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COMPUTER SIMULATION OF LOCAL THERMAL EFFECT ON BIOLOGICAL TISSUE

The physical, mathematical and computer models of local thermal effects on biological tissue are constructed in this paper. The thermal effect of thermoelectric device work instrument on biological tissue for destruction of cancer tumours is investigated by computer simulation. Temperature distributions in biological tissue and work instrument in cooling and heating modes are determined. The obtained results make it possible to optimize device design to achieve the required depth of biological tissue freezing, and the maximum effect at destruction of cancer tumours.

***Keywords**: thermoelectric cooling, cryodestruction, hyperthermia, biological tissue, computer simulation.*

Introduction

It is known that the use of cooling in surgery [2] promotes the reduction of blood loss, weakens pain severity and duration, and prevents microbial contamination and metastasis spread. A deeper cooling (to -60 °C), followed by destruction of biological tissue structure, is also used in medical practice for cryodestruction [2 – 11]. In so doing, blood circulation, oxygen delivery, nutrient enrichment, tissue respiration and all biochemical processes are completely stopped. As a result, the death of cells, wherein all vital processes have been continuously paralyzed, occurs. A similar effect with using elevated temperatures - hyperthermia (+39 ÷ +45 °C) for the destruction of benign neoplasms is observed, which leads to thermal damage and destruction of cancer cells, while healthy cells remain undamaged.

The above mechanisms of thermal effect on biological tissue show that thermoelectric cooling and heating have good prospects for surgery. It is due to their advantages, namely possibility of setting precisely the required temperature of work instrument surface, the duration of temperature effect on the corresponding part of human body, and a cyclic change of cooling and heating modes [12, 13]. However, the use of reduced and elevated temperatures in medical practice requires a comprehensive in-depth study of the characteristic features of thermal effects in healthy and affected tissues, which is a complicated task that requires creation of precise physical and mathematical models and the use of computer simulation.

Therefore, *the purpose of this paper* is to develop computer simulation method which will make it possible to predict the results of local thermal effect on biological tissue at the destruction of cancer tumours.

Physical, mathematical and computer models of local thermal effect on biological tissue

Prediction of thermal effect on biological tissue is an intricate multi-parameter task, which depends on the temperature and geometry of the work instrument, the cooling rate, the time of exposure, as well as the size and structure of biological tissue.

To solve this task, a method of analytical simulation of frozen tissue is generally used.

There is a fair amount of models describing the processes of freezing around cryotools which differ in complexity [14 – 17]. Analytical simulation for the estimation of the freezing area size is critical as a method for analysis and optimization of cryodestruction efficiency and as a basis for calculation of designs and systems of cryotools for deep cooling of cryosurgical equipment. Introduction of mathematical methods for prediction of thermal effect results will depend on the accuracy of description of thermal processes occurring during freezing of living tissues and accompanying phase transitions [18, 19].

In order to make preliminary prediction of the results of local thermal effects on biological tissue, physical (Fig. 1), mathematical and computer models have been developed.

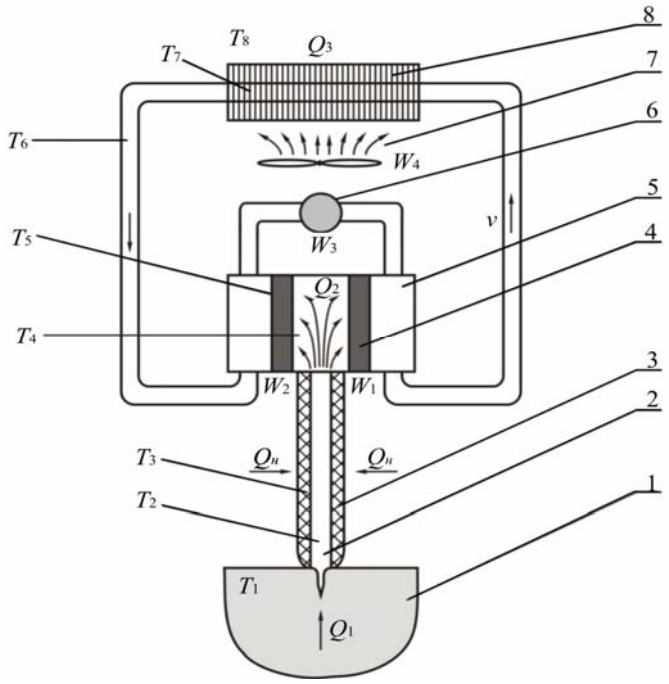

Fig. 1. A physical model of local thermal effect on biological tissue:
1 – biological tissue, 2 – work instrument, 3 - insulation,
4 - two-stage thermoelectric module, 5 - liquid heat exchanger,
6 - pump, 7 - fan, 8 - liquid-air heat exchanger.

Structurally, a model consists of a work instrument, thermal insulation of the work instrument lateral surface; 2 two-stage thermoelectric modules, on the hot sides of which liquid heat exchangers are mounted, a circulation pump, which pumps the water through the channels; a fan and a liquid-air heat exchanger, which cool the liquid pumped. Since it is necessary to achieve the lowest possible temperature values for the destruction, thermoelectric modules which are characterized by increased temperature difference ΔT_{max} are used.

The heat flow through the thermoelectric modules:

$$Q_2 = Q_1 + Q_H\,, \tag{1}$$

where Q_1 is the heat flow from the tumor, Q_{hg} is the heat inleak through the lateral surface of the work instrument.

The heat flow from the thermoelectric modules:

$$Q_3 = Q_2 + W\,, \tag{2}$$

where W is the power of thermoelectric modules.

The heat flux transferred to water from the hot side of the modules:

$$Q_4 = \alpha_4 S_4 (T_5 - T_6)\,. \tag{3}$$

The heat flux transferred to the liquid-air heat exchanger from water:

$$Q_5 = \alpha_5 S_5 (T_6 - T_7)\,. \tag{4}$$

The heat flux transferred to environment from the liquid-air heat exchanger:

$$Q_6 = \alpha_6 S_6 (T_7 - T_8)\,, \tag{5}$$

where α_4, α_5, α_6 are the heat transfer coefficients, S_4, S_5, S_6 are the heat exchange surface areas.

To determine the temperature distribution in the structural components of thermoelectric device for cancer tumours destruction in biological tissue (liver affected with cancer), Comsol Multiphysics software package is used [20, 21] which allows simulation of thermophysical processes in biological tissue with regard to blood circulation and metabolism. The heat transfer equation in biological tissue in this case will take on the form:

$$\rho C_p \frac{\partial T}{\partial t} + \nabla \cdot (-k \nabla T) = \rho_b C_b \omega_b (T_b - T) + Q_{met}\,, \tag{6}$$

where: ρ_b is blood density (kg/m^3), C_b is specific heat (J/kg*K), ωb is blood circulation rate (1/sec), T_b is arterial blood temperature (K), Q_{met} is the amount of heat due to metabolism (W/m^3).

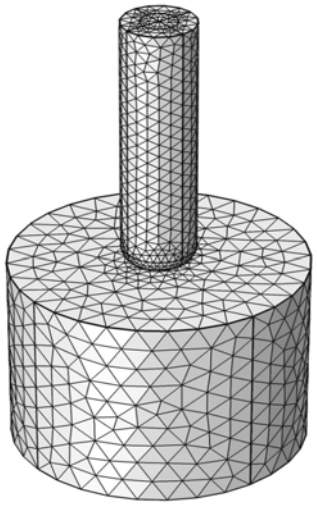

Fig. 2. Finite element method mesh.

Calculation of temperature and heat flux density distribution in biological tissue and work instrument is carried out by the finite element method [22] (Fig. 2).

Computer model is a bulk of the biological tissue with the isotropic thermal properties. A needle made of high thermal conductivity material fixed on the rod of the same material is placed inside the tissue. The rod, surrounded by thermal insulation, is in heat exchange state with the environment. The temperature at the end of the rod is preset and equals to -50 °C. The boundary condition in the area sufficiently distant from the probe, where the temperature should be equal to body temperature, is 37 °C. In the process of freezing, cells will be subject to phase change at freezing point. The tissue properties in frozen and unfrozen states are shown in Table 1. In the temperature range of -1 ÷ -8 °C, when the cells are frozen, the latent heat of the phase transition is absorbed and can be simulated by adding the corresponding value to specific heat [23].

Table 1.

The properties of biological tissue in frozen and unfrozen states [24]

	Unit	Value
Specific heat of frozen tissue	MJ/m^3 °C	1.8
Specific heat of unfrozen tissue	MJ/m^3 °C	3.6
Specific heat of blood	MJ/m^3 °C	3.6
Thermal conductivity of unfrozen tissue	W/ m °C	0.5
Thermal conductivity of frozen tissue	W/m °C	2
Latent heat	MJ/m^3	250
Body temperature	°C	37
Lower phase transition temperature	°C	-8
Upper transition temperature	°C	-1
Blood perfusion in healthy tissue	ml / s / ml	0.0005
Blood perfusion in tumour	ml/s/ml	0.002
Metabolism in normal tissue	W/m^3	4200
Metabolism in tumour	W/m^3	42000

Moreover, at freezing of biological tissue, blood vessels in capillaries are constricted to freezing of all the blood, the value ω_b tending to zero. In this case, the cells will not be able to generate metabolic heat, and Q_{met} will be equal to zero.

Work instrument optimization

Temperature distributions in the work instrument and tumor for heating and cooling modes were investigated, which allowed optimization of thermoelectric device work instrument for the destruction of oncologic neoplasms. Dependences of freezing depth and needle temperature on the diameter and length of work instrument and needle diameter were determined.

As an example, Fig. 3 shows a typical temperature distribution in the work instrument and biological tissue in cooling mode. For this case: rod diameter is 8 mm, rod length is 40 mm, insulation thickness is 5 mm, needle length is 7 mm, needle diameter is 2 mm. Curves 1, 2 are isotherms - 1°C and - 8 °C, respectively. Temperature distribution along half-sphere radius of the work instrument thermal effect (lines *R* in Fig. 3) is shown in Fig. 4.

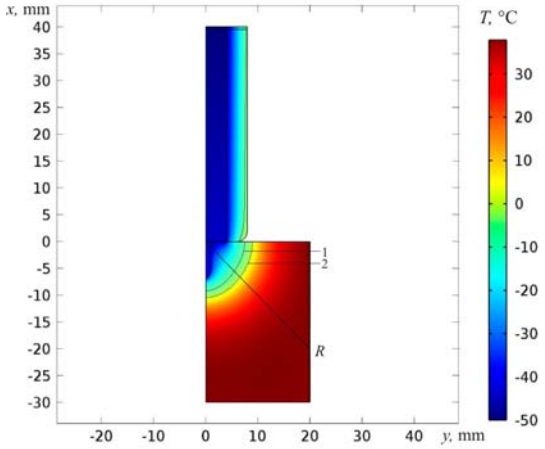

Fig. 3. Typical temperature distribution in the work instrument and biological tissue in cooling mode.

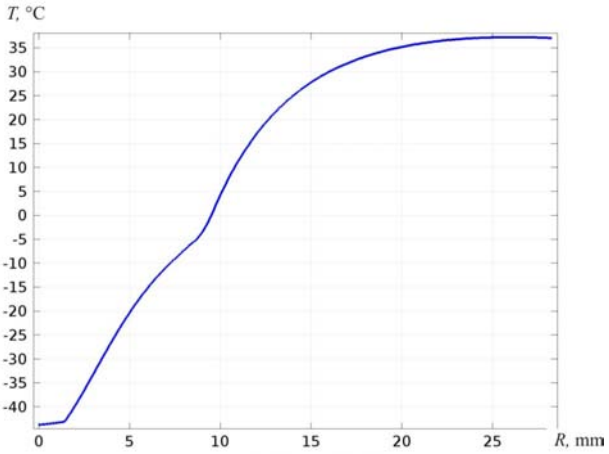

Fig. 4. Temperature distribution along half-sphere radius of device thermal effect for cooling mode.

Similar distributions for heating mode are shown in Figs. 5 – 6. Fig. 5 shows a typical temperature distribution in the work instrument and biological tissue, Fig. 6 – temperature distribution along line *R*.

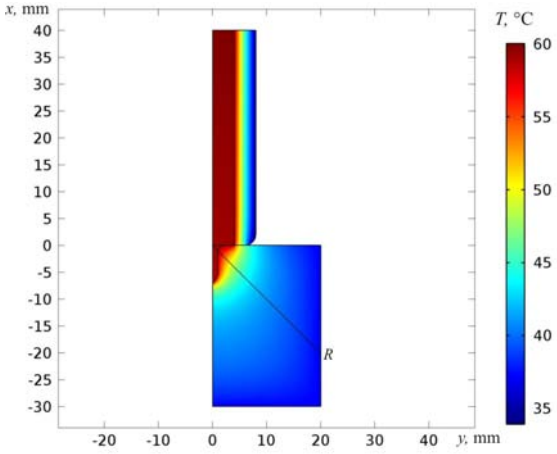

Fig. 5. Typical temperature distribution in the work instrument t and biological tissue in heating mode.

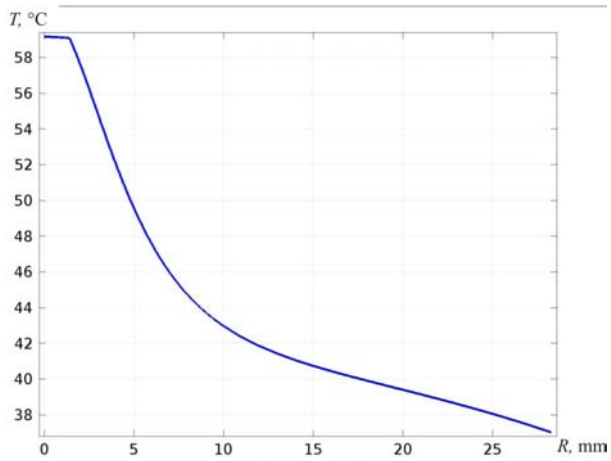

Fig. 6. Temperature distribution along half-sphere radius of device thermal effect for heating mode.

Figs. 7 – 11 show the dependences of needle temperature and freezing depth on the rod diameter at different rod lengths. Insulation thickness h_{is} = 5 mm, needle diameter d_n = 2 mm, needle length h_n = 7 mm.

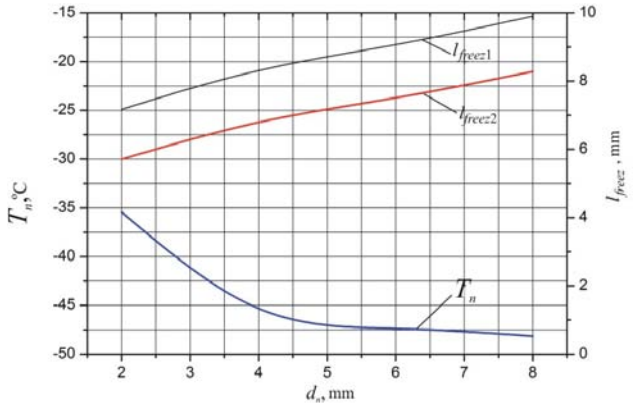

Fig. 7. Dependences of needle temperature (T_n) and freezing depth (l_{freez1} is the distance t o isotherm with temperature -1 ° C, l_{freez2} is the distance to isotherm with temperature -8 ° C) on rod diameter (rod length 10 mm).

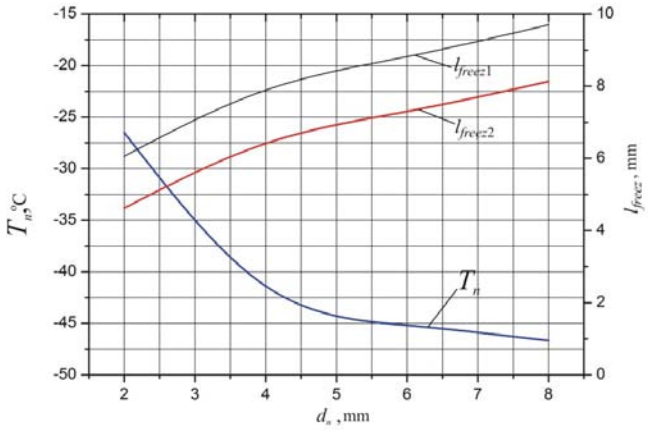

Fig. 8. Dependences of needle temperature (T_n) and freezing depth ($freez_1$ is the distance to isotherm with temperature -1 ° C, l_{freez2} is the distance to isotherm with temperature -8 ° C) on rod diameter (rod length 20 mm).

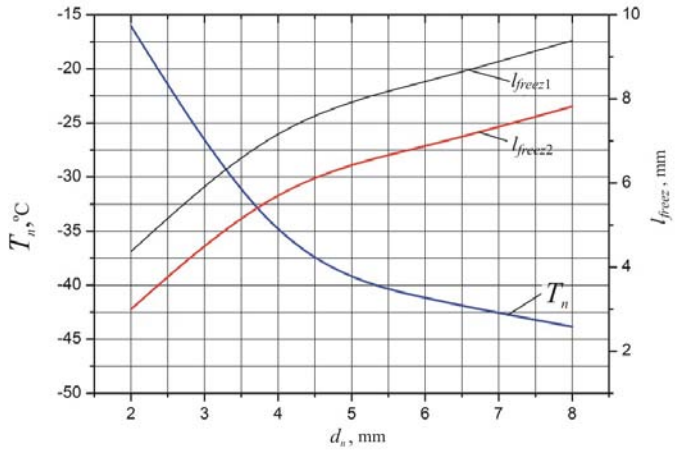

Fig. 9. Dependences of needle temperature (T_n) and freezing depth (l_{freez1} is the distance to isotherm with temperature -1 ° C, l_{freez2} is the distance to isotherm with temperature -8 ° C) on rod diameter (rod length 40 mm).

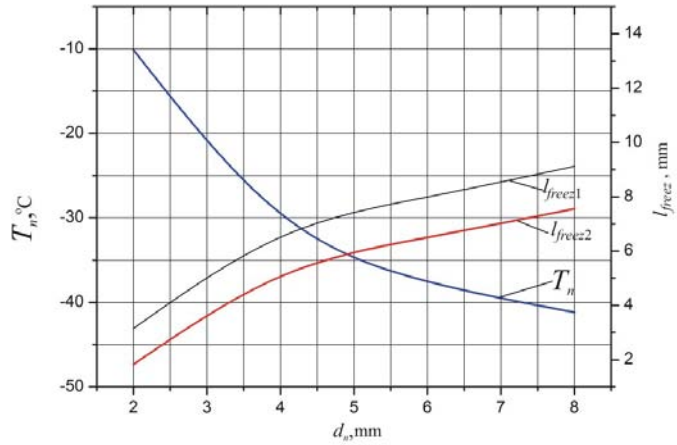

Fig. 10. Dependences of needle temperature (T_n) and freezing depth (l_{freez1} is the distance to isotherm with temperature -1 ° C, l_{freez2} is the distance to isotherm with temperature -8 ° C) on rod diameter (rod length 60 mm).

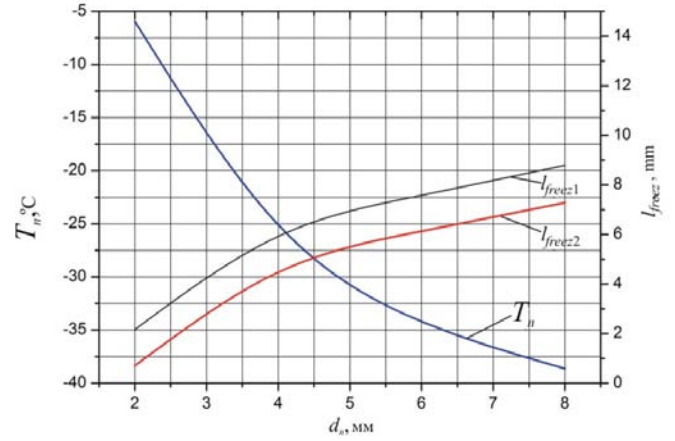

Fig. 11. Dependences of needle temperature (T_n) and freezing depth (l_{freez1} is the distance to isotherm at temperature -1 ° C, l_{freez2} – distance to isotherm with temperature -8 ° C) on rod diameter (rod length 80 mm).

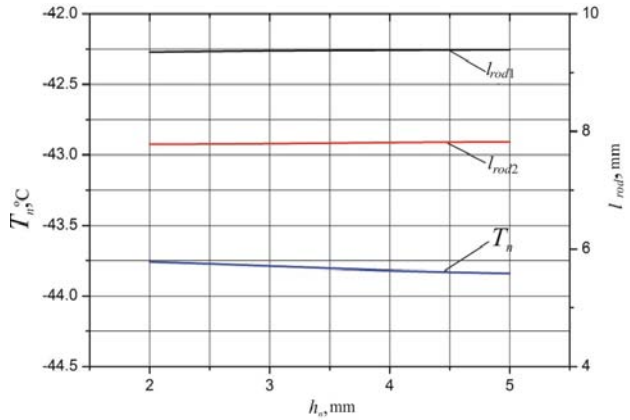

Fig. 12. Dependences of needle temperature (T_n) and freezing depth (h) on insulation thickness (d_n = 1 mm, h_n = 10mm, h_{rod} = 80 mm).

Dependences of needle temperature and freezing depth on insulation thickness are shown in Fig. 12. Rod diameter d_{rod} = 8 mm, rod length l_{rod} = 40 mm, needle diameter d_n = 2 mm, needle length h_n = 7 mm.

Dependences of needle temperature and freezing depth on needle diameter are shown in Fig. 13. Rod diameter d_{rod} = 8 mm, rod length l_{rod} = 40mm, insulation thickness h_{ins} = 5 mm, needle length h_n = 7 mm.

Dependences of needle temperature and freezing depth on needle length are shown in Fig.14. Rod diameter d_{rod} = 8 mm, rod length l_{rod} = 40 mm, insulation thickness h_{ins} = 5 mm, needle diameter d_n = 2 mm.

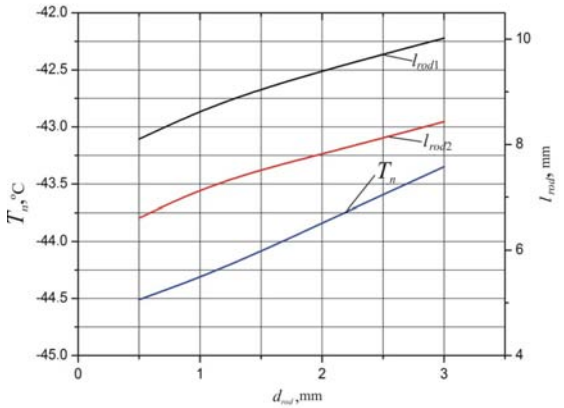

Fig. 13. Dependences of needle temperature (T_n) and freezing depth (h) on needle diameter (d_{rod} = 8, l_{rod} = 40, h_{ins} = 5, h_n = 7 mm).

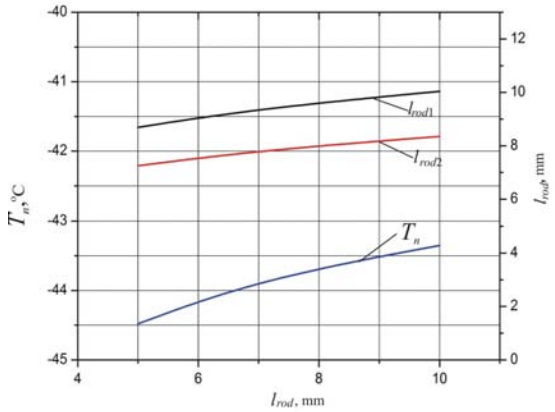

Fig. 14. Dependences of needle temperature (T_n) and freezing depth (h) on needle length (d_{rod} = 8, l_{rod} = 40, h_{ins} = 5, d_n = 2 mm).

The obtained results allow optimizing device design to achieve the required biological tissue freezing depth and maximum effect at destruction of oncologic neoplasms.

Conclusions

1. The physical, mathematical and computer models of local thermal effects on biological tissue have been constructed.
2. Computer simulation method has been developed, which allows predicting the results of local thermal effect on biological tissue at destruction of oncologic neoplasms.
3. Analysis of temperature distribution in biological tissue at cyclic cooling and heating effect has been performed. Optimal geometric size of work instrument to achieve maximum therapeutic effect has been determined.

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