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Title: Modeling the effect of adverse environmental conditions and clothing on temperature rise in a human body exposed to radio frequency electromagnetic fields
Year: 2014
Journal: IEEE Transactions on Biomedical Engineering
Volume: 62
Issue: 2
Pages: 627-637
URL: http://hdl.handle.net/1959.3/393077

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The definitive version is available at: https://doi.org/10.1109/TBME.2014.2362517
Abstract—This study considers the computationally determined thermal profile of a fully clothed, finely discretised, heterogeneous human body model, subject to the maximum allowable reference level for a 1 GHz radio frequency electromagnetic field for a worker, and also subject to adverse environmental conditions, including high humidity and high ambient temperature. An initial observation is that while electromagnetic fields at the occupational safety limit will contribute an additional thermal load to the tissues and subsequently cause an elevated temperature, the magnitude of this effect is far outweighed by that due to the conditions including the ambient temperature, relative humidity, and the type of clothing worn. It is envisaged that the computational modeling approach outlined in this paper will be suitably modified in future studies to evaluate the thermal response of a body at elevated metabolic rates, and for different body shapes and sizes including children and pregnant women.

Keywords—Computational electromagnetics, computational biophysics, occupational safety, protective clothing, electromagnetic safety standards.

I. INTRODUCTION

Whenever a person is exposed to a radiofrequency (RF) electromagnetic field (EMF), currents and fields will be induced inside the body that will heat tissue due to dielectric losses. The International Commission on Non-Ionizing Radiation (ICNIRP) recommend that RF-EMF induced tissue temperature rises do not exceed 1°C, and adherence to RF-EMF exposure limits in international safety guidelines and standards is generally assumed to ensure this [1], [2]. Such setting of the exposure limits only considers the magnitude of the temperature rise due to RF-EMF. This strategy neglects to directly consider the environmental conditions affecting the person, such as the ambient temperature and humidity, the clothing that the person is wearing, and their level of physical exertion. Such factors are taken into account through the setting of additional safety margins but are not well quantified. This raises the question of whether a given rise, that has no adverse effect in a comfortable environment, may possibly lead to thermal distress or tissue damage in a thermally stressful environment. Thus an alternative strategy for setting RF-EMF exposure limits may be to consider the environmental conditions, either on a temporal or durational basis. This topic is particularly pertinent for RF-EMF workers as the safety guidelines for adverse environmental conditions, either on a temporal or durational basis. This topic is particularly pertinent for RF-EMF workers as the safety guidelines are less stringent for those occupationally exposed to RF-EMF. The rationale for this research is consistent with the thoughts expressed in the review of Foster and Morrissey [3] who wrote that it would be useful to validate thermal models that incorporate the effects of RF-EMF under varying environmental conditions to “... allow RF exposure limits for occupational groups to be considered as part of a framework of recommendations of health agencies and industrial hygiene groups regarding physical labour in warm environments.” Furthermore they suggested it would be useful to understand the limits of tolerance of the human body for RF-induced heat loads, using a more reliable approach than extrapolation from rodent data as has been done in setting present exposure guidelines. This paper is planned to be the first in a series of studies and our current aim is to present the computational framework along with validation. The human body thermal model includes thermoregulation, the addition of a clothing model, as well as investigating the thermal effect of RF-EMF. Future studies will consider the influence of different body shapes and sizes including children, high levels of activity, various RF-EMF exposure scenarios, hot winds, and consideration of people with poor thermoregulation systems.

II. MODELS AND METHODS

A. Geometrical Modeling Review and Generation

Previous work estimating temperature variation within human tissues has involved various levels of complexity of geometric models. One approach is the use of compartmental models, which approximate the human body with a number of cylinders and spheres. These compartments are then decomposed into concentric tissue layers that have been used by Bernardi et al. [11], [12] and Nelson et al. [13], [14] have used a human body model of 2 – 5mm voxel resolution and with 30 or more tissues types developed from the Visible Human Project [15]. One final approach is the use of voxel-based models that make use of images of the human body and classification of the voxels into different tissue types. For example, both Bernardi et al. [11], [12] and Nelson et al. [13], [14] have used a human body model of 2 – 5mm voxel resolution and with 30 or more tissues types developed from the Visible Human Project [15]. One final approach is the use of voxel-based models that make use of images of the human body and classification of the voxels into different tissue types. For example, both Bernardi et al. [11], [12] and Nelson et al. [13], [14] have used a human body model of 2 – 5mm voxel resolution and with 30 or more tissues types developed from the Visible Human Project [15]. One final approach is the use of voxel-based models that make use of images of the human body and classification of the voxels into different tissue types.

For the present study an unstructured polyhedral finite volume mesh was generated in order to define the shape of the human body. The geometry was based on the ‘Duke’ member of the Virtual Family Dataset [21] that defines 78 tissue types. Starting from the image dataset (Fig. 1(a)) in-house software based on the ITK [22] and VTK [23] libraries was used to assign a unit intensity to all tissue and internal air voxels (Fig. 1(b)), then apply a direct threshold in order to extract a surface triangulation (Figure 1(c)). This surface was then used as an input to create the unstructured mesh with the OpenFOAM [24] mesh generator snappyHexMesh (Figure 1(d)). This mesh consisted of a total of approximately 4.6 million cells with a uniform resolution of 3.125mm hexahedral cells throughout the...
the majority of the domain and polyhedral cells around the skin boundary, which was found to give a mesh independent solution.

B. Thermal Framework

In order to assign thermal properties of the tissues, in-house software based on the ITK and OpenFOAM libraries was developed in order to interpolate these properties from the voxel based virtual family dataset onto the cell centroids of the polyhedral mesh. The interpolated properties include the density \( \rho \), specific heat capacity \( C \), thermal conductivity \( k \), metabolic heat production \( Q_m \), heat sink due to blood perfusion \( \beta \), and the additional heat source due to an electromagnetic (EM) load \( \rho \cdot SAR \). Using the tissue classification defined in the 1mm resolution Duke image dataset, the corresponding tissue properties outlined in [25], [26] were first assigned to each voxel (Fig. 2(a)). Then, for each cell in the polyhedral mesh a neighbour search was performed to find and flag voxels which were positioned within the cell. Using these flagged voxels, Sheppard interpolation was then used to interpolate each thermal property to the cell centroid (Fig. 2(b)). In order to ensure that the thermal properties of the atmospheric air did not affect the interpolated values in the skin cells connected to the skin boundary surface had the properties of the skin directly assigned to them.

The height of the model was 1.80m, surface area, 1.83m\(^2\), and volume, 0.068m\(^3\), all calculated from the triangulated surface extracted from the image dataset. The total mass, 72.5kg, metabolic rate, 119.5W, and heat loss from blood perfusion, 481.2Wm\(^{-3}\)K\(^{-1}\), were computed by the summation of \( \rho \cdot Q_m \) and \( \beta \) (defined at each cell centroid multiplied) by the volume of each cell (essentially a discrete form of the integral of the properties over the volume of the body model). The height and weight reported in [21] for the Duke body model are 1.74m and 70kg, which is in reasonable agreement. The discrepancy in height can be explained by noting that during the scan with which the dataset was obtained the feet were extended. The overall metabolic rate is in reasonable agreement with the 70Wm\(^{-2}\) reported in the ASHRAE standards [27] for a standing relaxed human, providing some confidence that the mesh and interpolated tissue properties are within a reasonable physiological range.

For this paper, we define the thermoneutral condition as the steady-state temperature profile of an unclothed person, standing at rest, subject to an ambient temperature of 28°C and 40% relative humidity (RH). The rationale for choosing these values follows the use of experimental data for validation purposes as discussed in the Section III-A. The temperature change, \( \Delta T \) (expressed in °C), is defined as the difference between the steady-state temperature in a given scenario being examined and the steady-state temperature at the thermoneutral condition. Finally, we define core temperature as the temperature of the blood in the circulation and its value obtained at a point in the pulmonary artery [28].

C. Calculation of Tissue and Blood Temperature using the Finite Volume Method

Throughout the tissue domain the temperature variation with time \( t \) and space \( x \) can be described by Pennes bioheat equation [4], [6], [29]:

\[
\frac{\partial \rho CT}{\partial t} = \nabla \cdot (k \nabla T) + Q_m + \beta (T_b - T) - Q_r + \rho \cdot SAR
\]

where \( T(x, t) \) is the tissue temperature field [K], \( Q_m(x, t) \) is the metabolic heat production [Wm\(^{-3}\)], \( Q_r(x, t) \) is the heat loss through the respiratory system [Wm\(^{-3}\)], \( \beta(x, t) \) is the heat sink due to blood perfusion [Wm\(^{-3}\)K\(^{-1}\)], and \( T_b(t) \) is the arterial blood temperature [K]. The major heat losses in the body occur through the skin and through the respiratory system. Although the skin boundary surface could be segmented from the Duke voxel dataset, the finer details of the respiratory system such as bronchioles, alveoli, etc could not be, as the dataset only includes lung voxels. The implication of this is that while the skin heat loss can be treated as a boundary condition imposed on the bioheat equation (described in Section II-D), the respiratory losses are treated as a source term. The particular form of these losses was taken from Fiala et al. [30] where the evaporative heat loss \( Q_{ev}[W] \) and dry heat loss \( Q_{re}[W] \) are given by:

\[
Q_{ev}(x,t) = \frac{\rho \cdot SAR(x,t)}{\rho_{water}C_{water}} \cdot (RH - 1) \cdot W
\]

\[
Q_{re}(x,t) = \frac{\beta(x,t) \cdot T_b(t)}{\rho \cdot SAR(x,t)} \cdot W
\]
\[ Q_{re} = 4.373 \int \rho_b V_b \frac{\partial T_b}{\partial t} dt + \int \rho_b V_b \beta (T - T_b) dV - Q_{bo} \] (3)

where \( \rho_b \) and \( C_b \) are the density and specific heat capacity of the blood respectively, and \( V_b \) is the total blood volume, taken as 5L. As with the respiratory losses in (2) the integral defining the heat acquisition by the blood is taken over the whole body and \( Q_{bo} \) defines this heat flow under thermoneutral conditions, necessary in order to model the correct variation in blood temperature [31].

The bioheat equation was solved using an in house OpenFOAM based solver, discretizing the Laplacian term in (4) with a Gauss linear corrected scheme and the unsteady term with the Crank-Nicolson method. A custom Neumann heat flux boundary condition was applied at the skin boundary using a clothing model described in Section II. The resulting system of algebraic equations was solved at every time step using a geometric-algebraic multigrid solver based on Gauss-Seidel smoothing with a time step size of 10s and a relative tolerance of 10^-9.

**D. Calculation of SAR using the Finite Difference Time Domain Method**

The fundamental metric for specifying RF-EMF heating is the specific absorption rate, which can be calculated at any point in human tissue from knowledge of the internal electric field using:

\[ SAR = \frac{\sigma |E|^2}{\rho} \] (4)

where \( \sigma \) is the tissue conductivity [S/m], and \( E \) is the electric field [V/m] root mean square (rms). When setting protective limits for localized tissue heating, \( SAR \) is mass averaged, in recognition of the thermal diffusion properties of tissues. For example, ICNIRP sets the localized exposure limit for RF-EMF occupational workers to be 10W/kg over 10g of contiguous tissue, in the frequency range 100kHz to 10GHz [31]. For the general public the limit is reduced further to 2W/kg. The whole-body average (WBA) \( SAR \) limit is 0.4W/kg for occupational exposure (0.008W/kg for general public). In this study, the induced \( SAR \) in the Duke model was calculated using the commercially available finite-difference time-domain (FDTD) software XFDTD [32]. The finite-difference mesh consisted of cubes with sides 2mm in length. The ability of FDTD methods to handle the highly irregular and heterogeneous structures in the human anatomy has made it a popular choice in human RF-EMF modeling [33].

**E. Thermoregulation Model**

Given the desire to model temperature change in the tissues under conditions of thermal stress, the incorporation of a thermoregulation model is vital if physiologically realistic results are to be obtained. There has been a significant body of work performed in the area of thermoregulation and comfort, with a good review of various models found in [34]. Again the model of Stolwijk et al. [4] presents an initial effort to which much modification and improvement has been performed over the years. One well known and widely used model is that of Fiala et al. [30], [35], [36], which defines control equations for sweating, shivering, vasodilation, and vasodilation and validated against numerous physical experiments, and as such was the model chosen for the present study. Examples of thermal modeling applications include the response for hot conditions [13], for RF-EMF exposure at the maximum allowable level including for adults and children [37], and at high levels of RF-EMF exposure [4].

From Fiala et al. [36], the equation for shivering \( Sh[W] \) is:

\[ Sh = 10(\tanh(0.48\Delta T_{sk,m} + 3.62) - 1)\Delta T_{sk,m} - 27.9\Delta T_{hy} + 1.7\Delta T_{sk,m} \frac{dT_{sk,m}}{dt} - 28.6 \] (5)

where \( \Delta T_{sk,m}[K] \) and \( \Delta T_{hy}[K] \) are the deviation in sensitivity weighted mean skin and hypothalamic temperature respectively from thermally neutral conditions, with an upper limit of 350W. The equation for vasodilation is \( Cs \) is:

\[ Cs = 35(\tanh(0.34\Delta T_{sk,m} + 1.07) - 1.0)\Delta T_{sk,m} + 3.9\Delta T_{sk,m} \frac{dT_{sk,m}}{dt} + 6.3\Delta T_{hy} \] (6)

which is dimensionless. The equation for sweating \( Sw[gmin^{-1}] \) is:

\[ Sw = (0.8 \tanh(0.59\Delta T_{sk,m} - 0.19) + 1.2)\Delta T_{sk,m} + (5.7 \tanh(1.98\Delta T_{hy} - 1.03) + 6.3)\Delta T_{hy} \] (7)

with an upper limit of 30g/min. Finally, the equation for vasodilation \( Dl[W K^{-1}] \) is:

\[ Dl = 21(\tanh(0.79\Delta T_{sk,m} - 0.70) + 1.0)\Delta T_{sk,m} + 32(\tanh(3.29\Delta T_{hy} - 1.46) + 1.0)\Delta T_{hy} \] (8)

The model includes a variation in metabolic rate in the tissues:

\[ Q_m = Q_{m,ba} + \Delta Q_m \] (9)

where the deviation in metabolic rate from basal conditions is given by:

\[ \Delta Q_m = Q_{m,ba} \times 2^{(\frac{\Delta T_{sk,m} - 1}{0.5})} + Q_{m,sk} + Q_{m,sw} \] (10)

which includes the van Hoff Q10 effect, the extra metabolic heat production from shivering, and the extra metabolic heat production from exercise. The variation in heat sink due to blood perfusion is given by:

\[ \beta = \beta_0 + \mu_h \Delta Q_m \] (11)

where the proportionality constant \( \mu_h \) is taken as 0.932Wm^{-3}K^{-1} [4]. To incorporate the vasodilation and vasodilation response the heat sink due to blood perfusion in the skin cells is computed as:
The clothing model implemented in the present study is based on the model of Wissler and Havenith [43] which considers a clothing ensemble to be comprised of an arbitrary number of fabric and air layers, where nodes are defined at the boundaries between each layer (Fig. 2). This model includes the effect of heat and mass transport through a clothing ensemble, and the effect of condensation within any of its layers, while avoiding the complexities of fiber orientation and multiphase flow within a textile. The rate of sensible heat transfer $Q_c [Wm^{-2}]$ through a layer $i$ is given by:

$$Q_{c,i} = \frac{T_{i} - T_{i+1}}{R_{c,i}}$$  \hspace{1cm} (13)

where $T_i[K]$ denotes the temperature at the $i^{th}$ node in the ensemble and $R_{c,i}[m^2KW^{-1}]$ denotes the resistance to sensible heat transfer, including the effect of conduction within the fabric layers, the additional effect of radiation within the inner air layers and the effects of convection and radiation with the ambient air. The rate of evaporative heat transfer $Q_e[Wm^{-2}]$ due to water transport through a layer $i$ is given by:

$$Q_{e,i} = \frac{p_{i} - p_{i+1}}{R_{e,i}}$$  \hspace{1cm} (14)

where $p_{i}[Pa]$ denotes the partial pressure of water at a node in the ensemble and $R_{e,i}[m^2PaW^{-1}]$ denotes the resistance to evaporative heat transfer. Using (13) and (14) a clothing ensemble can be created and the nodal temperatures and pressures computed, based on knowledge of both the skin and ambient temperatures and partial pressures of water. From the work of Jones and Ogawa [44] the partial pressure of water at the skin surface is given by:

$$p_{sk} = \frac{\lambda_{H_2O}S_{w} + \frac{P_{e,i}}{R_{e,i}} + \frac{P_{i}}{R_{e,i}}}{\frac{1}{R_{e,i}} + \frac{1}{R_{e,i}}}$$  \hspace{1cm} (15)

where $\lambda_{H_2O}$ is the partial molar heat capacity of water, $S_{w}$ is the surface water vapor pressure at the skin, $P_{e,i}$ and $P_{i}$ are the partial pressures of water vapor at the inner and outer surfaces of fabric layer $i$, respectively, $R_{e,i}$ is the resistance to evaporative heat transfer through fabric layer $i$, and $R_{e,i}$ is the resistance to evaporative heat transfer through the fabric layers.

F. Clothing Model

Under conditions of thermal stress, the sweat secreted under the clothed regions of a person will either be transported through the various layers as water vapor, or condense back into liquid water within one or more of the layers. This phenomenon will affect both sensible and evaporative heat transfer through the clothing and as such was considered to be an important feature to capture, given the purpose of the study. There has been a significant body of work performed in the area of measuring and modeling heat and mass transfer through clothing, with a number databases [38] presenting thermal properties of various fabrics and clothing ensembles. In terms of modeling mass transport through clothing, one approach has involved developing partial differential equations describing the unsteady transport of local moisture ‘regain’ through a fabric [39] while other studies have developed even more detailed systems of partial differential equations describing multiphase flow [40] and including details such as fiber orientation within a textile [7, 41].

The clothing model implemented in the present study is based on the model of Fiala thermoregulation model (the black rectangles are solely for privacy reasons and do not represent body compartments) (c) a cutaway through the 1 mm resolution Duke image dataset illustrating thermal conductivity of different tissues (d) the thermal conductivity interpolated onto the polyhedral mesh (e) the occupational EM load interpolated onto the polyhedral mesh.
Fig. 3. An illustration of the coupling between the clothing model and the polyhedral mesh. Multiple layers can be applied, defining an ensemble and then based on the skin and air temperatures and partial pressures of water, the temperatures at nodes between layers can be computed.

where \( \lambda_{H_2O} \) is the latent heat of vaporization of water, taken as 2256 kJ/kg \(^{-1} \), \( R_{e,i,k} \) is the evaporative resistance of the skin, taken as 333.3 m\(^2\) Pa W \(^{-1} \) \([30]\), and \( p_v \) is the vapour pressure of water, computed as:

\[
p_v = e^{20.386 - \frac{35.52}{T_v}}
\]

At every time step in a given simulation and over every boundary face of the polyhedral mesh, the nodal temperatures and pressures throughout the clothing ensemble are computed by ensuring that the they satisfy the conditions:

\[
\frac{p_{i-1} - p_i}{R_{e,i-1}} = \frac{p_i - p_{i+1}}{R_{e,i}} \quad \text{if} \quad p_i < p_v,
\]

\[
\frac{T_{i-1} - T_i}{R_{e,i-1}} = \frac{T_i - T_{i+1}}{R_{e,i}} = \frac{p_i - p_{i+1}}{R_{e,i}} - \frac{p_{i-1} - p_i}{R_{e,i-1}} \quad \text{otherwise}
\]

\[
E_i \geq E_{i+1}
\]

(17)

where \( E [kgm^{-1} s^{-1}] \) is the rate of water transport between nodes. These conditions produce a linear system of equations that are subject to a nonlinear constraint, which were solved iteratively using the Gauss-Seidel method. The skin temperature defined at the centroid of every boundary face was treated explicitly, using the value from the previous time step with the skin vapor pressure evaluated using (16) and then the skin partial pressure of water evaluated using (15). After the system of equations, a subset of (17) for a given face had converged, the sensible and evaporative heat flux could be computed using (13) and (14) for the first layer (i.e. adjacent to the skin boundary) and the total heat flux used to impose a Neumann boundary condition in the solution of the bioheat equation:

\[
\nabla T \cdot n \big|_f = \frac{Q_{e,1} + Q_{e,1}}{k_{sk}}
\]

(18)

where \( k_{sk} \) in the denominator of (18) is the thermal conductivity of the skin defined at the centroid of every boundary face. In order to investigate the effects of thermal stress under a range of environmental conditions and clothing ensembles, a subset of the database presented in [45] was used (Table I). Boots/shoes are set to have the property PVC coated outerwear (socks are set to permeable outerwear).

### III. RESULTS

#### A. Model Validation

Given the aim of investigating extreme environmental conditions coupled with the additional effects of electromagnetic loads it is important that the thermal model provide realistic temperatures and heat fluxes, and more importantly that the thermoregulation mechanism produce realistic responses under extreme conditions that match experimental data. One widely used source of experimental data is that of Stolwijk et al. [5] who performed a series of experiments on three young men, subjecting them to a number of two hour periods at increasingly elevated ambient temperatures, while recording key temperatures and measuring weight loss (to infer sweat rate and hence evaporative heat loss). These experiments were chosen for simulation in the present study since the experimental conditions such as the weight and height of the subjects, and the elevated temperatures to which they were exposed, are similar to the extreme conditions being investigated. Since the average metabolic rates of the subjects (who were seated in a reclined position for the duration of the experiments) was approximately 46.5 W \( m^{-2} \), the metabolic rate of the Duke body model was scaled by a factor of 0.712 in order to simulate this condition. The resulting total metabolic rate under these conditions was 34 W, which is close to the 87 W found in the model of Fiala et al. [30].

To perform the simulations a basal temperature field is needed under conditions of thermal neutrality, in order to provide a reference point for the thermoregulation mechanism. As such, the bioheat equation was solved subject to an ambient temperature of 28°C and 40% relative humidity (RH), similar to the neutral room used in the Stolwijk experiments, with the \( Q_m \) and \( \beta \) fields held constant, and allowing a period of 4 hours for the temperature to reach a steady state. We call this the thermoneutral condition (Fig. 4). The average skin temperature of 34.4°C and brain temperature of 37.0°C are both in close agreement with the model of Fiala et al. [30] and the thermoneutral conditions in [5]. These temperatures were acquired by computing the volume weighted mean of the temperatures in all

---

**TABLE I. GARMENT THERMAL PROPERTIES**

<table>
<thead>
<tr>
<th>Garment</th>
<th>( R_e [m^2 K W^{-1}] )</th>
<th>( R_p [m^2 Pa W^{-1}] )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Polypropylene underwear</td>
<td>0.026</td>
<td>3.7</td>
</tr>
<tr>
<td>Permeable outerwear</td>
<td>0.025</td>
<td>5.6</td>
</tr>
<tr>
<td>Semi-permeable outerwear</td>
<td>0.023</td>
<td>18.6</td>
</tr>
<tr>
<td>PVC coated outerwear</td>
<td>0.007</td>
<td>1000.0</td>
</tr>
<tr>
<td>1 mm interior air space</td>
<td>0.120</td>
<td>21.8</td>
</tr>
<tr>
<td>External boundary layer</td>
<td>0.096</td>
<td>9.0</td>
</tr>
</tbody>
</table>

---
Fig. 5. A comparison between the Duke body model and the experiments of Stolwijk et al. [5] where three young men were exposed to 2 hour periods of elevated ambient temperature and humidity. The solid curves illustrate the model outputs and the dotted curves illustrate the experimental data. Comparisons are made for rectal, tympanic and skin temperature, $T$, metabolic heat transfer $Q_m$, and evaporative heat transfer, $Q_e$. To enable a comparison with measured data the total evaporative heat loss through the skin and the total metabolic rate, are divided by the skin area, to present the data in units $Wm^{-2}$. In all experiments the initial ambient temperature and relative humidity were approximately $28^\circ C$ and $40\%$ respectively. Fig. 5(a) and 5(b) illustrate the results from a 2 hour period at $33.3^\circ C$ and $34\%RH$, Fig. 5(c) and 5(d) illustrate the results from a 2 hour period at $37.5^\circ C$ and $33\%RH$, Fig. 5(e) and 5(f) illustrate the results from a 2 hour period at $42.5^\circ C$ and $28\%RH$, and Fig. 5(g) and 5(h) illustrate the results from a 2 hour period at $47.8^\circ C$ and $27\%RH$. 

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cells within the mesh that had been classified to be of a particular tissue region (Fig. 2(a)). The average temperature in the eyes was 36.0°C, testes, 36.1°C, marrow, 36.0°C, and core, 37.0°C. The total evaporative heat loss was 20.8 W, which is also in reasonable agreement with the 18.1 W found in the model of Fiala et al.

The comparison between the simulated and experimental data is presented in Fig. 5(a)-5(h), illustrating the transient rectal, tympanic, blood, and mean skin temperatures, and the metabolic rate and evaporative heat loss throughout the 4 hour period of the experiment. As can be observed, the simulated variations in temperature show close agreement with the experimental data, and the variations in evaporative heat loss show reasonable agreement with experimental data. Given the minor differences in surface area and thermoneutral evaporative heat losses between the Duke body model and the test subjects, it is not unexpected that the whole body evaporative heat losses differ slightly. The major discrepancy in the validation is in the comparison between simulated and measured mean skin temperature in the 33.3°C experiment. Here, the measured skin temperature is lower than in other experiments and also lower than the simulated mean skin temperature obtained under thermoneutral conditions, but a similar relative variation is still observed when subject to the elevated temperature. Most importantly, under the most extreme conditions of 47.8°C and 27%RH the computational model is able to accurately capture the increases in key temperatures, providing some confidence that the Duke body model and thermoregulation mechanism are capable of providing physiologically realistic results.

In future studies, a valuable test will be to compare the model with the experimental data of Adair and colleagues, that provide human body temperature measurements for EM load conditions well above the reference levels and over an extended time period. A study of Foster and Adair [46] uses this data and shows that the Stolwijk et al. model accurately accounts for the resultant thermal effects.

**B. Effects of EM Load, Clothing, and Ambient Conditions**

In this section we present simulations of some sample environmental conditions, in order to illustrate the relative importance of the various sources of heating; namely EM load, clothing, and ambient conditions, but leave a more detailed parametric study to future work.

Fig. 6 illustrates the effect of an EM load under thermoneutral conditions. The EM exposure was set to be a vertically polarized 1 GHz plane-wave. Dielectric properties (electrical conductivity, \(\sigma\), and relative permittivity, \(\epsilon_r\)) for the tissues were sourced from [47]. This frequency was selected as it is around the values used internationally for mobile phone communications. Hence, a plane-wave may represent the exposure from telecommunications infrastructure such as a mobile phone base-station. The strength of the field was set to an input power flux density level of 25 W m\(^{-2}\), the maximum allowable reference level at 1 GHz for RF-EMF workers [41]. This level is set to ensure the SAR limits are not exceeded. For this reference field level, the WBA SAR was calculated to be 0.15 W kg\(^{-1}\) and the peak...
The EM exposure is at the maximum allowable reference level for occupational exposure.

Fig. 9. A comparison of temperature difference in some key tissues and the core, at 50°C and 70% RH, with the two layer impermeable clothing ensemble, with and without an additional EM load. The EM exposure is at the maximum allowable reference level for occupational exposure. ΔT represents the temperature rise above the thermoneutral condition.

10g SAR was 2.3 W/kg which occurred in the ankle. It can be observed that in the skin, brain, and core, there is a very minor increase of approximately 0.05°C, whereas the eye experiences the greatest increase of approximately 0.2°C. As can be observed in Fig. 2(c) the EM load is quite heterogeneous with relatively low heat input throughout most of the body, except for localised regions such as the eye and nasal area, hands and legs.

Fig. 7 illustrates the effect of the two layer clothing ensemble under thermoneutral conditions with no EM load. It can be observed that areas on the periphery of the body such as the skin and testes experience a temperature increase of approximately 0.7°C, while areas such as the brain and core experience no increase in temperature, qualitatively what would be expected from the addition of clothing. One anomalous result is the eye temperature, which experiences a decrease of approximately 0.1°C. This result is in fact due to the limitation of the thermoregulation model. As the skin temperature increases due to the addition of clothing, minor sweating and vasodilation will occur, but this will in fact occur uniformly over a body compartment, and weighted by the distribution coefficients of the Fiala model. This means that there will be vasodilation and sweating over the eyelids and since the environmental conditions haven’t changed, this will ultimately lead to a reduction in temperature.

Fig. 8 illustrates the effect of environmental conditions by comparing key temperatures under thermoneutral conditions to those at an ‘extreme’ condition of 50°C and 70% RH, without clothing and with no EM load. As can be observed, the high ambient temperature and relative humidity leads to and increase approximately 1.3°C in the skin, and approximately 1.3°C in the brain and core, with the largest increase of approximately 1.9°C occurring in the testes.

Finally, Fig. 9 illustrates the effect that an additional EM load could for a person experiencing extreme ambient conditions, while wearing a 2 layer clothing ensemble. In this case temperature increases in the key regions relative to thermoneutral temperatures are presented. It can be observed that the EM load presents an additional increase of approximately 0.2°C in the skin and approximately 0.3°C in the brain and core. An interesting result of this simulation is that the addition of EM load has a greater effect under adverse conditions than under thermoneutral conditions. The reason for this effect is that under thermoneutral conditions, the body’s thermoregulation mechanism has the capability to remove the heat input from the EM load by initiating minor vasodilation and sweating, limiting the temperature increase. Under adverse conditions however, the thermoregulation mechanism’s capacity to remove heat has become exhausted, meaning that the additional thermal loads can have a greater effect on steady state tissue temperature.

IV. DISCUSSION

Validation of the thermal model shows close agreement with time series experimental data of Stolwijk et al. [5] for evaporative heat loss, mean skin temperature, and rectal and tympanic temperatures, the latter surrogates for core body temperature. Under thermoneutral conditions, the mean skin temperature of 34.4°C and brain temperature of 37.0°C are in close agreement with the model of Fiala et al. [10]. These results provide confidence that the thermoregulation model is capable of generating physiologically realistic results. In the adverse case (Fig. 7), average increases in the skin temperature were up to 1.3°C. This level is comparable to the limit suggested in [2], [3]. Our assessments of the effect of various influences on this temperature change will be given in future papers. For the EM load component, the highest increase was around 0.2°C in the eye.

The two major limitations of the model presented in this study are related to the thermoregulation model and the approach for modeling the variation in blood temperature. One of the desirable features of this type of body model compared to the nodal type models of Stolwijk et al. [1] and Fiala et al. [10] is the ability to specify detailed and localised temperature information in particular parts of the body. This capability is especially important when investigating the heating due to EM loads which, by nature is very heterogeneous, depending on geometrical and histological factors, and can’t be modeled in a body is considered to be a collection of spheres. Despite the increased fidelity in terms of the bioheat calculation however, for lack of better information, the thermoregulation model is still based on a nodal/body compartment type approach, using distribution coefficients that were defined for these cylinders and spheres. One future improvement to the model could therefore include modification of the thermoregulation distribution coefficients from ‘say’ discrete sweating distribution by creating a sweat gland density field over the surface of the Duke body model based on the most recent and experimental data [48].

The limitation with the blood temperature model lies in the fact that fundamentally, the combination of Equations (2) and (3) as used in [11], [31] does not satisfy the physical principle of conservation of energy. Under thermoneutral conditions, the arterial blood temperature would remain constant indicating that the left hand side of (3) be zero. Any increase in tissue temperature (from ambient conditions, clothing, or EM load) will cause tissue temperatures to increase, and then the integral term over the body on the right hand side of (3) will become greater than zero, leading to an increase in blood temperature. When fed back into the bioheat equation, this increase in arterial blood temperature will lead to an increase in tissue temperature, ultimately leading to a positive feedback cycle so that over time, both blood and tissue temperatures can increase in an unphysical way that violates conservation of energy. This effect is in fact the reason for the $Q_{ao}$ term in Eqn. (4) and for only minor increases in tissue temperature, this modeling approach can produce realistic results that will compare with experimental data. One future improvement to the model could however be the incorporation of a bioheat model such as that of Shrivastava et al. [49] that satisfies conservation of energy.
by defining partial differential equations (PDEs) for both blood and tissue temperature.

V. CONCLUSION

This study has described the development of a novel computational modeling environment for the thermal analysis of a fully clothed, finely discretized, heterogeneous human body model, subject to RF-EMF, and also subject to adverse environmental conditions, including high humidity and high ambient temperature. An initial observation is that while EM loads at the occupational safety limit will contribute an additional thermal load to the tissues and subsequently cause an elevated temperature, the magnitude of this effect is far outweighed by that due to the conditions including the ambient temperature, relative humidity, and the type of clothing worn. As a result of this preliminary investigation we do not envisage any immediate change being required to the RF-EMF safety standards and guidelines for RF-EMF workers subject to adverse conditions. In future studies we aim to validate thermal models of the human body that incorporate effects of RF heating with varying levels of work (elevated metabolic rates), simulating the work environment of an RF-EMF worker. Future studies will also consider the influence of different RF-EMF exposure conditions including localized and extreme exposures, hot winds, and different body shapes and sizes including children and pregnant women. Outcomes of these studies may provide valuable input to future revisions of RF-EMF exposure guidelines and standards.

VI. ACKNOWLEDGMENT

This research was supported by a Victorian Life Sciences Computation Initiative (VLSCI) grant VR0204 on its Peak Computing Facility at the University of Melbourne, an initiative of the Victorian Government.

REFERENCES


