Induced heat transfer from human skin: Its relationship to blood perfusion in the tissue

Tadhg S. O'Donovan, Mohamed A. Atmane & Darina B. Murray
Department of Mechanical & Manufacturing Engineering, Trinity College Dublin, Ireland.

Abstract

The present research investigates the relationship between surface heat transfer from human skin and the physiological features of human tissue. The work is based on the assumption that changes in the tissue characteristics can be thermally sensed at the skin surface. For example, the presence of a malignant tumour in the tissue, known to be at a slightly different temperature than the surrounding tissue, is expected to have an effect on both temperature and heat flux at the skin surface.

A numerical model, based on the bioheat equation proposed by Pennes [1], is developed. The bioheat equation represents energy conservation within the tissue and is solved using finite difference schemes. The configuration adopted consists of a region of human tissue and a surface thermal probe at which is imposed an external convective heat flux. The usefulness and the feasibility of this non-intrusive probe in monitoring the behaviour of the human thermal regulatory system are assessed.

The input parameters of the model are the blood perfusion or volumetric flow rate within the tissue and the convective heat flux imposed at the probe. Results are monitored in the form of time varying heat flux and temperature at the skin surface and in the form of the temperature field within the tissue. It is shown that the time varying heat flux curve at the skin depends on the blood perfusion rate within the tissue. A similarity in the decay law of the heat flux and temperature at the skin is observed also. This introduces a redundancy and has as a consequence the possibility to access one of these quantities independently from the other. The existence of a tumour modifies, depending on its size and position with respect to the skin, the above-mentioned characteristics.
Nomenclature

\[ \rho \] density (kg/m\(^3\))
\[ c \] specific heat (J/kgK)
\[ k \] thermal conductivity (W/mK)
\[ q''_{met} \] metabolic heat generation, (W/m\(^3\))
\[ t \] time, (sec)
\[ T \] temperature, (K)
\[ \omega \] blood perfusion, (ml/ml/s)
\[ H \] depth of tissue, (m)
\[ x \] distance along horizontal axis
\[ y \] distance along vertical axis
\[ p \] iteration number
\[ M \] matrix containing nodal temperature coefficients
\[ A, B, C, D, E, F \] constant coefficients of nodal temperatures

Subscripts

\[ a \] artery
\[ b \] blood
\[ v \] vein
\[ t \] tissue
\[ f \] mesh numbering along x-axis
\[ g \] mesh numbering along y-axis

Introduction

Within the medical profession there is considerable interest in quantifying the blood perfusion rate of an individual using non-invasive methods. Essentially, the well-being of an individual can be inferred from his or her blood perfusion rate. A symptom of ill health is a high body temperature. This is an indirect measure of an individual’s blood perfusion rate since perfusion is increased when sick as a by-product of the immune process. The efficacy of integration of a skin graft could also be assessed by measuring the blood perfusion within the graft and comparing to the rate within normal healthy tissue. The microcirculation of blood is a means of transporting nutrients to cells and removing waste products. If it is possible to measure the rate at which blood passes through tissue the efficacy of this transport system can also be determined. One of the most important applications of blood perfusion measurements, however, is tumour detection. Tumours are known to have a different perfusion rate than normal healthy tissue. In general tumours are highly vascular and so blood flows through them more quickly. Therefore the ability to
detect or measure this abnormal perfusion rate could help evaluate the size and severity of a tumour.

The thermal regulatory system is a means by which the body maintains a relatively constant core temperature. The body is continually dissipating heat from its core under normal environmental conditions. The blood circulation throughout the body, a form of forced convection, controls this. Blood perfusion is defined as the blood volume exchange through a given volume of tissue. Therefore it is expected that the heat dissipation should vary with the blood perfusion rate. By inducing heat flux from the surface of the skin, it is possible to examine the effect that different volume flow rates of blood have on the heat transfer.

There is much on-going research in this area. Techniques used to quantify blood perfusion vary from the injection of a radioactive tracer into the tissue to Laser Doppler Anemometry. These techniques are effective but are either expensive or could cause further trauma to the patient. Thus, it is desirable to develop a non-invasive, low cost method of quantifying the blood perfusion rate of individuals. There have been many attempts made to quantify blood perfusion in the past. Efforts to describe the heat transfer through the tissue began in 1948 with a paper by Pennes [1], which develops an equation describing energy balance within the tissue. This equation is commonly referred to as the Pennes (1948) bioheat equation and includes a blood perfusion rate term.

Much work has continued in the area of blood perfusion and bioheat transfer ever since. A biomedical probe designed in 1991 by Michener et al. [2] induces a negative heat flux at the surface of tissue by means of a water jet. Surface measurements of heat flux and temperature were recorded. The bio-probe was tested on an anaesthetised dog's limb. When a vasodilator was administered to the animal, which would have the effect of increasing the blood perfusion within its limb, a noticeable change in the measured heat flux and temperature decay was found. In 1996, Diller et al. [3] devised a new non-invasive probe for tissue blood perfusion measurements. This probe induced a negative heat flux by way of a pressurised airflow. In a further continuation of this work, Scott et al. [4] constructed an analytical model to calculate the temperature and heat flux at the skin surface under the conditions applied by the probe. Most recently, in 1999, a paper by Liu et al [5] investigated the use of such an analytical model. Their model was constructed using a dual reciprocity boundary element method, (DRBEM). The main focus of the paper was the effect that tumours and other tissue abnormalities have on heat transfer. It was concluded that there was sufficient change in the temperature field to use a similar probe to detect tissue abnormalities such as tumours or skin lesions and to quantify the severity of burn injuries.

The present study uses a finite difference method to model the effect of a thermal surface probe in an effort to define parameters that will eventually lead to the estimation of the blood perfusion rate of individuals.
Description of Model

A finite difference model of the probe/tissue heat transfer problem has been constructed in MatLab. This model replicates the conditions of a blood perfusion test conducted using a thermal surface probe developed for this work. As illustrated in figure 1 the probe applies a cooling air flow to a plate in contact with the skin and is similar in design to that described in Diller et al. [3] although with a different type of sensor for surface heat flux measurements.

Figure 1: Schematic of Blood Perfusion Probe

Heat transfer from the tissue to the probe was modelled in two dimensions, as the system is symmetric. The model was treated as two domains, that of a cross-section of the probe and also of the tissue. Dimensions of the probe were taken directly from the thermal surface probe and the extent of the tissue domain was determined by the requirement to avoid interference with heat transfer from the tissue.

The nodes at the extremities of the tissue domain were assigned a constant temperature boundary condition; that of the initial domain temperature. The boundaries of the domain were set sufficiently distant from the probe to ensure that a constant temperature boundary condition does not interfere with heat transfer from the tissue. The boundary condition between the probe and the tissue is one of conservation of heat flux, in that heat flux from the tissue is set equal to the heat flux entering the probe.

The initial conditions for the probe and tissue domains were of constant temperature, which in this case is that of the body core temperature. This is consistent with the test procedure, which specifies that before the heat flux is applied to the probe, the probe must reach temperature parity with the tissue.

Inputs required by the model include the negative heat flux applied to the probe, which is conducted through to the surface of the tissue, and the perfusion rate of the tissue. The perfusion rate was also set spatially variable in the tissue for the simulations. This offers the possibility of simulating a tumour in the tissue, by setting the blood perfusion in that part to a different value than the surrounding tissue.

The bioheat equation defines the thermal behaviour of tissue. It includes four terms that influence the heat transfer at the tissue surface. These are the heat exchange between the tissue surface and the environment, conduction through the tissue, the energy transfer by blood circulation in the tissue and the heat generation due to local metabolism.
\[ (pc) \frac{\partial T_i}{\partial t} = k_i \nabla^2 T_i + (pc\omega) \left( T_a - T_v \right) + q_{met}^* \]

Heat generation due to the metabolism of the body was neglected from the numerical model because it does not play a major role in the part of the body (close to the skin) where the probe is to be applied. The following characteristic parameters were used to normalise the above bioheat equation:

\[ x^* = \frac{x}{H} \quad T^* = \frac{T_i}{\Delta T} \quad k^* = \frac{k}{k_i} \]

\[ y^* = \frac{y}{H} \quad t^* = t\omega \]

The normalised bioheat equation has the form:

\[ \frac{\partial T_i^*}{\partial t^*} = A_1 \nabla^2 T_i^* - A_2 T_i^* + C \]

where:

\[ A_1 = \frac{k_i}{H^2 \alpha (pc)_i} \quad A_2 = \frac{(pc)_b}{(pc)_i} \quad C = \frac{(pc)_b}{(pc)} \left( \frac{T_a}{\Delta T} \right) \]

The bioheat equation was then further broken down into nodal effects as follows:

\[ A_f T_{f-1,g}^{p+1} + B_g T_{f,g+1}^{p+1} - E_{f-g} T_{f-g}^{p+1} + C_f T_f^{p+1} + D_f T_{f+1}^{p+1} = FT_{f,g}^{p+1} + C \]

where:

\[ A_f, C_f = -\frac{A_1}{\partial x^2} \quad E = 2A_1 \left( \frac{1}{\partial x^2} + \frac{1}{\partial y^2} \right) + A_2 + \frac{1}{\partial t} \]

\[ B_f, D_f = -\frac{A_1}{\partial y^2} \quad F = \frac{1}{\partial t} \]

Similarly, finite difference equations were developed for nodes in the probe domain, using the heat conduction equation. All equations had the form:

\[ [M] [T^{p+1}] = [F] [T^p] + C \]

This overall equation is solved iteratively after each time interval to find the new nodal temperatures of both the tissue and the probe domains, \( T_{f,g}^{p+1} \). After each time interval both the temperature and heat flux at the surface of the skin was recorded. These values were then plotted and compared for different inputs or initial conditions.
Results & Discussion

Simulations were firstly conducted for a constant applied negative heat flux and for three different perfusion rates. The temperature and heat flux at the surface of the tissue was recorded and is graphed below:

Figure 2: Effect of Perfusion on Temperature Decay Curve

Figure 3: Effect of Perfusion on Heat Flux Decay Curve
These simulations were conducted over a period of 120 seconds. The plots of the heat flux and temperature at the surface of the skin indicate that there is a significant dependence on perfusion rate in the rate of decay of both temperature and heat flux. It is thought that this dependence is sufficient to enable the evaluation of the blood perfusion rate during experimental tests. To date the focus of blood perfusion studies has been on the flux decay curve. The results in this study suggest that the temperature decay curve could also be used to estimate the blood perfusion rate. Both sets of curves appear to be exponentially decaying and this would suggest that the curves could be described by a mathematical equation. This equation would include a blood perfusion term and both model and experimental data would be used to evaluate constants for the equation.

In order to validate the model, some experimental data was obtained using an existing thermal surface probe. Figure 2 below compares the model and experimental transient heat flux at the surface of the tissue. Both experimental and model curves show that the heat flux at the surface of the skin rises to a maximum and then decays with time.

The finite difference model did not include the effects of contact resistance at the probe/tissue interface. It is for this reason that the two curves do not agree exactly. According to Michener et al. [2], contact resistance has the effect of delaying the time to peak heat flux and reducing the rate of heat flux decay. The same paper also states that the effects of contact resistance are most pronounced in the early stages of the test when the thermal resistance of the tissue is small. With increasing penetration depth however, the relative significance of the contact resistance decreases. It is therefore expected that there would be better coherence between experimental and model data with the effects of contact resistance included.

![Figure 4: Comparison of model and experimental data](image-url)
The model was then used to simulate the effect that a tumour would have on heat transfer within the tissue. In attempting to develop a non-invasive method of tumour detection, it is the temperature inconsistencies at the surface of the tissue that are of interest. For three different sized tumours at various depths within the domain, the temperature increment that would be detected has been plotted. Figure 4 below shows the results obtained for the case where healthy tissue is modelled with a perfusion rate of 0.0005 ml/ml/s and tumours with a perfusion rate of 0.002 ml/ml/s.

![Figure 5: Tumour Detection & Analysis Curves](image)

The above plots give an indication of the temperature sensitivity of the thermocouples required to detect a tumour in the tissue. This data can also help estimate the size and depth of a tumour once it has been detected. For example, it is quite feasible to detect a temperature difference of 0.15K at the surface of the tissue and this may suggest the presence of a tumour of radius 0.02m at a depth of 0.02m in the tissue.

**Conclusions**

The ultimate objective of this investigation is to develop a method to definitively quantify the blood perfusion rates of individuals. A numerical model is essential to achieve this aim. It is because we can input the perfusion rate within the tissue for the model that it can later be used to approximate constants for a proposed mathematical model. These constants could then be used in calculating the perfusion rate from experimental data.

The results obtained indicate that the perfusion term in the bioheat equation is a controlling factor for the rate of decay of the heat flux. It has also been found that the temperature decay curve is influenced by the perfusion rate. A
A mathematical model that describes these decay curves would therefore include a blood perfusion term. As heat flux sensors are relatively complex and expensive, it would be favourable if the blood perfusion rate could be estimated using the more commonplace instrumentation of the thermocouples.

Results obtained from the simulation of a tumour shows that it affects both heat flux and temperature at the skin, as also found by Liu & Xu [5] in their simulations. The precise effect the tumour has on these surface parameters depends on its size, its position with respect to the skin and its nature (which mainly dictates the blood perfusion rate).

References


