In vivo measurements of heat transfer on the endocardial surface

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Abstract. A catheter-based instrument was used to measure the heat transfer on the right atrial and ventricular endocardial surfaces of two pigs in vivo. The heat transfer parameters will assist calculating the proper dose for radio-frequency ablation. The time-constant of the device was 0.05 s. It was found that the average heat convection coefficient varies significantly both spatially and temporally on the endocardium. The average heat convection coefficients found were between 510 and 4800 W m$^{-2}$K$^{-1}$.

Keywords: heat convection, instrumentation, heat flux sensor, cardiac ablation, endocardium, in vivo.

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1. Introduction

Catheter ablation is a non-surgical technique used to destroy portions of myocardium and the conducting system in an attempt to ameliorate cardiac tachyarrhythmias. Since its introduction in the early 1990s, a number of studies has reported the results of radio-frequency catheter ablation (RFCA) in the treatment of atrioventricular nodal reentrant tachycardia, atrial fibrillation, accessory pathway mediated tachycardia, atrial flutter, atrial tachycardia and idiopathic ventricular tachycardia. In RFCA the physician maneuvers a catheter to the target heart tissue and delivers a burst of radio-frequency current between the ablating electrode in the heart and a ground electrode placed on the patient’s back. The electric current flow through the tissue causes resistive heating of a small volume of myocardium around the electrode. This heating destroys a region of the myocardium and causes a scar to form in that region. The scar tissue cannot transmit electrical impulses. As a result, the abnormal electrical pathway may no longer be able to generate arrhythmias. Thus, the catheter ablation procedure can permanently cure the patient.

In order to eliminate the abnormal electrical pathway, the procedure must cause a lesion of specific volume in the myocardium. The myocardial tissue temperature can be used to predict the lesion size. The problem is that the internal tissue temperature itself must be predicted since it cannot be measured directly without damaging the tissue. As a result, to predict the temperature inside of myocardium, an accurate heat transfer model is required. An important step toward the prediction of the myocardial temperature is the accurate knowledge of the heat convection coefficient ($h$) as boundary condition on the endocardial surface of the heart.

There are only a small number of reports of direct measurements of $h$ between the endocardium and the circulating flow available in the scientific literature, e.g., Bhavaruju [1] measured in vitro using a plastic model of the heart and a pulsatile pump. The lack of direct measurements is probably due its many technical difficulties. Swan-Ganz catheters have been used for several decades to measure cardiac output and more recently to measure ejection fraction. It is a very convenient apparatus for performing measurements in the heart because it is mechanically reliable and readily available. Dos Santos et al. [2] proposed and evaluated a method that can be used to measure in vivo $h$ using modified Swan-Ganz catheters. This paper presents in vivo measurements of $h$ on the endocardial surface in swine using such instrument.

2. RFCA

Several distinct heat transfer mechanisms occur during RFCA. The heat transfer in the myocardium and the amount of radio-frequency power delivered together ultimately determine the size of the lesion achieved by the procedure.

During RFCA, a catheter is introduced into a heart chamber via percutaneous peripheral venous or arterial conduits and placed in contact with the target ablation
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region at the endocardial surface. A current with a frequency between 300 kHz and 1 MHz is applied between the catheter electrode and a dispersive electrode attached to the patient’s skin. The myocardium is heated by Joule heating and heat is conducted inside the myocardium. A temperature of 50 °C or higher causes irreversible loss of cellular excitability of tissue and forms a lesion in the site [3].

During RFCA six heat transfer mechanisms exist:

1. conductive heat transfer in the tissue,
2. resistive heating of the blood and tissue,
3. convective heat loss to an epicardial coronary artery,
4. convective heat loss from the endocardium to circulating blood pool,
5. convective heat loss from the electrode to circulating blood pool,
6. and conductive heat loss along the catheter.

There are three convective mechanisms that occur during RFCA. Consequently, there are three distinct heat convection coefficients that can be determined. Due to the complexity of measuring $h$ and despite the fact that several convective losses happen at the same time, this work is specific in scope. The data presented here are measurements of the convective heat loss from the endocardium to the circulating blood pool (number 4). This is the parameter that has a significant impact on the size of the lesion [1].

3. Review of previous works

Several works have addressed the cardiac ablation procedure in the past years. It is difficult to experimentally verify cardiac ablation because of the many factors that must be controlled, such as the thermal effects of flow in the chambers. Consequently, simulations are widely used to provide better understanding of the underlying biophysics of the cardiac ablation. Those simulations require the use of $h$ as an important input parameter. Because of the lack of direct measurements of $h$ in vivo, several investigators have used rough estimations in their simulations.

Labonté [4] assumed a laminar flow over a flat surface and found a value in the order of 2000 W m$^{-2}$ K$^{-1}$. In another study Labonté [5] used much smaller values of $h$ (25 and 100 W m$^{-2}$ K$^{-1}$).

Shahidi and Savard [6] performed a finite element model for radio-frequency cardiac ablation. Their value for $h$ was 2089 W m$^{-2}$ K$^{-1}$ assuming that the blood flow is laminar.

Jain and Wolf [7] used $h$ to be 1800 W m$^{-2}$ K$^{-1}$ in their simulations.

Jain et al. [8] assumed fully developed laminar flow through a duct and estimated $h$ to be 1300 W m$^{-2}$ K$^{-1}$.

Cao et al. [9] experimentally showed that the flow velocity influences the lesion size during RFCA. However, in their experiments they did not measure $h$. The blood velocity used by them in the experiments is in the range found in the heart. However, the geometry they used in the experiment is different from the heart geometry.
Consequently, the range of \( h \) in their experimental apparatus could greatly differ from the range found in the heart chambers.

The parameters used in the simulations in the previous works are either based on estimations that do not seem realistic or are estimated from fluid velocity without adequate care about the geometry. The accuracy and applicability of any simulation ultimately depends on the parameters used. Consequently, the results and conclusions based on those experiments are only as accurate as the parameters used in the simulation.

Bhavaraju [1] performed \textit{in vitro} experiments and found higher values of \( h \) than the ones used in the simulations in previous works. The great virtue of this work is the fact that it provides the researchers with more realistic values of \( h \) in different sites of a ventricle. Those values can be used in analytical models allowing for a better qualitative and quantitative insights about the temperature profile along the myocardium volume. He found values ranging from 44 to 3930 W m\(^{-2}\) K\(^{-1}\). Smaller values were found below the mitral valve, while larger values were found in the middle of the left ventricle.

Bhavaraju’s work still has room for improvement due to three important limitations. First, he used a physical model of the heart made of silicone that cannot change size like a real heart. This should cause some error in the estimation of \( h \). Second, the used procedure is highly invasive and cannot be used \textit{in vivo}. Third, the value of \( h \) probably varies from individual to individual and certainly varies from position to position. Therefore, \textit{in vivo} measurements are required.

4. Fundamentals of the measurement of heat convection coefficient

Several methods have been used to measure \( h \) elsewhere on different surfaces at different conditions. A widely used method is based on the Joule effect dissipated by a heater at steady-state [10, 11, 12, 13, 14, 15, 16]. In our case, in order to measure the \( h \), a thermo resistive sensor is employed as both heating element and temperature sensor. The probe is placed in close contact with the endocardium and self-heated by electric power above the temperature of the circulating blood pool as illustrated in Figure 1. The power lost by convection can, then, be used to measure \( h \) on the endocardial surface.

In this configuration, the instrumentation has two modes of operation: the sensing and the heating mode. The probe can be switched from one mode to another mode anytime. In the sensing mode, the probe measures the surrounding temperature. In the heating mode, the electrical resistance of the probe and hence its volume-averaged temperature is kept constant by a feedback circuit like the one described in [2].

In order to measure \( h \), in principle one could use the so called Newton’s law of cooling (Equation 1),

\[
P = hA(T_s - T_f),
\]

where \( P \) is the power dissipated by the probe, which can be calculated from the voltage across its terminal and its resistance; \( T_s \) is the temperature at the surface of the sensor
that is ideally kept constant by the anemometer circuitry in the heating mode; $T_f$ is the fluid temperature that is measured in the sensing mode; and $A$ is the surface area in contact with the fluid.

Since the probe is meant to be used inside of the heart, a Swan-Ganz catheter was chosen as a suitable instrument to make the measurements. A Swan-Ganz catheter has a thermistor embedded close to its tip that can be used to measure $h$. However, one should be aware that the use of this instrument for the measurement of $h$ is complicated by the following factors:

1. part of the heat is conducted through the back of the sensor,  
2. the catheter surface temperature is not kept constant for different values of $h$,  
3. the catheter surface temperature is not the sensor average temperature,  
4. the sensor area is not easy to measure because the sensor is mounted in the catheter,  
5. finally, the relationship between the physical and effective area is not clear.

Consequently, the measured $h$ could largely differ from the correct value. A calibration technique to overcome these problems has been devised by dos Santos et al. [2]. The method consists in using empirical equations to predict $h$ on the wall in circular tubes. The predicted value is, then, used to calibrate the instrument. Besides the fact that the calibration environment closely matches the environment where the measurement will be performed, turbulent flow was chosen because of three reasons. First, it facilitates calibration because the values of $h$ can be easily controlled under this condition by simply changing the flow rate. Second, the fully developed velocity profile is attained for small tube lengths. Third, long-time and widely used empirical equations are available to evaluate $h$ under this condition. Hence, knowing the tube diameter and shape, the mean flow velocity and thermal properties of the water, the evaluation of the heat transfer coefficient from the wall to the circulating flow inside of
the tube is possible. This means that the sensor output can be calibrated versus the evaluated average heat convection coefficient \( h \).

Figure 2 shows a 1D model of the probe behavior on convective medium. In the picture, \( T_c \) is the core sensor temperature in the heating mode; and \( T_f \) is the fluid temperature measured by the sensor in the sensing mode and \( T_b \) is the temperature at the back of the sensor.

\[ h = a \times \left( \frac{P}{T_c - T_f} \right)^3 + b \times \left( \frac{P}{T_c - T_f} \right)^2 + c \times \left( \frac{P}{T_c - T_f} \right) + d, \]  

(2)

where \( a, b, c \) and \( d \) are calibration constants.

The accuracy of this curve was evaluated in a previous work \[2\]. The average error of full scale found for this instrument in the range tested was 7.4 %.

The time constant of the instrument in the heating mode was measured in still water. The time constant of the instrument is defined here as the time required for a thermistor to change its body temperature by 63.2% of a specific temperature span. The experiment to measure the instrument time-constant was performed as follows. The probe working in the heating mode was inserted in still water. The overheating ratio of the probe was step changed and readings were taken until they stabilized. The time constant found was 0.05 s. Figure 3 is a plot of the result of the experiment in order to measure the instrument time-constant.

5. Materials and methods

5.1. Animal preparation

In order to demonstrate the feasibility of the method for the \textit{in vivo} measurement of \( h \), the technique described previously was applied in swine. The pigs were obtained from
the Department of Animal Science at the University of Wisconsin. The protocol for these studies was approved by the Animal Care and Use Committee and was in compliance with all NIH guidelines for humane use of animal in research. The pigs were injected intramuscularly with Telazol®. When the animals were sedated, they were masked to a surgical plane of anesthesia with halothane at a 5% level. An electrosurgical unit was used to make a cut-down through the skin and underlying tissue to expose the trachea and sternum. The trachea was isolated by blunt dissection, and the animal incubated via a tracheostomy. The animal was placed on a ventilator. The level of oxygen saturation was between 90 and 100 and the heart rate between 100 and 120 bpm. Because no fluoroscope was used, the chest was opened and any bleeding from the incisions ligated or stopped by electrocautery. The chest was held open by a surgical retractor exposing the intact heart within the pericardium. One of the advantages of the open chest preparation is the exact position of the probe can be manipulated by direct palpitation.

5.2. Experimental protocol

Figure 4 shows the endocardial locations where \( h \) was measured. Positions A1 and B1 are on the anterior surface of the right atrium located midway between superior vena cava inlet and the tricuspid valve. Position A2 is on the antero-lateral surface of the right atrium, just above the tricuspid valve. Position B5 is on the anterior surface of the right ventricle just below the pulmonary valve. Position A3 and B3 are at the midpoint of the antero-lateral surface of the right ventricle. Position A4 and B4 are on the intraventricular septal surface of the right ventricle. The catheter was inserted through superior vena cava and placed against the endocardium in different parts of the right atrium (RA) and ventricle (RV). The catheter position was, then, interpreted by tactile sensation on the epicardium. The probe was assumed to be against the target tissue when there was mechanical resistance to its further insertion. However, no fluoroscopy
was used. *In vivo* measurements were performed in two pigs.

Initially, we took 10 s measurements to see the behavior of the power dissipated by the sensor (periodicity, magnitude and shape). Based on these observations, the following experimental protocol was applied.

1. The catheter was placed in the chosen position.
2. The instrument was switched to the sensing mode. The temperature readings were taken for 3 s at a sampling rate of 960 Hz.
3. The circuit was switched from sensing to heating mode and measurements were taken for 3 s at a sampling rate of 960 Hz.
4. The waveform of the heating mode was analyzed. If the waveform did not show pulsatile variation, it was assumed that the thermistor was facing the endocardium, therefore, the catheter was rotated and a new set of measurements was taken.

Several important statements are made here regarding the geometry of the problem. First, the cylindrical shape of the probe is similar to the shape of the trabeculae. Second, the catheter curvature ratio is similar to the trabecular curvature ratios. The measurement will be more accurate in the ventricle since the catheter shape matches the ventricular surface better than the atrial surface. However, even though the atrium is smooth, the instrument is still able to measure the relative importance of the heat convection at various sites of both chambers. Finally, the catheter is placed in close contact with the endocardial surface during the experiments. Thus, we believe that the error caused by the invasiveness of the probe is kept small. Figure 5 shows a close up view of the catheter laid down on the endocardial surface of the right ventricle.
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Figure 5. A close up view of the catheter tip adjacent to the RV endocardium. Note in (a) that the catheter has dimensions similar to the trabeculae. Consequently, the catheter lays down on the endocardial surface as part of the trabeculae themselves. This means that the catheter is in close contact with the endocardium. Also note in (b) that the catheter can reach various locations in the heart.

<table>
<thead>
<tr>
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</thead>
<tbody>
<tr>
<td>Average $h$ (W m$^{-2}$ K$^{-1}$)</td>
<td>511</td>
<td>2520</td>
<td>4500</td>
<td>2200</td>
</tr>
<tr>
<td>Standard deviation (W m$^{-2}$ K$^{-1}$)</td>
<td>214</td>
<td>494</td>
<td>477</td>
<td>61</td>
</tr>
<tr>
<td>Cardiac rate (bpm)</td>
<td>120</td>
<td>110</td>
<td>120</td>
<td>120</td>
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Table 1. Experimental results for the pig A

6. Results and discussion

The heat convection coefficient was measured in several locations for each pig. For each location, three sets of measurements were taken and the standard deviation of the average values were calculated. A hypothesis is assumed here. For a short period of time (up to 15 heart beats) there are no changes in $h$ since the temperature, heart rate, cardiac output and ejection fraction are unlikely to change in such a short period of time. This means that the standard deviation of the average values can be used as an assessment of the performance of the instrument.

Tables 1 and 2 give the measured convective coefficient values for the pig A and B, respectively. Since the probe-electronics time-constant is 0.05 s, it is possible to plot the real-time heat transfer. A close up view of the graphics are also shown in order to clarify the characteristics of the curves. In the graphics, DT stands for $T_s - T_f$ and the periodicity of the curves is indicated by alphanumeric symbols.

Figure 6 shows the waveform collected at position B1. As can be seen, in this position, the magnitude of the heat transfer has great variation. It can also be seen the periodicity of the flow as is pointed out in the graphic. In Figure 6a, it is possible to see that the power dissipated and, hence, $h$ greatly varies in time but this variation
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<table>
<thead>
<tr>
<th>Position</th>
<th>B1 (RA)</th>
<th>B5 (RV)</th>
<th>B3 (RV)</th>
<th>B4 (RV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average $h$ (W m$^{-2}$ K$^{-1}$)</td>
<td>1010</td>
<td>570</td>
<td>4890</td>
<td>1510</td>
</tr>
<tr>
<td>Standard deviation (W m$^{-2}$ K$^{-1}$)</td>
<td>360</td>
<td>170</td>
<td>381</td>
<td>107</td>
</tr>
<tr>
<td>Cardiac rate (bpm)</td>
<td>120</td>
<td>120</td>
<td>110</td>
<td>110</td>
</tr>
</tbody>
</table>

Table 2. Experimental results for the pig B

Figure 6. Measurement above the tricuspid valve of the pig B (position B1). The numbers are used to show the periodicity of the waveform.

is periodic. One can see that the peaks are 0.5 s apart. This is due the cardiac rate. However, the higher peaks happen every 3 s. This suggests that the respiration also has an important role in the dissipation of heat. Figure 6b is a close up view of this waveform. The peaks have an “M” shape that is a characteristic of the flow profile close to the tricuspid valve [17]. Figure 6b also shows that “2” is higher than “2a” as expected in a normal heart. In fact, the early inflow of blood reaches a peak at the point “2”. Flow, then, decelerates until atrial systole, at which time the right atrial pressure rises above the right ventricular pressure and flow again passes through the tricuspid valve [17].

Figure 7a shows the measurements taken at the position B2. Again, it shows that the power dissipated by the probe varies significantly in time through the cardiac cycle. However, it does not vary as much with the respiration as B1. Figure 7b shows a close up view of a peak. It also shows the M shape characteristics of the right ventricular inflow [17].

Measurements performed at B3 on the right ventricular free wall of the right ventricle are shown in Figure 8a. It shows the periodic nature of the power dissipated varying in time with the cardiac cycle. It also shows some variation probably due the respiration. Figure 8b is a zoom view of one of the peaks. Again, it is possible to see that the peaks follow a pattern.
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(a) Measurement showing the periodicity of the curve.

(b) Close up view of (a)

\textbf{Figure 7.} Measurement below the tricuspid valve of the pig B (position B2).

(a) Periodical nature of the measurements.

(b) Close up view of (a).

\textbf{Figure 8.} Measurement on the free wall at the right ventricle of the pig B (position B3).

Since the experiment was performed for several hours without major complications, e.g., fibrillation, the instrument can be used safely for research purposes. The heat transfer coefficient for regions of high blood flow are higher than the values used by researchers [4, 5, 6, 8] obtained by theoretical estimation. However, the value of $h$ can also be much smaller for regions of low flow rate. This difference can be explained by the fact that they used some assumptions in their estimations that seem unlikely to exist in the real heart. The results found here are also higher than the ones obtained by Bhavaraju [1], who used a silicone model of the heart. One reason for this might be because the model does not expand nor contract like a real heart. Despite the fact that the idea of this work was to measure $h$ on the endocardial surface and the instrument was calibrated accordingly, the results found here show that the instrument can measure the
instantaneous variation of the blood flow in real-time. In particular, this device could be used to measure the instantaneous variation of $h$.

7. Conclusion

This work presented the feasibility of a technique for the in vivo measurement of the average heat convection coefficient ($h$). An analysis of the data shows that the method can be used in an animal model. Measurements were taken in regions of low and high flow in the right atrium and in the right ventricle. The instrument was capable of detecting instantaneous flow variation and measure the average $h$ in those regions. An analysis of the data showed that the average coefficient is not time-dependent. Moreover, the results found were higher than the ones estimated by other investigators or by in vitro measurements.

The instrument was calibrated in vitro and tested for the measurement of average heat convection. The results suggest that the instrument can be further improved by performing transient calibrations for the measurement of instantaneous $h$. In addition, another feature could be the ability to relate the waveforms collected with the cardiac cycle, i.e., the pumping and emptying of the cardiac chambers using our instrument. For doing this, it would be necessary to record the ECG and blood pressure in the chambers. It should be noticed that the amplitude of the signal will be a function of the rotation angle of the probe. This is an important source of error. The signal becomes completely absent if the catheter faces the endocardium. The proper technique is to rotate the catheter until the amplitude is maximal.

The results obtained here suggest that: (1) the heat convection coefficient significantly varies from location to location on the endocardial surface; (2) $h$ varies significantly in time through the cardiac cycle; (3) respiration plays an important role on the dissipation of heat; (4) the instantaneous value of $h$ can be measured using a Swan-Ganz catheter.

Acknowledgment

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