# ESTIMATION OF BLOOD FLOW SPEED AND VESSEL LOCATION FROM FUNCTIONAL INFRARED IMAGING

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**Abstract: In this paper we present a novel method, based on a bioheat transfer model and high-resolution thermal imaging, to compute blood flow speed and to estimate vessel location for a vessel close to the skin surface. The model is adapted from Pennes' model and it assumes the form of a partial differential equation (PDE) with boundary conditions. A direct numerical simulation has been used to test the soundness of the proposed model. Then, the inverse problem has been solved both in steady and dynamic states on data provided by a functional infrared imaging. The method may extend actual possibility offered by medical thermal imaging, also opening new frontiers in biometrics and biomedicine.**

## Introduction

Thermal imagery and functional infrared imaging (fIRI) [ref] open new possibilities both for psychometrics investigations [1, 2, 3] and for biomedical applications [4, 5, 6]. In fact, combined use of high resolution thermal imaging systems and bioheat modelling allows obtaining functional and quantitative parameters describing specific signatures of autonomic functions. In addition, fIRI is a touchless technique. Such a feature is very important in the context of pshyco-physiological measurements where it is crucial that the subject feels as comfortable as possible.
After appropriate processing, fIRI provides quantitative information about variables other than temperature, like blood flow speed, respiratory, and perspiratory function that constitute a set of vital signs, suitable for monitoring a variety of disorders of dysfunctions.

In this paper we propose a fIRI-based method to compute blood flow properties in vessels proximal to the skin using thermal video as source of data.

The method considers the heat exchange processes in the first few millimetres below the surface and the effects of the underlying tissue. For a general discussion on bioheat transfer processes and models, we refer to [7] and to [3] for the general background on the computation of perfusion rate in skin with an underlying dense vascular structure. Bioheat model dealing with surface temperature control can be decomposed roughly into two categories: *continuum* and *geometric*. Continuum models relate the blood perfusion rate to the temperature as a function of the effective conductivity of the tissue and the source of heat in the body core [3, 8, 9, 12], but they fail taking into account the position and shape of large vessels and the tissue heterogeneity. Geometric models are based on the analytic reconstruction of the vascular network, which accounts for all local variations of the temperature near the individual vessels [10, 11], but great complexity and variability of the vasculature in the tissue layers under the skin does not allow their immediate generalization

This paper presents a general procedure to solve the problem of recovering missing information on vessel morphology and blood flow fluctuations through a novel model combining geometric and continuum model features.

## Overview of the model

The proposed model describes heat transfer in the surrounding regions of a large superficial vessel, which serves as volumetric heat source for the surrounding four-layer tissue structure. These layers (see Figure 1) are successively, in positive z direction (depth direction), the skin, the fat, the muscle, and the core; and they are assumed being isotropic with respect to thermal conductivity K(z), metabolic heat rate $q^M(z)$, density ρ, and specific heat c of the tissue. Vessel heating effect of the on the skin temperature depends on the vessel's location and shape as well as its blood flow speed and temperature.
As simplifying hypothesis, we consider a single large vessel constrained along the x direction (see Figure 1), and arranged in a periodical manner along the y direction. Such an assumption may result realistic, for example, for radial vessels of the wrist (see Figure 2). The thermal conduction in the tissue surrounding the vessel is therefore dominant in directions parallel (x) and perpendicular (z) to the skin. Therefore, we will


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consider a 2D model assumes the following form:

$$\rho\, c\, \frac{\partial \Theta}{\partial t} - \frac{\partial}{\partial x}(K(z)\frac{\partial \Theta}{\partial x}) - \frac{\partial}{\partial z}(K(z)\frac{\partial \Theta}{\partial z}) = q^{BL}(x,t) + q^{M}(x,z),\ (x,z) \in (0,L)\times(0,D), \quad (1)$$

where $q^M$ is the volumetric metabolic heat while $q^{BL}$ is the heat due to blood flow speed ubl in a vessel assimilated to a line source z =S(x). K(z) is the thermal conductivity of a particular layer, while ρ and c are the tissue density and specific heat respectively.

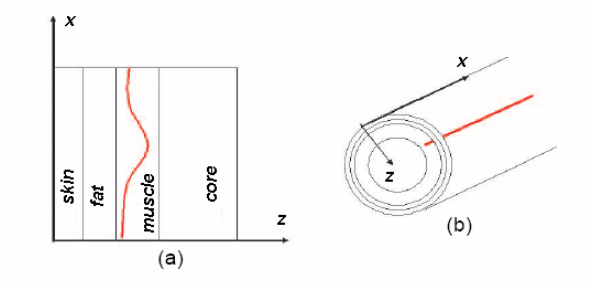


Figure 1: (a) Four layer tissue structure hypothesized by our model along with the coordinate system convention. The red curve represents the assumed position and shape of the vessel. (b) Cross section of the tissue.

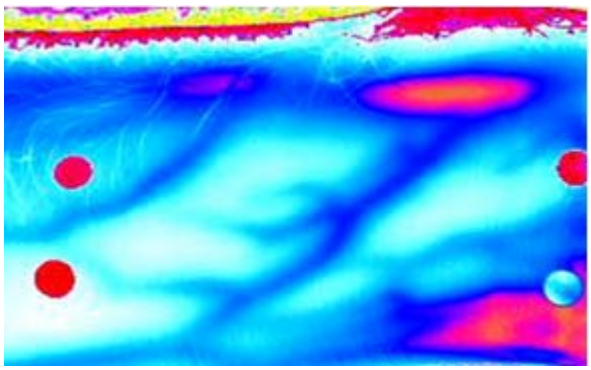

Figure 2: Thermal image of the wrist.

We impose the following boundary conditions:

$$\Theta(x,D,t) = \Theta_{core},\ x \in (0,L), \quad (2)$$

$$\Theta(x,0,t) = \Theta_{skin}(x,t),\ x \in (0,L), \quad (3)$$

$$\frac{\partial \Theta}{\partial z}(x,0,t) = \lambda\,(\Theta(x,0,t) - \Theta_{air}) + q_{ir},\ x \in (0,L), \quad (4)$$

$$\frac{\partial \Theta}{\partial x}(0/L,z,t) = 0,\ z \in (0,D). \quad (5)$$

λ is the convection heat transfer coefficient, which depends on air flow.
According to [13]: $\lambda = 2.7+7.4(v_{air})^{0.67}$ (W/m$^2$K), where $v_{air}$ is the air speed in (m/s). $q_{ir}$ is the radiation heat flux: $q_{ir} = \sigma\ \varepsilon\ (\Theta^4_{skin} - \Theta^4_{wall})$, where σ is the Stefan-Boltzmann constant and ε is the skin emissivity. $\Theta_{wall}$ is approximated by the temperature of the air.
The heat source term associated with blood flow is modelled as the composition of two factors: blood speed (variable in time) and vessel geometry (fixed), thus being represented as $q^{BL} = u_{bl}(t)r(x, z)$, where $u_{bl}$ is the unknown blood flow speed in the vessel. We assume that the vessel geometry r(x, z) is centred on the curve z =S(x) and represented by the modified bell function

$$r(x,z) = \mu \exp(-\frac{(z - S(x))^2}{\pi \nu^2_{app}}).$$

Where $\nu_{app}$ is the apparent radius of the vessel seen as a heat source. μ is defined as follows:

$$\mu = \rho_{bl} c_b \frac{A}{V}(\Theta_{vessel}(x,z,t) - \Theta(x,z,t))\ \ (J/m^4), \quad (6)$$

where $\rho_{bl}$ and $c_{bl}$ are the density and the specific heat of blood respectively, A is the vessel cross section, and V is the control volume of tissue. We assume that the temperature of the blood in the vessel is the same as the core temperature $\Theta_{vessel} = \Theta_{core}$.
In constructing our model, we have assumed that there is no heat flux between the domain of interest (0, L)×(0, D) with the rest of the body. Specific initial condition for the model will be defined later on.
The mathematical problem is to retrieve the unknown vessel location z =S(x) and blood flow speed $u_{bl}(t)$from the skin temperature $\Theta_{skin}$ obtained with an high-resolution infrared camera.
To facilitate the solution we derive a normalized formulation of the model:

$$\frac{\partial \theta}{\partial t} - \frac{\partial}{\partial x}(\tilde{K}(z)\frac{\partial \theta}{\partial x}) - \gamma\frac{\partial}{\partial z}(\tilde{K}(z)\frac{\partial \theta}{\partial z}) = \tilde{\mu}\ u_{bl}(t)\ (1-\theta)\exp(-\frac{(\gamma z - \tilde{S}(x))^2}{\eta}) + \tilde{q}^M(x,z),\ (x,z) \in (0,1)\times(0,1). \quad (7)$$

The normalized boundary conditions become:

$$\theta(x,1,t) = 1,\ x \in (0,1),\ t \in (0,T), \quad (8)$$

$$\theta(x,0,t) = \theta_s(x,t),\ x \in (0,1),\ t \in (0,T), \quad (9)$$

$$\frac{\partial \theta}{\partial z}(x,0,t) = \beta\theta(x,0,t),\ x \in (0,1),\ t \in (0,T). \quad (10)$$

$$\frac{\partial \theta}{\partial x}(0,z,t) = \frac{\partial \theta}{\partial x}(1,z,t) = 0,\ z \in (0,1),\ t \in (0,T). \quad (11)$$

The model becomes complete once we provide an initial condition $\theta(x, z, 0)$, where $x \in (0, 1)$ and $z \in (0, 1)$.


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**Model Computation**

We perform both direct simulation and solve the inverse problem. In each case we solve both the steady state and dynamic problems.

*Direct Simulation in Steady State*

We present hereby a direct simulation of the steady state problem for $(x, z) \in (0, 1)\times(0, 1)$ and $\gamma = 1$:

$$-\frac{\partial}{\partial x}(\tilde{K}(z)\frac{\partial\theta}{\partial x}) - \frac{\partial}{\partial z}(\tilde{K}(z)\frac{\partial\theta}{\partial z}) = \tilde{\mu}\, u_{bl}(t)\,(1-\theta)\exp(-\frac{(z-\tilde{S}(x))^2}{\tilde{\eta}}) + \tilde{q}^M(x,z), \quad (12)$$

with boundary conditions (8), (10), and (11).

This problem is well posed and has a unique continuous solution. We use a Finite Volume (FV) approximation with centred cells of size $h_x \times h_z$ on a regular space grid. Let us denote $\theta_{i,j}$ the average value of θ in the centred FV cells. The discrete version of Eq. (12) is:

$$-h_z\,(\Phi_{i+1/2,j} - \Phi_{i-1/2,j}) - h_x\,(\Phi_{i,j+1/2} - \Phi_{i,j-1/2}) = h_x\,h_z\,(\tilde{\mu}\,u(t)(1-\theta_{i,j})\exp(-\frac{(z_j-\tilde{S}(x_i))^2}{\tilde{\eta}}) + \tilde{q}^M(x_i,z_j)),$$

The heat flux values at the wall of the cells are approximated with:

$$\Phi_{i+1/2,j} = K_{i+1/2,j}\frac{\theta_{i+1,j}-\theta_{i,j}}{h_x},\ i = 1..N_x - 1,$$

$$\Phi_{i,j+1/2} = K_{i,j+1/2}\frac{\theta_{i,j+1}-\theta_{i,j}}{h_z},\ j = 1..N_z - 1,$$

and

$$K_{i+1/2,j} = \frac{1}{2}(K_{i+1,j}+K_{i,j}),\ K_{i,j+1/2} = \frac{1}{2}(K_{i,j+1/2}+K_{i,j}).$$

that can be solved numerically.

Figure 3 shows a solution of the steady problem (12) with boundary conditions (8), (10), (11) and the vessel location given by the formula:

$$S(x) = 1/2 \cdot S_0\,(1 - \cos(2\pi x)) + S_1 \quad (13)$$

We assume that the vessel is located in the muscle layer with parameter values S0=S1=0.2. Figure 4 shows the sensitivity of the skin temperature as a function of the blood flow speed in the vessel. Figure 5 shows the sensitivity of the skin temperature as a function of the vessel's depth. These results are in qualitative agreement with our expectations. As a matter of fact, the skin temperature varies monotonically as a function of the vessel's depth z as well as its blood flow speed ubl. Let us note that the skin temperature always stays bounded between the core temperature θ =1 and a minimum constant value corresponding to zero blood flow speed. We will denote this minimum value as θ0. Any dimensionless $\theta_{skin}$ obtained from thermal imaging outside the range (0, 1)will be a strong indication that the parameter values of the model or the model itself are not relevant.

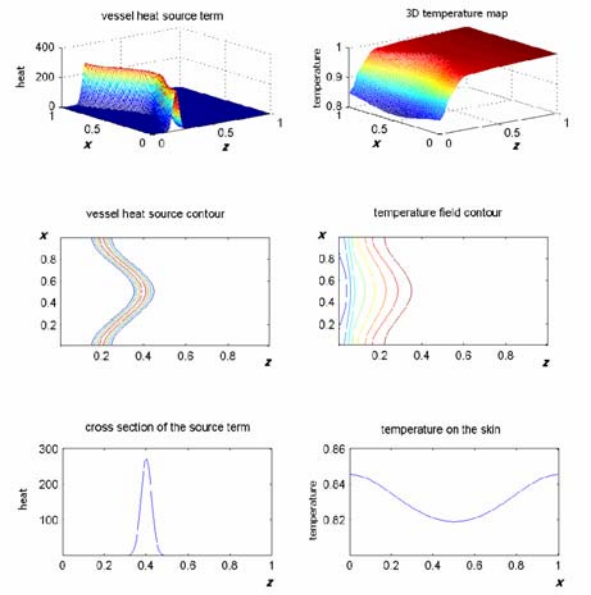


Figure 3: Steady state simulation: The top row shows on the left the vessel heat source $q^{BL}$ and on the right the 3D temperature map. The middle row shows the corresponding contour plots. The bottom row shows on the left a crosssection of the heat source and on the right the spatial variation of skin temperature.

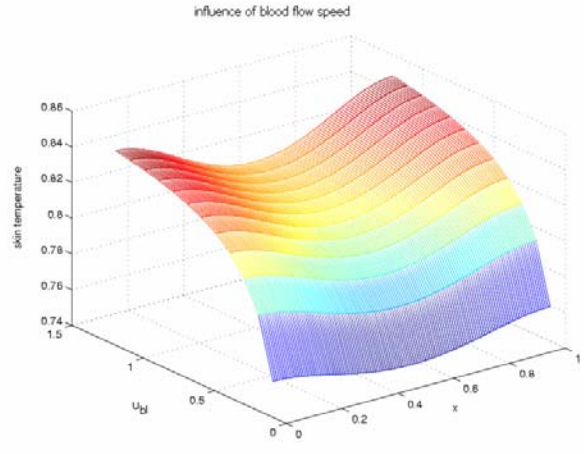


Figure 4: Steady state simulation: Influence of the blood flow speed on the skin temperature.


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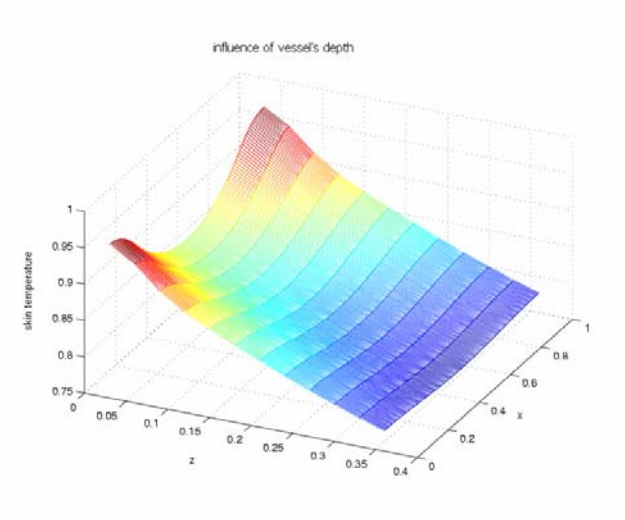


Figure 5: Steady state simulation: Influence of vessel's depth on the skin temperature.

*Direct Simulation in Dynamic State*

We compute the direct simulation in dynamic state with the same discretization in space as in the steady state case and a first order implicit Euler scheme in time:

$$\begin{aligned} \theta^{n+1} - \theta^n - dt\ h_z\ (\Phi^{n+1}_{i+1/2,j} - \Phi^{n+1}_{i-1/2,j}) \quad &- \\ dt\ h_x\ (\Phi^{n+1}_{i,j+1/2} - \Phi^{n+1}_{i,j-1/2}) \quad &= \\ dt\ h_x\ h_z\ (\tilde{\mu}\ u(t)(1-\theta^{n+1}_{i,j}) \exp(-\frac{(z_j - \tilde{(S)}(x_i))^2}{\tilde{\nu}}) \quad &+ \\ \tilde{q}^M(x_i, z_j)), \quad (15) \end{aligned}$$

where $\theta^n_{i,j}$ denotes the temperature at time *ndt*.
In the dynamic direct simulation we are interested to compute the oscillations of the temperature on the skin as a function of the oscillating blood flow speed (cardiac cycle). We consider the following heat source term related to blood flow:

$$u_{bl}(t) = 1. + 0.3 * (\exp(-7 * sin(\pi\omega t)^2) - 0.5).$$

We take ω =1 to simulate a cardiac pulse of 60 beats per minute. In our simulation we start with an initial condition that is the steady solution corresponding to the blood flow speed at t =0.
Figures 7 gives an example of our dynamic direct simulation. We consider here the same parameter values as in the steady state example in Figure 3.

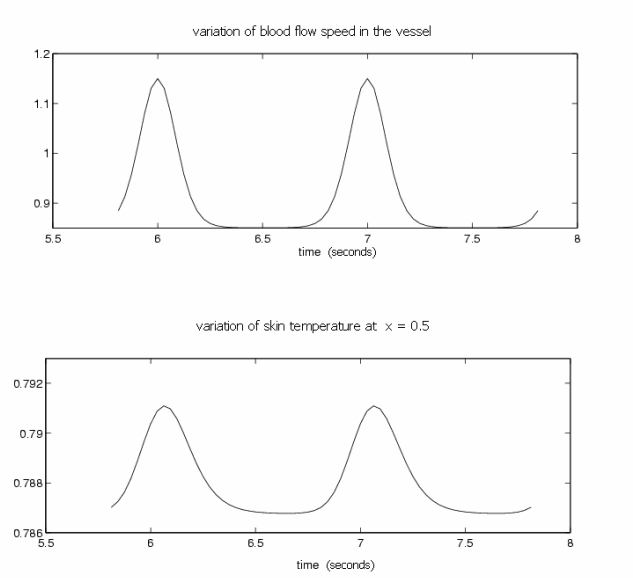


Figure 7: Skin temperature with pulsating blood flow in the vessel. The graph at the top shows the time variation of blood flow speed in the vessel. The graph at the bottom shows the time variation of skin temperature at location $x = 0.5$.

***Inverse Problem***

Initially, we solve the model in the steady state case. The steady state case solution provides the unknown position and shape of the vessel. Then, we use the steady state solution to solve the model in the time domain.

*Inverse Problem in Steady State*

The first step of the inverse problem procedure is to identify the average blood flow speed and the location of the vessel z, from the time average of the time dependent PDE model. Let us denote as $\bar{\theta}$ the average temperature in time. Since the PDE is linear, we get:

$$\begin{aligned} -\frac{\partial}{\partial x}(\tilde{K}(z)\frac{\partial\bar{\theta}}{\partial x}) - \frac{\partial}{\partial z}(\tilde{K}(z)\frac{\partial\bar{\theta}}{\partial z}) \quad &= \\ \tilde{\mu}\bar{u}_{bl}(1-\bar{\theta})\exp(-\frac{(z-\tilde{S}(x))^2}{\tilde{\eta}}) \quad &+ \\ \tilde{q}^M(z),\ (x,z) \in (0,l_x)\times(0,1), \quad (16) \end{aligned}$$

with boundary conditions,

$$\bar{\theta}(x,1) = 1,\ x \in (0,l_x), \quad (17)$$

$$\bar{\theta}(x,0) = \bar{\theta}_s(x),\ x \in (0,l_x), \quad (18)$$

$$\frac{\partial\bar{\theta}}{\partial z}(x,0) = \beta\,\bar{\theta}(x,0),\ x \in (0,l_x), \quad (19)$$

$$\frac{\partial\bar{\theta}}{\partial x}(0/1,z) = 0,\ z \in (0,1). \quad (20)$$


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Based on literature data we assume that we have apriori a reasonable estimate of the average blood flow speed and that the vessel of interest lays in the muscle layer. In order to minimize the dimension of the search space we approximate S(x) with a trigonometric polynomial:

$$S(x) = S_0 + \Sigma_{j=1..n} S_j \cos(j * \pi * x), \qquad (22)$$

under the constraint $S_{min} < S(x) < S_{max}$.
To facilitate the search on S(x) we proceed with an increasing degree n of the trigonometric expansion in (22) with $1 \leq n \leq m$. First, we look at the horizontal line S(x) = C for which we can match on average the observed skin temperature. Then, we look at the next order unknown term in the expansion, and so on.

*Inverse Problem in Dynamic State*

In our application, we are mostly interested in the modulation of blood flow speed in time. The identification process of S(x)with an apriori estimate of average $u_{bl}$ can be seen as a first order correction to compensate for the small temperature variation on the skin along the length of the vessel (see Figure 8).
We can now look for the even smaller temperature variation that is the consequence of the blood flow pulsation in the vessel. Therefore, we use the vessel location derived from the time average skin temperature $\overline{\theta}_{observed}$ and a given frequency ω of the blood flow variation to get the amplitude of the blood flow pulsation.

Assuming that

$$u_{bl} = \bar{u}_{bl} + Pf(\omega t), t \in (0,T),$$

With $f$ a given periodic function of period ω , we address the dynamic inverse problem, that is, to recover the amplitude of the signal P such that the skin temperature for the time periodic solution matches best the $\theta_{observed}(x,t)$.
We have implemented this procedure using for the signal f a simple sinus wave, i.e.,

$$u_{bl}(t) = \bar{u}_{bl} + P \sin(\omega t).$$

The objective function to minimize is:

$$||\hat{\theta}_{\omega} - \hat{\theta}_{observed,\omega}||, \qquad (23)$$

where $\hat{\theta}_{\omega}$ represents the coefficient in the trigonometric polynomial expansion of sin(ωt). The solution of the dynamic inverse problem is straightforward, since we are dealing with the search of a single parameter.

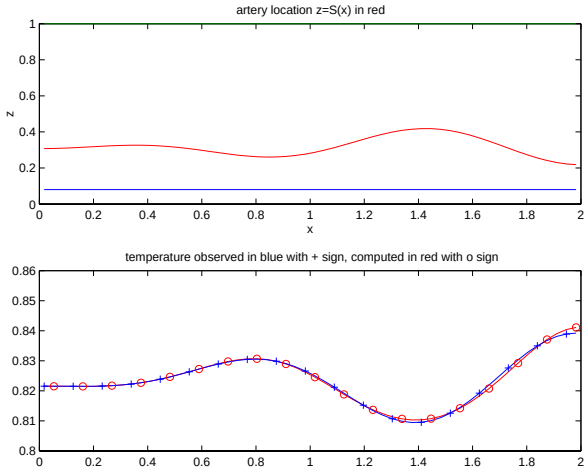


Figure 8: The solution of the inverse problem for the steady state model provides the vessel location (upper figure). In normalized coordinates, the vessel is located between z = 0.2 and z = 0.4, which corresponds to the muscle layer. The lower picture shows the experimental temperature profile over the skin atop of the vessel and compares it to the theoretical profile obtained by the model.

## Conclusions

We have used high-resolution thermal video to compute blood flow and location of large surface vessel. A general bioheat transfer model, aimed to describe the role of large vessels close to the skin, should take into account the geometry and the anatomy of the tissue surrounding the vessel, its thermal properties, and the general energy balance between the vessel, the tissue, and the environment. Moreover, hypotheses about the location and the shape of the vessel should also be considered for a realistic description of the bioheat transfer processes. The model we propose attempts to satisfy the above requirements since it starts from a general energy balance for the tissue surrounding the vessel, like in the Pennes´ approach [8], but modifies the simplified Pennes´ model as follows:

1. It takes into account a modified bell shape for the vessel. This shape formulation allows the model to adapt to arbitrarily complex vascular geometry.
2. It assumes a multilayer structure for the surrounding tissue.
3. It assumes a modulation of the heat power of the vessel through its blood flow speed control.
4. It considers the vessel as a heat source, since its actual temperature is higher than the surrounding tissue.

Method presented in this paper suggests the way for multidimensional passive and remote physiological measurements which can run continuously on a subject. This may have a profound effect on psychophysiology and preventive medicine. However, the present paper is methodological and offers validation only through simulation and limited laboratory testing. Extensive clinical tests are needed to further ascertain the experimental validity of our method.


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