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Monitoring temperature of a heating needle and surrounding blood during interventional whole body hyperthermia therapy

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Abstract

Recent studies confirmed the feasibility of interventional whole body hyperthermia (IWBH) through directly heating the blood flow via minimally invasive heating needles. Among the factors to determine successful IWBH, a most challenging task is to *in situ* monitor the temperatures of the heating needle and the blood running across it. In this paper, an indirect method was developed which can approximately measure the temperatures of the needle while performing its heating role. Meanwhile, the information acquired in the simulated experiments and theoretical analyses was combined to evaluate the temperature variation of the blood during IWBH. Finally, the measurement errors of this method were discussed and the significance of adopting IWBH for efficiently raising the body-core temperature was demonstrated. This study also raised quite a few new fundamental issues related to the heat transfer of large blood vessels.

Keywords: whole body hyperthermia, interventional blood heating, minimally invasive therapy, heating needle, temperature monitoring

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Whole body hyperthermia (WBH) is being regarded as a very promising way of efficiently treating patients with malignant tumors already disseminated throughout the whole body [1–3]. However, there is currently a strong lack of a simple, safe, easy to administrate and minimally invasive heating strategy. This is a bottleneck limiting the wide application of WBH in clinical practice. Early attempts such as using artificially induced fever, warm water, wax, hot air, or just wrapping the patient in heated blankets lack efficacy with regard to their capability of raising the body-core temperature to 40–42 °C [4]. As a noninvasive method, radiative whole body hyperthermia (RWBH) has been available in clinics since the mid-1980s. Its most important side effect lies in that it may cause peripheral neuropathy, psychogenic disorders and arrhythmias. Besides, it is difficult to achieve a core temperature up to 42 °C without causing burn risks to the patients [4]. As an alternative, extracorporeal whole body hyperthermia (EWBH), using the body's cardio-vascular system to uniformly distribute heat throughout the body, is more 'physiological' than that of an external heat source. This method can, however, cause serious damage to the blood and blood vessels of the tumor patients, due to the complex circuit of the blood taken out of the human body. A number of complications specially related to EWBH have been reported, including pulmonary edema, liver necrosis, peripheral neuropathy, transverse myelitis, renal dysfunction and infection [5]. In order to overcome such shortcomings of the traditional methods, we have established for the first time a new conceptual minimally invasive interventional whole body hyperthermia method which was termed IWBH [6]. The basic

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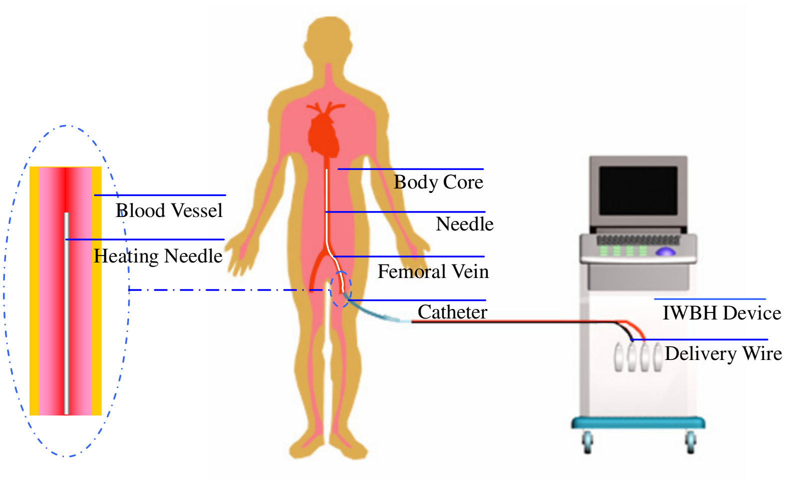

Figure 1. Schematic illustrating the principle of IWBH.

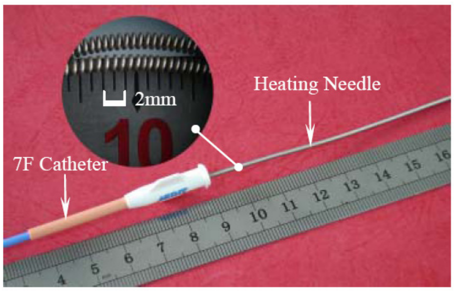

Figure 2. Micro helix needle for interstitial blood heating.

principle is based on directly heating the blood in a vessel by an inserted fine micro heating needle (figure 1). As shown in figure 2, the micro heating needle made of nickel can be inserted into a 5–7F catheter. The needle has both pliable and stiff properties, which make it easy for the physician to correctly position the needle. The surface of this needle can be coated with a thin layer of polyimide. Biocompatibility studies suggest that the polyimide is ideally suitable for biomedical applications. The catheter can then be endovascularly disposed within the patient's femoral vein (see figure 1) and advanced into the inferior vena cava by the percutaneous (Seldinger) technique. A direct electric current, typically in the range of 1–20 V at 1–10 A, may be applied on the needle via the delivery wires from a dc source. The blood flowing through the needle was then heated and the thermal energy was subsequently transported away to the whole body due to continuous blood circulation.

Clearly, accurate and reliable thermometry instrumentation is important in all types of hyperthermia procedures, particularly in IWBH. This is because a small temperature variation would affect the magnitude of side effects. Potential complications of excessive heating by the needle would lead to blood hemolysis, impairment of blood cells and the vessel wall, etc. Therefore, a most challenging task is to determine the temperature of the heating needle and the blood flowing across it. In this side, the thermocouple is perhaps the most commonly used sensing element for WBH temperature monitoring. However, it cannot be directly used here. This is because the size of a thermocouple's tip is approximately that of the heating needle, which may worsen the situation in blocking the vessel. Meanwhile, since the needle is submerged in water during the experiment or immersed in blood during the clinical operation, the infrared thermometer, fluoroscopic fiber optic temperature sensor and liquid crystal paint cannot be used here either [7, 8]. A platinum resistance thermometer may be inserted into the blood vessel to measure the blood temperature. But it will increase the complexity of this apparatus and cannot measure the temperature of the heating needle. In our previous trial, a thermocouple was attached to the needle tip for a temperature measurement. But its recorded information is only the temperature of the fluid flowing through the needle, which makes it insufficient in directly getting the temperature of the needle itself. Overall, the temperature monitoring technique should be as simple as possible.

In this paper, we aim to establish a method which is capable of simultaneously heating the blood and measuring the blood temperature. The needle used in the IWBH treatment is made of nickel, which has a positive temperature coefficient (PTC), i.e. its resistance increases proportionally with the temperature. The steep increase in the resistance as a function of the increased temperature guarantees a high sensitivity in detecting the temperature variation, even in small magnitude.

Besides, one can even use this method for more estimation with the help of heat transfer theory. For example, one can write out a general formula to characterize the heat transfer between the needle and blood:

$$Q = hA\Delta T, \tag{1}$$

where Q is the heat flux between the needle and the blood running across it, A is the heat transfer area, i.e. the surface area of the heating needle, h is the heat transfer coefficient and $\Delta T = \bar{T}_{\text{needle}} - \bar{T}_{\text{blood}}$ ($\bar{T}_{\text{needle}}$ is the average temperature of the needle and $\bar{T}_{\text{blood}}$ is the temperature of the blood). During a steady state of heating with a needle, A and Q are constants and can be easily measured in advance, and h is also a constant and can be obtained by an *in vitro* simulated experiment or theoretical evaluation (this will be discussed in more detail in a later section). It should be noted that the $\bar{T}_{\text{needle}}$ value can

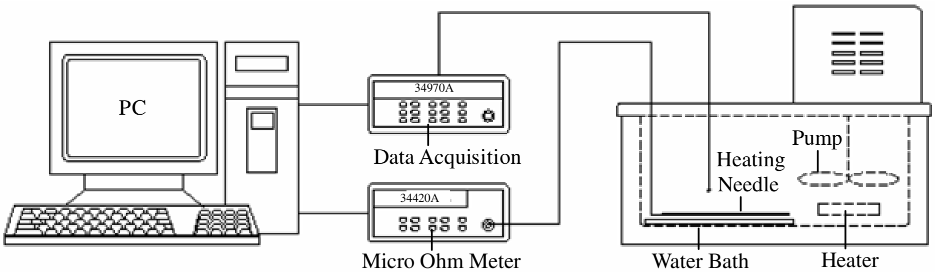

Figure 3. Schematic diagram for measuring the resistance of the needle at various temperatures.

be acquired by monitoring only the resistance of the needle. In this way, the average temperature $\bar{T}_{\text{blood}}$ of the blood can be calculated. In the following sections, a series of experiments were performed to validate the above strategy.

2. Experimental setup

Two conceptual experiments were performed: (1) calibration of the R–T (resistance–temperature) relation of the heating needle; (2) determination of the needle temperature during heating in the simulated blood vessel.

2.1. Calibration of the R–T relation of the heating needle

Although the R–T (resistance–temperature) relation of pure nickel can be obtained from a standard handbook for physical properties of metals, the manufacture condition for a specific needle may lead to a shift of such a typical value. In fact, the nickel thermal resistance varies with different fabrication conditions and stress [9]. Therefore, experiments were performed to calibrate the thermal resistance of the present needle. Figure 3 shows the setup of the experiment. A digital thermocouple system was calibrated to 0.1 °C resolution to monitor the temperature. A high resolution micro ohm meter ($7\frac{1}{2}$ digit nano volt/micro ohm meter, Agilent 34420A, USA) was used to calibrate the resistance of the heating needle at different temperatures. The resistance measurement by the micro ohm meter is a four-wire measurement which can eliminate the error caused by the lead wire. In order to guarantee a constant temperature, a water bath with a temperature stabilization accuracy of ±0.1 °C was adopted, which was filled with 10 l of clean water. It is important to note that the water in the bath is stirred by a pump to avoid a large temperature gradient. The micro helix heating needle (length: 400 mm, outer diameter: 1.5 mm) is made of nickel wire 0.4 mm in diameter and 2.5 m in total length. The needle was submerged in the water and kept on the surface of a 280 mm × 240 mm × 3 mm polytetrafluoroethylene dielectric plate, which is to avoid needle contact with the metallic bottom of the bath and provide suitable thermal insulation. The thermocouple was firmly attached on the surface of the insulating plate. Subsequently, the water was warmed from 35.0 °C to 55.0 °C via increments of 1.0 °C and held for 10 min at each temperature stage. The temperature of the bath and resistance of the needle were recorded by a data acquisition apparatus at a rate of 5 samples per second and then transferred to a PC. After calculating the mean resistance and standard deviation (SD $<$ 0.1 mΩ) at different temperatures,

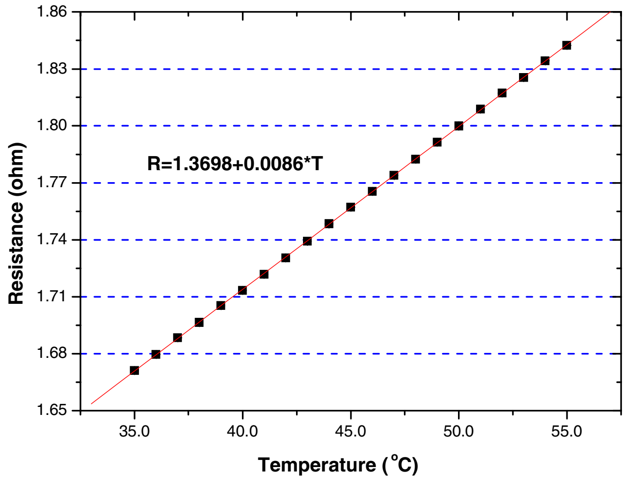

Figure 4. The R–T relation of the heating needle.

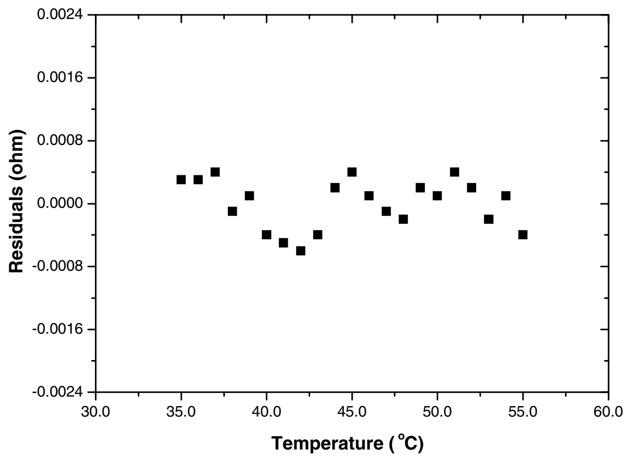

Figure 5. Plot of residuals versus temperature.

the results were finally converted to a curve of resistance versus temperature, which can be seen in figure 4. Clearly, a perfect linear correlation has been obtained. This is beneficial for accurately measuring the temperature. A plot of residuals (figure 5) was used to check the quality of the curve fitting which reflected the good measurement accuracy in the present experiment.

Figure 4 shows the resistance of the needle at different temperatures. It clearly indicates the excellent characteristics in linearity, sensitivity and stability of the needle. The standard pure nickel temperature coefficient is 6170 ×

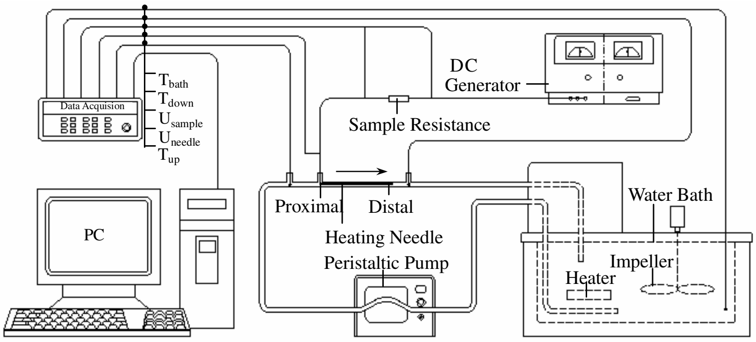

Figure 6. Schematic diagram for measuring the temperature of the needle during heating in the simulated blood vessel system.

$10^{-6}\,^{\circ}\mathrm{C}^{-1}$ (-40–$150\,^{\circ}$C) [9]. According to our experimental measurement, the thermal temperature coefficient of the needle can be generalized as

$$\alpha = \frac{R_{100} - R_0}{\Delta T \cdot R_0} = \frac{2.2288 - 1.3698}{1.3698} = 6271 \times 10^{-6}\,^{\circ}\mathrm{C}^{-1}, \tag{2}$$

where α is the thermal temperature coefficient, R_{100} and R_0 are respectively the resistance of the needle at 100 °C and 0 °C, and ΔT_{100} is the temperature difference equal to 100 °C.

The temperature coefficient of the needle is a little larger than that of standard nickel. The reason may be due to the stress caused by the specific manufacturing situation. We have tested an additional needle made under similar conditions, but still there is little difference. That means each needle should be calibrated before being used in a practical operation. Applying heat treatment can eliminate the stress and keep the temperature coefficient at a relatively uniform value. In a real situation, the relation between the resistivity of metal and the temperature may not always be linear. Therefore, detection of such a relation should be carefully done before applying a needle sensor for monitoring the temperature. Sometimes when the temperature is as low as 0 °C, the measurement error becomes large. In this case, one must always consider its uncertainty and the propagation error, in order to obtain an appropriate estimation.

2.2. Determination of the needle temperature during heating in the simulated blood vessel

In the above experiment, we have obtained the *R–T* relation of the heating needle. If the resistance of the needle during heating can be measured and compared with such data as given in figure 4, the temperature of the needle can then be estimated.

In a clinical operation, the heating needle should be inserted into the blood vessel. To simulate this situation, a flow model was experimentally constructed from clear plastic tubing with 8 mm internal diameter (figure 6), which is to represent the femoral vein. As far as practical, we took a glycerol solution containing 30 wt% glycerol and 70 wt% distilled water as the blood analog fluid, which is routinely adopted in previous blood simulation measurements [10]. A Brookfield viscometer (ND-1 Brookfield Viscometer, Tianjin, China) was used to measure the viscosity of the glycerol solution. The measured viscosity of the distilled water and glycerol solution is shown in figure 7. A peristaltic pump

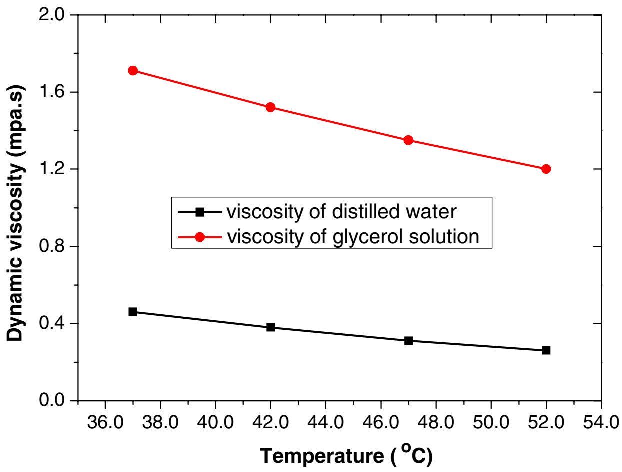

Figure 7. The dynamic viscosity of distilled water and glycerol solution.

served as the power to drive the aqueous glycerol flow in the circulation system. The solution runs with a constant pulsatile flow at the rate of 700 ml min^{-1}, which is similar to that of the blood flow of the femoral vein. An impeller was adopted to promote heat transfer of the fluid in the bath. A dc generator (10 A, 20 V) was used to supply power to the heating needle. The key component, a sample resistance (10 W, 50 mΩ) with low temperature drift (5 ppm), was connected to the needle for assisting the measurement. In this way, the electrical current flow in the circuit is given as

$$I_{\mathrm{sample}} = \frac{U_{\mathrm{sample}}}{R_{\mathrm{sample}}}, \tag{3}$$

where I_{sample} is the electric current, U_{sample} is the voltage applied on the sample resistance and R_{sample} is the resistance of the sample, which remains constant throughout the heating process.

Then, the resistance of the needle in the serial circuit can be calculated as

$$R_{\mathrm{probe}} = \frac{U_{\mathrm{probe}}}{I_{\mathrm{sample}}}, \tag{4}$$

where R_{probe} is the resistance of the needle and U_{probe} is the voltage applied on the needle.

At the beginning of the experiment, the water in the bath was set to 37 °C. After initiating the flow of the water, the voltage of the dc source is adjusted from 7 V to 14 V via

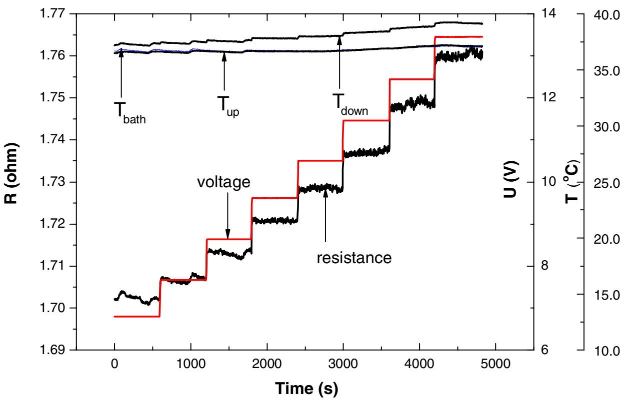

Figure 8. The temperature and resistance variation during heating.

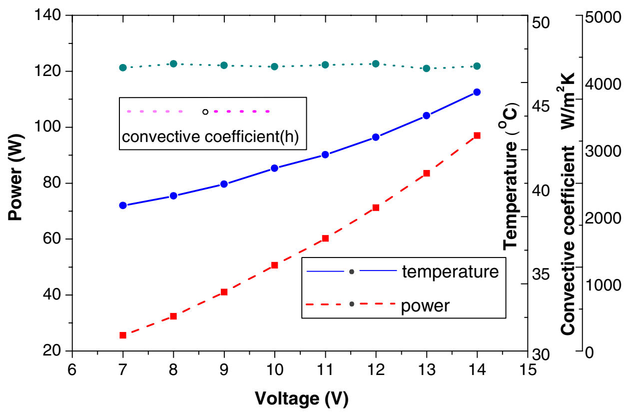

Figure 9. The heating power and the temperature of the needle as well as the h value versus the output voltage of the dc generator. For power, SD < 0.1 W. For temperature, SD < 0.1 °C. For convective coefficient, SD < 30 W m^{-2} K^{-1}.

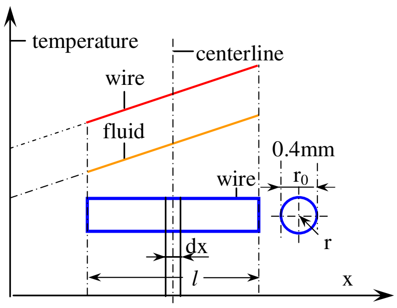

Figure 10. Schematic diagram of the temperature field of the nickel wire and the fluid.

1 V increments and held for 10 min at each temperature stage. The voltage applied on the sample resistance and the heating needle was continuously recorded by the data acquisition system. The water bath was kept at a steady state of 37 °C to simulate the physiological environment. The upstream and downstream temperatures of the water were recorded throughout the experiment.

3. Results and discussion

As shown in figure 8, the resistance of the needle keeps increasing with the voltage during heating in the simulated vessel. This finding is in accordance with expectation. The fluctuation of the temperature of the water upstream (T_{up}) and downstream (T_{down}) of the circulation system and the temperature of the water in the bath (T_{bath}), is caused by the alternative on–off of the heater in the bath. Similarly, the resistance of the heating needle also fluctuates. Due to heat transfer from the pipe to the ambient, T_{up} is a little lower than T_{bath}. Clearly, the water running through the heating needle has been heated by the needle. Therefore, T_{down} increases with increasing voltage.

The variation of the power applied on the heating needle is displayed in figure 9. The power is nearly 100 W at 14 V. By comparing the resistance of the needle during heating with the calibration chart, the resistance at different voltages can be converted into temperature. The results are illustrated in figure 9. It can be observed that the temperature of the needle increases slowly while the power increases rapidly. The temperature of the heating needle is 38.6 °C at 7 V and is merely 45.4 °C at 14 V.

One needs to be aware that the water flow in the tube was heated by the needle. So the temperature of the fluid downstream of the needle should be a little higher than the temperature upstream. Similarly, the temperature at the distal tip (position for this point can be found in figure 6) of the needle is higher than that at the proximal tip, which is close to the delivery wire. The needle serves as a uniform heating source; thus, the temperature of the fluid running through it increases linearly (a rigorous mathematical explanation can refer to a uniform heat flux as a boundary condition [11]). Strictly speaking, it may cause a certain degree of error for this assumption. But it can simplify the question and the error is tolerable. It should be pointed out that the temperature of the needle acquired by the present method represents that of the center point of the needle. In the following, we will explain this viewpoint in more detail. Before illustrating this problem, one additional hypothesis still needs to be made. That is, the temperature of the needle increases linearly with the fluid temperature. This is a simplified however intuitive assumption, which makes the analysis feasible.

A short nickel wire was taken as the analysis object. The heat exchange between the wire and the surrounding fluid can be described by the schematic diagram in figure 10. For simplicity, we only consider the temperature variation along the radius of the nickel wire. The theoretical model for the steady state heat transfer was then established as follows [11]:

$$\frac{d^2T}{dr^2} + \frac{1}{r}\frac{dT}{dr} + \frac{q_v}{\lambda} = 0, \tag{5}$$

where λ is the thermal conductivity of the needle, and q_v is the internal heat generation rate. q_v can be easily obtained as

$$q_v = \frac{I^2\rho\frac{l}{\pi r_0^2}}{\pi r_0^2 l} = \frac{I^2\rho}{(\pi r_0^2)^2}, \tag{6}$$

where I denotes the electric current flow in the wire and ρ, l, r_0 are respectively the resistivity, total length and radius of the wire.

Considering the practical situations, the boundary conditions are given as follows:

$$-\lambda \frac{\mathrm{d}T}{\mathrm{d}r} = h(T - T_f), \qquad r = r_0, \tag{7}$$

$$\frac{\mathrm{d}T}{\mathrm{d}r} = 0, \qquad r = 0. \tag{8}$$

The solution to equations (5)–(8) can be obtained as follows:

$$T = \frac{q_v r_0^2}{4\lambda} + \frac{q_v r_0}{2h} - \frac{q_v}{\lambda}\frac{r^2}{4} + T_f, \tag{9}$$

where T_f is the temperature of the fluid surrounding the wire. The Biot number of the nickel wire can be expressed as [11]

$$\mathrm{Biot} = \frac{hr_0}{2\lambda} \approx \frac{4400 \times 0.0002}{2 \times 94} = 0.005 < 0.05. \tag{10}$$

Clearly, the wire met the applicability criterion of the lumped method. Therefore, the surface temperature of the wire (T_{surface}) is uniform with the internal temperature. For any cross section of the analysis object, the temperature difference between the wire and the fluid can be expressed as

$$\Delta T = T_{\mathrm{surface}} - T_f = \frac{q_v r_0^2}{4\lambda} + \frac{q_v r_0}{2h} - \frac{q_v}{\lambda}\frac{r_0^2}{4} = \mathrm{constant}. \tag{11}$$

Next, we will demonstrate the first problem. The resistance of a differential element

$$\begin{aligned} \mathrm{d}R &= \rho_x \frac{\mathrm{d}x}{\pi r_0^2} = (k_1 T + b_1)\frac{\mathrm{d}x}{\pi r_0^2} \\ &= [k_1(k_2 x + b_2) + b_1]\frac{\mathrm{d}x}{\pi r_0^2} = (kx + b)\frac{\mathrm{d}x}{\pi r_0^2}, \end{aligned} \tag{12}$$

where k_1, k_2 and k are proportional coefficients and assumed here for simplicity; b_1, b_2 and b are also approximated as constants.

Then one has

$$\rho_x = kx + b. \tag{13}$$

The total resistance can be obtained as

$$\begin{aligned} R &= \int_l \mathrm{d}R = \int_l \rho_x \frac{\mathrm{d}x}{\pi r_0^2} = \int_l (kx + b)\frac{\mathrm{d}x}{\pi r_0^2} \\ &= \frac{1}{\pi r_0^2}\left(k\frac{l^2}{2} + bl\right). \end{aligned} \tag{14}$$

Assuming that the temperature of a nickel resistance which has the same resistance and geometry as the above is uniform along the length, its resistivity can be defined as ρ'. In reality, what one can measure experimentally is the total resistance R of the heating needle. The temperature of the needle is just approximately estimated from such information using the calibration R–T curve as explained before. This is a simplified however intuitive approach for understanding the interior thermal state of the blood vessel. There are still no other ideal ways for realizing this object. Therefore the present method may serve as a start up toward this object if more complex factors are included in the model in the near future. From equation (14), one can get

$$\rho' = \frac{R\pi r_0^2}{l} = \left(k\frac{l}{2} + b\right). \tag{15}$$

According to equation 13), the resistivity of the center point of the needle is

$$\rho_{\frac{1}{2}l} = k\frac{l}{2} + b. \tag{16}$$

As explained above, ρ' represents the measured temperature information of the needle. By comparing equations (15) and (16), we found that the temperature acquired by the present method in the experiment represents the temperature of the center of the needle. This allowed an evaluation of the temperature variation along the wire. For instance, the temperatures for the center of the needle, downstream and upstream of the glycerol solution are respectively measured as 45.4 °C, 39.2 °C and 37.1 °C, when the power is 97 W. The temperature difference between the center of the needle and the surrounding fluid can then be approximately obtained as

$$\Delta T = 45.4 - \frac{39.2 + 37.1}{2} = 7.3\ ^{\circ}\mathrm{C}. \tag{17}$$

Therefore, the temperatures at the distal and proximal points of the needle are then obtained as $39.2 + 7.3 = 46.5$ °C and $37.1 + 7.3 = 44.4$ °C, respectively. Clearly, the largest difference for the upstream and downstream temperatures is 2.1 °C, which is in a small scale. Therefore, the above linear assumption between the needle temperature and its length is acceptable. The measured temperature information for the needle and the blood can then serve as a valuable reference for managing the heating state during IWBH.

Such a method can possibly be used in future clinics. For example, taking into account that the temperature of the blood upstream of the needle in a real situation is nearly 40.0 °C, which is 3.0 °C higher than the temperature of the solution in the present experiment, the temperature of the needle in the clinical operation will be $46.5 + 3.0 = 49.5$ °C and $44.4 + 3.0 = 47.4$ °C, respectively. Here we assumed that the real thermal treatment condition is the same as the experiment. Of course, the result in this paper is only a fundamental work, some measured errors need to be calibrated in latter studies.

The temperature of the needle is a major concern during IWBH. In this study we have obtained this value when the needle is immersed in the tubing system. Since the thermal and rheological properties of the glycerol solution, the diameter of the tubing, and the flow rate of the blood analog fluid circulating in the tubing are similar to the properties of blood, the diameter of the femoral vein and the blood flow in the femoral vein respectively, one can possibly apply this result to predict the temperature of the needle in a real blood vessel in the near future. Herron *et al* reported that an inline microwave blood warmer may be used to heat blood safely to 49 °C [12]. Many studies have demonstrated that hemolysis is dependent not only on the temperature at which blood is exposed, but also on the duration of exposure [13]. In fact, the velocity of blood flow in the vein is nearly 20 cm s^{-1} and the length of the needle is less than 500 mm. So the blood cell flowing through the needle will be actively heated within 3 s. This temperature range is entirely safe for a clinical operation. Of course, blood has some characters different from the blood analog fluid. A further experiment or clinical trial will be conducted on animals to test the safety of IWBH.

In traditional WBH, only the temperatures of the surface or cavity of the human body were measured. The temperature of the blood however has never been monitored. In IWBH, such information can be simultaneously obtained through the flowing way. In this study, the experiment was conducted in

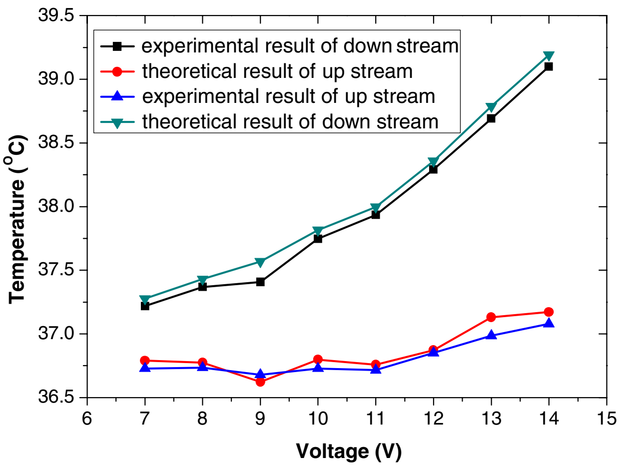

Figure 11. Theoretical and experimental results for the temperature of water downstream and upstream of the needle.

a quasi-steady state and the temperature of the needle was known. According to equation (1)

$$h = \frac{Q}{A(\bar{T}_{\text{needle}} - \bar{T}_{\text{blood}})}. \tag{18}$$

Considering that $\bar{T}_{\text{needle}}$ represents the average temperature of the needle and $\bar{T}_{\text{blood}}$ is the average temperature of the blood via the needle, h can be estimated as

$$h = \frac{Q}{A\left(\bar{T}_{\text{needle}} - \frac{T_{\text{down}}+T_{\text{up}}}{2}\right)}. \tag{19}$$

Here during the experiment, Q , A, $\bar{T}_{\text{needle}}$, $\bar{T}_{\text{blood}}$ are measurable parameters; thus the heat transfer coefficient h can be conveniently calculated and depicted in figure 11.

It is well known that h is dependent on the velocity, flow pattern, thermal property of the fluid and geometric size of the flow channel [14]. Thus, once this value is obtained, we can generalize it to the approximate situation or clinical operation which has a similar heat transfer condition as in the experiment.

Further, we can adopt the blood vessel positioned with the needle for an additional analysis. If we consider that the blood runs quickly through the needle, the heat loss from the large blood vessel to the tissue can be neglected. Based on the energy conservation equation and heat transfer equation, one has

$$\begin{cases} Q = \dot{m}c_p(T_{\text{down}} - T_{\text{up}}) \\ Q = hA(\bar{T}_{\text{needle}} - \bar{T}_{\text{blood}}) = hA\left(\bar{T}_{\text{needle}} - \frac{T_{\text{down}}+T_{\text{up}}}{2}\right) \end{cases} \tag{20}$$

where $\dot{m}$ and c_p are respectively the mass flow and the specific heat of blood.

The solution to equation (20) is

$$\begin{cases} T_{\text{up}} = \bar{T}_{\text{needle}} - \dfrac{Q}{hA} - \dfrac{1}{2}\dfrac{Q}{\dot{m}c} \\ T_{\text{down}} = \bar{T}_{\text{needle}} - \dfrac{Q}{hA} + \dfrac{1}{2}\dfrac{Q}{\dot{m}c}. \end{cases} \tag{21}$$

In a real clinical hyperthermia treatment procedure, $\bar{T}_{\text{needle}}$, Q, $\dot{m}$ and C_P can be obtained by special measuring equipment, and h can be evaluated by the experimental results *in vitro*. Therefore T_{up} and T_{down} are in fact known values.

Here, the validity of this equation might be verified with the data in the experiment. Figure 11 shows the theoretical and experimental results of the temperature of the water downstream and upstream of the needle, which were measured by thermocouple. Obviously, the experimental result is a little lower than the estimated result. The reason can be attributed to heat loss from the pipe to the ambient, which does not exactly follow the theoretically assumed adiabatic condition.

It is by its very nature that the temperature variation of the blood is hard to predict because it depends on a large and rather poorly defined set of influential factors, such as the disturbance of the blood flow, non-uniform radius of the blood vessel, etc. The aim of this study is to determine the temperature of the heating needle and establish a method for monitoring *in situ* the temperature variation of the blood running through the vessel during IWBH. Fortunately, some other temperature information, such as the temperature of the oesophagus or tympanum, can also help us to judge the blood temperature and calibrate the present method.

Overall, quantifying the temperature variation in the human body is an intricate procedure. Up to now, there was no clear ceiling to the amount of heat that should be input into the human body for WBH. Wust *et al* addressed the power surplus of the energy input into the human body as about 85 W to cause a systematic temperature increase of 1 °C for a patient weighing 70 kg undergoing WBH [15]. This is because in the process of traditional WBH, such as RWBH, water bath or wax bath, etc, energy is transported from the skin surface to the core. Therefore, to avoid burning the superficial skin tissue, only a small amount of heat can be absorbed via the poor heat conduction across the tissues, although a large amount of heat can be provided by the external heat applicator.

Some physiological procedures imply that increasing the temperature of the human body may not require a large amount of energy for the purpose of IWBH in some cases. For example, the body-core temperature of a person in fever may quickly exceed 40 °C. In such a situation the heat mainly comes from the metabolism. The metabolic basal rate is about 85 W at 37 °C and becomes twice as high at 42 °C [16]. That means a net energy input less than 170 W will quickly raise the patient's core temperature. By Joule heating, all the electrical energy will transform into thermal energy in IWBH. In the present study, a single heating needle can safely apply nearly 100 W thermal energy into the body. In our experiment, we did not run the fluid in the tube with a very high velocity. The experimental results, therefore, represent the worst case in which the blood flow rate is much lower than in a real IWBH treatment situation. Conservatively speaking, more than 100 W energy will be safely delivered into the body core for one needle. If four heating needles are simultaneously inserted into both sides of the femoral veins and the femoral arteries of the human body, over 400 W energy can be delivered into the body. In this way, the efficiency of the IWBH can still be significantly improved. Overall, the capacity of the IWBH for raising the body temperature has been demonstrated by the present experiment. Therefore, what one needs to clarify in the near future is perhaps only the issue whether the inserted needles would cause trouble for the running of the blood flow. This requires another different work.

4. Conclusion

This paper established an approach which can monitor several key temperatures during IWBH. The conceptual experiments show that the temperature of the heating needle can be flexibly controlled and determined within a certain range as desired, such as being less than 50 °C, which is safe for a clinical IWBH operation. This study demonstrates the possibility of using the micro helix heating needle as a multifunctional device. It serves not only as an endovascular heating device to produce heat for whole body hyperthermia, but also as a thermometry instrument to monitor the temperature variation of the heating needle and the blood running through it. The effort established the groundwork for a new IWBH method and latter work will further improve this method.

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