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A NEW TRANSIENT BIO-HEAT MODEL OF THE HUMAN BODY

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ABSTRACT

A new mathematical multi-segmented model based on an improved Stolwijk model is developed for predicting nude human thermal and regulatory responses within body segments and the environment. The passive model segments the body into the 15 cylindrical parts. Each body part is divided into four nodes of core, skin, artery blood, and vein blood. The body nodes interact with each other through convection, perfusion and conduction. In any body element, the blood exiting the arteries and flowing into the capillaries is divided into blood flowing in the core (exchanges heat by perfusion in the core) and blood flowing into the skin layer (exchanges heat by perfusion in the skin). The model calculates the blood circulation flow rates based on exact physiological data of Avolio [1], real dimensions, and anatomic positions of the arteries in the body. The circulatory system model takes into consideration the pulsatile blood flow in the macro arteries with its effect on the convective heat transport. The inclusion of calculated blood perfusion in both the tissue and the skin, based on the arterial system model and the heart rate is unique for the current model. The bio-heat human model is capable of predicting accurately nude human transient physiological responses such as the body's skin, tympanic, and core temperatures, sweat rates, and the dry and latent heat losses from each body segment.

The nude body model predictions are compared with published theoretical and experimental data at a variety of ambient conditions and activity. The current model agrees well with experimental data during transient hot exposures. The nude human model has an accuracy of less than 8% for the whole-body heat gains or losses and ±0.48°C for skin temperature values.

INTRODUCTION

Mathematical modeling of the human body thermal response is a valuable tool to understand the human body thermal behavior under different environmental conditions and activity levels. Bioheat models are needed in a variety of applications and have been successfully used [2] in thermal comfort research aimed for energy efficiency, and human thermal discomfort under exposure to extreme hot or cold conditions as in fire fighting or deep sea diving [3]. The continuing rise in energy demand resulted in the need for designing energy efficient HVAC systems for thermal comfort in transient and steady state conditions. The prediction of the human thermal response in transient or steady non-uniform radiant environment can benefit from the development of a realistic bioheat model [4-6]. In such applications, it has been necessary to build thermal models of the human body that are able to predict core and partitional body temperatures during time transients in any environment. The single-segment and multi-segmental models for the human body and its thermoregulatory responses have been developed based on the theories of physiology, thermodynamics and transport processes for the prediction of thermal behavior of either the entire human body or a part of it [4, 7-11].

Gagge developed a simple and an easy-to-implement one-segment two-layer (core and skin) lumped model of the human body [7, 12]. Gagge derived energy balances for the core and the skin. The control system equations were written as a function of temperature signals for shivering, sudomotor response (sweat generation), vasomotor response (control of skin blood flow rate) and for the fraction of total mass assigned to the skin layer. Gagge model is applicable for moderate activity levels and uniform environmental conditions [13], and is used to predict thermal comfort [14].

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The first multi-segmented mathematical model of the entire human body was developed by Stolwijk [9, 14]. His model uses five cylindrical segments to describe the human body: trunk, extremities, hands, feet, and a spherical head. Each segment is divided into four concentric lumped layers, representing core, muscle, fat and skin tissue layers. The segments are connected through blood flow in the arteries and veins offering an improved representation of the circulatory system and the distribution of heat within the body. Stolwijk's model of thermal control is described as functions of two tissue temperature signals as in Gagge's control system signals and the other is related to the rate of change of tissue temperatures [8, 9]. The Stolwijk model does not recognize the flow in individual arteries and veins but uses a unique central blood node or pool at a uniform temperature. The constant temperature blood node exchanges heat with tissue layers by convection only, while tissue layers exchange heat in the segment by conduction only. The Stolwijk model is still limited to constant environment conditions.

All subsequent contemporary multi-segment multi-mode bioheat models are modifications and improvements on Stolwijk model parameters and thermal control equations such as the Berkeley Comfort Model of Huizenga et al [4], Tanabe and Kobayashi model [6], and Fiala et al [15]. Several improvements were introduced in the Berkeley Comfort Model that include an increase in number of body segments, improved blood flow model for counter flow heat exchange in limbs, addition of a clothing node, and improved convection and radiation coefficients [4]. The Berkley Model improved Stolwijk constant blood node temperature by using a thermoregulatory system that maintains the core temperature within a narrow range by either inhibiting or enhancing heat production and heat loss. The blood perfusion flow rates in the muscle, skin, fat, and core tissues remain as input parameters to the Berkeley Comfort Model.

The blood flow is a very important or decisive part of the human thermal functions. About 50%-80% of the heat flow in the tissue is carried in or out of the tissue by the blood flow [16]. The Berkeley Comfort Model [4] improved some of Stolwijk model deficiencies but it did not use actual blood circulation network based on cardiac pulsating input that can predict exact core and skin blood flows. A detailed transient human body model known as the KSU Model was first developed by Smith [13] using a three-dimensional finite element method and later improved by Fu [17] to include a clothing layer. They decompose the human body into fifteen cylindrical parts. KSU model allows having non-uniform and non-symmetric environmental conditions for each body element. The circulatory system is modeled with branching that initiates from the heart. The location of arteries and veins are the boundaries of each 3D finite element. The control equations for shivering, sudomotor and vasomotor are derived by Smith based on real experimental data. KSU model has a heavy computational time [13, 17] and the blood system model parameters have been adjusted to fit experimental data using the blood perfusion flow in the capillary bed as input. Both Berkeley [4] and KSU [13] models ignore the effect of the pulsatile blood flow in the large arteries on heat transfer, and the perfusion blood flow rate is used as an input to the model.

The multi-segmented passive model proposed in this work will simulate the heat transfer within the body parts and the environment while combining the numerical simplicity of Stolwijk model [8-9] and using the most accurate and realistic representation of the arterial system including blood flow pulsation. The proposed model based on exact anatomical data of the human body will calculate the blood flow rates and their variations based on exact physiological data [1] to improve the heat exchange model. The circulatory system model will take into consideration the pulsatile blood flow in the arteries and the dependence of the arterial system on the heart rate and calculate the perfusion flow rates in the core and skin nodes. The model will be validated by comparisons with published experimental and simulation results on core and skin temperatures, and body latent heat loss when exposed to transient environment. This work contributes to research efforts aimed at providing reliable predictions of human thermal responses over a wide range of environmental conditions.

NOMENCLATURE

A	: area (m^2).
C	: thermal capacitance (J/K).
c	: specific heat (J/Kg·K).
h	: pulsating blood flow convective heat transfer coefficient ($W/m^2\cdot K$).
h_{mean}	:mean non-pulsating blood flow convection coefficient ($W/m^2\cdot K$)
K	: conductance (W/K).
M	: segmental metabolic rate (W).
$\dot{m}$	: blood mass flow rate. (kg/s).
P	: vapor pressure (Pa).
t	: time (s).
T	: temperature (^oC).

Subscripts

bl	: blood
bl,a	: artery blood.
bl,v	: vein blood.
cr	: core.
cr-sk	: between core and skin.
perfusion ,total	: total perfusion rate.
sk	: skin.

MATHEMATICAL FORMULATION

The human body model divides the body into the fifteen-body segment as shown in Fig. 1, where the trunk is divided into two segments, while the head and neck are combined in one segment. The other segments are the upper arms, thighs, forearms, calves, hands, and feet. Each segment is represented by a cylinder of a uniform temperature. The heat transfer model is based on four nodes: core, skin, artery and vein that exchange heat through convection, conduction and perfusion as illustrated in Fig. 2. The blood vessels included in each segment are determined from geometrical, physiological and anatomical considerations [1]. The human circulatory system transports blood to all body tissues through an intrinsic network of blood vessels. The proposed model uses the arteries system model of Avolio to represent the multi-branched human arterial system network of pulsatile flow that initiates from the heart and branches throughout the body [1]. The cardiac output and the heartbeat are correlated to the total metabolic rate [18].

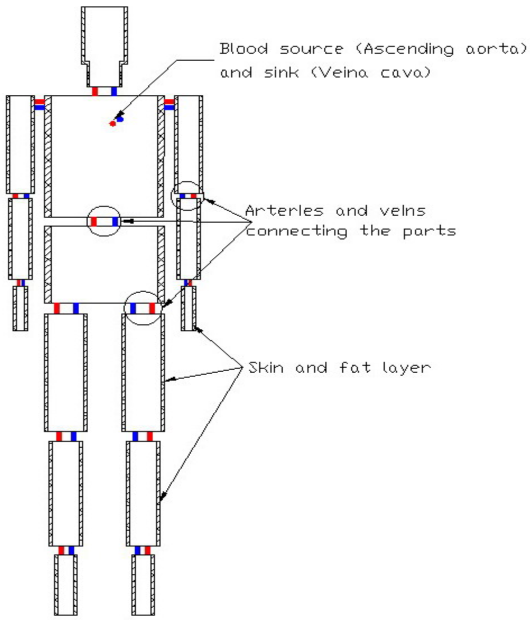


Fig. 1: Schematic of the Multi-segment Model

In any body segment, the blood exiting the arteries and flowing into the capillaries is divided into blood flowing in the core (exchanges heat by perfusion in the core) and blood flowing into the skin layer (exchanges heat by perfusion in the skin) after crossing the core tissue as shown in Fig. 3 [19-21]. The blood vessels (arteries and veins) contained in each segment of the body are identified [1, 21-23]. Therefore, the blood entering a segment (from an artery) will split into a perfusion flow (including the skin blood flow) in the considered segment and a blood flow entering the adjacent segment. The opposite takes place in the veins for the same segment (see Fig. 3). For peripheral segments (head, hand and foot), the blood entering rate is equal to the blood perfusion rate and the blood exiting rate. The perfusion blood flow rates are accurately calculated based on the cardiac output [23]. The thermal signals affect the blood circulation in two ways. First, it affects the quantity of blood flowing in the muscles tissue by increasing or decreasing the amplitude of the pulsating cardiac output referred to as the increase in the cardiac volume of blood ejected in one heart cycle [21, 23]. Secondly, the thermoregulation affects the skin blood flow by the mechanisms of vasodilation and vasoconstriction. Thermoregulatory control equations are based on KSU Model [13, 17].

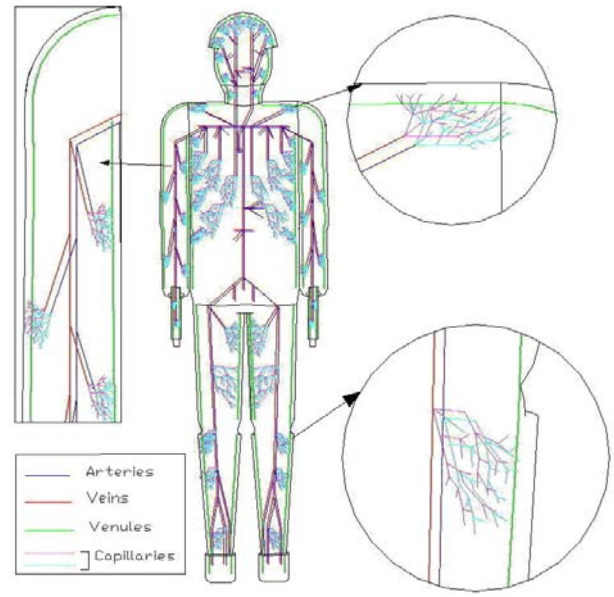


Fig. 2 The complete blood vessels model that includes arteries, veins, capillaries and venules

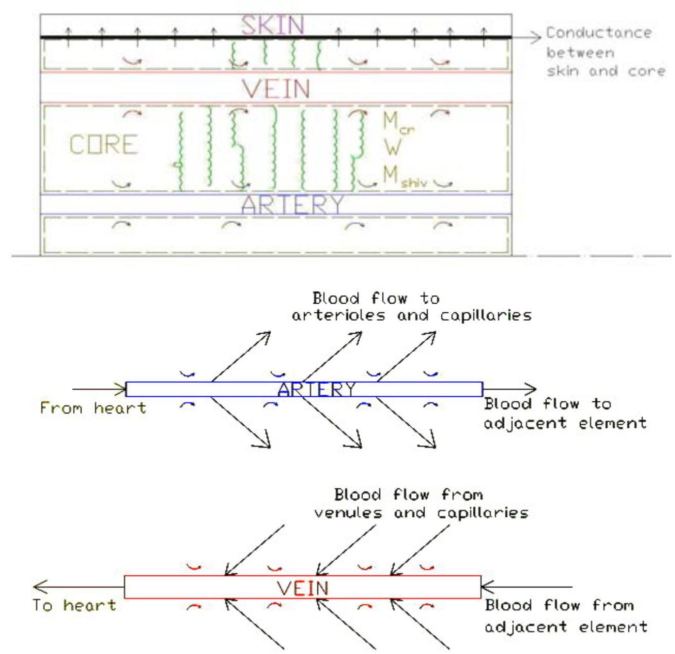


Fig. 3: Heat exchange in core, skin, artery blood, and vein blood nodes.

The heat generated by the human body is enhanced or dissipated to the environment through radiation, convection, and evaporation to maintain the core temperature within a narrow range. When the core temperature rises/drops above the reference temperature, the blood vessels under the skin are dilated/constricted in order to reject/store more heat to/from the environment. If the vasomotor thermoregulatory function is not sufficient to reject the heat from the body, the sudomotor function is triggered where heat is lost by sweating [13]. On the other hand, if vasomotor is not able to keep the heat in the body, metabolic thermoregulatory function takes place; the body will start shivering in order to heat the body.

In the following section, a description will be give of the human circulatory system, and the unsteady bioheat equations for the core, skin, artery and vein nodes of each segment.

Human Circulatory System

Arteries are divided into 128 segments based on Avolio's model and accounting for all the central vessels and major peripheral arteries supplying the extremities including vessels of the order of 2.0 mm diameter [1]. The velocity in each artery is computed using the momentum equation in cylindrical coordinates for unsteady unidirectional flow in a pipe [21, 24]. The principal variables of an arterial segment as derived by Womersely are the characteristic impedance and the propagation constant [25]. The equivalent impedance at the source (the heart) is the summation of all branches in series and parallel depending on their geometry. The expressions of the complex characteristic impedance and propagation constant for one artery can be found in the work of Avolio as a function of the blood density, the angular pulsating frequency, and the Poisson's ratio of the arterial wall [1, 21]. The impedance (the ratio of the pressure to the flow) for each artery is calculated as function of the physical geometry and flow parameters. The input impedance to the heart and the pressure ratios at the level of each artery are computed using Avolio's Algorithm which takes into consideration the elastic behavior of the artery wall and the wave reflections that take place whenever the impedance changes from an artery to another. The contribution of the veins system is taken into consideration as boundary conditions on the peripheral arteries. Knowing the cardiac ejection waveform at the heart, the flow in each artery segment can be computed [21].

The blood vessels are assumed such that each artery-vein pair is represented by two parallel cylinders, as reported by Weinbaum and Jiji [22-23] with branches to peripheral skin layer to form the venules. The network of arteries has a parallel network of veins with the same length as shown in Fig. 4. The flow in veins is assumed non-pulsatile, and is equal to the mean value of the corresponding artery flow [21, 24]. Veins are bigger than arteries where the vein diameter is assumed equal to twice its corresponding artery diameter [13]. The inlet and outlet arteries/veins are assigned to each body segment. The computed artery and vein blood flows are used for evaluating the heat transfer coefficients in the arteries and veins. The convective heat transfer coefficient of pulsating flow in main arteries is used in the model [26, 27]. Pulsating flow in main arteries induces 6-10% increase of the average unsteady heat transfer coefficient from steady state value while for arterioles the difference between unsteady and steady heat transfer is insignificant and is independent of the pulsation frequency [26]. For a large amplitude pulsation in laminar pipe flow, the heat transfer due to pulsation is always augmented [27].

The perfusion blood flow rate through the capillaries is calculated by subtracting the exiting flow from the entering blood flow of the considered segment. Larger arteries, veins, and arterioles are assumed to exchange heat by convection with the core of the body. The venules located in the tissue (skin and core) exchange heat between the core and the skin. These small vessels play a major role in thermoregulation where the skin blood flow varies as a function of the core and skin temperatures [13]. The circulatory system model is used to predict the blood flow in the arteries and veins while the skin blood flow is calculated based on empirical and experimental correlations reported by Smith [13]. The information is then used to update the energy balances of each body segment.

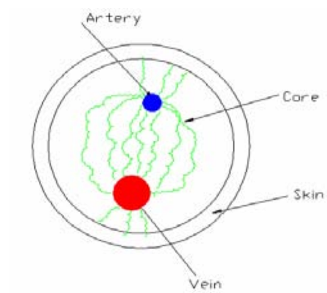


Fig. 4: Radial cross sections of a body segment.

Energy Balance Equations

For each segment, the heat balance equation is written for the four nodes (core, skin, artery blood, vein blood) [9]. The core node energy balance is given by

$$C_{cr}\cdot\frac{dT_{cr}}{dt} = M_{cr} + M_{shiv} - W - \alpha\cdot Q_{res} - Q_{cr-sk}$$
$$\underbrace{-\sum_{\substack{arteries\\arterioles}} h(t)\cdot A_{artery}\cdot(T_{cr}-T_{bl,a})}_{\text{Arteries heat exchange by convection}}$$
$$\underbrace{-\sum_{veins} h_{mean}\cdot A_{vein}\cdot(T_{cr}-T_{bl,v})}_{\text{Veins heat exchange by convection}} \underbrace{+\dot{m}_{perfusion,total}\cdot c_{bl}\cdot\left(T_{bl,a}-T_{cr}\right)}_{\text{Core perfusion heat exchange}}$$
$$\underbrace{+\dot{m}_{skin}\cdot c_{bl}\cdot\left(T_{sk}-T_{cr}\right)}_{\text{Skin perfusion heat exchange}} \tag{1}$$

Where **α** is equal to unity for the chest and zero for all other elements, *W* is the mechanical work generated by the body, Q_{res} is the heat dissipated by respiration calculated based on the known ASHRAE correlation [18], h(t) is the pulsating heat convection coefficient of the blood flow, h_{mean} is the mean non-pulsating convection coefficient of blood flow in veins,

$\dot{m}_{perfusion,\,total}$ is the total perfusion rate of blood entering the core; $\dot{m}_{skin}$ is the skin perfusion blood flow, M_{cr} is the basal metabolic heat generated in each segment [13], and M_{shiv} is the segmental thermoregulatory metabolic rate generated by shivering. The area of any blood vessel is $A_{vessel} = 2.\pi.r.l$ where l is the element length whether the vessel is an artery or a vein. The term on the left-hand side of Eq. (1) is the rate of accumulation of thermal energy per unit volume due to the changing temperature of core tissue. The last four terms on the right-hand side of Eq. (1) represent: the heat exchange between the core node and the arteries by convection, the heat exchange with veins by convection; the net heat flow associated with perfusion blood flow through capillaries entering the core at the arteries blood temperature and leaving at the core temperature [10, 19, 28]; and the net heat exchange associated with perfusion blood flow between the skin and the core node, $\dot{m}_{skin}$. The heat exchanged between the skin and the core through the contact thermal resistance is Q_{cr-sk} and is given by

$$Q_{cr-sk} = K \cdot (T_{cr} - T_{sk}) \tag{2}$$

where K is the skin to core conductance calculated from the two conductance's of the skin-fat and the muscle. The values of K are based on the correlations of Havenith [29] and the physiological data of Gordon on skin and fat layers thickness [30] given by

$$\frac{1}{K} = \frac{1}{K_{muscle}} + \frac{1}{K_{fat-skin}} \tag{3a}$$

where

$$K_{muscle} = \frac{A_{skin}}{0.05} \quad \text{(W/K)} \tag{3b}$$

and

$$K_{fat-skin} = \frac{A_{skin}}{\left(th_{fat+skin} - 2\right) \cdot 0.0048 + 0.0044} \tag{3c}$$

where $th_{fat+skin}$ is the thickness of fat and skin layers as reported by Gordon [30]. The muscle conductance K_{muscle} is based on the maximal muscle insulation. The overall conductance K in the current model expression does not show dependence on the metabolic rate, but Q_{cr-sk} depends on the metabolic rate through the core and the skin temperature difference. The core and the skin temperatures depend strongly on the change of the perfusion blood flow rate through the muscle as calculated from the circulatory system model based on the cardiac output and the heart rate. The model formulation assumes complete thermal equilibrium between blood and tissue [17, 19] for all body parts meaning that the blood exiting the core node and entering the vein node is at core temperature. The thermal equilibrium assumption proves its validity because the blood flow through the core tissue is very slow [2]. Another argument that supports this assumption is that the core and blood system can be assimilated to heat exchanger with a very high heat transfer area, area/volume ≈ 5000 as calculated from Milnor's data [21]. This large heat transfer area makes the outlet blood temperature equal to the core temperature.

The skin node energy balance is given by

$$C_{sk} \cdot \frac{dT_{sk}}{dt} = M_{sk} + Q_{cr-sk} - A_{sk} \cdot \left[h_c \cdot (T_{sk} - T_{amb}) + h_r \cdot (T_{sk} - T_o) + h_e \cdot (P_{sk} - P_\infty) \right] + \underbrace{\dot{m}_{skin} \cdot c_p \cdot \left(T_{cr} - T_{sk} \right)}_{heat\ transfer\ through\ the\ skin\ blood\ flow} \tag{4}$$

where h_c is the external convection heat transfer coefficient between the skin and the atmosphere, h_r is the radiation transfer coefficient, h_e is the evaporation coefficient deducted from h_c by Lewis formula [18], T_o is the surroundings radiant temperature, P_∞ is the ambient vapor pressure, and P_{sk} is the skin vapor pressure.

Arteries and vein blood nodes energy balance

The arteries node energy balance is given by

$$C_{bl,a} \cdot \frac{dT_{bl,a}}{dt} = -\sum_{\substack{arteries \\ arterioles}} h(t) \cdot A_{artery} \cdot (T_{bl,a} - T_{cr}) + \dot{m}_a \cdot c_{bl} \cdot \left(T_{bl,a,\,adjacent} - T_{bl,a} \right) \tag{5}$$

where $\dot{m}_a$ is the mass flow rate of blood entering the considered part through the inlet artery, and $\dot{m}_a$ is assumed to change with time within one period of pulsation like the cardiac ejection waveform. Note that for the chest, there is no adjacent element, but the blood entering the chest from the heart is the one of the veins that enters the chest through the vena cava.

The vein node energy balance is given by

$$C_{bl,v} \cdot \frac{dT_{bl,v}}{dt} = -\sum_{veins} h_{mean} \cdot A_{vein} \cdot (T_{bl,v} - T_{cr}) + \dot{m}_v \cdot c_{bl} \cdot \left(T_{v,\,adjacent} - T_v \right) + \dot{m}_{perfusion,\,total} \cdot c_{bl} \cdot \left(T_{cr} - T_v \right) \tag{6}$$

where $\dot{m}_v$ is the mass flow rate of blood entering the considered part through the inlet vein and $\dot{m}_{perfusion,\,total}$ is the perfusion blood flow rate in the considered part where it enters the venous bed at core temperature and exits at vein temperature. The total perfusion blood flow rate is divided into perfusion blood flow in the core and perfusion blood flow to the skin controlled by thermoregulatory functions described by Smith [13].

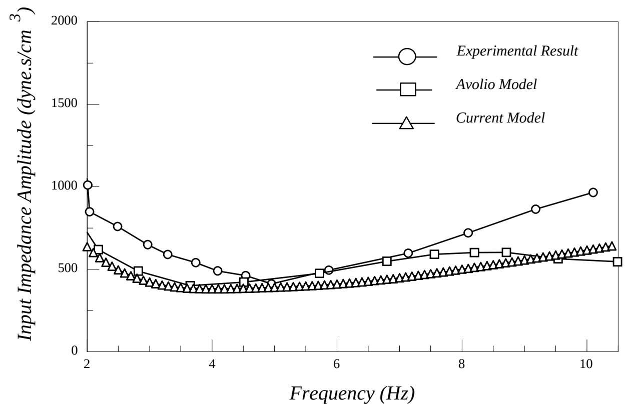
The heart rate and velocities in the arteries are computed based on the cardiac output and its correlation to the thermoregulatory response and to the volume of oxygen consumed. This cardiac output is equal to the mean value of the cardiac ejection waveform used as input to Avolio model [1] to calculate the velocity in each artery. The basal cardiac output corresponds to a heart rate of 75 beats per minute. Moreover, the volume of oxygen consumed is proportional to the metabolic rate and AHRAE correlation and data are used in the present model [18]. In previous models, the skin conductance is composed of two terms: one related to the direct contact between skin and core and the other term is related to the skin blood flow [7, 12, 19, 28]. In the current model, the skin conductance is modeled in the heat balance equations of the core and skin that take into consideration the perfusion blood flow rate.

NUMERICAL METHOD

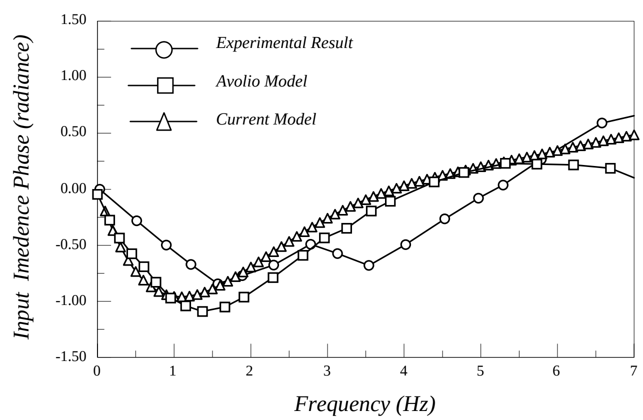
A simulation program was developed. The input to the program consists of the initial thermal state of the human body, the metabolic rate, the ambient conditions, and the physiological and physical parameters inherent in the bioheat model of the human body. A fully explicit Euler-Forward integration scheme with a time step of 0.02 seconds over the desired simulation period is used to solve the energy balance equations of the human body nodes for any segment where the respiratory heat loss, blood flow rates, thermoregulatory responses and skin vapor pressure are calculated from previous time step. The regional blood flow rates in the arteries veins and tissue are obtained from the output of Avolio model which consists of the velocity ratios for any considered artery to the cardiac output. Therefore, the temperatures of the core, skin, artery, and vein nodes, and the artery blood flow rate are updated for each segment at each time step. Avolio model blood flow that gives the arterial blood velocity and thereafter the blood perfusion rates is computed after each complete cycle of the heart which approximately takes 0.8s for the neutral state of the body and which can vary depending on the metabolic rate. Suitable initial conditions are determined starting from the neutral conditions of T_{cr}=36.8^{o}C, and T_{sk}=33.7^{o}C to simulate a relatively long exposure to an ambient temperature of 28^{o}C and 31% relative humidity. The obtained steady state values are then used as initial conditions for all other unsteady calculations of various node temperatures for al segments and the sensible and latent heat loss from the skin. The program simulates one hour of exposure in 20 minutes.

RESULTS AND MODEL VALIDATION

The arterial circulation system is validated against published experimental data of Mills et al [31] and simulation model of Avolio [1] by comparing the input impedance frequency response at the ascending aorta to the applied cardiac ejection waveform. Figure 5 shows (a) the input impedance amplitude and (b) the input impedance phase as function of the pulsation frequency for the current model and the published data. The current work simulation results agree well with the experimental data and the model of Avolio for the impedance amplitude for frequencies has better agreement with experimental data on impedance phase at high frequencies. The better results of our model stem from the treatment of the Bessel functions where they are calculated on analytical expressions and not from the use of empirical formulae for the tabulated values in Womersley solution [2]. The program is run at smaller frequency steps of 0.05 compared to 0.5 of Avolio model.



(a)



(b)

Fig. 5: A plot of (a) the input impedance amplitude and (b) the input impedance phase as function of the pulsation frequency for the current model and the published data [1].

The arterial system is used to calculate blood flow rate to each body segment and the amount of blood perfusion rates that carried from the core to the skin node. Table 1 presents the basal perfusion blood flow rate calculated using the current model and the published experimental perfusion flow rates of Gordon [30] and Ganong [32]. The circulatory model simulation results show good agreement with published data.

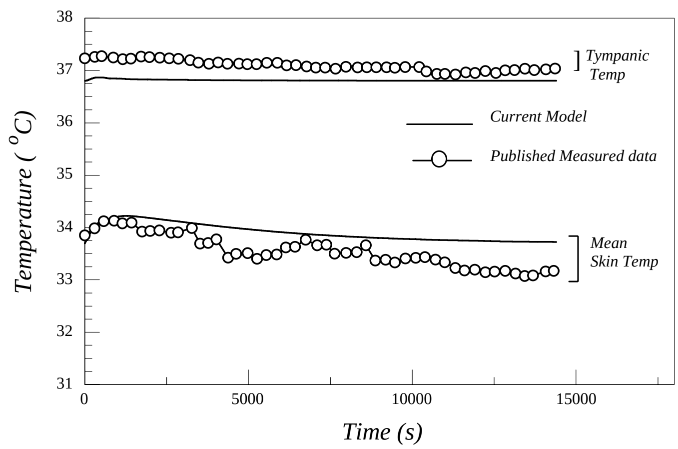
Table 1 Calculated basal perfusion blood flow rate of the current model and the published experimental data of Gordon [30] and Ganong [32].

Segment	Calculated perfusion flow rate from current model (cm^3/hr)	Experimental perfusion flow rate [25, 38] (cm^3/hr)
Head	58,829	56,526
Chest	194,352	198,164
Upper arm	3,577	3,852
Forearm	2,266	2,152
Hand	1,502	1,378
Thigh	6,520	6,196
Calf	2,531	2,741
Foot	1,288	1,339
Total cardiac Output	288,549	290,006

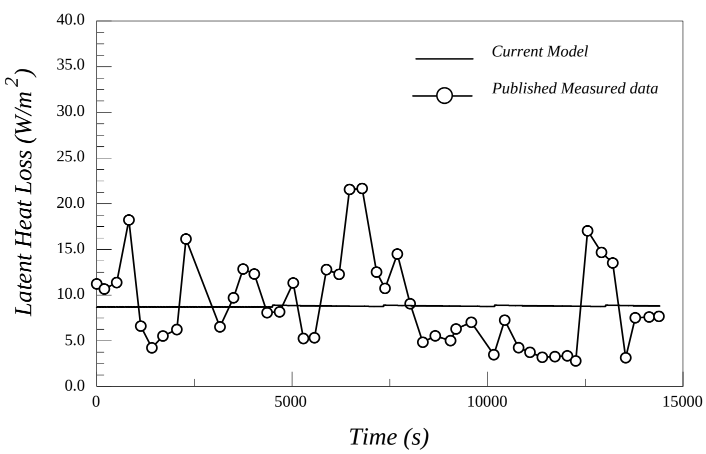
The multi-segmented model is validated by comparing simulated skin, tympanic and rectal temperatures with published experimental data and simulation data of other models for a variety of ambient and activity levels at steady-state and transient. In addition to experimental comparison, simulation results will be compared with results of a computationally detailed model predictions of Smith [13], and to the recent improved predictions of the Berkeley Model. For a sedentary human condition, Hardy and Stolwijk reported experimental data of the mean skin temperature for resting human subjects who wore only shorts [33] for an exposure period of four hours to ambient temperature of T_∞=28.5°C and relative humidity RH_∞=31% at metabolic rate M = 1 met at neutrality. Figure 6 presents the simulated results of current model of (a) tympanic and mean skin temperature, and (b) latent heat loss as a function of time in addition to the reported experimental data of Hardy and Stolwijk [33] at the same conditions. Under steady state conditions, the mean skin and tympanic temperature predictions are very close to the measured values within 0.5 ^oC. The latent heat loss agrees well with experimental measurement.

Figure 7 represents the steady state nodes temperature (for T_∞=28.5°C and RH_∞=31%) variation throughout the human body segments from (a) the head to foot nodes' temperatures, and (b) the head to hand nodes' temperatures. It can be inferred that for the peripheral parts (hand, foot etc.), the core temperature approaches the skin temperature. This is mainly due to the compact size of these parts which makes the muscles nearly attached to the skin. The venal blood temperature is approximately equal to the core temperature through all the parts since the model assumes a complete thermal equilibrium between the tissue and venal blood temperature. The constant arterial blood temperature throughout the segments is consistent with reported data of Chen [34].

The second comparison with measured data and models is for a hot-exposure transient of Hardy and Stolwjik [33]. The reported experimental data were done for a step change in environment temperature from 30°C at 40% RH to 48°C at 30% RH for an exposure period of two hours followed by one hour of environment at 30°C at 40% RH. Figure 8 shows the measured [33] and predicted (a) mean skin temperature and (b) evaporative heat loss in addition to the results of the recent Berkley Comfort Model [4]. During the hot interval the current model predicted well the mean skin temperature and latent heat loss, and produced comparatively similar results to Berkley Model predictions. The core-skin convection interaction is well represented in the current model due to the use of a more realistic circulatory blood flow system for the macro and micro circulation effect on convection heat exchange.

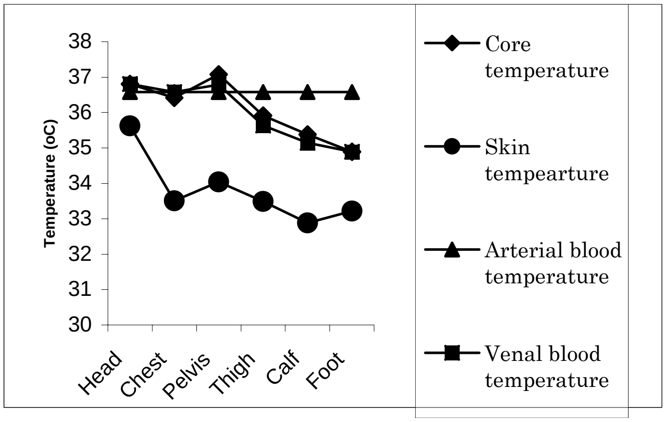


(a)

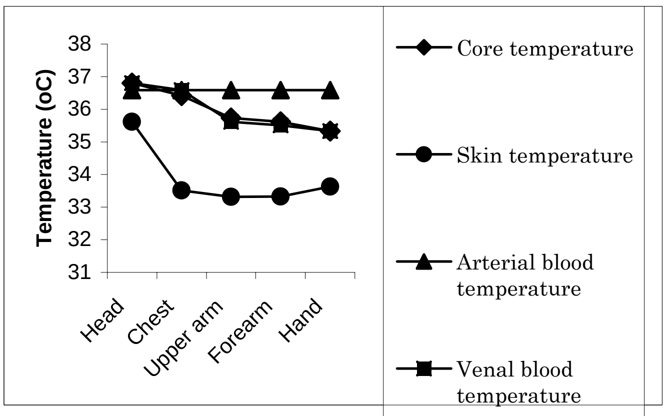


(b)

Fig. 6 The measured [33] and simulated results of (a) tympanic and mean skin temperature, and (b) latent heat loss as a function of time at T_∞=28.5°C and RH_∞=31%.



(a)

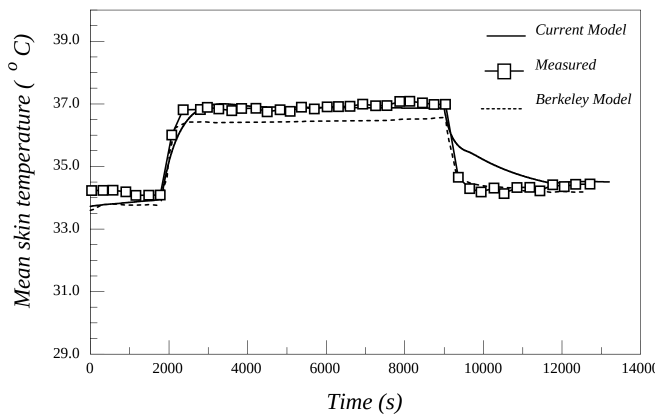


(b)

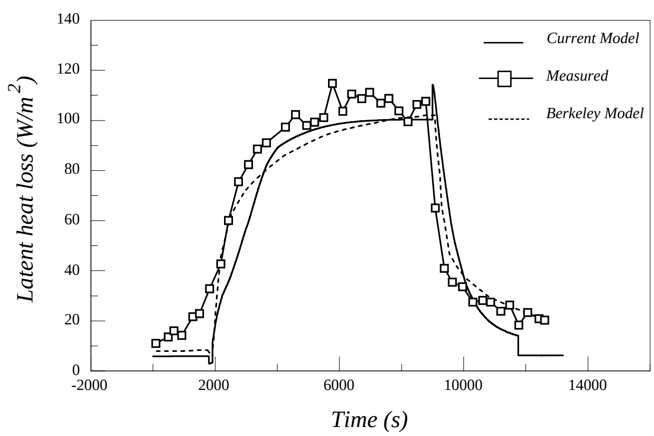
Fig. 7 The steady state nodes temperature (for T_∞=28.5°C and RH_∞=31%) variation throughout the human body segments from (a) the head to foot nodes' temperatures and (b) the head to hand nodes' temperatures.

The third comparison with measured data and models is for a cold-exposure transient of Raven et al. [35] from a cold step change in ambient temperature from 28.5°C, 45% RH to 4.7°C, 70% RH for a period of two and a half hours. Figure 9 shows the measured [35] and simulated temperatures for a step change from 28 to 4.7°C for (a) head skin temperature; (b) upper arm; and (c) thigh. The blood flow model has shown good agreement with experimental data in limbs and extremities skin temperature predictions comparable to the agreement shown in the Berkley model [4].

The last comparison is performed against measured data of Saltin [36] at constant environment temperature 20.5°C, 55% RH with a step change in metabolic rate from M = 4.14 Met continuing for a period of 60 minutes followed by a 20 minute rest period at M = 1 met. Figure 10 shows the measured [36] and predicted values of current model of (a) rectal and mean skin temperature; and (b) tympanic temperature in constant environment of 20.5°C, 55% RH for a step change in metabolic rate. On the same plot Smith Model published simulation data [13] are shown. The rectal and tympanic temperatures agreed well with experimental results and have shown comparable level of agreement with Smith 3-D model results with deviations from experimental data of less than 0.5 °C. However, the predicted mean skin temperature of current model and that of Smith ran two degrees higher than experimental values of Saltin [36]. The agreement with the mean skin temperature at a high sweat rate may not be critical since the reported experimental data have indicated an environmental chamber temperature between 19 and 22 °C and relative humidity between 45-65%, while the simulations are done at the mean fixed ambient condition of 20.5 °C and the mean relative humidity at 55%.



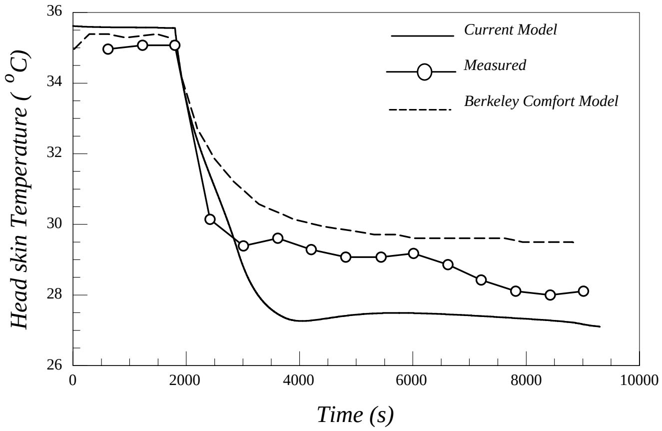
(a)



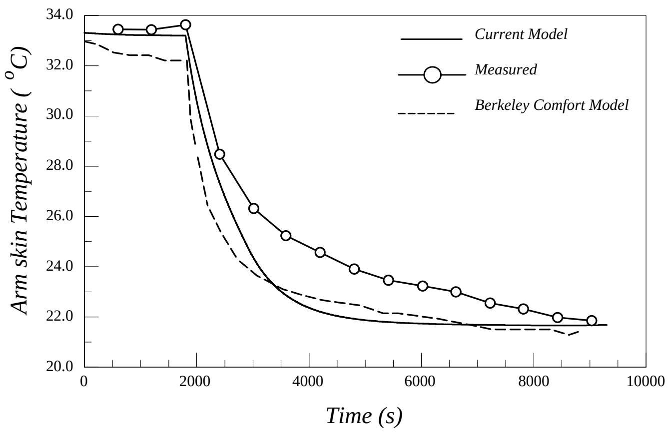
(b)

Fig. 8 The measured [33] and predicted (a) mean skin temperature and (b) evaporative heat loss in addition to the results of the Berkley Comfort Model [4].

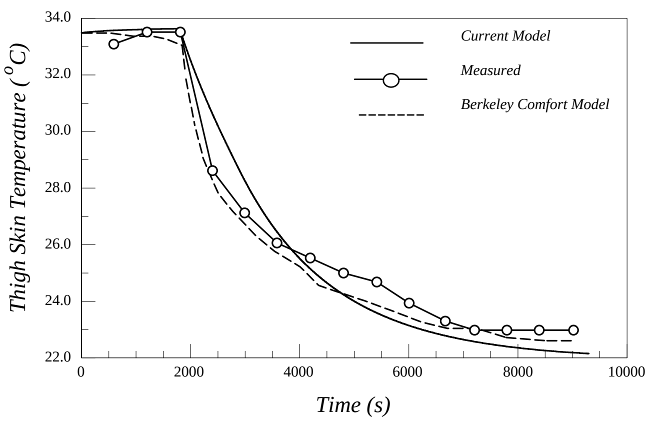
The current simple multi-node multi-segment bioheat model has shown good agreement with experimental data with less input parameters to a program that uses a realistic circulatory blood flow model. In addition to the initial thermal state of the human body, the metabolic rate is the main input parameter to the model that affects both the blood flow cardiac output, and the heat balances of each body segment. The model has produced accurate results that are comparable with recently published models results at a relatively lower computational cost and complexity. The model validations show that it can predict both core and skin temperatures for different body segments with reasonable accuracy under a range of environmental and exercise conditions (different metabolic rates).



(a)

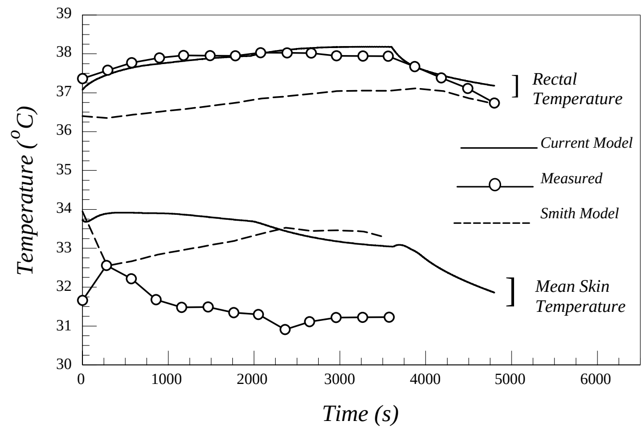


(b)

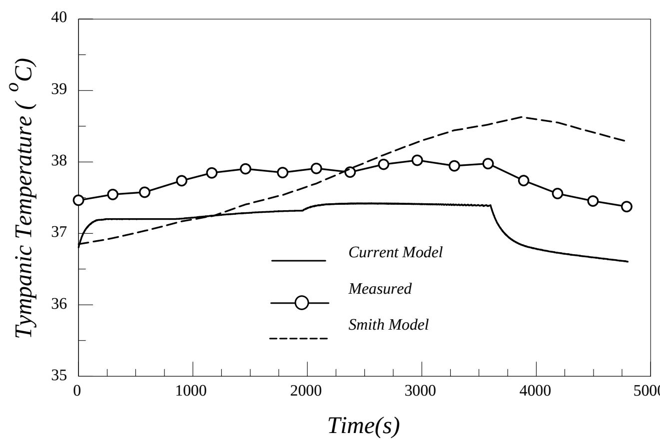


(c)

Fig. 9: The measured [35] and simulated temperatures of current model and Berkley model [4] for a step change from 28 to 4.7°C for (a) head skin temperature; (b) upper arm; and (c) thigh.



(a)



(b)

Fig. 10 The measured [36] and predicted values of current model and Smith Model [13] of (a) rectal and mean skin temperature; and (b) tympanic temperature; and (c) mean skin temperature in constant environment of 20.5°C, 55% RH for a step change in metabolic rate.

CONCLUSION

A bioheat model is developed to predict human thermal responses in steady and transient conditions. The model has accurately simulated the human circulatory system and the perfusion blood flow rate in the core and skin layers that are used in the transient bio-heat equations for the core, skin, and blood nodes in each body segment. The accuracy of the model is verified over a range of conditions for which it is applicable against published experimental data and other human body models simulation results. The model predicts well core, rectal, tympanic, and mean skin temperatures and the latent heat loss rate from the skin surface during heat and cold stress conditions and with reasonable accuracy during exercise and at rest.

Future work will extend the bioheat model to a clothed human which entails adding a cloth node for each segment using appropriate clothing models and to update the model to take into consideration non-uniform environment by incorporating angular variations of thermal response within the skin layer.

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