A Simple Instrument to Measure the Thermal Transport Properties of the Human Skin

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***Abstract*—The evaluation of the skin thermal properties is relevant for many applications ranging from clinical dermatology to cosmetology. We introduce a simple passive device, capable of rapidly measuring skin thermal parameters using transient surface temperature measurements. Thanks to the development of an analytic thermodynamic skin model, tissue thermal diffusivity can be extracted from experimental data. For validation purposes, the thermal response of the apparatus has been modelled using a layered finite-element 3D model of the skin in thermal contact with a metallic measuring tip. Simplified 1D analytical and semi-analytical models have also been developed with the intent of modelling the thermal properties of the skin surface. The simplified models can be used to fit the thermal response measured by the device and to extract the thermal diffusivity in real time.**

***Index Terms*—skin measurement device, skin thermal properties, numerical modelling, analytical model**

I. Introduction

The skin, the largest organ of the body, exhibits a mean surface area of over two square meters and a mass of 4 to 10 kg in adults [1]. Overall, this represents more than 8% of the total body mass, and is the primary line of defense against pathogens, injuries and excessive water or heat loss. The thermal transport properties of tissue (i.e., density, heat capacity, and thermal conductivity) can reflect physical/chemical states of the skin, and thus may yield information relevant to clinical medicine, basic skin physiology research and cosmetic/aesthetic applications [2].

The experimental setups designed to measure thermal attributes typically monitor the transient surface temperature subjected to a thermal stimulation [2]. Extracting the thermal properties is challenging and requires the implementation of complex heat transfer models. Most thermal stimulation schemes rely on potentially dangerous flash lamps, laser systems or airflow modulation techniques, which are difficult to implement in practice and incompatible with medical usage regulations [3], [4]. In addition, very little quantitative information exists about the relationship between the thermal transport properties of the skin and the relevant physiological parameters such as hydration, vascularization or structure.

In the light of these limitations, several proposals have been done for biomedical measurement devices for the skin thermal properties [5], [6]. Recently, Rogers and co-workers have proposed a small epidermal device for the quantitative mapping of temperature and thermal transport characteristics of the skin [7]–[9]. The apparatus consists of an ultra-thin photonic device combining colorimetric temperature indicators with wireless stretchable electronics for temperature measurement. This elegant technology monitors the skin surface temperature by contact measurements instead of measuring the thermal radiation as it is often the case when performing non-contact measurements. Besides, the compliant, skin-like device exhibits an anatomically perfect fit and allows the measurement on locations of the body with complex topology.

Nonetheless, mapping the thermal parameters is in many cases not necessary. Instead, local measurements of the thermal transport characteristics are sufficient [3]. In addition, the extraction of the thermal properties and their correlation to physiological parameters is based on fittings of the experimental data to a simplistic analytical skin thermal model using two strongly coupled parameters. Based on these observations, we propose an alternative approach where the skin surface is cooled by thermal contact with a thermal reservoir having a lower temperature than the skin, and a high thermal conductivity and heat capacity. The proposed setup retrieves the thermal properties of the skin using a fully passive technique. The thermal parameters are extracted by fitting to an analytical function derived from a 1D homogenized thermal model of the skin. In the present study, the measurement principle and the fitting method are validated using a 3D fully numerical model of the skin thermal characteristics.

II. Material and Methods

A. Measurement device

Fig. 1 depicts the measurement principle of the device. One side of a massive hollow cylinder element is placed into thermal contact with the skin surface. If the cylinder temperature is lower than the skin surface temperature, the skin surface in contact with the cylinder will cool down until thermal equilibrium is reached. The dimensions and the material of the hollow cylinder are carefully chosen so that its temperature remains largely unaffected for several seconds after being in contact with the skin. After extensive testing, stainless steel or aluminium blocks of approximately 100 cm^3

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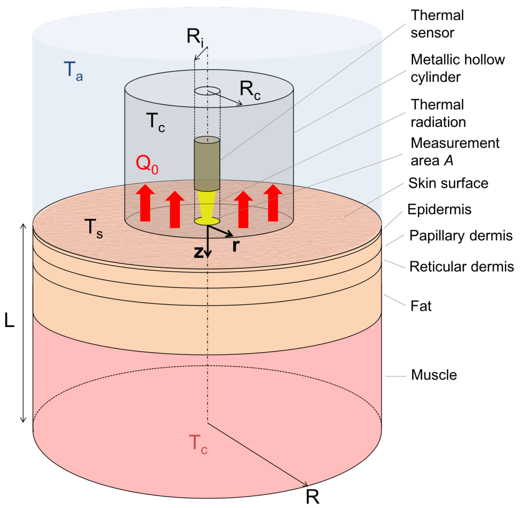

Fig. 1. **Scheme of the device**. The scheme describes the main characteristics of the measuring device and the measuring principles, and it also shows a typical layered skin model.

have been found to be sufficient to ensure that the skin in contact with the cylinder is subject to a constant temperature boundary condition during the measurement. In the middle of the cylinder, a small sensor monitors the transient thermal radiation of the skin. Inexpensive thermopiles can be used to achieve this task. The advantage of these type of sensors over contact temperature measurements lies in their acquisition speed and in the fact that they are independent of the heat transfer coefficient between the skin and the sensor. The transient thermal decay depends on the thermal properties of the skin. Many different other sensors can be easily incorporated into the device to achieve a better accuracy: electrical conductivity and impedance, reflection measurements of the stratum corneum in wavelengths ranging from UV to near IR, or even transepidermal water loss sensor (TEWL) [10], [11].

B. Thermal transport theory

The thermal response of the skin is well described by the so called Pennes model [12], the heat equation including the blood perfusion as a heat flux term,

$$\rho c_p \frac{\partial T}{\partial t} = \kappa \nabla^2 T + \rho_b c_p^b \omega \left(T_b - T\right) + Q_{ext} \tag{1}$$

ρ and c_p^b denote the mass density and the specific heat of blood, respectively, and T_b the blood temperature. The temperature of the deep skin layers, the dermis and the hypodermis, is stabilized by the blood perfusion. It is described by the blood perfusion rate ω. Any other external heat source or sink are included in Q_{ext}. They are mainly relevant in the boundaries of the calculation domain and act as heat reservoirs or sinks. The stationary temperature distribution of the skin is obtained from the stationary Pennes equation.

$$\kappa \nabla^2 T + \rho_b c_p^b \omega \left(T_b - T\right) + Q_{ext} = 0 \tag{2}$$

On the other hand, the temperature in a metal block is described accurately by the heat equation

$$\rho^{Al} c_p^{Al} \frac{\partial T}{\partial t} = \kappa^{Al} \nabla^2 T + Q_{ext} \tag{3}$$

The interface between the metal block and the skin, as well as between each skin layer, are assumed to be in perfect thermal contact. The temperature and the heat flux are continuous across the interface x_0:

$$\begin{aligned} T_A(x_0) &= T_B(x_0) \\ -\kappa_A \frac{\partial T}{\partial n_A}\bigg|_{x=x_0} &= -\kappa_B \frac{\partial T}{\partial n_B}\bigg|_{x=x_0} \end{aligned} \tag{4}$$

where A and B can be skin layers or the metal piece. The thermal contact with the surrounding air of the skin surface and the metal block is modelled by a convective heat flux assuming a heat transfer coefficient $h_{air} = 10\ W/(m^2 K)$. This value is typical for convective cooling by air at atmospheric pressure [13].

and an ambient temperature $T_{air} = 296\ K$.

C. Finite element 3D model

A 3D axisymmetric model of the skin in contact with a metallic block is built using COMSOL®. It is composed by an aluminium cylinder placed in contact with a four-layer (Stratum Cornea, Epidermis, Dermis, Hypodermis) skin model (see Fig. 1). The aluminium cylinder has an axial hole in the center which will be used for the temperature measurements. The dimensions of the model are detailed in a table.

TABLE I
DIMENSIONS OF THE ALUMINUM BLOCK AND THE SKIN LAYERS.

Name	**Dimension** (mm)
Cylinder radius	12.5
Cylinder height	31
Hole radius	1
SC depth	0.02
Epi depth	0.48
Derm depth	3.4
Hypo depth	5
Lateral width	50

The thermal response of each skin layer has been modelled using 3 parameters: the mass density ρ, the specific heat c_p and the thermal conductivity κ. For the deep layers of the skin, the dermis and the hypodermis, the blood perfusion acts as a temperature stabilizer. It is described by the blood perfusion rate ω. The thermal parameters of the skin model are detailed in table II.

The lateral limit of the calculation domain is thermally isolated. This condition is exact only far from the metal detector where the heat flux is vertical. To avoid numerical artifacts, the lateral width of the model has been set large enough so to ensure a vertical heat flux in the lateral boundaries. The lower limit of the Hypodermis has been assumed to have a fixed temperature equal to the blood temperature T_b. The outer skin layer (SC) and the walls of the aluminium detector contact

TABLE II
RELEVANT THERMAL PARAMETERS OF ALUMINUM, SKIN LAYERS AND BLOOD.

	ρ (kg/m^3)	C_p $(J/(kgK))$	κ $(W/(mK))$	ω (Hz)
Aluminum	2700	900	238	-
Stratum Cornea	1400	2000	0.28	0
Epidermis	1200	3600	0.24	0
Dermis	1200	3300	0.45	0.001
Hypodermis	1000	2700	0.19	0.002
Blood	1000	3700	-	-

with the air loose heat by convection. The heating of the surrounding air by the thermal contact and radiation to the skin or the Aluminum block is neglected. This simplification could be problematic within the hollow block where convection can be slow dissipating the extra heat, but the effect will be small for a typical measurement time (10-20 s).

D. Semi-analytic multilayer 1D model

The layered model can be solved in 1D using the variable separation method to find a solution to the Pennes equation. First, the solution is written as the sum of a stationary solution and a transient one $T(x,t) = T_0(x) + \Delta T(x,t)$. The full solution $T(x,t)$ is a solution to the full Pennes equation (1), and the stationary solution $T_0(x)$ is given by the stationary Pennes equation (2). The resulting equation for the transient perturbation $\Delta T(x,t)$ in a single layer is:

$$\rho c_p \frac{\partial \Delta T}{\partial t} = \kappa \nabla^2 \Delta T - \rho_b c_p^b \omega \Delta T \tag{5}$$

where the external heat source Q_{ext} has been set to zero. The equation for ΔT is linear and it can be solved by separation of variables $\Delta T(x,t) \equiv \chi(x)\tau(t)$:

$$\frac{\tau'}{\tau} = \frac{\kappa}{\rho c_p}\frac{\chi''}{\chi} - \frac{\rho_b c_p^b}{\rho c_p}\omega \tag{6}$$

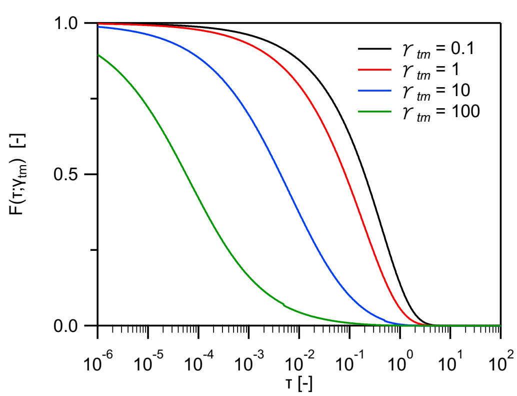

Fig. 2. Graphs of the function $F(\tau;\gamma_{tm})$.

The single layer solutions are exponentials with negative exponents for the time part and harmonic functions for the spatial part. The solutions depend on eigenvalues f and k that are related by (6). The coupling between the single layer solutions requires that the time part is common for all the layers $\tau_i(t) = \tau(t)$ and the spatial parts are joined requiring a full thermal coupling condition like in (4). The normal mode $u^n(x)$ comprises a linear combination of trigonometric functions within each layer defined by eigenvalues k_i^n, where i corresponds to a skin layer. The outer boundary conditions, convective heat flux with the air across the skin surface and fixed temperature $T(x_b,t) = T_b$ in the bottom, now yield an equation for the spatial eigenvalues which can be solved to obtain the eigenvalues corresponding to each normal mode.

The final solution $\Delta T(x,t)$ is a linear combination of normal modes $u^n(x)$ multiplied by a coefficient C^n and a time eigenfunction $\exp(-f^n t)$. Coefficients C^n are obtained by the initial conditions $\Delta T(x,0)$ using the scalar product $C^n = \rho c_p \int_{x_a}^{x_b} \Delta T(x,0) u^n(x) dx$ where x_a and x_b are the outer limits of the model.

E. Fully analytic single layer 1D model

An analytical one-dimensional (1D) model has also been developed. This 1D model uses the assumption that the lateral characteristic length scales are much larger than the vertical characteristic length scale, so that the three-dimensional geometry can be reduced to one space dimension (resolving in the vertical direction only) with a single interface, which is the tissue-air interface before contact and the tissue-metal interface after contact. In this reduction we also neglect the air channel inside the metal cylinder, which is justified by the assumption that the channel radius is much smaller than the cylinder radius. The 1D model for the evolution of the temperature distribution in the skin over time $T(z,t)$ [K] then takes the form of Pennes' bioheat equation [12] in a semi-infinite domain $z \geq 0$ (idealized "skin region"), together with a thermal contact condition at the skin surface and a prescribed initial temperature distribution.

The initial-boundary-value problem can be solved analytically using a Laplace transform in time. We obtain the following expression for the skin-surface temperature

$$T_s(t) = T_\infty(0) + (T_0(0) - T_\infty(0))\, F\left(\alpha\beta^2 t; \gamma_{tm}\right) \tag{7}$$

where the strictly monotonically decreasing function $F(\tau;\gamma_{tm})$ is

$$F(\tau;\gamma_{tm}) := e^{-\tau}\frac{\gamma_{tm}\,\mathrm{erfcx}\left(\sqrt{\gamma_{tm}^2\tau}\right) - \mathrm{erfcx}(\sqrt{\tau})}{\gamma_{tm} - 1} \tag{8}$$

Some graphs of F are shown in Fig. 2.

Equation (7) yield an analytical model for the evolution of the skin-surface temperature over time. The four model parameters $T_0(0), T_\infty(0), \alpha\beta^2, \gamma_{tm}$ can be used to fit the measured data.

III. OUTLINE OF EXPECTED RESULTS

The finite element model (Fig. 3a) describes in detail the spatial distribution of temperature within the skin and it will be used to model the thermal measurements of the **apparatus**. A detailed description of the skin temperature, shown in Fig. 3c),

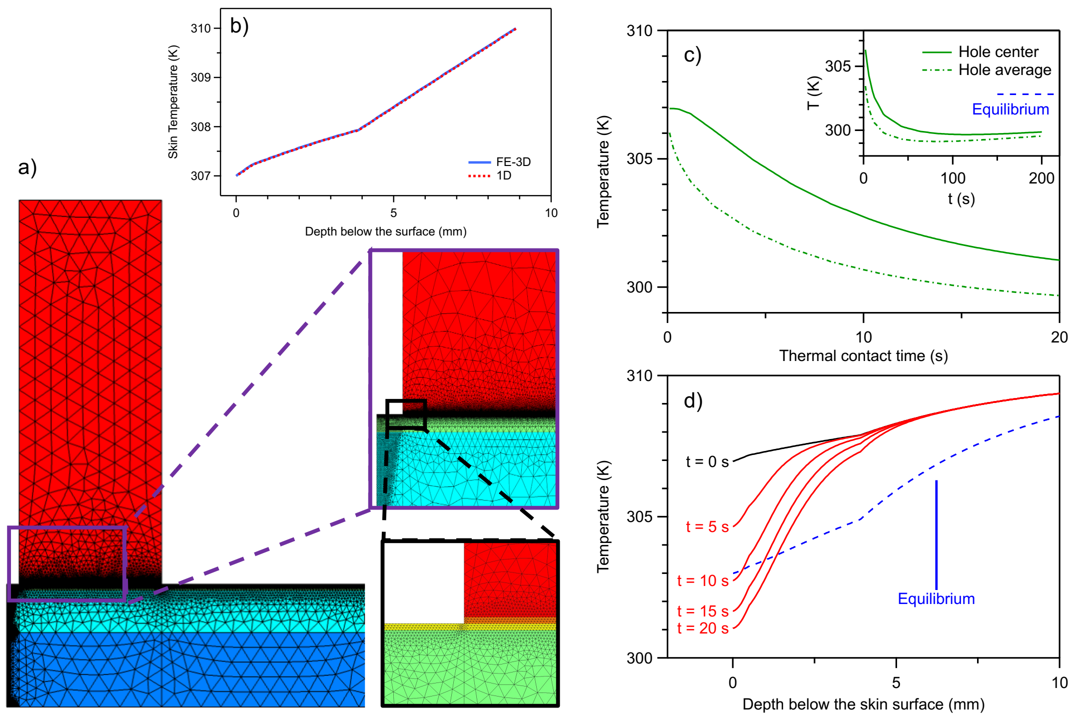

Fig. 3. **Finite-element mesh description, validation and model output**. a) Detail view of the geometry and finite element mesh used for the numerical simulation are shown. b) The finite-element stationary skin model temperature profile is validated using a multilayer semi-analytical model. c) Time evolution of the temperature in the skin surface at the center of the hole and averaged along the area of the hole. The inset shows the long times evolution and the thermal equilibrium value. d) The axial temperature distribution at different times after the beginning of the thermal contact with the Al block.

as well as, of the vertical temperature distribution, see Fig. 3d), have been calculated to assess the effect of the skin parameters on the thermal response. Special attention has been put to the variation of the thermal conductivity and the heat capacity within the physiologically relevant range. A four layer 1D model has been shown in Fig. 3b) to validate the finite-element 3D result for the stationary state before the thermal contact with the Al block is established.

A. Sensitivity study

The detector size is observed to also influence the thermal inertia of the Al block, limiting the duration of the measurement. In addition, the diameter of the central channel has a big influence on the sensitivity and the time response of the detector. The information extracted can be maximised by limiting the measured region to an area within the center of the measurement hole. When the thermal radiation from a sufficiently small area is collected, a delay in the cooling process is observed and at longer times, a speed-up of the cooling process is observed. From the measurement point of view, the delay of the cooling process can be used to increase the sensitivity to the thermal conductivity.

B. Model of the thermal response

A mathematical model to be used for fitting the detector response needs to be 1D in nature to be able to analyse the measured experimental data like the one in Fig. 3b. For that purpose, two approaches have been pursued. On one hand, an analytical single layer 1D model has been used to describe the temperature depth profile and the temperature evolution at the skin surface (Sec. II-E). Its main advantage is that it provides an analytical function that can be easily used to fit the data measured by the apparatus in-situ. On the other hand, a semi-analytical multi-layer model has been developed (Sec. II-D). It allows to describe the temperature depth profile in more detail at the price of a higher complexity. The main motivation for the second, more elaborate model, is to account for the radial heat flux near the skin surface. The heat flux obtained from the 3D model shows in Fig. 4 a strong outward radial flow towards the Al block right below the skin surface.

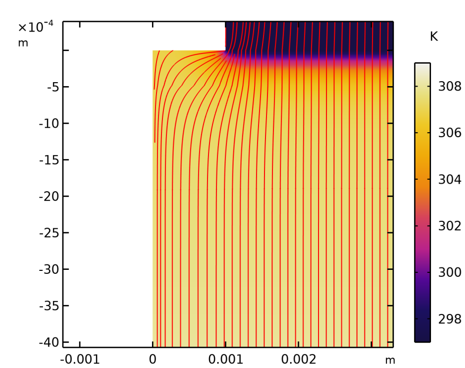

Fig. 4. **Heat flux**. The heat flux flows upwards towards the skin surface. Close to the surface a large of the heat flows radially outwards, towards the Al block.

The radial component of the flux will effectively remove heat from the skin surface and lower its temperature. The accurate description of the temperature evolution within a 1D model will require accounting for this effect as an effective heat sink near the skin surface. This can be implemented as a two layer model with the layer right below the skin surface being $\simeq 1\ mm$ thick and including a heat sink term, $A_s(T - T_s)$, similar to the perfusion term of the Pennes equation. The values of the parameters A_s and T_s will be empirically determined to fit the evolution of the skin temperature. The model will allow quantifying the influence of the radial heat flux in the skin surface. It will also help to determine the link between the relevant skin parameters (thermal conductivity and heat capacity mainly) and the measured values. The extraction of the thermal parameters from thermal response curves will be modelled and the obtained values will be discussed. In practice, the thermal diffusivity $\alpha = \kappa/(\rho c_p)$, will be extracted from the measured temperature evolution curves.

In a subsequent analysis the fully analytical single layer model will be used. The main interest on the model comes from its suitability to implement a fast reading of the apparatus measurement in real-time. The simplified model does not account for the heat flux flowing radially. As so, it cannot provide a direct estimation of the thermal parameters of the transient measurement. Despite this downside, it can be a good fit for the equilibrium states where the heat flux remains constant. It may possibly fit the new equilibrium reached after long enough contact has been set. In the first stage of the thermal contact, however, the temperature of the skin surface will not fit the time evolution described by the single layer 1D model. To overcome this difficulty, minimal modifications to the model accounting for the heat loss can be done. More specifically, a spatially non-uniform heat sink can be added which will be strongest near the surface. This can be implemented using a depth dependent heat sink temperature. The applicability of a model of this type will be explored. In light of the results, a measurement protocol and a data analysis algorithm will be proposed which can be used to extract the thermal properties of the skin. It will provide fast and easy to perform measurements wherefrom the skin physiological properties can be determined fast and non-invasively.

The proposed device when properly configured will allow getting fast measurements of the local thermal properties of skin. It will avoid using bulky measuring devices and elaborated thermal contacts attached to the skin surface. The accuracy of the temperature values will be enough for fast '*on the fly*' diagnosis in clinical applications. It can also be marketed as a consumer level device for the general public.

IV. Provisional conclusions

The measurement principle of a fast and non-invasive device to measure the skin thermal parameters has been studied. A finite-difference 3D model of the measurement device and the skin has been used to study in detail the thermal response of the skin and the temperature values measured by the device. Subsequently, the full temperature evolution information has been compared to a multilayer 1D model of the temperature evolution along the axial direction. This allowed us to understand the effect of the out of axis cooling device on the measured skin surface temperatures. Finally, a simplified 1D model of the skin surface temperature evolution has been developed with the aim of fitting the values measured by the device. For this purpose, a fully analytical single layer model has been developed and its range of application has been e.

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