

Article

Anatomical Asymmetry in Adipose Tissue Thickness Alters Heat Transfer During Superficial Thermotherapy: A Computational and Experimental Study

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Abstract

Computer models and protocols commonly used for superficial thermotherapy assume anatomically symmetric tissue geometries. However, anatomical variability, particularly the variability of the adipose tissue thickness, may have a significant impact on heat transfer and cause non-uniform thermal doses in the patient population. This paper studies the influence of anatomical asymmetry due to adipose tissue thickness on heat propagation during superficial thermotherapy based on combined computational and experimental analyses. A Finite Element Model of the lower limb was designed with anatomically representative layers and tissue thermal properties. Parametric simulations were conducted for medium (M) and extra-large (XL) models at hot-pack temperatures (38–44 °C) for 15 min. The simulation results were compared with experimental measurements on *ex vivo* porcine tissue. The results showed that the thickness of adipose tissue significantly affected deep-tissue heating. When the same heating condition was applied, the boundary temperature of the muscle was increased by ~1.38 °C for the 44 °C hot-pack in the M-size model and was restricted to 0.21 °C for the XL-size model. The experimental muscle temperature increased from 22.26 °C to approximately 23.3 °C after 15 min. Under the same initial temperature and nominal 38 °C heating condition, the perfused and zero-perfusion M-size simulations reached approximately 26.8 °C and 30.5 °C, respectively; these endpoint differences support a qualitative comparison rather than quantitative validation. These results show that anatomical asymmetry has a strong effect on heat transfer during superficial thermotherapy and challenge the precept that standard heating protocols provide similar thermal doses in different body morphologies.



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1. Introduction

Superficial thermotherapy is a non-invasive therapy for musculoskeletal pain, sports injuries and circulatory disorders due to its low cost and ease of application [1,2]. Thermotherapy causes vasodilation, increases local blood flow, improves metabolism and

promotes tissue healing and pain relief. However, the amount of heat transferred to deeper tissues depends on complex interactions between tissue layers, thermal properties, exposure time and physiological heat dissipation mechanisms; thus, the prediction of internal heat distributions is difficult [3,4]. Peripheral Artery Disease (PAD), which in recent epidemiological studies has been called lower extremity artery disease (LEAD) [5], is a condition that is frequently treated with this type of therapy. In 2015 around 234 million people had PAD and more than 230 million adults worldwide are affected [6,7]. Furthermore, the global incidence of PAD increased by 72% between 1990 and 2019 [8]. Superficial thermotherapy is a common method for pain relief and local blood flow improvement in LEAD patients by means of controlled heat application to the skin surface within a temperature range of 38 °C to 45 °C. Within this therapeutic range heat increases local metabolism, enhances blood flow and promotes relief of pain and restoration of function [1]. However, long exposures to temperatures above 45 °C may cause adverse effects like protein denaturation and thermal tissue damage [9]. Thus, precise control of both treatment temperature and exposure time is crucial for the safety and efficacy of thermotherapy, prompting the development of clinically optimized treatment protocols [10,11].

Direct measurement of temperature inside biological tissues is invasive and often impractical in a clinical environment. Therefore, the thermal response to superficial thermotherapy is often indirectly measured through empirical measurements of the surface temperature [9]. Computational modeling has become a useful means of investigating heat transfer within subcutaneous tissues. Finite Element Models (FEMs) based on the Pennes bioheat equation are able to simulate transient temperature distributions taking into account heterogeneous tissue properties, blood perfusion, metabolic heat generation and realistic boundary conditions. They provide a mathematical framework for the evaluation of treatment parameters, optimization of therapeutic regimens, and prediction of tissue thermal responses under conditions which would be difficult or impossible to evaluate experimentally [3]. Although widely used, many computational bioheat models assume symmetric tissue structures or simplified anatomical geometries. This may lead to an underestimation of the impact of patient-specific anatomical heterogeneity on heat transfer.

Heat transfer in biological tissues has been extensively studied using both experimental and computational approaches. FEMs have become the predominant tool for analyzing temperature propagation during thermotherapy because they enable controlled evaluation of tissue properties and treatment parameters. Nevertheless, many existing studies rely on simplified symmetrical geometries or homogeneous tissue representations, while only a limited number incorporate anatomically realistic models or experimental validation. More importantly, the influence of anatomical asymmetry associated with adipose tissue thickness on heat transfer has received little attention. Table 1 summarizes representative studies addressing computational and experimental modeling of heat transfer during thermotherapy.

Table 1. Comparison of representative experimental and computational studies on heat transfer during superficial thermotherapy.

Contribution	Validation	Computational Approach	Structure Representation	Heat Modality	Study
Thicker adipose tissue was associated with higher post-heating skin temperatures.	Skin temperature measurements	Experimental	Human lower back	Superficial heating	McLellan et al. [12].
Subjects with >25% body fat exhibited nearly 2× slower deep heating; muscle ΔT remained <1 °C after 20 min.	Muscle temperature	Experimental	Human lower limb	Hot pack	Petrofsky et al. [1].
ReBound produced 3.69 °C intramuscular heating compared with 2.82 °C for moist hot packs ($\approx 31\%$ increase).	Intramuscular thermocouples	Experimental	Human triceps	ReBound vs. moist hot packs	Hawkes et al. [2].
Predicted transient temperatures for 16 body segments with good agreement to published measurements.	Comparison with published data	FEM + Pennes bioheat equation	Simplified 16-segment human body	General thermoregulation	Geng et al. [11].
PCM splints achieved approximately a 1 °C increase in muscle temperature and deeper heating than water-filled splints.	No experimental validation	3D FEM	Simplified knee geometry	Thermal knee splints	Flores Cuautle et al. [13].
Skin temperature increased significantly, whereas core temperature remained unchanged after treatment.	Healthy subjects	Experimental	Human volunteers	Hot pack	Kominami et al. [14].
Demonstrated image-based patient-specific thermal simulations; quantitative thermal validation was not reported.	No validation	3D FEM	Patient-specific CT geometry	Lower-limb thermography	Pakarinen et al. [15].
Muscle ΔT decreased from 1.38 °C to 0.21 °C (84.8% reduction) as adipose thickness increased; the experimental comparison included perfused and zero-perfusion simulations.	Intramuscular thermocouples	FEM + Pennes bioheat equation	Anatomically representative multilayer lower limb with variable adipose thickness	Hot pack	Our proposal

As summarized in Table 1, superficial thermotherapy is affected by several physiological and anatomical variables such as body composition, tissue properties and the heating modality used. The thermal insulating effect of adipose tissue limits the depth of heat penetration and, consequently, the therapeutic effect. Similarly, computational studies have shown that FEMs are a useful tool to predict temperature distributions under different treatment conditions and anatomical scenarios. However, there are still several limitations. Most mathematical models are based on simplified symmetrical geometries, homogeneous tissue distributions or single patient-specific anatomies, not intended to evaluate the influence of anatomic variability. Thus, although the insulating effect of adipose tissue has been experimentally acknowledged, the influence of systematic changes in adipose tissue thickness, and the consequent anatomical asymmetry, on heat transfer during superficial thermotherapy has been limited. This limitation is particularly relevant since standardized thermotherapy protocols are usually prescribed without taking into account the patient's morphology, although the adipose tissue thickness varies substantially among individuals and even between the contralateral anatomical regions. Therefore, the common assumption of symmetric tissue geometries can lead to inaccurate prediction of heat propagation and, consequently, variability in the delivered thermal control.

In this work, by using a combined computational and experimental approach, we aim to study how the anatomical asymmetry induced by the thickness of adipose tissue influences heat propagation during superficial thermotherapy. An anatomically representative FEM of the lower limb was developed to simulate varying thicknesses of adipose tissue under clinically relevant heating conditions. The remainder of this paper is organized as follows. The computational model, the experimental protocol and the numerical methodology adopted to evaluate the heat transfer under different anatomic conditions are described in Section 2. Section 3 refers to computational and experimental results and their comparative analysis. Section 4 concludes with a discussion of the main findings, limitations of the proposed approach and directions for future research, and Section 5 summarizes the principal conclusions of this study.

In this study, anatomical asymmetry refers to the non-concentric arrangement of the tissue layers within the lower-limb cross-section. The M–XL comparison addresses a separate source of anatomical variability: differences in adipose tissue thickness between two representative morphologies. It does not compare contralateral limbs or quantify bilateral asymmetry.

2. Materials and Methods

This section describes the experimental and computational methodology to study the asymmetrical effect of adipose tissue thickness on heat transfer during superficial thermotherapy. Firstly, an anatomically representative FEM of the lower limb was reconstructed, and tissue-specific thermophysical properties were assigned. Subsequently, a parametric analysis was performed to simulate the transient heat transfer under clinically relevant thermotherapy conditions with different thicknesses of adipose tissue and different hot-pack temperatures. Finally, the numerical results were compared with measurements from *ex vivo* tissue under controlled heating conditions.

2.1. Anatomical Model Construction

The two-dimensional (2D) anatomical model was constructed based on a representative cross-sectional illustration of the lower limb derived from a human anatomy atlas [16], preserving realistic morphological proportions of the principal tissue layers. As shown in Figure 1, the model preserves the multilayer organization of the lower limb and the irregular internal distribution of the modeled tissues. The relative arrangement of bone,

muscle, adipose tissue, and skin was defined according to anatomical references and the geometrical configuration of the model, as summarized in Table 2.

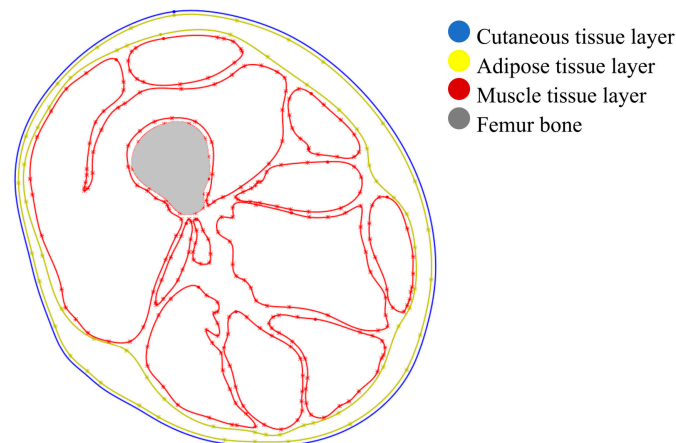


Figure 1. Two-dimensional cross-sectional geometry of the lower limb used for finite element modeling. The model preserves the anatomical arrangement of the main tissue layers, including skin (outer boundary), adipose tissue, muscle, and bone. Internal structures were manually delineated to reflect the irregular morphology of the limb, while maintaining a layered organization suitable for computational analysis.

Table 2. Anatomically representative distribution of bone, muscle, adipose tissue, and skin in the two-dimensional lower limb model.

Tissue	Radial Range (r/R)	Area Proportion (%)
Bone	0.00–0.20	~10–15%
Muscle	0.20–0.75	~55–65%
Adipose tissue	0.75–0.95	~20–30%
Skin	0.95–1.00	~3–5%

The asymmetrical 2D geometry was designed using SOLIDWORKS 2025 SP2.0 (Dassault Systèmes, Vélizy-Villacoublay, France). To maintain the morphology and relative spatial arrangement of each tissue layer, the boundaries of the primary anatomical structures including bone, muscle, adipose tissue, and skin were manually traced with polyline-based tools. For the FEM implementation, each tissue layer was defined as an independent geometrical domain; thus, the tissue-specific thermal properties could be assigned. The model was exported using the Drawing Exchange Format (DXF), a standard computer-aided design format that allows interoperability between CAD and FEM environments while preserving the geometric structures. After importing the DXF model into COMSOL Multiphysics® version 6.2 (COMSOL AB, Stockholm, Sweden), the multilayer geometry allowed for heterogeneously assigning thermal properties and independently controlling each domain during the multiphysics simulations. The proposed modeling approach offers sufficient flexibility to systematically vary the thickness of adipose tissue, while keeping other anatomical structures unchanged, thereby enabling a controlled evaluation of the effect of anatomical asymmetry on heat transfer in superficial thermotherapy. Though the external shape of the model preserves an anatomically realistic lower limb morphology, the tissue organization can be characterized by the normalized radial distribution summarized in Table 2. These radial ranges give an equivalent description of the multilayer anatomical structure for computational purposes and allow the proposed model to be reproduced. Importantly, this representation is not based on concentric symmetry but

rather represents an anatomically informed approximation where the tissue regions are defined according to the observed geometry, while maintaining their relative proportions within the computational domain.

The 2D representation was selected to examine heat propagation across the skin, adipose tissue, and muscle while retaining the irregular cross-sectional geometry. It permits a controlled comparison of the two anatomical configurations with a limited computational cost. However, the model does not resolve longitudinal heat spreading, variations in anatomy along the limb, or the end effects of a hot-pack with finite axial length. The present analysis therefore describes the response within the modeled cross-section; equivalence to a three-dimensional temperature field has not been established.

2.2. Bioheat Transfer Model

The transient form of the Pennes bioheat equation was used to describe the heat transfer in the multilayer model considering the combined effects of thermal conduction, blood perfusion, and metabolic heat generation. As context, this formulation has been used in computational studies of thermoregulation, hyperthermia, thermotherapy and biomedical heat transfer [4,9,17] due to its ability to describe the main mechanisms responsible for heat transport in perfused biological tissues. In our study, Equation (1) was used to study the effect of variations in adipose tissue thickness during superficial thermotherapy.

$$\rho_i c_i \frac{\partial T_i}{\partial t} = \nabla \cdot (k_i \nabla T_i) + w_{b_i} \rho_{b_i} c_{b_i} (T_{a_i} - T_i) + q_{met_i} \quad (1)$$

where i denotes the tissue layer, T_i is the local tissue temperature ($^{\circ}\text{C}$), ρ_i is the tissue density (kg/m^3), c_i is the specific heat capacity ($\text{J}/(\text{kg}\cdot\text{K})$), k_i is the thermal conductivity ($\text{W}/(\text{m}\cdot\text{K})$), w_{b_i} is the blood perfusion rate ($\text{kg}/(\text{m}^3\cdot\text{s})$), q_{met_i} is the metabolic heat generation (W/m^3), T_{a_i} is the arterial blood temperature ($^{\circ}\text{C}$), ρ_{b_i} is the blood density (kg/m^3), and c_{b_i} is the specific heat capacity of blood ($\text{J}/(\text{kg}\cdot\text{K})$).

The implementation assumes that each anatomical layer behaves as a homogeneous and isotropic medium with constant thermal properties during the simulation. The interfaces between the neighboring tissues were assumed to be in perfect thermal contact, ensuring the continuity of temperature and heat flux at the interfaces between adjacent domains. Also, the temperature of arterial blood and the metabolic heat generation were assumed constant during the heating period, while the thermophysical properties were considered temperature-independent.

The thermal properties assigned for the skin, adipose tissue and muscle domains are given in Table 3. The parameters, i.e., thermal conductivity, density, specific heat capacity and blood perfusion, were acquired from the study carried out by Flores-Cuautle et al. [18], ensuring that the heat transfer properties were physiologically appropriate for each tissue layer.

Table 3. Thermal properties assigned to each tissue layer in the FEM model.

Tissue	Skin	Fat	Muscle
Thermal Conductivity ($\text{W}/(\text{m}\cdot\text{K})$)	0.37	0.21	0.49
Specific Heat ($\text{J}/(\text{kg}\cdot\text{K})$)	3391	2348	3421
Blood perfusion rate, ω_b (s^{-1})	0.0018	0.0057	0.0006
Density (kg/m^3)	1109	911	1090

The bone region was retained to preserve the anatomical arrangement of the surrounding tissues. However, this region was represented as a void and assigned the thermal

properties of air rather than those of cortical bone. Thus, it remained within the computational domain, although heat transfer through cortical bone was not explicitly represented.

2.3. FEM and Simulation Conditions

Transient heat transfer analysis was employed to study the effect of temperature propagation through the asymmetrical multilayer model. The properties and boundary conditions were defined independently for each anatomical region. The independent variables studied were hot-pack temperature, treatment time, and lower-limb morphology characterized by variations in adipose tissue thickness. This allowed us to assess the effect of asymmetry on temperature propagation in a controlled manner, keeping the thermal properties and boundary conditions the same for all simulated cases.

The hot-pack was modeled as an external heat source applied over a localized region of the skin surface corresponding to the anatomical location of the vastus lateralis muscle as shown in Figure 2. The contact length was fixed at 25 cm, the dimensions of the heating device used in the experimental comparison. Two clinically relevant hot-pack temperatures, 38 °C and 44 °C, were evaluated according to reported thermal safety thresholds for superficial thermotherapy [19].

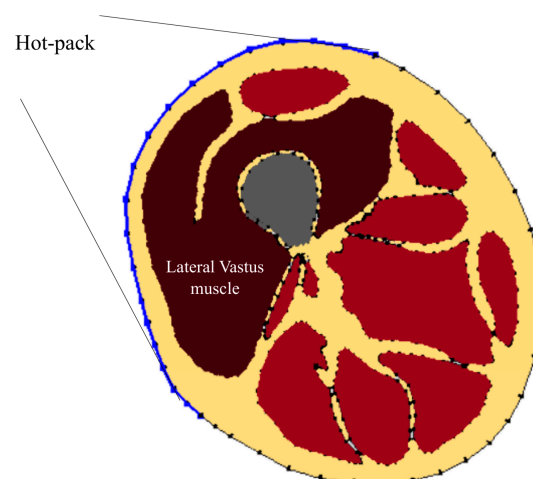


Figure 2. Location of the applied heat source (hot-pack) over the skin surface, represented by the blue contour. The heat source is positioned over the lateral region of the lower limb, corresponding to the vastus lateralis muscle. The contour length was set to 25 cm, approximating the dimensions of a hot-pack used in the experimental stage. The different colors represent the tissue layers included in the anatomical model: light yellow corresponds to adipose tissue, dark red to muscle, gray to bone, and the thin outer layer to skin. The blue contour indicates the location of the applied hot-pack.

The temperature of the heat source was evaluated at 38 °C and 44 °C, and remained constant throughout the simulation interval. The condition is expressed mathematically as Equation (2) [20]:

$$T(r, t) = T_{hp} \text{ for } r \in \Gamma_{contact}, t > 0 \quad (2)$$

where $T(r, t)$ is the temperature as a function of position (r) and time (t), T_{hp} is the constant temperature (38 °C and 44 °C) on the surface, and $\Gamma_{contact}$ is the specific contour where the contact occurs. The condition $t > 0$ indicates that it applies from the moment the process begins.

The prescribed temperature was retained as an idealized heating condition to compare the M and XL geometries under the same thermal loading. It does not represent the finite thermal capacity or progressive cooling of a passive hydrogel hot-pack. Relative to a cooling pack, this assumption may overestimate the heat supplied to the tissue and

the resulting temperature elevation. The predictions should therefore be interpreted as a constant-temperature heating scenario, rather than as a reproduction of the complete experimental hot-pack history.

Initially, all tissue domains were set to the baseline temperature of 37 °C and the treatment time was limited to 15 min, which is in line with conventional superficial thermotherapy protocols. The external limb perimeter was altered based on standardized anthropometric measures reported in the National Institute of Standards and Technology (NIST) [4] to accommodate anatomical variability. The analysis considered two representative morphologies, related to Medium (M) and Extra Large (XL) body sizes, with lower-limb perimeters of 52.4 cm and 62.3 cm, respectively. The adipose tissue layer was scaled to different sizes during the scaling procedure, while the remaining anatomical structures and material properties were kept intact. This made it possible to quantify the insulating impact of adipose tissue thickness on heat propagation. The parameters are listed in Table 4.

Table 4. Parameters and simulation conditions considered in the parametric finite element study of heat transfer in the lower limb model.

Parameter	Values	Description
Hot-pack temperature (°C)	38, 44	Applied heat source temperature
Initial tissue temperature (°C)	37	Physiological baseline condition
Treatment time (min)	15	Duration of thermal exposure
Heat source length (cm)	25	Approximate size of the hot-pack
Application region	Lateral skin surface	Over vastus lateralis region
Lower limb perimeter (cm)	52.4–62.3	Anthropometric variation (M–XL)
Analysis type	Transient	Time-dependent simulation

The M and XL configurations were selected to provide a controlled contrast in adipose tissue thickness while retaining the other modeled tissue structures and properties. They are representative comparison cases, not a continuous sampling of patient anatomy.

The parametric simulations were followed by a mesh convergence analysis to ensure numerical accuracy. The model was discretized with an unstructured triangular mesh with local refinement around the heating region and the tissue interfaces, where the highest thermal gradients were expected. Mesh convergence was evaluated from the stabilization of the temperature at the muscle boundary. The mesh convergence criterion was a variation of less than 5% between consecutive refinements. To express the reported results consistently, the absolute temperature differences in Table 5 were also normalized by the temperature rise of the finer mesh above the 37 °C baseline. The ten meshing configurations examined varied from 5441 to 42,181 finite elements. Additional mesh refinement showed insignificant variations in the computed temperature field, indicating mesh-independent numerical solutions.

Table 5. Comparison of tissue layer thickness between the numerical model and *ex vivo* samples used for experimental comparison.

Δ Thickness (mm)	Porcine Thickness (mm)	Model Thickness (mm)	Tissue
0.29	1.10	1.39	Skin
1.07	12.60	13.67	Fat
0.83	17.60	18.43	Muscle

After spatial discretization, the governing equation can be expressed as a system of first-order differential equations, where the temporal evolution of the temperature field is governed by the balance between thermal capacity, conduction, and heat sources, as shown in Equation (3).

$$M \frac{dT}{dt} + KT = F \quad (3)$$

where M represents the thermal capacity matrix, K the conduction matrix, and F the contribution of internal and external heat sources, including perfusion, metabolic heat, and the applied thermal load. The temporal discretization was performed using an implicit time integration scheme to ensure numerical stability under transient conditions.

The transient thermal problem was solved by the implicit time-dependent finite element solver implemented in COMSOL Multiphysics® which ensures stable numerical integration for parabolic heat transfer problems over the whole simulation interval. All the simulations were performed on an Intel® Core™ i7-3630QM workstation with a clock speed of 2.40 GHz and 8 GB RAM.

2.4. Ex Vivo Experimental Comparison

Ex vivo porcine tissue layers were used in a controlled setting to compare experimental and simulated temperature trends. Porcine tissue was chosen because of its multilayer anatomical structure and thermophysical properties that are like human skin, subcutaneous fat, and skeletal muscle, and is widely accepted as a surrogate for thermal studies involving superficial heat transfer [17]. The experimental study aimed to compare the transient temperature response predicted by the computational model with the direct temperature measurements taken within the muscle layer after superficial thermotherapy.

Before the experiments, the thicknesses of the skin, adipose tissue, and muscle layers were measured and compared with the corresponding sizes of the computational model. The layer thicknesses are presented in Table 5 for both configurations.

Temperature measurements were taken by placing Type J thermocouples in the tissue to a depth of about 25 mm, ensuring that the sensing junction reached the muscle layer after passing through the skin and adipose tissues. The thermocouples were connected to a National Instruments NI-9219 data acquisition unit, and tissue temperature was recorded at 2-min intervals. Figure 3 presents the diagram of the experimental setup.

Before each experiment, the hot-pack was immersed in a thermostatically controlled water bath for 10 min to reach a temperature of 38 °C. The hot-pack was applied to the surface of the tissue immediately after heating to apply superficial heating under the experimental protocol. A small temperature drop may be seen when going from the water bath to the tissue surface, but this was minimized by limiting the handling time and by the high heat capacity of the hydrogel pack. For the paired computational comparison, both the perfused and zero-perfusion simulations were initialized at 22.26 °C, matching the measured initial temperature of the porcine specimen.

The experimental muscle-temperature profile was compared with perfused and zero-perfusion simulations initialized at 22.26 °C and evaluated for 15 min under a prescribed 38 °C boundary. The perfused simulations used the tissue-specific rates in Table 3, whereas the zero-perfusion simulations set the perfusion term to zero. Because the experimental hydrogel pack cooled during application, the comparison was used to assess temporal behavior and endpoint differences rather than to establish quantitative validation.

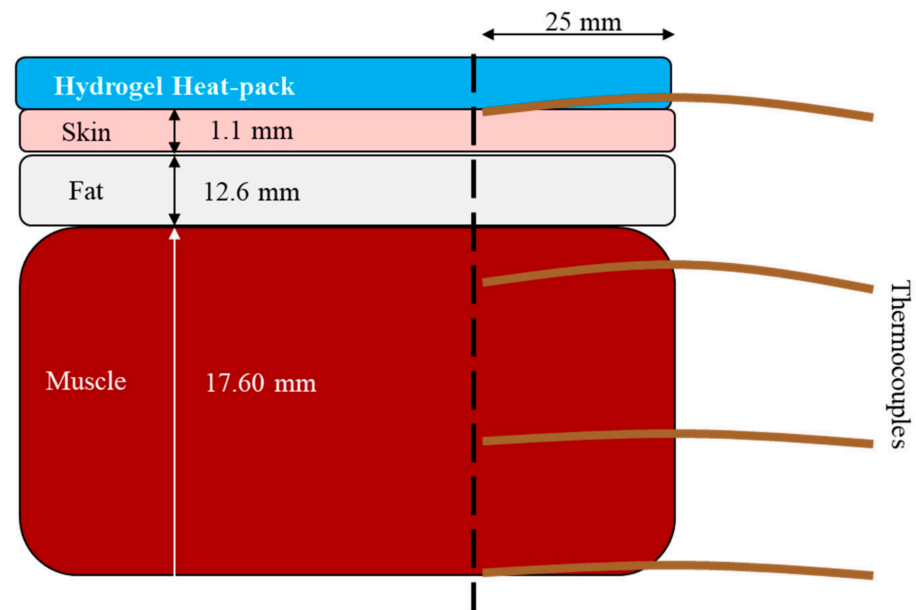


Figure 3. Diagram of the experimental measurement setup used for experimental comparison. The hot-pack is applied over the skin surface, and type J thermocouples are inserted laterally into the tissue at an approximate depth of 25 mm. The diagram illustrates the multilayer structure, including skin (1.1 mm), adipose tissue (12.6 mm), and muscle (17.60 mm), as well as the measurement path across the tissue layers.

3. Results

3.1. Numerical Model Setup and Convergence

A mesh convergence analysis was carried out to find a discretization that gave mesh-independent solutions and kept the computational cost reasonable. The temperature at the muscle boundary, which is the main region of interest, was monitored for assessing ten meshes ranging from 5441 to 42,181 elements. Also, as the mesh density increased, the predicted muscle temperature gradually converged, but the computational cost increased proportionally. The predicted temperature in the meshes considered differed only by 0.06 °C, indicating that further mesh refinement led to insignificant changes in the numerical solution. The solution time was 9 s for the coarsest mesh and 24 s for the finest mesh, respectively. The results are summarized in Table 6.

Table 6. Mesh convergence analysis showing the influence of mesh refinement on computational cost and predicted temperature at the muscle boundary.

Mesh ID	Number of Elements	Solution Time (s)	Temperature (°C)
M1	5441	9	38.33
M2	8732	11	38.35
M3	11,560	12	38.36
M4	14,982	13	38.37
M5	18,205	14	38.37
M6	22,183	15	38.38
M7	26,740	17	38.38
M8	31,506	19	38.38
M9	36,892	21	38.39
M10	42,181	24	38.39

For the rounded values in Table 6, the largest difference between consecutive meshes was $0.02\text{ }^{\circ}\text{C}$, equivalent to approximately 1.5% of the temperature rise of the finer mesh. The full range from M1 to M10 was $0.06\text{ }^{\circ}\text{C}$, corresponding to 4.3% of the $1.39\text{ }^{\circ}\text{C}$ rise predicted by M10. These percentages re-express the existing convergence results; no additional mesh simulations were performed.

These results indicate that mesh M6 was selected for all subsequent simulations and has 22,183 elements. The mesh provided a balance between computational efficiency and numerical accuracy. The calculated muscle temperature was within $0.01\text{ }^{\circ}\text{C}$ of the finest mesh (approximately 0.7% of its temperature rise above baseline), but the computational cost was reduced by ~38%.

3.2. Temperature Distribution and Thermal Response

The spatial temperature distribution in the multilayer tissue model was evaluated using the isothermal contours after 15 min of superficial thermotherapy. The thermal response of the M-size anatomical model exposed to hot-pack temperatures of $38\text{ }^{\circ}\text{C}$ and $44\text{ }^{\circ}\text{C}$ is shown in Figure 4.

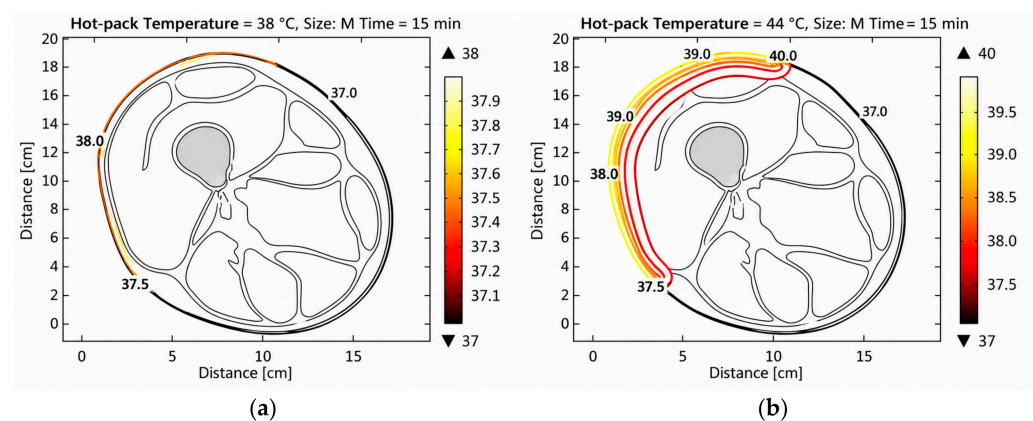


Figure 4. Isothermal contours illustrating the temperature distribution in the M-size anatomical model after 15 min of superficial thermotherapy using hot-pack temperatures of (a) $38.38\text{ }^{\circ}\text{C}$ and (b) $44\text{ }^{\circ}\text{C}$. The lower-temperature condition produced a muscle boundary temperature increase of approximately $0.14\text{ }^{\circ}\text{C}$, whereas the $44\text{ }^{\circ}\text{C}$ hot-pack increased the muscle boundary temperature to approximately $38.38\text{ }^{\circ}\text{C}$. The spatial axes are expressed in centimeters, and the color bars indicate temperature in degrees Celsius.

At an application temperature of $38\text{ }^{\circ}\text{C}$ (Figure 4a) the temperature elevation was mostly limited to the superficial tissue layers, with only limited heat penetration to deeper regions. The result was a marginal temperature increase in the muscle boundary, about $0.14\text{ }^{\circ}\text{C}$ from the initial condition.

A higher hot-pack temperature of $44\text{ }^{\circ}\text{C}$ (Figure 4b) resulted in a more widespread thermal diffusion into the tissue. The resulting thermal gradients penetrated deeper into the multilayer structure, increasing the temperature of the muscle boundary by $\sim 1.38\text{ }^{\circ}\text{C}$ after 15 min of treatment. The isothermal contours showed a smoother temperature gradient from the skin surface to the muscle layer than the $38\text{ }^{\circ}\text{C}$ situation.

The comparison between both heating conditions shows that an increase in hot-pack temperature results in increased heat penetration and higher temperatures in the deeper tissue layers, considering the evaluated conditions.

3.3. Effect of Adipose Tissue Thickness on Heat Transfer

To evaluate the effect of anatomical variability on heat transfer during superficial thermotherapy, the M-size and XL-size anatomical models were treated under the same

conditions (44 °C hot-pack for 15 min) and their thermal responses were compared. The only geometric difference between the models was the adipose tissue layer thickness, allowing us to isolate the influence of this anatomical asymmetry on heat propagation.

The time evolution of the temperature at the muscle boundary is shown in Figure 5. In the M-size model, the temperature at the muscle boundary increased by about 1.38 °C after 15 min of heating, indicating that the thermal stimulus applied reached the deeper layers of tissue effectively. Conversely, the temperature increase of the XL-size model was much lower, just 0.21 °C under the same boundary conditions. In this configuration, temperature gradients were mainly confined to the superficial tissues and did not propagate much towards the muscle layer.

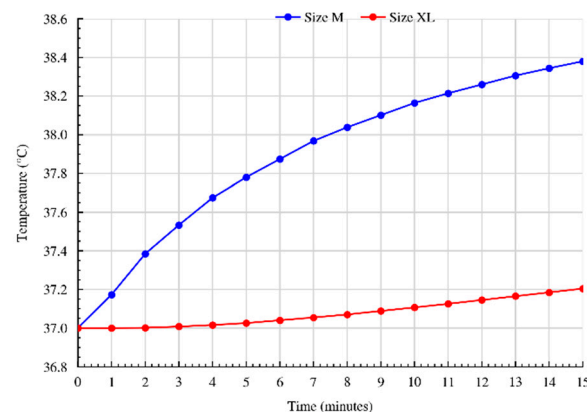


Figure 5. Temperature evolution at the muscle boundary for the M-size and XL-size anatomical models during 15 min of superficial thermotherapy using a 44 °C hot-pack. Increased adipose tissue thickness markedly reduced the temperature rise within the deeper tissue layers.

The comparison shows that geometric differences related to adipose tissue thickness significantly alter the predicted thermal response even when the heating conditions are identical. The time to reach the same temperature increase in the XL-size model was much longer than in the M-size model. From the results it was extrapolated that the duration of the treatment of the XL-size configuration should be longer than 1 h to increase by 1 °C at the muscle boundary under the same thermal conditions.

3.4. Experimental Comparison with the Finite Element Model

Figure 6 shows the temporal evolution of the experimentally measured temperatures at various depths in the tissue. A progressive increase in temperature in the muscle layer was observed during the heating period, with the final temperature increase being approximately 1.16 °C after 16 min of thermal stimulation.

The experimental results show a transient temperature increase in the multilayer tissue under the evaluated heating conditions. To examine the influence of the perfusion mismatch, paired simulations were evaluated with the tissue-specific perfusion rates and with the perfusion term set to zero.

Figure 7 compares the temporal evolution of muscle-boundary temperature during 15 min of heating with a prescribed 38 °C hot-pack temperature. All curves were initialized at 22.26 °C. The M-size zero-perfusion simulation produced the largest temperature increase, reaching approximately 30.5 °C, whereas the corresponding perfused simulation reached approximately 26.8 °C.

For the XL-size model, the zero-perfusion simulation reached approximately 24.2 °C, while the perfused simulation remained near 22.3 °C. The experimental muscle temperature increased from 22.26 °C to approximately 23.3 °C. At 15 min, the M-size perfused and zero-perfusion predictions exceeded the experimental value by approximately 3.5 °C and

7.2 °C, respectively. These are endpoint differences calculated from the rounded values reported above, not time-averaged error metrics.

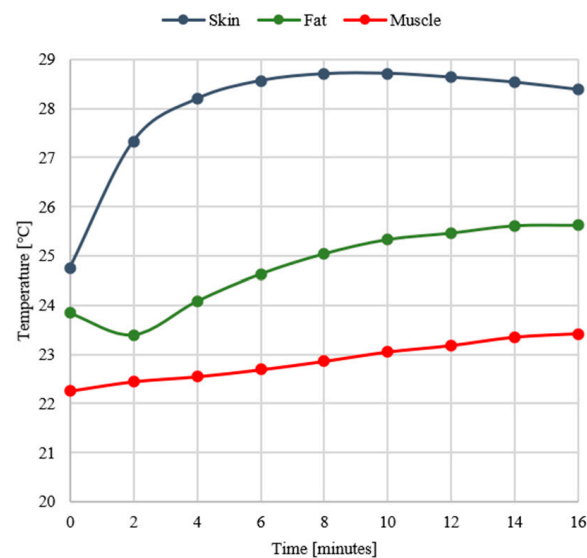


Figure 6. Experimental temperature evolution measured at different tissue depths in *ex vivo* porcine tissue during superficial thermotherapy.

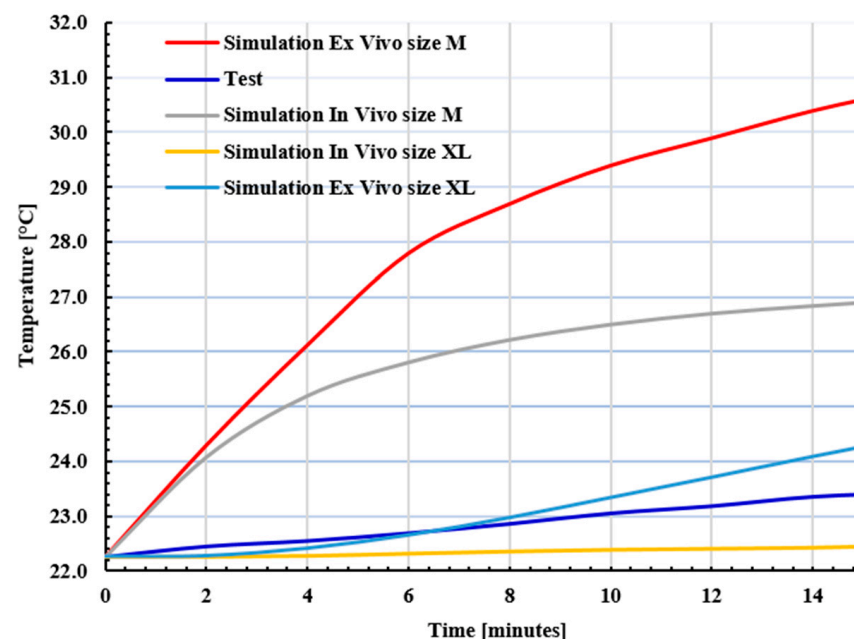


Figure 7. Temporal evolution of temperature at the muscle boundary during 15 min of superficial thermotherapy with a prescribed 38 °C hot-pack temperature. The perfused and zero-perfusion M-size and XL-size simulations are compared with the experimental porcine-tissue measurements.

4. Discussion

This paper demonstrates that anatomical variability, particularly in adipose tissue thickness, significantly affects heat transfer in superficial thermotherapy. The numerical simulations demonstrated a significantly different thermal response for the two anatomical configurations under the same heating conditions. In the M-size model, after 15 min of heating with a 44 °C hot-pack, the temperature at the muscle boundary increased by approximately 1.38 °C, whereas in the XL-size model, only a 0.21 °C increase was observed. These results suggest that geometrical differences related to the thickness of the adipose

tissue significantly impact heat propagation, indicating that standardized thermotherapy protocols based on the assumption of anatomically symmetric conditions may not result in equivalent thermal doses in patients with different body morphologies.

These findings are consistent with previous studies that have demonstrated less deep-tissue heating in subjects with greater adiposity. Petrofsky et al. [1] reported that the transfer of heat to deeper tissues was slowed by an increased thickness of the subcutaneous fat layer. Kominami et al. [14] reported that traditional hot-pack therapy increased superficial temperature mainly and had little effect on the underlying muscle. However, unlike these studies, which were mainly experimental, the present work combines an FEM with experimental comparison to quantify the effect of systematic changes in adipose tissue thickness on heat propagation. This combined framework offers a controlled setting to assess anatomical asymmetry, allowing comparisons that would be challenging to evaluate experimentally while providing an *ex vivo* comparison of the heating response.

The observed behavior is interpreted from a heat transfer point of view as a consequence of the relatively low thermal conductivity of adipose tissue, which limits the heat propagation into deeper regions and promotes heat retention in the superficial layers. Thus, the increased thickness of the adipose tissue decreases the effective thermal gradient to the muscle, leading to the attenuated thermal response observed in the XL-size model. These results are consistent with previous work that has found subcutaneous fat to be an important contributor to thermal diffusion during superficial heating [12]. The simulation results also demonstrate the increase in the insulating effect with the increase in adipose tissue thickness, emphasizing the significance of incorporating patient-specific anatomical features in computational bioheat models.

The analysis was limited to two anatomical configurations and cannot establish a continuous relationship between adipose tissue thickness and muscle heating. Intermediate morphologies may produce intermediate responses, but this was not tested, and linearity cannot be assumed. Additional configurations or a systematic sensitivity analysis are required to quantify the relationship and assess its generalizability.

These results have one important clinical implication: the duration of treatment. Extrapolation of the XL-size model suggests that under the same thermal conditions, more than one hour of continuous heating would be required to increase the temperature by 1 °C at the muscle boundary. These exposure times are not recommended clinically, since prolonged application of temperatures around 44–45 °C increases the risk of skin burns and thermal injury [19]. Thus, extending the treatment time cannot be considered a feasible way to compensate for the lower heat penetration into subjects with thicker adipose layers. These results instead indicate the need to consider patient morphology in the selection of heating conditions as part of treatment protocols to achieve consistent therapeutic outcomes.

The experimental measurements provide a descriptive comparison with the simulations under the matched initial temperature of 22.26 °C and the nominal 15-min, 38 °C heating condition. At 15 min, the experimental muscle temperature was approximately 23.3 °C, compared with 26.8 °C for the perfused M-size simulation and 30.5 °C for the zero-perfusion M-size simulation; the corresponding endpoint differences were approximately 3.5 °C and 7.2 °C. Removing perfusion alone therefore did not improve agreement. The idealized constant-temperature boundary supplies more heat than a cooling hydrogel pack and can overestimate tissue temperature. Postmortem changes in water content, moisture, tissue structure, and thermal properties may also affect the experimental response. Accordingly, the new simulations constitute a perfusion sensitivity analysis and a qualitative comparison, rather than quantitative validation of the physiological model. Additionally, it is important to note that the difference between the temperatures reached by the experi-

mental test and those from the simulation could also be attributed to changes in the sample of ex-vivo tissue after the animal's death. As time passes, the tissue can change its content and distribution of water, humidity losses, structural alterations, and cellular degradation; all these conditions can modify its thermal properties and consequently its capacity to absorb, store, and transfer the heat applied by the hot-pack. In the simulation, ideal tissue properties were used, while in the experimental evaluation the tissue properties are strongly related to the time after the animal's death and to the conditions of conservation of the ex-vivo tissue.

The paired simulations show the expected heat-removal influence of perfusion: setting the perfusion term to zero increased the predicted temperature, particularly in the M-size model. Because the zero-perfusion M-size prediction diverged further from the experiment, the perfusion mismatch alone does not explain the observed difference. Thermal models were performed with and without perfusion using the Pennes bioheat formulation. It was observed that perfusion plays an important role in reaching a more realistic heat distribution in the in vivo tissue, because blood perfusion contributes to transporting and dissipating thermal energy to regions far from the heating zone. For this reason, using parameters such as perfusion helps to estimate the thermal distribution and its evolution under physiological conditions with higher precision.

Finally, the small differences between the anatomical dimensions of the porcine specimens and the computational geometry and the measurement uncertainties may also contribute to the observed deviations.

To summarize, this study shows that anatomical asymmetry due to differences in adipose tissue thickness has a strong influence on heat transfer during superficial thermotherapy. The results highlight the limitations of standardized treatment protocols that do not consider patient morphology and the importance of anatomically informed computer models in predicting patient-specific thermal responses. In addition to the experimental comparison, this work provides quantitative evidence that systematic variations in the tissue geometry alone can significantly modify the delivered thermal dose, supporting the development of personalized thermotherapy protocols based on individual anatomical characteristics. A comparison between the present and previous experimental and computational studies is summarized in Table 7.

Table 7. Comparison of deep-tissue heating reported in thermotherapy studies.

ΔT (Muscle)	Condition	Approach	Study
$\sim 1^\circ\text{C}$	High body fat	Experimental	Petrofsky et al. [1].
Not reported	Hot-pack	Experimental	Kominami et al. [14].
$\sim 2.8^\circ\text{C}$	Hot-pack	Experimental	Hawkes et al. [2].
$\sim 1^\circ\text{C}$	Knee model	FEM	Flores Cuautele et al. [13].
Not reported	Patient	FEM	Pakarinen et al. [15].
$\sim 1.38^\circ\text{C}/\sim 0.21^\circ\text{C}$	Hot-pack	FEM + Experimental	Our study

5. Conclusions

In this study, we conducted a combined computational and experimental study of heat transfer during superficial thermotherapy. A finite element model of the lower limb with anatomically representative multilayer tissues was used. The experimental comparison included perfused and zero-perfusion simulations initialized at 22.26°C and evaluated

for 15 min under a prescribed 38 °C boundary. Their divergence from the experimental endpoint shows that the present measurements do not quantitatively validate the physiological model.

The results indicate that the thickness of the adipose tissue significantly impacts the propagation of heat to the deeper tissues. The M-size model under the same heating conditions showed an increase of about 1.38 °C at the muscle boundary after 15 min of heating with a hot-pack at 44 °C, while the XL-size model showed only an increase of 0.21 °C. These results show that anatomical asymmetry due to differences in adipose tissue thickness can significantly impact the delivered thermal dose, suggesting that standardized thermotherapy protocols based on uniform anatomical assumptions may not offer equivalent therapeutic effects across patients.

To summarize, this study provides quantitative support for considering patient morphology in the design of superficial thermotherapy protocols. The proposed FEM framework can assess heat propagation under different anatomical configurations and can be regarded as a useful tool for the development of anatomically informed and patient-specific treatment strategies. Future work will include the incorporation of temperature-dependent blood perfusion models, transient hot-pack cooling, and in vivo validation to further improve the physiological realism and clinical applicability of the proposed methodology.

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Abbreviations

FEM	Finite Element Method
CAD	Computer-Aided Design
M	Medium-size anatomical model
XL	Extra-large anatomical model
°C	Degrees Celsius
min	Minutes
mm	Millimeters
NI	National Instruments

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