Inverse Relationship Between Electrode Size and Lesion Size During Radiofrequency Ablation With Active Electrode Cooling

Hiroshi Nakagawa, MD, PhD; Fred H.M. Wittkampf, PhD; William S. Yamanashi, PhD; Jan V. Pitha, MD, PhD; Shinobu Imai, MD; Barclay Campbell, BS; Mauricio Arruda, MD; Ralph Lazzara, MD; Warren M. Jackman, MD

Background—Clinical efficacy has driven the use of larger electrodes (7F, length ≥4 mm) for radiofrequency ablation, which reduces electrogram resolution and causes variability in tissue contact depending on electrode orientation. With active cooling, ablation electrode size may be reduced. The purpose of this study was to examine the effect of electrode length on tissue temperature and lesion size with saline irrigation used for active cooling.

Methods and Results—In 11 anesthetized dogs, the thigh muscle was exposed and bathed with heparinized canine blood. A 7F ablation catheter with a 2- or 5-mm irrigated tip electrode was positioned perpendicular or parallel to the thigh muscle. Radiofrequency current was delivered at constant voltage (50 V) for 30 seconds during saline irrigation (20 mL/min) to 148 sites. Tissue temperature at depths of 3.5 and 7 mm and lesion size were measured. In the perpendicular electrode-tissue orientation, radiofrequency applications at 50 V with the 2-mm electrode compared with the 5-mm electrode resulted in lower power at 50 V (26 versus 36 W) but higher tissue temperatures, larger lesion depth (8.0 versus 5.4 mm), and greater diameter (12.4 mm versus 8.4 mm). Also, in the parallel orientation, overall power was lower with the 2-mm electrode (25 versus 33 W), but tissue temperatures were higher and lesions were deeper (7.3 versus 6.9 mm). Lesion diameter was similar (11.1 versus 11.3 mm) for both electrodes.

Conclusions—The smaller electrode resulted in transmission of a greater fraction of the radiofrequency power to the tissue and resulted in higher tissue temperature, larger lesions, and lower dependency of lesion size on the electrode orientation. (Circulation. 1998;98:458-465.)

Key Words: catheter ablation • tachyarrhythmias

When radiofrequency (RF) current was initially explored as an energy source for catheter ablation, these procedures were performed with conventional 6F, 2-mm electrodes.1-5 Clinical efficacy was limited by coagulum formation and impedance rise at relatively low power levels due to limited convective cooling by circulating endocavitary blood.1,2,4-5 The introduction of larger (7F to 8F, 4- to 8-mm) tip electrodes allowed for RF delivery at higher power levels because of better electrode cooling due to exposure of the larger noncontact electrode surface to the circulating endocavitary blood, which markedly improved clinical efficacy.11-27 However, there are multiple inherent disadvantages of larger electrodes: (1) a reduction in electrogram resolution, which makes it more difficult to identify the optimal ablation site; (2) greater variability in electrode-tissue coupling, depending on catheter tip orientation relative to the endocardium (parallel versus perpendicular); and (3) a reduction in flexibility and mobility of the catheter, which may impair positioning of the ablation electrode. Recently, active electrode cooling by saline irrigation has been introduced to prevent an impedance rise, allowing the use of higher RF power to produce significantly larger and deeper lesions.28-36

This study tested the hypothesis that because convective cooling by the blood stream is not required with a saline-irrigated electrode, a smaller electrode can be used without sacrificing ablation efficacy. Potential advantages of a smaller electrode include higher electrogram resolution (improving mapping accuracy to increase ablation efficacy and decrease the number of RF applications),37-39 increased ablation catheter flexibility and mobility, and the ability to design small ablation electrodes for use in small children and in small anatomic spaces, such as in the coronary veins. To test the hypothesis, we compared in vivo tissue temperature and lesion size between a 2-mm and a 5-mm ablation electrode oriented perpendicular and parallel to the tissue interface, respectively, during RF ablation with saline irrigation in a canine thigh muscle preparation.31

Received October 28, 1997; revision received February 2, 1998; accepted February 13, 1998.

From the Cardiovascular Section/Department of Medicine, University of Oklahoma Health Sciences Center and the Department of Veterans Affairs Medical Center, Oklahoma City; and the Department of Cardiology, Heart Lung Institute, University Hospital Utrecht, Netherlands (F.H.M.W.).


Correspondence to Hiroshi Nakagawa, MD, PhD, Department of Medicine/Cardiovascular Section, University of Oklahoma Health Sciences Center, 920 S.L. Young Blvd, WP3120, Oklahoma City, OK 73104. E-mail hiroshi-nakagawa@ouhsc.edu

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The experimental protocol was approved by the University of Oklahoma Institutional Animal Care and Use Committee. Eleven mongrel dogs weighing 20 to 26 kg were anesthetized with sodium pentobarbital (25 mg/kg) and mechanically ventilated with room air. General anesthesia was maintained with supplemental doses of sodium pentobarbital. The right carotid artery was cannulated for continuous monitoring of atrial pressure. The thigh muscle preparation was used as previously described (Figure 1). The dog was initially placed on its right side, and a 20-cm skin incision was made over the left thigh muscle. The skin, connective tissue, and thin superficial muscle were gently dissected, exposing the surface of the thicker underlying muscle. The edges of the skin were raised to form a cradle that was filled with heparinized canine blood from the same dog, maintained at 36°C to 37°C. The blood in the cradle was exchanged at a rate of 350 mL/min. A custom 7F catheter with a central lumen and either a 2-mm or 5-mm tip electrode with 6 irrigation holes (0.4-mm diameter) located radially around the electrode 1 mm from the tip electrode (Cordis Webster, Inc) was used in this study (Figure 2). The tip electrodes also contained a thermocouple for measurement of the tip electrode temperature (accuracy, ±0.2°C between 20°C and 120°C).

The 2- or 5-mm tip electrode was positioned perpendicular or parallel to the thigh muscle in different experiments (Figure 2). In both electrode-tissue orientations, a constant weight of 10 g was applied to the tip electrode. Tissue temperatures were measured with 2 fluoroptic thermal sensor probes (Luxtron Inc, model 3000-4; measurement range, 0°C to 125°C; accuracy, ±0.2°C) bundled together with shrink tubing. The probe extended 3.5 and 7.0 mm from the end of the shrink tubing. The probes were situated 3.5 and 7.0 mm below the surface directly adjacent to the ablation electrode, as illustrated in Figure 2.

The ablation electrodes were irrigated through the catheter lumen with room-temperature (20°C to 22°C), heparinized (2 U/mL) normal saline at 20 mL/min. Saline irrigation was started 3 to 5 seconds before the onset of RF application and was maintained until 5 seconds after the completion of RF delivery. RF energy (500 kHz) was delivered at constant voltage (Radionics, model 3D-J) between the ablation electrode and an adhesive electrosurgical dispersive pad (skin patch) applied to the shaven skin of the opposite thigh. During each RF application, current and impedance were monitored and recorded along with the electrode and tissue temperatures.

Ablation Protocol

The dogs were divided into 2 groups: perpendicular electrode-tissue orientation (group 1, 5 dogs) and parallel electrode-tissue orientation (group 2, 6 dogs). Five to 8 RF applications were delivered at constant voltage (50 V) to the 2-mm and 5-mm electrodes at separate sites on the left thigh muscle. The skin incision was closed, the dog was turned into its left side, and 5 to 8 RF applications were delivered to the right thigh muscle. Before each RF application, the skin cradle was depleted of blood and the electrode-tissue contact area was flushed with deionized water, creating a high-resistance barrier around the electrode except at the electrode-tissue contact site. A short (1-second), noninjurious (20-V) RF test application was used to measure the insulated electrode impedance to estimate electrode-tissue interface impedance. The temperature probes were then inserted, and the cradle was filled with heparinized canine blood. RF energy was delivered at 50 V for 30 seconds. The impedance was recorded at the onset of RF application. The RF application was terminated immediately in the event of an impedance rise >10 Ω. The occurrence of an audible “pop” was recorded, but the application of RF energy was not terminated if the pop was associated with an impedance rise of ≤10 Ω. After each RF application, the electrode was examined for coagulum formation. At the end of each experiment, the impedance was measured between a skin patch (18×7.5 cm) placed on the thigh muscle and the skin patch on the opposite thigh, as an approximate measurement of the body component of the ablation impedance (body impedance).

After Ablation

Two hours after the completion of the ablation protocol, 30 mL of 2% triphenyl tetrazolium chloride was administered intravenously. This dye stains intracellular dehydrogenase, which distinguishes viable and necrotic tissue. The dogs were euthanized, and the thigh muscles were excised and fixed in 10% formalin. The thigh muscles were sectioned, and the maximal depth, maximal diameter, and lesion surface diameter were measured.

Statistical Analysis

The values are expressed as mean±SD. The electrical parameters of RF delivery (voltage, current, impedance, and power), electrode and
tissue temperatures, and lesion dimensions were compared between the 2- and 5-mm electrodes in each electrode orientation by use of a two-tailed t test for unpaired variables. A χ² test was used to determine the significance of the difference in occurrence of audible pop between the 2 electrodes in each electrode orientation. A value of *P* = 0.05 was used as the level of statistical significance.

**Results**

A total of 148 RF lesions were produced in the thigh muscles of the 11 dogs; 31 and 32 RF applications were delivered in the perpendicular orientation with the 2-mm and 5-mm electrodes, respectively, and 42 and 43 RF applications in the parallel orientation with the 2-mm and 5-mm electrodes, respectively. The electrical RF parameters, temperatures, and lesion dimensions obtained in these 2 groups are shown in the Table and Figures 3 through 5. Because of active electrode cooling, neither an impedance rise >10 Ω nor coagulum formation occurred in any RF application. Because of saline irrigation, the electrode temperature remained below 53°C during all RF applications.

**Group 1: Perpendicular Electrode-Tissue Orientation**

The insulated electrode impedance, measured with deionized water, was not significantly different for the 2- and 5-mm electrodes (229±15 and 228±13 Ω, respectively, Table 1). During RF ablation (cradle filled with blood), the overall impedance with the 2-mm electrode was higher than with the 5-mm electrode (98±8 versus 70±9 Ω, *P* < 0.01). The delivered RF power was lower with the 2-mm electrode (26.0±2.1 versus 36.4±5.0 W, *P* < 0.01), but tissue temperatures at the depths of 3.5 and 7 mm were significantly higher with the 2-mm electrode (Figures 3 and 4 and Table 1). Consequently, lesions created with the 2-mm electrode were significantly wider (maximal diameter, 12.4±1.4 versus 8.4±0.9 mm, *P* < 0.01) and deeper (maximal depth, 8.0±1.0 versus 5.4±0.9 mm, *P* < 0.01, Figure 5). Small, sharp increases in impedance (<10 Ω) usually coinciding with an audible pop occurred in 5 of 31 RF applications (after 15.2±3.6 seconds) with the 2-mm electrode and in 0 of 32 RF applications with the 5-mm electrode (*P* < 0.05).

**Group 2: Parallel Electrode-Tissue Orientation**

In the parallel orientation, the insulated electrode impedance, measured with deionized water, was significantly different for the 2-mm and 5-mm electrodes (238±22 and 187±32 Ω, *P* < 0.01, Table 1). During RF ablation, the overall impedance was higher with the 2-mm electrode (101±7 versus 76±5 Ω, *P* < 0.01), resulting in lower RF power (25.0±1.7 versus

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### Radiofrequency Parameters and Temperatures During Ablation

<table>
<thead>
<tr>
<th></th>
<th>Perpendicular-Tissue Orientation and Electrode Length</th>
<th>Parallel-Tissue Orientation and Electrode Length</th>
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<tr>
<td></td>
<td>2 mm</td>
<td>5 mm</td>
</tr>
<tr>
<td>No. of lesions</td>
<td>31</td>
<td>32</td>
</tr>
<tr>
<td>Electrical parameters</td>
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<td>Voltage, V</td>
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<tr>
<td>Current, A</td>
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<td>0.73±0.09</td>
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<tr>
<td>Power, W</td>
<td>26.0±2.1*</td>
<td>36.4±5.0</td>
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<tr>
<td>Overall impedance, Ω</td>
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<td>70±9</td>
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<tr>
<td>Insulated electrode impedance, Ω</td>
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<td>228±13</td>
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<tr>
<td>Temperature, °C</td>
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<td>39.4±4.2</td>
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<tr>
<td>Electrode</td>
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<tr>
<td>Tissue at 3.5-mm depth</td>
<td>96.9±8.2*</td>
<td>65.3±8.3</td>
</tr>
<tr>
<td>Tissue at 7.0-mm depth</td>
<td>62.4±12.2*</td>
<td>44.0±4.4</td>
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*P* < 0.05 between the 2-mm tip electrode and the 5-mm tip electrode within the perpendicular (group 1) or parallel (group 2) electrode-tissue orientation.
33.2 ± 2.2 W, *P* < 0.01), but tissue temperatures were significantly higher (Table 1). Although lesions obtained with the 2-mm electrode were significantly deeper than with the 5-mm electrode (7.3 ± 0.6 versus 6.9 ± 0.5 mm, *P* < 0.01, Figure 5), there was no significant difference in the maximum lesion diameter (11.1 ± 1.1 versus 11.3 ± 1.1 mm, *P* = NS, Figure 5). Audible pops occurred in 7 of 42 RF applications (after 18.1 ± 8.8 seconds) with the 2-mm electrode but none of the 43 RF applications with the 5-mm electrode (*P* < 0.05).

3.2 Body Impedance

The body impedance, measured between a skin patch placed on the thigh muscle and the skin patch on the opposite thigh, was 26 ± 2 V in group 1 and 26 ± 1 V in group 2.

4. Discussion

In this study, we investigated the effects of electrode size on lesion formation when the requirement for a large electrode size (to maintain a low electrode-tissue interface temperature) is eliminated by active cooling with saline irrigation. With equal RF voltage, the impedance was higher and the power was lower with a 2-mm electrode than a 5-mm electrode. However, tissue temperatures and lesion sizes were significantly larger with the smaller (2-mm) electrode, especially in the perpendicular orientation.

5. Heating Efficacy

The unexpected finding of greater tissue heating with a smaller electrode at lower overall RF power may be explained on the basis of a more detailed analysis of the various components of the ablation circuit. The overall ablation circuit can be approximated as illustrated in Figure 6. *R*$_{\text{Remote}}$ represents the impedance of the connecting cables, catheter, skin patch, and animal body (all impedance except for the ablation electrode interface with the tissue and blood). The impedance of all of the cables, connected in series, was measured at 1 V. The impedance of the ablation catheter was measured at 3.5 V. The average body impedance was 26 V. Therefore, *R*$_{\text{Remote}}$ can be estimated at 30 V. The ablation electrode impedance (electrode-tissue interface [*$R$$_{\text{Tissue}}$] and electrode-blood interface [*$R$$_{\text{Blood}}$] connected in parallel) can be estimated by subtracting 30 V (*R*$_{\text{Remote}}$) from the overall impedance measured during RF ablation. In the perpendicular electrode orientation, the 2-mm ablation electrode impedance can be estimated at 98 Ω - 30 Ω = 68 Ω (Figure 7A). At 50 V, the voltage drop across the electrode-tissue interface (tissue voltage) is 50 V × (68 Ω/98 Ω) = 34.7 V, and the voltage drop across *R*$_{\text{Remote}}$ is 50 V × 30 Ω/98 Ω = 15.3 V. The insulated electrode impedance was measured with deionized water at 229 V. The electrode-tissue interface impedance (*R*$_{\text{Tissue}}$) can be estimated at 199 Ω by subtracting *R*$_{\text{Remote}}$ (30 Ω) from the insulated electrode impedance. The tissue power, which is

![Figure 4](http://circ.ahajournals.org/)

**Figure 4.** Electrical parameters and temperatures recorded during RF application at 50 V through 5-mm electrode with saline irrigation (20 mL/min) in perpendicular electrode-tissue orientation. Format and animal same as Figure 3. Note, longer (5-mm) electrode with larger electrode-blood contact area was associated with lower overall impedance (67 to 71 Ω) and higher overall current (0.70 to 0.75 A), but tissue temperatures reached only 64°C at depth of 3.5 mm and only 44°C at 7.0-mm depth.

![Figure 5](http://circ.ahajournals.org/)

**Figure 5.** Diagram of lesion dimensions for 2 groups studied. Values are millimeters (mean ± SD). (A) indicates maximal lesion depth; (B), maximal lesion diameter; (C), depth at maximal lesion diameter; and (D), lesion surface diameter. *P* < 0.05 between 2-mm and 5-mm ablation electrodes within same electrode-tissue orientation.

![Figure 6](http://circ.ahajournals.org/)

**Figure 6.** Circuit for RF ablation can be considered to have overall impedance consisting of nonablation electrode impedance (*R*$_{\text{Remote}}$) produced by cables, skin patch, and body, which is in series with impedance of ablation electrode consisting of electrode-tissue interface impedance (*R*$_{\text{Tissue}}$) and electrode-blood interface impedance (*R*$_{\text{Blood}}$) connected in parallel.

Body Impedance

The body impedance, measured between a skin patch placed on the thigh muscle and the skin patch on the opposite thigh, was 26 ± 2 Ω in group 1 and 26 ± 1 Ω in group 2.

Discussion

In this study, we investigated the effects of electrode size on lesion formation when the requirement for a large electrode size (to maintain a low electrode-tissue interface temperature) is eliminated by active cooling with saline irrigation. With equal RF voltage, the impedance was higher and the power was lower with a 2-mm electrode than a 5-mm electrode. However, tissue temperatures and lesion sizes were significantly larger with the smaller (2-mm) electrode, especially in the perpendicular orientation.

Heating Efficacy

The unexpected finding of greater tissue heating with a smaller electrode at lower overall RF power may be explained on the basis of a more detailed analysis of the various components of the ablation circuit. The overall ablation circuit can be approximated as illustrated in Figure 6. *R*$_{\text{Remote}}$ represents the impedance of the connecting cables, catheter, skin patch, and animal body (all impedance except for the ablation electrode interface with the tissue and blood). The impedance of all of the cables, connected in series, was measured at <1 Ω. The impedance of the ablation catheter was measured at 3.5 Ω. The average body impedance was 26 Ω. Therefore, *R*$_{\text{Remote}}$ can be estimated at 30 Ω. The ablation electrode impedance (electrode-tissue interface [*$R$$_{\text{Tissue}}$] and electrode-blood interface [*$R$$_{\text{Blood}}$] connected in parallel) can be estimated by subtracting 30 Ω (*R*$_{\text{Remote}}$) from the overall impedance measured during RF ablation. In the perpendicular electrode orientation, the 2-mm ablation electrode impedance can be estimated at 98 Ω - 30 Ω = 68 Ω (Figure 7A). At 50 V, the voltage drop across the electrode-tissue interface (tissue voltage) is 50 V × (68 Ω/98 Ω) = 34.7 V, and the voltage drop across *R*$_{\text{Remote}}$ is 50 V × 30 Ω/98 Ω = 15.3 V. The insulated electrode impedance was measured with deionized water at 229 Ω. The electrode-tissue interface impedance (*R*$_{\text{Tissue}}$) can be estimated at 199 Ω by subtracting *R*$_{\text{Remote}}$ (30 Ω) from the insulated electrode impedance. The tissue power, which is
Effective heating power, is (tissue voltage)/\(R_{\text{Tissue}}\) = (34.7 V)/199 \(\Omega\) = 0.17 W. For the 5-mm electrode in the perpendicular orientation, the overall impedance was 70 \(\Omega\). By the same calculations (Figure 7B), the 5-mm electrode had a tissue voltage of 28.6 V and tissue power of only 4.1 W. Therefore, the 2-mm electrode delivered 49% more heating power to the tissue, resulting a marked increase in tissue temperatures and lesion size (Table 1 and Figures 3 through 5). The larger electrode resulted in greater current shunting to the blood, increasing overall current. The ineffective current shunted through the blood pool also has to pass through the ineffective impedance (\(R_{\text{Remote}}\)). Consequently, a larger proportion of the RF voltage is lost to \(R_{\text{Remote}}\), decreasing the voltage available for tissue heating (28.6 versus 34.7 V with the 5-mm and 2-mm electrodes, respectively).

In the parallel orientation, the tissue voltage is still smaller with the 5-mm electrode (30.3 versus 35.1 V with the 5-mm and 2-mm electrodes, Figure 8), but the lower electrode-tissue interface impedance (due to a larger electrode-tissue contact area) leads to a tissue heating power similar to the 2-mm electrode (5.8 versus 5.9 W, Figure 8), resulting in similar lesion diameters (Figure 5). However, the similar tissue power over a smaller area with the 2-mm electrode resulted in greater tissue temperatures at the 3.5- and 7.0-mm depths and greater lesion depth.

Because of the difference between the length and diameter in the 5-mm electrode, the contact area differed greatly between the perpendicular and parallel orientations, confirmed by the insulated electrode impedance measurements with deionized water. In the 2-mm electrode, electrode length and diameter are similar, making tissue temperatures and lesion sizes relatively independent of the electrode-tissue orientation.

Constant-Voltage Versus Constant-Power Delivery

In this study, RF energy was delivered at constant voltage, whereas most previous experimental and clinical studies used constant RF power or variable power based on electrode temperature (“temperature control”). The use of constant power (instead of constant voltage) would have magnified the difference in tissue voltage and tissue power between the 2- and 5-mm electrodes and between the perpendicular and parallel orientations, because voltage is highly dependent on impedance when constant power is used. For example, at 25 W with the 2- and 5-mm electrodes in the perpendicular orientation, the overall voltages would be \(\approx 49.5\) and 41.8 V, the tissue voltages would be 34.3 and 23.9 V, and tissue power would be 5.9 and 2.9 W with the 2-mm and 5-mm electrodes, respectively. Therefore, at 25 W, the tissue power would be 103% higher for the 2-mm electrode than the 5-mm electrode, compared with a 49% increase in tissue power for a constant 50-V application. For the 5-mm electrode, the tissue power would be 52% greater in the parallel orientation than the perpendicular orientation at 25 W (4.4 versus 2.9 W) compared with 41% greater at 50 V (5.8 versus 4.1 W). Therefore, the use of constant-voltage applications should decrease the variability in lesion size with differences in electrode size and orientation and may be preferable to constant-power applications because of a greater predictability in lesion size.

Tissue Superheating

Small, sharp increases in impedance coinciding with audible pops occurred only with the 2-mm electrode. The
selective occurrence of a steam pop with the smaller electrode was most likely due to the higher tissue temperatures, resulting from greater tissue power at the same RF voltage setting, because similar pops were observed with the 5-mm electrode in another study that used 66 V. The steam pop is thought to be caused by the sudden release of steam from below the surface of the tissue. Steam formation occurs when the tissue temperature several millimeters below the surface reaches 100°C. This requires some degree of surface cooling to prevent the surface temperature from reaching 100°C (with an impedance rise) before deeper tissue temperatures reach 100°C to produce steam. Sufficient surface cooling can occur without irrigation when the ablation electrode is exposed to high blood flow or when the electrode is sliding with each cardiac contraction. Therefore, steam pops occur relatively often with conventional ablation electrodes, even in the temperature control mode, and can be recognized by a brief, sharp 5- to 10-Ω increase in impedance. The occurrence of pericardial tamponade after RF applications has been linked to the occurrence of a steam pop. Pericardial tamponade may occur when the steam vents through the epicardium rather than the endocardium. This has occurred more frequently with an irrigated electrode than with a conventional electrode, probably as a result of steam formation closer to the atrial epicardium due to greater endocardial cooling and higher sustained RF power applications. Preliminary studies indicate that tissue superheating can be prevented by pulsing the RF application with on and off cycles, such as 5 seconds on and 5 seconds off. The off periods allow the hottest region, relatively shallow sites, to cool, which prevents superheating and steam formation. The deeper areas cool less during off periods, allowing a progressive increase in deep tissue temperature, resulting in greater lesion depth. Alternatively, tissue superheating can be prevented while deep lesions are obtained by application of RF current at lower power for a prolonged period.

**Study Limitations**

This study used the canine thigh muscle preparation instead of a beating heart to control electrode-tissue orientation and contact pressure and to allow the measurement of tissue temperature at various depths beneath the electrode. The flat surface of the thigh muscle also allows accurate determination of lesion size and geometry. In contrast, delivering RF current to the trabeculated endocardium, especially in a beating heart, often results in an irregular lesion shape and significantly greater variation in lesion size than was found in this study, which would make it more difficult to quantify the effects of changing any parameter (such as electrode length) on RF lesion size.

Because the intramural blood flow may be less in the resting thigh muscle than in the myocardium, the thigh muscle may provide less of a heat sink and therefore may result in greater lesion size. However, a previous study suggested that the tissue temperature profile during RF ablation may be independent of intramyocardial perfusion. In addition, the lesion size in the present study is similar to lesions described in other studies in which RF current was delivered to the ventricular myocardium of a beating canine heart. More importantly, the degree of difference in lesion size between the 2- and 5-mm electrodes would be expected to be similar in the human heart, because the same basic electrical principles apply.

The 2- and 5-mm electrodes were compared in the perpendicular orientation in group 1 dogs and in the parallel orientation in group 2 dogs. Because perpendicular and parallel orientations were studied in 2 different groups of dogs, the tissue temperatures and lesion sizes for the 2 orientations were compared only qualitatively and not statistically. However, body weight (23.8 ± 3.0 versus 23.2 ± 2.5 kg) and body impedance (26 ± 2 versus 26 ± 1 Ω) were not significantly different between the group 1 and group 2 dogs. Therefore, except for catheter orientation, there were no procedural differences between the studies in the 2 groups, suggesting that a quantitative comparison may be reasonable.

**Clinical Implications**

This study confirms earlier findings that active electrode cooling allows for relatively deep lesions to be created with small electrodes at relatively low power levels. This study importantly demonstrates a relatively large improvement in heating efficiency with decreasing electrode size. Therefore, with active electrode cooling, the ablation electrode size can be decreased with multiple potential benefits, including (1) an improvement in electrogram resolution, which should increase mapping accuracy and decrease the number of RF applications; (2) a decrease in catheter tip stiffness, which may improve catheter flexibility and mobility to facilitate reaching the ablation site; and (3) the ability to make very small ablation electrodes for use in small children or small ablation spaces, such as the middle cardiac vein and other coronary veins. With active electrode cooling, the electrode length could ideally be reduced to the electrode diameter, resulting in the same electrode-tissue contact area regardless of electrode-tissue orientation. This would make lesion size independent of electrode orientation. Because electrode cooling is provided by irrigation, RF voltage or power can be chosen and maintained independent of the local blood flow ("extrinsic cooling"), leading to a more consistent and predictable lesion size. In contrast, the power delivered (and lesion depth) in the temperature control mode (without irrigation) varies greatly with local blood flow. In low-flow areas, such as in a dilated atrium during atrial fibrillation, the RF power is markedly reduced, producing small and nontransmural lesions.

The use of small electrodes will require a significant reduction in RF power, because of a reduction in the amount of RF power lost to the blood pool and thus a greater percentage of the overall power delivered to the tissue. When required, lesion size and depth can be increased without tissue superheating and the occurrence of a steam pop either by a prolonged RF application at lower power or, preferably, by higher RF power delivered in the pulsed mode, in which the off periods prevent tissue superheating in the hottest, shallow sites.
Another potential advantage of a small irrigated electrode would be the ability to make a small, discrete, accurately placed RF lesion with a very low RF power application, such as for ablation of an anteroseptal or midseptal accessory pathway or ablation in a small child.

Acknowledgments

This study was supported by a grant (RO1-HL-39670) from the National Institutes of Health and a grant (HRC-RRP-A028) from the Oklahoma Center for the Advancement of Science and Technology.

References


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doi: 10.1161/01.CIR.98.5.458

Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
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Print ISSN: 0009-7322. Online ISSN: 1524-4539

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