

## DETERMINING OF THE NEONATAL THERMAL MODEL PARAMETERS USING INVERSE THERMAL ANALYSIS

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**Abstract.** In this paper computational model of the numerical thermal model within the neonatal body cable of controlling the brain cooling process is presented. The main aim of the analysis was to find the proper parameters of the process. The simplified geometric model of newborn's body, consisting of the number of compartments, was built using Design Modeler. The neighbouring compartments exchange heat by conduction through tissues but also interact with each other through flowing blood. To simulate the natural heat transfer processes through the human tissues the Pennes bioheat equation was implemented into Ansys Fluent, using UDF (User Defined Function) capability. Model parameters are determined using inverse thermal analysis.

### 1 INTRODUCTION

The perinatal hypoxic-ischemic encephalopathy is a fairly frequent problem which the neonatologists have to cope with. Until recently there was not any method to remedy this disease. At present the therapeutic mild hypothermia of the infant's brain is more and more frequently used [3, 5]. Such hypothermal therapy generally stops apoptosis of neurons. However, to counteract the side effects of violent modifications of the temperature during the hypothermal therapy, the deep body temperature should be kept at the constant and secure level.

In order to perform the treatment in a proper way a distribution of the temperature inside the neonatal body (especially inside the brain) during the whole therapy has to be known. Since it was already proved [5] that temperature level of 34-34.5°C is secure during brain cooling, at the beginning of the therapy the deep body temperature is decreased to that level. In the case of neonate suffering from the hypoxia his temperature is usually decreasing spontaneously 0.5 K per hour, thus the deep body temperature at the certain moment is

appropriate to start therapy. If not, his temperature has to be reduced utilising cooling helmet or other device. Next, during the therapy, the deep body temperature is maintained at the constant level. Since the most vulnerable organ is the brain, the temperature of the scalp is kept by cooling helmet at the lowered level equal to about  $24^{\circ}\text{C}$ . Simultaneously, in order to maintain the deep body temperature at the secure level, the skin of the rest of the neonatal body has to be heated by the radiant warmer. The therapy is carried out for 72 hours and then the deep body temperature is gradually increased to the normal level (about  $37^{\circ}\text{C}$ ). It is very important to guarantee that temperature augments not faster than 0.5 K per hour, what entails many enormous difficulties.

In this work the attempt is made to develop a numerical thermal model within the neonatal body cable of controlling the brain cooling process. The main aim of the analysis was to find the proper parameters of the process. The simplified geometric model of newborn's body, consisting of the number of compartments [1, 3], was built using Design Modeler. As the shape of the head an ellipsoid was used. The forms of neck and trunk were simplified to elliptic cylinders. The limbs were modelled as cylinders. Inside the head the scalp, skull and brain were separated. The brain was surrounded by the brain cerebrospinal fluid. The remaining part of the head was modelled as a space occupied by the air. The neck consisted of the spine, muscle and skin. The spine was lengthened to the torso. Interior of the trunk was divided into lungs and viscera, which were surrounded by bone and skin. Limbs were split into three tissues: bone, muscle and skin. The built geometric model of the half of the newborn's body and the more detailed snapshot of the built geometric model of the head are presented in Figure 1.



**Figure 1:** The built geometric model of the neonate's body (left) and of the head (right)

The neighbouring compartments exchange heat by conduction through tissues but also interact with each other through flowing blood. To simulate the natural heat transfer processes through the human tissues the Pennes bioheat equation [4] was implemented into Ansys Fluent [2], using UDF (User Defined Function) capability.

## 2 GOVERNING EQUATIONS

The analysis of heat transfer processes during brain cooling should generally be carried out as a transient one, what eventually can lead to the steady-state. However, it should be noted that the heat capacity of the blood within one numerical cell is small comparing with the heat capacity of the tissue and as a consequence equations for arterial and vein blood temperature can be formulated as steady-state, but for the consecutive time steps.

As already mentioned, it is assumed that transient heat transfer in human tissues is

governed by Pennes bioheat equation [4]:

$$\rho_t c_t \frac{\partial T_t}{\partial t} = \nabla \cdot (k_t \nabla^2 T_t) + \rho_b c_b \omega_t (T_{a,k} - T_t) + \zeta_t \dot{q}_{met,t} \quad (1)$$

where subscripts  $()_t$ ,  $()_b$  and  $()_a$  refer to tissue, blood and artery, respectively. Symbol  $T$  stands for temperature (in K),  $k_t$ , W/(mK), is a tissue thermal conductivity,  $\rho$ , kg/m<sup>3</sup>, its density while  $c$ , J/(kgK) is a specific heat. Quantity  $\dot{q}_{met,t}$ , W/m<sup>3</sup> represents a metabolism and is an amount of energy which is generated within a unit time and a unit volume of a tissue. Blood perfusion is marked by  $\omega_t$ , 1/s, time is depicted by  $t$ , s, and  $\zeta_t$  is a weighting factor varying between 0 and 1.

Obviously, for steady-state one can write:

$$0 = \nabla \cdot (k_t \nabla^2 T_t) + \rho_b c_b \omega_t (T_{a,k} - T_t) + \zeta_t \dot{q}_{met,t} \quad (2)$$

The blood perfusion rate and the metabolic heat rate are depended on the tissue temperature and they can be calculated in the following way:

$$\omega_t = \omega_{t,bas} (3^{(T_t - T_0)/10}) \quad (3)$$

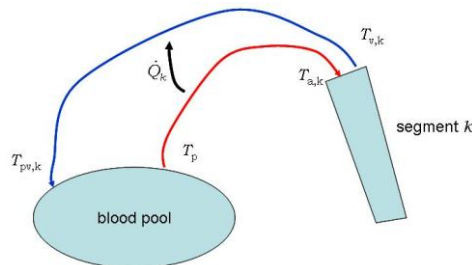
$$\dot{q}_{met,t} = \dot{q}_{met,t,bas} (3^{(T_t - T_0)/10}) \quad (4)$$

where  $\dot{q}_{met,t,bas}$ , W/m<sup>3</sup>,  $\omega_{t,bas}$  1/s, stand for the metabolic heat rate and the blood perfusion rate in the reference temperature  $T_0$ , K equal to 310.15 K.

As already mentioned, the geometric model is divided into sectors: a head, a face, a neck, a shoulder, a forearm, a left palm, a leg, a left foot, a thorax and an abdomen. Each sector, identified by index  $()_k$ , has its own arterial blood temperature,  $T_{a,k}$ , (in K) vein blood temperature,  $T_{v,k}$ , (in K), and blood mass flow,  $\dot{m}_k$ , kg/s. In the analysis it is assumed that mass flows of the artery and the vein bloods are equal. Each  $k$  sector is also divided into appropriate number of tissues, identified by index  $()_j$ , while index  $()_i$  in the derived equations refers to a particular numerical cell.

In the developed model it is assumed that the circulatory system contains the mixing spot called *the central blood pool* gathering the vein blood from all sectors and distributing the arterial blood to each sector. The temperature of the central blood pool,  $T_p$ , K, is common for the whole body and is identified with the rectal temperature of the newborn.

Let's consider now a single segment, identified by index  $k$ , which is receiving from the blood pool arterial blood having temperature  $T_{a,k}$  and sending to blood pool a vein blood having temperature  $T_{v,k}$ . This situation is schematically shown in Figure 2.



**Figure 2:** The built geometric model of the neonate's body (left) and of the head (right)

The temperature of the arterial blood leaving the blood pool is equalled to  $T_p$  and the temperature of the arterial blood reaching the segment  $k$  is equalled to  $T_{a,k}$ . The temperature of the vein blood leaving the segment  $k$  is equalled to  $T_{v,k}$  whilst the temperature of the vein blood reaching the blood pool is equalled to  $T_{pv,k}$ . While the arterial blood flows from the blood pool to the segment  $k$  and in the same time the vein blood flows from the segment  $k$  to the blood pool. Therefore, the amount of heat equalled to  $\dot{Q}_k$  is exchanged between the arterial and the vein vessel and can be described as:

$$\dot{Q}_k = \dot{m}_{a,k} c_b (T_p - T_{a,k}) \quad (5)$$

$$\dot{Q}_k = \dot{m}_{v,k} c_b (T_{pv,k} - T_{v,k}) \quad (6)$$

where  $\dot{m}_{a,k}$  stands for the arterial blood mass flow rate reaching the segment  $k$  whilst  $\dot{m}_{v,k}$  stands for the vein blood mass flow rate leaving the segment  $k$ .

The heat flow  $\dot{Q}_k$  can be also expressed as

$$\dot{Q}_k = k_k (T_{a,k} - T_{v,k}) \quad (7)$$

where  $k_k$ , W/(m<sup>2</sup>K) stands for a counter-current heat exchange coefficient for the segment  $k$  and takes into consideration the arrangement of the circulatory system. Connecting equations (5) - (7) and assuming that

$$\dot{m}_k = \dot{m}_{a,k} = \dot{m}_{v,k} \quad (8)$$

the following relationship can be received:

$$T_{a,k} = \frac{h_k \cdot T_{v,k} + c_b \cdot \dot{m}_k \cdot T_p}{h_k + c_b \cdot \dot{m}_k} \quad (9)$$

To determine the temperature of the arterial and vein blood, the each segment was divided into tissues. In the model the following 6 tissues have been distinguished: skin, muscle, brain, viscera, lungs and bone. They are identified by index ( $j$ ).

The blood mass flow for the segment  $k$  is related to the blood perfusion rate for the particular type of tissue  $j$  by the following relationship:

$$\dot{m}_k = \rho_b \dot{V}_k = \rho_b \sum_{j=1}^{J_k} \dot{V}_{j,k} = \rho_b \sum_{j=1}^{J_k} (\omega_{j,k} V_{j,k}) \quad (10)$$

where  $J_k$  represents the total amount of all types of tissue of the segment  $k$ ,  $\omega_{j,k}$  stands for the blood perfusion rate for the tissue  $j$  in the segment  $k$ ,  $\dot{V}_k$  is the blood volume flow rate in the segment  $k$ , whilst  $V_{j,k}$  stands for the volume of the tissue  $j$  in the segment  $k$  and is the sum of the volumes  $V_{i,j,k}$  of all numerical cells  $i$  in the tissue  $j$  in the segment  $k$ :

$$V_{j,k} = \sum_{i=1}^{N_{j,k}} V_{i,j,k} \quad (11)$$

where  $N_{j,k}$  represents the total amount of numerical cells  $i$  in the tissue  $j$  in the segment  $k$ . Thus  $T_{a,k}$  is eventually calculated by the following relationship:

$$T_{a,k} = \frac{h_k \cdot T_{v,k} + \rho_b c_b T_p \sum_{j=1}^{J_k} \left[ \omega_{j,k} \left( \sum_{i=1}^{N_{j,k}} V_{i,j,k} \right) \right]}{h_k + \rho_b c_b \sum_{j=1}^{J_k} \left[ \omega_{j,k} \left( \sum_{i=1}^{N_{j,k}} V_{i,j,k} \right) \right]} \quad (12)$$

The heat transfer  $h_k$  accounts for the differences of the arterial and blood temperatures as dependent on the distance from the blood pool. Since the neonate's body which is generally small, comparing to an adult's body, it is assumed here that the value of  $h_k$  is generally small (almost equal to 0) for each segment.

It is assumed that the vein blood temperature  $T_{v,i,j,k}$  leaving a given numerical cell  $i$  in tissue  $j$  in a segment  $k$  is equalled to the temperature of the considered numerical cell  $T_{t,i,j,k}$ . Additionally, the energy of the vein blood leaving the tissue  $j$  in a segment  $k$  is equalled to the total energy of the vein blood leaving all numerical cells in the tissue  $j$ , as well as that the heat capacity and density of blood is common for whole body, the temperature of the vein blood is expressed by:

$$T_{v,k} = \frac{\sum_{j=1}^{J_k} \omega_{j,k} \left[ \sum_{i=1}^{N_{j,k}} (V_{i,j,k} T_{t,i,j,k}) \right]}{\sum_{j=1}^{J_k} \omega_{j,k} \left[ \sum_{i=1}^{N_{j,k}} V_{i,j,k} \right]} \quad (13)$$

In the central blood pool the blood from each segment is mixed. After some simple algebra manipulations [6] one can obtain the equation to determine blood pool temperature  $T_p$  which is extension of result given by Fiala et. al., [1]:

$$T_p = \frac{\sum_{k=1}^K \left\{ \frac{\dot{m}_k \cdot c_b^2 \rho_b \sum_{j=1}^{J_k} \omega_{j,k} \left[ \sum_{i=1}^{N_{j,k}} (V_{i,j,k} \cdot T_{t,i,j,k}) \right]}{h_k + c_b \cdot \dot{m}_k} \right\}}{\sum_{k=1}^K \left\{ \frac{(c_b \cdot \dot{m}_k)^2}{h_k + c_b \cdot \dot{m}_k} \right\}} \quad (14)$$

The blood mass flow for the segment  $k$  is related to the blood perfusion rate for the particular type of tissue  $j$  by the following relationship:

Obviously, for steady-state one can write:

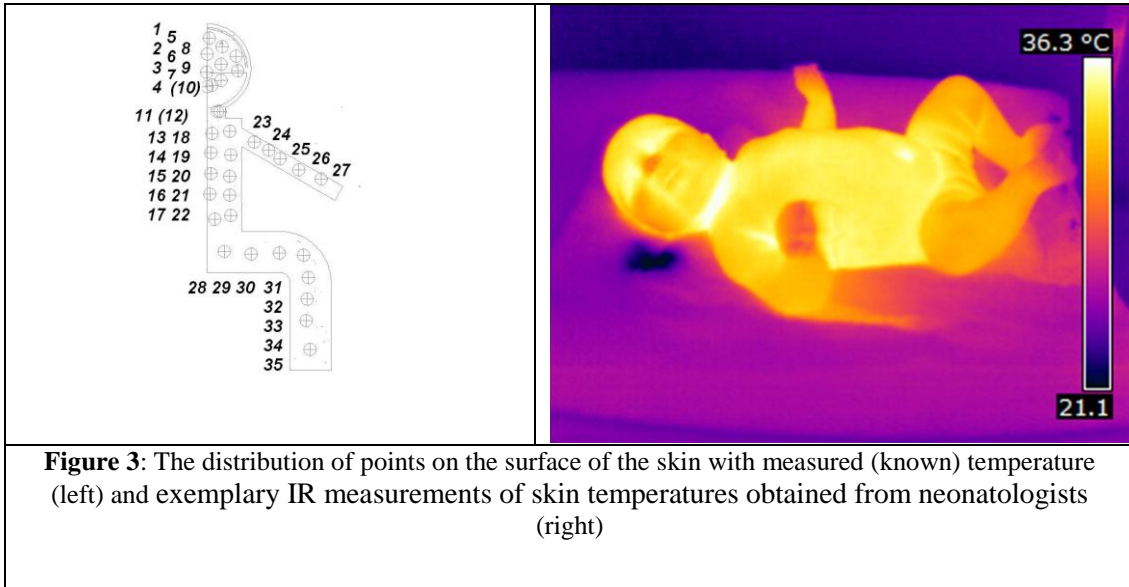
### 3 INVERSE ANALYSIS

Despite the adopted geometrical and mathematical simplifications the inverse analysis was used to retrieve the needed parameter values of the heat transfer model like blood perfusion rate  $\omega_{t,b,as}$  used in equation (3), internal heat generation  $\dot{q}_{mst,t,b,as}$ , used in equation (4) as well as the heat transfer coefficient on the skin. Generally the method consists of tuning the temperature field of the numerical model to the temperature field measured on the real object

by changing the set of parameters values. As the objective function  $\Delta$  the standard least-square function was applied. The objective function consisted of the sum of squares of deviations between temperature  $T$  predicted by the mathematical model and temperature  $U$  measured at the selected locations on the infant's skin which, introducing the vectors of temperatures values at considered locations, can be written as follows:

$$\Delta = (\mathbf{T} - \mathbf{U})^T (\mathbf{T} - \mathbf{U}) \rightarrow \min \quad (15)$$

where superscript T indicates matrix transposition. The measurements were collected from 35 spots on the surface of the skin and considering the deep body temperature (rectal temperature in case of a real baby and blood pool temperature in case of numerical simulation). The location of those 35 spots on the surface of the model, as it is schematically shown in Figure 3, as well as the temperature field  $\mathbf{U}$ , including the blood pool temperature (the rectal temperature in the real baby), was consulted with the neonatologists and approved by them.



**Figure 3:** The distribution of points on the surface of the skin with measured (known) temperature (left) and exemplary IR measurements of skin temperatures obtained from neonatologists (right)

Minimization of the objective function  $\Delta$  is carried out making use the sensitivity coefficients  $Z_{ij}$  determined numerically:

$$Z_{ij} = \frac{\partial T_i}{\partial Y_j} \approx \frac{T_i - T_i^*}{Y_j - Y_j^*} = \frac{T_i - T_i^*}{\Delta Y_j} \quad (16)$$

where  $Y_j$  stands for the selected  $j$  parameter (any of 5 blood perfusion rates, any of 5 internal heat generations or heat transfer coefficient) affecting temperature  $T_i$  at selected  $i$  location. In order to make the problem over-determined the number of measurements was more than three times higher than the number of estimated parameters. This should generally help in reducing the influence of experimental errors on results. Quantities  $Y^*$  and  $T^*$  stand for the initial guess of unknown model parameters.

Collecting all the parameters values into a vector  $\mathbf{Y}$ , all sensitivity coefficients into the sensitivity matrix  $\mathbf{Z}$  and expanding temperature  $\mathbf{T}$  into a Taylor series (cut after first-order derivatives):

$$\mathbf{T} = \mathbf{T}^* + \mathbf{Z}(\mathbf{Y} - \mathbf{Y}^*) \quad (17)$$

where  $\mathbf{Y}^*$  stands for the vector of initial guess of unknown model parameters. These model parameters were taken from the literature. In many cases the values of these model parameters have been adopted directly from data available only for adults. The vector of  $\mathbf{T}^*$  temperatures is the result of using derived model using the vector  $\mathbf{Y}^*$  of model parameters.

The final solution of optimisation problem (15) can now be obtained from the following system of equations [7]:

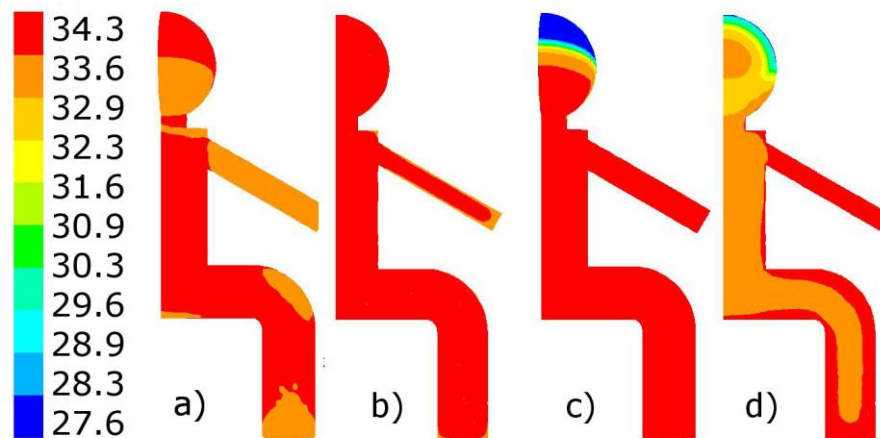
$$(\mathbf{Z}^T \mathbf{Z})\mathbf{Y} = \mathbf{Z}^T (\mathbf{U} - \mathbf{T}^*) + (\mathbf{Z}^T \mathbf{Z})\mathbf{Y}^* \quad (17)$$

where  $\mathbf{Y}^*$  stands for the vector of initial guess of unknown model parameters. These model parameters were taken from the literature. In many cases the values of these model parameters have been adopted directly from data available only for adults.

#### 4 NUMERICAL EXAMPLE

The initial temperature of the whole newborn body was assumed to be about 34°C. Besides it was presumed that the naked neonate laid on the heating mattress and was surrounded by the air. Thus on the bottom part of the skin the heat flux was estimated to be 0 W/m<sup>2</sup> (insulation boundary condition). Furthermore to the top part of the skin the convection with the heat transfer coefficient equalled to  $h = 5$  W/m<sup>2</sup>K was prescribed. The ambient temperature was assumed to be 25°C. Then the baby is treated by the hypothermal therapy which lasts 72 hours. Similarly to a real therapy, it was assumed that on the baby's head a cooling helmet was fixed whilst the rest of the body was irradiated by the radiant warmer. The cooling helmet was imitated by the convective boundary condition ( $t_a = 10.8^\circ\text{C}$ ,  $h = 19.5$  W/m<sup>2</sup>K). The radiant warmer and the free convection to the ambient air were simulated by the mixed boundary condition (convection:  $t_a = 25^\circ\text{C}$ ,  $h = 5$  W/m<sup>2</sup>K, radiation:  $\dot{q} = 100$  W/m<sup>2</sup>).

Many simulations confirmed that the proper temperature distribution during the therapy can be achieved using inverse thermal analysis. The example temperature fields at the beginning and during the therapy are displayed in Figure 4:



**Figure 4:** The temperature distribution (in °C) on the skin surface (a,c) and on the central cross section (b,d) at the beginning (a,b) and at the end of the cooling therapy (c,d).

## 5 CONCLUSIONS

The presented model of the hypothermal therapy includes a geometric model of neonate, a computer program describing the heat transfer in the human body and the file in Fluent in which the process of the therapy is preset. For model the most required are values of parameters both for the therapy and for the heat transfer in the neonate's body. Such parameters can be received using the inverse analysis presented in this work. Tuning the model to the data from the reality one can obtain the set of proper and reasonable parameters. Additionally it should be stressed that values of blood perfusion rate and heat generation depend on the tissue temperature so they are quite universal. It can be assumed that if the set of parameters fits to one experimental data and is validated with another data from experiment, the final set of parameters can be applied to test another situations e.g. test various settings or conditions like temporary change of ambient temperature or changes of metabolic heat. Summing up the presented model is very universal.

The results of the inverse analysis showed that the predicting of values of important parameters of bioheat transfer is possible, but it still requires some tests and improvements. The difference between the measured/assumed and the calculated temperatures using a new set of model parameters can still be observed. It could be valuable to include the weighting coefficients in the optimization procedure, e.g. the rectal temperature requires the higher value of weighting coefficient than the rest of measurements. Besides the optimization procedure can be improved by combination with constraints taken from the literature. Nevertheless, it can be stated that obtained results are valuable from the scientific and practical point of view.

The final values of parameters controlling the neonatal brain cooling process are extremely valuable from technical and scientific point of view.

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