Irrigated Versus Dry Radiofrequency Epicardial Ablations in the Treatment of Atrial Fibrillation

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Background: Dry radiofrequency (RF) ablation is an established form of surgical treatment of atrial fibrillation and many groups have reported good clinical results (¹-⁴). One limiting factor of this type of ablation is that transmural lesions are not consistently obtained (⁴-⁶). Several reports on percutaneous ablation have shown that catheter irrigation leads to larger volume lesions than non-irrigated catheters (⁷-⁹). When the catheter is irrigated with cold saline the temperature at the electrode/myocardium interface is artificially lowered and the generator keeps applying power to the tissue (up to 150W) in an attempt to bring the surface temperature up to the target value set by the user. By irrigating the catheter one is actually bypassing a very important safety mechanism of the generator, the temperature control, and one can expect lot more power to be delivered to the tissue during irrigated ablations than during dry ones. It is therefore important to determine the power that can be applied in order to ensure that tissue integrity is not compromised.

There are several in-vitro and in-vivo studies of the impact of irrigation on RF percutaneous ablation using animal models (¹⁰-¹⁵) and computation models simulating irrigated epicardial ablations (¹⁶). However, there are to our knowledge no experimental studies of the effect of catheter irrigation on the lesion created by epicardial ablation on human tissues. We studied in-vitro and in-vivo the effects of external irrigation of the catheter on the sub-endocardial (Tend) and sub-epicardial (Tepi) temperatures during epicardial ablation under power control. We assessed lesion depth and compared the results with dry RF epicardial ablation under temperature control.

Methods: Seventeen fragments of atrial tissue from organ donors were used. Three points were marked on the epicardium approximately 5 mm apart along a straight line. The atrial thickness was measured at these points, with an error of 0.3 mm, using a digitizer Threespace (Polhemus, Vermont, CA) and T-type thermocouples (0.10mm thick) were inserted sub-endocardially and sub-epicardially at the three points. The thermocouples were connected to a special unit (EP Technologies, San Jose, California) for signal processing. The unit was connected to a PC for data acquisition and real time graphical display of intra-tissue temperatures.

The atrial fragments were mounted on a custom built bath filled with circulating saline at 37°C. A roller pump was used to keep the saline circulating with a debit of 3l/min, in order to simulate conditions found in surgical RF ablation on a beating heart. RF currents were delivered to the tissue between the catheter placed over the three points marked on the epicardial surface and the dispersive electrode placed underneath the bath. We used a malleable Thermaline catheter connected to a RF generator (EPT Technologies, San Jose, California). The generator output was connected to another PC so that the variations of power, impedance and temperature of the tissue surface (measured by the sensors in the catheter) were displayed in real time by means of EPT graphics software. Instant and average root mean square values of current intensity were also displayed. It was therefore possible to compare the variation of the temperature at the endocardial surface with that of the temperatures measured simultaneously by the thermocouples placed sub-endocardially and sub-epicardially.

We performed five dry ablations always under temperature control with a target temperature of 85°C and 12 irrigated ablations under power control. For the latter the RF generator was set at a target temperature of 85°C with power limited to 30W (n=4), 50W (n=4) and 70W (n=4). Each ablation lasted 2 minutes and the electrodes were manually irrigated with saline at ≈ 12°C. Similar irrigated applications, plus one dry at Ttarget=85°C, were performed in each of 4 pigs. Thirteen in-vitro and eight in-vivo lesions were histologically evaluated.
**Results:** *In-vitro:* During the dry ablations the temperature at the tissue surface was kept at the target ablation value, whereas it never reached the target value during any of the irrigated ablations. In dry ablations the power delivered by the generator was much lower than in irrigated ones (Fig. 1).

![Fig. 1. Power, impedance and temperature at the tissue surface measured during the ablation. Irrigation of the catheter was stopped at the cursor. Note the sudden rise in the temperature at the tissue surface and the drop in power once the target temperature was attained. Once the catheter irrigation was stopped the current intensity (I\text{rms}) is nearly half the average current intensity during the irrigated ablation (I\text{rms}).](image)

Maximal $T_{\text{send}} = 95^\circ\text{C}$ for dry ablations and $98.2^\circ\text{C}$ for irrigated ones. Table I shows the average values of sub-endocardial temperatures, power and current intensity measured during the dry and irrigated ablations.

<table>
<thead>
<tr>
<th>Abl. type</th>
<th>$P_{\text{lim}}$ (W)</th>
<th>$T_{\text{send}}$ (°C)</th>
<th>$P_{\text{av}}$ (W)</th>
<th>$I_{\text{av}}$ (A)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dry</td>
<td>No</td>
<td>52.4±6.9</td>
<td>12.3±1.5</td>
<td>0.35±0.06</td>
</tr>
<tr>
<td>Irrig</td>
<td>30</td>
<td>63.6±7.0</td>
<td>28.5±0.6</td>
<td>0.69±0.04</td>
</tr>
<tr>
<td>Irrig</td>
<td>50</td>
<td>57.8±4.1</td>
<td>45.3±0.5</td>
<td>1.03±0.13</td>
</tr>
<tr>
<td>Irrig</td>
<td>70</td>
<td>62.5±4.0</td>
<td>62.7±1.5</td>
<td>1.28±0.14</td>
</tr>
</tbody>
</table>

$P_{\text{lim}}$: power limit; $T_{\text{send}}$: sub-endocardial temperature; $P_{\text{av}}$: average power; $I_{\text{av}}$: average current.

Lesions induced by dry ablations were $1.50±0.32\text{mm}$ deep and none were transmural whereas lesions from irrigated ablations were $1.25±0.00\text{mm}$ deep at $30\text{W}$, $1.84±0.04\text{mm}$ at $50\text{W}$ and $2.18±0.16\text{mm}$ at $70\text{W}$. Five out of the eight irrigated lesions were transmural: 3 induced with power limited to $30\text{W}$, 1 with power set at $50\text{W}$ and 1 at $70\text{W}$. Furthermore, histology showed that lesions induced by irrigated ablations with power limited to $70\text{W}$ showed a higher degree of tissue destruction, including loss of myocardial tissue, than the ones induced at $30$ and $50\text{W}$. However, tissue loss was also observed in one lesion induced at $30\text{W}$ in a very thin atrial wall ($1.63\text{mm}$). In applications at $70\text{W}$ some medium size vessels showed damage of the adventitia and of the media. These features were never observed in lesions induced by dry ablations under temperature control.

**In-vivo:** Similarly to the in-vitro ablations the temperature at the tissue surface was always below the target ablation temperature during the irrigated ablations. Irrigated ablations in pigs did not always produce deeper lesions than dry ablations: $2.19±0.65\text{mm}$ vs. $1.64±0.38\text{mm}$. Moreover, increasing the power in irrigated ablations did not always lead to deeper lesions. However, irrigated ablations caused more destruction of the atrial tissue than dry ones. Loss of tissue was never observed in dry ablations, whereas tissue loss and damage of medium size vessels were seen in samples from irrigated ablations particularly at higher values of power.

**Conclusions:** Throughout all the irrigated ablations the temperature at the tissue surface was kept well under the set temperature by the cold irrigating saline, which means that the temperature control mode was effectively bypassed. By limiting the maximum power applied and choosing a target ablation temperature sufficiently high ($85^\circ\text{C}$) we basically switched from the temperature control mode to a power control mode. We can thus avoid excessive power being applied to the tissue during the irrigated ablations.

The fact that under the same ablation conditions some lesion are transmural and others are not depends on the composition of the atrial wall and not just on its thickness. This may account for the large variability in the clinical results reported by different groups. Since the histological features vary from patient to patient it is logical to assume that this feature is one of the reason for such variability. It is presumed that the primary mechanism of tissue injury caused by radiofrequency currents is thermal. However, it is possible and it has been suggested, that besides...
an increase in temperature the actual conduction of current through the tissue will cause alterations at the cellular level, that are more difficult to quantify than the actual temperature increase and which may influence the formation of the lesion. Moreover the ischemia caused by the thrombosis of the microcirculation may be another mechanism of tissue injury. The presence and location of microcirculation thrombosis may explain why acute histology does not necessarily correlate exactly with chronic histology (R. Gouveia et. al, unpublished data). From a biophysical point of view, it is likely that atrial tissues with different compositions will have different electrical properties, which will affect the conduction of RF electrical current in the tissue.

A limitation of the present study lies in the small number of animals from which we could obtain histological information hence limiting the comparisons between dry and irrigated ablations in the same animal. We believe that the tissue destruction observed in-vitro and in-vivo is due to the higher values of the RF current flowing through the tissue during irrigated ablations. We conclude that switching from temperature to power control during irrigated ablations and setting a low upper limit to the power delivered to the tissue is extremely important if tissue integrity is to be maintained. Had we not worked under power control during the irrigated ablations, the generator would have kept applying as much power as possible in an attempt to bring the surface temperature up to the target value. The intensity of the current circulating through the tissue would have been much higher and serious tissue damage could have resulted.

Further studies are needed to evaluate the importance of the duration of the ablation. In fact the duration of the ablation is not a major issue in dry ablations under temperature control because little power is applied to the tissue once the temperature at the tissue surface has reached the value set by the user. Therefore the current intensity only reaches high values in the first few seconds it takes the set temperature to be reached. Conversely, in irrigated ablations the current intensity is higher throughout the ablation even under power control. It is possible that shortening the ablation time will avoid tissue destruction while still leading to higher intra-tissue temperatures than the ones reached with dry ablations.

References


