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A Phantom Tissue System for the Calibration of Perfusion Measurements

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Abstract

A convenient method for testing and calibrating surface perfusion sensors has been developed. A phantom tissue model is used to simulate the nondirectional blood flow of tissue perfusion. A computational fluid dynamics (CFD) model was constructed in Fluent® to design the phantom tissue and validate the experimental results. The phantom perfusion system was used with a perfusion sensor based on clearance of thermal energy. A heat flux gage measures the heat flux response of tissue when a thermal event (convective cooling) is applied. The blood perfusion and contact resistance are estimated by a parameter estimation code. From the experimental and analytical results, it was concluded that the probe displayed good measurement repeatability and sensitivity. The experimental perfusion measurements in the tissue were in good agreement with those of the CFD models and demonstrated the value of the phantom tissue system.

Keywords

blood flow; perfusion measurement; calibration system

Introduction

Blood perfusion is the local, nondirectional blood flow per volume of tissue. It represents the local flow through the capillary network and intracellular space of tissue and is responsible for providing the oxygen and nutrients required by cells along with removing waste products. Due to the vital role that blood perfusion plays in maintaining normal physiologic conditions, it is very valuable to be able to measure blood perfusion. Differences in local blood perfusion can be an indicator that the body responds to an abnormal physiologic or pathologic state. For example, blood perfusion can be used to assess the progression of skin graft healing or the

extent of burns. Low blood perfusion can also be an indicator of microcirculation problems in diabetes patients.

Blood perfusion is defined as the exchange of fluid per volume of tissue. It is so convoluted as to be considered nondirectional at the macroscopic level. Conversely, fluid flow rate and velocity are directional and are difficult to directly relate to perfusion. For example, the flow rate or velocity of a fluid through a straight pipe is well defined and constant in one direction. The corresponding perfusion, however, can only be defined for a specified length of pipe. Perfusion appears as if there is no entrance or exit in any direction. This makes the creation of a well defined flow field for perfusion testing and modeling difficult.

Background on Perfusion Measurement Techniques

Because of its importance, several methods have been developed for measuring blood perfusion. Noninvasive imaging techniques include positron emission tomography (PET)[1] and magnetic resonance imaging [2]. Prinzen and Bassingthwaite [3] reviewed microsphere deposition techniques to estimate blood perfusion. These systems have produced reliable perfusion estimates in deep tissues. They cannot be used in a surgical situation or for long term monitoring, however. They tend to be very large, costly, and require substantial processing time.

Laser Doppler systems use the scattering of light by the red blood cells just below the surface of tissue to measure the cell velocity [4]. Although there is no reliable method to convert this measurement into perfusion, laser Doppler flowmetry (LDF) has recently been used for monitoring dialysis [5], assessing the extent of burn wounds [6], and transplant rejection [7]. These systems are dependent on a number of factors other than perfusion, including the optical properties of the tissue, blood, and the optical fibers [8]. Although there is no standard calibration system, calibrations of laser Doppler systems have been attempted using latex microspheres in Brownian motion in water [9].

Other perfusion measurement systems work based on the clearance of a substance such as hydrogen [10,11]. The substance is injected into an artery and as the blood moves farther into the circulation system it diffuses to the surrounding tissue and the amount of tracer in the blood exponentially decays. Based on the arteriovenous concentration difference and the blood flow rate, perfusion can be determined.

Another technique for invasively measuring blood perfusion uses a thermal diffusion probe (TDP). This system works based on the clearance of thermal energy in the tissue [12,13]. A TDP uses a self-heated thermistor to measure absolute perfusion in real time by measuring the power required to keep the probe temperature higher than the tissue temperature [14].

The convective perfusion probe used in the present research works on a principle similar to the TDP system, although it is noninvasive. This probe induces a thermal event (convective cooling) on the surface of tissue and then measures the heat flux response of the tissue [15]. An absolute perfusion value is calculated based on the transient thermal response.

A common challenge with blood perfusion measurement systems is calibration and validation of the measurements. Several of these perfusion measurement techniques have been validated using the microsphere technique [3]. Microspheres are injected into the blood stream and become embedded in the tissue microcirculation. Then, the tissue is harvested and the distribution of the blood flow is determined based on the density of the microspheres trapped in the tissue relative to the density in a blood sample taken at a known flow rate [16]. The microspheres are usually labeled with a radioisotope. Unfortunately, the measurement is not convenient and the tissue is necessarily destroyed.

Experimental results have proven that there is a strong correlation between renal microsphere perfusion estimates and PET estimated perfusion [17]. Monnet et al. [18] showed a correlation between the LDF measurements and a microsphere technique on dog stomach walls. Martin and Bowman [19] were able to validate the thermal diffusion probe based on the microsphere technique.

A more convenient system for testing and calibrating perfusion measurement systems would be advantageous. Real-time verification of perfusion values without physiological variability would be particularly useful. Although Robinson et al.[20] were able to develop a phantom tissue system to test a thermal perfusion probe, the system was not able to verify absolute perfusion values.

The main objective of the current research is to develop a phantom tissue system to mimic the flow and thermal effects of perfusion while providing absolute perfusion values. Because of the difficulty in reproducing the fine blood vessel network of tissue, modeling of the flow is crucial to establish correspondence between the phantom and tissue. Commercial computational fluid dynamics (CFD) software (Fluent) was used to validate the fluid flow in the phantom tissue and the thermal transfer. There are four main criteria to be fulfilled by the phantom tissue model: (1)simulation of nondirectional perfusion, (2) properties similar to blood and tissue, (3) known perfusion values that are easily varied, and (4)repeatability.

Convective Perfusion Probe

The thermal response of tissue depends on the perfusion through the tissue. Therefore, the heat flux response of the tissue due to a thermal event at the surface is a function of the local blood perfusion. This is the basic principle behind both the thermal diffusion probe and the convective perfusion probe. The convective perfusion probe cools the tissue and then measures the thermal response of the tissue. The surface heat flux generates a temperature differential in the tissue, which is warmed by the incoming blood perfusion.

The current blood perfusion probe is a second generation design with a smaller housing and heat flux sensor than Cardinali et al.[15]. The convective perfusion probe is shown in Fig. 1. The probe housing dimensions are 12.7×12.7×10.2 mm³. A Vatell heat flux gage ×10×10×0.25 mm³ is mounted at the bottom of the housing. The thermal event is created by convectively cooling the top surface of the sensor using an array of nine impinging jets. Compressed air is passed through a water bath to maintain it at constant temperature before it flows through a plate with 0.37 mm diameter holes to create the jets. The air then exhausts through outlets in the housing. The shape of the resulting heat flux curve is dependent on the effects of the blood perfusion in the tissue.

The data acquisition system for the voltage signals from the heat flux sensor and an inlet air temperature was a laptop computer with National Instrument's LABVIEW 7.1. These signals were first amplified by an AMP-6 Vatell amplifier and then read with a 16 bit, multifunction National Instrument DAQ-Pad 6015. The other temperatures were recorded by a Doric thermocouple reader.

Finite-Difference Formulation and Parameter Estimation Technique

The predicted heat flux response from the tissue is developed by solving a two-dimensional, finite-difference model of the Pennes [21] bioheat equation

$$(\rho C_p)_t \frac{\partial T_t}{\partial t} = k_t \nabla^2 T_t + (\rho C_p \omega)_b (T_a - T_v) \quad (1)$$

where T is temperature, t is time, ρ is density, C_p is specific heat, k is thermal conductivity, and ω is blood perfusion. The subscripts t , b , v , and a correspond to tissue, blood, venous, and arterial, respectively. The heat generated due to metabolic activity is assumed to be negligible. In the Pennes bioheat equation, the volume of tissue is assumed to be homogenous and is supplied with arterial blood, which is at the body's core temperature. The critical assumption is that the venous blood fully equilibrates with the tissue, $T_v = T_t$. This has been shown over the years to accurately predict the effects of perfusion in many cases. The blood perfusion is also assumed to be uniform and the thermal properties of the tissue and the blood are constant.

A two-dimensional finite-difference solution was used to solve Eq. (1) using a cylindrical coordinate system around the probe. An alternating direction implicit (ADI) method [22] was used to solve the discretized equations for the temperature field. Details of the finite-difference model for the tissue and heat flux gage are given by Robinson et al. [20]. The thermal contact resistance, R_c , between the tissue and gage is defined in the standard manner

$$R_c = \frac{1}{q''} (T_t - T_{\text{gage}})_{\text{interface}} \quad (2)$$

where q'' is the heat flux

The parameter estimation code compares the recorded heat flux data to an estimate predicted by the bioheat transfer model to solve for the value of the blood perfusion term in the Pennes bioheat equation. A flowchart for the parameter estimation routine is shown in Fig. 2. The model is initially solved using guesses for the two unknown parameters, blood perfusion and contact resistance. This numerical solution is then compared to the data set gathered using the sensor. The code determines how closely correlated the estimated solution is to the experimental data via Gaussian minimization [23] and then, if it is not satisfactorily accurate, it changes the parameters and repeats the solution. This continues until the best match is achieved between the bioheat model and the recorded experimental data set.

A typical convergence plot from the parameter estimation routine for one of the test cases is shown in Fig. 3. Because the first few seconds of the heat flux is dominated by the thermal contact resistance and conduction, it is not important to the perfusion estimate. Consequently, it is given a smaller weighting to emphasize the later time data and improve the perfusion estimation process. The match between the experiments and computer model is quite good over most of the region of interest.

Phantom Tissue Test Stand Design

The phantom tissue design was initially modeled with commercial CFD software using FLUENT® 6.2.16. Several configurations and specifications were tested and the final model was developed, as illustrated in Fig. 4. The flow near the top surface where the heat flux gage is placed is shown in Fig. 5. This model has the perfusate (water) entering at the bottom through a flat plate. It was chosen over other configurations because it allows the water to be distributed through the porous matrix and gives multidirectional flow near the surface of the heat flux gage. Several porosities and porous media thicknesses were tested to determine the effects of these parameters on the model. The model showed that the width and length of porous matrix did not have any effect on the heat flux measured during the applied thermal event; however, the thickness did affect the heat flux response. The thickness and the porosity of the sponge used in experiments were measured to be 12.7 mm and 95%, respectively. The porosity of the sponge was determined based on the wet and dry weight of the sponge.

The complete design of the experimental setup can be seen in Fig. 6. The porous media is enclosed in a plexiglass tank filled with water. The system can be broken into two main parts:

the water regulator system and the actual tissue simulator. The water regulator system includes a pump, a flow controller, and a heater. The tissue simulator system includes a skin simulator, a porous matrix, and a structural support system.

The test stand is designed to supply perfusate into the tissue simulator at a controlled temperature and flow rate. The bottom area of the tank stores water maintained at a constant 37°C temperature (core body temperature) by a temperature controller in conjunction with a cartridge heater. The flow into the tissue is provided by a small centrifugal pump that pumps water from the water bath directly through a flowmeter (Dwyer Instruments Inc. Visi-Float® Model VFB-82-BV, 0-30 cm³/min flow range) into the tissue simulator. The test stand also has leveling feet to ensure the system is true.

The tissue simulator is made of seven parts, shown in Fig. 7. The system is mounted to the plexiglass tank by two plexiglass brackets. The inlet plate, which rests on the brackets, is a 127 × 127 mm² piece of plexiglass with a tapped hole in the center for the water inlet. A tube of 6.35 mm diameter and 100 mm length is connected to the tapped hole to allow the flow to develop inside the tube before entering the test section. Also, this piece supports the entire tissue simulator section. Above the center hole of the inlet plate is the 40 × 40 × 12.7 mm³ porous matrix. On top of the porous matrix lies the simulated skin, Teflon. The Teflon sheet of 0.05 mm thickness is kept taut by a fabric hoop. The supports for the fabric hoop are adjustable to ensure that the porous matrix is not compressed. The blood perfusion probe is then attached to the Teflon using a double-sided tape. The water flows through the inlet plate into the porous matrix, and exits through all sides of the porous matrix into the water bath. The use of tape is not necessary, but helps with consistency of the contact. The parameter estimation procedure accounts for differences in thermal contact resistance.

Thermocouples are used at a number of locations to document the temperature distribution throughout the system. Thermo-couples are placed at the top and bottom of the porous media, shown in Fig. 8. The bottom thermocouple is used to measure the arterial temperature. Two more thermocouples measure the temperature of the air entering the housing and the temperature of the top of the Teflon “skin.” The thermocouples are labeled as follows in Fig. 8: (1) airflow, (2) skin, (3) top of porous media, and (4) bottom (inlet) of the porous media. The air temperature measurement is very important for the parameter estimation because it provides the convective boundary condition.

Computational Flow Model

The computational flow model with the boundary conditions marked is shown in Fig. 9. Also shown are the temperature contours at the end of a thermal event (60 s) for a typical flow of 15 cm³/min. The thermal penetration in the tissue below the heat flux gage for this typical flow is approximately 1.25 mm and is independent of the thermal event time for times greater than 45 s.

The Reynolds number in the inlet tube for the maximum flow rate of 30 cm³/min is 100. Therefore, it takes only six diameters to get a fully developed flow in the inlet tube. Hence, a fully developed velocity profile was used in the Fluent model at the inlet to the test section. The outlet flow boundary condition was a constant atmospheric pressure. Hydrostatic and natural convection effects were considered and shown to have negligible effects on the flow and heat transfer.

The usual Navier-Stokes equations were solved with the addition of a momentum sink (pressure drop) for the porous media. This additional term used the Darcy model, with a scalar permeability (α) for the isotropic porous media [24].

$$\nabla P = - \frac{\mu}{\alpha} \mathbf{v} \quad (3)$$

Here, $\mathbf{v}$ is the velocity vector of the flow, μ is the fluid viscosity, and ∇P is the pressure gradient. The value of the permeability $\alpha = 8.5 \times 10^{-10} \text{ m}^2$ was determined from experimental pressure measurements of the viscous resistance of the phantom. In the energy equation, perfusion was initially included as the usual convective flow terms, but was also compared with results using the Pennes model, as shown in Eq. (1). The resulting momentum and energy equations for Fluent were solved using a second-order up-wind scheme and the unsteady formulation was carried out using the second-order implicit scheme.

The heat flux gage is a thermopile that measures the differential voltage between the two faces of the gage. To incorporate this into the model, the heat flux from the computational model is calculated as the conductive heat flux based on the area-weighted average temperature difference between the top and the bottom faces. Both circular and square models of the thermopile geometric pattern were used. When based on the same sensor areas, the results were nearly identical.

Grid and time-step independence tests were carried out before finalizing the computational model. Grid independence tests were carried out for steady-state conditions at $15 \text{ cm}^3/\text{min}$ flow rate, with results shown in Table 1. There is a negligible change in the heat flux for a 22% increase in the number of cells, implying that the model is grid independent. The time step of 0.1 s for both the parameter estimation program and the CFD model was based on the sampling frequency of the DAQ used to acquire heat flux signals, which is 10 Hz. Time-step independence was tested with a time step of 0.05 s. The maximum difference in the resulting heat flux was less than 4% for both CFD and parameter estimation programs, thus demonstrating time-step independence. These results are typical of the results for a range of different operating grids and time steps that were used.

Computational Flow Results

The computational flow model was used to evaluate and optimize a number of operational parameters and assumptions of the phantom tissue system. The typical heat flux response from the Fluent flow model is shown in Fig. 10. As the flow rate increases, the heat flux curves shift upwards and flatten out at longer times. At short times, there is very little influence of perfusion. This is the region of dominance by the thermal contact resistance. The effects of perfusion become more important at longer times.

The CFD results were used to estimate perfusion for the phantom tissue with the same parameter estimation procedure as used for the experimental results (outlined in Fig. 2). This was a convenient tool to investigate a number of questions and issues with using the phantom tissue system. Figure 11 shows the perfusion estimate as a function of thermal event duration for a typical flow of $15 \text{ cm}^3/\text{min}$. There is a little difference for perfusion thermal event durations longer than 60 s. Therefore, the thermal event duration was set to 60 s with a time step of 0.1 s for all the tests carried out experimentally and computationally.

Because the convective perfusion probe and tissue phantom are square, a 3D model was initially used in Fluent. However, to conserve computational time, a 2D axisymmetric model was also created having the same effective heat transfer area. The comparison of the resulting perfusion estimates for the axisymmetric and the 3D model is shown in Fig. 12 over the range of flow rates used. The 2D axisymmetric model results are in very good agreement with the 3D model. Consequently, the 2D axisymmetric model was subsequently used.

The CFD Fluent model solves for the flow and heat flux in the phantom tissue, which is used to determine the corresponding perfusion from the finite-difference parameter estimation routine. This establishes a correlation between the flow rate in the phantom tissue and the corresponding perfusion value, as demonstrated in Fig. 12. The Fluent model can also be modified to solve the Pennes bioheat model as expressed in Eq.(1). To do this, the perfusion term from Eq. (1) is included in Fluent as an energy source term and the convective flow is set to zero. To check the performance of Fluent, the thermal contact resistance and perfusion values from the CFD parameter estimation are input directly into this model of the Pennes equation. The resulting heat flux curve can then be compared with the Fluent CFD model to check the estimate of the perfusion and contact resistance by using the parameter estimation program. A sample of the results is shown in Table 2. For example, at a flow rate in the phantom of 15 cm³/min, the estimated perfusion from the parameter estimation program is 0.0347 (ml/s)/ml. This was used as input to the Pennes Fluent model to obtain a heat flux curve. This heat flux curve gave an effective perfusion value of 0.0333 (ml/s)/ml from the parameter estimation program. These estimates are in good agreement, with a maximum difference of 9% for all of the flow rates tested. Therefore, the Fluent models and the parameter estimation procedure have analytically confirmed the effective correspondence between the flow rate and perfusion for the phantom tissue system.

Sensitivity Analysis of the Parameter Estimates

The sensitivity of the perfusion values on the heat flux curves was quantitatively evaluated using curves such as those shown in Fig. 10 from the CFD model. The heat flux curves for different flow rates are almost identical for the first five seconds. After that, the heat flux curves spread out depending on the flow rates(perfusion values). The effects of both perfusion and contact resistance on the heat flux curves were analyzed. A dimensionless sensitivity coefficient, X_i^+ , was defined as

$$X_i^+ = \frac{\delta q''_i}{\delta \beta_i^+} = \frac{\Delta q''_i}{\Delta \beta_i^+} \quad (4)$$

$$\Delta q''_i = \left| \frac{q''_{Pi} - q''_{Ni}}{q''_{Ni}} \right| \quad (5)$$

where $\nabla q''_i$ is the dimensionless change in heat flux for a perturbed value of the parameters, β . The subscripts Ni and Pi represent the nominal and perturbed values, respectively. The dimensionless difference of the nominal and perturbed perfusion and contact resistance values are represented by $\nabla \beta_i^+$ as calculated by

$$\Delta \beta_i^+ = \left| \frac{\beta_{Pi} - \beta_{Ni}}{\beta_{Ni}} \right| \quad (6)$$

A set of typical results of the sensitivity analysis is shown in Fig. 13 for a flow rate of 15 cm³/min. The sensitivity coefficient for contact resistance starts high and then becomes constant after the first 5 s. The perfusion sensitivity coefficient keeps on increasing with time, making it clear that perfusion is best estimated at longer times.

Experimental Procedure

To ensure repeatability, the same experimental procedure was used for all of the tests. Table 3 summarizes the experimental procedure timeline that was used for all phantom tissue tests. The first step is to equilibrate the temperature within the porous matrix by flushing at a maximum flow rate for 20 s. Then, the flow rate is set and the system is allowed to equilibrate

for the next 40 s. Data acquisition begins after 1 min. The first minute of data acquisition is used to determine the offset in the heat flux signal. Next, the thermal event is started in the second minute of data acquisition and is recorded for 1 min.

Ninety-one total tests were completed on the phantom tissue test stand. Tests were completed at seven different flow rates ranging from 0 cm³/min to 30 cm³/min in increments of 5 cm³/min. At each flow rate, 13 tests were completed to test the repeatability of the phantom tissue test stand. The testing was completed on three different days to test the effects of changing environmental conditions (especially air temperature).

Results

Experimental Heat Flux Results

One of the main design criteria for the phantom tissue test stand is to give repeatable experimental conditions. Figure 14 shows an example of the heat flux curves produced by the convective perfusion probe over three different days of testing at a flow rate of 15 cm³/min. The curves all have the same shape, but the heat flux results from day 1 are shifted slightly lower than the other two days. This difference is due to a higher convective air temperature for the first day and a higher thermal contact resistance, possibly due to probe positioning. The parameter estimation program accounts for these differences when calculating the corresponding perfusion values. Overall, these curves showed very good repeatability. The curves are representative of the repeatability of the results from all of the different flow rates.

Another goal of the phantom tissue test stand is to have easily varied flow rates. To determine if the changes in the flow rates were detectable by the perfusion probe, a comparison of the experimental heat flux at different flow rates is presented in Fig. 15. The shapes of the heat flux curves for the three different flow rates are decidedly different. This correlates to the different perfusion values and shows the sensitivity of the blood perfusion probe to changes in the flow rate.

Results of the Perfusion and Contact Resistance Estimates

The parameter estimation code was used to determine the experimental perfusion in the phantom tissue test stand for all of the tests at different flow rates. The average experimental perfusion values with 95% confidence intervals are shown in Fig. 16, along with the perfusion estimates from the Fluent CFD model. The uncertainty bars shown for the CFD model results are based on a $\pm 5\%$ uncertainty of the program input parameters (density, specific heat and thermal conductivity of both the tissue and heat flux gage, temperatures of the tissue, arterial, initial and air, heat flux, contact resistance, and gage radius). The perfusion estimates for the no flow case in the experimental phantom tests were slightly negative. The offset was removed by subtracting the average perfusion estimated for the no flow case (-0.00652 (ml/s)/ml) from all of the experimental results. The CFD flow model perfusion was 0.00043 (ml/s)/ml for the no flow case. This value is small relative to the perfusion measurements for the other flow rates and hence was assumed to be negligible.

Figure 16 shows that perfusion increases almost linearly with flow rate, as expected. The experimental perfusion estimates are in good agreement with those of the CFD flow model including the uncertainties in both values. For flow rates of 25 cm³/min and 30 cm³/min, the uncertainties in the perfusion estimates are slightly higher. These results clearly indicate the direct relationship between perfusion and flow rate, and provide an experimental and a theoretical validation of the perfusion probe. The larger uncertainties at the higher flow rates are due to the uncertainty in the experimental temperature measurements. From the sensitivity analysis, it was found that the parameter estimation procedure is highly sensitive to these

measurements. The results indicate that the blood perfusion probe can be used to consistently detect changes in perfusion by surface measurements.

The thermal contact resistance estimated for the phantom tests is shown in Fig. 17 with the 95% confidence intervals. It appears that there is no strong correlation between the thermal contact resistance and the flow rate, although the points do exhibit some upward slope for flow rates greater than $10 \text{ cm}^3/\text{min}$. Statistical analysis was used to check the dependence of contact resistance on the perfusate flow rate. The statistical mean for all the flow rates is within 95% confidence intervals except for the flow rate of $10 \text{ cm}^3/\text{min}$. Therefore, it can be concluded that the differences in contact resistances are statistically insignificant and are generally independent of the inlet flow rate.

Conclusions

The phantom tissue system was proved to be a convenient, reliable test bed for surface perfusion measurements in an environment where physiological and pathological conditions are eliminated. The phantom tissue system was shown to be a valid calibration system using a combination of experimental and CFD results with the convective perfusion probe. The convective perfusion probe was shown to be sensitive and repeatable for perfusion measurements. The repeatability, sensitivity, and good comparison with the CFD models demonstrates that perfusion can be reliably measured by the convective perfusion probe. The phantom tissue system is an easy to use method for testing and development of perfusion measurement devices.

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Fig. 1.
Convective perfusion probe(left: 3D model of the housing; right: actual probe with the sensor)

Fig. 2. Model based estimation method used to estimate the unknown parameter, blood perfusion, ω , and contact resistance, R_c

Fig. 3.
Typical heat flux convergence plot from parameter estimation routine

Fig. 4.
The basic concept for a controlled perfusion phantom tissue

Fig. 5.
Flow near the surface of the heat flux gage in the phantom tissue

Fig. 6.
Schematic of phantom tissue test stand and probe

Fig. 7.
Photograph of the experimental test section setup.

Fig. 8.
Schematic arrangement of the thermocouples in the test setup

Fig. 9. The Fluent axisymmetric model of the phantom tissue shown with boundary conditions, streamlines, and temperature distribution

Fig. 10.
Heat flux response from the Fluent flow model

Fig. 11.
Perfusion estimates for different thermal event durations

Fig. 12.
Perfusion estimate comparison for Fluent 2D axisymmetric and 3D flow models

Fig. 13.
Sensitivity analysis for blood perfusion and contact resistance

Fig. 14.
Repeatability of the experimental heat flux for 15 cm³/min flow rate

Fig. 15.
Heat flux sensitivity for the phantom tissue test stand

Fig. 16.
Perfusion estimates of the experimental tests and the Fluent model

Fig. 17.
Contact resistance estimates for the phantom tissue tests

Table 1Grid independence test for 15 cm³/min

No. of cells	Heat flux (W/m ²)
7725	2472.77
9396	2473

Table 2

Comparison of the Fluent CFD flow model and the Pennes model

Phantom flow rate (cm ³ /min)	Phantom perfusion estimate ((ml/s)/ml)	Pennes perfusion estimate ((ml/s)/ml)
5	0.0057	0.0053
10	0.0185	0.0186
15	0.0347	0.0333
20	0.0524	0.0491
25	0.0709	0.0658
30	0.0901	0.0829

Table 3

Summary of the experimental procedure for the phantom test

Step	Procedure	Time (s)
1	Set the flow meter to maximum flow possible	0
2	Set the flow rate to the desired flow rate for experimental run	20
3	Start the data acquisition (heat flux and temperature)	60
4	Turn the air supply on	120
5	Turn the air supply off	180
6	Data acquisition stops. Save the data files.	180