

A COMBINED CRYOSURGICAL/HYPERTHERMIA SYSTEM USING THE PELTIER THERMOELECTRIC EFFECT

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ABSTRACT

To conveniently accommodate to the needs for the combined cryosurgical and hyperthermia treatments, a hand-held cryoprobe system for general-purpose thermotherapy (including both cryosurgery and hyperthermia) has been developed using multi-stage peltier thermoelectric coolers/heaters in this study. The advantages of the present cryosurgical/hyperthermia system consist of hand-held, fully electrically-controlled, no movement of cooling medium and refrigerant-free. To test the performance of the new system, a series of experiments are respectively performed in air and tissue phantom. The results indicate that the temperature at the probe tip of the new system can easily reach -60°C during freezing and above 60°C during heating. It suggests that the new cryosurgical/hyperthermia system would allow reliable combined cryosurgery and hyperthermia, especially when cryo-refrigerants are not available.

1. INTRODUCTION

Cryosurgery is using freezing temperatures for the therapeutic destruction of diseased tissues. Applications of this treatment are used quite widely in superficial diseases such as tumors, warts and some other dermatological conditions (Rubinsky, 2000). In contrast to cryosurgical therapy, heating is also an effective way of selective treatment of diseased tissues, which is termed as hyperthermia (Falk *et al.*, 2001). Up till now, a wide variety of cryosurgical and hyperthermia apparatus have been established separately such as freezing by dry ice, Joule–Thomson effect (expansion of compressed gas), liquefied gas, and Peltier thermoelectric (TE) cooler (Baust *et al.*, 1997; Holman *et al.*, 1997; Hewitt *et al.*, 1997), or heating by microwave, ultrasound, radiofrequency, electromagnetic source, laser, and hot medium (Roemer, 1999; Deng and Liu, 2002; Wood *et al.*, 2002).

Cryosurgical and hyperthermia treatments had been performed separately for several decades until several years ago it was realized that combining cryosurgery and hyperthermia would significantly improve the treatment effect of tumor. At that time, all of the available cryosurgical or hyperthermia systems are only capable of performing a single freezing or heating function. To accommodate the needs for the combined freezing and heating treatment for diseased tissues, Liu *et al.* (2003) developed the first cryoprobe system with a powerful heating feature. After this, Goldberg *et al.* (2004) reported a hybrid system of radiofrequency ablation and cryosurgery. Both systems have particularly emphasized treating the tumor in deep tissue through minimally invasive approaches (Liu *et al.*, 2004; Goldberg *et al.* 2004), and have very complex structures which will result in high cost. However, for the treatment of many superficial diseases,

such devices are too expensive and then unnecessary. Development of combined system of cryosurgery and hyperthermia for these applications is still needed. In addition, liquid nitrogen is indispensable to Liu's system while compressed argon is the necessity of Goldberg's system. Consequently, the freezing function of both systems cannot work when such cryo-refrigerants are not available.

In the past, besides expansion of compressed gases and liquefied gases were used as cryogens to design cryoprobes, other cryoprobes had been developed which used different cooling techniques. Alternative techniques have included the use of a Peltier TE cooler and a cryogenic heat pipe (Orpwood, 1981). The heat pipe cryoprobe however appears to have a limited heat extraction capability. The TE cryoprobe, which can easily generate freezing temperature with a relatively high capability, possesses the advantages of low-cost, moving-part-free, noise-free, refrigerant-free and temperature-precisely-controllable. At present, the TE cryosurgical device has been successfully used in treatment for many diseases, especially for dermatological conditions (Almond-Roesler and Zouboulis, 2000). Besides as cooler, a TE device can also be used as heater which is in principle the same as that used as TE cooler except that the polarity of the power supply is reversed (Xu *et al.*, 2007). Based on the above consideration, a hand-held cryoprobe system aiming at performing both cryosurgery and hyperthermia was then developed using multi-stage peltier TE modules in this study. To test the capabilities of freezing and heating of this cryoprobe system, several preliminary experiments were performed in air and tissue phantom, respectively.

2. DEVICE CONSTRUCTION AND EXPERIMENTAL DESIGN

2.1. Device Construction

As shown schematically in Fig. 1, the freezing-heating system consists of five major parts: DC power, probe, water-flow based heat sink, data acquisition unit and computer. The freezing-heating probe contacts with a plurality of stacked thermoelectric modules coupled to a water-flow based heat sink (which dissipates heat to water when the probe is cooled, and absorbs heat from water while the probe is heated). The polarity of the DC power supply can be reversed by the switch to achieve the conversion between freezing and heating for the probe. Using the switch, multiple cycles of freezing and heating can also be actualized with the aim of enhancing the treatment effect.

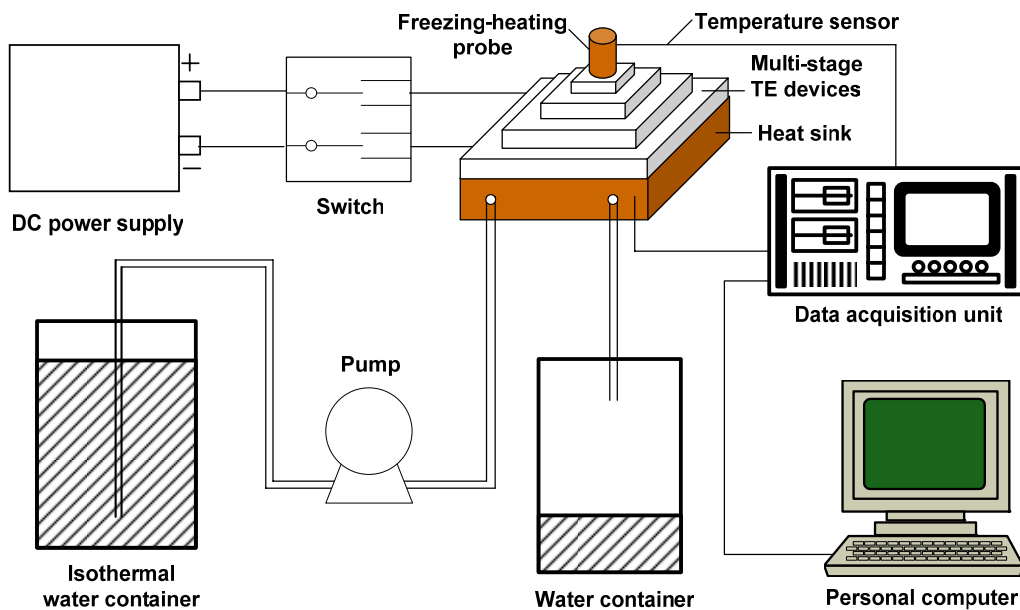


Figure 1. Schematic representation of the new freezing-heating probe system.

Figure 2 shows the prototype of the TE freezing-heating device. As the powers of freezing and heating supplied by the TE modules are transmitted to the probe through heat conduction, the probe was made of copper, and heat conducting oil was spread at the contacting surfaces of probe and the TE modules, to effectively conduct the freezing or heating power in this study. It would not be a major concern, although copper may be harmful to the human body, since the TE cryoprobe has addressed itself to the applications for dermatological conditions. From the clinical point of view, the size of the probe should be as small as possible for exactly treating diseased tissues. In the present study, the diameter of the probe was fabricated at 4.5 mm. For different sizes of diseased tissues, the probe can be easily fabricated smaller or larger.

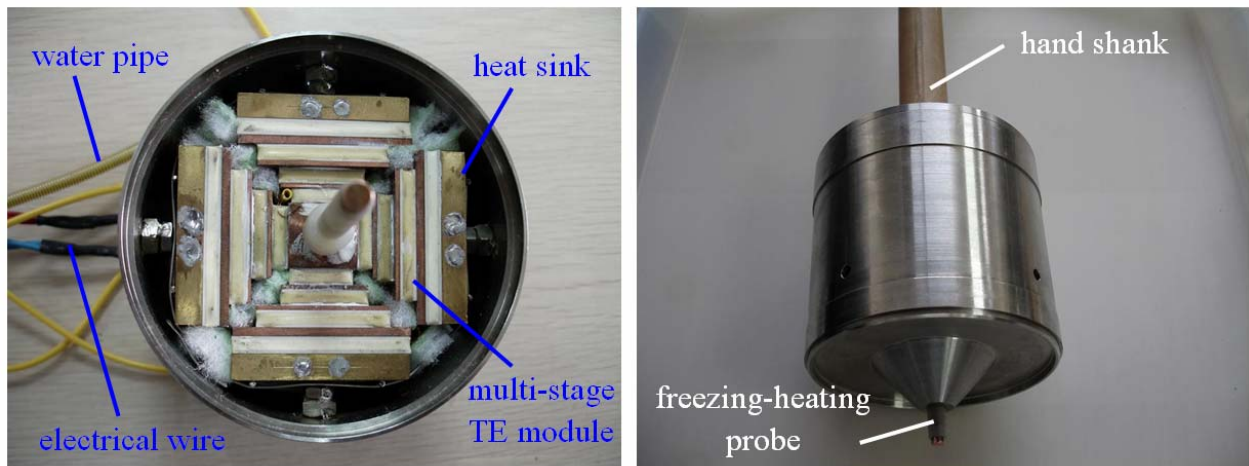


Figure 2. Prototype of the new freezing-heating device.

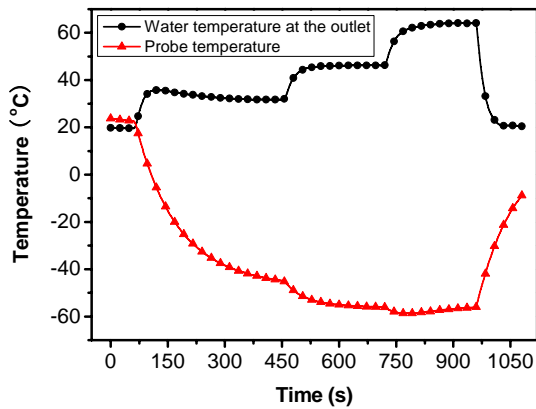
2.2. Experimental Design

In this study, temperatures were measured with copper-constantan thermocouples connected to an Agilent 34970A Data Logger. The thermocouples were calibrated and an accuracy of $\pm 0.1^\circ\text{C}$ was obtained. To test the capabilities of freezing and heating of this system, probe temperature and water temperature at the outlet were particularly monitored. Performances of freezing and heating of this TE probe were evaluated by a series of experiments in air and tissue phantom, respectively. During testing, the effects of electrical current supplied for the TE device, water temperature at the inlet and water flow flux on the performance of the TE cryoprobe were seriously taken into account.

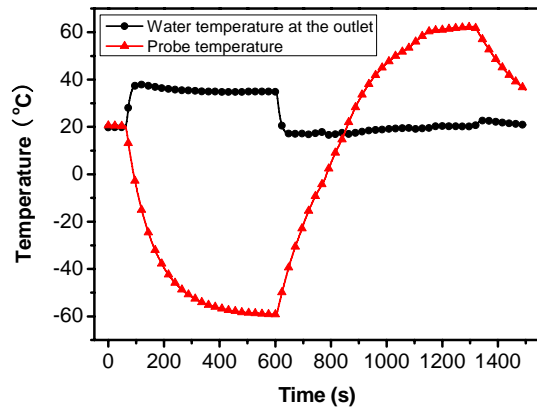
3. RESULTS AND DISCUSSION

Although a large amount of experimental data have been collected by considering the effects of electrical current (I_s), inlet water temperature (T_i) and water flow flux (Q_f), only several typical results (as shown in Fig. 3) are given here for illustration purpose. During experiments, the freezing performance of the TE cryoprobe was first tested in air under different electrical currents supplied for the TE device. The representative result is depicted in Fig. 3(a), in which the DC power was switched on at the time of 1min, and the initially supplied current was 2A. The electrical current was increased to 3A at 7.5min and 4A at 12min, respectively. At 16min, the DC power was switched off. It can be found from Fig. 3(a) that the probe is easily cooled to below -40°C even for relatively low supplied current. In dermatological applications, -40°C is low enough for effective treatment (Almond-Roesler and Zouboulis, 2000). When the current was increased to 3A, the temperature of probe tip quickly dropped from -40°C to -56°C . It indicates that the present TE cryoprobe

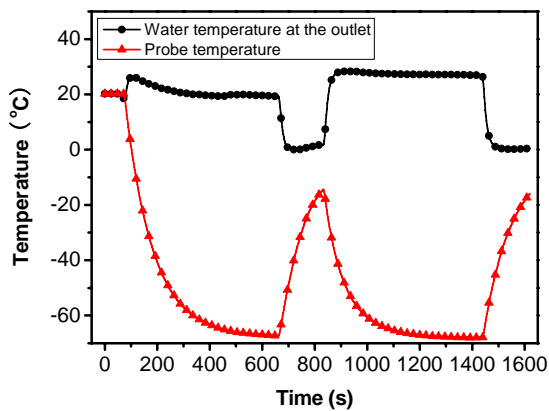
has a good freezing performance. Although further drop in the probe tip temperature was still realized by increasing the supplied current, the value of temperature drop is already very small (only about 1°C drop when the current was increased to 4A) for the given conditions. It can also be found from Fig. 3(a) that with the increase of the electrical current, the temperature difference between water at the outlet and probe tip also sharply increases, and the maximum temperature difference is about 121°C. Such a large temperature difference will consequently result in larger electrical energy consumption. In addition, high water temperature at the outlet accompanying with large temperature difference may limit the freezing performance of the TE cryoprobe.



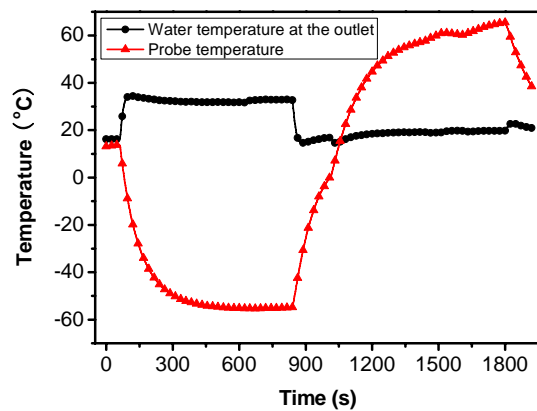
(a) $Q_f = 1.5\text{g/s}$, $T_i = 20^\circ\text{C}$



(b) $Q_f = 2.5\text{g/s}$, $T_i = 20^\circ\text{C}$



(c) $Q_f = 2.5\text{g/s}$, $T_i = 0^\circ\text{C}$



(d) $Q_f = 2.5\text{g/s}$, $T_i = 20^\circ\text{C}$

Figure 3. Temperature responses at the probe tip and the outlet of water, in which (a), b), and (c) are typical results of experiments in air, and (d) shows typical result of experiments performed on tissue phantom.

Theoretically, both increasing the water flow flux and decreasing the temperature of water at the inlet can reduce the temperature difference, and then save electrical energy and enhance the cooling performance of the TE system. Figure 3(b) shows the results in air including both freezing and heating for the case applying a larger water flow flux (2.5g/s). The DC power was switched on at the time of 1min, and the supplied current during freezing was 3A. At 10min, the DC power was switched off, and then the freezing was stopped. At 13min, the DC power was reversely switched on, and then heating of the TE probe started. The

supplied current during heating was just -0.5A . At 22min, the DC power was switched off. It can be found from Fig. 3(b) that by increasing water flow flux to 2.5g/s , the temperature of probe tip can reach to -60°C , which is 4°C lower than that when using 1.5g/s water flow flux at the same supplied current and inlet water temperature. In the same way, the maximum temperature difference between water at the outlet and probe tip, which reads 94°C from Fig. 3(b), also decreases largely. To avoid overheating of the inner layer of the Peltier elements during heating operation, the electrical current supplied for the TE device was only -0.5A . It is indicated in Fig. 3(b) that the temperature of probe tip can be easily elevated above 60°C , even applying such a low electrical current. In fact, the temperature of probe tip can be further elevated to a higher value, just increasing the supplied electrical current. The above results indicate that the present TE cryoprobe has excellent performances in both freezing and heating.

The results depicted in Fig. 3(c) are the temperature responses at the probe tip and the outlet of water for the case applying a lower water temperature (0°C) at the inlet. The DC power was switched on at the time of 1min, and the supplied electrical current was 3A . At 11min, the DC power was switched off. At 14min, the DC power was again switched on, and the supplied current was increased to 4A . At 24min, the DC power was switched off. It can be found from Fig. 3(c) that the temperature of probe tip can reach to -67°C when 3A current was supplied, which is 7°C lower than that when using 20°C of inlet water temperature at the same supplied current and water flow flux. Such results further indicate that the present TE cryoprobe has excellent freezing performance. When 4A current was supplied, the lowest temperature of probe tip is -68°C , only 1°C lower than that when 3A current was used. However, the maximum temperature difference between water at the outlet and probe tip increases from 86°C to 95°C . It indicates that most of the increased power supply has just been consumed to elevate the temperature of cooling water. Therefore, effective enhancement of the freezing performance of the TE cryoprobe cannot be achieved only by increasing the electrical current supplied to the TE modules. For a TE system, when inlet water temperature and water flow flux are given, there should be an optimal parameter of electrical current for the TE cryoprobe to produce an optimal freezing performance. Such issue is very important but beyond the focus of this study, and it will be addressed in the near future.

Besides the experiments performed in air, the experiments in tissue phantom were also conducted to test the capabilities of freezing and heating of the present TE probe. The reason for choosing tissue phantom as experimental material is that the ice ball formed during freezing will be visible in the tissue phantom, and that the tissue phantom will possess the thermal properties similar to that of human tissue through appropriate proportioning of gelatin and water. The typical result is depicted in Fig. 3(d), in which the DC power was switched on at the time of 1min, and the supplied current during freezing was 3A . At 14min, the DC power was switched off, and then the freezing was stopped. At 17min, the DC power was reversely switched on, and then heating of the TE probe started. The supplied current during heating was just -0.5A , same as that for experiments performed in air. At 30min, the DC power was switched off. It can be found from Fig. 3(d) that the temperature of probe tip can reach to about -56°C , which is only 4°C higher than that for experiments performed in air under the same parameters of electrical current, inlet water temperature and water flow flux. Figure 4 shows the picture of ice ball generated by the TE cryoprobe in tissue phantom, which was snapshotted at 11min (only after 10min' freezing). The diameter of the ice ball is about 18mm , which is large enough for treatment of warts, small superficial tumors, and some other dermatological diseases. During heating, the temperature of probe tip can also be easily elevated above 60°C by supplying -0.5A electrical current. It is further indicated from these results that the present TE cryoprobe has excellent performances in both freezing and heating.



Figure 4. Ice ball generated by the TE system in tissue phantom ($I_s = 3A$, $Q_f = 2.5g/s$, $T_i = 20^\circ C$)

The above results have shown that the performance of the present TE cryoprobe can be improved by appropriately changing the parameters of electrical current, inlet water temperature and water flow flux. It should be pointed out that besides these approaches, shortening of the length of probe (which used in this study is 30mm) is also a feasible method, since the powers of freezing and heating supplied by the TE modules are transmitted to the probe tip through heat conduction. In addition, optimization of the flow channel in the heat sink is another alternative technique for further improving the performance of the TE cryoprobe. Further efforts are strongly needed to probe into such important issues.

4. CONCLUSIONS

In this study, a new cryoprobe system capable of performing the combined treatments of cryosurgery and hyperthermia was successfully developed using multi-stage peltier TE modules. This closed system generates freezing temperatures by Peltier effect, and therefore does not involve the use of any cryo-refrigerants. The new system also represents a cost-effective improvement in the management of the instrument for the combined cryosurgical and hyperthermia treatments. The experiments performed in air and tissue phantom indicate that the present TE cryoprobe has excellent performances in both freezing and heating. It should be pointed out that, although the effects of electrical current, inlet water temperature and water flow flux on the capabilities of the TE cryoprobe were already included in the present study, additional efforts are still needed to further understanding the effects of the length of cryoprobe and the structure of flow channel of heat sink, to obtain optimal freezing/heating results when using this TE system.

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