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Modelling and validation of transient heat transfer processes in human skin undergoing local cooling

Abstract. *The analysis of transient heat transfer processes in human forearm skin undergoing local thermal stimulation (mild cooling by means of brass compress) is presented. A numerical model of heat transfer within: living tissues (Pennes bioheat equation) and metal compress used for skin cooling is proposed. Simulation results were validated against* in vivo *measurement data of heat flux and compress temperature for N=8 samples in four male adults (age 25–38 years).*

Streszczenie. *W pracy przedstawiono analizę niestacjonarnych procesów wymiany ciepła w skórze przedramienia poddanej lokalnej termostymulacji (łagodnego ochładzania przy pomocy mosiężnych kompresów). Zbudowano model numeryczny procesów przepływu ciepła w: tkankach (równanie biociepła Pennes'a) oraz kompresie używanym do chłodzenia. Otrzymane wyniki poddano walidacji na podstawie pomiarów* in vivo *gęstości strumienia ciepła oraz temperatury kompresu dla N=8 próbek w grupie czterech przebadanych dorosłych mężczyzn w wieku 25-38 lat.**(Modelowanie oraz walidacja niestacjonarnych procesów wymiany ciepła w skórze poddanej lokalnemu ochładzaniu)***

Keywords: modelling, validation, bioheat equation, thermal physiology
Słowa kluczowe: modelowanie, walidacja, równanie biociepła, termofizjologia

Introduction

The accurate numerical models of living tissue can play crucial role in modern biomedical engineering. Such models can not only allow one to understand processes itself, but they can help to develop a new treatment and/or equipment used to assist medical staff during diagnosis and controlled treatment process [7].

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

In previous work of the research team [2] thermographic (IR camera) measurements of skin recovering from local cooling was used to validate the numerical model. However, during above mentioned research, additional model validation possibilities were identified.

Current work is a follow up research targeted on model validation employing skin - cooling compress interfacial heat flux measurements. New experimental setup, including custom design cuboid brass cooling compress, was proposed to measure and record transferred heat flux using differential thermopile sensor.

Mathematical model

Bioheat equation. The passive part of heat transfer in the living tissues is described by Pennes' bioheat equation [3]

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

where: c, c_b – specific heat (tissue, blood respectively) [J·kg^{-1}·K^{-1}]; ρ, ρ_b – density (tissue, blood respectively) [kg·m^{-3}]; T, T_a – temperature (tissue, perfusing (artery) blood respectively) [K]; t – time [s], k – tissue heat conductivity [W·m^{-1}·K^{-1}]; q_m – metabolic heat production rate [W·m^{-3}]; ω_b – blood perfusion rate [s^{-1}], $\mathbf{r}$ – vector coordinate.

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

$$q_m = q_{m,0} + \Delta q_m \tag{2}$$

where $q_{m,0}$ is the metabolic heat production rate for thermoneutral conditions (i.e. body in thermal equilibrium with environment). Under non-neutral conditions metabolic rates

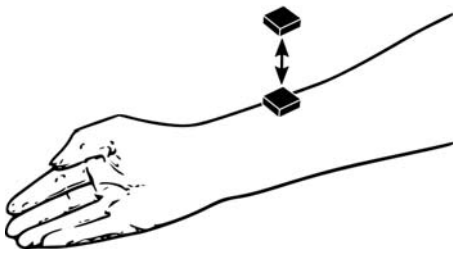

Fig. 1. Human forearm skin undergoing local cooling by means of cuboid brass compress

vary with the local tissue temperature. The dependency of metabolism on temperature is modelled according to the Q_{10} relation. It states that for every 10K reduction (change) in the tissue temperature, there is a corresponding reduction (change) in the cell metabolism Δq_m by the factor Q_{10}, as reported in [1]

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

where T_0 is basal temperature distribution (i.e. in thermoneutral conditions).

Perfusion. In non-neutral conditions the perfusion rate ω_b (i.e. tissue volumetric blood flow per unit volume of tissue per unit time) varies with changes in regional metabolic rates as well. The dependency of the perfusion rate ω_b change on variations of the metabolic heat production rate Δq_m (3) is linear [5]

$$\Delta\beta = \mu_b\Delta q_m \tag{4}$$

where $\beta = \omega_b\, c_b\, \rho_b$ is the blood perfusion energy equivalent [W·m^{-3}·K^{-1}] while $\mu_b = 0.932\text{K}^{-1}$ is empirically estimated proportionality constant [1, 6].

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

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

where $\omega_{b,0}$ is the basal perfusion rate for thermoneutral conditions, cf. $q_{m,0}$ arising in eq. (2).

Active thermoregulation. No vasoconstriction/vasodilatation, shivering thermogenesis, nor sweating was introduced in the model at hand.

Numerical model

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

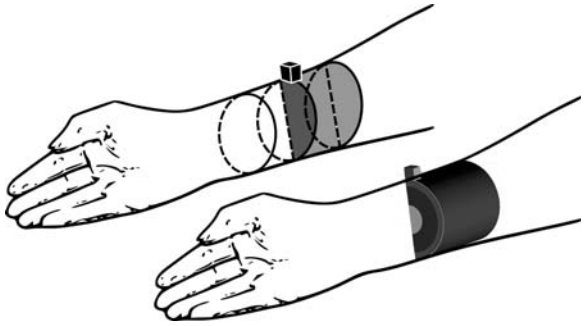

Fig. 2. Scope of modelled region (200mm long section of human forearm), due to symmetry only 1/4 of forearm geometry (gray) and compress (black) is considered (upper sketch). 3-D geometrical model of computational domain representing section of human forearm (lower sketch)

Model geometry. The geometrical model of human forearm and cuboid shape compress was developed. Common practice when modelling of human heat transfer and thermal physiology is to simplify complicated structure of human tissues. The part of human forearm under consideration is treated as cylinder with multiple concentric homogeneous layers representing different tissue types, namely: bone, muscle, fat, inner skin, outer skin (see Figure 3). The outer radius, as well as material properties of every tissue layer is shown in Table 1.

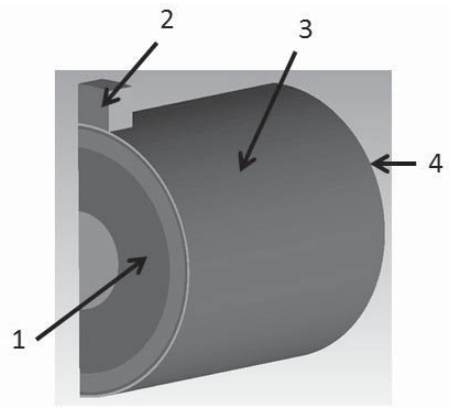

Fig. 3. 3-D geometrical model of computational domain (1 – tissues layers: bone, muscle, fat, inner skin, outer skin, 2 – cooling compress, 3 – forehand outer skin 4 – isolated wall, i.e back & front)

As the cooled area of skin is small (approx. 24mm x 24mm) and the cooling time is only 15s, the modelled length of arm was reduced to 200 mm of forearm. It is well enough to prove solution independence on the external boundaries of the model: thermal insulation for tissue sections planes and heat convection/insulation for skin surface (see Figure 3). Due to the symmetry of such model, only a quarter part of the model was considered (see Figure 2). Small cuboid volume on top of the forearm represents the cooling compress used in the thermal stimulation (cooling) of skin. Geometrical dimensions of the compress are: 12.3mm (height) x 24mm (width) x 24mm (length). Overview of the model and the boundary conditions are presented in Figure 3.

Numerical mesh was generated using ICEM CFD (Ansys Inc., USA). O-grid approach was performed in order to provide high quality hexahedral elements. To verify spatial discretization of mathematical model, three grids with different element sizes and total number of elements were examined. Each grid was refined in areas where high gradients of temperatures and fluxes where anticipated. Finally, based on results of comparison, 1.4M cells mesh was chosen for further simulations. Selected regions of numerical mesh are shown in Figure 4.

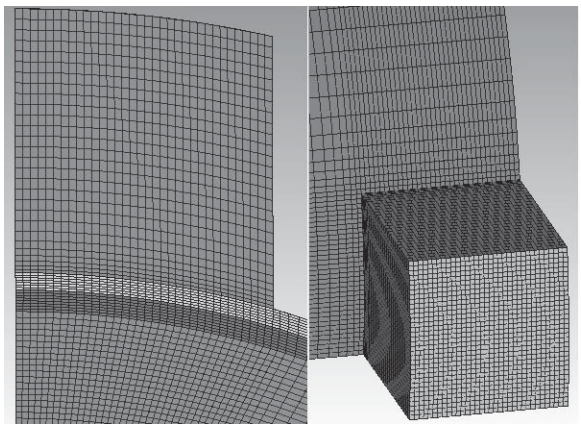

Fig. 4. Selected regions of numerical mesh: in symmetry plane (left) and seen from above the compress (right)

Numerical discretization, equations solved and material properties. The set of equations was solved for each time step (arising from mass, momentum in 3D and energy conservation principles). For transient part of the simulation the sensitivity analysis of time discretization was done as well. Taking into account obtained results and computation time, the time step 0.5s was selected for transient simulation in the current study.

Experimental data

The proposed numerical model of skin cooling process was validated against experimental data collected in course of pilot medical experiment (being part of wider research project).

Ethical approval. The medical ethical committee affiliated at 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.

Subjects. For the analysis at hand, the group of four adult males was selected. Subject's characteristics (mean $\pm$ SD) are: age 30.8 $\pm$ 5.4, height 1.83 m $\pm$ 0.07 m, weight 98.5 $\pm$ 16.5 kg. The studied skin sites were dorsal and ventral side of the left forehand halfway the wrist and inner side of the elbow, resulting in N=8 samples. The subjects were asked to stay sited for 15 minutes prior to the measurements.

Measurement equipment. The temperature of the brass compress (at mid and top plane) was measured using calibrated T-type thermocouples of diameter 0.5mm (CZAKI THERMO-PRODUCT, Poland).

The Micro-Fiol®(RdF Corp., USA) model 27160 Heat Flux Sensor was used to measure heat flux on brass compress/skin interface. This sensor is a differential thermopile.

Table 1. Tissue dimensions, properties and initial values of model variables (for steady state analysis)

Tissue	Outer radius r [mm]	Thermal conductivity k [W·m^{-1}·K^{-1}]	Density ρ [kg·m^{-3}]	Specific heat c [J·kg^{-1}·K^{-1}]	Perfusion rate ω [s^{-1}]	Metabolic heat production rate q_m [W·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

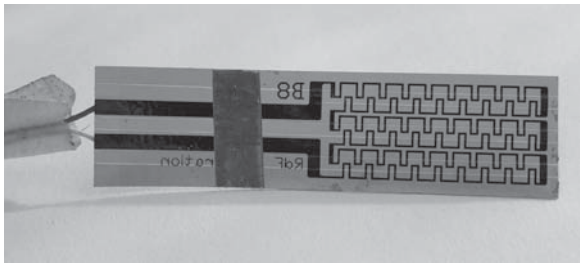

Fig. 5. Micro-Fiol®heat flux sensor (RdF Corp., USA)

Heat passing through a calibrated polyimide membrane produces a small temperature difference. The signal is proportional to the difference in temperature and the number of junctions in the thermopile [4]. Hence the active sensor ares is 12mm x 24mm, the set of 2 heat flux sensors was used to cover whole skin/compress interfacial area.

All data were acquired and recorded using personal computer (PC) running LabView Signal Express software and 24-Bit Universal Analog Input cards, type NI9214 (National Instruments, USA).

Laboratory setup. Cooling of skin was done by means of brass cooling compress at stabilized initial temperature 7(8)°C by means of thermoelectric cooling device (in house).

In order to prevent experiment from being disturbed by external heat sources (sinks), the cooling compress was isolated by extruded polystyrene foam (see figure 6). The compress covered (side & top) with expanded polystyrene block was placed on cooling device first. Once stable and uniform temperature of brass compress was reached, compress together with EPS insulation was put on the subject forearm skin. The temperatures of compress (center - mid & top plane) and heat flux were then recorded for 15 seconds.

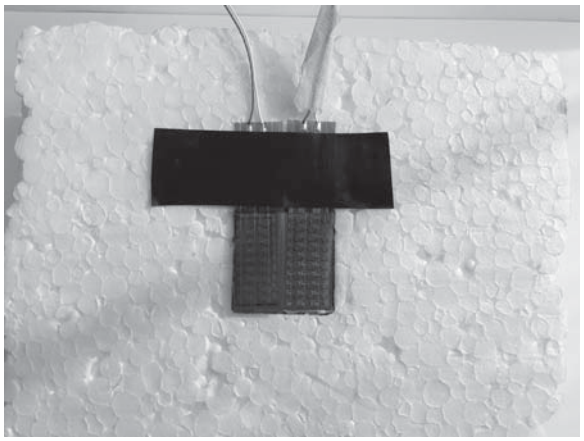

Fig. 6. Measurement setup - brass compress (not visible) located below heat sensors attached to the block of expanded polystyrene (heat insulation)

Numerical simulation

Numerical computations were carried in two steps:

- steady state simulation of thermoneutral conditions being initial state for following step;
- transient simulation of skin cooling.

Thermoneutral state. 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 ω_b as in the thermoneutral state (i.e. model in thermal equilibrium with environment). During this step type of volume representing cooling compress was turned off. On the outer skin the convection boundary conditions with convective heat transfer coefficient 5W/(m^2·K) and air temperature 23°C was specified. Tissue properties, initial metabolic heat production rates and perfusion rates used in steady state calculations were presented in Table 1 after [5].

The temperature distribution being result of steady state analysis (shown in Figure 7) is then prescribed as initial condition for further transient calculations. In addition, the resulting distributions of: temperature, metabolic heat production rate and perfusion rate are thereafter treated as:

- basal temperature distribution T_0,
- basal metabolic heat production rate $q_{m,0}$, and
- basal blood perfusion rate $\omega_{b,0}$

entering eqs. (2), (3) and (5).

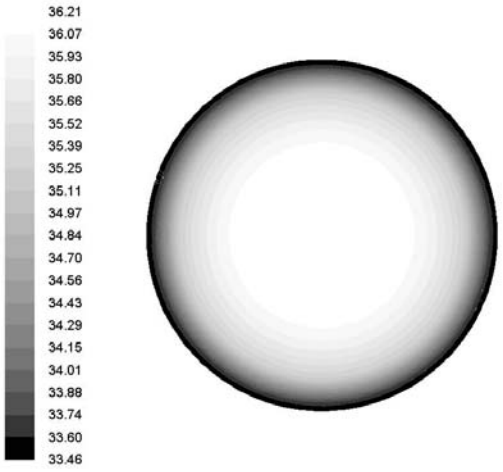

Fig. 7. Initial tissue temperature distribution (in symmetry plane) for transient analysis

Skin cooling. Transient computations were then conducted to mimic the skin and compress cooling procedure, the material type of volume representing cooling compress was turned on and switched to solid having properties of brass: density 6831.5kg·m^{-3}, specific heat 380J·kg^{-1}·K^{-1} and thermal conductivity 110W·m^{-1}·K^{-1} and prescribed uniform initial temperature 8°C. For this step, the skin surface as well as outer surface of compress were insulated (as the compress was covered with thermal insulation block). Then transient simulation was started to mimic 15s of local skin cooling.

Contact resistance. As first attempt, ideal contact between cooling compress and the skin was assumed in the model at hand. However simulated heat fluxes as well as compress temperature response was far of those measured,

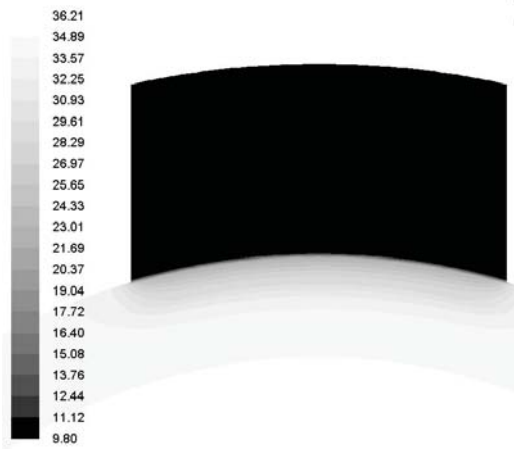

Fig. 8. Tissue ans compress temperature distribution (in symmetry plane) for time 15s of simulation

as reported in [2]. The origin of such behaviour are identified (among others) as being result of:

- the presence of hairs,
- limited allowable cooling compress pressure on skin,
- additional contact resistance introduced by MicroFoil® heat flux sensor itself.

To simulate non-ideal contact conditions taking place during experiment the contact resistance of 0.00025 $m^2 \cdot K \cdot W^{-1}$ was implemented in numerical model (on the skin-compress interface).

Results

Compress temperature. The tissue ans compress temperature distribution (in symmetry plane) for time 15s of simulation is shown in Figure 8. Figure 9 shows the CFD simulation result (solid black line) compared to the mean ±SD of N=8 recorded compress temperature history samples (compress temperature at the center midplane point).

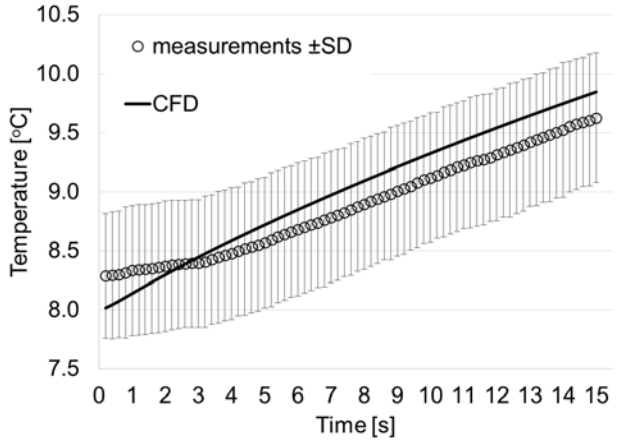

Fig. 9. Compress center midplane temperature vs. time for experiment (mean ±SD, N=8 samples) and CFD simulation

Total heat flux. Figure 10 shows the CFD simulation result (solid black line) compared to the mean ±SD of N=8 recorded total heat flux history samples (for compress-skin interface). Especially for first five seconds (the highest heat fluxes) good model agreement with measured data is observed.

Discussion and concluding remarks

The numerical model of the section of the human forearm undergoing the skin thermostimulation was developed and validated. Comparison of the numerical model response

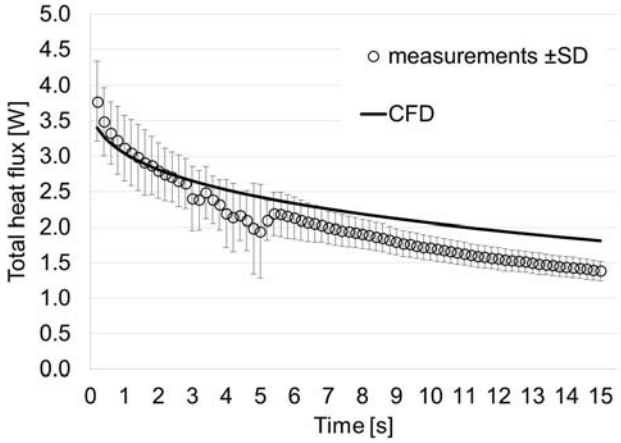

Fig. 10. Total heat flux on compress/skin interface vs. time for experiment (mean ±SD, N=8 samples) and CFD simulation

with experimetal data shows that the model roughly meets the *in vivo* measurements. Further model improvements are needed in the field of implementation of active thermoregulation models (including local & global vasoconstriction and vasodilatation).

Because relatively strong sensitivity of the modelled temperature field on input material properties was noticed, there is also room for improvements in the determination of compress material properties. Material laboratory tests are ongoing.

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