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Body Core Temperature Estimation Using New Compartment Model With Vital Data From Wearable Devices

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ABSTRACT With increasing heat-wave frequency, the prevention and public awareness of heat-related illnesses has become an essential topic. In the standard for heat strain and stress, empirical guidelines to prevent excess core temperature rise above 1 °C have been prescribed for workers. However, measuring core temperature change in our daily life or working place is not straightforward. The estimation of core temperature from measured vital signals in a non-invasive manner is thus essential for the management of heat stress or strain. Here, we propose an estimation method for core temperature change by a simplified thermodynamics model with the measured heart rate and ambient conditions (temperature and relative humidity). Our proposed model is based on a two-layer two-compartment model with tuned parameters, which were derived from comparison between the computations using high-resolution anatomical human body model. Our model exhibited good agreement with the measured core temperature rise; the computed and measured core temperature rise for the naked trial were 0.54 °C and 0.53 °C, whereas those for the clothed trial were 0.70 °C, and 0.71 °C, respectively. Furthermore, our compartment model with vital data measured from a wearable device achieved good estimation in real time for field measurement in addition to computational replication with a previous study.

INDEX TERMS Bioheat equation, wearable sensors, heart rate, thermoregulatory response, heat-related illness.

I. INTRODUCTION

The number of heat wave cases has been increasing in most areas of the world [1], [2]. With increasing frequency of extreme heat waves, morbidity of heat-related illnesses has been increased [3]. In Japan, the number of fatalities from heat stroke increased and reached 492 per year after 1994 [4]. The number of people transported by ambulances has been increasing gradually; 40,000−60,000 from 2010 to 2017 and reached a record high of 95,137 in 2018. Since then, it has remained high at 71,317 in 2019 and 64,869 in 2020 [5].

Several indexes, such as the heat index [6], wet-bulb globe temperature (WBGT) [7], and the universal thermal climate index [8], are used to estimate the heat stress and strain as well as thermal comfort [9], [10]. These indexes are metrics of heat-related risks assuming that an individual remains in

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the same environmental condition; however, their daily maximum values or instantaneous values, such as when workers start to work, are often used. In our daily life, environment changes with time, and the duration of heat exposure is case-specific.

Heat strain and tolerance may depend on age and environment [11], [12]. One typical case is that of athletes and workers in summer events or workplaces [13]. It is impossible to change the schedule of major sport events because of weather information (e.g., Summer Olympic Games, World Athletic Competition), which has been determined a priori, several years ago.

Protection of workers and volunteers from heat-related illnesses is essential for the success of these events. One difficulty associated with managing their health is partly attributable to the variability of the individuals, which is characterized by their ages, heat accumulation, etc.

For workers' safety, the American Conference of Government Industrial Hygienists (ACGIH) [14] publish and periodically maintain standards for managing work duration using an empirical equation in terms of the WBGT. In [14], workers are assumed to wear long sleeve shirts and pants (work clothes). Thus, this assumption may not be applicable in daily life activities and workplaces where the heat load is relatively mild. The main rationale of the standard is to maintain the core temperature below 1 °C (1.5 °C for heat accumulated) to avoid heat strain. Additionally, guidelines for water intake are suggested. It would be useful if highly-reliable risk management was possible for different and time-dependent cases and humans considering their individual activity.

Estimation of temperature and temperature change is one of the essential topics in different applications, including product development and management, comfort of buildings, and controlled greenhouses [15]–[18]. A possible solution to prevent the heat stroke for workers [19] is to estimate the time course of the core temperature and water loss in the time domain using vital data for each individual. However, the core temperature cannot be measured directly; thus, its estimation from available data is essential. Wearable sensors or devices are required for this purpose. The amount of vital data available from wearable devices are typically limited [20]; e.g., the heart rate, physical activity (gait change), and cardiac activity. One of the promising approaches is to apply textile materials as sensors [21]–[24]. Several attempts have been made to monitor heart rate [25], [26], gait [27], and so forth using wearable sensors.

The heat balance equation can then be used to estimate the body temperature based on metabolism including basal metabolism and physical activity [28]. Compartment models, which are comprised of several body parts, were used to estimate the time course of temperature elevation as well as thermoregulation (mainly for the evaporation and vasodilatation for heat stress) [29], [30]. Recently, computational models with an anatomical human body model have been proposed by combining thermodynamics and thermoregulatory response, in addition to electromagnetics (solar radiation). The change in the metabolic rate during exercise mainly occurs in muscles and is position-dependent. The anatomical human models have more flexibility to consider this effect. The computational model was validated empirically for different scenarios, including the assessment of the core temperature elevation during exercise in a warm room (e.g., [31]).

However, it is not realistic to use such anatomical models for real-time monitoring of the core temperature rise. Another possibility is to use an empirical model, e.g., multi-layer cylinder model or simple two-layer model. The limitation of such a model is that a set of parameters derived in one scenario may not be applicable to different scenarios, which is attributable to simplification.

The purpose of this study is to estimate core temperature rise, which is not directly measured, from measured vital data in real time. To be realized this, a two-layered two-compartment model is developed based on computations using anatomical human body models. Its applicability in real-time estimation is demonstrated by comparing the computed core temperature rise with those in experiment in chamber as well as a field experiment. The predicting system of individual indexes monitoring heat rate and skin temperature from wearable devices has been reported [15], [32]. However, to the best of authors knowledge, this is the first approach to estimate core temperature, which is the direct cause of heat related illness, in real time by wearable sensing of biological signal. The novelty of our method is that the significant reduction in computational resources/execution time by representing he conventional thermodynamics-based method [31] in a simplified model and is possible to be implemented on a wearable device.

II. REVIEW OF THERMAL COMPUTATION MODELS

Computational methods for modeling body temperature have been developed extensively to estimate the body temperature [33], [34]. The modeling methods are classified into two types: thermodynamics models based on the bioheat transfer equation, and non-physical models based on empirical equations such as regression analysis.

A. MODELS BASED ON THERMODYNAMICS

The thermodynamics-based method is based on the computation of temperature change by solving Pennes' bioheat transfer equations [35] considering thermoregulation. First, Stolwijk [29] proposed a hybrid thermodynamics and thermophysiological response model, which consisted of three cylinders arranged in concentric layers. Subsequently, the model was improved to the six compartments model. A twelve spherical compartments model, whose skin had two layers—the blood-perfused inner layer and outer skin containing sweat glands—was proposed in [30]. The compartment model was extended to consider the anatomical human model, composed of voxels with a resolution of 2 mm, which was segmented into 50 tissues [36]. We extended the model to include hybrid thermal models, considering anatomical body models for local temperature and a compartment model for the core temperature [31].

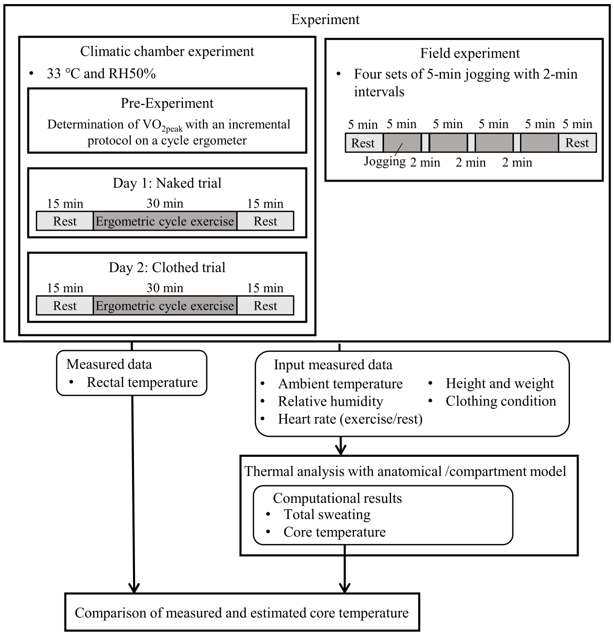

FIGURE 1. Block diagram to summarize two experimental conditions. The first experiment was designed for validation. The second experiment was designed for demonstrating our proposal using data obtained from wearable device.

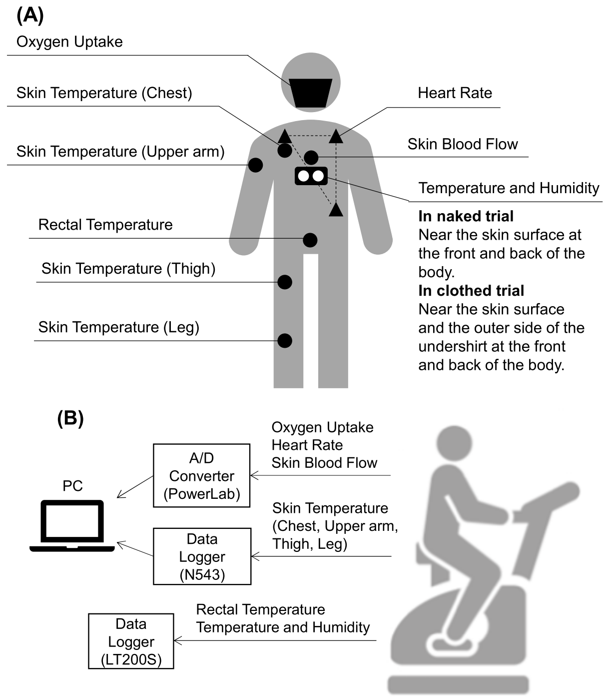

FIGURE 2. Schematic of experimental setup for naked and clothed trials. (A) Measurement site and (B) Experiment setup in the first experiment. In the second experiment, the wearable sensor attached to the chest was used for obtaining the input data, in addition to the rectal temperature measurement for verification.

A feature of these models is that they are applicable to different environmental conditions and individual activity without adjustment. Instead, the models are often unsuitable for real-time estimation due to the trade-off between the prediction accuracy and computational resources/execution time.

B. NON-PHYSICAL MODELS

Methods for core temperature estimation with non- or less invasive measures, using the heart rate, respiration rate, and skin temperature, are proposed [37]–[39]. The zero heat flux method was developed for estimating core temperature [40]. A heat flux device, which mathematically predicted the core temperature was proposed considering skin temperature [41]. An estimation algorithm with the Kalman filter, using a series of heart rates during a road march was developed, which is used as a real-time thermal indicator in militaries [38]. The method was extended to estimate the resting core temperature by combining a sigmoid function [42]. Core temperature estimation using the measured skin temperature, skin heat fluxes, and heart rate, by multiple regression analysis was proposed [37]. The estimation of the circadian body temperature rhythm was proposed using the extended Kalman filter combined with the heart rate [43].

The models have a simple algorithm and are suitable for real-time estimation with small computational resources. Most models are developed based on the data processing of measurements [37]–[39], [42], [43], and thus their applicability to different scenarios are unknown. The extended Kalman filter requires an additional protocol for accumulating initial values in real-world usage [43].

III. EXPERIMENTAL CONDITION

The block diagram of this study is summarized in Fig. 1. Two experiments with exercise were performed. The first experiment is designed to validate a new two-layer two-compartment model. Then, in the second experiment, the effectiveness of the proposed techniques was demonstrated by comparing the measured and computed core temperatures using measured vital data and weather data as input data.

A. SUBJECTS AND ETHICAL APPROVAL

Eight healthy men with a mean (±SD) age of 23 ± 1 years, height of 169 ± 3 cm, and body mass of 59.0 ± 7.5 kg voluntarily participated in the laboratory experiment of this study. Additionally, two subjects were selected from the eight subjects to conduct a field experiment. Each subject provided written, informed consent after all potential risks and procedures were explained. All experimental procedures and protocols conformed to the Declaration of Helsinki and were approved by the Institutional Review Boards of Shigakkan University (IRB # 132). None of the subjects were taking any medication that may have influenced hemodynamic responses to exercise.

B. PEAK OXYGEN UPTAKE ($\dot{V}O_{2peak}$) MEASUREMENT

As a pre-experimental study, the $\dot{V}O_{2peak}$ was determined using an incremental protocol on a cycle ergometer (Corical cpet; Lode, Netherlands), one week before the main experiments. The subjects were exposed to an initial work rate of 30 W at a pedal rate of 60 rpm. The subjects were asked to maintain the pedaling frequency, and the work rate was increased by 15–20 W every minute until volitional exhaustion. The respiratory variables were determined using a breath-by-breath system (AE-310S; Minato Medical Science, Japan). The highest value obtained for $\dot{V}O_2$ over a 60-s interval was regarded as the $\dot{V}O_{2peak}$.

C. EXPERIMENTAL DESIGN

This study was performed in two trials. One was a naked trial, in which the subjects were given a pair of shorts, socks, and shoes. The other was a clothed trial in which the subjects were given a long-sleeved shirt, an undershirt, a pair of long pants, socks, and shoes. Each trial was separated by at least 1 week from the preceding trial to avoid any effects of thermal adaptation and fatigue by exercise on the experiment.

The subjects were requested to refrain from taking food or drinks for at least 12 h and avoid caffeinated beverages, alcohol, and strenuous physical activity for at least 24 h before each experiment. Drinking water was allowed as needed.

On the experiment day, the subjects were weighed in the nude and given clothes and shoes after arriving at the laboratory. They were then made to insert a rectal thermistor. Subsequently, the subjects entered a climatic chamber controlled at 33 °C ambient temperature and 50% relative humidity. After the subjects were instrumented, they were seated on the cycle ergometer and rested quietly to allow for cardiovascular stability. Following 15 min of data collection, the subjects continuously conducted 30 min of ergometric cycle exercises at 50% $\dot{V}O_{2peak}$ frequency of 60 rpm, followed by 15 min of recovery. They then wiped themselves and were weighed again in the nude to estimate the total sweat loss.

Experimental setup is shown in Fig. 2. Oxygen uptake, heart rate and skin blood flow signals were sampled using analog-to-digital converter (PowerLab, ADInstruments, Milford, MA) interfaced with a computer. Data for skin temperature on the chest, upper arm, thigh and leg were collected with a data logger (N543; Nikkiso-Thermo Co., Tokyo, Japan) interfaced with the computer. Data for temperature and humidity near the skin surface and on the outer side of the undershirt, and rectal temperature were stored in a data logger (LT-200S; Gram Co, Saitama, Japan).

For two subjects, a field experiment was conducted from 10:00 to 10:50 on Oct 26, 2020, Ishigaki, Japan. In this experiment, following the seated condition for 5 min, four sets of 5-min jogging were conducted with 2-min intervals. Subsequently, measurements were taken for an additional 5 min in the recovery phase. The subjects wore a short-sleeved T-shirt, a pair of long pants, socks, and shoes. Wearable sensor.

D. MEASUREMENTS

The rectal temperature (T_{re}) was measured using a rectal thermistor (LT-ST08-11; Gram Co, Saitama, Japan) inserted 10 cm into the rectal ampulla. The skin temperature was measured using thermistors (409J, Nikkiso therm Co., Japan) placed on the right side of the upper arm (T_{arm}), chest (T_{chest}), thigh (T_{thigh}), and leg (T_{leg}). The mean skin temperature (T_{sk}) was calculated from the body surface area distribution and thermal sensitivity of each skin area using the following formula, which was proposed in [44]:

$$T_{sk} = 0.3 \times (T_{arm} + T_{chest}) + 0.2 \times (T_{thigh} + T_{leg}), \quad (1)$$

The temperature and humidity near the skin surface were measured using a thermistor (LT-2N-00; Gram Co., Saitama, Japan) and a hygrometer (LT-HM4S+LT-2HM4; Gram Co., Saitama, Japan) attached to the front and back of the body.

In the clothed trial, the temperature and humidity on the outer side of the undershirt were measured using a thermistor (LT-2N-00; Gram Co., Saitama, Japan) and a hygrometer (LT-HM4S+LT-2HM4; Gram Co., Saitama, Japan) attached to the front and back of the body. The heart rate (HR) was monitored using a lead II electro-cardiogram. The skin blood flow at the chest was measured by laser Doppler flowmetry (ALF21, Advance, Japan). Oxygen uptake (VO_2) was measured by a metabolic gas analyzer system (AE-310S; Minato Medical Science, Japan). The total sweat loss was estimated by determining the change in the dry body weight, which was measured immediately before and after the experiment.

In the field experiment, 'hitoe™,[1]' which integrated different types of sensors including the heart rate, temperature, and relative humidity, was used [45]. The heart rate is measured by wearing a garment in which hitoe™ is sewn in and an ECG transmitter which is electrically connected to hitoe™ is attached [45]. Temperature and humidity can be measured through sensors in the transmitter. Among others, the data of the heart rate were used as an input parameter to estimate the metabolic equivalents (METs). From a wearable device based on hitoe™ was transferred wirelessly to a smart phone. Ambient conditions, temperature, humidity, and wind velocity reported by the Japan Weather Association (JWA) were used to estimate the heat exchange between the body and air.

E. ESTIMATION OF METABOLIC RATE FROM HEART RATE

From the measured heart data, METs can be estimated by the following equation.

$$\mathrm{METs}(t) = VO_2/3.5 = 0.085 * (HR(t) - HR_{rest}) + 1, \quad (2)$$

where VO_2 is the oxygen consumption, and HR_{rest} is the heart rate in the resting condition. Using (2), the metabolic rate during exercise can be estimated from the following equation. In this study, METs is averaged for 6 minutes to simulate heat transfer delay in the core. Using (2), the metabolic rate during exercise $Ex(t)$ can be estimated as follows:

$$Ex(t) = \mathrm{METs}(t) * 1.05 * W * 4184/3600 \text{ [W]}. \quad (3)$$

[1] The performance fabric hitoe™ is a fiber material jointly developed by Toray Industries, Inc. and Nippon Telegraph and Telephone Corporation, and is the registered trademark of the two companies.

Note that for the estimation of the metabolic rate during exercise, pre-knowledge of the heart rate during resting condition is necessary.

IV. MODEL AND METHODS

A. HUMAN BODY MODEL

In this study, a Japanese adult male model was used for the reference temperature change for the heat load [46]. The height and weight of the model are 173 cm and 65 kg, respectively. The body surface area is estimated as 1.86 m^2 [47]. The adult model was divided into 51 anatomic regions (e.g., skin, muscle, bone, brain, and heart) with a resolution of 2 mm for temperature computation.

B. THERMAL ANALYSIS IN ANATOMICAL MODELS

The algorithm for computing the temperature variation considering thermoregulatory responses is summarized in Fig. 3. The review of our computational modeling is presented in the following subsections. It should be noted that the change in metabolism rate and the effect of age as well as computational implementation are shown in our previous study [31]. The computational results with the anatomical model are used as reference to derive parameters in a simplified model.

The temperature elevation in the numerical human models was calculated by solving the bioheat transfer equation [35] in a time domain, which models the thermodynamics of the body. A generalized bioheat equation considering thermoregulation and core temperature change [48] is expressed as:

$$C(\boldsymbol{r})\rho(\boldsymbol{r})\frac{\partial T(\boldsymbol{r},t)}{\partial t} = \nabla\cdot(K(\boldsymbol{r})\nabla T(\boldsymbol{r},t)) + M(\boldsymbol{r},t) + Ex(\boldsymbol{r},t) - B(\boldsymbol{r},t)\,(T(\boldsymbol{r},t) - T_B(m,t)), \quad (4)$$

where $T(\boldsymbol{r},t)$ and $T_b(m,t)$ denote the tissue temperature and blood temperature, respectively, of different body parts (where $m = 1, 2, 3, 4, 5, 6, 7, 8, 9, 10, 11, 12$, and 13 represent the head and trunk, right hand, right forearm, right upper arm, left hand, left forearm, left upper arm, right feet, right shin, right thigh, left feet, left shin, and left thigh, respectively); C [J/kg/°C] is the specific heat of the tissue; K [W/m/°C] is the thermal conductivity of the tissue; M [W/m^3] is the basal metabolic rate per unit volume; and B [W/m^3/° C] is a term associated with blood perfusion.

The boundary condition between the air and tissue for (1) is expressed by

$$-K(\boldsymbol{r})\frac{\partial T(\boldsymbol{r},t)}{\partial n} = H(\boldsymbol{r},t)\,(T(\boldsymbol{r},t) - T_e(t)) + EV(\boldsymbol{r}), \quad (5)$$

where H [W/m^2/°C], T [°C], and T_e [°C] denote the heat transfer coefficient, body surface temperature, and air temperature, respectively. H includes the convective and radiative heat loss, and EV [W/m^2] is the evaporative heat loss.

A key feature of our computational modeling is that, unlike conventional bioheat modeling, both the body core temperature variation and temperature in the shallow regions can be tracked. The blood temperature depends on the surrounding tissues, according to the formula in [30] and was validated in our previous study [49]. The heat transfer of the blood flow between different body parts was then considered.

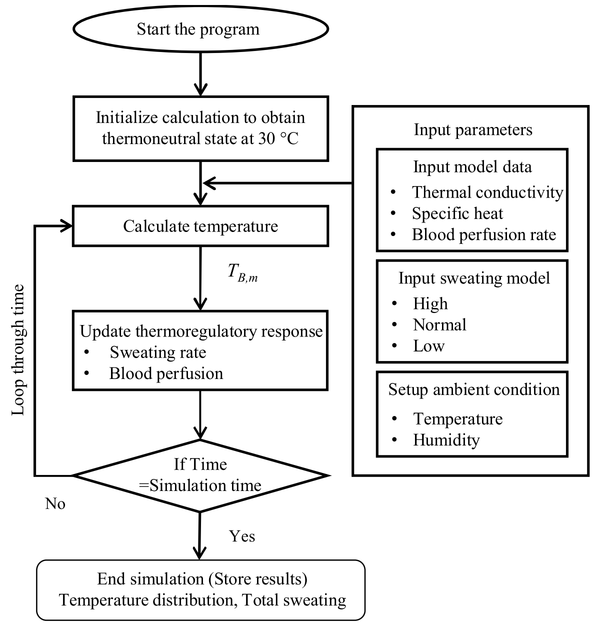

FIGURE 3. Flowchart of bioheat modeling using thermoregulatory response in computational domain.

Sweating is modeled using the formulas in [30]. The sweating rate is assumed to depend on the temperature rise on the skin and in the hypothalamus. The dependence of the sweating rate on the body parts is considered. For the anatomically based model, its parameters were derived in [50]. The evaporative heat loss on the skin depends on the ambient conditions. The heat loss via sweating was determined in the equation given in [30].

The thermal constants of human tissues and the heat transfer coefficients used in this study are identical to those in [51]. The heat transfer coefficient from the skin to the air, including the insensible heat loss is considered. The numeric phantom used is discretized by voxels; thus, its surface is approximately 1.4 times as large as that of an actual human [52]. The heat transfer coefficient is adjusted by the ratio between the actual and voxelized body surface area.

The computational code using an anatomical human model was optimized to a supercomputer with high vector processing performance as same in [31]. The 1-h estimation of core temperature is possible in 30 s on the supercomputer with 64 nodes (12.5 GB).

C. EMPIRICAL COMPARTMENT MODEL

Our main aim is to estimate the time course of the core temperature elevation. One of the dominant factors affecting the core temperature elevation is sweating, which results in evaporative heat loss. As mentioned above, the thermoregulatory response or sweating is characterized by the average skin and core temperature elevation. Thus, we attempted to estimate the core and skin temperature change using the two-layered

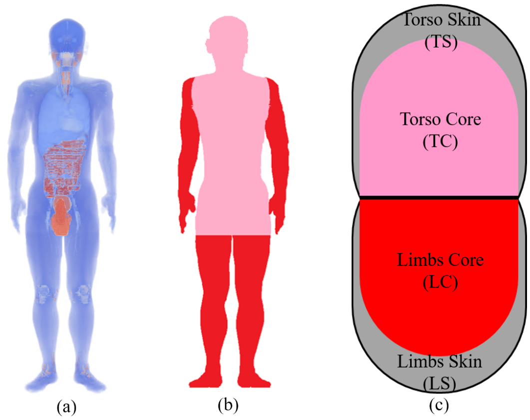

FIGURE 4. (a) Anatomical human model, (b) its compartment separated into head and core, and limbs, and (c) schematic diagram of proposed two-layer two-compartment model.

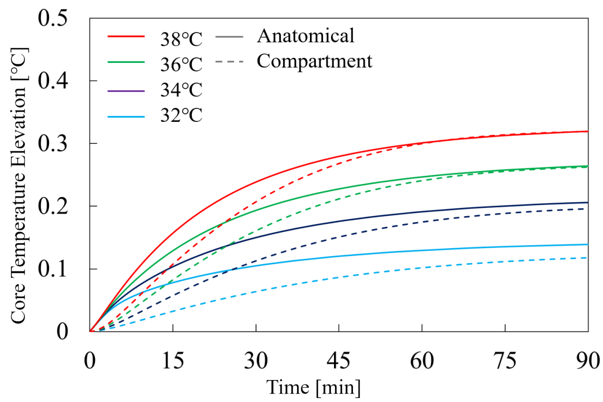

FIGURE 5. Time course of body-core temperature elevations in the anatomical and compartment models. The models are assumed to be resting and naked.

two-compartment model, as shown in Fig. 2. The surface area is empirically derived from the DuBois equation using the weight and height of each subject [47].

The core temperature is assumed to be similar to that in the "Torso Core" layer. The body temperature change is then obtained from the following equations. The skin temperature is assumed to be characterized by the evaporative heat loss and heat transfer between the air and skin layer. Assuming that the effect of the blood flow is not considered in the layer, this implicitly assumes uniform temperature distribution in each layer. The empirical heat exchange equations are given by the following (6)–(9), as shown at the bottom of the page, where hx_{SC} and hx_{CC} [W/°C] are the heat exchange coefficients between the skin and body and that between the torso core and limbs core, respectively, which depend on the body temperature. W [kg] denotes the weight of each layer, and the subscript denotes the value for the layer. The thermal parameters for the body layer were empirically estimated as the averaged values of the anatomical human body model. From (10) and (11), the skin and body temperature change are estimated empirically with very low computational costs.

The term expressing the heat exchange between the skin and body layers as well as its coefficient are assumed. The heat exchange coefficient is assumed to be characterized by the following equations in terms of body core and skin temperature elevations in an analogy of [29]:

$$hx_{SC} = hx_0 + hx_1^{\Delta T_{core} + \Delta T_{skin}/\alpha + (METs-1)/\beta}, \tag{10}$$

$$hx_{CC} = hx_2 + hx_3^{\Delta T_{core} + \Delta T_{skin}/\alpha + (METs-1)/\beta}, \tag{11}$$

where $hx_0 = 27.5$ W/°C, $hx_1 = 24.4$ W/°C, $hx_2 = 16.5$ W/°C, and $hx_3 = 13.5$ W/°C are the constants which are derived so that the computational results derived with the anatomical and compartment model coincide with each other. The first terms on the right-hand side of Eqs. (10) and (11) correspond to the thermal conduction with the adjacent layer (the first term on the right-hand in (4)). The second terms correspond to the heat exchange to blood perfusion and heat generation from exercise (second and third terms on the right-hand in (4)), which exponentially increase with increasing temperature. The parameters α (°C^{-1}) and β are determined by comparing the computed results with the anatomical model in the following section. Because the experimental values obtained from the sensor are used in determining these parameters, these parameters implicitly consider the effect of noises from the sensor.

The estimation code can be implemented to a device with ≥50 MB of computational memory. The execution time required to estimate the core temperature using the pre-stored heart rate for one hour was 0.56 s.

V. RESULTS

A. DERIVATION OF PARAMETERS IN THE COMPARTMENT MODEL

The parameters of α and β used in (10) and Eq. (11) for the compartment model are derived by comparing the computed

$$T_{TC,m} = T_{TC,m-1} + \frac{\left\{hx_{SC}\cdot\left(T_{TS,m-1}-T_{TC,m-1}\right) + hx_{CC}\cdot\left(T_{LC,m-1}-T_{TC,m-1}\right) + M_{TC} + Ex(t)\right\}}{W_{TC}C_{TC}}\Delta t \tag{6}$$

$$T_{TS,m} = T_{TS,m-1} + \frac{\left\{hx_{SC}\cdot\left(T_{TC,m-1}-T_{TS,m-1}\right) + M_{TS} - H\cdot\left(T_{TS,m-1}-T_e\right)S_{TS}/f_{pcl} - EV(t)\right\}}{W_{TS}C_{TS}}\Delta t \tag{7}$$

$$T_{LC,m} = T_{LC,m-1} + \frac{\left\{hx_{SC}\cdot\left(T_{LS,m-1}-T_{LC,m-1}\right) + hx_{CC}\cdot\left(T_{TC,m-1}-T_{LC,m-1}\right) + M_{LC} + Ex(t)\right\}}{W_{LC}C_{LC}}\Delta t \tag{8}$$

$$T_{LS,m} = T_{LS,m-1} + \frac{\left\{hx_{SC}\cdot\left(T_{LC,m-1}-T_{LS,m-1}\right) + M_{TS} - H\cdot\left(T_{LS,m-1}-T_e\right)S_{LS}/f_{pcl} - EV(t)\right\}}{W_{LS}C_{LS}}\Delta t \tag{9}$$

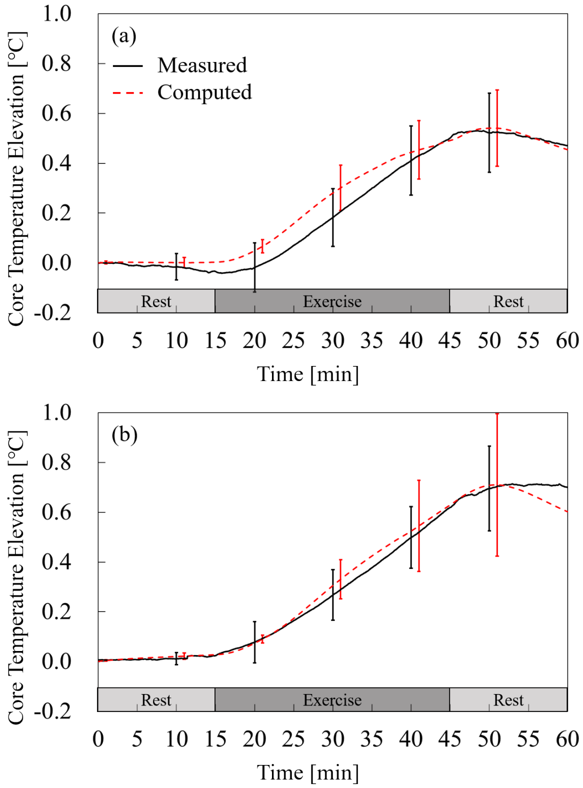

FIGURE 6. Computed and measured core temperature elevations for (a) naked and (b) clothed trials. Error bars represent standard deviation.

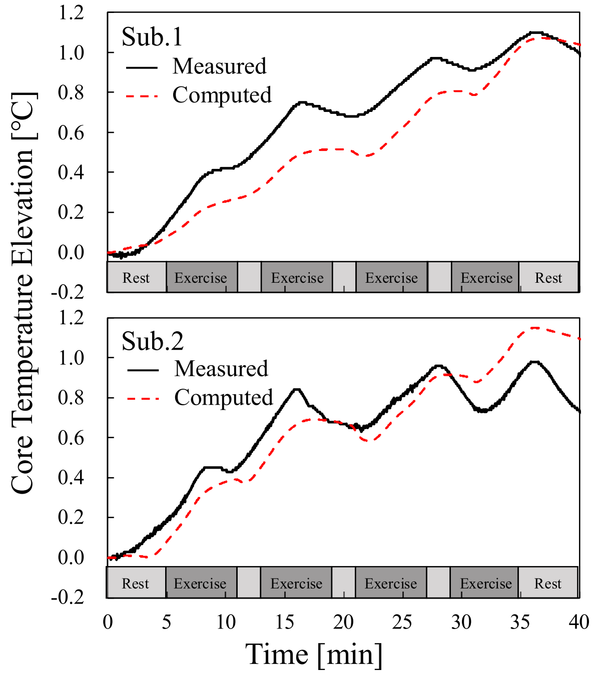

FIGURE 7. Computed and measured core temperature elevations during exercise in field experiments in two selected subjects. The subjects wore a short-sleeved T-shirt, a pair of long pants, socks, and shoes.

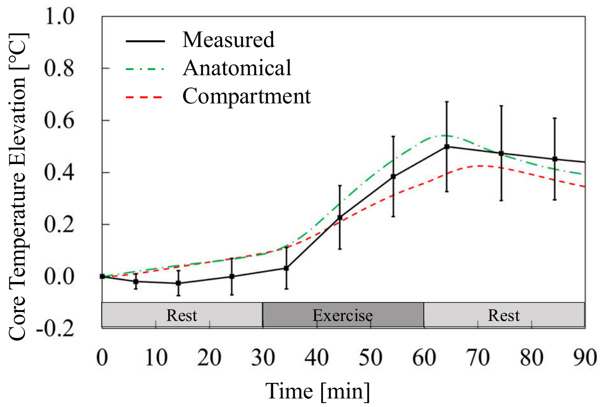

FIGURE 8. Computed and measured core temperature elevations during exercise when naked. The measured data were obtained from [46]. Error bars represent the standard deviation of the measured data.

results with the anatomical human model. In this derivation, the computation using the anatomical model is used as a reference. The range of temperature was chosen as 32–38 °C, and the relative humidity was set to 50%. The parameters in (10) and (11) were derived such that the time course of the temperature was fitted with the least mean square method. The resultant parameters of α and β were 7.2 and 4.6, respectively. As shown in Fig. 5, a good agreement is observed between the results of the compartment and anatomical models in the core temperature elevation during rest. The difference in the core temperature, especially in the first 30 min is attributable to the difference in the modeling. The core temperature defined in Eq. (6) was used in the compartment model, whereas the arterial blood temperature was used in the anatomical model, which may have had a retarded effect in the order of several minutes.

B. VALIDATION WITH MEASUREMENT

To confirm the effectiveness of the model, the core temperature elevation computed using the compartment model was compared using the measurement data in the experiment (Section II). The core temperature elevation measured and computed by the compartment model is shown in Fig. 6.

As shown in Fig. 6, good agreement is observed between the computed and measured core temperature elevations during exercise under heat load. The core temperature elevation in the computation and measurement data for the naked trial were 0.54 °C and 0.53 °C, respectively. For the clothed trial, the corresponding figures were 0.70 °C and 0.71 °C. The root mean square error (RMSE) for the naked and clothed trials were 0.16 °C and 0.14 °C, respectively.

Additionally, we applied the proposed compartment model to the field experiment. Fig. 7 shows the time series of the core temperature in two subjects as typical results. The RMSE in all the subjects was <0.18 °C.

C. VERIFICATION WITH PREVIOUS STUDY

To verify the compartment model, the exposure scenario, whose experimental results have been published, and the field testing were considered.

The exposure scenario in [53] was as follows: 1) The subject rested for 90 min in a thermoneutral room with an ambient temperature of 25 °C and relative humidity of 50%; 2) the subject moved and stayed for 90 min in another room with ambient temperature and relative humidity of 33 °C and 50%, respectively. There, the subject initially sat on a seat for 30 min, then walked at 4.5 km/h for 30 min on a treadmill, and rested on the seat again for 30 min.

As shown in Fig. 8, a good agreement is observed between the computed and measured core temperature elevations during exercise under the heat load. The core temperature rise in the anatomical and compartment models at 65 min were 0.54 °C and 0.43 °C, respectively, which is in good agreement with the measurement of 0.50 °C.

VI. DISCUSSION AND SUMMARY

We have developed a two-layer two-compartment model, which characterized the core temperature rise. The coefficients for expressing the heat transfer between the two compartments, corresponding to the head and trunk, and the limbs were derived by comparing the computed temperature with that of the anatomical model. The coefficient was characterized in terms of the body core and skin temperature rises. The feature of this estimation method is attributable to its non-intrusive and robust manner. Specifically, the estimation based on the bioheat modeling follows the physics law. The thermoregulation works to cool down temperature rise, resulting in the robustness of computation. In our tests for 834 person-days [54], its robustness has been demonstrated. Under the environment where the heat of vaporization is ineffective, such as under extremely high humidity, high core temperature rises might be expected. In such a case, it may result in excessive core temperature rise, which may reach the level where the alert is needed. In our system, the heart rate is then chosen as an additional factor [54] for the health condition to be monitored.

To validate our proposal, we conducted two trials, in which eight subjects were naked and clothed. Our computational results with the proposed model were in good agreement with the measured results. Comparing the naked and clothed cases, the agreement of the latter was better than the former. This may be caused by the modeling condition before exercise. In the case of the naked subjects, the core temperature was decreased by 0.04 °C, owing to the reduced metabolic rate. Furthermore, our computational model was found to be useful for field experiments, in which vital data measured with the wearable device was used (Fig. 7), and replication of a previous study (Fig. 8).

The two-layered one-compartment model based on signal-processing was provided in [55]. The averaged absolute error in that study was 0.1 to 0.2 °C. Because of different algorisms and ambient conditions and activities, straightforward comparison with the existing study is not feasible. To overcome this difficulty, a two-layered one-compartment model was applied in the same condition as Fig. 6 for comparison. The RMSE was 0.18 °C for the naked trial and 0.30 °C for the clothed trial, respectively, higher than those of the two-layered two-compartment model. One uncertainty factor to be noted is the modeling of the clothing. In this study, for simplicity, a fixed constant was given to characterize the heat transfer between the inner and ambient temperature. Nonetheless, a good agreement was obtained for appropriate choices. In addition, limitation of the compartment model, which is attributable to the simplification of the body shape and tissue composition, can be found in Fig. 5 where the temperature difference in the early stage of the exercise is found as compared to anatomical model.

The uncertainty of our estimation is mostly caused by the sweating, which governs the cooling using heat of vaporization. Variability of sweating was reported in the order of 30%, which is much larger than those of the remaining factors [56]. Note that the estimation of core temperature and amount of sweating at a population level has also been shown to be related to the number of patients with heat-related illness transported by the ambulance [57], [58] as well as the athlete performance [59]. For such biological applications, the uncertainly of variability in individual is more significant. Consideration of the difference in individual sweating due to adjustment of parameters by data assimilation might be possible to estimate core temperature with more accuracy, though this is out of scope in this study where one of the targets is on the occupational safety management. The measurement uncertainty is mainly related to the heart rate. When the sensor was not well attached to the body, accurate measurement of heart rate may become difficult. However, the pressure of the sensing garment is designed to maintain sufficient contact between the skin and the electrodes [45], so that the heart rate data was rather stable and may not be needed to consider during our measurement. When applying our wearable system in real world, additional uncertainty might be added.

Additional factor to be noted is the application of machine learning to core temperature rise estimation. However, individual variability is mainly attributable to the sweating, in addition to the individual morphological difference and heart rates. Potential method would be recurrent neural networks or long short-term memory (LSTM) structures [60]. However, as the individual differences are non-negligible, in addition to the difficulty to take potential data for individual variability is not straightforward. Our approach in Fig. 5 is considered to conduct measurement in computational domain. Thus, it would not be realistic to apply such approach.

The number of subjects considered here was limited to eight, with ages 23 ± 1 years. The aim of this study was to develop a method for core temperature estimation toward preventing exertional heat-related illness, which primarily occurs younger or active person. In previous studies [61], [62], minor revision of the sweating model provided good estimation in elderly for the resting condition. Thus, this method may be extended to non-exertional heat-related illness, which primarily occurs in the elderly and those with chronic illness [63]. In addition, further field testing would be needed for different environmental conditions that vary with time. Future study also includes the development of a system that remotely monitors the physical condition of each people wearing the sensor and sends the alert via the cloud, which will enable them to proactively keep out of heat to rest when uncomfortable.

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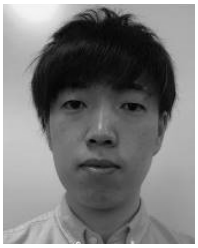

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