Chapter

Technique and Technology of Whole-Body Cryotherapy (WBC)

Alexander Baranov, Oleg Pakhomov, Alexander Fedorov,
Vladimir Ivanov, Andrew Zaitsev and Ruslan Polyakov

Abstract

Whole-body cryotherapy (WBC) is a highly effective treatment method of a number of serious diseases. The therapeutic effect of WBC is achieved by stimulating cold receptors of the patient's skin, which provide supercooling of the skin surface to the level of $-2°C$. To achieve such a temperature of the skin surface, it is necessary to ensure heat removal with intensity not less than 3500 W/m^2. Such a heat flux can remove gas with the temperature not higher than $-130°C$. Procedures lasting less than 2 minutes do not form therapeutic effect. Procedures lasting more than 3 minutes are dangerous for the patient's health. WBC procedures are carried out in single- and multi-seat devices. Due to the compact placement of the patient in the WBC area, the share of useful heat load on the cryostatting system is up to 70%. In multi-seat installations, the useful heat load share is not more than 50%. During the WBC procedure, consumption of liquid nitrogen per patient is 3.77 kg. For the effective use of WBC technology, it is necessary to determine the general requirements for the power of cooling systems and the temperature of cryostatting of the WBC area.

Keywords: whole-body cryotherapy, WBC, supercooling of the skin surface, heat removal, useful heat load, WBC minimum procedure, cooling gas temperature

1. Introduction

Equipment for whole-body cryotherapy (WBC) has been used in clinics around the world for over 40 years [1, 2]. Despite this, until today there is no universally accepted concept describing the mechanism for achieving the healthcare effect of this physiotherapeutic procedure, and the physical conditions of safety and effectiveness of cryogenic cooling of the patient's skin surface have not been determined [3–6]. Temperature of the cooling gas and the duration of its contact with the patient's skin, being the most important technological parameters of WBC, vary over a wide range. The requirements for the power supply capacity of equipment for the implementation of WBC technology are not defined. In such conditions, manufacturers of devices for WBC procedures gradually increase the value of the minimum gas temperature in the WBC cab. Over 40 years of cryotherapeutic system production, the gas temperature declared by manufacturers of devices for WBC has doubled from 98 K in 1978 [1, 2] to 192 K [4–7]. By increasing the operating temperature of the equipment, manufacturers significantly reduce the cost of its production. For 40 years, the cost of devices for group WBC has

decreased by 30 times. Low prices for equipment provide a high level of sales, so the trend of increasing operating temperature of WBC devices persists. An increase in the temperature level is accompanied by a decrease in the power of systems for cryostatting the WBC zone. The newest installations are equipped with refrigerators with a specific power of the electric driver of not more than 1 kW/m³. At a temperature level of 170 K, a refrigerator with such a power has a heat-removing capacity of not more than 400 W/m³, which is comparable with the physiological heat release of a patient under thermal comfort conditions (150 W) [7].

Unreasonable changes in WBC technology affect the effectiveness of the procedures. Recently, more and more articles appear, the authors of which express doubt that cryotherapy can provide the healthcare effects described in papers published before 1990 [7, 8]. The reason that many modern WBC systems are not able to provide the conditions for obtaining the healthcare effects described in the last century [1, 2] is the increase in gas temperature in the working zone of new installations. This can be seen even from the titles of the articles [1, 8]. The temperature increase from $-170°C$ (102 K) to $-110°C$ (163 K) changes the absolute value of the temperature by 1.6 times, which cannot but affect the intensity of heat removal, the degree of supercooling of the patient's body surface, etc. From a thermophysical point of view, it is obvious that from 1978 to 2018 the technology, which is commonly referred to as WBC, has qualitatively changed. And, judging by contemporary publications, this qualitative change had a negative impact on the healthcare effectiveness of the procedures, which until recently were successfully used to treat a number of severe diseases: rheumatoid arthritis, bronchial asthma, psoriasis, etc. [9, 10].

In such conditions, the determination of cause–effect relationships between the WBC technological parameters and the magnitude of the healthcare effect acquires high scientific and social significance. Formation of the thermophysical theory of WBC creates a scientific basis for restoring the production of effective cryotherapeutic installations at the modern technical level.

2. Historic reference

The WBC method is based on providing the total contact of the patient's skin surface with a cryogenic gas. With a contact duration of up to 3 minutes and a gas temperature of less than 140 K, the WBC procedure provides a number of positive effects that are used in treatment practice [11, 12]. The most demonstrative and controlled sign of the WBC effectiveness is the duration of analgesic action, which can last 6–8 hours [7]. The analgesic effect of WBC was first described and used in treatment practice by a Japanese doctor Yamauchi. For the WBC procedures, a special installation was made, called "Cryotium" by the author of the method [2]. The "Cryotium" design was analogously a refrigeration chamber for long-term storage of perishable products. The chamber was separated from the environment by the lock chamber (**Figure 1**), which was supposed to reduce the loss of cold air from the main chamber.

Given the relatively large size of the chamber, several patients were undergoing the WBC procedures simultaneously (from 5 to 12). To obtain cryogenic temperatures, liquid nitrogen was fed to the "Cryotium" heat exchangers instead of freon. The "Cryotium" design appeared by chance. Yamauchi believed that in order to obtain the maximum treatment outcome, the maximum decrease in temperature should be used. This condition became the basis of the design. To reduce the cost of manufacturing "Cryotium," Japanese engineers used the insulating structure and heat exchangers of the serial refrigeration chamber. The temperature regime in the chamber volume was

Figure 1.
Multi-seat cab for WBC.

determined by the requirement of the inadmissibility of air condensation on the surface of a heat exchanger. The temperature of the outer surface of the heat exchanger T_{HC} must be higher than the condensation temperature T''_A:

$$T_{HC} > T''_A = 81 \ K. \tag{1}$$

The removal of heat from air to the surface of the heat exchanger was carried out by natural convection.

With natural convection, the calculated temperature gradient between the gas and the heat-removing surface is 20 K:

$$T_{A-HC} = T_A - T_{HC} \approx 20 \ K. \tag{2}$$

Minimum possible air temperature in the cab:

$$T_A = T''_A + \Delta T_{A-HC} \approx 101 \ K \left(-172 °C\right). \tag{3}$$

The value of the air temperature during the WBC sessions specified by the method's author [1] is the lowest possible temperature that could be achieved in this cab design. It is important to note that in "Cryotium" the temperature was maintained by choosing the pressure of liquid nitrogen (LN) vapor in the heat exchanger tubes (**Figure 2**). The boiling point of LN depends on pressure; by increasing the vapor pressure to a level of $P \geq 0.2$ MPa, it is possible to ensure the fulfillment of condition (1) without using the temperature control systems. The lack of a temperature control system has provided "Cryotium" with unique operational advantages over modern WBC devices. Heat exchangers filled with liquid nitrogen successfully dealt with an increase in heat load when patients entered, and the correct choice of operating pressure prevented air condensation.

Thus, the "Cryotium" design determined the WBC technology. Perhaps, that is why the author of the method did not give any reason for the WBC temperature regime in his works. The ratios of the boiling points of nitrogen and air, as well as the design features of the device in which the procedures were performed, have randomly created the conditions for a safe and highly efficient procedure.
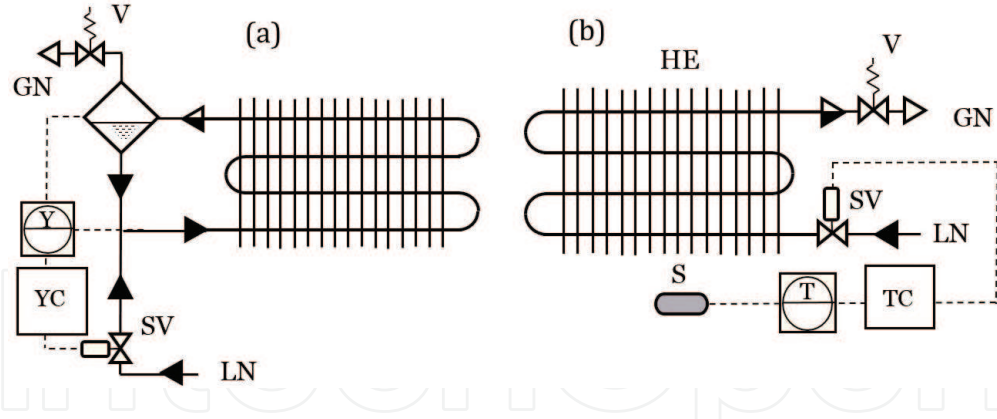
Figure 2.
The scheme of supplying liquid nitrogen to the heat exchangers of the installations for the WBC: (a) "Cryotium";
(b) "KR-2005."

Yamauchi used the WBC method for the treatment of rheumatoid arthritis [1]; the technique was so effective that it quickly spread to the countries of Western Europe. In Poland and Germany, devices similar to "Cryotium" were put in production. European manufacturers have tried to reproduce the Japanese installation on the base of available information but, for unknown reasons, have changed the basic operating principle of "Cryotium." This is translated into an increase in the minimum operating temperature from −170 to −160°C [13]. A slight increase in temperature led to a whole chain of changes in cooling technology, which caused a gradual decrease in the efficiency of European (Polish and German) installations for WBC. As already shown above, the temperature level of −170°C was maintained in "Cryotium" without using a temperature control system, only through relief of excess pressure in the LN vapor line (**Figure 2a**).

The liquid level controller (YC) in the heat exchangers of the "Cryotium" installation according to the sensor signals of the Y level controls the operation of the solenoid valve (SV), through which LN enters the system. The cryogenic liquid enters the heat exchanger (HE) tubes, where it partially evaporates due to the supply of heat from the procedural room air. Vaporization reduces the flow density in the tubes; the vapor–liquid mixture is pushed out from the top of the heat exchanger to the liquid separator (LS). In this apparatus, liquid and vapor are separated. The liquid flows into the lower section of the heat exchanger (HE) and again participates in the removal of heat. LN vapors accumulate at the top of the liquid separator (LS). The vapor pressure is controlled by a safety valve (V), which opens at a pressure of 0.22 MPa. The vapor pressure determines the LN boiling point and the temperature of the tubes of the heat exchanger (HE), which must meet condition (1). The air temperature in the main "Cryotium" cab at the presence of patients rises to −170°C. In the pauses between the procedures, when the heat load on the cooling system is reduced by 10 times [7], the air temperature in the main cab approaches the temperature of the heat exchanger tubes:

$$\Delta T_{A-HC} \rightarrow 5 \text{ K}; \quad T_A \rightarrow T_A^{''} + \Delta T_{A-HC} \approx 86 \text{ K} \left(-187^\circ C \right). \tag{4}$$

The air temperature in the cab remains at the minimum possible level. In European installations, the air temperature in the cab is controlled by the temperature controller (TC), which, by signals from the temperature sensor (S), opens the liquid nitrogen supply valve (SV). In order to maintain the temperature at −160°C, the TC limits the supply of LN to the heat exchanger (HE) in the period when there are no patients in the cab. The LN level decreases until the sensor (S) registers the set

temperature. The upper sections of the heat exchanger (HE), through which the nitrogen passes in the vapor state, are heated to a temperature close to the air temperature in the chamber. The temperature of the inner tube surfaces exceeds the LN boiling point by more than 20 K:

$$T_{HC} \rightarrow T_A \approx 110 \ \text{K}; \quad T_{HC} > T_A^{''} + 20 \ \text{K}. \tag{5}$$

When patients enter the main cab, relatively warm air enters from the lock chamber ($T_s = 210$ K). Because of this, the air temperature in the cab increases by 50 K or more. The temperature controller opens the SV and resumes the supply of LN to the heat exchanger (HE). However, it takes some time to fill the tubes of the heat exchanger; moreover, the temperature of the upper sections of the heat exchanger (HE) exceeds the boiling point of nitrogen by more than 20 K. Because of this, the LN film boiling occurs at which the intensity of heat removal to the liquid is much lower; therefore, the vapor–liquid mixture passes the heat exchanger (HE) and is discharged through valve (V) into the environment. Under such conditions, the heat exchanger does not cope with heat generation from the surface of the patient's bodies and cannot restore the specified temperature mode of the chamber until the patients enter the lock chamber. The WBC procedure takes place at a temperature significantly higher than the nominal −160°C [7]. After the patients leave, the thermal load on the cryostatting system is reduced 10 times, the air temperature drops to the nominal level, and the temperature controller stops the supply of LN to the heat exchanger (HE).

The increase in the air nominal temperature in the cab for WBC from −170°C to −160°C fundamentally changed the temperature algorithm of the procedure. The transition to the LN film boiling regime caused a significant overrun of the cryoagent. The operational drawbacks of the nitrogen cooling system and the uncertainty of the air temperature requirements in the main procedural cab created conditions for use in the WBC cryostatting system of three-stage chillers and steam cycles on gas mixtures. Refusing LN resulted in an increase in the nominal temperature in the main cab to −110°C.

Specialists in the field of WBC did not only pay attention to this but also actively promoted the "modernization" of cryotherapy equipment [13–15]. The ability to refuse to use LN and significantly reduce the costs of WBC procedures turned out to be so attractive that the specialists "did not notice" that the efficiency of the procedures in the "nitrogen-free" installations was 10 times lower than in "Cryotium" [7]. At the beginning of the twenty-first century, "Criohome" "cryotherapeutic" devices with a nominal temperature of −85°C was used for WBC procedures, i.e., the tendency to increase the temperature persists. Since 1985, the Russian direction of devices for WBC has been developing independently, based on the use of single-seat installations with a nitrogen cooling system (cryosaunas). The temperature in the cab of a single-seat cryosauna during the whole procedure is no higher than −130°C. The conditional constancy of temperature fundamentally changes the degree of supercooling of the skin surface; therefore, cryosaunas ensure the effectiveness of WBC at the level of the original technology implemented in "Cryotium" but with less energy loss. The current state of WBC in Europe is a consequence of the 40-year use of the method in the absence of a reliable concept of the method for obtaining the cryotherapeutic effect and the uncertainty of the technological requirements to specialized equipment [3–5]. In such conditions, manufacturers of WBC installations have flooded Europe with installations that, by their therapeutic efficacy, do not differ from traditional hypothermia. The popularization of the thermophysical theory of WBC will stop the regression of cryotherapy in Europe and the world.

3. Thermophysical theory of WBC

The WBC thermophysical theory was formulated at St. Petersburg National Research University of Information Technologies, Mechanics and Optics (ITMO University) in order to overcome the uncertainty of the technological requirements for specialized devices for WBC. In developing the theory, information on the conduct of WBC procedures and their effectiveness was used [1, 2, 7, 13]. As a criterion for optimizing the WBC technology, it is reasonable to use the duration of the analgesic effect of cryotherapy. The duration of the analgesic effect or effective time (WBC ET) is easy to determine in practice. To carry out computational experiments at ITMO University, a method was developed for calculating the WBC ET [16], which made it possible to perform studies on the optimization of the WBC technology in the mode of numerical experiment.

To calculate the WBC ET, a formula is proposed that relates the positive effect with the degree of approaching the skin surface temperature (T_s) to the temperature of the cryogenic damage onset (T_{cr} = 270.5 K), as well as with the area (f_s) and the duration (τ_{max}) of body surface contact with cryogenic gas:

$$\tau_E = f_s \int_{\tau=0}^{\tau_{max}} \frac{A}{(T_s - T_{cr})^2} \partial\tau, \tag{6}$$

where A is an empirical constant, providing the calculation of the WBC ET in seconds, A = 1200.

The maximum duration of WBC (τ_{max}) is determined taking into account the requirements of the patient's hypothermic safety, which limit the permissible changes in body temperature on the surface (T_s) and on the inner boundary of the subcutaneous fat layer (T_f) (**Figure 3**).

Compliance with the established limitations of changing the value of T_s and T_f protects the patient from the danger of frostbite and frigorism, respectively:

$$T_s \geq 271 \text{ K } (-2°C); \ T_f > 309 \text{ K } (36°C).$$

Figure 3 shows a graphical representation of the patient's body shell (BS). BS is the outer layer of the body, the mass of which is 30% of the total body mass. BS consists of three types of tissues: epithelium 1, adipose tissue 2, and muscle tissue 3.
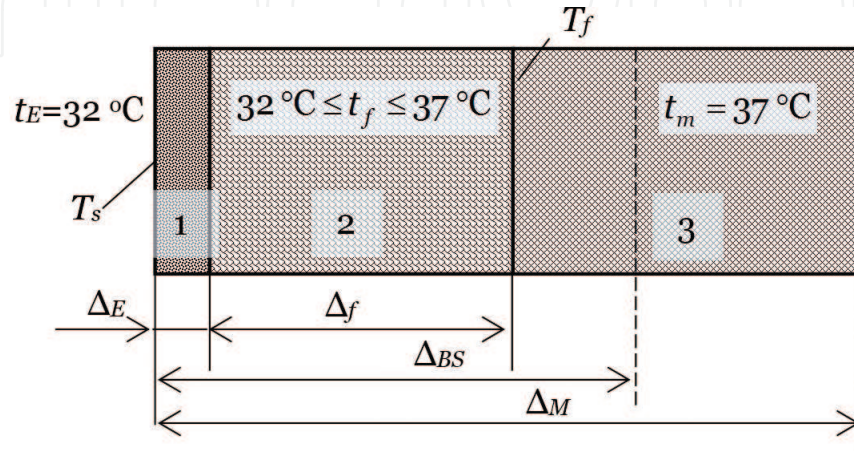

Figure 3.
Physical model of the human body shell (BS).

Layers 1 and 2 endure significant hypothermia without any harm; patient's safety is ensured when the violation of the normal temperature distribution does not extends beyond the inner boundary of the BS [13]. In the normal state $t_E = 32°C$, within the fat layer, the BS temperature rises $32°C \leq t_f \leq 37°C$; the temperature of the muscle layer is equal to the human body core (BC) temperature $t_m = t_{BC} = 37°C$. It has been assumed that the thickness of layers 1 and 2 for an average patient has the following values: $\Delta_E = 2$ mm; $\Delta_f = 2$ mm.

The simplified physical model of a human BS has become the basis of a mathematical model.

4. Mathematical model of the human body shell

BS is a relatively thin surface layer. The calculated thickness of the BS of an average human is $\Delta_{BS} = 16$ mm. In this case, the effective diameter of the body is 280 mm [7]. This allows us to describe the processes of heat transfer through the BS tissues by one-dimensional energy equation:

$$\rho \frac{\partial h}{\partial \tau} = \frac{\partial q_x}{\partial x} + q_v,$$ (7)

where h means the tissue enthalpy, kJ/kg; τ means the time, sec; q_x means the heat flux through BS, W/m^2; x means the coordinate along which heat is transferred, m; and q_v means the heat of metabolism, W/m^3.

The energy equation allows you to simulate thermal processes associated with significant changes in the temperature of the study object. The energy equation describes well the processes of changing the state of aggregation; this provides the mathematical model with certain advantages compared to models built based on the heat transfer equation [13]. When replacing derivatives with difference approximants, it is possible to obtain an algebraic expression suitable for automated calculations:

$$\rho \frac{\Delta h}{\Delta \tau} = \frac{\Delta q_x}{\Delta x} + q_v.$$ (8)

Solving Eq. (8) with respect to the value of enthalpy at the new time layer h', we get:

$$h' = h + \frac{(\Delta q_x + q_v \Delta x)\Delta \tau}{\Delta x \rho},$$ (9)

where $\Delta h = h' - h$, h means the substance enthalpy value on the current time layer; Δx means the step of area elements along the x coordinate, m; and $\Delta \tau$ means the time step. Eq. (9) allows you to calculate the enthalpy of the nodal points of the simulated object under known boundary and initial conditions.

The structure of the cooling object and temperature distribution in the BS layers are described above (**Figure 3**). Thermophysical properties of the human body shell tissues are shown in **Table 1**.

Due to the high water content (φ), all human BS tissues have a high heat capacity, which ensures the accumulation of a significant amount of heat. The heat accumulated in the tissues protects the organs of the body core (BC), heart, lungs, kidneys, and liver from frigorism at a sharp decrease in ambient temperature. Thermal balance of the BS area element

Tissue	ρ, kg/m^3	φ, %	c, J/(kg·K)	λ, W/(m·K)	q_v, W/m^3
Skin	1093	72.0	3600	0.35	10,996
Muscles	1041	80.0	3458	0.475	7277
Adipose tissue	916	20	2250	0.21	—

Table 1.
Properties of the human BS tissues [17].

$$\Delta h = h' - h = \frac{(\Delta q_x + q_v \Delta x)\Delta \tau}{\Delta x \, \rho}, \tag{10}$$

determined by the ratio of the intensity of the heat fluxes transferred by the thermal conductivity of the tissue along the x coordinate Δq_x and the heat released by internal sources q_v. The intensity and direction of heat transfer by thermal conductivity depend on the temperature distribution along the x coordinate:

$$\Delta q_x = q_{i+1} + q_{i-1}; \, q_{i+1} = \lambda \frac{T_{i+1} - T_i}{\Delta x}; \, q_{i-1} = \lambda \frac{T_{i-1} - T_i}{\Delta x}. \tag{11}$$

At BS boundaries, heat transfer is described by boundary conditions. For the outer boundary, the intensity of the convective heat removal is calculated:

$$i = 1; \quad q_{i-1} = \alpha (T_g - T_i), \tag{12}$$

where α means the heat transfer coefficient at the natural convection of gas or liquid, $\alpha = f(T_g, T_i)$, and W/(m^2·K); T_g means the temperature of the heat-removing medium, K.

On the inner BS boundary, the temperature of the tissues during WBC does not change and is equal to the body core temperature (0):

$$i = n; \quad T_{i+1} = T_{BC} = const. \tag{13}$$

Eq. (10) describes the change in the heat content of tissues over time $h_i = f(\tau)$, and in order to form the trial of the WBC technology issues, a similar dependence for temperature should be obtained $\mathbf{T}_i = f(\tau)$. At each time step, the temperature value is calculated from the known value of the enthalpy of the array elements:

$$T_i = f(h_i). \tag{14}$$

The described algorithm of computations forms the mathematical model of the human BS, which is suitable for studies of processes of the therapeutic effect of low-temperature liquids and gases.

5. Thermophysical bases of achievement of the WBC healthcare effect

The mathematical model of the human BS allowed us to perform a numerical experiment to study the physical bases of the WBC healthcare efficacy. Quite often [4, 5] WBC is compared with the cold water immersion procedures. The basis for this comparison is that cryogenic gas and cold water remove a significant amount of heat from the body surface. Moreover, under conditions of natural convection, the coefficient of heat transfer from a source of heat to gas is usually 10 times lower than when heat is removed by water [7]. In reliance on this information, attempts to

replace WBC with cheaper water procedures have been ongoing for 40 years [18]. Proponents of such a replacement do not take into account the fact that the WBC ET is more than 360 minutes, and water treatments provide pain relief for a maximum of 30 minutes. Such a difference in efficiency should be based on the fundamental differences between the results of heat removal by liquid and gaseous heat carrier.

Simulation of the BS surface cooling process with water with a temperature of 273 K and a cryogenic gas with a temperature of 140 K has allowed revealing such a difference (**Figure 4**), which reduces to the level of the minimum surface temperature of the cooling object. In cold water, the minimum surface temperature of the cooling object is at least 5.5°C. As the temperature difference between the water and the cooled surface decreases, the intensity of convective heat removal to water decreases to less than 950 W/m^2 [19]. The intensity of heat supply from the inner layers of the body to the surface, on the contrary, approaches the level of 880 W/m^2. Due to the small difference in heat fluxes at the boundary of the cooling object, the rate of temperature decrease $T_{i=1}$ reduces to 0.01 K/s; therefore, an increase in the duration of water hypothermia leads only to overcooling the patient's body. The estimated duration of the analgesic effect after water hypothermia is 31 minutes [16].

The use of a cryogenic gas with a temperature of 140 K for cooling the BS surface gives a completely different picture. The graph of surface temperature changes (**Figure 3**) is almost a straight line, which means that the temperature decreases almost without the rate change:

$$\partial T_{i=1}/\partial \tau \sim const. \tag{15}$$

The minimum temperature value $T_{i=1}$ = 271 K was obtained due to the termination of the numerical experiment on the patient's safety conditions during WBC: $T_s \geq 271$ K (−2°C); $T_f > 309$ K (36°C).

In the case of cooling with a cryogenic gas, the BS surface is supercooled to the minimum acceptable level. WBC technology is based on the use of this hypothermia to stimulate the cold receptors of the skin. Expression (6) for calculating the duration of positive effects contains a term that allows us to illustrate the intensity of stimulation of cold receptors by changing the temperature of the BS surface. This parameter of the WBC procedure is called the intensity of stimulating action (ISA):
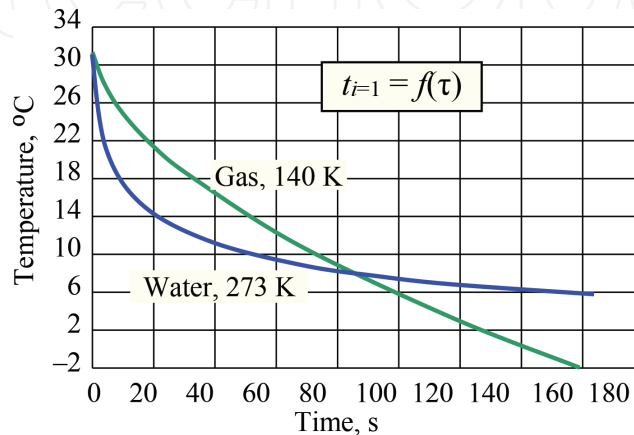
$$I_{SA} = A/(T_{i=1} - T_{kr})^2. \tag{16}$$

Figure 4.
Dependence of skin temperature on time.

The dependence graph $I_{SA} = f(T_{i\,=\,1})$ (**Figure 5**) shows how the receptor signal increases with the skin surface supercooling. At the lowest possible temperature of the skin surface under conditions of water hypothermia (5°C), I_{SA} = 21 s/s, while at WBC the maximum value is 225 times higher than I_{SA} = 4800 s/s. Differences in the intensity of stimulation of cold receptors determine the therapeutic benefits of WBC.

Data on the amount of heat Q_{HC} removed from a BS surface unit by a heat carrier (HC) and heat flux intensity has the fundamental importance for the development of WBC technology q_{HC}. The total heat removal is determined by integrating the instantaneous values $q_{HC}q_{HC}$, which were calculated by Eq. (12):

$$Q_{HC} = \int_{\tau=0}^{\tau_{max}} q_{HC}\,\partial\tau. \tag{17}$$

The calculated values are given in **Table 2**. In cryogenic gas, heat removal was 440 kJ/m², which is 10% more than in cold water. The result obtained is significantly less than that supposed by some WBC popularizers who estimate heat removal from the patient's body at 1250–2500 kJ/m² [20].

At the same time, the result obtained is significantly more than can be removed from a WBC device by using 2 kg of liquid nitrogen for one patient's cooling [6, 10].

The assessment of the power of the specific heat flux, which the BS surface gives to the heat carrier, has essential practical importance. **Table 1** shows the maximum
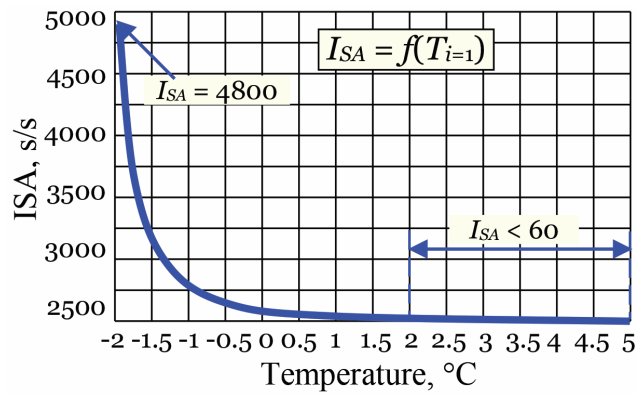

Figure 5.
Dependence of ISA value on body surface temperature.

Results		Heat carrier	
		gas	water
Cooling time, τ_{max}, sec		159	177
Minimum surface temperature of the object, T_s, °C		−2.0	5.5
The minimum temperature at the inner boundary of the fat layer, T_f, °C		309.2	309.0
Heat removed by heat carrier from the body surface, Q_{HC}, kJ/m²		440	410
Heat removed through the inner boundary of the fat layer, Q_f, kJ/m²		10.2	12.5
Heat flux from the body surface at the beginning of the cooling process, q_{HC}^{max}, kW/m²		3.5	11.3
Heat flux from the body surface at the end of the cooling process, q_{HC}^{min}, kW/m²		2.3	0.95

Table 2.
The results of a numerical experiment on the simulation of heat removal by water with a temperature of 273 K and gas with a temperature of 140 K [16].

values, at the beginning of the procedure, and the minimum values, at the time of completion of the cooling powers of the heat flux to the cold water and the gaseous heat carrier. For designing WBC devices, it is useful to know the mean value of the heat flux, which the heat carrier must remove in a single procedure, 2.9 kW/m2. This value is 29 times greater than the nominal calorific capacity of the human body; therefore, it is often challenged by manufacturers of WBC devices [13]. Estimation of the heat reserve in BS tissues before and after the procedure shows that the heat flux to the heat carrier is provided by the heat capacity of the body shell tissues (**Figure 6**).
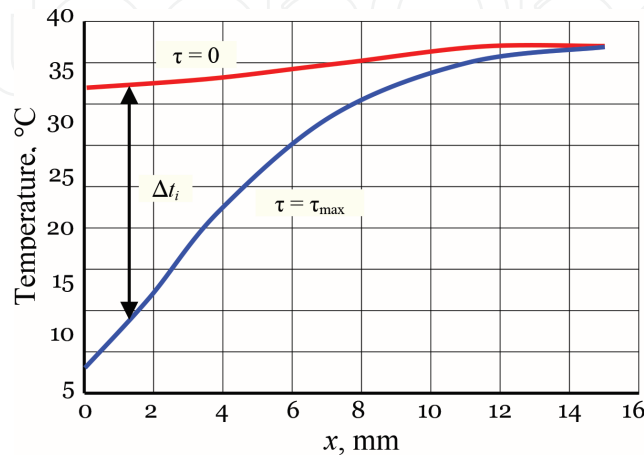

Figure 6.
The temperature of the body shell before and after the WBC procedure.

Lowering the surface temperature of the BS creates conditions for increasing heat transfer with thermal conductivity from the deep to the periphery of the body. As a result, there is a change in the distribution of the tissue temperature throughout the entire thickness of the BS. The amount of heat removed from different BS tissues is determined by the enthalpy difference before and after the WBC procedure. Taking into account the constancy of the heat capacity of the tissues in the temperature range from −2 to 40°C [17], the amount of heat removed can be calculated from the temperature difference:

$$\Delta T_i = T_i^{\tau=0} - T_i^{\tau=\tau_{\max}}. \tag{18}$$

The amount of accumulated heat removed from one area element:

$$Q_{Ai} = \Delta T_i \Delta x \rho_i c_i. \tag{19}$$

The total amount of heat released due to supercooling of each of the three types of BS tissues is the sum of portions of heat released in the area elements of this tissue layer:

$$
\begin{aligned}
&\text{epithelial layer,} \quad 1 \leq i \leq n_E, \quad Q_{AE} = \sum_{i=1}^{n_E} \Delta T_i \Delta x \rho_i c_i; \\
&\text{fat layer,} \quad n_E + 1 \leq i \leq n_F, \quad Q_{AF} = \sum_{i=n_E+1}^{n_F} \Delta T_i \Delta x \rho_i c_i; \\
&\text{muscular layer,} \quad n_F + 1 \leq i \leq n_{BS}, \quad Q_{AM} = \sum_{i=n_F+1}^{n_{BS}} \Delta T_i \Delta x \rho_i c_i,
\end{aligned}
\tag{20}
$$

where n_E, n_F, and n_{BS} means the number of area elements located between the outer surface and the inner boundary of the epithelial layer, the fat layer, and the patient's body shell, respectively.
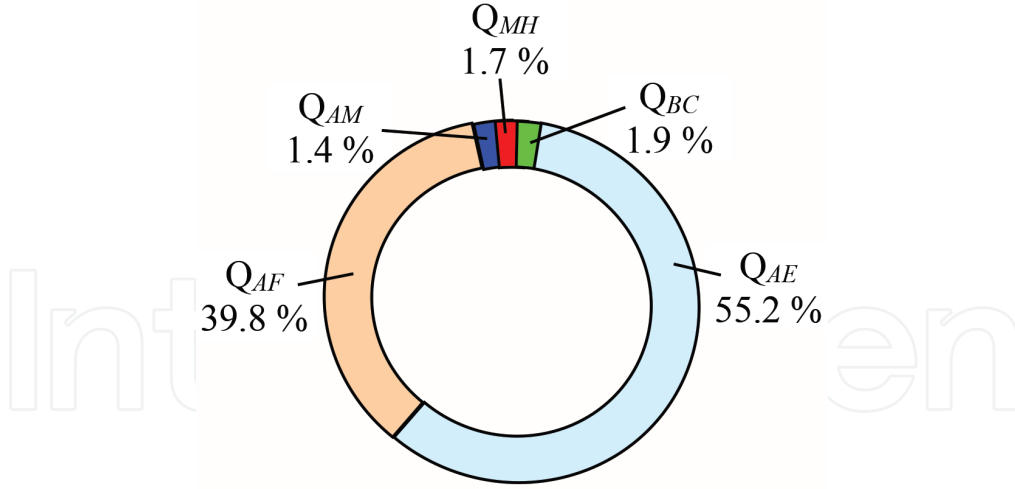
Figure 7.
Sources of the heat gained the WBC procedure.

Part of the heat removed was obtained from internal sources in the epithelial and muscle layers, heat of metabolism Q_{MH}. This heat was calculated by a known value q_v (**Table 1**):

$$Q_{MHE} = \tau_{max}\Delta x(n_E - 1)q_{VE}; \quad Q_{MHM} == \tau_{max}\Delta x(n_{BS} - n_F - 1)q_{VM}, \quad (21)$$

where q_{VE} and $q_{VM}q_{VM}$ means the specific heat of metabolism of epithelium and muscles, respectively, W/m^3.

Some of the heat removed came from the patient's body core; the amount of heat gained can be determined by numerical integration and instantaneous values of the heat flux transferred by thermal conductivity through the inner boundary of the body shell:

$$Q_{BC} = \int_{\tau=0}^{\tau_{max}} q_{n_{i+1}} \partial\tau. \quad (22)$$

The histogram on **Figure 7** gives an idea of what is the source of heat removed from the surface of the patient's body shell. The main share of the heat of 55.2% was gained due to supercooling the epithelial layer. The heat gained by supercooling the fat layer Q_{AF} is 39.8%. The supply of heat from the body core Q_{BC} and internal sources Q_{MH} in the body shell tissues is less than 2%. This supply of heat is gained by supercooling the muscle layer Q_{AM}.

The calorific capacity of the body does not play any role in the formation of the heat load on the cooling system of the WBC device, which is determined by the heat storage capacity of the body shell tissues. The safety of the WBC procedures is ensured by the correct choice of the contact duration of the body surface with a cryogenic gas. The thermal control system of the body does not affect the safety of procedures.

6. Selecting optimal gas temperature in the WBC zone

In practice, there are two options for carrying out WBC procedures in multi-seat and single-seat installations [7, 21–23]. The cooling conditions in these installations

differ significantly; therefore, the technology of group and individual WBC should be developed separately.

Contrary to the popular belief [7, 13], GWBC and IWBC provide effects on only a fraction of the skin area. In a group installation, the contact area of the cooling gas with the patient's body in a multi-seat cab is up to 70.5% of the total surface area of the body. In an individual cab, the contact area reaches 66% [7]. Temperature regimes of GWBC and IWBC are fundamentally different.

The GWBC technology was influenced by the design of the device for performing the procedures (**Figure 1**). Using a low-temperature food storage chamber for WBC procedures, Japanese engineers and doctors were forced to carry out WBC procedures in groups. The dimensions of the chamber were too large for individual procedures. This forced solution is contrary to the general practice of physiotherapy; treatment is always carried out individually.

Systems for implementing technology I, individual cryosaunas, were developed 20 years after multi-seat installations [7] with consideration of the experience of their operation. Modern installations for IWBC use a nitrogen cooling system (NCS), so they quickly reach a given temperature level and allow you to adjust the temperature of the gas in the WBC zone.

It is impossible to develop universal recommendations on selecting the optimal temperature of the gaseous heat carrier for GWBC and IWBC, since in multi-seat and single-seat installations the algorithm for changing the temperature of the cooling gas during the procedure is different.

To conduct a preliminary analysis of the effect of gas temperature in the WBC zone on the magnitude of the positive effect achieved, it can be assumed that the procedure takes place in isothermal conditions:

$$0 < \tau \leq \tau_{\max}; \quad T_1 = \text{const.} \tag{23}$$

It is impossible to implement WBC in the isothermal mode, since it takes some time for the patient to enter the low-temperature zone and exit from it. However, the study of WBC processes in ideal temperature conditions allows us to formulate the general technological conditions of efficiency.

To determine the optimal gas temperature in the WBC zone, the calculated values of the WBC ET obtained by Eq. (6) were used. Simulation of the BS cooling process under conditions of natural gas convection with a temperature from 90 to 190 K allowed us to plot the dependence of the ET value on the gas temperature in the WBC zone (**Figure 8**).

When isothermally cooling the surface of the patient's body, the maximum value of ET (325 min) is achieved at a temperature of 140 K. At temperatures below 140 K, the WBC efficiency gradually decreases. At a temperature of 100 K, the value of WBC ET is almost three times lower than the maximum [7]; therefore, when conducting WBC procedures, it is advisable to use a gas with a temperature from 120 to 140 K [16]. At temperatures above 140 K, the WBC efficiency rapidly decreases. At a temperature of 160 K, the WBC ET of the procedures is 10 times lower than the maximum value and is close to the results achieved during water procedures. The results of the computational experiment on simulation of cooling the body surface with gas with a temperature of 160 K (−110°C) ideally coincide with the results of tests performed by doctors in sports medicine [8], which in comparing the therapeutic effect of WBC procedures at a temperature of −110°C and water baths with a temperature of 8°C, did not reveal any advantages of the WBC. The results obtained have clear thermophysical reasons. As the temperature of the gaseous heat carrier increases, the intensity of heat removal from the BS surface decreases, and the safe cooling time increases.
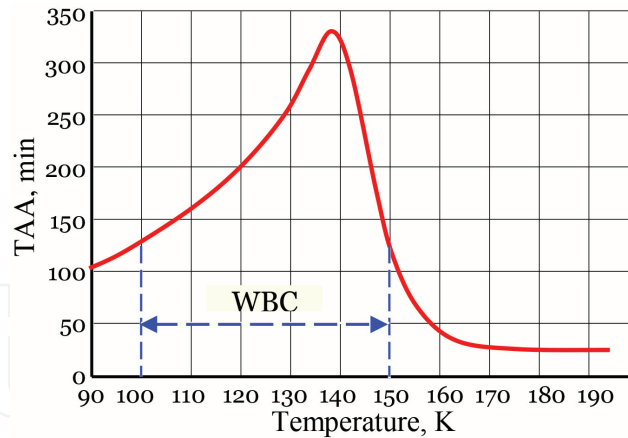
Figure 8.
The estimated duration of the effect of WBC at different gas temperature.

When the gas temperature is above 150 K, the danger of supercooling of the body core ($T_f \rightarrow$ 309 K) occurs before the surface of the body shell is supercooled. The reason for the termination of the WBC procedures becomes a violation of the condition $T_f \geq$ 309 K. At the same time, the temperature of the body shell surface remains at a sufficiently high level of $T_s \geq$ 275 K (2°C), due to which the cold receptors of the skin do not experience significant irritation and the accumulation of a positive WBC effect is extremely slow. The picture described is identical to what is observed at the time of completion of the water hypothermia procedure. Under conditions of isothermal cooling of the body with gas with a temperature of 160 K (−110°C), the estimated duration of the WBC procedure is 207 sec. During this time, the BS surface temperature drops only to 275 K. At this BS surface temperature, the I_{SA} value is 80 times less than the maximum value (**Figure 5**). Under actual conditions, the WBC procedures in installations with a minimum temperature of 160 K (−110°C) do not ensure the constancy of the gas temperature, so the BS surface temperature after the procedure is much higher than the calculated one and is 15–20°C [24]. Such a temperature on the surface of the skin can be obtained using water baths with a temperature of 8°C; therefore, the doubts of some authors [11, 18, 25] on the advisability of using cryogenic technologies are fully justified.

According to the results of simulating the process of cooling the BS surface with a cryogenic gas, it can be argued that for effective procedures the gas temperature in the WBC zone should be not lower than 140 K.

7. Selecting the optimal duration of the WBC procedure

The author of the WBC method, Yamauchi, limited the exposure of the body contact with a cryogenic gas to a period of 180 sec [1, 2]. The minimum air temperature in the Japanese installation was −175°C. According to the contemporary idea that WBC technology is based on metered supercooling of the body shell, the choice of the cooling exposure should be related to the temperature of the gas in the WBC zone. Using the assumption of the constancy of the gas temperature in the WBC zone, it is possible to determine the maximum duration of cooling at different gas temperatures. Computational experiments on the mathematical model of the human BS showed that with an increase in the heat carrier temperature from 90 to 190 K, the safe duration of a patient's stay in the WBC zone increases from 54 to

237 sec [3]. At a temperature of 140 K, the safe exposure time for cooling is 161 sec. The practice of using WBC has shown that, along with the maximum duration of cooling, it is necessary to limit the minimum duration of stay of patients in a cryotherapeutic installation [7].

The reasons for this limitation are explained by the graph of dependence $ET = f(\tau)$ (**Figure 8**). The graph shows that there is a fairly long period in the WBC procedure when a positive effect is not formed. At a gas temperature of 140 K, this phase of the procedure accounts for almost 80%, but 93% of the positive effect is formed after its completion. The reason for the low efficiency of WBC at the beginning of the procedure is the relatively high skin temperature (T_S) (**Figure 8**), which drops to 275 K (2°C) only by the end of the first phase of the procedure. As it can be seen from the graph of dependence $I_{SA} = f(T_S)$ (**Figure 5**), at a skin temperature of $T_S > 275$ K, the intensity of the WBC stimulating effect is negligible $I_{SA} < 60$. The first phase of the WBC procedure reduces the surface temperature of the skin to a temperature of $T_S = 275$ K, so it is called the cooling time (τ_{cool}) (**Figure 9**). It is obvious that the duration of the WBC procedure must be longer than the duration of the cooling time but less than the time of the violation of safety conditions $\tau_{cool} < \tau < \tau_{max}$. The second, effective, phase of the procedure ensures the formation of the main positive result, the longer the duration of the effective phase, the greater the effect of the procedure.

$$\tau_{EP} = \tau_{max} - \tau_{cool} \tag{24}$$

The calculated dependences of the WBC safe exposure (τ_{max}) and the duration of the cooling time (τ_{cool}) on the gas temperature (T_g) (**Figure 10**) show that increasing the gas temperature from 90 to 150 K increases the effective phase of the procedure. A further increase in temperature increases the duration of the cooling time. At temperatures above 160 K, the estimated duration of the cooling time exceeds the safe WBC exposure. Even with isothermal cooling, it is impossible to provide an effective WBC when using gas with a temperature $T_g > 150$ K; in real conditions the gas temperature should be no higher than 140 K.

Numerical experiments on a mathematical model of the human body shell allowed to formulate general ideas about the technological foundations of effective WBC. When developing technological recommendations on the design of installations for the implementation of GWBC or IWBC methods, it is necessary to take into account the algorithm for changing the temperature of the gas in contact with the patient's body surface.
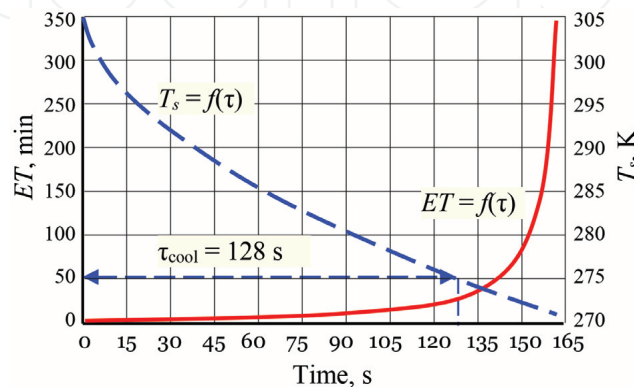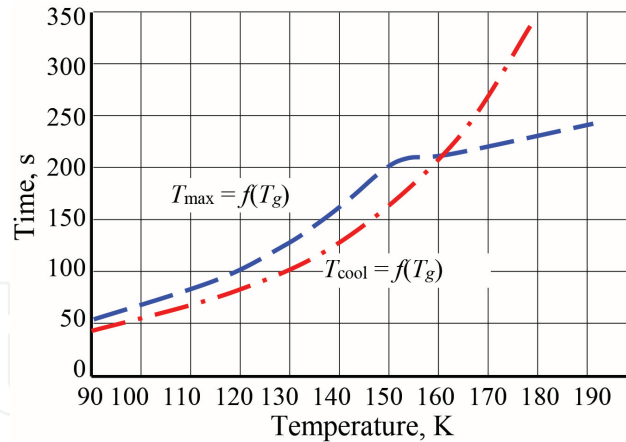

Figure 9.
The change of the body temperature surface T_s *and the value of* ET *during the WBC procedure with gas temperature of 140 K.*

Figure 10.
The dependence of the cooling phase duration and safe exposure WBC on the gas temperature.

8. Algorithm for changing the cooling gas temperature during the individual and group WBC procedures

Installations for GWBC consist of two or three heat-insulated rooms with different air temperatures [7]. Patients pass from the treatment room to the chamber with the minimum temperature (main chamber, MC) and back through the lock chambers (LC). In most modern installations, the temperature in the main chamber is maintained at 160 K and in the lock chamber means at 210 K. At the time of entry (exit) of patients into the LC or MC, warmer air enters from adjacent volumes. Because of this, the air temperature in the MC volume increases by at least 25 K. From the body surface of each patient, 3.5 to 4.5 kW of heat is released into the MC volume. Taking into account these factors, the actual GWBC temperature regime depends not only on the choice of the nominal temperatures in MC and LC but also on the power of the cooling system. Another uncertainty factor is the duration of stay of patients in the main cab. There are different opinions about the advisability of pre-cooling the body surface at an intermediate temperature of 210 K. Some researchers believe that a gradual decrease in temperature increases subjective comfort and safety [7]. In other works it is proposed to reduce the time of stay of patients in LC to a minimum [13]. Given all the reasons presented, it is obvious that it is extremely difficult to simulate the GWBC process. The temperature of the cooling gas varies according to a complex schedule, which consists of at least eight stages (**Figure 11**).

The algorithm of changing the gas temperature in IWBC is much simpler (**Figure 11**). The patient enters the cab filled with atmospheric air, which is quickly replaced by vapors of liquid nitrogen with a temperature not higher than 140 K. The time to reduce the gas temperature in the IWBC cab to the optimum level depends on the power of the cooling system and is at least 20 sec.

Taking into account the results of simulating the WBC process under conditions of a constant gas temperature, it can be argued that the GWBC procedures using the algorithm shown in **Figure 11** do not provide significant therapeutic outcomes. To restore the effectiveness of GWBC, it is necessary to significantly reduce the minimum air temperature in the main treatment cab. Experiments on a mathematical model of the body shell showed that the effectiveness of GWBC reaches the optimal level when the air temperature in the main cab drops to 130 K. However, modern installations for GWBC cannot maintain the temperature at this level, since they use
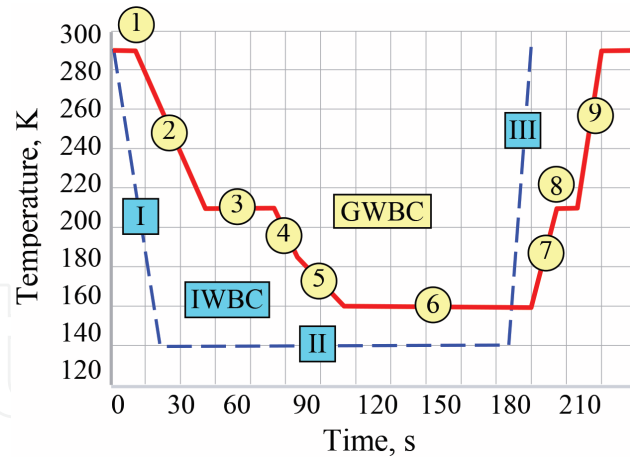
Figure 11.
Algorithm for changing the temperature of the of the cooling gas with the technology GWBS and IWBS.

compression cooling systems on gas mixtures [7]. To lower the temperature, it is necessary to use other heat transformation cycles in the cooling system, the power of which will allow compensating for the heat load associated with WBC procedures.

9. The heat load on the cooling system of the WBC zone

When designing cooling systems of the WBC zone, it is necessary to adequately estimate the power of the heat fluxes that need to be compensated. It was shown above that during the WBC procedure, 440 kJ/m^2 of the heat is released from the patient's body surface, and the mean heat flux from the body to the cryogenic heat carrier varies from 3.5 to 2.3 kW/m^2 (**Table 2**). Taking into account the surface area of the body (1.6 m^2 [7]), the heat input from one patient will be 700 kJ; the mean power of the heat input is 4.6 kW. It is necessary to spend at least 2.7 kg of liquid nitrogen only to remove the heat released from a patient's body surface with a gas with the temperature of T_g = 140 K:

$$G_{LN} = \frac{Q_{HC}}{\left[r_{LN} + c_{LN}\left(T_g - T''_{LN}\right)\right]'} \tag{25}$$

where r_{LN} means the heat of vaporization of nitrogen, r_{LN} = 199 kJ/kg; c_{LN} means the heat capacity of nitrogen vapor, c_{LN} = 1.002 kJ/kg; and T''_{LN} means the boiling point of liquid nitrogen at atmospheric pressure, T''_{LN} = 78 K.

Estimated nitrogen flowrate for removal of the heat from the body surface is 2.7 times higher than in modern nitrogen-cooled WBC installations [13]. To restore the WBC effectiveness, it is necessary to provide cryotherapy installations with suffi-ciently powerful cooling systems.

The heat input from the patients Q_{HC} is only the useful part of the heat load on the cooling system. In addition, it is necessary to compensate for the heat input from the walls of the thermal fencing of the WBC zone and Q_{TI} the heat that the warm gas fluxes bring from the adjacent volumes (lock chamber or environment) to the volume of the treatment cab Q_{GC}. The total heat load is defined as the sum of heat received from different sources:

$$Q_\Sigma = Q_{HC} + Q_{TI} + Q_{GC}. \tag{26}$$

The energy efficiency of the installation design for WBC can be estimated by the share of the useful load on the cryostatting system, the coefficient of thermal efficiency:

$$\eta_H = Q_{HC}/Q_\Sigma. \tag{27}$$

To estimate the expenditure of energy and select the optimal technology for WBC procedures, it is necessary to conduct a numerical experiment on a mathematical model of a cryotherapeutic device.

10. Mathematical model of cryotherapeutic device

Given the variety of design solutions used in the manufacture of devices for WBC, the mathematical model should have the most generalized form. It is necessary to stop considering particular design features and focus on the fundamental issues. It becomes possible with a one-dimensional model of the WBC zone (**Figure 12**). The model considers the processes occurring in a volume unit of the WBC zone.

The surface of the patient's body 1 is cooled with a gaseous heat carrier 2 which fills the volume of the thermal fencing 3. The heat flux is removed from the patient's body surface q_{HC}. The heat flux is supplied to the gas 2 from the surface of the thermal fencing 3 q_{TI}. The movement of gas fluxes creates an additional source of the heat load q_{GC}. To keep the gas temperature in the WBC zone at a constant level, it is necessary to ensure the removal of total heat q_Σ to the cryostatting system. To form a one-dimensional model, it is necessary to relate all system indicators to the volume unit of the WBC zone. The specific heat transfer surface of the thermal insulation f_3 and the patient's body f_1 are determined taking into account the volume of the treatment cab V_3:

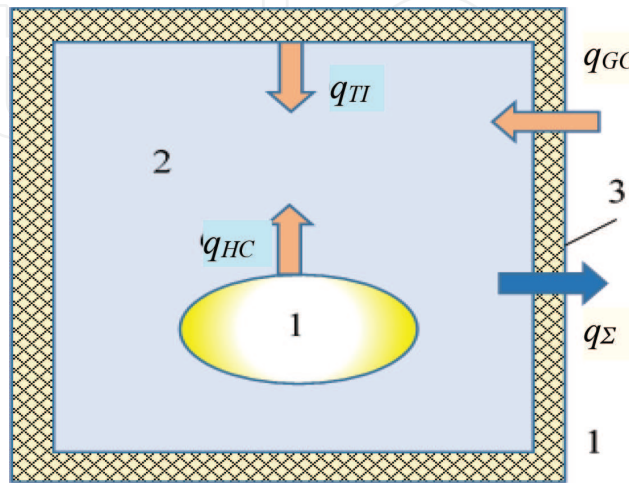$$f_3 = F_3/V_3; \ f_1 = nF_1/V_3, \tag{28}$$

Figure 12.
Heat fluxes in the WBC zone.

where F_3 means the total area of the internal surface of the cab's thermal fencing, n means the number of patients in the cab, and F_1 means the surface area of a patient's body.

The specific heat input from the patient's body and the thermal fencing is calculated considering the temperatures of their surfaces:

$$q_{TI} = \alpha_{3-2}\, f_3(T_3 - T_2); q_{HC} = \alpha_{1-2}\, f_1(T_1 - T_2), \qquad (29)$$

where α_{3-2} and α_{1-2} means the heat transfer coefficients from the thermal fencing and the patient's body, respectively; T_1 and T_3 means the temperatures of surfaces of the body and the fencing; and T_2 means the temperature of the heat carrier gas.

Specific characteristics of devices designed for the implementation of GWBC and IWBC technologies have large differences. In multi-seat installations, the patient accommodation density is 0.4–0.7 person/m^3, and the specific volume of free space V_0 is at least 98%. The specific surface area of the patient's body is 0.6–1.0 m^2/m^3; the thermal fencing area of the WBC zone is 2.4–3.0 m^2/m^3. Low compactness of accommodation of patients in multi-seat installations is necessary so that they can move from one low-temperature chamber to another.

In single-seat cryosaunas, the patient accommodation density reaches 2.0 persons/m^3, the specific surface area of the patient's body is 3.2 m^2/m^3, the thermal fencing area of the WBC zone is 6.4 m^2/m^3, and the specific free space volume is 84% [7]. High compactness of the patient accommodation is ensured by the fact that the patient does not move during the procedure; therefore, the cab size is comparable to the size of the patient's body.

The heat input with gas fluxes is determined by the intensity of convective mass transfer of warm gas to the volume of the WBC zone. The heat input by gas convection across the boundary of the WBC zone is determined from the expression:

$$q_{GS} = c_p g_G(T_g - T_2), \qquad (30)$$

where g_G means the specific transfer of the gas mass into the volume of the WBC zone, kg/(m^3 sec); c_p means the heat capacity of the gas, kJ/(kg K); and T_g means the temperature of the gas entering the WBC zone.

Large heat flows with gas fluxes are supplied into the WBC zone as patients enter and exit. For example, a multi-seat lock chamber and a cab of a single-seat cryosauna are filled with atmospheric air at the moment patients enter. 93 kJ/m^3 of heat enters the lock cab with atmospheric air. When the temperature recovers to the nominal level, the air density in the lock cab increases by 40%; this is accompanied by supplying additional air from the atmosphere, which contributes another 27 kJ/m^3 of heat. In one procedure, 120 kJ/m^3 of heat transferred by gas convection enters the lock chamber.

The basis of the mathematical model of the WBC zone is a one-dimensional energy equation:

$$\rho \frac{\partial h}{\partial \tau} = \frac{\partial q_x}{\partial x} + q_v, \qquad (31)$$

where q_v means the heat from internal sources:

$$q_v = q_\Sigma + q_{HC} + q_{TI} + q_{GC}. \qquad (32)$$

In the ideal case, $q_v = 0$, since $q_\Sigma = -(q_{HC} + q_{TI} + q_{GC})$, i.e., the cooling system compensates for the heat input per volume unit of the WBC zone.

To account for material balance in the mathematical model of the WBC zone, the continuity equation is used:

$$\frac{\partial \rho}{\partial \tau} + \frac{\partial g_x}{\partial x} = 0. \tag{33}$$

So, the transfer of heat by the thermal conductivity of gas is small; expression (30) is simplified and can be transformed by replacing the derivatives with differential approximants:

$$\rho \frac{\Delta h}{\Delta \tau} = q_v; h' = h + \frac{q_v \Delta \tau}{\rho}. \tag{34}$$

The numerical solution of the continuity (Eq. (33)) allows to take into account the input of gas mass to compensate for the change in density:

$$g'_x = g_x + \frac{(\rho' - \rho)\Delta x}{\Delta \tau}. \tag{35}$$

Eqs. (29), (34) and (35) allow to analyze the processes occurring in the WBC zone during the implementation of individual or group technology. To perform a computational experiment, it is necessary to adopt an algorithm for changing the temperature of the cooling gas for IWBC and GWBC.

Formulation of a temperature algorithm for the IWBC process is relatively simple. let us take the time of filling the zone with a cryogenic gas (**Figure 11**) τ_I = 20 sec, τ_{II} = 150 sec, and τ_{III} = 10 sec and the gas temperature in the isothermal phase II T_{II} = 140 K. The specific heat transfer surfaces of heat sources are f_1 = 3.2 m^2/m^3, and f_3 = 6.4 m^2/m^3; the specific free space volume is 84% [7]. When simulating the IWBC process, heat fluxes from various sources (28) and (29) are calculated, and the integral heat input is determined:

$$Q_{HC} = f_2 \int_{\tau=0}^{\tau=\tau_{max}} q_{HC}\partial\tau; Q_{TI} = f_3 \int_{\tau=0}^{\tau=\tau_{max}} q_{TI}\partial\tau; Q_{GC} = \int_{\tau=0}^{\tau=\tau_{max}} q_{GC}\partial\tau. \tag{36}$$

By Eqs. (26) and (27), the total heat load on the cooling system and the coefficient of thermal efficiency are calculated. It is assumed that the cooling system covers all types of heat load, so the specific power of the refrigerator electric drive can be determined by the heat load and the value of the coefficient of performance at the current temperature level:

$$N_5 = q_\Sigma/\varepsilon_5; \varepsilon_5 = f(T_g), \tag{37}$$

where ε_5 means the coefficient of performance and the ratio of the heat removed to expenditure of energy in the refrigerator at the temperature level of 140 K, ε_5 = 0.25 W/W.

For the instantaneous values of the calculated power of the system refrigerator, the specific expenditure of energy for cooling the IWBC zone per procedure is calculated:

$$Q_5 = \int_{\tau=0}^{\tau=\tau max} N_5 \partial\tau \tag{38}$$

Let us determine the specific values of the liquid nitrogen flowrate per procedure:

$$g_{LN} = \frac{q_\Sigma}{\left[r_{LN} + c_{LN}\left(T_g - T''_{LN}\right)\right]}; \; G_a = \int_{\tau=0}^{\tau=\tau_{max}} g_a \partial\tau. \qquad (39)$$

The results of the numerical experiment are summarized in **Table 3**. In the experiment on simulating the GWBC process, the time algorithm presented on the graph (**Figure 11**) was used; the nominal gas temperature in the main cab was 130 K. Energy indicators for the main and lock chambers were calculated.

The energy efficiency of the technology was estimated by the total energy expenditures in the main and lock chambers. The results of the numerical experiment on simulating the GWBC process are summarized in **Table 3**.

The data in **Table 3** show that WBC procedures require the removal of large amounts of heat from the low-temperature zone. Specific heat input to the IWBC zone is Q_Σ = 2012 kJ/m3. Given the short duration of the procedure (τ_{max} = 180 sec), the mean heat load on the cooling system of the IWBC zone was 11.8 kW/m^3. Considering that this heat load must be removed at a temperature level of 140 K, the estimated power of the cooling system of the low-temperature zone was 45.3 kW/m^3. Refrigerators of such power are rather expensive; therefore, it is economically feasible to use the nitrogen cooling system of the WBC zone. Specific liquid nitrogen flowrates are 7.503 kg/m^3.

Indicators	IWBC	GWBC		
		Cab	Lock chamber	Total
Features of WBC zone				
f_1, m^2/m^3	3.2	0.62	0.62	
f_2, m^2/m^3	6.4	2.4	2.4	
V_0, %	0.84	0.97	0.97	
Heat input to the WBC zone				
q_Σ^{max}, kW/m3	33	7.43	3.97	
Q_Σ, kJ/m^3	2012	422	144	566
Q_{GC}, kJ/m3	92	142	96	238
Q_{HC}, kJ/m3	1427	246	33	279
Q_{TI}, kJ/m3	493	33	14	47
Q_Σ/τ_{max}, kW/m3	11.8	2.76	1.18	3.94
η	0.71	0.58	0.23	0.49
The expenditure of electrical energy and liquid nitrogen flowrate for cooling				
N_5^{max}, kW/m3	136	11.04	4.72	15.76
Q_5, kW/hour/m^3	2.15	0.50	0.07	0.57
Q_5/τ_{max}, kW/m3	45.3	11.04	4.72	15.76
G_{LN}, kg/m^3	7.503	1.56	0.42	2.02
G_{LN}/τ_{max}, g/(sec·m3)	0.04	0.0029	0.009	0.0038

Table 3.
Energy features of devices for IWBC and GWBC.

The energy indicators of the GWBC zone are much lower (**Table 3**). The specific heat input is Q_Σ = 566 kJ/m^3, the mean heat load on the cooling system is 15.6 kW/m^3, and the power of the cooling system is determined by the sum of the inflows of heat into the main cab and lock chamber.

Due to the low compactness of the accommodation of patients in the treatment area, the GWBC thermal efficiency coefficient was 0.49. Under conditions of a single-seat installation, the thermal efficiency coefficient was 0.71, which indicates a more rational expenditure of energy. This is clearly illustrated by the histogram of the structure of the heat load on the cooling system of the IWBC and GWBC zones (**Figure 13**).

In single-seat installations, the heat storage capacity of the thermal fencing makes a significant contribution to the heat load, due to which the share of heat removed from thermal insulation reaches 24%. At the beginning of each procedure, a single-seat cab is filled with atmospheric air, which heats the inner surface of the thermal insulation. When implementing the GWBC technology, the heat load from the insulation is insignificant means of 9%, but the convective heat supply is 24%. The negative impact of convective heat transfer is determined by a large share of the free space in the low-temperature zone.

The data in **Table 3** do not allow giving an unambiguous preference for a particular technology. This is due to the fact that all indicators are related to the volume unit of the WBC zone, while the technological task of the process is to cool the surface of the patient's body shell. If we calculate the specific heat load values and the expenditure of energy for cooling a unit of the shell surface (**Table 4**), the advantages of the IWBC technology become indisputable. According to all energy indicators, the IWBC technology is 1.5 times more efficient than the GWBC process.
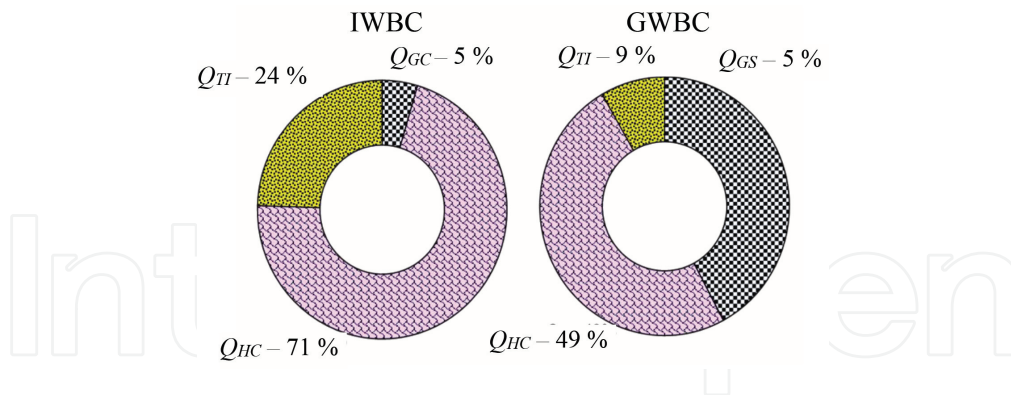

Figure 13.
The structure of the heat load on the cooling system zones IWBC and GWBC.

Indicators	IWBC	GWBC
Q_Σ/f_1, kJ/m2	629	913
$Q_5/f_1, Q_5/f_1$, kW/hour/m2	0.67	0.92
G_{LN}/f_1, kg/m2	2.34	3.24
$G_{LN}/f_1 \cdot F1$, kg/m2	3.77	5.18

Table 4.
Energy features of devices for IWBC and GWBC.

11. Conclusion

The performed analysis of the healthcare and energy efficiency of the two options for the implementation of the WBC technology allows us to reasonably give preference to individual procedures that not only combine high healthcare efficiency with relatively low expenditure of energy but also to a greater extent correspond to the traditional principle of individuality of therapeutic techniques.

The effectiveness of WBC technology depends on the choice of the duration of contact with cryogenic gas. The minimum duration of WBC procedure at the optimum gas temperature ($-130°C$) is 120 s. Meanwhile, one should remove 440 kJ/m^2 with an average intensity of at least 2.4 kW/m^2 and spend not less than 1.7 kg/m^2 of liquid nitrogen on heat removal. The electric drive of the cooling system of WBC zone should have an average power of at least 9.3 kW/m^2, and in the case of using nitrogen cooling system, the cryoagent consumption should be not less than 2.4 kg/m^2.

Other declarations

Author details

Alexander Baranov*, Oleg Pakhomov, Alexander Fedorov, Vladimir Ivanov, Andrew Zaitsev and Ruslan Polyakov
Saint Petersburg National Research University of Information Technologies, Mechanics and Optics, Saint Petersburg, Russian Federation

*Address all correspondence to: abaranov@corp.ifmo.ru

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