COMPUTER SIMULATION OF THE RESPONSE OF THE HUMAN BODY
TO IMMERSION IN COLD WATER

by

Graham Richardson

Thesis submitted for the degree of Doctor of Philosophy
at the University of Surrey

December 1988

A report on work carried out by the author at the

Royal Air Force Institute of Aviation Medicine

VOLUME 1
ABSTRACT

Many military and civilian personnel are required to work in situations where there is a risk of accidental immersion in the sea. Since immediate rescue may not be possible, it is important to predict the time for which survivors may remain alive. A computer-based mathematical model may provide a means of simulating the change in body temperature with time. The need for such a model and the physiological basis for its development have been investigated.

A mathematical model has been developed in which the human body is visualised as 15 cylindrical or spherical segments, each divided into 10 radial shells of tissue. Passive heat flows are simulated at the surface and internally. Transport of heat by blood flow is represented in 120 arterial and venous compartments. The physiological mechanisms of thermoregulation are simulated, using existing physiological data.

The model is implemented in structured FORTRAN 77 code. Although it is primarily configured for cold water immersion, infrastructure is included to permit adaption to simulate heat or cold stress in air. Code has been included for heat transfer through clothing and for exercising as well as resting conditions.

Comparisons of the model predictions have been made against experimental data obtained from semi-nude immersions in water at 12, 18 and 24°C. For subjects with a relatively high body mass and fat content, the predicted body core temperature is generally within plus or minus one standard error of the experimental mean. For small, thin subjects at 12 and 18°C, the prediction is within two standard errors. The model does not cope well with sudden large changes in exercise but predictions for clothed subjects appear adequate.
ACKNOWLEDGEMENTS

This thesis is an account of original work carried out at the Royal Air Force Institute of Aviation Medicine, Farnborough, under the University of Surrey scheme for collaboration with industry and research establishments. I should like to thank the retired Commandant of the Institute, Air Vice Marshal P Howard RAF (Retd.) and his successor Air Commodore J Ernsting RAF for permission to undertake and submit this work for the degree of Doctor of Philosophy. I have also benefited from the support of Dr J R Allan, Director of Research.

I am indebted to my joint supervisors Prof G A Parker and Mr S C Hughes at the University and Dr C T Kirkpatrick at the Institute for their help and encouragement. A number of my colleagues have provided invaluable advice, notably Dr P A Hayes on human physiology, Dr D Burgess and Capt J B Cohen USAF on computing techniques and Dr A J Belyavin on statistical analysis. In particular, I should like to thank Sqn Ldr P J Sowood RAF for permission to use his experimental results in validating the model.

Finally, I must thank my wife Patricia and daughter Jenny for their tolerance as I indulged this mid-life yearning for recognition.
TABLE OF CONTENTS

VOLUME 1

<table>
<thead>
<tr>
<th>Title Page</th>
<th>Page No.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Abstract</td>
<td>ii</td>
</tr>
<tr>
<td>Acknowledgements</td>
<td>iii</td>
</tr>
<tr>
<td>Table of contents</td>
<td>iv</td>
</tr>
<tr>
<td>List of figures</td>
<td>vi</td>
</tr>
<tr>
<td>List of tables</td>
<td>ix</td>
</tr>
<tr>
<td>Abbreviations and symbols</td>
<td>x</td>
</tr>
</tbody>
</table>

CHAPTER 1 - THE NEED TO PREDICT RESPONSES TO THERMAL STRESS

1.1. Background 1
1.2. Heat stress in military environments 4
1.3. Cold stress in military environments 9
1.4. Relative merits of modelling and experimentation 13

CHAPTER 2 - ELEMENTS OF THERMAL PHYSIOLOGY

2.1. Passive heat exchange 18
2.2. The active control system 23
2.3. Energy balance 27

CHAPTER 3 - ASSESSMENT OF EXISTING MODELS

3.1. Brief history 30
3.2. Steady state models 33
3.3. Empirical models 37
3.4. Simple dynamic models 43
3.5. Multi-segment, lumped temperature models 51
3.6. Recent Developments 74
3.7. Rationale for future development 85
# LIST OF FIGURES

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page No.</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1</td>
<td>Earth's climatic zones</td>
<td>2</td>
</tr>
<tr>
<td>1.2</td>
<td>Comfort zones for normally clothed adults</td>
<td>5</td>
</tr>
<tr>
<td>1.3</td>
<td>Upper limits of exposure for unimpaired mental performance</td>
<td>7</td>
</tr>
<tr>
<td>1.4</td>
<td>Estimated cockpit heat loads</td>
<td>8</td>
</tr>
<tr>
<td>1.5</td>
<td>Effects of reduced core temperature</td>
<td>10</td>
</tr>
<tr>
<td>1.6</td>
<td>Effect of wind on temperature</td>
<td>12</td>
</tr>
<tr>
<td>1.7</td>
<td>Survival times in cold water</td>
<td>14</td>
</tr>
<tr>
<td>1.8</td>
<td>British coastal water temperatures</td>
<td>15</td>
</tr>
<tr>
<td>2.1</td>
<td>Heat flow mechanisms</td>
<td>29</td>
</tr>
<tr>
<td>3.1</td>
<td>Classification of models</td>
<td>31</td>
</tr>
<tr>
<td>3.2</td>
<td>Infinite slab model geometry</td>
<td>45</td>
</tr>
<tr>
<td>3.3</td>
<td>Single cylinder model geometry</td>
<td>47</td>
</tr>
<tr>
<td>3.4</td>
<td>Geometry of Stolwijk's model</td>
<td>52</td>
</tr>
<tr>
<td>3.5</td>
<td>Stolwijk's parameters for body segments</td>
<td>53</td>
</tr>
<tr>
<td>3.6</td>
<td>Validation of Stolwijk's model</td>
<td>57</td>
</tr>
<tr>
<td>3.7</td>
<td>Hancock's validation of Stolwijk's model</td>
<td>58</td>
</tr>
<tr>
<td>3.8</td>
<td>Geometry of Gordon's model</td>
<td>61</td>
</tr>
<tr>
<td>3.9</td>
<td>Validation of Gordon's model</td>
<td>63</td>
</tr>
<tr>
<td>3.10</td>
<td>&quot; &quot; &quot;</td>
<td>64</td>
</tr>
<tr>
<td>3.11</td>
<td>Geometry and blood flow in Wissler's model</td>
<td>65</td>
</tr>
<tr>
<td>3.12</td>
<td>Heat stress validation of Wissler's model</td>
<td>71</td>
</tr>
<tr>
<td>3.13</td>
<td>Validation of Wissler's model for cold water immersion</td>
<td>72</td>
</tr>
<tr>
<td>3.14</td>
<td>Rectal temperature as a function of mean weighted skinfold</td>
<td>73</td>
</tr>
<tr>
<td>3.15</td>
<td>Cooling of the leg when exposed to -10°C air, comparison of experimental and model data</td>
<td>75</td>
</tr>
<tr>
<td>3.16</td>
<td>Huckaba's control system diagram</td>
<td>77</td>
</tr>
<tr>
<td>3.17</td>
<td>Validation of Huckaba's model</td>
<td>78</td>
</tr>
<tr>
<td>3.18</td>
<td>Validation of Volpe and Jain's model</td>
<td>80</td>
</tr>
<tr>
<td>3.19</td>
<td>Smith and Twizell's spatial discretization</td>
<td>81</td>
</tr>
<tr>
<td>3.20</td>
<td>Smith and Twizell's radial profile, example</td>
<td>83</td>
</tr>
<tr>
<td>Section</td>
<td>Title</td>
<td>Page</td>
</tr>
<tr>
<td>---------</td>
<td>----------------------------------------------------------------------</td>
<td>------</td>
</tr>
<tr>
<td>4.1</td>
<td>State space representation of a system with n variables</td>
<td>93</td>
</tr>
<tr>
<td>4.2</td>
<td>Two dimensional illustration of linearisation about a point</td>
<td>95</td>
</tr>
<tr>
<td>4.3</td>
<td>Vector/matrix representation of the human thermal system</td>
<td>98</td>
</tr>
<tr>
<td>4.4</td>
<td>Geometrical representation of the body</td>
<td>100</td>
</tr>
<tr>
<td>4.5</td>
<td>Radii for heat conduction between shells</td>
<td>106</td>
</tr>
<tr>
<td>4.6</td>
<td>Blood volume as a function of body mass</td>
<td>109</td>
</tr>
<tr>
<td>4.7</td>
<td>Distribution of venous and arterial blood</td>
<td>110</td>
</tr>
<tr>
<td>4.8</td>
<td>Arrangement of blood vessels</td>
<td>112</td>
</tr>
<tr>
<td>4.9</td>
<td>Conductances affecting blood vessels</td>
<td>116</td>
</tr>
<tr>
<td>4.10</td>
<td>Combination of segmental blood flows</td>
<td>125</td>
</tr>
<tr>
<td>4.11</td>
<td>Estimation of surface temperature</td>
<td>129</td>
</tr>
<tr>
<td>4.12</td>
<td>Scheme of blood distribution</td>
<td>133</td>
</tr>
<tr>
<td>5.1</td>
<td>Distribution of body mass</td>
<td>142</td>
</tr>
<tr>
<td>5.2</td>
<td>A subject undergoing voluntary immersion (upper view)</td>
<td>143</td>
</tr>
<tr>
<td>5.3</td>
<td>A subject undergoing voluntary immersion (lower view)</td>
<td>144</td>
</tr>
<tr>
<td>5.4</td>
<td>Rectal temperature change for Group 3 at 24°C</td>
<td>152</td>
</tr>
<tr>
<td>5.5</td>
<td>Metabolic rate for Group 3 at 24°C</td>
<td>153</td>
</tr>
<tr>
<td>5.6</td>
<td>Surface heat flow for Group 3 at 24°C</td>
<td>154</td>
</tr>
<tr>
<td>5.7</td>
<td>Rectal temperature change for Group 2 at 24°C</td>
<td>155</td>
</tr>
<tr>
<td>5.8</td>
<td>Skin temperature for Group 2 at 24°C</td>
<td>156</td>
</tr>
<tr>
<td>5.9</td>
<td>Rectal temperature change for Group 1 at 24°C</td>
<td>157</td>
</tr>
<tr>
<td>5.10</td>
<td>Surface heat flow for Group 1 at 24°C</td>
<td>158</td>
</tr>
<tr>
<td>5.11</td>
<td>Metabolic rate for Group 1 at 24°C</td>
<td>159</td>
</tr>
<tr>
<td>5.12</td>
<td>Rectal temperature change for Group 3 at 18°C</td>
<td>161</td>
</tr>
<tr>
<td>5.13</td>
<td>Metabolic rate for Group 3 at 18°C</td>
<td>162</td>
</tr>
<tr>
<td>5.14</td>
<td>Surface heat flow for Group 3 at 18°C</td>
<td>163</td>
</tr>
<tr>
<td>5.15</td>
<td>Rectal temperature change for Group 2 at 18°C</td>
<td>164</td>
</tr>
<tr>
<td>5.16</td>
<td>Rectal temperature change for Group 1 at 18°C</td>
<td>165</td>
</tr>
<tr>
<td>5.17</td>
<td>Surface heat flow for Group 1 at 18°C</td>
<td>166</td>
</tr>
<tr>
<td>5.18</td>
<td>Metabolic rate for Group 1 at 18°C</td>
<td>167</td>
</tr>
<tr>
<td>5.19</td>
<td>Rectal temperature change for Group 3 at 12°C</td>
<td>168</td>
</tr>
</tbody>
</table>
5.20 Skin temperature for Group 3 at 12°C
5.21 Rectal temperature change for Group 2 at 12°C
5.22 Skin temperature for Group 2 at 12°C
5.23 Rectal temperature change for Group 1 at 12°C
5.24 Surface heat flow for Group 1 at 12°C
5.25 Skin temperature for Group 1 at 12°C
5.26 Rectal temperature for an exercising subject at 12°C
5.27 Heat flow for an exercising subject at 12°C
5.28 Metabolic rate for an exercising subject at 12°C
5.29 Rectal temperature for a clothed subject at 6.6°C
5.30 Heat flow for a clothed subject at 6.6°C
5.31 Mean skin temperature for a clothed subject at 6.6°C
5.32 Rectal temperature for a clothed subject at 6.2°C
5.33 Heat flow for a clothed subject at 6.2°C

A1 General arrangement of a subject performing underwater exercise
A2 Wheel, vane and leg geometry
A3 Phase relationships of thigh and calf geometry
A4 Metabolic rate vs. water temperature for 5 pedalling speeds
### LIST OF TABLES

| 4.1 Properties and distribution of tissue | 103 |
| 4.2 Fractional distribution of blood volume | 111 |
| 4.3 Starting and set point temperatures | 118 |
| 4.4 Metabolic weighting factors and resting values | 120 |
| 4.5 Distribution of work and shivering metabolism | 123 |
| 5.1 Characteristics of subjects | 145 |
| 5.2 Characteristics of the three subject groups | 148 |
| 5.3 Numbers of subjects at specific temperatures | 149 |
| 5.4 Details of clothing assemblies | 182 |
| A1 Physical data on cycle and body | 220 |
| A2 Comparison of predicted and measured metabolic rate at 30°C | 221 |
LIST OF SYMBOLS AND ABBREVIATIONS

(VARIABLES IN COMPUTER PROGRAMS NOT INCLUDED)

a  radius, constant
A  area, system matrix
b  constant
B  input matrix
°C degrees Celsius
C  thermal capacity, drag coefficient
c  specific heat
d  differential operator, fluid drag
D  total fluid drag
e  emissivity
E  error signal
exp exponential
f  permeability efficiency
F  fraction, weighting factor
ft  feet
g  gramme, rate of heat generation
G  gradient
h  hour, heat flow coefficient, radius increment
H  height, heat flow coefficient
i  integer
I  insulation, integer, permeability index
j  integer
J  Joule, integer
K  thermal conductance, Kelvin, constant quantity
k  kilo, thermal conductivity, integer
kt  knot
l  length
m  metre, milli, mass of blood, molecular weight
M  mass of body, metabolic rate
min minute
n  rotational speed
N  integer
NBC nuclear, biological & chemical
P  mechanical power, pressure, percent body fat
Q  rate of heat transfer
r  radius
R  ideal gas constant
RH relative humidity
s  second, signal
S  signal, specific gravity
t  time
T  temperature, mechanical torque
u  input vector
U  heat transfer coefficient
v  velocity
V  volume
w  rate of flow of blood
W  watt
WBGT wet bulb globe temperature
x  distance, state vector
X  absolute humidity

α  constant
β  constant
δ  partial differential operator
Δ  a small increment
f  function
λ  latent heat of water
μ  micro, viscosity
η  mechanical efficiency
π  pi
ρ  density
σ  Stefan’s constant, standard deviation
Σ  sum
τ  time constant
ω  angular velocity
ʃ  integral
√  square root
∞  infinity
1. THE NEED TO PREDICT HUMAN THERMOREGULATORY RESPONSES

1.1. Background

All living organisms depend, for their existence, on chemical reactions. Because no chemical reaction is 100% efficient, it follows that all living organisms produce heat. This heat can be stored in the organism or lost to the environment. If it is stored, the heat content and therefore the temperature of the organism will rise. A temperature gradient between the core of the organism and the immediate environment will be created, down which heat will flow. Organisms which possess no thermal control system will reach a steady state condition and their temperature will vary with the environmental temperature. Such organisms are termed poikilotherms and, as Hardy (1979) points out, these comprise the majority of species on Earth.

A relatively small number of species, including most birds and mammals, control the production and loss of metabolic heat. As a result, body temperature is maintained within narrow limits of a 'normal' value, despite wide variations of environmental temperature. This ability seems to have evolved because of its advantages for survival and exploitation of ecological niches. Organisms possessing physiological control of temperature are termed homeotherms. Man is a homeotherm.

Current evidence suggests that man originally developed in central Africa in a fairly hot and moist environment. Over the past million years or so, man has spread to all regions of the world from the tropics to the polar regions. Nevertheless, the human body remains well adapted to life in tropical environments. Edholm (1978) states that apart from industrial regions in Western Europe and Japan, the most densely populated areas are the island of Java and the Ganges valley. Figure 1.1
FIG. 1.1 EARTH'S CLIMATIC ZONES

- Isotherm of 6°C for coldest month
- Isotherm of 18°C for coldest month
- Isopleth of 6°C below 0°C
- Isopleth of 3°C below 0°C

ARCTIC
COOL
TEMPERATE
WARM

80°N 60°N 40°N 20°N 0 20°S 40°S 60°S

160°W 120°W 80°W 40°W 0 40°E 80°E 160°E
illustrates the Earth's climatic zones. Man's success in spreading to sub-tropical, temperate and finally cold latitudes has been dependent largely on protection and technology rather than biology. The simple but effective survival techniques employed by Eskimos and the considerable technological challenge in setting up a laboratory at the South Pole provide two examples. Eskimos remain tropically adapted to the micro-climate beneath their clothing despite living in a climate where sub-zero temperatures are normal.

The rapid growth of technology over the past few decades has improved thermal protection but has also increasingly exposed man to hostile environments. Next to manned space flight, this is perhaps most evident in the field of defence. The aviation environment, in particular, is one in which the physiological control system is sometimes unable to maintain a normal body temperature. Aircraft operate over the whole range of naturally occurring climates at ground level, from extreme arctic cold (-60°C, 0%RH) to tropical heat (+40°C, 50%RH). For a reasonable proportion of the time, aircrew will be exposed to local climatic conditions. For groundcrew, the problem is considerably worse, owing to longer exposure and higher workloads. The natural environment is of great importance when aircrew have to survive a crash, ejection or ditching. However, both heat and cold stress can occur in aircraft cockpits or helicopter cabins. In the following chapters, the term 'thermal stress' is used to indicate the external factors influencing the human thermal system. The term 'thermal strain' is the amount by which the system is actually perturbed by the stress.
1.2. Heat stress in military environments

Any resident of a cold or temperate zone who visits a country with a hot climate will experience discomfort or even illness. Comfort zones are illustrated in Figure 1.2, and vary with acclimatisation. It is of great military importance to know the effects of such changes on individual performance in wartime operations, when many thousands of men may be rapidly deployed to unfamiliar climates. The combinations of exercise and heat can stress the cardiovascular system to its limit in attempting to supply blood to working muscle and to the skin for heat dissipation.

The most common form of heat illness is sudden collapse called 'heat syncope'. It is caused by pooling of blood in the muscles and skin in an attempt to lose heat, thus reducing the blood pressure in the brain. Consciousness quickly returns when the victim lies down or is moved to a cooler environment. Heat stroke is a much more serious condition and is characterised by a rapid rise in body temperature following a loss of the ability to lose heat by sweating. If the core temperature exceeds 43°C, recovery may not be possible.

Thermal problems in high speed aircraft can be quite severe. It might be thought that modern air conditioning systems would be capable of providing a thermally comfortable environment. However, a number of factors combine to make it barely possible for systems to keep up with the demands placed on them. The major sources of heat input to the cockpit are aerodynamic heating of the airframe, dissipation of electrical power in avionic equipment, solar radiation and the pilot's and navigator's own metabolic heat. The rate of aerodynamic heating depends on airspeed, air density and ambient temperature. As Allnutt and Allan (1976) point out, the current emphasis on high speed, low level flight for many
FIG 1.2 COMFORT ZONES FOR NORMALLY CLOTHED ADULTS

From Lemaire (1960)
operational situations increases the likelihood of high cockpit temperatures. The degree of concentration required for flying is high and, as Wing (1965) shows, mental performance is affected by exposure to hot environments. The curve in Figure 1.3 shows the time limit for unimpaired performance of a mental multiplication task at various Wet Bulb Globe Temperatures (WBGT). WBGT=0.7T_{wb}+0.3T_{g}, where $T_{wb}$=wet bulb temperature and $T_{g}$=150mm globe temperature.

The need for all-round visibility leads to large cockpit canopies which not only expose the aircrew to solar radiation, but increase cockpit temperatures by the 'greenhouse effect'. Radiant heat is also absorbed in the rest of the structure and can reach the cockpit by conduction and convection. The increasing use of sophisticated avionic aids affects cockpit temperature in two ways. Firstly, waste heat is generated electrically both in the cockpit and equipment bays elsewhere. Secondly, the demand for cooling air from the conditioning system is increased, leaving less of the system capacity for the cockpit.

The final source of heat is the aircrew themselves. At rest, each crew member will generate approximately 100W. This can increase by a factor of 2 to 3 during high performance flying. An example of worst case heat input to a modern fighter aircraft is shown in Figure 1.4.

In modern warfare it is necessary to provide whole body protection against nuclear, biological and chemical (NBC) threats. Some items of NBC clothing are impermeable to air and water, therefore sweating quickly causes a saturated environment from which heat loss can only be achieved by conduction to the outer surface. It is clearly advisable to investigate thoroughly any such clothing assembly in its intended range of environments, both by experiment and by modelling. Clothing assemblies
FIG 1.3 UPPER LIMITS OF EXPOSURE FOR UNIMPAIRED MENTAL PERFORMANCE

From Wing (1965)
### FIG 1.4 ESTIMATED COCKPIT HEAT LOADS

<table>
<thead>
<tr>
<th>Heat Load Description</th>
<th>Heat Load (kW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Aerodynamic heat load</td>
<td>9.0 kW</td>
</tr>
<tr>
<td>2. Solar radiation load</td>
<td>2.5 kW</td>
</tr>
<tr>
<td>3. Electrical heat load (cockpit instruments)</td>
<td>1.2 kW</td>
</tr>
<tr>
<td>4. Pilot metabolic heat production</td>
<td>0.1 kW</td>
</tr>
<tr>
<td>5. Electrical heat load (equipment bays)</td>
<td>9.5 kW</td>
</tr>
</tbody>
</table>

Total direct heat input to cockpit = 12.8 kW

Indirect heat input
worn by aircrew generally increase problems of thermal stress in cockpits, owing to their high degree of insulation. Protection must be provided in the form of anti-G suits, NBC and immersion coveralls and lifepreservers. This results in a far lower air temperature being required for thermal comfort than would be the case for a shirtsleeve environment. One notable conflict arises where the degree of insulation required to ensure adequate survival times in the sea would cause intense heat stress in some flight regimes.

1.3. Cold stress in military environments

Cold stress is not commonly encountered in modern aircraft because of the large volumes of hot air available from the engines, but older jet aircraft may have poor cabin conditioning systems. The temperature at 35,000 ft altitude is typically -55°C and certain aircraft flying at this level with low power settings can only maintain a cockpit temperature which is less than 0°C. Occasionally, helicopters are flown with the doors open or with low cabin temperatures to prevent heat stress in troops wearing full arctic kit. Here, too, the aircrew can suffer cold stress, and the prediction of its onset is of vital interest.

If the body is unable to maintain thermal equilibrium in a cold environment then the amount of heat stored will reduce and the core temperature will fall. Death would eventually occur from hypothermia but long before this, performance of tasks such as those required to fly an aircraft is impaired. The 'normal' temperature of most core tissue in man is between 36 and 37°C - Harrison (1978). Deterioration of mental ability and slowing of speech are evident at an oral temperature of 35°C, as illustrated in Figure 1.5 from Maclean and Emslie-Smith (1977).
FIG 1.5 EFFECTS OF REDUCED CORE TEMPERATURE

From Maclean & Emslie-Smith (1977)
Cold stress is more usually a result of exposure to the outside environment. Apart from hypothermia, this can cause a number of conditions ranging from chilblains to frostbite. Chilblains are the mildest form of 'non-freezing cold injury' a term covering also conditions such as trench or immersion foot. These conditions result from an insufficient supply of blood to a poorly insulated extremity. Quite severe injury can occur after prolonged immersion of the feet and legs in cold mud or water, either in trenches or in a life raft. The condition, known as 'trench foot' or 'immersion foot', consists of damage to nerves and muscles followed by swelling, which further restricts the supply of blood.

Vital actions involving the use of the hands and feet can be slowed by reduced extremity temperatures, as shown by Allan, Marcus and Saxton (1974). Such actions are needed to fly an aircraft or operate survival aids. It is therefore important to develop the ability to predict thermal conditions at extremities as well as in the body core.

Frostbite is the name given to the condition when the affected part actually freezes. Damage can be trivial if thawing occurs within minutes but prolonged freezing can cause gangrene and the loss of fingers, toes or even parts of limbs. The effect of wind chill, shown in Figure 1.6, greatly increases the rate at which extremities cool.

One of the most serious hazards is that of whole-body immersion in cold water after a plane crash or shipwreck. Water is a much better conductor of heat than air and will quickly produce hypothermia in a poorly insulated individual. The hypothermic effects of cold water have been studied in some detail by Keatinge (1969) who demonstrated the relationship between body size and rate of cooling. The analysis by Molnar (1946) of survival
1 Comfort with normal precautions
2 Very cold, travel becomes uncomfortable on overcast days
3 Bitterly cold, travel becomes uncomfortable even on clear sunny days
4 Freezing of human flesh begins, travel and life in temporary shelter becomes disagreeable
5 Exposed flesh will freeze in less than one minute

FIG 1.6 EFFECT OF WIND ON TEMPERATURE
From Burton & Edholm (1955)
times for Second World War casualties, showed that these were closely related to sea temperature, as depicted in Figure 1.7. Since that time, major efforts have been made to find means of preventing or delaying body cooling in water, for example Golden (1976). Figure 1.8 shows that the open sea around Britain is cold, even in summer. Winter temperatures in estuaries, into which water has flowed from frozen land, can be as low as 3°C.

A continuing programme of experiments and refinements to methods of prediction is therefore justified in support of military and offshore oil operations.

1.4. Relative merits of modelling and experimentation

The foregoing sections have given a brief description of the range of thermal stresses to be found in military environments, particularly in the field of aviation, towards which the present investigation is directed. If military personnel are to survive and function effectively in such environments, adequate protection will be required. For relatively mild stresses, an appropriate level of clothing insulation will allow physiological thermoregulation to maintain a tolerable core and surface temperature. For severe stresses, however, this may not be possible and an external heating or cooling mechanism may be required. In either case, extensive knowledge of internal heat transfer, and its interaction with the environment, is required.

It cannot be denied that this knowledge could be obtained by experimentation alone. For many simple cases, experiments may be attractive when compared with the major task of developing a whole-body thermal model. For example, it is relatively easy to deduce experimentally the amount of clothing insulation required in order to maintain a core temperature high enough for survival in cold water over a given time. This essentially requires
FIG 1.7 SURVIVAL TIMES IN COLD WATER

From Molnar (1946)
immersion experiments to be carried out on a number of volunteer subjects appropriately dressed and with the core temperature (rectal, oesophageal or auditory) monitored. However, whilst the solution may at first sight appear attractive, there are a number of drawbacks to even this simple case.

In the first instance, it may be difficult to find a sufficient number of willing volunteers, since experiments of this kind are far from comfortable for the subjects. A small sample size from a limited range of body morphologies or ethnic characteristics may yield results that are misleading. Large samples of subjects lead to more general results but take proportionally more time to conduct. Most importantly, ethical considerations prevent the experimenter from allowing his subjects to approach the physiological limit for survival. This limit must instead be predicted from trends, i.e., from rates of fall of core temperature. A number of similar arguments can be applied to other types of investigation in thermal physiology, and it is partly the difficulties experienced by experimenters which have led to the development of mathematical models of the human thermal system. In addition, models may contribute to future understanding of the functioning of the human thermal system.

A theoretical model summarises the current understanding of a physical process. Although the life sciences are often less quantitative than the physical sciences, most models of thermoregulation are based on physical principles and therefore tend to be quantitative. In order to quantify the physiological processes so that the model can be constructed, it is necessary to draw on experimental data. After development, further data are needed to validate the performance of the model. Modelling cannot therefore replace experiments entirely, but once a validated model exists, the process of
investigation of a wide range of cases is greatly simplified. A number of comprehensive and well-validated models have appeared over the last 20 years. Relevant examples are discussed in Section 3 of this thesis. These models have been applied with varying success to many physical situations.

While the science of thermal modelling is mature and many major problems have been solved, knowledge of thermal physiology is continually improving, therefore providing fresh numerical data. New ways of expressing the thermal system mathematically and more powerful techniques of solution also need to be explored. Before embarking on a new phase of development, however, it is appropriate to determine whether existing models can be used to advantage. In some cases, the structure of a model may limit its prospects for further development. In others, lack of accuracy or of documentation may limit its usefulness. It is desirable that a model should impart understanding of the system it is designed to simulate. As an aid to achieving this aim in the new model, the physical and physiological elements of the human thermal system are reviewed in the following Chapter.
2. ELEMENTS OF THERMAL PHYSIOLOGY

2.1. Passive heat gains and losses

In the context of thermoregulation, the term 'passive' implies that there is no physiological control over the heat transfer mechanism by the autonomic nervous system. The primary means by which heat is generated in the body is from metabolic reactions.

The rate of heat generation varies considerably with different levels of muscular activity. Sedentary activity by an average sized man will generate about 100W and this is generally accepted as the resting or basal metabolic rate. Heavy industrial work can generate a continuous 300W, whilst athletes can attain peaks of 750W for short durations. Fan, Hsu and Hwang (1971) note that, if basal metabolic heat were not dissipated, the temperature of a 75 kg body would rise by approximately 1°C per hour. For the purpose of modelling, the basal rate can be treated as a passive heat gain, since it appears to have little variation except during the fever conditions described by Cooper (1972). Additional metabolic heat generated by muscular activity is best treated as a separate input since, as Cabanac (1972) shows, there is no predictable behavior pattern in man which can be related to a particular level of thermal stress.

The primary means of passive heat exchange between body tissues is by conduction. Conduction also takes place in layers of clothing and at the periphery heat may be exchanged with the environment by conduction, although this is not a major effect. The conductivity of clothing in which air is trapped is low. Clothing wetted by sweat or immersion will have little value as insulation, however.
The rate of conductive heat transfer is dependent on the temperature gradient, area of contact, tissue conductivity and thickness.

\[ Q_c = A \cdot k \cdot \frac{\Delta T}{x} \]  

(2.1)

where \( Q_c \) = rate of heat transfer (W), 
\( k \) = conductivity (W m\(^{-1}\) °C\(^{-1}\)), 
\( A \) = area (m\(^2\)), 
\( \Delta T \) = temperature (°C).

For core to skin conduction, for example, \( \Delta T = T_1 - T_2 \), 
\( T_1 \) = core temperature and \( T_2 \) = skin temperature. The direction of heatflow depends on the relationship between core and peripheral temperatures, ie on the sign of \( (T_1 - T_2) \).

Strictly, \( Q_c = A \cdot k \cdot \frac{dT}{dx} \)  

(2.2)

where \( \frac{dT}{dx} \) is the temperature gradient.

Heat is lost or gained at the skin surface or clothing surface primarily by convective exchange with surrounding air or by a combination of conduction and convection in water. Convection can take two forms, depending on whether the fluid is stationary or moving. If there is no relative motion between the body and surrounding fluid, convection takes place. Molecules in contact with a hot body surface are heated and form a boundary layer. The fluid becomes less dense near the surface and rises, to be replaced by cooler molecules. The reverse effect can occur for a cool body in a warm stationary fluid. Heat transferred in this way follows Langmuir's law (Kendall 1968) which has been determined experimentally for air as:

\[ Q_{cn} = h_{cn} \cdot A \cdot \Delta T^{5/4} \]  

(2.3)

where \( Q_{cn} \) = heat flow due to natural convection (W)
\[ h_{cn} = \text{natural convection coefficient (Wm}^{-2}\text{.C}^{-5/4}) \]
\[ A = \text{area of surface (m}^2) \]
\[ \Delta T = \text{temperature difference (°C)} = (T_1-T_2) \]

Normally, \( T_1 = \text{surface temperature, } T_2 = \text{fluid temperature.} \)

It is more usual for there to be some relative motion, however, and in this case forced convection takes place. For forced convection, Newton's law applies (Kendall 1968):

\[ Q_{cf} = h_{cf}.A(T_1-T_2) \quad (2.4) \]

where \( Q_{cf} = \text{heatflow due to forced convection (W)} \)
\[ h_{cf} = \text{forced convection coefficient (Wm}^{-2}\text{.C}^{-1}). \]

It has been discovered that, for a clothed or unclothed body in air, there is an approximate relationship between \( h_{cf} \) and air velocity, stated by Allan and Harrison (1978) as:

\[ h_{cf} = 8.3 \sqrt{v} \quad (2.5) \]

where \( v = \text{air velocity (ms}^{-1}). \)

Convection can account for 40% of the heat lost by a nude man in thermoneutral conditions, as estimated by Tamari & Leonard (1972).

The final passive heat exchange mechanism not involving mass transfer to the environment is radiation. All objects with temperatures above absolute zero radiate in the electromagnetic spectrum between the visible and far infra-red bands. Similarly, an object whose surroundings are above absolute zero will receive radiant heat from the surroundings. Radiant heat exchange takes place both at exposed skin and clothing surfaces. The amount of heat radiated or absorbed at a given temperature depends on the emissivity, \( e \). A perfect black body has an
emissivity of 1 and is a perfect emitter and absorber. Real surfaces radiate and absorb with e<1. Emissivity is related to surface roughness but colour of the surface in the visible spectrum is a poor indicator of emissivity in the infra-red region. Mitchell (1970) has shown that both black and white skin behave approximately as black body radiators. The rate of heat exchange with the environment is given by Stefan's law (Kendall 1968):

\[ Q_r = A \cdot e \cdot \sigma (T_1^4 - T_2^4) \quad (2.6) \]

where \( Q_r \) = heatflow due to radiation (W)
\( A \) = area exposed (m²)
\( e \) = emissivity (0 < e < 1)
\( \sigma \) = Stefan's constant = \( 5.7 \times 10^8 \) (Wm⁻² K⁻⁴)
\( T \) = absolute temperature (K)

\( T_1 \) is the surface temperature, \( T_2 \) is the mean radiant temperature.

The effective radiant temperature of the surroundings can vary widely, so that heat gains or losses are possible. Effective radiant temperatures are not necessarily related to ambient temperatures. For example, solar radiation is considerable on snow-covered mountains, where the air temperature is low. Radiation can account for 50% of the heat lost by a nude man in thermo-neutral conditions.

Three mechanisms of passive heat exchange by mass transfer are diffusion of water vapour through the skin, dry respiration heat loss and latent heat loss in respiration. Sweating also involves mass transfer but is under active control by the thermoregulation system. The amount of water vapour diffusing through the skin is normally assumed to be proportional to the difference between saturated vapour pressure at skin temperature and partial pressure of water vapour in the ambient air. The diffusion resistance of normal clothing can be assumed to be small, compared with skin resistance, except in
impervious clothing. An empirical formula for heat loss by skin diffusion is given by Fanger (1970) as follows:

\[ Q_d = 0.41A (1.92 T_S - 25.3 - P_w/7.5) \] (W) \hspace{1cm} (2.7)

where
- \( Q_d \) = heat loss by diffusion (W),
- \( A \) = surface area of skin (\( m^2 \)),
- \( T_S \) = skin temperature (°C),
- \( P_w \) = partial pressure of water vapour (kPa).

In respiration, heat is exchanged with the body owing to the temperature difference between inspired air and the respiratory tract. This so-called 'dry respiration loss' is only a few percent of that due to other mechanisms, hence it is often valid to assume that the temperature of expired air is 34°C according to Fanger (1970), who gives the following expression:

\[ Q_{xd} = 0.00163 Q_m (34 - T_a) \] (W) \hspace{1cm} (2.8)

where
- \( Q_{xd} \) = heat exchanged by dry respiration (W),
- \( Q_m \) = metabolic rate (W),
- \( T_a \) = ambient temperature (°C).

The need to calculate the humidity ratio in expired air may thus be avoided.

A further respiratory heat loss exists, due to evaporation of water from the mucous surface of the respiratory tract. Expired air is normally close to the deep body temperature and is always saturated with water vapour. Again according to Fanger (1970), a simple expression is available to describe latent heat loss in respiration:

\[ Q_{xw} = 0.0027 Q_m (44 - P_{ws}/7.5) \] (W) \hspace{1cm} (2.9)

where
- \( Q_{xw} \) = exchanged by latent respiration (W),
2.2. The active control system

The primary purpose of the effector mechanisms in thermoregulation is to minimise temperature variations of the vital organs in the core of the body. This purpose is accomplished by four distinct means, namely cardiovascular (bloodflow), sudomotor (sweating), metabolic (shivering) and to some extent respiratory changes. The skin of the body plays a large part in sensing and controlling temperature. In this context, the term 'skin' includes structures such as surface blood vessels, sweat glands and nerve endings.

The rate of bloodflow in surface vessels affects the temperature of the skin. With minimum blood flow, the surface temperature will be close to that of the environment, while with maximum flow the temperature will be close to body core temperature. Blood flow is adjusted by variations in vasomotor tone. An increase in tone causes vasoconstriction and a reduction in bloodflow, while a reduction in tone causes vasodilatation and hence an increase in flow. The segments of artery possessing muscular control are termed arterioles. These can act directly in series with capillary beds or in parallel shunts which open and close to divert bloodflow, termed 'arteriovascular anastomoses' as described by Burton (1972).

In a whole body model, the physical details of vasomotor action may be represented by relating changes in skin blood flow to changes in skin and core temperature. In a given area of skin blood flow can exhibit a range of 100:1. In a man exercising in hot conditions it can reach 1000 ml min\(^{-1}\) for each kg of skin, while in cold
conditions it can fall to less than 10 ml min\(^{-1}\). Since heat loss from the skin depends on skin temperature, which is controlled by bloodflow, vasomotor action is a very effective means of control. Edholm (1978) states that, owing to changes in blood flow, a range of 6:1 can be observed in heat loss by radiation and convection.

An empirical formula for control of blood flow is given by Stolwijk and Hardy (1977):

\[
\frac{w_i = w_{bi} + K_1(117.0E_c + 7.5\sum E_{si})}{1 + K_2(-5.0E_c - 5.0\sum E_{si})} \cdot 2^{E_{si}/6} \tag{2.10}
\]

where

- \(w_i\) = blood flow in the skin of body segment \(i\) (litre hr\(^{-1}\))
- \(w_{bi}\) = basal value of blood flow (litre hr\(^{-1}\))
- \(K_1\) = a constant for vasodilation intensity, which varies between segments
- \(K_2\) = a constant for vasoconstriction intensity, which also varies between segments
- \(E_c\) = the error between core temperature and its 'set point' value
- \(\sum E_{si}\) = a weighted sum of errors in skin temperature, accounting for variation in thermoreceptor density.

Vasomotor action also occurs in muscles in response to the demand for oxygen and nutrients during exercise or shivering. No relationship appears to have been established between muscle blood flow and temperature.

Sudomotor action is the term given to the process of secreting sweat by the sweat glands. The distribution of sweat glands over the body is not uniform. They are concentrated more on the extremities than on the limbs or trunk. Concentrations vary between 1 to 6 glands mm\(^{-2}\), making a total of more than \(2 \times 10^6\) glands. The glands are situated in the deeper layers of the skin and sweat.
reaches the surface via ducts. There, it evaporates and extracts an amount of latent heat proportional to the sweating rate, skin temperature and environmental temperature. There has been considerable debate on the extent to which sweating rate depends upon local skin temperature, core temperature and their rates of change.

There are two types of sweat gland; eccrine glands which are concerned with thermoregulation, and apocrine glands whose effect is trivial. Eccrine glands produce sweat, which is 99% water, in quantities which depend on the impulse rate received from the autonomic nervous system. The blood supply to the glands is more than adequate and there is no evidence to suggest that sweat gland fatigue ever occurs. However, an effect known as hidromeiosis reduces the sweating rate when the skin is wetted, as can occur at high exercise levels in hot, humid environments. For a given sector of the body, the heat lost by evaporation at the skin surface can be calculated from the following formula, also by Stolwijk and Hardy (1977):

\[
Q_e = [Q_{eb} + K_3(33.7E_c + 372.0E_s)] \cdot 2E_s^{10}
\]

where \(Q_e\) = heat loss from evaporation (W),
\(Q_{eb}\) = basal evaporative heat loss (W),
\(K_3\) = constant for sweating intensity which varies between sectors.

The apocrine glands are concentrated mainly in the armpits and are not innervated by thermoregulatory nerves.

The physiological effects described so far contribute to heat gains and losses at the body surface. In addition, heat can be produced within the body to compensate for losses by raising the metabolic rate. This is done by causing the muscles to work and hence consume oxygen. Shivering is a specialised form of muscular activity.
which is evoked by cooling the body. Rather than contracting and relaxing to produce a smooth movement, groups of muscle fibres act out of phase with one another. The result is an increase in metabolic heat production which may reach five times the resting rate, i.e. 500 watts approximately, for short periods. However, shivering usually occurs in bursts and the average increase is usually from two to three times the resting rate.

Shivering is not a particularly efficient thermoregulatory mechanism. Because the same arteries supply both skin and underlying muscle, some of the heat produced is immediately lost through the skin. Furthermore, the random muscular movements tend to destroy the boundary layer of warmer air or water which builds up around a stationary body. However, it appears to be the only major source of heat production not under voluntary control in human adults. Sowood (1984) points out that the core and skin temperatures at which shivering begins and the maximum metabolic increase appear to be affected by body composition, fitness and cold acclimatisation.

Shivering normally continues until a state of exhaustion is reached, after which core temperature may drop rapidly. There is little agreement among physiologists as to the form of temperature signal which stimulates a shivering response. Hayward, Eckerson and Collis (1977) suggest that the total metabolic rate during shivering may be described by the empirical relationship:

\[ Q_s = 0.0314 \cdot M \cdot (T_s - 42.2) \cdot (T_r - 41.4) \tag{2.12} \]

where \( Q_s \) = metabolic rate (W)
\( M \) = mass of body (kg)
\( T_s \) = mean skin temperature (°C)
\( T_r \) = rectal temperature (°C),
However, this relationship does not provide an accurate prediction in small individuals or during the transient phase just after immersion in cold water. Alternative relationships may be appropriate in some situations, as described in later sections of the thesis.

There remain two thermoregulatory mechanisms which, in human beings, are of minor importance: nonshivering thermogenesis and respiration. Non-shivering thermogenesis (NST) is the production of heat by chemical reactions in muscle, liver, gastrointestinal tract and brown fat. It appears to occur in most mammals. In man, however, NST is thought to be insignificant except during the first year of life, before the thermoregulatory system is fully developed (Hardy 1979).

2.3. **Energy Balance**

The purpose of the thermoregulatory system is to maintain a stable internal body temperature. The implication of a stable temperature is that the energy created within the body and received by the body is balanced by the energy lost. If this condition exists the rate of storage of energy, and therefore the rate of change of temperature, will be zero. Using the energy exchange mechanisms described above and adding terms to describe work and heat storage, an energy balance equation for the whole body may be developed as follows:

\[
Q_m + Q_w - Q_e - Q_d - Q_{wx} + Q_{xd} + Q_{cn} \text{ (or } Q_{cf} \text{)} + Q_r + Q_c
\]

\[
= M \cdot c \cdot dT/dt \quad (= 0 \text{ for energy balance})
\]

\[Q_m = \text{metabolic rate, including basal metabolism, shivering and exercise (W)}\]

\[Q_w = \text{external work done by the body (negative) or done on the body (positive) (W)}\]

\[Q_e = \text{heat loss due to evaporation (W)}\]
\( Q_d \) = heat loss rate due to diffusion (W)
\( Q_{xw} \) = heat loss rate due to evaporation in respiration (W)
\( Q_{xd} \) = dry heat exchange rate in respiration (positive in hot environments) (W)
\( Q_{cn} \) = heat exchange rate with environment due to natural convection (for a static body) (W)
\( Q_{cf} \) = heat exchange rate by forced convection (moving body or fluid) (W)
\( Q_r \) = radiant heat exchange rate (W)
\( Q_c \) = conduction heat exchange rate (W)
\( M \) = mass of body (kg)
\( c \) = mean specific heat of body (J kg\(^{-1}\)°C\(^{-1}\))
\( \frac{dT}{dt} \) = rate of change of mean body temp. (°C s\(^{-1}\))

Figure 2.1 shows pictorially the terms in Equation 2.13. The equation is a statement of the first law of thermodynamics which is fundamental to all thermoregulation. Given sufficient time, a body generating heat at a fixed rate in a fixed environment will reach thermal equilibrium and therefore satisfy the condition \( M.c.dT/dt = 0 \). However, equilibrium rarely exists for very long, especially during exercise in hot conditions or during immersion in cold water. It is therefore necessary to apply the physical laws of heat exchange to each variable separately in Equation 2.13 and construct a dynamic model of temperature changes with time.

Relationships similar to Equation 2.13 can be written for individual body segments. Not all the terms will necessarily be applicable to each segment. It is further necessary to add terms describing the blood flow to and from each segment, and to construct additional energy balance equations from which blood temperatures are derived. From this basis, the development of a dynamic model may proceed.
FIG 2.1 HEAT FLOW MECHANISMS
3. ASSESSMENT OF EXISTING THERMOREGULATION MODELS

3.1. Brief History

The subject of human thermal modelling is one of considerable complexity and spans quite wide areas of biology, medicine, psychology and engineering. Earlier reviews of work in the field have taken a number of different approaches, from a philosophical overview by Hardy (1972) to a detailed exposition of mathematical techniques by Fan, Hsu and Hwang (1971) and later by Hwang and Konz (1977). A number of comprehensive reviews have been published over the years dealing with the physiological mechanisms, for example Mitchell (1977), Mitchell, Atkins and Wyndham (1972). More recently, evaluations of existing models have been made, either for a specific purpose such as coastguard operations, (Baker, Harnett and Ringuest (1979)) or 'to define the state of knowledge' (Wissler (1984)). The most recent review available, by Haslam and Parsons (1987), evaluates four air-exposure models.

It would be inappropriate, in the present thesis, to repeat these detailed reviews of the topic. Nevertheless, in justifying the development of a further model, it is necessary to trace the development of techniques through some of the earlier examples. The following brief review attempts to identify those techniques likely to be successful, while rejecting those which have little to offer.

Figure 3.1 illustrates the adopted scheme of classification of models. The first apparent division is between mathematical and physical models. Physical models have occasionally been constructed but difficulties in construction have prevented their widespread adoption. An early example is a water-in-glass model of the hand, constructed by Aschoff
Models of the Human Thermal System

- Mathematical
- Physical

- Steady state
- Dynamic

- Theoretical
- Empirical

- Single region or mechanism
- Whole body

- Single segment
- Multi segment

- Lumped temperature
- Finite element

FIG 3.1 CLASSIFICATION OF MODELS
in 1958. This was used to study counter-current heat exchange between veins and arteries.

Another example is the electrical circuit proposed by MacDonald and Wyndham in 1950 and improved by Cornew et al in 1967. This network simulated whole body thermal behaviour, using electric current as an analogue of heat flow, voltage as temperature, capacitance as heat storage and resistance as insulation. Although the approach may have led to later analogue computer models, inflexibility after construction limits its usefulness.

It is desirable to choose an information-based rather than physically-based means of representing the problem. A computer is therefore used to solve a set of equations repetitively, providing a prediction of changes in the variables of interest over a period of time. Some, though not all, of the earliest attempts employed analogue computers. One of the most successful was developed by Smith and James in 1964. The model gave adequate agreement with experiments particularly for cardiac output during work/rest cycles at normal ambient temperatures.

The main advantage of an analogue computer is its ability to perform many integrations in parallel, rather than sequentially by numerical methods as on a digital computer. It is therefore possible to run problems in real time and gain good visualisation of controls and effects. Problems occur, however, with accuracy due both to the electronic signal-to-noise ratio and to amplifier drift with time. In human thermoregulation models, it is necessary to achieve an accuracy of about ± 0.01°C, since temperature gradients can be small.

The last analogue model for which details are available was reported by Behling et al in 1971. The model was based on a simple two-compartment cylindrical geometry.
It predicted steady state oesophageal temperatures with sufficient accuracy but was not successful for high workload cases. This effect was thought by the authors to be due to the use of linear functions to represent metabolic rate-dependencies which are in fact non-linear. Non-linear functions can be generated on analogue computers, using diode function generators to obtain a piecewise approximation. However, the numerical accuracy of digital computers, and their ability to generate non-linear functions by algorithms or look-up tables, makes their use preferable.

The remainder of this review will concentrate on models implemented in software.

3.2. Steady state models

An important conceptual division occurs between steady state models and dynamic models. Steady state models attempt to predict the spatial temperature distribution in the body or a segment of it. They apply to situations where a heat balance exists, so that all temperatures have a zero rate of change with time. Dynamic models, on the other hand, attempt to predict the time history of temperatures.

One of the earliest thermal models reported was a steady-state model, due to Pennes in 1948, which related temperature in the human forearm to local metabolic rate and bloodflow. The forearm was approximated as a cylinder with heat loss at the surface by convection, radiation and evaporation. The temperature profiles were validated experimentally by inserting a needle thermocouple into the arm of a volunteer subject.

Although developed for the forearm, the cylindrical visualisation is equally applicable to other elements of the body. The technique was adopted by Wissler (1961) in
his first steady state model of the whole body. Geometrically, this model consisted of six cylindrical segments representing the head, trunk, arms and legs, connected by blood circulation. Venous and arterial blood temperatures were assumed to be uniform within each segment. Heat exchange between veins and arteries was not simulated.

More recently, two papers by Smith and Twizell (1980, 1982) have proposed a technique for determining the temperature profile within a body segment, using finite element analysis. The body segment under consideration is represented by a cylinder divided into four annular regions denoting core, muscle, fat and skin. The annular regions are triangulated into an arbitrary number of elements, each representing a region of uniform temperature. Because each triangle is small, a close approximation to the surface boundary is obtained, in the form of a many-sided polygon. An accurate representation of the temperature profile can be computed by solving a partial differential equation for heat flow from the axis to the surface of the cylinder.

The method could be extended to predict whole body temperature distributions by including further cylinders, but computing time would increase considerably. The advantage of finite element analysis is that the model can be made effectively two-dimensional. This can be useful where heat loads are uneven, for example when solar radiation falls on one side of the segment. However, although a steady state model can be made to represent the active elements of the thermal system as well as the passive heat exchanges, it can only do so for equilibrium conditions, or for a single instant in time under dynamic conditions.

One further class of part-body or single-mechanism models can be identified, not falling within the main sequence.
of whole-body digital computer models. It is sometimes desirable to predict heat and blood flow patterns dynamically in a single body segment, for reasons of cold injury, surgery or thermal treatment. Data on models with this capacity are sparse, since very few concentrate solely on one part of the body. Although Fan, Hsu and Hwang (1971) reviewed four models which they classed as 'single element of the body, non-steady state', they were in fact referring to single cylinder representations of the whole body.

The only part-body dynamic models for which details have become available are reported by Mitchell et al (1970) and by Hsia (1975). The Mitchell model was developed to predict temperatures as functions of time in the human leg during cooling for anaesthesia prior to amputation. It was based on the more general models by Perl (1965) and Wissler (1963), using the same heat flow mechanisms.

Hsia visualised a single limb as an arbitrary number of cylindrical segments, divided radially into two annular regions, representing muscle and bone together and fat and skin together. He derived separate equations for the circulatory system and the thermal system within each segment. Only radial heat flow was considered. A novel representation of arterial resistance was devised, using overlapping exponential functions for vasoconstriction and vasodilatation. The transient response of skin perfusion was also accounted for in a third-order differential equation. Even without accounting for metabolic heat production in muscle, excellent agreement was obtained with skin temperatures observed during limb cooling experiments. The model is valuable, in that it indicates the importance of modelling bloodflow accurately and suggests that a complex representation of radial heatflow is not necessary.
A number of sub-models have also been developed, often as simple written formulae, in order to predict the performance of a particular feature of the thermo-regulation system. Among these are the models for sweating developed by Timbal et al (1969) and by Ibamoto (1971). The Timbal model provided a formula for weight loss with time, as a function of rates of diffusion through skin, respiratory perspiration and sweating. It was suggested that weight loss could be related to time by a first order equation of the form:

\[
\Delta m = -(m_2 - m_1) (\tau \exp(-t/\tau) - 1) + t \quad (3.1)
\]

where \( \Delta m \) = total weight loss of fluid (mg),

\( m_1 \) = rate of diffusion and respiration loss (mg s\(^{-1}\)),

\( m_2 \) = rate of weight loss due to sweating (mg s\(^{-1}\)),

\( \tau \) = first order time constant (s),

\( t \) = time from start of exposure (s).

The formula was verified experimentally with an error of 2.5%, where the actual values of parameters varied from subject to subject. Typically, \( m_1 = 16 \text{ mg s}^{-1} \), \( m_2 = 130 \text{ mg s}^{-1} \) and \( \tau = 660 \text{s} \) for nude subjects in exposures between 35 and 45°C.

By contrast, Ibamoto (1971) developed a theoretical relationship for skin heat loss under layers of clothing, taking account of the wetted area of skin. This takes the form:

\[
W_s = (T_s - T_a) + 2.6 f_p h_c (X_w - X_a) A_w \quad (3.2)
\]

\[
\frac{(I_c + I_a)}{A_a}
\]

where \( W_s \) = total skin heat flow density (W m\(^{-2}\)),

\( T_s \) = skin temperature (°C),

\( T_a \) = ambient temperature (°C),

\( I_c \) = clothing insulation value (m\(^2\cdot\text{°C}^{-1}\cdot\text{W}^{-1})\),
\[
I_a = \text{insulation from clothing surface to ambient temp (m}^2\text{oCw}^{-1}),
\]
\[
f_p = \text{water vapour permeability efficiency of clothing},
\]
\[
h_c = \text{convective heat transfer coefficient (Wm}^{-2}\text{oC}^{-1})
\]
\[
X_w = \text{absolute humidity of wet skin area},
\]
\[
X_a = \text{absolute ambient humidity},
\]
\[
A_w = \text{wetted skin area (m}^2\text{)},
\]
\[
A_a = \text{total skin area (m}^2\text{)}. 
\]

Such relationships, although they do not, in themselves, constitute complete mathematical models, can be incorporated in larger models. Other analytical approaches have led to a dynamic model for bloodflow in working muscle by Mitchell, Stolwijk and Nadel (1972) and a quantitative prediction of heat exchange between veins and arteries by Mitchell and Myers (1968). A mathematical sub-model for shivering thermogenesis has recently been developed by Mekjavic and Morrison (1984). This uses least-squares regression to fit parameters to empirical data from experimental immersions.

To bring to a close the topic of sub-models, the work of Kitney (1975, 1979) on the vasomotor system is worthy of note for its mathematical sophistication. A digital simulation has been developed using z-transform techniques, in order to investigate the interaction between the respiratory and blood pressure control systems. It has been demonstrated that the periodic cold-induced dilatations and constrictions of the peripheral blood vessels can be explained by a non-linear model of the system controlling them.

3.3. Empirical methods

A further fundamental division in approaches to human thermal modelling occurs between empirical and
theoretical models. All models are to some extent theoretical, since they must start from the first law of thermodynamics, as expressed in Equation 2.13. However, some do attempt to include physiological phenomena descriptively by using relationships derived from experimental data. These are empirical models. Others use physical representations of the body, based on various solid geometries, and attempt to calculate each variable from heatflow equations and control theory.

These are theoretical models. Having defined the theoretical model, it is necessary to point out that all models, whilst representing passive heatflow theoretically, must contain empirical statements in the sections representing the active control system. This is so, since it is not possible to measure directly each interaction of the nervous system and organs producing sweating, shivering and vasomotor action, but only to observe the global effect of a known stimulus.

An empirical model, if one can be made to describe a particular situation accurately, offers the advantage of simplicity. In a few cases, empirical models have been developed which will run on a hand-held programmable calculator. The main disadvantage is that they apply only to small ranges of body morphology and thermal conditions, outside which inaccuracies quickly develop. If care is taken in the development of physiological and geometrical relationships, a theoretical model can offer a description of human thermal behaviour in a wide variety of conditions.

An interesting illustration is provided by an early empirical model, which was developed before computers became widely available for this type of task. It was developed by Machle and Hatch in (1947) and employed a variation of Equation 2.13 to represent whole-body heat exchange:
\[ Q_m + m \cdot c \cdot \frac{dT}{dt} - Q_x = Q_r + Q_{cf} + Q_v \quad (3.3) \]

where

- \( Q_x \) = total respiratory heat loss (W) = \( Q_{xw} + Q_{xd} \)
- \( Q_v \) = total evaporative heat loss (W) = \( Q_d + Q_e \)

Radiation and convection were calculated theoretically but for evaporation an empirical straight-line relationship of the form:

\[ P_s = aT_s + b \quad (3.4) \]

was assumed,

where

- \( P_s \) = water vapour pressure at skin (\( P_a \)),
- \( T_s \) = skin temperature (°C),
- \( a \) and \( b \) are constants.

Furthermore, it was assumed that the rate of heat storage could be expressed as a linear combination of rectal and skin temperature rates:

\[ m \cdot c \cdot \frac{dT}{dt} = m \cdot c (p \cdot \frac{dT_r}{dt} + q \cdot \frac{dT_s}{dt}) \quad (3.5) \]

where \( p \) and \( q \) are constants.

The most significant empirical assumption in the model, however, was that rectal and skin temperatures were mutually dependent in a way which was linear over the range of interest:

\[ T_r = x + yT_s \quad (3.6) \]

where \( x \) and \( y \) are constants.

This, and further empirically-derived proportionality constants for radiation, convection and evaporation, enabled the authors to develop a simple exponential relationship for a situation in which metabolic rate and
respiration losses were constant:

\[ T_{eq} - T_s = (T_{eq} - T_0)\exp\left(-\frac{AK}{m c(p + q)}t\right) \]  

(3.7)

where \( T_{eq} = \) equilibrium skin temperature (°C),
\( T_0 = \) initial skin temperature (°C),
\( K = \) a constant derived from proportionality constants for convection, radiation and evaporation.

The model was useful in only a limited range of circumstances, owing to difficulties in determining skin wetness and hence the area over which evaporation could occur. Kerslake and Waddell (1958) later modified the model to include complete skin wetness. This early example serves to illustrate the simplicity but highly specific application, often associated with empirical models.

The principal model in the highly empirical category is the heat model of Givoni and Goldman (1972). This model computes rectal temperatures as a function of metabolic rate, for various combinations of work, environmental conditions and clothing. It was formulated at a time when no other model in existence predicted the effects of clothing. The equilibrium rectal temperature is assumed to be determined by three factors: the rate of metabolic heat production, environmental heat exchange rate and the maximum evaporative capacity of the environment. The general form of the empirical equation predicting the final equilibrium temperature is as follows:

\[ T_{re(f)} = 36.75 + 0.004(Q_m - Q_w) + 0.025/C (T_a - 36) \]
\[ + 0.8 \exp 0.0047(Q_{er} - Q_{em}) \]  

(3.8)

where \( T_{re(f)} = \) final equilibrium rectal temperature (°C),
\( Q_m = \) rate of metabolic heat production (W),
\( Q_w = \) rate of heat exchange with environment (W),
\( Q_{er} = \) rate of evaporative heat loss (W),
\( Q_{em} = \) rate of evaporative heat gain (W),
\( C = \) specific heat capacity of body, \( m \) and \( p \) are proportional constants, and \( q \) is a proportionality constant for body mass and heat capacity.
\( Q_w = \) rate of external work performed (W),
\( T_a = \) ambient temperature (°C),
\( Q_{er} = \) required rate of evaporative heat loss (W),
\( Q_{em} = \) maximum available evaporative heat loss (W),
\( C = \) clothing insulation value (clo).

In practice, it may be found that the available capacity for evaporation is not sufficient to maintain equilibrium during exercise. This is very likely to be true under water vapour-impervious clothing where vapour pressures can be so high that condensation may even take place on the skin. The metabolic rate \( Q_m \) has been evaluated for walking on a gradient as:

\[
Q_m = M\{ (2.7+3.2(v-0.7)^{1.65}) + G[0.23+0.29(v-0.7)] \}
\]  \( (3.9) \)

where \( M = \) body mass incl. clothing (kg),
\( v = \) walking velocity (ms\(^{-1}\)),
\( G = \) gradient in %.

The external work \( Q_w \) performed in walking is taken as

\[
Q_w = 0.098 MvG \quad (3.10)
\]

The required heat loss by evaporation, \( Q_{er} \) is simply the sum of radiant and convective heat inputs, plus metabolic heat generated, minus external work. The maximum available evaporative heat loss is determined empirically for a man with 1.8m\(^2\) surface area as:

\[
Q_{em} = 25.5 \frac{I}{C} (44-P_a) \quad (3.11)
\]

where \( I = \) permeability index of clothing
\( P_a = \) water vapour pressure in atmosphere (mm Hg).
The time history of temperature $T_{re}(t)$ from an initial value $T_{re}(0)$ and the final value in Equation 3.8 depend upon whether the subject is resting or working. For a resting subject there is an initial half-hour time lag and the transition is described by an exponential function:

$$T_{re}(t) = T_{re}(0) + [T_{re}(f) - T_{re}(0)] 0.1^u \quad (3.12)$$

where $u = 0.4(t-0.5)$ \quad (3.13)

and $t$ is the time in hours from the initial state.

For a sudden onset of work, the initial time lag is inversely proportional to the metabolic rate and the relationship takes a different form:

$$T_{re}(t) = T_{re}(0) + (T_{re}(f) - T_{re}(0)) \cdot (1 - \exp[-(2.0 - 0.5 \sqrt(T_{re}(f) - T_{re}(0)))]. \quad (t - 58/Q_m)) \quad (3.14)$$

Equation 3.14 only accounts for the response of rectal temperature during transitions from rest to work. On ceasing work, a different relationship applies, in which the initial time delay and exponential time constant are both dependent on a quantity referred to as 'effective cooling power'. This quantity combines the effects of convection, radiation and evaporation and is expressed as:

$$W = 0.27I/C (44-P_a) + 0.174/C (36-T_a) -1.57 \quad (3.15)$$

The time constant then becomes

$$a = 1.5[1 - \exp(-1.5W)] \quad (3.16)$$

and the delay
\[ t_d = 0.25 \exp(-0.5W). \] (3.17)

The final work to rest temperature profile is then:

\[ T_{re}(t) = T_{re}(0)-[T_{re}(0)-T_{re}(f)]\{1-\exp[-a(t-t_d)]\} \] (3.18)

The model has been stated in some detail, since it shows how useful relationships can be derived for metabolic and heat transfer rates for various conditions. It has been well validated by the authors for various metabolic rates up to 470 W and ambient temperature up to 35°C, using groups of subjects not included in the empirical derivations.

Further assessment occurred during a working party held at the University of Texas in 1982, when it was found that, allowing for individual variations, the model performed well on seven sets of constant temperature data and four sets of step changes. Inputs required are the subject's body mass and surface area, rate of work and number of days of acclimatisation. Wet and dry bulb temperatures, air velocity and clothing insulation values are also required. The computed variable is rectal temperature and the model gives little in the way of physical visualisation of the processes. In other words, it behaves as a 'black box'. However, it does have a short execution time and requires only a small computer.

3.4. Simple dynamic models

The majority of mathematical models of thermoregulation contain an embedded geometrical structure which, to a greater or lesser degree, represents the body being simulated. The purposes of the structure are firstly to provide a basis for formulating passive heatflow equations and secondly to visualise the regions in which active thermoregulatory effects take place. The degree of complexity in the geometry tends to be associated with
the purpose of the model, particularly the number of sites at which it is desired to predict temperature. It is theoretically possible to derive heatflow equations for a body represented by a very large number of infinitesimal elements. The finite element analysis of Smith and Twizell (1980, 1982) approximates to this approach but has mainly been applied to the steady state. In practice, in order to limit complexity, a 'lumped parameter' approach is normally adopted, in which the body is divided into a number of segments and layers. Each segment normally represents an identifiable part of anatomy (torso, legs etc) whilst layers are normally associated with tissue types (muscle, fat, skin).

The simplest geometry of all is that of the single segment, representing the whole body. This has been utilised where it is desired to predict changes in mean core temperature and mean skin temperature. In this category, an early attempt was made by Crosbie et al in 1963 to represent the body by a three-layered infinite slab. The three layers, shown in Figure 3.2, represent core tissues, muscle and skin. Fat was not represented separately. Temperatures were assumed to be uniform within each layer, with discontinuities at the boundaries. A mean weighted body temperature was calculated from the formula:

\[ T_b = \frac{X_s T_s + X_m T_m + X_c T_c}{X_s + X_m + X_c} \]  

(3.19)

where \( T_s, T_m \) and \( T_c \) are skin, muscle and core temperatures (°C)

\( X_s, X_m \) and \( X_c \) are skin, muscle and core thicknesses (mm)

A heat balance equation was written for each layer, including metabolic reactions, evaporation and blood perfusion rate changes where appropriate. Metabolic rate was assumed to be a function of \( T_b \) for negative heat loads. Evaporation was similarly a function of \( T_b \) for
FIG 3.2 INFINITE SLAB MODEL GEOMETRY

From Crosbie et al (1963)
positive heat loads. A tissue conductivity $K$ was specified to represent blood perfusion rate, and was a function of both $T_b$ and $dT_b/dt$. An analogue computer was used in the simulation. Agreement to within $1^\circ C$ for core and mean skin temperature transient responses was achieved in a comparison with experiments using nude human subjects.

An alternative geometrical visualisation is that of a cylinder with several annular layers, providing a close analogue of tissue layers in the body. This has been adopted by Atkins and Wyndham (1969) and Gagge (1973) for heat stress, and by Timbal et al (1976) for cold water immersion. The fundamental heat flow equation for radial conduction in unit volume of a cylinder is:

$$\rho c \frac{\delta T}{\delta t} = \frac{1}{r} \left[ \frac{\delta}{\delta r} (K r \frac{\delta T}{\delta r}) \right] \left( \text{W}_m^{-3} \right) \quad (3.20)$$

where

- $\rho$ = density of tissue ($\text{kgm}^{-3}$),
- $c$ = specific heat of tissue ($\text{J kg}^{-1} \text{C}^{-1}$),
- $T$ = temperature ($\text{C}$),
- $r$ = radius ($\text{m}$),
- $K$ = conductivity of tissue ($\text{W}_m^{-2} \text{C}^{-1}$).

To this must be added terms for blood flow, metabolism and surface heat transfer. In models employing a cylindrical geometry, it is generally assumed that there is no axial heat flow - a good approximation in practice. This is directly equivalent to the semi-infinite slab approach and both are essentially one-dimensional in heat flow terms. Figure 3.3 shows a generalised version of a cylindrical segment in cross section. This geometry has subsequently been adopted for general use in single and multisegment models.

The Atkins model had four concentric layers, representing core (viscera and skeleton), muscle, subcutaneous fat and skin. The fat layer was assumed to be avascular.
FIG 3.3 SINGLE CYLINDER MODEL GEOMETRY
(bloodless). An independent central blood pool was postulated, which collected returned blood in the heart/lung region and took account of mixing before redistribution to the periphery. While radiation, convection and evaporation were modelled in a more or less theoretical way, it was not considered necessary to model explicitly the mechanisms of the controlling system. Instead, these were represented by empirically-derived relationships between skin and core temperature, in many ways similar to those of Givoni and Goldman (1972), which they preceded. Comparing simulation results with core and skin temperatures observed during exposures to air at 13, 24, 35, 41 and 49°C, agreement is generally within a few percent. In particular, the initial overshoot of skin temperature before sweating occurs is predicted for heat exposure. So also is the continuing fall in core temperature in 13°C air after 2 hours. The control equations were highly non-linear and an analogue computer was used for their solution.

The model by Gagge (1973) is geometrically even simpler, consisting of a single cylinder with only two concentric layers. Metabolic rate and external work performed are independent variables. Another variable accounts for the combined effects of surface radiation and convection. Rather than varying conductance of the surface layer to represent blood flow variations, as in the Crosbie model, the heat transferred between layers depends specifically on a blood flow rate. This rate is assumed to vary in sympathy with temperature signals from the core and shell:

\[
\begin{align*}
\Sigma_s &= T_s - 34.1 \\
\Sigma_c &= T_c - 36.6
\end{align*}
\]  

where \( \Sigma_s \) and \( \Sigma_c \) are dimensionless skin and core signals.
$T_s$ and $T_c$ are skin and core temperatures.

The values of 34.1 and 36.6 are significant, in that they can be considered as equilibrium points. The rate of sweat production also depends on the two temperature signals. This model apparently has limited use for heat exposures of less than 1 hour. It does, however, have the ability to utilise all the principal parameters associated with heat exposure in order to predict probable body temperature states.

The cold immersion model of Timbal et al (1976) was again a two-compartment cylinder. The value of heat transfer coefficient at the surface in water has been the subject of some debate, since it is affected by relative motion, convection and shivering and, in the sea, by wave motion. The following values were used in this model for still water:

\[
\begin{align*}
    h_c &= 43 \text{ Wm}^{-2} \quad \text{for } T_c > 28^\circ\text{C} \\
    h_c &= 60 \text{ Wm}^{-2} \quad \text{for } T_c < 15^\circ\text{C}.
\end{align*}
\]

Between these limits there is a linear slope which is assumed to be dependent on shivering intensity. For moving water, the empirical relationship:

\[ h_c = 497.1 v^{0.65} \left( \text{Wm}^{-2} \right) \quad (3.23) \]

was used, where $V = \text{velocity of water (ms}^{-1}).$

Owing to the much higher specific heat of water, heat flows from the body at a higher rate than in air. The metabolic response is therefore also higher, ie there is a greater intensity of shivering. The Timbal model took account of this and also of the change in respiration rate with metabolism, which in turn caused a change in respiratory heat loss. Intensity of shivering determined the length of time for which it could be maintained in
the model, with the maximum of 5 times normal metabolic rate being maintained for only 90 minutes. Also, a reduction in intensity was put into effect when rectal temperature fell below 35°C.

One of the most important factors in determining survival time in cold water is the thickness of the layer of subcutaneous fat. The authors used this as an input to the model and predicted a set of curves of time to cool to a certain rectal temperature, versus water temperature. This was done for five average skinfold thicknesses from 4 to 20mm and four rectal temperatures from 20 to 35°C. The resulting curves are similar to those plotted by Molnar (1946) as in Figure 1.7. Naturally, an experimental validation of the model cannot be made at rectal temperatures below 35°C. However, above this figure, simulations were carried out for various morphological types, which gave agreement to within 0.2°C of experimental observations.

Finally, on single cylinder models, a three-layer model was evaluated by Baker et al (1979), having been proposed by Winton and Linebarger. Unfortunately, the evaluation report gives no source reference and does not compare simulation with experiment. The description given in the report indicates that the model is very similar to that of the Timbal model just described and apparently appeared at about the same time.

Single element models, whether they use cylindrical or slab geometries, are only capable of predicting mean skin and mean core temperatures. They do not allow for variations in fluid temperature or velocity over the body surface. Neither do they allow variations in external insulation, and are therefore only applicable to whole-body protective clothing. As they have no representation of body extremities, they can neither predict cold injury hazards nor take account of sites of
temperature receptors in the nervous system. These limitations suggest that, both for flexibility of application and for accuracy of simulation, a multi-segment model is the preferred approach.

3.5. Multi-segment, lumped-temperature models

The remaining models examined here form what may be viewed as the mainstream of recent development and possess a number of common characteristics. The effects of heat transfer by conduction and blood convection are evaluated by dividing the body into geometric regions and writing energy balance equations for each one. All multi-segment models have been heavily influenced by the 1966 paper by Stolwijk and Hardy. Stolwijk has used five cylindrical segments to represent the trunk, arms, legs, hands and feet, and a sphere to represent the head. Each segment is divided into four layers to represent core, muscle, fat and skin. Conductive heat transfer is calculated in terms of conductances between adjacent layers and their temperatures. The temperature in each layer is assumed to be homogeneous, as are the physical properties of the layer. Hence the model may be termed 'lumped variable' as well as 'lumped parameter', where the term 'variable' refers to a time-varying quantity such as temperature or flow rate and 'parameter' applies to conductivities, specific heats, etc.

The Stolwijk model represented the first attempt to formulate physiological thermoregulation equations using feedback control theory. It evolved under NASA sponsorship and is still in use in a form substantially the same as that published by Stolwijk and Hardy in 1977. A schematic representation of division of the body is shown in Figure 3.4 and a representation of the four layers of any segment (I), where I = 1 to 6, in Figure 3.5. The central blood pool is denoted as compartment number 25 and the compartment numbers (N) for other
FIG 3.4 GEOMETRY OF STOLWIJIK'S MODEL

From Stolwijk (1966)
FIG 3.5 STOLWIJK'S PARAMETERS FOR BODY SEGMENT (1)

From Stolwijk (1971)
layers are functions of I. The variables and parameters in the diagram are defined in the FORTRAN 66 program as follows:

\[
\begin{align*}
C(N) &= \text{thermal capacity of compartment } N \ (J \ °C^{-1}) \\
TC(N) &= \text{thermal conductance between } N \text{ and } N+1 \ (W °C^{-1}) \\
Q(N) &= \text{total metabolic heat production in } N \ (W) \\
BF(N) &= \text{total effective bloodflow to } N \ (l \ s^{-1}) \\
E(N) &= \text{total evaporative heat loss from } N \ (W) \\
H(I) &= \text{total environmental heat transfer coefficient for } I \ (W °C^{-1}).
\end{align*}
\]

The sensors in the thermoregulatory control system generate afferent signals of the form:

\[
\text{ERROR}(N) = T(N) - T\text{SET}(N) + \text{RATE}(N) \cdot F(N) \tag{3.24}
\]

where
\[
\begin{align*}
T(N) &= \text{temperature of compartment } N \ (°C) \\
T\text{SET}(N) &= \text{equilibrium or 'set point' temperature } (°C) \\
F(N) &= \text{rate of change of temperature } (°C \ s^{-1}) \\
\text{RATE}(N) &= \text{a controlling system gain term.}
\end{align*}
\]

In order to evaluate different controller concepts, the authors found it easier to assume that thermoreceptor structures were present in all tissues. The model primarily uses inputs from skin receptors, however, and distributes these in realistic densities over the surface by incorporating appropriate gain terms. The head core temperature is used to represent hypothalamic temperature error, ERROR(1).

If \( \text{ERROR}(41) \), the skin temperature change for element I, is positive, its value is taken on by a variable WARM. This is then weighted for density of skin receptors to form a signal which contains the sum of all skin receptor effects, called WARMS. A similar procedure is followed.
to derive COLD and COLDS from negative errors. The
subcontraction (WARMS - COLDS) is then a linear skin
error signal. These signals are used to derive the four
controller outputs SWEAT, DILAT, STRIC and CHILL which
represent respectively sweating, vasodilatation,
vasoconstriction and shivering. For example,

\[
\text{SWEAT} = \text{CSW.\text{ERROR}(I) + SSW.\text{(WARMS-COLDS) + PSW.WARM(I).WARMS}} \quad (3.25)
\]

where CSW, SSW and PSW are weighting constants for the
error signals. (In practice, PSW = 0, so the expression
remains linear.)

Similar expressions are stated for the other three
controller outputs. These outputs are then modified
peripherally according to local conditions. For example,
the sweating command is modified by an exponential
relationship which doubles the rate of sweat production
in a segment for every 10°C rise in its skin temperature.
Bloodflow in muscle is made proportional to the local
metabolic rate. Skin bloodflow is controlled by DILAT
and STRIC and is modified in a linear fashion by local
skin temperature. Vasodilatation is affected eight times
as much as vasoconstriction by core temperature. It is
therefore necessary to allow both controller outputs to
operate simultaneously. The shivering controller output
CHILL is derived from a product relationship for segment
I:

\[
\text{CHILL} = \text{PCHILL.COLD(I).COLDS} \quad (3.26)
\]

The gain term PCHILL is set to 21.0. The expression is
non-linear.

Having established thermoreceptor outputs, formed
weighted sums and determined controller outputs as
described above, the model calculates the heat flows in
each compartment from a standard heat balance equation. The new temperatures are then calculated by numerical integration. The step size is normally set to 1 minute but if any temperature change is more than 0.1°C, the step size is reduced and the calculation repeated.

Although the model contains shivering and vasoconstriction effects, it was primarily designed for heat stress and it is in this area that most of the validation has taken place. The 1971 report gives comparisons between simulations and experiments in which subjects were exposed to step changes in air temperature. Figure 3.6 shows that there is close correlation under certain conditions. Further examples are given by Stolwijk in a 1980 paper which describes a later version of the model.

An independent validation has been carried out by Hancock (1980) who compared results of work-rest cycles on a bicycle ergometer. Records of rectal temperature in Figure 3.7 show that despite a large initial discrepancy, the long term prediction is useful. Stolwijk has concluded that similar results in an earlier comparison occurred because the model supplies all the muscle instantly with the blood necessary to replace oxygen used in exercise.

The model was tested against experimental data from both hot and cold exposures at the workshop reported by Wissler (1984). Although results for exercise in 30°C air were good, computed central temperatures seemed to decrease more than they should when the ambient temperature was 10°C. Examination of Figure 3.5 shows that blood flows to peripheral layers from the central blood pool and presumably returns there directly, when it is at tissue temperature. Re-routing the blood in a more representative manner may yield more realistic temperatures.
FIG 3.6 VALIDATION OF STOLWIJK'S MODEL
Redrawn from Stolwijk (1971)
FIG 3.7 HANCOCK'S VALIDATION OF STOLWIJK'S MODEL
Redrawn from Hancock (1980)
The original model has been adapted for cold water immersion by Montgomery (1974). Instead of four layers in each segment, Montgomery used ten. This appears necessary because temperature gradients are much larger for water immersion than in air and the lumped approach gives poor definition, which may lead to errors. The coefficients of the physiological control equations were also changed, and this led to agreement within 0.4°C between measured and computed auditory canal temperatures in a relatively mild 28°C immersion. Rectal temperatures agreed more closely, to within 0.2°C. By contrast, an immersion model by Miller and Seagrave (1974), published in the same year and using only four layers, often showed discrepancies of 1°C for similar ambient temperatures. Too coarse a definition of gradients is therefore to be avoided, despite the attraction of simplicity.

A further adaptation of Stolwijk’s model was made by Kuznetz (1975, 1979), whose doctoral thesis reported the first known attempt to model non-uniform environmental conditions and internal heat generation. The cylindrical concentric layers are divided into four 90 degree sectors so that the model effectively becomes two-dimensional. Since this resulted in 361 unknown isothermal zones, it was necessary to find ways of reducing computer run time, while also minimising stability problems. Derivatives in the differential equations were approximated by Taylor series expansions in which second and higher order terms were neglected. Experiments have been carried out in which subjects stood facing a wall with a high radiant temperature for a period of time and then turned to face away from it. The Kuznetz model tracks the posterior and anterior skin temperature profiles to better than 1.5°C. It has been extensively used in the Space Shuttle programme, particularly for liquid conditioned clothing simulations and extravehicular activities.

A similar approach to Stolwijk’s was taken by Gordon,
Roemer and Horvath (1976) and defines more than one temperature node for each tissue region. Their model employed ten body segments to represent the passive system. Cylinders are used for the neck, thorax, abdomen, arms, hands, legs and feet. A sector of a cylinder represents the face and two spherical sectors represent the forehead and cranium as illustrated in Figure 3.8. The tissue type divisions were similar to Stolwijk’s in that core, muscle, fat and skin were identified, except that here, bone was separately defined so that each segment had four or five tissue layers. Each of these layers was further subdivided into two to five annular shells, the number depending on tissue conductivity and thickness. These shells provided integration nodes for the temperature calculations.

Bloodflow in the Gordon model was treated in a similar way to Stolwijk’s, using a central blood pool. Before reaching some particular tissues, however, blood was passed through counter-current heat exchangers to simulate the proximity of certain large veins and arteries to one another. Although the model contains Stolwijk’s equations for vasodilatation and sweating, it was primarily designed for cold air exposure simulations. Another unique feature is the use of heat flux through the skin, as well as head core and skin temperature, as a sensory input to the controller. The existence of such sensors has been debated for some time but no positive proof has been forthcoming. In the same manner as Stolwijk, Gordon uses a thermoneutral value for each flux and temperature to form an error signal and these signals are weighted according to density of receptors.

Validation of the model was carried out by the authors for exposures to air environments at 28°C changing rapidly to 4.7°C. It was found that, over a period of 120 min, rectal and skin temperatures for model and experiment agreed to within 1.5°C. However, foot and
FIG 3.8 GEOMETRY OF GORDON’S MODEL
From Gordon et al (1976)
hand temperatures differed by up to 5°C, as shown in Figures 3.9 and 3.10. Independent validations have been carried out by Baker, Harnett and Ringuest (1979) and by Wissler (1984), for cold water immersion. In both cases, major discrepancies between experiment and simulation became evident. The model exhibited the expected initial rise in core temperature, but it continued to increase, reaching approximately 42°C after five hours in Baker’s observations. Wissler has concluded that the numerical integration method used for the solution fails to converge when heat fluxes are large enough to simulate water immersion.

A further disadvantage with the Gordon model is the large amount of effort required to prepare data. As is normal with numerical integration techniques, initial conditions are set up and the solution is advanced over small time steps. This process is known as 'forward marching'. Unless the environment changes between successive time steps, there is normally no need to call for new data at each time step. However, as written, the Gordon model does so, and hence imposes a considerable penalty in running time.

Probably the most comprehensive thermoregulation model developed to date is due to Wissler at the University of Texas. The history of this model can be traced over many years, beginning with a steady state, six-cylinder model first reported in the open literature in 1961. This model consisted of 25 compartments and had the advantage of considering respiration heat exchange and counter-current heat exchange between large arteries and veins. By 1963, it had been revised to solve dynamic changes, whilst maintaining the same geometric configuration (Wissler, 1963). Shortly afterwards, the geometry was changed so that the body was represented by 15 cylindrical elements as shown in Figure 3.11. Finally, the number of compartments within the 15
FIG 3.9 VALIDATION OF GORDON'S MODEL
Redrawn from Gordon et al (1976)
FIG. 3.10 VALIDATION OF GORDON'S MODEL
Redrawn from Gordon et al. (1976)
FIG 3.11 GEOMETRY AND BLOODFLOW IN WISSLER'S MODEL

From Wissler (1984)
elements has now been increased to a total of 250, so that each element has between 15 and 21 radial shells. The purpose in doing so has been to define more accurately the temperature gradient in each segment (Wissler, 1984).

Wissler, in common with other authors, uses a statement of the first law of thermodynamics as the starting point for his model. This is formulated in the following equation, which simply states that heat generated by metabolic reactions is either stored in the tissue, carried away by circulating blood or conducted to the surface where it is transferred to the environment. For the element (i) therefore:

\[ \rho_i c_i \frac{\delta T_i}{\delta t} = (k_i r \frac{\delta T_i}{\delta r} + g_{mi} + \rho_b c_b w_{bi}(T_{ai} - T_i) + h_{ai}(T_{ai} - T_i) + h_{vi}(T_{vi} - T_i) \ (W_m^{-3}) \ (3.27) \]

where
- \( \rho \) = density of tissue or blood (kgm\(^{-3}\)),
- \( c \) = specific heat of tissue (Wm\(^{-3}\) °C\(^{-1}\)),
- \( T \) = temperature (°C),
- \( t \) = time (sec),
- \( r \) = radius of radial shell (m),
- \( k \) = thermal conductivity of tissue (Wm\(^{-1}\) °C\(^{-1}\)),
- \( g \) = rate of generation of metabolic heat in muscle (Wm\(^{-3}\)).

The suffixes (a) and (v) refer to arterial and venous quantities respectively, and the suffix (b) refers to blood, so that:

- \( w_{bi} \) = volumetric flow rate of blood entering the capillary beds (m\(^3\)s\(^{-1}\)),
- \( h_{ai} \) = heat transfer coefficient between arteries and tissue (Wm\(^{-3}\) °C\(^{-1}\)),
- \( h_{vi} \) = heat transfer coefficient between veins and tissue (Wm\(^{-3}\) °C\(^{-1}\)).
It is assumed that there is perfect heat transfer between capillaries and surrounding tissue, with the result that the temperature of blood leaving capillaries is equal to tissue temperature. It is also assumed that the rate of heat transfer between large blood vessels and surrounding tissue is proportional to the difference between blood and tissue temperatures. The effect of varying flow rate is not accounted for. Since the temperature of blood in large vessels varies with time, it is necessary to write energy balance equations for the venous and arterial blood in each element. The rate of change of heat energy in the blood pool of an element is equal to the sum of the net rate at which heat is carried into the pool by blood flow, the rate at which heat is transferred to tissue and the rate of counter-current heat exchange between large arteries and veins running parallel to one another. This is expressed as follows:

\[
\frac{d}{dt} \left( m_{ai} c_b T_{ai} \right) = \rho_b c_b w_{ai} (T_{am} - T_{ai}) + 2\pi l_i \int_0^{a_i} h_{ai} (T_i - T_{ai}) r \, dr + H_{avi} (T_{vi} - T_{ai})
\]

(3.28)

where \(T_{am}\) = temperature of blood entering the \(i^{th}\) element from the \(m^{th}\) element (°C)

\(m_{ai}\) = mass of blood in the arterial pool of the \(i^{th}\) element (kg)

\(w_{ai}\) = flow rate of blood entering \(i^{th}\) element (m\(^3\)s\(^{-1}\))

\(l_i\) = length of \(i^{th}\) element (m)

\(H_{avi}\) = heat transfer coefficient between large arteries and veins (W°C\(^{-1}\)).

The integral is necessary, since the exact position of blood vessels in the element is not defined, and since tissue temperature is a function of radius. A similar equation can be written for the venous pool:

\[
\frac{d}{dt} \left( m_{vi} c_b T_{vi} \right) = \rho_b c_b w_{vi} (T_{vn} - T_{vi}) + 2\pi l_i \int_0^{a_i} (\rho_b c_b w_{vi} + h_{vi}) (T_i - T_{vi}) r \, dr + H_{avi} (T_{ai} - T_{vi})
\]

(3.29)

where \(w_{vi}\) = flow rate into venous pool of element \(i\)
It can be seen from the second term on the right hand side that blood returning to the pool from capillaries is accounted for. The equation for venous temperature in the abdominal section is slightly different from Equation 3.29 since a vein from each leg flows into each section. Similarly, a vein from each arm flows into the thoracic section and, moreover, all venous streams terminate and arterial streams originate in this section.

Wissler has used the general arrangement of thermoreceptors and peripheral effectors proposed by Stolwijk, but includes rate of change of temperature as a stimulus. The signal produced by the rate-sensitive receptors is proportional to the difference between instantaneous temperature and an exponentially decreasing weighted mean of previous temperatures:

\[ S(t) = \beta T(t) - \beta \int_{-\infty}^{t} T(\tau) \exp[-\beta(t-\tau)] \, d\tau \quad (3.30) \]

where

- \( S(t) = \) afferent signal (function of time, dimensionless)
- \( \beta = \) a constant
- \( T(t) = \) temperature (function of time) (°C)
- \( \tau = \) exponential time constant (s)
- \( t = \) time (s)

Skin blood flow is also treated in a similar way to that of the Stolwijk model, in that vasodilatation occurs in response to raised hypothalamic and skin temperature. The ratio of core to peripheral temperature influence is 20:1 but there is also a term for rate of change of core temperature. Skin blood flow is also influenced by an afferent signal for vasoconstriction and in this case the ratio of core to peripheral influence is 1:1. When considering bloodflow to organs and tissues other than skin, account is taken of the conflicting needs for cardiac
output between thermoregulation and other functions such as exercise. The model also accounts for venomotor action, so that blood may return via superficial veins in hot conditions or deep veins in cold conditions. Counter-current heat exchange between veins and arteries is therefore greater under cold conditions.

The afferent signal for sweating, again similar to Stolwijk's, depends on core and peripheral temperatures with a weighting of 10:1. However, it also contains a rate term for core temperature. The signal is modified peripherally by local skin temperature. Similarity between the two models ends at shivering, which is treated in a more complex fashion by Wissler. The afferent signal for shivering contains contributions from lowered core temperature and skin temperature and from rates of change of core and skin temperature. This is apparently the only model to account for exhaustion in shivering, and will not allow it to proceed indefinitely. The cut-off point is determined by depletion of glycogen in muscle, which is explicitly modelled.

As was stated earlier, the Wissler model is the most complete model of the human thermal system for which details have been published. In fact, it is not merely a thermal model but in addition computes material balances for oxygen, carbon dioxide and lactic acid at each node where temperature is calculated. Furthermore, variation down the length of capillaries is accounted for, by dividing them into four regions for each segment. The addition of material balances considerably increases the computational requirements of the model, both in memory size and central processor time. However, these problems are becoming less significant as computing technology develops.

The model has been validated for a wide variety of conditions including warm and cold air exposure, cold
water immersion and even for divers breathing helium/oxygen mixture at high pressures. Wissler (1984) quotes a comparison with experiments by Saltin et al (1970), in which 3 subjects exercised at 25%, 50% and 75% of their maximum oxygen uptake for 30 minutes in air temperatures of 10, 20 and 30°C. A comparison of one subject’s core and mean skin temperatures with the model prediction is shown in Figure 3.12. For core temperature, agreement is excellent, although mean skin temperature does show some deviations. A further rigorous test of the model’s performance is provided by the data of Hayward (1975) who immersed 8 nude subjects at 10°C for one hour, followed by re-warming at 28-40°C. Figure 3.13 shows rectal and tympanic temperatures compared to the model’s central blood temperature and indicates agreement, generally to within 0.5°C.

Notwithstanding the good agreements quoted above, there are conditions under which the model gives anomalous predictions. As a part of the input data, the model calls for weight and mean skinfold thickness of the subject. Sizes are first given to the cylindrical elements to match the weight, based on known tissue densities. The subcutaneous fat is then distributed according to a stored table of data. Cylinder length must then be adjusted to correct the weight, which inevitably has change in this process. It seems this may result in an unrealistic surface area.

Simulation results are shown in Figure 3.14 for a thin subject with a 5mm mean skinfold thickness, suggesting that his core temperature will fall less rapidly than for other, fatter subjects (Hayes,P.A. 1985 - private communication). The experiments of Sowood (1988) show that this is unlikely. Further, modelling of an experiment by Kirkpatrick (1985), in which a single leg was exposed to air at -10°C, predicted a calf temperature after 30 minutes which differed by 8°C from the observed
FIG 3.12 HEAT STRESS VALIDATION OF WISSLER'S MODEL

Redrawn from Saltin et al (1970)
FIG 3.13 VALIDATION OF WISSLER'S MODEL FOR COLD WATER IMMERSION
Redrawn from Hayward (1975)
FIG 3.14 RECTAL TEMPERATURE PLOTS FROM THE WISSLER MODEL SHOWING ANOMOLOUS BEHAVIOUR AT LOW FAT VALUES (Hayes 1985)
result, as shown in Figure 3.15. This suggests either that blood flow or blood temperature in the extremities is overestimated, which may, in itself result from an under-estimate of losses from major blood vessels or too high a control gain on vasoconstriction in the skin layer.

The model has evolved to its present comprehensive state over a period of 27 years. The FORTRAN code is largely unstructured as a result, making it difficult to assimilate and to modify. Computation of material balances in the blood probably accounts for the long run-times which have been observed. Thus, although the Wissler model is applicable to a wide variety of exposures and gives accurate predictions for certain conditions, it has shortcomings which limit its usefulness.

3.6. Recent developments

Thermoregulation, if judged by volume of published work, has been extensively researched. Even in 1973, Bligh estimated that over 4000 papers had been published on the subject. A similar picture emerges when the attention is focussed on human thermal modelling. Writing in 1972, Hardy observed that details of more than 20 models were published in the decade 1960-70. He predicted that the 1970s might produce between 50 and 100 further models. The extent to which this prediction has been fulfilled depends on one's viewpoint as to what constitutes a model of thermoregulation. If one applies the caveats 'human' and 'whole body', the literature reveals only a few published works. It seems that the effect of a worldwide economic recession on scientific research may have been to slow the rate of development in this, as in many other fields.

In the present decade, the rate of publication on the
FIG 3.15 COOLING OF THE LEG WHEN EXPOSED TO -10°C
COMPARISON OF EXPERIMENTAL AND MODEL DATA
(Experiments by Kirkpatrick 1985)
subject appears to have declined still further and, with a few exceptions, most workers have made improvements to models already in existence. A notable exception is the model by Huckaba and Tam (1980). Although they employed a relatively simple geometry of eleven 4-layer cylinders, these authors devoted considerable attention to the control loops in the system. They attempted to apply classical control theory in order to visualise which variables are controlled, which are sensed, and what relationships exist between them. For example, the concept of a 'dead band', for which no output change occurs during limited input variations, was applied to the control of sweating and shivering. Thus, it was possible to model the observations that neither occurs without a certain minimum temperature change, but both may continue after the change has reduced below that threshold. An example of this approach is illustrated in the block diagram for thermoregulation during exercise in Figure 3.16. A major advantage of the model is that it uses weighting coefficients and gains derived from new experiments by the two authors and their co-workers. It therefore provides an independent check on other models.

By way of validation, Huckaba and Tam (1980) give detailed comparisons between the model’s temperature predictions and observations made during mild heat and cold stress exposures on seven male subjects. The heat stress consisted of an ambient temperature increase from 30 to 45°C over a period of 30 minutes with subjects nude and resting. Oral and model head core temperatures agreed to within 0.2°C and skin temperatures generally agreed to within 0.5°C. However, the changes in oral temperature only spanned approximately 0.5°C and skin temperature only 2°C. The cold exposure consisted of a fall in ambient temperature from 30 to 20°C, over a 20 minute period. Again, the model yielded good predictions but again the changes were small, the maximum being 6°C at the fingers. All these results are shown in Figure
FIG 3.16 HUCKABA'S CONTROL SYSTEM DIAGRAM
From Huckaba & Tam (1980)
FIG 3.17 VALIDATION OF HUCKABA'S MODEL
From Huckaba & Tam (1980)
3.17. It must be concluded that, whilst the techniques show promise, the model has not been validated far enough away from the comfort zone or for sufficiently long exposures to be useful in the context of this study.

Taking Huckaba's formulation, Volpe and Jain (1982) extended the model to account for whole-body hyperthermia as applied to the treatment of cancer patients. The majority of the changes were made to accommodate the dramatic skin bloodflow increases which occur during hyperthermia. Metabolic rate also increases with the resulting demand for cardiac output, and it is necessary to derive a relationship between core temperature and metabolic rate. The resulting model gave reasonable agreement with oral temperatures measured in the treatment of one patient by means of a liquid-conditioned suit. In this case, core temperature was maintained above 41°C for nearly an hour. The results of the comparison are shown in Figure 3.18. No comparisons were made for skin temperatures. Although the upward slope is reproduced by the model, it can be seen that errors of up to +1°C can occur. Since critical hyperthermia is being approached, it would therefore be dangerous to rely too closely on the predictions given by the model.

Smith and Twizell (1984) have supplemented their previous work on the steady state by extending their finite element technique to a dynamic model. Using a geometrically simple arrangement, these authors represent the head by a sphere, the torso by one cylinder, both arms by another and both legs by a third cylinder. Each of these elements is divided in the standard way into concentric shells for core, muscle, fat and skin. Two further shells are added to the cylinders to represent a layer of clothing and a layer of trapped air. For calculations of temperatures, the radius of each element is divided into 102 mesh points as shown in Figure 3.19.
FIg. 3.18 Validation of Volpe & Jain's Model

From Volpe & Jain (1982)

Oral temperature (°C)

Time (minutes)
FIG 3.19 SMITH AND TWIZELL'S SPATIAL DISCRETIZATION
From Smith & Twizell (1984)
The energy balance equations used are essentially those of Wissler (Equations 3.26 to 3.28 inclusive). However, whilst changes of bloodflow with time are represented and respiration is included, no mention is made of the physiological regulation mechanisms of sweating and shivering. The earlier paper (Twizell and Smith, 1982), in fact provides a validation only for a non-sweating, non-shivering subject. The latest paper follows essentially the same methods of obtaining a solution for each of the 102 mesh points, which depends on replacing partial derivatives by central difference operators as follows:

\[ \frac{\delta^2 T_{ijk}}{\delta r^2} = \frac{1}{h_i} \left( T_{ij(k+1)} - 2T_{ijk} + T_{ij(k-1)} \right) \]  

(3.32)

and  

\[ \frac{\delta T_{ijk}}{\delta r} = \frac{1}{2h_i} \left( T_{ij(k+1)} - T_{ij(k-1)} \right) \]  

(3.33)

where  

\[ T = \text{temperature} \ (°C) \]

\[ h = \text{radius increment} \ (m) \]

\[ r = \text{radius} \ (m) \]

The suffices \( i, j \) and \( k \) refer to the element number, tissue type and mesh point number, respectively. A simulation is reported in which a clothed subject is initially in a 23°C environment and has basal metabolic and sweating rates. The environmental temperature then changes to 10°C. The model computes 'snapshots' of temperature profiles after specified time intervals, as shown for example in Figure 3.20. The method of solution apparently does not lend itself to creating a time history for particular variables, although given that the information exists in data output files, it could presumably be re-organised to do so. The radial profiles are extremely well resolved, owing to the large number of mesh points. However, as no validation is reported and as subcutaneous temperatures are difficult to measure, no comment is made here on accuracy. The model currently simulates only the comfort zone.
FIG 3.20 SMITH & TWIZELL'S RADIAL PROFILE, EXAMPLE
From Smith & Twizell (1984)
While the 1970s saw a period of sustained activity in the field, developments in the 1980s have been sparse. An active group does exist, however, in the Institute for Physiology at the Ruhr University, West Germany. Mathematical modelling of human thermoregulation began in 1974 with a multi-element lumped-parameter model similar to Stolwijk's (Werner, 1975). Simpler single-node and core-shell models have also been produced for specific purposes in 1975 and 1983. The main thrust of this group, however, has been in distributed parameter modelling. A multi-element model showing the radial dependency of temperature profiles was developed between 1974-81 (Buse and Werner, 1984). Single element versions have also been produced to investigate the dynamics of fever and dependency on locally distributed parameters, such as those occurring in internal organs.

The most recent project is the first approach towards a highly sophisticated 3-dimensional model including axial heat flow (Buse and Werner, 1985). The body is represented by a set of 400,000 elements. Radial and axial functions of parameters, like tissue density for example, can be specified (if they are known). The model uses vector algebra to define the spatial relationships between elements. It attempts to define a flexible control strategy, in that the state of any element can affect the control of the state of any other.

There are a number of problems associated with this model, including the calculation of temperatures at the real body surface. Profiles within the body are apparently much easier. The morphology used is that of a standard person - only the size can be changed. The model runs on a Cyber 205 computer which has a 1/2M byte memory. This memory size is a limiting factor and overlaying has to be employed to overcome it. A time step of 1 minute requires 3 sec of CPU time but to reach the steady state may take 9 hours on a multi-user system.
The motivation for development of the model has been to define temperature profiles for clinical work. It has therefore been necessary to model organs explicitly and to include 64 tissue types. Together with the large number of elements, this is the cause of complexity and very long run-time. It is concluded that the clinical basis of this model makes it unlikely to be adaptable for use in survival and heat stress applications, where accurate internal profiles are not required. Validation is proceeding, using a small climatic chamber with a range of -10 to 60°C and humidity control. It appears that this is the only modelling development in West Germany at the present time.

3.7. Rationale for future development

The desirability of using a computer model to simulate the effects of thermal stress on military personnel has been established in Chapter 1 of the thesis. The need to predict the effects of immersion on survivors in cold water is high on the list of priorities in research. The ultimate goal is to achieve a model which will simulate accurately the effects of exposure not only to cold water but to hot or cold climates. The effects of clothing insulation and of work/rest cycles need to be predicted and extremity, as well as core temperatures are required.

It is apparent from the literature reviewed that modelling of the human thermal system is by no means easy. The development of a new model should not, therefore, be undertaken unless the need for one can be established beyond doubt. In assessing this need, the attributes of existing models require consideration under the following headings:
applicability to the exposure requiring simulation (air/water, heat/cold, clothed/nude)
availability of required variables (core/skin temperature, mean/individual site)
resolution and validity of predictions
processing factors (running speed, memory size)
user-related factors (structure, documentation)

Physical models are rejected without further discussion, the justification being their lack of flexibility. The empirical model by Givoni & Goldman (1972) appears capable of accurate predictions of core temperature response to heat exposure. However, it has no facilities for predicting skin temperatures and contains no empirical relationships for water immersion. The simple theoretical models of Crosbie et al (1963), Atkins & Wyndham (1969) and Gagge (1973) apply to air exposure. They are capable of providing core and mean skin temperature predictions but not those of extremities. This also applies to the similar model by Timbal et al (1976), which simulates cold water immersion.

Since predictions of extremity temperatures will be required, a multiple-segment geometry is mandatory. The simplest model in this category is due to Stolwijk and Hardy (1966). This six-segment model has been adapted for cold water exposure by Montgomery (1972) and for unequal radiant heat inputs by Kuznetz (1975). Many of the physiological assumptions in the original model have survived and are used in more recent models. However, the simplicity of the geometry is thought to be the cause of some of the observed inaccuracies. The treatment of blood distribution is also over-simplified, there being only a single blood pool.
A model which extends the Stolwijk approach has been developed by Gordon et al (1976). This embodies a 14-segment geometry and can simulate heat and cold stress in air or water. However, as described above, there are major discrepancies in some of its predictions. The 15-segment model by Wissler (1984) also uses, to some extent, the original ideas proposed by Stolwijk and Hardy and has a structure of 15 shells within each segment. This allows for a more accurate representation of internal temperature gradients. The model can be used to simulate heat or cold exposure, nude or clothed in air or water. However, as indicated above, there are anomalies in its predictions for certain exposures. In addition, the computer program contains little or no structure and is virtually undocumented. This makes it unattractive from the point of view of development.

The model by Huckaba & Tam (1980) and its adaptation by Volpe & Jain (1982) apply respectively to the comfort region in air and to hyperthermic treatment of cancer patients. The latter model shows errors in core temperature approaching 1°C. It is possible that the 4-shell structure within the 11-cylinder representation of the body is a limitation. The finite element model of Smith & Twizell does not simulate the physiological responses to thermal stress and is therefore not applicable to the present purpose. The final model in the review is by Buse & Werner (1984). This contains 400,000 temperature nodes and is therefore too complex to consider for adaption from its intended clinical purpose.

None of the above models appears to fulfil all the requirements of accuracy, applicability to the exposure and adequate documentation. In addition, most have no means of adjusting body size and morphology. The simulation of heat transfer by blood flow also appears to be over-simplified in some cases. Nevertheless, some contain features which it is desirable to adopt. This
applies particularly to the physiological equations by Stolwijk & Hardy (1966, 1977) and the 15-segment geometry by Wissler (1984). It therefore appears justified to undertake the development of a structured computer-based model of the responses of the human thermal system to immersion in cold water. While fulfilling an immediate research objective, this will also provide the basis for extension to the simulation of heat and cold exposure.
4. DEVELOPMENT OF A MODEL FOR COLD WATER IMMERSION

4.1. Mathematical background

The development of a multi-segment computer model of the human thermal from first principles is a task of considerable magnitude. The task has been undertaken by previous workers, as described in Chapter 3. The strategy for development will be to extend upon, rather than to re-create these earlier attempts.

A logical first step in formulating a new model is to determine the mathematical basis on which to proceed. Researchers in thermal physiology and survival are likely to make the most use of model which will predict the time history of the body's responses to thermal stress. The variables on which information is normally required are temperatures, both internally and at various points on the skin surface. It is therefore necessary to represent the body mathematically in such a way as to assign a temperature to each region or compartment of interest. This is the 'lumped parameter' approach referred to in Chapter 3 and has been adopted by the majority of previous workers in the field of human thermal modelling.

The evolution of the geometrical representation of the body may be traced from the original 6-segment proposal by Stolwijk and Hardy (1966). This was modified by Montgomery (1972) to contain 10 shells per segment rather than the original four, but still had only one blood pool. Wissler (1984) increased the number of segments to 15 and introduced an arterial and venous blood pool in each but treated left and right limbs together. It is proposed that the new model will separate the limbs and introduce minor blood pools in viscera, muscle and skin for each segment. Blood heat transfer coefficients will be calculated as functions of geometry.
It is proposed to adopt data on thermal properties of the body and thermoregulatory control from the improved model by Stolwijk and Hardy (1977). Additional data from the model by Gordon, Roemer and Horvath (1976) will also be included. The data on blood flow and distribution will, however, be taken from physiological, rather than modelling sources.

Other respects in which the new model will differ from its predecessors include the estimation of true surface temperature and accounting for the effects of temperature on the thermal properties of fluids and tissues. One of the primary purposes in developing a new model will be to provide a sound theoretical basis on which to superimpose the physiological mechanisms of thermoregulation.

Each compartment is given physical properties such as mass, density, volume, specific heat and thermal conductivity, so that a set of heat balance equations similar to Equation 2.13 can be written. Examining Equation 2.13, it is seen that the rate of change of tissue temperature depends on the sum of heat flows into and out of a compartment. These heat flows are themselves dependent on the temperature in the compartment and its surroundings. As a result, a first order differential equation in temperature may be written for each compartment, of the form:

\[ \frac{dT}{dt} + a \cdot T = b \cdot u(t) \]  

(4.1)

where \( T \) = temperature (°C)
\( u \) = an input function depending on external factors 
\( a \) & \( b \) = constants depending on physical properties 
\( t \) = time (s)

Alternatively, Equation (4.1) may be written:

\[ T = a \cdot T + b \cdot u \]  

(4.2)
Huckaba and Tam (1980), among other authors, have shown that classical control theory may be applied to a set of such equations in order to represent the thermoregulation system of the body as a series of coupled feedback loops. The controlling elements in these loops are the phenomena of shivering, sweating, vasoconstriction and vasodilation, which act to maintain the body temperature at a 'normal' value.

Since a digital computer is to be employed to carry out the simulation, a continuous time solution is not possible. The thermal state of the body will be predicted at a succession of discrete instants in time, as a function of conditions in each of the previous instants. If a large number of separate compartments is used to represent the body, the sequential solution of the heat balance equations becomes cumbersome. An alternative approach has been developed, known as 'state space analysis', which represents each variable in a system as an element in a vector space, as described by Ogata (1967). Equation 4.2 is, in effect, one of a system of cross-coupled equations which may be written in general vector/matrix form:

$$\dot{x} = A.x + B.u$$

(4.3)

where $x = \text{an n-dimensional state vector (whose elements could represent temperatures in the body)}$

$A = \text{an n x n matrix containing system coefficients}$

$B = \text{an m x n matrix coupling inputs to the system}$

$u = \text{an m-dimensional vector whose elements represent inputs to the system}$
The concept is illustrated in Figure 4.1. Its attraction for a computer-based simulation is that, using an array processor, a single vector/matrix calculation would be needed to determine the thermal state at each discrete time interval.

The primary limitation of state space is that it applies only to linear systems or over linear portions of the characteristics of non-linear systems. Considering a general state transformation $x_3 = f(x_1, x_2)$ in which a particular state variable $x_3$, say, is a function of two of the system variables $x_1$ and $x_2$, the transformation will be linear if the function can be expressed as a linear combination using constants such as $\alpha$ and $\beta$:

$$f(x_1, x_2) = \alpha x_1 + \beta x_2 \quad (4.4)$$

Thus, if changes $\Delta x_1$ and $\Delta x_2$ occur, then:

$$f(x_1+\Delta x_1, x_2+\Delta x_2) = \alpha x_1 + \alpha \Delta x_1 + \beta x_2 + \beta \Delta x_2$$
$$= f(x_1, x_2) + f(\Delta x_1, \Delta x_2) \quad (4.5)$$

which indicates linearity. However, if:

$$x_3 = x_1 \cdot x_2$$

then

$$f(x_1+\Delta x_1, x_2+\Delta x_2) = x_1 \cdot x_2 + x_1 \cdot \Delta x_2 + x_2 \cdot \Delta x_1 + \Delta x_1 \cdot \Delta x_2$$
$$\neq f(x_1, x_2) + f(\Delta x_1, \Delta x_2) \quad (4.6)$$

and the system is non-linear.

Applying this argument to the human thermal system, it is apparent from the consideration of thermoregulatory mechanisms in Section 2 that their control equations contain a number of non-linear terms (exponentials, discontinuities, etc). Furthermore, if heat transfer by blood flow is examined, it is seen that the quantity of heat supplied by blood to a tissue compartment is given by:
FIG 4.1 STATE SPACE REPRESENTATION OF A SYSTEM WITH N VARIABLES
\[ Q_b = w c (T_{\text{blood}} - T_{\text{tissue}}) \quad (4.7) \]

where \( w \) = volumetric flow rate of blood (ml s\(^{-1}\))
\( c \) = specific heat of blood (J ml\(^{-1}\) °C\(^{-1}\))
\( T \) = temperature (°C)

The blood flow rate is dependent on local and central body temperatures, with the result that squared terms in temperature appear in Equation 4.7. These are non-linear, as shown in Equation 4.6. The rate of change of tissue temperature will, of course, depend upon heat flows such as \( Q_b \) and it is the vector containing these rates of change which forms the first derivative of state vector. It is therefore concluded that the state space approach cannot be used directly in the simulation of the human thermal system.

A possible method of overcoming the problem of non-linearity is to linearise about each point in the vector space using the first method of Liapunov (Wiberg 1971). The method may be illustrated by the following example of a non-linear function \( x_3 = f(x_1, x_2) \), illustrated geometrically in Figure 4.2. The non-linear function is represented by the surface \( S \) in the \( x_1, x_2, x_3 \) plane. The total differential may be found:

\[ dx_3 = \left( \frac{\delta f}{\delta x_1} \right) dx_1 + \left( \frac{\delta f}{\delta x_2} \right) dx_2 \quad (4.8) \]

where \( dx \) = a small change in a state
\( \frac{\delta f}{\delta x} \) = the partial derivative in a single direction \( x \)
FIG 4.2 TWO DIMENSIONAL ILLUSTRATION OF LINEARISATION ABOUT A POINT
An initial point \( Q \) may be defined on the surface, such that:

\[
x_3 - x_3(0) = (\delta f/\delta x_1)_Q(x_1 - x_1(0)) + (\delta f/\delta x_2)_Q(x_2 - x_2(0))
\]  

(4.9)

This is the equation of the tangent plane \( P \) to the surface \( S \) at point \( Q \). In an \( n \)-dimensional system similar to the one expressed in Equation 4.3,

\[
\dot{x} = F \cdot x
\]

and \( F \) is a matrix of functions \( f_i \) such that:

\[
\begin{bmatrix}
\dot{x}_1 \\ \vdots \\ \dot{x}_n
\end{bmatrix} =
\begin{bmatrix}
f_1(x_1, x_2, \ldots, x_n) \\ \vdots \\ f_n(x_1, x_2, \ldots, x_n)
\end{bmatrix}
\]

it is necessary to define a new state vector with respect to an initial point \( x_0 \), say \( y = x - x_0 \). Then:

\[
\dot{y} = A \cdot y \text{ where } A = \begin{bmatrix}
\delta f_1/\delta x_1 & \cdots & \delta f_1/\delta x_n \\
\vdots & \ddots & \vdots \\
\delta f_n/\delta x_1 & \cdots & \delta f_n/\delta x_n
\end{bmatrix}
\]

If the \( \delta f_i/\delta x_i \) terms are numbers, the point \( Q \) is stationary. If they contain terms in \( x \), it is moving with time. In this case, the \( A \) matrix must be re-evaluated at each new time step. Equation 4.7 implies that this will be the case for the simulation of the human thermal system. The need to re-evaluate the elements of the system matrix before performing the transformation for each time step makes this method of simulation potentially computationally intensive. In addition, it is necessary to perform an additional transformation to find the original state vector (temperature) from the new state vector. Such a simulation would bear little conceptual relation to the real process of thermoregulation and so would do little to aid its understanding. It is therefore concluded that
pursuit of a linearised state space simulation is not profitable and a more conventional approach is adopted.

While the state space approach is not specifically employed in the simulation, a vector/matrix representation provides a useful description of the human thermal system. Figure 4.3 illustrates the concept. A set of temperatures for which sensory mechanisms exist is compared to 'set point' values. The existence of these set points has been questioned by Werner (1975) but until this physiological issue is resolved, no attempt is made here to implement a new theory. An error vector is formed, which drives the thermoregulatory mechanisms. These result in heat flows which must be summed with those caused by environmental changes and exercise and with the passive heat flows within the body. The net heat flows cause rates of change of temperature in all compartments, which are integrated to obtain new temperatures, as in the state space method. As a computer-based simulation, the solution proceeds in a series of time steps, temperatures at a particular step being calculated as a result of conditions which obtained at the previous step. The temperature feedback is used not only to form the error signal but also to modify the coefficients of the control equations. Hence the non-linearity of the system is represented. The system equations are solved sequentially in a conventional manner, as described in more detail in the following Sections.

4.2. Geometrical representation of the body

The following description of the model may be traced in Fortran 77 code by referring to the listings in Volume 2 of the thesis. The underlying physical and physiological equations are presented in this chapter.
FIG 4.3 VECTOR/MATRIX REPRESENTATION OF THE HUMAN THERMAL SYSTEM
In common with previous multi-segment models, the present model uses cylindrical and spherical segments to represent various anatomical features of the body. There are 15 segments in all, as shown in Figure 4.4. Segment 1, the head, is spherical as in Stolwijk and Hardy’s model (1977). The torso, however, is represented by two separate cylindrical segments (2 and 3) corresponding to the thorax and abdomen. This has the advantage of enabling the internal organs to be represented more accurately.

Each leg is divided into a proximal, medial and distal segment (4, 5, 6, 7, 8 and 9) corresponding approximately to the thigh, calf and foot. Wissler (1984) also adopts this approach but does not treat the limbs separately in his heat flow calculations, despite using 15 segments. The arms are also divided into three segments, representing the upper arm, fore-arm and hand (10, 11, 12, 13, 14 and 15). In the FORTRAN 77 program, the segments are numbered by the integer variable I. The use of separate segments for the left and right limbs potentially enables unequal exposures to be simulated.

Each segment is sub-divided into 10 radial shells. Stolwijk and Hardy used only four, as did Gordon et al (1976) but this number is not sufficient to represent accurately the steep temperature gradients which can occur during cold water immersion. Wissler’s use of 15 internal shells plus up to 7 for clothing on each segment imposes a considerable computational overhead. The shells in each segment are numbered by the integer J in the program. I and J together determine which of the 150 tissue compartments is being referred to. Hence, many of the variables are held in 10 or 15-element arrays or in two-dimensional 15x10 arrays. The four innermost shells represent core tissue (viscera, bone and internal fat).
FIG 4.4 GEOMETRICAL REPRESENTATION OF THE BODY

Body segments
I = 1 to 15

Muscle
J = 5 to 8

Skin
J = 10

Fat
J = 9

Cross-section
Shells 5 to 8 represent muscle, 9 represents subcutaneous fat and 10 represents skin.

The geometrical formulation allows for variation in the height, mass and specific gravity of the subject under simulation. If neither specific gravity nor percent body fat are given as inputs, the equations of Pierson and Eagle (1969) are used:

\[
S = 0.8 \frac{(100 \ H)^{0.242}}{(1000 \ M)^{0.1}} + 0.162
\quad (4.10)
\]

and

\[
P = 100 \left( \frac{5.548}{S} - 5.044 \right)
\quad (4.11)
\]

where

- \(S\) = specific gravity
- \(P\) = percent body fat by weight
- \(H\) = height (m)
- \(M\) = body mass (kg)

The percentage body fat by volume is found from a knowledge of the density of fat. This density and other thermophysical properties of tissues have been obtained from Gordon et al (1976) and Charny et al (1987). Table 4.1 gives the numerical values. The fraction of the volume of each segment taken up by a particular tissue type is calculated from data used by Montgomery (1972). These numerical values, shown in Table 4.1, are the mass fractions obtained by dissection and are therefore converted to volumetric fractions in the program. However, these fractions apply only to the fat distribution for the 'standard man' and must be modified for individual subjects as follows:

\[
F_{fat}(i) = F_{fat}(i) \left[ 1 + \frac{(P_v - P_{vs})}{100} \right]
\quad (4.12)
\]

\[
F_{nat}(i) = 4[F_{bone}(i) + F_{visc}(i) + F_{musc}(i) + F_{skin}(i)]
\quad (4.13)
\]

and, for example,
\[ F_{\text{bone}}(i) = F_{\text{bone}}(i) - \frac{[F_{\text{fat}}(i) (P_v - P_{vs})/100] F_{\text{bone}}(i)}{F_{\text{nat}}(i)} \]  

where \( F_{\text{fat}}(i) \) = volume fraction of fat in segment \( i \)  
\( F_{\text{nat}}(i) \) = volume fraction of non-adipose tissue in segment \( i \)  
\( P_v \) = percent total body fat by volume for given subject  
\( P_{vs} \) = standard percent body fat by volume

The remaining \( F \) terms are volumetric fractions of other tissue types in a segment and are similarly modified. The total body fat content is divided into a subcutaneous fraction and an internal fraction which is distributed among the four core shells of each segment. This division conforms to the data of Hayes and Sowood (1985).

Having established the tissue proportions of each segment, it is possible to estimate the volume of each shell. The total surface area of the body is first estimated from the formula of Dubois and Dubois (1915):

\[ A = 7.184 \times 10^{-3} M^{0.425} (100 H)^{0.725} \ (m^2) \]  

The Quetelet index, the ratio of body mass to the square of height, is often used to describe the somatotype of an individual (Light & Gibson 1986). A shape factor, defined as \( M/H^2/25.15 \), is used to modify the shapes of the cylinders in accordance with the somatotype of the subject. The value 25.15 is \( M/H^2 \) for the standard man defined by Stolwijk and Hardy (1966, 1977). The distribution of surface areas in Table 4.1 is used to determine the radii, lengths and hence volumes of the segments. The shell volumes are then determined from the fractions in Table 4.1, using the defined proportions of bone, viscera and internal fat in the core shells. The mean density of each shell is estimated from the individual tissue densities and volume fractions.
TABLE 4.1 - PROPERTIES AND DISTRIBUTION OF TISSUE

<table>
<thead>
<tr>
<th>Tissue Type</th>
<th>Density (kg m⁻³)</th>
<th>Specific Heat (J kg⁻¹ K⁻¹)</th>
<th>Thermal Conductivity (W m⁻¹ K⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone</td>
<td>1300.0</td>
<td>1591.0</td>
<td>1.1630</td>
</tr>
<tr>
<td>Brain</td>
<td>1050.0</td>
<td>3696.9</td>
<td>0.5275</td>
</tr>
<tr>
<td>Lung</td>
<td>550.0</td>
<td>3717.9</td>
<td>0.2820</td>
</tr>
<tr>
<td>Viscera</td>
<td>1050.0</td>
<td>3696.0</td>
<td>0.5466</td>
</tr>
<tr>
<td>Muscle</td>
<td>1050.0</td>
<td>3768.1</td>
<td>0.4200</td>
</tr>
<tr>
<td>Fat</td>
<td>850.0</td>
<td>2512.1</td>
<td>0.1600</td>
</tr>
<tr>
<td>Skin</td>
<td>1000.0</td>
<td>3768.1</td>
<td>0.2093</td>
</tr>
<tr>
<td>Blood</td>
<td>1000.0</td>
<td>3744.0</td>
<td>0.5466</td>
</tr>
</tbody>
</table>

Fractions of segment mass taken by each tissue type, per radial shell:

<table>
<thead>
<tr>
<th></th>
<th>Bone</th>
<th>Viscera</th>
<th>Muscle</th>
<th>Fat</th>
<th>Skin</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>0.075875</td>
<td>0.111325</td>
<td>0.023000</td>
<td>0.092000</td>
<td>0.067200</td>
</tr>
<tr>
<td>Thorax</td>
<td>0.018375</td>
<td>0.060725</td>
<td>0.116225</td>
<td>0.183600</td>
<td>0.035100</td>
</tr>
<tr>
<td>Abdomen</td>
<td>0.018375</td>
<td>0.060725</td>
<td>0.116225</td>
<td>0.183600</td>
<td>0.035100</td>
</tr>
<tr>
<td>Each thigh</td>
<td>0.060600</td>
<td>0.023100</td>
<td>0.123100</td>
<td>0.115000</td>
<td>0.057800</td>
</tr>
<tr>
<td>Each calf</td>
<td>0.060600</td>
<td>0.023100</td>
<td>0.123100</td>
<td>0.115000</td>
<td>0.057800</td>
</tr>
<tr>
<td>Each foot</td>
<td>0.095350</td>
<td>0.015475</td>
<td>0.018050</td>
<td>0.236800</td>
<td>0.247700</td>
</tr>
<tr>
<td>Each upper-arm</td>
<td>0.053375</td>
<td>0.026100</td>
<td>0.119225</td>
<td>0.137300</td>
<td>0.067900</td>
</tr>
<tr>
<td>Each forearm</td>
<td>0.053375</td>
<td>0.026100</td>
<td>0.119225</td>
<td>0.137300</td>
<td>0.067900</td>
</tr>
<tr>
<td>Each hand</td>
<td>0.085825</td>
<td>0.011200</td>
<td>0.026125</td>
<td>0.223800</td>
<td>0.283600</td>
</tr>
</tbody>
</table>

Fractions of total body surface area in each segment:

<table>
<thead>
<tr>
<th></th>
<th>Head 0.070</th>
<th>Thorax 0.180</th>
<th>Abdomen 0.180</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thigh</td>
<td>0.095</td>
<td>Calf 0.063</td>
<td>Foot 0.034</td>
</tr>
<tr>
<td>U. arm</td>
<td>0.034</td>
<td>F. arm 0.034</td>
<td>Hand 0.025</td>
</tr>
</tbody>
</table>

TABLE 4.1 - PROPERTIES AND DISTRIBUTION OF TISSUE

103
mass of each shell can then be estimated.

Up to this point, the masses and dimensions of the body segments have been determined in an 'open loop' sense. It is possible to close this loop by applying corrections based on the known body mass and specific gravity, thus minimising any inaccuracies in the data on individual tissues. An estimated body mass is obtained from the sum of individual shell masses and is used in conjunction with the true body mass to obtain a corrected set of shell masses:

\[ M_{\text{shell}}(i,j) = M_{\text{shell}}(i,j) \cdot M/M_{\text{est}} \]  \hfill (4.16)

where \( M_{\text{shell}} = \) mass of each individual shell (kg)
\( M_{\text{est}} = \) mass estimate for whole body (kg)

The new shell masses are used to find intermediate estimates for individual shell volumes and hence whole body volume. A final correction is applied to these by using the specific gravity for the whole body:

\[ V_{\text{shell}}(i,j) = V_{\text{shell}}(i,j) \cdot 1000.S.V_{\text{body}}/M_{\text{body}} \]  \hfill (4.17)

where \( V_{\text{shell}} = \) volume of an individual shell (m³)
\( V_{\text{body}} = \) volume of whole body (m³)
\( 1000.S = \) apparent density of body (kgm⁻³)

It is now possible, by using cylindrical and spherical geometry, to find the outer radius of each shell and hence to determine the segment surface areas and volumes.

4.3. Thermal conduction within the body

A fundamental feature of the lumped parameter approach is that the variables and constants in each compartment of the model are considered to act at a single point or plane in the compartment. For a solid cylindrical or
spherical shell, there exists a radius such that the mass within it is equal to the mass outside. This is known as the centre-of-mass radius $r_{cm}$. For a spherical shell, this is derived from:

$$r_{cm} = \left\{ \frac{[r(i,j-1)^3 + r(i,j)^3]}{2} \right\}^{1/3}$$

(4.18)

and for a cylinder:

$$r_{cm} = \left\{ \frac{[r(i,j-1)^2 + r(i,j)^2]}{2} \right\}^{1/2}$$

(4.19)

where $r(i,j-1) = \text{inner radius of shell } j \text{ in segment } i$

$r(i,j) = \text{outer radius}$

The temperatures within the tissue shells are defined at these radii and the conduction path lengths ($x$ in Equation 2.1) are the distances between them. The area through which heat conduction takes place is considered to be the cylindrical or spherical mid-plane half way between the two centres of volume. These areas are calculated at $r = r_{cm} + x/2$, as shown in Figure 4.5A. This completes the geometrical representation sufficiently for conductive heat flow calculations to be made.

For the purpose of heat balance calculations, it is necessary to assign a thermal capacitance (mass x specific heat) to each tissue shell. The specific heats of the core shells are a combination of those for their constituent tissue types and are therefore calculated proportionally, using the volume fractions in Table 4.1. The thermal capacitances of blood compartments in each segment must also be calculated and their values subtracted from the tissue capacitances, when the blood volumes are known. Blood volume determination is described in detail in Section 4.4.

In Equation 2.1, the quantity $A.k/x$ is known as the
FIG 4.5a RADII FOR HEAT CONDUCTION BETWEEN SHELLS OF IDENTICAL TISSUE

FIG 4.5b RADII FOR CONDUCTION BETWEEN SHELLS OF DIFFERING TISSUE
thermal conductance, which, when multiplied by a
temperature difference, gives the heat flow in Watts
directly. Accordingly, the thermal conductance between
each tissue shell and the next is calculated. This is a
simple matter in cases where neighbouring shells are of
the same tissue type and therefore have the same thermal
conductivity. In crossing a tissue boundary (for
example, between shells i,8 and i,9 - muscle to fat) a
mean conductance is estimated. Figure 4.5B illustrates
the procedure. A new mid- plane radius is found in each
shell, half way between the centre-of-mass radius and the
boundary:

\[ r_{mp}(i,j) = r_{cm}(i,j) + \frac{r(i,j) - r_{cm}(i,j)}{2} \] (4.20)

and \[ r_{mp}(i,j+1) = r(i,j) + \frac{r_{cm}(i,j+1) - r(i,j)}{2} \] (4.21)

where \( r_{mp} \) = new mid-plane radius (m)

The mid-plane areas are then calculated:

\[ A_{mp}(i,j) = 2\pi r_{mp}(i,j) \cdot l_{seg} \text{ (for cylinder)} \] (4.22)

and \[ A_{mp}(i,j+1) = 2\pi r_{mp}(i,j+1) \cdot l_{seg} \] (4.23)

The conduction distances are:

\[ x(i,j) = r(i,j) - r_{cm}(i,j) \] (4.24)

and \[ x(i,j+1) = r_{cm}(i,j+1) - r(i,j) \] (4.25)

The thermal conductances on either side of the boundary
are then:

\[ K(i,j) = A_{mp}(i,j) \cdot k(i,j) / x(i,j) \] (4.26)

and \[ K(i,j+1) = A_{mp}(i,j+1) \cdot k(i,j+1) / x(i,j+1) \] (4.27)

The harmonic mean of the conductances is then calculated:

\[ K(j,j+1) = \frac{K(i,j) \cdot K(i,j+1)}{K(i,j) + K(i,j+1)} \] (4.28)
where \( K(j,j+1) = \) the mean thermal conductance between
dissimilar tissue shells \( i,j \) and \( i,j+1 \)
\((W °C^{-1})\).

This procedure was proposed by Montgomery (1972) but was
only applied to six tissue boundaries, since only two
different tissue conductivities were used. Seven are
used in the model under discussion and the process is
applied at 45 separate boundaries. In the program, the
initial conductances are stored and are later modified as
tissue conductivity changes with temperature.

4.4. The circulatory system

The total volume of blood in the body varies with body
mass and can be predicted from the curve given by Guyton
(1986), shown in Figure 4.6. For the purposes of
modelling this curve has been quantised into a number of
linear sections to give the following relationships:

<table>
<thead>
<tr>
<th>Mass of body (kg)</th>
<th>Volume of blood (l)</th>
</tr>
</thead>
<tbody>
<tr>
<td>( M &lt; 45 )</td>
<td>0.079 ( M )</td>
</tr>
<tr>
<td>( 45 &lt; M &lt; 50 )</td>
<td>3.7 + 0.12(M - 45)</td>
</tr>
<tr>
<td>( 50 &lt; M &lt; 60 )</td>
<td>4.2 + 0.09(M - 50)</td>
</tr>
<tr>
<td>( 60 &lt; M &lt; 70 )</td>
<td>5.06 + 0.05(M - 60)</td>
</tr>
<tr>
<td>( 70 &lt; M &lt; 80 )</td>
<td>5.56 + 0.035(M - 70)</td>
</tr>
<tr>
<td>( 90 &lt; M )</td>
<td>5.91 + 0.025(M - 80)</td>
</tr>
</tbody>
</table>

The total volume of blood is distributed among major and
minor arteries and veins, the heart and pulmonary
vessels, liver, spleen and capillaries. This
distribution changes with exercise, posture and the
thermal state of the body, among other factors. Owing to
lack of data, it has not been possible to simulate
changes in the distribution of volume. The fixed
distribution is determined from the data published by
Guyton (1986) with allowances made for the liver and
FIG 4.6 BLOOD VOLUME AS A FUNCTION OF BODY MASS

From Guyton (1986)

Lean male - 79ml kg
FIG 4.7 DISTRIBUTION OF VENOUS AND ARTERIAL BLOOD

Modified from Guyton (1986)

Large veins 30.5%
Venous 64%
Small veins 20%
Arterial 31%
Capillaries 5%
Large arteries 8%
Small arteries 7%
Pulmonary 6%
Heart 3.5%
Liver 5%
Spleen 1.5%
Pulmonary 3.5%
Spleen 1.5%
spleen as shown in Figure 4.7. From Guyton’s distribution, a table of fractions of the total volume is formed, as listed in Table 4.2. This is then used in conjunction with the proportions of total body volume in each segment to determine the quantities of blood per segment in each type of blood vessel.

<table>
<thead>
<tr>
<th>Arteries</th>
<th>Veins</th>
</tr>
</thead>
<tbody>
<tr>
<td>Large</td>
<td>0.080</td>
</tr>
<tr>
<td>Small</td>
<td>0.070</td>
</tr>
<tr>
<td>Pulmonary</td>
<td>0.060</td>
</tr>
<tr>
<td>Heart</td>
<td>0.035</td>
</tr>
<tr>
<td>Liver</td>
<td>0.050</td>
</tr>
<tr>
<td>Spleen</td>
<td>0.015</td>
</tr>
</tbody>
</table>

**TABLE 4.2 - FRACTIONAL DISTRIBUTION OF BLOOD VOLUME**

It is assumed that there are two major arteries and two major veins in each body segment and that these are situated at the axis, as shown in Figure 4.8A. This assumption, although not valid throughout the lengths of all segments, appears to be a reasonable approximation when compared to the anatomical cross-sections displayed in Williams & Warwick (Eds.) (1980). The diameters of the major vessels are simply obtained by cylindrical geometry.

Blood in the minor arteries and veins is distributed among a network supplying the internal organs, muscles and skin. Fat is assumed to be avascular. In order to determine the length of minor vessels in each tissue shell, it is necessary to assume mean diameters for arteries and veins. For arteries, this is given the value 500 μm, based on the assumptions of Keller and Seiler (1971). The vein diameter is increased in the ratio of minor venous to arterial volumes. For simplicity, it is assumed that all minor blood vessels run in a radial direction through the thickness of each shell and that they are equally spaced from one another. Figure 4.8B illustrates this arrangement. The total length of minor blood vessels in a segment can then be
Two artery/vein pairs close to axis

FIG 4.8a ARRANGEMENT OF MAJOR BLOOD VESSELS

Artery/vein pairs running radially

FIG 4.8b ARRANGEMENT OF MINOR BLOOD VESSELS IN A TYPICAL TISSUE SHELL
calculated. For example, in the four muscle shells of a particular segment:

\[
1_{\text{am}}(i) = \frac{v_{\text{am}}(i)}{1000\pi r_a^2}
\]  

(4.29)

\[
1_{\text{vm}}(i) = \frac{v_{\text{vm}}(i)}{1000\pi r_v^2}
\]  

(4.30)

where \(1_{\text{am}}(i)\) = total length of minor arteries in the muscle shells in segment \(i\) (m)  
\(1_{\text{vm}}(i)\) = total length of minor veins (m)  
\(v_{\text{am}}(i)\) = volume of blood in minor arteries in the muscle shells of segment \(i\) (l)  
\(v_{\text{vm}}(i)\) = volume in minor veins (l)

Similar calculations are made for viscera and skin. The number of small arteries or veins in a tissue region such as muscle can be obtained by dividing the total length by the thickness of the region, which is the difference between the radii of the 5th and 8th shells:

\[
N_m(i) = \frac{1_{\text{am}}(i)}{(r(i,8) - r(i,5))}
\]  

(4.31)

A representative mid-plane area for the region, say \(A_{\text{mp}}(i,6)\) is then divided by this number and the square root of the result is taken to find the distance between one minor artery or vein and the next:

\[
x_{\text{m}}(i) = \left[\frac{A_{\text{mp}}(i,6) \cdot (r(i,8) - r(i,5))}{1_{\text{am}}(i)}\right]^{1/2}
\]  

(4.32)

where \(x_{\text{m}}(i)\) = distance between vessels of one type (artery or vein) (m)

Similar calculations are carried out for skin and viscera. On the basis of studies of the vascular system of the rabbit, Weinbaum et al (1984) suggest that arteries and veins run in pairs, with their axes spaced 1.5 times the mean diameter apart. A proportion, rather
than a measurement, may readily be scaled from rabbit to human. Having determined the dimensions and spacing of the blood vessels, it is possible to assign values to the heat conductances between the vessels and surrounding tissue and to the counter-current conductances between arteries and veins. The rate of heat transfer is not considered to vary with blood flow. This feature was originally incorporated but since it required the conductances to be re-evaluated at each time step, the running speed of the simulation was almost halved.

The conductances are calculated according to the equations of McAdams (1954), who examined heat flow in buried pipes. Taking the minor arteries in muscle as an example, conduction is assumed to take place radially in a region whose outer radius is half the inter-artery spacing, as shown in Figure 4.9A. Thus:

\[
\frac{h_{am}(i)}{2\pi k_m l_{am}(i)} = \frac{1}{\log\left(\frac{x_m(i)}{2r_a}\right)}
\]  

(4.33)

where \( h_{am}(i) \) = thermal conductance between arteries and tissue in muscle of segment \( i \) (W °C⁻¹)

The heat exchange between veins and surrounding tissue may be expressed in a similar way. For counter-current heat exchange, the conductance between the two cylindrical vessels is given by an expression:

\[
h_{ccm}(i) = \pi k_m l_{am}(i)/1.1
\]  

(4.34)

(Since the natural logarithm of 1.5, the ratio of the diameters, is 1.1).

where \( h_{ccm}(i) \) = thermal conductance between neighbouring arteries and veins in the muscle of segment \( i \) (W °C⁻¹)

The conductances for vessels in the skin and viscera may be calculated from equations of the same form with the
FIG 4.9a CONDUCTANCES AFFECTING MINOR BLOOD VESSELS

FIG 4.9b CONDUCTANCES AFFECTING MAJOR BLOOD VESSELS
appropriate parameters substituted. For the large vessels at the axes of each segment, radial conduction is assumed to take place within a cylindrical region bounded by the radius of the 4th shell, as shown in Figure 4.9B. Furthermore, there are two of each type of vessel in each segment so the conductance for arteries is given by:

\[ h_{a1}(i) = 2 \frac{2\pi k_v \cdot I_{seg}(i)}{\log[r(i,4)/r_{a1}(i)]} \]  \hspace{1cm} (4.35)

where \( h_{a1}(i) \) = thermal conductance between the two large arteries and tissue in segment \( i \) (W °C\(^{-1}\))

\( k_v \) = conductivity of viscera (W m\(^{-1}\) K\(^{-1}\))

\( r_{a1}(i) \) = radius of large arteries in segment \( i \) (m)

A similar expression may be written for venous conductance. The counter-current conductance in large blood vessels is expressed in a similar manner to that between minor vessels, as in Equation 4.34.

4.5. Representation of thermoregulatory mechanisms

The three primary mechanisms of thermoregulation - sweating, shivering and control of skin blood flow - are included in the model. Since the main purpose is to simulate immersion in cold water, the effects of increased evaporation at the skin surface are not simulated. Nevertheless, the equations performing the integration of afferent inputs for control of sweating proposed by Stolwijk and Hardy (1977) are included. The infrastructure necessary to develop a heat exposure model is therefore preserved. The concept in classical control theory of a 'set point' temperature, i.e. a temperature which the thermoregulatory system attempts to maintain, is employed. In fact, error signals are formed at each time step between temperatures in each tissue compartment and their initial values:
\[ E(i, j) = T(i, j) - T_{set}(i, j) \]  \hspace{1cm} \text{(4.36)}

where \( E(i, j) \) = error signal for shell \( j \), segment \( i \) (°C)
\( T(i, j) \) = instantaneous value of temperature (°C)
\( T_{set}(i, j) \) = set point temperature (°C)

The set point temperatures are listed in Table 4.3. Their values were obtained after the completion of model development by simulating immersion of the 'standard man' in water at a thermoneutral temperature. All thermoregulation mechanisms were inhibited so that resting conditions were in force. The model was run until the sum of surface and respiratory heat losses was equal to the resting rate of metabolic heat production. Using a water temperature of 34.4°C, the heat balance occurred after approximately four hours. The temperatures reached by all tissue and blood compartments were then recorded. The afferent error signals are integrated as in Stolwijk and Hardy's program to form two signals \( S_{\text{warm}} \) and \( S_{\text{cold}} \):

\[
S_{\text{warm}} = \sum_{i=1}^{15} E(i, 10) \cdot R_{\text{skin}(i)} \quad E(i, 10) > 0 \hspace{1cm} \text{(4.37)}
\]

\[
S_{\text{cold}} = \sum_{i=1}^{15} E(i, 10) \cdot R_{\text{skin}(i)} \quad E(i, 10) < 0 \hspace{1cm} \text{(4.38)}
\]

where \( R_{\text{skin}(i)} \) = a weighting factor proportional to the density of temperature receptors in the skin of segment \( i \)

The above signals, together with the temperature error in the head core \( E(1, 1) \), are used to obtain efferent signals for vasoconstriction and vasodilation separately in the skin:

\[
S_{\text{vcon}} = -5.0 \ E(1, 1) - 5.0 \ (S_{\text{warm}} - S_{\text{cold}}) \hspace{1cm} \text{(4.39)}
\]

\[
S_{\text{vdil}} = 32.5 \ E(1, 1) + 2.08 \ (S_{\text{warm}} - S_{\text{cold}}) \hspace{1cm} \text{(4.40)}
\]
Tissue set point and starting temperatures (before modification at run time):

<table>
<thead>
<tr>
<th>segment</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
<th>10</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>37.08</td>
<td>37.04</td>
<td>36.94</td>
<td>36.72</td>
<td>36.44</td>
<td>36.35</td>
<td>36.26</td>
<td>36.18</td>
<td>35.68</td>
<td>35.06</td>
</tr>
<tr>
<td>2</td>
<td>37.01</td>
<td>37.00</td>
<td>37.00</td>
<td>36.98</td>
<td>36.94</td>
<td>36.83</td>
<td>36.66</td>
<td>36.45</td>
<td>35.78</td>
<td>34.97</td>
</tr>
<tr>
<td>3</td>
<td>37.06</td>
<td>37.06</td>
<td>37.05</td>
<td>37.05</td>
<td>37.01</td>
<td>36.90</td>
<td>36.73</td>
<td>36.52</td>
<td>35.85</td>
<td>35.00</td>
</tr>
<tr>
<td>4</td>
<td>36.94</td>
<td>36.84</td>
<td>36.73</td>
<td>36.61</td>
<td>36.34</td>
<td>36.02</td>
<td>35.76</td>
<td>35.53</td>
<td>35.19</td>
<td>34.83</td>
</tr>
<tr>
<td>5</td>
<td>36.52</td>
<td>36.42</td>
<td>36.32</td>
<td>36.21</td>
<td>35.98</td>
<td>35.71</td>
<td>35.50</td>
<td>35.21</td>
<td>35.04</td>
<td>34.76</td>
</tr>
<tr>
<td>6</td>
<td>36.42</td>
<td>36.38</td>
<td>36.34</td>
<td>36.30</td>
<td>36.28</td>
<td>36.26</td>
<td>36.25</td>
<td>36.24</td>
<td>36.04</td>
<td>35.70</td>
</tr>
<tr>
<td>7</td>
<td>36.94</td>
<td>36.84</td>
<td>36.73</td>
<td>36.61</td>
<td>36.34</td>
<td>36.02</td>
<td>35.76</td>
<td>35.53</td>
<td>35.19</td>
<td>34.83</td>
</tr>
<tr>
<td>8</td>
<td>36.52</td>
<td>36.42</td>
<td>36.32</td>
<td>36.21</td>
<td>35.98</td>
<td>35.71</td>
<td>35.50</td>
<td>35.21</td>
<td>35.04</td>
<td>34.76</td>
</tr>
<tr>
<td>9</td>
<td>36.42</td>
<td>36.38</td>
<td>36.34</td>
<td>36.30</td>
<td>36.28</td>
<td>36.26</td>
<td>36.25</td>
<td>36.24</td>
<td>36.04</td>
<td>35.70</td>
</tr>
<tr>
<td>10</td>
<td>36.50</td>
<td>36.41</td>
<td>36.31</td>
<td>36.21</td>
<td>36.02</td>
<td>35.78</td>
<td>35.58</td>
<td>35.41</td>
<td>35.11</td>
<td>34.80</td>
</tr>
<tr>
<td>11</td>
<td>36.25</td>
<td>36.17</td>
<td>36.08</td>
<td>36.00</td>
<td>35.82</td>
<td>35.62</td>
<td>35.44</td>
<td>34.29</td>
<td>34.03</td>
<td>34.76</td>
</tr>
<tr>
<td>12</td>
<td>36.07</td>
<td>36.04</td>
<td>36.01</td>
<td>35.99</td>
<td>35.96</td>
<td>35.95</td>
<td>35.93</td>
<td>35.92</td>
<td>35.78</td>
<td>35.52</td>
</tr>
<tr>
<td>13</td>
<td>36.50</td>
<td>36.41</td>
<td>36.31</td>
<td>36.21</td>
<td>36.02</td>
<td>35.78</td>
<td>35.58</td>
<td>35.41</td>
<td>35.11</td>
<td>34.80</td>
</tr>
<tr>
<td>14</td>
<td>36.25</td>
<td>36.17</td>
<td>36.08</td>
<td>36.00</td>
<td>35.82</td>
<td>35.62</td>
<td>35.44</td>
<td>34.29</td>
<td>34.03</td>
<td>34.76</td>
</tr>
<tr>
<td>15</td>
<td>36.07</td>
<td>36.04</td>
<td>36.01</td>
<td>35.99</td>
<td>35.96</td>
<td>35.95</td>
<td>35.93</td>
<td>35.92</td>
<td>35.78</td>
<td>35.52</td>
</tr>
</tbody>
</table>

Starting temperatures (segments 1 to 15) before modification at run time:

Major arterial pools
36.70 36.70 36.70 36.66 36.57 36.49 36.66 36.57
36.49 36.62 36.49 36.35 36.62 36.49 36.35

Major venous pools
36.82 36.70 36.75 36.22 36.03 35.89 36.22 36.03
35.89 35.97 35.87 35.74 35.97 35.87 35.74

Visceral arterial pools
36.91 36.98 36.98 36.77 36.38 36.37 36.77 36.38
36.37 36.38 36.16 36.04 36.38 36.16 36.04

Visceral venous pools
36.93 36.99 37.03 36.78 36.37 36.78 36.37 36.78
36.37 36.37 36.14 36.03 36.37 36.14 36.03

Muscle arterial pools
36.31 36.72 36.79 35.91 35.63 36.26 35.91 35.63
36.26 35.70 35.55 35.94 35.70 35.55 35.94

Muscle venous pools
36.31 36.72 36.79 35.91 35.63 36.26 35.91 35.63
36.26 35.70 35.55 35.94 35.70 35.55 35.94

Skin arterial pools
35.26 35.25 35.27 34.94 34.88 34.82 34.94 34.88
34.82 34.99 34.94 35.65 34.99 34.94 35.65

Skin venous pools
35.14 35.09 35.11 34.88 34.81 34.74 34.88 34.81
34.74 34.88 34.84 35.57 34.88 34.84 35.57

TABLE 4.3. - STARTING AND SET-POINT TEMPERATURES (°C).

118
A resting or basal blood flow \( w_b(i,10) \) is required to each skin shell. These rates are dependent on the volume of tissue and are discussed in following paragraphs. The rate of blood flow to the skin of each segment is determined from:

\[
w(i,10) = w_b(i,10) + \frac{F_{vdil}(i) \cdot S_{vdil}}{1 + F_{vcon}(i) \cdot S_{vcon}} \cdot \frac{E(i,10)}{Q_{sbf}}
\]

where

- \( w(i,10) \) = flow rate in skin of segment \( i \) (ml s\(^{-1}\))
- \( F_{vdil}(i) \) = weighting factor for vasodilation effectors
- \( F_{vcon}(i) \) = weighting factor for vasoconstriction effectors

The values of \( R_{skin}, F_{vdil} \) and \( F_{vcon} \) are shown in Table 4.3. The term \( E(i,10)/Q_{sbf} \) is necessary to simulate the effect of local skin temperature on skin blood flow. Montgomery (1972) used a value of 6.0 for \( Q_{sbf} \), such that the rate of flow was halved for every 6°C fall in skin temperature. Charny et al have proposed a value of 4.5. However, there is evidence from the validations described in Chapter 6 of this thesis that a value of 3.0 may be appropriate for subjects with a high percentage body fat, since their skin temperatures are generally low. Accordingly, \( F_{sbf} \) is made a function of body mass:

\[
F_{sbf} = 6.0 - 3.0(M-50.0)
\]

This formula has not been validated, owing to difficulties in measuring blood flow in the skin, especially when immersed in water. However, its inclusion does improve the accuracy of predictions of fall in core temperature for a group of subjects whose body masses span a wide range.

The resting or basal values of both metabolic rate and blood flow are adapted from those used by Gordon et al (1976). The metabolic heat generated per unit volume of
<table>
<thead>
<tr>
<th>Head</th>
<th>Thorax</th>
<th>Abd.</th>
<th>Thigh</th>
<th>Calf</th>
<th>Foot</th>
<th>U.arm</th>
<th>F.arm</th>
<th>Hand</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_{\text{skin}}(i) =$ fraction of thermoreceptors in each segment</td>
<td>0.210</td>
<td>0.210</td>
<td>0.210</td>
<td>0.060</td>
<td>0.040</td>
<td>0.015</td>
<td>0.025</td>
<td>0.025</td>
</tr>
<tr>
<td>$F_{\text{vdil}}(i) =$ fraction of vasodilation effectors in each segment</td>
<td>0.132</td>
<td>0.161</td>
<td>0.161</td>
<td>0.069</td>
<td>0.046</td>
<td>0.050</td>
<td>0.024</td>
<td>0.024</td>
</tr>
<tr>
<td>$F_{\text{vcon}}(i) =$ fraction of vasoconstriction effectors in each segment</td>
<td>0.050</td>
<td>0.075</td>
<td>0.075</td>
<td>0.015</td>
<td>0.010</td>
<td>0.175</td>
<td>0.013</td>
<td>0.013</td>
</tr>
<tr>
<td>$F_{\text{mb}}(i) =$ skin resting metabolic rate per unit volume (W m$^{-3}$)</td>
<td>370.4</td>
<td>348.2</td>
<td>348.2</td>
<td>308.3</td>
<td>308.3</td>
<td>333.3</td>
<td>312.5</td>
<td>312.5</td>
</tr>
<tr>
<td>$F_{\text{bfbs}}(i) =$ skin resting blood flow per unit volume (ml s$^{-1}$ m$^{-3}$)</td>
<td>1482</td>
<td>1667</td>
<td>1667</td>
<td>660</td>
<td>660</td>
<td>3472</td>
<td>1157</td>
<td>1157</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Bone</th>
<th>Brain</th>
<th>Lung</th>
<th>Viscera</th>
<th>Muscle</th>
<th>Fat</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_{\text{mb}} =$ resting metabolic heat generated per unit volume (Wm$^{-3}$)</td>
<td>0.0</td>
<td>8769.6</td>
<td>368.9</td>
<td>5910.3</td>
<td>0340.8</td>
</tr>
<tr>
<td>$F_{\text{bfbs}} =$ resting blood flow per resting metabolic rate (ml s$^{-1}$ W$^{-1}$)</td>
<td>0.0</td>
<td>0.836</td>
<td>0.669</td>
<td>1.108</td>
<td>0.287</td>
</tr>
</tbody>
</table>

**TABLE 4.4 - METABOLIC WEIGHTING FACTORS AND RESTING VALUES**
each tissue type is shown in Table 4.4. These figures are used in combination with the volume fractions calculated previously to estimate the metabolic rate of each tissue shell. The basal blood flow $BFB(i,j)$ is made proportional to the metabolic rate, again using Gordon's data, except in the case of the skin shells. Here, the blood supply has a thermoregulatory function even under resting conditions, rather than purely sustaining life in the tissue. It is therefore made proportional to skin mass after Stolwijk & Hardy (1977).

Individual metabolic rates are summed to find the total rate for the body. If the actual rate for the particular subject being simulated is known, then the individual rates are modified in this model in the following way:

$$M_b(i,j) = M_b(i,j) \cdot \frac{M_r \cdot A}{Me}$$

(4.43)

where $M_b(i,j) =$ rate of metabolic heat generation for tissue shell $j$ in segment $i$ (W)

$M_r =$ real metabolic rate (Wm$^{-2}$)

$A =$ surface area of body (m$^2$)

$Me =$ sum of original estimates for shells (W)

The total metabolic rate naturally changes as a result of immersion in cold water. Shivering and exercise both raise the rate of heat production in the muscle shells of all segments. The Stolwijk model contains an expression for the additional metabolism due to shivering, which is added to the basal rate. This has taken a number of forms, the most applicable to cold water immersion being the one employed by Montgomery (1972):

$$M_{shiv} = 21.0 \cdot E(1,1) \cdot S_{cold}$$

(4.44)

Although this expression uses a product of the change in head core and mean skin temperature, it tends to
underestimate the metabolic rate measured in immersion experiments, as described in Chapter 5. A number of other workers have developed expressions, based on their own experimental results, five of which have been evaluated by Mekjavik and Morrison (1986). Their conclusion was that the metabolic control equation of Hayward et al (1977) provided the best overall prediction for 10 male subjects immersed in water at 10°C. The Hayward controller has therefore been adopted in the present model:

\[ M_e = 0.0314 \times 10^2 (T_s - 42.4) (T_c - 41.4) \]  

The total metabolic rate of the muscle shells is a function of the amounts of shivering and exercise taking place. Sowood (1988) has shown that this function is not simply the sum of the two contributions. It is found that:

If \( M_{rc} < 0.816 M_{ew} \),
then \( M_e = 0.438 M_{rc} + 0.901 M_{ew} \) \hspace{1cm} (4.46)

If \( M_{rc} > 0.816 M_{ew} \),
then \( M_e = M_{rc} + 0.442 M_{ew} \) \hspace{1cm} (4.47)

where \( M_e = \) total metabolic rate (W)  
\( M_{rc} = \) metabolic rate in cold water at rest (W)  
\( M_{ew} = \) metabolic rate when working in thermoneutral conditions (W)

Since it is the total metabolic rate which is predicted, the basal rates of viscera, fat and skin must be subtracted proportionally in order to obtain the total muscular components \( M_{ms} \) and \( M_{mw} \). The metabolic heat generated by working and by shivering is then distributed among the 60 muscle shells according to the fractions in Table 4.5, adapted from Stolwijk and Hardy (1977):
Figures apply to each of four muscle shells in each segment, repeated for each limb separately.

**TABLE 4.5 - DISTRIBUTION OF WORK AND SHIVERING METABOLISM**

The metabolic rate of each muscle shell is therefore found from:

\[ M_m(i,j) = F_{\text{chill}} \cdot M_{ms} + F_{\text{work}} \cdot M_{mw} (1 - \eta/100) \] (4.48)

where \( M_m(i,j) \) = metabolic rate of muscle shell \( j \) in segment \( i \) (W)

\( \eta \) = efficiency of muscles in performing external work (\%)

The factor \((1 - \eta/100)\) is the fraction of work which is evident as internal metabolic heat. The blood flow required by a muscle compartment, however, must be calculated on the basis of the total work and shivering performed:

\[ w_{mr}(i,j) = 0.322 (F_{\text{chill}} \cdot M_{ms} + F_{\text{work}} \cdot M_{mw}) \] (4.49)

where \( w_{mr} \) = required blood flow to a muscle shell (ml s\(^{-1}\))

The figure 0.322 is based on the assumption that each ml of arterial blood contains approximately 0.2 ml of oxygen. If this is completely scavenged by the muscle, it can produce 3.1 Joules of heat. A flow rate of \( 1/3.1 \) = 0.322 ml s\(^{-1}\) is therefore required for each Watt of muscular metabolism. When there is a change in the rate of working or shivering, the blood flow does not immediately change to the required value according to Wissler (1985). Instead, it follows a first-order time
lag of the form:

\[ w_m(i,j) = w_{mp}(i,j) \cdot e^{-\Delta t/\tau} + w_{mr}(i,j) \cdot (1 - e^{-\Delta t/\tau}) \quad (4.50) \]

where \( w_{mp}(i,j) \) = value of muscle shell blood flow at previous time step (ml s\(^{-1}\))
\( \Delta t \) = time increment (s)
\( \tau \) = time constant (s) - typically 60s

Thus, on an increase of work, an oxygen debt builds up in the working muscle but is later repaid. A similar first order lag is applied to the blood flow in the skin but here, the time constant is only of the order of 5s. Blood flows in the fat and visceral organs are assumed to remain at their basal values. The 10 shell flows in each segment are summed to form a segmental blood flow \( w_{seg} \) which, in the case of the head, hands and feet, is the flow in the major arteries and veins:

\[ w_{seg}(i) = \sum_{j=1}^{10} w(i,j) \quad (4.51) \]

In the major blood vessels of the other segments, the flows of more distal segments are added to the segment's own flow to obtain the total, as shown in Figure 4.10. The arterial or venous flow in the thorax is thus the total cardiac flow.

Respiratory heat exchange, although not under control of the thermoregulatory system, must be evaluated in order that heat balance equations may be solved. It is reasonable to assume that the respiration rate is driven by the demand for oxygen, which is proportional to the metabolic rate of the body. The first order lag in Equation 4.50 is applied to changes in the metabolic rate so that the respiration rate will be proportional to the total blood flow rate. There are two components to the heat loss - dry heat exchange and loss by evaporation of water vapour.
Venous blood return

Arterial blood supply

Segment

\( W_{\text{seg}} \)

\( i-1 \)

\( W_{\text{seg}} \)

\( i \)

\( W_{\text{seg}} \)

\( i+1 \)

(One artery and vein shown per segment for clarity)

FIG 4.10  COMBINATION OF SEGMENTAL BLOOD FLOWS
The following expressions have been developed by Montgomery (1972), who gives a detailed analysis. According to Ruch and Patton (1965) approximately 1.0 l hr\(^{-1}\) of oxygen are consumed for every 5.6\(\text{W}\) of metabolism. Asmussen and Nielsen (1946) estimate that 22.0 l of air are expired for every 1 litre of oxygen consumed in exercise. The minute-volume of expired air is therefore:

\[
V_{\text{emin}} = \frac{22.0 \ M_b}{60.0 \times 5.62}
\]  

where \(V_{\text{emin}}\) = volume of air expired per minute (1 min\(^{-1}\))

\(M_b\) = lagged metabolic rate for breathing (W)

The inspired air is assumed to be at ambient air temperature \(T_a\):

\[
T_i = T_a + 273 \ (K)
\]  

The expired air is assumed to contain contributions from the head core, the thorax core and the venous blood in the thorax:

\[
T_e = \frac{T(1,1) + T(2,1) + T_v(2)}{3} + 273 \ (K)
\]  

The product of density and specific heat for the expired air must be corrected to expiration temperature and for variations in barometric pressure:

\[
\rho_a c_a = \frac{0.2905 \ P_b}{101324 \ 293/T_e /1000} \ (J \ l^{-1} K^{-1})
\]  

where \(\rho_a\) = corrected density of air (kg m\(^{-3}\))

\(c_a\) = corrected specific heat of air (J kg\(^{-1}\) K\(^{-1}\))

\(P_b\) = barometric pressure (Pa)

0.2905 = \(\rho_a c_a\) at standard temperature & pressure

The water vapour pressure in expired air is estimated in the following way. An integer \(N\) is determined from the following expression and is used to address a look-up table containing vapour pressures:
\[ N = \left| \frac{T_e}{5} \right| + 1 \] (4.56)

<table>
<thead>
<tr>
<th>N</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pe</td>
<td>610.5</td>
<td>872.3</td>
<td>1227.8</td>
<td>1704.9</td>
<td>2337.8</td>
<td>3167.2</td>
<td>4242.8</td>
<td>5622.9</td>
</tr>
<tr>
<td>N</td>
<td>9</td>
<td>10</td>
<td>11</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pe</td>
<td>7375.9</td>
<td>9583.2</td>
<td>12333.6</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

where \( P_e \) = vapour pressure in expired air (Pa)

The rate of loss of water in the expired air can be calculated from:

\[
M_{we} = \frac{V_e}{3600} \frac{P_b m_e}{R T_e} \left( \frac{P_e}{P_b} \right) \frac{m_w}{m_e} \quad (g \ s^{-1}) \quad (4.57)
\]

where \( m_e \) = molecular weight of expired air = 99.962 g/mol

\( m_w \) = molecular weight of water = 18.01534 g/mol

\( R \) = ideal gas constant 8315 1 Pa mol^{-1} K^{-1}

The respiratory heat loss can then be calculated from:

\[
Q_{rl} = \frac{V_e}{3600} p_a c_a (T_e - T_i) + \lambda M_{we} \quad (W) \quad (4.58)
\]

where \( \lambda \) = latent heat of water = 2428 kJ kg^{-1}

There is also a quantity of heat generated by the effort of breathing air while exercising. Milic-Emili and Petit (1960) have shown that this may be estimated from the equation:

\[
Q_{rg} = 7.8 \times 10^{-3} \ v_{emin}^2 + 7.2 \times 10^{-5} \ v_{emin}^3 \quad (W) \quad (4.59)
\]

The net heat exchange is therefore:

\[ Q_r = Q_{rl} + Q_{rg} \quad (Q_{rl} \ is \ normally \ negative) \]

These variables are evaluated by the model at each time step.
4.6. Heat transfer

Before the beginning of the first time step, the model gives initial values to a number of variables. The tissue and blood pool temperatures are modified from their standard values Table 4.3, if the rectal temperature $T_r$ at the start of the exposure period is known. This is achieved simply by multiplying all temperatures by the fraction $T_r/T_{set}(3,1)$.

The skin temperatures $T(i,10)$ are referred to the centres of mass of the skin shells and do not indicate the true surface temperatures. These are calculated in the manner indicated in Figure 4.11. The rate of change of temperature with radius between shells 8 and 9 and between shells 9 and 10 is calculated:

\[
\frac{\delta T(i,8)}{\delta r} = \frac{T(i,9) - T(i,8)}{r_{cm}(i,9) - r_{cm}(i,8)} \tag{4.60}
\]

\[
\frac{\delta T(i,9)}{\delta r} = \frac{T(i,10) - T(i,9)}{r_{cm}(i,10) - r_{cm}(i,9)} \tag{4.61}
\]

The second rate of change of temperature with radius at shell 9 may then be found:

\[
\frac{\delta_2 T(i,9)}{\delta r^2} = \frac{[\delta T(i,9)/\delta r - \delta T(i,8)/\delta r]}{r(i,10) - r(i,9)} \tag{4.62}
\]

The rate of change in shell 10 can be estimated:

\[
\frac{\delta T(i,10)}{\delta r} = \frac{\delta T(i,9)}{\delta r} + \frac{\delta_2 T(i,9)}{\delta r^2} [r(i,10) - r(i,9)] \tag{4.63}
\]

Finally, an estimate of surface temperature is made:

\[
T_s(i) = T(i,10) + \frac{\delta T(i,10)}{\delta r} [r(i,10) - r_{cm}(i,10)] \tag{4.64}
\]

A similar procedure is followed to obtain the surface temperatures of clothing.

Although the current use of the model is to simulate cold
FIG 4.11 ESTIMATION OF SURFACE TEMPERATURE
water immersion, the infrastructure necessary to develop an air-exposure model has been included. In either case, the subject's head is in air. The fluid properties of water and air vary with temperature and compensations must be made to preserve accuracy. These are calculated for a temperature which is the mean of the surface and bulk fluid temperatures:

\[ T_f = \frac{T_s + T_{env}}{2} \]  \hspace{1cm} (4.65)

where

- \( T_f \) = fluid temperature at body surface (°C)
- \( T_s \) = mean area-weighted surface temperature (°C)
- \( T_{env} \) = environmental temperature (°C)

The density of water is taken as \( \rho_w = 1000 \text{ kg m}^{-3} \) and specific heat \( c_w = 4185.5 \text{ J kg}^{-1} \text{ K}^{-1} \) (Kaye & Laby 1973). The viscosity is calculated from the empirical equation:

\[ \mu_w = \frac{0.431}{(24.0 + T_f)} \text{ (kg m}^{-1} \text{ s}^{-1}) \]  \hspace{1cm} (4.66)

and the conductivity:

\[ k_w = 0.56 + 0.0014 T_f \text{ (W m}^{-1} \text{ K}^{-1}) \]  \hspace{1cm} (4.67)

Prandtl's number for water may therefore be calculated:

\[ N_{Prw} = \frac{c_w \cdot \mu_w}{k_w} \]  \hspace{1cm} (4.68)

and Reynold's number:

\[ N_{Rew} = \frac{\rho_w \cdot v_w \cdot d_{seg}}{\mu_w} \]  \hspace{1cm} (4.69)

where

- \( v_w \) = velocity of water w.r.t. subject (ms\(^{-1}\))
- \( d_{seg} \) = diameter of body segment (m)

For air, the specific heat \( c_a = 1040 \text{ J kg}^{-1} \text{ K}^{-1} \). The following expressions for density, viscosity and conductivity have been used by the National Aeronautics
and Space Administration (1966) in defining the 'US Standard Atmosphere':

\[
\rho_a = \frac{m_a P_b}{R(T_f+273)} \quad (\text{kg m}^{-3}) \tag{4.70}
\]

where \( m_a \) = molecular weight of air = 29.8 g mol\(^{-1}\)

\[
\mu_a = \frac{1.458 \times 10^{-6} (T_f+273)^{1.5}}{(T_f+273) + 110.4} \quad (\text{kg m}^{-1} \text{s}^{-1}) \tag{4.71}
\]

\[
k_a = \frac{0.6325 \times 10^{-6} (T_f+273)^{1.5}}{(T_f+273) + 245.4 \times 10^{-12}/(T_f+273)} \quad (\text{W m}^{-1} \text{K}^{-1}) \tag{4.72}
\]

McAdams (1954) gives the following formulae for the heat transfer coefficients to a fluid flowing past a sphere and parallel to a cylinder:

\[
h_s = k/d_{seg} 0.6 N_R^{0.5} N_P^{0.31} \quad (\text{W m}^{-2} \text{°C}^{-1}) \tag{4.73}
\]

\[
h_c = k/d_{seg} 0.35 N_P^{0.3} + 0.56 N_R^{0.52} N_P^{0.3} \quad " \tag{4.74}
\]

These equations are used to determine the heat flow at the surface of each segment. This heat flow must, in each case, be equalled by an internal conductive heat flow in the tissue immediately beneath the surface, governed by a heat transfer coefficient \( K \) as in Equation 4.27. The total heat transfer coefficient between the centre-of-mass radius \( r_{cm(i,10)} \) and the fluid is therefore found from the harmonic mean of the conductive and convective coefficients:

\[
H(i) = K(i).h(i)/[K(i) + h(i)] \tag{4.75}
\]

and

\[
U(i) = H(i).A_{seg(i)} \tag{4.76}
\]

where \( U(i) \) = total surface heat transfer coefficient for segment \( i \) (\text{W °C}^{-1})

\( A_{seg(i)} = \) surface area of segment (m\(^2\))
Equation 4.76 enables the surface heat flows to be evaluated from a knowledge of skin and fluid temperatures.

Conduction of heat normally takes place from the core to the periphery of each segment. The quantity of heat conducted to or from each segment is found from the expression:

$$Q_{c}(i,j) = K(i,j) \cdot [T(i,j) - T(i,j+1)] \quad \text{(W)} \quad (4.77)$$

Blood flow plays a major role in convective heat transfer within the body. In the model, calculations are made of the heat transferred to each of the 30 major venous and arterial pools and the 90 minor pools in the viscera, muscle and skin. Each of these has its own assigned temperature, which is calculated from a heat balance equation. The scheme adopted for blood heat flow is illustrated in Figure 4.12.

Blood flows from each major arterial pool into the three minor arterial pools in the segment, where it exchanges heat with the surrounding tissue and with the minor veins. From a minor arterial pool, it flows through the capillaries and returns to the minor venous pool, where it again exchanges heat with tissue and with the minor arteries. It then returns to the major vein. The quantity of heat transported by blood away from a pool is expressed in equations whose general form is:

$$Q_{b}(i) = w_{b}(i) \cdot T_{b}(i) \cdot \rho_{b} \cdot c_{b} \times 10^{-6} \quad (4.78)$$

where
- $Q_{b}(i)$ = heat transported from pool i (W)
- $w_{b}(i)$ = blood flow rate in pool i (ml s$^{-1}$)
- $\rho_{b}$ = density of blood (kg m$^{-3}$)
- $c_{b}$ = specific heat of blood (J kg K$^{-1}$)
Capillaries and other vessels of dia<100 μm

Axis

1 - Major artery
2 - Major vein
3 - Minor visceral artery
4 - Minor visceral vein
5 - Minor muscle artery
6 - Minor muscle vein
7 - Minor skin artery
8 - Minor skin vein

FIG 4.12 SCHEME OF BLOOD DISTRIBUTION
An equation such as this is solved for each blood flow in the simulation. Heat balance equations are then solved to indicate the net heat flow into and out of each tissue compartment and blood pool, using the conductive and convective exchanges developed above. The exact form of the equation depends upon the location of the compartment it applies to. As examples, the equations for a core tissue compartment, a skin tissue compartment, a major arterial pool and a minor arterial pool are shown.

Core:

\[ Q_f(i,1) = Q_m(i,1) - Q_r/12 + Q_a(i,1) - Q_v(i,1) - Q_c(i,1) + Q_{atl}(i)/4 - Q_{vtl}/4 + Q_{atv}(i)/4 - Q_{vtv}(i)/4 \]  

(4.79)

where

- \( Q_f \) = net heat flow to or from a shell (W)
- \( Q_m \) = metabolic heat generated (W)
- \( Q_r/12 \) = proportion of respiratory heat exchange (in core shells of head and thorax) (W)
- \( Q_a \) = heat supplied by arterial blood (W)
- \( Q_v \) = heat lost to venous blood (W)
- \( Q_c \) = heat conducted away from shell (W)
- \( Q_{atl}/4 \) = proportion of heat exchanged with large arteries in this shell (W)
- \( Q_{vtl}/4 \) = proportion of heat exchanged with large veins in this shell (W)
- \( Q_{atv}/4 \) = proportion of heat exchanged with small visceral arteries in this shell (W)
- \( Q_{vtv}/4 \) = proportion of heat exchanged with small visceral veins in this shell (W)

Skin:

\[ Q_f(i,10) = Q_m(i,10) - Q_e(i) + Q_a(i,10) - Q_v(i,10) + Q_c(i,9) + Q_{ats}(i) - Q_{vts}(i) - H(i)[T(i,10) - T_w] \]  

(4.80)

where

- \( Q_f(i,10) \) = heat flow to or from a skin shell (W)
- \( Q_e(i) \) = heat lost by evaporation at skin surface (zero for immersion) (W)
- \( Q_{ats}(i) \) = heat exchanged between tissue and small
skin arteries (W)

\[ Q_{vts}(i) = \text{heat exchanged between tissue and small skin veins (W)} \]

Major arterial pool in thorax:

\[
Q_{fal}(2) = \{w_{seg}(2) \cdot T_{al}(2) - [w_{seg}(3) + w_{seg}(10) + w_{seg}(13) + w_{seg}(1)] \cdot T_{al}(2) \} \cdot \rho_b \cdot c_b \times 10^{-6}
\]

\[
- H_{al}(2) \cdot [T_{al} - T_{visc}(i)] - Q_{aseg(i,j)} \quad (4.81)
\]

where \( Q_{fal}(2) \) = net heat flow to or from thorax arterial blood pool (W)

\( w_{seg}(i) = \text{blood flow in major vessel } i \text{ (ml s}^{-1}\text{)} \)

\( T_{al}(2) = \text{temperature of large arterial blood pool in thorax (°C)} \)

\[
Q_{aseg(i)} = \sum_{j=1}^{10} Q_{a(i,j)}, \text{the total arterial heat flow to tissue in segment } i \text{ (W)}
\]

\[
T_{visc}(i) = \sum_{j=1}^{4} T(i,j), \text{the mean visceral temp. (°C)}
\]

Minor arterial pool in skin:

\[
Q_{fas}(i) = Q_{as}(i) - Q_{a(i,10)} - Q_{ats}(i) - Q_{ccs}(i) \quad (4.82)
\]

where \( Q_{as} = \text{heat supplied to arterial pool in skin by blood from large arteries (W)} \)

\( Q_{ats} = \text{heat exchanged between arterial pool and tissue (W)} \)

\( Q_{ccs} = \text{counter current heat exchanged between arteries and veins in skin (W)} \)

The principles illustrated above are used in the construction of heat balance equations for all other tissue regions arterial and venous blood pools in the model.
The final region in which heat transfer must be evaluated is the clothing layer. In models simulating heat stress in an air environment, clothing requires a detailed treatment. Evaporation of sweat into air layers between the fabric and the permeability of fabric to water vapour will affect the overall thermal conductance and must therefore be evaluated. In cold water immersion sweating is not expected to occur and air layers will be minimal, owing to hydrostatic pressure. It is therefore considered adequate to represent clothing by a single additional shell at the skin surface.

A thermal conductivity is assigned to this layer equal to that of air, since the air layer trapped under impermeable layers of clothing is the most significant component of insulation. This conductivity is calculated by using Equation 4.72 and substituting in the water temperature. The conductance of the clothing on each segment is obtained from the clothing insulation value in Togs simply by dividing by 10:

\[ K_{\text{clo}}(i) = I_{\text{clo}}(i)/10.0 \]  

(4.83)

The clothing thickness is then:

\[ X_{\text{clo}}(i) = K_{\text{clo}}(i) \cdot k_a \]  

(4.84)

The segment diameters are increased by this amount and the new diameters used in obtaining heat transfer coefficients at the clothing surface, in a similar manner to Equations 4.73 to 4.76. Knowing the conductances of the skin and clothing shells, the method of Equation 4.28 is again used to find the mean. The heat flow from skin to water through the clothing can thus be evaluated if the temperature of each is known.
4.7. **Mathematical integration**

A heat flow entering or leaving a tissue or blood compartment will cause an increase or decrease in temperature. The rate of increase or decrease is evaluated by dividing the heat flow by the thermal capacity of the compartment. The thermal capacities are obtained from the volumes of each compartment and the data for density and specific heat data from Table 4.1:

\[ C(i,j) = \Delta \text{mass}(i,j) \cdot \rho(i,j) \]  

where \( C = \) thermal capacity \((\text{J} \, ^\circ\text{C}^{-1})\)  
\( \Delta \text{mass} = \) mass of compartment \((\text{kg})\)  
\( \rho = \) density \((\text{kg} \, \text{m}^{-3})\)

Similar calculations are made for the blood pools. In the case of the core tissue shells, in which more than one type of tissue is present, the thermal capacities are calculated on a proportional basis. The rates of change of temperature are then:

\[ \dot{T}(i,j) = \frac{Q_f(i,j)}{C(i,j)} \]  

where \( \dot{T} = \) rate of change of temperature \((^\circ\text{C} \, \text{s}^{-1})\)  
\( Q_f = \) heat flow \((\text{W})\)

Since the set of differential equations to be solved is of the first order, it is possible to use the following trapezoidal routine, as originally proposed by Stolwijk and Hardy (1966). More advanced routines exist but are generally applied to higher order systems. To obtain the temperature of a compartment one time instant in advance, the rate of change is multiplied by the time step and the result added to the previous temperature:

\[ T_{\text{new}} = T_{\text{old}} + \dot{T}_{\text{old}} \cdot \Delta t \]  

(4.87)
A time step of 1 minute is chosen initially. However, if the time step is too large, the prediction will be inaccurate. Instability can also occur. To avoid this, the temperature changes are limited to 0.1 °C by examining all the rates of change and successively reducing the size of the time step to enforce this limit:

\[ \Delta t = \min\{0.1/T\} \quad (4.88) \]

Having found the minimum value, Equation 4.87 is applied to all tissue compartments and blood pools, also to clothing shells, if present. The initial estimates of surface temperatures provided by Equation 4.64 are used in the model to determine fluid properties in the absence of prior knowledge. At the end of each time step, conductive heat flows in the skin shells and convective flows at the surface are known. These may be equated to obtain an equation in which the only unknown is surface temperature:

\[ K(i,10) \cdot [T(i,10)-T_{\text{surf}}(i)] = U(i) \cdot [T_{\text{surf}}(i)-T_w] \quad (4.89) \]

Transposing,

\[ T_{\text{surf}}(i) = \frac{K(i,10) \cdot T(i,10) + U(i) \cdot T_w}{K(i,10) + U(i)} \quad (4.90) \]

A new area-weighted mean surface temperature may then be calculated. Finally, the model accounts for the changes in thermal conductivity which occur in skin, fat and muscle as their temperature changes. This is estimated by applying Equation 4.67 to the water content in these tissues. The previous value of conductivity is saved and modified:

\[ k_{\text{new}}(i,j) = k_{\text{old}}(i,j) + 0.0014 \cdot k_{\text{old}}(i,j) \cdot \frac{[T(i,j)-T_{\text{set}}(i,j)]}{0.56} \quad (4.91) \]
Core conductivities are not modified, since the relatively small temperature changes in core shells do not justify it.

The foregoing description completes the physical and physiological description of the model. A further description of the progress made in computational terms appears in Volume 2 of the thesis. In the following chapter, the comparisons between model predictions and experimental observations used to investigate the model’s performance are described.
5. VALIDATION OF THE MODEL

5.1. Description of experimental data

The validation of the model is a process which has, to some extent, proceeded throughout the development phase. As new features have been introduced, comparisons have been made between the model output and experimental data to assess the effect. However, these comparisons were made informally on a limited number of runs. In order to validate the completed model, it has been necessary to obtain sound experimental data covering as wide a range of conditions as possible, within those the model is intended to simulate.

The model is currently configured to simulate nude or clothed immersion in cold water. The nude configuration has been investigated in detail deliberately, since it results in the highest heat flows and therefore potentially the greatest inaccuracies. It was felt to be important to establish an adequate nude model before simulating clothing, since errors in both areas may cancel to give a falsely accurate result. Simulations of clothed immersion have been validated by comparison with a small number of experiments. Outline code for air exposure is also included but is not complete, in that it does not account for heat loss by sweat evaporation. One of the primary aims has been to provide a structured basis on which future workers may extend new ideas and deal with new situations.

The conclusion to the project is therefore the formal comparison of the predictions provided by the model with existing experimental data. One of the most comprehensive sets of experiments in recent years has been performed by Sowood (1988) at the RAF Institute of Aviation Medicine, in which the author has participated as a volunteer subject. A mathematical analysis of the
workload provided by the underwater cycling exercise used by Sowood has also been carried out and is reported in Appendix 1. It has therefore been profitable to base the model validation mainly on data from these experiments, which are described in the following paragraphs.

As discussed in Chapter 1, the length of time for which survivors in the sea can remain conscious and, ultimately, remain alive is determined to a large extent by their ability to maintain a central body temperature above 34°C. This ability is influenced by two processes of metabolic heat generation - exercise (swimming, for example) and shivering. The relationship between these two processes has been investigated in the Sowood experiments. Fifteen volunteer male subjects performed exercising and non-exercising immersions in water at 12, 18 and 24°C and also under thermo-neutral conditions (>30°C). The subjects wore only swimming trunks and immersion was to the level of the neck.

The morphological characteristics of the 15 subjects are shown in Table 5.1. It may be seen that there is a wide variation between individual body masses and fat contents. Figure 5.1 shows that the set of masses spans the range from less than the first percentile to greater than the 99th percentile in a survey of 2000 RAF air crew carried out by Bolton et al (1973).

The experimental procedure consisted of instrumentation of the subjects, followed by 20 min rest in air at 21°C. The subjects then entered the pool and remained for 2 hours or until rectal temperature fell below 35°C, unless the experiment was terminated through discomfort or medical concern. Figure 5.2 shows a view from above the water surface of a subject undergoing immersion and Figure 5.3 shows the same subject as seen through an observation window below the surface.
FIGURE 5.1 - DISTRIBUTION OF BODY MASS
FIG 5.2 - A SUBJECT UNDERGOING VOLUNTARY IMMERSION (UPPER VIEW)
FIG 5.3 – A SUBJECT UNDERGOING VOLUNTARY IMMERSION (LOWER VIEW)
<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Mass (kg)</th>
<th>Height (m)</th>
<th>% fat</th>
<th>Mass/Ht²</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>68.8</td>
<td>1.83</td>
<td>7.4</td>
<td>20.5</td>
</tr>
<tr>
<td>2</td>
<td>50.5</td>
<td>1.71</td>
<td>7.4</td>
<td>17.3</td>
</tr>
<tr>
<td>3</td>
<td>73.0</td>
<td>1.80</td>
<td>9.5</td>
<td>22.5</td>
</tr>
<tr>
<td>4</td>
<td>101.4</td>
<td>1.78</td>
<td>19.5</td>
<td>32.0</td>
</tr>
<tr>
<td>5</td>
<td>79.5</td>
<td>1.88</td>
<td>6.8</td>
<td>22.5</td>
</tr>
<tr>
<td>6</td>
<td>84.2</td>
<td>1.73</td>
<td>17.2</td>
<td>28.1</td>
</tr>
<tr>
<td>7</td>
<td>84.0</td>
<td>1.88</td>
<td>22.2</td>
<td>23.8</td>
</tr>
<tr>
<td>8</td>
<td>60.5</td>
<td>1.79</td>
<td>9.0</td>
<td>18.9</td>
</tr>
<tr>
<td>9</td>
<td>72.0</td>
<td>1.68</td>
<td>23.6</td>
<td>25.5</td>
</tr>
<tr>
<td>10</td>
<td>101.4</td>
<td>1.81</td>
<td>21.4</td>
<td>31.0</td>
</tr>
<tr>
<td>11</td>
<td>73.4</td>
<td>1.94</td>
<td>11.0</td>
<td>19.5</td>
</tr>
<tr>
<td>12</td>
<td>73.5</td>
<td>1.83</td>
<td>10.0</td>
<td>22.0</td>
</tr>
<tr>
<td>13</td>
<td>70.1</td>
<td>1.78</td>
<td>10.7</td>
<td>22.1</td>
</tr>
<tr>
<td>14</td>
<td>64.0</td>
<td>1.65</td>
<td>8.2</td>
<td>23.5</td>
</tr>
<tr>
<td>15</td>
<td>92.6</td>
<td>1.82</td>
<td>26.3</td>
<td>28.0</td>
</tr>
<tr>
<td>Mean</td>
<td>76.5</td>
<td>1.80</td>
<td>13.7</td>
<td>23.8</td>
</tr>
<tr>
<td>S.D.</td>
<td>12.5</td>
<td>0.08</td>
<td>7.1</td>
<td>4.2</td>
</tr>
</tbody>
</table>

Mean age of subjects 29.5 years s.d. 7.2 years.

Percentage body fat is the ratio of total fat volume to total body volume, obtained by magnetic resonance imaging.

**TABLE 5.1 - CHARACTERISTICS OF SUBJECTS (SOWOOD 1988)**
For each subject, the oxygen consumption was measured both at rest and while exercising. Hence, it was possible for metabolic rate to be inferred. In addition, rectal and aural temperatures were monitored by means of thermistors, to indicate core temperature. Surface heat flow and temperature were also monitored at 9 sites on the body surface by means of thermopile heat flow transducers. The immersions were carried out between 09.00 and 12.00 hrs each day, no subject being immersed more than once in any week. Of the 15 subjects, all participated in the 18°C experiments, 14 took part in the 24°C immersions and 7 subjects were immersed in 12°C water.

The exercise consisted of pedalling a Monark unicycle, modified for underwater use as suggested by Shapiro et al (1981). Three vanes were attached to one face of the main wheel and the friction belt was removed. The fluid drag provided by the vanes replaced the normal frictional workload. It was therefore not possible to obtain a direct measurement of the workload, the only available method being inference from measurements of the subject’s rate of uptake of oxygen (VO₂).

Five appropriate rates of pedalling were determined under thermo-neutral conditions, spanning the range for each subject from rest to VO₂max, which had previously been determined during a treadmill test. Four of these rates were used for each subject during the later, colder immersions. It was desirable that the work load for a given pedalling rate should be the same at the lower temperatures as at 30°C, so that any change in metabolic rate could be attributed to shivering. The analysis of fluid drag in Appendix 1 was carried out to provide an estimate of the dependence of workload on pedalling rate and water temperature.
The analysis, while attempting to provide as precise an estimate as possible of workload and its temperature dependence, is derived from input data which may be unreliable in some cases. This is particularly true of the efficiencies of the cycle mechanism and the muscles and of the degree of circulation affecting relative velocity between the wheel and surrounding water. Nevertheless, by making appropriate estimates, it has been possible to obtain good agreement between the predicted and measured mean metabolic rates for a range of subjects. The predicted variation with pedalling speed agrees well with the variation observed at 30°C. It is therefore possible to predict with reasonable confidence the effect on workload of reducing the water temperature. The analysis suggests that the variation between 12°C and 30°C is less than 6%, thus allowing valid conclusions to be drawn on the relative effects of exercise and shivering thermogenesis in maintaining body temperature.

5.2. Methods of comparison between model and experiment

Since it is the temperature of the body core which determines the survival time, the most appropriate comparison is between the core temperatures predicted by the model and those measured during the experiments. Of the two core temperature measurements made experimentally, the aural temperature proved less reliable. This is thought to be due in part to the lack of insulation between the interior of the auditory canal and the atmosphere and also to the fact that the thermistor periodically touches the wall of the canal, thus increasing the reading. Rectal temperature was therefore adopted as the primary measurement.

The model can also provide a prediction of the mean skin temperature. In the experiments, thermistors were only attached to the skin below the water line and none to the...
head. The temperature of the immersed skin is lower than on the head. In the model calculation of mean area-weighted skin temperature, the head temperature and head skin area were therefore omitted. The model also predicts the mean surface heat flow and again it is the below-water value which has been computed for comparison with measurements made with heat flow discs. The final variable which may be used for model validation is the total predicted metabolic rate of the body, which is compared to the rate calculated from measurements of oxygen uptake made during the immersion.

Variations between individual responses to immersion are wide, owing to differences in body composition, health, age and temperament. The purpose of most whole-body thermal models is to show the likely mean response of a group of individuals with relatively similar characteristics. It is therefore appropriate to divide the subjects into categories by body mass and fat content. Accordingly, the 15 subjects who took part in the experiments are divided into three groups, whose characteristics are listed in Table 5.2.

Group 1 consists of subjects whose body mass was less than 70 kg and fat content less than 10%. Group 2 had masses between 70 and 80 kg and fat content less than 20%. Group 3 had either body masses in excess of 80 kg or fat contents in excess of 20%.

<table>
<thead>
<tr>
<th>GROUP NO.</th>
<th>SUBJECT NOS.</th>
<th>MEAN MASS</th>
<th>MEAN HT.</th>
<th>MEAN % FAT</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1,2,8,&amp;14</td>
<td>61.0 kg</td>
<td>1.75 m</td>
<td>8.0%</td>
</tr>
<tr>
<td>2</td>
<td>3,5,11,12&amp;13</td>
<td>73.9</td>
<td>1.85</td>
<td>9.6</td>
</tr>
<tr>
<td>3</td>
<td>4,6,7,9,10&amp;15</td>
<td>89.3</td>
<td>1.78</td>
<td>21.7</td>
</tr>
</tbody>
</table>

TABLE 5.2 - CHARACTERISTICS OF THE THREE SUBJECT GROUPS
The experimental data for the individual subjects were processed to obtain the mean values for each group of the variables of interest as functions of time. These variables are change in rectal temperature, mean skin temperature, mean heat flow and metabolic rate. The standard error was computed for each variable, defined as:

\[ S = \frac{\sigma}{\sqrt{N}} \] (5.1)

where

- \( S \) = standard error of the group mean
- \( \sigma \) = standard deviation from group mean
- \( N \) = number of subjects in the group

The use of the standard error rather than standard deviation was necessary since the number of subjects was small. For each variable, it was then possible to plot the mean, mean plus standard error and mean minus standard error as functions of time.

Some or all of the subjects from each group undertook immersions at 12, 18 and 24°C. The numbers taking part for each temperature are shown in Table 5.3

<table>
<thead>
<tr>
<th>TEMPERATURE OF WATER</th>
<th>12°C</th>
<th>18°C</th>
<th>24°C</th>
</tr>
</thead>
<tbody>
<tr>
<td>GROUP 1</td>
<td>3</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>GROUP 2</td>
<td>1</td>
<td>5</td>
<td>5</td>
</tr>
<tr>
<td>GROUP 3</td>
<td>4</td>
<td>6</td>
<td>5</td>
</tr>
</tbody>
</table>

TABLE 5.3 - NUMBERS OF SUBJECTS AT SPECIFIC TEMPERATURES

It is not appropriate statistically to create imaginary subjects for each group whose bodily characteristics are the group means. The model was therefore run for each
subject/temperature combination - a total of 37 runs. Mean values of the variables of interest as functions of time were then computed for each group. This was achieved by means of a FORTRAN 77 program 'AVMOD', which was written for the purpose.

The model means were superimposed on plots of the experimental means and standard errors, to give a visual indication of the model's performance. In addition, the root mean square difference between the model output and experimental data was computed over the immersion period. This gave a numerical indication of the value of the predictions, and was achieved by another FORTRAN 77 program 'RMS'.

Data on clothed immersions are less plentiful and only a limited number of comparisons have been made. In these cases, the model output is compared directly with records of individual immersions and the RMS difference computed as before.

5.3. Details of specific model/experiment comparisons

Plots of the predicted and observed mean changes in rectal temperature, mean skin temperature, metabolic rate and surface heat flow appear in Figure 5.4 and those following. In these plots, the following marking convention is adopted:

++++++ experimental mean for group
****** experimental mean plus standard error
oooooo experimental mean minus standard error
....... model mean for group

All rectal temperature comparisons are shown. The remaining variables - mean skin temperature, surface heat flow and metabolic rate, are displayed when required for explanation of a particular effect. The standard errors
for skin temperature are less than 0.5°C and, to aid clarity, only the experimental mean is plotted. The comparisons are presented in reverse order of severity, i.e. the mildest combination of factors first. The order is:

1. Fat 24°C  
2. Medium 24°C  
3. Thin 24°C  
4. Fat 18°C  
5. Medium 18°C  
6. Thin 18°C  
7. Fat 12°C  
8. Medium 12°C  
9. Thin 12°C  

For Group 3, Figure 5.4 shows that the model gives a slight underestimate of the mean fall in core temperature for heavily built subjects in water at 24°C. The rms difference is only 0.15°C in 100 minutes, however. The metabolic rate for these subjects is slightly overestimated, as Figure 5.5 shows. In Figure 5.6, a good estimate of surface heat flow is seen. It is probable that these subjects had a lower skin blood flow at 24°C than suggested by current physiological data.

Figure 5.7 indicates that the prediction of rectal temperature change for Group 2 (medium) at 24°C is within the standard error range of the experiments for the entire 100 minute exposure. As Figure 5.8 shows, the prediction for skin temperature is within 0.2°C of the experimental mean over the entire 100 minutes.

The rectal temperature prediction in Figure 5.9 for Group 1 (thin) at 24°C is within the standard error range for approximately two thirds of the 80 minute exposure. Towards the end, it shows signs of continuing to fall, whereas the experimental values do not. Figure 5.10 indicates reasonable agreement between the predicted and measured heat flows. However, in Figure 5.11, it is seen that the model underestimates the extent to which thin subjects shiver in this relatively mild exposure. The rate of heat replacement by increased metabolism is therefore insufficient.
Figure 5.4 - Rectal temp. Group 3, 24 deg C.

Rms difference = 0.15 deg C.
Figure 5.5 - Metabolic rate, Group 3, 24 deg C.

Rms difference = 37.9 W
Figure 5.6 - Heat flow, Group 3, 24 deg C.

Rms difference = 66 W m-2.
Rms difference = 0.07 deg C.

Figure 5.7 - Rectal temp. Group 2, 24 deg C.
Rms difference = 0.2 deg C.

Figure 5.8 - Skin Temp. for Group 2 at 24 deg C.
Rms difference = 0.18 deg C.

Figure 5.9 - Rectal temp. Group 1, 24 deg C.
Figure 5.10 - Heat flow, Group 1, 24 deg C.

Rms difference = 62 Wm⁻²
Rms difference = 109.6 W

Figure 5.11 - Metabolic rate, Group 1, 24 deg C.

Metabolic rate (Watts)
In the case of Group 3 at 18°C, the model, for the first time, gives a slight overestimate of the fall in core temperature as shown in Figure 5.12. Examination of Figure 5.13 shows that there is reasonable agreement between the predicted and observed metabolic rates, the rms difference being only 23.9 Watts. The heat flow comparison in Figure 5.14 reveals that, apart from being unable to track the initially high experimental value, the model provides only a slight overestimate. This is probably due to the skin blood flow rate remaining too high in the model, thus transporting too much heat to the surface from deeper tissue layers.

The rectal temperature prediction in Figure 5.15 for Group 2 at 18°C is initially higher than the experimental mean. Increased shivering after about 40 minutes has the result that the experimental fall begins to level off. The model prediction once again falls within the standard error range of the experiments after 55 minutes.

For Group 1 at 18°C, Figure 5.16 shows good initial agreement between mean rectal temperature falls but after 50 minutes the model prediction continues to fall where the experimental value levels off. This is surprising in view of the fact that heat flow, in Figure 5.17, is underestimated. However, Figure 5.18 shows a severe underestimate by the model of the increase in metabolic heat due to shivering in the early part of the immersion. The resulting lack of heat replacement results in the core temperature dropping too low, once the first order time lags of heat conduction have been overcome.

For the four subjects in Group 3 at 12°C, Figure 5.19 shows that again the model's prediction for rectal temperature is good, particularly in the early stages. The rms difference is only 0.06°C. The skin temperature plot in Figure 5.20 shows a rms difference of only 0.5°C.
Figure 5.12 - Rectal temp. Group 3, 18 deg C.
Figure 5.13 - Metabolic rate, Group 3, 18 deg C.

Rms difference = 23.9 W
Rms difference = 55 W m$^{-2}$.

Figure 5.14 - Heat flow, Group 3, 18 deg C.
Rms difference = 0.30 deg C.

Figure 5.15 - Rectal temp. Group 2, 18 deg C.
Rms difference = 0.29 deg C.

Figure 5.16 - Rectal temp. Group 1, 18 deg C.
Figure 5.17 - Heat flow, Group 1, 18 deg C.
Figure 5.18 - Metabolic rate, Group 1, 18 deg C.

Rms difference = 167.5 W
Rms difference = 0.06 deg C.

Figure 5.19 - Rectal temp. change for Group 3 at 12 deg C.
Rms difference = 0.50 deg C.

Figure 5.20 - Skin Temp. for Group 3 at 12 deg C.
The change in rectal temperature for Group 2 (medium) at 12°C is shown in Figure 5.21. In this group, only one subject undertook the immersion at this low temperature. Nevertheless, the agreement between model and experiment is good, with a rms difference of only 0.09°C over 65 minutes. The skin temperature prediction in Figure 5.22 is also acceptable, at 0.7°C rms, but is an under- rather than over-estimate.

Figure 5.23 shows the comparison of rectal temperatures for Group 1 (thin) in 12°C water - the most severe combination of factors investigated. The model over-estimates the initial rise in core temperature, which is a feature often observed in the early part of a cold exposure. Nevertheless, the later slope of approximately 0.07°C min\(^{-1}\) matches the slope of the experimental mean reasonably well. The rms difference between the two means is 0.56°C over a period of 35 minutes. Examination of Figure 5.24 reveals that the model initially underestimates the surface heat flow, thus accounting for the initially high core temperature. During the later part of the exposure, however, the predicted heat flow falls within the standard error of the experimental mean. In Figure 5.25, the skin temperature in the model is slightly high in the later part of the exposure, in accordance with the heat flow.

It has also been attempted to simulate cold water immersions in which the subjects were performing intermittent exercise. However, the results have not been satisfactory, as Figure 5.26 illustrates for a thin subject in 12°C water. Here, it is seen that the rectal temperature prediction begins by rising in a similar way to the resting prediction in Figure 5.23. At the onset of the first 10 minute period of exercise, which begins after 10 minutes, there is a marked fall. An even steeper fall occurs at the more intense exercise, beginning at 30 minutes. Figures 5.27 and 5.28 show that
Rms difference = 0.09 deg C
(One subject)

Figure 5.21 - Rectal temp. Group 2, 12 deg C.
Rms difference = 0.70 deg C (one subject)

Figure 5.22 - Skin Temp. for Group 2 at 12 deg C.
Rms difference = 0.56 deg C.

Figure 5.23 - Rectal temp. change for Group 1 at 12 deg C.
Figure 5.24 - Surface heat flow for Group 1 at 12 deg C.

Rms difference = 168 W m⁻².
Figure 5.25 - Skin Temp. for Group 1 at 12 deg C.
Figure 5.26 - Rectal temp. 1 subject exercising, 12 deg C.
Figure 5.27 - Heat flow, 1 subject exercising, 12 deg C.
Figure 5.28 - Metabolic rate, 1 subject exercising, 12 deg C.
the surface heat flow and metabolic rate are no more in error than for the resting immersions.

The effect which seems to occur in the model is as follows: when the leg muscles are at rest, they will contribute to the total amount of shivering activity but, since their metabolic rate is relatively mild, they will cool fairly rapidly in the cold water. This is particularly true for the thin subject illustrated. When a step change in exercise occurs, there is a sudden demand for oxygen replacement in these muscles, which calls for an increased blood flow. The arterial blood gives up its heat and the venous return is approximately at muscle temperature. This cool blood returns to the large blood vessels in the abdomen, where it naturally affects rectal temperature.

In the model, all the muscles in shells 5 to 8 are equally active and have a blood flow requirement based on this activity level. It is probable that, in a real leg, only those muscles near the axis of the limb, where it is warmer, are active and receive the initial increase in flow. The outer layers become gradually more perfused by blood, at which time they will have warmed up, owing to the exercise. The venous return of blood therefore remains at a reasonably high temperature. It is recognised that the modelling of large changes in exercise rate is difficult. In cold water, this seems particularly so. A more detailed scheme of control of blood supply to the muscles would be required in order to cope with this situation. However, rest or steady levels of exercise seem more likely to occur than rapid changes during survival in cold water.

The third and final part of the validation exercise concerns immersion of clothed subjects. A detailed validation has not been carried out, since this is an area in which further development of the model is likely
Clothing assembly 1 - insulation 1.05 Tog.

Long cotton vest and drawers
Aircrew cotton T shirt
Inner knitted coverall
Immersion coverall (water-impermeable)
Aircrew flying coverall
Flying helmet
Woollen socks
Aircrew boots
Water-resistant gloves
Inflatable lifepreserver

Clothing assembly 2 - insulation 0.5 Tog.

Long cotton vest and drawers
Aircrew cotton T shirt
Heavy woollen jersey
Immersion coverall (water-impermeable)
Aircrew flying coverall
Flying helmet
Woollen socks
Aircrew boots
Water-resistant gloves
Inflatable lifepreserver

TABLE 5.4 - DETAILS OF CLOTHING ASSEMBLIES
to be pursued. Two model runs have been carried out to simulate experimental immersions by Sowood, Allan and Cohen (1987). The first subject was No.13 in Table 5.1 who was relatively heavily built. The immersion was in water at 6.6°C and the clothing assembly had an insulation value of 1.05 Togs. The second was No.8, who was slightly built and was immersed in 6.2°C water with an insulation of 0.5 Togs. The following Table 5.4 gives details of the clothing worn.

The instrumentation was identical to that in the semi-nude immersions described earlier. The comparison of rectal temperatures is shown in Figure 5.29. The model maintains a good prediction, even beyond two hours of immersion, although it cannot simulate the 50 minute period for which this subject maintained a high core temperature at the beginning. Figure 5.30 shows that the skin (rather than clothing surface) heat flow was also in reasonable agreement. The mean skin temperature prediction in Figure 5.31 is close for the first hour but is over-estimated to the extent of 1°C after two hours. Data from measurements of metabolic rate made during these experiments were not available.

For the second immersion, Figure 5.32 shows that the agreement of rectal temperatures was again good. The model overestimates the skin heat flow at the beginning, as seen in Figure 5.33. On the basis of these brief comparisons, it appears that the simulation of immersed clothing insulation is reasonably adequate. This concludes the description of the validation work which has been carried out to the date of writing.
Figure 5.29 - Subject clothed with 1.05 Tog at 6.6 deg C.
Figure 5.30 - Subject clothed with 1.05 Tog at 6.6 deg C.
Figure 5.31 - Subject clothed with 1.05 Tog at 6.6 deg C.
Figure 5.32 - Subject clothed with 0.5 Tog at 6.2 deg C.
Figure 5.33 - Subject clothed with 0.5 Tog at 6.2 deg C.
6. CONCLUSIONS

6.1. Summary and discussion

Thermal stresses facing military personnel have been reviewed and the effects of accidental immersion in cold water discussed. The benefits of using a computer-based model to predict the thermal time history of the human body during immersion have been identified. A survey has been conducted of past and current work in modelling of the human thermal system. The models described include steady state and dynamic simulations, both of parts and the whole of the body. The absence of a dynamic model capable of providing sufficiently valid predictions whilst being sufficiently well documented to allow further development has become apparent.

The development of the first stages of a new model of the human thermal system has been completed. The model is based, to some extent, on principles established by previous workers in the field but also contains a number of new features. The body is modelled in 15 segments, left and right limbs being treated separately in order to account for unequal exposures. The minimum inputs required to determine the dimensions of these segments are the subject’s height and body mass. However, if either % fat or specific gravity are known, then these may be included as input and will result in greater accuracy.

A total of 150 tissue regions have been defined, using distribution and density data on which a number of earlier sources agree. The masses of these regions are estimated, summed and compared to the known body mass, which is used as a correction factor. It is suggested that an individual geometrical representation of the subject being simulated is likely to improve considerably the accuracy of the predictions.
Differences exist in the representation of blood circulation between this and previous models. The total volume of blood is calculated as a function of body mass. Thirty venous and arterial pools are created to represent the major blood vessels, liver, spleen, heart and pulmonary circulation. A further 90 pools are used to represent arterial and venous circulation in the remaining visceral organs, muscle and skin compartments. The dimensions of all blood vessels are specifically calculated so that heat exchange coefficients may be determined. In all of the 150 blood pools, counter-current heat exchange between arteries and veins is simulated. The flow of blood in each vessel is computed and used to calculate the transfer of heat.

Thermal properties are assigned to each compartment, using weighted mean values where more than one tissue type is present. These are compensated at regional boundaries where the tissue type changes. Using a lumped parameter approach, thermal properties and temperatures are considered uniform throughout each region and are considered to apply at the centre of area, as in most previous models. The degree of compensation applied according to tissue type is more comprehensive, however.

In the skin, conduction towards the surface and convection away from it are equated to give a prediction of true surface temperature. In addition, the convective heat flow coefficients are calculated as functions of fluid properties, which are themselves functions of the mean of the surface and bulk fluid temperatures. A further feature is the modification of thermal properties of the body as tissue temperatures change.

Resting metabolic rates and blood perfusion rates are influenced by the subject’s actual metabolic rate at the start of the exposure, if known. An array of starting temperatures for a ‘standard man’ has been determined by
running the program for thermo-neutral conditions and disabling metabolic responses. This array is also modified by the starting core temperature of the subject, if known. In the absence of information on initial surface heat flow, a method employing second derivative of temperature with radius has been developed to predict the initial surface temperature of each segment.

The equations predicting the metabolic responses to temperature deviations are essentially those published from earlier physiological investigations. Alterations have been introduced to account for the changes in the metabolism of tissue with temperature. Respiratory volume, heat loss and heat generation are calculated and accounted for in the appropriate tissue and blood compartments of the thorax.

For each of the 150 tissue shells and 120 blood compartments, a heat balance is calculated at each time step. Established techniques have been expanded to cope with the additional complexity of this model. Particular care has been taken to ensure accuracy in convective exchange by blood, which is known to have a shorter time constant than conduction within the body. The rates of heat flow and thermal properties are used to compute a rate of change of temperature for each region and hence a new set of temperatures for each time step.

The time interval is controlled to prevent changes in temperature greater than 0.1°C between time steps, thus ensuring stability. The trapezoidal integration routine has the advantage of simplicity but tends to result in long running times for small subjects in cold water, where rates of change of temperature are initially large. The application of the multivariable state space technique to simulation of the human thermal system has been examined but found to be inappropriate, owing to
non-linearities. Linearisation would appear to impose an unacceptable computational load.

In the computer program, the FORTRAN 77 file has been arranged to contain a master segment and fourteen subroutines, each performing a specific function. Comment lines have been liberally applied to explain the purpose of each group of statements, so that subsequent users will understand the program with little difficulty. Where possible, mnemonic variable names have been used. The simulation runs almost entirely in SI units of measurement.

The program reads data from two files; one containing details of the subject and the exposure, the other containing initial conditions and coefficients necessary for the calculations. The input files contain comments giving the identity of each variable and stating its units. Output is directed either to a third file or to the screen and is in two parts. The first part, which is user-selectable, shows physical and thermal features of the body as modelled by the program. The second, which is mandatory, gives the time-history of the physiological variables as the simulation proceeds. To date, the program has been compiled and run under five different operating systems on various computers as a test of portability.

The model was initially developed to simulate resting nude immersion in cold water. This configuration has deliberately been investigated in depth since it results in high heat flows and therefore potentially the greatest inaccuracies. The importance of establishing an adequate nude model before simulating clothing has been recognised and the majority of validation has been carried out on this configuration. Code for clothed immersion has been added and validated briefly. Exercising immersions may also be simulated. Outline code for air exposure is also
included but is not complete, in that it does not yet account for heat loss by sweat evaporation. One of the primary aims has been to provide a structured basis on which future workers may extend new ideas and deal with new situations as they are required to be simulated.

The conclusion to the project has been the comparison of the predictions provided by the model with existing experimental data. Data have been made available from a comprehensive set of experiments covering resting and exercising nude immersion and resting clothed immersion in water at a variety of temperatures. A mathematical analysis of the workload provided by underwater cycling exercise in these experiments has been carried out. Validation of the model has been based on this set of data. Comparisons of the predicted and observed body temperature/time history have been made for fat, medium and thin groups of subjects. The mean and its standard error have been calculated for each group in both the experimental data and model output.

At 24°C, the prediction of core temperature fall and mean skin temperature is satisfactory for all groups. At 18°C, core temperature prediction for thin subjects is satisfactory over the first hour but becomes unreliable thereafter. For medium and heavy subjects, the prediction is again acceptable. The principal reason for the deficiencies seems to be the lack of a satisfactory means of predicting the metabolic rate increase due to shivering. Simple functions of core and mean skin temperature do not appear to suffice. During the course of model development, no published shivering control equation was discovered which yielded a better simulation over the range of water temperatures and subject characteristics.

For medium and fat subjects at 12°C, the predictions appear acceptable over a time period of just over one.
hour. Immersion of thin individuals at rest in water at 12°C without protection is a relatively severe exposure. Although the model indicates the slope of the core temperature fall quite well, the equations available to predict shivering activity do not allow an overall accuracy better than approximately 0.5°C to be achieved.

The target group of subjects for this initial phase of model development comprises survivors immersed in the sea, wearing protective clothing. The brief validation carried out for clothed immersion has provided encouraging results. Close agreement has been achieved between the simulated and observed falls in core temperature for two subjects in water at approximately 6°C. There was a marked contrast between the two sets of characteristics: the first subject had a high body fat content and wore a highly insulative clothing assembly. The second had less than half as much body fat and less than half the insulation. The method chosen to simulate immersed clothing insulation therefore appears valid.

The only remaining situation in which the model is unreliable is in simulation the core temperature response to large changes between work and rest in cold water. Any improvement would be dependent upon new physiological data on the way groups of muscles operate in this situation and hence the way in which heat is transported by blood flow. No such data appear to have been published at the time of writing. The simulation of immersions with steady exercise rates should not present a problem, since the muscle compartments in the model would not undergo cooling followed by a rapid increase in blood flow. However, this situation has not yet been investigated.
6.2. Recommendations for further research

Mathematical modelling of the human thermal system can avoid the need for experiments. Nevertheless, all such models require experiments to have been carried out initially, in order to produce the data upon which they are based. In view of the fact that the science is now at least 40 years old, there is a surprising lack of numerical data in certain areas. Where data do exist, they are often in a form not readily usable by models, having been obtained for some other purpose. A coordinated effort within the modelling community to fill such gaps therefore appears justified.

So far as cold water immersion modelling is concerned, the main deficiencies appear to be in the areas of shivering metabolism, control of skin blood flow and dynamic control of muscular blood flow. Control equations are required which predict these effects for a wide range of body sizes and water temperatures, particularly during the early part of an exposure. A wider range of input variables than merely mean skin and core temperature may be required.

Regarding development of the model itself, a new shivering controller, when available, should be incorporated and validated for semi-nude immersion. Following a successful demonstration, further validation could then be undertaken for clothed immersion. The exposures for which experimental data are available have been rather short, particularly for the smaller subjects and lower temperatures. Within the bounds of an ethical code of conduct, it would be of benefit if longer exposures could be undertaken so that more advanced stages of hypothermia could be simulated and validated.

Although the model is currently configured to simulate cold water immersion, it has been developed in such a way
as to permit adaption to air exposure. For cold air exposure, all that is required is to specify an air environment instead of water at the appropriate line of the first input file. The model has not yet been run in this way but predictions should be as accurate as for cold water exposure, provided the clothing insulation value used is obtained from a measurement made in air. For warm air exposure, the model requires modification to account for heat loss by sweating and for the degree of permeability of clothing to water vapour.

In the course of the investigation, the foundations of a structured cold water immersion model have been laid. It is recommended that the model is further refined by the additions proposed above. By means of additional code, it should eventually be possible to extend the basic configuration to simulate nude or clothed exposure to warm or cold air or water. The listings of code and operating instructions necessary are contained in Volume 2 of the thesis.
REFERENCES


APPENDIX 1 - VARIATION OF WORK WITH WATER TEMPERATURE

The general arrangement of the subject and underwater cycle in the experiments of Sowood (1988) is shown in Figure A.1. The physical properties of water, principally viscosity, vary with temperature. Hence it was believed possible that, for the same pedalling rate, the actual rate of working might vary significantly over the range of water temperatures used. Equation A.1, taken from Kaye & Laby (1973), shows that the viscosity of water varies from approximately $12 \times 10^{-3}$ kgms$^{-1}$ at 12°C to $8 \times 10^{-3}$ at 30°C.

$$\mu = \frac{0.431}{24 + T}$$ \hspace{1cm} (A.1)

Where: $\mu = \text{viscosity of water (kgms}^{-1})$

$$T = \text{temperature of water (°C)}$$

The density, however, varies by only 0.3% over the same temperature range and in the following analysis, it is considered to be constant at 1000 kgm$^{-3}$ (Kaye & Laby, 1973).

When a relatively blunt object, such as a vane, moves through a fluid, the low pressure in the turbulent flow region on the downstream side, combined with the high stagnation pressure on the front of the body, results in a high contribution to the total drag known as 'pressure drag'. This often amounts to more than 90% of the total drag for blunt bodies at high Reynold’s numbers (Potter & Foss, 1975). For streamlined bodies at low or zero angles of attack, such as a rotating wheel, the turbulent region is either small or non-existent. Here, the drag is the result of shear stresses acting on the surface and is known as 'viscous drag'. Viscous drag varies directly with the fluid viscosity, whereas pressure drag is virtually independent of it.
All dimensions in metres

FIG A.1    GENERAL ARRANGEMENT OF A SUBJECT PERFORMING UNDERWATER EXERCISE
Fluid drag occurs on all moving parts of the man/cycle assembly. It is convenient to analyse these separately, considering contributions from the wheel face and rim, the vanes and the subject’s thighs, calves and feet.

There are two components of drag on the wheel surface; one on the flat rim and one on the smooth wheel face. The viscous drag on the face on which the vanes are mounted will be ignored, to an approximation, since the turbulence caused by the vanes is likely to reduce relative motion between the water and the wheel surface almost to zero. On the smooth face of the wheel, consider an elemental strip of width dr at a radius r, as shown in Figure A.2A. The wheel is rotating at $n_w$ revolutions per second and the radial velocity at radius r is $v$ (ms$^{-1}$). Using the standard drag equation, referred to for example by Binder (1973), the drag on the strip is:

$$d_s = 0.5\rho v^2 C_{d_s} 2\pi r \, dr$$

(A.2)

Where: $d_s =$ viscous drag (N)

$\rho =$ density of fluid (Kgm$^{-3}$)

$v =$ radial velocity (ms$^{-1}$)

$C_{d_s} =$ viscous drag coefficient for surface (dimensionless)

$r =$ radius (m)

But $v = 2\pi r n_w$

Where: $n_w =$ wheel speed (s$^{-1}$)

So that $d_s = 4\pi^2 \rho C_{d_s} n_w^2 r^3 \, dr$

(A.3)

Summing the contributions from all such strips,
A.2a - Integration across wheel surface

A.2b - Integration across vane surface

A.2c - Integration along thigh (and calf) length

FIG A.2 - WHEEL, VANE AND LEG GEOMETRY
\[
D_g = \sum 4\pi^3 \rho \cdot C_{ds} \cdot n_w^2 \cdot r^3 \cdot dr \quad (A.4)
\]

Where: \( D_g \) = total drag on smooth surface of wheel (N)

\( R \) = radius of wheel (m)

and, as \( dr \) tends to zero,

\[
D_S = \int_{0}^{R} 4\pi^3 \rho \cdot C_{ds} \cdot n_w^2 \cdot 2^2 \cdot r^3 \cdot dr \quad (A.5)
\]

ie \( D_S = \pi^3 \rho \cdot C_{ds} \cdot n_w^2 \cdot R^4 \quad (A.6) \)

For the rim of the wheel, the viscous drag is similarly:

\[
D_r = 0.5 \rho \cdot v^2 \cdot C_{dr} \cdot (2\pi \cdot R \cdot x_r) \quad (A.7)
\]

or \( D_r = 4\pi^3 \rho \cdot C_{dr} \cdot n_w^2 \cdot 2^3 \cdot x_r \quad (A.8) \)

Where \( x_r \) = width of rim (m)

\( C_{dr} \) = viscous drag coefficient for rim

Considering a single vane viewed face-on as in Figure A.2B, the drag is first calculated for an elemental strip:

\[
d_v = 0.5 C_{dv} \cdot \rho \cdot v^2 \cdot w \cdot dr \quad (A.10)
\]

or \( d_v = 2\pi^2 C_{dv} \cdot \rho \cdot r^2 \cdot n_w^2 \cdot w \cdot dr \quad (A.11) \)

Where: \( x_v \) = width of vane (m)

\( C_{dv} \) = pressure drag coefficient for a flat plate

As before, the total drag is obtained by summing individual contributions:
\[
D_v = \sum_{r=r_1}^{r_2} 2\pi^2 C_{dv\rho} r^2 n_w^2 x_v dr \quad (A.12)
\]

and, as \(dr\) tends to zero,

\[
D_v = \int_{r_1}^{r_2} 2\pi^2 C_{dv\rho} r^2 n_w^2 x_v dr \quad (A.13)
\]

\[
= \frac{2}{3\pi^2} C_{dv\rho} n_w^2 x_v (r_2^3 - r_1^3) \quad \text{(one vane).} \quad (A.14)
\]

The above equation gives the drag on a flat plate in free space. For a vane mounted on a surface, it is necessary to account for the effect of the boundary layer on the surface. This will produce a circulation effect and reduce fluid drag on the plate and on the wheel rim. In the absence of any reliable theoretical means of estimating the magnitude of this effect, it is considered that the apparent rotational speed of the wheel with respect to the water \(n_a\) is simply reduced to a fraction of \(n_w\). It has been found that a good fit between predicted and measured metabolic rates is obtained by using the relationship:

\[
n_a = 0.75 n_w \quad (A.15)
\]

Since both the rim and vanes are likely to be affected by circulation, Equations A.8 and A.14 are modified to become:

\[
D_r = 4\pi^3 \rho C_{dr} (0.75 n_w)^2 R^3 x_r \quad (A.16)
\]

and

\[
D_v = 2/9\pi^2 C_{dv\rho} (0.75 n_w)^2 x_v (r_2^3 - r_1^3) \quad (A.17)
\]

(for one vane).
When analysing the leg movement of a cycling subject, there are three aspects of motion to be considered, each with an associated component of fluid drag. These comprise the cyclic vertical motion of the thighs (pivoted at the hip joints), the cyclic horizontal/vertical motion of the calves (pivoted at the knees) and the circular motion of the feet. Since each of these limb segments presents a relatively blunt cross-section to its direction of motion, the fluid drag will consist primarily of pressure drag. The cross-sectional area of the feet presented to the direction of circular motion of the pedals is, in fact, constantly changing. A valid simplification may be achieved by taking the mean dimensions of the feet and representing each by a cylinder of constant cross-sectional area and shape.

Considering the feet first, the drag on each representative cylinder may be determined by an expression similar to Equation A.2 such that:

\[ D_f = 0.5 \rho C_{df} V_p^2 l_f x_f \]  

(A.18)

Where: \( D_f \) = pressure drag on a cylinder representing a foot (N)

\( C_{df} \) = drag coefficient for flow normal to a cylinder

\( V_p \) = radial velocity of pedals \( (ms^{-1}) \)

\( l_f \) = length of foot cylinder (m)

\( x_f \) = diameter of foot cylinder (m)

The radial velocity \( V_p = 2\pi n_p r_p \)  

(A.19)

Where \( n_p \) = rotational speed of pedals \( (ms^{-1}) \)

\( r_p \) = radius of pedal crank (m)

Hence \( D_f = 2\pi^2 n_p r_p^2 \rho C_{df} l_f x_f \)  

(A.20)
which is constant for a given pedalling speed. The motion of
the thighs is a maximum at the knees and zero at the hip joints. The velocity will therefore vary
down the length of the thighs and so will the drag force.
Viewing one thigh from above in Figure A.2C, consider an
elemental strip of length $dl_t$ and width $x_t$. An
expression for the pressure drag on this strip may be
written:

$$d_t = 0.5 \rho C_{dt} v^2 x_t dl_t$$  \hspace{1cm} (A.21)

Where:  
- $d_t =$ drag (N)  
- $C_{dt} =$ drag coefficient for thigh cylinder  
- $v_t =$ velocity at point under consideration (ms$^{-1}$)  
- $x_t =$ mean diameter of thigh (m)  
- $l_t =$ length of thigh (m)

The velocity $v_t$ is cyclic and radial and has a sinusoidal
waveform which peaks half way down each pedal stroke and
reaches zero at the top and bottom. At the knee, the
peak velocity is equal to the constant radial velocity of
the pedals:

$$|v_t| = 2\pi r_p n_p$$  \hspace{1cm} (A.22)

Where $|v_t|$ = the peak velocity of the thigh at the knee
(measured)

As a function of time, $t$,

$$v_t(t) = 2\pi n_p r_p \sin 2\pi n_p t$$  \hspace{1cm} (A.23)

At the elemental strip, a distance $r_t$ from the hip joint,

$$v_t(t) = 2\pi n_p r_p r_t / l_t \sin 2\pi n_p t$$  \hspace{1cm} (A.24)

Substituting for $v_t$ in Equation A.19 gives:
To obtain the total drag on each thigh, the contributions from all such strips are summed:

\[ D_t = \sum_{r_t=0}^{l_t} 2\pi^2 \rho C_{dt} n_p^2 r_p^2 r_t^2 / l_t^2 x_t \sin^2 2\pi n_p t . dr_t \] (A.26)

and, as \( dr_t \) tends to zero,

\[ D_t = \int_{r_t=0}^{l_t} 2\pi^2 \rho C_{dt} n_p^2 r_p^2 r_t^2 / l_t^2 x_t \sin^2 2\pi n_p t . dr_t \] (A.27)

Hence \( D_t = 2/3 \pi^2 \rho C_{dt} n_p^2 r_p^2 l_t^2 x_t \sin^2 2\pi n_p t . dr_t \) (A.28)

The motion of the calves is more complex, since they move vertically as well as horizontally. However, the vertical movement will only result in a small amount of viscous drag and can be ignored with minimal loss of accuracy. The cyclic horizontal motion is very similar to that of the thighs but is 90° out of phase since its maximum occurs at the top and bottom of each pedal stroke, when the thigh motion is zero. By analogy with Equation A.26 and using a cosine term to account for the 90° phase difference, we may write:

\[ D_c = 2/3 \pi^2 \rho C_{dc} n_p^2 r_p^2 x_c l_c \cos^2 2\pi n_p t \] (A.29)

Where: \( D_c = \text{drag on each calf (N)} \)
\( C_{dc} = \text{drag coefficient for calf} \)
\( x_c = \text{width of calf (m)} \)
\( l_c = \text{length of calf (m)} \)

The total pressure drag on the legs must now be obtained by summing the contributions from the thighs, calves and feet. For a constant rate of pedalling, the drag on the
feet is considered constant. On the thighs and calves, however, the drag is cyclic, as indicated by Equations A.28 and A.29. Plotting these two equations for drag against time, the sequence in Figure A.3A is obtained. The two sinusoidally varying forces are 90° out of phase with one another and cannot be added directly. However, by using a trigonometric identity, it is possible to re-write Equations A.28 and A.29 as follows:

\[ D_t = \frac{2}{3} \pi \rho C_d n_p^2 r_p^2 x_t^l t_{1/2} (1-\cos 2\pi n_p t) \quad (A.30) \]

and

\[ D_c = \frac{2}{3} \pi \rho C_d n_p^2 r_p^2 x_c^l c_{1/2} (1+\cos 2\pi n_p t) \quad (A.31) \]

These drag components each contain a cosine term at the same frequency and may therefore be added. For each leg:

\[ D_l = D_t + D_c \]

\[ = \frac{1}{3} \pi^2 \rho n_p^2 r_p^2 \left[ (C_d t x_t^l t_c + C_d c x_c^l c) 
+ (C_d t x_t^l t_c - C_d c x_c^l c) \cos 2\pi n_p t \right] \quad (A.32) \]

The result is a zero-frequency term, to which is added a co-sinusoidal term at twice the pedalling frequency. These terms may be considered as the addition of a static vector and a rotating vector, as depicted in Figure A.3B. The mean amplitude of the resultant of these two vectors over a full cycle of pedal rotation is equal to the amplitude of the static vector. Hence:

\[ |D_l| = \frac{1}{3} \pi^2 \rho n_p^2 r_p^2 (C_d t x_t^l t_c + C_d c x_c^l c) \quad (A.33) \]

Where: \(|D_l|\) = the modulus of the varying drag \(D_l\) on each leg (N)

In order to obtain a numerical solution for each drag equation, it is necessary to find the values of the drag coefficients \(C_{dW}, C_{dR}, C_{dV}, C_{dt}, C_{dc}\) and \(C_{df}\). \(C_{dW}\) and \(C_{dR}\) apply to skin friction between the fluid and flat surfaces. According to Binder (1973), such a coefficient
FIG A.3 PHASE RELATIONSHIP OF THIGH AND CALF DRAG
varies inversely with the square root of the Reynold’s number:

\[ C_d = \frac{1.292}{\sqrt{N_R}} \]  

(A.34)

and \[ N_R = \frac{\rho \cdot v \cdot d}{\mu} \]  

(A.35)

Where: \( N_R \) = Reynold’s number
\( \rho \) = density of fluid (Kgm\(^{-3}\))
\( v \) = velocity at point under consideration (ms\(^{-1}\))
\( d \) = a dimension of the object under consideration (see below) (m)
\( \mu \) = viscosity of the fluid (Kgms\(^{-1}\))

For the face of the wheel, \( d \) is taken as the radius at the centre of area, \((d = R/\sqrt{2})\) and:

\[ v = 2\pi n_w R / \sqrt{2} \]  

(A.36)

For the rim, \( d \) is taken as the width \( x \) and \( v \) is the rim radial velocity:

\[ v = 2\pi n_w R \]  

(A.37)

The viscosity \( \mu \) may be obtained for each temperature of interest, using Equation A.1. Density \( \rho \) is taken to be constant at 1000 Kgm\(^{-3}\). By performing the appropriate substitutions from Equations A.1, A.36 or A.37 and A.35 in Equation A.34, it is possible to find values for \( C_{dw} \) and \( C_{dr} \) for any speed and temperature combination.

The vanes, although attached to the wheel face at one side, are considered as isolated flat plates normal to the fluid flow. According to Potter and Foss (1975), such a plate will exhibit an almost constant pressure drag coefficient of 1.2 for Reynold’s numbers greater than 10\(^3\). The Reynold’s number for the vanes may be calculated using their dimensions and the velocity which
occurs $1/\sqrt{2}$ of the way along their length. Taking into account the chainwheel ratio and considering pedalling speeds between 20 and 40 rpm, $N_R$ varies between approximately $1.5 \times 10^3$ and $4.5 \times 10^3$ over the temperature range 12 to 30°C. Accordingly, $C_{dv}$ is taken as 1.2.

The thighs, calves and feet are all considered as cylinders. Potter and Foss again show that the drag coefficient for a cylinder normal to a fluid flow is almost constant at 1.0 between Reynold’s numbers of $10^3$ and $5 \times 10^5$. Similar calculations to those made for the vanes show that $N_R$ for the legs and feet varies around $4 \times 10^3$. $C_{dt}$, $C_{dc}$ and $C_{df}$ are therefore taken as 1.0.

To estimate the power being supplied to the cycle by the subject, it is simply necessary to take the product of torque and angular speed for each drag contribution and refer these to the pedal shaft:

$$P = \omega T \quad \text{(A.38)}$$

Where: $P =$ power (W)

$\omega =$ angular speed $= 2\pi n \ (\text{rad s}^{-1})$

$T =$ torque (Nm)

To calculate the torque at the wheel shaft produced by each drag component, it is first necessary to find the distance from the shaft at which each drag force acts. In the case of the wheel face drag, this can be considered to act through the radius of the centre of area $R/\sqrt{2}$. For the rim drag, the distance is simply the wheel radius $R$. For each vane, the centre of drag will be situated at a point $1/\sqrt{3}$ of the length from the shaft end. Thus:

$$P_w = P_S + P_r + 3P_v \quad \text{(A.39)}$$

Where: $P_w =$ power input at wheel shaft (W)

$P_S =$ power requirement of smooth surface

216
$P_r =$ power requirement of rim
$P_v =$ power requirement of each vane

From the above considerations:

$$P_s = 2\pi n_w D_s R / \sqrt{2} \quad (A.40)$$

$$P_r = 2\pi n_w D_r R \quad (A.41)$$

$$P_v = 2\pi n_w D_v [r_1 + (r_2 - r_1) / \sqrt{3}] \quad (A.42)$$

The power $P_w$ will be transferred from the pedal shaft through the cycle mechanism, which has an efficiency $\eta_c$. At the pedals:

$$P_p = P_w / \eta_c \quad (A.43)$$

Where $P_p =$ power input to pedals.

The drag force on the feet acts at the radius of the pedal crank, such that:

$$P_f = 2\pi n_p D_f r_p \quad (A.44)$$

Where: $P_f =$ power input to overcome foot drag (W)

For the thighs and calves, the estimation is less straightforward but may be simplified by considering velocities to be linear, rather than radial. Considering a downward pedal stroke, the knee will have a peak velocity of $2\pi r_p n_p$ ms$^{-1}$ and rms velocity $\sqrt{2} \pi r_p n_p$. The drag force $D_t$ will act at a point $1/\sqrt{3}$ of the distance from the hip joint to the knee. A similar result can be obtained by analogy for the calf. The steady component of power needed to overcome the total leg drag is therefore:

$$P_l = \sqrt{2\pi} r_p n_p |D_1| / \sqrt{3} \quad (A.45)$$
The total power input required from the body may now be summed:

\[ P_b = \frac{(P_g + P_r + 3P_v)}{\eta} + 2(P_f + P_l) \]  \hspace{1cm} (A.46)

Finally, the metabolic rate may be estimated by assuming an efficiency \( \eta_b \) for the muscles of approximately 22\% (Dickinson, 1929) and a basal metabolic rate \( M_b \) for a non-exercising subject of approximately 100W:

\[ M = M_b + \frac{P_b}{\eta_b} \]  \hspace{1cm} (A.47)

Where: \( M = \) total metabolic rate (W)  
\( M_b = \) basal metabolic rate (W)  
\( \eta_b = \) efficiency of muscles

A computer program performs the above calculations. The input data are shown in Table A.1. A comparison of predicted metabolic rate in 30°C water with the actual rate derived from oxygen uptake measurements on 15 subjects is made in Table A.2. It is seen that, for the input parameters chosen, the prediction is within the standard error of the mean of the measurement in three out of the five pedalling speeds studied. The metabolic rate prediction is plotted as a function of temperature for 5 pedalling speeds in Figure A.4.

The analysis may now be used in the determination of the effect on the metabolic load of pedalling at the same speed at varying temperatures. The fluid drag acting on the wheel surface and rim will be almost entirely viscous and will therefore vary with temperature. The power input required to overcome this viscous component is obtained by summing:
FIG A.4 - METABOLIC RATE VS. WATER TEMPERATURE FOR 5 PEDALLING SPEEDS
Wheel outer radius 0.30 m

Wheel rim thickness 0.04 m

Height of vane 0.04 m

Inner radius of vane 0.05 m

Outer radius of vane 0.25 m

Length of calf 0.40 m

Diameter of calf 0.10 m

Length of foot 0.20 m

Diameter of foot 0.05 m

Length of thigh 0.40 m

Diameter of thigh 0.15 m

Radius of pedal crank 0.15 m

Efficiency of muscles 0.22

Efficiency of mechanism 0.70

Temperature of water 12.00, 18.00, 24.00, 30.00°C

Speed of pedalling 20.00, 25.00, 30.00, 35.00, 40.00 rpm

Basal metabolic rate 100.0 W

**TABLE A.1 - PHYSICAL DATA ON CYCLE AND BODY**
<table>
<thead>
<tr>
<th>PEDALING FREQUENCY (RPM)</th>
<th>MEASURED METABOLIC RATE (W)</th>
<th>STANDARD ERROR OF MEAN (W)</th>
<th>PREDICTED METABOLIC RATE (W)</th>
<th>PERCENT ERROR (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>20</td>
<td>221.56</td>
<td>6.51</td>
<td>200.86</td>
<td>-9.3</td>
</tr>
<tr>
<td>25</td>
<td>303.61</td>
<td>6.82</td>
<td>289.26</td>
<td>-4.7</td>
</tr>
<tr>
<td>30</td>
<td>421.54</td>
<td>7.18</td>
<td>417.18</td>
<td>-1.0</td>
</tr>
<tr>
<td>35</td>
<td>601.80</td>
<td>9.39</td>
<td>591.50</td>
<td>-1.7</td>
</tr>
<tr>
<td>40</td>
<td>835.86</td>
<td>15.30</td>
<td>819.03</td>
<td>-2.0</td>
</tr>
</tbody>
</table>

(MEAN OF 15 SUBJECTS)

**TABLE A.2** - COMPARISON OF PREDICTED AND MEASURED METABOLIC RATE AT 30°C
Where \( P_{visc} = P_s + P_r \) 

Similarly, the unvarying pressure drag power is obtained by summing the remaining components:

\[
P_{press} = 3P_v + 2P_f + 2P_l \tag{A.49}
\]

Examination of the predictions suggests that at 20 rpm the increase in predicted metabolic rate when the water temperature is reduced from 30°C to 12°C is 4.2%. This implies that the subjects were working only 4.2% harder at 12°C than at 30°C when pedalling at 20 rpm. Similarly at 40 rpm, there is an increase of 5.8% in metabolic rate over the same temperature range.
COMPUTER SIMULATION OF THE RESPONSE OF THE HUMAN BODY
TO IMMERSION IN COLD WATER

by

Graham Richardson

Thesis submitted for the degree of Doctor of Philosophy
at the University of Surrey

December 1988

A report on work carried out by the author at the
Royal Air Force Institute of Aviation Medicine

VOLUME 2
TABLE OF CONTENTS

VOLUME 2

<table>
<thead>
<tr>
<th>Title page</th>
<th>i</th>
</tr>
</thead>
<tbody>
<tr>
<td>Table of contents</td>
<td>ii</td>
</tr>
</tbody>
</table>

CHAPTER 1 - DESCRIPTION OF THE COMPUTER PROGRAM

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1. Language and structure</td>
<td>1</td>
</tr>
<tr>
<td>1.2. Subroutine descriptions</td>
<td>4</td>
</tr>
<tr>
<td>1.3. Notes on operation</td>
<td>8</td>
</tr>
<tr>
<td>1.4. Input and output file listings</td>
<td>12</td>
</tr>
<tr>
<td>1.5. Model listing</td>
<td>26</td>
</tr>
</tbody>
</table>

LIST OF FIGURES

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1 Flowchart illustrating program operation</td>
<td>3</td>
</tr>
</tbody>
</table>

LIST OF TABLES

<table>
<thead>
<tr>
<th>Table</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1 Compilation and storage statistics</td>
<td>10</td>
</tr>
</tbody>
</table>

APPENDIX 1 - List of variables in main computer program | 86   |
1. DESCRIPTION OF THE COMPUTER PROGRAM

1.1. Language and structure

The mathematical model of the human thermal system is described in Volume 1 of the thesis. This volume provides details of the computer program used to implement the model. It begins with a discussion of language and structure, followed by a description of the purpose of each subroutine. Operating instructions are then given, followed by a listing of the program source code. Listings are then given of typical input and output files.

It is hoped that the program will be sufficiently easy to run and modify that future users will find it possible to adapt the code for cold and warm air exposure. The version published has been numbered 1.1. It would be appreciated if any errors found by future users or any successful simplifications or modifications could be communicated to the author or to the Director of Research at the Royal Air Force Institute of Aviation Medicine, Farnborough, Hampshire, GU14 6SZ, United Kingdom.

The program is written in FORTRAN 77 code. This language was chosen for a number of reasons. A higher proportion of workers in the field of thermal physiology appear to have a working knowledge of FORTRAN than any other language. There are still a few computers with no compilers for Pascal or 'C', while the majority can run FORTRAN. Portability is therefore improved by its use. Existing models, such as those described in Volume 1, are mainly written in FORTRAN. The option of inserting sections of code from these models is therefore available. Finally, although the aim has been to produce structured code, the rigid structure imposed by the alternative high-level scientific languages, Pascal and
'C' may have been too restrictive.

The source code file 'MODEL.FOR' is approximately 3100 lines in length. It contains a considerable amount of spacing and tabulation as an aid to clarity. Comment lines have been applied frequently to explain the purpose of each group of statements. Wherever possible, mnemonically significant names have been given to the variables. All are six characters or less in length. Declarations of the variables are made in a separate file 'COMMON.FOR'. This is arranged in 26 COMMON blocks in alphabetical order for the numeric variables, plus a block for character variables.

The program consists of a master segment and fourteen subroutines for specific purposes. The master segment carries out all timing functions and calls the subroutines as required. INCLUDE functions for the COMMON file appear in the master segment and each subroutine so that all variables are transferred and are stored once only.

The program reads data from two files. One contains details of the subject and the exposure. The other contains initial conditions and coefficients necessary for the calculations. In the input files, the values of the variables are interspersed with comments giving their identity and units. Output is directed to a third file and is in two parts. The first part, which is user-selectable, shows the physical and thermal features of the body as modelled by the program. Its usefulness is confined mainly to development. The second part, which is mandatory, gives the time history of selected variables as the simulation proceeds.
Figure 1.1 - Flowchart illustrating program operation
1.2. Subroutine descriptions

Figure 1.1 shows the sequence of operation of the subroutines in the program. The first subroutine to be called is READ1. This reads data from any file assigned to Logical Unit 1 by the user. This file gives the model a title for the simulation and states whether the exposure is to air or water. (If water is specified, the head is nevertheless above the surface). The subject's name, and physical characteristics are then read, followed by initial conditions. Variables associated with clothing, the length of exposure and the length of each interval of unchanging environmental conditions are then read. Finally, the air and water temperatures and velocity, the relative humidity and barometric pressure are read. These are held in arrays of 36 elements, allowing 36 different condition periods of arbitrary length to be simulated during the exposure.

The second subroutine, READ2, reads data from a file on Logical Unit 2. This file contains the physical and thermal properties and the distribution of the eight types of tissue used in the simulation. The starting temperatures of all tissue regions and blood pools and the set-point temperatures are also included. Next are the coefficients for receptors and effectors used in the physiological equations in the model. Finally, the routine reads miscellaneous coefficients which can conveniently be removed from the main code as variables without loss of clarity.

The subroutine PHYSIC uses the above data to build a geometrical representation of the body, as described in Section 4.2 of Volume 1. The size and shape of the spherical and cylindrical elements will vary according to the body characteristics of the subject being simulated. All dimensions and masses of the 150 tissue shells, required later in the heat flow calculations, are
calculated in this routine.

The subroutine BLOOD calculates the total blood volume of a male subject. It then distributes this volume among the 30 major and 90 minor venous and arterial pools. The dimensions of all blood vessels are then calculated. These dimensions allow the calculation of heat transfer coefficients between arteries and veins and between either type of vessel and surrounding tissue.

The subroutine THERMO calculates the thermal capacitances of the tissue shells and blood pools, using weighted mean values where more than one tissue type is present in a shell. The thermal conductances between shells are then determined, again using mean values at shell boundaries between different tissue types.

At this point in the sequence, the main program segment calls the subroutine COVER if clothing is present. This subroutine determines the conductance of the clothing layer on each of the 15 body segments. It also calculates the new outer diameters and conduction path lengths from the skin centres of mass to the clothing surface.

Before the iterative simulation sequence begins, subroutine SETUP is called. In this routine, the initial values of relevant variables are set to their starting values (which may be different from the values at rest). The variables to be set include temperatures and metabolic rates for all tissue shells and temperatures of blood pools. An estimate is also made of the true surface temperature of each segment, as opposed to the temperature at the centre of mass within the skin shells.

The subroutine DETAIL prints the values of variables which are of use during program development. These include the outer dimensions of the segments and all the
internal dimensions of the tissue shells. Selected thermal characteristics and starting conditions are also printed. The routine may be altered by a programmer to include any variable of interest since all are passed by the COMMON blocks. If BPRINT is set to 'YES' in the input file, this subroutine is called. Normally, it is set to 'NOT'.

Subroutine CONDIT is called each time a new condition interval begins, i.e., each time the exposure conditions are allowed to change. It re-calculates the properties of the fluid (viscosity, conductivity, etc.) as functions of temperature, using the mean of the body surface and bulk fluid temperatures. This enables new convective surface heat transfer coefficients to be determined for each new set of conditions. In addition, the conductances of the skin layers are re-calculated, since these vary with temperature. Finally, the vapour pressure of expired air is re-calculated, to account for any change in body core temperature which has occurred since the last period.

Subroutine METAB determines the physiological responses of the body to the exposure conditions. It gives values to the afferent inputs to the metabolic controllers and then calculates the efferent signals to the effectors for sweating, shivering, vasodilation and vasoconstriction. It also calculates the new metabolic rate for each tissue shell at each time interval, taking into account work as well as shivering. The new blood flow rates are determined and then summed to find the total flow in each blood vessel. Finally, the respiration volume and hence respiratory heat exchange are calculated.

Subroutine HEATEX calculates the heat exchange rates in all body regions by conduction and convection from blood flow. It then solves a heat balance equation for each tissue shell and blood pool to determine their net heat flows. In so doing, it takes account of all the active
and passive sources of heat exchange evaluated by previous subroutines.

Subroutine UPDATE takes the heat flow rates and, using the thermal capacitance for each shell and blood pool, calculates the rates of change of temperature. These then undergo a simple trapezoidal integration routine to determine the temperature changes and hence the new set of temperatures. However, the size of the time step is first reduced by a series of tests to ensure that no temperature changes by more than 0.1°C. This ensures numerical stability. The true surface temperature is then re-calculated and, if the body is unclothed, the changes in conductivity with temperature of muscle, fat and skin are accounted for.

Subroutine OUTPUT is called at the completion of each print interval. If it is being called for the first time, it prints the title string and the headings for the columns of figures. It also calculates initial values of the relevant variables and prints these. At each subsequent call, it prints a row of values for the selected variables. Currently, these are: Time, Environmental Temperature, Heat Flow, Metabolic Rate, Respiratory Heat Flow, Mean Body Temperature, Mean Skin Temperature, Head Core Temperature, Thorax Core Temperature, Abdomen Core Temperature and Skin Blood Flow. A programmer could select a different set of variables, if desired.

The simulation thus encompasses a body representation phase, a setting-up phase and an inner and outer repetitive loop. The outer loop is traversed each time the environmental condition period is completed. The inner loop is traversed at each time step and its subroutines form a sub-set of those in the outer loop. When the simulated exposure is complete, the master segment prints an end message and sounds a 'beep' at the
1.3. Notes on operation

The program is designed to be run either on-line from a terminal or off-line as a background job. Data preparation is not carried out interactively in the present version. A text editor, available as a part of most computer operating systems, is used to prepare a file such as the example '0724R.DAT' shown in Section 1.4. This file contains all the necessary data on the subject and exposure. It is inadvisable to include the subject's name or initials in the file containing his/her personal data. The necessity to register the system, as required in the United Kingdom by the Data Protection Act, 1984, will then be avoided. A numbering system may be employed instead.

The file also gives the user a description of the input variables, the units in which they are expressed and the format required for each by the program. In practice, it is a simple matter to edit an earlier file to provide the data required. The .DAT qualifier is used by the Digital Equipment Corporation in their VAX family of computers to indicate a data file. The model also requires as input the file MODEL.DAT, which is also listed in Section 1.4. This will not require to be changed unless further development of the program is taking place or new data are being substituted.

The program reads these two files from Logical Units 1 and 2 respectively. Assignments must therefore be made, which will take the form: 'ASSIGN 0724R.DAT FOR001' and 'ASSIGN MODEL.DAT FOR002' in the VMS operating system (VAX Management System). The output is directed to Logical Unit 5, which is the terminal display in VMS. However, a further assignment of an output file may be made to FOR005 as an alternative. If more than one terminal, if this is the output destination.
simulation is required, it is best to construct a command file to carry out the assignments and run commands sequentially:

$ASS MODEL.DAT FOR002
$ASS 0112R.DAT FOR001
$ASS 0112R.OUT FOR005
$RUN MODEL
$ASS 0212R.DAT FOR001
$ASS 0212R.OUT FOR005
$RUN MODEL
etc.

The form of the output file is shown in Section 1.4. The first part is the detailed print-out of body characteristics, which is not normally selected. The second part is the time history of the most usually required variables.

As a test of portability, the program has been compiled and run successfully on six different computers:

VAX 11-750
VAXstation II
Sun 3/110
Harris 500
BBC 'B'/Cambridge co-processor
IBM 'AT'

The statistics from a compilation, link and run on the VAXstation II are shown in Table 1.1. The VAX family of computers are 'virtual storage' machines. Not all of the executable code or data is held in working memory at any one time but the required portions are called from disc storage as required. On a non-virtual storage machine, the working memory size required is likely to be similar to the virtual memory size indicated in Table 1.
File storage requirements (bytes)

FORTRAN source code file 71680
COMMON.FOR file 4608
MODEL.DAT file 11766
Object code 40960
Executable code 30208
Input data file 3584
Output data file 1536

164342

Storage requirements for processing (Mbytes)

<table>
<thead>
<tr>
<th></th>
<th>Working Memory</th>
<th>Virtual Memory</th>
</tr>
</thead>
<tbody>
<tr>
<td>Compilation</td>
<td>0.677</td>
<td>2.470</td>
</tr>
<tr>
<td>Linking</td>
<td>0.396</td>
<td>2.441</td>
</tr>
<tr>
<td>Running</td>
<td>0.265</td>
<td>2.021</td>
</tr>
</tbody>
</table>

CPU time to compile 113.59 sec on VAXstation II.

TABLE 1.1 - COMPILETION AND STORAGE STATISTICS
It is seen that the time taken to compile was less than two minutes. The total storage requirement for simply running the compiled program and storing only one file of each type is just over 300 kbyte. If it is desired to compile and link the program in addition, then nearly 850 kbyte is required.

The running speed on a particular computer depends upon the body size of the subject and on the temperature of the water. This is because the size of the time step is reduced to prevent large changes in temperature. Thus, the simulation of a small subject in cold water will run more slowly than one of a large subject in warmer water. As an example of speed, the simulation runs at approximately 15% real time on the VAXstation II for a 50.5 kg subject unclothed in water at 12°C. (That is to say, 100 sec in real life takes 15 sec in the simulation).

In addition, the running speed on virtual memory machines depends on the maximum storage size allocated to a user. The smaller the working memory allocation, the more time the computer will need to spend swapping code and data between physical memory and disc.

The program is computationally intensive, with comparatively little input and output. If the computer in use has a floating point processor, the simulation will run faster. For example, on the VAXstation II, it runs approximately four times as fast as on the VAX 11-750, which has a similar architecture in other respects.

During program development, compiling and running the program in 'de-bug' mode has been found extremely useful. This enables single steps to be made in the computation, breaks to be inserted and the value of any variable to be examined. The use of a single COMMON file, included in
each subroutine, makes all variables available. Thus, the inner workings of the simulation may be made visible to the operator.

1.4. Listings of input and output files

NOTE: In the following tables, spacing has been altered in order to display the data on the printed page. Input tables should be constructed in accordance with the FORMAT statements, which give the correct spacing.

A typical input table would take the following form:

INPUT FILE 1 - SUBJECT AND ENVIRONMENTAL DATA

TITLE=STRING UP TO 80 CHAR. CHOSEN BY USER FORMAT(5/,A80)
SUBJECT NO 7 IN WATER AT 24 DEG C

FLUID=ENVIRONMENT ("AIR" OR "WAT"). FORMAT( //,A3)
WAT

HT=HEIGHT IN METRES. FORMAT(//,F 4 .2)
1.85

MBODY=MASS IN KILOGRAMMES. FORMAT(//,F5.1)
84.0

PBF=% BODY FAT IF KNOWN; 00.0 IF NOT. FORMAT(//,F4.1)
22.2

SG=SPECIFIC GRAVITY KGM-3 00.0 IF UNKNOWN FORMAT(//,F4.2)
0.00

RMR=REST MET RATE. WM-2 00.0 IF UNKNOWN FORMAT(//,F4.1)
47.0

TR=START RECT. TEMP °C 00.00 IF UNKNOWN FORMAT(//,F5.2)
37.16

OUTFIT=CLOTHING PRESENCE "YES" OR "NOT". FORMAT(//,A3)
NOT

TOG=CLOTHING INSULATION TOGS. FORMAT(//,7F5.2,/8F5.2)
0.00 0.00 0.00 0.00 0.00 0.00 0.00
0.00 0.00 0.00 0.00 0.00 0.00 0.00

SPARE VARIABLE POSITIONS FORMAT(//7F5.2,/8F5.2)
0.00 0.00 0.00 0.00 0.00 0.00 0.00
0.00 0.00 0.00 0.00 0.00 0.00 0.00
SPARE VARIABLE POSITIONS FORMAT(//7F5.2,/8F5.2)
0.00 0.00 0.00 0.00 0.00 0.00 0.00
0.00 0.00 0.00 0.00 0.00 0.00 0.00

SPARE VARIABLE POSITIONS FORMAT(//7F5.2,/8F5.2)
0.00 0.00 0.00 0.00 0.00 0.00 0.00
0.00 0.00 0.00 0.00 0.00 0.00 0.00

BPRINT=BODY DATA PRINT-OUT "YES" OR "NOT" FORMAT(///,A3)
YES

***NOTE: PRINT < CONINT & CONINT = M*PRINT, (M=INTEGER) ***

PRINT=PRINT INTERVAL MIN. FORMAT(///,F6.1)
5.0

*** NOTE: CONINT < EXINT & EXINT = N*CONINT, (N=INTEGER)***

CONINT=TIME INTERVAL FOR CONDITIONS MIN FORMAT(///,F6.1)
5.0

****** NOTE: NUMBER (L) OF INTERVALS = EXINT/CONINT******

EXINT=TIME INTERVAL FOR EXPOSURE MIN. FORMAT(///,F6.1)
130.0

TW=WATER TEMP. °C FORMAT(///12F5.1,12F5.1,12F5.1)
24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0
24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0
24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0 24.0

TAIR=AIR TEMP °C. FORMAT(///12F5.1,12F5.1,12F5.1)
21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0
21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0
21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0 21.0

ARH=RELATIVE HUMIDITY %. FORMAT(///12F5.1,12F5.1,12F5.1)
30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0
30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0
30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0 30.0

AV=FLUID VELOCITY MS-1 FORMAT(///12F5.2,12F5.2,12F5.2)
0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20
0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20
0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20 0.20

AWORK=WORK RATE W FORMAT(///12F6.1,12F6.1,12F6.1)
0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0
0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0
0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0 0.0

BP=BAROMETRIC PRESSURE, PASCALS. FORMAT(///,F8.1)
101324.0

******************************************************************************
The model data table is as follows:

<table>
<thead>
<tr>
<th>INPUT FILE 2 - PARAMETERS AND COEFFICIENTS USED</th>
</tr>
</thead>
<tbody>
<tr>
<td>DENSE(K) = TISSUE &amp; BLOOD DENSITIES KGM-3 FORMAT(//////,8F7.1)</td>
</tr>
<tr>
<td>1300.0 1050.0 0550.0 1050.0 1050.0 0850.0 1000.0 1000.0</td>
</tr>
<tr>
<td>CONDB(K) = TISSUE &amp; BLOOD TH. COND. WM-1°C-1 FORMAT(///,8F7.4)</td>
</tr>
<tr>
<td>1.1630 0.5275 0.2820 0.5466 0.4200 0.1600 0.2093 0.5466</td>
</tr>
<tr>
<td>FA(I) = FRACT. OF BODY AREA PER SEG. FORMAT(///,8F6.3,7F6.3)</td>
</tr>
<tr>
<td>0.070 0.180 0.180 0.095 0.063 0.034 0.095 0.063</td>
</tr>
<tr>
<td>0.034 0.034 0.034 0.025 0.034 0.034 0.025</td>
</tr>
<tr>
<td>FBONE(I) = FRACT. OF VSEG REP. BONE FORMAT(///,5F9.6,2(5F9.6))</td>
</tr>
<tr>
<td>0.075875 0.018375 0.018375 0.060600 0.060600</td>
</tr>
<tr>
<td>0.095350 0.060600 0.060600 0.095350 0.053375</td>
</tr>
<tr>
<td>0.053375 0.085825 0.053375 0.053375 0.085825</td>
</tr>
<tr>
<td>FVISC(I) = FRACT. VSEG REP. VISCERA FORMAT(///,5F9.6,2(5F9.6))</td>
</tr>
<tr>
<td>0.111325 0.060725 0.060725 0.023100 0.023100</td>
</tr>
<tr>
<td>0.015475 0.023100 0.023100 0.015475 0.026100</td>
</tr>
<tr>
<td>0.026100 0.011200 0.026100 0.026100 0.011200</td>
</tr>
<tr>
<td>FMUSC(I) = FRACT. VSEG REP. MUSC. FORMAT(///,5F9.6,2(5F9.6))</td>
</tr>
<tr>
<td>0.023000 0.116225 0.116225 0.123100 0.123100</td>
</tr>
<tr>
<td>0.018050 0.123100 0.123100 0.018050 0.119225</td>
</tr>
<tr>
<td>0.119225 0.026125 0.119225 0.119225 0.026125</td>
</tr>
<tr>
<td>FFAT(I) = FRACTION OF VSEG REP. FAT FORMAT(///,5F9.6,2(5F9.6))</td>
</tr>
<tr>
<td>0.092000 0.183600 0.183600 0.115000 0.115000</td>
</tr>
<tr>
<td>0.236800 0.115000 0.115000 0.236800 0.137300</td>
</tr>
<tr>
<td>0.137300 0.223800 0.137300 0.137300 0.223800</td>
</tr>
<tr>
<td>FSkin(I) = FRACT. OF VSEG REP. SKIN FORMAT(///,5F9.6,2(5F9.6))</td>
</tr>
<tr>
<td>0.067200 0.035100 0.035100 0.057800 0.057800</td>
</tr>
<tr>
<td>0.247700 0.057800 0.057800 0.247700 0.067900</td>
</tr>
<tr>
<td>0.067900 0.283600 0.067900 0.067900 0.283600</td>
</tr>
<tr>
<td>SPHEAT(K) = SP HT TISSUE &amp; BLOOD JKG-1°C-1 FORMAT(///,8F7.1)</td>
</tr>
<tr>
<td>1591.0 3696.9 3717.9 3696.9 3768.1 2512.1 3768.1 3744.0</td>
</tr>
<tr>
<td>FBA(K) = FRACT. TOTAL BLOOD VOL IN ART. POOLS FORMAT(///,6F6.3)</td>
</tr>
<tr>
<td>0.080 0.070 0.060 0.035 0.050 0.015</td>
</tr>
<tr>
<td>FBV(K) = FRACT. TOTAL BLOOD VOL IN VEN. POOLS FORMAT(///,6F6.3)</td>
</tr>
<tr>
<td>0.305 0.200 0.035 0.035 0.050 0.015</td>
</tr>
<tr>
<td>FQB(K) = REST MET. RATE PER UNIT TISSUE VOL FORMAT(///,6F7.1)</td>
</tr>
<tr>
<td>0000.0 8769.6 0368.9 5910.3 0340.8 0299.4</td>
</tr>
<tr>
<td>FBFB(K) = REST BLOOD FLOW PER UNIT MET RATE FORMAT(///,6F7.3)</td>
</tr>
<tr>
<td>00.000 00.836 00.669 01.108 00.287 00.286</td>
</tr>
</tbody>
</table>
**FQBS(I)** = REST MET. RATE PER SKIN VOL FORMAT(///, 8F7.1, /7F7.1)

<table>
<thead>
<tr>
<th>I</th>
<th>370.4</th>
<th>348.2</th>
<th>348.2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>308.3</td>
<td>333.3</td>
<td>308.3</td>
</tr>
<tr>
<td></td>
<td>333.3</td>
<td>312.5</td>
<td>312.5</td>
</tr>
<tr>
<td></td>
<td>315.8</td>
<td>312.5</td>
<td>315.8</td>
</tr>
</tbody>
</table>

**FBFBS(I)** = REST BLOOD FL PER SKIN VOL FORMAT(///, 8F7.1, /7F7.1)

<table>
<thead>
<tr>
<th>I</th>
<th>1481.5</th>
<th>1666.7</th>
<th>1666.7</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>659.7</td>
<td>3472.2</td>
<td>659.7</td>
</tr>
<tr>
<td></td>
<td>3472.2</td>
<td>1157.4</td>
<td>2924.0</td>
</tr>
<tr>
<td></td>
<td>2924.0</td>
<td>1157.4</td>
<td>2924.0</td>
</tr>
</tbody>
</table>

**SAT. VAP. PRESS. ARRAY PASCAL. FORMAT (///,**6F8.1, /5F8.1)**

<table>
<thead>
<tr>
<th></th>
<th>610.5</th>
<th>872.3</th>
<th>1227.8</th>
<th>1704.9</th>
<th>2337.8</th>
<th>3167.2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>4242.8</td>
<td>5622.9</td>
<td>7375.9</td>
<td>9583.2</td>
<td>12333.6</td>
<td></td>
</tr>
</tbody>
</table>

**TSET(I,J)** = SET-POINT TEMP, °C. I=1,15 J=1,10 FORMAT(F6.2)

<table>
<thead>
<tr>
<th>I</th>
<th>37.08</th>
<th>37.04</th>
<th>36.94</th>
<th>36.72</th>
<th>36.44</th>
<th>36.35</th>
<th>36.26</th>
<th>36.18</th>
<th>35.68</th>
<th>35.06</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>37.01</td>
<td>37.00</td>
<td>37.00</td>
<td>36.98</td>
<td>36.94</td>
<td>36.83</td>
<td>36.66</td>
<td>36.45</td>
<td>35.78</td>
<td>34.97</td>
</tr>
<tr>
<td></td>
<td>37.06</td>
<td>37.06</td>
<td>37.05</td>
<td>37.05</td>
<td>37.01</td>
<td>36.90</td>
<td>36.73</td>
<td>36.52</td>
<td>35.85</td>
<td>35.00</td>
</tr>
<tr>
<td></td>
<td>36.94</td>
<td>36.84</td>
<td>36.73</td>
<td>36.61</td>
<td>36.34</td>
<td>36.02</td>
<td>35.76</td>
<td>35.53</td>
<td>35.19</td>
<td>34.83</td>
</tr>
<tr>
<td></td>
<td>36.52</td>
<td>36.42</td>
<td>36.32</td>
<td>36.21</td>
<td>35.98</td>
<td>35.71</td>
<td>35.50</td>
<td>35.21</td>
<td>35.04</td>
<td>34.76</td>
</tr>
<tr>
<td></td>
<td>36.42</td>
<td>36.38</td>
<td>36.34</td>
<td>36.30</td>
<td>36.28</td>
<td>36.26</td>
<td>36.25</td>
<td>36.24</td>
<td>36.04</td>
<td>35.70</td>
</tr>
<tr>
<td></td>
<td>36.42</td>
<td>36.42</td>
<td>36.32</td>
<td>36.21</td>
<td>35.98</td>
<td>35.71</td>
<td>35.50</td>
<td>35.21</td>
<td>35.04</td>
<td>34.76</td>
</tr>
<tr>
<td></td>
<td>36.42</td>
<td>36.38</td>
<td>36.34</td>
<td>36.30</td>
<td>36.28</td>
<td>36.26</td>
<td>36.25</td>
<td>36.24</td>
<td>36.04</td>
<td>35.70</td>
</tr>
<tr>
<td></td>
<td>36.50</td>
<td>36.41</td>
<td>36.31</td>
<td>36.21</td>
<td>36.02</td>
<td>35.78</td>
<td>35.58</td>
<td>35.41</td>
<td>35.11</td>
<td>34.80</td>
</tr>
<tr>
<td></td>
<td>36.25</td>
<td>36.17</td>
<td>36.08</td>
<td>36.00</td>
<td>35.82</td>
<td>35.62</td>
<td>35.44</td>
<td>34.29</td>
<td>34.03</td>
<td>34.76</td>
</tr>
<tr>
<td></td>
<td>36.07</td>
<td>36.04</td>
<td>36.01</td>
<td>35.99</td>
<td>35.96</td>
<td>35.95</td>
<td>35.93</td>
<td>35.92</td>
<td>35.78</td>
<td>35.52</td>
</tr>
</tbody>
</table>

**T(I,J)** = STARTING TEMP, °C. I=1,15 J=1,10 FORMAT(F6.2)

<table>
<thead>
<tr>
<th>I</th>
<th>37.08</th>
<th>37.04</th>
<th>36.94</th>
<th>36.72</th>
<th>36.44</th>
<th>36.35</th>
<th>36.26</th>
<th>36.18</th>
<th>35.68</th>
<th>35.06</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>37.01</td>
<td>37.00</td>
<td>37.00</td>
<td>36.98</td>
<td>36.94</td>
<td>36.83</td>
<td>36.66</td>
<td>36.45</td>
<td>35.78</td>
<td>34.97</td>
</tr>
<tr>
<td></td>
<td>37.06</td>
<td>37.06</td>
<td>37.05</td>
<td>37.05</td>
<td>37.01</td>
<td>36.90</td>
<td>36.73</td>
<td>36.52</td>
<td>35.85</td>
<td>35.00</td>
</tr>
<tr>
<td></td>
<td>36.94</td>
<td>36.84</td>
<td>36.73</td>
<td>36.61</td>
<td>36.34</td>
<td>36.02</td>
<td>35.76</td>
<td>35.53</td>
<td>35.19</td>
<td>34.83</td>
</tr>
<tr>
<td></td>
<td>36.52</td>
<td>36.42</td>
<td>36.32</td>
<td>36.21</td>
<td>35.98</td>
<td>35.71</td>
<td>35.50</td>
<td>35.21</td>
<td>35.04</td>
<td>34.76</td>
</tr>
<tr>
<td></td>
<td>36.42</td>
<td>36.38</td>
<td>36.34</td>
<td>36.30</td>
<td>36.28</td>
<td>36.26</td>
<td>36.25</td>
<td>36.24</td>
<td>36.04</td>
<td>35.70</td>
</tr>
<tr>
<td></td>
<td>36.42</td>
<td>36.42</td>
<td>36.32</td>
<td>36.21</td>
<td>35.98</td>
<td>35.71</td>
<td>35.50</td>
<td>35.21</td>
<td>35.04</td>
<td>34.76</td>
</tr>
<tr>
<td></td>
<td>36.42</td>
<td>36.38</td>
<td>36.34</td>
<td>36.30</td>
<td>36.28</td>
<td>36.26</td>
<td>36.25</td>
<td>36.24</td>
<td>36.04</td>
<td>35.70</td>
</tr>
<tr>
<td></td>
<td>36.50</td>
<td>36.41</td>
<td>36.31</td>
<td>36.21</td>
<td>36.02</td>
<td>35.78</td>
<td>35.58</td>
<td>35.41</td>
<td>35.11</td>
<td>34.80</td>
</tr>
<tr>
<td></td>
<td>36.25</td>
<td>36.17</td>
<td>36.08</td>
<td>36.00</td>
<td>35.82</td>
<td>35.62</td>
<td>35.44</td>
<td>34.29</td>
<td>34.03</td>
<td>34.76</td>
</tr>
<tr>
<td></td>
<td>36.07</td>
<td>36.04</td>
<td>36.01</td>
<td>35.99</td>
<td>35.96</td>
<td>35.95</td>
<td>35.93</td>
<td>35.92</td>
<td>35.78</td>
<td>35.52</td>
</tr>
</tbody>
</table>

**TAL(I)** = MAJ ART POOL START TEMP, °C FORMAT(///, 8F6.2, /7F6.2)

<table>
<thead>
<tr>
<th>I</th>
<th>36.70</th>
<th>36.70</th>
<th>36.70</th>
<th>36.66</th>
<th>36.57</th>
<th>36.49</th>
<th>36.66</th>
<th>36.57</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>36.49</td>
<td>36.62</td>
<td>36.49</td>
<td>36.35</td>
<td>36.62</td>
<td>36.49</td>
<td>36.35</td>
<td></td>
</tr>
</tbody>
</table>
TVL(I)=MAJ VEN POOL START TEMP, °C FORMAT(/••,8F6.2,/F6.2)
36.82 36.70 36.75 36.22 36.03 35.89 36.22 36.03
35.89 35.97 35.87 35.74 35.97 35.87 35.74

TAV(I)=VISC ART POOL START TEMP °C. FORMAT(/••,8F6.2,/F6.2)
36.91 36.98 36.98 36.77 36.38 36.37 36.77 36.38
36.37 36.38 36.16 36.04 36.38 36.16 36.04

TVV(I)=VISC VEN POOL START TEMP, °C FORMAT(/••,8F6.2,/F6.2)
36.93 36.99 37.03 36.78 36.37 36.37 36.78 36.37
36.37 36.37 36.14 36.03 36.37 36.14 36.03

TAM(I)=MUSC ART POOL START TEMP, °C FORMAT(/••,8F6.2,/F6.2)
36.31 36.72 36.79 35.91 35.63 36.26 35.91 35.63
36.26 35.70 35.55 35.94 35.70 35.55 35.94

TVM(I)=MUSC VEN POOL START TEMP, °C FORMAT(/••,8F6.2,/F6.2)
36.31 36.72 36.79 35.91 35.63 36.26 35.91 35.63
36.26 35.70 35.55 35.94 35.70 35.55 35.94

TAS(I)=SKIN ART POOL START TEMP, °C FORMAT(/••,8F6.2,/F6.2)
35.26 35.25 35.27 34.94 34.88 34.82 34.94 34.88
34.82 34.99 34.94 35.65 34.99 34.94 35.65

TVS(I)=SKIN VEN POOL START TEMP, °C FORMAT(/••,8F6.2,/F6.2)
35.14 35.09 35.11 34.88 34.81 34.74 34.88 34.81
34.74 34.88 34.84 35.57 34.88 34.84 35.57

EB(I)=REST EVAPORATION LOSS, WATTS. FORMAT(/••,8F6.1,/F6.1)
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0

SKIP 4 RECORDS TO START OF RATE TABLE, FORMAT(/••/)

RATE(I,J)=SENS TO RATE OF TEMP. FORMAT(F6.1)
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
000.0 000.0 000.0 000.0 000.0 000.0 000.0 000.0
<table>
<thead>
<tr>
<th>Variable</th>
<th>Description</th>
<th>Units</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>SKINR(I)</td>
<td>Fract Thermoreceptors in I Format</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.21000 0.21000 0.21000 0.06000 0.04000 0.01500 0.06000 0.04000</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.01500 0.02500 0.02500 0.02000 0.02500 0.02500 0.02000</td>
</tr>
<tr>
<td>SKINS(I)</td>
<td>Fract Sweat Effectors in I Format</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.08100 0.24050 0.24050 0.06540 0.04360 0.01750 0.06540 0.04360</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.01750 0.03850 0.03850 0.01550 0.03850 0.03850 0.01550</td>
</tr>
<tr>
<td>SKINV(I)</td>
<td>Fract Vasodilat. Effectors in I Format</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.13200 0.16100 0.16100 0.06900 0.04600 0.05000 0.06900 0.04600</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.05000 0.02375 0.02375 0.06050 0.02375 0.02375 0.06050</td>
</tr>
<tr>
<td>SKINC(I)</td>
<td>Fract Vasoco Effectors in I Format</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.05000 0.07500 0.07500 0.01500 0.01000 0.17500 0.01500 0.01000</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.17500 0.01250 0.01250 0.17500 0.01250 0.01250 0.17500</td>
</tr>
<tr>
<td>FWORK(I)</td>
<td>Fract Musc Work in Segment I. Format</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.00000 0.03750 0.03750 0.05250 0.02250 0.00125 0.05250 0.02250</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.00125 0.00500 0.00500 0.00125 0.00500 0.00500 0.00125</td>
</tr>
<tr>
<td>FCHIL(I)</td>
<td>Fract Musc Shivering in I Format</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.005000 0.106250 0.106250 0.006125 0.002625 0.000000 0.006125 0.002625</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.000000 0.003125 0.003125 0.000000 0.003125 0.003125 0.000000</td>
</tr>
<tr>
<td>CSW</td>
<td>Afferent Sweat Coeff Head Core, °C-1 Format</td>
<td></td>
<td>372.160</td>
</tr>
<tr>
<td>SSW</td>
<td>Afferent Sweat Coeff Skin, °C-1 Format</td>
<td></td>
<td>033.727</td>
</tr>
<tr>
<td>PSW</td>
<td>Sweat Product Skin &amp; Head Core, °C-1 Format</td>
<td></td>
<td>000.000</td>
</tr>
<tr>
<td>CDIL</td>
<td>Vasodil Coeff Head Core MLS-1°C-1 Format</td>
<td></td>
<td>032.500</td>
</tr>
<tr>
<td>SDIL</td>
<td>Vasodil Coeff Skin MLS-1°C-1 Format</td>
<td></td>
<td>002.083</td>
</tr>
<tr>
<td>PDIL</td>
<td>Vasodil Coeff Skin &amp; Head MLS-1°C-2 Format</td>
<td></td>
<td>000.000</td>
</tr>
<tr>
<td>CCHIL</td>
<td>Shivering Coeff Head Core °C-1 Format</td>
<td></td>
<td>000.000</td>
</tr>
<tr>
<td>SCHIL</td>
<td>Shivering Coeff From Skin, °C-1 Format</td>
<td></td>
<td>000.000</td>
</tr>
<tr>
<td>PCHIL</td>
<td>Shivering Prod Coeff Skin &amp; Head °C-2 Format</td>
<td></td>
<td>021.000</td>
</tr>
<tr>
<td>CCON</td>
<td>Vasocon Coeff Head Core, °C-1 Format</td>
<td></td>
<td>005.000</td>
</tr>
<tr>
<td>SCON</td>
<td>Vasocon Coeff Skin, °C-1 Format</td>
<td></td>
<td>005.000</td>
</tr>
<tr>
<td>PCON</td>
<td>Vasocon Coeff Skin &amp; Head °C-2 Format</td>
<td></td>
<td>000.000</td>
</tr>
</tbody>
</table>
QSBF=EXPONENT SKIN BLOOD FLOW TEMP S-1 FORMAT(//,F7.3)
006.000

RAS=RAD. SMALL ARTERIES, MICRO M. FORMAT(//,F4.0)
250

TBF=TIME CONSTANT OF BLOOD FLOW LAG, S (FORMAT(//,F4.0)
60

PBFINT=% BODY FAT CONSIDERED TO BE INTERNAL FORMAT(//,F4.1)
24.9

EFF= % SPECIFIED WORK CONSIDERED EXTERNAL. FORMAT(//,F4.0)
22

DENW = DENSITY OF WATER KG M-3 FORMAT(//,F4.0)
1000

SHW = SPECIFIC HEAT OF WATER J KG-1 DEG C-1 FORMAT(//,F6.1)
4185.5

SHA = SPECIFIC HEAT OF AIR J KG-1 DEG C-1 FORMAT(//,F4.0)
1040

PBFS = % BODY FAT BY WEIGHT FOR STANDARD MAN FORMAT(//,F4.1)
11.9

DENS = DENSITY OF STANDARD MAN KG M-3 FORMAT(//,F8.3)
1074.569

RMRS = REST MET RATE OF STANDARD MAN W M-2 FORMAT(//,F4.1)
45.8

RLP(15)=LEN/PERIMETER FOR CYLINDERS FORMAT(//,8F5.2,/7F5.2)
0.00 0.40 0.40 0.80 1.00 0.75 0.80 1.00
0.75 0.75 1.00 0.75 0.75 1.00 0.75

CONVW = CONVERSION FACTOR KCAL HR-1 TO WATTS FORMAT(//,F5.3)
1.163

********************************************************************
The output of the model, given the two input files above, would be as listed below. First, the thermal and physical data on the simulated body:

**BODY S.G.**=1.05355 % **BODY FAT**=22.200

**SURFACE AREA**= 1.9696 SQ M **VOLUME**= 0.0872 METRE**3**

**WEIGHT**= 1.85 METRES **BODY MASS** = 83.9999 KG

**HEAD RADIUS**= 0.1041 M **HEAD VOL.**= 0.0047 METRE**3**

**THORAX LENGTH**= 0.3917 M **THORAX VOL.**= 0.0256 METRE**3**

**ABDOMEN LENGTH**= 0.3917 M **ABDOMEN VOL.**= 0.0256 METRE**3**

**THIGH LENGTH**= 0.4024 M **THIGH VOL.**= 0.0069 METRE**3** EACH

**CALF LENGTH**= 0.3664 M **CALF VOL.**= 0.0034 METRE**3** EACH

**FOOT LENGTH**= 0.2331 M **FOOT VOL.**= 0.0015 METRE**3** EACH

**U.ARM LENGTH**= 0.2331 M **U.ARM VOL.**= 0.0015 METRE**3** EACH

**L.ARM LENGTH**= 0.2691 M **L.ARM VOL.**= 0.0013 METRE**3** EACH

**HAND LENGTH**= 0.1999 M **HAND VOL.**= 0.0010 METRE**3** EACH

**SEGMENT DIAMETERS IN METRES**

0.2083 0.2884 0.2883 0.1482 0.1079 0.0915 0.1482 0.1079
0.0915 0.0915 0.0793 0.0784 0.0915 0.0793 0.0784

**TISSUE SHELL OUTER RADII IN METRES**

0.059 0.075 0.085 0.094 0.095 0.096 0.097 0.098 0.102 0.104
0.045 0.064 0.078 0.091 0.101 0.111 0.120 0.128 0.142 0.144
0.042 0.059 0.073 0.084 0.097 0.108 0.118 0.127 0.142 0.144
0.021 0.030 0.037 0.042 0.050 0.056 0.062 0.067 0.072 0.074
0.015 0.022 0.027 0.031 0.036 0.041 0.045 0.049 0.052 0.054
0.015 0.021 0.026 0.030 0.031 0.031 0.032 0.032 0.040 0.046
0.021 0.030 0.037 0.042 0.050 0.056 0.062 0.067 0.072 0.074
0.015 0.022 0.027 0.031 0.036 0.041 0.045 0.049 0.052 0.054
0.015 0.021 0.026 0.030 0.031 0.031 0.032 0.032 0.040 0.046
0.013 0.018 0.022 0.026 0.030 0.034 0.037 0.040 0.044 0.046
0.011 0.016 0.019 0.022 0.026 0.029 0.032 0.035 0.038 0.040
<table>
<thead>
<tr>
<th>TISSUE SHELL CENTRE-OF-MASS RADII IN METRES</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.0470 0.0679 0.0805 0.0900 0.0946 0.0956 0.0965 0.0975 0.0999 0.1031</td>
</tr>
<tr>
<td>0.0320 0.0555 0.0716 0.0847 0.0961 0.1064 0.1157 0.1243 0.1353 0.1430</td>
</tr>
<tr>
<td>0.0297 0.0515 0.0664 0.0786 0.0905 0.1022 0.1127 0.1223 0.1345 0.1429</td>
</tr>
<tr>
<td>0.0149 0.0258 0.0334 0.0395 0.0460 0.0528 0.0588 0.0643 0.0694 0.0730</td>
</tr>
<tr>
<td>0.0109 0.0188 0.0243 0.0287 0.0335 0.0385 0.0429 0.0468 0.0505 0.0531</td>
</tr>
<tr>
<td>0.0106 0.0183 0.0237 0.0280 0.0302 0.0308 0.0314 0.0320 0.0361 0.0428</td>
</tr>
<tr>
<td>0.0149 0.0258 0.0334 0.0395 0.0460 0.0528 0.0588 0.0643 0.0694 0.0730</td>
</tr>
<tr>
<td>0.0109 0.0188 0.0243 0.0287 0.0335 0.0385 0.0429 0.0468 0.0505 0.0531</td>
</tr>
<tr>
<td>0.0106 0.0183 0.0237 0.0280 0.0302 0.0308 0.0314 0.0320 0.0361 0.0428</td>
</tr>
<tr>
<td>0.0091 0.0157 0.0203 0.0240 0.0280 0.0321 0.0357 0.0389 0.0424 0.0450</td>
</tr>
<tr>
<td>0.0079 0.0136 0.0176 0.0208 0.0242 0.0278 0.0309 0.0337 0.0367 0.0389</td>
</tr>
<tr>
<td>0.0085 0.0148 0.0191 0.0225 0.0245 0.0253 0.0260 0.0267 0.0303 0.0363</td>
</tr>
<tr>
<td>0.0091 0.0157 0.0203 0.0240 0.0280 0.0321 0.0357 0.0389 0.0424 0.0450</td>
</tr>
<tr>
<td>0.0079 0.0136 0.0176 0.0208 0.0242 0.0278 0.0309 0.0337 0.0367 0.0389</td>
</tr>
<tr>
<td>0.0085 0.0148 0.0191 0.0225 0.0245 0.0253 0.0260 0.0267 0.0303 0.0363</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CONDUCTANCE PATH LENGTHS IN METRES</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.0209 0.0126 0.0095 0.0046 0.0010 0.0010 0.0009 0.0024 0.0032 0.0010</td>
</tr>
<tr>
<td>0.0234 0.0161 0.0131 0.0114 0.0102 0.0093 0.0086 0.0110 0.0077 0.0011</td>
</tr>
<tr>
<td>0.0217 0.0150 0.0122 0.0119 0.0117 0.0105 0.0096 0.0121 0.0085 0.0012</td>
</tr>
<tr>
<td>0.0109 0.0075 0.0061 0.0065 0.0068 0.0060 0.0055 0.0051 0.0036 0.0011</td>
</tr>
<tr>
<td>0.0080 0.0055 0.0045 0.0048 0.0050 0.0044 0.0040 0.0037 0.0026 0.0008</td>
</tr>
<tr>
<td>0.0077 0.0053 0.0043 0.0022 0.0006 0.0006 0.0006 0.0042 0.0067 0.0029</td>
</tr>
<tr>
<td>0.0109 0.0075 0.0061 0.0065 0.0068 0.0060 0.0055 0.0051 0.0036 0.0011</td>
</tr>
<tr>
<td>0.0080 0.0055 0.0045 0.0048 0.0050 0.0044 0.0040 0.0037 0.0026 0.0008</td>
</tr>
<tr>
<td>MID-PLANE RADII IN METRES</td>
</tr>
<tr>
<td>---------------------------</td>
</tr>
<tr>
<td>0.077 0.053 0.043 0.022 0.006 0.006 0.006 0.042 0.067 0.029</td>
</tr>
<tr>
<td>0.067 0.046 0.037 0.039 0.041 0.036 0.0033 0.0034 0.026 0.008</td>
</tr>
<tr>
<td>0.058 0.040 0.032 0.034 0.035 0.031 0.0028 0.0030 0.023 0.007</td>
</tr>
<tr>
<td>0.062 0.043 0.035 0.019 0.0008 0.0007 0.0007 0.036 0.060 0.029</td>
</tr>
<tr>
<td>0.067 0.046 0.037 0.039 0.041 0.036 0.0033 0.0034 0.026 0.008</td>
</tr>
<tr>
<td>0.058 0.040 0.032 0.034 0.035 0.031 0.0028 0.0030 0.023 0.007</td>
</tr>
<tr>
<td>0.062 0.043 0.035 0.019 0.0008 0.0007 0.0007 0.036 0.060 0.029</td>
</tr>
</tbody>
</table>

MID-PLANE AREAS IN METRE**2

<table>
<thead>
<tr>
<th>MID-PLANE AREAS IN METRE**2</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.124 0.180 0.222 0.260 0.030 0.039 0.0373 0.0406 0.0437 0.0454</td>
</tr>
<tr>
<td>0.124 0.180 0.222 0.260 0.030 0.039 0.0373 0.0406 0.0437 0.0454</td>
</tr>
<tr>
<td>0.077 0.040 0.032 0.034 0.035 0.031 0.0028 0.0030 0.023 0.007</td>
</tr>
<tr>
<td>0.067 0.046 0.037 0.039 0.041 0.036 0.0033 0.0034 0.026 0.008</td>
</tr>
<tr>
<td>0.058 0.040 0.032 0.034 0.035 0.031 0.0028 0.0030 0.023 0.007</td>
</tr>
<tr>
<td>0.062 0.043 0.035 0.019 0.0008 0.0007 0.0007 0.036 0.060 0.029</td>
</tr>
</tbody>
</table>

21
<table>
<thead>
<tr>
<th>Thermal Conductances in W/deg C</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.443  4.006  6.980  15.313  48.763  5.784  52.826  8.571  7.306  0.000</td>
</tr>
<tr>
<td>2.102  4.434  6.711  8.565  1.227  12.298  14.369  6.101  7.368  0.000</td>
</tr>
<tr>
<td>2.674  5.641  8.538  8.452  8.498  1.571  12.643  5.465  6.687  0.000</td>
</tr>
<tr>
<td>4.116  8.682  13.141  8.954  7.708  9.840  11.969  7.782  8.678  0.000</td>
</tr>
<tr>
<td>3.747  7.905  11.964  8.152  7.018  8.959  1.897  7.085  7.901  0.000</td>
</tr>
<tr>
<td>2.458  5.184  7.847  14.969  31.919  33.149  34.379  1.990  1.550  0.000</td>
</tr>
<tr>
<td>4.116  8.682  13.141  8.954  7.708  9.840  11.969  7.782  8.678  0.000</td>
</tr>
<tr>
<td>3.747  7.905  11.964  8.152  7.018  8.959  1.897  7.085  7.901  0.000</td>
</tr>
<tr>
<td>2.458  5.184  7.847  14.969  31.919  33.149  34.379  1.990  1.550  0.000</td>
</tr>
<tr>
<td>2.248  4.741  7.176  5.171  4.532  5.767  7.000  3.907  4.219  0.000</td>
</tr>
<tr>
<td>2.595  5.475  8.287  5.971  5.233  6.659  8.083  4.511  4.872  0.000</td>
</tr>
<tr>
<td>2.248  4.741  7.176  5.171  4.532  5.767  7.000  3.907  4.219  0.000</td>
</tr>
<tr>
<td>2.595  5.475  8.287  5.971  5.233  6.659  8.083  4.511  4.872  0.000</td>
</tr>
</tbody>
</table>

22
<table>
<thead>
<tr>
<th>TISSUE SHELL MASSES IN KG</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.928 0.928 0.928 0.928</td>
</tr>
<tr>
<td>2.014 2.014 2.014 2.014</td>
</tr>
<tr>
<td>2.196 2.196 2.196 2.196</td>
</tr>
<tr>
<td>0.633 0.633 0.633 0.633</td>
</tr>
<tr>
<td>0.306 0.306 0.306 0.306</td>
</tr>
<tr>
<td>0.186 0.186 0.186 0.186</td>
</tr>
<tr>
<td>0.633 0.633 0.633 0.633</td>
</tr>
<tr>
<td>0.306 0.306 0.306 0.306</td>
</tr>
<tr>
<td>0.186 0.186 0.186 0.186</td>
</tr>
<tr>
<td>0.134 0.134 0.134 0.134</td>
</tr>
<tr>
<td>0.116 0.116 0.116 0.116</td>
</tr>
<tr>
<td>0.103 0.103 0.103 0.103</td>
</tr>
<tr>
<td>0.134 0.134 0.134 0.134</td>
</tr>
<tr>
<td>0.116 0.116 0.116 0.116</td>
</tr>
<tr>
<td>0.103 0.103 0.103 0.103</td>
</tr>
<tr>
<td>0.204 0.204 0.204 0.204</td>
</tr>
<tr>
<td>0.176 0.176 0.176 0.176</td>
</tr>
<tr>
<td>0.057 0.057 0.057 0.057</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>RESTING METABOLIC HEAT PRODUCTION RATES IN WATTS</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.032 0.032 0.032 0.032 0.032 0.032 0.032 0.032</td>
</tr>
<tr>
<td>0.121 0.121 0.121 0.121 0.121 0.121 0.121 0.121</td>
</tr>
<tr>
<td>0.008 0.008 0.008 0.008 0.008 0.008 0.008 0.008</td>
</tr>
<tr>
<td>0.053 0.053 0.053 0.053 0.053 0.053 0.053 0.053</td>
</tr>
<tr>
<td>0.046 0.046 0.046 0.046 0.046 0.046 0.046 0.046</td>
</tr>
<tr>
<td>0.007 0.007 0.007 0.007 0.007 0.007 0.007 0.007</td>
</tr>
<tr>
<td>0.060 0.060 0.060 0.060 0.060 0.060 0.060 0.060</td>
</tr>
<tr>
<td>0.074 0.074 0.074 0.074 0.074 0.074 0.074 0.074</td>
</tr>
<tr>
<td>RESTING BLOOD FLOW RATES IN ML/SEC</td>
</tr>
<tr>
<td>-----------------------------------</td>
</tr>
<tr>
<td>2.127 2.127 2.127 2.127 0.009 0.009 0.009 0.009 0.035 0.500</td>
</tr>
<tr>
<td>2.800 2.800 2.800 2.800 0.215 0.215 0.215 0.215 0.330 1.348</td>
</tr>
<tr>
<td>5.887 5.887 5.887 5.887 0.234 0.234 0.234 0.234 0.361 1.468</td>
</tr>
<tr>
<td>0.287 0.287 0.287 0.287 0.072 0.072 0.072 0.072 0.064 0.277</td>
</tr>
<tr>
<td>0.139 0.139 0.139 0.139 0.035 0.035 0.035 0.035 0.031 0.134</td>
</tr>
<tr>
<td>0.025 0.025 0.025 0.025 0.002 0.002 0.002 0.002 0.029 1.315</td>
</tr>
<tr>
<td>0.287 0.287 0.287 0.287 0.072 0.072 0.072 0.072 0.064 0.277</td>
</tr>
<tr>
<td>0.139 0.139 0.139 0.139 0.035 0.035 0.035 0.035 0.031 0.134</td>
</tr>
<tr>
<td>0.025 0.025 0.025 0.025 0.002 0.002 0.002 0.002 0.029 1.315</td>
</tr>
<tr>
<td>0.081 0.081 0.081 0.081 0.015 0.015 0.015 0.015 0.017 0.124</td>
</tr>
<tr>
<td>0.070 0.070 0.070 0.070 0.013 0.013 0.013 0.013 0.015 0.107</td>
</tr>
<tr>
<td>0.010 0.010 0.010 0.010 0.002 0.002 0.002 0.002 0.017 0.799</td>
</tr>
<tr>
<td>0.081 0.081 0.081 0.081 0.015 0.015 0.015 0.015 0.017 0.124</td>
</tr>
<tr>
<td>0.070 0.070 0.070 0.070 0.013 0.013 0.013 0.013 0.015 0.107</td>
</tr>
<tr>
<td>0.010 0.010 0.010 0.010 0.002 0.002 0.002 0.002 0.017 0.799</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>THERMAL CAPACITANCES IN KJ/DEG C</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.542 2.542 2.542 2.542 0.404 0.404 0.404 0.404 0.979 1.177</td>
</tr>
<tr>
<td>5.413 5.413 5.413 5.413 1.236 1.236 1.236 1.236 9.985 3.087</td>
</tr>
<tr>
<td>0.610 0.610 0.610 0.610 1.510 1.510 1.510 1.510 0.859 0.708</td>
</tr>
<tr>
<td>0.338 0.338 0.338 0.338 0.095 0.095 0.095 0.095 0.800 1.294</td>
</tr>
<tr>
<td>0.610 0.610 0.610 0.610 1.510 1.510 1.510 1.510 0.859 0.708</td>
</tr>
<tr>
<td>0.338 0.338 0.338 0.338 0.095 0.095 0.095 0.095 0.800 1.294</td>
</tr>
</tbody>
</table>
SKIN AND WHOLE BODY THERMAL CAPACITANCES IN KJ/DEG C

CSKIN= 17.310  CBODY= 250.52

The time history of the variables is as follows:

SUBJECT NO 7 IN WATER AT 24 DEG C

<table>
<thead>
<tr>
<th>T</th>
<th>TENV</th>
<th>HFLW</th>
<th>MR</th>
<th>QR</th>
<th>TBODY</th>
<th>TSKIN</th>
<th>TAC</th>
<th>TO</th>
<th>TR</th>
<th>SBF</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>34.6</td>
<td>-44.2</td>
<td>92.9</td>
<td>-11.9</td>
<td>36.38</td>
<td>34.88</td>
<td>36.74</td>
<td>37.11</td>
<td>37.16</td>
<td>8.8</td>
</tr>
<tr>
<td>5</td>
<td>24.0</td>
<td>-262.7</td>
<td>200.5</td>
<td>-27.4</td>
<td>35.57</td>
<td>24.49</td>
<td>36.81</td>
<td>37.04</td>
<td>37.13</td>
<td>1.5</td>
</tr>
<tr>
<td>10</td>
<td>24.0</td>
<td>-211.2</td>
<td>201.7</td>
<td>-27.8</td>
<td>35.28</td>
<td>24.41</td>
<td>36.89</td>
<td>36.99</td>
<td>37.12</td>
<td>1.4</td>
</tr>
<tr>
<td>15</td>
<td>24.0</td>
<td>-189.1</td>
<td>202.6</td>
<td>-27.9</td>
<td>35.06</td>
<td>24.37</td>
<td>36.96</td>
<td>36.96</td>
<td>37.11</td>
<td>1.3</td>
</tr>
<tr>
<td>20</td>
<td>24.0</td>
<td>-175.8</td>
<td>203.9</td>
<td>-28.0</td>
<td>34.87</td>
<td>24.34</td>
<td>36.99</td>
<td>36.93</td>
<td>37.09</td>
<td>1.3</td>
</tr>
<tr>
<td>25</td>
<td>24.0</td>
<td>-166.8</td>
<td>205.9</td>
<td>-28.2</td>
<td>34.72</td>
<td>24.33</td>
<td>36.99</td>
<td>36.90</td>
<td>37.05</td>
<td>1.3</td>
</tr>
<tr>
<td>30</td>
<td>24.0</td>
<td>-160.1</td>
<td>208.3</td>
<td>-28.5</td>
<td>34.58</td>
<td>24.31</td>
<td>36.97</td>
<td>36.85</td>
<td>37.01</td>
<td>1.3</td>
</tr>
<tr>
<td>35</td>
<td>24.0</td>
<td>-155.0</td>
<td>211.1</td>
<td>-28.7</td>
<td>34.46</td>
<td>24.30</td>
<td>36.93</td>
<td>36.80</td>
<td>36.95</td>
<td>1.2</td>
</tr>
<tr>
<td>40</td>
<td>24.0</td>
<td>-151.0</td>
<td>214.2</td>
<td>-29.0</td>
<td>34.35</td>
<td>24.30</td>
<td>36.88</td>
<td>36.74</td>
<td>36.89</td>
<td>1.2</td>
</tr>
<tr>
<td>45</td>
<td>24.0</td>
<td>-147.7</td>
<td>217.3</td>
<td>-29.3</td>
<td>34.26</td>
<td>24.29</td>
<td>36.82</td>
<td>36.67</td>
<td>36.82</td>
<td>1.2</td>
</tr>
<tr>
<td>50</td>
<td>24.0</td>
<td>-145.1</td>
<td>220.6</td>
<td>-29.6</td>
<td>34.17</td>
<td>24.29</td>
<td>36.76</td>
<td>36.60</td>
<td>36.75</td>
<td>1.2</td>
</tr>
<tr>
<td>55</td>
<td>24.0</td>
<td>-142.9</td>
<td>223.7</td>
<td>-29.9</td>
<td>34.09</td>
<td>24.28</td>
<td>36.69</td>
<td>36.54</td>
<td>36.69</td>
<td>1.2</td>
</tr>
<tr>
<td>60</td>
<td>24.0</td>
<td>-141.1</td>
<td>226.8</td>
<td>-30.2</td>
<td>34.02</td>
<td>24.28</td>
<td>36.63</td>
<td>36.47</td>
<td>36.62</td>
<td>1.2</td>
</tr>
<tr>
<td>65</td>
<td>24.0</td>
<td>-139.6</td>
<td>229.9</td>
<td>-30.5</td>
<td>33.95</td>
<td>24.28</td>
<td>36.57</td>
<td>36.40</td>
<td>36.16</td>
<td>1.2</td>
</tr>
<tr>
<td>70</td>
<td>24.0</td>
<td>-138.3</td>
<td>232.7</td>
<td>-30.7</td>
<td>33.89</td>
<td>24.27</td>
<td>36.51</td>
<td>36.34</td>
<td>36.50</td>
<td>1.2</td>
</tr>
<tr>
<td>75</td>
<td>24.0</td>
<td>-137.1</td>
<td>235.4</td>
<td>-30.9</td>
<td>33.84</td>
<td>24.27</td>
<td>36.45</td>
<td>36.28</td>
<td>36.44</td>
<td>1.2</td>
</tr>
<tr>
<td>80</td>
<td>24.0</td>
<td>-136.2</td>
<td>238.0</td>
<td>-31.2</td>
<td>33.79</td>
<td>24.27</td>
<td>36.39</td>
<td>36.22</td>
<td>36.39</td>
<td>1.1</td>
</tr>
<tr>
<td>85</td>
<td>24.0</td>
<td>-135.3</td>
<td>240.4</td>
<td>-31.4</td>
<td>33.74</td>
<td>24.27</td>
<td>36.34</td>
<td>36.16</td>
<td>36.34</td>
<td>1.1</td>
</tr>
<tr>
<td>90</td>
<td>24.0</td>
<td>-134.6</td>
<td>242.7</td>
<td>-31.5</td>
<td>33.70</td>
<td>24.27</td>
<td>36.29</td>
<td>36.11</td>
<td>36.29</td>
<td>1.1</td>
</tr>
<tr>
<td>95</td>
<td>24.0</td>
<td>-133.9</td>
<td>244.8</td>
<td>-31.7</td>
<td>33.66</td>
<td>24.27</td>
<td>36.24</td>
<td>36.06</td>
<td>36.25</td>
<td>1.1</td>
</tr>
<tr>
<td>100</td>
<td>24.0</td>
<td>-133.3</td>
<td>246.8</td>
<td>-31.8</td>
<td>33.63</td>
<td>24.26</td>
<td>36.20</td>
<td>36.02</td>
<td>36.20</td>
<td>1.1</td>
</tr>
<tr>
<td>105</td>
<td>24.0</td>
<td>-132.8</td>
<td>248.6</td>
<td>-32.0</td>
<td>33.60</td>
<td>24.26</td>
<td>36.15</td>
<td>35.97</td>
<td>36.17</td>
<td>1.1</td>
</tr>
<tr>
<td>110</td>
<td>24.0</td>
<td>-132.4</td>
<td>250.3</td>
<td>-32.1</td>
<td>33.57</td>
<td>24.26</td>
<td>36.12</td>
<td>35.93</td>
<td>36.13</td>
<td>1.1</td>
</tr>
<tr>
<td>115</td>
<td>24.0</td>
<td>-132.0</td>
<td>251.8</td>
<td>-32.2</td>
<td>33.55</td>
<td>24.26</td>
<td>36.08</td>
<td>35.90</td>
<td>36.10</td>
<td>1.1</td>
</tr>
<tr>
<td>120</td>
<td>24.0</td>
<td>-131.6</td>
<td>253.2</td>
<td>-32.3</td>
<td>33.52</td>
<td>24.26</td>
<td>36.05</td>
<td>35.86</td>
<td>36.07</td>
<td>1.1</td>
</tr>
<tr>
<td>125</td>
<td>24.0</td>
<td>-131.3</td>
<td>254.5</td>
<td>-32.4</td>
<td>33.50</td>
<td>24.26</td>
<td>36.02</td>
<td>35.83</td>
<td>36.04</td>
<td>1.1</td>
</tr>
<tr>
<td>130</td>
<td>24.0</td>
<td>-131.0</td>
<td>255.8</td>
<td>-32.5</td>
<td>33.48</td>
<td>24.26</td>
<td>35.99</td>
<td>35.80</td>
<td>36.02</td>
<td>1.1</td>
</tr>
</tbody>
</table>

END OF SIMULATION
1.5. Program listing

NOTE: Tabulation stops are rejected by some compilers.
The required number of spaces should be used instead.

The source code file is as follows:

```
PROGRAM MASTER

C THIS PROGRAM SIMULATES THE RESPONSE OF THE HUMAN BODY
C TO HEAT OR COLD STRESS IN WATER, EITHER NUDE OR
C CLOTHED. IT WAS DEVELOPED BY GRAHAM RICHARDSON AT THE
C RAF INSTITUTE OF AVIATION MEDICINE, FARNBOROUGH,

INCLUDE 'COMMON.FOR'

C SEQUENCE STARTS HERE

CALL READ1
CALL READ2
CALL PHYSIC
CALL BLOOD
CALL THERMO
IF(OUTFIT.EQ.'YES') CALL COVER
CALL SETUP
IF(BPRINT.EQ.'YES') CALL DETAIL

L=0
PRNTME=0.0
CONTME=0.0
TIME=0.0
PTIME=0.0

C NEW CONDITION INTERVAL BEGINS HERE

0010 L=L+1
CALL CONDIT

C NEW INTEGRATION AND PRINT INTERVALS BEGIN HERE
```
CALL METAB

IF(TIME .EQ. 0.0) CALL OUTPUT

CALL HEATEX
CALL UPDATE

PRNTME = PRNTME + INT / 60.0
TIME = TIME + INT / 60.0

************************
C END OF PRINT INTERVAL?
************************

IF(PRNTME-PRINT) 0020,0030,0030

0030 PTIME = PTIME + PRINT
PRNTME = 0.0
CALL OUTPUT
CONTME = CONTME + PRINT

************************
C START NEW CONDITION INTERVAL?
************************

IF(CONTME-CONINT) 0020,0040,0040

****************************
C END OF SIMULATION?
****************************

0040 CONTME = 0.0

IF(TIME-EXINT) 0010,0050,0050

0050 BEEP = CHAR(7)
WRITE(5,0060)BEEP

0060 FORMAT (/,' END OF SIMULATION',/,'1X,1A1)
STOP
END

C END OF MASTER SEGMENT
SUBROUTINE READ1

C READS SUBJECT & ENVIRONMENT DATA FROM FILE 1

INCLUDE 'COMMON.FOR'

C READ SUBJECT DATA

READ (1,0010) TITLE
READ (1,0020) FLUID
READ (1,0030) HT
READ (1,0040) MBODY
READ (1,0050) PBF
READ (1,0030) SG
READ (1,0050) RMR
READ (1,0060) TR

0010 FORMAT (/////,A80)
0020 FORMAT (///,A3)
0030 FORMAT (///,F4.2)
0040 FORMAT (///,F5.1)
0050 FORMAT (///,F4.1)
0060 FORMAT (///,F5.2)

C READ CLOTHING DATA

READ (1,0070) OUTFIT
READ (1,0080) TOG

C THREE SPARE READ STATEMENTS FOR ADDITIONAL C CHARACTERISTICS

READ (1,0090)
READ (1,0100)
READ (1,0110)

0070 FORMAT (///,A3)
0080 FORMAT (///,7F5.2,/8F5.2)
0090 FORMAT (///)
0100 FORMAT (///)
0110 FORMAT (///)

C READ PRINT-OUT REQUIREMENT

READ (1,0120) BPRINT
0120 FORMAT (///,A3)

C READ VALUES FOR TIME INTERVALS

READ (1,0130) PRINT
READ (1,0130) CONINT
READ (1,0130) EXINT

0130 FORMAT (///,F6.1)
C READ ENVIRONMENTAL VARIABLES

READ(1,0140) TW
READ(1,0140) TAIR
READ(1,0140) ARH
READ(1,0150) AV
READ(1,0160) AWORK
READ(1,0170) BP

0140 FORMAT(//12F5.1,/12F5.1,/12F5.1)
0150 FORMAT(//12F5.2,/12F5.2,/12F5.2)
0160 FORMAT(//12F6.1,/12F6.1,/12F6.1)
0170 FORMAT(///,F8.1)
RETURN
END

C END OF SUBROUTINE READ1

********************************************************************
SUBROUTINE READ2
********************************************************************
C READS NON-SPECIFIC COEFFICIENTS AND PARAMETERS FROM
C FILE 2

********************************************************************
INCLUDE 'COMMON.FOR'

C READ TISSUE DENSITIES
READ(2,0010)DENSE
0010 FORMAT(/////,8F7.1)

C READ TISSUE CONDUCTIVITIES
READ(2,0020)CONDB
0020 FORMAT(///,8F7.4)

C READ SEGMENT AREA DISTRIBUTION
READ(2,0030)FA
0030 FORMAT(///,8F6.3,/7F6.3)

C READ TISSUE TYPE DISTRIBUTIONS
READ(2,0040)FBONE
READ(2,0040)FVISC
READ(2,0040)FMUSC
READ(2,0040)FFAT
READ(2,0040)FSKIN
0040 FORMAT(///,5F9.6,2(/5F9.6))

C READ SPECIFIC HEAT VALUES
READ(2,0050)SPHEAT
0050 FORMAT(///,8F7.1)
C READ BLOOD DIST. IN ARTERIES: LARGE, SMALL, PULMONARY, C HEART, LIVER, SPLEEN.
READ(2,0060)FBA

C READ BLOOD DIST. IN VEINS: LARGE, SMALL, PULMONARY, C HEART, LIVER, SPLEEN.
READ(2,0060)FBV

0060 FORMAT(//,6F6.3)

C READ RESTING METABOLIC RATE PER UNIT MASS OF BONE, C BRAIN, LUNG, ABDOMEN CORE, MUSCLE AND FAT.
READ(2,0070)FQB

C READ RESTING BLOODFLOW DISTRIBUTION FACTORS
READ(2,0080)FBFB

0070 FORMAT(//,6F7.1)
0080 FORMAT(//,6F7.3)

C READ SKIN RESTING METABOLIC RATE PER UNIT MASS
READ(2,0090)FQBS

C READ SKIN RESTING BLOOD FLOW PER UNIT MASS
READ(2,0090)FBFBS

0090 FORMAT(//,8F7.1,/7F7.1)

C READ VAPOUR PRESSURE CALCULATION ARRAY
READ(2,0100)P

0100 FORMAT(//,6F8.1,/5F8.1)

C READ BLANK LINES TO IGNORE INPUT FILE COMMENTS
READ(2,0110)

0110 FORMAT(////)

C READ SET-POINT TEMPERATURES
DO 0120 I=1,15
READ(2,0140) (TSET(I,J),J=1,10)

0120 CONTINUE

C READ BLANK LINES TO IGNORE INPUT FILE COMMENTS
READ(2,0110)
C READ STARTING TEMPERATURES

    DO 0130 I=1,15
    READ(2,0140) (T(I,J),J=1,10)
0130    CONTINUE

0140    FORMAT(10F6.2)
C READ STARTING ARTERIAL AND VENOUS TEMPERATURES IN MAJOR
C AND MINOR VESSELS

    READ(2,0150) TAL
    READ(2,0150) TVL

    READ(2,0150) TAV
    READ(2,0150) TVV

    READ(2,0150) TAM
    READ(2,0150) TVM

    READ(2,0150) TAS
    READ(2,0150) TVS

0150    FORMAT(///,8F6.2,/7F6.2)
C READ COEFFICIENTS RELATING TO METABOLISM

    READ(2,0160) EB
0160    FORMAT(///,8F6.1,/7F6.1)
C READ BLANK LINES TO IGNORE INPUT FILE COMMENTS

    READ(2,0110)
    DO 0170 I=1,15
    READ(2,0180) (RATE(I,J),J=1,10)
0170    CONTINUE

0180    FORMAT(10F6.1)
C READ COEFFICIENTS FOR AFFERENT RECEPTORS

    READ(2,0190) SKINR
    READ(2,0190) SKINS
    READ(2,0190) SKINV
    READ(2,0190) SKINC
    READ(2,0200) FWORK
    READ(2,0200) FCHIL
0190    FORMAT(///,8F8.5,/7F8.5)
0200    FORMAT(///,8F9.6,/7F9.6)
C READ COEFFICIENTS FOR METABOLIC CONTROLLERS

READ(2,0210) CSW
READ(2,0210) SSW
READ(2,0210) PSW
READ(2,0210) CDIL
READ(2,0210) SDIL
READ(2,0210) PDIL
READ(2,0210) CCHIL
READ(2,0210) SCHIL
READ(2,0210) PCHIL
READ(2,0210) CCON
READ(2,0210) SCON
READ(2,0210) PCON
READ(2,0210) FATINT

0210 FORMAT(///,F7.3)

C READ RADIUS OF SMALLEST THERMALLY SIGNIFICANT ARTERIES

READ(2,0220) RAS

0220 FORMAT(///,F4.0)

READ(2,0220) TBF
READ(2,0230) INT
READ(2,0220) EFF

0230 FORMAT(///,F4.1)

READ(2,0220) DENW
READ(2,0240) SHW

0240 FORMAT(///,F6.1)

READ(2,0220) SHA
READ(2,0230) PBFS
READ(2,0250) DENS

0250 FORMAT(///,F8.3)

READ(2,0230) RMRS
READ(2,0260) RLP

0260 FORMAT(///,8F5.2,/7F5.2)

READ(2,0270) CONVW

0270 FORMAT(///,F5.3)

RETURN

END

C END OF SUBROUTINE READ2
**SUBROUTINE PHYSIC**

*****************************************************************************
C CALCULATES GEOMETRICAL PROPERTIES OF THE SIMULATED BODY
*****************************************************************************
INCLUDE 'COMMON.FOR'

C SPECIFIC GRAVITY & PERCENT FAT FROM PIERSO & EAGLE’S C FORMULAE

IF(PBF.EQ.0.0)THEN
  IF(SG.EQ.0.0)THEN
    SG=0.8*(((HT*100.0)**0.242)  
1     /((MBODY*1000.0)**0.1))+0.162
    PBF=100*((5.548/SG)-5.044)
  ELSE
    PBF=100*((5.548/SG)-5.044)
  END IF
ELSE
  SG=5.548/(PBF/100.0+5.044)
END IF

C CONVERT PBF TO A VOLUME PERCENTAGE AND FIND VOLUME % C FAT FOR STANDARD MAN.

PBFV=PBF*1000.0*SG/DENSE(6)
PBFVS=PBF*1000.0*SG/DENSE(6)

C CONVERT TISSUE MASS PROPORTIONS IN EACH SEGMENT TO C VOLUME PROPORTIONS

DO 0010 I=1,15
  FBONE(I)=1000.0*FBONE(I)/DENSE(1)
  FMUSC(I)=1000.0*FMUSC(I)/DENSE(5)
  FFAT(I)=1000.0*FBONE(I)/DENSE(6)
  FSkin(I)=1000.0*FBONE(I)/DENSE(7)
  FVISC(1)=1000.0*FBONE(I)/DENSE(2)
  FVISC(2)=1000.0*FBONE(I)/DENSE(4)
0010
DO 0020 I=3,15
0020 FVISC(I)=1000.0*FVISC(I)/DENSE(4)
C  RESTORE  SUM  OF  FRACTIONS  TO  1.0
DO 0030 I=1,15
   FVSEG(I)=4.0*(FBONE(I)+FVISC(I)+FMUSC(I))
   +FFAT(I)+FSKIN(I)
   FBONE(I)=FBONE(I)/FVSEG(I)
   FVISC(I)=FVISC(I)/FVSEG(I)
   FMUSC(I)=FMUSC(I)/FVSEG(I)
   FFAT(I)=FFAT(I)/FVSEG(I)
0030  FSKIN(I)=FSKIN(I)/FVSEG(I)
C  FIND  PROPORTIONS  OF  FAT  AND  OTHER  TISSUE  VOLUME
C  COMPARED  TO  STANDARD  MAN
DO 0040 I=1,15
   FFAT(I)=FFAT(I)*(1.0+(PBFV-PBFVS)/100.0)
   FNAT(I)=4.0*(FBONE(I)+FVISC(I)+FMUSC(I))+FSKIN(I)
   FBONE(I)=FBONE(I)-FFAT(I)*(PBFV-PBFVS)
   /100.0*FBONE(I)/FNAT(I)
   FVISC(I)=FVISC(I)-FFAT(I)*(PBFV-PBFVS)
   /100.0*FVISC(I)/ FNAT(I)
   FMUSC(I)=FMUSC(I)-FFAT(I)*(PBFV-PBFVS)
   /100.0*FMUSC(I)/ FNAT(I)
0040  FSKIN(I)=FSKIN(I)-FFAT(I)*(PBFV-PBFVS)
       /100.0*FSKIN(I)/FNAT(I)
C  DIVIDE  FAT  INTO  INTERNAL  AND  SUBCUTANEOUS  FRACTIONS
   FATSUB=1.0-FATINT
C THE BODY IS REPRESENTED BY 15 SEGMENTS. THE HEAD IS A
C SPHERE, ALL OTHERS ARE CYLINDERS. SEGMENTS ARE
C NUMBERED BY THE INTEGER (I). THE ORDER IS: HEAD,
C THORAX, ABDOMEN, RIGHT THIGH, CALF & FOOT, LEFT THIGH,
C CALF & FOOT, RIGHT UPPER ARM, FOREARM & HAND,
C LEFT UPPER ARM, FOREARM & HAND.

C CALCULATE SURFACE AREA OF WHOLE BODY USING THE DUBOIS
C FORMULA AND OF EACH SEGMENT ON A FRACTIONAL BASIS.

\[
ABODY = \frac{7.184}{1000.0} \times MBODY^{0.425} \times (HT\times100.0)^{0.725}
\]

DO 0050 I=1,15

0050 ASEG(I) = FA(I) \times ABODY

C DEFINE A SHAPE FACTOR RELATING THE SUBJECT'S
C MASS/HEIGHT**2 TO THAT OF STOLWIJK AND HARDY'S STANDARD
C MAN AND USE TO MODIFY LIMB LENGTH/PERIMETER RATIOS

\[
SHAPE = \frac{MBODY}{(HT^{2.0})/25.15}
\]

C USING SURFACE AREAS, FIND PERIMETER AND LENGTH OF EACH
C SEGMENT, ASSUMING CROSS-SECTION IS CYLINDRICAL (HEAD
C SPHERICAL).

C HEAD (SPHERICAL)

\[
R(1,10) = \left(\frac{ASEG(1)}{4.0/\pi}\right)^{0.5}
\]

\[
VSEG(1) = \frac{4.0}{3.0} \times \pi \times R(1,10)^{3.0}
\]

C REMAINING SEGMENTS (CYLINDRICAL)

DO 0060 I=2,15

PERI(I) = (SHAPE \times ASEG(I) / RLP(I))^{0.5}

LSEG(I) = ASEG(I) / PERI(I)

0060 CONTINUE

C FIND THE RADIUS OF CIRCULAR CROSS-SECTION

DO 0070 I=2,15

0070 R(I,10) = PERI(I) / 2.0/\pi

C SUM-UP SEGMENT AND WHOLE BODY AREAS AND VOLUMES

\[
VBODY = VSEG(1)
\]

DO 0080 I=2,15

VSEG(I) = \pi \times R(I,10)^{2.0} \times LSEG(I)

0080 VBODY = VBODY + VSEG(I)

35
C EACH SEGMENT IS DIVIDED INTO 10 RADIAL SHELLS, NUMBERED
C BY THE INTEGER (J). SHELLS 1 TO 4 FROM THE CENTRE
C OUTWARDS ARE COMPOSED OF CORE TISSUE. SHELLS 5 TO 8
C ARE MUSCLE, 9 IS FAT AND 10 IS SKIN.

C CALCULATE APPARENT VOLUMES OF SHELLS ON A FRACTIONAL
C BASIS

    DO 0110 I=1,15
         FVCORE(I)=FBONE(I)+FVISC(I)+FFAT(I)*FATINT/4.0
         DO 0090 J=1,4
             VSHELL(I,J)=FVCORE(I)*VSEG(I)
         DO 0100 J=5,8
             VSHELL(I,J)=FMUSC(I)*VSEG(I)
         VSHELL(I,9)=FFAT(I)*FATSUB*VSEG(I)
         VSHELL(I,10)=FSKIN(I)*VSEG(I)
    0110 CONTINUE

C CALCULATE MEAN DENSITY OF EACH SHELL. BLOOD DENSITY IS
C THE SAME AS THAT OF THE TISSUE IN WHICH MAJOR VESSELS
C ARE SITUATED.

    DO 0140 I=1,15
         DO 0120 J=1,4
             DEN(I,J)=(FBONE(I)*DENSE(1)+FVISC(I)*DENSE(4)
                +FFAT(I)/4.0*FATINT*DENSE(6))/FVCORE(I)
         DO 0130 J=5,8
             DEN(I,J)=DENSE(5)
         DEN(I,9)=DENSE(6)
         DEN(I,10)=DENSE(7)
    0140 CONTINUE

C CORRECT HEAD AND THORAX CORES

    DO 0150 J=1,4
        DEN(I,J)=(FBONE(I)*DENSE(1)+FVISC(I)*DENSE(2)
        +FFAT(I)/4.0*FATINT*DENSE(6))/FVCORE(I)
DEN(2,J) = (FBONE(2) * DENSE(1) + FVISC(2) * (0.6 * DENSE(3) + 0.4 * DENSE(4)) + FFAT(2)/4.0 * FATINT * DENSE(6)) / FVCORE(2)

C ESTIMATE MASS OF EACH SHELL, USING APPARENT SHELL VOLUMES. FIND APPARENT BODY MASS.

MTOTAL = 0.0
DO 0160 I = 1, 15
   DO 0160 J = 1, 10
      MSHELL(I,J) = VSHELL(I,J) * DEN(I,J)
      MTOTAL = MTOTAL + MSHELL(I,J)
   0160 CONTINUE

C CORRECT SHELL MASSES USING KNOWN BODY MASS.

DO 0170 I = 1, 15
   DO 0170 J = 1, 10
      MSHELL(I,J) = MSHELL(I,J) * MBODY / MTOTAL
   0170 CONTINUE

C FIND NEW TISSUE SHELL VOLUMES, USING NEW SHELL MASSES.

DO 0180 I = 1, 15
   DO 0180 J = 1, 10
      VSHELL(I,J) = MSHELL(I,J) / DEN(I,J)
   0180 CONTINUE

C SUM-UP SHELL VOLUMES FOR INTERMEDIATE ESTIMATE OF BODY VOLUME.

VBODY = 0.0
DO 0190 I = 1, 15
   DO 0190 J = 1, 10
      VBODY = VBODY + VSHELL(I,J)
   0190 CONTINUE
C APPLY FINAL CORRECTION TO TISSUE SHELL VOLUMES, USING
C KNOWN BODY MASS & S.G.

DO 0210 I=1,15
VSEG(I)=0.0
   DO 0200 J=1,10
      VSHELL(I,J)=VSHELL(I,J)*1000.0*SG*VBODY/MBODY
0200      VSEG(I)=VSEG(I)+VSHELL(I,J)
0210 CONTINUE

C FIND FINAL ESTIMATE FOR WHOLE BODY MASS

MBODY=0.0
   DO 0220 I=1,15
      DO 0220 J=1,10
         MBODY=MBODY+MSHELL(I,J)
0220 CONTINUE

C FIND OUTER RADIUS OF EACH SHELL, WORKING OUTWARDS FROM
C THE CENTRE

VSUM(1)=0.0
   DO 0230 J=1,10
      VSUM(1)=VSUM(1)+VSHELL(1,J)
      R(1,J)=(3.0*VSUM(1)/4.0/PI)**0.333
0230 CONTINUE

ASEG(1)=4.0*PI*R(1,10)**2.0
ABODY=ASEG(1)
VSEG(1)=VSUM(1)
VBODY=VSEG(1)
   DO 0250 I=2,15
      VSUM(I)=0.0
         DO 0240 J=1,10
            VSUM(I)=VSUM(I)+VSHELL(I,J)
            R(I,J)=(VSUM(I)/PI/LSEG(I))**0.5
0240 CONTINUE

38
ASEG(I) = 2.0 * PI * R(I,10) * LSEG(I)
ABODY = ABODY + ASEG(I)
VSEG(I) = VSUM(I)
VBODY = VBODY + VSEG(I)

0250 CONTINUE

C CALCULATE RADIUS TO CENTRE OF MASS OF EACH SHELL
C HEAD
RCM(1,1) = R(1,1) / (2.0**0.333)
DO 0260 J = 2, 10
0260 RCM(1,J) = ((R(1,J-1)**3.0 + R(1,J)**3.0)**0.333)
               1 / (2.0**0.333)
C REMAINING SEGMENTS
DO 0270 I = 2, 15
RCM(I,1) = R(I,1) / (2.0**0.5)
DO 0270 J = 2, 10
RCM(I,J) = ((R(I,J-1)**2.0 + R(I,J)**2.0)**0.5)
               1 / (2.0**0.5)
0270 CONTINUE

C CALCULATE THERMAL CONDUCTION PATH LENGTHS BETWEEN
C SHELLS
DO 0290 I = 1, 15
   DO 0280 J = 1, 9
0280 X(I,J) = RCM(I,J+1) - RCM(I,J)
X(I,10) = R(I,10) - RCM(I,10)
0290 CONTINUE

C CALCULATE RADIUS TO MID-PLANE FOR EACH SHELL
DO 0300 I = 1, 15
   DO 0300 J = 1, 10
      RMP(I,J) = RCM(I,J) + X(I,J) / 2.0
0300 CONTINUE
C CALCULATE AREA AT MID-PLANE OF EACH SHELL FOR THERMAL
C CONDUCTION

DO 0310 J=1,10
   I=1
   AMP(I,J)=4.0*PI*RMP(I,J)**2.0
   DO 0310 I=2,15
      AMP(I,J)=2.0*PI*RMP(I,J)*LSEG(I)
0310 CONTINUE

C CALCULATE OUTER DIAMETERS OF SEGMENTS

DO 0320 I=1,15
   DSEG(I)=2.0*R(I,10)
0320 RETURN

END

C END OF SUBROUTINE PHYSIC

***************************************************************************
*** SUBROUTINE BLOOD
***************************************************************************
C DETERMINES PHYSICAL PROPERTIES OF BLOOD VESSELS

INCLUDE 'COMMON.FOR'

C CALCULATE TOTAL BLOOD VOLUME AS A FUNCTION OF BODY MASS
C ACCORDING TO GUYTON (A MALE SUBJECT IS ASSUMED)

IF(MBODY.LE.45.0) THEN
   VBLOOD=0.079*MBODY
ELSE
   IF(MBODY.LE.50.0) THEN
      VBLOOD=3.7+0.12*(MBODY-45)
   ELSE
      IF(MBODY.LE.60.0) THEN
         VBLOOD=4.2+0.09*(MBODY-50)
      ELSE
         IF(MBODY.LE.70.0) THEN
            VBLOOD=4.7+0.07*(MBODY-60)
         ELSE
            IF(MBODY.LE.80.0) THEN
               VBLOOD=5.2+0.05*(MBODY-70)
            ELSE
               IF(MBODY.LE.90.0) THEN
                  VBLOOD=5.7+0.03*(MBODY-80)
               ELSE
                  VBLOOD=6.2+0.01*(MBODY-90)
               END IF
            END IF
         END IF
      END IF
   END IF
END IF
IF (MBODY .LE. 70.0) THEN
  VBLOOD = 5.06 + 0.05 * (MBODY - 60)
ELSE
  IF (MBODY .LE. 80.0) THEN
    VBLOOD = 5.56 + 0.035 * (MBODY - 70)
  ELSE
    VBLOOD = 5.91 + 0.025 * (MBODY - 80)
  END IF
END IF
END IF
END IF
END IF

C VOLUME OF BLOOD IN MAJOR ARTERIES AND VEINS IN EACH
C SEGMENT IS CALCULATED AS A FRACTION OF THE TOTAL VOLUME
C IN LITRES

DO 0010 I = 1, 15
  VBAL(I) = FBA(1) * VBLOOD * VSEG(I) / VBODY
0010  VBVL(I) = FBV(1) * VBLOOD * VSEG(I) / VBODY

C FOR THORAX, PULMONARY AND HEART BLOOD IS ADDED TO MAJOR
C VESSELS

  VBAL(2) = VBAL(2) + (FBA(3) + FBA(4)) * VBLOOD
  VBVL(2) = VBVL(2) + (FBV(3) + FBV(4)) * VBLOOD

C FOR ABDOMEN, LIVER AND SPLEEN BLOOD IS ADDED TO MAJOR
C VESSELS

  VBAL(3) = VBAL(3) + (FBA(5) + FBA(6)) * VBLOOD
  VBVL(3) = VBVL(3) + (FBV(5) + FBV(6)) * VBLOOD

41
C VOLUME OF BLOOD IN MINOR ARTERIES AND VEINS IN SKIN AND
C MUSCLE IS CALCULATED IN PROPORTION TO TISSUE VOLUME

DO 0020 I=1,15
VVISC(I)=4.0*FVISC(I)*VSEG(I)
VMUSC(I)=4.0*FMUSC(I)*VSEG(I)

DO 0030 I=1,15
VBAV(I)=FBA(2)*VBLOOD*VSEG(I)/VBODY
  *VVISC(I)/(VSHELL(I,10)+VMUSC(I)+VVISC(I))
VBVV(I)=FBV(2)*VBLOOD*VSEG(I)/VBODY
  *VVISC(I)/(VSHELL(I,10)+VMUSC(I)+VVISC(I))
VBAM(I)=FBA(2)*VBLOOD*VSEG(I)/VBODY
  *VMUSC(I)/(VSHELL(I,10)+VMUSC(I)+VVISC(I))
VBVM(I)=FBV(2)*VBLOOD*VSEG(I)/VBODY
  *VMUSC(I)/(VSHELL(I,10)+VMUSC(I)+VVISC(I))
VBAS(I)=FBA(2)*VBLOOD*VSEG(I)/VBODY*VSHELL(I,10)
  /(VSHELL(I,10)+VMUSC(I)+VVISC(I))
VBVS(I)=FBV(2)*VBLOOD*VSEG(I)/VBODY*VSHELL(I,10)
  /(VSHELL(I,10)+VMUSC(I)+VVISC(I))

0030 CONTINUE

C FIND DIMENSIONS OF EACH MAJOR VEIN AND ARTERY, ASSUMING
C 2 OF EACH PER LIMB SEGMENT AND 4 PER SEGMENT IN THORAX
C & ABDOMEN, HANDS & FEET.
RAL(1)=(VBAL(1)/1000.0/2.0/PI/2.0/R(1,10))**0.5
RVL(1)=(VBVL(1)/1000.0/2.0/PI/2.0/R(1,10))**0.5

DO 0040 I=2,15
RAL(I)=(VBAL(I)/1000.0/PI/2.0/LSEG(I))**0.5
RVL(I)=(VBVL(I)/1000.0/PI/2.0/LSEG(I))**0.5

0040 CONTINUE

C FIND TOTAL LENGTH OF MINOR VEINS AND ARTERIES IN EACH
C SKIN AND MUSCLE SHELL, GIVEN MINOR ARTERY RADIUS IN
C MICRO-METRES
RAS=RAS*1E-6
RVS=RAS*(FBV(2)/FBA(2))**0.5

DO 0080 I=1,15
LBVVV(I)=VBAV(I)/1000.0/(PI*RAS**2.0)
LBVM(I)=VBAM(I)/1000.0/(PI*RAS**2.0)
LBVS(I)=VBAS(I)/1000.0/(PI*RAS**2.0)

0080 CONTINUE

C CALCULATE HEAT EXCHANGE COEFFICIENTS BETWEEN WALLS OF
C BLOOD VESSELS AND SURROUNDING TISSUE FOR LARGE VESSELS.

        HAL(1)=4.0*PI*CONDB(2)*2.0*R(1,10)
        1/LOG(R(1,4)/RAL(1))
        HVL(1)=4.0*PI*CONDB(2)*2.0*R(1,10)
        1/LOG(R(1,4)/RVL(1))
        HCCL(1)=2.0*PI*CONDB(4)*2.0*R(1,10)/LOG(3.0)

        DO 0100 I=2,15
           HAL(I)=4.0*PI*CONDB(6)*LSEG(I)/LOG(R(I,4)/RAL(I))
           HVL(I)=4.0*PI*CONDB(6)*LSEG(I)/LOG(R(I,4)/RVL(I))
           HCCL(I)=2.0*PI*CONDB(4)*LSEG(I)/LOG(3.0)
        0100 CONTINUE

C CALCULATE HEAT EXCHANGE COEFFICIENTS BETWEEN SMALL
BLOOD VESSELS IN VISCERA,
C MUSCLE AND SKIN AND SURROUNDING TISSUE.

        DO 0120 I=1,15
           XBVV(I)=(AMP(I,2)*R(I,4)/LBVV(I))**0.5
           XBVM(I)=(AMP(I,6)*(R(I,8)-R(I,5))/LBVM(I))**0.5
           XBVS(I)=(AMP(I,10)*(R(I,10)-R(I,9))/LBVS(I))**0.5
           HAV(I)=2.0*PI*CONDB(4)*LBVV(I)
           1/LOG(XBVV(I)/2.0/RAS)
           HVV(I)=2.0*PI*CONDB(4)*LBVV(I)
           1/LOG(XBVV(I)/2.0/RVS)
           HCCV(I)=PI*CONDB(4)*LBVV(I)/LOG(3.0)
           HAM(I)=2.0*PI*CONDB(5)*LBVM(I)
           1/LOG(XBVM(I)/2.0/RAS)
           HVM(I)=2.0*PI*CONDB(5)*LBVM(I)
           1/LOG(XBVM(I)/2.0/RVS)
           HCCM(I)=PI*CONDB(5)*LBVM(I)/LOG(3.0)
           HAS(I)=2.0*PI*CONDB(7)*LBVS(I)
           1/LOG(XBVS(I)/2.0/RAS)
HVS(I) = 2.0 * PI * CONDB(7) * LBVS(I) / LOG(XBVS(I) / 2.0 / RVS)

HCCS(I) = PI * CONDB(7) * LBVS(I) / LOG(3.0)

0120 CONTINUE

RETURN

END

C END OF SUBROUTINE BLOOD

******************************************************************************

SUBROUTINE THERMO

******************************************************************************

C CALCULATES THERMAL PROPERTIES OF THE SIMULATED BODY

******************************************************************************

INCLUDE 'COMMON.FOR'

C CALCULATE SPECIFIC HEAT OF CORE SEGMENTS
C THORAX VISCERA CONSIST OF 60% LUNG AND 40%
C OTHER VISCERAL TISSUE).

SHCORE(1) = (SPHEAT(1) * FBONE(1) + SPHEAT(2) * FVISC(1)
1 + SPHEAT(6) * FFAT(1) * FATINT/4.0)
2 / FVCORE(1)

SHCORE(2) = (SPHEAT(1) * FBONE(2) + (0.6 * SPHEAT(3)
1 + 0.4 * SPHEAT(4)) * FVISC(2)
2 + SPHEAT(6) * FFAT(2) * FATINT/4.0)
3 / FVCORE(2)

DO 0010 I = 3,15

SHCORE(I) = (SPHEAT(1) * FBONE(I)
1 + SPHEAT(4) * FVISC(I)
2 + SPHEAT(6) * FFAT(I) * FATINT/4.0)
3 / FVCORE(I)

0010 CONTINUE

C CALCULATE THERMAL CAPACITANCE OF EACH SHELL

DO 0040 I = 1,15

DO 0020 J = 1,4

0020 C(I,J) = SHCORE(I) * MSHELL(I,J)

DO 0030 J = 5,8

0030 C(I,J) = SPHEAT(5) * MSHELL(I,J)
C(I,9)=SPHEAT(6)*MSHELL(I,9)
C(I,10)=SPHEAT(7)*MSHELL(I,10)

0040 CONTINUE
C FIND BLOOD THERMAL CAPACITANCES FROM VOLUMES.
DO 0050 I=1,15
CBA(I)=SPHEAT(8)*VBAL(I)*DENSE(8)/1000.0
CBV(I)=SPHEAT(8)*VBVL(I)*DENSE(8)/1000.0
CBAV(I)=SPHEAT(8)*VBAV(I)*DENSE(8)/1000.0
CBVV(I)=SPHEAT(8)*VBVV(I)*DENSE(8)/1000.0
CBAM(I)=SPHEAT(8)*VBAM(I)*DENSE(8)/1000.0
CBVM(I)=SPHEAT(8)*VBVM(I)*DENSE(8)/1000.0
CBAS(I)=SPHEAT(8)*VBAS(I)*DENSE(8)/1000.0
CBVS(I)=SPHEAT(8)*VBVS(I)*DENSE(8)/1000.0
0050 CONTINUE
C SUBTRACT MAJOR BLOOD VESSEL CAPACITANCES FROM CORE
C SHELLS IN HEAD, THORAX, ABDOMEN, HANDS & FEET AND FROM
C MUSCLE SHELLS IN REMAINDER. ALSO MINOR VESSELS FROM
C CORE, MUSCLE AND SKIN IN ALL SEGMENTS.
DO 0080 I=1,15
   DO 0060 J=1,4
0060 C(I,J)=C(I,J)-CBA(I)/4.0-CBV(I)/4.0-CBAV(I)/4.0-CBVV(I)/4.0
   DO 0070 J=5,8
0070 C(I,J)=C(I,J)-CBAM(I)/4.0-CBVM(I)/4.0
C(I,10)=C(I,10)-CBAS(I)-CBVS(I)
0080 CONTINUE
C SUM THERMAL CAPACITANCES OF SKIN AND WHOLE BODY
CSKIN=0.0
DO 0090 I=1,15
0090 CSKIN=CSKIN+C(I,10)
CBODY=0.0

45
DO 0100 I=1,15
DO 0100 J=1,10

0100   CBODY=CBODY+C(I,J)

C EVALUATE THERMAL CONDUCTANCE OUTWARDS FROM EACH SHELL
C HAVING NO CHANGE IN CONDUCTANCE AT THE BOUNDARY

C HEAD

   I=1
   DO 0110 J=1,3

0110   TC(I,J)=(FBONE(I)*CONDB(1)+FVISC(I)*CONDB(2)
   1 +FFAT(I)/4.0*CONDB(6)*FATINT)
   2 /FVCORE(I)
   3 *AMP(I,J)/X(I,J)
   DO 0120 J=5,7

TCMUSC(I,J-4)=CONDB(5)*AMP(I,J)/X(I,J)

0120   TC(I,J)=TCMUSC(I,J-4)

C THORAX

   I=2
   DO 0130 J=1,3

0130   TC(I,J)=(FBONE(I)*CONDB(1)
   1 +0.6*CONDB(3)+0.4*CONDB(4))
   2 *FVISC(I)+FFAT(I)/4.0*CONDB(6)*FATINT)
   3 /FVCORE(I)
   4 *AMP(I,J)/X(I,J)
   DO 0140 J=5,7

TCMUSC(I,J-4)=CONDB(5)*AMP(I,J)/X(I,J)

0140   TC(I,J)=TCMUSC(I,J-4)

C ABDOMEN

   I=3
   DO 0150 J=1,3

0150   TC(I,J)=(FBONE(I)*CONDB(1)+FVISC(I)*CONDB(4)
   1 +FFAT(I)/4.0*CONDB(6)*FATINT)
   2 /FVCORE(I)
   3 *AMP(I,J)/X(I,J)
DO 0160 J=5,7
TCMUSC(I,J-4)=COND(5)*AMP(I,J)/X(I,J)
0160 TC(I,J)=TCMUSC(I,J-4)

C ALL LIMB SEGMENTS
DO 0180 I=4,15
DO 0170 J=1,3

0170 TC(I,J)=(FBONE(I)*COND(1)+FVISC(I)*COND(4)
1 +FFAT(I)/4.0*COND(6)*FATINT)
2 /FVCORE(I)
3 *AMP(I,J)/X(I,J)

DO 0180 J=5,7
TCMUSC(I,J-4)=COND(5)*AMP(I,J)/X(I,J)
TC(I,J)=TCMUSC(I,J-4)
0180 CONTINUE

C THERMAL CONDUCTANCES AT BOUNDARIES
C WHERE TISSUE CONDUCTIVITIES CHANGE

C HEAD, CORE/MUSCLE BOUNDARY

RMPA=RCM(1,4)+(R(1,4)-RCM(1,4))/2.0
AA=4.0*PI*RMPA**2.0
XA=R(1,4)-RCM(1,4)
YA=(FBONE(1)*COND(1)+FVISC(1)*COND(2)
1 +FFAT(1)/4.0*COND(6)*FATINT)
2 /FVCORE(I)*AA/XA
RMPB=R(1,4)+(RCM(1,5)-R(1,4))/2.0
AB=4.0*PI*RMPB**2.0
XB=RCM(1,5)-R(1,4)
YB=COND(5)*AB/XB

TC(1,4)=YA*YB/(YA+YB)

C HEAD, MUSCLE/FAT BOUNDARY

RMPA=RCM(1,8)+(R(1,8)-RCM(1,8))/2.0
AA=4.0*PI*RMPA**2.0
XA=R(1,8)-RCM(1,8)
YA=COND(5)*AA/XA
RMPB=R(1,8)+(RCM(1,9)-R(1,8))/2.0
AB=4.0*PI*RMPB**2.0
XB=RCM(1,9)-R(1,8)
YB=COND(6)*AB/XB

TCMUSC(1,4)=YA*YB/(YA+YB)
TC(1,8)=TCMUSC(1,4)
C HEAD, FAT/SKIN BOUNDARY

\[
\begin{align*}
\text{RMPA} &= \text{RCM}(1, 9) + \frac{(\text{R}(1, 9) - \text{RCM}(1, 9))}{2.0} \\
\text{AA} &= 4.0 \times \pi \times \text{RMPA}^2 \\
\text{XA} &= \text{R}(1, 9) - \text{RCM}(1, 9) \\
\text{YA} &= \text{CONDB}(6) \times \frac{\text{AA}}{\text{XA}} \\
\text{RMPB} &= \text{R}(1, 9) + \frac{\text{RCM}(1, 10) - \text{R}(1, 9)}{2.0} \\
\text{AB} &= 4.0 \times \pi \times \text{RMPB}^2 \\
\text{XB} &= \text{RCM}(1, 10) - \text{R}(1, 9) \\
\text{YB} &= \text{CONDB}(7) \times \frac{\text{AB}}{\text{XB}} \\
\text{TCFAT}(1) &= \frac{\text{YA} \times \text{YB}}{\text{YA} + \text{YB}} \\
\text{TC}(1, 9) &= \text{TCFAT}(1)
\end{align*}
\]

C THORAX, CORE/MUSCLE BOUNDARY

\[
\begin{align*}
\text{RMPA} &= \text{RCM}(2, 4) + \frac{(\text{R}(2, 4) - \text{RCM}(2, 4))}{2.0} \\
\text{AA} &= 2.0 \times \pi \times \text{RMPA} \times \text{LSEG}(2) \\
\text{XA} &= \text{R}(2, 4) - \text{RCM}(2, 4) \\
\text{YA} &= (\text{FBONE}(2) + \text{CONDB}(1) + \text{FVISC}(2) \\
&\quad \times \frac{0.6 \times \text{CONDB}(3) + 0.4 \times \text{CONDB}(4)}{2} + \text{FFAT}(2) \times \frac{0.4 \times \text{CONDB}(6) \times \text{FATINT}}{\text{FVCORE}(2)} \times \frac{\text{AA}}{\text{XA}} \\
\text{RMPB} &= \text{R}(2, 4) + \frac{\text{RCM}(2, 5) - \text{R}(2, 4)}{2.0} \\
\text{AB} &= 2.0 \times \pi \times \text{RMPB} \times \text{LSEG}(2) \\
\text{XB} &= \text{RCM}(2, 5) - \text{R}(2, 4) \\
\text{YB} &= \text{CONDB}(5) \times \frac{\text{AB}}{\text{XB}} \\
\text{TC}(2, 4) &= \frac{\text{YA} \times \text{YB}}{\text{YA} + \text{YB}}
\end{align*}
\]

C ABDOMEN CORE/MUSCLE BOUNDARY

\[
\begin{align*}
\text{RMPA} &= \text{RCM}(3, 4) + \frac{(\text{R}(3, 4) - \text{RCM}(3, 4))}{2.0} \\
\text{AA} &= 2.0 \times \pi \times \text{RMPA} \times \text{LSEG}(3) \\
\text{XA} &= \text{R}(3, 4) - \text{RCM}(3, 4) \\
\text{YA} &= (\text{FBONE}(3) + \text{CONDB}(1) + \text{FVISC}(3) \times \text{CONDB}(4) \\
&\quad + \text{FFAT}(3) \times \frac{0.4 \times \text{CONDB}(6) \times \text{FATINT}}{\text{FVCORE}(3)} \times \frac{\text{AA}}{\text{XA}} \\
\text{RMPB} &= \text{R}(3, 4) + \frac{\text{RCM}(3, 5) - \text{R}(3, 4)}{2.0} \\
\text{AB} &= 2.0 \times \pi \times \text{RMPB} \times \text{LSEG}(3) \\
\text{XB} &= \text{RCM}(3, 5) - \text{R}(3, 4) \\
\text{YB} &= \text{CONDB}(5) \times \frac{\text{AB}}{\text{XB}} \\
\text{TC}(3, 4) &= \frac{\text{YA} \times \text{YB}}{\text{YA} + \text{YB}}
\end{align*}
\]

C LIMB SEGMENTS CORE/MUSCLE BOUNDARY

\[
\begin{align*}
\text{DO 0190 I}=4,15 \\
\text{RMPA} &= \text{RCM}(I, 4) + \frac{(\text{R}(I, 4) - \text{RCM}(I, 4))}{2.0} \\
\text{AA} &= 2.0 \times \pi \times \text{RMPA} \times \text{LSEG}(I) \\
\text{XA} &= \text{R}(I, 4) - \text{RCM}(I, 4)
\end{align*}
\]
YA = (FBONE(I) * CONDB(1) + FVISC(I) * CONDB(4) + FFAT(I) / 4.0 * CONDB(6) * FATINT) / FVCORE(I) * AA / XA

RMPB = R(I, 4) + (RCM(I, 5) - R(I, 4)) / 2.0
AB = 2.0 * PI * RMPB * LSEG(I)
XB = RCM(I, 5) - R(I, 4)
YB = CONDB(5) * AB / XB

TC(I, 4) = YA * YB / (YA + YB)

C ALL SEGMENTS EXCEPT HEAD, MUSCLE/FAT BOUNDARY

DO 0200 I = 2, 15
RMPA = RCM(I, 8) + (R(I, 8) - RCM(I, 8)) / 2.0
AA = 2.0 * PI * RMPA * LSEG(I)
XA = R(I, 8) - RCM(I, 8)
YA = CONDB(5) * AA / XA

RMPB = R(I, 8) + (RCM(I, 9) - R(I, 8)) / 2.0
AB = 2.0 * PI * RMPB * LSEG(I)
XB = RCM(I, 9) - R(I, 8)
YB = CONDB(6) * AB / XB

TCMUSC(I, 4) = YA * YB / (YA + YB)
TC(I, 8) = TCMUSC(I, 4)

0200 CONTINUE

C ALL SEGMENTS EXCEPT HEAD, FAT/SKIN BOUNDARY

DO 0210 I = 2, 15
RMPA = RCM(I, 9) + (R(I, 9) - RCM(I, 9)) / 2.0
AA = 2.0 * PI * RMPA * LSEG(I)
XA = R(I, 9) - RCM(I, 9)
YA = CONDB(6) * AA / XA

RMPB = R(I, 9) + (RCM(I, 10) - R(I, 9)) / 2.0
AB = 2.0 * PI * RMPB * LSEG(I)
XB = RCM(I, 10) - R(I, 9)
YB = CONDB(7) * AB / XB

TCFAT(I) = YA * YB / (YA + YB)
TC(I, 9) = TCFAT(I)

0210 CONTINUE

RETURN

END

C END OF SUBROUTINE THERMO
SUBROUTINE COVER

C CALCULATES PHYSICAL PARAMETERS OF CLOTHING

INCLUDE 'COMMON.FOR'

C CALCULATE THICKNESSES, RADII AND SURFACE AREAS AT CLOTHING SURFACE ASSUMING CLOTHING IS REPRESENTED BY A THICKNESS OF AIR GIVING THE SAME INSULATION. AIR CONDUCTIVITY OBTAINED FROM N.A.S.A. EMPIRICAL FORMULA.

\[
CONDA = \frac{0.6325 \times 10^{-6} \times (TW(1) + 273.0)^{1.5}}{((TW(1) + 273.0)^2 + 245.4 \times 10^{12} / (TW(1) + 273.0)^{12.0})}
\]

DO 0010 I=1,15
CONDC(I)=TOG(I)/10.0
0010 THCL0(I)=CONDC(I)\times CONDA
ATCLO=0.0
RCLO(1)=R(1,10)+THCLO(1)
DCLO(1)=2.0*RCLO(1)
ACLO(1)=4.0*PI*(RCLO(1)**2.0)
ATCLO=ATCLO+ACLO(1)
DO 0020 I=2,15
RCLO(I)=R(I,10)+THCLO(I)
DCLO(I)=2.0*RCLO(I)
ACLO(I)=2.0*PI*RCLO(I)*LSEG(I)
ATCLO=ATCLO+ACLO(I)
0020 CONTINUE

C SKIN THERMAL CONDUCTION PATH LENGTHS ARE NOW INCREASED BY CLOTHING

DO 0030 I=1,15
0030 X(I,10)=X(I,10)+THCLO(I)
RETURN
END

C END OF SUBROUTINE COVER
SUBROUTINE SETUP

C SETS INITIAL CONDITIONS READY FOR SIMULATION

INCLUDE 'COMMON.FOR'

C INITIALISE VARIABLES

TSOLD=T(2,10)
TSNEW=T(2,10)
TCOLD=T(3,1)
TCNEW=T(3,1)
TVISC(1)=(T(1,1)+T(1,2)+T(1,3)+T(1,4))/4.0
TVISC(2)=(T(2,1)+T(2,2)+T(2,3)+T(2,4))/4.0

C FIND RESTING METABOLIC RATE FOR EACH SHELL, AND TOTAL C MR IN WATTS. MODIFY IF RMR IS KNOWN. RMR IS IN WM-2.

MR=0.0
DO 0010 J=1,4
   QB(1,J)=VSEG(1)*(FBONE(1)*FQB(1)+FVISC(1)*FQB(2)
   +FFAT(1)*FATINT/4.0*FQB(6))
0010   MR=MR+QB(1,J)
DO 0020 J=1,4
   QB(2,J)=VSEG(2)*(FBONE(2)*FQB(1)
   +FVISC(2)*(0.6*FQB(3)
   +0.4*FQB(4))+FFAT(2)*FATINT/4.0*FQB(6))
0020   MR=MR+QB(2,J)
DO 0030 I=3,15
   DO 0030 J=1,4
      QB(I,J)=VSEG(I)*(FBONE(I)*FQB(1)
      +FVISC(I)*FQB(4)
      +FFAT(I)*FATINT/4.0*FQB(6))
0030   MR=MR+QB(I,J)
DO 0040 I=1,15
   QB(I,9)=VSHELL(I,9)*FQB(6)
   MR=MR+QB(I,9)
DO 0040 J=5,8
QB(I,J)=VSHELL(I,J)*FQB(5)

0040 MR=MR+QB(I,J)

DO 0050 I=1,15
QB(I,10)=VSHELL(I,10)*FQBS(I)

0050 MR=MR+QB(I,10)

IF(RMR .NE. 0.0) THEN

DO 0060 I=1,15

DO 0060 J=1,10
QB(I,J)=QB(I,J)*RMR*ABODY/MR

0060 CONTINUE

MR=0.0

DO 0070 I=1,15

DO 0070 J=1,10
MR=MR+QB(I,J)

0070 CONTINUE
ELSE

RMR=MR/ABODY
END IF

C CALCULATE RESTING BLOODFLOW FOR EACH SHELL IN MLS-1 AS
C A FUNCTION OF METABOLIC RATE (FUNCTION OF VOLUME FOR
C SKIN).

DO 0080 J=1,4

BFB(1,J)=QB(1,J)*(FBONE(1)*FBFB(1)
1 +FVISC(1)*FBFB(2)
2 +FFAT(1)*FATINT/4.0*FBFB(6))/FVCORE(1)

DO 0090 J=1,4

BFB(2,J)=QB(2,J)*(FBONE(2)*FBFB(1)
1 +FVISC(2)*(0.6*FBFB(3)+0.4*FBFB(4))
2 +FFAT(2)*FATINT/4.0*FBFB(6))/FVCORE(2)

52
DO 0100 I=3,15
    DO 0100 J=1,4
0100   BFB(I,J)=QB(I,J)*(FBONE(I)*FBFB(1) +FVISC(I)*FBFB(4) +FFAT(I)*FATINT/4.0*FBFB(6))/FVCORE(I)
    DO 0110 1=1,15
0110   BFB(I,9)=QB(I,9)*FBFB(6)
    DO 0018 J=5,8
0110   BFB(I,J)=QB(I,J)*FBFB(5)
    DO 0120 1=1,15
0120   BFB(I,10)=VSHELL(I,10)*FBFBS(I)
C COMPENSATE FOR DIFFERENCE IN ACTUAL STARTING
C TEMPERATURE, UNLESS TR IS READ AS 0.0. GRADIENTS ARE
C USED IN OUTER REGIONS.
    IF (TR.NE.0.0) THEN
        DO 0130 I=1,15
0130   TAL(I)=TAL(I)*TR/TSET(3,1)
        TVL(I)=TVL(I)*TR/TSET(3,1)
        TAV(I)=TAV(I)*TR/TSET(3,1)
        TVV(I)=TVV(I)*TR/TSET(3,1)
        TAM(I)=TAM(I)*(1.0+((TR-TSET(3,1))/TSET(3,1))
1           *(1.0-(RCM(I,6)-RCM(I,1)))
2           /(R(I,10)-RCM(I,1))))
        TVM(I)=TVM(I)*(1.0+((TR-TSET(3,1))/TSET(3,1))
1           *(1.0-(RCM(I,6)-RCM(I,1)))
2           /(R(I,10)-RCM(I,1))))
        TAS(I)=TAS(I)*(1.0+((TR-TSET(3,1))/TSET(3,1))
1           *(1.0-(RCM(I,10)-RCM(I,1)))
2           /(R(I,10)-RCM(I,1))))
        TVS(I)=TVS(I)*(1.0+((TR-TSET(3,1))/TSET(3,1))
1           *(1.0-(RCM(I,10)-RCM(I,1)))
2           /(R(I,10)-RCM(I,1))))
        CONTINUE
      END
DO 0140 I=1,15
  DO 0140 J=1,10
    T(I,J)=T(I,J)*(1.0+(TR-TSET(3,1))/TSET(3,1))
    * (1.0-(RCM(I,J)-RCM(I,1))
    / (R(I,10)-RCM(I,1)))
  CONTINUE
ELSE
  IF(RMR .NE. 0.0) THEN
    DO 0160 I=1,15
      TAL(I)=TAL(I)*RMR/RMRS
      TVL(I)=TVL(I)*RMR/RMRS
      TAV(I)=TAV(I)*RMR/RMRS
      TVV(I)=TVV(I)*RMR/RMRS
      TAM(I)=TAM(I)*RMR/RMRS
      TVM(I)=TVM(I)*RMR/RMRS
      TAS(I)=TAS(I)*RMR/RMRS
      TVS(I)=TVS(I)*RMR/RMRS
    CONTINUE
    DO 0150 J=1,10
      T(I,J)=T(I,J)*RMR/RMRS
  CONTINUE
  ELSE
    DO 0180 I=1,15
      TAL(I)=TAL(I)*MR/ABODY/RMRS
      TVL(I)=TVL(I)*MR/ABODY/RMRS
      TAV(I)=TAV(I)*MR/ABODY/RMRS
      TVV(I)=TVV(I)*MR/ABODY/RMRS
      TAM(I)=TAM(I)*MR/ABODY/RMRS
      TVM(I)=TVM(I)*MR/ABODY/RMRS
      TAS(I)=TAS(I)*MR/ABODY/RMRS
      TVS(I)=TVS(I)*MR/ABODY/RMRS
    END
DO 0170 J=1,10

0170 T(I,J)=T(I,J)*MR/ABODY/RMRS

0180 CONTINUE

END IF

END IF

C CALCULATE SURFACE & MEAN SKIN TEMPERATURE, FOR USE IN
C FIRST PASS ONLY. TS IS MEAN SKIN OR CLOTHING SURFACE
C TEMPERATURE, TSNEW IS MEAN SKIN TISSUE TEMPERATURE.

TSNEW=0.0

ASKIN=ABODY-ASEG(I)

DO 0190 I=2,15

0190 TSNEW=TSNEW+T(I,10)*ASEG(I)/ASKIN

C CALCULATE INITIAL TEMPERATURES OF SKIN SURFACE &
C CLOTHING, IF PRESENT, BASED ON INTERNAL GRADIENTS.

TS=0.0

DO 0200 I=1,15

TRATE8(I)=(T(I,9)-T(I,8))/X(I,8)
TRATE9(I)=(T(I,10)-T(I,9))/X(I,9)
TRTDOT(I)=(TRATE9(I)-TRATE8(I))/(R(I,9)-R(I,8))
TRTE10(I)=TRATE9(I)+TRTDOT(I)*(R(I,10)-R(I,9))
TSURF(I)=T(I,10)+TRTE10(I)*X(I,10)

0200 CONTINUE

IF(OUTFIT.EQ.'YES') THEN

TS=0.0

CONDA=(0.6325E-6*(TAIR(L)+273.0)**1.5)
CONDB=1/(TAIR(L)+273.0)**2+245.4*10**(-12.0/(TAIR(L)+273.0)))

DO 0210 I=1,15

TCLRTE(I)=TRTE10(I)*CONDB(7)/CONDA
TSURF(I)=TSURF(I)+TCLRTE(I)*THCLO(I)

0210 CONTINUE

END IF
C CALCULATE MEAN TEMPERATURE OF SKIN SURFACE, EXCLUDING C HEAD.

    DO 0220 I=2,15
    IF(OUTFIT.EQ.'NOT') THEN
        TS=TS+TSURF(I)*ASEG(I)/ASKIN
    ELSE
        TS=TS+TSURF(I)*ACLO(I)/ATCLO
    END IF
0220 CONTINUE

C SET BASAL EVAPORATION RATE FROM SKIN SURFACE (ZERO FOR C COLD WATER IMMERSION).

    DO 0230 I=1,15
    0230 E(I)=EB(I)
RETURN
END

C END OF SUBROUTINE SETUP

***********************************************************************
SUBROUTINE DETAIL
***********************************************************************

C PRINTS SUBJECT CHARACTERISTICS & RESULTS OF GEOMETRICAL C AND THERMAL CALCULATIONS, IF CALLED.

***********************************************************************
INCLUDE 'COMMON.FOR'

    WRITE(5,0010)SG,PBF,ABODY,VBODY
0010 FORMAT(1X,'BODY S.G.=','F7.5,2X,'% BODY FAT=','F7.4,1X,'SURFACE AREA=','F7.4,2X,'SQ M',2X,'VOLUME=','F7.4,2X,'METRE**3')

    WRITE(5,0020)HT,MBODY
0020 FORMAT(/,1X,'HEIGHT=','F5.2,2X,'METRES',2X,'BODY MASS =','F8.4,1X,'KG')

    WRITE(5,0030)R(1,10),VSEG(1)
0030 FORMAT(/,1X,'HEAD RADIUS=','F7.4,1X,'M',T30,1X,'HEAD VOL.=','F7.4,1X,'METRE**3')

56
WRITE (5, 0040) LSEG(2), VSEG(2)
0040 FORMAT (1X, 'THORAX LENGTH=', F7.4, 1X, 'M', T30,
      1   'THORAX VOL.=', F7.4, 1X, 'METRE**3')

WRITE (5, 0050) LSEG(3), VSEG(3)
0050 FORMAT (1X, 'ABDOMEN LENGTH=', F7.4, 1X, 'M', T30,
      1   'ABDOMEN VOL.=', F7.4, 1X, 'METRE**3')

WRITE (5, 0060) LSEG(4), VSEG(4)
0060 FORMAT (1X, 'THIGH LENGTH=', F7.4, 1X, 'M', T30,
      1   'THIGH VOL.=', F7.4, 1X, 'METRE**3 EACH')

WRITE (5, 0070) LSEG(5), VSEG(5)
0070 FORMAT (1X, 'CALF LENGTH=', F7.4, 1X, 'M', T30,
      1   'CALF VOL.=', F7.4, 1X, 'METRE**3 EACH')

WRITE (5, 0080) LSEG(6), VSEG(6)
0080 FORMAT (1X, 'FOOT LENGTH=', F7.4, 1X, 'M', T30,
      1   'FOOT VOL.=', F7.4, 1X, 'METRE**3 EACH')

WRITE (5, 0090) LSEG(10), VSEG(10)
0090 FORMAT (1X, 'UPPER ARM LENGTH=', F7.4, 1X, 'M', T30,
      1   'UPPER ARM VOL.=', F7.4, 1X, 'METRE**3 EACH')

WRITE (5, 0100) LSEG(11), VSEG(11)
0100 FORMAT (1X, 'LOWER ARM LENGTH=', F7.4, 1X, 'M', T30,
      1   'LOWER ARM VOL.=', F7.4, 1X, 'METRE**3 EACH')

WRITE (5, 0110) LSEG(12), VSEG(12)
0110 FORMAT (1X, 'HAND LENGTH=', F7.4, 1X, 'M', T30,
      1   'HAND VOL.=', F7.4, 1X, 'METRE**3 EACH')

WRITE (5, 0120)
0120 FORMAT ('\/, 'SEGMENT DIAMETERS IN METRES', \\)

WRITE (5, 0130) (DSEG(I), I=1, 15)
0130 FORMAT (2 (8F8.4,/))

WRITE (5, 0140)
0140 FORMAT ('\\/,' 'TISSUE SHELL OUTER RADII IN METRES', \\)

WRITE (5, 0150) ((R(I,J), J=1, 10), I=1, 15)
0150 FORMAT (1(10F7.3,\/) )

WRITE (5, 0160)
0160 FORMAT ('\\/,' 'TISSUE SHELL CENTRE-OF-MASS RADII IN', \\ 'METRES', \\)

WRITE (5, 0170) ((RCM(I,J), J=1, 10), I=1, 15)
0170 FORMAT (1(10F7.4,\/) )

WRITE (5, 0180)
0180 FORMAT ('\\/,' 'CONDUCTANCE PATH LENGTHS IN METRES', \\)

WRITE (5, 0190) ((X(I,J), J=1, 10), I=1, 15)
0190 FORMAT (1(10F7.4,\/) )
WRITE (5, 0200)
FORMAT (/,' MID-PLANE RADII IN METRES',/)  
WRITE (5, 0210) ((RMP(I,J), J=1,10), I=1,15)
FORMAT (1(10F7.4, /))
WRITE (5, 0220)
FORMAT (/,' MID-PLANE AREAS IN METRE**2',/)  
WRITE (5, 0230) ((AMP(I,J), J=1,10), I=1,15)
FORMAT (1(10F7.2, /))
WRITE (5, 0240)
FORMAT (/,' THERMAL CONDUCTANCES IN W/DEG C',/)  
WRITE (5, 0250) ((TC(I,J), J=1,10), I=1,15)
FORMAT (1(10F7.3, /))
WRITE (5, 0260)
FORMAT (/,' TISSUE SHELL MASSES IN KG',/)  
WRITE (5, 0270) ((MSHELL(I,J), J=1,10), I=1,15)
FORMAT (1(10F7.3, /))
WRITE (5, 0290)
FORMAT (/,' RESTING METABOLIC HEAT PRODUCTION', 1  'RATES IN WATTS',/)  
WRITE (5, 0300) ((QB(I,J), J=1,10), I=1,15)
FORMAT (1(10F7.3, /))
WRITE (5, 0320)
FORMAT (/,' RESTING BLOOD FLOW RATES IN ML/SEC',/)  
WRITE (5, 0330) ((BFB(I,J), J=1,10), I=1,15)
FORMAT (1(10F7.3, /))
C SCALE C(I,J) FOR PRINT-OUT IN KJ/DEG C
DO 0340 I=1,15
  DO 0340 J=1,10
    C(I,J)=C(I,J)/1000.0
  CONTINUE
WRITE (5, 0350)
FORMAT (/,' THERMAL CAPACITANCES IN KJ/DEG-C',/)  
WRITE (5, 0360) ((C(I,J), J=1,10), I=1,15)
FORMAT (1(10F7.3, /))

58
C RE-SCALE C(I,J) TO J/DEG C

DO 0370 I=1,15
  DO 0370 J=1,10
    C(I,J)=C(I,J)*1000.0
  0370 CONTINUE

C SCALE CS AND CB FOR PRINT-OUT IN KJ/DEG C

CSKIN=CSKIN/1000.0
CBODY=CBODY/1000.0

WRITE(5,0380)
0380 FORMAT (/,' SKIN AND WHOLE BODY THERMAL',
          1 'CAPACITANCES',
          2 IN KJ/DEG C'/)

WRITE(5,0390)CSKIN,CBODY
0390 FORMAT(IX,'CSKIN=',F10.3,T30,'CBODY=',F10.3)

C RE-SCALE CSKIN AND CBODY TO J/DEG C

CSKIN=CSKIN*1000.0
CBODY=CBODY*1000.0

RETURN
END

C END OF SUBROUTINE DETAIL

******************************************************************************

SUBROUTINE CONDIT
******************************************************************************

C SETS ENVIRONMENTAL AND PHYSICAL CONDITIONS WHICH APPLY
C FOR THE DURATION OF EACH PERIOD, WHOSE LENGTH =
C 'CONINT'.

******************************************************************************

INCLUDE 'COMMON.FOR'

C SET UP ENVIRONMENTAL CONDITIONS FOR INTERVAL

IF(FLUID.EQ.'WAT') THEN
  TENV=TW(L)
ELSE
  TENV=TAIR(L)
END IF
\begin{verbatim}
TEMPW = (TW(L) + TS) / 2.0
TEMPA = (TAIR(L) + TSURF(1)) / 2.0
TATM = TAIR(L)
V = AV(L)
RH = ARH(L) / 100

C SET FLUID PROPERTIES (AS FUNCTIONS OF TEMPERATURE, C WHERE APPROPRIATE), USING DATA FROM KAYE & LABY.

C VISCOSITY OF WATER

\[ \mu_W = \frac{0.431}{(24.0 + T_{\text{tempw}})} \]

C THERMAL CONDUCTIVITY OF WATER

\[ \lambda_W = 0.56 + 0.0014 \cdot T_{\text{tempw}} \]

C CALCULATE PRANDTL'S NUMBER FOR WATER

\[ Pr_W = \frac{\nu_W \cdot \lambda_W}{\alpha_W} \]

C DENSITY OF AIR

\[ \rho_A = \frac{29.8 \cdot B_P}{8315.0 \cdot (T_{\text{tempa}} + 273.0)} \]

C VISCOSITY OF AIR

\[ \nu_A = \frac{(1.458 \times 10^{-6} \cdot (T_{\text{tempa}} + 273.0)^1.5)}{((T_{\text{tempa}} + 273.0) + 110.4)} \]

C THERMAL CONDUCTIVITY OF AIR

\[ \lambda_A = \frac{(0.6325 \times 10^{-6} \cdot (T_{\text{tempa}} + 273.0)^1.5)}{((T_{\text{tempa}} + 273.0) + 245.4 \times 10^2 \cdot (12.0 / (T_{\text{tempa}} + 273.0)))} \]

C CALCULATE PRANDTL'S NUMBER FOR AIR

\[ Pr_A = \frac{\nu_A \cdot \lambda_A}{\alpha_A} \]

C CALCULATE FORCED CONVECTION COEFFICIENTS, DEPENDING ON C ENVIRONMENT, USING EMPirical FORMULAE OF MCADAMS.

IF (OUTFIT .NE. 'YES') THEN
  IF (FLUID .NE. 'AIR') THEN

C SURFACE HEATFLOW CALCULATIONS

C CONVECTION, NUDE, HEAD (OUT OF WATER)

\[ \text{Re(1)} = \frac{\rho_A \cdot \nu_A \cdot \delta(1)}{\nu_A} \]
\[ \text{Ho(1)} = \frac{\lambda_A}{\delta(1)} \cdot 0.6 \cdot \text{Re(1)}^{0.5} \]
\[ \text{Ho(1)} = \frac{\lambda_A}{\delta(1)} \cdot 0.6 \cdot \text{Re(1)}^{0.5} \cdot Pr_A^{0.31} \]
\end{verbatim}
C CONVECTION, NUDE, REST OF BODY (IMMERSED)

DO 0010 I=2,15

\[ \text{RE}(I) = \frac{\text{DENW} \times V \times \text{DSEG}(I)}{\text{MUW}} \]
\[ \text{HO}(I) = \frac{\text{CONDW} / \text{DSEG}(I) \times (0.35 \times \text{PRW}^{0.3} + 0.56 \times \text{RE}(I)^{0.52} \times \text{PRW}^{0.3})}{1} \]

0010 CONTINUE

C CONVECTION, NUDE, HEAD (IN AIR)

ELSE

\[ \text{RE}(I) = \frac{\text{DENW} \times V \times \text{DSEG}(I)}{\text{MUW}} \]
\[ \text{HO}(I) = \frac{\text{CONDW} / \text{DSEG}(I) \times 0.6 \times \text{RE}(I)}{1} \times \text{PRA}^{0.31} \]

C CONVECTION, NUDE, REST OF BODY (ALSO IN AIR)

DO 0020 I=2,15

\[ \text{RE}(I) = \frac{\text{DENW} \times V \times \text{DSEG}(I)}{\text{MUW}} \]
\[ \text{HO}(I) = \frac{\text{CONDW} / \text{DSEG}(I) \times (0.35 \times \text{PRA}^{0.3} + 0.56 \times \text{RE}(I)^{0.52} \times \text{PRA}^{0.3})}{1} \]

0020 CONTINUE

END IF

C CLOTHED BODY CALCULATIONS

ELSE

IF(FLUID.NE.'AIR') THEN

C CONVECTION, CLOTHED, HEAD (IN AIR)

\[ \text{RE}(I) = \frac{\text{DENW} \times V \times \text{DCLO}(I)}{\text{MUA}} \]
\[ \text{HO}(I) = \frac{\text{CONDA} / \text{DCLO}(I) \times 0.664 \times \text{RE}(I)}{1} \times \text{PRA}^{0.33} \]

C CONVECTION, CLOTHED, REST OF BODY (IMMERSED)

DO 0030 I=2,15

\[ \text{RE}(I) = \frac{\text{DENW} \times V \times \text{DCLO}(I)}{\text{MUA}} \]
\[ \text{HO}(I) = \frac{\text{CONDW} / \text{DCLO}(I) \times 0.664 \times \text{RE}(I)}{1} \times \text{PRW}^{0.33} \]

0030 CONTINUE
C CONVECTION, CLOTHED, HEAD (IN AIR).
ELSE

RECL0(1)=DENA*V*DCLO(1)/MUA
HOCLO(1)=CONDA/DCLO(1)*0.664*RECL0(1)**0.5
1 *PRA**0.33

C CONVECTION, CLOTHED, REST OF BODY (ALSO IN AIR).

DO 0040 I=2,15

RECL0(I)=DENA*V*DCLO(I)/MUA
HOCLO(I)=CONDA/DCLO(I)*0.664*RECL0(I)**0.5
1 *PRA**0.33

0040 CONTINUE

END IF

END IF

C HEAT TRANSFER TO SURFACE

IF (OUTFIT.NE.'YES') THEN

C CONDUCTION, NUDE, HEAD.

RMPA=RCM(1,10)+(R(1,10)-RCM(1,10))/2.0
AA=4.0*PI*RMPA**2.0
XA=R(1,10)-RCM(1,10)
TCSKIN(1)=CONDB(7)*AA/XA

TC(1,10)=TCSKIN(1)

C CONDUCTION, NUDE, REST OF BODY.

DO 0050 I=2,15

RMPA=RCM(I,10)+(R(I,10)-RCM(I,10))/2.0
AA=2.0*PI*RMPA*LSEG(I)
XA=R(I,10)-RCM(I,10)
TCSKIN(I)=CONDB(7)*AA/XA

TC(I,10)=TCSKIN(I)

0050 CONTINUE

ELSE

C CONDUCTION, CLOTHED, HEAD.

RMPA=RCM(1,10)+(R(1,10)-RCM(1,10))/2.0
AA=4.0*PI*RMPA**2.0
XA=R(1,10)-RCM(1,10)
YA=CONDB(7)*AA/XA
RMPB = R(1,10) + THCLO(1)/2.0
AB = 4.0*PI*RMPB**2.0
XB = THCLO(1)
YB = CONDA*AB/XB

TCSKIN(1) = YA*YB/(YA+YB)
TC(1,10) = TCSKIN(1)

C REMAINING SEGMENTS

DO 0060 I=2,15
RMPA = RCM(I,10) + (R(I,10) - RCM(I,10))/2.0
AA = 2.0*PI*RMPA*LSEG(I)
XA = R(I,10) - RCM(I,10)
YA = CONDB(7)*AA/XA

RMPB = R(I,10) + THCLO(I)/2.0
AB = 2.0*PI*RMPB*LSEG(I)
XB = THCLO(I)
YB = CONDA*AB/XB

TCSKIN(I) = YA*YB/(YA+YB)
TC(I,10) = TCSKIN(I)

0060 CONTINUE

ENDIF

C FIND SURFACE HEATFLOW COEFFICIENTS

IF(OUTFIT .NE. 'YES') THEN

DO 0070 I=1,15
U(I) = HO(I) * ASEG(I)
H(I) = (U(I) * TC(I,10) / (U(I) + TC(I,10))

0070 CONTINUE

ELSE

DO 0080 I=1,15
UCLO(I) = HOCLO(I) * ACLO(I)
HCLO(I) = (UCLO(I) * TC(I,10) / (UCLO(I) + TC(I,10))

0080 CONTINUE

ENDIF
C VAPOUR PRESSURE OF EXPIRED AIR

\[ M = (0.333 \times (TVISC(1) + TVISC(2) + TVL(2)) / 5.0) + 1.0 \]

\[ POUT = P(M) + (P(M+1) - P(M)) \times (0.333 \times (TVISC(1) + TVISC(2) + TVL(2)) - 5.0 \times (M-1)) / 5.0 \]

RETURN

END

C END OF SUBROUTINE CONDIT

**********************************************************************
SUBROUTINE METAB
**********************************************************************
C DETERMINES PHYSICAL RESPONSES TO PREVAILING CONDITIONS
**********************************************************************
C AFFECTER INPUTS FOR METABOLIC CONTROLLERS

DO 0030 I=1,15
   DO 0030 J=1,10
      WARM(I,J)=0.0
      COLD(I,J)=0.0
      IF (TDOT(I,J)) 0020,0010,0010
      0010 TDOT(I,J)=0.0
      0020 CONTINUE
      ERROR(I,J)=T(I,J)-TSET(I,J)+RATE(I,J)*TDOT(I,J)
      IF (ERROR(I,J).LT.0.0) THEN
         COLD(I,J)=-ERROR(I,J)
      ELSE
         IF (ERROR(I,J).GT.0.0) THEN
            WARM(I,J)=ERROR(I,J)
         END IF
      END IF
   END IF
0030 CONTINUE
WARMS=0.0
COLDs=0.0

DO 0040 I=1,15

WARMS=WARMS+WARM(I,10)*SKINR(I)
COLDs=COLDs+COLD(I,10)*SKINR(I)

0040 CONTINUE

C CALCULATE EFFERENT SIGNALS (STOLWIJK & HARDY).
SWEAT=CSW*ERROR(1,1)+SSW*(WARMS-COLDs)
+PSW*WARM(1,1)*WARMS

DILAT=CDIL*ERROR(1,1)+SDIL*(WARMS-COLDs)
+PDIL*WARM(1,1)*WARMS

STRIC=-CCON*ERROR(1,1)
-SCON*(WARMS-COLDs)+PCON*COLD(1,1)*COLDs

C SHIVERING

C HAYWARD METABOLIC CONTROLLER

CHILL=0.0314*MBODY*(TSNEW-42.4)*(TCNEW-41.4)

C ELIMINATE NEGATIVE VALUES

IF(SWEAT) 0050,0060,0060
0050 SWEAT=0.0

0060 IF(DILAT) 0070,0080,0080
0070 DILAT=0.0

0080 IF(STRIC) 0090,0100,0100
0090 STRIC=0.0

0100 IF(CHILL) 0110,0120,0120
0110 CHILL=0.0

0120 CONTINUE

C ALLOCATE METABOLIC HEAT GENERATION AND BLOOD FLOW TO EACH SHELL

MR=0.0

DO 0150 I=1,15
C CORE

DO 0130 J=1,4

QM(I,J)=QB(I,J)*2.0**( (T(I,J)-TSET(I,J))/10.0 )
MR=MR+QM(I,J)

BFREQ(I,J)=BFB(I,J)
1 *2.0**( (T(I,J)-TSET(I,J))/10.0 )

0130 CONTINUE

C MUSCLE (RESTING)

DO 0140 J=5,8

QM(I,J)=QB(I,J)*2.0**( (T(I,J)-TSET(I,J))/10.0 )
MR=MR+QM(I,J)

BFREQ(I,J)=BFB(I,J)
1 *2.0**( (T(I,J)-TSET(I,J))/10.0 )

0140 CONTINUE

C FAT

J=9

QM(I,J)=QB(I,J)*2.0**( (T(I,J)-TSET(I,J))/10.0 )
MR=MR+QM(I,J)

BFREQ(I,J)=BFB(I,J)*2.0**( (T(I,J)-TSET(I,J))/10.0 )

C SKIN

J=10

QM(I,J)=QB(I,J)*2.0**( (T(I,J)-TSET(I,J))/10.0 )
MR=MR+QM(I,J)

BFREQ(I,J)=((BFB(I,J)+SKINV(I)*DILAT)
1 /(1.0+SKINC(I)*STRIC))
2 *2.0**( (T(I,J)-TSET(I,J))/10.0 )

0150 CONTINUE

C MUSCLE (WORKING OR SHIVERING)

C CONSTRAIN METABOLIC RATE TO RESTING VALUE

IF (CHILL .LT. MR) THEN

CHILL=MR

END IF
C SUBTRACT RESTING MUSCLE METABOLISM

DO 0160 I=1,15
    DO 0160 J=5,8
0160   MR=MR-QM(I,J)

C DETERMINE CONTRIBUTIONS TO TOTAL METABOLIC RATE FROM
C WORK AND SHIVERING (FIGURES FROM SOWOOD).

IF(AWORK(L).EQ.0.0) THEN
    SHIVM=CHILL
    WORKM=0.0
ELSE IF(CHILL.LT.(0.816*AWORK(L))) THEN
    SHIVM=0.438*CHILL
    WORKM=0.901*AWORK(L)
ELSE IF(CHILL.GT.(0.816*AWORK(L))) THEN
    SHIVM=CHILL
    WORKM=0.442*AWORK(L)
END IF

C PROPORTIONALLY SUBTRACT ACCUMULATED METABOLIC RATE FOR
C PREVIOUS TISSUES IN ORDER TO OBTAIN MUSCULAR COMPONENT.

    SUBS=SHIVM/(SHIVM+WORKM)
    SUBW=WORKM/(SHIVM+WORKM)
    SHIVM=SHIVM-MR*SUBS
    WORKM=WORKM-MR*SUBW

C METABOLIC RATE DEPENDS ON INTERNAL WORK, BLOOD FLOW ON
C TOTAL WORK.

DO 0170 I=1,15
    DO 0170 J=5,8
    IF((FCHIL(I).NE.0.0).OR.(FWORK(I).NE.0.0)) THEN
        QM(I,J)=FCHIL(I)*SHIVM+FWORK(I)*WORKM*(1-EFF/100)
        BFREQ(I,J)=FBFB(5)*(FCHIL(I)*SHIVM+FWORK(I)*WORKM)

76
ELSE
    BFREQ(I,J)=FBFB(5)*QM(I,J)
END IF
    MR=MR+QM(I,J)
0170 CONTINUE

C SAVE BLOOD FLOWS FROM PREVIOUS TIME STEP
C AND APPLY FIRST-ORDER LAG

    IF(TIME.EQ.0.0) THEN
    DO 0180 I=1,15
    DO 0180 J=1,10
0180    BFNEW(I,J)=BFREQ(I,J)
    ELSE
    DO 0210 I=1,15
        BFVISC(I)=0.0
        BFMUSC(I)=0.0
        DO 0190 J=1,4
0190        BFNEW(I,J)=BFREQ(I,J)
        BFVISC(I)=BFVISC(I)+BFNEW(I,J)
        DO 0200 J=5,8
0200            BFOLD(I,J)=BFNEW(I,J)
            BFNEW(I,J)=BFOLD(I,J)*EXP(-INT/TBF)+BFREQ(I,J)
            1*(1-EXP(-INT/TBF))
        BFMUSC(I)=BFMUSC(I)+BFNEW(I,J)
        BFNEW(I,9)=BFREQ(I,9)
        BFOLD(I,10)=BFNEW(I,10)
0210        BFNEW(I,10)=BFOLD(I,10)
            1*EXP(-INT/5.0)+BFREQ(I,10)
            2*(1-EXP(-INT/5.0))
    END IF
C SUM BLOOD FLOWS TO FIND TOTAL FLOW THROUGH EACH SEGMENT

DO 0220 I=1,15
BFSEG(I)=0.0
DO 0220 J=1,10
BFSEG(I)=BFSEG(I)+BFNEW(I,J)
0220 CONTINUE

BFTSEG(1)=BFSEG(1)
BFTSEG(6)=BFSEG(6)
BFTSEG(9)=BFSEG(9)
BFTSEG(12)=BFSEG(12)
BFTSEG(15)=BFSEG(15)
BFTSEG(5)=BFSEG(5)+BFSEG(6)
BFTSEG(8)=BFSEG(8)+BFSEG(9)
BFTSEG(4)=BFSEG(4)+BFTSEG(5)
BFTSEG(7)=BFSEG(7)+BFTSEG(8)
BFTSEG(3)=BFSEG(3)+BFTSEG(4)+BFTSEG(7)
BFTSEG(11)=BFSEG(11)+BFSEG(12)
BFTSEG(14)=BFSEG(14)+BFSEG(15)
BFTSEG(10)=BFSEG(10)+BFTSEG(11)
BFTSEG(13)=BFSEG(13)+BFTSEG(14)
BFTSEG(2)=BFSEG(2)+BFTSEG(1)+BFTSEG(3)
BFTSEG(10)=BFSEG(10)+BFTSEG(13)

C CALCULATE RESPIRATION VOLUME RATE (ASMUSSEN & NIELSEN).
C OXYGEN REQUIREMENT IS SUBJECT TO A FIRST-ORDER LAG.

IF(TIME.EQ.0.0 .OR. MR .LT. ABODY*RMR) THEN
BWORK=ABODY*RMR
ELSE
BWORK=BWORK*EXP(-INT/TBF)+MR*(1-EXP(-INT/TBF))
END IF

VE=22.0*BWORK/4.83/CONVW
VEMIN=VE/60.0
TI=TATM+273.0
TE=0.333*(TVISC(1)+TVISC(2)+TVL(2))+273.0

C CALCULATE RESPIRATORY HEAT LOSS & GAIN (RUCH & PATTON).

AIRDCCP=0.2905*BP/101324.0*293.0/TE/1000.0
WVOUT=VE*BP*0.75/62.4/TE*POUT/(BP-POUT)
1 *18.01534/99.962

69
RHGEN = 6.7E-3 * VMIN**2.0 + 6.2E-5 * VMIN**3.0
1 * 60.0 * CONVW / 1000.0

RHLOSS = (VE * AIRDCP * (TE - TI) + 0.58 * WVOUT) * CONVW

QRESP = RHLOSS - RHGEN

RETURN

END

C END OF SUBROUTINE METAB

***********************************************************************

SUBROUTINE HEATEX

***********************************************************************

C PERFORMS HEAT EXCHANGE CALCULATIONS

***********************************************************************

INCLUDE 'COMMON.FOR'

C CALCULATE HEAT EXCHANGES BY CONDUCTION

DO 0040 I = 1, 15
  DO 0010 J = 1, 9
    0010 QC(I, J) = TC(I, J) * (T(I, J) - T(I, J + 1))

C HEAT EXCHANGES BY BLOOD SUPPLY TO CAPILLARIES IN TISSUE

FACTOR = SPHEAT(8) * DENSE(8) / 1.0E6

DO 0020 J = 1, 4
  QA(I, J) = BFNEW(I, J) * TAV(I) * FACTOR

DO 0030 J = 5, 8
  QA(I, J) = BFNEW(I, J) * TAM(I) * FACTOR

QA(I, 9) = BFNEW(I, 9) * TAL(I) * FACTOR

QV(I, 9) = BFNEW(I, 9) * T(I, 9) * FACTOR

QA(I, 10) = BFNEW(I, 10) * TAS(I) * FACTOR

QV(I, 10) = BFNEW(I, 10) * T(I, 10) * FACTOR

QAV(I) = BFVISC(I) * TAL(I) * FACTOR
QVV(I) = BFVISC(I) * TVV(I) * FACTOR
QAM(I) = BFMUSC(I) * TAL(I) * FACTOR
QVM(I) = BFMUSC(I) * TVM(I) * FACTOR
QAS(I) = BFNEW(I, 10) * TAL(I) * FACTOR
QVS(I) = BFNEW(I, 10) * TVS(I) * FACTOR

0040 CONTINUE

C CALCULATE HEAT EXCHANGE COEFFS. IN LARGE BLOOD VESSELS
C BETWEEN BLOOD AND VESSEL WALLS. ALSO, COUNTER-CURRENT
C HEAT EXCHANGE COEFFS BETWEEN ARTERIES AND VEINS, EXCEPT
C IN THORAX AND ABDOMEN.

DO 0060 I=1,15
TVISC(I) = 0.0
   DO 0060 J=1,4
      TVISC(I) = TVISC(I) + T(I, J) / 4.0
0060 CONTINUE

DO 0070 I=1,15
TMUSC(I) = 0.0
   DO 0070 J=5,8
      TMUSC(I) = TMUSC(I) + T(I, J) / 4.0
0070 CONTINUE

DO 0080 I=1,15
QATL(I) = HAL(I) * (TAL(I) - TVISC(I))
0080 QVTL(I) = HVL(I) * (TVISC(I) - TVL(I))

DO 0090 I=1,15
QCCL(I) = HCCL(I) * (TAL(I) - TVL(I))
QATV(I) = HAV(I) * (TAV(I) - TVISC(I))
QVT(I) = HVV(I) * (TVISC(I) - TVV(I))
QCCV(I) = HCCV(I) * (TAV(I) - TVV(I))
QATM(I) = HAM(I) * (TAM(I) - TMUSC(I))
QVTM(I) = HVM(I) * (TMUSC(I) - TVM(I))
QCCM(I) = HCCM(I) * (TAM(I) - TVM(I))
QATS(I) = HAS(I) * (TAS(I) - T(I,10))
QVTS(I) = HVS(I) * (T(I,10) - TVS(I))
QCCS(I) = HCCS(I) * (TAS(I) - TVS(I))

0090 CONTINUE

C PERFORM HEAT BALANCES FOR EACH SHELL

DO 0120 I=1,2

QF(I,1) = QM(I,1) - QRESP/12.0 + QA(I,1) - QV(I,1) - QC(I,1)
  + QATL(I)/4.0 - QVTL(I)/4.0 + QATL(I)/4.0 - QVTL(I)/4.0

DO 0100 J=2,4

QF(I,J) = QM(I,J) - QRESP/12.0 + QA(I,J) - QV(I,J)
  + QATL(I)/4.0 - QVTL(I)/4.0 + QC(I, (J-1)) - QC(I,J)
  + QATM(I)/4.0 - QVTM(I)/4.0

DO 0110 J=5,8

QF(I,J) = QM(I,J) + QA(I,J) - QV(I,J)
  + QC(I, (J-1)) - QC(I,J)
  + QATM(I)/4.0 - QVTM(I)/4.0

QF(I,9) = QM(I,9) + QA(I,9) - QV(I,9)
  + QC(I,8) - QC(I,9)

IF (OUTFIT .NE. 'YES') THEN

C SKIN SHELLS, NUDE

QF(I,10) = QM(I,10) - E(I) + QA(I,10) - QV(I,10)
  + QATS(I) - QVTS(I)
  + QC(I,9) - H(I) * (T(I,10) - TENV)

ELSE

C SKIN SHELLS CLOTHED

QF(I,10) = QM(I,10) - E(I) + QA(I,10) - QV(I,10)
  + QATS(I) - QVTS(I)
  + QC(I,9) - HCLO(I) * (T(I,10) - TENV)

END IF

0120 CONTINUE
DO 0150 I=3,15
QF(I,1)=QM(I,1)+QA(I,1)-QV(I,1)-QC(I,1)
1 +QATL(I)/4.0-QVTL(I)/4.0
2 +QATV(I)/4.0-QVT(I)/4.0
DO 0130 J=2,4
0130 QF(I,J)=QM(I,J)+QA(I,J)-QV(I,J)
1 +QATL(I)/4.0-QVTL(I)/4.0
2 +QC(I,(J-1))-QC(I,J)
3 +QATV(I)/4.0-QVT(I)/4.0
DO 0140 J=5,8
0140 QF(I,J)=QM(I,J)+QA(I,J)-QV(I,J)
1 +QC(I,(J-1))-QC(I,J)
2 +QATM(I)/4.0-QVTM(I)/4.0
QF(I,9)=QM(I,9)+QA(I,9)-QV(I,9)
1 +QC(I,8)-QC(I,9)
IF(OUTFIT .NE. 'YES') THEN
C SKIN SHELLS, NUDE
QF(I,10)=QM(I,10)-E(I)+QA(I,10)-QV(I,10)
1 +QATS(I)-QVTS(I)
2 +QC(I,9)-H(I)*(T(I,10)-TENV)
ELSE
C SKIN SHELLS, CLOTHED
QF(I,10)=QM(I,10)-E(I)+QA(I,10)-QV(I,10)
1 +QATS(I)-QVTS(I)
2 +QC(I,9)-HCLO(I)*(T(I,10)-TENV)
END IF
0150 CONTINUE
C CORRECT FOR AIR ENVIRONMENT AT HEAD SURFACE
IF(OUTFIT .NE. 'YES') THEN
QF(1,10)=QM(1,10)-E(1)+QA(1,10)-QV(1,10)
1 +QATS(1)-QVTS(1)
2 +QC(1,9)-H(1)*(T(1,10)-TENV)
ELSE
QF(1,10)=QM(1,10)-E(1)+QA(1,10)-QV(1,10)
1 +QATS(1)-QVTS(1)
2 +QC(1,9)-HCLO(1)*(T(1,10)-TATM)
END IF
C HEAT BALANCE FOR ARTERIAL AND VENOUS POOLS

C FOR EACH SEGMENT, SUM INDIVIDUAL BLOOD FLOWS OF SHELLS
C AND FIND ARTERIAL AND VENOUS HEAT EXCHANGE WITH TISSUE

DO 0160 I=1,15
   QASEG(I)=QAV(I)+QAM(I)+QA(I,9)+QAS(I)
   QVSEG(I)=QVV(I)+QVM(I)+QV(I,9)+QVS(I)
0160 CONTINUE

C BLOOD HEAT BALANCES

DO 0170 I=1,15

C VISCERAL POOLS

   QFAV(I)=QAV(I)-BFVISC(I)*FACTOR*TAV(I)-QCCV(I)
             -QATV(I)
   QFVV(I)=BFVISC(I)*FACTOR*TVISC(I)-QVV(I)+QCCV(I)
             +QVT(I)

C MUSCULAR POOLS

   QFAM(I)=QAM(I)-BFMUSC(I)*FACTOR*TAM(I)-QCCM(I)
             -QATM(I)
   QFVM(I)=BFMUSC(I)*FACTOR*TMUSC(I)
             -QVM(I)+QCCM(I)+QVTM(I)

C SKIN POOLS

   QFAS(I)=QAS(I)-QA(I,10)-QCCS(I)
             -QATS(I)
   QFVS(I)=QV(I,10)-QVS(I)+QCCS(I)
             +QVT(I)
0170 CONTINUE

C LARGE VESSEL POOLS

C HEAD

   QFAL(1)=BFTSEG(1)*TAL(2)*FACTOR
             -QASEG(1)-QATL(1)-QCCL(1)
   QFVL(1)=QVSEG(1)-BFTSEG(1)*TVL(1)*FACTOR
             +QVTL(1)+QCCL(1)
C THORAX

QFAL(2) = (BFTSEG(2) * TVL(2) - (BFTSEG(3) + BFTSEG(10) + BFTSEG(13) + BFTSEG(1)) * TAL(2)) * FACTOR - QATL(2) - QASEG(2)

QFVL(2) = (BFTSEG(1) * TVL(1) + BFTSEG(3) * TVL(3) + BFTSEG(10) * TVL(10) + BFTSEG(13) * TVL(13) - BFTSEG(2) * TVL(2)) * FACTOR + QVSEG(2) + QVTL(2) - QRESP/3.0

C ABDOMEN

QFAL(3) = (BFTSEG(3) * TAL(2) - BFTSEG(4) * TAL(3)) - BFTSEG(7) * TAL(3)) * FACTOR - QASEG(3) - QATL(3)

QFVL(3) = (BFTSEG(4) * TVL(4) + BFTSEG(7) * TVL(7) - BFTSEG(3) * TVL(3)) * FACTOR + QVSEG(3) + QVTL(3)

C BOTH THIGHS

DO 0180 I=4,7,3

QFAL(I) = (BFTSEG(I) * TAL(I) - BFTSEG(I+1) * TAL(I)) * FACTOR - QASEG(I) - QATL(I) - QCCL(I)

QFVL(I) = (BFTSEG(I+1) * TVL(I+1) - BFTSEG(I) * TVL(I)) * FACTOR + QVSEG(I) + QVTL(I) + QCCL(I)

C BOTH CALVES & FORE-ARMS.

DO 0190 I=5,14,3

QFAL(I) = (BFTSEG(I) * TAL(I-1) - BFTSEG(I+1) * TAL(I)) * FACTOR - QASEG(I) - QATL(I) - QCCL(I)

QFVL(I) = (BFTSEG(I+1) * TVL(I+1) - BFTSEG(I) * TVL(I)) * FACTOR + QVSEG(I) + QVTL(I) + QCCL(I)

C BOTH HANDS & FEET

DO 0200 I=6,15,3

QFAL(I) = BFTSEG(I) * TAL(I-1) * FACTOR - QASEG(I) - QATL(I) - QCCL(I)

QFVL(I) = (BFTSEG(I) * TAL(I-1) * FACTOR - QASEG(I) - QATL(I) - QCCL(I)
0200  QFVL(I)=QVSEG(I)-BFTSEG(I)*TVL(I)*FACTOR
1    +QVTL(I)+QCCL(I)

C BOTH UPPER ARMS

   DO 0210 I=10,13,3

    QFAL(I)=(BFTSEG(I)*TAL(2)
1    -BFTSEG(I+1)*TAL(I))*FACTOR-QASEG(I)
2    -QATL(I)-QCCL(I)

0210  QFVL(I)=(BFTSEG(I+1)*TVL(I+1)
1    -BFTSEG(I)*TVL(I))*FACTOR
2    +QVSEG(I)+QVTL(I)+QCCL(I)

   RETURN

   END

C END OF SUBROUTINE HEATEX

******************************************************************************

SUBROUTINE UPDATE

******************************************************************************

C PERFORMS INTEGRATION TO OBTAIN NEW TEMPERATURES

*******************************************************************************

INCLUDE 'COMMON.FOR'

C OBTAIN RATES OF CHANGE OF TEMPERATURE

   DO 0020 I=1,15

       DO 0010 J=1,10

0010   TDOT(I,J)=QF(I,J)/C(I,J)

       TALDOT(I)=QFAL(I)/CBA(I)
       TVLDOT(I)=QFVL(I)/CBV(I)
       TAVDOT(I)=QFAV(I)/CBAV(I)
       TVVDOT(I)=QFVV(I)/CBVV(I)
       TAMDOT(I)=QFAM(I)/CBAM(I)
       TVMDOT(I)=QFVM(I)/CBVM(I)
       TASDOT(I)=QFAS(I)/CBAS(I)
       TVSDOT(I)=QFVS(I)/CBVS(I)

0020   CONTINUE
C INTEGRATION INTERVAL = 60 SECONDS, INITIALLY.

INT=60

C PREVENT TEMPERATURE CHANGES OF > 0.1 DEG C, TO MAINTAIN
C STABILITY.

DO 0060 I=1,15
    TDOTM=ABS(TALDOT(I))
    IF(TDOTM*INT-0.1) 0040,0040,0030
    0030 INT=0.1/TDOTM
0040  TDOTM=ABS(TVLDOT(I))
    IF(TDOTM*INT-0.1) 0060,0060,0050
    0050 INT=0.1/TDOTM
0060 CONTINUE

DO 0080 I=1,15
    DO 0080 J=1,10
        TDOTM=ABS(TDOT(I,J))
        IF(TDOTM*INT-0.1) 0080,0080,0070
        0070 INT=0.1/TDOTM
0080     CONTINUE

DO 0120 I=1,15
    TDOTM=ABS(TAVDOT(I))
    IF(TDOTM*INT-0.1) 0120,0120,0110
    0110 TDOTM=ABS(TWDOT(I))
    IF(TDOTM*INT-0.1) 0130,0130,0120
    0120 CONTINUE

DO 0160 I=1,15
    TDOTM=ABS(TAMDOT(I))
    IF(TDOTM*INT-0.1) 0160,0160,0150
    0150 TDOTM=ABS(TVMDOT(I))
    IF(TDOTM*INT-0.1) 0170,0170,0160
    0160 CONTINUE

DO 0200 I=1,15
    TDOTM=ABS(TASDOT(I))
    IF(TDOTM*INT-0.1) 0180,0180,0170
    0170 INT=0.1/TDOTM
TDOTM = ABS (TVSDOT (I))
IF (TDOTM * INT - 0.1) 0200, 0200, 0190
INT = 0.1 / TDOTM

CONTINUE

C SAVE PREVIOUS MEAN SKIN & CORE TEMPERATURES
TCOLD = TCNEW
TSOLD = TSNEW

C INTEGRATION TO OBTAIN TEMPERATURES
DO 0210 I = 1, 15
TAL (I) = TAL (I) + TALDOT (I) * INT
TVL (I) = TVL (I) + TVLDO T (I) * INT
TAV (I) = TAV (I) + TAVDOT (I) * INT
TVV (I) = TVV (I) + TVVDOT (I) * INT
TAM (I) = TAM (I) + TAMDOT (I) * INT
TVM (I) = TVM (I) + TVMDOT (I) * INT
TAS (I) = TAS (I) + TASDOT (I) * INT
TVS (I) = TVS (I) + TVSDOT (I) * INT
DO 0210 J = 1, 10
T (I, J) = T (I, J) + TDOT (I, J) * INT
CONTINUE

C ESTIMATE TRUE SKIN SURFACE TEMPERATURE
TCNEW = T (3, 1)
TSNEW = 0.0
IF (OUTFIT .NE. 'YES') THEN
TSURF (I) = (TC (I, 10) * T (I, 10) + U (I) * TATM) / (U (I) + TC (I, 10))
DO 0220 I = 2, 15
TSURF (I) = (TC (I, 10) * T (I, 10) + U (I) * TENV) / (U (I) + TC (I, 10))
TSNEW = TSNEW + TSURF (I) * ASEG (I) / ASKIN

78
C COMPENSATE FOR CHANGE IN CONDUCTIVITY OF SKIN, FAT & MUSCLE WITH TEMPERATURE

\[
TC(I,10) = TCSKIN(I) + 0.0014 \times TCSKIN(I) / 0.56 \times (T(I,10) - TSET(I,10))
\]

\[
TC(I,9) = TCFAT(I) + 0.0014 \times TCFAT(I) / 0.56 \times (T(I,9) - TSET(I,9))
\]

DO 0220 J=5,8

\[
TC(I,J) = TCMUSC(I,J-4) + 0.0014 \times TCMUSC(I,J-4) / 0.56 \times (T(I,J) - TSET(I,J))
\]

\[
H(I) = U(I) \times TC(I,10) / (U(I) + TC(I,10))
\]

0220 CONTINUE

ELSE

\[
TSURF(1) = (TC(1,10) \times T(1,10) + UCLO(1) \times TATM) / (UCLO(1) + TC(1,10))
\]

DO 0230 I=2,15

\[
TSURF(I) = (TC(I,10) \times T(I,10) + UCLO(I) \times TENV) / (UCLO(I) + TC(I,10))
\]

\[
TSNEW = TSNEW + T(I,10) \times ASEG(I) / ASKIN
\]

0230 CONTINUE

ENDIF

C CALCULATE NEW VALUES & RATES OF CHANGE OF CORE, FAT & SKIN TEMPERATURE

\[
DTCORE = (TCNEW - TCOLD) / INT
\]

\[
DTSKIN = (TSNEW - TSOLD) / INT
\]

RETURN

END

C END OF SUBROUTINE UPDATE
SUBROUTINE OUTPUT

C PRINTS VALUES OF VARIABLES AT END OF EACH PRINT
C INTERVAL OF LENGTH='PRINT'

INCLUDE 'COMMON.FOR'

C CALCULATE VALUES OF OUTPUT VARIABLES

TEMP=TENV
EVAP=0.0
TS=0.0
TB=0.0
HFLOW=0.0
SBF=0.0

C EVAPORATION LOSS IS PRINTED AS NEGATIVE IF OUTWARDS

DO 0010 I=1,15
EVAP=EVAP-E(I)
0010 CONTINUE

IF(OUTFIT .NE. 'YES') THEN
DO 0020 I=2,15
0020 HFLOW=HFLOW-H(I)*(T(I,10)-TENV)
ELSE
HFLOW=-HCLO(1)*(T(1,10)-TATM)
DO 0030 I=2,15
0030 HFLOW=HFLOW-HCLO(I)*(T(I,10)-TENV)
END IF

SBF=SBF+BFNEW(1,10)
DO 0040 I=2,15
SBF=SBF+BFNEW(I,10)
C CALCULATE MEAN TEMPERATURE OF SKIN SURFACE, EXCLUDING C HEAD.

IF(OUTFIT .NE. 'YES') THEN
   TS=TS+TSURF(I)*ASEG(I)/ASKIN
ELSE
   TS=TS+T(I,10)*ASEG(I)/ASKIN
END IF

0040 CONTINUE

C CALCULATE MEAN WEIGHTED BODY TEMPERATURE

DO 0050 I=1,15
   DO 0050 J=1,10
      TB=TB+T(I,J)*C(I,J)/CBODY
0050 CONTINUE

C HEATFLOW IS PRINTED AS NEGATIVE IF OUTWARDS

QRESP=-QRESP

C IF FIRST PRINT-OUT, PRINT TITLE AND HEADING

IF(PTIME) 0060,0060,0100

C INITIAL RESPIRATORY HEAT LOSS

0060 VE=22.0*ABODY*RMR/4.83/CONVW

VESEC=VE/3600.0

TI=30.0+273.0

TE=0.333*(TVISC(1)+TVISC(2)+TVL(2))+273.0

AIRDCP=0.2905*BP/101324.0*293.0/TE/1000.0

WVOUT=VE*BP*0.75/62.4/TE*POUT/(BP-POUT)*

1 18.01534/99.962

RHGEN=(6.7E-3*((VESEC*60)**2.0))

1 + (6.2E-5*((VESEC*60)**3.0))

2 *60*CONVW/1000.0

RHLOSS=(VE*AIRDCP*(TE-TI))+(0.58*WVOUT)*CONVW

QRESP=(RHLOSS-RHGEN)

HFLOW=-(MR-QRESP)
QRESP = -QRESP

TEMP = 34.6

SBF = 0.0

DO 0070 I = 1, 15

0070 SBF = SBF + BFB(I, 10)

WRITE (5, 0080) TITLE

0080 FORMAT (///, 1X, A80, /)

WRITE (5, 0090)

0090 FORMAT (2X, 'T', 2X, 'TENV', 2X, 'HFLW', 3X,
1 'MR', 4X, 'QR', 4X, 'TBODY', 2X,
2 'TSKIN', 2X, 'TAC', 4X, 'TO', 5X, 'TR', 4X, 'SBF')

0100 WRITE (5, 0110) PTIME, TEMP, HFLOW/ASKIN,
1 MR, QRESP, TB, TS,
2 T(l, 4), T(2, 1), T(3, 1), SBF

0110 FORMAT (I4, F5.1, F7.1, 2F6.1, 4F7.2, 2F6.2)

RETURN

END

C END OF SUBROUTINE OUTPUT

*********************************************************
C END OF PROGRAM
*********************************************************
The file COMMON.FOR is as follows:

NOTE: The COMMON blocks would contain fewer than nine extension lines on a FORTRAN page. This limitation is imposed by some compilers. They have been altered for display purposes only.

**********************************************************
C FILE "COMMON.FOR" - CONTAINS DECLARATIONS OF ALL
C VARIABLES IN ALPHABETICAL ORDER.
**********************************************************

| REAL INT, INTMIN, LBVM, LBVS, LBV, LSEG, 1 MBODY, MR, MSHELL, MTOTAL, MUA, MUW |
| INTEGER PTIME |
| CHARACTER*1 BEEP |
| CHARACTER*3 BPRINT, FLUID, OUTFIT |
| CHARACTER*80 TITLE |
| COMMON/ALPHA/BEEP, BPRINT, FLUID, OUTFIT, TITLE |
| COMMON/NUMA/AA, AB, ABODY, ACLO (15), AIRDCP, 1 AMP (15, 10), AMPCLO (15), ARH (36), 2 ASEG (15), ASKIN, ATCLO, AV (36), AWORK (36) |
| COMMON/NUMB/BFB (15, 10), BFUSC (15), BFNEW (15, 10), 1 BFOLD (15, 10), BFREQ (15, 10), BFSEG (15), 2 BFTSEG (15), BFVISC (15), BP, BWORK |
| COMMON/NUMC/C (15, 10), CBA (15), CBAM (15), 1 CBAS (15), CBAV (15), CBODY, 2 CBV (15), CBVM (15), CBVS (15), 3 CBV (15), CCHIL, CCLO (15), 4 CCON, CDIL, CHILL, CONDA, 5 CONDB (8), CONDC (15), COND, 6 COLD (15, 10), CONINT, CONTME, 7 CONVW, CSKIN, CSW |
| COMMON/NUMD/DCLO (15), DENO (15, 10), DENA, 1 DENSE (8), DENV, DEN, 2 DILAT, DSEG (15), DTCORE, DTSKIN |
| COMMON/NUME/E (15), EB (15), EFF, 1 ERROR (15, 10), EVAP, EXINT |
| COMMON/NUMF/FA (15), FATSUB, FATINT, 1 FBA (6), FBFB (6), FBFS (15), 2 FBONE (15), FBV (6), FCNIL (15), 3 FCLO (15), FFAT (15), 83 |
FMUSC(15), FNAT(15), FQB(6),
FQS(15), FSKIN(15),
FVCORE(15), FVISC(15), FVSEG(15),
FWORK(15)

COMMON/NUMH/H(15), HAL(15), HAM(15), HAS(15),
HAV(15), HCCL(15),
HCCM(15), HCCS(15), HCCV(15), HCLO(15),
HFCLO(15), HFLOW,
HO(15), HOCLO(15), HT, HVL(15), HVM(15),
HVS(15), HVV(15)

COMMON/NUMI/INT

COMMON/NUML/L, LBVM(15), LBVS(15), LBVV(15), LSEG(15)

COMMON/NUMM/M, MBODY, MR, MSHELL(15, 10),
MTO, MUA, MUW

COMMON/NUMN/NBV

COMMON/NUMP/P(11), PBF, PBFIN, PBFS,
PBFV, PBFVS, PCHIL, PCON, PDIL,
PERI(15), PI, POUT, PRA, PRINT, PRNTME,
PRW, PSW, PTIME

COMMON/NUMQ/Q(15, 10), QA(15, 10), QAM(15),
QAS(15), QASEG(15), QATL(15), QATM(15),
QATS(15), QATV(15), QAV(15), QB(15, 10),
QC(15, 10), QCCL(15), QCCM(15), QCCS(15),
QCCV(15), QF(15, 10), QFAL(15), QFAM(15),
QFAS(15), QFAV(15), QFVL(15), QFVM(15),
QFVS(15), QFVV(15), QM(15, 10), QRESP,
QSBF, QV(15, 10), QVM(15), QVS(15),
QVSEG(15), QVTL(15), QVTM(15), QVTS(15),
QVTX(15), QVV(15)

COMMON/NUMR/R(15, 10), RA(15), RAL(15), RAS,
RATE(15, 10), RB(15), RCLLO(15),
RCM(15, 10), RCMCLO(15), RE(15),
RECLO(15), RH, RHGEN, RHLOSS, RLP(15),
RMP(15, 10), RMPA, RMPE, RMPCLO(15), RMR,
RMRS, REM(15), RVL(15), RVS

COMMON/NUMS/SBF, SCHIL, SCON, SDIL, SG, SHA, SHAPE,
SHCORE(15), SHVM, SHW, SKINC(15),
SKINR(15), SKINS(15), SKINV(15),
SPHEAT(8), SSW, STRIC,
SUBS, SUBW, SWEAT

COMMON/NUMT/T(15, 10), TAIR(36), TAL(15), TALDOT(15),
TAM(15), TAMDOT(15), TAS(15), TADSOT(15),
TATM, TAV(15), TAVDOT(15), TB, TBF,
TC(15, 10), TCCLO(15), TCFAT(15),
COMMON/NUMU/U(15), UCLO(15)

COMMON/NUMV/V, VBAL(15), VBAM(15), VBAS(15),
1 VBAV(15), VBLEOD, VBODY,
2 VBVL(15), VBVM(15), VBVS(15),
3 VBVV(15), VE, VEMIN, VESEC,
4 VMUSC(15), VSEG(15), VSHELL(15, 10),
5 VSUM(15), VVISC(15)

COMMON/NUMW/WARM(15, 10), WARM, WORKIN, WORKM, WVOU

COMMON/NUMX/X(15, 10), XA, XB, XBVM(15),
1 XBVV(15), XBV(15), XCLO(15)

COMMON/NUMY/YA, YB

PI=3.14159

**********************************************************
APPENDIX 1 - LIST OF VARIABLES IN MAIN COMPUTER PROGRAM
(SIZES OF ARRAYS SHOWN IN BRACKETS)

AA  Temporary area inside shell boundary
AB  Temporary area outside shell boundary
ABODY  Total surface area of body
ACLO(15)  Surface area of a clothed segment
AIRDCP  Density x specific heat of air
AMP(15,10)  Mid-plane area in a shell
AMPCLO(15)  Mid-plane area in a clothing shell
ARH(36)  Relative humidity of air
ASEG(15)  Surface area of a segment
ASKIN  Surface area of skin excluding head
ATCLO  Total surface area of clothed body
AV(36)  Velocity of fluid (air or water)
AWORK(36)  Total working metabolic rate
BEEP  Audible signal at end of simulation
BFB(15,10)  Resting blood flow rate in a shell
BFMUSC(15)  Blood flow in muscles of a segment
BFNEW(15,10)  Updated blood flow rate in a segment
BFOLD(15,10)  Previous blood flow rate in a shell
BFREQ(15,10)  Blood flow required for oxygen debt
BFSEG(15)  Total blood flow to tissue in a segment
BFTSEG(15)  Total blood flow to & through a segment
BFVISC(15)  Blood flow in viscera of a segment
BP  Barometric pressure
BPRINT  Detailed print-out 'yes' or 'not'
BWORK  Work rate used in respiratory calculations
C(15,10)  Thermal capacitance of a shell
CBA(15)  Th. cap. of blood in major arteries
CBAM(15)  Th. cap. of blood in muscle minor arteries
CBAS(15)  Th. cap. of blood in skin minor arteries
CBAV(15)  Th. cap. of blood in visceral arteries
CBODY  Total thermal capacitance of body
CBV(15)  Th. cap. of blood in major veins
CBVM(15)  Th. cap. of blood in muscle minor veins
CBVSM(15)  Th. cap. of blood in skin minor veins
CBVV(15)  Th. cap. of blood in visceral minor veins
<table>
<thead>
<tr>
<th>Variable</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>CCHIL</td>
<td>Coeff. for core temp. effect on shivering</td>
</tr>
<tr>
<td>CCON</td>
<td>&quot; &quot; &quot; &quot; vasoconstriction</td>
</tr>
<tr>
<td>CDIL</td>
<td>&quot; &quot; &quot; &quot; vasodilation</td>
</tr>
<tr>
<td>CHILL</td>
<td>Total metabolic rate during shivering</td>
</tr>
<tr>
<td>CONDA</td>
<td>Thermal conductivity of air</td>
</tr>
<tr>
<td>CONDB(8)</td>
<td>Thermal conductivities of 8 tissue types</td>
</tr>
<tr>
<td>CONDC(15)</td>
<td>Thermal conductances of clothing layers</td>
</tr>
<tr>
<td>CONDW</td>
<td>Thermal conductivity of water</td>
</tr>
<tr>
<td>COLD(15,10)</td>
<td>Inputs from cold receptors</td>
</tr>
<tr>
<td>CONINT</td>
<td>Time interval for one set of conditions</td>
</tr>
<tr>
<td>CONTME</td>
<td>Elapsed time during condition interval</td>
</tr>
<tr>
<td>CONVW</td>
<td>Conversion factor Kcal hr&lt;sup&gt;-1&lt;/sup&gt; to Watts</td>
</tr>
<tr>
<td>CSKIN</td>
<td>Total thermal capacitance of skin</td>
</tr>
<tr>
<td>CSW</td>
<td>Coeff. for core temp effect on sweating</td>
</tr>
<tr>
<td>DCLO(15)</td>
<td>Diameter of clothed segments</td>
</tr>
<tr>
<td>DEN(15,10)</td>
<td>Estimated densities of tissue shells</td>
</tr>
<tr>
<td>DENA</td>
<td>Density of air</td>
</tr>
<tr>
<td>DENSE(8)</td>
<td>Densities of 8 tissue types</td>
</tr>
<tr>
<td>DENS</td>
<td>Density of water</td>
</tr>
<tr>
<td>DILAT</td>
<td>Skin vasodilation control signal</td>
</tr>
<tr>
<td>DSEG(15)</td>
<td>Diameters of unclothed segments</td>
</tr>
<tr>
<td>DTCORE</td>
<td>Rate of change of core temperature</td>
</tr>
<tr>
<td>DTSKIN</td>
<td>Rate of change of skin temperature</td>
</tr>
<tr>
<td>E(15)</td>
<td>Evaporative heat losses from skin</td>
</tr>
<tr>
<td>EB(15)</td>
<td>Resting evaporative heat loss</td>
</tr>
<tr>
<td>EFF</td>
<td>Mechanical efficiency of muscular work</td>
</tr>
<tr>
<td>ERROR(15,10)</td>
<td>Difference between temperature &amp; set point</td>
</tr>
<tr>
<td>EVAP</td>
<td>Total evaporative heat loss</td>
</tr>
<tr>
<td>EXINT</td>
<td>Total time interval for exposure</td>
</tr>
<tr>
<td>FA(15)</td>
<td>Fractions of surface area in each segment</td>
</tr>
<tr>
<td>FATSUB</td>
<td>Fraction of body fat which is subcutaneous</td>
</tr>
<tr>
<td>FATINT</td>
<td>Fraction of body fat which is internal</td>
</tr>
<tr>
<td>FBA(6)</td>
<td>Blood vol. fractions in major arteries</td>
</tr>
<tr>
<td>FBFB(6)</td>
<td>Resting blood flow per unit mass of tissue</td>
</tr>
<tr>
<td>FBFBS(15)</td>
<td>Resting blood flow per unit mass of skin</td>
</tr>
<tr>
<td>FBONE(15)</td>
<td>Fraction of bone in segment mass</td>
</tr>
<tr>
<td>FBV(6)</td>
<td>Blood vol. fractions in major veins</td>
</tr>
</tbody>
</table>

87
FCHIL(15) Distribution of shivering activity
FFAT(15) Fraction of fat in segment mass
FLUID Exposure medium ‘air’ or ‘water’
FMUSC(15) Fraction of muscle in segment mass
FNAT(15) “ non-adipose tissue in segment mass
FQBS(15) Resting metabolic rate per unit mass
FSKIN(15) Fraction of skin in segment mass
FVISC(15) Fraction of viscera in segment mass
FVSEG(15) Sum of volume fractions in a segment
FWORK(15) Fraction of work in muscles of a segment
H(15) Body surface heat transfer coefficient
HAL(15) Ht. trans. coeff. large arteries to tissue
HAM(15) " " " muscle " " "
HAS(15) " " " skin " " "
HAV(15) " " " visceral " " "
HCCL(15) Counter-current trans. coeff. lge. arteries
HCCM(15) " " " muscle "
HCCS(15) " " " visceral "
HCCV(15) " " " skin "
HCLO(15) Clothing surface heat transfer coefficient
HFLOW Total heat flow at body surface
HO(15) Convective surface heat transfer coeff.
HOCLO(15) Convective clothing heat transfer coeff.
HT Height of subject
HVL(15) Ht. trans. coeff. large veins to tissue
HVM(15) " " " muscle " " "
HVS(15) " " " skin " " "
HVV(15) " " " visceral " " 
I Integer segment number
INT Length of integration interval
J Integer shell number
K Integer tissue type number
L Integer condition interval number
LBVM(15) Length of blood vessels in segment muscle
LBVS(15) " " " " " " skin
LBVV(15) Length of blood vessels in segment viscera
<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>LSEG(15)</td>
<td>Length of a segment</td>
</tr>
<tr>
<td>M</td>
<td>General purpose integer variable</td>
</tr>
<tr>
<td>MBODY</td>
<td>Mass of body</td>
</tr>
<tr>
<td>MR</td>
<td>Metabolic rate</td>
</tr>
<tr>
<td>MSHELL(15,10)</td>
<td>Mass of each tissue shell</td>
</tr>
<tr>
<td>MTOTAL</td>
<td>Uncorrected body mass</td>
</tr>
<tr>
<td>MUA</td>
<td>Viscosity of air</td>
</tr>
<tr>
<td>MUW</td>
<td>Viscosity of water</td>
</tr>
<tr>
<td>N</td>
<td>General purpose integer variable</td>
</tr>
<tr>
<td>NBV</td>
<td>Number of blood vessel paths in a segment</td>
</tr>
<tr>
<td>OUTFIT</td>
<td>Presence of clothing 'yes' or 'not'</td>
</tr>
<tr>
<td>P(11)</td>
<td>Table of pressures for interpolation</td>
</tr>
<tr>
<td>PBF</td>
<td>Percent body fat by mass</td>
</tr>
<tr>
<td>PBFS</td>
<td>Percent body fat by mass for standard man</td>
</tr>
<tr>
<td>PBFV</td>
<td>Percent body fat by volume</td>
</tr>
<tr>
<td>PBFVS</td>
<td>Percent body fat by volume for std. man</td>
</tr>
<tr>
<td>PCHIL</td>
<td>Core &amp; skin product coeff. for shivering</td>
</tr>
<tr>
<td>PCON</td>
<td>&quot; &quot; &quot; &quot; &quot; &quot; vasodilation</td>
</tr>
<tr>
<td>PDIL</td>
<td>&quot; &quot; &quot; &quot; &quot; vasoconstriction</td>
</tr>
<tr>
<td>PERI(15)</td>
<td>Perimeter length of a segment</td>
</tr>
<tr>
<td>PI</td>
<td>3.14159</td>
</tr>
<tr>
<td>POUT</td>
<td>Pressure of expired air</td>
</tr>
<tr>
<td>PRA</td>
<td>Prandtl’s number for air</td>
</tr>
<tr>
<td>PRINT</td>
<td>Interval for print-out of results</td>
</tr>
<tr>
<td>PRNTME</td>
<td>Elapsed time within a print interval</td>
</tr>
<tr>
<td>PRW</td>
<td>Prandtl’s number for water</td>
</tr>
<tr>
<td>PSW</td>
<td>Core &amp; skin product coeff for sweating</td>
</tr>
<tr>
<td>PTIME</td>
<td>Time at which print occurs</td>
</tr>
<tr>
<td>Q(15,10)</td>
<td>Metabolic rate of each shell</td>
</tr>
<tr>
<td>QA(15,10)</td>
<td>Heat transported to tissue by blood flow</td>
</tr>
<tr>
<td>QAM(15)</td>
<td>Heat transported to muscle arterial pools</td>
</tr>
<tr>
<td>QAS(15)</td>
<td>&quot; &quot; &quot; skin</td>
</tr>
<tr>
<td>QASEG(15)</td>
<td>Total seg. heat supply by large arteries</td>
</tr>
<tr>
<td>QATL(15)</td>
<td>Heat cond. to tissue from large arteries</td>
</tr>
<tr>
<td>QATM(15)</td>
<td>&quot; &quot; &quot; muscle</td>
</tr>
<tr>
<td>QATS(15)</td>
<td>&quot; &quot; &quot; skin</td>
</tr>
<tr>
<td>QATV(15)</td>
<td>&quot; &quot; &quot; visceral</td>
</tr>
<tr>
<td>QAV(15)</td>
<td>Heat transported to visc. arterial pools</td>
</tr>
</tbody>
</table>
Resting metabolic rates of tissue shells
Heat conducted to or from a tissue shell
Counter current heat exch. large vessels
" muscle"
" skin"
" visceral"
Total heat flow to or from a tissue shell
Heat transported to or from large arteries
" muscle"
" skin"
" visceral"
" large veins
" muscle"
" skin"
" visceral"
Actual metabolic rate of a tissue shell
Heat exchanged by respiration
Heat removed from tissue by blood flow
Heat removed from muscle venous pools
" skin"
Total seg. heat returned to large veins
Heat cond. to tissue from large veins
" muscle"
" skin"
" visceral"
Heat removed from visceral venous pools
Outer radius of each shell
Radius of each large artery
Radius of all small arteries
Term providing temp. rate sensitivity
Outer radius of clothed segment
Radius at centre of mass in a shell
Radius at centre of mass in clothing
Reynold’s number for unclothed segments
Reynold’s number for clothed segments
Relative humidity
Heat generated by respiration effort
Respiratory heat loss
<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>RLP(15)</td>
<td>Ratio of segment lengths to perimeters</td>
</tr>
<tr>
<td>RMP(15,10)</td>
<td>Mid-plane radii of shells</td>
</tr>
<tr>
<td>RMPA</td>
<td>Temporary radius inside tissue boundary</td>
</tr>
<tr>
<td>RMPB</td>
<td>&quot;                              &quot; outside &quot;                              &quot;</td>
</tr>
<tr>
<td>RMR</td>
<td>Resting metabolic rate</td>
</tr>
<tr>
<td>RMRS</td>
<td>&quot;                              &quot; of standard man</td>
</tr>
<tr>
<td>RVL(15)</td>
<td>Radii of large veins</td>
</tr>
<tr>
<td>RVS</td>
<td>Radius of all small veins</td>
</tr>
<tr>
<td>SBF</td>
<td>Total skin blood flow</td>
</tr>
<tr>
<td>SCHIL</td>
<td>Skin coefficient for shivering</td>
</tr>
<tr>
<td>SCON</td>
<td>&quot;                              &quot; vasoconstriction</td>
</tr>
<tr>
<td>SDIL</td>
<td>&quot;                              &quot; vasodilation</td>
</tr>
<tr>
<td>SG</td>
<td>Specific gravity of subject</td>
</tr>
<tr>
<td>SHA</td>
<td>Specific heat of air at const. pressure</td>
</tr>
<tr>
<td>SHAPE</td>
<td>Shape factor of cylinders</td>
</tr>
<tr>
<td>SHCORE(15)</td>
<td>Specific heat of core tissue</td>
</tr>
<tr>
<td>SHIVM</td>
<td>Shivering content of metabolic rate</td>
</tr>
<tr>
<td>SKINC(15)</td>
<td>Fraction of vasoconstriction in a segment</td>
</tr>
<tr>
<td>SKINR(15)</td>
<td>&quot;                              &quot; skin receptors &quot;                              &quot;</td>
</tr>
<tr>
<td>SKINS(15)</td>
<td>&quot;                              &quot; sweating command &quot;                              &quot;</td>
</tr>
<tr>
<td>SKINV(15)</td>
<td>&quot;                              &quot; vasodilation &quot;                              &quot;</td>
</tr>
<tr>
<td>SPHEAT(8)</td>
<td>Specific heat of 8 tissue types</td>
</tr>
<tr>
<td>SSW</td>
<td>Skin coefficient for sweating</td>
</tr>
<tr>
<td>STRIC</td>
<td>Skin vasoconstiction control signal</td>
</tr>
<tr>
<td>SUBS</td>
<td>Shivering part of non-muscular met. rate</td>
</tr>
<tr>
<td>SUBW</td>
<td>Working &quot;                              &quot; &quot;                              &quot;</td>
</tr>
<tr>
<td>SWEAT</td>
<td>Skin sweating control signal</td>
</tr>
<tr>
<td>T(15,10)</td>
<td>Temperature of a tissue shell</td>
</tr>
<tr>
<td>TAIR(36)</td>
<td>Temperature of air at each interval</td>
</tr>
<tr>
<td>TAL(15)</td>
<td>Temperatures in large arterial pools</td>
</tr>
<tr>
<td>TALDOT(15)</td>
<td>Rate of change of above</td>
</tr>
<tr>
<td>TAM(15)</td>
<td>Temperatures in Muscle arterial pools</td>
</tr>
<tr>
<td>TAMDOT(15)</td>
<td>Rate of change of above</td>
</tr>
<tr>
<td>TAS(15)</td>
<td>Temperatures in skin arterial pools</td>
</tr>
<tr>
<td>TASDOT(15)</td>
<td>Rate of change of above</td>
</tr>
<tr>
<td>TATM</td>
<td>Temperature of air at head</td>
</tr>
<tr>
<td>TAV(15)</td>
<td>Temperatures of visceral arterial pools</td>
</tr>
<tr>
<td><strong>Variable</strong></td>
<td><strong>Description</strong></td>
</tr>
<tr>
<td>----------------------</td>
<td>-------------------------------------------------------------------</td>
</tr>
<tr>
<td>TAVDOT(15)</td>
<td>Rate of change of above</td>
</tr>
<tr>
<td>TB</td>
<td>Mean body temperature</td>
</tr>
<tr>
<td>TBF</td>
<td>Time constant of lag in blood flow</td>
</tr>
<tr>
<td>TC(15,10)</td>
<td>Thermal conductances between tissue shells</td>
</tr>
<tr>
<td>TCFAT(15)</td>
<td>Thermal conductance in fat</td>
</tr>
<tr>
<td>TCLRTE(15)</td>
<td>Rate of change of cloth. temp. with radius</td>
</tr>
<tr>
<td>TCMUSC(15,4)</td>
<td>Thermal conductance in muscle</td>
</tr>
<tr>
<td>TCNEW</td>
<td>Updated value of core temperature</td>
</tr>
<tr>
<td>TCOLOD</td>
<td>Previous value of core temperature</td>
</tr>
<tr>
<td>TCSKIN(15)</td>
<td>Thermal conductance in skin</td>
</tr>
<tr>
<td>TDOT(15,10)</td>
<td>Rate of change of tissue temperature</td>
</tr>
<tr>
<td>TDOTM</td>
<td>Modulus of above</td>
</tr>
<tr>
<td>TE</td>
<td>Temperature of air in respiratory tract</td>
</tr>
<tr>
<td>TEMP</td>
<td>Environmental temperature for print-out</td>
</tr>
<tr>
<td>TEMPA</td>
<td>Mean of air and head surface temperatures</td>
</tr>
<tr>
<td>TEMPW</td>
<td>Mean of water and body surface temps.</td>
</tr>
<tr>
<td>TENV</td>
<td>Temp. of body environment (air or water)</td>
</tr>
<tr>
<td>THCLO(15)</td>
<td>Thickness of clothing layer</td>
</tr>
<tr>
<td>TIME</td>
<td>Elapsed time of exposure</td>
</tr>
<tr>
<td>TITLE</td>
<td>Character string read from subject file</td>
</tr>
<tr>
<td>TMUSC(15)</td>
<td>Mean muscle temperature in a segment</td>
</tr>
<tr>
<td>TOG(15)</td>
<td>Insulation values of clothing layer</td>
</tr>
<tr>
<td>TR</td>
<td>Rectal temperature</td>
</tr>
<tr>
<td>TRATE8(15)</td>
<td>Rate of change of temp. with radius in 8</td>
</tr>
<tr>
<td>TRATE9(15)</td>
<td>&quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; &quot; 9</td>
</tr>
<tr>
<td>TRTDOT(15)</td>
<td>2nd rate of change of temp. with radius</td>
</tr>
<tr>
<td>TRTE10(15)</td>
<td>Rate of change of temp. with radius in 10</td>
</tr>
<tr>
<td>TS</td>
<td>Mean skin surface temperature</td>
</tr>
<tr>
<td>TSET(15,10)</td>
<td>Set point temperatures for tissue shells</td>
</tr>
<tr>
<td>TSNEW</td>
<td>Updated mean skin shell temperature</td>
</tr>
<tr>
<td>TSOLD</td>
<td>Previous mean skin shell temperature</td>
</tr>
<tr>
<td>TSURF(15)</td>
<td>Skin surface temperatures</td>
</tr>
<tr>
<td>TVISC(15)</td>
<td>Mean segment visceral temperature</td>
</tr>
<tr>
<td>TVL(15)</td>
<td>Temperatures in large venous pools</td>
</tr>
<tr>
<td>TVLDOT(15)</td>
<td>Rate of change of above</td>
</tr>
<tr>
<td>TVM(15)</td>
<td>Temperatures in muscle venous pools</td>
</tr>
<tr>
<td>TVMDOT(15)</td>
<td>Rate of change of above</td>
</tr>
<tr>
<td>TVS(15)</td>
<td>Temperatures in skin venous pools</td>
</tr>
<tr>
<td>Variable</td>
<td>Description</td>
</tr>
<tr>
<td>----------</td>
<td>-------------</td>
</tr>
<tr>
<td>TVSDOT(15)</td>
<td>Rates of change of above</td>
</tr>
<tr>
<td>TVV(15)</td>
<td>Temperatures in visceral venous pools</td>
</tr>
<tr>
<td>TVVDOT(15)</td>
<td>Rates of change of above</td>
</tr>
<tr>
<td>TW(36)</td>
<td>Temperature of water in each interval</td>
</tr>
<tr>
<td>U(15)</td>
<td>Surface heat transfer coeff. per unit area</td>
</tr>
<tr>
<td>UCLO(15)</td>
<td>Clothed</td>
</tr>
<tr>
<td>V</td>
<td>Fluid velocity in each interval</td>
</tr>
<tr>
<td>VBAL(15)</td>
<td>Volume of blood in large arteries</td>
</tr>
<tr>
<td>VBAM(15)</td>
<td>&quot;muscle&quot;</td>
</tr>
<tr>
<td>VBAS(15)</td>
<td>&quot;skin&quot;</td>
</tr>
<tr>
<td>VBAV(15)</td>
<td>&quot;visceral&quot;</td>
</tr>
<tr>
<td>VBLOOD</td>
<td>Total blood volume</td>
</tr>
<tr>
<td>VBODY</td>
<td>Total body volume</td>
</tr>
<tr>
<td>VBVL(15)</td>
<td>Volume of blood in large veins</td>
</tr>
<tr>
<td>VBVM(15)</td>
<td>&quot;muscle&quot;</td>
</tr>
<tr>
<td>VBVS(15)</td>
<td>&quot;skin&quot;</td>
</tr>
<tr>
<td>VBVV(15)</td>
<td>&quot;visceral&quot;</td>
</tr>
<tr>
<td>VE</td>
<td>Volume of air expired per hour</td>
</tr>
<tr>
<td>VEMIN</td>
<td>&quot;minute&quot;</td>
</tr>
<tr>
<td>VESEC</td>
<td>&quot;second&quot;</td>
</tr>
<tr>
<td>VMUSC(15)</td>
<td>Volume of muscle in a segment</td>
</tr>
<tr>
<td>VSEG(15)</td>
<td>Volume of a segment</td>
</tr>
<tr>
<td>VSUM(15)</td>
<td>Volume of shell volumes used to find radii</td>
</tr>
<tr>
<td>VVISC(15)</td>
<td>Volume of visceral tissue in a shell</td>
</tr>
<tr>
<td>WARM(15,10)</td>
<td>Output from warm receptors in a shell</td>
</tr>
<tr>
<td>WARMW</td>
<td>Integrated output from skin warm receptors</td>
</tr>
<tr>
<td>WORKIN</td>
<td>Internal muscular work resulting in heat</td>
</tr>
<tr>
<td>WORKM</td>
<td>Fraction of work in metabolic rate</td>
</tr>
<tr>
<td>WOUT</td>
<td>Heat loss due to expired water vapour</td>
</tr>
<tr>
<td>X(15,10)</td>
<td>Conduction path lengths between shells</td>
</tr>
<tr>
<td>XA</td>
<td>Temporary path length inside boundary</td>
</tr>
<tr>
<td>XB</td>
<td>&quot;outside&quot;</td>
</tr>
<tr>
<td>XBVM(15)</td>
<td>Distance between blood vessels in muscle</td>
</tr>
<tr>
<td>XBVS(15)</td>
<td>&quot;skin&quot;</td>
</tr>
<tr>
<td>XBVV(15)</td>
<td>&quot;viscera&quot;</td>
</tr>
<tr>
<td>YA</td>
<td>Temporary th. conductance inside boundary</td>
</tr>
<tr>
<td>YB</td>
<td>&quot;outside&quot;</td>
</tr>
</tbody>
</table>