The study of temperature fields of the WBC object in three-dimensional formulation using numerical methods

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Abstract. In this paper, we reviewed the temperature fields and heat flux emanating from the target tissues of the object. The influence of low temperatures on the integumentary tissues of the object with cryotherapy treatment on the given object or the whole-body cryotherapy (WBC) procedure was analysed. WBC is a method connected with the use of cryogenic gas for the treatment and prevention of certain diseases. The results that are shown in this article are derived using a mathematical model of heat exchange between the flow of a gaseous medium (liquid nitrogen vapor) and a biological object (an object of WBC impact) in a three-dimensional non-stationary formulation. The study and construction of the model was carried out using the finite element method. The computer model allows you to calculate the dynamics of the thermal field in the skin of the patient’s body during and after the WBC. Computer simulations in the future could increase both safety and comfort of cryotherapy.

1. Introduction

Thermal imaging is a harmless method of diagnosing various diseases, based on the registration of a person’s own infrared radiation in the form of a temperature field of the skin surface, which reflects the characteristics of the processes occurring in the body. The high efficiency of the method is confirmed by years of research and its widespread use in medical practice. The method can be recommended for inclusion in a national health project.

It is known that the temperature field on the surface of anybody (including on the human skin) reflects many internal processes occurring in the body. The first publication on the issue was a report by Dr. Ray Lawson in a medical journal in Canada and it was called “Implications of Surface Temperatures in the Diagnosis of Breast Cancer” [1]. However, the term “thermal imaging” itself, or rather “thermography”, appeared in the article of the same author in the following year, 1957, and the article was called “Thermography; a new tool in the investigation of breast damages” [2]. Since the 60s of the last centuries, the number of publications on the use of thermal imaging in medicine has grown like an avalanche [3-6].

Measurement of the temperature state of a biological object in conditions of extremely low temperatures is an important task. Particularly challenging is the determination of unsteady heat flux deriving from the epithelial tissues of an object at low temperatures. The solution to this problem can be applied in medical systems for monitoring and diagnosing various diseases.

One of the actual directions of heat metering is the measurement of the thermal state of the skin of the subject of cryotherapeutic effects. Cryotherapy, also called whole-body cryotherapy (WBC), is
performed in a cryogenic chamber or cryochamber at a temperature of cryogenic gas (from -110 to -140 °C), exposure varies from 1 to 3 minutes [7].

WBC is widely used as a universal treatment for several diseases, such as: rheumatoid arthritis, oncology, obesity, etc. [8-9]. The most striking manifestation of the positive effect of WBC is the drug suppression of pain for 6–8 hours, due to which this procedure is popular with professional athletes [10-11].

The practice of cryotherapy, including in Russia, has shown that the physiotherapeutic use of cold gas does not always give the necessary result. Empirical observations suggested the existence of a causal relationship between the curative effect of the WBC, primarily the duration of the analgesic action, and the technological options of the cooling process: gas temperature, contact time, cooling uniformity, contact area of the gaseous medium with the surface of the object.

For the cooling process, the terminal threshold for the skin surface is \( t_{\text{term}} = -2.5 \, ^\circ\text{C} \) [12]. The intensity of the stimulation of the thermoreceptors increases hyperbolically as the temperature of the skin approaches the threshold values:

\[
I = \frac{a}{(t_i - t_j)^n},
\]

where: \( t_i \) — epithelial surface temperature; \( t_k = t + 0.5 \, ^\circ\text{C}, t \) - rated temperature.

The signals of thermoreceptors integrated into the center of thermoregulation of the subcortical region of the brain are determined by the law of changes in the temperature of the surface of the epithelium. They can be calculated by the formula for the duration of the therapeutic effect of WBC:

\[
\tau^* = \int_{\tau=0}^{\tau^*} \frac{a}{(t_j - t_{sp})^n} \, d\tau,
\]

where \( n = 2 \) — constants that allow you to calculate the time of the healing effect and the effective time of the WBC in minutes.

The maximum exposure of WBC \( \tau_{\text{max}} \) depends on compliance with hypothermic safety conditions on the body surface \( T_i > 271 \, \text{K} \) and at the boundary of the fat and muscle layers \( T_{\text{con}} > T_{\text{con max}} -1 = 309 \, \text{K} \).

The WBC process should provide the optimal heat removal rate from the body surface, in which internal and external safety conditions are not violated. The intensity of convective heat removal is proportional to the difference between the temperature of the gas and the surface of the body. Due to the limitations resulted by the conditions of hypothermic safety, the surface temperature of the object varies relatively a few.

In this regard, the urgent task is to reduce the time that is taken to measure the temperature of biological objects without losing accuracy and working in low temperature ranges. The solution of the problem of reducing the measurement time to 0.1-1 seconds will expand the scope of application of the known method of thermal testing for assessing the state of an object at low temperatures.

There are difficulties in conducting physical studies of this kind, also because of the problems with the measurement of the instantaneous values of gas temperature. The use of traditional temperature sensors is difficult, since during a single WBC session, the gas temperature varies with an amplitude above 150 K and a rate of more than 1 K/s [13]. Thermal inertia of serial temperature sensors does not allow one to reliably determine the temperature of the gaseous medium or the surface of the patient's skin during the WBC procedure.

To measure the temperature fields of the cryotherapeutic impact object, we use the simulation of this process using the finite element method. Numerical simulation allows you to experiment with the modes of exposure to WBC without causing harm to the patient and analyze the optimal temperature regimes with maximum effect.

2. Simulation model

Computer modeling is of particular interest in conducting research, which allows to obtain the distribution of temperature fields and to establish the calculated values of the temperature of the skin surface in the process of WBC. The computer 3D-model was created by the finite element method. A
schematic representation of the geometry of the model is shown in Fig. 1a. The model was considered as time-dependent. The mesh consisted of 15226600 domain elements of triangular shape. The construction of the grid in the three-dimensional model is shown in Fig. 1b,1c. The cryochamber material has the following properties: \( \kappa = 0.04 \) [W/(m⋅K)]; \( \rho = 80 \) [kg/m\(^3\)]; \( C_p = 1470 \) [J/(kg⋅K)].

![Figure 1](image)

**Figure 1.** Schematic representation of the model and the division of the model into finite elements: a – general view of the model; b, c – general view of the model, which is divided into finite elements and a more detailed view.

The thermophysical properties of the layers forming the simulated object are listed in Table 1 [14].

| Organ or tissue | \( \rho \) [kg/m\(^3\)] | Water content, % | \( C_p \) [J/(kg⋅K)] | \( \kappa \) [W/(m⋅K)] | Heat dissipation, W/kg |
|----------------|--------------------------|------------------|-----------------------|----------------------|-----------------------|
| Epithelium     | 1093                     | 53,5 – 72,5      | 3600                  | 0,389                | 10,06                 |
| Muscle         | 1041                     | 68,5 – 80,3      | 3456                  | 0,439                | 6,99                  |
| Fatty tissue   | 916                      | 15 – 20          | 2250                  | 0,200                | –                     |

Bioheat Transfer module was used for mathematical description of biological tissues [15]. This module is used to model heat transfer in biological tissue. It considers such heat sources as: blood perfusion and metabolism, which is included in the classical heat transfer equation in the form of \( Q_{bio} \) [16].

A general heat equation for a solid:

\[
\rho \cdot C_p \cdot \frac{\partial T}{\partial t} + \nabla \cdot q = Q + Q_{bio},
\]

where \( Q_{bio} \) is the heat release of biological tissues:

\[
Q_{bio} = \rho_b \cdot C_{p,b} \cdot \omega_b (T_b - T) + Q_{met},
\]

where \( \rho_b \) - blood density [17], \( C_{p,b} \) - specific heat of blood at constant pressure, \( \omega_b \) - blood perfusion rate [18], \( T_b \) - arterial blood temperature, \( Q_{met} \) - metabolic heat source.
The general heat equation for a liquid (in a gas N\textsubscript{2}):

\[ \rho \cdot C_p \frac{\partial T}{\partial t} + \rho \cdot C_p \cdot \mathbf{u} \cdot \nabla T + (\nabla \cdot \mathbf{q}) = Q, \]  

where \( \rho \) - density, \( C_p \) - heat capacity, \( T \) - absolute temperature, \( t \) - time, \( \mathbf{u} \) - velocity vector, \( \mathbf{q} \) - heat flux, \( Q \) - heat sources.

The heat flux density:

\[ \mathbf{q} = -\kappa \cdot \nabla T, \]  

where \( \kappa \) - a coefficient of thermal conductivity.

In this paper, the gas flow was specified as a turbulent flow of the k-\( \varepsilon \) model. This model is one of the most used models of turbulence for industrial applications. This module includes the standard model k-\( \varepsilon \) [19]. The model introduces two additional transport equations and two dependent variables: turbulent kinetic energy, \( k \) and turbulent dissipation rate, \( \varepsilon \). Turbulent viscosity is modeled as:

\[ \mu_t = \rho C_{\mu} \frac{k^2}{\varepsilon}, \]  

where \( C_{\mu} \) is a model constant.

The transport equation for \( k \) reads:

\[ \rho \cdot \frac{\partial k}{\partial t} + \rho \mathbf{u} \cdot \nabla k = \nabla \left[ (\mu + \frac{\mu_t}{\sigma_k})\nabla k \right] + P_k - \rho \varepsilon, \]  

where the production term is:

\[ P_k = \mu_t \left[ \nabla \mathbf{u} \cdot \left( \nabla \mathbf{u} + (\nabla \mathbf{u})^T \right) - \frac{2}{3} (\nabla \cdot \mathbf{u})^2 \right] - \frac{2}{3} \rho k \nabla \cdot \mathbf{u}. \]  

The transport equation for \( \varepsilon \) reads:

\[ \rho \cdot \frac{\partial \varepsilon}{\partial t} + \rho \mathbf{u} \cdot \nabla \varepsilon = \nabla \left[ (\mu + \frac{\mu_t}{\sigma_\varepsilon}) \nabla \varepsilon \right] + C_{\varepsilon_1} \frac{\varepsilon}{k} P_k - C_{\varepsilon_2} \rho \frac{\varepsilon^2}{k}, \]  

where \( \mathbf{u} \) is the velocity, \( \mu_t \) is the eddy viscosity.

The model constants in Eq. (7), Eq. (9), and Eq. (10) are determined from experimental data [19] and the values are listed in Table 2.

| Table 2. Model Constants |
|--------------------------|
| Constant     | Value  |
| \( C_{\mu} \)  | 0.09   |
| \( C_{\varepsilon_1} \) | 1.44   |
| \( C_{\varepsilon_2} \) | 1.92   |
| \( \sigma_k \)  | 1.0    |
| \( \sigma_\varepsilon \) | 1.3    |

3. Results and Discussion

Computer simulation of the cooling process of the patient’s body surface using WBC in a gaseous medium was carried out in the time interval from 0 to 180 seconds, the next interval from 180 to 1500 seconds was simulated to restore the human body temperature to normal temperatures in room conditions.

The dependence of temperature change in different layers of biological tissue in the time interval from 0 to 180 seconds and the subsequent interval from 180 to 1500 seconds of a cryotherapy session is shown in Fig. 2-3. The temperature of the outer surface of the body depends on the intensity of convective heat removal to the environment. It is believed that under comfortable conditions the temperature of the skin surface is 32 °C. When exposed to low temperatures, irreversible tissue damage occurs when the temperature of the epidermis reaches \( t = -2 \) °C and the temperature of the core tissues decreases by more than 1 °C [20].
As far as we know from sources, the functioning of any living system occurs under conditions of continuous exchange with the environment, matter, energy, impulse, which is accompanied by a change in the physiological state of the living system [21]. A characteristic reflection of bioenergy processes is the heat flux from the surface of human skin. In our work, we also reflected the heat flux from the surface of the human body, and showed in Fig. 3, curve 2.

Figure 2. Time dependency graph: 1 – the temperature value in the epithelium, 2 – the value of the minimum temperature in fatty tissue.

Figure 3. Time dependency graph: 1 – the value of the average temperature in fatty tissue, 2 – heat flux from integumentary tissues.

4. Conclusion
In this paper, a 3D model was simulated, which is a structure of a layered biological tissue that interacted with a cryogenic gas. The WBC procedure that will provide an opportunity to create a convenient platform to study the effect of cryotherapy options (coolant speed and temperature, session exposure, etc.) on the thermal processes that take place in the tissues of the human body is modeled. In the present work, it was shown that a decrease of temperature on the upper surface of the epithelium (Fig.2-3) during the WBC time period (from 0 to 180 s) leads to an increase in the heat flux from the target tissues of the object (Fig.3). These processes confirm the theory of the therapeutic properties of cryotherapeutic effects.
The increase in temperature is due to increase blood circulation caused by exposure to extremely low temperatures on the surface of the human body, which confirms the developed model. In the future, this 3D model will make it possible to determine the nature of the dependence of the temperature of each layer on the time of the WBC cryosecession and the time it takes for the thermal field of the object to return to its normal state. The magnitude of the temperature change $\Delta T$ will vary depending on the feed rate and nitrogen temperature, session time and individual patient characteristics, which this model does not take into account. These results can be used to determine the duration and degree of cryotherapeutic effect, and further modernization of cryosauna.

5. Conflict of interest
The authors declare that they have no conflict of interest on the content of this paper.

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