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Non-invasive Muscle Temperature Control during Cooling

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***Abstract*—The aim of this work was to investigate the possibilities of using the non-invasive method to control the temperature of deep layers in limb tissues during cooling, using thermistors as temperature sensors.**

The invasive temperature control of the deep tissue layers is not always acceptable to the test subject. The temperature distribution was simulated using computer software, when the limb was cooled in the +15 °C temperature water bath and frozen by adding the ice pack. The temperature was measured experimentally by using thermistor as a temperature sensor. The uncertainty of calibrated temperature transducer was ±0.05 °C. The 3 cm thickness layer of thermal insulation separated the temperature sensor from the cooling agent.

After 30 min of cooling the tissue surface temperature at the sensor attachment point was 31.5±0.65 °C. During modeling it was determined, that in case when temperature underneath the sensor was 31.5 °C, then the tissue temperature in 30 mm depth should be 31.9 °C, what corresponds to the temperature values determined by other researchers using invasive measurements under analogous conditions. On the basis of the study results the structure of a hand-held device for control of temperature in deeper muscle layers via measurement of the skin surface temperature was offered.

***Index Terms*—Biomedical equipment, temperature control, thermistor, human tissue.**

I. INTRODUCTION

Oral Thermometry and Tympanic Thermometry methods are used for non-invasive body temperature measurements [1]. Results of measurement using Oral Thermometry method correlate with the results of invasive measurements better than using Tympanic Thermometry method [1]. For non-invasive body temperature measurement a method of human forehead temperature evaluation using double-sensor measurement was offered. Measurement results obtained using this method matched the oesophageal temperature measurements within the limits of ±0.5 °C [2]. When high-precision NTC thermistors were used to measure body surface temperature, the temperature measurement accuracy of up to 0.02 °C was achieved in the temperature range 16–42 °C [3].

Zero heat flow method is used for non-invasive measurements of body temperature. During local heating of skin up to the temperature of deep layers a condition of heat low balance or so-called 'zero heat flow' is reached. The skin temperature measured in this state can be regarded as a body (deep layer) temperature [4]. The coincidence of the measurement results obtained by this method with the results of invasive measurement techniques was confirmed by many studies [5], [6]. There are attempts to measure the temperature of deeper layers using ultrasound method [7].

Diffuse optical spectroscopy [8] and magnetic resonance spectroscopy [9] methods are used for temperature control in deeper tissue layers. It has been indicated [10], that by application of IR technology and using special algorithm for data processing (Spatial Temperature Algorithm) it is possible to measure the temperature in the deeper tissue layers. However, these methods require costly hardware and often require operating room or special laboratory environment.

When reducing the temperature of entire body, the temperature is typically controlled using invasive methods [1], [2].

A number of devices are proposed for non-invasive body temperature measurement, in which two temperature sensors, separated by a layer of thermal insulation with a known heat transfer coefficient *k*, are used. Knowing the empiric human tissue heat transfer coefficient k_t and by measuring the temperatures on both sides of the thermal insulation layer it is possible to calculate the body temperature [2]. The drawback of the method is that human tissue heat transfer coefficient k_t has to be determined empirically and depends largely on the structure of the tissue, thickness of fat layer, skin thickness and moisture, distribution of blood vessels and a number of other factors. In other cases, a heater for local body tissue heating is installed additionally [6], [11].

Temperature of deeper layers of body tissues is usually referred to as the 'core temperature' and for healthy human under normal conditions 37±0.6 °C is considered as stable. Body surface temperature may vary over a wide range, depending on environmental conditions.

It is believed that the core body temperature is measured most accurately using the thermistor on the pulmonary artery catheter [12].

When studying effects of cold on muscle function, invasive method is typically used by inserting the

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temperature sensor into appropriate depth inside the muscle [13], but this way is not always acceptable to patient. In an investigative practice, body part is usually cooled by immersing it into the water at a temperature of 15 °C where it is stored for a specified period of time. Invasive temperature measurements have shown that prior to cooling the muscle temperature was 36,8 ± 0,2 °C, and after cooling for 30 min. the muscle temperature at a depth of 3 cm decreased to 32.5±0.3 °C [13].

When cooling muscles or other parts of the body after traumatic injuries, various applications are possible, when ice of temperature 0 °C, frozen gel or other materials are used as a cooling agent [14]. When using such cooling, the temperature of deeper layers usually remains unknown and only cooling time is being limited.

Therefore it is important to find a way to control the muscle temperature changes when cooling a particular part of body (arm, hand or leg) using non-invasive measurement methods. So the aim of this work was to investigate the possibilities to control the temperature of deep layers in limb tissues during cooling via non-invasive method.

II. Modeling of Thermal Fields in Muscles

Heat propagation is an important process for living organisms, especially to a human body in order to maintain a nearly constant body temperature.

Heat generated inside the body is commonly named as bioheat, the propagation process of which has been described by so-called Pennes equation or its various modifications [15]–[17]. Analysis of bioheat transfer processes is applicable when researching the effect of heat on physiological and physical properties of tissues [16], [17].

Thus the bioheat transfer process taking place in muscles can be described by

$$\delta_{ts}\,\rho C\frac{\delta T}{\delta t}+\nabla\cdot(-k\nabla T)=\rho_b C_b\,\omega_b(T_b-T)+Q_m+Q_{ex}, \quad (1)$$

where δ_{ts} – a time-scaling coefficient (dimensionless); ρ – the tissue density (kg/m^3); C – the specific heat of tissue (J/(kg·K)); k – the tissue's thermal conductivity tensor (W/(m·K)); ρ_b – the density of blood (kg/m^3); C_b – the specific heat of blood (J/(kg·K)); ω_b – the blood perfusion rate (m^3/(m^3·s)); T_b is the arterial blood temperature (K); Q_{met} – the heat source from metabolism (W/m^3); Q_{ex} – the spatial heat source (W/m^3).

For a steady-state problem we have

$$\nabla\cdot(-k\nabla T)=\rho_b C_b\omega_b(T_b-T)+Q_m+Q_{ex}. \quad (2)$$

In the absence of spatial heat sources inside tissues, (2) can be transformed as

$$\nabla\cdot(-k\nabla T)=\rho_b C_b\omega_b(T_b-T)+Q_m \quad (3)$$

and it can be used to model thermal fields in tissues.

Boundary conditions

$$-\vec{n}\cdot(-k\nabla T)=q_0+h(T_{ext}-T), \quad (4)$$

heat flux (upper and lower boundary of the model)

$$\vec{n}\cdot(k_1\nabla T_1-k_2\nabla T_2)=0, \quad (5)$$

continuity on the all interior boundary between layers of the model

$$-\vec{n}\cdot(-k\nabla T)=0, \quad (6)$$

symmetry condition, used to reduce model size by taking advantage of symmetry (on the left outer boundary of the model)

$$T=T_0, \quad (7)$$

prescribed temperature (water or ice pack), where T_{ext} – external temperature, h – heat transfer coefficient.

The structure of tissue layer model is shown in Fig. 1.

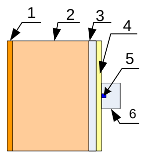

Fig. 1. Tissue layers model with affixed temperature sensor and thermal insulation. Here: 1 – the deepest layer with temperature +37^0 C; 2 – muscle; 3 – subcutaneous fat; 4 – skin; 5 – temperature sensor; 6 – thermal insulation.

Bioheath transfer model properties used in modeling are given in Table I, and bioheat equation inputs are provided in Table II.

TABLE I. TISSUE LAYERS PROPERTIES.

Layer	**ρ, (kg/m^3)**	***k*, (W/m*K)**	***Cp*, (J/kg*K)**	**Thickness, (m)**
Tissue	1050	0.5	3766	0.12
Fat	850	0.16	2510	0.0025
Skin	1100	0.21	3250	0.002

TABLE II. BIOHEAT EQUATION INPUTS.

Tissue	**Q_{met}, (W/m^3)**	**ω_b, (1/s)**	**T_b, (K)**
Muscle	5	0,0001	310.15
Fat	0	4.5e-6	310.15
Skin	4	7.2e-8	310.15

In order to determine the distribution of temperature field in muscles during cooling, a modeling was performed using *COMSOL Multiphysics* software package; two cases which are typically relevant to researchers were analyzed: 1. Limb is cooled in a bath containing water of temperature 15 °C, i.e. temperature is set at T_0 = +15 °C. 2. Limb is cooled using ice pack or chilled gel pack, T_0 = 0 °C.

The temperature distribution in a muscle model, when the prescribed temperature T_0 was +15 °C, for illustration purposes is given in Fig. 2.

The temperature changes in tissues, from deep layers to the cooling agent (water), according to modeling results, are shown in Fig. 3.

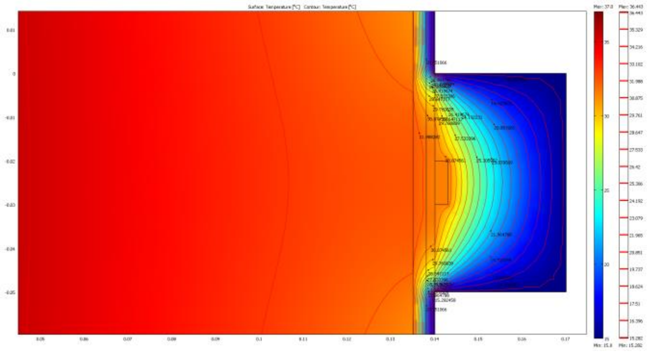

Fig. 2. Temperature distribution in tissues, under temperature sensor and in thermal insulation.

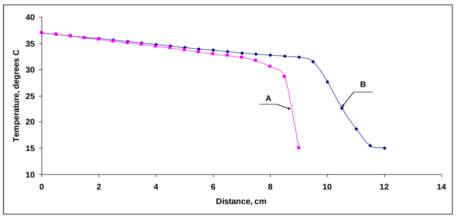

Fig. 3. Results of modeling, when T_0 = + 15^0 C: A – temperature distribution from deep layers to the surface of skin; B – temperature distribution from deep muscle layers to the outer surface of thermal insulation (perpendicular to the body surface).

It can be determined from modeling results, that the skin temperature underneath the sensor was 31.5 °C. The muscle temperature in an area not covered by thermal insulation at a depth of 30 mm was 31.9 °C. Hence the temperature difference is 0.4 °C.

III. RESULTS OF EXPERIMENTS

The structure diagram of temperature control experiment is shown in Fig. 4. Sensor–thermistor was used to measure temperature; its housing was protected from the cooling water by 3 cm thick layer of thermal insulation of *Thermoflex* type, with a thermal conductivity coefficient k = 0.028 (W/m*K).

The operation of temperature control system used for the experiments. Values of thermistor resistance R(T) are measured every 30 s. Then resistance data is converted to the temperature values using resistance–temperature converter and digitized using A/D converter. The temperature control data is transmitted wirelessly in the digital form to the data acquisition device (5), from where the data is transferred to PC for storage and visualization.

Data transmission via wireless network was used for patient electrical safety reasons. A temperature change versus time as indicated on a PC screen for illustrative purposes is visualized in Fig. 5.

The temperature control gear (device for processing results of measurements with temperature sensor-thermistor) was calibrated using thermostat which maintained the water temperature at ±0.02 °C and it was used reference thermometer type *Black Stack* with Thermistor scanner type 2554 and temperature sensor probe type 5610, accuracy ±0.015 °C. The uncertainty of calibrated temperature control gear was ±0.05 °C.

The visualization of temperature measurement results during experiment for illustrative purposes is given in Fig. 5.

Temperature change of tissue surface (skin) underneath the sensor during cooling versus time is shown in Fig. 6.

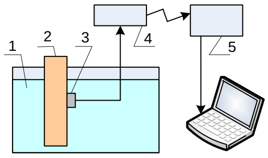

Fig. 4. The structure diagram of temperature control experiment in cooled tissues. Here: 1 – water, +15±1 ^{0}C; 2 – cooled limb; 3 – temperature sensor under the thermal insulation; 4 – device for processing results of measurements and data transmission over wireless network; 5 – data acquisition device.

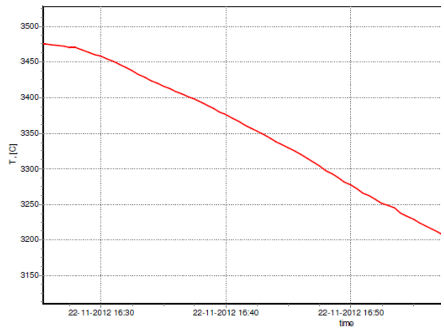

Fig. 5. Temperature change versus time, when it measuring underneath the thermal insulation layer. Y–axis contains temperature in ^{0}C x 100; X–axis contains cooling time.

Tissue surface (skin) temperature change versus time can be described by

$$T = 34.8359 - 0.1046t, \tag{8}$$

where T – tissue surface (skin) temperature, °C, t – time, min.

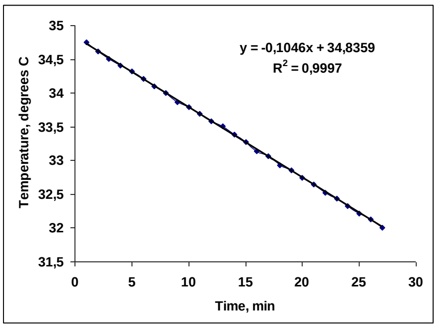

Fig. 6. Tissue surface temperature change during cooling versus time.

During limb cooling, the temperature of deeper muscle layers is expected to increase almost linearly when going deeper from the surface, as is can be seen from the simulation results (Fig. 3). The skin surface temperature under the sensor during cooling decreases almost linearly over time in accordance with the experimental data (Fig. 6). It is known [18] that by heating the surface of the skin the temperature of deeper muscle layers over the depth increases linearly. It was found [19] that, when the temperature of the surface (skin) decreases linearly over time during cooling, the temperature of muscles at the depth of 2 cm also decreases linearly. According to our and [19] experimental data, when cooling the limb for 30 minutes, the linear surface (skin) temperature decrease starts in 10–12 minutes after the cooling is initiated.

So we can expect that if the temperature of the surface (skin) under the sensor decreases linearly over time during cooling, the temperature of muscle layers should decrease linearly both over time and going deeper from the surface.

By using (8) it is possible to calculate time over which it is possible to cool deeper layers of muscles down to required temperature.

The muscle temperature control during cooling was repeated 10 times. Average skin surface temperature at the spot of sensor placement was 31.5 ± 0.65 °C at the end of cooling.

On the basis of the study results the structure of a hand-held device (Fig. 7) for control of temperature in deeper muscle layers via measurement, using thermistors as temperature sensor [20] of the skin surface temperature was designed.

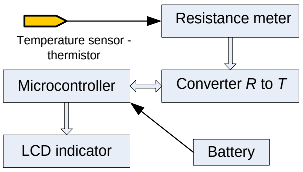

Fig. 7. Structure of hand-held device for control of temperature in deeper muscle layers.

It is relatively difficult to compare measurement results obtained in various studies, since different authors provide thermal conductivity of muscular tissue k in a range from 0.16 W/m·K [6], [16] to 0.51 W/m·K [17], and measurements using invasive method are made at such depth which is of interest to the researcher at particular time, ranging from 1 cm to 4 cm [13], [17], [21].

Temperature change in deeper layers is associated not only with the temperature of cooling agent, but also with the way the cooling packet is placed [19] or by placing persons in a climate chamber for immersion in a cold water [22].

Therefore the temperature control using non-invasive method is very important when proceeding with cooling after traumatic injuries, since it is also possible to control the cooling duration, which is related to the temperature decrease level in deeper layers.

IV. Conclusions

During limb cooling, it is possible to control the temperature changes in deeper layers of limb tissues using non-invasive method.

Non-invasive temperature control method can be used when performing cold therapy or research work.

On the basis of the study results the structure of a hand-held device for control of temperature in deeper muscle layers via measurement of the skin surface temperature, using thermistors as temperature sensors, was offered.

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