RADIOMETRIC METHOD FOR ASSESSING THE DEPTH OF THERMAL INJURIES

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Abstract. This study indicates the need for early diagnosis of the depth of thermal injuries. It was noted that burning injuries account for more than 10% of the total occurrences of injuries. Such statistics indicate the relevance of the use of standard operating procedures for assessing thermal injuries. The analysis of existing standard operating procedures for assessing the depth of thermal injuries has been conducted. The analysis results indicate that at present laser and ultrasonic flowmetry, radiographic and thermographic methods are the most widely used methods for assessing the depth of thermal injuries. Laser and ultrasonic flowmetry methods are based on the study of microcirculation of blood in body tissues. The use of laser radiation sources ensures the differentiation of healthy and damaged tissues at a depth of several millimeters. Complexity and its absence in diagnostic and surgical centers is the main problem of using such equipment. In addition, surface lesions of the skin cover leads to significant errors. The ultrasonic range allows one to increase the depth of diagnosis. However, the equipment used presupposes providing mechanical contact with the affected tissue, which is not always possible. Radiographic methods make it possible to evaluate the depth of thermal damage to biological tissues with high accuracy. Their widespread use for practical diagnosis of burn depth limits the cost of equipment. Besides that, such methods are unsafe for humans. The use of thermographic methods in studying the surface of a burning lesion is very promising. They make it possible to visualize the condition of tissues, do not require the use of radiation, which is harmful to the body, dyes and contrast agents. Their main disadvantage is a

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strong influence on the result of the condition of surface tissues. The results of the analysis indicated that in the domestic and foreign literature there is no information on the use of specialized measuring systems in the diagnosis of burning injuries. This predetermines the urgent need of creating such equipment. Main objective of research – to develop new methods and instruments that allow such an assessment to be performed non-invasively and notouch during the early stages. This study contains the substantiation of the possibility of using its own microwave electromagnetic radiation for assessing the depth of thermal injuries. A scheme, which allows such measurements to be made, is suggested. An algorithm for its operation is described.

1. Introduction

One of the common forms of injuries in industrial and domestic conditions is thermal injuries (burns). According to the information of the World Health Organization in world practice, thermal injuries take third place among other injuries. And in the post-Soviet states they account for more than 10% of the total structure of injuries [1, p. 20]. About 60 thousand people die from burns every year in the world [2, p. 8]. Treatment of burns nowadays remains one of the most difficult problems of surgery. It has not only medical, but also socio-economic significance. This is due to the relatively large relative share of burns among other injuries, high mortality and disability among victims, the cost of their treatment.

In case of burn shock, peripheral vascular spasm occurs. It is followed by expansion, slowing of blood flow, impaired haemopexis, micro thrombosis, impaired metabolic processes, hypoxia, impaired permeability of vascular and cell membranes, pulmonary artery spasm due to the release of catechol amines and impaired vascular permeability with the release of water into the lung parenchyma and other pathophysiological changes.

These changes occur within 6…8 hours after suffering a thermal injury. Therefore, the early start of therapeutic measures for preventing and compensating for them increases the likelihood of a favorable course of burn disease and reduces the occurrence of severe complications.

The main problem in this case is associated with the choice of the correct treatment technique, which depends on the severity of thermal injury. A present there is no single international classification of burn severity

[3, p. 21]. Thus, abroad thermal injuries are usually divided into 3 groups. According to the classification adopted at the II Congress of Surgeons of Ukraine in 2002, thermal injuries shall be divided into 4 groups. In the countries of the former Soviet Union, it is customary to distinguish 5 groups of thermal injuries. Such an approach makes it extremely difficult to formulate universal principles for assessing the severity of thermal injuries and developing unique methods for their treatment.

In all cases, the severity of thermal injury shall be determined by two main parameters – area and depth. Assessing the area of injury is not that problematic. There are many methods available for determining it. Assessing the depth of thermal injury is very problematic at present. This is due to many factors, among which one can distinguish the following ones:

– absence of specialized measuring instruments and standards for simulating thermal injuries of different depths;

– the difference in the surface condition of the injured areas depending on the type of heat agent, etc.

Thus, the diagnosis of deep thermal injuries using non-invasive methods is extremely difficult, especially in the early stages after getting injured. Therefore, when determining the severity of the condition, one often focuses on the total area of the burn, especially in case of a mass flow of injured. In most cases, this approach does not allow one to choose the correct method of treatment.

In this regard, we can conclude the relevance of the development of methods and instruments for non-invasive assessment of the depth of thermal injuries of biological tissues.

2. Analysis of methods for assessing the depth of thermal injuries

According to the terminology, which is accepted in combustiology, methods for assessing the depth of thermal injury are divided into physical methods and standard operating procedures. The first ones involve the use of various kinds of mechanical impacts on damaged tissue and are not considered in this study. The use of standard operating procedures that allow assessing the depth of the injury without direct contact with injured tissue is of greatest interest.

The temperature threshold for human tissue viability is 45...50°C. In case of their overheating, irreversible changes (coagulation) of proteins

occur; cellular enzymes are inactivated, metabolic processes are disrupted, blood circulation ceases. This leads to tissue necrosis. For most of the standard operating procedures discussed below, it is the circulation of blood in the blood vessels that is used as an informative parameter that indicates the depth of the lesion [4, p. 105].

Various methods and methods for diagnostics of skin vitality using laser Doppler flowmetry have been suggested in medical practice [5, p. 11; 6, p. 1]. The authors state that in case if blood microcirculation is during tissue examination, a superficial dermal burn shall be diagnosed. In case of tissue death at a depth of more than 1 mm and, accordingly, absence of blood microcirculation, a deep dermal burn shall be diagnosed.

However, in order to obtain reliable results, indicators of blood microcirculation only in a separate area of the skin are not enough. One shall take into account the general condition of the vascular system and the individual characteristics of the body. In addition, there is a disadvantage, because surface damage prevents the penetration of the laser beam into the tissue. This causes significant errors in the diagnosis of the depth of thermal injury. It is also worth mentioning that the diagnosis of the depth of thermal injury using laser Doppler flowmetry firstly requires the use of complex stationary devices. At the same time, in many cases they do not provide an unambiguous response about the depth of thermal injury and require further research. The consequence of this is that laser Doppler flowmetry methods are not used for the quantitative evaluation of the depth of thermal injury, but for obtaining a qualitative characteristic – the severity of thermal injury.

Some authors use the ultrasonic range of electromagnetic radiation instead of the optical spectrum for the diagnostics of the depth of thermal injury [7, p. 1]. The use of ultrasonic radiation helps increasing the depth of penetration of the probe signal into the tissue under study and allows using additional informative parameters. Thus, the study [8, p. 1] suggests assessing the condition of tissues according to the combined data of Doppler and Elastography ultrasound studies. This approach allows one to replicate the picture of blood flow in the tissues adjacent to the damage area, and thereby draw a conclusion about the depth and degree of tissue injury. However, this is true only for tissues that are located near the passage of the great vessels, which are achievable for hardware research. In other areas, this data is either impossible to obtain or not reliable. Another significant drawback is

that Elastography data can only be obtained in the depth of dissected coagulation-necrotic tissues. That is, such a diagnosis is not only traumatic, but also delayed in time.

Some authors [9, p. 1] suggest assessing the depth of tissue injury using contact method. In this case, the amplitude of the reflected ultrasonic wave on the damaged and intact tissues of morphologically identical zones is being compared. This approach allows non-invasively, in any condition of the victim, even in the presence of burn shock, to identify the boundaries of the burn injury. But at the same time it is possible to detect only gross tissue changes in the form of coagulation necrosis at a depth of not more than 10 mm. This is due to the capabilities of the selected devices, namely surface ultrasonic waves at a frequency of 1.25 MHz.

The absence of unambiguous positive results of assessing the depth of thermal injuries leads to an expansion of the methods used for diagnosis. Thus, in the study [10, p. 1] authors suggest inject a contrast agent and a water-soluble dye directly into the tissue of the supposed injury. X-ray examination of damaged tissue shall be conducted in 12…24 hours. In this case a residual accumulation of contrast is a sign of tissue death. However, in this case, only dead tissue is contrasted even with the existing border zone, that is, in the distant period. The width of the paranecrosis zone and the functional-morphological changes in it, that are observed during the first 3...5 days after thermal injury, cannot be determined in this way.

In the study [11, p. 1] the authors conduct a computed tomographic study and determine the signs of tissue death and coagulation necrosis zones according to their x-ray density. In this case the delimitation of the coagulation necrosis zone shall be carried out by determining the density of tissues in Hounsfield Units (HU). Areas with reduced density (12-15 HU), as well as areas with increased density (65-68 HU), shall be diagnosed as areas of tissue necrosis.

This method allows determining the volume of dead tissue, but does not display their functional condition in the zone, which is qualified as a zone of necrosis. That is, it does not allow delimiting the area of necrosis itself and irreversible changes from areas of reversible changes (paranecrosis areas).

X-ray methods generally allow assessing the depth of thermal injury to biological tissues with high reliability. However, the cost of such equipment and the need for qualified personnel significantly limit their use in the

practical diagnosis of burn depth. In addition, these methods alone are not beneficial to the human body.

Thermographic methods occupy an important place in assessing the depth of thermal injuries. The basis of their use is the fact that, in case of burn tissue injury, coagulation of blood vessels occurs. As a result of this blood circulation is disturbed. This reduces the temperature of the surface layers of the skin. According to some authors [12, p. 761] a temperature difference of homologous skin segments of 1°C indicates the presence of superficial dermal burns. A temperature difference of 2°C and a wound temperature of less than 34°C indicate the presence of deep dermal burns.

Thermographic studies of the surface of the burn injury, as well as the previously considered x-ray studies, make it possible to visualize the state of body tissues. It is worth mentioning that the cost of equipment, if compared with the previously discussed methods, the thermographic method is one of the most affordable. In addition, the use of thermographic studies does not require the use of harmful radiation, dyes and contrast agents that are introduced into the body.

A classic example of thermographic study is described in the study [13, p. 1]. The authors determine the temperature of the body and wounds using a thermal imager. Studies are carried out remotely using special infrared optics. The radiation of the human body is converted into electrical signals, which either give images on the monitor screen, or are registered on special paper. In this way, a temperature pattern of the surface of the human body can be obtained with all its main features and shades that are caused by physiological and pathological processes. The disadvantages of this solution are the need to use expensive specialized equipment that burn and surgical departments do not possess.

In the study [14, p. 1] authors suggest performing preliminary temperature measurements in various parts of the body: thermal injuries and unaffected skin of one homologous segment. It uses a notouch digital infrared thermometer. According to the authors, when performing such studies, the temperature difference of the sections of the thermal injury is from 0,5°C to 3,5°C compared to the intact skin surface. This allows differentiating the depth of burn. But, it is difficult to obtain a complete picture of a burn during spot measurements using an infrared thermometer. In addition, the measurement result largely depends on the surface condition of both thermal injury and the skin itself.

The conducted analysis allows drawing several conclusions regarding the use of equipment for clinical laboratory evaluation of the depth of thermal injuries. Firstly, in the domestic and foreign literature there is no information on the use in practice of diagnosis and treatment of thermal injuries of specialized measuring systems. For the most part, equipment of a different functional purpose, not intended for such studies, is used to diagnose the depth of thermal injuries. Secondly, the use of the considered methods often implies the need for the introduction of additional drugs into the body and surgical interventions, which often does not contribute to the acceleration of treatment. Thirdly, some methods make it possible to assess the degree of thermal injury only 3...5 days after receiving the injury. In most cases this is unacceptable.

Thus, taking into account the state of the problem of assessing the depth of tissue injuries during thermal exposures, it is extremely important to develop new methods and instruments that allow such an assessment to be performed non-invasively and notouch during the early stages.

3. Assessment of the depth of thermal injuries according to the level of inherent electromagnetic radiation of body tissues

Radiometric method is one of the promising methods for assessing the depth of the location of vital biological tissues.

It is well known [15, p. 311], that one of the manifestations of the activity of living cells of organisms is its inherent electromagnetic radiation (EMR) of the microwave range. The intensity of inherent microwave radiation shall be determined by the activity of biological processes in the cells of living organisms. The level of this radiation is very low (10^{-14}…10^{-15} W). It is comparable with the level of thermal radiation of dielectrics in the indicated frequency range.

According to the current laws of physics, anybody, which is heated to a certain temperature, emits a wide range of thermal radiation. For objects having a temperature in the range of 20...40°C, a frequency band is characteristic in the infrared region of the spectrum, where the EMR intensity has maximum values. However, infrared radiation from the internal structures of biological tissues is shielded by the skin cover. That is why the use of thermographic methods gives the temperature difference between living and non-living tissue at the level only 1…2°C.

While analyzing the components of the thermal radiation of living organisms, the following features should be mentioned. The emissivity of any heated bodies in the microwave range obeys the Rayleigh-Jeans law

$$B = 2f^2kT\beta / C^2,$$

where: f – frequency;
k – Boltzmann constant;
T – temperature;
β – radiation value;
C – speed of light in vacuum.

Biological tissues that contain viable cells have non-thermal EMR in the microwave range (30...500 GHz). Acoustoelectric processes that occur in cell membranes are the source of EMR. Their intensity characterizes the interaction of protein molecules and intracellular structures. The density of coherent electromagnetic waves emitted by cells already at a small distance from their surface decreases many times. Moreover, this decrease is accompanied by stochastization of radiation. This is due to the small size of the cells in comparison with the emitted wavelength.

As a result of this, the coherent radiation of individual cells is converted into a noise EMR, which is added to the original radio thermal EMR, which also has a noise character. Therefore, in order to analyze the level of radiation of living tissue, it is necessary to measure a weak bioinformatic electromagnetic radiation in the microwave range against the background of more powerful radio thermal radiation. This is a difficult task, especially taking into account the fact that the total power of thermal and bioinformatic radiation is several orders of magnitude lower than the noise power of modern electronic measuring equipment.

However, it is worth mentioning that the skin for the EMR microwave range is not a screen, in contrast to thermal radiation. Due to this, the microwave component of EMR is more informative in terms of assessing cell viability, if compared with the infrared component.

Thus, the objective of this study is creating such a device for the analysis of the level of electromagnetic radiation of biological tissues, which provides the measurement of the bioinformatic (microwave) component of electromagnetic radiation regardless of the temperature of the analyzed material.

Figure 1 shows a functional diagram of a device for assessing the level of electromagnetic radiation of biological tissues.

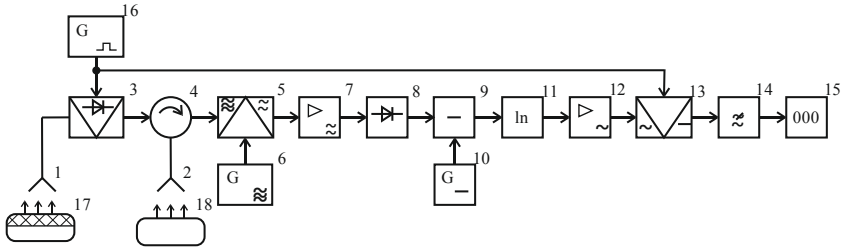

Figure 1. Functional diagram of a device for assessing the level of electromagnetic radiation of biological tissues

The following legend is used in the diagram: 1, 2 – measuring and support receiving antennas; 3 – controlled microwave reflector; 4 – circulator; 5 – mixer; 6 – microwave oscillator; 7 – intermediate frequency amplifier; 8 – amplitude detector; 9 – subtractor; 10 – constant voltage source; 11 – logarithmer; 12 – low frequency amplifier; 13 – synchronous detector; 14 – low pass filter; 15 – voltmeter; 16 – low frequency multivibrator; 17 – the analyzed fragment of the affected tissue; 18 – healthy tissue fragment.

The device indicated in the diagram works as follows.

Radiothermal and biological EMR from the measuring receiving antenna 1 while interacting with the analyzed fragment of the affected biological tissue, is fed to the input of a controlled microwave reflector 3. The output of the controlled reflector is connected to one of the inputs of the circulator 4. Radio thermal radiation from the support receiving antenna 2, while interacting with unaffected fragment of biological tissue 18, is fed directly to the second input of the circulator 4.

The microwave reflector 3 is made on p-i-n diodes and works according to the principle of total reflection when voltage is applied to its control input. A periodic change in the reflector mode (reflects – transmits) is carried out by rectangular pulses of modulating voltage, which is created by a low frequency multivibrator 16.

During one half-period of the low-frequency modulating voltage, when the microwave reflector 3 transmits radiation, a dispersion noise signal is input to the mixer 5

$$\bar{U}_{11}^2 = K_1\left(\bar{U}_{21}^2 + \bar{U}_3^2 + \bar{U}_4^2\right), \qquad (1)$$

where $\bar{U}_{21}^2$ – dispersion of radio thermal radiation of the analyzed tissue fragment;

$\bar{U}_3^2$ – dispersion of bioinformatic radiation of viable cells of the analyzed tissue fragment;

$\bar{U}_4^2$ – dispersion of inherent noises of the mixer reduced to its input;

K_1 – waveguide transmission coefficient.

During the second half-cycle of the modulating voltage, when the microwave reflector 3 is closed, i.e. fully reflects the radiation; the noise signal from the output of the measuring receiving antenna 1 is reflected. At the same time, the output signal of the support receiving antenna 2 passes through the circulator 4, is reflected from the closed microwave reflector 3, and again through the circulator 4 it is fed to the input of the mixer 5. The dispersion of the input signal of the mixer during this half-cycle takes on the value

$$\bar{U}_{12}^2 = K_1\left(\bar{U}_{22}^2 + U_4^2\right), \tag{2}$$

where $\bar{U}_{22}^2$ – dispersion of radiothermal radiation of unaffected tissue fragment.

In case if the switching frequency of the multivibrator is determined according to the value of Ω. Then, during periodic operation of the microwave reflector 3, an amplitude modulated signal is formed at the input of the mixer 5, consisting of packets of noise signals with dispersions (1) and (2).

Given the real ratio of the component signals (1) and (2), the modulation depth M of the input signal of the mixer is small $\bar{U}_4^2 \rangle\rangle \bar{U}_{21}^2 = \bar{U}_{22}^2 \rangle \bar{U}_3^2$ and is represented by the expression

$$M_1 = \frac{\bar{U}_{11}^2 - \bar{U}_{12}^2}{\bar{U}_{11}^2 + \bar{U}_{12}^2} = \frac{\bar{U}_3^2}{2\left(\bar{U}_4^2 + \bar{U}_{21}^2\right)}. \tag{3}$$

Using the signal of microwave oscillator 6 the microwave radiation spectrum is transferred to the intermediate difference frequency ω_0. It is at this frequency that the frequency amplifier 7 of the intermediate frequency is tuned. The width of the transferred spectrum $\Delta\omega$ is determined by bandwidth $\Delta\omega_0$ of the intermediate frequency amplifier with the central frequency $\omega_0 \rangle\rangle \Omega$. During one half-period of the reflector 3, the dispersion of the narrow-band noise signal is at the output of the intermediate frequency amplifier

$$\bar{U}_{51}^2 = K_1 S_1 K_2 \left[\bar{U}_{21}^2(\omega_0) + \bar{U}_3^2(\omega_0) + \bar{U}_4^2(\omega_0)\right]. \tag{4}$$

During the second half-period

$$\bar{U}_{52}^2 = K_1 S_2 K_2 \left[\bar{U}_{22}^2 (\omega_0) + \bar{U}_4^2 (\omega_0) \right]. \quad (5)$$

Signal packages of intermediate frequency ω_0 with dispersions (4) and (5) alternately are fed at an amplitude detector with a quadratic characteristic. At the detector output, video pulses with amplitudes that are proportional to the dispersion of the detected signals are formed

$$U_{61} = K_1^2 S_1^2 K_2^2 S_2 \left[\bar{U}_{21}^2 (\omega_0) + \bar{U}_3^2 (\omega_0) + \bar{U}_4^2 (\omega_0) \right], \quad (6)$$

$$U_{62} = K_1^2 S_1^2 K_2^2 S_2 \left[\bar{U}_{22}^2 (\omega_0) + \bar{U}_4^2 (\omega_0) \right], \quad (7)$$

where S_2 – steepness of the quadratic detector conversion.

Video pulses U_{61} and U_{62} alternately affect one input of the subtractor 9, constant voltage from the source is fed on the other input of it 10. Constant voltage U_7 of the source 10 is set from the condition for compensation of the noise of the mixer, which affect the depth of modulation

$$U_7 = K_1^2 S_1^2 K_2^2 S_2 \bar{U}_4^2 (\omega_0). \quad (8)$$

When condition (8) is fulfilled, the amplitudes of the video pulses at the output of the subtractor take values

$$U_{81} = K_1^2 S_1^2 K_2^2 S_2 K_3 \left[\bar{U}_{21}^2 (\omega_0) + \bar{U}_3^2 (\omega_0) \right], \quad (9)$$

$$U_{82} = K_1^2 S_1^2 K_2^2 S_2 K_3 \bar{U}_{22}^2 (\omega_0), \quad (10)$$

where K_3 – subtractor transmission coefficient/

After the functional conversion of the video pulses in the logarithmer, their amplitudes take the final form

$$U_{91} = S_3 \ln\{K_1^2 S_1^2 K_2^2 S_2 K_3 \left[\bar{U}_{21}^2 (\omega_0) + \bar{U}_3^2 (\omega_0) \right]\}, \quad (11)$$

$$U_{92} = S_3 \ln[K_1^2 S_1^2 K_2^2 S_2 K_3 \bar{U}_{22}^2 (\omega_0)], \quad (12)$$

where S_3 – steepness of a logarithmic transformation.

The depth of the amplitude modulation of the sequence of video pulses increases and taking into account the equality $\bar{U}_{21}^2 = \bar{U}_{22}^2$ takes the value

$$M_2 = \frac{U_{91} - U_{92}}{U_{91} + U_{92}} = \frac{\ln\left[\dfrac{\bar{U}_{21}^2 (\omega_0) + \bar{U}_3^2 (\omega_0)}{\bar{U}_{21}^2 (\omega_0)} \right]}{\ln\left\{ \left(K_1^2 S_1^2 K_2^2 S_2 K_3 \right)^2 \left[\bar{U}_{21}^2 (\omega_0) + \bar{U}_3^2 (\omega_0) \right] \bar{U}_{21}^2 (\omega_0) \right\}}. \quad (13)$$

The low frequency amplifier from the sequence of video pulses (12) isolates and amplifies the variable frequency component Ω with amplitude

$$U_{10} = K_4 \frac{U_{91} - U_{92}}{2} = \frac{1}{2} S_3 K_4 \ln \frac{\bar{U}_{21}^2 + \bar{U}_3^2(\omega_0)}{\bar{U}_{21}^2(\omega_0)}. \quad (14)$$

where K_4 – low frequency amplifier gain ratio.

Expression (14) can be represented as

$$U_{10} = \frac{1}{2} S_3 K_4 \ln\left[1 + \frac{\bar{U}_3^2(\omega_0)}{\bar{U}_{21}^2(\omega_0)}\right]. \quad (15)$$

After expanding it into a power series, we obtain

$$U_{10} = \frac{1}{2} S_3 K_4 \left[\frac{\bar{U}_3^2(\omega_0)}{\bar{U}_{21}^2(\omega_0)} - \frac{\bar{U}_3^4(\omega_0)}{2\bar{U}_{21}^4(\omega_0)} + ...\right]. \quad (16)$$

The power of bioinformatic radiation is much lower than the power of thermal radiation ($\bar{U}_3^2 \langle\langle \bar{U}_{21}^2$).Therefore, the second and subsequent terms of expansion terms (16) can be neglected. Then there is the following voltage at the output of the amplifier

$$U_{10} = \frac{1}{2} S_3 K_4 \frac{\bar{U}_3^2(\omega_0)}{\bar{U}_{21}^2(\omega_0)}. \quad (17)$$

Voltage U_{10} is rectified by a synchronous detector, which is controlled by the square voltage of the multivibrator, and is smoothed by a low-pass filter. DC output voltage is measured using a voltmeter.

From the expression (17) it is clear that the voltmeter readings are proportional to the ratio of bioinformatic and radiothermal radiation powers

$$\alpha = K_0 \frac{\bar{U}_3^2(\omega_0)}{\bar{U}_{21}^2(\omega_0)}. \quad (18)$$

where α – voltmeter readings;

$K_0 = \frac{1}{2} S_3 K_4$ – coefficient of proportionality.

The power ratio is a measure of the EMR level of viable biological tissue cells. The resulting ratio shall be determined by the intensity of biochemical processes in living tissues and allows assessing the depth of thermal injury.

The ability to use receiving antennas with different geometries allows localizing the area of the tissue fragment under study in the range from several square millimeters to several tens of centimeters.

Independence of the ratio η from weight due to the fact that the power of both bioinformatic and radio thermal radiation is proportional to the effective area of the receiving antennas. In addition, the measured power ratio does not depend on the transfer properties of the connecting wave-

guides (K_1), the inconsistency of the steepness of the heterodyne conversion of the spectrum of the compared signals (S_1), the instability of the gain of the selective intermediate-frequency amplifier (K_2), the sensitivity of the amplitude detector (S_2) and the gain ratio of the subtractor (K_3).

Proportionality factor K_0 depends only on the stability of the low-frequency blocks of the circuit and is determined during the calibration process using samples of microorganisms, the emissivity of which shall be estimated according to the results of biochemical studies.

4. Differential radiometer for recording the difference values of radiation intensities

To study the gradients of the electromagnetic field of biological objects, you can use differential radiometers that measure the difference in radiation intensities from neighboring or remote biologically active points, as well as various (for example, symmetrical) body parts.

To increase the sensitivity of differential radiometers, it is reasonably to use power feedback to small difference intensities, which provides a deep modulation [16, p. 220].

Figure 2 shows a functional diagram of a differential radiometer for recording the difference values of radiation intensities.

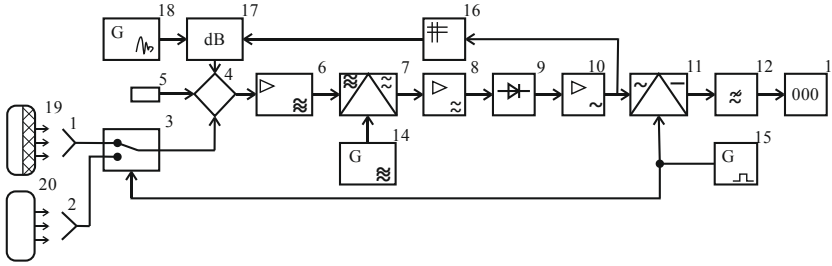

Figure 2. Functional diagram of a differential radiometer for recording the difference values of radiation intensities

The following legend is used in the diagram: 1, 2 – measuring and support receiving antennas; 3 – microwave switch; 4 – hybrid tee; 5 – matched load; 6 – microwave amplifier; 7 – microwave mixer; 8 – intermediate frequency amplifier; 9 – amplitude detector; 10 – low-frequency amplifier;

11 – synchronous detector; 12 – low pass filter; 13 – voltmeter; 14 – microwave oscillator; 15 – low frequency multivibrator; 16 – control unit; 17 – controlled attenuator; 18 – noise generator; 19 – the analyzed fragment of the affected tissue; 20 – healthy tissue fragment.

The device indicated in the diagram works as follows.

The thermal radiation from the surface of an object is received by microwave antennas 1 and 2. The output signals of an antenna can be represented as dispersions of random signals:

$$\bar{U}_1^2 = ST_1, \tag{19}$$

$$\bar{U}_2^2 = ST_2, \tag{20}$$

where S – the antennas sensitivity;

T_1 and T_2 – a temperature of controlled areas of an object surface.

The signals $U_1(t)$ и $U_2(t)$ through arms of the microwave switch 3 periodically arrive at one input of a hybrid tee 4. The signal from the noise generator 18 is fed to the second input of the tee through the attenuator 17, controlled by an electrical voltage of only one polarity – negative or positive one.

Since the controlled input of the attenuator is connected through the control unit 15, which is a rectangular voltage generator and a power amplifier that is connected to the output of the low-frequency amplifier 10, the attenuator opens for a time equal to the half-period of a low-frequency voltage, which is the envelope of the modulated microwave signal. In this case, the half-cycle of the low frequency is equal to the half-cycle of the switching frequency of the microwave switch 3.

In the first switching half-period, when a signal $U_1(t) > U_2(t)$ is fed to the input of the microwave switch, the attenuator 17 opens.

Independent noise signals are summed in a hybrid tee, the sum dispersion of which can be represented in the following way:

$$\bar{U}_4'^2 = K_1\left(\bar{U}_1^2 + K_2\bar{U}_3^2 + \bar{U}_5^2\right), \tag{21}$$

where K_1 – the power transmission coefficient of the hybrid tee 4;

K_2 – the attenuator power transfer coefficient 17;

$U_3(t)$ – signal of generator noise 18;

$U_5(t)$ – the intrinsic noise of a one-link path of a differential radiometer, brought to the input of the amplifier 6.

In the second switching half-cycle, the signal $U_2(t) < U_1(t)$ is supplied to the input of the microwave switch 3, the attenuator 17 is closed by a

low-frequency voltage of the opposite polarity. Therefore, the sum noise signal can be represented with dispersion:

$$\bar{U}''^{2}_{4} = K_1\left(\bar{U}^{2}_{2} + \bar{U}^{2}_{5}\right). \quad (22)$$

Signals $U'_4\ (t)$ and $U''_4\ (t)$, that create one modulated signal, are periodically fed to mixer 7. An amplifier with a filter 8 selects a signal of difference frequency, the spectrum of which is determined by the passband of the filter.

The quadratic detector 9 receives packets of noise signals of an intermediate frequency, which can be represented as dispersion:

$$\bar{U}'^{2}_{6} = K_3K_1\left(\bar{U}^{2}_{1} + K_2\bar{U}^{2}_{3} + \bar{U}^{2}_{5}\right), \quad (23)$$

$$\bar{U}''^{2}_{6} = K_3K_1\left(\bar{U}^{2}_{2} + \bar{U}^{2}_{5}\right), \quad (24)$$

where K_3 – a gain of the amplifier 8 in the filter bandpass.

The voltage of the envelope of the switching frequency with the amplitude is allocated at the output of the quadratic detector 9:

$$U_7 = K_1K_3K_4\left(\bar{U}^{2}_{1} + K_2\bar{U}^{2}_{3} - \bar{U}^{2}_{2}\right), \quad (25)$$

where K4 – a conversion coefficient of the quadratic detector 9.

An alternating voltage with an amplitude $U_7(t)$ controls the operation of an attenuator 17 and is simultaneously rectified by a synchronous detector 11. The rectified voltage, which is fixed by indicator 13, can be written as follows:

$$U_8 = K_1K_3K_5K_6\left(\bar{U}^{2}_{1} - \bar{U}^{2}_{2} + K_2\bar{U}^{2}_{3}\right) = \alpha\left(\bar{U}^{2}_{1} - \bar{U}^{2}_{2} + K_2\bar{U}^{2}_{3}\right), \quad (26)$$

where K_5 – the gain of amplifier 10;

K_6 – a conversion coefficient of a synchronous detector 11;

$\alpha = K_1K_3K_5K_6$ – a resulting coefficient of a direct conversion of radiometer.

The gain of the controlled attenuator A3 is proportional to the amplitude of the low-frequency voltage $U_7(t)$:

$$K_2 = K_7U_7, \quad (27)$$

where K_7 – the gear ratio of a control unit 16.

If a control voltage $U_7(t)$ is expressed through the output voltage $U_8(t)$, then:

$$K_2 = \frac{K_7}{K_6}U_8 = \beta U_8, \quad (28)$$

where $\beta = \dfrac{K_7}{K_6}$ – a coefficient of the inverse transformation of a radiometer output voltage.

Substituting the value a transmission coefficient K_2 of the controlled attenuator 17 from expression (28) into the expression (26), we get:

$$U_8 = \frac{\alpha}{1 - \alpha\beta\bar{U}_3^2}\left(\bar{U}_1^2 - \bar{U}_2^2\right). \quad (29)$$

Using expressions (29), (19) and (20), which take into account the temperature of controlled areas of an object, we finally get:

$$U_8' = \frac{\alpha S}{1 - \alpha\beta\bar{U}_3^2}\left(T_1 - T_2\right). \quad (30)$$

A change in the sign of the measured temperature difference $(T_1 < T_2)$ leads to a change in the phase of the low-frequency voltage at the output of the 10 amplifier by 180°.

As a result, the switching half-period changes, in which the controlled attenuator 17 opens, and the noise signal from the generator 18 is summed not with the signal $U_1(t)$, but with the signal $U_2(t)$ (with higher power). Thus, the operation of differential radiometer is not disturbed, and its indications will be equal to:

$$U_8'' = \frac{\alpha S}{1 - \alpha\beta\bar{U}_3^2}\left(T_2 - T_1\right). \quad (31)$$

The rectified voltage can be written in the following form because when the phase of the voltage of a switching frequency changes by 180°, the polarity of a rectified voltage changes at the output of a synchronous detector 11:

$$U_8 = \pm\frac{\alpha S}{1 - \alpha\beta\bar{U}_3^2}\left(T_1 - T_2\right). \quad (32)$$

Thus, the considered differential radiometer works consistently at any temperature ratio of the controlled object ($T_1<T_2$ or $T_1>T_2$), and the polarity of the measured voltage determines the sign of the controlled temperature difference. If the feedback coupling is absent in differential RS ($\beta = 0$), then the output voltage has the form:

$$U_9 = \alpha S\left(T_1 - T_2\right). \quad (33)$$

The introduction of positive feedback ($\beta > 0$) leads to the appearance of an output voltage, which is described by expression (32), and an increase in the sensitivity of the differential radiometer by a factor equal to:

$$\gamma = \frac{U_8}{U_9} = \frac{1}{1 - \alpha\beta\bar{U}_3^2}. \quad (34)$$

If a denominator of expression (34), for example, is equal to 0.01, then the sensitivity of differential radiometer will increase 100 times due to the feedback coupling.

The maximum gain in sensitivity with the condition $\left(\left[1-\alpha\beta\,\bar{U}_3^2\right]\to 0\right)$ is limited by the possibility of auto-oscillations in the positive feedback circuit. Phase compensation chains and amplitude limiting elements are introduced to suppress auto-oscillations in control unit 17.

In practice, the fluctuation threshold of sensitivity of a differential radiometer can be reduced to $10^{-22}...10^{-23}$ W/Hz by introducing feedback coupling, which corresponds to sensitivity by temperature difference $10^{-4}...10^{-5}$ K.

The usage of the considered differential radiometer allows to study the gradients of the temperature fields of biological objects in the range of their electromagnetic radiation.

4. Conclusions

As a result of the analysis of the state of the problem of diagnosing the depth of thermal injury of biological tissues, it can be concluded that it is necessary to develop specialized diagnostic equipment. The fundamental possibility and prospects of using radiometric measurement methods for these purposes, that make it possible to assess the level of intrinsic electromagnetic radiation of living tissues in the microwave range, are indicated. A functional diagram of such device is provided. It provides the possibility to implement such measurements. The use of two receiving antennas ensures the elimination of the influence of radio thermal component of the radiation and the allocation of the bioinformatic component due to the activity of body cells. Comparison of bioinformatic and radio thermal components allows one to ultimately assess the depth of thermal injury.

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