shown in Figure 1. Necessary distribution of E-field can be reached by means of independent feeding of each dipole that permits us to vary amplitudes and phases of electromagnetic field. In other words we can concentrate absorption energy of E-field and deliver therapeutic heat in tumor and at the same time prevent extra heating of normal tissues.

Deionized water filling space between patient and array is for cooling outer side of body and for better matching. The E-field energy is extremely concentrated in the inner side of a shell due to the electric field energy density inside the shell is higher by a factor ε (the relative dielectric constant of the medium) than outside the shell.

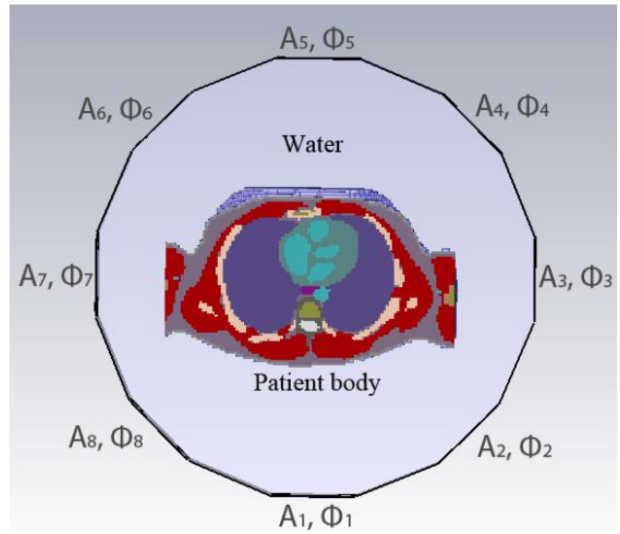

Figure 1: Patient body surrounded by phased dipoles array. Dipole antennas have amplitudes $A_1...A_8$ and phases $\Phi_1...\Phi_8$ respectively.

EXPERIMENTAL SETUP

In this paper the prototype of the phased array of two dipoles is presented. The RF power system schematic layout of this prototype is shown in Figure 2, where 1 – driving generator Agilent N5181A, 2 – preamplifier Analog Device 5545 (fixed gain 25 dB, 30 MHz-6 GHz, 5 V), 3 – Wilkinson power divider, 4 – phaseshifter Mini-Circuit JSPHS-150 (100-150 MHz, 0-12 V), 5 – amplifier Toshiba S-AV32A (134-174 MHz, 12.5 V, 60 W), 6 – matching circuit. Single phaseshifter is enough to produce any phase lag between two dipoles. The operating principle of such layout is the following. The RF signal at 150 MHz from signal source splits into two channels by microstrip power divider. Then by means of controlled one phase shifters and two solid state amplifiers we can adjust phase and amplitude of every signal. Due to these adjustments peak temperature moving is available. Solid state RF amplifier Toshiba S-AV32A is simple and stable in operation but it needs intensive cooling system.

For impedance measurements a commercial network analyzer system Agilent Technologies E5061A was used. Impedance matching is provided with short circuit stub. The stub is positioned a distance from the load. This distance is chosen so that at that point the resistive part of the load impedance is made equal to the resistive part of the characteristic impedance by impedance transformer action of the length of the main line. The length of the stub is chosen so that it exactly cancels the reactive part of the presented impedance. Return loss plot for one

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single channel after matching is depicted in Figure 3. The operating frequency is about 150 MHz.

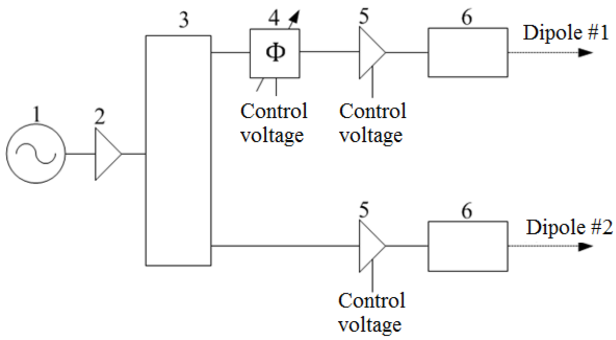

Figure 2: RF power schematic layout.

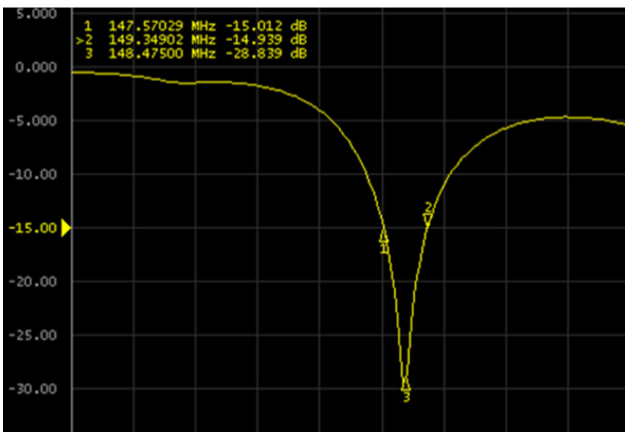

Figure 3: Return loss plot after matching with short circuit stub for one single channel.

As it noted above this phased array consist of two dipoles, thus temperature peak can move along the line connecting these dipoles. Water is used as an absorbing medium because it has dielectric properties and density similar with the muscle tissue. A number of dielectric tubes were positioned along the line connecting dipoles like it is shown in Figure 4 to prevent water blending during heating. These tubes were also filled by water and temperature measurements were carried out inside tubes. Tubes are made of aluminum oxide (permittivity ε=9.4 and density ρ=3990 kg/m^3). Tubes diameter is 10 mm and wall thick is 2 mm.

Thermocouple-sensing element ATE-9380 was used to perform temperature measurement. Whereas it has metallic compounds TSE introduce alternations in electric distribution when RF is on. Thus temperature measurement was carried out after switching RF off.

In these experiments, all channels were driven separately with a power of 15 W. Initially water temperature inside shell and tubes was equal. After switching RF on temperature starts to increase. In the first series of experiments dipoles have equal phases. Phase lag became 60° to move temperature peak. That was the second set of experiments.

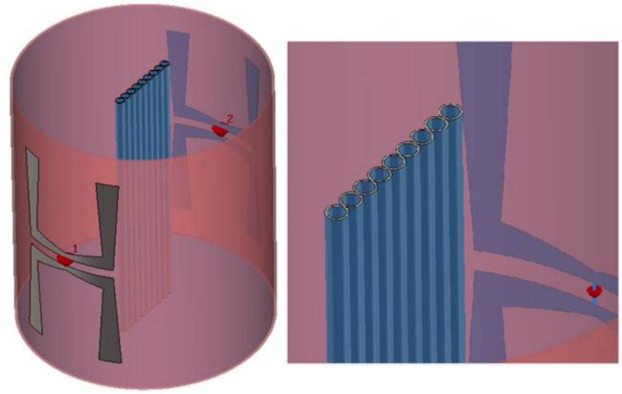

Figure 4: Experimental setup schematic view.

RESULTS

Experimental results were compared with simulation one. Simulations results are performed by CST Studio Suite [5]. Both dielectric and thermal properties of plastic enclosure, water, dielectric tubes were included. The simulations are performed with CST Microwave Studio which uses the Finite Element Method (FEM). Their solvers feature curved elements of arbitrary order. These elements enable a conformal representation of the geometry which improves the simulation accuracy. In combination with the unstructured FEM grid, which can resolve small structure details very efficiently, it can increase the simulation performance dramatically.

Temperature distributions are shown in Figure 5. Red line is for experimental data and black line is for simulation results. “0” in Figure 5 is the midway between dipoles. Initial water temperature was equal 23.4°C. When phases stay equal the temperature peak is situated in the middle tube. After 12 minutes of in-phase excitation temperature in the middle tube has reached 24.4°C.

When phase lag is equal 60 the temperature maximum is moved away from the dipole that has phase delay and after 20 minutes maximum temperature was situated in the third tubes at the left and it equal to 25.2°C.

Also cross section temperature distributions for both cases are presented in Figure 6. As it expected peak temperature is observed in the middle tube when phases are equal (Fig. 6a). Peak is moved away from the center when phase lag is 60° (Fig. 6b). According this picture we can conclude that dipole №2 has the phase delay, because the temperature maximum shifted away from this dipole. Since this array consist of two dipoles, the resulting temperature distribution is spread out and the peak is not so clearly identified as it could be if we would use eight dipoles.

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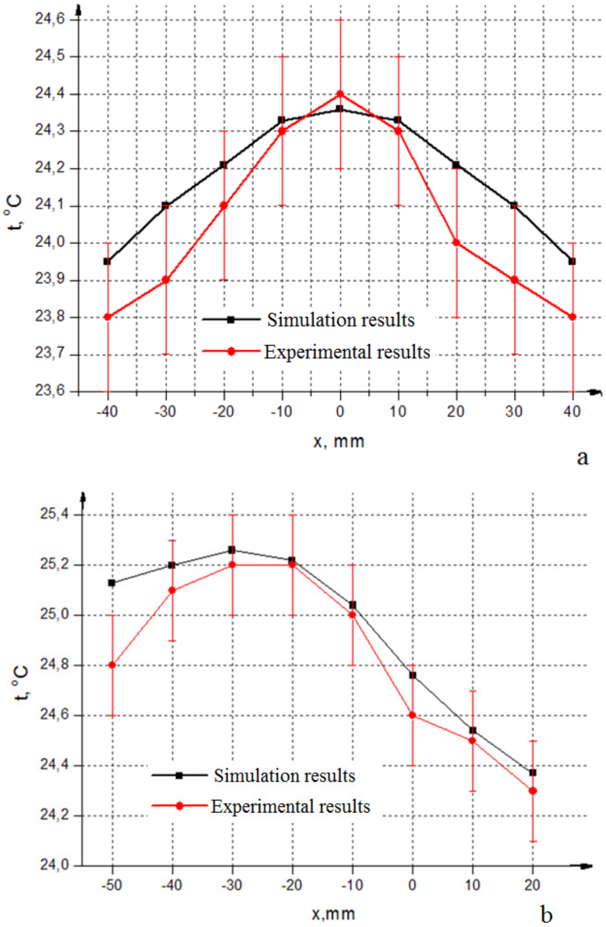
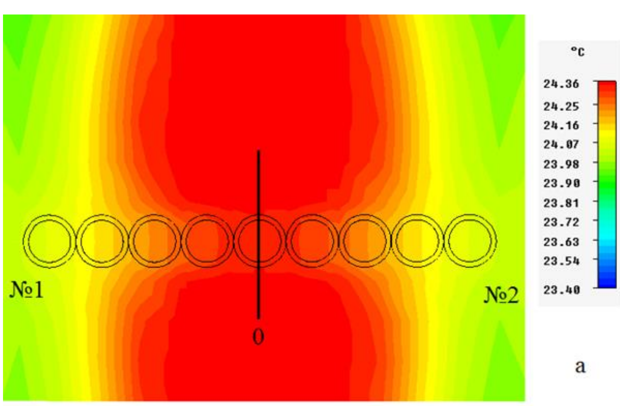


Figure 5: Temperature distribution along the line connecting two dipoles for a) in-phase excitation b) phase lag 60°.



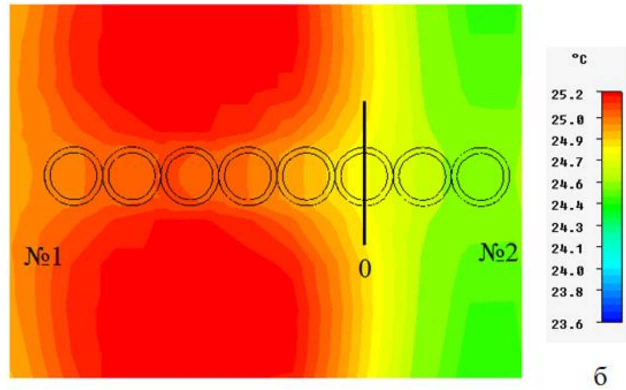


Figure 6: Cross-section temperature distributions for a) in-phase excitation and b) phase lag 60°.

CONCLUSION

To heat deep-seated tumor without overheating of healthy tissues is a complex technical challenge. Heating of deep-seated tumors can be realized by means of focusing of radiofrequency energy inside the patient body. The phased array was suggested to create desirable temperature distribution inside patient body. The first step in realization of this facility was done. The prototype of the phased array of two dipoles has been shown. Experimental and simulation results on water heating are presented. It is proved that the phased array can produce temperature gradient by introducing phase lag between dipoles. Experimental and simulation results agree within the experimental error.

The phased array prototype of eight dipoles construction and software developing for the hyperthermia planning system are the further steps in our project.

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