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Radiofrequency ablation with a vibrating catheter: A new method for electrode cooling

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ABSTRACT

A new electrode cooling system using a vibrating catheter is described for conditions of low blood flow when saline irrigation cannot be used. Vibrations of the catheter are hypothesized to disturb blood flow around the electrode, leading to increased convective cooling of the electrode. The aim of this study is to confirm the cooling effect of vibration and investigate the associated mechanisms. As methods, an in vitro system with polyvinyl alcohol-hydrogel (PVA-H) as ablated tissue and saline flow in an open channel was used to measure changes in electrode and tissue temperatures under vibration of 0–63 Hz and flow velocity of 0–0.1 m/s. Flow around the catheter was observed using particle image velocimetry (PIV). Results show that under conditions of no flow, electrode temperatures decreased with increasing vibration frequency, and in the absence of vibrations, electrode temperatures decreased with increasing flow velocity. In the presence of vibrations, electrode temperatures decreased under conditions of low flow velocity, but not under those of high flow velocity. PIV analyses showed disturbed flow around the vibrating catheter, and flow velocity around the catheter increased with higher-frequency vibrations. In conclusion, catheter vibration facilitated electrode cooling by increasing flow around the catheter, and cooling was proportional to vibration frequency.

1. Introduction

Radiofrequency (RF) catheter ablation is used in the treatment of several tachycardias and is delivered by electrodes on catheters to produce lesions on cardiac tissues. Deeper lesions may be necessary for post-infarction ventricular tachycardia because portions of the re-entry circuit are often located deep in the endocardium [1,2]. Although lesion sizes can be increased with higher power [3], high power delivery is limited by overheating of the electrode–tissue interface (interface temperature $\geq$ 80 °C), which causes thrombus formation [4,5]. Thus, to deliver higher RF power to cardiac tissue, electrode–tissue interface overheating must be prevented.

Several strategies have been developed to prevent overheating at the electrode–tissue interface, such as temperature control, larger electrodes, and active electrode cooling. In the temperature-control approach, a thermistor inside the electrode is used to monitor electrode temperature. The RF generator is adjusted to deliver the maximum output power, which maintains the electrode at a target temperature (commonly 55 C or 60 °C [5]). The temperature of the electrode–tissue interface is limited by that of the electrode and is determined by heating of the tissue that is in contact with the electrode and cooling from the blood flow around the electrode [4,6,7]. Because cooling decreases under conditions of low blood flow, maintenance of target electrode temperatures requires reduced output power. Thus, RF power is automatically reduced by the RF generator under conditions of low blood flow.

Larger electrodes (8 or 12 mm in length) may produce deeper lesions, owing to increased passive (as opposed to active) electrode cooling by the effect of the circulating blood and increased electrode–tissue interface areas [8]. However, increased electrode cooling is effective only in temperature control mode [9], and the cooling effect is limited under conditions of low blood flow.

Abbreviations: PVA-H, polyvinyl alcohol-hydrogel; PIV, particle image velocimetry.

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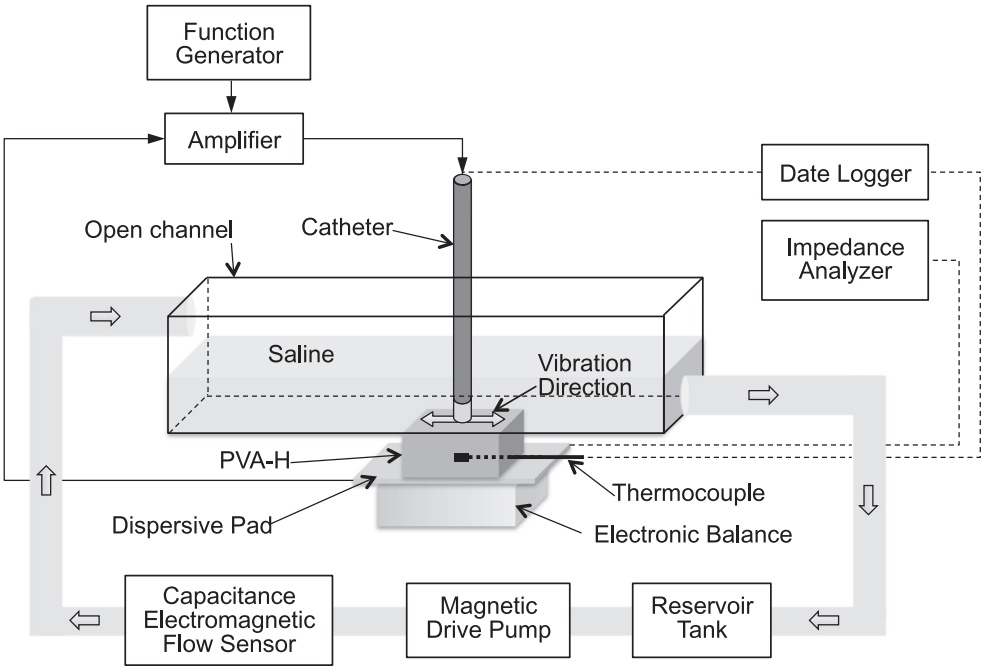

Fig. 1. Schematic representation of the in vitro ablation system.

Increases in electrode–tissue interface area depend on catheter orientation, which is hard to control in the clinical setting [9].

Currently, active electrode cooling is achieved using internal closed-loop cooling or open-irrigation cooling, and both increase lesion size by actively cooling the electrode [10–15]. Internal closed-loop cooling uses circulating fluid inside the catheter to directly cool the interface. However, the cooling may be limited at contact sites of only metal and tissue [16] and the majority of interface cooling is still derived from blood flow [5,16]. This insufficient cooling may result in a high incidence of thrombus formation under conditions of low blood flow [4]. With open-irrigation cooling, saline is flushed through holes arranged in the electrode and cools the electrode, the surrounding blood, and the tissue surface. This method provides sufficient environmental cooling from irrigation for high-power delivery to myocardial tissue, with markedly decreased incidence of thrombus formation even under conditions of low blood flow [5,11,13,14,17]. However, approximately 1500-ml saline infusions during open irrigation cooling can aggravate heart failure and pulmonary edema [18,19].

Thus, when open irrigation cannot be used under conditions of low blood flow, alternative strategies for preventing overheating of the electrode–tissue interface may be required. In the present study, we hypothesized that electrode cooling can be increased without use of cooling fluid and that convective cooling can passively occur via blood circulating past the electrode or by sliding the catheter contact [20]. Larger electrodes increase convective cooling by passively increasing the surface area of blood flow. However, the proposed novel strategy may actively cool the electrode using a vibrating catheter tip that increases convective cooling by blood flow. Thus, we hypothesized that catheter vibrations disturb flow around the electrode and increase convective cooling of the electrode. The present experiments showed a cooling effect of vibration, and mechanisms of cooling were investigated. We used a flow visualization system using poly (vinyl alcohol) hydrogel (PVA-H) as a model of ablated tissue and used an open channel to measure electrode and tissue temperature at a depth of 2 mm under various vibration frequencies and flow velocities. Flow around the catheter was then observed by particle image velocimetry (PIV).

2. Materials and methods

2.1. Materials

Tissue temperature is an important parameter for the study of dynamic heating processes during RF ablation [21]. Several in vitro studies have measured tissue temperatures in animal myocardium [6,12,13,22,23], although the electrical and thermal properties of these tissues are not uniform and measurements of internal myocardial temperatures lack accuracy because of irregular shapes and opacities [21]. An in vivo model of canine thigh muscle has been widely used [4,5,8–10,24–26] but has limitations similar to those of animal myocardium. In the present study, PVA-H models provided more uniform electrical and thermal properties, also (1) higher transparency, and (2) lower surface friction for catheter placement than myocardium [27], (3) dynamic viscoelasticity similar to biological soft tissue [28], and (4) catheter contact similar to that of heart tissue, with comparable levels of distortion due to catheter load [29].

2.2. PVA-H preparation

PVA (15 wt%, JF17, JAPAN Vam & Poval Co., Ltd., Japan) and NaCl (2 wt%) were dissolved in dimethyl sulfoxide aqueous solution (80 wt%, Toray Fine Chemicals Co., Ltd., Japan) at 120 °C for 2 h. NaCl was used to adjust the impedance of PVA-H to that of myocardium [30]. Molten PVA solution was poured into molds and cooled in a freezer at −20 °C for 24 h. The resulting PVA-H was then stored at room temperature (25 °C) for 2 h before experiments.

2.3. In vitro ablation system

The in vitro ablation system used is shown in Fig. 1. Saline in the reservoir tank was driven through a capacitance electromagnetic flow sensor (FD-M5AY, Keyence Corporation, Japan) into the open channel using a magnetic drive pump (CP20-PPRV-10, Nikkiso Eiko Co., Ltd., Japan). The open channel (L × W × H,