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# NUMERICAL MODEL OF HEAT TRANSFER IN SKIN LESIONS

Preliminary results of numerical modelling of skin undergoing thermal stimulation (mild cooling) in a human forearm is presented. Small brass compress was used for cooling purposes. The skin recovery process was then analysed. Temperature history for $N = 14$ samples was recorded using IR camera. The samples come from 8 male adults (age 25-38 years). A numerical model of heat transfer in tissues and CFD model of surrounding air (natural convection) was proposed. Simulation results were validated against experimental data.

**Keywords:** numerical modelling, validation, bioheat equation, skin lesions, malignant melanoma

## 1. Introduction

The accurate numerical models of living tissue are of high importance among numerous modern biomedical engineering challenges. Such models can not only allow one to understand processes involved, but they can help to develop a new treatment and/or equipment used to assist medical staff during diagnosis and controlled treatment process [3, 6].

The work presented here is a part of wider research project targeted in verifying the possibility of early diagnosis of skin lesions, with special interest in malignant melanoma identification.

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# 2. Materials and methods

## 2.1. Mathematical model

**Bioheat equation.** The passive part of heat transfer in the living tissues describes Pennes' bioheat equation [2]:

$$c\rho\frac{\partial T(\mathbf{r},t)}{\partial t}=\nabla\left[k(T)\nabla T(\mathbf{r},t)\right]+q_m(\mathbf{r},t)+\omega_b(\mathbf{r},t)c_b\rho_b\left[T_a-T(\mathbf{r},t)\right] \quad (1)$$

where: $c$, $c_b$ – specific heat (tissue, blood), $\rho$, $\rho_b$ – density (tissue, blood), $T$, $T_a$ – temperature (tissue, perfusing (artery) blood), $t$ – time, $k$ – tissue heat conductivity, $q_m$ – metabolic heat production rate, $\omega_b$ – blood perfusion rate, $\mathbf{r}$ – vector coordinate.

**Metabolism.** The metabolic heat production rate $q_m$ is a sum of the basal value $q_{m,0}$ and additional $\Delta q_m$ part, being result of autonomic thermoregulation:

$$q_m = q_{m,0} + \Delta q_m \quad (2)$$

where $q_{m,0}$ – metabolic heat production rate for thermoneutral conditions (i.e., body in thermal equilibrium with environment).

Under non-neutral conditions metabolic rates vary with the local tissue temperature. The dependency of temperature on metabolism is modelled according to the $Q_{10}$ relation. It states that for every 10 K reduction (change) in the tissue temperature, there is a corresponding reduction (change) in the cell metabolism $\Delta q_m$ by the factor $Q_{10} = 2$, as reported in [1]:

$$\Delta q_m = q_{m,0}\cdot\left[2^{(T-T_0)/10}-1\right] \quad (3)$$

where $T_0$ – basal temperature distribution (i.e., in thermoneutral conditions).

**Perfusion.** In non-neutral conditions the perfusion rate $\omega_b$ (i.e., tissue blood flow related to tissue value) varies with changes in regional metabolic rates as well. The dependency of the perfusion rate $\omega_b$ change on variations of the metabolic heat production rate $\Delta q_m$ (3) is linear [4]:

$$\Delta\beta = \mu_b \Delta q_m \quad (4)$$

where: $\beta = \omega_b c_b \rho_b$ – blood perfusion energy equivalent, while $\mu_b = 0.932\ \mathrm{K}^{-1}$ – empirically estimated proportionality constant [1, 5].

Using the above definition of $\beta$, the most right term of eq. (1), which is a source term arising from the arterial blood perfusing the tissue, reads:

$$q_p = \beta(T_a - T) = (\beta_0 + \Delta\beta)(T_a - T) = (\omega_{b,0} c_b \rho_b + \mu_b \Delta q_m)(T_a - T) \tag{5}$$

where $\omega_{b,0}$ – basal perfusion rate for thermoneutral conditions.

**Active thermoregulation.** Neither vasoconstriction/vasodilatation, shivering thermogenesis, nor sweating was introduced in the model at hand.

## 2.2. Numerical model

The numerical model of human forearm, cooling compress and surrounding air was developed. The numerical simulations were carried out using ANSYS Fluent 14 commercial CFD package (ANSYS Inc., USA). The additional source terms of heat conduction equation arising in bioheat transfer eq. (1) were introduced by means of UDF (user-defined function) functionality of the ANSYS Fluent code.

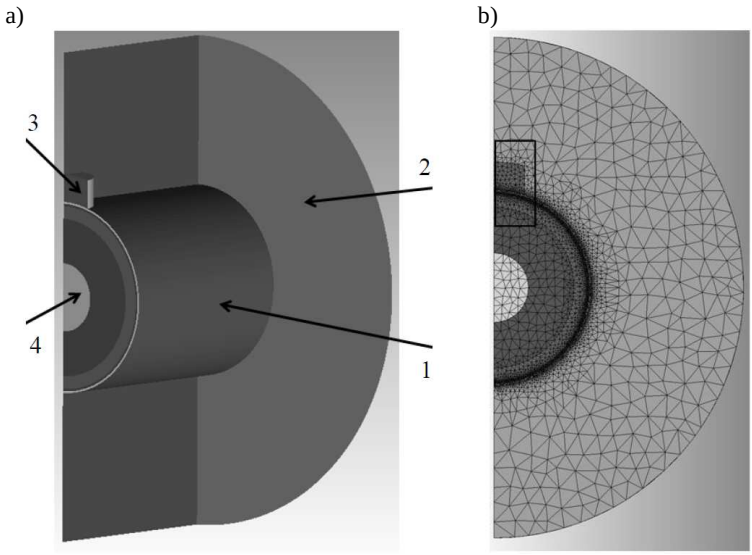


Fig. 1. Three-dimensional model of computational domain; 1 – forehand outer skin, 2 – surrounding air, 3 – cooling compress, 4 – tissues layers: bone, muscle, fat, inner skin, outer skin (a) and mesh of numerical partition (b)

**Geometry.** The 3-D geometrical model of the computational domain is presented in fig. 1. The human forearm was modelled as a cylinder surrounded by a cylindrical volume of air. Multiple concentric homogeneous layers represent-

ing different tissues types (bone, muscle, fat, inner skin, outer skin) were considered. The outer diameter of each layer is shown in tab. 1. The modelled length of arm was 200 mm which is enough to provide solution of independence of the boundary condition applied on the external boundaries: thermal insulation for tissue and pressure outlet for surrounding air (backflow air temperature 23°C). Due to the symmetry of the model, only a quarter part of the model was considered. Small cylindrical volume on top of the forearm represents the cooling compress used in the thermal stimulation (cooling) of skin.

**Numerical mesh** was generated using ICEM CFD (Ansys Inc., USA). Tetrahedral meshing scheme was used. In order to verify the influence of mesh size on the numerical solution, three grids were tested (namely: 0.3 M, 1.2 M and 5.5 M elements) in steady state simulations. For each case, the mesh was refined in cooling compress vicinity (i.e., where steepest temperature gradients were expected). Finally, based on results comparison, 1.2 M tetrahedral elements mesh was selected for further simulations. To check final quality of grid aspect ratio was tested (i.e., scaled ratio between the volume of the element and the radius of its circumscribed sphere power three). The mesh quality histogram is presented in fig. 2., where the worst element has aspect ratio above 0.3 while over 50% of elements have quality 0.6 and higher.

Table 1. Dimensions and properties of tissues and initial values of model variables (for steady state analysis)

| Tissue | Outer radius | Thermal cond. | Density | Specific heat | Perfusion rate | Metabolic heat production rate |
|---|---|---|---|---|---|---|
| | mm | $W \cdot m^{-1} \cdot K^{-1}$ | $kg \cdot m^{-3}$ | $J \cdot kg^{-1} \cdot K^{-1}$ | $s^{-1}$ | $W \cdot m^{-3}$ |
| Outer skin | 42.9 | 0.47 | 1085 | 3680 | 0 | 0 |
| Inner skin | 42.1 | 0.47 | 1085 | 3680 | 0.0011 | 631 |
| Fat | 41.1 | 0.16 | 850 | 2300 | 0.0000036 | 58 |
| Muscle | 35.3 | 0.42 | 1085 | 3768 | 0.000538 | 684 |
| Bone | 15.3 | 0.75 | 1357 | 1700 | 0 | 0 |

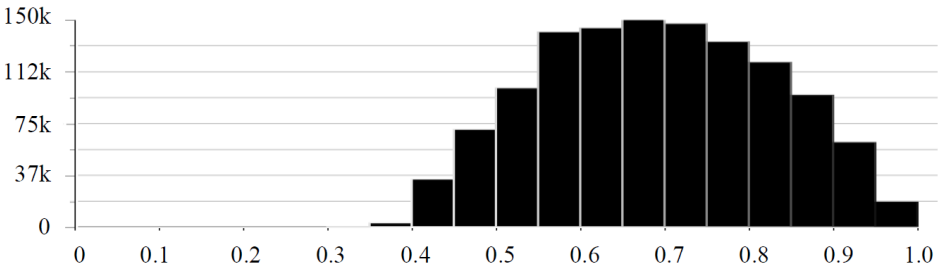


Fig. 2. Histogram of quality of numerical partition elements; 1 – the best, 0 – the worst

**Numerical discretization, equations solved and material properties.** The set of 4 equations was solved for each time step (arising from mass, momentum in 3D and energy conservation). Air flow was assumed laminar, with second order upwind scheme. Air density for natural convection was modelled using Boussinesq model approach. Operating conditions were taken as: pressure 101.325 kPa, temperature 288.15 K, density 1.225 $kg \cdot m^{-3}$. Gravity was set to 9.81 $m \cdot s^{-2}$ in vertical direction (cooling compress was placed on top of the forearm). Constant air properties were assumed: specific heat 1006.43 $J \cdot kg^{-1} \cdot K^{-1}$, thermal conductivity 0.0242 $W \cdot m^{-1} \cdot K^{-1}$, viscosity 1.79E-05 $kg \cdot m^{-1} \cdot s^{-1}$ and thermal expansion coefficient 0.00343 $K^{-1}$.

**Time step.** For transient part of the simulation the sensitivity analysis of time discretization was done as well. Three time steps were selected and tested against their influence on the results, namely: 0.25 s, 0.5 s and 1 s. Taking into account obtained results and computation time, the time step 0.5 s was selected for transient simulation in the current study.

### 2.3. Experiment

The proposed numerical model of skin cooling and rewarming processes was validated against experimental data collected from subjects examined in course of pilot medical experiment (being part of wider research project). The medical ethical committee of Maria Skłodowska-Curie Memorial Cancer and Institute of Oncology Gliwice Branch approved the study. Each subject gave written consent prior to participation in the study. For the initial analysis at hand, the group of eight adult males was selected. Subject's characteristics (mean ± SD) are: age 31.1 ± 5.0, height 1.80 m ± 0.07 m, weight 102.0 ± 10.4 kg. The studied skin sites were dorsal and ventral side of the left forehand halfway the wrist and inner side of the elbow. The subjects were asked to stay sited for 15 minutes prior to the measurements.

The skin temperature history was recorded using PI160 (Optris GmbH, Germany) infrared camera (160 x 120 px, 120 Hz, LWIR, 7.5-13.0 μm detector, standard lens 23° x 17°). Cooling of skin was done by means of brass cooling compress at stabilized initial temperature 6-7°C. The basal (thermoneutral) skin temperatures were measured by wireless iButton DS1922L (Maxim Integrated, USA) temperature data logger attached using adhesive tape near cooling zone (approx. 3 cm). Room temperature was measured using Almemo 2390-5 multifunction measuring instrument (AHLBORN, Germany) equipped with K-type thermocouple sensor.

## 3. Numerical simulations

Numerical computations were carried out in three steps: (a) steady state simulation tuning model variables to mimic thermoneutral conditions being ini-

tial state for following step; (b) transient simulation of skin cooling; (c) transient simulation of skin recovering from local cooling.

Steady state simulation was performed in order to get an initial temperature distribution and tune model variables: metabolic heat production rate $q_m$ and perfusion rate $\omega_b$ as in the thermoneutral state (i.e., model in thermal equilibrium with environment). During this step the type of volume representing cooling compress was set to *fluid* having the same properties as surrounding air. Uniform constant temperature of 23°C was prescribed for the whole computational domain (both forearm and air) and uniform zero velocity was prescribed within the air domain. Outer radius of all tissue layers of the modelled forearm, as well as tissue properties [4] and initial metabolic heat production rates and perfusion rates are presented in tab. 1.

The temperature (see fig. 3.) and velocity distribution being the result of steady state analysis is then prescribed as initial condition for further transient calculations. In addition, the resulting temperature distribution, metabolic heat production rate and perfusion rate are thereafter treated as the basal metabolic heat production rate $q_{m,0}$, basal blood perfusion rate $\omega_{b,0}$ and basal temperature distribution $T_0$ entering eqs. (2), (3) and (5).

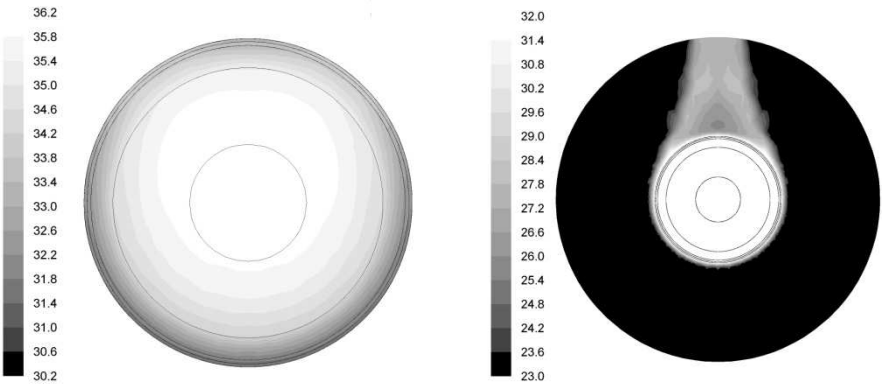


Fig. 3. Initial temperature distribution (in symmetry plane of the compress) for transient analysis: tissues (top) and surrounding air (bottom)

**Cooling.** To mimic the skin cooling procedure, during the first step of transient analysis, the material type of volume representing cooling compress was temporarily switched to *solid* having properties of brass: density 8500 $kg \cdot m^{-3}$, specific heat 380 $J \cdot kg^{-1} \cdot K^{-1}$ and thermal conductivity 110 $W \cdot m^{-1} \cdot K^{-1}$ and prescribed uniform initial temperature 7°C. Then transient simulation was run to simulate 15 s of local skin cooling.

**Recovery.** Once simulation of skin cooling is finished, the temperature and velocity field as well as model variables were stored and used as initial values for the next step. The material type of volume representing cooling compress is

then switched back to *fluid* having properties of air (uniform temperature 23°C). Then transient simulation of skin recovering from local cooling is executed for total time 100 s after cooling is finished.

## 4. Results

The mean skin temperature history ± SD of $N$ = 14 samples is shown in fig. 4. (each temperature represents the coldest skin surface point in area where cooling compress was applied).

**Contact resistance.** As first attempt, ideal contact between cooling compress and the skin was assumed. However simulated temperature response was far below those measured (see dashed black line in fig. 4.). To simulate non-ideal contact conditions taking place during experiment (being result of hair presence, limited cooling compress pressure, etc.) contact resistance was added. The simulation results for contact resistance 0.001 $m^2 \cdot K \cdot W^{-1}$ are shown in fig. 4. (solid black line).

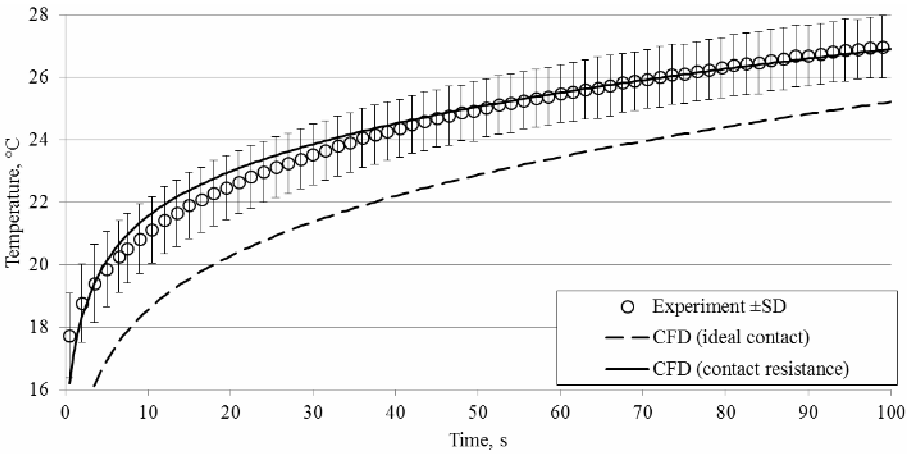


Fig. 4. Skin temperature variation vs. time for measuring data and CFD simulation, the results are presented for time after cooling compress is removed

## 5. Conclusion

As the result of preliminary research the numerical model of the human forearm undergoing the skin thermostimulation was developed and validated. However, further research is needed towards the proper contact resistance determination, as well as in implementation of additional active thermoregulation models (including vasoconstriction and vasodilatation).

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## MODEL NUMERYCZNY WYMIANY CIEPŁA W USZKODZENIACH SKÓRNYCH

**S t r e s z c z e n i e**

W pracy zaprezentowano wstępne wyniki modelowania numerycznego procesów wymiany ciepła w rejonie skóry przedramienia poddanej termostymulacji (łagodnego ochładzania). Do ochładzania użyto kompresów mosiężnych. Przeanalizowano proces powrotu skóry do stanu sprzed termostymulacji. Przy użyciu kamery termowizyjnej zarejestrowano rozkład temperatury dla $N$ = 14 próbek w grupie 8 przebadanych dorosłych mężczyzn (w wieku 25-38 lat). Zaproponowano model numeryczny przepływu ciepła w tkankach przedramienia oraz w otaczającym je powietrzu (w warunkach konwekcji swobodnej). Wyniki symulacji zostały poddane walidacji przy użyciu danych pochodzących z pomiarów.

**Słowa kluczowe:** modelowanie numeryczne, walidacja, równanie biociepła, zmiany skórne, czerniak złośliwy

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