Measurement of *in vivo* Endocardial and Hepatic Convective Heat Transfer Coefficient

by

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MEASUREMENT OF *IN VIVO* ENDOCARDIAL AND HEPATIC

CONVECTIVE HEAT TRANSFER COEFFICIENT

by

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ABSTRACT

The convective heat transfer coefficient ($h$) is a measure of how well heat transfers from a solid surface to a fluid (gas or liquid) by convection. The coefficient is a function of the fluid, the flow field, the geometry of the surface, and its thermal boundary condition. Endocardial convective heat transfer coefficients and hepatic convective heat transfer coefficients are needed for the simulation of heat loss from tissue to blood flow in the cardiac chambers and in the large vessels of the liver. This simulation of heat loss is needed for finite element method (FEM) modeling of radiofrequency cardiac catheter ablation, radiofrequency hepatic probe ablation and hepatic cryoablation in order to study and improve ablation probes and methods. Radiofrequency cardiac catheter ablation has been successfully used for the treatment of many forms of cardiac arrhythmias as a preferable alternative to the conventional “Mesh” operation and drug therapy. Radiofrequency hepatic catheter ablation and hepatic cryoablation have been used to treat both primary and metastatic liver tumors.

I designed and built a measuring circuit modified from a hot wire/film anemometer, which is normally used for velocity measurement. The measuring circuit is composed of a Wheatstone bridge circuit with one of its arms connected to a temperature sensor and the other arms connected to three wire-wound resistors and one potentiometer. The sensor is a Pt thin film temperature sensor overcoated with glass mounted at the tip of a steerable cardiac ablation catheter.

Using a flow rig, a water pump and a water bath, I simulated various flow rates from 0 to 5 L/min at 39 °C and performed in vitro measurements using normal water, distilled water and saline. By placing the catheter sensor axially in a 20 mm solid plastic tube that contained the
flowing liquid and using the measuring circuit to heat up the sensor to approximately 5 °C above the fluid temperature, I measured the power consumed by the sensor and the resistance of the sensor during the experiments and analyzed these data to determine the value of the convective heat transfer coefficient at various flow rates. From 0 to 5 L/min, experimental values of $h$ in W/m²·K were for distilled water 5100 to 13000, for normal water 5500 to 12300, and for saline 5400 to 13600. Theoretical values were 1900 to 10700.

In order to use this catheter sensor for *in vivo* measurement of endocardial and hepatic convective heat transfer coefficient, we had to ensure that little or no blood components deposited on the surface of the sensor. Therefore, we coated a catheter sensor with heparin–carbon coating to minimize the deposition of blood components and performed *in vivo* experiments in 5 pigs with the coated and uncoated sensor. The coated sensor resisted deposition of blood components for longer than 60 min with very little or no blood deposition, while the uncoated sensor resisted it for only 20 min before a significant amount of blood components were deposited on the surface of the sensor. However, the values of $h$ obtained from the coated sensor were noticeably higher than those obtained from the uncoated sensor due to the additional unknown thermal resistance of the coated layer; hence coating with heparin–carbon coating is not practical for *in vivo* $h$ measurement. In order to avoid blood deposition on the surface of the uncoated sensor, we cleaned the sensor frequently using normal water or saline and did not leave the sensor in contact with blood for longer than 20 min.

I measured the endocardial convective heat transfer coefficient, $h$, at 22 locations in the cardiac chambers of 15 pigs *in vivo*. Using fluoroscopy, the steerable catheter sensor tip and sensor orientation in pigs’ cardiac chambers could be located. I measured at least 3 separate
measurements (from different pigs) at each measuring location and calculated the median, average and the standard deviations. For each measurement, I measured at least 30 s and calculated to yield the average value of $h$. The waveforms I obtained from each measurement were very similar to waveforms from pressure measurement. With flows, $h$ varies from 2500 to 9500 $\text{W/m}^2\cdot\text{K}$. With zero flow, $h$ is approximately 2400 $\text{W/m}^2\cdot\text{K}$. These values of $h$ can be used for FEM modeling of radiofrequency cardiac catheter ablation.

I measured in vivo the hepatic convective heat transfer coefficient, $h$, in 8 locations in the large vessels of the liver, the portal veins and the hepatic veins of 8 pigs. The steerable catheter sensor could be located in those large vessels in the pig’s liver using fluoroscopy. The experimental results of $h$ at 4 locations in the hepatic veins vary from 4400 to 7180 $\text{W/m}^2\cdot\text{K}$. The experimental results of $h$ at 4 locations in the portal veins vary from 3000 to 6100 $\text{W/m}^2\cdot\text{K}$. These values of $h$ can be used for the FEM modeling of radiofrequency hepatic ablation and hepatic cryoablation.
CHAPTER 1

INTRODUCTION
I. GOAL

The goal of this research is to measure *in vivo* the value of the endocardial and hepatic convective heat transfer coefficients, $h$, for finite element method (FEM) modeling of radiofrequency cardiac catheter ablation, radiofrequency hepatic ablation, and hepatic cryoablation.

II. BACKGROUND

A. Cardiac arrhythmias and RF ablation

According to the American Heart Association [17], more than 2 million people suffer from some form of cardiac arrhythmia (such as atrial fibrillation and tachycardia) caused by abnormalities in the formation and/or the conduction of the cardiac electrical impulse. Atrial fibrillation (AF) is an irregular heartbeat caused by abnormal electrical impulses firing randomly in the atria. Although it is not fatal, it can lead to stroke, heart failure and increased cardiovascular morbidity and mortality. Ventricular tachycardia (VT) is a condition in which the heart beats too fast at rest and the electrical signals originate in the ventricles instead of the SA node. This type of arrhythmia can lead to sudden cardiac death.

Treatment of AF presents a challenging problem in clinical practice because AF may be triggered from multiple focal sites and sometimes from larger diseased areas in the atrium. Conventional treatments for AF were the “Mesh” surgical operation and drug therapy. However, the operation is highly invasive and very complicated with high cost. Usage of drug therapy causes unacceptable side effects and is a very expensive treatment [1, 8, 13].

Treatments for ventricular tachycardia vary with the symptoms, the situation, and the underlying cardiac disorder. In emergency situations, CPR, electrical defibrillation or
cardioversion (electric shock), or intravenous anti-arrhythmic medications may be required. For long-term treatment, the use of oral anti-arrhythmic medications may be required. The medications can cause severe side effects [18]. Predictable and successful RF ablation is needed in order to reduce the arrhythmia load in patients who have this condition. This in turn would lessen the reliance on anti-arrhythmic drugs and diminish the frequency of treatments delivered by implantable defibrillators.

RF ablation has been used to cure focal cardiac arrhythmia problems. The technique of RF catheter ablation utilizes an RF generator and a catheter-tip-based delivery system. RF power, typically between 5 and 50 W, is delivered for up to 120 s to heat the tissue to a temperature above 50 °C to create thermal lesions [8, 14]. Local resistive “Joule effect” heating occurs, followed by conduction of heat to deeper tissues. Convective heat loss occurs from the endocardium and electrode to blood flow in the cardiac chamber. Simulation of heat loss due to blood flow is needed for FEM modeling of radiofrequency cardiac catheter ablation in order to improve ablation probes and methods. The values of the endocardial convective heat transfer coefficient are essential for the simulation of the heat loss [7, 14, 15].

**B. Primary and metastatic hepatic tumors and RF ablation**

The most common types of hepatic tumors are primary or hepatocellular carcinoma (HCC) and secondary or metastatic hepatic tumors. A primary tumor may result from long-term damage to the liver from viral infections such as hepatitis B and hepatitis C or from alcoholism. A secondary tumor is caused by the spread of the cancer from other areas. These two types of hepatic tumors have been a major worldwide public health problem. The conventional treatment for these tumors is surgical resection. However, in many cases, such as patients with multiple
tumors, surgical resection is not an option. Even in patients who are suited for surgical resection, long-term survival rates are low (only 30 to 45% for 5-year survival). Other treatments, such as systemic chemotherapy, ethanol injection, microwave therapy, laser therapy, cryosurgical ablation and radiofrequency (RF) ablation, are also available. Cryosurgical ablation and RF ablation are the most widely used in the US for treatments of those tumors.

In cryosurgery treatment, subzero temperatures are used to destroy the malignant tissues. The process of cryosurgery treatment normally includes two or three cycles of freeze and thaw with 7 to 30 min for the last freeze cycle. Because the organs nearby the surgical site must be protected from the low temperature, cryosurgery is commonly performed as open surgery. Target tissues are destroyed by the cellular dehydration and denaturation of proteins [5]. Survival rates of this treatment vary from 30 to 50% due to many complications including hemorrhage or bleeding, thrombosis or blood clot within a blood vessel and liver surface cracking.

RF ablation has been used for hepatic tumor treatment. However, a major limitation for this method is the small lesion size (the largest lesions created are about 4 to 4.5 cm in diameter. The treatment usually lasts between 10 to 35 min depending on the lesion size. To destroy the target tissue, RF ablation uses high-frequency current (normally around 500 kHz) with voltages around 100 V and applied power up to 150 W to heat up the target tissue. This method can be minimally invasive since the RF ablation probe may sometimes be introduced percutaneously [4, 10].

C. Bioheat equation and finite element method model

During RF ablation (both cardiac and hepatic RF ablation), the electrical-thermal behavior of tissue can be described by the bio-heat equation:
\[
\rho c \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + J \cdot E - Q_p + Q_m
\]  

(1)

where: \( \rho \) is the density (kg/m³)

\( c \) is the specific heat capacity (J/kg·K)

\( T \) is the temperature (K)

\( t \) is time (s)

\( k \) is the thermal conductivity (W/m·K)

\( J \) is the current density (A/m²)

\( E \) is the electric field intensity (V/m)

\( Q_p \) is the heat loss due to blood perfusion within myocardium or in hepatic tissue (W/m³)

\( Q_m \) is the rate of metabolic heat production (W/m³) (it is small compared to the other terms so it can be neglected)

The bioheat equation states that the heat gained per second by a unit volume \((\rho c \partial T / \partial t)\) equals the heat conducted into this unit volume due to the temperature gradient \((\nabla \cdot k \nabla T)\) plus the Joule heat generated by the current in this unit volume \((J \cdot E)\) minus the heat loss due to blood perfusion \((Q_p)\) plus metabolic heat \((Q_m)\). For RF cardiac catheter ablation, the heat loss due to blood perfusion \((Q_p)\) is small and it can be neglected.

The bioheat equation can be solved analytically, however, it is very complicated due to the complicated configuration. Therefore, finite-element method (FEM) modeling has been used to simulate the process. The FEM divides the entire volume into a large number of volumes and solves the equation for each volume. This way, various lesion configurations can be modeled.
D. Heat loss due to blood perfusion within tissue ($Q_p$)

From the bioheat equation (1), a small part of the energy is lost to blood perfusion. Blood perfusion is the flow of blood in small vessels (capillary size) inside the tissues. Heat loss due to perfusion ($Q_p$) occurs due to the heat carried by the blood in those small vessels. This can be found by:

$$Q_p = h_{bl}(T - T_{bl})$$  \hspace{1cm} (2)

$$h_{bl} = \rho_{bl} c_{bl} \varpi_{bl}$$  \hspace{1cm} (3)

where:

- $h_{bl}$ is the convective heat transfer coefficient accounting for the blood perfusion in the tissue (W/m$^3$·K)
- $T_{bl}$ is the temperature of the blood (assumed to be constant at 37 °C)
- $\rho_{bl}$ is the blood density (kg/m$^3$)
- $c_{bl}$ is the specific heat of the blood (J/kg·K)
- $\varpi_{bl}$ is the blood perfusion (s$^{-1}$)

In cardiac RF ablation, this $Q_p$ (heat loss due to blood perfusion in the myocardium wall) is very small and can be neglected since the ablation time is short (up to 2 min.) [13]. However, in radiofrequency hepatic ablation, this $Q_p$ cannot be neglected because the liver is highly perfused and the ablation takes a longer time (from 12 to 30 min.).
E. Effect of convective heat transfer coefficient \( (h) \) on FEM models

During cardiac RF ablation, the ablation catheter is in contact with the myocardium, and they both are in contact with the blood in the chamber. This flowing blood causes heat fluxes that cause a cooling effect, which depends upon the location in the cardiac chamber. A boundary condition for the bioheat equation (1) is required for calculation of these heat fluxes at the interfaces of the blood–catheter and the blood–tissue and can be calculated by:

\[
k \frac{\partial T}{\partial n} = h(T - T_{bl})
\]

(4)

where: \( h \) is the convective heat transfer coefficient due to the blood flow in the cardiac chambers or in the large vessels in the liver (W/m\(^2\)·K)

\( n \) is unit vector normal to the myocardial or the hepatic tissue or the electrode surface

In order to find this convective heat transfer coefficient due to the blood flow in the cardiac chambers or in the large vessels in the liver, we perform experiments to measure this value. A sensor, which is placed inside the cardiac chambers or the large vessels in the liver, is heated about 5 °C above the blood temperature, and then the resistance is measured. Then \( h \) can be calculated by:

\[
Q_h = hA(T_s - T_b)
\]

(5)

where: \( Q_h \) is the power consumed by heating the sensor (W)

\( h \) is the convective heat transfer coefficient (W/m\(^2\)·K)
\( A \) is the sensor area (m\(^2\))

\( T_s \) is the temperature of the heated sensor (K)

and \( T_b \) is the temperature of the blood that flows adjacent to the site (K)

In order to simulate the RF ablation process effectively using the FEM, an accurate value of \( h \) is needed. Currently, the values of \( h \) that have been used mostly come from theoretical calculations using the boundary conditions. The myocardial convective heat transfer coefficient inside the cardiac chamber (from the theoretical calculations) that has been used for the FEM ranges from 500 to 2000 W/m\(^2\)·K [13, 14].

**F. Hot wire/film anemometer**

I studied the hot wire/film anemometer since it uses the same heated sensor and electric circuit that I used to measure \( h \). The hot wire/film anemometer has normally been used for measurement of the velocity of fluid by detecting the changes of the heat transfer from a small heated temperature sensor that is exposed to the flowing fluid. A resistive thermal detector (RTD) is extensively used for this application. This anemometer was used for studying flow details, such as turbulent and pulsatile blood flow *in vivo* [9, 11, 12, 16]. The sensor for these applications must be small and have good frequency response.

There are two modes for this method, constant-power mode, and constant-temperature mode. For a constant-power mode, the circuit provides the sensor constant electric power by selecting a large resistance of \( R_1 \) compared to that of \( R_s \). Therefore, \( I \) is virtually constant even when the sensor gets cooler as the heat transfer rate increases (i.e. the velocity of the fluid increases). Increased velocity increases the convective heat transfer, which cools \( R_s \), which
decreases its resistance, which decreases $V_1$, which increases the output $V_o$. Thus, if the convective heat-transfer decreases, $V_o$ decreases.

For the constant-temperature mode (as shown in Fig. 1), feedback from $V_o$ is needed to maintain the resistance of the sensor constant, which keeps the temperature constant. How this system operates is given below:

- First, when the fluid flows past the hot wire, it cools the hot wire. Decreasing temperature decreases the resistance ($R_s$) of the hot wire, hence decreases the voltage at point 1 ($V_1$).
- The decreased voltage at point 1 makes the input voltage $V_{12}$ become more negative, which increases the output voltage of the amplifier ($V_o$).
- The larger $V_o$ increases the current that flows through the sensor. This current heats up the sensor, resulting in an increase of the temperature and the resistance of the sensor ($R_s$).
- The higher $R_s$ increases the voltage at point 1 which then makes $V_{12}$ become more positive.

This system continues to adjust the value of $R_s$, $V_1$, $V_{12}$, and $V_o$ until equilibrium is reached.

**G. Convective heat-transfer**

There are two mechanisms in convective heat transfer, random molecular motion (diffusion) and the bulk or macroscopic motion of the fluid. This fluid motion is related to the large numbers of molecules moving in the presence of a temperature gradient. The total heat transfer by convection is then caused by a superposition of energy transport by the random motion of the molecules and by the bulk motion of the fluid.

Convective heat transfer occurs between two different temperatures of a fluid in motion and a boundary surface. Figure 2 shows that fluid flows over the heated surface. As a result of the fluid–surface interaction, there is a region in the fluid in which the velocity varies from zero
(at the surface) to a finite value $u_\infty$ related to the flow (this region is known as the hydrodynamic or velocity, boundary layer). Additionally, because of the temperature difference between the surface and the flow, this fluid region will also express a thermal boundary layer, through which the temperature varies from $T_s$ at $y = 0$ to $T_\infty$ in the outer flow. This thermal boundary layer can be smaller, larger, or the same size as that through which the velocity varies. Convective heat transfer will occur between the surface and the outer flow when $T_s > T_\infty$.

According to the nature of the flow, convective heat transfer may be classified into two categories, forced convection and free or natural convection. Forced convection occurs when external means (such as a pump, a fan, or winds) cause the flow. On the other hand, free or natural convection occurs when the flow is induced by buoyancy forces (which occur from the density differences caused by temperature variation in the fluid. Table 1 lists some typical values of the forced and free convective heat transfer coefficients [6].

The appropriate equation for convective heat transfer is:

$$q'' = h(T_s - T_\infty)$$

where $q''$ is the convective heat flux (W/m$^2$)

$T_s$ is the surface temperature (°C)

$T_\infty$ is the fluid temperature (°C)

$h$ is the convective heat transfer coefficient (W/m$^2$·K)
III. SUMMARY

Chapter 2 presents the detailed information of the measuring system, the modified catheter sensor and the calculation correction for the $h$ measurements. I describe the structure of the measuring system, the configuration of the catheter sensor and the correction equation for the $h$ measurement. This chapter also presents the experimental results from \textit{in vitro} measurements in normal water, distilled water and saline, along with the theoretical calculations. I used a flow rig and a water bath to simulate the blood flow in the cardiac chamber by setting the temperature of the fluid at $39 \, ^\circ \text{C}$ and varying the flow rate of the fluid from 0 to 5 L/min. The goal was to prepare the measuring system and the catheter sensor for \textit{in vivo} measurements.

Chapter 3 presents one potential solution for the problems caused by the deposition of blood components on the surface of the catheter sensor during the \textit{in vivo} $h$ measurement. By coating the surface of the sensor with heparin–carbon coating, the coated sensor could resist the deposition of blood components much longer than without the coating. I describe the mixture of the heparin–carbon coating, the coating procedure and the experiments I did to test the thrombus resistance properties of the coated sensor. I conducted \textit{in vivo} experiments in 4 pigs using both the coated and the uncoated sensors and compared the results obtained by both catheter sensors. I conclude that coated sensors are not practical because the coated layer added an additional unknown thermal resistance into the system. This unknown thermal resistance would cause error in the calculation to yield the value of $h$.

Chapter 4 presents the experimental results of the endocardial convective heat transfer coefficient from \textit{in vivo} measurements in 22 locations in the cardiac chambers of 15 pigs. I used the measuring system and the catheter sensor described in chapter 2 for these \textit{in vivo} measurements by advancing the catheter sensor into the cardiac chambers of the pigs and
collected the data with the data acquisition programs, Biobench and Labjack. I calculated the median, average and the standard deviations of the results. This chapter also presents the waveforms of the voltage across the sensor obtained from 8 locations in the cardiac chambers.

Chapter 5 presents the experimental results of the hepatic convective heat transfer coefficient from *in vivo* measurements in 8 locations in the large vessels of the liver of 8 pigs. I used the measuring system and the catheter sensor described in chapter 2 for these measurements by advancing the catheter sensor into the portal veins and the hepatic veins of the test animals and collected the data with the data acquisition programs, Biobench and Labjack. I also calculated the median, average and the standard deviations of the results.

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**Figure and Table Captions:**

**Figure 1.** A hot-wire anemometer keeps the sensor resistance and temperature constant.

**Figure 2.** Development of boundary layer in convective heat transfer

**Table 1.** Typical values of the convective heat transfer coefficient [6]
Table 1. Typical values of the convective heat transfer coefficient [6]

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<td>Boiling or condensation</td>
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Figure 1. A hot-wire anemometer keeps the sensor resistance and temperature constant.
Figure 2. Development of boundary layer in convective heat transfer
CHAPTER 2

MEASURING ENDOCARDIAL CONVECTIVE HEAT TRANSFER COEFFICIENT:
MEASURING SYSTEM AND IN VITRO EXPERIMENTS

This chapter has been submitted for publication as:

ABSTRACT — We built an instrument to measure the endocardial convective heat transfer coefficient, $h$, using a Wheatstone-bridge circuit, similar to a hot-wire anemometer circuit. One arm is connected to a steerable catheter sensor whose tip is a 1.9 mm $\times$ 3.2 mm thin film resistive temperature detector (RTD) sensor. We used a circulation system to simulate different flow rates at 39 °C for in vitro experiments using distilled water, normal water and saline. We heated the sensor approximately 5 °C above the fluid temperature to prevent blood coagulation and cell damage. We measured the power consumed by the sensor and the resistance of the sensor during the experiments and analyzed these data to determine the value of the convective heat transfer coefficient at various flow rates. From 0 to 5 L/min, experimental values of $h$ in W/(m²·K) were for distilled water 5100 to 13000, for normal water 5500 to 12300, and for saline 5400 to 13600. Theoretical values were 1900 to 10700.

Index terms — convective heat transfer coefficient, RF ablation, cardiac ablation, hepatic ablation, heat convection, hot-film anemometer

I. INTRODUCTION

Radiofrequency (RF) catheter ablation has been used to treat many types of cardiac arrhythmias such as atrial tachyarrhythmias (atrial tachycardia, atrial flutter, atrial fibrillation), atrioventricular nodal re-entrant tachycardia, Wolf-Parkinson-White syndrome, symptomatic supraventricular and ventricular tachycardia (fascicular VT, bundle branch re-entrant VT, idiopathic VT, ischemic VT) [3, 10, 14, 15, 20] with high success rate because of its controllability, high efficacy, low complications and minimal invasiveness. The technique of radiofrequency catheter ablation is to deliver high-frequency alternating electric current from
350 kHz to 1 MHz through the electrode catheter to generate a thermal lesion in myocardial tissue [9]. The tissue in direct contact with the catheter is heated by resistive (ohmic) heating. Resistive heating in tissue is proportional to the power density [9]. The thermal energy from the directly contacted tissue is then transferred to its vicinity by means of conduction and forms an RF lesion. Flowing blood near the tissue transfers heat away by convection and is a major cause of heat loss from the RF lesion. Convective heat transferred by the epicardial coronary artery is considered a minor heat loss in RF catheter ablation [9, 22].

In order to improve the electrodes and success rate of RF catheter ablation, we perform finite element method (FEM) simulations of the ablation [2, 13, 22]. The value of endocardial convective heat transfer coefficient \( h \) is essential for the simulation of heat loss from the endocardium to the blood pool in the cardiac chambers [13, 22]. Researchers have used values of \( h \) ranging from 44 to 6090 W/m\(^2\)-K for various locations in the cardiac chambers [1, 13, 21, 22], however, none of those values came from in vivo measurement in animals. Because we did not find in vivo measurements of \( h \) in animals, we built our instrument, which is a Wheatstone-bridge circuit connected to a thin film resistive temperature detector (RTD) sensor, similar to a hot-wire anemometer circuit, and used it to measure the endocardial \( h \) first in vitro and then in vivo.

Hot wire/film anemometer systems have been used to measure the blood velocity in vivo. Nerem et al. [16] used a hot film probe as described by Seed to study the velocity distribution in the aortas of dogs. Paulsen and Nissen [19] developed a safety system for a hot-film anemometer for blood-velocity measurement in humans. Yamaguchi et al. [24] performed turbulence measurements in the center of the canine ascending aorta using a hot-film anemometer. Paulsen
et al. [18] analyzed the dynamic properties of a hot-film anemometer system for blood velocity measurements in humans.

II. METHOD AND MATERIALS

In order to obtain the value of $h$ inside the cardiac chambers, we performed *in vitro* experiments to ensure the capability of our measuring system and to calibrate the system. Then, we measured the value of $h$ *in vivo*. Fig. 1 shows a block diagram of the *in vitro* experiment setting, which consisted of a circulation system (shown in Fig. 2), the sensor (shown in Fig. 4 and Fig. 5), the measuring circuit (shown in Fig. 3) and the data acquisition program. The circulation system simulated flow rates of blood from 0 to 5 L/min to provide different flow rates for distilled water, normal water and saline. The catheter sensor, whose tip is a Pt thin film resistive temperature detector (RTD) sensor, was placed in the fluid flow, and formed one arm of the Wheatstone bridge circuit (the measuring circuit). The electric power consumed by the sensor and the temperature of the sensor were measured and saved using a data acquisition program, BioBench, and were then analyzed to yield $h$ (using equation (1) which is in section B, Measuring circuit).

A. Circulation system

Fig. 2 shows that the circulation system was composed of a fluid bath, a centrifugal pump, a flow meter and a fluid container. The fluid bath maintained the temperature of the fluid as measured by the sensor at the sensor location at 39 °C (swine body temperature). From the fluid bath, the pump pumped the fluid to the flow meter. A valve at the flow meter was adjusted to vary the flow rate from 0 to 5 L/min. The fluid then flowed to the container through a 20 mm diameter, 500 mm long, solid PVC tube with the heated sensor on the axis in the tube. Since the flow in the
tube would follow a parabolic pattern (laminar flow), the sensitivity of the sensor would be maximal at the site near the axis of the tube, and decrease slowly when the sensor moves off axis.

The tube was more than 20 times longer than the diameter of the tube, which suggests that the flow was fully developed. The flow of the fluid at the measuring point was expected to be laminar. As the fluid flowed past the sensor, it dissipated heat from the sensor and cooled the sensor by convection. The higher the flow rate, the faster the heat was dissipated.

**B. Measuring circuit**

Fig. 3 shows our measuring system (similar to a hot wire/film anemometer system) using a resistive temperature detector (RTD), \( R_s \), forming one arm of a Wheatstone bridge. This circuit maintained the resistance of the sensor at a constant value, hence holding the temperature of the sensor constant (constant temperature mode). During the measurement, the sensor was heated approximately 5 °C above the temperature of the flowing fluid (distilled water, normal water or saline). The electric power consumed by the sensor and the resistance of the sensor during the experiment were collected and analyzed to yield the value of \( h \).

The Wheatstone bridge had a resistive temperature detector (RTD) sensor in one arm of the bridge, and three wire-wound resistors (1 W), and a precision potentiometer in the other arms. A power op amp supplied the power to the circuit and the sensor. The resistance of the sensor was maintained constant by the bridge at

\[
R_s = R_R \frac{(G - 2)}{(G + 2)}
\]

where \( R_R \) is the total resistance of the upper right arm of the bridge, and \( G \) is the gain of the op-amp [5]. The sensor was electrically heated about 5 °C above the fluid temperature because higher temperatures may
cause blood coagulation on the surface of the sensor and may damage the surrounding cells [9]. Dissipation of the heat occurred because the flowing fluid carried the warmed fluid away from the surface of the heated sensor. Increased velocity of the fluid reduced the fluid temperature next to the sensor and therefore the ohmic resistance of the sensor. Restoration of the sensor to its original working temperature was achieved by feedback controlled by the op amp. The sensor current increased to increase the power and the temperature of the sensor.

The voltage across the sensor ($V_1$) and the voltage across $R_1$ and $R_s$ ($V_A$) were recorded by the BioBench program through the analog-to-digital converter (ADC) (12 bit, 100 kS/s, 8 analog inputs). $V_A$ and $V_1$ were used to determine the current that flowed through the sensor. $V_1$ was also used to determine the resistance and the temperature of the sensor. Once we know the value of the current, the resistance and the temperature of the sensor, we calculate $h$ using:

$$Q_h = hA(T_s - T_b)$$

(1)

where: $Q_h$ is the electric power consumed by heating the sensor (W)

$h$ is the convective heat transfer coefficient (W/m$^2$-K)

$A$ is the sensor area (m$^2$)

$T_s$ is the temperature of the heated sensor (K)

and $T_b$ is the temperature of the blood that flows adjacent to the site (K)

However, since the sensor we use is not a bare Pt thin film, we must calculate a correction.
C. Catheter sensor design

To measure in vivo $h$, we require a catheter that can be externally steered and placed against chamber and vessel walls. The sensor (model TFD, from Omega Company) is at the tip of a cardiac ablation catheter and Loctite seals the electric connection, covers the backside of the probe and rounds the sharp edges of the sensor. Dr. Dorin Panescu of Boston Scientific Corporation supplied the probe. Fig. 4 shows the structure of the thin film Pt sensor. Fig. 5 shows the structure of the catheter with the sensor at the tip.

D. Temperature vs. resistance for the sensor

When electrically connecting the bare sensor to a catheter, the overall resistance of the catheter sensor is greater than that of the bare sensor because of the added resistance of the lead wire. We used an adjustable temperature water bath and a digital multimeter (model# HP34401A) to plot the resistance versus the temperature shown in Fig. 6. The average resistance difference between the bare sensor and the sensor with the lead wire is about 8.51 $\Omega$. The resistance differences of the lead wire in air (20 °C) and in water (37 °C) was less than 0.1 $\Omega$. Using this information, we use the DIN EN 60751 [25], resistance vs. temperature table. The resistance of the catheter sensor is $R_c = R_s + 8.51 \, \Omega$, where $R_s$ is the resistance of the bare sensor.

III. RESULT AND ANALYSIS

A. Theoretical calculation of forced convection at the center of a 20 mm diameter tube

We calculate the value of forced $h$ for laminar flow:

$$\text{Nu} = \frac{hL}{k}$$

(2)
where: $\text{Nu}$ is Nusselt number

$k$ is thermal conductivity (W/m·K)

$L$ is the length of the probe (parallel to the direction of the flow) (m)

$h$ is the convective heat transfer coefficient (W/m²·K)

The Nusselt number ($\text{Nu}$) can be estimated from the following equation, valid for a sharp edged plate heated from $x = 0$ to $x = L$:

$$\text{Nu} = 0.664(\text{Re}^{1/2})(\text{Pr}^{1/3}) \quad (3)$$

$$\text{Re} = \frac{\rho u L}{\mu} \quad (4)$$

$$\text{Pr} = \frac{C_T \mu}{k} \quad (5)$$

where: $\text{Re}$ is Reynolds number

$\text{Pr}$ is Prandtl number

$\rho$ is the density of fluid (kg/m³)

$u$ is the velocity of the flow at the sensor location (m/s)

$\mu$ is dynamic viscosity (N·s/m²)

$C_T$ is specific heat at constant temperature (J/kg·K)

$F$ = Flow

$A_t$ is the cross-sectional area of the 20 mm diameter tube = $\pi(0.01)^2$

$$= 3.1416 \times 10^{-4} \text{ m}^2$$

$$u = \frac{2F(\text{m}^3/\text{s})}{A_t(\text{m}^2)} = \frac{2F(\text{L/min})}{60(1000)(A_t)} \quad \text{(m/s) (on the axis)}$$
At the axis of the tube, the velocity of the fluid is two times the average velocity in a laminar flow pattern. However, if the flow is not laminar, the velocity of the fluid at the axis of the tube would be different and it would require a different method for calculation of the velocity of the fluid at the axis.

B. Theoretical calculation of free convection for a vertical plane

When placing a heated sensor in a still liquid, heat generated by the sensor is dissipated by free or natural heat convection. Free or natural convection is caused by the fluid movement resulting from the change of the fluid density due to the heating process [11].

Because of the heating process, the density of the fluid adjacent to the heat transfer surface is decreased, resulting in buoyancy forces that cause the movement of the fluid in free convection. Some external force fields, such as gravity or centrifugal force, must act upon the fluid so that the buoyancy forces and free convection current are present if one or more heated surfaces are in contact with the fluid. The free convection currents, which arise from the buoyancy forces, are called body forces.

We calculate the average value of free $h$:

$$\overline{Nu_f} = C(Gr_f Pr_f)^m = \frac{hx}{k_f}$$

(6)

where the subscript $f$ indicates that the properties in the dimensionless groups are evaluated at the film temperature ($T_f$), where:

$$T_f = \frac{T_e - T_w}{2}$$

(7)

where
$T_{\infty}$ is the fluid temperature (°C)

$T_w$ is the wall temperature (of the heated plate) (°C)

and

$$Gr_f Pr_f = \left( \frac{g \beta \rho^2 C_p}{\mu k} \right) x^3 \Delta T$$

(8)

where: $Gr_f$ is the Grashof number at the film temperature.

$Pr_f$ is the Prandtl number at the film temperature.

$g$ is acceleration of gravity (m/s²).

$\beta$ is the volume coefficient of expansion (1/K).

$\rho$ is density (kg/m³).

$C_p$ is specific heat at constant pressure (J/kg·K).

$\mu$ is dynamic viscosity (N·s/m²).

$k$ is thermal conductivity (W/m·K).

$\Delta T$ is temperature difference of the fluid and the wall ($T_{\infty} - T_w$).

$C$ is a constant, and can be evaluated by the value of $Gr_f Pr_f$.

$m$ is a constant, and can also be evaluated by the value of $Gr_f Pr_f$.

$h$ is the local free convective heat transfer coefficient (W/m²·K).

For free convection from isothermal vertical planes, the value of local $h$ and $Nu_f$ can be analyzed according to Table 2.

The average convective heat transfer coefficient ($\bar{h}$) can be evaluated by:
\[
\bar{h} = \frac{1}{L} \int_0^L h_x \, dx = \frac{5}{4} h_{x=L}
\]  

(9)

**Example:** Calculation of free convection in still normal water.

In a water bath with constant temperature of 39 °C, a sensor is placed vertically. The sensor is heated to a constant 5 °C above the water temperature. Calculate the free \( h \).

From the information given, the Grashof number and the Prandtl number are:

\[ Gr_f Pr_f = 1200 \]

\[ Gr_f = 1.3 \times 10^3 \]

\[ Pr_f = 4.42 \]

From Fig. 7-7 of [11], \( Nu_f = 4.47 \)

\[ h_{local} = 1500 \text{ W/m}^2\cdot\text{K} \]

and \( \bar{h} = 1870 \text{ W/m}^2\cdot\text{K} \)

Where \( h_{local} \) is the value of \( h \) at \( x = L \) and \( \bar{h} \) is the average value of \( h \) for over the exposed area of the sensor.

Table 3 shows the value of \( u, Re, Nu \) and \( \bar{h} \). Note that at zero flow rate, \( \bar{h} \) is calculated using free convection equations.

**C. Calculation correction**

We calculated a correction for the sensor to obtain a more accurate value of \( h \) since a glass layer covers the top of the Pt thin film. A thermal circuit analogous to an electric circuit was used for
this correction. We treat the heat transfer-rate \((q)\) as a flow. We calculated the thermal resistances from the thermal conductivity, convective heat transfer coefficient, and thickness of the material and the area of the material. The temperature difference is analogous to the potential difference. The Fourier equation [6] may be written as:

\[
\text{Heat flow} = \frac{\text{thermal potential difference}}{\text{thermal resistance}} \quad \text{or} \quad q = \frac{\Delta T_{\text{overall}}}{\Sigma R_{\text{th}}} \quad (10)
\]

When the sensor is heated, the heat from the Pt thin film conducts through this thin layer of glass, which is on the top of the Pt film. The equivalent thermal resistance of the glass layer is \(R_G\). The equivalent thermal resistance of the ceramic layer is \(R_C\) and the equivalent thermal resistance of the Loctite layer is \(R_L\). Convection occurs on the surface of this layer, and is included in this thermal circuit as \(R_{\text{conv}}\). Fig. 7 shows the one-dimensional structure of the catheter sensor and the equivalent circuit of this system. The values of \(R_G\) and \(R_{\text{conv}}\) can be calculated using:

\[
R_G = \frac{d_G}{k_G A} \quad (11)
\]

\[
R_C = \frac{d_C}{k_C A} \quad (12)
\]

\[
R_L = \frac{d_L}{k_L A} \quad (13)
\]

and

\[
R_{\text{conv}} = \frac{1}{h_{\text{conv}} A} \quad (14)
\]

where: \(d_G\) is the thickness of the glass layer = 0.027 mm.

\(k_G\) is the thermal conductivity of the glass = 1.38 W/m·K
$d_C$ is the thickness of the ceramic substrate = 0.45 mm.

$k_C$ is the thermal conductivity of the ceramic = 6.06 W/m·K

$d_L$ is the thickness of the Loctite = 1.00 mm.

$k_L$ is the thermal conductivity of the Loctite = 0.55 W/m·K

$A$ is the surface area of the glass layer = the surface area of the ceramic substrate = 3.2 mm × 1.9 mm = 6.08 × 10^{-6} \cdot \text{m}^2

$T_A$ is the temperature at point A

$T_B$ is the temperature at point B

$T_C$ is the temperature at point C

$T_D$ is the temperature at point D

Heat from the layer of thin film Pt also conducts through the ceramic substrate and Loctite layer as shown in Fig. 7. The combination of the thermal resistance of the Loctite layer ($R_L$) and the thermal resistance of the ceramic substrate ($R_C$) is parallel to the thermal resistance of the glass layer. Thus for the catheter sensor, we calculate $h$, using:

\[
 h = \frac{1}{A \left[ \frac{\Delta T}{q} - \frac{R_G + R_L + R_C}{2} \right] + \sqrt{\frac{(R_G + R_L + R_C)^2}{4} + \left( \frac{\Delta T}{q} \right)^2} - R_G (R_L + R_C)}
\]

where $A$ is the exposed surface area of the catheter sensor = 1.9 mm × 3.2 mm
D. Heat dissipation through each side of the sensor

The heat does not dissipate equally through each side of the sensor. The heat dissipation is proportional to the result of total thermal resistance of the opposite site of the sensor divided by the total thermal resistance of both sides of the sensor. From equations (11), (12) and (13), the thermal resistances of glass, ceramic, Loctite and are 3.22 K/W, 12.21 K/W and 299.04 K/W respectively. The thermal resistance of the convection, $R_{\text{conv}}$, could be calculated using equation (14). However, $R_{\text{conv}}$ varies depending on the value of $h$, which depends on the flow rate. Using the theoretical values of $h$ at various flow rates, $R_{\text{conv}}$ at 0 L/min is 82.24 K/W and $R_{\text{conv}}$ at 5 L/min is 16.45 K/W.

The percentage of heat dissipation could then be estimated using the following equations:

\[
\% \text{ Heat dissipation through the glass side} = \frac{\text{thermal resistance of the ceramic and loctite side}}{\text{total thermal resistance of both sides}} = \frac{R_c + R_L + R_{\text{conv}}}{R_g + R_{\text{conv}} + (R_c + R_L + R_{\text{conv}})} \quad (16)
\]

\[
\% \text{ Heat dissipation through the ceramic and Loctite side} = \frac{\text{thermal resistance of the glass side}}{\text{total thermal resistance of both sides}} = \frac{R_g + R_{\text{conv}}}{R_g + R_{\text{conv}} + (R_c + R_L + R_{\text{conv}})} \quad (17)
\]

For a low value of $h$ ($\approx 2000 \text{ W/m}^2\text{-K}$), the percentage of heat dissipation through the glass side and the ceramic and Loctite side are 82% and 18% respectively. For a high value of $h$ ($\approx 10000 \text{ W/m}^2\text{-K}$), the percentage of heat dissipation through the glass side and the ceramic and Loctite side are 94% and 6% respectively.
E. In vitro experimental results

Fig. 8 shows the value of $h$ from 16 experiments at various flow rates in three different types of media: distilled water, normal water and saline. All of the in vitro experiments were performed at 39 °C similar to the swine body temperature.

The experimental results show that from 0 to 5 L/min, values of $h$ in W/(m²·K) were for distilled water 5100 to 13000, for normal water 5500 to 12300, and for saline 5400 to 13600 (as shown in Fig. 8). At low flow rates (from 1 to 3 L/min), distilled water yielded the highest value of $h$ though at the higher flow rates (from 4 L/min or higher), saline yielded the highest value of $h$. Nevertheless, we found that there is no significant difference of $h$ among these three media.

When we compared the experimental results with the theoretical values, we found that at all flow rates, the $h$ resulting from the in vitro experiments of those three types of media yielded a higher value of approximately 1500 W/m²·K above the theoretical value.

IV. CONCLUSIONS AND DISCUSSION

We have built an instrument for the measurement of $h$ and tested it in distilled water, normal water and saline. The measured $h$ varies significantly with flow rate, as theory suggests it should, but does not follow the square root dependence on Reynolds number that is expected. The present data vary approximately with the 0.3 power, whereas theory suggests 0.5. This difference might be explained by the fact that the condition of the experiments is different from the theoretical condition; the flow we created might not have been perfect laminar flow. Surface roughness might have been a factor. Another reason for this difference might be due to heat transfer from the bottom side of the heated film through the support structure. This would introduce a heat loss path whose heat flow is only partly responsive to the Reynolds number (the
boundary layer resistance on the bottom side of the structure is in series with a large resistance through the substrate) and might also contribute to the difference of the experimental results.

The data change slightly with the type of medium. All three data sets lie within ±7% of the mean line (excluding the free convection results, which are not expected to lie on the same lines as forced convection).

In order to use this instrument to measure \( h \) \textit{in vivo}, we must ensure that no components of the blood will deposit on the surface of the sensor and form a coating that would change the thermal resistance between the Pt film and the blood. To prevent this from happening, the test animal may need to be heparinized. However, it is not a desirable solution because the heparin would prevent the blood from any open wounds from coagulating causing the test animal to hemorrhage and might jeopardize the survival of the animal.

One alternative is to coat the catheter tip (the sensor) with heparin and carbon solution [7, 23]. Although this may solve the blood deposition problem, a more complicated correction factor would be needed to compensate for the change of the thermal resistance caused by the heparin-coating layer. The other alternative is not leaving the sensor in the blood pool for longer than 20 min and to clean the surface of the sensor every time we change the measuring location.

**ACKNOWLEDGEMENTS**

I would like to thank Prof. Robert J. Moffat, emeritus professor from Stanford University, Department of Mechanical Engineering, for his valuable suggestions and comments towards this paper. I also would like to thank Dr. James A. Will from Department of Animal Health and Biomedical Science, University of Wisconsin-Madison for helping me in many ways.
REFERENCES


**Figure and Table Captions:**

**Figure 1.** The RTD sensor is placed in a circulation system to measure $h$. 

**Figure 2.** The circulation system consists of a pump connected to a flow meter, a container, which is the measuring site for the RTD sensor probe, and a fluid bath, which maintains the temperature of the fluid at the measuring site at 39 °C.

**Figure 3.** Circuit diagram of constant temperature measuring system, \( R_1, R_2, \) and \( R_3 \) are wire-wound resistors with 1% tolerance, ±20 ppm/°C temperature coefficient. \( R_s \) is a resistive temperature detector.

**Figure 4.** The RTD sensor has a thin film Pt layer overcoated with glass, (a) Top view of the RTD sensor, and (b) Cross-sectional view of the RTD sensor through line C.

**Figure 5.** The Pt sensor is placed in a catheter probe with exposed area surrounded by epoxy.

**Figure 6.** Temperature vs. resistance of the bare sensor and the catheter sensor with the catheter wire immersing into the 37 °C fluid.

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**Figure 8.** *In vitro* experimental results of \( h \) at various flow rates in distilled water, pure water and saline performed at 39 °C.

**Table 1.** The value of Prandtl number, density, dynamic viscosity, specific heat, and thermal conductivity of water at 37 °C and 39 °C.

**Table 2.** Constants for use with equation (6) for isothermal surface.

**Table 3.** Theoretical value of \( \overline{h} \), Re, Nu, and \( u \) for laminar flow of the catheter sensor in a 20 mm diameter tube.
Table 1. The value of Prandtl number, density, dynamic viscosity, specific heat, and thermal conductivity of water at 37 °C and 39 °C.

<table>
<thead>
<tr>
<th></th>
<th>At 37 °C</th>
<th>At 39 °C</th>
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<tbody>
<tr>
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<td>4.4</td>
</tr>
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<td>ρ (kg/m³)</td>
<td>993.9</td>
<td>992.47</td>
</tr>
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<td>μ (N·s/m²)</td>
<td>6.9 × 10⁻⁴</td>
<td>6.7 × 10⁻⁴</td>
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<tr>
<td>Cₚ (J/kg·K)</td>
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</tr>
<tr>
<td>k (W/m·K)</td>
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Table 2. Constants for use with equation (6) for isothermal surface.

<table>
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<th>GrfPrf</th>
<th>C</th>
<th>m</th>
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<tr>
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<td>Use Fig. 7-7 of [6]</td>
<td>Use Fig. 7-7 of [6]</td>
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<tr>
<td>10¹ – 10⁵</td>
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<td>10⁹ – 10¹³</td>
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Table 3. Theoretical value of $\bar{h}$, Re, Nu, and $u$ for laminar flow of the catheter sensor in a 20 mm diameter tube.

<table>
<thead>
<tr>
<th>Flow (L/min)</th>
<th>$u$ (m/s)</th>
<th>Re</th>
<th>Nu</th>
<th>$\bar{h}$ (W/m²·K)</th>
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<tr>
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<td>0.5</td>
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<td>0.265</td>
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<td>54.6</td>
<td>10700</td>
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</table>
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CHAPTER 3

MINIMIZING BLOOD DEPOSITION THROUGH HEPARIN–CARBON COATING ON THE SURFACE OF CATHETER SENSORS FOR IN VIVO MEASUREMENT OF THE ENDOCARDIAL CONVECTIVE HEAT TRANSFER COEFFICIENT

This chapter has been submitted for publication as:

**ABSTRACT**—We coated a catheter sensor with heparin–carbon coating to minimize the deposition of blood components and used it for in vivo measurement of the endocardial convective heat transfer coefficient, $h$. We performed in vivo experiments in 5 pigs with the coated and the uncoated sensor and compared the results. The coated sensor resisted the deposition of the blood components for longer than 60 min with very little or no blood deposition, while the uncoated sensor resisted it for only 20 min before significant blood components were deposited on the surface of the sensor. The values of $h$ obtained from the coated sensor were noticeably higher than the ones obtained from the uncoated sensor due to the additional thermal resistance of the coated layer.

**Index Terms**—heparin–carbon coating, blood deposition, anticoagulant, radiofrequency cardiac catheter ablation, endocardial convective heat transfer coefficient.

**I. INTRODUCTION**

Radiofrequency (RF) cardiac catheter ablation has been widely used for treatment of many forms of arrhythmias [1, 5, 12, 13, 15]. To improve the success rate of RF ablation, researchers utilize the finite element method (FEM) to simulate ablation [6, 10, 11, 14, 18, 19, 20, 24]. The values of the endocardial convective heat transfer coefficient, $h$, are crucial to the simulation of heat loss due to the flow of blood inside the cardiac chambers. To obtain the values of $h$, in vivo measurements in pigs were performed [16, 17] by advancing the catheter sensor to the designated sites and heating the sensor about 5 °C above the pig’s body temperature. However, blood components were deposited on the surface of the sensor during the measurement after 20 min (see Fig. 1).
When devices were in blood, for different materials it took a different period of time before blood components deposited on the surface [3, 4, 21, 22]. For example, polyethylene catheters did not maintain their patency for more than one day when placed in the canine renal artery [3, 4], a siliconized flowmeter, used for an in vitro blood system, did not function properly after 30 min because of the fibrin deposition [4], severe thrombosis occurred within 2 h of placing a plain graphite surface or a graphite–benzalkonium surface in the canine vena cava [4, 21], the presence of a plain plastic surface or a siliconized surface in the canine vena cava resulted in severe clotting and moderate clotting, respectively, within 2 h [4, 21]. Coagulation of blood occurred on glass surfaces within 7 min in a low blood flow environment [3, 4].

Heparin or sulphated mucopolysaccharide, an anticoagulant drug, has been widely used to prevent blood clotting by injecting it into the bloodstream, termed heparinization. Researchers created formulas or mixtures for thrombus-resistant coating that imitates the chemicals found in the endothelium of blood vessels [2, 4, 8]. The major components in the thrombus-resistant coatings are heparin, an active cationic surface and a graphite layer [3, 4, 21, 22]. The heparin coating has been widely used to prevent or minimize the deposition of blood components for prosthetic devices, cardiopulmonary bypass devices, ventricular assist devices, oxygenators for coronary artery bypass grafting, coronary stents, etc. [4, 7, 9, 21 23]. Gott et al. and Whiffen et al. [4, 21] created several heparin–carbon coating formulas and coated them on prosthetic valves for the human and canine heart. They showed that the coated valves lasted in the canine heart for 12 to 30 months with no thrombus.

Since we needed to perform in vivo measurements of $h$ in pigs, we had to ensure that little or no blood components deposited on the surface of the catheter sensor in order to avoid adding any unknown thermal resistance into the system. Heparinizing the test animal was not an option
because the blood from any open cut would not coagulate, hence jeopardizing the survival of the test animal during the in vivo experiment. Thus, we coated our catheter sensor with heparin–carbon coating following Gott et al.’s recipe [3, 4] and tested its thrombus resistant property by performing in vivo experiments in 5 pigs with the coated and the uncoated sensor and then compared the results.

II. METHOD AND MATERIALS

We followed the formula of heparin–carbon coating created by Gott et al. [3, 4] using colloidal graphite Dag 154 (Acheson Colloids Company, Port Huron, MI), benzalkonium chloride and heparin to coat on the surface of the catheter sensor. Dag 154, colloidal graphite with an average graphite particle diameter of 2 µm, provides lubrication and a smooth and uniform adsorptive layer. Benzalkonium chloride, normally used as a surgical disinfectant and a main component of detergents, was used as a cationic surface-active agent for bonding heparin to the graphite layer. Heparin, an anticoagulant drug, inactivates the thrombin and other clotting factors, thus preventing the deposition of the blood components.

A. Heparin–carbon coating procedure

To coat the surface of the catheter sensor with the heparin–carbon coating, the sensor was coated first with the graphite mixture. The mixture contained two parts 95% ethyl alcohol and one part Dag 154. The surface of the sensor was thoroughly cleaned with alcohol. The sensor was immersed into the graphite mixture for 5 min, and then removed. The sensor was quickly immersed again into the graphite mixture for 1 min, followed by shaking and brief air-drying for 5 min. It was then dried by forced air for 48 h at room temperature to remove all of the solvents.
After it was completely dry, the graphite-coated sensor was dipped in 1:1000 aqueous benzalkonium chloride for about 5 min. It was briefly rinsed in saline and left in a solution of three parts saline and one part pure heparin for 10 min. The coated sensor was air-dried for another 10 min and it was then ready for the *in vivo* experiments. Fig. 2 shows the tips of coated and uncoated sensors.

**B. Catheter sensor and measuring system**

Fig. 3a shows the catheter sensor, which was a modified temperature ablation catheter (Blazer II Temp. ablation catheter, Boston Scientific Corp.) with a Pt thin film resistive temperature sensor (TFD, Omega Engineering) [16, 17] (Fig. 2b) on its tip. The temperature sensor was covered by a thin layer of glass ($\approx 27 \, \mu m$) on top of the thin film Pt to protect it from erosion and any mechanical impact. Another layer of ceramic substrate ($\approx 0.45 \, mm$) was underneath the Pt thin film. The temperature sensor was mounted on the tip of an ablation catheter and Loctite, a biocompatible epoxy, sealed the electric connection, covering the backside of the sensor and rounding the sharp edges of the sensor. Dr. Dorin Panescu of the Boston Scientific Corp. supplied the catheter. Fig. 4 shows the measuring circuit used to measure $h$ [16, 17]. Fig. 5 shows the equivalent structure of the catheter sensor and the electrical analogy. The values of $h$ can be calculated using [16, 17]:

$$h = \frac{1}{A \left[ \frac{\Delta T}{q} - \frac{R_G + R_L + R_C}{2} \right] + \sqrt{\left( \frac{R_G + R_L + R_C}{4} \right)^2 + \left( \frac{\Delta T}{q} \right)^2 - R_G (R_L + R_C)}}$$ (1)

where: $A$ is the exposed surface area of the catheter sensor $= 3.2 \, mm \times 1.9 \, mm$

$$= 6.08 \times 10^{-6} \, m^2.$$
$\Delta T$ is the temperature difference between the heated sensor and the blood (K).

$q$ is the electric power consumed by the heated sensor (W).

$R_G$ is the thermal resistance of the glass layer $= \frac{d_G}{k_G A}$ (K/W).

$R_C$ is the thermal resistance of the ceramic substrate $= \frac{d_C}{k_C A}$ (K/W).

$R_L$ is the thermal resistance of the Loctite layer $= \frac{d_L}{k_L A}$ (K/W).

and: $d_G$ is the thickness of the glass layer = 0.027 mm.

$k_G$ is the thermal conductivity of the glass = 1.38 W/m·K.

$d_C$ is the thickness of the ceramic substrate = 0.45 mm.

$k_C$ is the thermal conductivity of the ceramic = 6.06 W/m·K.

$d_L$ is the thickness of the Loctite = 1.00 mm.

$k_L$ is the thermal conductivity of the Loctite = 0.55 W/m·K.

Note that equation (1) does not include the thermal resistance of the heparin–carbon coating layer which would be a layer on top of both the glass layer and the loctite layer in Fig. 5. However, because we do not have information about the thermal properties of these main components (heparin, benzalkonium and Dag 154) and we do not know the real thickness of the coating layer, we did not add the thermal resistance of that layer into the calculation for $h$. As a result, the values of $h$ calculated from the coated sensor would be slightly smaller than the ones obtained from the uncoated sensor because of the additional thermal resistance of the coating layer.
C. Pig preparation

We obtained pigs, weighing from 12 to 34 kg, from the Department of Animal Science at University of Wisconsin-Madison. The protocol for these studies was approved by the Animal Care and Use Committee and was in compliance with all NIH guidelines for the humane use of animals in research. The pigs were sedated by injecting Telazol®, a narcotic pre-anesthetic, intramuscularly at an approximate dose of 4 mg/kg. They were then masked to a surgical plane of anesthesia with a 5% halothane. The anesthetic was then reduced to 3% to 4% and incisions were made with an electrosurgical unit through the skin and underlying tissues to expose the trachea and the sternum. The tracheostomies were achieved by dissection and intubation of the pigs’ trachea. The pig was then ventilated and the anesthetic levels adjusted to maintain an oxygen saturation near 100% and a heart rate was between 80 to 130 beats per minute for the balance of the experiment. The chest of the animal was opened by cutting the sternum from the xiphoid process to the thoracic inlet. Any bleeding from the cuts was ligated or stopped by electrocautery. A surgical retractor kept the chest open, exposing the intact heart within the pericardium. The catheters were introduced into the jugular vein for the right heart measurements and either through the carotid artery or the left atrium directly for the left heart measurements. Placement was verified by palpation and visual observation.

D. In vivo experiments

In order to perform in vivo experiments, the pericardium was opened and the catheters were advanced to the designated measuring sites in the atria, the ventricles and the ascending aorta. We performed in vivo experiments during three steps:

Step 1: In vivo experiment in the ascending aorta using the uncoated catheter sensor.
To study the time the blood components deposit on the surface of the uncoated catheter sensor, we performed one *in vivo* experiment in the ascending aorta of one pig using the uncoated catheter sensor. By advancing the uncoated catheter sensor to the ascending aorta through the carotid artery, the sensor was used to measure the value of $h$ at the ascending aorta. The measurement was done every 5 min during a 60 min period.

**Step 2: In vivo experiment in the right ventricle using both the coated and the uncoated catheter sensors.**

To compare the blood deposition resistance properties of both the coated and the uncoated catheter sensors, we performed two *in vivo* experiments in the right ventricle of one pig using both catheter sensors. The catheter sensors were advanced to the right ventricle via the jugular veins. Both catheter sensors were placed close to each other in the right ventricle and they were used to measure the values of $h$ in the right ventricle every 5 min consecutively for the period of 60 min. Since these catheters are stiff, we anticipated little or no movement of the catheter sensors during the measurement. The measuring time of both sensors was slightly different from each other since we recorded approximately 30 s using one sensor and then recorded from the other sensor after we finished the measurement from the first sensor.

**Step 3: In vivo measurement of $h$ in 6 locations in two pigs using both the coated and the uncoated catheter sensors.**

To compare the values of $h$ obtained from both the coated and the uncoated catheter sensors in various locations, we performed more $h$ measurements in 6 locations in the cardiac chambers of three pigs using both catheter sensors. The measuring locations are the floating position in the right atrium, against the lateral wall of the right atrium, the
AV node, the lateral wall of the right ventricle, the floating position in the right ventricle and in the left ventricle. Both sensors were cleaned with saline frequently to ensure that no blood components covered the surface of the catheter sensors while they were used to measure the values of $h$ in those designated locations.

III. RESULTS AND DISCUSSIONS

A. In vivo experiment in the ascending aorta using the uncoated catheter sensor.

Fig. 6 shows the result from Step 1 using the uncoated catheter sensor. The value of $h$ measured in the ascending aorta remained almost constant at 6300 W/m$^2$·K for the first 20 min and then dropped dramatically. After 25 min, the value of $h$ remained almost constant at 2000 W/m$^2$·K. We hypothesize that the surface of the catheter sensor was clean and free of blood deposition during the first 20 min. After that, a significant amount of blood deposition occurred on the surface of the catheter sensor causing the major change of the value of $h$. The dramatic change of the value of $h$ after 20 min happened after the first layer of the blood components (platelets) deposited on the surface of the sensor, this layer then activated a cascade of clotting causing a major amount of blood components deposited on the surface of the sensor. Because the deposition of the blood components covered the surface of the catheter sensor, it increased the thermal resistance between the sensor and the blood. As a result, less heat transfer occurred, which lowered the value of $h$.

B. In vivo experiment in the right ventricle using both the coated and the uncoated catheter sensor
Fig. 7 shows the experimental results obtained from Step 2. The values of $h$ were measured in the right ventricle of one pig for 60 min using both the heparin–carbon coated catheter sensor and the uncoated catheter sensors. We found that for the coated sensor, the values of $h$ measured over the 60 min period fluctuated between 6700 and 8400 W/m$^2$·K. The change of the values of $h$ over the 60 min period is noticeable but not significant (less than a 20% difference). This may be caused by the variations in blood velocity during that time period. Observation of the catheter sensor after removing from the right ventricle showed no sign of the blood deposition on the surface of the coated sensor. In contrast, the values of $h$ measured with the uncoated sensor varied between 5500 and 8900 W/m$^2$·K (about a 40% difference). This may be explained by the fact that the deposition of the blood components occurred on the surface of the uncoated catheter sensor. The layer of the blood deposition created an additional thermal resistance to the original system (as shown Fig. 5); and therefore reduced the value of $h$. The blood deposition might also have detached from the surface of the sensor because of the high blood velocity; hence a higher value of $h$ was detected. A large amount of blood deposition on the surface of the uncoated sensor was observed after removing it from the right ventricle.

Because the blood velocity at the ascending aorta is much higher than that in the right ventricle, it would take a longer time for blood components to deposit on the surface of the sensor in the ascending aorta than it would in the right ventricle. As a result, the values of $h$ in the ascending aorta remained almost constant for about 20 min, whereas, those in the right ventricle dropped within 10 min.

**C. In vivo measurement of h in 6 locations in two pigs using both the coated and the uncoated catheter sensors**
Table 1 shows the values of $h$ in 6 locations obtained from both the coated sensor and the uncoated sensor. We found that the $h$ values obtained from the coated catheter sensor were different those obtained from the uncoated catheter. This can be explained by the fact that the coated catheter sensor has another layer of the heparin–carbon coating. This layer added thermal resistance into the original system. However, equation (1) does not include that additional thermal resistance.

In theory, the additional thermal resistance from the coating layer should reduce the value of $h$. The anticipated results from the coated sensor should be lower than those from the uncoated sensor. However, our results show the opposite. This may due to the fact that the roughness of the coating layer might be different from the roughness of the glass layer of the sensor, which might contribute to the difference of the results. The second reason might be that most of the results from the coated sensor were obtained from only one or two measurements; hence they might not be able to serve as a good representation of the data. Another reason might be because of the different time after entering the blood environment. The values of $h$ obtained from the catheter sensors were not measured at the exact same time. Therefore the blood velocities at time of measurement were different, hence different values of $h$.

**IV. CONCLUSION**

We coated our catheter sensor with a heparin–carbon coating to improve its thrombus resistant property. We performed in vivo experiments using both the coated and the uncoated catheter sensors and compared results. We found that in the ascending aorta (high blood velocity), the uncoated sensor could resist blood deposition for only 20 min. In the right ventricle (lower blood velocity), the uncoated sensor could resist blood deposition for only 10 min, while the coated
sensor could resist blood deposition for more than 60 min. This demonstrated that the heparin–carbon coating can really improve the thrombus resistance property of the coated object.

There is, however, a negative aspect of this heparin–carbon coating for \textit{in vivo} \( h \) measurement. This layer of heparin–carbon coating created an additional thermal resistance in the system, and we have no information about the thermal properties of the coating ingredients or the coating layer. As a result, we could not correct equation (1) to yield accurate values of \( h \).

We concluded that the heparin–carbon coating can successfully improve the resistance to the blood deposition. Nevertheless, it is not practical for \textit{in vivo} \( h \) measurement because of the unknown thermal properties of the coating layer. It is, however, very useful for coating any vascular prosthetic devices and any \textit{in vivo} or \textit{in vitro} experimental instruments. However, if we really need to use the coated sensor for our \( h \) measurements, we might be able to calibrate the coated sensor by performing a fair amount of \textit{in vitro} measurements using both coated and the uncoated sensor and then compare the results and calculate for the additional unknown thermal resistance of the coated layer. We could then use that thermal resistance to calculate for the correction for more accurate result.

**REFERENCES**


Figure and Table Captions:

Figure 1. The blood deposition on uncoated catheter sensors: (a) after 20 min, (b) after 50 min. The arrows point to the dark part on the surface of the sensor where the blood deposited.

Figure 2. The tips of two catheter sensors. The one on the left was coated with heparin-carbon coating. The one on the right was uncoated.
Figure 3. The catheter sensor: (a) The steerable catheter has Loctite coating except where the thin film Pt sensor (RTD) at its tip is, (b) Top view of the RTD sensor, and (c) Cross-sectional view of the RTD sensor through line C.

Figure 4. Circuit diagram of constant temperature measuring system, $R_1$, $R_2$, and $R_3$ are wire-wound resistors with 1% tolerance, ±20 ppm/$^\circ$C temperature coefficient. $R_s$ is a resistive temperature detector.

Figure 5. One-dimensional heat transfer through the catheter sensor, (a) the equivalent structure of the catheter sensor, (b) the electrical analogy.

Figure 6. The value of $h$ measured at the ascending aorta dropped after 20 min using the uncoated catheter sensor.

Figure 7. The values of $h$ measured in the right ventricle remained stable using the heparin-carbon coated sensor and dropped after 10 min for the uncoated catheter sensor.

Table 1. Experimental results of the in vivo measurement of $h$ in 6 locations in three pigs using both the coated and the uncoated catheter sensor.

<table>
<thead>
<tr>
<th>Location</th>
<th>$h$ from coated sensor (W/m²·K)</th>
<th>SD</th>
<th>$h$ from uncoated sensor (W/m²·K)</th>
<th>SD</th>
<th>% difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right atrium (floating)</td>
<td>9500</td>
<td>N/A</td>
<td>8900</td>
<td>350</td>
<td>6.3</td>
</tr>
<tr>
<td>Right atrium (lateral wall)</td>
<td>4500</td>
<td>1350</td>
<td>4000</td>
<td>600</td>
<td>11.1</td>
</tr>
<tr>
<td>Right ventricle (floating)</td>
<td>7400</td>
<td>1000</td>
<td>7000</td>
<td>620</td>
<td>5.4</td>
</tr>
<tr>
<td>Right ventricle (lateral wall)</td>
<td>5300</td>
<td>N/A</td>
<td>4800</td>
<td>750</td>
<td>9.4</td>
</tr>
<tr>
<td>AV node (ventricle side)</td>
<td>3900</td>
<td>N/A</td>
<td>3800</td>
<td>350</td>
<td>2.6</td>
</tr>
<tr>
<td>Left ventricle</td>
<td>10400</td>
<td>N/A</td>
<td>9500</td>
<td>1200</td>
<td>8.7</td>
</tr>
</tbody>
</table>
Figure 1. The blood deposition on uncoated catheter sensors: (a) after 20 min, (b) after 50 min.

The arrows point to the dark part on the surface of the sensor where the blood deposited.
Figure 2. The tips of two catheter sensors. The one on the left was coated with heparin–carbon coating. The one on the right was uncoated.
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![Circuit Diagram](image)

Figure 4. Circuit diagram of constant temperature measuring system, $R_1$, $R_2$, and $R_3$ are wire-wound resistors with 1% tolerance, ±20 ppm/°C temperature coefficient. $R_s$ is a resistive temperature detector.
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CHAPTER 4

IN VIVO MEASUREMENT OF SWINE ENDOCARDIAL CONVECTIVE HEAT TRANSFER COEFFICIENT

This chapter has been submitted for publication as:

**ABSTRACT**—We measured the endocardial convective heat transfer coefficient, \( h \), at 22 locations in the cardiac chambers of 15 pigs *in vivo*. A thin film Pt catheter tip sensor in a Wheatstone-bridge circuit, similar to a hot wire/film anemometer, measured \( h \). Using fluoroscopy, we could precisely locate the steerable catheter sensor tip and sensor orientation in pigs’ cardiac chambers. With flows, \( h \) varies from 2500 to 9500 W/m\(^2\)-K. With zero flow, \( h \) is approximately 2400 W/m\(^2\)-K. These values of \( h \) can be used for the FEM modeling of radiofrequency cardiac catheter ablation.

**Index Terms**—convective heat transfer coefficient, radiofrequency ablation, cardiac radiofrequency ablation, heat convection, endocardial convective heat transfer coefficient

**I. INTRODUCTION**

According to the American Heart Association [23], as many as two million people in the United States suffer from atrial fibrillation and more than 100,000 people suffer from tachycardia. Radiofrequency (RF) catheter cardiac ablation has been successfully used for the treatment of these cardiac arrhythmias as a preferable alternative of the conventional “Mesh” operation (in which long surgical cuts interrupt excitation pathways) and drug therapy because of its controllability, high efficacy, low complication rate and minimal invasiveness [2, 3, 9, 10, 15, 18]. In order to improve the electrodes and enhance the success rate of RF catheter ablation, researchers have performed finite element method (FEM) modeling of RF catheter ablation [6, 7, 17, 18, 19, 21]. The value of endocardial convective heat transfer coefficient (\( h \)) is important to accurately simulate heat loss from the endocardium since the blood flow in the cardiac chambers carries away a large amount of heat from the ablation site during the ablation [6, 18, 19]. Researchers have used the value of \( h \) ranging from 44 to 6090 W/m\(^2\)-K for different locations in
the cardiac chambers [1, 6, 7, 8, 13, 14, 17, 19]. However, none of these values came from in vivo measurements in animals. Bhavaraju obtained the values of $h$ from in vitro experiments [1]. Shahidi et al. and Labonte estimated the values of $h$ from mathematical calculations assuming the blood flow is laminar [7, 14]. Tungjitkusolmun et al. estimated the values of $h$ using the blood velocities in the cardiac chambers obtained by Doppler ultrasound [19]. Jain et al. used the value of $h$ of 1800 W/m²·K for their FEM analysis but did not mention their sources of the value of $h$ [6].

Few attempts to measure $h$ in vivo have been reported. Bhavaraju [1] constructed a physical model of the swine heart from silicone rubber and embedded thermistors (BR11, Thermometrics Inc.) into the wall of the model heart. He then pumped a blood substitute, consisting of 40% glycerol and 60% water by weight, at various flow rates through the model heart and measured in vitro $h$, in several locations. His experiments yielded values of $h$ ranging from 44 to 3930 W/m²·K. Since his heart model was stiff, and not as flexible as the real heart, at many locations, especially under the valves, his experimental results of $h$ are extremely low and questionable. Santos et al. [12, 13] used a Swan-Ganz catheter with a thermistor embedded near its tip to measure $h$ in eight locations in two pigs in vivo. The values of $h$ obtained from his measurements range from 510 to 4800 W/m²·K. However, his experimental results are questionable for several reasons. First, the Swan-Ganz catheter he used is neither stiff nor steerable, making it extremely difficult to accurately locate the measuring sites. Second, he did not use fluoroscopy. Third, the orientation of the measuring surface of his catheter could face the endocardial wall instead of the flowing blood during the measurement, which could significantly alter the results. Fourth, because the thermistor that is embedded on the surface of his catheter is
as thick as 0.2 mm, the average temperature he measured was different from the surface temperature, which would result in miscalculation of $h$.

Because we could not find any reliable values of $h$ in previous research, we measured $h$ using a Wheatstone bridge circuit, similar to a hot wire/film anemometer circuit, connecting it to a steerable catheter sensor, using fluoroscopy and using the system to perform in vivo measurements of the endocardial convective heat transfer coefficient, $h$, in 22 locations inside the cardiac chambers of 15 pigs.

**II. METHOD**

Fig. 1 shows the flow chart of in vivo measurement of $h$. The test animals (pigs) were prepared for in vivo measurement. The catheter sensor shown in Fig. 2 was advanced into the cardiac chambers of the pigs to the measuring locations. The catheter sensor was connected to the measuring system, which heated the sensor about 5 °C above the pig’s body temperature. The temperature of the heated sensor and the electric power consumed by the sensor were measured and collected by data acquisition programs, Biobench and Labjack. Calculation from those data yielded $h$ [16].

**A. Pig preparation**

We obtained pigs, weighing from 12 to 45 kg, from the Department of Animal Science at University of Wisconsin-Madison (for open chest measurement) and from UW Hospital, University of Wisconsin-Madison (for measurement under fluoroscopy). The protocol for these studies was approved by the Animal Care and Use Committee and was in compliance with all NIH guidelines for the humane use of animals in research. The pigs were sedated by injecting
Telazol®, a narcotic pre-anesthetic, intramuscularly at an approximate dose of 4 mg/kg. They were then masked to a surgical plane of anesthesia with 5% halothane. The anesthetic was then reduced to 3% to 4% and incisions were made with an electrosurgical unit through the skin and underlying tissues to expose the trachea and the sternum. The tracheostomies were achieved by dissection and intubation of the pigs’ trachea. The animals were then ventilated and the anesthetic levels adjusted, maintained at oxygen saturation near 100% and the heart rate was between 80 to 130 beats per minute for the balance of the experiment.

For open chest measurements (without fluoroscopy), the chest of the animal was opened by cutting the sternum from the xiphoid process. Any bleeding from the cuts was ligated or stopped by electrocautery. A surgical retractor kept the chest open, exposing the intact heart within the pericardium. The catheters were introduced into the jugular vein for the right heart measurements and either through the carotid artery or the left atrium directly for the left heart measurements. Placement was verified by palpation and visual observation.

For the measurements under fluoroscopy, the external jugular of the test animal was canulated and the catheter sensor was advanced to the right atrium and then the right ventricle through the tricuspid valve. The catheter sensor was introduced through the carotid artery to reach the left ventricle and then the left atrium through the mitral valve. Placement of the catheter sensor was verified by fluoroscopy. The Pt layer of the catheter sensor could be seen under fluoroscopy thus we could rotate the sensor to the proper orientation (facing the flowing blood) for our h measurements.
B. The catheter sensor

Fig. 2a shows the catheter sensor, which was a modified temperature ablation catheter (Blazer II TM Temp. ablation catheter, Boston Scientific Corporation) with a Pt thin film resistive temperature sensor (TFD, Omega Engineering Inc.) as shown in Fig. 2b on its tip. The temperature sensor was covered by a thin layer of glass (≈ 27 µm) on top of the thin film Pt to protect it from erosion and any mechanical impact. Another layer of ceramic substrate (≈ 0.45 mm) was underneath the Pt thin film. The temperature sensor was mounted on the tip of an ablation catheter and Loctite, a biocompatible epoxy, sealed the electric connection, covering the backside of the sensor and rounding the sharp edges of the sensor. Dr. Dorin Panescu of the Boston Scientific Corporation supplied the catheter.

C. The measuring system

We used the circuit shown in Fig. 3 [16] to measure $h$. The measuring circuit, similar to a constant temperature hot wire/film anemometer, is a Wheatstone bridge circuit with the catheter sensor in one of its arms and three wire-wound resistors (1 W) and a precision potentiometer in the other arms. The bridge circuit maintains the resistance of the sensor constant; thus the temperature of the heated sensor remains constant (in the constant temperature mode) during measurement. The potentiometer is adjusted so the sensor heats 5 °C above the flowing fluid temperature.

To acquire the value of $h$, we measured the voltage across the sensor ($V_1$) and the voltage across $R_1$ and $R_s$ ($V_A$) and recorded them using the BioBench program and Labjack program. $V_A$ and $V_1$ were used to determine the current that flows through the sensor. $V_1$ was also used to
determine the resistance and the temperature of the sensor. Once we knew the current, the resistance and the temperature of the sensor, we could estimate the electric power consumed by the heated sensor. Hence we calculated $h$ using:

$$Q_h = hA(T_s - T_b)$$

(1)

where: $Q_h$ is the electric power consumed by heating the sensor (W)

$h$ is the convective heat transfer coefficient (W/m$^2$·K)

$A$ is the sensor area (m$^2$)

$T_s$ is the temperature of the heated sensor (K)

and $T_b$ is the temperature of the blood that flows adjacent to the site (K)

However, a correction is needed to obtain a more accurate value of $h$ since there is a glass layer cover on top of the Pt thin film and another layer of ceramic substrate underneath the Pt thin film [16]. The value of $h$ was calculated using:

$$h = \frac{1}{A\left[\frac{\Delta T}{q} - \frac{R_G + R_L + R_c}{2}\right] + \sqrt{\frac{(R_G + R_L + R_c)^2}{4} + \left(\frac{\Delta T}{q}\right)^2 - R_G (R_L + R_c)}}$$

(2)

where: $A$ is the exposed surface area of the catheter sensor (as shown in Fig. 2(a))

$= 3.2 \text{ mm} \times 1.9 \text{ mm} = 6.08 \times 10^{-6} \text{ m}^2$

$\Delta T$ is the temperature difference between the heated sensor and the blood (K).

$q$ is the electric power consumed by the heated sensor (W).

$R_G$ is the thermal resistance of the glass layer = $\frac{d_G}{k_G A}$ (K/W).
$R_C$ is the thermal resistance of the ceramic substrate $= \frac{d_C}{k_C A}$ (K/W).

$R_L$ is the thermal resistance of the Loctite layer $= \frac{d_L}{k_L A}$ (K/W).

and: $d_G$ is the thickness of the glass layer = 0.027 mm.

$k_G$ is the thermal conductivity of the glass = 1.38 W/m·K.

dc is the thickness of the ceramic substrate = 0.45 mm.

$k_C$ is the thermal conductivity of the ceramic = 6.06 W/m·K.

$d_L$ is the thickness of the Loctite = 1.00 mm.

$k_L$ is the thermal conductivity of the Loctite = 0.55 W/m·K.

**D. In vivo measurement**

In order to perform in vivo measurements, the pericardium was opened and the catheters were advanced to the designated measuring sites in the atria, the ventricles and the valves.

We measured the values of $h$ in 22 locations in the pig’s cardiac chambers. The measuring locations in the right atrium were the lateral and the medial walls, the floating position in the right atrium and at the AV node. For the right ventricle, we measured the floating position in the right ventricle, the lateral and the septal walls, the apex of the right ventricle, the AV node (ventricular side), underneath the tricuspid valve, and the floating position in the tricuspid valve and in the pulmonary valve.

For the left atrium, the measuring locations were the lateral and medial walls of the left atrium and the floating position inside the left atrium. The measuring locations in the left
ventricle included the floating position in the left ventricle, the lateral and septal walls, the apex, underneath the mitral valve and the floating positions in the mitral valve and in the aortic valve.

When we measured the values of $h$ at the endocardial walls under fluoroscopy, we pressed the back of the catheter sensor against the walls leaving the sensing area facing the flowing blood. We utilized fluoroscopy to ensure the proper orientation of the sensor and the precise measuring sites. Fig. 4 shows the images from fluoroscopy, illustrating the placement of the catheter sensor and the proper orientation of the sensor thin film, which faced the flowing blood in the cardiac chambers. We also ensured that the catheter sensor was not placed in the blood for more than 20 min. After 20 min the deposition of blood components on the sensor began to affect our measurements (data not shown). We also cleaned the surface of the sensor with water and dried it every time we reinserted the sensor to remove any blood deposition. Each measurement was recorded for at least 30 s using either Biobench or Labjack data acquisition programs. The results were then analyzed using equation (2) to yield the value of $h$.

III. RESULTS

A. Theoretical calculation of $h$ in zero flow in the cardiac chambers

For free convection in still blood (at the body temperature of 39 °C) for vertical planes at an isothermal surface [16], the sensor was heated to a constant 5 °C above the blood temperature. By using the thermal properties of blood [17, 18, 21] and the volume coefficient of expansion, $\beta$, of water, the Grashof number and the Prandtl number are:

$$\text{Gr}_f \text{Pr}_f = 260$$
From Fig. 7-7 of [4], \( N_u = 3.31 \)

\[
\bar{h} = 1200 \text{ W/m}^2\cdot\text{K}
\]

where \( \bar{h} \) is the average value of \( h \) over the exposed area of the sensor.

**B. In vivo experimental results**

Table 1 shows the experimental results of the *in vivo* measurements in 22 locations in the cardiac chambers of 15 pigs. We pooled the data obtained from both the open-chest animals and under fluoroscopy since there were no significant differences in data. The values of \( h \) range from 2500 W/m\(^2\)·K to 9500 W/m\(^2\)·K. We performed at least three measurements at each location (except at the medial walls of the atria and the floating position in the aortic valve) and calculated the median values of \( h \), the average values of \( h \) and the standard deviations of the results from all measurements at each location. Fig. 5 presents the median values of \( h \) mapping on a heart diagram. The largest value of \( h \) was obtained from the floating position in the left atrium and the smallest value of \( h \) was obtained from the right atrial wall near the AV node (AV node in the right atrium). Table 2 shows the values of \( h \) at zero flow in each cardiac chamber measured after the heart stopped. The value of \( h \) in each cardiac chamber at zero flow was approximately 2400 ± 50 W/m\(^2\)·K. There was no significant difference of the free \( h \) measured from each cardiac chamber. Fig. 6 shows the waveforms from the *in vivo* measurements of \( h \) at 8 locations in the cardiac chambers recorded by the Biobench program.
IV. DISCUSSION

The goal of this project was to obtain accurate *in vivo* measurements of $h$. Several factors were critical to the design of the probe we utilized. We selected a thin sensor so the surface temperature was close to the average temperature. We determined the thermal resistance of any backing and protective coating. We needed to ensure that little or no blood components were deposited on the surface of the sensor to avoid added thermal resistance. We considered two alternatives, coating the sensor with heparin–carbon solution [20] or frequently cleaning the sensor. Coating with heparin solution is desirable but not very practical since it adds an unknown thermal resistance. As a result, every 20 min we cleaned the sensor to remove any blood deposition during the measurement. Finally, we needed to confirm the location and proper orientation of the sensor. This was accomplished both by palpation in the open chest and by fluoroscopy in the closed chest.

The median values of $h$ are reported in 22 locations in the heart. For the floating position in each chamber, the values of $h$ in the left side of the heart are higher than those in the right side. This is reasonable since blood velocity is higher in the left cardiac chambers [5]. The values of $h$ measured at the endocardial wall in the ventricles and the values of $h$ at the apices of both ventricles are higher than those at the septum and the outer walls. This may reflect that the majority of the blood from both atria flows directly to the apices while a smaller amount of blood with lower velocity flows past the septum and outer walls.

For floating positions in each valve, the highest value of $h$ is at the pulmonary valve (8000 W/m²·K) and the lowest value of $h$ is at the aortic valve (5000 W/m²·K). However, theory suggests that higher velocities should yield a higher value of $h$ and the average velocity at the ascending aorta is higher than that at the pulmonary artery [5, 22] as shown in Table 3. This low
experimental value of $h$ at the aortic valve might result from the improper placement of the catheter sensor in the ascending aorta during the measurement and this location is the only location where we did not measure under fluoroscopy. The catheter sensor might have been close to the wall of the ascending aorta, which has lower blood velocity, instead of floating in the middle of the valve. For zero flow rate, all of the values of $h$ in each cardiac chamber are very similar at approximately $2400 \pm 50 \, \text{W/m}^2\cdot\text{K}$. This value is about twice as large as the theoretical calculation value of free convection in the blood mentioned earlier. The lack of variability between measurements suggests that our result is accurate. We measured the values of $h$ when the heart had already stopped its activity thus we do not expect there to be any residual flow, though, the blood might still move through the valves. Furthermore, some minor inaccuracy of the theoretical calculation might have occurred because the thermal properties we used for the theoretical calculation were the values at $37 \, ^\circ\text{C}$, since we could not find those values at $39 \, ^\circ\text{C}$ (pig’s body temperature). Finally, we estimated some of the values, such as the volume coefficient of expansion from the value for water since we could find no values measured in blood.

Fig. 6 shows that the waveforms from the \textit{in vivo} measurements of $h$ are very similar to the waveforms from pressure measurements. Fig 6h, measured in the aortic valve, shows a similar waveform as the aortic pressure recorded by Nerem et al. [11].

\textbf{V. CONCLUSIONS}

This study shows that $h$ can be measured \textit{in vivo} using a hot wire/film anemometer system.

We measured $h$ at 22 locations in the cardiac chambers of 15 pigs. At each location, we measured at least 3 separate measurements (from different pigs) and calculated the median,
average and the standard deviation. For each measurement, we measured at least 30 s and calculated to yield the average value of $h$. We also obtained waveforms from each measurement and we found that they are very similar to the waveforms from pressure measurement.

**ACKNOWLEDGEMENT**

We thank Patrick V. Farrell of the Mechanical Engineering Department, University of Wisconsin-Madison, for his helpful suggestions about the theoretical calculation of free convection.

**REFERENCES**


**Figure and Table Captions:**

**Figure 1.** Flow chart of *in vivo* measurement where the catheter is located inside the pig’s cardiac chambers.

**Figure 2.** The catheter sensor: (a) The steerable catheter has Loctite coating except where the thin film Pt sensor (RTD) at its tip is exposed, (b) Top view of the RTD sensor, and (c) Cross-sectional view of the RTD sensor through line C.

**Figure 3.** Circuit diagram of constant temperature measuring system, $R_1$, $R_2$, and $R_3$ are wire-wound resistors with 1% tolerance, ± 20 ppm/°C temperature coefficient. $R_s$ is a resistive temperature detector.

**Figure 4.** Fluoroscopy images from the *in vivo* measurements of $h$: (a) at the apex of right ventricle, (b) at the lateral wall of right ventricle, (c) underneath the tricuspid valve, and (d) at the lateral wall of the right atrium.

**Figure 5.** The median values of $h$ from *in vivo* measurements in 15 pigs in 22 locations mapping on a heart diagram.
Figure 6. Waveforms of the voltage across the catheter sensor, $R_s$, detected during the in vivo measurement of $h$ using the Biobench data acquisition program: (a) in the right atrium, (b) in the left atrium, (c) in the right ventricle, (d) in the left ventricle, (e) in the tricuspid valve, (f) in the mitral valve, (g) in the pulmonary valve and (h) in the aortic valve. All the waveforms are not from the same pig; hence there are different heart rates.

Table 1. Experimental results of the in vivo endocardial convective heat transfer coefficient, $h$, in 22 locations from 15 pigs.

Table 2. Experimental results of the in vivo endocardial convective heat transfer coefficient, $h$, at zero flow in each cardiac chamber.

Table 3. The velocities of the blood in the human heart and some large vessels [5, 22].
Table 1. Experimental results of the *in vivo* endocardial convective heat transfer coefficient, $h$, (W/m²·K) in 22 locations from 15 pigs.

<table>
<thead>
<tr>
<th>Location</th>
<th># of data</th>
<th>Range</th>
<th>Median $h$</th>
<th>Average $h$</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right atrium (lateral wall)</td>
<td>8</td>
<td>3700–5900</td>
<td>4000</td>
<td>4360</td>
<td>820</td>
</tr>
<tr>
<td>Right atrium (medial wall)</td>
<td>1</td>
<td>N/A</td>
<td>7900</td>
<td>7900</td>
<td>N/A</td>
</tr>
<tr>
<td>Right atrium (floating)</td>
<td>5</td>
<td>5400–9400</td>
<td>7200</td>
<td>7340</td>
<td>1800</td>
</tr>
<tr>
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<td>2900–7400</td>
<td>4300</td>
<td>4890</td>
<td>1820</td>
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<tr>
<td>Right ventricle (septum)</td>
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<td>3300–6300</td>
<td>4750</td>
<td>4780</td>
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<td>Right ventricle (apex)</td>
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<td>5350</td>
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<td>1030</td>
</tr>
<tr>
<td>Right ventricle (floating)</td>
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<td>4900–8900</td>
<td>7000</td>
<td>6820</td>
<td>1480</td>
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<tr>
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<td>8000</td>
<td>1320</td>
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<tr>
<td>Tricuspid valve (underneath)</td>
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<td>3600–9500</td>
<td>6550</td>
<td>6400</td>
<td>2280</td>
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<tr>
<td>Pulmonary valve (floating)</td>
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<td>8000</td>
<td>7830</td>
<td>1160</td>
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<td>Left atrium (lateral wall)</td>
<td>8</td>
<td>3700–7800</td>
<td>5350</td>
<td>5540</td>
<td>1600</td>
</tr>
<tr>
<td>Left atrium (medial wall)</td>
<td>1</td>
<td>N/A</td>
<td>7000</td>
<td>7000</td>
<td>N/A</td>
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<tr>
<td>Left atrium (floating)</td>
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<td>8500–10000</td>
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<td>2900–6600</td>
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<td>4430</td>
<td>1690</td>
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<tr>
<td>Left ventricle (apex)</td>
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<td>3700–8500</td>
<td>5700</td>
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<td>2000</td>
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<td>Left ventricle (floating)</td>
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<td>6530</td>
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<td>Mitral valve (underneath)</td>
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<td>6800</td>
<td>6830</td>
<td>512</td>
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<td>Aortic valve (floating)</td>
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<td>3900–7100</td>
<td>5500</td>
<td>5500</td>
<td>2260</td>
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Table 2. Experimental results of the *in vivo* endocardial convective heat transfer coefficient, $h$, (W/m$^2$·K) at zero flow in each cardiac chamber.

<table>
<thead>
<tr>
<th>Location</th>
<th># of data</th>
<th>Median $h$</th>
<th>Average $h$</th>
<th>SD</th>
</tr>
</thead>
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<tr>
<td>Right atrium</td>
<td>3</td>
<td>2400</td>
<td>2430</td>
<td>58</td>
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<tr>
<td>Right ventricle</td>
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<tr>
<td>Left atrium</td>
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<td>2450</td>
<td>2430</td>
<td>96</td>
</tr>
<tr>
<td>Left ventricle</td>
<td>4</td>
<td>2300</td>
<td>2280</td>
<td>377</td>
</tr>
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</table>

Table 3. The velocities of the blood in the human heart and some large vessels (cm/s) [5, 22].

<table>
<thead>
<tr>
<th>Location</th>
<th>Range</th>
<th>Average</th>
<th>Standard deviation</th>
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<tbody>
<tr>
<td>Superior vena cava</td>
<td>28 – 80</td>
<td>51</td>
<td>13</td>
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<tr>
<td>Tricuspid valve</td>
<td>33 – 81</td>
<td>53</td>
<td>12</td>
</tr>
<tr>
<td>Pulmonary artery</td>
<td>52 – 131</td>
<td>81</td>
<td>17</td>
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<tr>
<td>Mitral valve</td>
<td>44 – 128</td>
<td>77</td>
<td>16</td>
</tr>
<tr>
<td>Ascending aorta</td>
<td>76 – 155</td>
<td>104</td>
<td>19</td>
</tr>
<tr>
<td>Descending aorta</td>
<td>70 – 160</td>
<td>101</td>
<td>17</td>
</tr>
</tbody>
</table>

Figure 1. Flow chart of *in vivo* measurement where the catheter is located inside the pig’s cardiac chambers.
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CHAPTER 5

IN VIVO MEASUREMENT OF SWINE HEPATIC CONVECTIVE HEAT TRANSFER COEFFICIENT

This chapter has been submitted for publication as:

**ABSTRACT**—We measured *in vivo* the hepatic convective heat transfer coefficient, \( h \), in 8 locations in the large vessels of the liver, the portal veins and the hepatic veins, of 8 pigs using our measuring system. This system consists of a Wheatstone-bridge circuit, similar to a hot wire/film anemometer with one of its arms connected to a catheter sensor. We could precisely locate the steerable catheter sensor in those large vessels in the pig’s liver using fluoroscopy. The experimental results of \( h \) at 4 locations in the hepatic veins vary from 4400 to 7180 W/m\(^2\)·K. The experimental results of \( h \) at 4 locations in the portal veins vary from 3000 to 6100 W/m\(^2\)·K. These values of \( h \) can be used for the FEM modeling of radiofrequency hepatic catheter ablation and hepatic cryoablation.

**Index Terms**—convective heat transfer coefficient, radiofrequency catheter ablation, hepatic RF ablation, hepatic convective heat transfer coefficient, hepatic cryoablation.

**I. INTRODUCTION**

In 2003, more than 16,000 deaths in the United States caused by liver/intrahepatic bile duct cancer were estimated (CANCER STATISTICS, 2003). The most common types of liver tumors are primary and secondary liver tumors. Primary hepatocellular carcinoma or cholangiocarcinoma may result from long-term damage to the liver from viral infections such as hepatitis B and C or from alcoholism. Secondary or metastatic liver tumors are commonly found in patients who suffer from tumors that develop somewhere else in the body and spread to the liver. Since liver tumors respond poorly to chemotherapy and radiation, the conventional treatment of liver tumors is surgical resection (PATTERSON *et al.*, 1998; CURLEY *et al.*, 1997). This conventional surgery (resection) is potentially curative but only possible for selected patients (less than 30%) with low medical complications. However, surgery is impossible for
patients who face complications due to age, comorbidity or extent of disease (TUNGJITKUSOLMUN et al., 2002; BC CANCER AGENCY, 2003; CURLEY et al., 1997). Radiofrequency (RF) ablation and cryoablation have gained more interest as alternative treatments of liver tumors because of their effectiveness, minimal invasiveness and lower cost and morbidity (CURLEY et al., 1997; GAZELLE et al., 2000; GOLDBERG et al., 1998; 2001; JIAO et al., 1999; PATTERSON et al., 1998; ROSSI et al., 1996). In order to improve the ablation techniques, the finite element method (FEM) modeling is often used to simulate the ablation (HAEMMERICH et al., 2001; 2002; 2003; CURLEY et al., 1997).

Because the liver has a dual blood supply from both the hepatic artery and portal vein, it is highly perfused. Researchers report many incidents of high recurrence rates of tumor cells’ survival next to those large vessels after ablation (HAEMMERICH et al., 2003; LU et al., 2002; ROSSI et al., 2000). One option for reducing the recurrence rates of tumor cells is to occlude the tumor’s blood supply using the Pringle Maneuver during ablation (ROSSI et al., 2000). However, this procedure is complicated and requires extra time and cost. Furthermore, ablated tissues obtained under this maneuver were larger than expected (PATTERSON et al., 1998; ROSSI et al., 2000). Therefore, the Pringle maneuver is not a practical solution for the treatment of tumor cells next to large vessels.

To overcome this problem, simulations must include the heat loss due to the high blood flow at those locations. Values of the convective heat transfer coefficient, $h$, in the large vessels of the liver are needed for these simulations. HAEMMERICH et al. (2003) used the value of $h$ at the portal vein ranging from 600 to 800 W/m$^2$·K for their simulations. However, none of those values came from in vivo measurements in animals.
Because we could not find any reliable values of $h$ in previous research, we measured $h$ using a Wheatstone bridge circuit, similar to a hot wire/film anemometer circuit. We connected it to a steerable catheter sensor, using both an open-abdomen technique and fluoroscopy to perform \textit{in vivo} measurements of the hepatic convective heat transfer coefficient, $h$, in 8 locations inside the large vessels of the liver of 8 pigs.

\section*{II. METHODS}

Fig. 1 shows the block diagram of the \textit{in vivo} $h$ measurement. The pigs were prepared for \textit{in vivo} measurement and the catheter sensor, as shown in Fig. 2, was advanced into the designated measuring sites in the large vessels of the pig’s liver. Fig. 3 shows the measuring system, which was connected to the catheter sensor and heated the sensor to about 5 °C above the pig’s body temperature. The electric power consumed by the sensor and the temperature of the sensor were determined and collected by the data acquisition programs BioBench and LabJack, and were calculated to yield $h$ (TANGWONGSAN \textit{et al.}, 2003).

\subsection*{A. Pig preparation}

We obtained 8 pigs, weighing from 12 to 45 kg, from the Department of Animal Science at University of Wisconsin-Madison (for the open abdomen measurement without fluoroscopy) and from UW Hospital, University of Wisconsin-Madison (for the measurement under fluoroscopy). The protocol for these studies was approved by the Animal Care and Use Committee and was in compliance with all NIH guidelines for the humane use of animals in research. The pigs were sedated by injecting Telazol®, a narcotic pre-anesthetic intramuscularly at an approximate dose of 4 mg/kg. They were then masked to a surgical plane of anesthesia with 5% halothane. The
anesthetic was then reduced to 3% to 4% and incisions were made through the skin and underlying tissues with an electrosurgical unit to expose the trachea. The tracheostomies were achieved by dissection and intubation of the pigs’ trachea. The animal was then placed on a ventilator and by adjusting the anesthetic levels, the balance of the experiment was maintained where the oxygen saturation was near 100% and the heart rate was between 80 to 130 beats per minute. The abdomens of the pigs were then cut open. Any bleeding from the cuts was ligated or stopped by electrocautery. A surgical retractor was used to keep the abdomen open, the intestines and stomach were moved and covered with a plastic bag or wet towels to expose the liver and the main portal vein.

For the measurement using an open abdomen technique (without fluoroscopy), the incision was extended from the xiphoid cartilage to the pelvic symphysis. The catheter sensor was threaded directly through the main portal vein. A hemostatic dressing was applied to stop bleeding. To reach the hepatic veins, the catheter sensor was threaded directly into the inferior vena cava and the proceeded retrograde into the hepatic vein. The measuring sites were verified by visual observation and palpation.

For the measurement with fluoroscopy at the UW-hospital, the catheter sensor was introduced into the hepatic vein via the external jugular by threading the catheter through the inferior vena cava to reach the hepatic veins. For the portal vein, the opening spot was on the portal vein about 2 cm distal to the bifurcation. The measuring sites were verified by fluoroscopy as shown in Fig. 4. The Pt layer of the catheter sensor could be seen under fluoroscopy thus we could rotate the sensor to the proper orientation (facing the flowing blood) for our h measurements.
B. The catheter sensor

Fig. 2 shows the catheter sensor, which is a temperature ablation catheter (Blazer II TM Temp. ablation catheter, Boston Scientific Corporation) with its tip connected to a Pt thin film resistive temperature sensor (TFD, Omega Engineering Inc.). The temperature sensor has a thin layer of glass ($\approx 27$ $\mu$m) covering the thin film Pt to protect the thin film Pt from erosion and any mechanical impact. There is another layer of ceramic substrate ($\approx 0.45$ mm) underneath the Pt thin film. The temperature sensor is connected to the tip of an ablation catheter. Loctite, a biocompatible epoxy, seals the electric connection, covers the backside of the sensor and rounds the sharp edges of the sensor. Dr. Dorin Panescu of the Boston Scientific Corporation supplied the catheter.

C. The measuring system

We used the measuring circuit (as shown in Fig. 3) (TANGWONGSAN et al., 2003) to measure in vivo the value of the hepatic convective heat transfer coefficient, $h$. The measuring circuit, similar to a hot wire/film anemometer, is composed of a Wheatstone bridge circuit with the catheter sensor forming one of its arms and three wire-wound resistors (1 W) and a precision potentiometer forming the other arms. The bridge circuit maintained the resistance of the sensor; thus the temperature of the heated sensor remained constant (in constant temperature mode). The potentiometer was set so that the sensor was heated $5$ $^\circ$C above the flowing fluid temperature.

In order to acquire the values of $h$, we measured the voltage across the sensor ($V_1$) and the voltage across $R_1$ and $R_S$ ($V_A$) and those values were recorded by the BioBench program and Labjack program. $V_A$ and $V_1$ were used to determine the current that flows through the sensor.
$V_1$ was also used to determine the resistance and the temperature of the sensor. Once we knew the value of the current, the resistance and the temperature of the sensor, we could estimate the electric power consumed by the heated sensor. Hence we calculated $h$ using:

$$Q_h = hA(T_s - T_b)$$

(1)

where: $Q_h$ is the electric power consumed by heating the sensor (W)

$h$ is the convective heat transfer coefficient (W/m$^2\cdot$K)

$A$ is the sensor area (m$^2$)

$T_s$ is the temperature of the heated sensor (K)

and $T_b$ is the temperature of the blood that flows adjacent to the site (K)

However, a correction of equation (1) was needed to obtain a more accurate value of $h$ since there is a glass layer cover on top of the Pt thin film and another layer of ceramic substrate underneath the Pt thin film. The value of $h$ could be calculated using:

$$h = \frac{1}{A \left[ \frac{\Delta T}{q} - \frac{R_G + R_L + R_C}{2} \right] + \sqrt{\frac{(R_G + R_L + R_C)^2}{4} + \left( \frac{\Delta T}{q} \right)^2} - R_G (R_L + R_C)}$$

(2)

where: $A$ is the surface area of the glass layer = the surface area of the ceramic substrate

$= 3.2 \text{ mm} \times 1.9 \text{ mm} = 6.08 \times 10^{-6} \text{ m}^2$.

$\Delta T$ is the temperature difference between the heated sensor and the blood (K).

$q$ is the electric power consumed by the heated sensor (W).


\( R_G \) is the thermal resistance of the glass layer = \( \frac{d_G}{k_g A} \) (K/W).

\( R_C \) is the thermal resistance of the ceramic substrate = \( \frac{d_C}{k_c A} \) (K/W).

\( R_L \) is the thermal resistance of the Loctite layer = \( \frac{d_L}{k_L A} \) (K/W).

and: \( d_G \) is the thickness of the glass layer = 0.027 mm.

\( k_g \) is the thermal conductivity of the glass = 1.38 W/m·K.

\( d_C \) is the thickness of the ceramic substrate = 0.45 mm.

\( k_c \) is the thermal conductivity of the ceramic = 6.06 W/m·K.

\( d_L \) is the thickness of the Loctite = 1.00 mm.

\( k_L \) is the thermal conductivity of the Loctite = 0.55 W/m·K.

\( D. \textit{In vivo measurement} \)

We measured \textit{in vivo} the hepatic convective heat transfer coefficient, \( h \), at 8 locations in the large vessels of the livers of 8 pigs. The measuring locations were separated into two groups, the portal veins and the hepatic veins. For the portal veins, the measuring sites were the floating position in the main portal vein, against the walls of the main portal vein, the left branch of the portal vein, and the right branch of the portal vein. For the hepatic veins, the measuring locations were the floating position in the mid hepatic vein, against the walls of the midhepatic vein, the left hepatic vein, and the right hepatic vein.

When measuring the values of \( h \) at the wall of the veins, we pressed the back of the catheter sensor against the wall leaving the sensing area of the sensor facing the flowing blood.
By utilizing fluoroscopy, we could precisely locate the proper orientation of catheter sensor at the designated measuring sites. Fig. 4 shows the fluoroscopy images of the measuring locations in the large vessels of the liver.

III. RESULTS

Table 1 shows the experimental results of $h$ from in vivo measurements in 8 locations in the large vessels of the livers of 8 pigs. Fig. 5 shows the median values of $h$ mapped on a liver diagram. Since there were no significant differences in the data obtained from both the open-abdomen technique and those under fluoroscopy, we pooled the data. The values of $h$ range from 3100 to 7200 W/m$^2$·K. We performed at least 3 measurements to obtain at least 3 results from each location and used them to calculate the median values of $h$, the average values of $h$ and the standard deviations of those results. The median values of $h$ at the floating position in the main portal vein and the midhepatic vein are 6300 and 7200 W/m$^2$·K respectively. The median values of $h$ at the wall of the main portal vein, the right branch of the portal vein and the left branch of the portal vein are 3400, 3200, and 3100 W/m$^2$·K respectively. The median values of $h$ at the wall of the midhepatic vein, the right hepatic vein and the left hepatic vein are 6400, 5400 and 4200 W/m$^2$·K respectively. Fig. 6 shows the waveforms of the voltage across the sensor, $R_s$, detected during the in vivo $h$ measurement using BioBench data acquisition program.

IV. DISCUSSION

Fig. 5 shows the median values of $h$ in 8 locations, mapped on a liver diagram. For the floating position, the value of $h$ obtained from the midhepatic vein is higher than that obtained from the
main portal vein. For the values of $h$ obtained from the walls of the vessels, those from the hepatic veins are higher than those obtained from the portal veins. These results are reasonable since blood velocities in the hepatic veins are higher than blood velocities in the portal veins (as shown in Table 2).

For the results from the walls of the hepatic veins, the highest value of $h$ is at the wall of the midhepatic vein and the lowest value of $h$ is at the wall of the left hepatic vein. This result can be explained by the fact that the blood velocity in the midhepatic vein is the highest (as shown in Table 2) and theory suggests that higher velocity should yield higher value of $h$.

For the experimental results from the portal veins, the highest value of $h$ was obtained from the wall of the main portal vein and the lowest value of $h$ was obtained from the wall of the left branch of the portal vein. These results suggest that the blood velocity in the main portal vein should be higher than the blood velocity in the left portal vein. However, the estimated blood velocity shown in Table 2 does not agree. This may be due to several reasons. First, the blood flow rates in the portal vein are roughly estimated (RAPPAPORT et al., 1963); hence they may not be very accurate. Second, the diameters of the portal veins were measured from CT scan images in a few pigs, thus they might not represent the appropriate data (STRIGEL, 2003). Finally, the blood velocities shown in Table 2 are from the calculation based on the blood flow rates and the diameters of the veins, if any of this information is incorrect, the calculated blood velocities would be incorrect.

Since the hepatic veins drain into the inferior vena cava, which is connected to the right atrium, the blood flow in the hepatic veins fluctuates with each heart beat and with ventilation. In contrast, the blood flow in the portal veins has less fluctuation and is not associated with the heart beat and ventilation.
V. CONCLUSION

We measured the values of $h$ in 8 locations in the large vessels of the liver in 8 pigs using a system similar to a hot wire/film anemometer. To obtain accurate results, fluoroscopy was used to locate the precise measuring locations in the liver with proper orientation of the catheter sensor. The sensor was frequently cleaned with water or saline during the measurements to remove any blood deposition in order to avoid adding any thermal resistance into the system.

For each $h$ measurement, we measured at least 30 s and used equation (1) to calculate the average value of $h$. At each measuring location, we performed at least 3 separate measurements (from different pigs) and calculated the median, average and the standard deviation of the experimental results. Most of the results obtained from measurement agree with theory, which suggests that the higher values of $h$ should be detected at the locations with the higher blood velocities.

ACKNOWLEDGEMENTS

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SATHIANPITAYAKUL, E. (2003), personal communication, Department of Internal Medicine, Mahidol University, Bangkok, Thailand

STRIGEL, R. (2003), personal communication, Department of Surgery, University of Wisconsin, Madison, USA.


Figure and Table Captions:

**Figure 1.** Flow chart of *in vivo* measurement where the catheter is located inside the pig’s large vessels of the liver.

**Figure 2.** The catheter sensor: (a) The steerable catheter has Loctite coating except where the thin film Pt sensor (RTD) at its tip is exposed, (b) Top view of the RTD sensor, and (c) Cross-sectional view of the RTD sensor through line C.

**Figure 3.** Circuit diagram of constant temperature measuring system, $R_1$, $R_2$, and $R_3$ are wire-wound resistors with 1% tolerance, ± 20 ppm/$^\circ$C temperature coefficient. $R_s$ is a resistive temperature detector.

**Figure 4.** Fluoroscopy images from the *in vivo* measurements of $h$: (a) at the main portal vein, (b) at the left branch of the portal vein, (c) at the right branch of the portal vein, (d) at the mid hepatic vein, (e) at the left hepatic vein, and (f) at the right hepatic vein.

**Figure 5.** The median values of $h$ from *in vivo* measurements in 8 pigs in 8 locations mapping on a liver diagram.

**Figure 6.** Waveforms of the voltage across the catheter sensor, $R_s$, detected during the *in vivo* measurement of $h$ using the Biobench data acquisition program: (a) in the middle hepatic vein, (b) in the main portal vein.
Table 1. Experimental results of $h$ in 8 locations in the large vessels of the liver from 8 pigs.

<table>
<thead>
<tr>
<th>Location</th>
<th># of data</th>
<th>Range (W/m²·K)</th>
<th>Average $h$ (W/m²·K)</th>
<th>Median $h$ (W/m²·K)</th>
<th>SD (W/m²·K)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Main portal vein (floating)</td>
<td>5</td>
<td>5000 – 6700</td>
<td>6100</td>
<td>6300</td>
<td>670</td>
</tr>
<tr>
<td>Main portal vein (wall)</td>
<td>5</td>
<td>2600 – 5700</td>
<td>3840</td>
<td>3400</td>
<td>1290</td>
</tr>
<tr>
<td>Right portal vein (wall)</td>
<td>3</td>
<td>3000 – 4000</td>
<td>3400</td>
<td>3200</td>
<td>530</td>
</tr>
<tr>
<td>Left portal vein (wall)</td>
<td>3</td>
<td>2700 – 3200</td>
<td>3000</td>
<td>3100</td>
<td>270</td>
</tr>
<tr>
<td>Mid hepatic vein (floating)</td>
<td>5</td>
<td>4500 – 9900</td>
<td>7180</td>
<td>7200</td>
<td>3090</td>
</tr>
<tr>
<td>Mid hepatic vein (wall)</td>
<td>4</td>
<td>5300 – 7000</td>
<td>6240</td>
<td>6400</td>
<td>690</td>
</tr>
<tr>
<td>Right hepatic vein (wall)</td>
<td>3</td>
<td>3600 – 5500</td>
<td>4830</td>
<td>5400</td>
<td>1070</td>
</tr>
<tr>
<td>Left hepatic vein (wall)</td>
<td>3</td>
<td>4100 – 4800</td>
<td>4400</td>
<td>4300</td>
<td>360</td>
</tr>
</tbody>
</table>

Table 2. The estimated blood velocities in the large vessels in the human liver.

<table>
<thead>
<tr>
<th>Location</th>
<th>Diameter range (mm)</th>
<th>Average diameter (mm)</th>
<th>Blood flow (L/min)</th>
<th>Estimated blood velocity (cm/s)</th>
<th>Peak Blood velocity (cm/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Main portal vein</td>
<td>10-30</td>
<td>14.5*</td>
<td>1200**</td>
<td>12.1</td>
<td>18* .23**</td>
</tr>
<tr>
<td>Right portal vein</td>
<td>8-15</td>
<td>10*</td>
<td>600**</td>
<td>12.7</td>
<td>N/A</td>
</tr>
<tr>
<td>Left portal vein</td>
<td>6-14</td>
<td>8*</td>
<td>600**</td>
<td>19.9</td>
<td>N/A</td>
</tr>
<tr>
<td>Mid hepatic vein</td>
<td>7-8</td>
<td>7.3***</td>
<td>500***</td>
<td>19.9</td>
<td>N/A</td>
</tr>
<tr>
<td>Right hepatic vein</td>
<td>N/A</td>
<td>8.7*</td>
<td>500***</td>
<td>14.0</td>
<td>N/A</td>
</tr>
<tr>
<td>Left hepatic vein</td>
<td>N/A</td>
<td>N/A</td>
<td>500***</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>


** RAPPAPORT, A. M. et al. (1993)


+ TOMIC, D. et al. (2001)

++ CIONI, G. et al. (1992)

Figure 1. Flow chart of \textit{in vivo} measurement where the catheter is located inside the pig’s large vessels of the liver.
Sensor area
Biocompatible epoxy (Loctite)
Steering control
Probe tip

(a)

Thin film Pt
2.0 mm
3.0 mm
1.9 mm

6.5 mm
9.4 mm
Line C

(b)

Glass cover
Thin film Pt layer
Ceramic substrate

0.5 mm
0.45 mm
27 μm

(c)
**Figure 2.** The catheter sensor: (a) The steerable catheter has Loctite coating except where the thin film Pt sensor (RTD) at its tip is exposed, (b) Top view of the RTD sensor, and (c) Cross-sectional view of the RTD sensor through line C.

![Catheter Sensor Diagram]

**Figure 3.** Circuit diagram of constant temperature measuring system, $R_1$, $R_2$, and $R_3$ are wire-wound resistors with 1% tolerance, ±20 ppm/°C temperature coefficient. $R_s$ is a resistive temperature detector.
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APPENDIX A

Correlation between the endocardial convective heat transfer coefficients \((h)\) and the heart rate

By comparing the results of the \(h\) obtained from each location in the cardiac chambers, I found some correlations between the \(h\) values and the heart rate. Fig. 1 to Fig. 11 show the correlation between the \(h\) values and the heart rate.

I. Floating position in the right atrium

![Graph showing correlation between \(h\) and heart rate]

\[ y = 57.832x + 619.93 \]

**Figure 1.** \(h\) at the floating position in the right atrium vs. heart rate.

For the floating position in the right atrium, 5 experimental results were obtained from 5 pigs. Fig. 1 shows the change of the \(h\) values with the change of the heart rate. For the test animal with
higher heart rate, the value of $h$ tends to be high. For the test animal with lower heart rate, the value of $h$ tends to be low.

II. At the wall of the right atrium

![Graph showing the relationship between heart rate and $h$](image)

**Figure 2.** $h$ at the wall of the right atrium vs. heart rate.

For the measurement at the wall of the right atrium, 8 experimental results were obtained from 8 pigs. Fig. 2 shows the change of the $h$ values with the change of the heart rate. For the test animal with higher heart rate, the value of $h$ tends to be high. For the test animal with lower heart rate, the value of $h$ tends to be low.
III. Floating position in the right ventricle

For the floating position in the right ventricle, 5 experimental results were obtained from 5 pigs. Fig. 3 shows the change of the $h$ values with the change of the heart rate. For the test animal with higher heart rate, the value of $h$ tends to be high. For the test animal with lower heart rate, the value of $h$ tends to be low.
IV. Floating position in the tricuspid valve

![Graph showing the relationship between heart rate and $h$ values.](image)

*y = 77.32x - 505.15*

**Figure 4.** $h$ at the floating position in the tricuspid valve vs. heart rate.

For the floating position in the tricuspid valve, 3 experimental results were obtained from 3 pigs. Fig. 4 shows the change of the $h$ values with the change of the heart rate. For the test animal with higher heart rate, the value of $h$ tends to be high. For the test animal with lower heart rate, the value of $h$ tends to be low.
V. Underneath the tricuspid valve

For the measurement underneath the tricuspid valve, 6 experimental results were obtained from 6 pigs. Fig. 5 shows the change of the $h$ values with the change of the heart rate. For the test animal with high heart rate, the value of $h$ tends to be low. For the test animal with low heart rate, the value of $h$ tends to be high.

**Figure 5.** $h$ underneath the tricuspid valve vs. heart rate.
VI. Floating position in the pulmonary valve

For the floating position in the pulmonary valve, 3 experimental results were obtained from 3 pigs. Fig. 6 shows a small change of the $h$ values with the change of the heart rate. The $h$ values measured at this location are almost constant at 8000 W/m$^2$·K with less than 15% difference.
VII. Floating position in the left atrium

For the floating position in the left atrium, 4 experimental results were obtained from 4 pigs. Fig. 7 shows the change of the $h$ values varies with the change of the heart rate. For the test animal with higher heart rate, the result of $h$ tends to be higher than those obtained from the test animal with lower heart rate.

**Figure 7.** $h$ floating in the left atrium vs. heart rate.
VIII. At the wall of the left atrium

For the measurement at the wall of the left atrium, 8 experimental results were obtained from 8 pigs. Fig. 8 shows the values of $h$ vary with the change of the heart rate. For the test animal with higher heart rate, the result of $h$ tends to be higher than those obtained from the test animal with lower heart rate.
IX. Floating in the left ventricle

For the floating position in the left ventricle, 4 experimental results were obtained from 4 pigs. Fig. 9 shows the values of $h$ vary dramatically with the change of the heart rate. For the test animal with higher heart rate, the result of $h$ tends to be higher than those obtained from the test animal with lower heart rate.

**Figure 9.** $h$ floating in the left ventricle vs. heart rate.
X. Floating in the mitral valve

**Figure 10.** $h$ floating in the mitral valve vs. heart rate.

For the floating position in the mitral valve, 3 experimental results were obtained from 3 pigs. Fig. 10 shows the values of $h$ vary with the change of the heart rate. For the test animal with higher heart rate, the result of $h$ tends to be higher than those obtained from the test animal with lower heart rate.
XI. Underneath in the mitral valve

![Graph showing the relationship between heart rate and h value with the equation y = 36.384x + 1190 drawn on it.]

**Figure 11.** $h$ underneath the mitral valve vs. heart rate.

For the measurement underneath the mitral valve, 5 experimental results were obtained from 5 pigs. Fig. 11 shows the change of the $h$ values varies with the change of the heart rate. For the test animal with higher heart rate, the result of $h$ tends to be higher than those obtained from the test animal with lower heart rate.
APPENDIX B

How to measure blood velocity using a hot film sensor system

Because this $h$ measurement system (explained in chapter 2, page 20) is adapted from a hot wire/film anemometer, it can also be used to measure blood velocity in the body. In order to do so, calibration using the flow system (Fig. 2 on page 38) will be needed.

To calibrate this system, approximately 12 L of blood is needed to create blood flow at various velocities in the flow system. Heparin or any anticoagulant agent will be needed to mix with the blood in order to prevent the blood from clotting.

The catheter sensor must be placed axially in the 20 mm tube where the flow has the highest velocity, which is approximately two times the average velocity (in laminar flow). The flow rig would be adjusted from 0 to 5 L/min to create the maximum velocity from 0 to 0.531 m/s. The electric power consumed by the heated sensor would be measured at various velocities. A plot of power consumed by the sensor vs. blood velocity would be created from this calibration and it would serve as a calibration graph for the blood velocity measurement in the body. To use this calibration graph, just input to the graph the electric power consumed by the heated sensor and read from the graph the blood velocity.