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NUMERICAL SIMULATION OF HEAT TRANSFER IN CANINE KNEE JOINT DURING THERAPEUTIC HEATING

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Abstract. *This study aimed to simulate the transient heat transfer in a canine knee joint during application of a therapeutic heating. The geometric model of the canine knee joint was created by the software SOLIDWORKS®, based on a photographic record of an anatomical piece. The heat diffusion equation was used to model the heat transfer phenomenon. The blood perfusion rate was constant in all tissues. The simulations were performed using the software ANSYS-CFX® 12.1. A maximum percentage difference of 26% was found between the numerical simulations and the experimental in vivo data. The results suggest that some improvements must be performed in the considerations adopted in the simulations. Nevertheless, the results of numerical simulation can help understanding the biological tissue temperature profile during application of therapeutic heating.*

Keywords: *bioheat transfer; Pennes equation; computational simulation; thermal resource; knee joint.*

1. INTRODUCTION

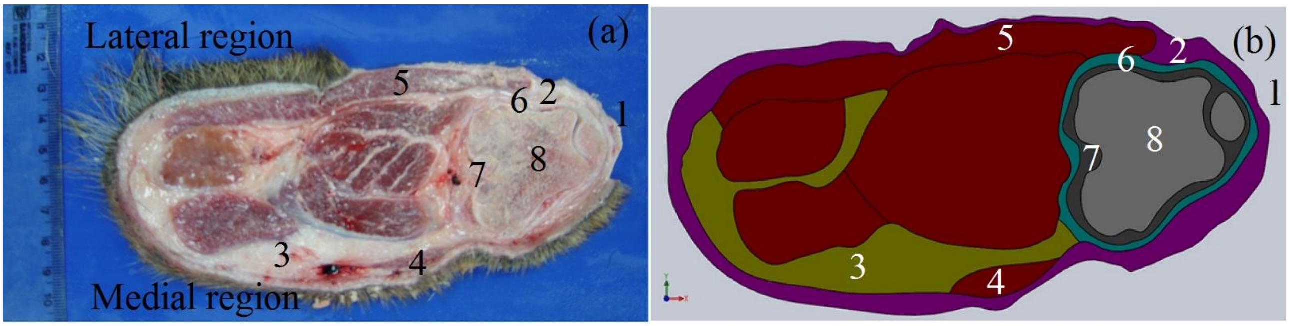
The knee joint is commonly affected by traumatic and / or degenerative lesions, hence it is one of the joints that most frequently receives physiotherapy interventions (Martin *et al.*, 2001; Denegar, 2003; Warren *et al.*, 2004). Among these interventions, thermal heating resources are widely used in clinical practice of physical therapy. These resources, mainly due to its effects on blood flow, increasing it, are commonly applied when it aims to promote the progress of the tissue repair process, which is, however, contraindicated in the first 24 to 48 hours after the trauma (Halvorson, 1990). Other effects, such as muscle relaxation (Lessard *et al.*, 1997), easier joint move and mobilization (Chen *et al.*, 1997) and decreased of pain in chronic-degenerative character injuries (Sluka *et al.*, 1999) are also achieved with the use of tissue heating. However, the resulting therapeutic effects of the application of thermal heating capabilities are dependent on the tissue temperature. According to the literature, it is necessary that the target tissue or structure reaches an optimum temperature range for that therapeutic benefits may occur (Deal *et al.*, 2002). Changes below this range will induce placebo treatments, which has no clinical effectiveness, as well as excessive variations can damage tissues (Draper *et al.*, 2004; Greenhalgh *et al.*, 2004). However, the application of these resources is performed mostly without any monitoring the target tissues temperatures. *In vivo* measurements are usually invasive and often impossible to be performed when deep tissues are the target (Trobec *et al.*, 2008). In this context, numerical simulation can be an interesting tool to assist in understanding the temperature profile in different tissues of the joint and thus contribute to better planning the applications of thermal resources used in clinical practice of physical therapy. This study aimed to simulate the heat transfer in a canine knee joint during the application of a therapeutic heating and evaluate the temperature profile in different tissues due to this application.

2. MATERIAL AND METHODS

2.1 Geometric model and mesh generation

The geometric model of the canine knee joint was created based on a photographic record of a cross-section of an anatomical part, Fig. 1a. The different tissues that comprise the canine knee joint were established by visual inspection. After, the geometric model was developed with the help of SolidWorks® software. For this study, we developed a geometric model considering the epidermis, the subcutaneous, fat and muscle tissues, pericapsular and cruciate ligaments regions and bone. The geometric model was developed with a thickness of 0.5 mm, and can be seen in Fig. 1b.

After the development of computational domain, the meshes were built using ANSYS Meshing, a tool available in the ANSYS Workbench® platform. According to the literature, meshing is one of the most critical aspects of the simulation. A large number or very small elements can result in extensive simulation time or inaccurate results, respectively (Versteeg and Malalasekera, 2007). The simulations were started with less refined mesh (mesh 1), composed of prismatic quadrangular elements. Subsequently, the initial coarse meshes were gradually refined until the average changes in temperature results in tissues were lower than 1%. Refinements followed the criteria of the length value of mesh element described in literature (Celik *et al*., 2008). For the choice of the optimal mesh was considered that the solution is independent of the mesh with a lower percentage than 1% difference between the results obtained with coarse and subsequent refined mesh. Among the three evaluated meshes, the intermediate one (mesh 2) met the previously mentioned criteria and, therefore, was chosen to perform the simulations. The mesh chosen for the simulation of the heating can be seen in Fig. 2.



1- Epidermis; 2- Subcutaneous; 3- Fat tissue; 4- Medial muscle; 5- Fat muscle; 6- Pericapsular region; 7- Cruciate ligaments; 8- Bone

Figure1. (a) Cross-sectional and (b) Geometric model of the canine knee joint

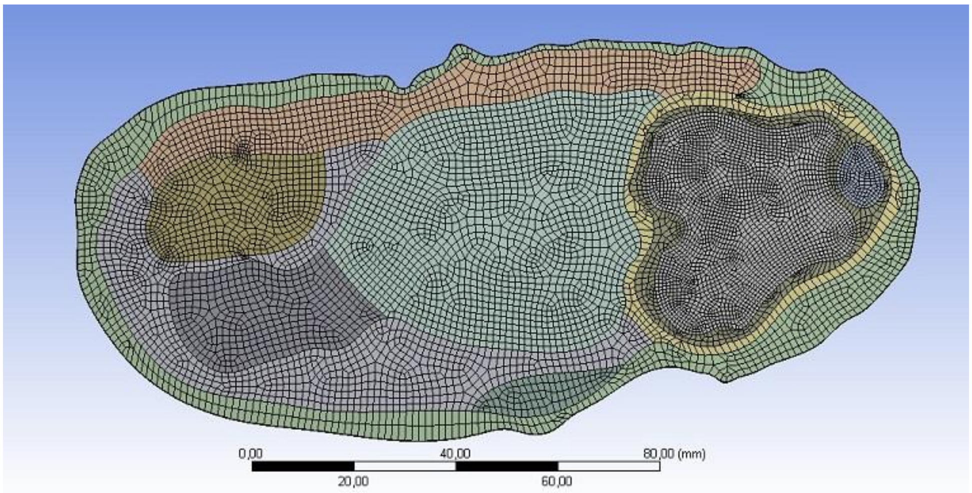


Figure 2. Mesh 2 generated on the geometric model

2.2 Mathematical model

The heat diffusion equation was used to model the heat transfer phenomenon in solid tissues during therapeutic heating condition. This equation can be described as:

$$\frac{\partial}{\partial x}\left(k\frac{\partial T}{\partial x}\right) + \frac{\partial}{\partial y}\left(k\frac{\partial T}{\partial y}\right) + \frac{\partial}{\partial z}\left(k\frac{\partial T}{\partial z}\right) + \dot{q_m} + \dot{q_w} = \rho_L c_{pL}\frac{\partial T}{\partial t} \quad (1)$$

$$\dot{q_w} = w\rho_s c_{ps}(T_{ref} - T_{tec}) \quad (2)$$

where: k is the thermal conductivity of each tissue [$Wm^{-1}°C^{-1}$]; $\dot{q_m}$ is metabolic heat rate of the tissue [Wm^{-3}]; $\dot{q_w}$ is the blood perfusion rate of the tissue [Wm^{-3}]; w is the blood perfusion [$m^3s^{-1}\ m^{-3}$]; ρ_L and c_{pL} are the specific mass [kgm^{-3}] and the specific heat [$Jkg^{-1}\ °C^{-1}$] of the layer; ρ_s corresponds to the blood specific mass [kgm^{-3}]; c_{ps} is the specific heat of blood [$Jkg^{-1}\ °C^{-1}$] and T_{ref} and T_{tec} are rectal and tissue temperatures, respectively.

The rectal temperature T_{ref} (38.1°C) and the temperatures used as initial condition (T_i) of each tissue layer were taken from Araújo (2009) study (Tab. 1). According to the author, the initial temperatures were considered to be equal to the average of the experimental temperatures obtained in thermal neutrality condition (24.7ºC). The bone initial temperature was considered based on the innermost region in which the temperature was measured in this study (cruciate ligaments region).

In this study, the blood perfusion rate ($\dot{q_w}$) in each tissue was considered to be constant, not varying with the temperature of each tissue layer during the simulation.

2.3 Numerical simulation

The physiological properties used to perform the simulations were considered by the average of published data and were obtained from Araújo (2009) study. The values of the variables used for each of the layers are shown in Tab. 1. All properties were considered constant for the implementation of the simulations.

Table 1. Physiological properties and thermo-physical layers of tissue and blood.

Region	c_p ($Jkg^{-1o}C^{-1}$)	ρ (kgm^{-3})	k ($Wm^{-1o}C^{-1}$)	q_m (Wm^{-3})	w ($m^3s^{-1}m^{-3}$ tec)	Initial temperature (T_i) (°C)
Epidermis	3593	1200	$2.28x10^{-1}$	0	0	34.9
Subcutaneous	3365	1200	$4.64x10^{-1}$	200	$1.3x10^{-3}$	35.5
Fat tissue	2678	937	$2.03x10^{-1}$	3.9	$2.9x10^{-4}$	36.1
Muscle	3684	1097	$5.29x10^{-1}$	716	$5.8x10^{-4}$	36.5
Pericapsular	3500	1051	$4.98x10^{-1}$	0	$1.8x10^{-3}$	35.2

Cruciate ligaments	4190	1000	6.10×10^{-1}	0	0	37.1
Bone	1785	1585	7.35×10^{-1}	368.3	4.0×10^{-4}	37.1
Blood	3800	1060	-	-	-	38.1

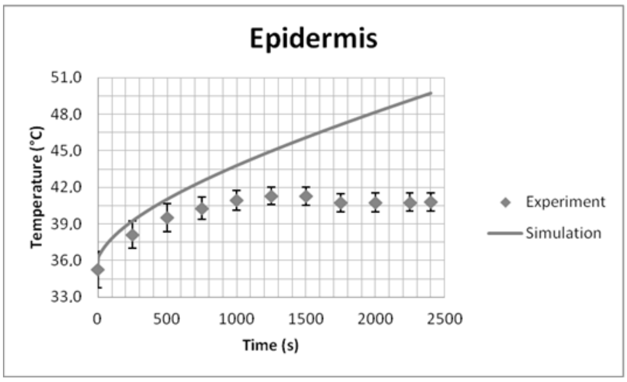
The simulation of the temperature profile of the knee joint was carried out in the heating transient condition. To perform the simulation it was adopted at the upper and lower joint face a boundary condition of second kind, corresponding to a perfectly isolated or adiabatic surface. Thus, it was assured that the heat transfer occurs only in the two-dimensional cross sectional plane of the canine knee, although the geometric model is three-dimensional. On the external surface of the geometric model was considered the average heat flux from a heating element (qx" = 215.7 W/m^2), considering that all energy emitted by the blanket is transferred to the knee joint. The simulations were performed considering a time of 40 minutes of treatment. The information about the external sources and the time used to heat the canine knee, which was used in the simulations, were taken from the experimental study by Araújo (2009). Once this model is composed of different tissue layers became necessary to define the interface condition between the various tissues. Thus, it was assigned conservative heat flow between the tissue interfaces. The numerical simulation of the canine knee in the heating condition was performed using the ANSYS CFX® 12.1 software. The results of the simulation were evaluated and compared to average values of temperature obtained from an experimental study *in vivo* (Araújo, 2009) .

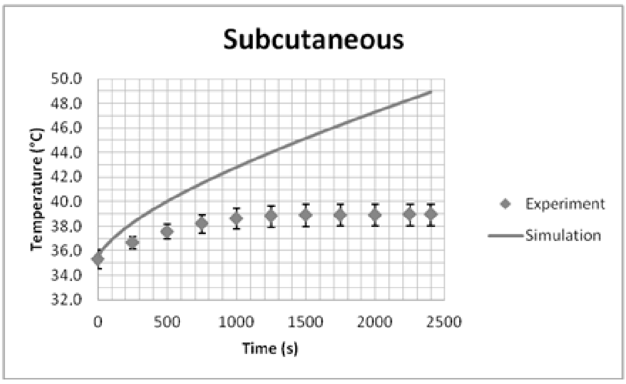
The average percentage difference between the simulation and the experimental data was calculated for each of the tissue layers at different time instants. A smaller or equal average percentage difference of 10 % was considered acceptable, since the thermo-physics and physiological parameters related to bio heat transfer process are quite divergent in the literature. These differences show that there are still uncertainties in the experimental measurement of these parameters.

3. RESULTS AND DISCUSSION

The numerical simulations performed on this study aimed to appraise the transient profile of the tissue temperature during the application of a heating feature commonly used in clinical practice of physical therapy. According to the literature, for tissue heating, it is recommended to maintain the target tissue for at least 5 minutes in the temperature between 40ºC to 45ºC (Draper *et al*., 1999; Kitchen, 2003; Robertson *et al*., 2005). In temperatures above 45°C, many proteins are damaged and, therefore, there is destruction of the cytoskeleton, cell membrane and the regional microvasculature. Such injuries can cause disruption of cellular reproductive cycle and blood flow at the site, which predisposes to tissue necrosis (Knight *et al*., 2001). However, for both heating and for cooling, the literature reports that local temperature changes in deep tissues should be limited to about 5 or 6 ° C above or below the temperature of the tissue (Low and Reed, 2001).

The profile of the simulated temperature of the layers during heating period can be observed in Fig. 3. The average temperature profile referring to experimental data and their respective standard deviations are also represented in the same figure. Exception is made to the bone that did not have its temperature measured during the experiment.





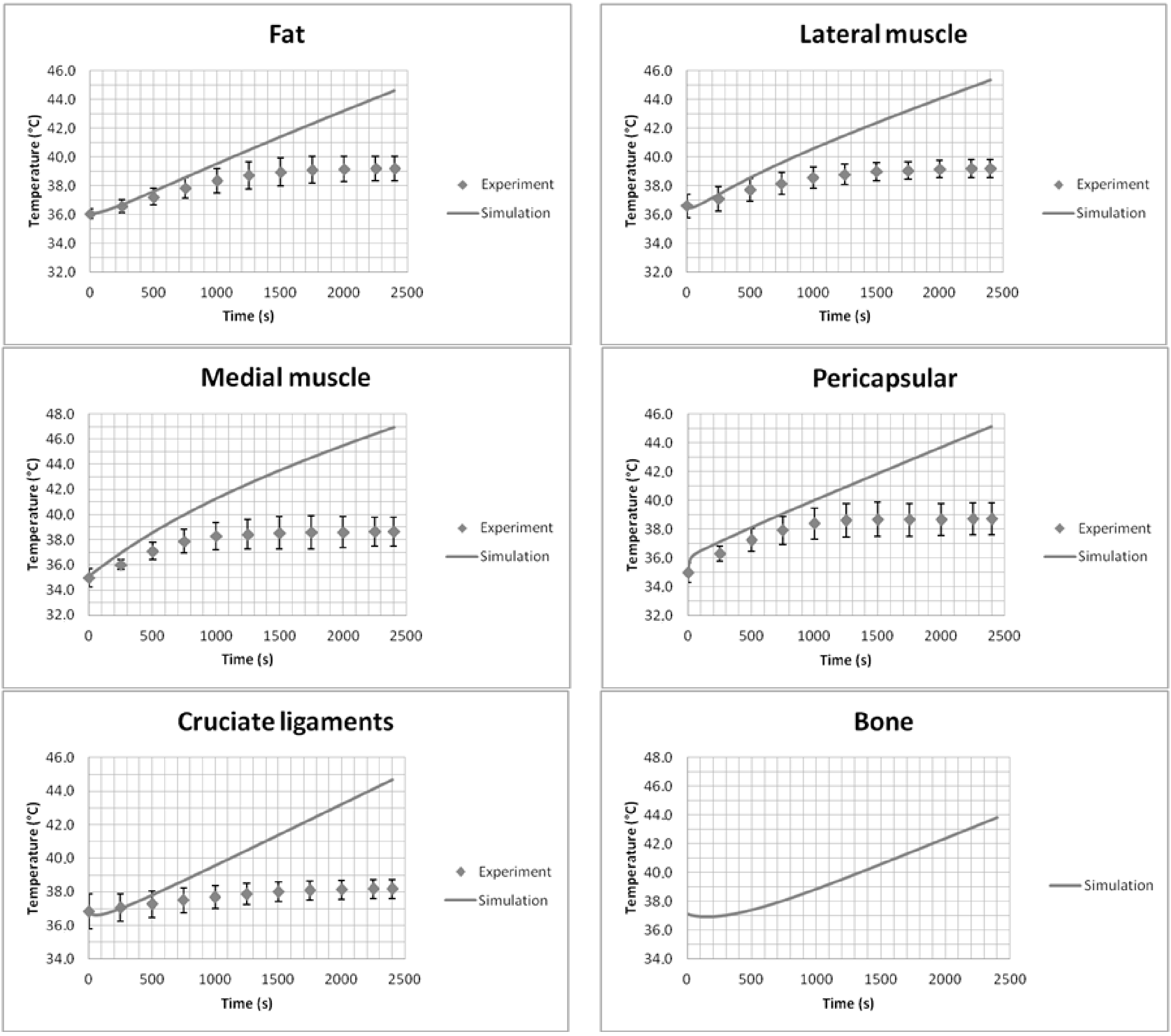


Figure 3 - Variation of local tissue temperatures with time during heating

In general, it can be observed that in all tissues the simulated temperatures are higher than the experimental temperatures during the whole period of analysis. Note that up to 500 seconds the behavior of simulated temperature is similar to the first three measurements in the experiment. However, after this period, the simulated temperature curve tends to move away from *in vivo* measurements and grows almost linearly throughout the rest of the simulation. On the other way, the experimental temperatures grow more when the application starts the heating action and after 1000 seconds it tends to stabilize at a given value.

The final temperature of the simulation, the experiment and the percentage difference between both is available in Tab. 2.

Table 2. Experimental end temperatures, and simulated percentage difference between the two.

Tissue	Final experimental temperature (°C)	Final simulated temperature (°C)	Difference (%)
Epidermis	40.8	49.8	22
Subcutaneous	38.9	48.9	26
Fat tissue	39.2	44.6	14
Lateral muscle	39.2	45.3	16
Medial muscle	38.6	46.9	21
Pericapsular	38.7	45.2	17
Cruciate ligaments	38.5	44.7	16

Bone	-	43.8	-

The analysis of experimental measurements show a clear trend for internal tissues, pericapsular and the region of the cruciate ligaments, to remain at lower temperatures, which may indicate a difficulty for the heat from the thermal blanket to achieve these tissues. In the simulation results, these two tissues also showed the lowest temperatures of the model, as well as in the experimental data. However, these regions simulated temperatures are above the values expected.

The results show that all final simulated temperatures are above the ones measured in the experiment. The tissue temperatures are at a level above the ideal range for therapeutic heating. It is noticed that the subcutaneous tissue temperatures had the most difference from *in vivo* measurements, with a difference of 26% and a temperature increase of 10°C, the highest among all tissues. The lateral muscle and fat tissue were the closest to the experimental temperature, however, their difference were 14% and 16 %, respectively. Fat tissue also had the lowest increase in temperature, 5.4 ° C, which may be due to its low thermal conductivity, acting as an insulating within the model. The higher final temperature in the simulation and in the experiment was in the epidermis layer, which was expected due to this proximity with the boundary condition of the simulation and thermal blanket applied in the experiment.

High values of simulated temperatures indicate that the model did not provide a good approximation to the real behavior of biological tissues. Thus, some points must be taken into consideration:

1) The thermo-physics properties (thermal conductivity, specific heat, etc.) and physiological (metabolism and blood perfusion), linked to the bio heat transfer process, are quite disparate in the literature. These differences show that there are still uncertainties in the experimental measurement of these properties. Thus, the wrong approach by any of the used properties values may have been one of the factors responsible for the differences found. According to Cui and Barbanel (1990 and 1991), among the physiological and thermo-physics properties, the blood perfusion and thermal conductivity are the one who most influence behavior and thermal responses. They have significant influence on the behavior of the insulating response in relation to the rise in temperature, the time to achieve balance and the different values of the temperature profile of the assessed region (Lima *et al.*, 2006).
2) The use of blood perfusion rate of each tissue as a constant source of energy certainly contributed for the simulated temperatures to be higher than the experimental. It was due to the fact that, during the therapeutic heating of the tissues, blood exceeded the reference temperature of blood. Consequently, the blood present in the tissue ceases to be a power supply to the model and becomes a sink, removing energy by taking out the heat of the tissue. Thus, modeling the blood perfusion rate of each tissue as a function of tissue temperature will generate more consistent results with the experimental data.

Although some differences between the experimental results and the simulation data have been found, the numerical simulation is the first step performed to comprise the temperature distribution in biological tissues during therapeutic heating process. The improvement of the simulations will be an important step for the noninvasive evaluation of the effectiveness of thermal heating resources.

4. CONCLUSION

The study sought to simulate the therapeutic heating of the canine knee joint in a transient regime by a numerical simulation. The simulated temperature values were compared to the experimental data showed that the model should be improved. Changes in the thermo-physical and physiological properties used can be the first step to get closer to the experimental results. Moreover, simulate the blood perfusion rate as a function of temperature may also modify the results founds. The developed model showed that the numerical simulation can be an important tool to assess the temperature profile of biological tissues during the application of thermal-therapeutics resources.

5. ACKNOWLEDGEMENTS

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