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IMPROVEMENT OF THE MODEL OF TEMPERATURE DISTRIBUTION AND REGISTRATION OF NATIVE RADIATION OF BIOLOGICAL OBJECTS

V. Katrych
Doctor of physical and mathematical sciences, Professor*
E-mail: vkatrich@karazin.ua
M. Mustetsov
PhD, Professor*
E-mail: mustetsovn@ukr.net
V. Kozheshkurt
Postgraduate student*
E-mail: vkozeskurt@gmail.com
*Department of Physical and Biomedical Electronics and Complex Information Technologies
V. N. Karazin Kharkiv National University
Svobody sq., 4, Kharkiv, Ukraine, 61022

Більшість захворювань в організмі людини призводить до порушення розподілу теплових полів. Неінвазивний контроль глибинних температур може підвищити ефективність діагностики. Розглянуто модель розподілу температури при наявності області з пониженою температурою в глибинних шарах однорідної біологічної тканини. Показано можливість визначення характеристик температурної аномалії за допомогою багаточастотної мікрохвильової радіотермометрії

Ключові слова: аномалія температурна, випромінювання теплове, поле електромагнітне, радіотермометрія, термографія багаточастотна

Большинство заболеваний в организме человека приводит к нарушению распределения тепловых полей. Неинвазивный контроль глубинных температур позволяет повысить эффективность диагностики. Рассмотрена модель распределения температуры при наличии области с пониженной температурой в глубинных слоях однородной биологической ткани. Показана возможность определения характеристик температурной аномалии с помощью многочастотной микроволновой радиотермометрии

Ключевые слова: аномалия температурная, излучение тепловое, поле электромагнитное, радиотермометрия, термография многочастотная

1. Introduction

Maintaining a constant body temperature – homeothermy – is one of the necessary and complex functions of the body, which possesses complex mechanism and involves the interaction of different body systems [1]. Temperature, as a diagnostic indicator, is used in almost any medical procedure. A rise in temperature of body surface is a signal of the development or presence of pathological processes in the body. Deep temperature is an integral indicator of the level of bioenergetic processes and it can serve as an indicator of the morphofunctional state of separate organs. The reasons of increasing of the local temperature may be [2]: inflammation of any genesis, malignant neoplasms [3]; disorders of venous outflow and venous congestion; increase in the metabolic rate of various organs. Of particular importance is the measurement of deep temperatures for the diagnosis of cancer.

In some cases, a local decrease in the deep temperatures is observed. The reasons of such decrease [2] can be:

– disturbances of arterial blood supply and a decrease in blood microcirculation;

– decrease in the metabolism rate of various organs of an age or pathological nature;

– degenerative processes with the replacement of functionally active tissue with a connective tissue.

Existing invasive methods for measuring deep temperatures are based on the introduction of various types of sensors, which traumatizes tissues and does not allow determining anomalous regions with a sufficient degree of accuracy. Non-invasive measurement of deep temperatures is possible with the use of microwave radiometers [4] – devices that register their own radiation of biological tissues in the radio range. By the power level of this radiation, it is possible to estimate the deep temperatures of organs and tissues.

For practical implementation of radiothermometry in medical practice, the following conditions are necessary:

– a presence of models of distribution of deep temperatures in human tissues;

– theoretical justification of the requirements to the radiothermometer equipment;

– development of measurement methodology.

2. Literature review and problem statement

Radiothermometry is a promising method of non-invasive diagnosis of diseases accompanied by a disturbance of the temperature balance. The most developed methods of diagnosis are the methods of diagnosis of malignant mammary neoplasms. At the moment, developments are carried out to improve the method. In paper [5], additional factors that had not been taken into account in the initial measurement model and that affect the detection accuracy of the tumor have been identified and investigated.

The development of antennas for the purposes of mammary radiothermometry continues. One of such developments has been presented in paper [6], the authors have developed a wide-band thin-film antenna coordinated with the multilayered structure of biological tissue. In article [7], authors have presented an antenna at the open end of the waveguide that meets the frequency requirements and can be used in microwave diagnostics of breast cancer.

The results of clinical studies of a method combining infrared thermometry of the skin surface and internal temperature radiothermometry have been presented in paper [8]. The method has shown reliability of the diagnosis, comparable to the mammographic method, which is considered as the reference method. In article [9] results of the study of the physiological changes of the mammary gland during pregnancy by the methods of infrared and microwave thermography have been presented.

One of the developing areas of research is the combined use of radiothermometry with therapeutic actions. Experimental studies of the use of radiothermometry in combination with microwave hyperthermia have been given in the paper [10], where a test of a combined apparatus for radiothermometry and hyperthermia on a model of a head, electrophysical properties of which are correlated with the real ones, has been described. Report [11] has described the application of radiothermometry in combination with laser action.

Existing technical means of non-invasive temperature measurement make it possible to identify areas of elevated temperature without an exact determination of the depth of the source of the anomalous temperature. At the same time, the authors have set the task of measuring zones of elevated temperature, while in the human body zones of local lower temperature can also be considerably dangerous. To create technical means to determine the depth of anomalous temperature, it is necessary to consider the model of temperature fields on the basis of which it is possible to estimate the existing technical means of radiothermometry.

3. The study objective and tasks

The objective of present work is to model the distribution of thermal fields in the human body in the presence of lower temperature zones to justify the requirements to the parameters of radiothermometers. The results of the modeling enable the technical implementation of the radiothermometry method to solve medical tasks.

To achieve the objective, the following tasks were set:

– to improve the model of temperature distribution in biological tissue and analyze the distribution obtained;

– to evaluate the possibility of determination of the characteristics of the temperature anomaly by the method of two-frequency radiothermometry;

– to evaluate the accuracy of the method and justify the required technical parameters of the radiometer.

4. Model of distribution of deep temperatures

In the existing models of the distribution of temperature fields in biological objects, a source of increased heat release (local inflammation, cancer neoplasm [12]) is assumed. The temperature fields in the region of lower temperature have not been studied sufficiently. This is possible if there is a region with disturbed blood circulation and slow metabolic processes, not accompanied by inflammation, in the deep layers of tissues.

To construct the temperature distribution model in the presence of a source with a lower temperature, we make the following assumptions [13]:

– affected area is a homogeneous sphere of radius r_0, located at a distance $a>r_0$ from the surface of the skin;

– heat release in the sphere is negligibly small in comparison with the heat release of surrounding healthy tissues;

– thermal conductivity inside the sphere is less than the thermal conductivity of the surrounding tissues due to a decrease in blood flow in the affected area.

An exact solution to the problem of heat transport and distribution in the area can be obtained by using the methods of mathematical physics. In this case, the solution of the thermal conductivity equation reduces to solution of the Cauchy problem and allows us to determine the heat distribution function in the area with a high degree of accuracy. In the case of biological tissue, this is extremely difficult for several reasons. First, the area is not homogeneous, in terms of thermal properties. Second, the source function cannot be uniquely determined. In addition, the process of formation of a source of elevated temperature (cancer neoplasm) is of an individual character.

For approximate estimates of the thermal fields there is no need in accurate calculations. Therefore, we consider the problem using the example of electrostatic field distribution, using the analogy between electrostatic (potential) and thermal fields. When modeling thermal processes by electrostatic ones, the dielectric permeability ε corresponds to the coefficient of thermal conductivity κ, and the charge density ρ – to the specific heat release $\rho_т$. The electrostatic model is a dielectric sphere of radius r_0 and a dielectric constant ε_1 in area with a dielectric constant ε_2.

The specific heat release in the sphere, as indicated above, is negligible comparing to the specific heat release of the surrounding tissue. Accordingly, within the framework of the electrostatic model, the region under consideration will be the region with zero charge in area with a positive charge density ρ. By the principle of superposition of the electric field, the electric field potential N of the sources is equal to the sum of the field potentials of each source:

$$\varphi = \sum_{i}^{N} \varphi_i. \quad (1)$$

Conditionally, we place a fictitious negative charge with density ρ in each point of biological tissue. Using the principle of superposition of the electric field and calculating the total charge at each point, we obtain a model. This model has been brought to the following form: a charged dielectric sphere with a distributed negative charge with a density ρ in an uncharged dielectric area. Since heat release of intact tissue is taken as zero level, the solution of this problem will be not the absolute value of temperature, but the temperature change relative to the normal temperature of healthy tissue ΔT, caused by the anomaly.

The total charge of the sphere is:

$$q = \rho V = \frac{4}{3}\pi\rho r_0^3. \quad (2)$$

By solving the problem of electrostatics for a charged sphere in a homogeneous space, we obtain the potential ϕ at a point at a distance r from the center of the sphere:

$$\varphi(r)=\frac{\rho r_0^2}{3\varepsilon_2}+\frac{\rho}{6\varepsilon_1}\cdot\left(r_0^2-r^2\right)\ \text{ for }\ 0\le r\le r_0, \tag{3}$$

$$\varphi(r)=\frac{\rho r_0^3}{2\varepsilon_2 r}\ \text{ for }\ r\ge r_0. \tag{4}$$

In the case when the temperature anomaly is located near the surface of the body at a distance a comparable to the radius of the anomaly, the temperature distribution is affected by the thermophysical parameters of the skin. The temperature anomaly is located inside a biological tissue with a high thermal conductivity near the plane interface with air, which has a low thermal conductivity. In the electrostatic model, we set the problem of the distribution of the charge field located near the plane interface of two areas with a dielectric permeability ε_2 and $\varepsilon_3<\varepsilon_2$ respectively. The charge is uniformly distributed inside the dielectric sphere with dielectric permeability ε_1 and radius r_0.

The potential at a random point P of the space depends on the distance between the point P and the center of the charged sphere and the distance from the point P to the plane interface of the areas. The effect of the plane interface is taken into account using the image method [14]. We place the charge $Q'=Q$ symmetrically to the center of the charge with respect to the plane interface, that is, to the point $(-a;0;0)$ (Fig. 1).

The solution of the problem is the distribution of potentials only in the half-space $x>0$. The potential at a random point of this half-space is equal to the sum of the potentials of the charged sphere ϕ_1 and the fictitious symmetric charge φ_1':

$$\varphi=\varphi_1+\varphi_1'. \tag{5}$$

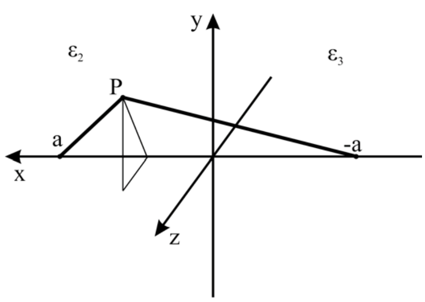

Fig. 1. Electric field at point P created by two charges located at (a; 0; 0) and ($-a$; 0; 0) (x, y, z – are the spatial values)

The potential ϕ_1 is calculated from formulas (3), (4). The distance from the center of the sphere r is:

$$r=\sqrt{(x-a)^2+y^2+z^2}. \tag{6}$$

The potential φ_1' is:

$$\varphi_1'=\frac{\lambda Q}{\varepsilon_2 r'}, \tag{7}$$

where

$$\lambda=\frac{\varepsilon_2-\varepsilon_3}{\varepsilon_2+\varepsilon_3},$$

proceeding from the conditions for the continuity of the electric field at the plane interface [14]. The distance from the center of the fictitious charge r' is:

$$r'=\sqrt{(x+a)^2+y^2+z^2}. \tag{8}$$

We substitute (3), (4), (7) in (5) and obtain the distribution of potentials:

$$\varphi(x,y,z)=\begin{cases}\dfrac{\rho}{6\varepsilon_1}\cdot\left(r_0^2-r^2\right)+\dfrac{r_0^2\rho}{3\varepsilon_2}\cdot\left(1+\dfrac{\varepsilon_2-\varepsilon_3}{\varepsilon_2+\varepsilon_3}\cdot\dfrac{r_0}{r'}\right) & \text{for } 0\le r\le r_0,\\[2ex] \dfrac{r_0^3\rho}{2\varepsilon_2}\cdot\left(\dfrac{1}{r}+\dfrac{\varepsilon_2-\varepsilon_3}{\varepsilon_2+\varepsilon_3}\cdot\dfrac{1}{r'}\right) & \text{for } r\ge r_0.\end{cases} \tag{9}$$

Switching from electrostatic to thermophysical parameters, we obtain the following temperature distribution:

$$T(x,y,z)=\begin{cases}\dfrac{\rho_T}{6\kappa_1}\cdot\left(r_0^2-r^2\right)+\dfrac{r_0^2\rho_T}{3\kappa_2}\cdot\left(1+\dfrac{\kappa_2-\kappa_3}{\kappa_2+\kappa_3}\cdot\dfrac{r_0}{r'}\right) & \text{for } 0\le r\le r_0,\\[2ex] \dfrac{r_0^3\rho_T}{2\kappa_2}\cdot\left(\dfrac{1}{r}+\dfrac{\kappa_2-\kappa_3}{\kappa_2+\kappa_3}\cdot\dfrac{1}{r'}\right) & \text{for } r\ge r_0.\end{cases} \tag{10}$$

The distribution obtained is shown in Fig. 2. The results obtained correspond to the results of the study of temperature fields using IR thermal imagers for the case of a two-dimensional model [15, 16].

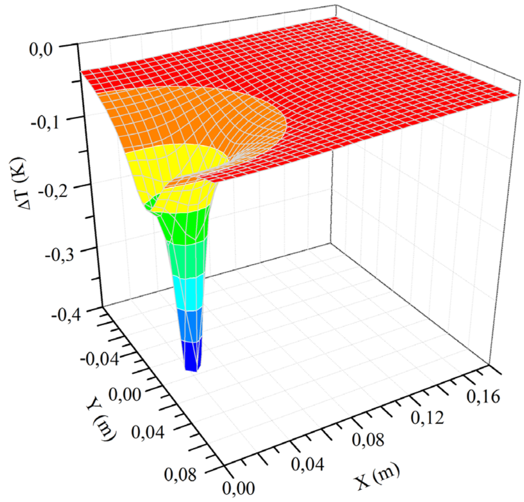

Fig. 2. Distribution of the temperature deviation ΔT inside the biological tissue at a=2.5 cm (x is the axis directed perpendicular to the skin, y is the axis directed along the skin)

As can be seen, the biological plane interface contributes to the formation of the temperature field, and a thermal spot appears on the surface of the skin. At small depths of the location of the temperature anomaly, the temperature spot is clearly discernable (Fig. 3) and can be investigated by infrared thermometry. Modeling of the temperature spot on the skin surface with different sizes and depths of the temperature anomaly allows us to determine the boundary depth of the location of the temperature anomaly.

This condition is met when the temperature anomaly is located at a depth of 2–3 cm, depending on the parameters of the biological area and the parameters of the temperature anomaly, which determines the choice of operating frequencies of the radiometer.

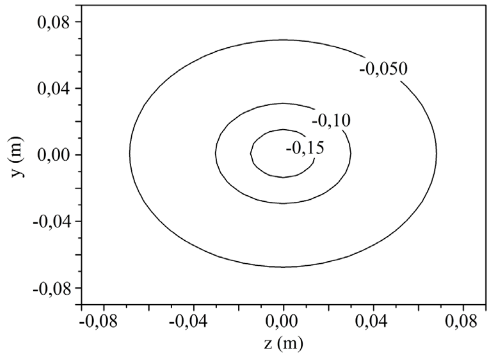

Fig. 3. Temperature spot on the surface of the skin (y, z are the spatial values on the skin surface) for a=2 cm

If a depth of occurrence is greater (Fig. 4, 5), the temperature spot becomes less discernible with the IR thermal imager [16]. Therefore, with an increase in the depth of the source of the temperature anomaly, the main method of investigation is microwave radio thermometry.

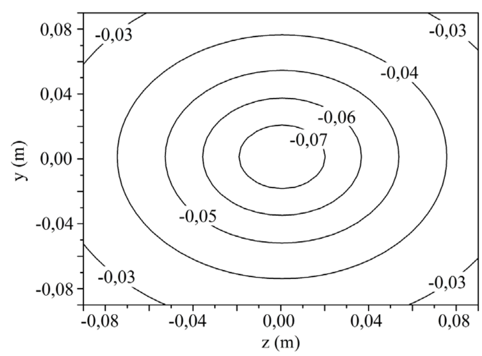

Fig. 4. Temperature spot on the surface of the skin (y, z are the spatial values on the surface of the skin) for a=5 cm

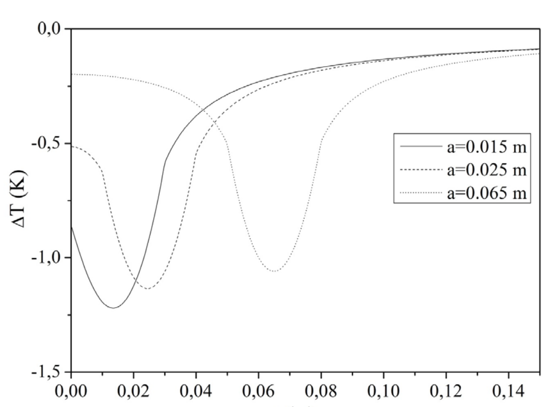

Fig. 5. Distribution of temperature deviation ΔT by depth x in a plane perpendicular to the surface of the skin and passing through the center of the temperature anomaly at different depths of the anomaly

It can be seen how the effect of the temperature anomaly on the temperature of the skin surface (x=0) varies with the depth change on Fig. 5. Expectedly maximum influence on the surface temperature is provided by not to deep disposed regions of anomalous temperature.

5. Determination of the depth and size of the temperature anomaly

The power of electromagnetic radiation of any point of matter having a nonzero temperature located at depth x is proportional to temperature. Spreading inside a layer of tissue of x thickness, the radiation decays exponentially [17] and the radiation power at the surface is equal to:

$$P(x) = P_0 e^{-\frac{x}{l}}, \quad (11)$$

where l is the thickness of the tissue layer, in which the radiation power decreases by e times.

The power allocated in an element of dx length is equal to:

$$dP = -\frac{1}{l} \cdot P_0 e^{-\frac{x}{l}} \mathrm{d}x. \quad (12)$$

The radiation power conditioned by the temperature of the biological tissue is directly proportional to temperature. Therefore, the radiation power is uniquely determined by the noise temperature T_n. Integrating (12), we obtain an expression for the general case of the temperature distribution inside the object:

$$T_{\mathrm{n}} = \frac{1}{l} \cdot \int_0^{\infty} T(x) e^{-\frac{x}{l}} \mathrm{d}x. \quad (13)$$

Existing solutions of the problem of determination of the deep temperatures [12, 17–19] use different models of temperature distribution $T(x)$ corresponding to the real distribution with the required degree of accuracy.

It has been shown in paper [12] that the use of a rectangular temperature distribution model makes it possible to connect the presence of an elevated temperature region with the results of radiometric measurements. In [19], a model of Gauss distribution of temperature has been applied, which makes possible to localize the temperature anomaly more accurately.

The temperature T_n measured by the radiometer is a function of three variables: the depth of the temperature anomaly, its temperature and dimensions (in the case of a spherical model – the radius). The radius of the temperature anomaly can be brought into dependence on the peak value of its temperature, in this case T_n is a function of two independent variables. Thus, it is impossible to localize the temperature anomaly and determine its temperature accurately with the help of one equation and without the assumption of a fixed depth of occurrence, and this is a disadvantage of the method. This problem restricts the practical application of radiothermometry in diagnosis. Since only a qualitative determination of the presence of an anomaly without accurate quantitative characteristics is possible, the method is most widely used for the diagnosis of breast cancer [3, 6–8]. The mammary gland in terms of electrophysical properties is a homogeneous area. Being a pair symmetrical organ, it enables the qualitative detection of a temperature anomaly by differences in the parameters obtained for the two glands.

A possible method of determination of two variables (the size of the temperature anomaly and the depth of its occurrence) is to perform measurements at different frequencies. The measurement at each frequency adds an additional equation to the system of equations, which allows us to calculate the required number of independent variables [19]. Let us investigate the possibilities of multifrequency radiothermometry for the considered model of temperature distribution in the biological area $T(x, y, z)$ with temperature anomalies (10).

We calculate the noise temperature on the surface of the skin according to formula (13) at a point directly above the center of the temperature anomaly, that is, we use the formula for $T(x, y, z)$ with $y=z=0$. In the formula, we assume that the depth of occurrence of the temperature anomaly a and its radius r_0 are unknown variables.

The depth of penetration of radiation in biotissue l is approximately equal to the wavelength of radiation in the tissue, that is, it depends on the radiation frequency and the dielectric constant of the area, which in general also depends on the frequency [20]:

$$l(f)=\frac{c}{f\cdot\sqrt{\varepsilon(f)}}. \tag{14}$$

We obtain an expression for the noise temperature on the surface of the skin, which depends on the depth and the radius of the temperature anomaly, as well as on the frequency at which the measurement has been carried out:

$$T_{\mathrm{n}}(a,r_0,f)=\frac{1}{l(f)}\cdot\int_0^\infty T(x,a,r_0)e^{-\frac{x}{l(f)}}\mathrm{d}x. \tag{15}$$

Thus, in order to find two unknowns a and r_0, a system of two equations is necessary [21]. To obtain a system of equations, it is necessary to measure the noise temperature on the surface of the skin at two frequencies:

$$\begin{cases} T_{\mathrm{n}}(a,r_0,f_1)=T_1, \\ T_{\mathrm{n}}(a,r_0,f_2)=T_2. \end{cases} \tag{16}$$

In the Mathcad software, area calculations have been made for two different frequencies. The choice of frequency is justified by the dependence of the penetration depth of electromagnetic radiation to biological tissue. Since the temperature anomaly located at a depth of less than 2 cm can be studied by infrared thermography, the following condition must be fulfilled for a radiothermometer:

$$\frac{c}{f\cdot\sqrt{\varepsilon(f)}}\geq 2\ \text{cm}. \tag{17}$$

Having solved inequality (17) for adipose tissue, we find that the frequency of the radiometer should satisfy the condition $f\leq 2.8$ GHz. For muscle tissue, with increased water content and high dielectric permeability, – $f\leq 1.8$ GHz. It is advisable to choose a frequency even less, without going over to boundary cases.

The results of calculations at a noise temperature of 0.18 K and a frequency of f_1=1 GHz and at a noise temperature of 0.175 K and a frequency of f_2=0.8 GHz are shown in Fig. 6.

Given

$$\mathrm{T_j}(\mathrm{a},\,10^9,\,\mathrm{ro})=-0.18$$

$$\mathrm{T_j}(\mathrm{a},\,0.8\times10^9,\,\mathrm{ro})=-0.175$$

$$\mathrm{Find}(\mathrm{a},\,\mathrm{ro})=\begin{pmatrix}0.062\\0.012\end{pmatrix}$$

Fig. 6. Results of calculations. The calculated depth is 6.2 cm, the anomaly radius is 1.2 cm

Technical implementation of the method of two-frequency radiothermometry is a separate task [22–25]. The problem is the development of a broadband antenna coordinated with a biological object at different frequencies, or the synchronization and matching of several antennas used for simultaneous measurements at different frequencies.

6. Estimation of the resolving power of the method

The resolving power ΔT of determination of the temperature of the anomaly at a fixed accuracy of the radiometer depends on the depth of the anomaly: with increasing of the depth – ΔT increases and the accuracy decreases. Thus, important information on the estimation of the accuracy of the method is contained in the dependence of the resolvability of detection of the temperature anomaly, on the depth at which the anomaly is located. Since the value measured directly by the radiometer is the noise temperature T_n, we fix the accuracy of its measurement at the level ΔT_n=0.01 °C, which corresponds to the order of accuracy of the existing radiometers.

Let us write the equation connecting the change of the depth of occurrence of the temperature anomaly Δx with the change of the noise temperature:

$$T_{\mathrm{n}}(x,r_0,f)-T_{\mathrm{n}}(x+\Delta x,r_0,f)=\Delta T_{\mathrm{n}}. \tag{18}$$

Assuming the frequency at which the measurements are taken to be f=1 GHz and the radius of the temperature anomaly to be 15 mm, we obtain the dependence of the measurement error of the spatial value Δx on the depth of the temperature anomaly x (Fig. 7).

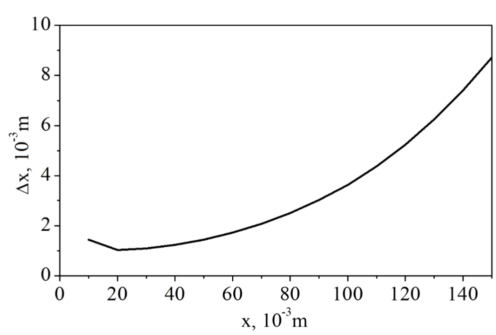

Fig. 7. Dependence of the measurement error by depth Δx on the absolute value of depth x

As a result of the simulation carried out, the possibility of using the laws of electrostatics has been demonstrated, which greatly simplifies the modeling process and allows the simulation of thermal fields. Unlike the existing models [12, 17–19], the case of the presence of a zone of lower temperature in the human body has been considered.

The resolving power of the method of radiothermometry has been estimated and the limiting depths of detection of zones of anomalous temperature have been evaluated. As a result of numerical modeling, it has been shown that the resolving power of the method has made up to 5 % of the depth at different depths of occurrence of the temperature anomaly.

The principal possibility of the method of radiothermometry for determination of the depth of occurrence of an anomalous temperature source has been shown. For these purposes, it is necessary to use a two-wave radiothermometry.

7. Discussion of results of the study of formation and recording of thermal electromagnetic radiation

The developed model covers two stages of investigation of the electromagnetic radiation of biological tissue: the distribution of the temperature of biological tissue and the formation of electromagnetic radiation. Electromagnetic radiation is formed by each elementary layer of tissue with subsequent absorption of radiation by the overlying layers.

One of the advantages of the temperature distribution model is the relationship between temperature and physiological parameters. In contrast to the empirical study of temperature distribution, this model allows us not only to ascertain the temperature values at the point of space $x, y, z,$ but also to predict the temperature values depending on the physiological processes in the tissue. In particular, the temperature distribution in the case of necrotic processes has been considered when there is a region with reduced heat release in the tissue.

In the final system of equations obtained by performing measurements at different frequencies, the characteristics of the necrotic region of the temperature anomaly are unknown. In the case of highly specialized diagnostics, this allows to determine the main physiological indicators accurately. In the general case, when causes of temperature change are unknown, a measurement method that determines the temperature values at different points in the tissue is required. This problem has been solved by reducing the temperature distribution model to the form where the variable is the peak value of the anomaly temperature.

In the present study, a single-layer model of biological tissue has been considered. With its help it is possible to model the simplest cases of homogeneous tissue, however, for most real biological tissues, a multilayer structure is required. The study of the distribution of temperature in multilayer biological tissue could become the subject of further research in this field.

8. Conclusions

1. The application of the electrostatic model for determination of the temperature distribution has been substantiated. The model for the distribution of thermal fields within a biological object with the presence of zones with a lowered temperature in the biological object has been improved. Unlike existing models, a three-dimensional model has been considered, this allows to formulate requirements to the equipment more correctly. In addition to areas with increased temperature, this model allows to explore zones with a lower temperature. The model takes into account the physiological processes of formation of temperature fields. It is shown that the determination of sources with a lower temperature is impossible by infrared thermography at depths greater than 3 cm.

2. The possibility of determination of the depth and size of the anomalous temperature zone using two-wave radiothermometry has been confirmed by results of numerical modeling of the formation of thermal radiation and by the solution of the system of equations.

3. The working frequencies of the radiometer are justified taking into account the penetrating power of electromagnetic waves and the requirements to the accuracy of temperature measurement. The error in measuring of the depth of occurrence of the temperature anomaly, based on the available accuracy of determination of the noise temperature by modern radiometers, is no more than 5 % of the depth.

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