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ELECTROMAGNETIC AND THERMAL MODELS OF EMISSION OF A COOLED DIRECTIONAL FOR LOCAL ENDOCAVITY HYPERTHERMIA

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Possibilities of creation of controlled temperature fields in deep-seated biological tissue with the use of a microwave endocavity applicator with directed emission and surface cooling are explored. Mathematical models are proposed and calculated that make it possible to construct electromagnetic and thermal fields in biotissue depending on the specific thermophysical and ultrasound characteristics of the medium being irradiated, and to reveal situations and effects which should be achieved to solve problems of practical medicine in the field of local microwave hyperthermia of tissues.

The growing interest in methods of local hyperthermia and thermotherapy of biological tissue is determined by promising clinical results in treatment of a number of pathological states of the organism and stimulates the creation of specialized techniques for local heating of tissues and monitoring of their status. At the same time, a number of basic problems connected with the choice of conditions of heating control are still solved ambiguously and, in fact, are still unclear. This situation, along with the currently used controversial methods of statistical analysis of experiments [1, 2], requires the development of well-founded models of heating processes for particular clinical conditions.

Among all practical applications of hyperthermic treatment, a number of investigators have started to use the natural cavities of the organism for heat delivery, with the additional specific action of one or another physical field, if possible, to deep-seated biological tissue with the aim of its heating to particular temperatures or destruction.

The endocavity heating technique, due to its apparent simplicity, makes it possible to carry out a number of simple experiments, and the possibilities of its clinical applications are rather wide. In particular, endocavity microwave hyperthermia, as applied to treatment of inflammatory processes of the prostate gland, has been in last two or three years a subject of 30–50 papers. Most of these investigations employ similar methods; however, the main conclusions drawn from the results obtained are contradictory. Thus, the problem emerges of theoretical modeling of endocavity microwave heating and revealing the conditions under which it is possible to achieve heating of the greatest local volume of tissue and to avoid its overheating and burning, which is achievable with the use of a cooling liquid (water).

Let us construct the distribution of the electromagnetic field in the vicinity of an asymmetric emitting dipole embedded in a scattering medium: biological tissue (Fig. 1).

We consider the most general case: the dipole is situated asymmetrically in a cylindrical insulator whose diameter is comparable with the radiation wavelength, and fabriacated from a material with a small loss tangent and dielectric constant. A cooling liquid – distilled water – circulates between the inner insulator and the outer casing of the emitter. Thus, the model of the microwave applicator is a four-layer system for which the following condition must be satisfied:

$$\frac{|k_2|^2}{|k_3|^2} << 1 , \tag{1}$$

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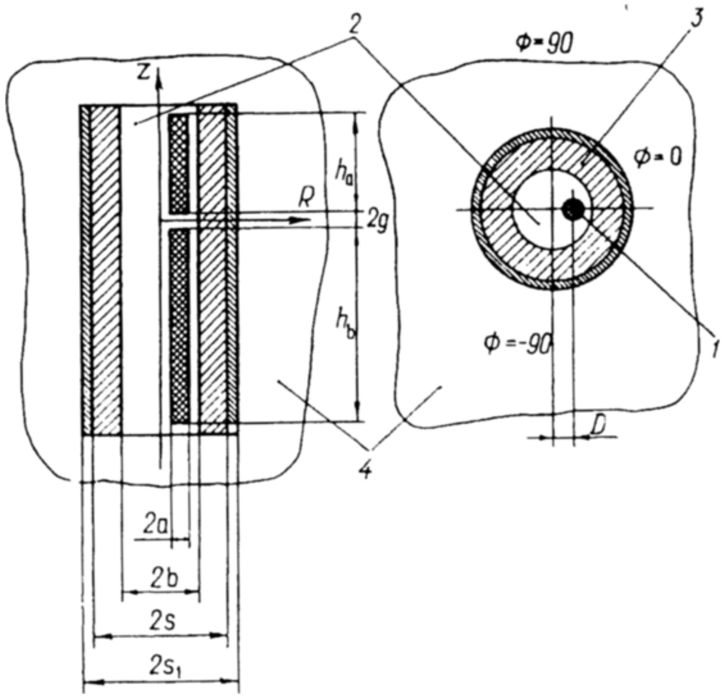


Fig.1. Schematic diagram of asymmetric applicator with cooling liquid: region 1) dipole; 2) insulator; 3) water; 4) biological tissue.

where k_2 is the wave number of the insulator; k_3 is the wave number of the coolling liquid. Here and in what follows, the inner conductor, dielectric, cooling liquid, and the outer medium comprise areas 1, 2, 3, and 4. We denote the outer medium as 4 even when region 3 is missing.

In [3], expressions for the components of the electric field E_{r4}, $E_{\varphi 4}$, and E_{z4} and magnetic field B_{r4}, $B_{\varphi 4}$, and B_{z4} in a cylindrical coordinate system are presented for a dipole placed in an absorbing medium without cooling. They were obtained by solving the Maxwell equations under conditions $|k_4| >> |k_2|$ and $|k_2 d| << 1$, where $d = b - D$. The latter make it possible to assume that the field within the dielectric 2 ($r < b$) satisfies the Maxwell equations for low frequencies; the external field satisfies the Maxwell equations with wave number k_4.

Inasmuch as the condition $|k_3| >> |k_2|$ is satisfied for the cooling water, the following equalities hold: $E_{r4} = E_{r3}$, $E_{\varphi 4} = E_{\varphi 3}$, $E_{z4} = E_{z3}$, $B_{r4} = B_{r3}$, $B_{\varphi 4} = B_{\varphi 3}$, $B_{z4} = B_{z3}$.

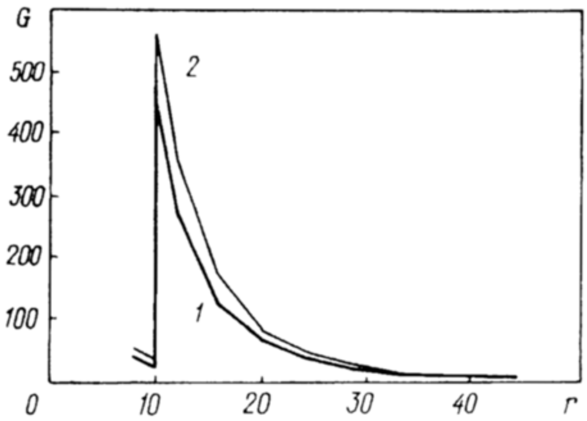
Let us assume that the external casing can be neglected, since its thickness is much smaller than the wavelength ($s_1 = s$). Then, in order to obtain the components of the fields in zone 4, we use the boundary conditions [4] for $r = s = s_1$:

$$E_{r4}(s, \varphi, z) = \frac{k_3^2}{k_4^2} E_{r3}(s, \varphi, z), \quad E_{z4}(s, \varphi, z) = E_{z3}(s, \varphi, z),$$

$$E_{\varphi 4}(s, \varphi, z) = E_{\varphi 3}(s, \varphi, z);$$

$$B_{r4}(s, \varphi, z) = \frac{k_3^2}{k_4^2} B_{r3}(s, \varphi, z), \quad B_{z4}(s, \varphi, z) = B_{z3}(s, \varphi, z), \tag{2}$$

$$B_{\varphi 4}(s, \varphi, z) = B_{\varphi 3}(s, \varphi, z),$$



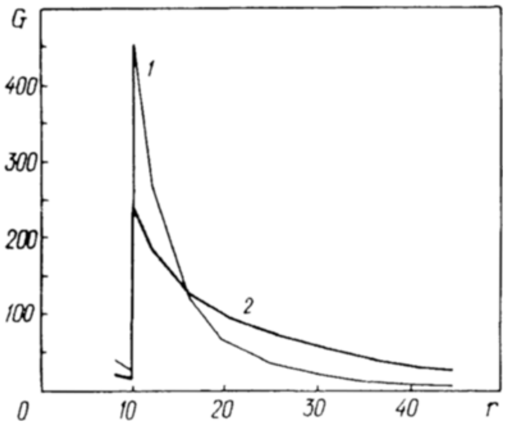


Fig. 2. Distribution of the specific power dissipation G, dB, of the cooled applicator with allowance for refraction of the field at the water-medium interface (1) and with refraction not taken into account (2). Frequency F = 2.45 GHz, power P = 10 W, $z = 0$, $\varphi = 0$. r, mm.

Fig. 3. Distribution of the specific power dissipation G, dB, of cooled applicator at frequencies F = 2.45 GHz (1) and 915 MHz (2). Power P = 10 W, $z = 0$, $\varphi = 0$.

TABLE 1. Dimensions of Applicator

Regions	Radius, mm	Material	Electric parameters
Emitter	$a = 0.5$	Ideal conductor	$\varepsilon = 0, \sigma = \infty$
Insulator	$b = 0.8$	Teflon	$\varepsilon = 2.1, \sigma = 0$
Cooling liquid	$s = 10.0$	Distilled water	$\varepsilon = 80, \sigma = 0.0864$, cm/m
Outer casing	$s_1 = 10.5$	Polyethylene	–

where E_{r3}, E_{z3}, $E_{\varphi 3}$ and B_{r3}, B_{z3}, $B_{\varphi 3}$ are components of the electric field strength and magnetic induction in water, respectively; E_{r4}, E_{z4}, $E_{\varphi 4}$ and B_{r4}, B_{z4}, $B_{\varphi 3}$ are components of the electric field strength and magnetic induction in the medium, respectively.

The electric field at an arbitrary point is defined as follows [5]:

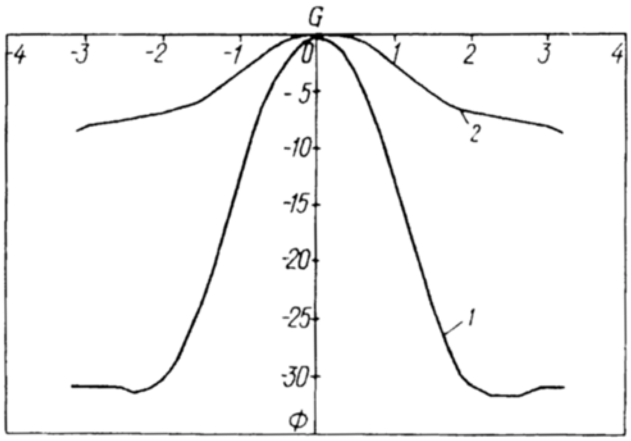
$$\mathbf{E}_4 = \frac{1}{4\pi} \int\limits_{\Sigma'} \left\{ \begin{array}{l} i\omega\, [\bar{n} \times \mathbf{B}_4 (s, \varphi', z')]\, \Psi + [\bar{n} \times \mathbf{E}_4 (s, \varphi', z')] \times \\ \times \operatorname{grad}' \Psi + [\bar{n} \times \mathbf{E}_4 (s, \varphi', z')] \operatorname{grad}' \Psi \end{array} \right\} d\Sigma', \tag{3}$$

where $i = \sqrt{-1}$; W is the frequency; r, φ, z are coordinates of the point in the medium at which the field is calculated; r', φ', z' is a point on the surface of the cylindrical insulator with radius s; n is the normal to the cylinder surface;

$$\Psi = \frac{\exp(-ik_4 R)}{R}; \quad R = \sqrt{r^2 + s^2 + (z - z')^2 - 2rs\cos(\varphi - \varphi')}.$$

If we neglect the field at the ends of the cylinder, we obtain an expression for the electric-field components in the biological tissue:

$$\mathbf{E}_4 (r, \varphi, z) \cong \frac{1}{4\pi} \int\limits_{z' = -h_b}^{h_a} \int\limits_{\varphi' = -\pi}^{\pi} \Big\{ i\omega\, [\bar{n} \times \mathbf{B}_4 (s, \varphi', z')]\, \Psi +$$



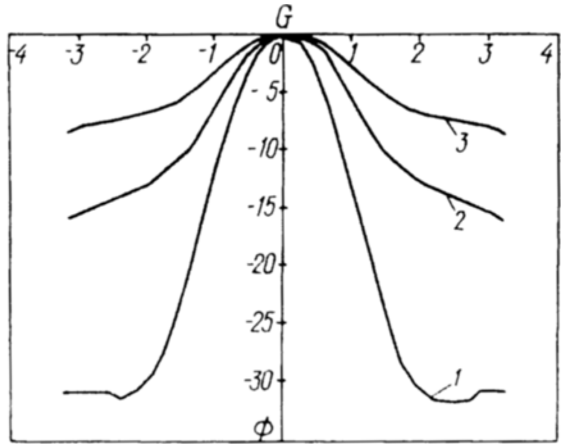


Fig. 4. Distribition of the specific power dissipation G, dB, in biological tissue at frequencies of 2.45 GHz (1) and 915 MHz (2); $z = 0$, $r = 12$ mm, power $P = 10$ W, eccentricity $D = 0.7b$. φ, rad.

Fig. 5. Distribution of the specific power dissipation G, dB, in biological tissue at various eccentricities $D = 0.7b$ (1), $0.4b$ (2), and $0.2b$ (3); frequency 2.45 GHz, power $P = 10$ W, $z = 0$, $r = 12$ mm.

$$+ [\bar{n} \times \mathrm{E}_4 (s, \varphi', z')] \,\mathrm{grad}'\, \Psi + [\bar{n} \times \mathrm{E}_4 (s, \varphi', z')] \,\mathrm{grad}'\, \Psi \Big\} \, s d\varphi' dz' . \tag{4}$$

The specific power dissipated in the tissue:

$$G (r, \varphi, z) = \sigma_4 \, |\mathrm{E}^2| , \tag{5}$$

where $|\mathrm{E}^2| = E_{r4}^2 + E_{\varphi 4}^2 + E_{z4}^2$; σ_4 is the conductivity of the biological tissue. Numerical methods are used for calculation of the power dissipation.

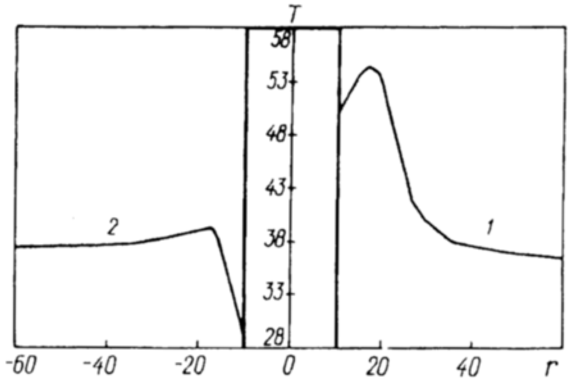
On the basis of the expressions obatined we model the operation of an applicator at a frequency of 2.45 GHz with the parameters of the muscle tissue (dielectric constant $\varepsilon_4 = 53$, specific conductivity $\sigma_4 = 1.43$ cm/m). The dimensions of the applicator are presented in Table 1.

The distribution of the specific power dissipation G is shown in Fig. 2 as a function of the coordinate r. Curve 1 presents the distribution of G calculated without taking into account refraction of the fields at the water-medium interface. It is evident that the difference between the maxima is about 25%. The discontinuitv of the distribution curves at the water-medium interface is explained by the fact that the conductivity of water is two orders of magnitude smaller than that of the biological tissue.

Figure 3 presents the distribution of the specific power dissipation G in the biological tissue as a function of the coordinate r for two different radiation frequencies

Calculations show that attenuation at 2.45 GHz is higher in biological tissue than attenuation of an electromagnetic field (EMF) with a frequency of 915 MHz. However, the maximum value of G is higher at 915 MHz. From the practical viewpoint of heating of most deep-seated tissue, it is worthwhile, as follows from the model built, to choose a radiation frequency of 915 MHz. However, at this frequency the specific power dissipation G depends weakly on the angle φ (Fig. 4). Thus, an EMF with a frequency of 2.45 GHz is more directional than that with a frequency of 915 MHz for equal radiation powers.

The dependence of the specific power dissipation G on the eccentricity is shown in Fig. 5. It is evident that with increasing D the radiation becomes more directional; the optimum value is $D = 0.7b$. This agrees qualitatively with results of investigations for an insulated dipole [5].



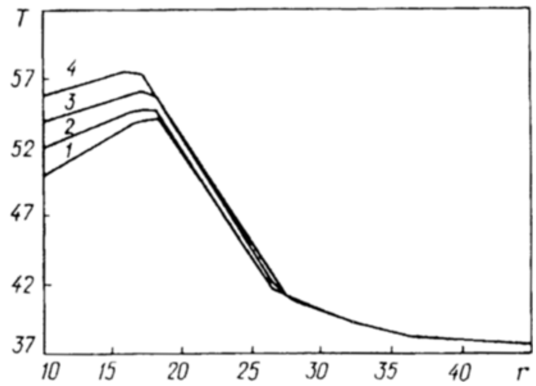


Fig. 6. Distribution of temperature T_w, °C, in biological tissue in the plane $z = 0$ with cooling. Radiation frequency 2.45 GHz, power 10 W, water temperature $T = 5$°C, blood flow $m_b = 0.45 \cdot 10^{-5}$ m^3/kg·sec, $\varphi = 0$ (1), 180° (2).

Fig. 7. Distribution of the temperature T, °C in the biological tissue in the plane $z = 0$ for various temperatures of the cooling water: $T_w = 5$°C (1), 10 (2), 15 (3), and 20 (4). Radiation frequency 2.45 GHz, power 10 W, blood flow $m_b = 0.45 \cdot 10^{-5}$ m^3/kg·sec.

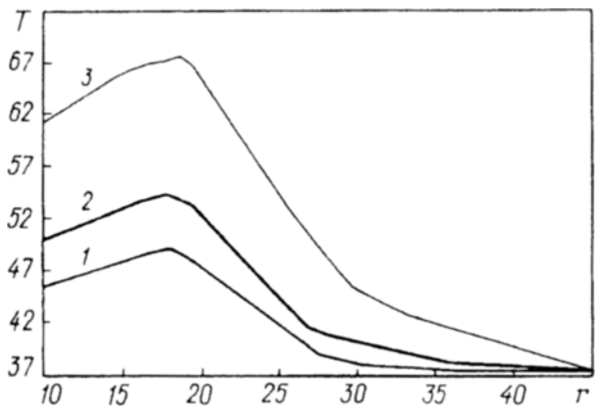


Fig. 8. Distribution of the temperature T, °C, in the biological tissue in the plane $z = 0$ for blood flow: $m_b = 9.00 \cdot 10^{-6}$ m^3/kg·sec (1), $0.45 \cdot 10^{-6}$ (2), and $0.45 \cdot 10^{-5}$ (3). Radiation frequency 2.45 GHz, power 10 W.

To obtain the temperature distribution in biological tissue with an operating heating source we use the classical two-dimensional biothermal equation [6] that takes into account metabolic processes and blood flow. In cylindrical coordinates the equation can be written as follows:

$$\frac{\partial^2 T}{\partial r^2} + \frac{1}{r}\frac{\partial T}{\partial r} + \frac{\partial^2 T}{\partial z^2} + \frac{1}{k_t} G(r, z) + \frac{1}{k_t} W_m - C(T - T_b) = \frac{1}{\alpha}\frac{\partial T}{\partial t}, \qquad (6)$$

where $\alpha = k_t/\rho_t c_t$; $C = c_b \rho_b m_b \rho_t / k_t$; k_t is the thermal conductivity of the tissue; ρ_t and ρ_b are the specific densities of the tissue and blood, respectively; c_t and c_b are the heat capacities of the tissue and blood, respectively; m_b is the value of the blood flow per unit tissue weight; $G(r, z)$ is the electromagnetic power dissipation; W_m is the power of metabolic processes; T_b is arterial temperature of blood. To solve this equation numerically under boundary conditions of the first and the third kinds, we used the finite-element method.

The temperature distribution in the biological tissue is shown in Figs. 6-8 for the steady-state regime. Calculations of the effect of the radiation pattern on the temperature distribution have shown that in the plane $\varphi = 0$ the tissue temperature is 20% higher than in the plane $\varphi = 180^0$ (Fig. 6).

The dependence of the temperature of the tissue being heated on the temperautre of the cooling water is illustrated by Fig. 7. Its is evident from the figure that the position of the heating maximum varies with the water temperature. A water temperature of $5-10^0$C can be considered optimal. In this case it is important to note that deeper layers of tissue are heated as the temperature of the cooling liquid decreases.

The heating maximum is more sensitive to changes in the blood flow value than to the temperature of the cooling liquid. Thus, a decrease in blood flow by one order leads to an increase in tissue temperature of more than 10^0C (Fig. 8).

Theoretical investigations of an applicator with directional radiation and a cooling system make it possible to draw the following conclusions:

1. To determine the temperature of tissue heating by the applicator, one should carry out an analysis of combined theoretical electromagnetic and thermal models. This applicator makes it possible to achieve the necessary tissue heating in the desired direction.

2. The efficiency of the cooling system can be optimized by the choice of the temperature and velocity of the cooling liquid.

3. Small changes in blood flow during hyperthermia can lead to substantial changes in the temperature distribution at a constant input power of the applicator. Variations in the temperature in the tissue and the use of electronic power control make it possible to avoid overheating and burning of the tissue in hyperthermia.

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