Ablation of cardiac arrhythmias –
energy sources and mechanisms
of lesion formation

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Throughout the history of the use of antiarrhythmic drugs to suppress cardiac arrhythmias, the prevailing opinion has been that this therapeutic option produces less than optimal results in terms of durable arrhythmia suppression and the magnitude and the spectrum of side effects. The era of ablative therapy of cardiac arrhythmias began in 1968 when Cobb et al. reported successful surgical interruption of the Kent bundle in a patient with Wolff-Parkinson-White syndrome [9]. The profoundly invasive character of antiarrhythmic surgery involving open thoracotomy limited the widespread use of such therapy and accelerated the efforts for development of closed-chest, catheter-based ablative therapy.

In general, all ablative-type therapies for cardiac arrhythmias consist of the delivery of some source of energy within the heart at such a magnitude that it causes local myocardial destruction of anatomic regions critical for abnormal impulse generation and/or propagation. The ultimate aim of these destructive lesions is either silencing the foci responsible for abnormal automaticity or interruption of the reentry circuits responsible for arrhythmia genesis or continuation. The first energy form used for ablation was high voltage, direct current catheter ablation [74]. With this technique, internal shocks were applied to specific regions in the heart which led to local destruction through a combination of electrical, thermal and mechanical (barotrauma) factors. Catheter-based direct current ablation had a limited use due to its uncontrolled character, reported serious side effects and the fear of using it on thin-walled atrial or coronary sinus structures for ablation of supraventricular tachycardias. Soon after the introduction of the direct current ablation technique, several other sources of ablative energy such as radiofrequency [3, 31], cryotherm [22], microwave [87], laser [49] and intracoronary alcohol infusion [6] were used for ablation of cardiac arrhythmias. However, only catheter-based radiofrequency (RF) ablation and to some extent cryothermal ablation found the widest clinical applications, and they are currently the most commonly used and the most standardized catheter-based ablation techniques for ablation of cardiac arrhythmias. In this Chapter, only catheter-based ablation procedures will be described.

2.1 Radiofrequency ablation

2.1.1 Physical aspects

Radiofrequency ablation (RF) is the most widely accepted catheter-based treatment for supraventricular and ventricular arrhythmias and is the energy source most familiar to cardiologists. The frequency of the RF current, mostly used in ablation of cardiac arrhythmias is 300 to 1000 kHz. Lower frequency alternating currents (<100 kHz) usually stimulate excitable cells and produce pain and muscle contractions or ventricular fibrillation when applied to myocardium. RF current is delivered to specific regions within the heart through transvenous electrode catheters with a catheter tip between 4 and 10 mm (Figure 2.1). As the high-frequency current passes through the living tissues, electrically charged carriers (ions) tend to follow the changes in the direction of the alternating current. This leads to conversion of the electromagnetic energy into the mechanical energy of ions and heat production. This type of current-mediated heat production is called Ohmic or resistive heating. Resistive heating is the primary mechanism by which cardiac lesions are produced. The RF energy is emitted from the catheter tip over a very small area and, thus, has high current density. This high-density cur-
rent encounters the tissue, which acts as a resistor leading to heat generation. RF current may be delivered in the unipolar or bipolar mode. In the unipolar mode which is most commonly used, the RF is concentrated at the ablation surface (catheter tip–tissue contact), disperses throughout the body and exits to a large surface electrode (indifferent, groundpad or dispersive electrode) positioned distally on the body surface. The surface area of the groundpad electrode should be 100 to 250 cm$^2$. The electrode geometry and size as well as the close skin contact helped by applying electrocardiographic gels produce a very low density current avoiding any substantial local heating and potential skin burning. The groundpad electrode may be placed in any convenient place on the patient’s skin; however, the placement on the posterior aspect of the chest is preferred. Since the RF current is alternating, the selection of polarity of the connections with the generator source has no importance.

2.1.2 Factors that influence lesion formation

The effects the RF energy on myocardial tissue depend on multiple factors such as the current density, the surface area of the active electrode, the quality of electrode-tissue contact, the duration of current application, histological characteristics of the tissue including blood supply and proximity to major blood vessels, the degree of tissue heating and the degree of heat dissipation (proximity to intramyocardial major blood vessels or ablating in cardiac regions with rapid blood flow).

As stated above the main mechanism of lesion formation with RF current ablation is by resistive heating. However, the resistive heating is effective only for distances less than 2 mm from the RF current source (electrode tip). This is due to the fact that current density, the main driving force of resistive heating, is diminished markedly with the increase in the distance from RF current source. Deeper penetration of heat within the myocardium is enabled by thermal conduction or heat transfer from the zone with higher temperature to zones with lower temperatures. While resistive heating in myocardial regions close to the RF current source is rapid, passive heat transfer to deeper layers is a slow process. An experimental study by Wittkampf et al. showed that the intramyocardial temperature at a distance 3 mm from the catheter tip increases progressively with the increase in the time during which RF current is applied from 10 to 60 seconds [91]. This study showed that in order to produce an effective RF ablation lesion the current should be applied for at least 60 seconds because a steady state is not achieved until 40–50 seconds of RF energy application. Another factor that conditions the amount of heat transferred to deeper layers is the temperature at the zone of resistive heating. The greater the temperature in the resistive heating zone the greater is the amount of heat transfer to deeper myocardial layers. The heat transfer continues even after discontinuation of RF current delivery. This may result in lesion volume expansion after RF current cessation which may have clinical consequences (i.e., arrhythmia termination or side effects seconds after current delivery interruption) [90]. The optimal temperature for human RF ablation is not entirely clear. In general catheter-based endocardial ablation is performed for 60 seconds at a target temperature of 50 to 70 °C (Table 2.1). However, there is great variability in the tissue characteristics according to the nature of structural heart disease. As a general rule, however, temperatures greater than 95 °C should be avoided due to risk of tissue disruption.

The dimensions and the volume of the ablated tissue are proportional to the delivered power [89]. The increase of power invigorates the heat production and results in deeper penetration of heat with destructive capability. Energy delivery is regulated by temperature control that is based on fixing a target temperature and adjusting the RF energy to maintain the