35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Improvement of the Dual-heat-flux Method for Deep Body Temperature Measurement Based on a Finite Element Model

Ming Huang, *Member*, *IEEE*, Wenxi Chen, *Member*, *IEEE*, Kei-ichiro Kitamura, Tetsu Nemoto, and Toshiyo Tamura, *Senior Member*, *IEEE*

Abstract— **Based on finite element method, this study was dedicated to improve the performance of the newly proposed dual-heat-flux method (DHF) for noninvasive deep body temperature (DBT) measurement. This study focused mainly on the dimensions, e.g., the height and radius, of the probe of DHF to improve its performance, in terms of the measurement depth beneath the skin. This paper was also engaged in confirmation of the improved probe's performance on a physiological significant temperature range, 35 to 40 °C. The results showed that a probe with lower in height and larger in radius would have better performance. It suggests that we should reduce the height of the probe to half of the original and keep the radius unchanged for a better performance.**

I. Introduction

Daily healthcare involves the long-time monitoring of vital signs, e.g., body temperature, heart rate, blood pressure and respiratory rate, continuously and unconstrainedly. This specific requirement excludes the possibility of using invasive method for the measurement, which is a real challenge for the measurement of temperature. Since the heat can be exchanged with ambient environment by means of radiation, conduction and convection. The temperature on skin is prone to fluctuate. Although the temperature value on specified site, such as the toe, would give important information on a certain physiological condition [1]. A more reliable index is necessary for a reliable monitoring system.

Unlike the body surface temperature, which will vary along with the ambient situation, deep body temperature (DBT) is kept within a narrow range around 37 °C and considered as one of the barometers reflecting human health conditions. For intermittent measurement, a 2 °C deviation of DBT from the normal 37 °C is considered as a serious threat for the subject's health [2]. Besides, DBT monitoring and management are of vital importance during surgery or therapeutic intervention. For continuous monitoring, DBT rhythms are also thought to be particularly significant in chronobiological studies, and there is a wealth of evidences that the ebb and flow of human mental and physical performance follows the rhythm of the deep body temperature [3].

Since the 1970s, methods that sought to measure the deep body temperature noninvasively have been proposed. In 1971, Fox *et al*. proposed the so-called zero-heat-flux (DHF) method, which is reported to be capable in measuring the temperature under the measuring site [4], [5]. Its fatal problem for daily healthcare monitoring is that it is power consuming, which make the demand of unconstrained measurement hard to satisfy.

In 2009, Kitamura *et al*. developed a new dual-heat-flux (DHF) method that could measure the deep body temperature from the surface of the skin without using an extra heater. Through simulation experiment, the dual-heat-flux method has shown competitive with the conventional ZHF method in performance, in terms of measurement accuracy and step response to changes of DBT [6]. This method was theoretically proven by M. Huang and W. Chen with 3D model based on finite element method (FEM) [7]. The measured value of this method is reported to be 6 mm under the prototype (would also be called as probe hereafter) in the skin layer. It is a crucial parameter for the performance of the prototype. Because a deeper site where the prototype is measuring means that the measured temperature value by the prototype is closer to the real DBT. This study also provided an idea of improving the design of the probe since that a computer model is readily for dimensions and materials modification and the FEM is capable of handling models with complex geometry by nature. It greatly reduces the expenditure and life cycle of prototype development.

In this study, we would like to make adjustment of the probe dimensions so as to improve the measurement performance, which is mainly evaluated by the measured depth by the method. Through this modification, we hope to increase the measured depth beneath the skin layer.

II. Procedure

Bioheat Conduction

Being different from the heat transfer in the abiotic materials, heat transfer in biological tissues is a complicated procedure since it involves thermal conduction in the tissues, convection and perfusion of blood, delivery of arterial blood to a capillary bed in tissues, and metabolic heat generation. The physical equation describing this kind of heat transfer is called bioheat equation. The well-acknowledged bioheat equation was proposed by Pennes in 1948 [8], with a form of partial differential equation to describe the spatial and

Manuscript received Feb, 4, 2013. This work was supported in part by the University of Aizu Competitive Research Funds P-11 and P-21. *Asterisk indicates corresponding author*.

M. Huang is with QRS Co., Ltd., Funabashi, Chiba, Japan (e-mail: alexhuang0604@gmail.com).

*W. Chen is with the University of Aizu, Aizu-wakamatsu, Fukushima 965-8580, Japan (phone: 81-242-37-2606; fax: 37-2728; e-mail: wenxi@u-aizu.ac.jp).

K. Kitamura and T. Nemoto are with the Kanazawa University, Kanazawa, Ishikawa, 920-1192, Japan (phone: 81- 076-265-2595; fax: 234-4369).

T. Tamura is with the Osaka Electro-Communication University, Shijonawate, Osaka Prefecture 575-0063 (e-mail: tamurat@isc.osakac.ac.jp).

temporal behavior of heat transfer inside biological tissues as shown below:

$$\rho c_{p} \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + q_{p} + q_{m} . \quad (1)$$

ρ, c_p, and k are the density of the local tissue (kg/m^3), the specific heat of tissue (J/kg·°C), and the thermal conductivity (W/m·°C), respectively. The terms q_p and q_m are the heat transfer rate from the blood to the tissue (the reverse situation, in which heat transfers from tissue to the blood may also occur) and the unified rate of metabolic heat generation, respectively.

In Pennes's model, the heat transfer between the blood and tissue, q_p is assumed to be proportional to the temperature difference between the arterial blood entering the capillary bed in tissues, with the following form:

$$q_{p} = \omega_{b} \rho_{b} c_{b} (T_{b} - T) , \quad (2)$$

where T_b is the blood temperature. The terms ω_b, ρ_b and c_b are the blood perfusion rate (m^3/m^3·s), blood density (kg/m^3) and specific heat of blood (J/kg·°C), respectively. For most of the times, it is assumed that the arterial blood is delivered to the capillary bed at a constant temperature, 37 °C and this heat effect was imported as a heat source in the corresponding layer, to whom this study also conformed.

The DHF probe used in the Kitamura's experiment was actually covered by a urethane sponge to reduce the variance of the surface temperature of the probe due to the surrounding airflow. Thereby as for the boundary conditions in this model, the radiation at the boundaries of the probe was taken into account while the convection was not, as the sponge cover would prevent the probe from being influenced by the surrounding airflow. Thus, heat exchange between the probe and the ambient environment occurs at the boundaries of the probe and can be described by

$$\nabla \cdot (k \nabla T) = \varepsilon\sigma (T_{amb}^{4} - T_{s}^{4}) , \quad (3)$$

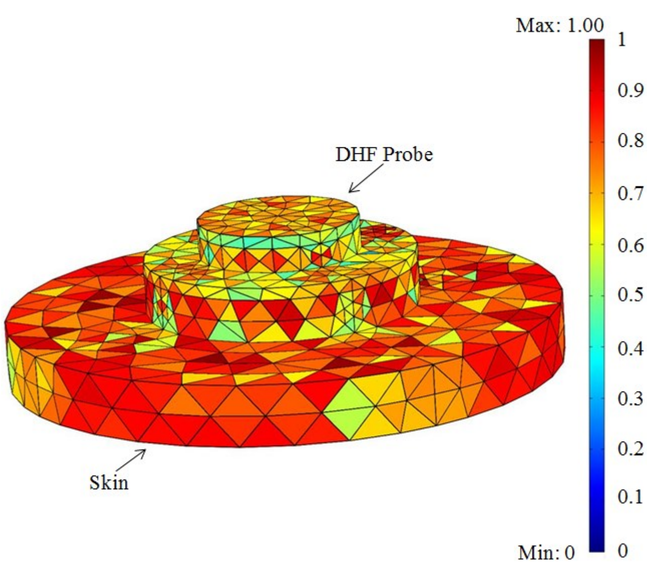

Fig. 1. Mesh elements on the periphery of the model and their qualities. The elements qualities can be discriminated by their colors and the legend on the right-hand side. The minimum element quality of this model is 0.29.

where T_s and T_{amb} are the temperatures on the boundaries and of the surrounding air (°C), respectively, σ is the Stefan-Boltzmann constant, and ε is the emissivity of the material.

3D Model Based on FEM

These biophysical equations were then applied to a 3D model based on FEM, which was constructed with FEM software COMSOL Multiphysics (COMSOL Inc., Sweden). In the finite element discretization, the Lagrange quadratic tetrahedral element was used. Quadratic element was used here for the reason that in comparison to the linear element, it provided a better approximation to the heat transfer problem.

When meshing the model, the mesh quality, q, which is a number between 0 and 1, can have a considerable impact on the computational analysis in terms of the quality of the solution and the time needed to obtain it. In 3D model, the value of q was computed using [9]:

$$q = \frac{72\sqrt{3}V}{(h_1^2 + h_2^2 + h_3^2 + h_4^2 + h_5^2 + h_6^2)^{3/2}} , \quad (4)$$

where V is the volume, and h_1, h_2, h_3, h_4, h_5, and h_6 are the edge lengths of the tetrahedron elements. The mesh quality should not affect the solution's quality if $q > 0.1$. Hence the minimum element quality of this model: 0.29 was adequate to generate an accurate solution. The element qualities of the model can be observed in Fig. 1, where the qualities of those elements on the periphery of the model can be discriminated by their colors and the legend.

Fig. 1 also gave an overall image of the model. The probe was applied to a cutaneous site, around which the skin was modeled as a cylinder, whose height was 10.0 mm. The temperature of its lower boundary was assumed to be DBT (T_D), whose default value was 37 °C. The upper boundary exchanged heat with the surrounding environment by radiation as described by (3), while the peripheral boundary is given as heat insulation according to the assumption of the DHF that the heat flow vertically from the deep body through skin onto the cutaneous surface [6].

In this model, it was assumed that the thermal properties of each component remained homogeneous and constant with changes in temperature. The contact resistance between the skin surface and the probe was neglected. The relevant thermal properties of the materials are tabulated in Table I.

TABLE I. THERMAL PROPERTIES OF MATERIALS

Part	Conductivity (W/m·°C)	Density (kg/m^3)	Specific heat (J/kg·°C)	Emissivity
Skin	0.17	1100	3500	0.95
Blood	*	1000	4175	*
Nylon	0.28	1150	1700	0.85
Copper	400	8700	385	0.03

* denotes the corresponding parameters is unnecessary in this study

Simulation Protocol

Based on the assumption that the heat flux flows from the tissue deep inside the human body onto the probe vertically manner without heat loss, the DHF uses the temperature values acquired by four ICs embedded in the probe to estimate the DBT. This engineering process can be easily simulated in 3D model of FEM, as the temperature value of every spot of the model is readily to be inspected. The details could be found in [7], thereby we make no effort to relate this aspect.

Once the simulation gives the analogue (called as measured value hereafter) of the measured value by the probe, the measured value can be compared with the temperature distribution in the skin layer along the axis of the probe. Then the corresponding depth of the measured temperature can be determined.

Fig. 2 is the lateral view of the model, which shows the definition of the corresponding depth. The Y-axis is parallel to the axis of the probe with its base point at the height of the cutaneous surface. Hence the corresponding height in the skin layer is negative, and the bigger its absolute is, the deeper in the skin.

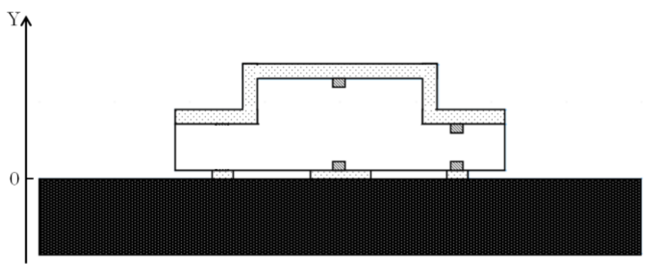

Fig. 2.Lateral view of the model. The Y-axis is parallel to the axis of the probe and the base point of the y-axis is set at the cutaneous surface.

In order to improve the performance of the DHF probe, especially the corresponding depth of the measured value, it height and the radius were investigated. The model trials were carried out in such a manner that probe's height was changed by multiplying a ratio to the height of the prototype. The ratio = 0.5, 0.6, 0.7, 0.8, 0.9 and 1 respectively, while the radius was kept constant. Whereafter, the radius of the probe was changed consistently in the similar manner: by multiplying a ratio, which was the same series as the one used in the previous trials, to the radius of the original prototype while its height unchanged. Trials were carried under three different ambient temperatures: 20, 25 and 30 °C. Thus, there a total of 30 trials of different situations in our study. We wished to generate a general relation between the measured depth and the dimensions, so as to provide suggestions for further improvement.

Why we chose the ratio series from 0.5 to 1 with 0.1 intervals is because that the size of the probe should consider the mobility and implementability. The size of the original prototype is appropriate for daily usage; however, a bigger size may cause unpleasant experience of the user. Reversely, a much smaller size may increase the cost and the developing cycle. Admittedly, some other ratio seems reasonable as well, the numbers we try here would somehow figure out the general relations.

III. Result

We show the results of two scenarios in this part. In the first scenario, we would like to figure out the relations between the dimensions of the probe and its performance through measured depth when the DBT (T_D) unchanged. In second scenario, we would like to change the DBT to examine and confirm the insights acquired in the first scenario.

Results when T_D = 37 °C

In this scenario, we, firstly, modified the height of the probe by keeping the radius constant, and then modified the radius by keeping the height unchanged. In all of these trials, T_D was unchanged at 37 °C.

The influence of the height of the probe on the measured depth was concluded in Fig. 3. Results of three kinds of ambient temperature shows consistently that higher the probe

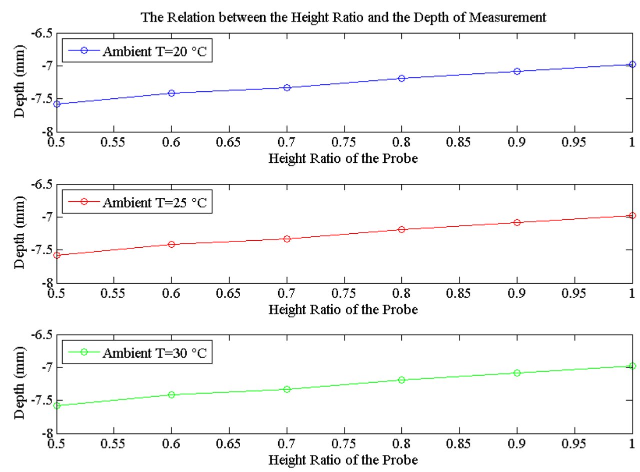

Fig. 3. Results of height modification. Three subfigures show the relation between the measured depth by the probe and the corresponding height with different ambient temperatures.

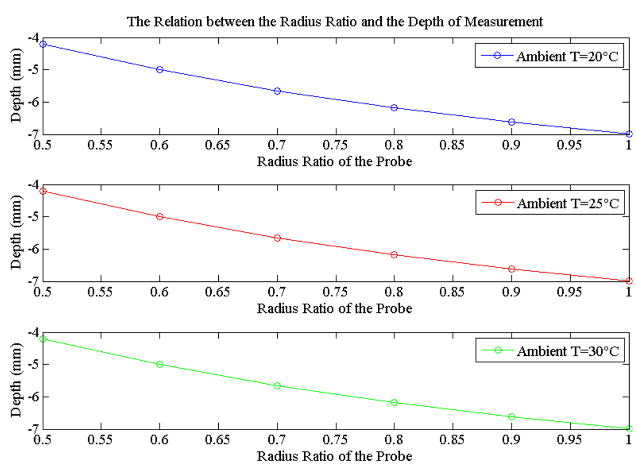

Fig. 4. Results of radius modification. Three subfigures show the relation between the measured depth by the probe and the corresponding radius with different ambient temperatures.

is, shallower the measured depth is. When the height is half of the original one, the difference of the measured temperature value can be as large as about 0.1 °C higher (figure not shown here because of space limitation) than the value measured by the original probe when the ambient temperature was 20 °C, and the difference of the measured depth can be as large as 0.6 mm (lower height, deeper depth).

The effect of the radius is summarized with Fig. 4. The measured depth of the probe would nonlinearly decrease with the decrease of the radius. When the radius is half of the original one, the measured temperature value can be as large as 0.8°C lower than the value measured by the original probe when the ambient temperature is 20 °C (figure not shown here), and the difference of measured depth can be as large as 2.7 mm (longer radius, deeper depth).

It is accessible for us to combine the above two insights to propose an improved prototype, whose height is half of the original one and the radius unchanged. The conjunct improvements of these two modifications are that the measured depth can be extended to 7.6 mm in the skin.

Results when T_D is changed

Based on the improved design of the DHF probe, we further investigated the measuring situations where T_D changed from 35 to 40 °C with 1 °C intervals. It is also accessible for the DBT would not be a constant but fluctuate around 37 °C [10], and this range is physiologically significant.

The results of this trials show that though the measured temperature is changed accordingly (higher the T_D, bigger the measured value). However, the measured depth was totally unchanged at 7.6 mm.

IV. Discussion

About the height of the probe

The reason why a lower prototype means a better performance may lie in that higher the prototype, the more space between the upper and lower ICs inside the probe. It, on the one hand, extends the initial time of the probe [6], and, on the other hand, gives more space to inevitable horizontal heat flux to bring out the heat flux. Nevertheless, the horizontal heat flux would distort the basic assumption of the DHF that the heat flux flow vertically from the deep body on the probe with no heat loss.

The reduction of the height of the probe becomes understandable from the results and we should try to limit it height under the implementability of the prototype.

About the radius of the probe

The influence of the radius is obvious, and the results may suggest us to enlarge it for a better performance. However, in considering the original radius of the probe (22 mm), a larger radius's design should go through comprehensive balance between the mobility and the accuracy.

Situations when T_D is changed

The insight that the measured depth is unchanged at 7.6 mm in the skin layer when the T_D was varying in the range from 35 to 40 °C, while the measured temperature value would change accordingly is in line with the results in [7]. If we can obtain the relation between the measured value and the temperature at 7.6 mm over a temperature range of ambient temperature covering most of the situation when the probe is applied, say, 0 °C – 35 °C, we may use this relation to further estimated the temperature of the much deeper sites.

Other Factors

The physical properties of the materials constructing the probe are important in the improvement of performance as well. As it is mentioned in [7], the conduction of the probe's insulator and the emissivity of the probe's conductor should also be comprehensively analyzed.

Only when the improvement on both the dimensions and the physical properties are conducted, can an improved prototype be designed and then fabricated.

V. Conclusion

By a close examination of the effect of the dimensions of the probe on the measurement performance, in terms of measured depth, the results of this study show that a lower and larger probe would improve its performance.

In considering the mobility and the implementability, this study suggests that we should reduce the height of the probe to half of the original and keep the radius unchanged for a better performance.

References

[1] H. Joly, and, M. Weil, "Temperature of the Great Toe as an Indication of the Severity of Shock," *Circulation*, vol. 39, no. 1, pp.131–38, 1963

[2] M. Neuman, "Measurement of vital signs: Temperature," *IEEE Pulse*, vol. 1, no. 2, pp. 40-49, 2011.

[3] K. Blendstrup, "A sheathed micro thermoprobe for local temperature measurements in tissue," *Medical progress through technology*, vol. 8, no. 3, pp. 149-55, 1981.

[4] R. Fox and A. Solman, "A new technique for monitoring the deep body temperature in man from the intact skin surface," *The Journal of physiology*, vol. 212, no. 2, pp. 8-10, Jan. 1971.

[5] R. Fox, A. Solman, R. Isaacs, A. Fry, and I. MacDonald, "A new method for monitoring deep body temperature from the skin surface," *Clinical science*, vol. 44, no. 1, pp. 81-6, Jan. 1973.

[6] K. Kitamura, X. Zhu, W. Chen, T. Nemoto, "Development of a new method for the noninvasive measurement," *Med Eng Phys,* vol. 32, pp. 1–6, Jan. 2010.

[7] M. Huang and W. Chen, "Theoretical simulation of the dual-heat-flux method in deep body temperature measurements," *Conference proceedings :Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, vol. 2010, pp. 561–4, Jan. 2010.

[8] H. Pennes, "Analysis of tissue and arterial blood temperatures in the resting human forearm," *Journal of applied physiology*, vol. 1, no. 2, pp. 93-122, 1948.

[9] COMSOL Multiphysics, "COMSOL multiphysics user's guide," Version: COMSOL 3.5a, Nov. 2008

[10] R. Fox, A. Solman, "A new technique for monitoring the deep body temperature in man from the intact skin surface," *J Physiol,* vol. 212, pp. 8–10, 1971.