Optimal Design for Microwave Hyperthermia Applicator

Mihaela Morega, Laura Mogos, M. Neagu, Al. Morega
POLITEHNICA University of Bucharest, Department of Electrical Engineering, Bucharest, Romania

Abstract— **The thermal effects that occur when the human body is exposed to electromagnetic radiation in the microwave frequency range could be used in medicine to treat tumors through hyperthermia. The coupled phenomena, electromagnetic and thermal, are analyzed here in a numerical experiment that introduces a simple and efficient configuration for a minimally invasive microwaves applicator and suggests the method for the optimization of its design. Dosimetric parameters are computed and the energetic efficiency of the applicator is evaluated.**

I. INTRODUCTION

Hyperthermia based therapies represent noninvasive or minimally invasive medical procedures for the treatment of tumors in soft tissues (liver, kidney, breast, etc.); it is still an experimental method, but the perspectives are promising because it ensures therapeutic efficiency as much as patient comfort. The method provides the rise of the temperature in the tumorous tissue to 43 – 45 ^{0}C for extended periods of time, on the order of minutes or even hours, while metabolic changes are expected to occur in metastatic cells, aiming to their destruction [1, 2].

The use of heat to reduce or eliminate tumors has been known for more than a century, but only in the past three decades the means for accomplishing reasonably controlled heating have been available. The first period of intense research in hyperthermia occurred during the 1970s and 1980s, and in 1984, it was legally adopted in the USA as a medical procedure, specifically recommended for locally recurrent tumors, and for primary cancer. After that success, at least in the USA, there was a decrease in research activity, primarily caused by the difficulty in obtaining reliable and uniform heating at deep tissue sites. In the late 1990s, there has been resurgence in hyperthermia interest, and new approaches and new design for the applicators were developed. Some of these new systems use ultrasound instead of electromagnetic energy for generating heat. Currently in USA and Japan significant progress in microwave hyperthermia is reported [3 - 7]. For better results, hyperthermia is associated with ionizing radiation and chemotherapy.

Heating through exposure to microwaves (electromagnetic field (EMF) radiation in the frequency range of 10^8 – 10^{11} Hz) is explained by the energy transferred to the exposed tissue, primarily via capacitive coupling, which causes vibration of polar particles, mainly the water molecules. Soft tissues have high water content, and more than that, tumorous tissue is especially hydrated and becomes the region that preferentially concentrates heat. The frequency of the electric field determines the rate at which energy is delivered to the tissue. The higher the frequency, the faster the wave will lose energy as it propagates through the tissue, and the shallower the heating. This is a key factor in determining the operating frequency for various applicators.

For the treatment of superficial tumors the radiation is applied through external antennas, while internal tumors are exposed to invasive applicators. The operating frequency is usually 2.450 GHz, which is one of the ISM (Industrial, Scientific, and Medical) dedicated frequencies. Microwave radiation has a low penetration depth in anatomical tissues, on the order of 0.015 – 0.025 m [8]; consequently, external applicators manifest the inability to deliver uniform thermal doses to tumor volumes. As a solution to that problem, local interstitial techniques have been developed that are proving to be safe and effective. These techniques employ implanted minimally invasive thin antennas for the delivery of local thermal doses; they are inserted through the skin, into a biocompatible catheter, under the guidance provided with an imaging monitoring procedure (for example ecography) [5-7].

The propensity of the tissue to produce heat in the presence of the time variable electric field is determined by the values of its electric properties, the electric conductivity σ, and the dielectric permittivity ε, and by other physical properties. The dosimetric quantity that relates to heat generation in tissues is the Specific energy Absorption Rate (*SAR*), defined as

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

where ρ represents the mass density of the tissue and E is the rms value of the time harmonic electric field strength inside the exposed tissue. The distribution of the electric field is determined from the electromagnetic waves equation solved here for time harmonic electromagnetic source (with the angular frequency $\omega = 2\pi f$), and expressed in complex form for the electric, respectively for the magnetic field strength

$$\begin{aligned} &\nabla \times \left(\frac{1}{\mu_0} \nabla \times \underline{\mathbf{E}} \right) - \omega^2 \underline{\varepsilon} \underline{\mathbf{E}} = 0, \text{ or} \\ &\nabla \times \left(\frac{1}{\underline{\varepsilon}} \nabla \times \underline{\mathbf{H}} \right) - \omega^2 \mu_0 \underline{\mathbf{H}} = 0. \end{aligned} \tag{2}$$

The wave equations are applied for lossy linear media, characterized by the complex electric permittivity $\underline{\varepsilon} = \varepsilon - j\sigma/\omega$, while the biological tissue behave like a nonmagnetic material ($\mu_0 = 4\pi 10^{-7}$ H/m). The heating is illustrated by the temperature distribution estimate, expressed with the bio-heat equation, first introduced by Pennes [9]

$$\rho\, C\, \frac{\partial T}{\partial t} = k\, \nabla^2 T - \rho\, \rho_b\, C\, F\left(T - T_b\right) + \rho\, SAR\,, \qquad (3)$$

where T and T_b are the temperatures of the thermally treated tissue and of the blood, t is the time, ρ and ρ_b represent the densities of the tissue and blood, C and C_b are the specific heat coefficients of the tissue and blood, k represents the thermal conductivity of the tissue and F is the blood flow rate. The local temperature rise due to electromagnetic energy deposition is limited through heat transfer (conduction and convection) by blood flow, as (3) shows.

An aspect that remains problematic in hyperthermia with microwaves is the ability to heat deeply and uniformly the target tissue, in a well-controlled localized manner; the temperature is high in the vicinity of the applicator and decreases rapidly with the distance. For external applicators, surface overheating accompanies deep heating, and for interstitial applicators a nonuniform temperature distribution occurs in the tumorous tissue volume. An attempt to solve this problem looks for adequate configurations of the applicator; the most popular solution was found to be the array antenna, with specific distribution of pins, either for external, or for interstitial application. In particular, minimally invasive microwave thermal therapies using thin applicators (pins or needles) are of a great interest [6, 7]. Their design (the number of the needles, dimensions and distances, working frequency, etc.) is very much related to the characteristics of the target region (shape, dimensions and electric properties of the tumor and of the surrounding tissue).

II. Models and Methods

The research presented further refers to the optimization of the hyperthermia procedure, applied to hepatic tumors as a minimally invasive microwave therapy. The study follows two steps: (*A*) numerical modeling of the electromagnetic and thermal fields for a thin coaxial antenna inserted into the liver tissue, where the penetration depth for both fields is evaluated and the design of the antenna is optimized for maximum power transfer and (*B*) the optimization of the spatial configuration of an array interstitial applicator, with regard to the temperature distribution in a tumorous liver tissue with known characteristics (dimensions and physical properties). A 2D finite element model (FEM) is developed to determine the absorbed power and the temperature distribution surrounding the single coaxial antenna; further, a 3D FEM model is used for the optimization of the array applicator's configuration.

A single thin coaxial antenna used in this study is presented in fig. 1.a. The cable is short-circuited at the tip and a ring shaped slot, 1.5 mm wide, is cut at 6 mm distance from the tip. This particular geometry of the coaxial antenna is the result of an optimization process with regard to the efficiency of the energetic transmission from the antenna to the exposed tissue. The radial dimensions of the pin, as much as the position and axial dimension of the slot were found to provide the optimal energetic coupling between the microwave source and the liver tissue. Several geometries were subjected to trials before the dimensions were set to those presented in fig. 1.a. The antenna is inserted with the aid of a catheter (made of poly-tetra-flour-ethylene - PTFE, a biocompatible dielectric material), for both hygienic purposes and guidance; after the procedure, the catheter could be maintained inserted into the tissue for a later intervention. The array applicator is made with several thin coaxial antennas of the same type; the optimal configuration should provide the most uniform distribution of the temperature in the tumor and therefore consists of several pins, equally spaced if possible, as fig. 1.b shows for the geometry with seven, respectively three pins. The number of pins depends on the dimensions of the targeted tissue volume.

The physical properties of the materials involved in the model and specified in (1), (2) and (3) were selected from literature [6,7,10]. The electric properties are considered at

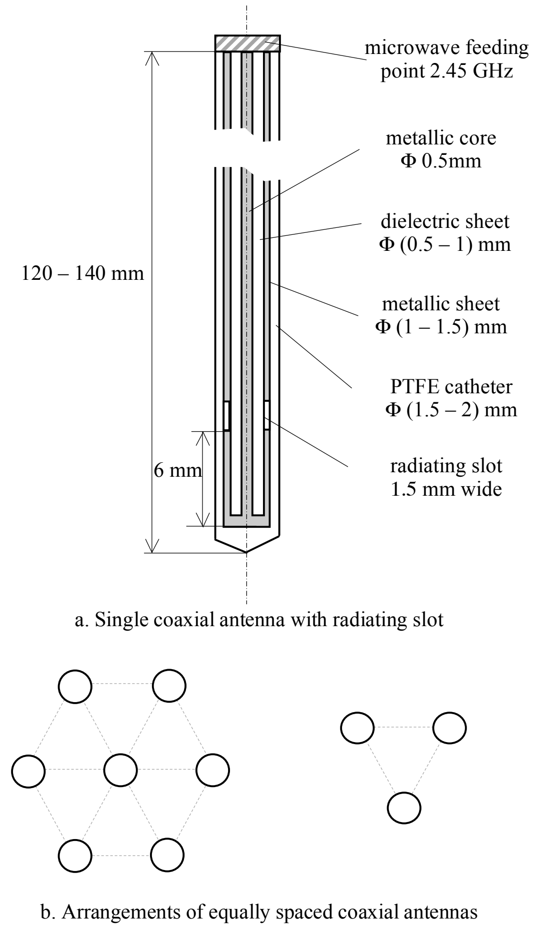


Figure 1. Design characteristics for the array microwave applicator

2.45 GHz, the working frequency; the liver electric conductivity and dielectric relative permittivity are $\sigma_{liver} = 1.66$ S/m and $\varepsilon_{liver} = 42.6$; the temperature dependence of the electric properties is not considered, because in hyperthermia, temperatures do not exceed 43–45°C. In [6] it is shown that, at temperatures below 45°C, the change in the electric field strength, current density, electrical conductivity, *SAR*, and the heat flux are about or less than 10%, which is inside the range of data dispersion due to natural variations. The PTFE catheter and the sheet of the coaxial cable are nonconductive materials and the values of their dielectric relative permittivities are $\varepsilon_{PTFE} = 2.6$ and $\varepsilon_{diel} = 2.03$.

One could consider that the PTFE catheter is a thermal insulator and the thermal problem is limited to the tissue domain; therefore, thermal properties are only needed for liver and blood. The specific heat, the thermal conductivity and the density of the liver have the values: $C_{liver} = 3600\ \mathrm{J/\left(kg{\cdot}^{o}C\right)}$, $k_{liver} = 0.502\ \mathrm{W/\left(m{\cdot}^{o}C\right)}$, $\rho_{liver} = 1060\ \mathrm{kg/m^3}$. The blood flow rate affects the thermal field distribution, as (3) shows, and it is considered here at the constant value $F = 6.4 \cdot 10^{-6}\ \mathrm{m^3/(kg \cdot s)}$; other properties of the blood, necessary for the thermal problem solution are the specific heat $C_b = 4180\ \mathrm{J/\left(kg{\cdot}^{o}C\right)}$ and the mass density $\rho_b = 1000\ \mathrm{kg/m^3}$ [4]. While we consider here that σ_{liver} and ε_{liver} do not change with temperature, the electromagnetic and thermal problems are unidirectional coupled, through the thermal source, *SAR* in (3), computed as (1) shows, with the solution of the EMF equations (2).

III. Numerical Computation and Results

We applied here the finite element method (FEM) for the numerical modeling of the two problems. The Comsol Multiphysics software package [11] is used for the FEM implementation.

A. *Numerical modeling for one thin coaxial antenna inserted into the liver tissue*

The antenna described in fig. 1a, fed at 2.45 GHz, inserted in a volume of material with liver properties is represented as a 3D FEM model. The liver tissue has the shape of a cylinder, coaxial with the antenna, with a radius of 30 mm, large enough to enclose the volume affected by the EMF source. The external surface of the cylinder acts as boundary for the computation domain. In the EMF problem, a low reflecting boundary condition is set on that surface, which means that the boundary does not disturb the EMF distribution, while in the heat transfer problem the boundary is thermally isolated. The microwave source is set at the upper end of the coaxial cable and the power emitted is adjusted to 1W. The antenna design was previously optimized, using the same numerical model, in order to maximize the transferred energy between the source and the liver tissue. Taking advantage of the axial symmetry of the domain, a much more economical 2D FEM model was also produced and intensively used in the preliminary study of the antenna design and in the primary estimates of the EMF parameters and temperature. Fig. 2 shows the temperature distribution at the end of the heating process (the steady state), in a longitudinal plane,

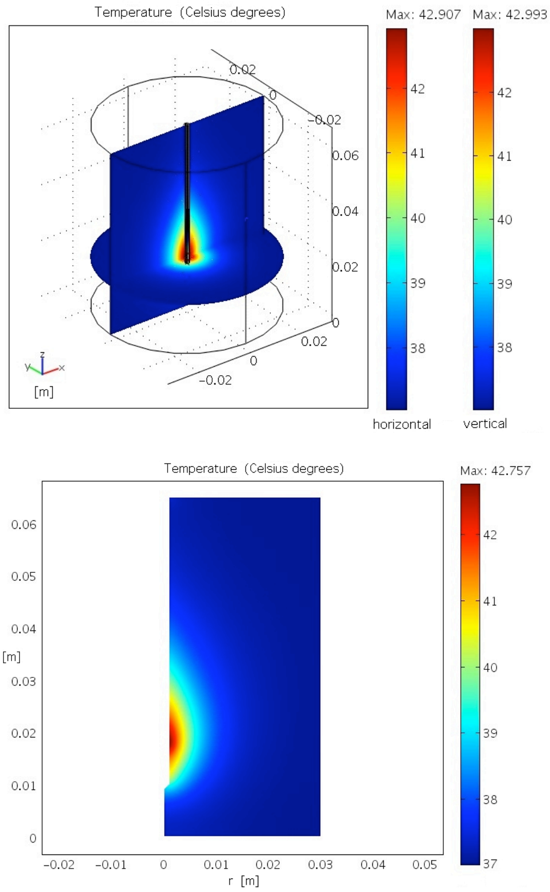


Figure 2. Temperature distribution in the tissue, 3D FEM model (up) and 2D FEM model based on axial symmetry (down)

which includes the symmetry axis of the 3D model (up), and in the 2D FEM model (down).

Fig. 3 presents the *SAR* and temperature distributions computed with the 2D FEM model, on the axis with maximal values, perpendicular to the antenna, at the upper end of the radiating slot. The penetration depth of the absorbed power is found to be 1.55 mm (evaluated as the distance where *SAR* decreases at 1/e of its incident value). An array applicator with several pins could be used to get a larger heated volume and a more uniform temperature distribution in that volume. Fig. 3 suggests that the distance between two adjacent pins should be at least the double of the penetration depth, which means 3.55 mm between the antenna longitudinal axes. This supposition will be checked further with a 3D FEM model for an array applicator.

B. *Optimization of the spatial configuration of an array interstitial applicator*

For the optimization study presented here, a three pins array applicator (with the coaxial-slot antennas equally spaced, as in fig. 1.b) is analyzed. The distance between the longitudinal axes of the adjacent identical pins is varied in the range 5 - 15 mm. For a better evaluation of the results, the emitted power is 1W for each pin in all analyzed cases.

The computation model of the array applicator with 7mm distance between pins is shown in fig. 4, and includes two *SAR* observation planes; one is a cross-

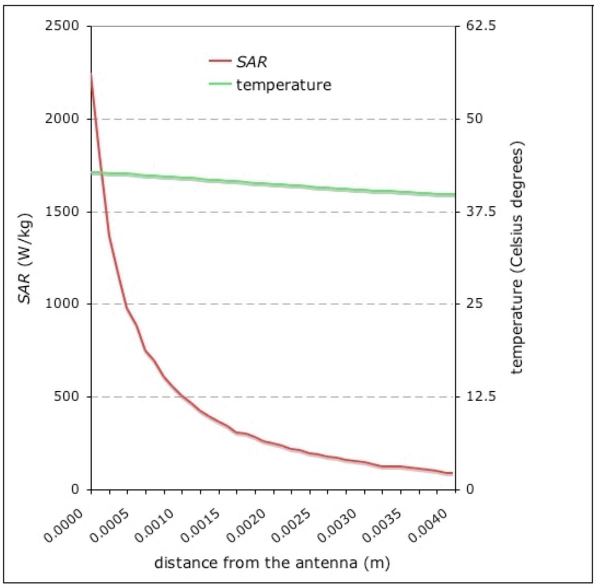


Figure 3. *SAR* and temperature decrease with the distance from the coaxial-slot antenna

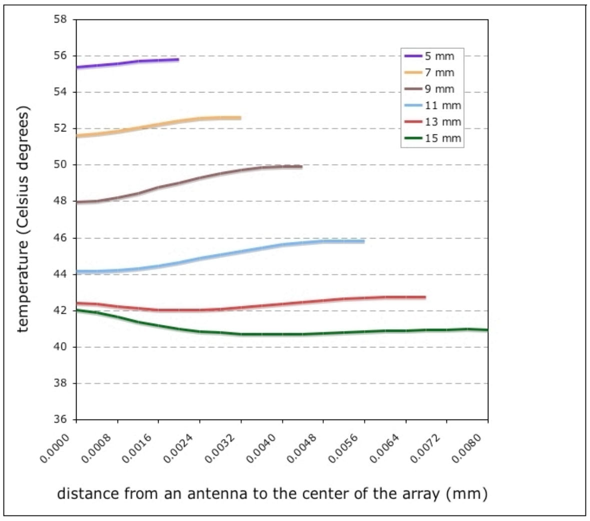


Figure 5. Temperature distributions from one antenna to the center of the array, for several 3-pins array applicators with different distances between pins (5 to 15 mm)

section at the level that corresponds to the upper end of the radiating slots (horizontal plane where the maximal values occur), and the other is longitudinal and includes one of the three pins axes (vertical plane).

The FEM model allows the post processing of the EMF and thermal field solutions. Fig. 5 shows the distribution of the temperature, as main dosimetric parameter, on the axis of maximal values; this axis is perpendicular to one of the antennas, at the upper end of the slot and it is directed to the array's center of symmetry. Several three-pin array applicators, with different distances between pins (5 to 15 mm), have been considered. Each distribution in fig. 5 begins on the antenna surface and ends in the center of symmetry of the corresponding array applicator.

Fig. 6 presents the temperature spectra in the same cross-sectional plane, for several distances between coaxial-slot antennas. Each antenna emits 1W.

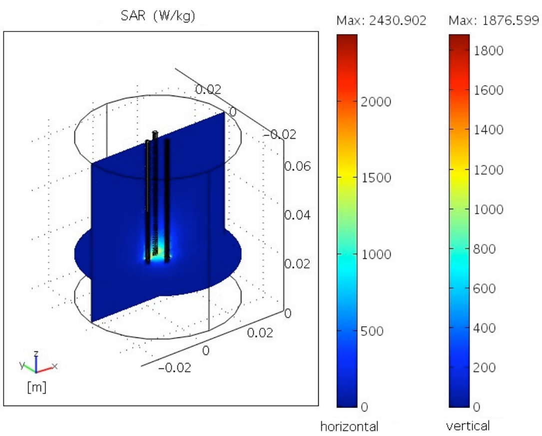


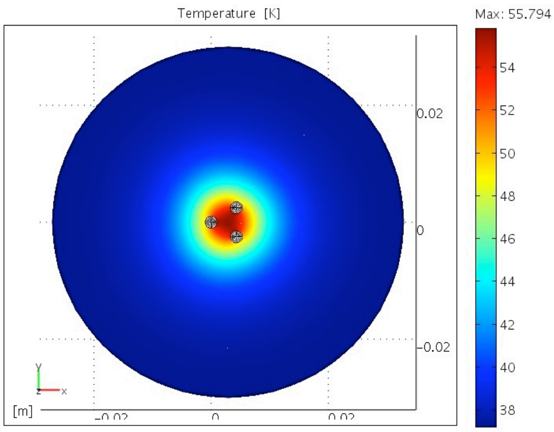
Figure 4. *SAR* distribution for a three pins applicator (7 mm distance between pins, 1W power emitted by each pin)

As one could see in figures 5 and 6, the temperature distribution is more or less no uniform and the temperature spectrum changes with the distance. When the temperature uniformity criterion is considered, the array applicator with the pins spaced at 13 mm appears to be the best one, because the target region is heated almost uniformly, in the range 42 - 42.5 ^{o}C (see the red curve in fig. 5). In all other cases, the dispersion of the temperature values in the target region is larger.

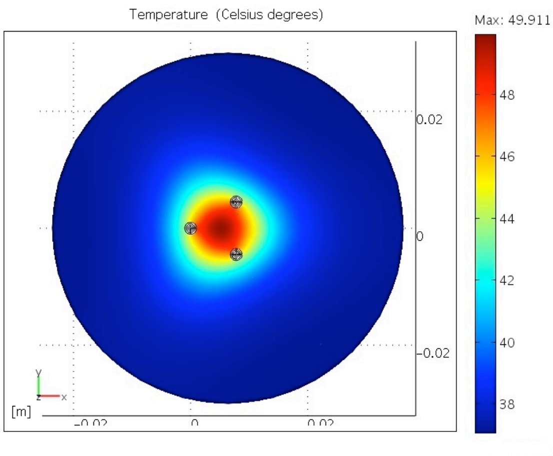
Fig. 7 shows the temperature spectra in the two observation planes (horizontal and vertical) for the array applicator with the antennas spaced at 13 mm. This array structure ensures the almost uniform heating at 42 ^{o}C of a tissue volume, in the shape of a sphere, of 1436 mm^3 (7 mm radius). The side effects are also eliminated, because the maximal temperatures are localized inside the target volume, not at the surface of the pins (as for the array applicator with 15 mm distance between pins, see figures 5 and 6) or in other regions of the tissue. Since the results presented are for the power of 3 x 1W injected in the exposed tissue, the heating level could be controlled through the applied power.

IV. Conclusions

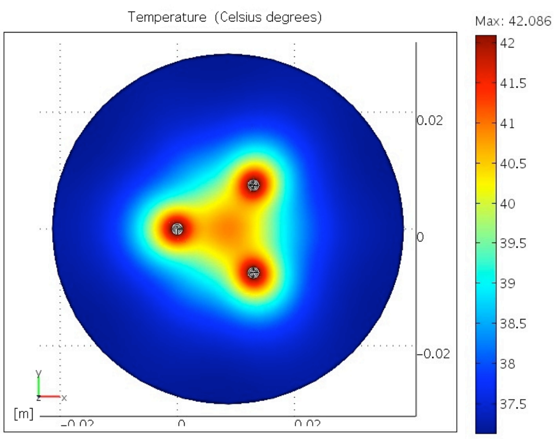
This work presents the analysis of the thermal output produced by some microwave devices, with the objective to approach general criteria for the optimal design of performant minimally invasive hyperthermia microwave applicators used in medical therapy. A technical solution for a three pins array applicator, made of coaxial-slot antennas is analyzed by numerical modeling. The design of a single antenna is first optimized with a 2D FEM model, based on axial symmetry, and verified with the original 3D FEM model. The dimensions of the coaxial cable and the position of the radiating slot were first fixed through several trials, aiming at the most efficient energy transfer from the microwave source to the exposed tissue.



5 mm distance



9 mm distance



15 mm distance

Figure 6. Temperature spectra in the cross-sectional plane with maximal values, for several distances between pins.

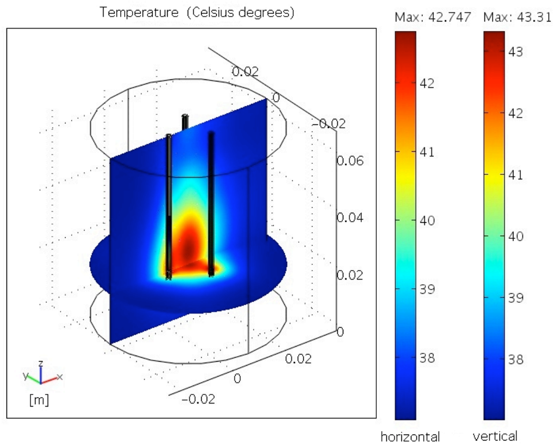


Figure 7. Temperature spectra in two observation planes for the 3-pins array with 13mm distance between pins

The optimal configuration of the coaxial-slot antenna was then used at the construction of an array applicator with three identical pins, equally spaced. The most suited distance between each two adjacent pins was found around the value of 13 mm, after the comparison of thermal spectra produced by several arrangements, with different distances between pins. The optimal solution is characterized by an almost uniform temperature distribution in a target volume of tissue placed at the center or the array and by the absence of overheating outside the target area.

We will further develop medical applications of the optimal array applicator designed here. The research presented in this paper will continue with an analysis of the relation between the applied power and the temperature level and with the investigation of non-homogeneous media heating (for example a tumoral tissue modeled as tumorous sub domains embedded in normal tissue, or the insertion of ferromagnetic grains in the target tissue in order to better concentrate de radiated energy into it).

The advantage of invasive probes is that the heat can be localized with higher precision, in a smaller and more deeply situated volume, than with external applicators. One disadvantage, of course, is that it could be uncomfortable for the patient. Even using multiple probes does not assure uniform heating; there still may be considerable nonuniformity in the power deposition pattern, depending on the placement and individual patterns from probes. Other engineering issues remain too; these include the need for multiple-point temperature measurements for accurate and thorough monitoring, or the enhance of treatment efficiency by combining methods (hyperthermia and radiation therapy, or two hyperthermia techniques: microwaves and ultrasound), using the same array of implanted probes, in order to reduce patient's discomfort. Treatment planning will require accurate characterization of the applicator deposition pattern and the tissue parameters, as well as a numerical technique to predict the resultant heating pattern. Tissue perfusion significantly modifies the temperature distribution for any given power deposition pattern, often in a time-variable and unpredictable way. Still, the promise for even a partially successful therapy for cancer spurs the continued study of hyperthermia.

ACKNOWLEDGMENT

This work was supported through the CNCSIS research grant A - 357/2005

REFERENCES

[1] K. R. Foster, A. Lozano-Nieto, P. J. Riu, "Heating of tissues by microwaves: a model analysis," *Bioelectromagnetics*, vol.19, pp. 420-428, 1998

[2] Urano M., Douple E. (Editors), *Hyperthermia and Oncology, Thermal Effects on Cells and Tissues*, Brill Academic Pub. 1988, ISBN 9067640875

[3] Bini M., Feroldi P., Olmi R., Pasquini A., “Electromagnetic diathermy: a critical review,” Phys Med 1994, X: 39-46.

[4] Hildebrandt B., Wust P., Ahlers O. et al., “The cellular and molecular basis of hyperthermia,” Critical Reviews in Oncology/Hematology, vol 43, pp.33–56, 2002

[5] Kenneth R. Foster, Albert Lozano-Nieto, Pere J. Riu, “Heating of Tissues by Microwaves: A Model Analysis,” Bioelectromagnetics, vol. 19, pp.420-428, 1998

[6] I. Chang. "Finite element analysis of hepatic radiofrequency ablation probes using temperature-dependent electrical conductivity," *BioMedical Engineering OnLine* , 2:12, 2003

[7] K. Ito, K. Saito, T. Taniguchi, H. Yoshimura, "Temperature distribution in and around array applicator for interstitial microwave hyperthermia combined with interstitial radiation therapy," Proc. 27th Intl. URSI Gen. Assembly, Maastricht, 2002 (http://www.ursi.org/Proceedings/ProcGA02/papers/p1433.pdf)

[8] M. Morega, A. Morega, A. Machedon, "Simplified numerical models for microwave dosimetry in human tissue," 3rd European Medical and Biological Engineering Conf. EMBEC'05&IFMBE, Praga, Rep. Ceha, IFMBE Proceedings CD-ROM, ISSN 1727-1983, nov. 2005

[9] H. H. Pennes, "Analysis of tissue and arterial blood temperature in the resting human forearm," *J. Appl. Phys*., vol. 1, pp. 93-122, 1948

[10] C. Gabriel, S. Gabriel, "Dielectric properties of body tissues at RF and microwave frequencies," Report for Armstrong Laboratory (AFMC), Occupational and Environmental Health Directorate - Radiofrequency Radiation Division, USA, 2002G. (http://www.brooks.af.mil/AFRL/HED/hedr/reports/dielectric)

[11] COMSOL Multiphysics™ v. 3.2., Users' Guide, Electromagnetic Module, Heat Transfer Module, 2005.