Nonuniform Heating During Radiofrequency Catheter Ablation With Long Electrodes: Monitoring the Edge Effect

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Background Long, narrow electrodes are being considered for radiofrequency ablation of atrial fibrillation; however, preliminary work revealed coagulum formation on the electrodes and lack of lesion continuity. This may be due to the "edge effect," which concentrates radiated energy at sharp geometric gradients. It is proposed that temperature sensors at electrode edges are preferable to a single centered sensor for temperature feedback and monitoring of long electrode geometries.

Methods and Results A finite element model was used to predict the heating properties of new long electrode geometries. Sixteen dogs with atrial fibrillation underwent left and right atrial ablation using catheters with multiple 12.5-mm coil electrodes. Electrodes with a single thermistor were compared with electrodes with dual thermocouples placed at opposite ends and on opposing sides of the electrode. Power, temperature, and impedance were recorded for all lesions, and coagulum adhesion and magnitude were noted in a subset of lesions. Finite element analysis shows uneven heating, with the main heating concentrated at the electrode edges and a propensity toward temperatures >100°C with single-thermistor feedback control. Ablations with dual thermocouple electrodes achieved higher measured temperatures at lower power levels than those that used single-thermistor electrodes. Impedance rises and coagulum adherence occurred less frequently with dual thermocouple electrodes than with single, centered thermistor electrodes (176 of 395 versus 9 of 425 lesions; \( P < 0.001 \); 46 of 98 versus 7 of 150 lesions; \( P < 0.001 \), respectively).

Conclusions Maximum heating from radiofrequency energy occurs at the electrode edges, particularly with long electrodes. The safety of temperature-feedback atrial ablation with these electrodes is significantly improved by
It has been proposed that the ablation of atrial fibrillation may be achieved with transcatheter radiofrequency ablation techniques but may require the use of long, narrow electrodes to create long, linear, continuously transmural lesions in a pattern similar to the maze procedure. Radiofrequency ablation of supraventricular arrhythmia has become a standard procedure in electrophysiology, and many biophysical aspects of radiofrequency ablation, such as the effects of electrode radius, tissue temperature, and electrode impedance, have been defined. However, the power and temperature distributions from long electrode geometries may differ from those observed with standard ablation catheters. The advantages of temperature monitoring for the prevention of impedance rises and thrombus and for judging both tissue contact and lesion efficacy have been shown with conventional catheters, but the characteristics of long coil electrodes have not been defined.

The distribution of electric potential from a radiating source is governed by the Laplace equation. This equation explains observations of an "edge effect" in electrodes and antennae; that is, that there is a higher current and power density at areas of high geometric gradients such as the edges of electrodes (see "Appendix"). Because heating increases with power density during radiofrequency ablation, peak temperatures should also occur at the electrode edge at the junction between the electrode and insulator. Thus, it was hypothesized that positioning multiple temperature sensors at the edges of the electrode would be preferable to a single, centrally located sensor and should reduce the likelihood of overheating with associated impedance rise and thrombus formation. The purpose of this investigation was to compare the characteristics of temperature-controlled ablations performed with a single temperature sensor with those performed with sensors located at the electrode edges by use of a computer model and in vivo experimentation.

**Methods**

**Electrode Description**

The electrode studied during this experiment was a 7F, 12.5-mm coil of wire with a rectangular cross section, 0.5 mm wide and 0.1 mm thick, coiled around the catheter in 19 windings spaced 0.17 mm apart. The ends of the coil electrode were covered by a thin electrically and thermally insulated layer of UV adhesive. Two temperature sensor configurations were tested (Fig 1): first, a single thermistor was positioned in the center of the electrode under but in good thermal contact with the electrode (single-thermistor electrode); second, two thermocouples were placed on opposite edges and opposing sides of the electrode in good thermal contact with the electrode (dual thermocouple electrode). The single-thermistor electrode catheter was designed such that when a right-hand bend is placed in the catheter, the thermistor is on the outside of the bend. This bend could then be fluoroscopically maneuvered to achieve good placement against the atrial wall with confidence that the thermistor was tangential to the endocardium. With the dual thermocouple electrode catheter design, the thermocouples were placed on opposite ends and opposing sides of the electrode in the same plane as the electrode bend. Thus, with the catheter in either a left- or right-hand bend, one of the thermocouples was on the outside of the curve and could be positioned tangential to the endocardium during radiofrequency energy delivery.

**Figure 1.** Schematic of portions of the two catheters used for the comparison of electrode temperature-sensor configurations examined during this study. Catheter A was designed with a single centered thermistor flush with the electrode coils. Catheter B was designed with dual thermocouples placed on...
Finite Element Model of a Coil Electrode

Finite element analysis is a numerical method for solving differential equations under circumstances that restrict empirical solutions, such as complex geometries or varying initial and boundary conditions. A complex geometry is subdivided into finite elements with a number of nodal solution sites. Numerical solutions to the differential equations are calculated at the nodes systematically, starting at the known boundaries and initial conditions. The nodal values are then interpolated over each element via element-interpolation functions. The finite element technique and its validation have been described previously. In this experiment, the Laplace equation and the bioheat transfer function are solved at the nodes to determine the electric field and then the temperatures that result from the radiofrequency energy delivery. A three-dimensional finite element model of the 7F, 12.5-mm coil electrode was used for this model. The coil was modeled as stainless steel and the catheter body as an insulator. The insulated layer of UV adhesive at the ends of the electrodes was modeled as an area of very low electrical and thermal conductivities. The model simulated an electrode lying on a 4-cm-thick slice of cardiac tissue. The region modeling the blood extended 4 cm above the tissue, and the blood-electrode-tissue region was 4 cm long. It was assumed that there were negligible effects of both radiofrequency energy–derived heating and negligible electric field strength at a great distance from the electrode (the external boundary of the model), so conditions were defined as 37°C for the bioheat transfer equation and 0 V for the Laplace equation on the model boundary. The initial node temperatures were set at 37°C, assuming temperature equilibrium within the model. Two models of radiofrequency energy delivery were investigated emphasizing the steady-state temperature distributions. The first model represented an electrode with a centered temperature sensor for temperature feedback, so energy was applied until the center of the electrode reached 70°C. The second model represented electrodes with edge temperature sensors, so edge conditions were set to 70°C. An electrode-tissue contact surface of 40% of the electrode surface area was assumed, and a flow velocity of blood causing convective cooling of 18 cm/s was selected. The overall finite element model was formed of 9029 nodes and used 9056 hexahedral elements in a nonuniform mesh. Regions of interest, such as boundaries between different materials or material edges, were accorded a high concentration of smaller-sized elements for increased regional model resolution.

Animal Model

All experimental protocols observed the position of the American Heart Association on research animal use and were accepted by an Internal Animal Research Review Committee. Sixteen dogs with atrial fibrillation were anesthetized with 0.5% to 1.0% halothane, 60% nitrous oxide, and 25 to 100 µg per hour intravenous fentanyl. All dogs received 1 mg of intravenous atropine during induction of anesthesia. The right femoral artery and vein were exposed and cannulated via surgical cutdown. We performed transseptal catheterization using a modified Brockenbrough needle and Mullins sheath to provide access to the left atrium. Arterial pressure and a six-lead ECG were continuously monitored and periodically recorded on a physiological recorder during the ablation procedure.

Catheters and Ablation System

The ablation of atrial fibrillation was accomplished with the use of steerable 8F ablation catheters (EP Technologies) composed of a series of two to six of the previously described coil electrodes spaced 2 mm apart along the distal, deflectable portion of the catheter (Fig 2). Radiofrequency energy was delivered to appropriate electrodes from an

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Experimental high-power generator (maximum output, 150 W root mean square; EP Technologies). Power was controlled by a feedback control algorithm in the generator that adjusted power to maintain a preset target temperature. For multiple temperature sensors, a temperature-monitoring unit compared the input temperatures and controlled the generator power with the highest monitored temperature. The generator allowed the on-line recording of voltage, current, impedance, and temperature. It was also programmed to terminate radiofrequency energy delivery if the impedance became >300 Ω.

**Figure 2.** A representative photograph of the catheters tested in this study. The catheters had 7F shafts with 8F electrodes. They used bidirectional steering and two to six coil electrodes per catheter.

![Catheter Image](image-url)

**Experimental Protocol**

Ablation lesions were created in both left and right atria. During lesion creation, power delivery was automatically controlled to maintain a target temperature of 70°C for 60 seconds. The lesions were created with sequential unipolar radiofrequency energy delivery to selected electrodes that appeared to be in good contact with the atrial wall in right anterior oblique and left anterior oblique fluoroscopic views. An electrode was judged to be in good contact if the electrode was moving in concert with the atrial wall. The impedance, voltage, current, and electrode-tissue interface temperature were recorded continuously on a personal computer during each delivery and stored in a file for processing after the experiment. After a set of lesions was created, the catheter was removed from the heart and examined for the presence of adherent thrombus. Preliminary studies showed that clots on the coil electrodes tend to propagate from the edges toward the center, so the amount of electrode thrombus was assessed by measuring the distance along the electrode that the thrombus progressed. The instances of electrode thrombus adherence were assessed semiquantitatively as follows: 0, clean electrode; 1, trace fibrin only; 2, <1 mm of coagulum; 3, >1 mm of coagulum and <2 mm of coagulum; and 4, >2 mm of coagulum. After the completion of energy delivery, the dogs were euthanatized and the hearts removed. The endocardial surface of each atria was inspected grossly for charring, pitting, perforation, and the apparent continuity of the lesions. The lesions were then bisected and stained with nitro-blue tetrazolium, which demarcates viable from nonviable tissue. The surface continuity was checked throughout the myocardium, and the transmural portion of each lesion was recorded and tabulated as a percentage of the gross endocardial surface length. Any areas of lesion dropout were noted qualitatively.

**Data Analysis and Statistics**

The impedance, temperature, and power characteristics were analyzed off-line after the experiment. An increase in impedance >20 Ω above that observed at the initiation of radiofrequency energy delivery was defined as a significant impedance rise. Automatic power shutdowns when impedance exceeded 300 Ω were recorded. The observations of coagulum adherent to the electrode were tabulated after the experiment and compared with impedance-rise observations. Observations in which more than a single radiofrequency delivery was made between catheter inspections were censored from this analysis.

Raw data were stored in a computerized database. Normally distributed continuous data are presented as mean±SD. Comparisons between grouped data were made by use of the Student's *t* test. The comparison of alterations in event frequency was judged with Fisher's exact test. Statistical significance was determined as values of *P*<.05.
Results

Finite Element Analysis
The finite element analysis of radiofrequency energy delivery to an electrode with a centered temperature sensor for temperature control (Fig 3) required 55.8 W of power (system impedance was 72 Ω) to raise the temperature at the center temperature sensor to 71°C. The maximum tissue temperature was 161°C ≈ 0.5 mm directly below the edge of the electrode, and the temperature at the edge of the electrode was 136°C. A finite element technique cannot model the impedance rise subsequent to the boiling and charring that occurs at the electrode-tissue interface at temperatures >100°C, which restricts energy transfer and limits the effective lesion dimensions. The analysis of radiofrequency energy delivery to an electrode with a temperature sensor on the edge for temperature feedback power control (Fig 4) required 28.6 W of power (system impedance of 72 Ω) to raise the edge temperature sensor to 75°C. The temperature at the center of the catheter was 56°C, and the maximum tissue temperature was 97.7°C ≈ 0.5 mm directly below the electrode edge. The 50°C isotherm predicts a lesion depth of 6.5 mm and a length of 20.5 mm.

Figure 3. The steady-state temperatures derived from the finite element analysis of radiofrequency ablation with a 12-mm-long coil electrode. In this analysis, the electrode temperature at the center of the electrode was maintained at 71°C. The legend of temperatures is shown at the right of the graph and ranges from the physiological normal (violet=37°C) to the maximum tissue temperature (red=161°C) located below the electrode edges.

Figure 4. The steady-state temperatures derived from finite element analysis of radiofrequency ablation with a 12-mm-long coil electrode. In this analysis, the electrode temperature at the edges of the electrodes was maintained at 75°C. The legend of temperatures is shown at the right of the graph and ranges from the physiological normal (violet=37°C) to the maximum tissue temperature (red=97.7°C) located below the electrode edges, as in Fig 3, but does not exceed 100°C, thus preventing boiling and coagulum formation.

In Vivo Testing
Radiofrequency energy was delivered in a unipolar fashion to single coil electrodes 833 times in 16 dogs. The single centered thermistor electrode catheters were used in 8 dogs for a total of 395 energy deliveries. Ablation catheters with thermocouples on opposite sides of the electrodes were used in 8 dogs for a total of 438 energy deliveries. The mean measured electrode-tissue interface temperatures were significantly higher and delivered powers associated with these ablations were significantly lower with dual thermocouple electrodes than with the single thermistor electrodes (Table 1).
The lesions were examined pathologically. There was no evidence of pitting or perforation by either electrode catheter, although there was one "pop" lesion associated with each electrode type. The lesions from the single-thermistor electrodes were continuous in 44 (69%) of 64 lesion lines versus 29 (53%) of 55 lines of lesion created with catheters with dual thermocouple electrodes \((P=\text{NS})\). Char was adhered to 36 (56%) of the single-thermistor electrode lesions and 20 (35%) of the dual thermocouple electrode lesions. Most of the char was found at the edges of the lesions. With single-thermistor electrodes, 93.9% of the epicardial length of each lesion was found to be transmural with nitro-blue tetrazolium staining, compared with 96.1% of the dual thermocouple controlled electrodes \((P=\text{NS})\). The transmural dropout regularly occurred in the center of the dual thermocouple electrode lesions and in the intercoil spaces of the single-thermistor electrode lesions.

**Impedance Rise and Coagulum Adherence**

Ablation performed with single centered thermistor electrodes showed an impedance rise \(>20\,\Omega\) above baseline measurements during 176 (45%) of the 395 radiofrequency deliveries versus 9 (3.6%) of 247 radiofrequency deliveries with dual-edge thermocouples \((P<.0001)\). Only 44 (27%) of those cases had an impedance rise that exceeded the 300 \(\Omega\) automatic generator cutoff. Temperatures of 100°C are associated with a rise in impedance, but high temperatures \((>90^\circ\text{C})\) were only observed in 13 (7%) of 176 impedance rises with single-thermistor electrodes and 2 (22%) of 9 impedance rises using dual thermocouple electrodes \((P=\text{NS})\). Sudden \((>7\,\Omega/s)\) impedance rises followed by a rapid return to baseline impedance were observed in 46 (24%) of the 185 cases of impedance rise and were associated with temperatures of 82.2±5.1°C and powers of 45.7±25.3 W. During the energy deliveries in which this type of impedance rise occurred, the phenomenon of rapid rise and return to baseline could oscillate throughout the energy delivery (Fig 5) or rise and fall only a single time. Slow impedance rises in which peak values did not exceed the 300-\(\Omega\) power cutoff were seen in 95 (49%) of 185 cases of impedance rise and were associated with lower temperatures of 67.9±6.4°C \((P<.0001)\) and higher powers of 77.8±31.3 W \((P<.0001;\text{Fig 5})\). These phenomena usually presented as a gradual and consistent rise in impedance from baseline to end impedance (between 40 and 150 \(\Omega\) above baseline).

**Coagulum Adherence to the Electrode**

The catheters were removed and inspected for coagulum adherence after a single radiofrequency energy delivery in 98 cases with single-thermistor and 157 cases with dual thermocouple electrode ablation. The coagulum found on the single-thermistor electrodes rated 2.3±1.5 versus that on the dual thermistor electrodes, which rated 0.6±0.9 \((P<.001)\). A breakdown of the biophysical data comparing those energy deliveries resulting in coagulum formation and those without...
Coagulum formation is shown in Table 2. There were 10 impedance rises (5%) out of 196 energy deliveries that did not result in coagulum adhesion versus 31 (52%) of 59 with adherent coagulum ($P<.0001$). Coagulum adherent to the electrode due to rapid impedance rises amounted to 2.1±0.06 versus that due to a slow impedance rise, which was 2.9±0.06 ($P<.03$).

### Discussion

The present investigation examined the safety and efficacy of radiofrequency ablation using long ablation electrodes with single or dual temperature sensors. Two finite element models indicated the highest temperature values were at the coil edges and were 20°C to 30°C higher than the temperature at the center of the electrode. The set of parameters that modeled a 70°C temperature for temperature feedback power control from the center of the electrode calculated temperatures >100°C at the electrode edges. However, if the temperature feedback power control is performed using temperatures measured at the electrode edges where the current density is the greatest, the heating is controlled much more efficiently and the risk of boiling is minimized. These configurations were tested in vivo. Electrodes using dual thermocouples at the electrode edges required less average power to achieve the targeted temperature and had a significantly lower prevalence of impedance rise and adherent coagulum without a significant decrease in lesion continuity. Frequent impedance rises and consistently severe coagulum adherence to the electrode were observed with the single centered thermistor, which indicated the lack of power control provided by a central temperature sensor. Unlike radiofrequency ablation with conventional 4-mm-tip electrode catheters, significant coagulum accumulation resulted in a slow rise in impedance in 23% of cases undetected by the temperature monitor. In these cases, it is likely that temperatures >100°C were achieved outside the effective range of the temperature sensors. The resulting desiccation limited the effective surface area available for ablation to the center portion of the electrode, which was then temperature controlled for the remainder of the energy delivery while coagulum continuously accreted at the edges.

### Previous Studies

It has been shown in vivo and in vitro that there is a strong association between temperatures of 100°C and a sudden rise in impedance. It is believed that when temperatures of 100°C are reached at an electrode-tissue interface, boiling occurs and tissue contiguous to the electrode denatures and forms an insulating layer that resists the flow of current. The resultant loosely adherent thrombus is subsequently at risk for embolization. To control the thrombus, radiofrequency ablation system designs are incorporating integrated temperature feedback power control. Closed-loop control has been added in some clinical tools and is expected to be of primary importance in most systems. The inclusion of closed-loop temperature control makes the location of temperature sensors very important. Blouin et al showed that the accuracy of temperature monitoring could be improved by electrically insulating the sensor from the electrode and placing it in contact with the tissue in the center of the area to be heated. A single centered-tip temperature sensor was considered adequate for standard 8F 4-mm electrodes. Tip-sensor temperature-monitored 8F electrodes of 8 mm and 12 mm lengths produced larger lesions than a standard 4-mm tip, but the 12-mm lesions were smaller than the 8-mm lesions, indicating an overestimation of the tissue temperature by the tip temperature sensor or partial electrode contact. The 12-mm-tip lesions were also associated with char, indicating impedance rises and temperatures of 100°C at points on the electrode distal of the tip.

The coil electrode was chosen to increase catheter flexibility and diminish the partial electrode contact without loss of the electrical and thermal properties studied in solid-ring electrodes. To this end, these electrodes were designed using finite element analysis such that the electric field formed during the delivery of radiofrequency energy would be similar to that.
of a long-ring electrode. Despite the improved flexibility, the irregular pattern of heating in this experiment indicated poor power control with a single centered temperature sensor. Subsequently, the temperature sensors were placed to improve the safety and control of atrial ablation. Other alterations in the coils have been considered, such as doubling the winding pitch (increasing the interwinding space) at the edges of the electrodes to decrease local current density and temperatures. However, preliminary finite element models of these coil electrodes demonstrated no change in the resultant current density and only a minimal reduction in edge temperature. Analysis also concluded that the pitch would have to be increased tenfold for a decrease to be observed, which would drastically reduce the electrode surface area available for the creation of lesions and decrease the solid-ring–like properties of the coils. It is also believed that a further reduction in the incidence of impedance rises, the amount of coagulum, and the efficacy of the lesions can be realized if electrode-tissue contact is optimized.

Monitoring the edges of the electrodes resulted in a dramatic decrease in the coagulum adherent to the electrode and the char on the lesions. However, there was a trend toward a reduction in lesion continuity. The advantage of greater control of power output is that it may allow a safe increase in target temperature from 70°C, as in the present study, to 80°C, which may improve the lesion continuity. As efforts to increase lesion size and length with new electrode geometries continue for the ablation of arrhythmias such as atrial fibrillation and atrial flutter, the risks of coagulum formation and embolism may increase if temperature monitoring is not designed around geometric concerns such as the edge effect.

Limitations
There were a limited number of temperature sensors on each electrode of the catheters used in this protocol. Even with the dual-edge thermocouple design, it is likely that only one temperature sensor in each electrode was in contact during each ablation, because the temperature sensors are on opposite sides of the electrodes. If the electrode-tissue contact on one electrode edge has poor contact, then the peak surface temperature may still be underestimated. In fact, when coil electrode catheters were positioned under fluoroscopic guidance in good contact and then judged with intravascular ultrasound, it was concluded that those electrodes judged in good contact fluoroscopically were not necessarily in contact along the entire length of the electrode. Even with tissue contact ensured along the entire length of the electrode, it is possible that the peak temperature may occur at a position different from that of the temperature sensor. This is an inherent limitation in the design of multiple electrode catheters in which there is a trade-off between the efficacy of temperature monitoring and the ability to fit multiple wires and steering mechanisms inside a transvenous catheter. To counter the possibility of unmonitored hot spots, the temperature selected for temperature-controlled power regulation is usually limited to 70°C to 80°C. Because a number of ablation sites and catheter positions were overlapping, it was not possible to correlate a specific radiofrequency energy delivery to a specific pathological lesion. Therefore, only summary pathological data were reported.

Conclusions
Using finite element analysis and in vivo testing in a canine model, we determined that there is a significant disparity between peak temperatures achieved at the electrode edges versus their midpoint. The consequences of the edge effect may be excessive heating and coagulum formation during temperature-feedback power-controlled radiofrequency energy delivery if the temperature sensor is not optimally placed. With long coil electrodes, use of dual-edge temperature sensors results in comparable lesion efficacy but a much lower risk of overheating and coagulum or char formation. To optimize the safety profile of new electrode geometries, temperatures should be monitored at all electrode edges or transition points to prevent excess power delivery.

Footnotes
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The temperature change at any point in the body is represented by a partial differential equation called the bioheat transfer equation, which may be reduced to

\[ \rho c \frac{\partial T}{\partial t} = \nabla \cdot (\kappa \nabla T) + Q_p - Q_h \]  

(1)

Equation 1 equates the rate of change of temperature per time \( \frac{\partial T}{\partial t} \) to the heat conduction caused by a temperature gradient \( \nabla T \) (a derivative of temperature over three-dimensional space) and added heat from a source, \( Q_p \) (in the present study, radiofrequency energy), and the loss to convection, \( Q_h \). In Equation 1, \( \rho \) is the density, \( c \) is the specific heat, and \( \kappa \) is the thermal conductivity of tissue. During radiofrequency ablation, the heat loss due to convection, \( Q_h \), is proportional to the difference between tissue temperature and the temperature of the blood. However, the energy derived from resistive radiofrequency energy delivery, \( Q_p \), is dependent on geometric factors and is the product of the current density, \( J \), and the electric field, \( E \):

\[ Q_p = JE \]  

(2)

Ohm's law states that the current density is equal to the product of the electric field and the tissue conductance, \( \varsigma \):

\[ J = \varsigma E \]  

(3)

The substitution of Ohm's law into the energy equation gives a representation the heat energy from the electric field in terms of the square of the current density:

\[ Q_p = \frac{J^2}{\varsigma} \]  

(4)

By reducing the Laplace equation, which governs current density and electric field distribution, an expression for the current density can be derived:

\[ -\nabla J = 0 \]  

(5)

The \( \nabla \) term simply represents the derivative of its associated variable in three-dimensional space, so this equation expands to

\[ \delta J_x / \delta x + \delta J_y / \delta y + \delta J_z / \delta z = 0 \]  

(6)

This equation describes the current density at a point on the electrode surface that is related through Equation 4 to the heat produced in the tissue in the vicinity of that point on the electrode. Thus, the current density is distributed in the direction of and in the area of the highest spatial geometric gradient. The cylindrical portion of long electrodes is relatively smooth and would have a low spatial derivative compared with the geometrically sharp edges with high derivatives. This is known as the edge effect for antennae or electrodes. The finite element model solved in the present study confirms the edge effect during radiofrequency ablation with long electrodes. The heating (dependent on the square of the current density) is concentrated at the edges, where the geometric gradient changes acutely.

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References


2. Mitchell MA, McRury ID, Haines DE. Linear atrial ablations in a canine model of chronic atrial


26. Langberg JJ, Gallagher M, Strickberger SA, Amirano O. Temperature-guided radiofrequency catheter ablation with
nonuniform heating during radiofrequency catheter ablation with long electrodes: Monitoring the...
**Figure 2.** A representative photograph of the catheters tested in this study. The catheters had 7F shafts with 8F electrodes. They used bidirectional steering and two to six coil electrodes per catheter.
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Table 1. Biophysical Data Comparing Single-Thermistor Electrodes With Dual Thermocouple Electrodes

<table>
<thead>
<tr>
<th></th>
<th>Single-Thermistor Electrodes</th>
<th>Dual-Thermistor Electrodes</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean power, W</td>
<td>61.6 ± 38.0</td>
<td>50.7 ± 32.3</td>
<td><em>P</em> &lt; .0001</td>
</tr>
<tr>
<td>Mean temperature, °C</td>
<td>59.7 ± 9.3</td>
<td>68.1 ± 5.8</td>
<td><em>P</em> &lt; .001</td>
</tr>
<tr>
<td>RF energy duration, s</td>
<td>53.3 ± 15.6</td>
<td>59.3 ± 5.6</td>
<td><em>P</em> &lt; .001</td>
</tr>
<tr>
<td>Any impedance rise</td>
<td>176 of 395 (45%)</td>
<td>9 of 425 (2%)</td>
<td><em>P</em> &lt; .0001</td>
</tr>
<tr>
<td>Auto power cutoff (&gt;300 Ω)</td>
<td>41 of 395 (10%)</td>
<td>3 of 425 (1%)</td>
<td><em>P</em> &lt; .0001</td>
</tr>
<tr>
<td>Rapid rise in impedance</td>
<td>46 of 395 (12%)</td>
<td>0 of 425 (0%)</td>
<td><em>P</em> &lt; .0001</td>
</tr>
<tr>
<td>Slow rise in impedance</td>
<td>89 of 395 (23%)</td>
<td>6 of 425 (1%)</td>
<td><em>P</em> &lt; .0001</td>
</tr>
<tr>
<td>Temperature at impedance rise, °C</td>
<td>75.5 ± 9.0</td>
<td>77.8 ± 11.9</td>
<td><em>P</em> &lt; .0001</td>
</tr>
<tr>
<td>Coagulum on electrode¹</td>
<td>2.3 ± 1.5</td>
<td>0.6 ± 0.9</td>
<td><em>P</em> &lt; .001</td>
</tr>
</tbody>
</table>

RF indicates radiofrequency.

¹ See text for details on quantifying coagulum adherent to electrode.
Figure 5. Impedance (Imp), temperature (Temp), and power (Pow) tracings from atrial ablations with electrodes with a single centered thermistor. The top tracing is an example of sudden intermittent increases in impedance. Each rapid impedance rise is followed by a rapid rise in temperature causing a sudden decrease in power. The impedance then returns to baseline. The phenomenon causing this tracing could be intermittent boiling (which stops power reduction), changing catheter contact, or formation and embolization of adherent coagulum. The lower tracing is an example of a slow rise in impedance. In this example, the impedance rises slowly and continuously throughout energy delivery despite the steady-state maintenance of the 70°C target temperature, suggesting gradual accumulation of coagulum on the electrode.
**Table 2. Factors Influencing Coagulum Adherent to Long Electrodes**

<table>
<thead>
<tr>
<th>Factor</th>
<th>No Coagulum Present</th>
<th>Coagulum Present</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single-thermistor electrode</td>
<td>46</td>
<td>52</td>
<td>p&lt;0.0001</td>
</tr>
<tr>
<td>Dual thermocouple electrode</td>
<td>150</td>
<td>7</td>
<td>p=NS</td>
</tr>
<tr>
<td>Auto power cutoff (&gt;300 Ω)</td>
<td>2 of 196 (1%)</td>
<td>0 of 59 (0%)</td>
<td>p=NS</td>
</tr>
<tr>
<td>Rapid rise in impedance</td>
<td>4 of 196 (2%)</td>
<td>12 of 59 (20%)</td>
<td>p&lt;0.0001</td>
</tr>
<tr>
<td>Slow rise in impedance</td>
<td>4 of 196 (2%)</td>
<td>19 of 59 (32%)</td>
<td>p&lt;0.0001</td>
</tr>
<tr>
<td>Mean power, W</td>
<td>46.5 ±32.8</td>
<td>61.4 ±35.8</td>
<td>p&lt;0.003</td>
</tr>
<tr>
<td>Mean temperature, °C</td>
<td>67.9 ±6.6</td>
<td>62.2 ±6.7</td>
<td>p&lt;0.001</td>
</tr>
<tr>
<td>Temperature at impedance rise, °C</td>
<td>86.4 ±14.2</td>
<td>87.7 ±27.1</td>
<td>p=NS</td>
</tr>
</tbody>
</table>

1 Only those electrodes inspected after a single radiofrequency delivery were included in this analysis (see text for details).

2 Absent or trace fibrin.

3 Mild, moderate, or severe coagulum.