Skin-Attachable Sensor for Continuous Core Body Temperature Monitoring

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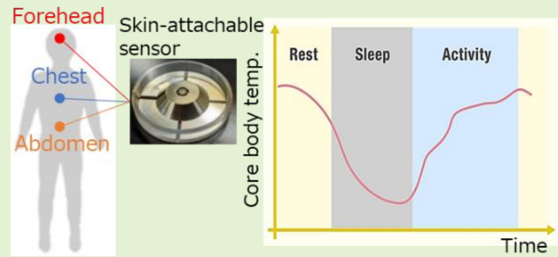
***Abstract*—This paper presents an in-vivo study on a skin-attachable core body thermometer. One of the biggest issues with skin-attachable sensors is lateral heat flow induced by convection, which results in overestimation of heat flow from the body. The proposed sensor has a unique aluminum structure to reduce lateral heat flow, which provides robustness against ambient convections. The ability of the sensor was verified in an in-vitro study in our previous work. In the in-vivo study reported here, core body temperature measurements were performed under conditions of fanning and light exercise. Although there were some transient and systematic errors at high blood flows or right after fanning, the sensor attached to the forehead showed good agreement with the reference tympanic temperature, with a correlation coefficient of 0.96 and RMSE of 0.1℃.**



***Index Terms*—core body temperature, skin-attachable, non-invasive.**

I. Introduction

CORE body temperature (CBT), i.e., the temperature of internal organs located deep in the body, such as the brain, heart, and liver, is an important indicator of health. CBT usually shows a 24-hour rhythm linked with the internal clock of the body [2]. However, CBT becomes arrhythmic in some diseases. For example, patients suffering from depression often show a lack of rhythm whereby CBT during sleep at nighttime is the same as in the daytime. Therefore, a continuous CBT measurement could be used as a tool for detecting some diseases in their early stages, even when patients are asymptomatic. Moreover, a continuous CBT measurement may detect a phase gap between one's internal clock and the social clock, which has been reported to induce metabolic aberrations and depression related to diabetes, cancer, and sleep disorders, etc., and also decrease exercise and work performance [1-8].

It was reported that a reliable procedure for continuous monitoring of CBT is to place a thermometer probe in the rectum [5]. However, this procedure is stressful to the patient, with accompanying risks of infection and damage to the rectal cavity. For these reasons, in-ear, ingestible, and skin-attachable sensors have been explored as alternatives [9-18]. Here, long-term placement of an in-ear sensor can cause discomfort. In addition, the measurement is sometimes influenced by environmental conditions and movements. Ingestible sensors have a limitation on the length of time they can measure in addition to the problem of collecting the sensor after use. Different from these sensors, skin-attachable sensors indirectly measure CBT by estimating heat flux from the skin surface. Although their methodology has not been fully developed, we think that the skin-attachable sensors are feasible for daily CBT monitoring. There are two types of skin-attachable sensor. One type heats the skin to obtain a thermal equilibrium state between the skin and body core (the zero-heat flux method) [11, 12]. The zero-heat flux method heat the skin and close the skin temperature to CBT. This method provides a highly accurate and traceable CBT measurement. However, its usage is assumed to be for patients in hospitals. Additionally, its high energy consumption makes it unsuitable for portable and wearable usage. The other type estimates CBT by using the skin temperature and the heat flax measured on the skin surface. This type of the CBT sensors using two or four temperature sensors was proposed in [14, 15]. Several sensors using this method are commercially available [19, 20]. These sensors provide fully passive CBT measurements and are suitable for continuous measurements, although these sensors are vulnerable to the effects of surrounding environmental changes. It is assumed that the sensors are to be used in stable environments or in a hospital like the first type.

First, let us clarify the issue we tackled by referring to a classical model. According to Pennes's mathematical models of heat transfer in humans [21], the principle of CBT estimation is expressed as follows:

$$C_{Tissue}\rho_{Tissue}\frac{\partial T}{\partial t} = k_{Tissue}\Delta T + \rho_{Blood}C_{Blood}w\left(T_{Artery} - T\right) + Q, \quad (1)$$

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where T is temperature, t is time, k is thermal conductivity, ρ is density, C is specific heat, w is circulation, Q is internal heating, and Δ is the Laplacian operator. Here, we make three assumptions to derive a CBT estimation equation. First, we assume that the left side of (1) can be ignored since the temperature of the body changes slowly (hereafter referred to as the steady approximation). The second assumption is that the second and third terms on the right side of (1) can also be ignored when the circulation and internal heating are small; in other words, we can ignore the circulation and internal heating. Third, when the thermal properties of the tissue are uniform and the temperature values are distributed only in the depth direction, namely when we assume a one-dimensional (1D) approximation, the first term on the right side of (1) becomes

$$\frac{d^2T}{dz^2} = 0 \tag{2}$$

where z is the coordinate in the depth direction. Since the temperature distribution becomes linear from the body core to the skin, CBT can be obtained by measuring the skin temperature and the heat flux at the skin. This situation can be explained with an analogy to the heat equivalent circuit shown in Fig. 1, and it can be simply realized with a sensor attached to the skin that measures skin temperature T_{Skin} and heat flux H_{Body}:

$$T_{CBT} = T_{Skin} + R_{Body}H_{Body}, \tag{3}$$

where R_{Body} is a constant coefficient related to the thermal properties of the body and the sensor. For example, R_{Body} increases with decreasing thermal conductivity or increasing depth of the body core region. R_{Body} is obtained at the start of the monitoring by using a reference temperature (e.g., tympanic temperature). However, the assumptions mentioned above are easily disrupted in daily life. For instance, as depicted in Fig. 1 [22], ambient convection induces a lateral heat flow H_{2D} and the estimation error increases immediately. The color map shows the temperature in the body. Because of the presence of the sensor, the temperature distribution near the sensor is distorted, which induces a lateral heat flow. We have demonstrated that a truncated cone structure made of a material with high thermal conductivity can reduce the lateral heat flow, and we have verified this ability in in-vitro studies [22]. On the other hand, addressing the assumptions of the steady approximation and that circulation and internal heating can be ignored requires considerable discussion based on an in-vivo study. In this work, we addressed them by examining the effect of the sensor's location on the body (forehead, chest, and abdomen) and compared blood-flow-based temperature measurements with a reference temperature (tympanic temperature).

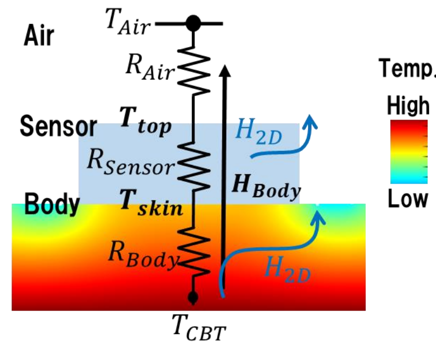

Fig. 1. CBT estimation model. The color map in the body area shows the temperature. T_{air} is the ambient temperature, and T_{skin} and T_{top} are the temperature of the skin and that above the skin. H_{body} is the heat flux transferred from the body core. H_{2D} is lateral heat flow induced by the ambient convection. T_{CBT} is CBT. R_{air}, R_{sensor}, and R_{Body} are the thermal resistances of the ambient air, sensor, and body, respectively.

II. EXPERIMENTAL

The positions of the sensor and a photograph and illustration of the skin-attachable sensor designed via topology optimization are shown in Fig. 2. The diameter and thickness of the sensor are 30 and 5 mm, respectively. As shown in Fig. 2 (a, b), the sensor has a truncated-cone structure made of aluminum with a thickness of 0.5 mm to reduce lateral heat flow. Two thermistors coated by Teflon (LT-2T-02, Gram, Japan) are integrated in the cylindrical polymer part made of acrylonitrile butadiene styrene (Fig 2 (c)). The diameter of the thermistor is 1 mm (Fig 2 (d)). It consists of main body to hold the thermistors and beams to hold chassis (Fig 2 (e)). The main body has a diameter of 8 mm and has 2 holes to hold thermistors. The vertical distance between the two thermometers is 2.5 mm. The thermistors were connected to a data logger (LT200, Gram, Japan) as shown in fig 3. The thermistors were calibrated by the manufacture with an accuracy of ±0.01°C under the condition that the temperature is between 10 and 50°C. The temperature difference between the thermometers is taken to be H_{Body} in the CBT estimation (3); the temperature difference is assumed to be proportional to the heat flux. Six healthy volunteers participated in the study. The experimental protocol was approved by the Ethical Committee of Human Research, Waseda University **(**2018-287**)** and was conducted in accordance with the Declaration of Helsinki. The skin-attachable sensor was placed on the forehead, chest, and abdomen with double-coated tape. Blood flow at the forehead was measured by a laser Doppler blood flow meter (ALF21, Advance, Japan). Tympanic temperature was monitored by an infrared thermometer (NIPRO CE THERMO, Nipro, Japan) as the CBT reference temperature since it can be easily measured and is highly correlated with the temperature of the body core [23, 24]. However, the tympanic temperature is easily affected by the ambient circumstances [25-27]. For this reason, the volunteers put on earmuffs to prevent the environment from affecting the tympanic temperature. The study was performed at an ambient temperature of 28°C and 50% humidity. To accommodate practical usage scenarios, we set the temperature relatively high. This decision was made to mitigate physiological responses, ensuring that individuals can comfortably exist without needing to engage in temperature regulation measures while nearly naked at the neutral temperature. After 1-h of stabilization of the skin-attachable sensor, each volunteer was exposed to wind (1 m/s) produced by an electric fan and performed light exercise on a bicycle ergometer. The wind exposure induced a lateral heat flow, and the exercise promoted circulation and internal heating. At the start of the experiment, R_{Body} was calculated from the reference tympanic temperature.

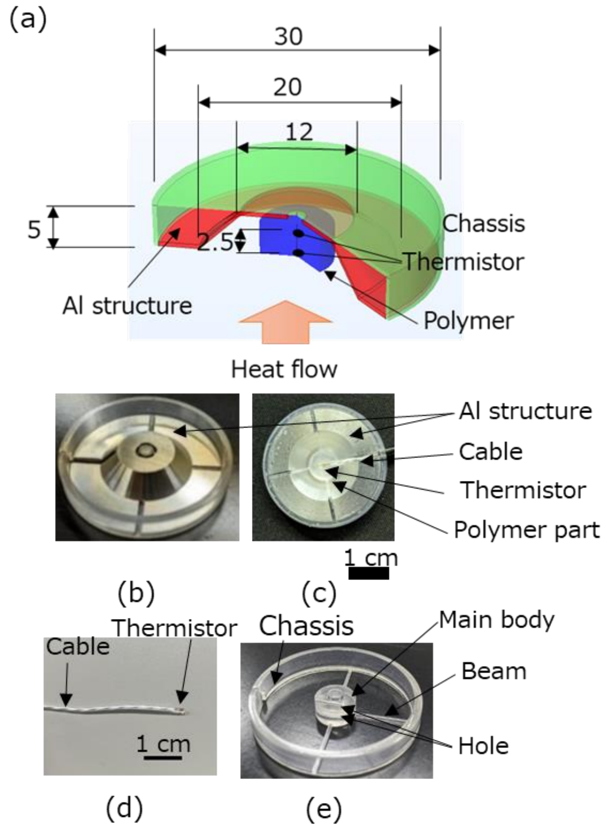

Fig. 2 Skin-attachable sensor for core body temperature monitoring. (a) Schematic illustration of the sensor. The light-green, red, and blue areas are the chassis, Al structure, and cylindrical polymer with two integrated thermometers. (b) Bird view of the sensor. (c) bottom view of the sensor. (d) Photograph of thermistor and its cable. (e) Photograph of cylindrical polymer part to integrate thermistors.

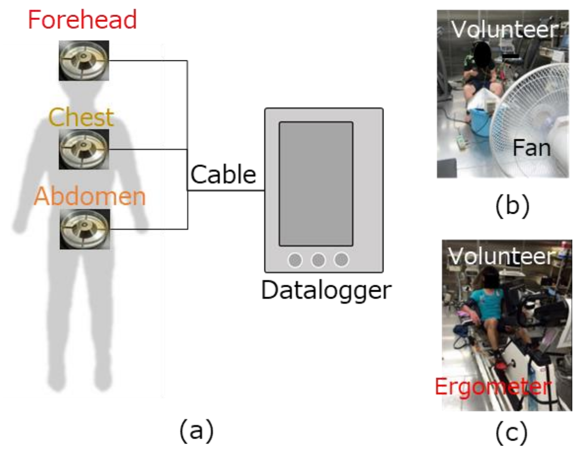

Fig. 3 Experimental configuration of in-vivo study. (a) Experimental configuration. Thermistors integrated in the sensors are connected to the datalogger. (b, c) Photograph of the attached sensor under fanning and ergometer.

III. Results and Discussion

R_{Body} values for the six volunteers are shown in Fig. 4. The coefficient was calibrated at the beginning of the experiment by using the reference tympanic temperature. The red, blue, and orange bars are the coefficients at the forehead, chest, and abdomen, respectively. The averages and standard deviations for the sensor-attachment positions are shown in Table I. Focusing on the position dependence of the R_{Body}, the average R_{Body} at the forehead was three times smaller than that at the other positions. We are utilizing the tympanic temperature nearest to the forehead, as it demonstrates the least dispersion in the R_{Body} across the forehead. This implies that the distance between the skin surface and body core, i.e., the “skin-body core length”, at the forehead was also three times smaller than at the other parts, which makes it easy to satisfy the 1D approximation at the forehead. In terms of the components of the tissue at each position, the forehead is mostly composed of bone with thickness of 10 mm, and the abdomen has a lot of fat tissue, which has low thermal conductivity, and also has many more capillaries compared with the forehead. The standard deviation of R_{Body} at the forehead was the smallest. In other words, the individual differences of the forehead were the smallest, while those of the abdomen were ten times larger than that of the forehead. R_{Body} at the abdomen easily depended on the body shape. As for the experimental error, R_{Body} might also be influenced by the contacting condition between the measurement device and skin surface, which varied among subjects because of the anatomical shape of the forehead. The standard deviation may also reflect for the influence.

TABLE I
AVERAGE AND STANDARD DEVIATION OF COEFFICIENT R_{BODY}

Position	Average	Standard deviation
Forehead	3.2	0.6
Chest	8.0	2.6
Abdomen	9.7	5.6

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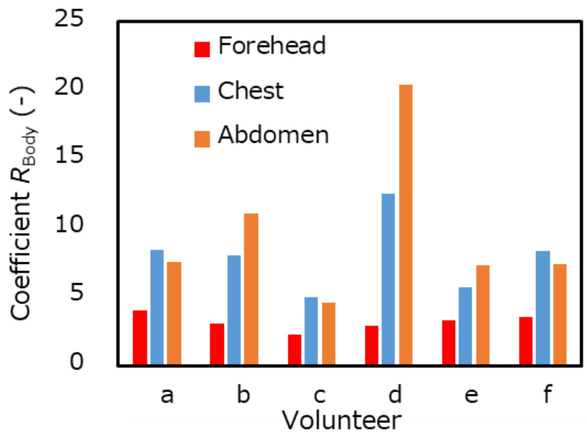

Fig. 4 Coefficient R_{Body} for CBT estimation of six volunteers. Red, blue, and orange bars are respectively R_{Body} values at the forehead, chest, and abdomen used in the CBT estimation.

TABLE II
THERMAL CHARACTERISTICS OF BODY. [26]

Component	k (W/m/K)	C (J/kg/K)	ρ (kg/m^3)	τ (min)
Skin	0.47	3680	1085	14
Subcutaneous fat	0.21	2300	920	17
Muscle	0.51	3800	1085	13
Lungs	0.28	3520	560	12
Bone	0.75	1700	1357	5

Thermal conductivity k, specific heat C, density ρ, thermal time constant τ calculated under condition that the skin-body core length is 10 mm.

As a typical result, the skin temperature, temperature difference H_{Body}, estimated CBT, and blood flow profiles for volunteer 1 are shown in Fig. 5, where the red, blue, and orange, lines are the values recorded with the skin-attachable sensor attached to the forehead, chest, and abdomen, respectively. The black line in Fig. 5 (c) is the reference tympanic temperature. The green and yellow labels show the duration of the fanning and exercise. Here, the skin temperature immediately decreased right after fanning by a few degrees at each sensor position. The decrease in skin temperature at the abdomen was moderate. Here, the time constant of heat transfer can be evaluated as

$$\tau = \frac{L}{k} \cdot \rho L C, \tag{4}$$

where L is the skin-body core length. Typical characteristics of the body components and the thermal time constant under the assumption that the skin-body length is 10 mm are shown in Table II. The abdomen has more subcutaneous fat than other parts and it has a larger time constant. This induces a moderate response at the abdomen. Additionally, the abdomen was covered with the cloth of the volunteers' clothing.

On the other hand, the relation $H_{Body} = T_{Skin} - T_{Top}$ increased right after fanning. As for the estimated CBT in Fig. 5 (c), the skin temperature was compensated by 1 to 2 degrees by the effect of R_{Body} at each position. The estimated CBT at the forehead was in good agreement with the tympanic temperature, while the estimated temperatures at the chest and abdomen were somewhat different from the tympanic temperature, especially, during the fanning and the exercise. This implies that the apparent coefficient at those positions may vary due to lateral heat flow. Thus, it is difficult to satisfy the 1D approximation when the skin-body core length is large and the wind exposure due to the fanning enhances the lateral heat flow. Another source of error is gas exchange in the lung. However, the gas exchange reduces the internal temperature. In that case, the temperature would have been underestimated since the room temperature is lower than the internal body temperature. Accordingly, we assumed that the effect of gas exchange is small.

As for this evaluation protocol of the study, the forehead is the best position for the CBT estimation with the skin-attachable sensor.

Uncertainty of this result includes not only the accuracy of the thermometer but also the uncertainty associated with the tympanic temperature sensor. Although, the uncertainty in temperature measurements requires assessment against calibration standards, the accuracy of the thermometer integrated in the skin-attachable sensor is 0.01°C according to the manufacturer's specifications. The maximum permissible error for the tympanic temperature sensor is 0.2°C. Therefore, challenges still remain in the assessment of accuracy.

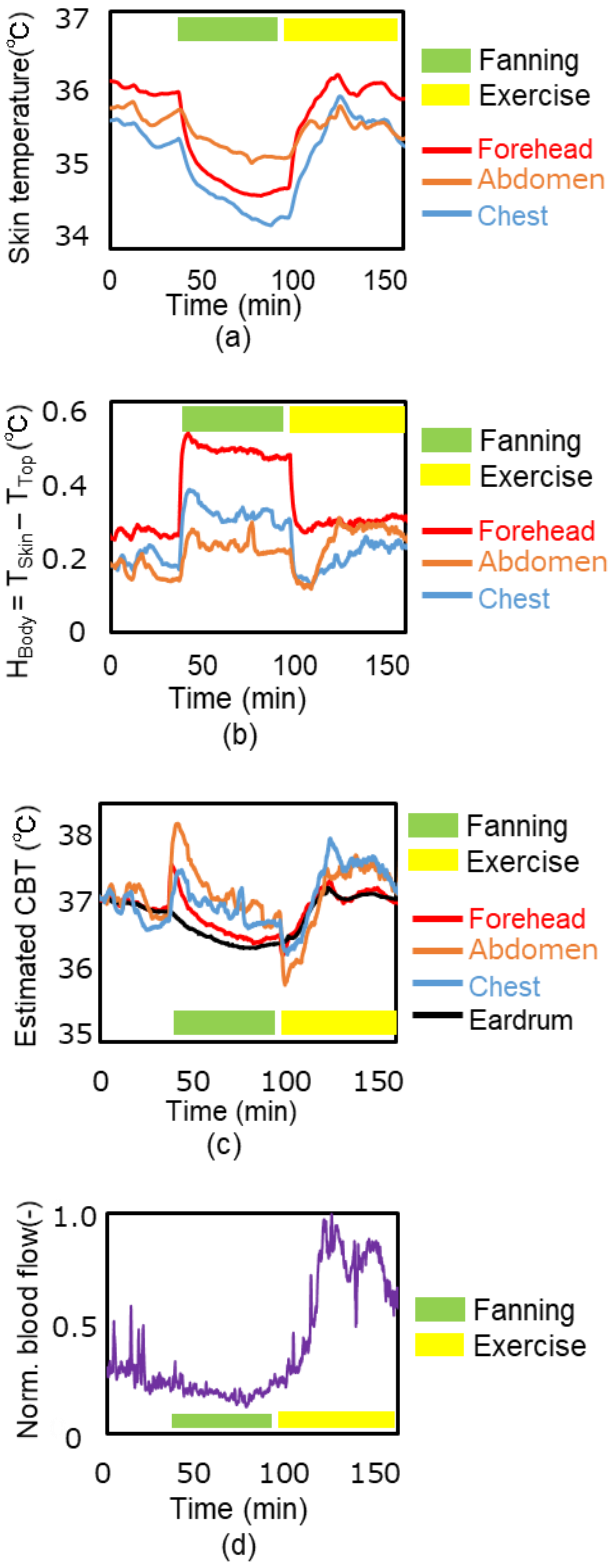

Fig. 5 Typical results of the skin temperature, temperature difference H_{body}, estimated CBT and reference tympanic temperature. Red, blue, orange, and black lines are the values measured with the skin-attachable sensor attached to the forehead, chest, and abdomen, and the reference tympanic temperature, respectively. The green and yellow labels show the durations of the fanning and the exercise.

Typical blood flow profiles at the forehead are shown in Fig. 5 (d). The maximum (minimum) blood flow rate is normalized to 1 (0). The blood flow was measured by a laser Doppler flow meter, which measures the blood flow in small capillaries in tissues from the skin surface to a few millimeters depth [29, 30]. As for the forehead, the thickness of the skin is about 2 mm [31]. Accordingly, the blood flow measured by the laser Doppler flow meter may cover the whole thickness of the scalp, although it gives a relative value, and does not directly relate to a quantifiable value, such as blood perfusion. However, it could relate to an increase or decrease in blood perfusion. The blood flow gradually increased after exercise. In particular, it increased by more than two times from the beginning. The estimated temperature at the forehead was in good agreement with the reference tympanic temperature. This implies that the capillary layers are small enough to not change the R_{body} coefficient even when the blood flow changes in the forehead, whereas the capillaries in the chest and abdomen are large enough that CBT might swing a lot due to changes in blood flow. On the other hand, there is a gap between the estimated temperature and reference tympanic temperature right after the fanning. The estimated CBTs show a large 10-min transient deviation from the tympanic temperature by about 0.5℃ right after the fanning. The disruption of the steady-approximation assumption caused the deviation, since the temperature under the skin moves rapidly with the time constant of heat transfer. From (4), the time constant is 10 min. This implies that the continuous change in the wind condition decreased the accuracy, or this skin-attachable sensor required 10 min for stabilization. The variables used in the calculation are shown in Table II. The transient error could be reduced by using the filter described in [22]. We determined the transient region by using the standard deviation of the estimated temperature σ in 10 min. If the estimated temperature exceeded 3 σ, we regarded it as transient error and removed it by using a fitting function in [22]. The fitting range extends from where it exceeds 3 σ to where it returns to the same temperature. The filtered estimated CBT is obtained by subtracting the fitted result from the estimated CBT. An example of estimating the CBT filtering is shown in Fig. 6. The black and red lines show results with and without filtering, respectively, and the green line is the reference tympanic temperature. The filtering process works both at the start and the end of the fanning. The transient errors right after the fanning are clearly removed by the filtering. However, this method requires sufficient time for the convection to transition from a change to a steady state. Therefore, further refinement is needed for cases where convection changes continuously.

The relationship between tympanic temperature and CBT estimated at the forehead with this filtering process are shown in Fig. 7. The root mean square error was 0.14℃. The correlation coefficient was 0.96. The skin-attachable sensor attached to the forehead showed good agreement with the tympanic temperature. It is expected that the estimation error would increase at an ambient temperature lower than in this work (28°) because it would be more difficult to satisfy the 1D approximation and the lateral heat flow would increase.

Additionally, there is still room to consider the differences from reference temperature, although the heat source of both the tympanic temperature and the deep forehead temperature is the blood supply from the internal and external carotid arteries and the brain.

TABLE II

PARAMETERS TO EVALUATE HEAT TRANSFER TIME CONSTANT

Variable	value
Skin-body core length	10 mm
Thermal conductivity	0.6 W/m/K
Density	1000 kg/m^3
Specific heat	4200 J/Kg/K

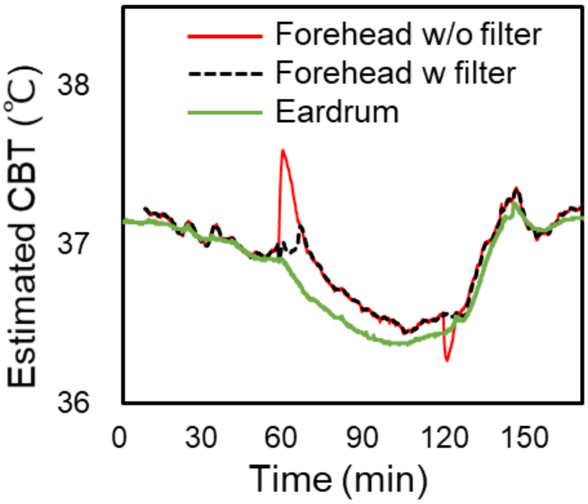

Fig. 6 Typical results of transient error filtering. The black and red lines indicate CBT estimated with and without filtering, and the green line is the reference tympanic temperature.

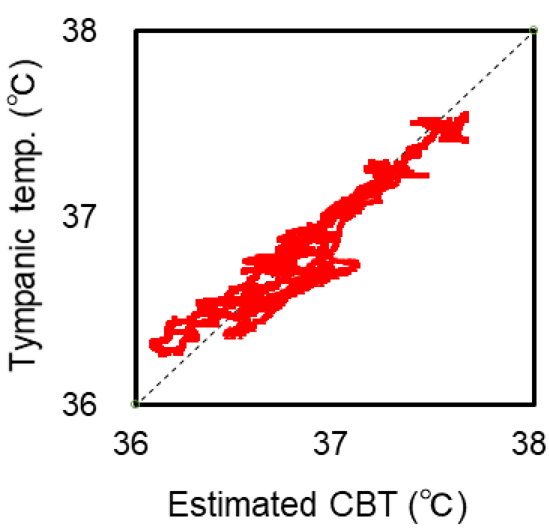

Fig. 7 Relationship between reference tympanic temperature and CBT estimated at the foreheads of the six volunteers.

IV. Conclusion

A skin-attachable sensor with a unique aluminum structure to reduce lateral heat flow was verified in an in-vivo study. CBTs estimated with the proposed sensor attached to the forehead, chest, and abdomen were compared with the tympanic temperature under the conditions that volunteers were at rest sitting in a chair, exposed to wind, and performing light exercise. R_{Body} of the CBT estimation equation at the chest and the abdomen was more than twice as large as at the forehead, which is related to the skin-body core length. Therefore, the sensor showed large differences (about 1℃) from the tympanic temperature when it was attached to the chest and abdomen during the wind exposure and exercise, which was caused by a violation of the 1D approximation.

The CBT estimated with the sensor attached to the forehead showed good agreement with the reference tympanic temperature, with an RSME of 0.1℃ and correlation coefficient of 0.96, while a small difference was found due to the change in circulation or difference in the measured positions between the forehead and ear. The skin-attachable sensor thus shows high potential as a CBT monitoring tool. However, the uncertainty of this result includes not only the accuracy of the thermometer but also the uncertainty associated with the tympanic temperature sensor.

Although the ambient temperature and humidity were controlled in this experiment, in our future work it will be necessary to perform tests under various conditions. Moreover, the initial calibration of the coefficient R_{Body} is troublesome for practical use; a “self-calibration” method will be incorporated in the future.

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