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Quantification of Local Convectional Cooling During Cardiac Radiofrequency Catheter Ablation

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Abstract- **Radiofrequency (RF) catheter ablation is an effective, minimally invasive treatment method in clinical use for treatment of different cardiac arrhythmia. Studies have shown that lesion dimensions strongly depend on blood flow mediated convective cooling at the ablation site. We present a simple method to quantify convective cooling. A brief pulse of RF energy (10 W for 5 s) is applied, and catheter tip temperature is measured during and after energy application. Two parameters are extracted: 1) maximum tip temperature increase, and 2) slope of temperature decay 8ºC above initial temperature. We tested whether these parameters can quantify convective cooling in *ex vivo* experiments. A RF catheter was inserted into a tissue phantom placed in a saline bath. Flow at different rates of 0, 1, 2 and 3 L/min was injected towards the catheter, and the parameters were extracted. Both parameters correlated with flow rate. Slope of temperature decay showed linear dependence on flow rate, maximum temperature increase showed exponential dependence. The parameters are potentially useful in quantifying convective cooling before ablation to predict lesion dimensions.**

***Keywords* - radiofrequency catheter ablation; RF ablation; cardiac ablation; cardiac arrhythmia**

I. INTRODUCTION

Radiofrequency (RF) catheter ablation has become a standard clinical treatment for different types cardiac arrhythmia such as atrial tachycardia, atrial flutter, and some forms of ventricular tachycardia, due to it's efficacy and minimally invasiveness. A catheter is steered to the treatment site in the heart guided by fluoroscopy, and RF energy is applied to the tissue. Tissue near the catheter electrode is heated by resistive heating, resulting in irreversible tissue damage above ~50 ºC [1]. In many catheter types, catheter tip temperature is measured by a temperature sensor located inside the tip. Applied RF energy is controlled to keep the tip temperature at typically 60 – 80 ºC (maximum tissue temperature is higher than the measured tip temperature).

During RF ablation, heat is carried away by several different mechanisms (see Figure 1). Part of the heat energy is conducted into the myocardium resulting in tissue temperature elevation. In the myocardium, some heat is lost from blood perfusion. At the endocardium, convective cooling resulting from blood flow carries heat away from the tissue surface. Some heat is conducted through the catheter electrode to the electrode surface that is in direct contact with the blood, where it is also carried away by convection.

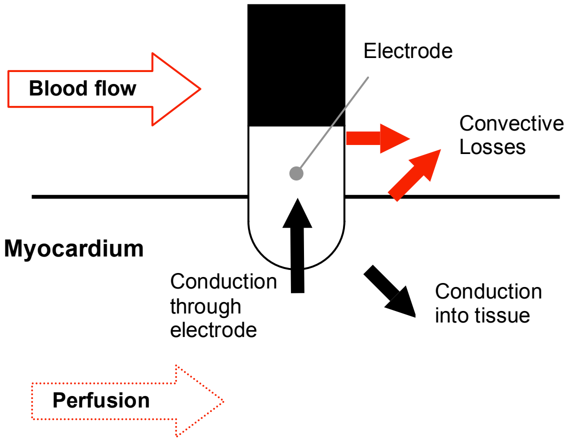

Fig.1. During RF ablation, heat is removed by different mechanisms.

Several studies have shown that lesion size (lesion = zone of tissue damage) is strongly dependent on local blood flow inside the heart [2-5]. Higher convective cooling results in larger and deeper RF lesions, since heat is carried away from the catheter and tissue surface, allowing more energy to be deposited deeper into the tissue. It has been shown in *ex vivo* experiments and in computer models, that higher flow rate results in wider and deeper lesions [2, 3, 5, 6]. Similarly, *in vivo* experiments in a porcine model have shown that lesion size depends on ablation site in the heart. Mukherjee et al. classified ablation sites in the heart according to flow condition (low/medium/high) and found significant differences in RF lesion dimensions [4]; Peterson et al. also found significant differences of RF lesion dimensions depending on ablation site [5]. A recent publication presents treatment planning software for cardiac RF ablation, estimating RF lesion dimensions depending on different parameters such as target temperature, ablation time, and blood flow [6]. However, they only used estimates for convective cooling since no detailed data are available; furthermore it can be expected that blood flow in the heart and related cooling is different between individuals, depending on size of the heart among other factors.

In this paper we present a simple method to quantify convective cooling at the ablation site before RF ablation is carried out. This information can potentially be used as input parameter in computer models to obtain more accurate estimates of lesion dimensions.

II. METHODOLOGY

We created tissue phantoms with same electrical conductivity as myocardium at 500 kHz and 37 ºC (0.54 S/m), by mixing 5% Agar-Agar and 0.25% NaCl with deionized water; this phantom has same thermal properties as water, similar to those of tissue. We placed the phantom (10 x 10 x 2 cm) inside a frame that held the phantom from two sides to avoid alteration of RF current flow. We immersed the phantom in 0.3% saline, having same conductivity as blood at 500 kHz and 37 ºC (0.67 S/m). We built a flow rig (see Figure 2) that allows adjustment of the flow rate between 0 and 12 L/min. A flow meter (Model 7200, King Instrument) regulates the flow rate. Saline is injected through a rigid tube (20 mm inner diameter, Polyethylene) with a cross sectional area comparable to mitral and tricuspid valves into the water bath. The bottom of the injecting tube aperture and phantom surface are at the same level. The distance between tube outlet and catheter is 30 mm. We measured maximum flow velocity at different flow rates by injecting a droplet of dye at the inlet of the rigid tube, and recording the wavefront at the phantom surface with a digital camera at 30 frames/s. Maximum flow velocity was 15.5 cm/s at 3 L/min flow rate. This velocity is comparable to blood flow velocities inside the beating heart as measured by an ultrasound Doppler transducer [3].

We used a commercial cardiac ablation generator (Boston Scientific, EPT-1000 XP), and an ablation catheter with electrode 4 mm long, and 6 FR (2 mm) diameter (Boston Scientific, Blazer II). The ablation catheter was inserted 2 mm deep perpendicular into the tissue phantom. We used a piece of alminum foil (~10 x 15 cm) placed at ~30 cm from the catheter as ground electrode. Experiments were carried out between 35 ºC – 40 ºC saline temperature, starting at 40 ºC.

We confirmed that phantom temperature was equilibrated with saline temperature before beginning of experiments. Total saline volume was ~20 L, and due to the slow cooling rate (2 hours from 40 ºC to 35 ºC), phantom and saline temperatures were always within 0.2 ºC. We describe later in the text how we correct for different initial temperatures of phantom and saline in our experiments.

We applied 10 W of RF power for 5 s to the ablation catheter, and measured catheter tip temperature for 60 s following start of power application. We performed 6 experiments each at 0, 1, 2 and 3 L/min flow rate. Each experiment was carried out at a different location on the phantom, at least 2 cm distant from the previous position to prevent contamination.

From the temperature time course of each trial we determined two parameters:

- Maximum change in tip temperature ($T_{\max}$-T_{ini})
- Slope of temperature decay at 8 ºC over initial temperature (dT/dt ($T = T_{\mathrm{ini}} + 8$ºC)).

The initial temperature T_{ini} is the temperature of saline and phantom at the beginning of each trial.

We assume that a significant amount of heat energy is removed by convection, and that by measurement of these parameters we can quantify convective cooling.

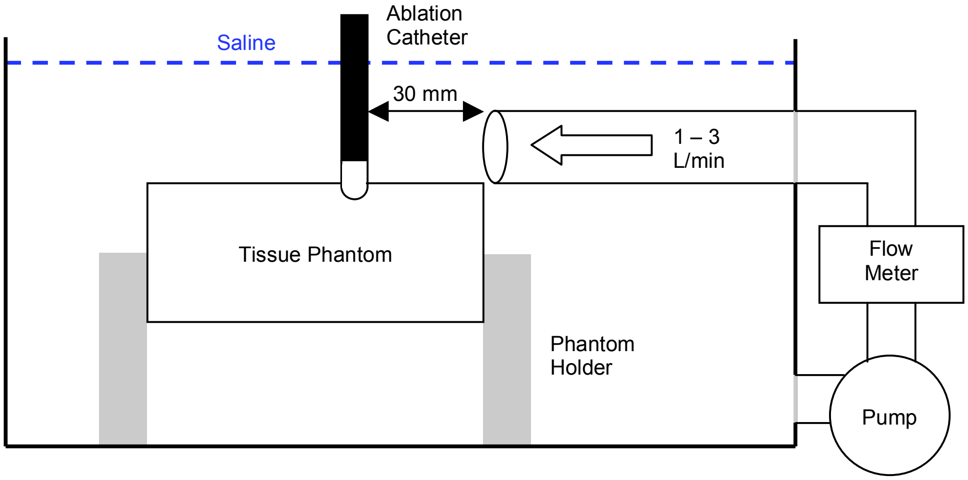

Fig. 2. Schematics of experimental setup.

Correction for different initial temperatures

For a potential clinical application it is of importance that the measured parameters are independent of body temperature (i.e. initial saline and phantom temperature in our experiments). Otherwise, different body temperatures will constitute an error source.

The heat transfer equation describing our experimental setup is:

$$\rho c \frac{\partial T}{\partial t} = \nabla \cdot k \nabla T + Q_{RF} \qquad (1)$$

where:

T	phantom temperature (ºC)
ρ	phantom density (kg / m^3)
c	phantom specific heat (J/(kg K))
k	phantom thermal conductivity (W/(m K))
Q_{RF}	power density deposited by RF (W/m^3)

From equation (1) we see that:

- Application of a certain power to the phantom always results in the same temperature elevation, independent of initial temperature
- Thermal conduction in the phantom (1st right-hand side term) is only dependent on the temperature gradient, but independent of absolute temperature

The amount of convective cooling depends both on flow rate, and temperature of the liquid (i.e. saline). The heat flux removed from a surface by convection is:

$$Q_{\mathrm{conv}} = h\,(T - T_{\mathrm{L}}) \qquad (2)$$

Where:

h	convective heat transfer coefficient (W/(m^2 K))
T	surface temperature
T_{L}	temperature of liquid (saline)

The heat transfer coefficient h depends on flow velocity, and flow profile. From equation (1) we saw that the relative temperature elevation everywhere in the phantom is independent of absolute initial temperature. In (2) we note that flow conditions (i.e. flow rate), and difference between surface and fluid (=initial) temperatures determine convective cooling. From above we see that this temperature difference is independent on initial temperature.

By measuring parameters relative to initial temperature ((T_{max}-T_{ini}), and dT/dt (T = T_{ini} + 8ºC)) we assure independence of our parameters on initial (=fluid) temperature.

We fit the relation between the parameters and flow rate to 2nd order polynomial expressions. We performed student's t-tests to check whether there is significant difference of the two parameters between different flow rates. We tested between 0 and 1 L/min, 1 and 2 L/min, 2 and 3L/min.

III. RESULTS

Figure 3 shows typical time-temperature relationships for flow rates of 1 and 3 L/min.

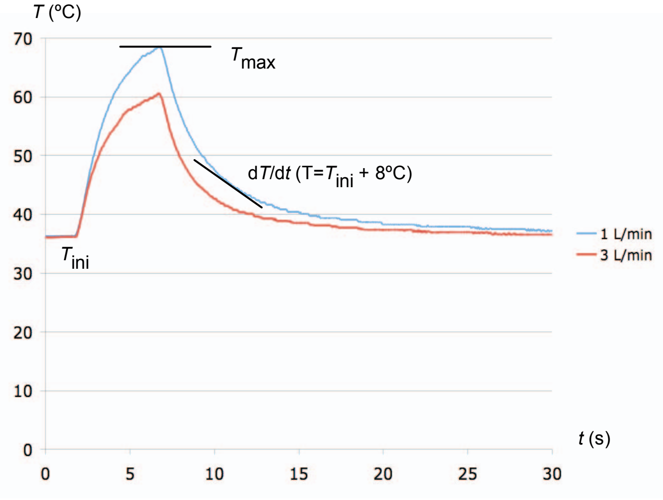

Fig. 3. Time course of catheter tip temperature for flow rates of 1 and 3 L/min. Measured parameters are indicated.

Figures 4 and 5 show dependence of the two parameters on flow rate. For each flow rate, results of the six trials, average, and 2nd order polynomial approximation are shown. In Figure 5, the 2nd order term was negligible resulting in a linear approximation. There was significant difference ($p < 0.05$) of both parameters between different flow conditions.

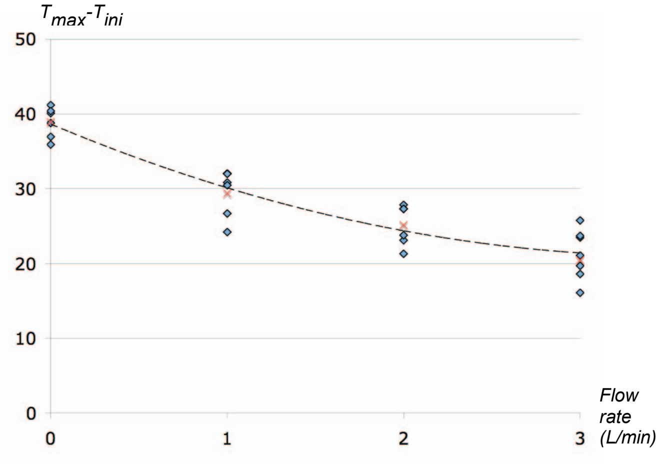

Fig. 4. Parameter ($T_{\mathrm{max}} - T_{\mathrm{ini}}$) at different flow rates. 2nd order polynomial approximation (line) and averages (red crosses) are shown in graph.

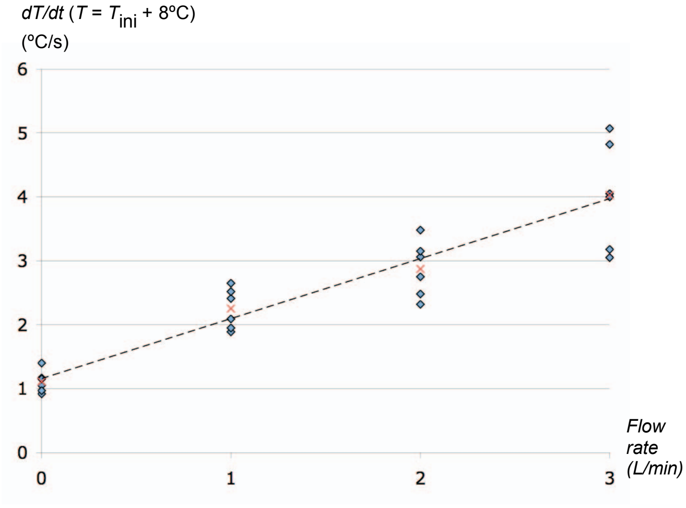

Fig. 5. Parameter (dT/dt ($T = T_{\mathrm{ini}}$ + 8ºC)) at different flow rates. Linear approximation (line) and averages (red crosses) are shown in graph.

IV. DISCUSSION

Both parameters showed significant dependence on flow rate. Temperature decay showed higher sensitivity to flow rate compared to maximum temperature increase. The maximum temperature increase showed an exponential dependence on flow rate, whereas the temperature decay showed linear dependence. However, flow rate is not necessarily related linearly with the associated convective heat transfer coefficient h, i.e. twice the flow rate does not result in h being twice as large (see equation (2)). To use the information provided from the measured parameters in computer models [6], the relation between the parameters and the heat transfer coefficient has to be established (see Figure 6). Another way to use the parameters would be to investigate the relationship between lesion dimensions and parameters directly.

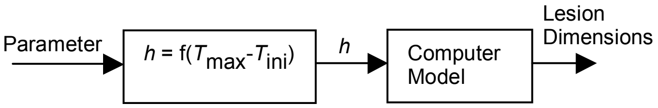

Fig. 6. Convective heat transfer coefficient h is deducted from one of the measured parameters, and used in a bioheat transfer model to estimate lesion dimensions.

Limitations

In the experimental setup we used constant flow; it should be noted that pulsatile flow will make measurement of the parameters more difficult, and may require signal processing or modification of the parameters. In our setup the catheter was inserted at constant depth, perpendicular to the phantom surface. Insertion depth and angle vary in clinical practice, and likely have a significant impact on both lesion dimensions, and measured parameters. One possibility is to add a second thermal sensor near the corner of the electrode may provide valuable additional information since that sensor would directly be in contact with blood and could allow measurement of convective cooling independent on insertion depth and angle of the catheter.

V. CONCLUSION

We present a simple method that allows quantification of convective cooling at the ablation site before treatment. This information is potentially useful in treatment planning.

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