Computational aspects in numerical simulation of skin tissues

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Abstract – The aim of this paper is to present some computational aspects in numerical simulation of the human skin tissue. A multi-layered model is considered for the skin tissue with emphasis on the mathematical modelling and numerical models in space 2D and 3D (for axisymmetric fields). A coupled model involves an interdisciplinary study including physiology, biology, neurology, heat transfer, fluid mechanics and computer-aided analysis of the coupled phenomena.

We limit our study to some particular cases with emphasis on the numerical simulation of the heat transfer in the skin and its thermomechanical behaviour.

Keywords: Skin tissue; Biothermomechanics; Numerical methods; Thermal systems; Coupled problems; Finite element method.

1 Introduction

Skin is the largest single organ of the human body that plays important roles in the thermoregulation, host defence and sensory system. Accurate evaluation of the spatial and temporal distribution of the temperature in the skin is of great importance in the development and applications of different treatments of the diseases involving the skin tissue. It is obviously that the effectiveness of these treatments is governed by the coupled thermal, mechanical, biological and neural responses of the affected skin tissue.

The analysis of the mechanical and thermal phenomena in a skin is of great importance and contributes to a variety of the medical applications and space and military activities. In Fig. 1 the structure of the skin is presented. This organ is the house of many phenomena including heat transfer, blood circulation, sweating, metabolic heat generation and the interaction with the surrounding environment. The skin properties are influenced by a variety of the factors such as temperature, pressure, damage, age, gender, hydration, site, etc.

The skin is an active, self-regulating system: heat transfer through skin dramatically affects the state of the skin, which can lead to the redistribution of skin blood flow over the cutaneous vascular network, whereby influencing the thermal response of the skin tissue. For example, in a neutral environment, the skin receives 5-10% of the cardiac output, which can be zero in a cold environment and can increase to 50-70% in a hot environment.

There is a natural coupling of the strain, stress and temperature in the skin tissue. A non-uniform distribution of the temperature can cause a thermal stress that contributes to the thermal pain sensation.

The blood perfusion has little effect on thermal damage but large influence on skin temperature distribution, which, in turn, influences significantly the resulting thermal stress field. The stratum corneum is very thin but has a large effect on the thermomechanical behaviour of skin, especially in the modelling of skin thermal stresses. Thus, the thermally induced mechanical stress appears due to the thermal denaturation of the collagen, the major constituent of dry skin. This constituent is macroscopic thermal shrinkage and its hydration level is changed in the denaturation process. Thermal denaturation of a collagenous tissue can lead to important changes of mechanical, thermal, electrical and optical properties of the skin tissue. When collagen is heated, the heat-labile highly organized crystalline structure to a random, gel-like state, which is the denaturation process.
From a simple visual analysis of the Fig. 1 we conclude that each layer has distinct physical properties, especially in the thermal behaviour of the skin. More, even within the same layer, there is a large non-homogeneity and anisotropy due to presence of the blood vessels. Consequently, a linear mathematical model or an analytical solution of the skin behaviour is not possible. Only a numerical model can lead to an approximate solution, can predict the effects of different factors on the skin behaviour.

2 Models for the skin thermomechanics

In the professional literature for the modelling of the human body and thermal comfort, there is a large variety of models on the heat transfer in different tissues of the human body, including the influence of the blood flow in the vascular network. These models can be included in one of these four classes: continuum models, vascular models, hybrid models and models based on the porous media theory.

In the numerical simulation of the skin thermomechanics, numerical models are used because of the large computing power of the advanced computers nowadays. Our work presents a numerical model based on the finite element method (FEM) with emphasis on the thermomechanical phenomena of the skin. The heat transfer in skin tissue is mainly a heat conduction process coupled to complicated physiological processes as blood perfusion that represents a heat source (or a sink). We consider the multi-layer structure of the skin with emphasis on three- and four-layer models. A three-layer structure includes the epidermis, dermis and subcutaneous tissue. A four-layer model is presented in Fig. 1 with stratum corneum, living epidermis, dermis and fat.

In the numerical simulation of the thermomechanical analysis of the skin we considered a two-dimensional (2D) physical model presented in Fig. 2, where the skin tissue is considered as a perfect, infinitely wide/long plate of thickness \( H \) and width \( L \). In a Cartesian coordinate system \( Oxyz \), \( z \) denotes the depth of the plate and \( x \) and \( y \) define the plane of a cross section in the skin layer.

3. Modelling of the heat transfer

Mathematical model for the thermal field is Pennes equation of the bioheat [1]. The Pennes bioheat equation describes the thermal behaviour based on the classical Fourier’s law and has the following form in a space two-dimension (2D):

\[
\frac{\partial}{\partial x} \left( k_x \frac{\partial T}{\partial x} \right) + \frac{\partial}{\partial y} \left( k_y \frac{\partial T}{\partial y} \right) + q + w \rho_b c_b (T - T_b) = \rho c \frac{\partial T}{\partial t}
\]

Here, \( \rho, c \) and \( T \) denote density, specific heat, and temperature of tissue. The specific heat, density and perfusion rate of blood are denoted by \( c_b, \rho_b \) and \( w_b \), respectively. The heat source is denoted by \( q \). \( T_b \) is the arterial temperature and we regard as a constant.

The heating source \( q \) is viewed as the sum of two components: \( q = q_{\text{met}} + q_{\text{ext}} \). The first component \( q_{\text{met}} \) is due to the metabolic heat generation in the skin tissue and \( q_{\text{ext}} \) is the heat generated by other heating methods.

The equation (1) is solved with specified initial and boundary conditions. Initially, the skin tissue has a temperature distribution \( T(x, y, 0) = T_0 \), and at time \( t=0 \) the skin surface (at \( y=L \)) is suddenly exposed to a thermal agitation, derived from either a hot contacting plate, a convective medium, or a constant heat flux, whereas on the bottom, \( y=0 \) the surface is held at the core temperature, \( T_c \), or thermally insulated. The effect of blood perfusion is regarded as a heat source.
under heating (or heat sink under cooling) distributed uniformly inside the tissue.

In the model from the Fig. 2, we did not included the hair /fur of the skin, although the hair strands can be so dense that they can trap a layer of air and thus work as an insulation layer. Since the thermal conductivity of hair is about 14 times greater than that of air (about 0.37 W/mK for human hair and 0.026W/mK for air), every single hair strand also works as fin, which enhances the heat transfer from the skin and thus is an unwanted effect in the thermal insulation sense due to the heat loss. In many models the air-hair layer is treated as an orthotropic material, that is, it is considered uniform on the plane parallel to the skin surface.

3.1 Modelling of the sweat gland
The sweating contributes significantly to the thermoregulation of the skin. Heat is lost from the body through skin by two mechanisms[1]:
- Insensible perspiration
- Sweat vaporization from the skin surface

The sweating can be:
- latent sweat
- sensible sweat

For latent sweat, in the secretory portion, the sweat absorbs heat and changes to vapor, which flows along the straight dermal duct into the superficial layer of the skin where, due to the low pressure and low temperature, it releases the heat and changes to fluid in the spiralled duct. For sensible sweat, the sweat flows into the skin surface and then evaporates.

4 Thermomechanical modelling of skin
As previously discussed, skin can be modelled as a multi-layer structure. From the mechanical viewpoint the layers are similar, that is the thermal properties have the same order of magnitude. The mechanical properties vary greatly from one layer to another (up to three orders of magnitude). Consequently, the mechanical behaviour of the skin can be assimilated to a laminated composite structure, with each layer assumed to be uniform with linear, orthotropic thermoelastic properties.

In a simplified model of the skin tissue, the thermal and mechanical behaviours of the skin can be treated as uncoupled problems, that is, the mechanical behaviour has no influence on thermal behaviour and vice versa. In this approach, a sequential algorithm can be developed. Firstly, the temperature field in skin tissue is obtained from solving the governing equations of biological heat transfer; which is then used as the input to the mechanical model, from which the corresponding thermal stress field can be obtained.

4.1 Skin biothermomechanics
Thermal damage (denaturation) of the skin can be modelled as a chemical rate process. In this area many models were proposed, but most of them have similar format using a first order Arrhenius rate equation. In modelling of the damage we can calculate the time for the appearance of the irreversible damage at temperature T. The damage is related to the rate of protein denaturation and exposure time at a given absolute temperature T.

Skin biothermomechanics is defined as the response of the skin under thermomechanical loading. Collagen is the major thermal constituent of the skin and the thermomechanical loading leads to damage – the thermal denaturation of collagen. The effects of heating on collagen can be reversible or irreversible and the behaviour of the collagenous tissue and shrinkage depend on several factors, ones of them
being: the collagen content, the maximum temperature reached and exposure time, the mechanical stress applied to the tissue during heating, and aging.

To measure the thermal denaturation and heating-induced damage a set of metrics were proposed and used, including biological metrics and mechanical metrics. As biological metrics we remember enzyme dezactivation and extravasation of fluorescent-tagged plasma proteins; as mechanical metrics we have the thermal shrinkage or optical metrics such as thermally induced loss of birefringence.

4.2 Thermal damage of the skin tissue

The skin damage can be represented as a chemical rate process that can be calculated by using a model that was proposed by Arrhenius and developed by other researchers [5]. Damage is related to the rate of protein denaturation (k) and exposure time (τ) at a given absolute temperature T. The measure of thermal injury Ω was introduced and its rate k is defined by:

\[ k(T) = \frac{d\Omega}{dT} = A \cdot e^{-\frac{E_a}{RT}} \]  

where A [1/s] is a material parameter equivalent to a frequency factor, E_a is the activation energy, R=8.314 [J/mol K] is the universal gas constant and T(x, t) is local tissue temperature (K). The constants A and E_a are obtained experimentally.

Many researchers proposed other models. In [1] the reaction rate of the thermal damage process is computed with formula:

\[ k(T) = e^{-\frac{E_a}{RT}(T-x)} \]  

The following correlations between Ω-values and degrees of burn injury were found:

- Ω=0.53 is the first degree burn (irreversible epidermal injury)
- Ω=1.00 is the second degree burn
- Ω=10^4 is the third degree burn (complete trans-epidermal necrosis)

5. Some cases of study

We consider some practical cases using 2D models and we seek the solution of the problems for different boundary conditions.

**Case 1: a three-layer model for a hot contacting plate**

The first case is when the skin is heated at the surface by a heat source with a constant temperature, e.g. in contact with a hot plate, while the bottom of the skin tissue is kept at body temperature. T_c, T_x is the ambient temperature.

Because of the symmetry we consider a half of the analysis domain as in Fig. 4 for a three-layer model. The skin is divided into three layers with different properties: the epidermis with thickness of 0.1 mm, dermis with thickness of 1.5 mm and subcutaneous fat with thickness of 4.4 mm.

Boundary conditions are the following:

1. Dirichlet condition:
   \[ T = T_c \quad \text{at} \quad y = 0 \]  

2. Convective boundary condition:
   \[ -k \frac{\partial T}{\partial y} = h(T - T_x) \quad \text{at} \quad y = H \]  

3. Neumann condition:
   \[ -k \frac{\partial T}{\partial x} = 0 \quad \text{at} \quad x = 0; x = L \]  

In Eq. (4) h is the convective coefficient with value 7 in our target example. The values of the geometric and thermal physical properties for the thermal analysis are presented in Table 1. For the stress analysis the parameters of the system are presented in Table 2.

Table 1

<table>
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<th>Epidermis</th>
<th>Dermis</th>
<th>Fat</th>
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<td>Thickness [mm]</td>
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<td>1.5</td>
<td>4.4</td>
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<tr>
<td>Density [Kg/m³]</td>
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Table 2

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<th>Parameters</th>
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<th>Fat</th>
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<tbody>
<tr>
<td>Thermal expansion coefficient</td>
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<td>0.0001</td>
<td>0.0001</td>
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<td>Young’s modulus</td>
<td>100</td>
<td>10</td>
<td>0.01</td>
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![Fig. 4 Analysis domain for case 1](image-url)

The skin is divided into three layers with different properties (see Fig.1): the epidermis with thickness of 0.1 mm, dermis with thickness of 1.5 mm and subcutaneous fat with thickness
of 4.4 mm. Epidermis can be divided into 2 layers: stratum corneum and living epidermis but in our study we use an idealized skin model with the structure from Fig. 1.

At the final time the map of the temperature field is shown in Fig. 6. The distribution of the temperature along the axis Oy is illustrated in Fig. 7. It is obviously that the temperature value decreases in direction of the subcutaneous fat.

In the numerical simulation we used the finite element method [3]. The analysis domain is divided in finite elements as in Fig. 5.

The initial temperature of the skin tissue is defined by normal parameters. At the initial moment a hot steel plate at the temperature of 90 °C is applied on the skin and a Dirichlet condition is imposed. We consider a transient process for a time interval of 20 s.

The temperature versus time can be computed in different interest points. For example, at the interface epidermis-dermis, in the point (0, 5.9) the temperature vs. time is shown in Fig. 8 for a time interval of 20 s.

Case 2: a two-layer model

In many medical treatments for a multitude of diseases and injuries involving skin tissue, the heating is mainly limited to the top of epidermis layer (laser heating) due to the exponential decrease of heat generation along skin depth. In other words the stratum corneum layer dominates the thermomechanical response of skin tissue so
that the analysis domain can be limited to epidermis.

The skin is reduced at two layers (Fig. 9): corneum with thickness of 0.00002 m and living epidermis with thickness of 0.00008 m. The spectrum of temperature and vectors of the heat flux are presented in Fig. 9. The temperature along the axis Oy is linearly (see Fig. 10) so that 1D-models can be used in the prediction of the temperature distribution in the medical treatments. More, analytical solutions can be used for temperature distribution in the skin tissue.

![Fig. 10 Temperature along the axis Oy](image)

In our simulations the temperature profile for initialization was considered the temperature in skin tissue under normal conditions. For this initialization, the steady-state equation of the bioheat equation was solved by a finite element model.

5.1 Damage profiles of the skin tissue

The burn simulation is based on Arrhenius burn integration using equations (2) and (3). The values of the constants R and E were: R=8.314 [J/mol K] and E=6.27 x 10^5 [K/mol]. In our simulations we considered the heat source intensity as being 60 °C and exposure time 40 s. We increased the source intensity at 61 °C and the damage function increased (see Fig. 11 with the damage function for two distinct values: solid line is for 60 °C and dotted line for 61 °C). From these graphs we can estimate the shortest time at which constant predetermined cutaneous surface temperature produces transepidermal necrosis (denoted by Ω=1). This time can be computed analytically from Eq. (2). Thus, for a constant temperature T and a specified value of E, the thermal damage function is [6]:

$$\Omega(t) = k(T) \cdot t$$

Assuming that Ω=1.0 denotes the beginning of irreversible damage, we can calculate the time for the appearance of irreversible damage at temperature T, as:

$$t_\Omega = \frac{1}{k(T)}$$

![Tissue damage function](image)

**Fig. 11** Damage function Ω vs. time

For our example, at T=60 °C, the value of \(t_\Omega=2.709 \text{ [s]}\), and for T=61 °C the value of time \(t_\Omega=1.376 \text{ [s]}\).

5.2 Thermal stresses

From the thermal viewpoint, a one-layer continuum model for skin tissue can be used in a numerical simulation because of the properties of the layers that have the same order of magnitude. From the mechanical viewpoint the mechanical properties are different from one layer to another. For this reason, the distribution of heat-induced stresses must be computed using a multi-layer model, with each layer assumed to be uniform with linear, orthotropic thermoelastic properties. The strain, stress and temperature are highly correlated so that the study of skin thermomechanics is a problem which is fully coupled.

The skin tissue can be modelled as a coupled problem where the interaction between heat transfer and thermomechanical model is natural and tightly. In a simplified approach, this coupled problem can be solved sequentially with assumption that the mechanical behaviour has no influence on thermal behaviour and vice versa. In this simplified model
the algorithm for numerical simulation involves two stages:
- The temperature field is obtained from solving biological bioheat transfer
- A thermomechanical model is solved where the temperature field is considered as input
  In thermomechanical model the parameters from the Table 2 are used. The vectors of strains are computed as a combination between thermal and mechanical loading.
  In this work we limited the study to the heat transfer model. In a future work we shall develop a model for coupled fields.

6 Conclusions
Thermally induced damage plays an important role in causing thermal pain so that the evaluation of the temperature distribution in the human skin, the heat transfer between different compartments of the human body and related thermomechanics in skin, play an important role in the study of the causes of pain and its relief.

The main reason for numerical simulation of the thermomechanics of skin tissue is the complex structure of the skin tissue that limits the accessibility needed for a detailed experimental solution and theoretical solutions. An accurate analysis of the thermal response in biological tissues is not possible because of the internal mechanisms that maintain body temperature, such as blood flow and metabolic generation. Almost all models in the thermomechanics of the skin tissue are developed for a specific type of therapy (hypothermia, hyperthermia, cryosurgery) and the numerical results are applied only to a specific treatment. Many aspects of the bioheat transfer during thermal therapy remain to be elucidated.

The objective of this paper was to understand the mechanisms of the biothermomechanics of the skin by using theoretical and numerical methods and to apply these results to special treatments as the thermal therapies. The difficulty of the development of accurate models of the human body leads to the idea that the numerical results must be compared with clinical data to validate or not our models.

Our models were developed in some simplified assumptions as:
- The material properties of the skin tissue are uniform within each of the three well-defined layers. In reality the skin does not consist of perfectly homogeneous layers.
- The material properties were considered as independent of temperature. In reality, the material properties depend on the current temperature T.
  These aspects can be included easily in the numerical model but there is insufficient data in the professional literature to model these real properties of the skin tissue.

References