Journal of Thermal Biology 47 (2015) 26–31

Contents lists available at ScienceDirect

Journal of Thermal Biology

journal homepage: www.elsevier.com/locate/jtherbio

Evaluation of structural and thermophysical effects on the measurement accuracy of deep body thermometers based on dual-heat-flux method

Ming Huang [a,*], Toshiyo Tamura [b], Wenxi Chen [c], Shigehiko Kanaya [a]

[a] Graduate School of Information Science, Nara Institute of Science and Technology, Ikoma, Japan
[b] Department of Biological Engineering, Osaka Electro-Communication University, Shijonawate, Japan
[c] Biomedical Information Technology Laboratory, the University of Aizu, Tsuruga, Aizu-wakamatsu City, Japan

ARTICLE INFO

Article history:
Received 30 July 2014
Received in revised form
4 November 2014
Accepted 6 November 2014
Available online 11 November 2014

Keywords:
Unconstrained measurement
Deep body temperature
Transcutaneous thermometer
Daily healthcare
Ubiquitous monitoring
Dual-heat-flux method

ABSTRACT

To help pave a path toward the practical use of continuous unconstrained noninvasive deep body temperature measurement, this study aims to evaluate the structural and thermophysical effects on measurement accuracy for the dual-heat-flux method (DHFM). By considering the thermometer's height, radius, conductivity, density and specific heat as variables affecting the accuracy of DHFM measurement, we investigated the relationship between those variables and accuracy using 3-D models based on finite element method. The results of our simulation study show that accuracy is proportional to the radius but inversely proportional to the thickness of the thermometer when the radius is less than 30.0 mm, and is also inversely proportional to the heat conductivity of the heat insulator inside the thermometer. The insights from this study would help to build a guideline for design, fabrication and optimization of DHFM-based thermometers, as well as their practical use.

1. Introduction

Smart devices and cloud computing provide opportunities for the ubiquitous monitoring of vital signs and for personalized daily healthcare. As one of the vital signs, the deep body temperature (DBT) is that of the abdominal, thoracic or cranial cavities. The ubiquitous monitoring of DBT requires that the measurement devices be low power, portable and human friendly, although usually at a cost of reduced accuracy. These rigorous requirements have eliminated several of the techniques available, such as microwave thermography (Levick et al., 2011) or MRI temperature mapping (Kickhefel et al., 2010). The tympanic infrared thermometer (Kyriacou, 2010) is a promising alternative technique to the "gold standard" of invasive DBT measurement at the sites of the pulmonary artery, the nasopharynx or the distal esophagus (Kimberger et al., 2009). However, it is not suited to being worn long term because of the risk of perforation.

A scenario for which long-term unconstrained DBT monitoring may be desired is the prevention of heatstroke. According to the Japanese Ministry of Health, Labour and Welfare, the mortality from heatstroke was 1,731 in 2010, 79.3% of whom were over the age of 65 (http://www.mhlw.go.jp/toukei/saikin/hw/jinkou/suii10/dl/s08.pdf, 2011). For the prevention of heatstroke, DBT may be one of the most reliable indicators. Therefore, finding a convenient technique for the continuous monitoring of DBT for high-risk groups both indoors and outdoors is imperative.

Long-term DBT measurement is also sought for chronobiological studies. The circadian rhythm of DBT is one of the most important endogenous rhythms of the human body (Koch et al., 2010). It can be fitted to a cosine curve with a period of about 24 h, with a peak in the afternoon and a trough in the early morning (Reilly and Waterhouse, 2009; Baehr et al., 2000). The sleep/wake rhythm and the DBT rhythm are closely related, and a measurement of the DBT rhythm should help to guide the timing of treatment for sleep disorders (Bjorvatn and Pallesen, 2009).

The requirements for DBT measurement in these cases can be met by skin-contact thermometers based on thermophysical principles. An original prototype, called the zero-heat-flux method, was proposed by Fox and Solman (1971). Because this prototype needs an AC-power heater, it is mainly used in hospitals. The double-sensor method (DSM) (Gunga et al., 2008) and the dual-heat-flux method (DHFM) (Kitamura et al., 2010) were developed more recently. Both methods are suited to long-term DBT measurement because they do not need heaters.

* Corresponding author. Fax: +81 (0)743 72-5329.
E-mail address: alex-mhuang@is.naist.jp (M. Huang).

http://dx.doi.org/10.1016/j.jtherbio.2014.11.004
0306-4565/

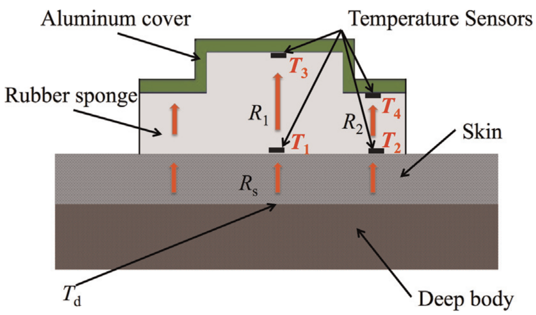

Fig. 1. Illustration of the DHFM. The thermometer has a convex structure with two concentric cylinders of different heights. Heat flow caused by a temperature difference is assumed to flow longitudinally from the deep body to the skin and into the thermometer.

However, because both methods are based on the idealized situation that heat flows from the inner part of the human body through the subcutaneous layers and skin layer into the thermometer longitudinally as shown in Fig. 1, a systematic error caused by transverse heat flow inside the probe is inevitable. Therefore, a straightforward way to improve its accuracy is to suppress the transverse heat flow inside the thermometer. The DSM includes a term in its estimation formula to compensate for the transverse heat loss, whereas the original DHFM has yet to consider this problem. Although satisfactory results (about 0.1 °C measurement error) were reported, an additional urethane sponge cover was found to be necessary in the experiments. To bring the DHFM to the stage of practical application for daily DBT monitoring, a miniaturized design will be indispensable. Furthermore, general insights about its physical mechanisms and possible disadvantages (and their origins) remain unclear in this method.

Our study will focus on the DHFM because it uses two heat paths to eliminate the unknown parameter, namely the thermal resistance of the skin and subcutaneous layers, which differs individually and may vary even for the same person. In this study, we aimed to evaluate the effects of structural parameters (height and radius) and thermophysical parameters (conductivity, density and specific heat) of the components constituting the thermometer on measurement accuracy. In this way, we should achieve a comprehensive understanding for potential improvements in this method, which will ultimately establish a reliable and convenient means for long-term noninvasive DBT monitoring.

2. Materials and methods

In this study, structural parameters (height (h) and radius (r)) and thermophysical parameters (conductivity (k), density (ρ) and specific heat (c_p)) of the thermometer were investigated on how they affect the measurement accuracy. The structural parameters were readily available, whereas the thermophysical parameters were more difficult to measure. Aiming for better thermometer structure and accuracy, an investigation of all combinations of these factors would be a time-consuming and expensive task. We therefore developed 3-D models based on the finite element method (FEM) and drew conclusions from simulations. Our models mimicked the measurement circumstances when applying the thermometer to human skin.

2.1. The DHFM

To calculate the DBT, the DHFM uses at least four temperature sensors mounted inside the DHFM-based thermometer. This can be explained in terms of the following formulas and Fig. 1.

There is a temperature difference between the deep body and the skin surface. If we assume that the inner boundary of the skin is at the same temperature as the deep body, heat will flow from the inner boundary to the skin surface and then into the thermometer, as shown in Fig. 1. Assuming that the heat flow from the deep body into the thermometer remains the same, Fourier's law applies, giving

$$(T_d - T_1)/R_s = (T_1 - T_3)/R_1, \quad (1)$$

$$(T_d - T_2)/R_s = (T_2 - T_4)/R_2, \quad (2)$$

where T_d is the temperature of the inner boundary of the skin, i.e., the DBT, while T_1 and T_2 are skin temperatures measured by two cutaneous temperature sensors inside the thermometer. T_3 and T_4 are the temperatures measured by the other two sensors. R_s is the heat resistance of the skin, with R_1 and R_2 being the heat resistances of the two heat paths inside the thermometer. According to (1) and (2), T_d then can be expressed as

$$T_d = T_1 + \frac{(T_1 - T_2)(T_1 - T_3)}{k(T_2 - T_4) - (T_1 - T_2)} \quad (3)$$

where $k = R_1/R_2$ and can be represented as the ratio of the heights of the two concentric cylinders used in the thermometer fabrication.

It should be noted that, the last term in the denominator of Eq. (3) was $(T_1 - T_3)$ originally, we changed it to $(T_1 - T_2)$ for the adoption of a correction term to compensate the transverse heat flow. The justice has been validated by the simulation study and the prototype experiment that being carried out. In both studies, this correction showed superior results to stabilize the estimation.

2.2. Bioheat transfer formulation

Bioheat transfer in tissues is complicated by blood perfusion and metabolic processes. It can, however, be well described by the Pennes equation (Pennes, 1948), as follows.

$$\rho c_p \frac{\partial T(\mathbf{X}, t)}{\partial t} = \nabla[k(\mathbf{X})\nabla T(\mathbf{X}, t)] + \omega_b \rho_b c_b (T_b - T(X, t)) + q_m(\mathbf{X}, t) \quad X \in \Omega_T \quad (4)$$

This is a formulation of the temporal and spatial evolution of the temperature distribution, which is affected by the heat from blood perfusion and metabolism. The parameters T, ρ, c_p, and k are the temperature, the density (kg/m^3), the specific heat (J/kg °C), and the thermal conductivity (W/m °C) of the local tissue, respectively. The heat exchange between the local tissue and capillary blood flow is described by the second term on the right side, where T_b, ω_b, ρ_b and c_b are the blood temperature, blood perfusion rate (m^3/m^3 s), the blood density (kg/m^3) and the specific heat of blood (J/kg °C), respectively.

Modeling simplifications have been applied to the bioheat transfer formula, as listed below.

a. Heat exchange by convection was ignored. this was acceptable when both the skin and the thermometer were assumed covered by suitable clothing, while radiation was inevitable. hence, the boundary conditions of the thermometer and its surrounding skin could be described by Stefan–Boltzmann law:

$$-\vec{n}[-k(\mathbf{X})\nabla T(\mathbf{X}, t)] = \varepsilon\sigma\left(T_{amb}^4 - T_{boun}^4(\mathbf{X}, t)\right), \quad X \in \Gamma$$

where T_{amb} is the ambient temperature; T_{boun} the local temperature value at the boundary, σ the Stefan–Boltzmann constant and ε the emissivity of the boundary.