Consideration of the effectiveness of cryotherapy for different thicknesses of the skin using numerical modelling

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Abstract. In this paper, a method for measuring the temperature of the integumentary tissues of a person during a general cryotherapy treatment (WBC) is considered. The authors investigated the physical processes in the device for cryotherapeutic effects on the human body (cryosauna). Also we constructed the computer model of the object of study, using the finite element method. The mathematical model allows you to calculate the temperature of the integumentary tissue during the WBC procedure and at its end. The dependence of the intensity of heat loss on the thickness of the covering tissues was studied. The Bioheat Transfer module was used, which allowed taking into account thermal effects. Recommendations on the choice of cryosauna modes are formulated. Research aims to improve the efficacy and safety of WBC due to the individual characteristics of patients.

1. Introduction

The technology, which is carried out by cryostimulation of thermoreceptors of the skin by cryogenic gas, is called Whole Body Cryotherapy (WBC). In the Russian Federation, this method is called cold therapy or cryotherapy [1]. The WBC is widely used in various industries, including sports, cosmetics and medicine. The operation and implementation of the WBC process requires special cryogenic devices (cryosauna or cryochamber) that ventilate the patient's cabin with cryogenic gas with a temperature of -140 to -110 °C. The exposure of the cooling system is 1 to 3 minutes [2-3]. There are cabins for patients of various designs. Analysis of the effectiveness of these structures was carried out in [4]. The WBC procedure starts the mechanisms of self-testing and correction in the human body, stimulates the improvement of metabolic processes in injured organs, and speeds up the process of treatment [5-8]. Cryotherapy is widely used as a universal remedy for the prevention of socially significant diseases, oncology, rheumatoid arthritis, pollinosis, osteoporosis, etc. [9-11]. This determines the high social significance, providing medical institutions in Russia with efficient and safe equipment for the WBC.
2. Problem definition

In accordance with the WBC thermophysical theory [1], to achieve the maximum therapeutic effect of
the procedures, it is necessary to cool the surface of the patient's skin to a temperature \( t_{\text{term}} = -2 \) °C. Cooling the surface below \(-2.5\) °C will cause irreparable damage to the skin of the WBC object [12-13]. Given there is risk of frostbite on the surface, the task of controlling the temperature of the skin of an
object during the WBC is extremely relevant. There is a risk of local overcooling of the skin of the WBC
object due to the non-uniform distribution of the cryogenic gas flow in the cryochamber. The mode of
supply of cryogenic gas to the WBC zone is designed for a patient with average body characteristics;
this ensures the safety of the majority of patients but can significantly reduce the subjective effectiveness
of the procedure. Since patients have skin of different thickness. Analysis of the effects of WBC
exposure on the skin of various thicknesses will allow the development of special modes of gas supply
to the cryochamber.

Physical studies of this kind are difficult to organize, including due to problems with the measurement
of instantaneous gas temperature values. The use of traditional temperature sensors is difficult, since
during a single WBC session, the gas temperature varies with amplitude of more than 150 K and a rate
of more than 1 K/s [14]. 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. You can analyze changes in the temperature of the patient's skin using computer simulation.

3. Description of the computer model

Computer simulation of the WBC procedure allows determining the optimal temperature conditions of
the procedure. The main interest is the temperature distribution over the thickness of the skin and along
the axis of gas movement, therefore a two-dimensional array of finite elements is formed in the
mathematical model. The cross-section of the research object (cryochamber with the patient) is shown
in Fig. 1. In this study, the thickness of the subcutaneous layer (Fig. 1, position 7) varied from 10 mm
to 30 mm in 5 mm increments. The WBC object is represented as a system of ellipses that mimic the
layer of muscle (Fig. 1, position 6) and adipose tissue (Fig. 1, position 7).

![Figure 1](image.png)

Figure 1. A cryochamber scheme: 1 – air outlet, 2 - air
inlet, 3 – chamber, 4 - wall, 5 – free air outlet into the
ambient, 6 - muscular layer, 7 – subcutaneous layer
(varies during the experiment)

Numerical analysis of the heat flux from the body surface and temperature fields over the thickness of
the skin of the WBC object is performed by using the finite element method. The simulation the object
described the grid, which on average consisted of seven thousand elements of a triangular shape and
more than seven hundred boundary elements.
Cryogenic gas was continuously supplied from channel 2 (see Fig. 1). The values of temperature and velocity of cryogenic gas are reduced in the form of Table I. After completion of the procedure, the chamber was filled with atmospheric and air velocity 0.2 m/s. The properties of the cryogenic gas and the properties of the cryochamber material were taken from [2]. The \( \text{N}_2 \) gas was inflated from the hole shown in Fig. 1 (pos. 2), the cryogenic gas blowing was continued for 160 seconds, and the boundary conditions described the gas flow at the inlet and its temperature (see Table I). After passing the flow of nitrogen, starting from 160 seconds of calculation, the chamber was filled with air. The air flow had a room temperature and a velocity of 0.2 m/s, which corresponded to the presence of a person in the room, detailed characteristics of the velocity and temperature of the gas [2]. Gas N2 properties are presented in [4]. The cryochamber material has the following properties: \( \kappa = 0.04 \text{ W/(m} \cdot \text{K)}; \rho = 80 \text{ kg/m}^3; \text{C}_p = 1470 \text{ J/(kg} \cdot \text{K)}.

The physical properties of the skin layers of the WBC object used in this model are listed in Table 1, determined by literature data [15].

Table 1. Dimensions of the layers and material properties used in modelling.

| №  | Type of fabric            | Thickness, [mm] | \( \rho [\text{kg/m}^3] \) | \( \kappa [\text{W/(m} \cdot \text{K)}] \) | \( \text{C}_p [\text{J/(kg} \cdot \text{K)}] \) |
|----|---------------------------|-----------------|-----------------------------|--------------------------------------------|-----------------------------------------------|
| 1  | Muscle layer              | 30              | 1041                        | 0.439                                      | 3456                                          |
| 2  | Subcutaneous fat layer    | 10;15;20;25;30  | 916                         | 0.200                                      | 2250                                          |

To describe biological processes, assumptions are introduced, and the bioheat equation is solved using the Penns approximation (Pennes). The Penns approximation is described in more detail in [16]. This approximation is used to simulate the transfer of heat in biological tissue, considers the sources of heat from blood perfusion and metabolism in the classical heat transfer equation. Bioheat Transfer module was used for mathematical description of biological tissues [17]. 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 \( Q_{\text{bio}} \) [18].

The general heat equation for a solid:

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

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

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

where \( \rho_b \) - blood density [19], \( C_{p,b} \) - specific heat of blood at constant pressure, \( \omega_b \) - blood perfusion rate [20], \( T_b \) - arterial blood temperature, \( Q_{\text{met}} \) - metabolic heat source.

The general heat equation for a liquid (in a gas \( \text{N}_2 \)):

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

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 \( k-\varepsilon \) [21]. 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:

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

where the production term is:

$$P_e = \mu_t \left[ \nabla \mathbf{u} \cdot \left( \nabla \mathbf{u} + (\nabla \mathbf{u})^T \right) - \frac{2}{3} (\nabla \cdot \mathbf{u})^2 \right] - \frac{2}{3} \rho \kappa \nabla \cdot \mathbf{u}$$

The transport equation for $\varepsilon$ reads:

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

where $\mathbf{u}$ is the velocity, $\mu_t$ is the eddy viscosity. The model constants in Eq. (5), Eq. (6), and Eq. (8) are determined from experimental data [21].

4. Results and Discussion

In a numerical analysis of the effectiveness of cryotherapy of the whole body, thirty models were modeled and analyzed. Fig. 2 shows the distribution of the velocity of a cryogenic gas under different operating conditions of the cryochamber at times when the gas velocity reaches its maximum value (Fig. 2, a), as well as with a decrease in gas supply (Fig. 2, b). The increase in the rate of cryogenic gas at the exit to the cryochamber causes an increase in the velocity of nitrogen vapor along the entire perimeter in the cross section of the cryochamber.

Fig. 3 shows the dependence of the temperature of the skin of the object WBC in the time interval from 0 to 2000 s. Intensive cooling of the skin of the WBC object occurred in the time interval from 0 to 180 s.

Achieving maximum exposure WBC $\tau_{\text{max}}$ is limited by the patient's safety conditions [15]. Hypothermic safety conditions are limitations of the allowable temperature drop on the patient's body surface $T_{i, >} > 271$ K and at the boundary of the fat and muscle layers $T_{i = n} > T_{i = n_{\text{max}} - 1} = 306$ K [1]. From the external condition of the hypothermic patient safety, it follows that the surface temperature of the object can vary relatively in a small temperature range: $271 \leq T_0 \leq 306$ K, i.e. $\Delta T_0 \leq 36$ K.

Explanatory notes to Fig. 3: c - the time dependence of the temperature of the subcutaneous layer (varies during the experiment); d - time dependence of the temperature of the muscle layer; 1 - thickness of subcutaneous layer h = 10 mm; 2 - thickness of subcutaneous layer h = 15 mm; 3 - thickness of subcutaneous layer h = 20 mm; 4 - thickness of subcutaneous layer h = 25 mm; 5 - thickness of subcutaneous layer h = 30 mm; 6 - critical temperature of the layer of the skin due to hypothermic safety.
Figure 2. Cryogenic gas velocity distribution depending on the mode: a - t = 160s.; b - t = 200s.

Figure 3. Temperature distribution of the skin over time, depending on the mode of operation of the cryochamber.

Based on the conditions of hyperthermal safety, we can conclude that the first and the second cryotherapy regimen contraindicated this exposure to a group of patients with a skin thickness of 10 mm (see Fig. 3, modes 1-2).

Simulation of three modes of the cryotherapy process was considered, in which the maximum gas velocity reached 18 m/s (Fig. 2). The maximum cryotherapeutic effect is obtained for patients with a skin thickness of 25–30 mm (see Fig. 3, mode 2). However, a group of patients with a skin thickness of 10–20 mm would have received critical damage to the skin if they had taken cryotherapy procedures with the third mode of operation selected.
To assess the heat load on the cab cryostat system, it is important to determine the heat flux from the surface of the skin of the WBC object. Fig. 4 presents the results of calculating the magnitude of the heat flux.

![Figure 4](image)

**Figure 4.** Distribution of the density of heat flux emanating from the surface of the skin over time, depending on the operating mode of the cryochamber: 1 - thickness of the subcutaneous layer $h = 10$ mm; 2 - thickness of subcutaneous layer $h = 15$ mm; 3 - thickness of subcutaneous layer $h = 20$ mm; 4 - thickness of subcutaneous layer $h = 25$ mm; 5 - thickness of subcutaneous layer $h = 30$ mm.

5. Conclusion

To achieve the maximum cryotherapeutic effect for patients with skin thickness of 25-30 mm, the third cryotherapy regimen was chosen (Fig. 3, mode 2). However, from the graph (Fig. 3, mode 2), we can conclude that the hypothermic safety condition is not satisfying for patients with skin of 10-20 mm. This group of patients with a skin thickness of 10-20 mm would receive critical damage to the skin if they took cryotherapy procedures with the selected third mode of operation. These results prove that for each group of patients, individual cryotherapy regimens should be developed to achieve the maximum cryotherapeutic effect without causing harm to the patient.

Fifteen models of the cryochamber were used in the studies. The process of cryotherapeutic effects was modeled using the finite element method. The conditions that provide a high physiotherapeutic effect while observing hypothermic safety conditions are shown. This makes it possible to make recommendations on the choice of the mode of gas movement in a cryochamber, considering the patient’s subjective indicators.

Individual patient characteristics were not considered. The loss of heat flow depends on the degree of cooling of the layers of human skin, and the degree of cooling, in turn, depends on the design of the cryochamber wall. By changing the parameters of the cryochamber and considering the individual characteristics of the patient (metabolic rate, skin thickness, thick layer, muscle tissue, height, age and vascular status), individual recommendations can be developed to achieve the maximum analgesic effect.

6. Conflict of interest

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

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