Temperature Measurement within Myocardium During In Vitro RF Catheter Ablation

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Abstract—While most commercial ablation units and research systems can provide catheter tip temperature during ablation, they do not provide information about the temperature change inside the myocardium, which determines the lesion size. We present the details of a flow simulation and temperature measurement system, which allows the monitoring of the temperature change inside the myocardium during in vitro radio frequency (RF) cardiac catheter ablation at different blood flow rates to which the catheter site may be exposed. We set up a circulation system that simulated different blood flow rates of 0 to 5 L/min at 37 °C. We continuously measured the temperature at the catheter tip using the built-in thermistor and inside the myocardium using a three-thermocouple probe. The system provides a means for further study of the temperature inside myocardium during RF catheter ablation under different flow conditions and at different penetration depths.

Index Terms—Electrical interference, radiofrequency cardiac ablation, temperature measurement, thermistor, thermocouple.

I. INTRODUCTION

Radio-frequency (RF) catheter ablation of cardiac tissue is highly effective in the treatment of supraventricular cardiac arrhythmias, such as Wolff–Parkinson–White syndrome, atrioventricular (AV) node reentry, atrial tachycardia, and atrial flutter [1]. Several groups are also investigating it as a cure for ventricular tachycardia and atrial fibrillation [2], [3]. During RF cardiac ablation, a catheter is introduced into a heart chamber via percutaneous peripheral venous or arterial conduits and placed in contact with the target ablation 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 heat and heat conduction inside the myocardium and heat convection at the myocardial-blood and electrode-blood interfaces. Usually this heating process is modeled by the finite element method (FEM) or finite difference method due to the requirements for electric and thermal modeling and its complex geometry [5]–[9]. These numerical methods can provide information such as lesion dimension, delivered power, impedance, temperature distribution and temperature change during ablation. They are helpful in the analysis of RF ablation methods and the design of ablation catheters. But the numerical models still require in vitro and in vivo experiments to verify their correctness because of 1) lack of accurate myocardial electrical and thermal properties, 2) the complex heat convection at the endocardial surface, and 3) the simplified geometry of the model. These factors make it difficult to match the numerical results with experimental results.

Many research groups have used in vitro ablation models to study the lesion process during RF ablation with parameters such as lesion dimension, applied power, impedance, and catheter tip temperature [5]–[8]. The myocardial temperature profile is an important parameter for the study of the dynamic heating process during the RF ablation. It takes about 5 s for the catheter tip to reach the target temperature. Then the ablation generator regulates catheter tip temperature, applied power and impedance to be almost constant for the rest of the ablation. But inside the myocardium, the temperature increases gradually, depending on the distance from the catheter tip. It also takes more than 5 s to reach equilibrium. The myocardial temperature is not available unless thermal sensors are imbedded in the myocardium. Temperature change during ablation can help us to better understand lesion formation and improve efficacy. However, it is difficult to monitor the myocardial temperature because 1) There is strong RF interference around the catheter, 2) The distance between the catheter and sensor is difficult to determine because the myocardium is opaque.

Although some ablation systems measure the temperature of the catheter tip for monitoring and control of ablation [4]. Normally, with standard ablation systems in clinical use, RF power is delivered to the tissue up to 50 W for up to 120 s. In clinical practice, RF power may be delivered single or multiple times, at the same or different locations and with varying durations. The clinical end point is the elimination of an arrhythmia and, typically, there is little knowledge of the lesion size created to achieve this end.
we sought to measure the temperature change inside the myocardium during ablation. Labonté [6] used a thermographic camera to measure the surface temperature profile of a tissue-equivalent material sectioned at the end of ablation. It displayed the temperature distribution. But it was only a snapshot of the center cross section at the end of the ablation instead of temperature change during ablation. Kaouk et al. [7] measured the temperature change during microwave ablation in a transparent tissue equivalent material using a fluoroptic thermometer. The microwave probe was immersed in the phantom and no convection was introduced during its measurement. Nakagawa et al. measured the tissue temperature using two separate fluoroptic thermal probes during ablation on canine thigh tissue using a saline irrigated catheter [10]. The two probes were inserted 3.5 mm and 7.0 mm deep into tissue adjacent to the catheter electrode. They were far away from the directly heated rim and deep in the tissue. Note that normal lesions are about 5 mm to 8 mm deep and the fluoroptic sensors were close to that boundary. Once the probes are inserted into tissue, it is difficult to judge the distance between the catheter tip and sensor from outside the tissue. Also, the fluoroptic probe has a relative longer response time (~1 s) and larger dimension (0.75-mm diameter for a single probe). Jain reported temperature mapping of in vitro catheter ablation using thermocouples [11]. He estimated sensor positions by marks on the thermocouples, which introduced unknown error in measuring the distance between the catheter electrode and the thermocouples. Hence, the temperature recordings have standard deviations as high as 13 °C.

We designed a probe containing three thermocouples, which could be imbedded in the myocardium and reproducibly measure the tissue temperature at given distances from the RF source during application of RF current. We also designed an adjustable flow system to simulate the blood flow at various locations within the heart chamber. The system provides us means for further study of RF catheter ablation.

II. MEASUREMENT SYSTEM

A. System Setup

To study the myocardial temperature change during RF catheter ablation, we record both the temperature at the catheter tip and temperature inside the tissue for comparison. Fig. 1 shows a block diagram of the measurement system. It contains a probe of three thermocouples inside the myocardial tissue and a thermistor at the catheter tip. The sensor signals are amplified and filtered by low-pass filters to minimize the 60 Hz power line and 500 kHz RF interference. A 12-bit analog-to-digital converter (ADC) (DI-220 by Data Instruments, Akron, OH) samples the data at ten samples/s for each channel. A LabVIEW program collects the sampled data from four channels (one for the thermistor and three for the thermocouples) and saves data to storage. It converts the voltage to temperature using a calibration curve for each channel and displays the temperature changes in real-time on a Dell Optixel 200-MHz Pentium personal computer.

Fig. 2 shows the circuit diagram for thermocouple amplifiers. The amplifier has a gain of 2500 and the output is about 120 mV/°C. The signal is shifted so that the amplified signal is between the ±5 V range of the ADC for the temperature range from 25 °C to 105 °C.

During ablation, the thermocouples are exposed to very strong 500 kHz RF ablation voltage and pick up the RF interference signals and power line interference. We reduce the RF and power line interference as follows.

1) The circuit has low-pass filters positioned prior to the amplifier stage, in the amplifier stage and in the output stage to minimize the RF and power line interference from entering the circuit. The three first-order low-pass filters have corner frequencies of 37, 10, and 37 Hz, respectively.

2) The circuits and battery power supply are put in a grounded shielding box to shield from the 60-Hz power line and RF interference [12].
3) Shields around the thermocouple extension cables are connected to the shielding box.
4) The ground lines are placed carefully to form a star network to avoid forming loops and interference signals travelling in the loop [12].

B. Thermistor Measurement and Calibration

The Blazer II catheter (EP Technologies, San Jose, CA) used in our in vitro experiment has a thermistor inside the catheter electrode. The EPT-1000XP ablation unit monitors the catheter tip temperature through the thermistor and controls delivered power during the temperature-controlled ablation. The EPT-1000XP ablation controller displays impedance, catheter tip temperature, applied voltage, applied current and applied power during ablation. But it does not display information after the ablation is over. Fortunately, the ablation unit always applies a current to the thermistor measurement circuit and the thermistor is floating with respect to ground. We tap two wires on the two ends of the thermistor and feed the thermistor voltage to the ADC for sampling as shown in Fig. 1. There is also strong 500 kHz RF interference at the two ends of the thermistor during ablation (peak value about 1 V). A low-pass filter, with corner frequency at 180 Hz, is placed before the ADC to minimize the RF interference. As the ADC has a very high input impedance (100 MΩ), it does not affect either the measurement system inside the ablation unit or the performance of the ablation unit.

We measured the voltage versus temperature of the thermistor and used third-order polynomial curve fitting to obtain a calibration curve for the thermistor such as

\[ T = aV^3 + bV^2 + cV + d \]  

where \( T \) is temperature (°C) and \( V \) is the voltage at the two ends of the thermistor. We did not use the Steinhart–Hart equation for calibrating the thermistor because from the data we deduce there may be another resistor inside the generation unit in parallel with the thermistor to linearize the characteristics of the thermistor. Experiment shows that a third-order polynomial curve fits the voltage and temperature very well, with a standard deviation of 0.12 °C between the range from 25°C to 95°C.

C. Thermocouple Probe and Calibration

There are many sensors available for temperature measurement, such as the thermistor, fiber optic sensor, and thermocouple. We did not select the fiber optic sensor because of its large size (0.75 mm for a single fiber probe and 0.9 mm for a four-fiber probe) and low response time (0.25 s for a single fiber probe and 1 s for a four-fiber probe). We want the sensor as small as possible so that it least disturbs the electric field and thermal transfer inside the tissue. The temperature probe should also have a small time constant so that it can respond to the rapid temperature rise at the beginning of the RF ablation. The thermocouple is smaller than other thermal sensors and has a faster response time than other sensors. We use IT-21 copper/constantan T-type thermocouples (Physitemp Instrument Inc., Clifton, NJ) with a time constant of 0.08 s. The thermocouple is small, 0.23 mm in diameter without Teflon coating and 0.41 mm with Teflon coating. The Teflon coating, as an insulator, minimizes electrical and thermal disturbance inside the tissue.

As thermocouples are soft and easy to bend in the myocardium, the multithermocouple probe uses a silver wire 10 mm long as a shaft for support. The bare silver is 0.25 mm in diameter and with Teflon coating, its diameter is 0.33 mm. It is strong enough to avoid bending during introduction. Three thermocouples surround the silver wire to form a bundle as shown in Fig. 3. They are bonded together by Super 77 spray Adhesive (3M, St. Paul, MN). The distances between the thermocouple tips are measured under magnification. In our experiment, the distances between the probe tip and thermocouple tip are 0.8, 2.0, and 2.9 mm, respectively. The thermocouple probe has a diameter of 1.5 mm and can be inserted into a 14-Ga needle. Fig. 4 shows the procedure to introduce the thermocouple probe into the tissue.

1) We first use a 14-Ga shielded I.V. Catheter to punch a hole from the bottom of the tissue block through to the surface.
2) We remove the metal needle and leave the plastic needle in the tissue. Then we insert the thermocouple probe into the plastic needle and introduce it to the surface of the tissue block.
3) The plastic needle is pulled back and only the thermocouple is left in the tissue. After mounting the tissue to the plastic frame, we push the probe top back with a flat aluminum block so that it is just at the surface of the tissue. The probe tip is visible for us to adjust the catheter on top of it.

4) We align the catheter and thermocouple. Then, we lower down the catheter using the depth meter.

We calibrate the thermocouple circuits and use second-order polynomial curve fitting to obtain the relationship between the output voltage and temperature as

$$T = aV^2 + bV + c$$  \hspace{1cm} (2)

where $T$ is temperature ($^\circ$C) and $V$ is the output voltage. Each channel has its own fitting curve with a specific thermocouple.

III. FLOW SYSTEM SETUP

We set up a circulation system to simulate blood flow inside the heart chamber for RF ablation. On the surface of the tricuspid valve and mitral valve the endocardium experiences the total blood flow of cardiac output, which is about 5 L/min. But right underneath these valves, there is little blood flow and just local turbulence. To simulate the different flow conditions inside the heart chamber at different locations, we set up a circulation system (Fig. 5) to carry out in vitro ablation on bovine myocardium. It consists of a 12 L water bath (Model 180, Precision Scientific, Winchester, VA) to maintain a constant temperature of saline, and a transparent plastic container ($30 \times 20 \times 18$ cm) for flow simulation. The 400-W water bath is able to keep all the saline in the circulation system (both the water bath and plastic container) at $37 \pm 1^\circ$C. By regulating a flow meter (7200 series, King Instrument, Huntington Beach, CA) we adjust the flow from 0 to 6 L/min to simulate the blood flow condition at different locations inside the heart chamber. The hose injecting saline to the tissue has an 18-mm diameter, which corresponds to 250 mm² area and is comparable to the areas of the mitral and tricuspid valves. We use 0.5% saline to approximate the electric conductivity of blood inside the heart chambers, which is about 150 $\mu$S/cm.

A Plexiglas frame is attached to the bottom of the plastic container. A block of bovine myocardium ($30 \times 30 \times 15$ mm) sits on the crossbar (2 mm width) of the plastic frame. Two sidebars of 5 mm height hold the bovine myocardium as a clamp to avoid buoyancy and fix the position of the myocardium. The base of the frame is hollow except for the crossbar so that electric current can pass through the holes to the dispersive electrode at the bottom of the container. Another plastic holder attaches to the base and holds the injecting hose with adjustable height. During the experiment, the bottom of the injecting hose aperture and the surface of myocardium are at the same level. The distance from the hose to the catheter is 18 mm, which is comparable to the distance between ablation sites and the mitral valve. The in vitro heart wall moves and changes thickness during a cardiac cycle. The catheter also moves with the beating heart. Hence, the catheter electrode and lesion spot remain relatively stationary during the cardiac cycle. To simulate the moving heart and change of thickness is not essential for in vitro study and it is too complex to design such a system. We also do not implement perfusion because the perfusion in the myocardium is so small that the heat loss due to perfusion in RF ablation is negligible [6].

We use an EPT-1000XP (EP Technologies, San Jose, CA) ablation unit to carry out in vitro ablation. The 7-Fr catheter electrode has a 2.6 mm diameter and 4 mm length. It also has a thermistor attached to the inner surface at the tip of the catheter electrode to measure the temperature at the catheter tip. The catheter contacts the myocardium at a normal angle. The dispersive electrode is fixed at the bottom of the plastic container and is about 10 cm away from the catheter tip to approximate the real surgical situation. The catheter is housed in a plastic tube, which is attached to a depth meter. The depth meter is formed of a micrometer, which advances the electrode with a resolution of 0.001 in (0.0254 mm).

We introduce the thermocouple probe into the myocardium as described in Fig. 4. As the catheter moves down into the tissue, it pushes the thermocouple back at the same time. But the relative position of the thermocouple probe and catheter electrode remains the same over a range of several millimeters.

IV. RESULTS AND DISCUSSION

A. Calibration of the Measurement System

A VWR 100 A digital thermometer (VWR Scientific, West Chester, PA) was used as the temperature standard, which has an accuracy of $\pm 0.2^\circ$C and resolution of 0.1 $^\circ$C. We bundled the thermometer, catheter and thermocouple probe together and immersed them in a stirred constant temperature water bath.

We calibrated the measurement at nine temperature points ($25^\circ$C, $30^\circ$C, $35^\circ$C, $40^\circ$C, $50^\circ$C, $60^\circ$C, $70^\circ$C, $80^\circ$C, $90^\circ$C, and $95^\circ$C). At boiling temperature ($\approx 100^\circ$C), the voltage readouts are not stable because of the turbulence at this temperature. So beyond...
95 °C, we use curve fitting to extrapolate the temperature. The standard deviation of temperature is calculated as

$$\sigma_T = \frac{1}{N} \sum (f(V) - T)^2$$

(3)

where

- $f(V)$: temperature calculated from the calibration curve;
- $T$: temperature measured by the digital thermometer;
- $N$: number of calibration points (nine here);
- $\sigma_T$: standard deviation of $T$.

The standard deviations are 0.12 °C for the thermistor and 0.09 °C, 0.10 °C, and 0.07 °C for the thermocouples. Or the full-scale errors are 0.17% for thermistor and 0.13%, 0.14%, and 0.1% for the thermocouples.

### B. System Response Time

The thermocouple itself has a time constant of 0.08 s. But the glue used to bundle the thermocouples and the low-pass filters in the circuit increase the time constant. To measure the time constant, we dip the thermocouple probe from room temperature (22 °C) into 70 °C water. The time constants (63% of the change) for three thermocouples are 0.25, 0.28, and 0.3 s. Also the time constant for the catheter thermistor is 0.4 s. The measurement system’s time constant is small to respond to the fast temperature rise at the beginning of ablation.

### C. Temperature Measurement of In Vitro Ablation

With the current system setup, we carried out in vitro catheter ablation on bovine heart tissue obtained from a local grocery store. The 0.5% saline was kept at 37 °C ± 1 °C by the water bath during ablation. We used temperature-controlled ablation with a preset target tip temperature of 60 °C, maximum power of 50 W, catheter insertion depth of 4 mm and ablation duration of 60 s at a flow rate of 3 L/min. The 4 mm insertion depth corresponds to 20 g force on the catheter. During ablation, we recorded the temperatures of one thermistor at the catheter tip and three thermocouples located inside the tissue.

Fig. 6 shows that the catheter tip temperature (T0) rises rapidly at the beginning. It approaches the target temperature (60 °C) in about 2 to 3 s. Then the ablation unit adjusts the delivered power to maintain T0 at the target temperature and T0 fluctuates around the target temperature. When the ablation stops, T0 drops at a faster rate because of convection at the electrode and myocardial surface.

The thermocouple temperatures (TC1, TC2, and TC3) rise more slowly than T0. They are inside the tissue and it takes time for heat to transfer from the hotter rim around the catheter into the tissue. TC1 is even higher than the catheter temperature T0 under high flow conditions, which is predicted by FEM models [8], [9]. This suggests that the tip temperature is not always higher than the tissue temperature. Operators must use caution in interpreting catheter tip temperature for monitoring to avoid charring during surgery.

### D. Reproducibility

To measure the reproducibility of the measurement system, we performed 6 ablations on the same spot without moving the thermocouple probe. The ablation setting is the same as above. We assumed the first ablation killed the tissue. The next five ablations should be the same since the myocardium was already ablated and it should not change during ablation. We use two parameters to evaluate temperature change of the lesion 1) $t_{50}$-time for the thermocouple to reach 50 °C, which represents the dynamic temperature change, 2) $T_m$: the maximum temperature of thermocouple at the end of ablation. Table I shows the distributions of these two parameters are quite narrow when we do not move the thermocouple probe. The deviation is mainly due to variation of the ablation control unit and the turbulence of the flow instead of the error of the temperature measurement system.

We also performed 5 ablations on different blocks of myocardium with the same ablation setting. Table I also shows the statistical distributions of $t_{50}$ and $T_m$ are slightly wider compared with the first case. This wider deviation is mostly due to the differences among the ablations, such as catheter contact, myocardial properties, tissue alignment and probe insertion. The ratio of standard deviation to mean is more significant for $t_{50}$ since it represents the dynamic change when the myocardial temperature rises. The steady state temperature $T_m$ is not as sensitive as the dynamic change.

### V. Conclusion

We built a system to measure the temperature change at the catheter tip and inside the myocardial tissue during RF ablation at different simulated flow rates. The results show that the system is capable of measuring the myocardial temperature during RF catheter ablation. It provides us a method for further study of RF catheter ablation.

![Fig. 6. Temperature measurement during temperature controlled catheter ablation with these settings: maximum allowed power 50 W, target temperature 60 °C and ablation duration 60 s. T0 is the thermistor temperature at the catheter tip. TC1, TC2, and TC3 are thermocouple temperatures at 0.8, 2.0, and 2.9 mm from the catheter tip.](image-url)
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