Chapter 2

Non-invasive continuous core temperature measurement by zero heat flux

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ABSTRACT

Purpose
Reliable continuous core temperature measurement is of major importance to monitor patients. The zero heat flux (ZHF) method can potentially fulfil the requirements of non-invasiveness, reliability and short delay time that current measurement methods lack. The purpose of this study was to determine the performance of a new ZHF device on the forehead regarding these issues.

Methods
Seven healthy subjects performed a protocol of 10 min rest, 30 min submaximal exercise (average temperature increase about 1.5°C) and 10 min passive recovery in ambient conditions of 35°C and 50% relative humidity. ZHF temperature \( T_{zhf} \) was compared to esophageal \( T_{es} \) and rectal \( T_{re} \) temperature.

Results
\( T_{zhf} - T_{es} \) showed an average bias ± standard deviation of 0.17 ± 0.19°C in rest, -0.05 ± 0.18°C during exercise and -0.01 ± 0.20°C during recovery, the latter two being not significant. The 95% limits of agreement ranged from -0.40 to 0.40°C and \( T_{zhf} \) had hardly any delay compared to \( T_{es} \). \( T_{re} \) showed a substantial delay and deviation from \( T_{es} \) and \( T_{zhf} \) when core temperature changed rapidly.

Conclusion
Results indicate that the studied ZHF sensor tracks \( T_{es} \) very well in hot and stable ambient conditions and may be a promising alternative for reliable non-invasive continuous core temperature measurement in hospital.
INTRODUCTION

Body core temperature ($T_{\text{core}}$) is one of the most common and important clinical measures. Substantial deviations from the normal $T_{\text{core}}$ of around 37°C, especially in the brain and the gut, form a serious threat to a subject’s health (1). Further, abnormal $T_{\text{core}}$ can indicate illness at an early stage and guide appropriate action. Beside this, controlled $T_{\text{core}}$ manipulation is used during surgery or as a therapeutic intervention the last few years. For example, mild therapeutic hypothermia is thought to improve the outcome of cardiac arrest and ischemic insult to the brain (2; 3). Therefore, reliable $T_{\text{core}}$ measurement is of major importance to monitor patients.

$T_{\text{core}}$ is actually not a single value and depends on the site of measurement (Pušnik and Miklavec, 2009). Two measures are accepted as gold standard for $T_{\text{core}}$: central blood temperature with a Schwan-Ganz catheter in the pulmonary artery and esophageal temperature ($T_{\text{es}}$) (4). As both are not acceptable in an operational setting, there has been a search for alternative non-invasive measures. However, current non-invasive measurement methods all have major disadvantages concerning reliability, delay time, convenience and/or usability (5-9). So there clearly is a need for a continuous measurement method that is reliable and has a small time delay. Further it needs to be safe, convenient and easy to use. The zero heat flux (ZHF) method may be a suitable method to fulfil these requirements. A ZHF sensor insulates the skin locally, ensuring the skin surface to be heated to deep body temperature and creating a region of zero heat flow from the body core to the skin (10). In that way, it allows measuring $T_{\text{core}}$ at the skin surface. Typical body locations for ZHF sensors have low skinfold thickness and few large veins (11) like the sternum, forehead and occipital region of the head. Zero heat flux sensors are acceptable for subjects because of their non-invasive nature and quickly respond to temperature changes (4; 10).

In previous research, reliability and measurement location of ZHF sensors varied. Fox et al. (10) developed the first ZHF sensor for the sternum. They measured temperatures that were somewhat lower than rectal and ear canal measurements, but unfortunately exact numbers are not reported. Response time was rapid and differing skin temperatures appeared not to affect the measured value. In working condition and/or a cooler climate results were less satisfactory. Studies of Ball et al. (12) and Tsuji et al. (13)
showed rather large deviations from rectal temperature ($T_{re}$) with a ZHF device at the sternum ($\Delta T_{mean}$: 0.54 ± 0.3°C) and the forehead ($\Delta T_{mean}$: 0.9 ± 0.4°C) respectively. Togawa (14) got better results with a ZHF at the occipital region ($\Delta T_{mean}$: 0.1 ± 0.2°C). This suggests that the occipital region is a reliable location, though it is not the most practical one. Unfortunately, in all studies $T_{re}$ was used as a reference instead of $T_{es}$.

In the past few decades little has been published concerning $T_{core}$ estimation by heat flux sensors, apart from the work of Gunga et al. (15; 16) and Kimberger et al. (17). However, they developed a heat flux device in which a heat element that compensates for changing internal and external conditions was omitted. Their ‘double sensor’ predicted $T_{core}$ mathematically by considering skin temperature ($T_{sk}$), heat flux through the sensors and heat losses through the exterior surface. A benefit over a zero heat flux sensor is that the lack of requiring zero heat flux reduces measurement time. Whether the result is more easily affected by changing internal and external conditions has not been established yet (18). Kimberger (17) reported some good clinical results with perioperative and intensive care patients with quite stable $T_{core}$ (98% of the heat flux measurement was within ±0.5°C of $T_{es}$). However, in their last report on healthy subjects during bed-rest, Gunga et al. (16) concluded that the sensor was not accurate enough for performing single individual core body temperature measurements under resting conditions at normal ambient room temperature (95% limits of agreement of −0.72 and +0.55°C).

Recently, Zeiner et al. (19) tested a new prototype non-invasive continuous cerebral temperature sensor (NICCT) using the ZHF method on the forehead. They monitored 19 patients undergoing mild therapeutic hypothermia after cardiac arrest. Compared to $T_{es}$, this resulted in reasonable 95% limits of agreement of -0.59 and +0.36°C. However, the study only investigated comatose patients in a temperature range of 33.5-36°C under clinical conditions. As justly brought up by Opatz (18), the device has not shown its capabilities under different ambient and physiological conditions. Therefore, the purpose of this study was to determine whether this device can also give a reliable estimation of $T_{es}$ when $T_{core}$ is stable, rapidly increasing or rapidly decreasing in the common human core temperature range of 36.5-38.5°C under hot ambient conditions. For that purpose we tested healthy subjects in rest, during exercise and during recovery after exercise in
an ambient temperature of 35°C. We hypothesized that the ZHF device would provide a good estimation of $T_{es}$ with 95% limits of agreement within ±0.5°C.

**METHODS**

**Subjects**

Ten healthy and moderately fit subjects (six males and four females) with a mean age of 28.3 ± 5.3 years and a mean weight of 68.4 ± 9.3 kg participated in this study. Subjects were requested to follow their usual diets and lessen physical activities the last day before each trial. Each subject was fully informed of the purposes, protocol, experimental procedures and any associated risks and benefits before giving their written consent to participate. The experiment was approved by the institutional Ethics Committee at TNO.

**Protocol**

The test procedure consisted of two sessions on separate days, one preparatory session lasting about one hour and one experimental session lasting about three hours. The experimental sessions took place in a warm climatic chamber without any wind at TNO Soesterberg.

At the first meeting subjects completed an informed consent and anamnesis form. Then subjects tried to insert the esophageal probe. This probe had to be introduced via the nose and was then swallowed by drinking water to enter the esophagus. In case of severe gagging reflexes they were excluded from the study, after which ten subjects remained.

For the experimental session, subjects first redressed into sport clothes and inserted a rectal probe. Then a heart rate sensor and skin temperature sensors were attached. After that, the subjects started with a 20-min habituation period within the climatic chamber at 35°C. Ambient temperature during the entire protocol was maintained at 35°C and a relative humidity of 50%. At the start of the habituation period, the $T_{zhf}$ sensors were attached to the forehead, giving them 20 min for stabilization. Then the esophageal probe was inserted and connected to the data acquisition system. After
habituation, the experimental protocol started with a 10-min rest measurement: 5 min in supine position and 5 min in erect position (offered in balanced order) to detect a possible effect of body orientation. Then a 30-min cycling trial was carried out. Subjects started at an intensity of 2 W/kg body mass. The purpose was to increase a subject’s $T_{\text{core}}$ by about 0.05°C/min, reaching an end temperature of around 38.5°C. $T_{\text{core}}$ was monitored every 2 min during the trial. If a subject’s $T_{\text{core}}$ was increasing distinctly too slow or fast for two consecutive 2-min periods (>0.2°C deviation), intensity was adjusted by 20 W. After the cycling trial, subjects got a recovery of 10 min by sitting on a chair in the climatic chamber before the measurement ended.

**Measurement methods and materials**

*Climatic chamber and cycle ergometer.* Experiments were carried out in a custom made climatic room (Weiss Enet, Tiel, The Netherlands). Temperature was set at 35°C with 50% humidity. The 30-min exercise protocol was performed on a Lode Excalibur bicycle ergometer (Lode, Groningen, The Netherlands).

*ZHF sensor.* This study makes use of the ZHF method, measuring deep tissue temperature. In the human body, there is a natural heat flux from the body core to the skin surface as long as $T_{\text{core}}$ is greater than $T_{\text{sk}}$. By locally insulating the skin, blocking all heat from going out, the temperature gradient between core and skin will decrease. $T_{\text{sk}}$ directly under the insulated area will rise until it reaches equilibrium with the warmest region under the insulation ($T_{\text{core}}$). At that moment, zero heat flux is established and $T_{\text{core}}$ can be measured at the skin. For the equations that base the ZHF method, see Zeiner et al. (19).

The ZHF system contains a prototype non-invasive continuous cerebral temperature sensor (NICCT, Philips, Eindhoven, Netherlands). This sensor consists of a patch (40x50x5 mm) that is placed on the forehead and comprises a layer of thermal insulation and electronics. A specific feature of the patch is its flexibility, which enables the sensor to follow the contours of the skin surface. This prevents the occurrence of air pockets between sensor and skin and optimizes thermal contact. Flexibility is ensured by choosing a flexible material (neoprene) for the thermal insulation. In addition, the electronics are mounted on a kapton® layer which is cut in a specific pattern to allow for deformation.
Two thermistors are placed at the top side of the insulation layer and one on the bottom side, continuously monitoring temperature on both sides of the sensor. The heat flux is defined as proportional to the difference between the average top temperature and bottom temperature. Heating elements are located at the top side of the sensor. The heating element is controlled by a proportional integral (PI) controller which is set to drive the heat flux to zero in order to eliminate heat loss from the skin. As heaters are adjusted in response to the continuously monitored temperatures, the sensor is shielded from external and internal influences.

Before the trials, two temperature probes were attached firmly to the skin just above the eyebrows by means of a dual-sided medical-grade adhesive tape (MP 597 MacTac 9710) and an adjustable headband. The second probe was added for testing an alternative sensor, but data from this probe was not used in the analysis. Wounded or inflamed skin at the measurement location was used as a contraindication. Several safety precautions have been built in, to prevent the patient’s skin from overheating in case of technical failure. Further, there is no galvanic contact between the electrical circuitry of the temperature sensor and the skin. The probes were connected via a wired connection to a logging system which displayed and stored all measurements.

*Esophageal, rectal and skin temperature sensors.* $T_{es}$ and $T_{re}$ were measured using thermistors (Yellow Springs Instruments 400 and 700 series respectively, Yellow Springs, OH, USA). Thermistors were calibrated before data acquisition in a thermal water bath (TLC 15, Tamson Instruments, Bleiswijk, The Netherlands) using a Pt100 digital temperature indicator (P650, Dostmann Electronic, Wertheim-Reicholzheim, Germany) with resistance temperature probe (PD-13/S, Tempcontrol, Voorburg, The Netherlands). This certified combination of calibration instruments had an accuracy of ± 0.03°C. The subjects inserted the esophageal sensor themselves through the nasal passage. The insertion depth beyond the nostrils was determined according to the formula: insertion depth (cm) = (0.479 * sitting height (cm)) – 4.44 (20) assuring that the esophageal sensor was located at the level of the left ventricle. The rectal probe was inserted to a depth of ten centimetres beyond the anal sphincter and fixed with tape. Sensors were attached to a custom-made data acquisition system (VU, Amsterdam), consisting of a data logger with medical power supply and Labview software (National Instrument, Austin TX, USA). Sample frequency was 0.5 Hz.
**T	extsubscript{sk}** was determined using iButtons (DS1922L, Maxim Integrated Products Inc, Sunnyvale, CA, USA) at eight locations, as described by ISO 9886 (21). The iButton on the forehead was placed between the headband and the hair line. A weighted average of the eight iButtons resulted in the mean T	extsubscript{sk}. A sample frequency of 0.1 Hz was used.

*Other measures.* To get an indication of the intensity at which the subject was performing, heart rate was measured using a Polar Vantage NV sport tester (Polar Electro, Finland) at a 5 s interval. The mass of the subjects was determined on a weighing scale prior to exercise (Sartorius F300S, Göttingen, Germany) for determination of the initial power output at the ergometer.

**Data analysis**

T	extsubscript{es} data were processed with a gating routine to remove the negative peaks due to swallowing relatively cool saliva. Then individual averages per 10 s, per 5 min and of the last minute of each trial phase were calculated for all temperature parameters, as well as 10 s group averages. These values have been used for statistical analysis in SPSS statistical software (SPSS 17.0, SPSS Inc, Chicago IL, USA). Concerning T	extsubscript{sk}, both mean and forehead skin temperature (T	extsubscript{fh}) were included in the analysis. T	extsubscript{fh} was of particular interest in comparison to the ZHF sensor which was also located on the forehead. Besides, an elevated T	extsubscript{fh} is often seen as an indicator of fever.

*Differences.* ANOVA for repeated measures was performed on each sensor’s averaged temperature values of the last minute of a phase, to determine significant temperature changes in response to a phase transition. T-tests for paired comparison were performed on the 5-min averaged values to determine differences between temperature sensors at different intervals. Significance level was set at \( p<0.05 \). Further, bias may underestimate the real difference when a delta value crosses the x-axis. In that case, positive and negative values average towards zero. To check the relevance of this effect, root mean square (RMS) calculations were made in addition.

*Bland-Altman diagram.* To quantify the deviation between T	extsubscript{re}-T	extsubscript{es} and T	extsubscript{zhf}-T	extsubscript{es} a Bland-Altman diagram (22) was constructed for all individual 5-min values. In this diagram, the average value of two compared temperatures is depicted against their difference. It also
indicates the 95% limits of agreement (LoA) for these measurements at two standard deviations of the difference. We considered 95% LoA’s of less than ±0.5°C as acceptable, as has been done in previous validation studies (16; 17; 23).

*Cross correlation.* All measurement methods have a certain time delay compared to $T_{es}$, i.e. it takes some time before a change in $T_{core}$ will be detected by one of the alternative measures. Especially in changing conditions like exercise, it is important to keep the delay time as small as possible. Therefore cross-correlation on the 10 s group averaged values was used to figure out how much each temperature pattern must be shifted along the x-axis to make it maximally identical to the pattern of $T_{es}$. In fact, the formula slides the studied graph along the x-axis, calculating the integral of their product for each amount of sliding.

**RESULTS**

Seven subjects (four males, three female) finished the experimental protocol with a complete dataset and have been included in the statistical analysis. For three subjects, the ZHF data were unreliable due to technical problems.

**Cycling intensity**

Subjects started their cycling trial at an average intensity of 133 ± 27 W and finished it on average at 146 ± 40 W. One subject started her trial at 1.5 instead of 2.0 W/kg body weight because she had recently gone through a period of inactivity. Subjects cycled at an average heart rate of 152 ± 10 beats per minute (bpm) and reached an average maximum of 174 ± 11 bpm. For individual values see Table 2.1.

**Absolute temperature patterns**

An overview of the temperature patterns during the complete trial, averaged over seven subjects, is depicted in Figure 2.1. The first 10 min are in rest, followed by a 30-min exercise protocol and 10 min of passive recovery. There were no significant differences ($p<0.05$) in any parameter between the rest measurement in supine and erect body orientation. Average $T_{es}$ in rest was 36.86 ± 0.13°C and rose during exercise significantly
Table 2.1. Subject characteristics and intensity parameters. M = male; F = female; P = power output; HR avg = average heart rate; HR max = maximal heart rate.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age (yr)</th>
<th>Weight (kg)</th>
<th>P start (W)</th>
<th>P end (W)</th>
<th>HR avg (bpm)</th>
<th>HR max (bpm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>M</td>
<td>27</td>
<td>74</td>
<td>148</td>
<td>170</td>
<td>157</td>
<td>181</td>
</tr>
<tr>
<td>2</td>
<td>F</td>
<td>26</td>
<td>83</td>
<td>166</td>
<td>186</td>
<td>144</td>
<td>172</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>38</td>
<td>56</td>
<td>84</td>
<td>90</td>
<td>135</td>
<td>154</td>
</tr>
<tr>
<td>4</td>
<td>F</td>
<td>24</td>
<td>66</td>
<td>132</td>
<td>110</td>
<td>152</td>
<td>169</td>
</tr>
<tr>
<td>5</td>
<td>M</td>
<td>23</td>
<td>67</td>
<td>134</td>
<td>160</td>
<td>161</td>
<td>185</td>
</tr>
<tr>
<td>6</td>
<td>M</td>
<td>33</td>
<td>74</td>
<td>148</td>
<td>188</td>
<td>156</td>
<td>179</td>
</tr>
<tr>
<td>7</td>
<td>M</td>
<td>27</td>
<td>58</td>
<td>116</td>
<td>116</td>
<td>162</td>
<td>181</td>
</tr>
</tbody>
</table>

to 38.41 ± 0.41°C, so the manipulation to raise T<sub>es</sub> about 1.5°C succeeded. T<sub>re</sub> and T<sub>zhf</sub> also increased significantly during exercise, but the increase in T<sub>fh</sub> was not significant (p<0.05). T<sub>es</sub> and T<sub>zhf</sub> decreased significantly (p<0.05) during the 10-min recovery phase, but T<sub>re</sub> and T<sub>fh</sub> did not. Average T<sub>sk</sub> during rest was 35.24 ± 0.56°C and increased till an average of 36.31 ± 0.42°C during the last minute of exercise.

Figure 2.1. Esophageal temperature (T<sub>es</sub>), rectal temperature (T<sub>re</sub>), zero heat flux temperature (T<sub>zhf</sub>) and forehead skin temperature (T<sub>fh</sub>) patterns during the complete experimental trial.
Differences between measurement methods

Figure 2.2 shows the average temperature differences between measurement methods for each 5-min interval. At all intervals, $T_{zhf}$ tracked $T_{es}$ better than $T_{re}$. Especially during the recovery phase, $T_{re}$ deviated considerably.

![Figure 2.2. Average temperature differences between different measurement methods - zero heat flux ($T_{zhf}$), esophageal ($T_{es}$), rectal ($T_{re}$) and forehead ($T_{fh}$) - during each 5-min interval of the experimental trial. For clarity of the graph, only positive error bars have been depicted.](image)

Table 2.2 gives the average temperature difference ± standard deviation (SD) between different methods for the three main phases: rest, exercise and recovery. $T_{es}$ and $T_{re}$ are in all phases significantly different, while $T_{es}$ and $T_{zhf}$ only differ during the rest phase ($p<0.05$). During the whole trial there is a large difference and a large variation between $T_{zhf}$ and $T_{fh}$ while measuring temperature at the same location. The RMS values gave a similar impression and have therefore been omitted.

<table>
<thead>
<tr>
<th>Method</th>
<th>Average Temperature Difference (°C) ± Standard Deviation (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T_{zhf}$-$T_{es}$</td>
<td>0.17 ± 0.19*</td>
</tr>
<tr>
<td>$T_{re}$-$T_{es}$</td>
<td>0.25 ± 0.20*</td>
</tr>
<tr>
<td>$T_{zhf}$-$T_{re}$</td>
<td>-0.09 ± 0.31</td>
</tr>
<tr>
<td>$T_{zhf}$-$T_{fh}$</td>
<td>0.97 ± 0.58*</td>
</tr>
</tbody>
</table>

*significant difference ($p<0.05$)
In Figure 2.3, Bland-Altman diagrams for $T_{zhf} - T_{es}$ and, for comparison, $T_{re} - T_{es}$ are depicted. The graphs show a mean difference of $0.00 \pm 0.20^\circ C$ and $0.01 \pm 0.32^\circ C$ respectively for all individual 5-min average values of the experimental trials. The 95% limits of agreement thus ranged from -0.40 to 0.40 for $T_{zhf} - T_{es}$ and from -0.62 to 0.64 for $T_{re} - T_{es}$.

**Delay time**

Cross correlation analysis revealed a substantial delay of $T_{re}$ compared to $T_{es}$. A maximal $R$ value (0.914) was reached for a delay of 3.30-4.10 min. $T_{zhf}$ did not show any delay compared to $T_{es}$ ($R$=0.992).

**Figure 2.3.** Bland-Altman diagrams showing the differences of A) the zero heat flux and esophageal temperatures ($\Delta T_{zhf-T_{es}}$) and B) the rectal and esophageal temperatures ($\Delta T_{re-T_{es}}$). Both figures consist of all individual 5-min values for the rest phase (circles), the exercise phase (squares) and the recovery phase (triangles).

**DISCUSSION**

The ZHF sensor used in this study gave a reliable estimation of $T_{es}$ during both stable and changing body core temperature in a hot windless environment. The 95% limits of agreement of $\pm 0.40^\circ C$ are well within our acceptable level of agreement of $\pm 0.5^\circ C$ and the sensor showed no delay time at all compared to $T_{es}$. This indicates that the studied ZHF sensor has potential for reliable non-invasive continuous $T_{core}$ measurement.
As the rectum is one of the most often used methods for $T_{\text{core}}$ determination, as well in hospital, as in laboratories and at home, it is interesting to compare the results of $T_{\text{re}}$ and $T_{\text{zhf}}$. At rest, $T_{\text{zhf}}$ and $T_{\text{re}}$ performed quite similar. In line with previous studies, $T_{\text{re}}$ in rest was slightly higher than $T_{\text{es}}$ (24; 25) and so did $T_{\text{zhf}}$. During exercise and recovery $T_{\text{zhf}}$ tracked $T_{\text{es}}$ in all intervals better and faster than $T_{\text{re}}$. The rectum is often used for $T_{\text{core}}$ measurement because it consists of a large mass of deep body tissue and is not affected by environmental conditions (7; 26; 27). It can be a useful measure, as it reflects the local temperature in the vulnerable abdominal cavity (28). However, as blood flow to the rectum is low and the mass of organs in the body cavity is large, it requires a great amount of energy to change temperature (7; 26; 28-30). As a result, $T_{\text{re}}$ is unreliable for monitoring quickly changing central blood temperature, reflected by $T_{\text{es}}$. This became most obvious in the recovery phase when individual differences between $T_{\text{re}}$ and $T_{\text{es}}$ rose up to 0.9°C. The time delay of about 4 min, calculated by cross correlation, was actually an underestimation of the real time delay in the recovery phase, as $T_{\text{re}}$ had hardly started its decrease when measurements stopped. At that moment the gap between $T_{\text{es}}/T_{\text{zhf}}$ and $T_{\text{re}}$ was still widening as can be seen in Figure 2.1. This seems to support the suggestion that the time lag of $T_{\text{re}}$ may rise up to 20 min, depending on environment, physical situation etc. (5). Especially in emergency situations where $T_{\text{core}}$ may be fluctuating fast, this can have serious consequences. Proulx (30) already showed that it can lead to serious hypothermia during cooling procedures after heat stress. In this respect, $T_{\text{zhf}}$ seems to be a promising alternative. In contrary to the statement of Yamakage and Namiki (8), $T_{\text{zhf}}$ was still reliable at rates of change above 0.3°C/min.

Figure 2.3 shows that the deviation of $T_{\text{zhf}}$ does not depend on trial phase, while the deviations of $T_{\text{re}}$ are clearly grouped per trial phase, mostly as a result of the discussed delay. Nevertheless, in rest there was a significant difference between $T_{\text{zhf}}$ and $T_{\text{es}}$, which was absent during exercise and recovery. This was largely due to two subjects with substantial average deviations in rest (±0.4°C). Individual analysis could not reveal a technical cause for this deviation, as temperatures seemed stabilized and the other phases looked normal.

Concerning limits of agreement, the current ZHF sensor seems to perform better than previous (zero) heat flux systems (12-16; 31). Previous studies are hard to compare though, as most studies measured under different ambient conditions and at different
locations. Most comparable was the set-up of Gunga et al. (15), as they estimated $T_{core}$ with a heat flux device on the forehead during rest and exercise in 10, 25 en 40°C. However, this was not a zero heat flux device and $T_{re}$ was used as reference. Our experiment shows that $T_{re}$ has an inconsistent deviation from $T_{es}$ when $T_{core}$ is not stable. This may have caused the larger limits of agreement in their study, also in the warm and thus more comparable ambient temperature conditions (-0.08 ± 0.35 and -0.01 ± 0.37°C for work and rest in 25°C; -0.11 ± 0.34 and 0.10 ± 0.42°C for work and rest in 40°C (15)).

The role of measurement technique remains unknown; the exact performance difference between heat flux and zero heat flux should be investigated further under different conditions with equal reference temperatures.

A possible pitfall in the analysis of temperature differences is underestimation of $\Delta T$ when averaging positive and negative differences. Therefore also RMS values were calculated on the temperature differences (10 s averages). Although this resulted in slightly higher deviations, differences of $T_{zhf}$ and $T_{es}$ were still within the set (acceptable) limits. In addition, as the upper limit of the 95% confidence interval of the RMS (0.41°C) for $T_{zhf}-T_{es}$ was very similar to that of $\Delta T$ (0.40°C), the limits of agreement depicted in Figure 2.3 can be considered as a reliable reflection of the expected deviation.

Although there is a good match with $T_{es}$, the question remains which temperature the ZHF device actually measures. From comparison of $T_{zhf}$ and $T_{fh}$, it is clear that $T_{zhf}$ is not simply a reflection of $T_{fh}$, which even in a hot environment appeared to be a poor estimator of $T_{es}$. But viewing the size of the probe and the thermoregulatory mechanisms within the head, it is not likely either that the ZHF probe can penetrate to the deep cerebral structures. Yamakage and Namiki (8) assume reliable measurements to no more than 9 mm, but deduced this from a model using the thermal conductivity of unperfused tissue. Brajkovic and Ducharme (11) used ZHF to estimate muscle temperature and showed that the ZHF probe tracked the muscle temperature to a depth of up to 2 cm below the skin surface. As the probe surface was of similar size, one could assume a similar measurement depth, which would imply that it measures temperature just within the skull. A more precise estimation of measurement depth would be interesting though.

It must be acknowledged that current measurements took place under nearly ideal hot and stable ambient conditions. It is plausible that performance of the ZHF device of the
current size and configuration would deteriorate in more unfavourable conditions. The results of Zeiner et al. (19), who established limits of agreement of -0.59 and +0.36°C with the same ZHF device at a climate controlled intensive care unit (around 23°C, 40% humidity), suggest that the device still performs at an acceptable level near room temperature. However, it remains to be seen whether this also holds good for cooler conditions. Then the temperature gradient between core and skin becomes bigger and insulation of the skin more difficult. Also, the relatively cold shell becomes larger, requiring deeper measurement. Possibly a larger and more powerful ZHF sensor would be necessary. Further, windy conditions form a potential disturbance for the ZHF measurement. Unpublished results of the authors indicate that substantial wind (4 m/s) perpendicular to the sensor seems to double the standard deviation of $\Delta T_{zhf} - T_{es}$.

Nevertheless, the studied ZHF sensor seems to have potential for practical applications, especially for clinical continuous $T_{core}$ monitoring in stable ambient conditions. Currently, several non-invasive measurement methods like rectal, tympanic and oral are used in clinic. The latter two are known to be unreliable as they can easily be affected by external factors (24; 32-34). $T_{re}$ has been shown to deviate from $T_{es}$ during $T_{core}$ changes (7; 30) and is regularly considered as uncomfortable. Besides, all of these methods are not suited for continuous measurement. For application during exercise, the ZHF method has in this study shown its merits under hot windless conditions in a lab. However, for operational application, there are several issues that need to be solved. A mobile ZHF system needs to be compact and wireless, while still providing sufficient energy supply. Further, performance in cold and windy conditions has to be optimized. And finally, for all applications a shortening of the 20 min stabilization period would be a big improvement. Comparison to a simple (non-zero) heat flux device would be useful in that respect.

Concluding, the studied ZHF device tracked $T_{es}$ well with little or no time delay during stable, increasing and decreasing $T_{core}$ in ambient conditions of 35°C without wind, estimating $T_{es}$ better than the traditional rectal measurement. Therefore, the ZHF method may have potential for practical application, at least under warm and stable ambient conditions.
REFERENCES


Chapter 3

Infrared thermal imaging of the inner canthus of the eye as an estimator of body core temperature

Teunissen LPJ, Daanen HAM
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ABSTRACT

Purpose and methods
Several studies suggest that the temperature of the inner canthus of the eye (T_{ca}), determined with infrared thermal imaging, is an appropriate method for core temperature estimation in mass screening of fever. However, these studies used the error prone tympanic temperature as a reference. Therefore, we compared T_{ca} to esophageal temperature (T_{es}) as gold standard in ten subjects during four conditions: rest, exercise, recovery and passive heating.

Results and conclusion
T_{ca} and T_{es} differed significantly during all conditions (mean ΔT_{es}-T_{ca} 1.80 ± 0.89°C) and their relationship was inconsistent between conditions. Also within the rest condition alone, intersubject variability was too large for a reliable estimation of core temperature. This poses doubts on the use of T_{ca} as a technique for core temperature estimation, although generalization of these results to fever detection should be verified experimentally using febrile patients.
INTRODUCTION

The outbreaks of pandemic infections such as SARS in 2002/2003 have called for a method that allows mass screening for fever detection. Infrared thermal imaging is mentioned as an appropriate technique for mass screening of fever (1). The temperature of the inner canthus of the eye ($T_{ca}$) seems the most suited spot (2) although others argue that the average temperature of the face may be valuable as well, albeit in combination with other physiological parameters (3). A recent review observed a wide range in fever detection sensitivity from 4 to 90%, while specificity ranged from 75 to almost 100% (4). All studies in the review used (infrared) tympanic measurements as a reference. However, it is well documented that these measurements may deviate considerably from the core temperature as assessed using more reliable methods such as esophageal temperature (5). Therefore, this study aimed at determining the value of infrared measurements of the inner canthus of the eye ($T_{ca}$) compared to esophageal measurements ($T_{es}$). In order to obtain reproducible conditions, exercise and heat exposure were used to modify the body core temperature instead of fever.

METHODS

Subjects

Ten healthy and fit subjects (six males and four females) with a mean age of 25.8 ± 3.9 years and a mean weight of 72.3 ± 4.6kg participated in this study. Subjects were requested to follow their usual diets and lessen physical activities the last day before each trial. Each subject was fully informed of the purposes, protocol, experimental procedures and any associated risks and benefits before giving their written consent to participate. The experiment was approved by the Ethics Committee at TNO.

Protocol

The test procedure consisted of three sessions on separate days with at least one day in between: one preparatory session, one experimental session in which the subject was actively heated by exercise and one experimental session in which the subject was passively heated by a water perfused suit. The experimental sessions were offered in balanced order.
In the preparatory session subjects completed an informed consent and anamnesis form. Subjects not familiar with the esophageal probe tested their tolerance by inserting this probe. In case of severe gagging reflexes they were excluded from the study.

At the active heating session, subjects first redressed into sport clothes and inserted the rectal and esophageal probe. The heart rate sensor and skin temperature sensors were attached. After about five minutes, when the esophageal probe had stabilized, the measurement started with twenty minutes rest in the climatic chamber (30°C). This was followed by a ten minute submaximal exercise test that started at an intensity of 130W which was, if necessary, increased till subjects reached a heart rate of about 150 beats per minute. Then subjects got two minutes rest, before they performed a maximal exercise trial of eight minutes. They were instructed to cover as much distance as possible during these eight minutes. Hereafter, they got ten minutes of recovery (pedalling quietly at low intensity) before the experimental session stopped.

The passive heating session started with a rest measurement of ten minutes in the climatic chamber (30°C). Then the subject put on the water perfused suit which was set at a temperature of 45°C. The subjects sat down for forty minutes while their core temperature was increased passively. The complete experimental protocol is summarized in Table 3.1.

**Table 3.1.** Experimental protocol of active heating session and the passive heating session. Sessions were offered in balanced order.

<table>
<thead>
<tr>
<th>Time (min)</th>
<th>Activity</th>
<th>Intensity</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Active heating session</strong></td>
<td>0-20 Rest</td>
<td></td>
</tr>
<tr>
<td></td>
<td>20-30 Submaximal exercise</td>
<td>HR ~150 bpm</td>
</tr>
<tr>
<td></td>
<td>30-32 Break</td>
<td></td>
</tr>
<tr>
<td></td>
<td>32-40 Maximal exercise</td>
<td>8 min self-paced</td>
</tr>
<tr>
<td></td>
<td>40-50 Recovery</td>
<td></td>
</tr>
<tr>
<td><strong>Passive heating session</strong></td>
<td>0-10 Rest</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10-20 Putting on tubed garment suit</td>
<td></td>
</tr>
<tr>
<td></td>
<td>20-60 Passive heating (sitting)</td>
<td>T set at 45°C</td>
</tr>
</tbody>
</table>
Materials

Experiments were carried out in a custom made climatic room (Weiss Enet, Tiel, The Netherlands). Temperature was set at 30°C with 50% relative humidity. The 30-min exercise protocol was performed on a Lode Excalibur bicycle ergometer (Lode, Groningen, The Netherlands). To get an indication of the intensity at which the subject was performing, heart rate was measured using a Polar Vantage NV sport tester (Polar Electro, Finland) at a 5 s interval.

\( T_{ca} \) of the eye was measured using a FLIR ThermaCAM SC2000 PAL infrared camera (Flir, Breda, The Netherlands). The camera was positioned at about 1.5 m from the face. IR measurements were made at different time intervals during the active and/or passive heating sessions (Table 3.2).

**Table 3.2.** Overview of infrared (IR) images made during the rest, exercise and recovery phase of the active heating sessions and during the passive heating sessions.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Rest</th>
<th>Exercise</th>
<th>Recovery</th>
<th>Passive heating</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td></td>
<td></td>
<td></td>
<td>X</td>
</tr>
<tr>
<td>2</td>
<td>X</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>X</td>
<td></td>
<td>X</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>X</td>
<td></td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>5</td>
<td>X</td>
<td></td>
<td></td>
<td>X</td>
</tr>
<tr>
<td>6</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>7</td>
<td>X</td>
<td></td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>8</td>
<td>X</td>
<td></td>
<td>X</td>
<td></td>
</tr>
<tr>
<td>9</td>
<td>X</td>
<td></td>
<td>X</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>X</td>
<td>X</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

\( T_{es} \) was measured using a thermistor (Yellow Springs Instruments 700 series, Yellow Springs, OH, USA). This thermistor was calibrated before data acquisition in a thermal water bath (Tamson TLC-15, Tamson instruments, Bleiswijk, The Netherlands) using a certified Pt100 calibration thermometer (P650, Dostmann electronic, Wertheim-Reicholzheim, Germany) with resistance temperature sensor (PD-13/S, Tempcontrol, Voorburg, The Netherlands). The subjects inserted the esophageal sensor themselves through the nasal passage. The insertion depth beyond the nostrils was determined according to the formula of Mekjavic et al. (6) based on sitting height. The \( T_{es} \) sensor was attached to a custom-made data acquisition system (VU, Amsterdam), consisting of a
data logger with medical power supply and Labview software (National Instrument, Austin TX, USA). Sample frequency was 1Hz.

Mean skin temperature \((T_{sk})\) of the body was determined by averaging the results of four iButtons (DS1922L, Maxim Integrated Products Inc, Sunnyvale, CA, USA) as described by ISO 9886 (7). See for an evaluation regarding the use of iButtons Van Marken Lichtenbelt et al. (8). A sample frequency of 0.1Hz was used.

**Data analysis**

Maximal \(T_{ca}\) on each image was determined with ThermaCAM Explorer software (Flir, Breda, The Netherlands). \(T_{es}\) data were gated to remove the negative peaks due to swallowing. Then \(T_{ca}\) measurements were matched with the \(T_{es}\) and \(T_{sk}\) measurements at the exact moment of the IR image. Averages per phase of the experimental sessions were calculated for \(T_{es}\), \(T_{ca}\) and \(T_{sk}\) as well as differences and standard deviations (SD) for each \(T_{es}-T_{ca}\) data pair.

A Bland-Altman diagram (9) was constructed for all data pairs to visualize the deviation between \(T_{es}\) and \(T_{ca}\). In this diagram, the average value of two compared temperatures is depicted against their difference. It also indicates the 95% limits of agreement (LoA) for these measurements at two standard deviations of the difference.

**RESULTS**

Table 3.3 gives the measured values of \(T_{es}\), \(T_{ca}\) and \(T_{sk}\) averaged over the different phases of the experimental sessions.

**Table 3.3.** Values (± SD) of esophageal temperature \((T_{es})\), infrared canthus temperature \((T_{ca})\) and skin temperature \((T_{sk})\) averaged per phase of the different experimental sessions.

<table>
<thead>
<tr>
<th>Phase</th>
<th>N</th>
<th>(T_{es})</th>
<th>(T_{ca})</th>
<th>(T_{sk})</th>
</tr>
</thead>
<tbody>
<tr>
<td>rest</td>
<td>6</td>
<td>36.87 ± 0.29</td>
<td>35.60 ± 1.04</td>
<td>32.61 ± 1.01</td>
</tr>
<tr>
<td>exercise</td>
<td>4</td>
<td>38.35 ± 0.90</td>
<td>35.53 ± 0.43</td>
<td>33.67 ± 1.12</td>
</tr>
<tr>
<td>recovery</td>
<td>6</td>
<td>37.87 ± 0.33</td>
<td>35.87 ± 0.74</td>
<td>34.14 ± 0.82</td>
</tr>
<tr>
<td>passive</td>
<td>4</td>
<td>37.03 ± 0.30</td>
<td>35.75 ± 0.44</td>
<td>36.61 ± 0.42</td>
</tr>
</tbody>
</table>
The differences between $T_{es}$ and $T_{ca}$ at different trial conditions are shown in Figure 3.1. The differences were significant for all periods ($p<0.05$). The Bland-Altman plot (Figure 3.2) for all collected data points shows a mean difference of 1.80°C and a standard deviation of 0.89°C, resulting in 95% limits of agreement of 0.03 to 3.57°C.

**Figure 3.1.** Difference ($\pm$ SD) between infrared temperature of the inner canthus of the eye ($T_{ca}$) and esophageal temperature ($T_{es}$) during rest, exercise and recovery of the active heating session and during the passive heating session in 30°C ambient temperature.

**Figure 3.2.** Bland-Altman diagram for esophageal ($T_{es}$) and infrared temperature of the inner canthus of the eye ($T_{ca}$), showing the difference between $T_{es}$ and $T_{ca}$ ($\Delta T_{es}-\Delta T_{ca}$) as a function of the average of both temperatures. Symbols indicate during which intervals a measurement was made: circles for rest, squares for exercise, triangles for recovery and crosses for passive heating.
In Figure 3.3, the same data points are depicted as a function of $T_{es}$ only. The solid line is the linear fit which describes the relationship between $T_{es}$ and the difference between $T_{es}$ and $T_{ca}$ ($\Delta T_{es}-T_{ca}$). The dotted line shows the best fitting regression line with a fixed slope of 1.0. This line reflects the situation in which the change in $T_{es}$ would be fully responsible for the change in $\Delta T_{es}-T_{ca}$.

**Figure 3.3.** Scatter plot for the difference between esophageal and infrared temperature of the inner canthus of the eye ($\Delta T_{es}-\Delta T_{ca}$) as a function of esophageal temperature ($T_{es}$). Symbols indicate during which intervals a measurement was made: circles for rest, squares for exercise, triangles for recovery and crosses for passive heating. The solid line shows the linear trend for all data points, the dotted line is the best fitting line of identity (slope=1.0) which would indicate that the increase in $\Delta T_{es}-T_{ca}$ is entirely due to the increase in $T_{es}$.

**DISCUSSION**

The study shows that the radiant temperature of the inner canthus of the eye ($T_{ca}$) has a poor and inconsistent relation with esophageal temperature ($T_{es}$) during rest, exercise and recovery. In rest, $T_{ca}$ was about 1.3°C lower than $T_{es}$ (Figure 3.1). During exercise, the average difference between $T_{ca}$ and $T_{es}$ increased to 2.8°C. The increase in core temperature during exercise, as reflected by the rise in $T_{es}$, was largely invisible in $T_{ca}$. So $\Delta T_{es}-\Delta T_{ca}$ increased almost proportionally to $T_{es}$ (Figure 3.3), while a consistent
relationship between $T_{es}$ and $T_{ca}$ would have been reflected by a horizontal line. Therefore, it can be concluded that $T_{ca}$ is not suited for estimating body core temperatures during exercise. Looking at the measurements in rest separately, adding a constant to $T_{ca}$ does not yield a reliable estimator of body core temperature either, because of the large intersubject variation. The passive heating trials indicated that this variation was decreased when subjects wore a water perfused suit, which possibly created a more homogeneous temperature distribution among subjects. However, the standard deviation of 0.46°C does still not allow for reliable core temperature estimation. Unfortunately the passive heating protocol was not forceful enough to result in hyperthermic core temperatures, so a comparison with fever is not feasible.

Although exercise and fever both result in increased body core temperatures, one could argue that the measurements made during and after exercise are not representative for fever because of differing thermoregulatory mechanisms. In fever, the core temperature setpoint is increased due to pyrogens that enter the blood stream and trigger the hypothalamic neurons [10]. Therefore, thermoregulatory responses are directed at the attainment and maintenance of an elevated core temperature. In human exercise, it is generally believed that the increased body temperature results from a delayed onset of heat loss mechanisms and that the setpoint does not increase. After exercise, heat loss exceeds heat production and this induces a core temperature drop back towards the fixed setpoint (10; 11). The following considerations regarding this discrepancy between fever and exercise hyperthermia are relevant for the current experiment.

$T_{ca}$ depends on the skin temperature ($T_{sk}$) of the inner canthus. If exercise and fever have a comparable effect on $T_{sk}$ of the inner canthus, our results on exercise hyperthermia could presumably be generalized to fever hyperthermia. However, to our knowledge, inner canthus $T_{sk}$ has not yet been determined during exercise and fever in the same subject. Mean $T_{sk}$ data suggest that $T_{sk}$ during fever may be higher than during exercise (12). Lenhardt et al. (12) induced fever (38.0-38.5°C) in eleven subjects during a control condition (supine position, only covered by a cotton blanket) and a self-adjust condition (subjects could control their warming themselves). In these conditions, skin temperatures started at a level comparable to the current study (32.5-33.0°C), but reached about 1 (control) to 2.5°C (self-adjust) higher peak values than during exercise/recovery of the current study. Considering these data, generalization from
exercise to fever seems unwarranted. Nevertheless, if $T_{ca}$ really is a reliable, broadly applicable measurement method, it should reflect esophageal temperature for every thermal state of the body, regardless of the way this status is achieved (rest, exercise, passive heating or fever). This is not the case and therefore it poses serious doubts on $T_{ca}$ as an estimator of body core temperature.

Further, as Cabanac (13) pointed out, there may not be a fixed setpoint in human thermoregulation. He argues that the setpoint is continuously adjusted, for instance for body fluid control. This makes fever and exercise hyperthermia more comparable in thermoregulatory aspects. In both cases an increase in body core temperature and skin temperature is observed. In addition, Kenny et al. (14) showed that whole body heat loss is rapidly reduced after exercise, despite the fact that body heat content, muscle temperatures and esophageal temperatures are still elevated (47% of the heat stored during exercise was not dissipated after one hour recovery, while heat loss mechanisms were back at baseline). Therefore, it is possible that a setpoint increase occurred during and after exercise, albeit of minor amplitude. In conclusion, $T_{ca}$ is not a good estimator of core temperature during exercise and recovery hyperthermia. Possibly, this conclusion extends to fever as well, but experimental verification using febrile patients is necessary.

In many studies concerning fever screening with infrared imagery, (infrared) tympanic temperature has been used as a reference to establish the validity of the method. However, the use of (infrared) tympanic temperature as a reference during fever is not acceptable, since tympanic temperature is error prone and dependent on variables as ambient temperature and ear canal morphology (5). For properly establishing validity, it is therefore recommended that future studies use a reliable method, such as esophageal or intravenous temperature measurements.

Unfortunately, it was not feasible to measure $T_{ca}$ for each subject for each thermal condition. However, in mass screening, also many different people in different physical states pass the system. Therefore, a consistent bias for different physical states with a small intersubject variation is a prerequisite. This study suggests that infrared imagery may not fulfil this prerequisite. Future studies should provide more insight into this issue, varying external and internal conditions for larger groups of subjects and structurally using reliable core temperature references.
REFERENCES

Chapter 4

Limitations of temperature measurement in the aural canal with an ear mould integrated sensor

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ABSTRACT

Purpose
Aural canal temperature measurement using an ear mould integrated sensor (T_{ac}) might be a suited method for continuous non-invasive core temperature estimation in operational settings. We studied the effect of ambient temperature, wind and high intensity exercise on T_{ac} and its ability to predict esophageal (T_{es}) and rectal temperatures (T_{re}).

Methods
Seven subjects performed a protocol of rest in 21, 10 and 30°C, followed by exercise and recovery in 30°C. Subjects performed the protocol twice: with and without face-wind from halfway the 30°C rest period. Extra auricle insulation was applied at one side.

Results
Ambient temperature changes affected T_{ac} significantly, while T_{es} and T_{re} remained stable. Insulating the auricle reduced but not abolished this effect. Wind had an immediate cooling effect on T_{ac} independent of auricle insulation. During exercise and recovery in 30°C, T_{ac} provided acceptable group predictions of T_{re} in trials without wind (bias \(-0.66 \pm 0.21°C\) covered, \(-1.20 \pm 0.15°C\) uncovered). Bias was considerably higher with wind, but variability was similar (\(-1.73 \pm 0.11°C\) covered, \(-2.49 \pm 0.04°C\) uncovered). Individual predictions of T_{es} and T_{re} showed more variation, especially with wind.

Conclusion
We conclude that T_{ac} may be used for core temperature assessment of groups in warm and stable conditions.
INTRODUCTION

Both in clinical and in operational settings, there is a need for continuous non-invasive temperature monitoring to quantify hyper- and hypothermia. The tympanic membrane is potentially a good location to accomplish this. It is located close to the hypothalamus, the main human thermoregulatory node, but still easily accessible. Further, it is well perfused with blood vessels circulating to and from the brain (1; 2), suggesting the tympanum might reflect core temperature. However, there are two major problems with tympanic temperature ($T_{ty}$) measurement. First, $T_{ty}$ may be affected locally by ambient conditions like temperature, wind and local cooling/heating of the head (2; 3). Second, it is difficult to measure tympanic temperature properly, especially concerning continuous measurement in an operational setting.

$T_{ty}$ has been compared regularly to esophageal temperature ($T_{es}$), which is often considered as gold standard for temperature measurement. In studies, measuring $T_{ty}$ by direct contact, a reasonable agreement with $T_{es}$ has been found during rest in stable conditions (4-7). However, significant deviations have been reported during passive heating, active heating and/or facial cooling (4; 6; 8-11), although Sato et al. (12) reported similar values for $T_{ty}$ and $T_{es}$ after slight rotation of the probe. It remains uncertain to which extent ambient conditions confounded these measurements. The deviation may also reflect a real difference between brain and trunk temperature. It is reported that in some mammals during heat stress, brain temperature levels off while trunk temperature continues to rise, a phenomenon often referred to as selective brain cooling (13-15). Further, the deviation might be caused by wrong thermocouple positioning. $T_{ty}$ should be measured at the lower anterior quarter of the tympanum to obtain reliable measurements (16). Nevertheless, if measured accurately at the right spot, directly contacting the tympanum with a temperature sensor seems the most reliable method to measure $T_{ty}$ (16). It is not a safe and comfortable method though. The tympanum may be damaged and a slight touch of the skin at the end of the aural canal, which is richly innervated with pain sensors, may cause severe pain. Therefore, direct tympanic measurements may be considered unsuited for practical application in an operational setting.
T<sub>ty</sub> can also be measured indirectly by detecting the emitted heat from tympanum and aural canal with an infrared (IR) sensor. This method is safer, more comfortable and more acceptable for subjects (17) and is therefore frequently used in hospitals and at home. Again, results concerning the relationship of T<sub>ty</sub> and T<sub>es</sub> are conflicting and mostly not convincing (18-32). A major concern is the fact that the shape of the aural canal, poor aiming and/or limited insertion depth may prevent a proper view at the tympanum (1; 20). In addition, by a lack of insulation, environmental influences can cause an inconsistent relationship between T<sub>es</sub> en T<sub>ty</sub> over subjects and over time (2). Further, in-ear IR is not suited for continuous measurement, as heating of the IR sensor itself and condensation of sweat on the lens cause serious technical problems (33 unpublished observations by the authors).

An alternative in-ear method to estimate core temperature is measuring temperature in the external aural canal (T<sub>ac</sub>) or against the wall of the aural canal. This method seems more suited for operational application than contact and infrared measurements. However, a strong relationship between core temperature and T<sub>ac</sub> is not well established (34). Ambient conditions affect accuracy of T<sub>ac</sub> measurement even more than T<sub>ty</sub> unless appropriate insulation is applied (35). Daanen and Wammes (36) measured T<sub>ac</sub> at several locations in the ear and observed, even at room temperature, a temperature difference of >1°C for two points that were 9 mm apart. So for proper measurement, a small temperature sensor has to be placed close to the tympanum and sufficient insulation is required to prevent environmental influences. House (37) indicated that measuring T<sub>ac</sub> with an individualized ear mould could provide a stable measure of core temperature in several different conditions, at least relatively. Recently Nagano et al. (38) presented T<sub>ac</sub> measurements with a thermocouple inserted in a sponge-type ear plug. Subjects performed a 120-min protocol with intermittent rest (15 min) and exercise (20 min, 75 W) periods in 25, 30 and 35°C. T<sub>ac</sub> deviated 0.45 ± 0.08°C, 0.36 ± 0.11°C and 0.30 ± 0.12°C respectively from T<sub>re</sub>.

Although this is a promising result for continuous operational measurement, it has only been obtained in warm and stable ambient conditions. It is not clear yet whether an ear mould (and additional auricle insulation if necessary) can provide sufficient protection to maintain reliable estimations of core temperature in cool or windy conditions. Further, in the study of Nagano et al. (38) core temperatures changed only gradually and over a
rather small range (about 0.8°C), while in operational settings, detection of rapid increases in core temperature is of major importance. Therefore this study aimed to get insight into the behaviour of $T_{ac}$, measured with an ear mould integrated sensor (EMS), during ambient temperature changes, wind application and high intensity exercise performance. For that purpose, we developed individual silicon ear moulds with a thermistor at the proximal side. In that way the thermistor could be brought close to the tympanum in a comfortable way, while the earplug insulated the aural canal. $T_{ac}$ was measured in rest in different ambient temperatures, during heavy exercise in a hot environment, with and without wind application. To study the indirect effect that wind might have by conduction via the auricle and other surrounding tissue, or possibly via selective brain cooling, one ear was protected from the environment by an insulating ear cover. $T_{es}$ was used as reference for body core temperature. We hypothesized that external conditions would still affect EMS measured $T_{ac}$ significantly compared to $T_{es}$, although less pronounced for the ear with the covered auricle. Further, based on the results of Nagano et al. (38), we hypothesized that $T_{ac}$ would track $T_{es}$ properly during exertional hyperthermia.

**METHODS**

**Subjects**

Seven healthy and fit subjects (five males and two females) with a mean age of 25.4 ± 1.8 years and a mean weight of 72.3 ± 5.2 kg participated in this study. Subjects were requested to follow their usual diets and lessen physical activities the last day before each trial. Each subject was fully informed of the purposes, protocol, experimental procedures and any associated risks and benefits before giving their written consent to participate. The experiment was approved by the institutional Ethics Committee.

**Protocol**

The test procedure consisted of one introductory and two experimental sessions on separate days with at least one day in between. At the two experimental sessions, subjects performed an identical protocol of rest and exercise in the climatic chambers at TNO. In one of the experimental sessions, wind was applied during the second half of the trial. Body core temperature was measured with ear mould-integrated thermistors in
both ears, with one auricle protected from the environment, and compared to esophageal, rectal and skin temperatures.

**Introductory session.** At the first meeting subjects who were not familiar with the esophageal probe, tested their tolerance. The probe had to be inserted via the nose and was then introduced into the esophagus by swallowing the sensor with water. In case of severe gagging reflexes subjects were excluded from the study. For the seven subjects who passed this test, silicon ear moulds for both auditory canals were made. Before the experimental sessions, a thermistor was mounted in the ear mould to measure the temperature of the auditory canal.

**Experimental sessions.** First, subjects redressed into sport clothes and inserted the rectal and esophageal probe themselves. Heart rate and skin temperature sensors were attached and the ear moulds were inserted. After about 5 min, when the esophageal probe had stabilized and the ear moulds were largely habituated to the environment, the measurement started with ten minutes rest at room temperature (~21°C), followed by 10 min rest in the cold (10°C) and warm (30°C, 50% relative humidity) climatic chamber. Then subjects stayed in the warm chamber and in session 2 the wind tunnel was turned on, before another 10-min rest measurement was being done. This was followed by a 10-min submaximal exercise test that started with an intensity of 130 W, which was, if necessary, increased till subjects reached a heart rate of about 150 beats per minute (bpm). After 2 min rest, the subjects performed a maximal exercise trial of 8 min. They were instructed to cover as much distance as possible during these 8 min. Each session ended with 10 min of recovery (pedalling quietly at low intensity). Sessions were offered in balanced order. The experimental protocol is summarized in Table 4.1.

**Measurement methods and materials**

**Climatic chamber, cycle ergometer and wind tunnel.** Experiments were carried out in a custom made climatic chamber (Weiss Enet, Tiel, The Netherlands). In the cold chamber, temperature was set at 10°C, relative humidity was not controlled. Temperature in the warm chamber was set at 30°C with 50% humidity. The 30-min exercise protocol was performed on a Lode Excalibur bicycle ergometer (Lode, Groningen, The Netherlands), which was placed in a wind tunnel (DCTLL 850-8, Ziehl-Abegg, Künzelsau, Germany). Wind speed was measured with a flow meter (LV110, Kimo Instruments, France).
Table 4.1. Experimental protocol. $T_{amb} = \text{ambient temperature}$.

<table>
<thead>
<tr>
<th>Session</th>
<th>Time (min)</th>
<th>Activity</th>
<th>$T_{amb}$ (°C)</th>
<th>Wind</th>
<th>Intensity</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0-10</td>
<td>Rest</td>
<td>21</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10-20</td>
<td>Rest</td>
<td>10</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>21-31</td>
<td>Rest</td>
<td>30</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>31-41</td>
<td>Rest</td>
<td>30</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>41-51</td>
<td>Submaximal exercise</td>
<td>30</td>
<td>No</td>
<td>HR ~150 bpm</td>
</tr>
<tr>
<td></td>
<td>51-53</td>
<td>Rest</td>
<td>30</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>53-61</td>
<td>Maximal time trial</td>
<td>30</td>
<td>No</td>
<td>8 min self-paced</td>
</tr>
<tr>
<td></td>
<td>61-71</td>
<td>Recovery</td>
<td>30</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>0-10</td>
<td>Rest</td>
<td>21</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10-20</td>
<td>Rest</td>
<td>10</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>21-31</td>
<td>Rest</td>
<td>30</td>
<td>No</td>
<td></td>
</tr>
<tr>
<td></td>
<td>31-41</td>
<td>Rest</td>
<td>30</td>
<td>4 m/s</td>
<td></td>
</tr>
<tr>
<td></td>
<td>41-51</td>
<td>Submaximal exercise</td>
<td>30</td>
<td>4 m/s</td>
<td>HR ~150 bpm</td>
</tr>
<tr>
<td></td>
<td>51-53</td>
<td>Rest</td>
<td>30</td>
<td>4 m/s</td>
<td></td>
</tr>
<tr>
<td></td>
<td>53-61</td>
<td>Maximal time trial</td>
<td>30</td>
<td>4 m/s</td>
<td>8 min self-paced</td>
</tr>
<tr>
<td></td>
<td>61-71</td>
<td>Recovery</td>
<td>30</td>
<td>4 m/s</td>
<td></td>
</tr>
</tbody>
</table>

**Ear moulds and thermistors.** An individualized ear mould was made for each subject. First, the aural canal was inspected with an otoscope for suitability (no injury, inflammation or severe obstruction and tympanum looks normal). In case of doubt a doctor checked the aural canal and if necessary cleaned it. Then a small wad of cotton was brought into the aural canal just beyond the second turn to protect the tympanum. Subsequently, the subjects’ aural canals were filled with a silicon ear impression material (Addition Ultra, Detax, Ettlingen, Germany) that cured within a few minutes. When solid, the ear print was removed from the ear and a small canal was drilled through the mould. Before each trial a thermistor (P-8432, ICBT, Tokyo, Japan) was mounted into this canal and fixed with tape, with the tip just sticking out of the mould at about 5 mm from the tympanum. In one of the moulds (balanced left or right) a small hollow tube (2 mm) was inserted next to the thermistor channel (Figure 4.1). This open connection to the environment functioned as an air channel to keep sufficient audibility. Just before the experimental trials, the ear moulds were reinserted into the subjects’ ear. In addition, the ear with the mould without air channel was extra protected from the environment with a cotton patch covering the complete auricle (Figure 4.2).
Esophageal, rectal and skin temperature sensors. $T_{es}$ and $T_{re}$ were measured using thermistors (Yellow Springs Instruments 400 and 700 series respectively, Yellow Springs, OH, USA). Thermistors were calibrated before data acquisition in a thermal water bath (TLC 15, Tamson Instruments, Bleiswijk, The Netherlands) using a certified Pt100 digital temperature indicator (P650, Dostmann Electronic, Wertheim-Reicholzheim, Germany) with resistance temperature probe (PD-13/S, Tempcontrol, Voorburg, The Netherlands). Accuracy of the calibration instruments was ± 0.03°C. The subjects inserted the esophageal sensor themselves through the nasal passage. The insertion depth beyond the external nares was determined according to the formula of Mekjavic et al. (39) based on sitting height. The rectal probe was inserted to a depth of 10 cm beyond the anal sphincter and fixed with tape. The esophageal and rectal sensors as well as the other thermistors were attached to a custom-made data acquisition system (VU University Amsterdam, The Netherlands), consisting of a data logger with a medical power supply and Labview software (National Instrument, Austin TX, USA). Sample frequency was set at 1 Hz.

Mean skin temperature of the body was determined by averaging the results of four iButtons (DS1922L, Maxim Integrated Products Inc, Sunnyvale, CA, USA) placed on the neck, scapula, hand and shin, as described by ISO 9886 (40). See for an evaluation regarding the use of iButtons Van Marken Lichtenbelt et al. (41). A sample frequency of 0.1 Hz was used.
Other measures. To get an indication of the intensity at which the subject was exercising, heart rate was measured using an Equivital Life Monitor (Equivital Hidalgo Ltd, Cambridge, UK) at 15-s intervals. The mass of the subjects was determined on a weighing scale prior to exercise (Sartorius F300S, Göttingen, Germany) and used to calculate the initial power on the ergometer.

Data analysis

$T_{es}$ data were processed with a gating routine to remove the negative peaks due to swallowing. Then individual and group averages per 30 s were calculated for all temperature parameters, as well as individual averages per 5 min. These values have been used for statistical analysis in SPSS statistical software (SPSS 17.0, SPSS Inc, Chicago IL, USA).

T-tests for paired comparisons were performed on the 30 s and 5 min averages to calculate bias and standard deviation between the $T_{ac}$, $T_{es}$ and $T_{re}$ sensors for different intervals. ANOVA for repeated measures was used to determine significant temperature changes of all sensors in response to a phase transition. For that purpose the final 30 s of a trial phase was compared to the final 30 s of the next phase. A $2 \times 2$ ANOVA for repeated measures was applied to discover a possible interaction between wind application and wearing an ear cover on $T_{ac}$. Significance level for all tests was set at $p<0.05$.

In view of the different response times of the different temperature measurement locations, cross-correlation on the 10-s averaged values was used to figure out how much each temperature pattern must be shifted along the x-axis to make it maximally identical to each other.
RESULTS

All seven subjects finished the experimental protocol with a complete dataset and have been included in the statistical analysis.

Average temperature patterns

In Figure 4.3, the different average temperature patterns are depicted for the entire trials. The upper panel A contains the trials without wind, panel B contains the trials where the wind tunnel was turned on at 31 min.

Clearly, both the covered and uncovered $T_{ac}$ values ($T_{ac,c}$ en $T_{ac,unc}$ respectively) differed significantly from $T_{es}$ in each phase ($p<0.05$). The absolute difference varied considerably depending on ambient temperature, wind and activity. Except for the rest period at room temperature, $T_{ac,unc}$ differed also significantly from $T_{ac,c}$, especially in the cold room. Changes in ambient temperature while subjects were in rest induced both $T_{ac}$ values to change in an exponential way as a result of the cooling and heating of the ear mould and aural canal. Both core temperatures remained stable. Also note the quick decrease in $T_{es}$ during the break between the exercise bouts in the wind condition (Figure 4.3B).

$T_{ac}$ response to phase transitions

The uncovered ear had a much stronger $\Delta T$ response (and larger standard deviation) to transitions in ambient temperature than the covered ear (Figure 4.4). Further, wind had a significant effect on $T_{ac}$ of both ears. When no wind was present during the second ‘rest 30°C’ phase, $T_{ac}$ continued its significant increase (Figure 4.4, left panel). When wind was turned on at the start of the second ‘rest 30°C’ phase, the increase in $T_{ac}$ stopped instantly (Figure 4.4, right panel). All other phase transitions produced a significant change in $T_{ac}$ values of both ears, except for maximal exercise to recovery.
Figure 4.3. Average rectal ($T_{re}$), esophageal ($T_{es}$), aural canal covered ($T_{ac_c}$), aural canal uncovered ($T_{ac_unc}$) and skin ($T_{sk}$) temperature patterns during the 71-min trial. Panel A shows the protocol entirely without wind, panel B shows the protocol in which the wind tunnel is turned on at 31 min. Vertical lines indicate transitions in protocol phase. The short intervals at 20-21 and 51-53 min are transfer time and exercise break respectively. Submax = submaximal exercise; max = maximal exercise; rec = recovery from exercise.
Figure 4.4. Average covered and uncovered aural canal temperatures ($T_{ac}$) during all trial phases for the condition without any wind (NW) and with wind (W). In the W-condition, wind started at the second ‘rest 30°C’ phase. *Not significantly different (p>0.05) from previous phase.

$T_{ac}$ compared to reference core temperatures during exercise in the heat

During the exercise and recovery periods under consistent ambient conditions, group averages of $T_{ac}$ show a reasonable tracking of $T_{re}$ over time. This is reflected in the acceptable standard deviation (SD) of the group averaged $\Delta T_{re}$-$T_{ac,c}$ and $\Delta T_{re}$-$T_{ac,unc}$ with and without wind (Table 4.2). Bias ± SD of $T_{ac}$ compared to $T_{es}$ was larger, but note that these results are influenced by the delay of $T_{ac}$ compared to $T_{es}$. Cross correlation revealed that a maximal R value (0.88) was attained for a delay of 3.5 min. Correcting this delay by shifting $T_{ac}$ data 3.5 min backwards, resulted in substantially smaller SD’s ($\Delta T_{es}$-$T_{ac,c}$ 0.83 ± 0.18°C and $\Delta T_{es}$-$T_{ac,unc}$ 1.43 ± 0.18°C without wind; 1.78 ± 0.19°C and 2.56 ± 0.19°C respectively with wind).
Further, it is clear that wind strongly increased the absolute difference of $T_{ac}$ with its references, but variability was not severely affected (Table 4.2, row one compared to row two). Finally, individual instead of group averaged calculation reveals a higher variability over time for individual values, especially in de wind condition (Table 4.2, row one and two compared to row three and four respectively).

Table 4.2. Average differences ± standard deviation of the two aural canal temperatures (covered and uncovered: $T_{ac,c}$ and $T_{ac,unc}$ respectively) with esophageal temperature ($T_{es}$) and rectal temperature ($T_{re}$) during the exercise and recovery period in the no wind and wind conditions. Differences are given for 30-s group averaged values (N=60) and 30-s individual values (N=420).

<table>
<thead>
<tr>
<th></th>
<th>$\Delta T_{es}-T_{ac,c}$ (°C)</th>
<th>$\Delta T_{es}-T_{ac,unc}$ (°C)</th>
<th>$\Delta T_{re}-T_{ac,c}$ (°C)</th>
<th>$\Delta T_{re}-T_{ac,unc}$ (°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>No wind (group)</td>
<td>0.89 ± 0.35</td>
<td>1.49 ± 0.33</td>
<td>0.66 ± 0.21</td>
<td>1.20 ± 0.15</td>
</tr>
<tr>
<td>Wind (group)</td>
<td>1.87 ± 0.46</td>
<td>2.63 ± 0.41</td>
<td>1.73 ± 0.11</td>
<td>2.49 ± 0.04</td>
</tr>
<tr>
<td>No wind (individual)</td>
<td>0.89 ± 0.47</td>
<td>1.49 ± 0.51</td>
<td>0.66 ± 0.32</td>
<td>1.20 ± 0.29</td>
</tr>
<tr>
<td>Wind (individual)</td>
<td>1.87 ± 0.71</td>
<td>2.63 ± 0.73</td>
<td>1.73 ± 0.58</td>
<td>2.49 ± 0.47</td>
</tr>
</tbody>
</table>

Ear cover and wind

Figure 4.5 shows the average $T_{ac}$ for the covered and the uncovered ear per 5-min interval (4-min interval for maximal exercise), both in the NW and the W condition. Till the second ‘Rest 30°C’ phase, the protocol of the two conditions (NW and W) was identical and there was neither for the uncovered nor for the covered ear a significant difference between conditions. However, from the moment the wind tunnel was turned on, both the uncovered and the covered side showed a significant difference between the NW and W conditions at each interval. Remarkably, there was no interaction between wind and ear coverage on $T_{ac}$; the wind effect was not significantly different for the uncovered and covered ear in any phase. So although $T_{ac,unc}$ was structurally lower than $T_{ac,c}$, wind had a similar significant temperature effect on both ears.
Figure 4.5. Average covered and uncovered aural canal temperatures ($T_{ac,c}$ and $T_{ac,unc}$) per 5-min trial interval (4-min interval for max exercise), both in the no wind (NW) and the wind (W) condition. In the W-condition, wind started at the arrow. To the right of this line, data points of the W-condition are significantly different from the NW-condition for both the covered and the uncovered ear.

DISCUSSION

This study addressed some potential limitations of $T_{ac}$ measurement with a thermistor at the tip of an ear mould as an indicator of core temperature. Results showed that varying ambient temperature in the range of 10-30°C clearly affected $T_{ac}$ and led to a poor estimate of core temperature. Further, it appeared that wind had an immediate significant effect on $T_{ac}$, which was independent of extra insulation of the auricle. This suggests that wind provides fast local cooling of $T_{ac}$ without the need for direct input on the auricle or aural canal. Finally, during exercise and recovery in stable ambient conditions, $T_{ac}$ gave acceptable group predictions of $T_{re}$ and, when its faster response was taken into account, $T_{es}$. However, increased SD’s of the mean differences (Table 4.2) indicate that individual predictions were less reliable, especially when wind was applied.

Ambient temperature

$T_{ac}$ measured by the EMS in the air chamber between ear mould, tympanum and aural canal wall is a weighted average of its surrounding structures. Ambient temperature
appeared to influence this temperature substantially independent of core temperature changes. This ambient temperature dependence of EMS-measured $T_{ac}$ agrees with previous reports for diverse in-ear measurement methods (2; 36; 42-44).

The influence of ambient temperature on $T_{ac}$ may have been caused by several mechanisms. First, the ear mould probably conducted some of the external cold or heat, which affected the measured temperature at the proximal tip of the ear mould. In that perspective it must be noted that the silicon ear mould had a large time constant causing a transient effect in the temperature pattern (Figure 4.3). Secondly, it has to be noted that the inside of the tympanic membrane is in contact with air of the cavities in the human head and thus, the tympanic temperature may be affected by ambient temperature from the inside as well. Thirdly, ambient conditions may have influenced the temperature of the tympanum and aural canal wall indirectly by conduction via the auricle and surrounding tissue. An accompanying effect of this mechanism would be vasoconstriction of the vessels of the aural canal which leads to less heat emission from the aural canal wall to the air pocket. Finally, ambient temperature may have affected the temperature of the local blood circulation. This mechanism may be induced by superficial cooling of the external carotid artery, which is a main supplier of arterial blood to this area, or passage of cool venous blood from the scalp and face (1; 2; 45).

As expected, covering the entire auricle led to a significantly higher $T_{ac}$ which was closer to core temperature than $T_{ac}$ uncovered. Apparently, the insulation of the ear mould itself was not sufficient to accomplish this. It suggests that the auricle plays a major role in conducting ambient temperature to the aural canal. Admittedly, the ear mould in the uncovered ear also contained an air channel. However, unpublished results without wearing the ear cover suggest that having an air channel in the ear mould does not make a substantial difference; the ear cover seemed fully responsible for the differences by stabilizing the temperature of the outer ear. The merits of stabilizing the temperature of the outer ear during $T_{ac}$ measurements were already appreciated by Keatinge and Sloan (42), who kept the outer ear at the same temperature level as the aural canal by servo-controlled heating. They found that $T_{ac}$ stabilized within 0.35°C of $T_{es}$ in an ambient temperature of 18-45°C and moderate wind; this often held good for cooler temperatures as well, albeit with slower stabilization, and thus prevented serious $T_{ac}$ depression in cold air.
**Wind application**

Next to ambient temperature, facial cooling by wind also had a significant effect on $T_{ac}$. As soon as the wind tunnel was turned on, $T_{ac}$ started to differ significantly from the values during the trial without wind. This agrees with Thomas et al. (2) who measured infrared $T_{ty}$ and concluded that facial cooling by fanning altered the relationship between $T_{ty}$ and $T_{es}$. Remarkable in the current results was the fact that, in contrast to ambient temperature, the wind effect appeared to be independent of the ear cover. Apparently the wind-induced decrease in $T_{ac}$ has not been caused by a direct decrease of the air temperature in the aural canal or around the auricle. This suggests a cooling mechanism by convection via the blood and/or by conduction via surrounding tissue. As mentioned before, convective cooling via the blood can be accomplished by cooled venous blood from the scalp and face or by superficially cooled external carotid blood. As there was a fast distinct response of $T_{ac}$ to wind (within 30 s) and conduction is a slower process than convection, convection might be the primary mechanism.

**Prediction of core temperature**

Considering the small SD of the difference, group averages of $T_{ac}$ provided a quite reliable prediction of $T_{re}$ during exercise and recovery in stable conditions. This held good for the wind condition as well, although a larger bias had to be taken into account. $T_{re}$ is an important and generally accepted practical measure for core temperature, although it does not indicate rapid changes in central blood temperature, as reflected by $T_{es}$. Nagano et al. (38) found a $\Delta T_{re}$-$T_{ac}$ of $0.36 \pm 0.11 ^\circ C$ with a thermocouple insulated by an ear plug under comparable ambient conditions, but during longer and lower intensity exercise. So despite the higher exercise intensity in our protocol, the SD was similar. The mean difference in this study was substantially higher than in Nagano et al. (38), possibly because our $T_{ac}$ was still affected by the cold interval earlier in the protocol. $T_{es}$-$T_{ac}$ differences seemed more variable, but after correction for $T_{ac}$’s delayed response, variability was reduced and predictive reliability of $T_{es}$ was close to $T_{re}$.

Individual predictions of $T_{re}$ during exercise and recovery in conditions without wind have to be judged critically because of the substantial variability of the difference between $T_{re}$ and $T_{ac}$ (SD’s of 0.29 and 0.32°C for the uncovered and covered ear). This is in line with the study of Muir (43) who compared $T_{ac}$ and $T_{re}$ during exercise in a hot environment.
without wind to evaluate worker safety guidelines. Their group mean predictions were satisfactory, but individual variability was rather large for setting effective guidelines (SD's of 0.28°C for an ear thermistor insulated by an ear plug and 0.36°C for an operationally used personal heat stress monitor). Individual predictions of $T_{re}$ in windy conditions cannot be considered reliable, as variability of the difference was at an unacceptable level (SD’s of ~0.5°C). As individual predictions are most relevant and could considerably extend the field of application, improvements on this issue are necessary. Variability might be reduced by bringing the sensor still closer to the tympanum and gluing the thermistors solidly into the ear mould. Applying some artificial heating to the outer ear is also likely to decrease variability in prediction, mainly by reducing external influences.

Predictions during unstable ambient conditions (e.g. changing ambient temperature and/or wind conditions) might be improved when a compensatory calculation would be available, taking into account the environmental situation. Future studies should reveal whether it is possible to predict core temperature reliably from $T_{ac}$ under changing ambient conditions when multipoint measures are used to account for these conditions. A prediction model will have to include the transient effect of the specific ear mould.

**Selective brain cooling**

One could argue that the results with wind support the concept of selective brain cooling (SBC). SBC may be defined as cooling of the brain temperature (often assumed to be reflected in the tympanic temperature) below arterial blood temperature. For SBC to take place, first venous blood has to be cooled superficially at the upper respiratory tract and at the surface of the head (11; 46-49). Secondly, this cooled venous blood would have to cool the inner brain by 1)countercurrent heat exchange with the internal carotid (50), 2)heat exchange with the cortical cerebral arteries mediated by the cerebrospinal fluid (51) and/or 3)direct contact with brain tissue (13). It is probable that superficial venous blood cooling is stimulated by face fanning. However, it seems improbable that within half a minute from starting fanning, the brain is cooled by one of these mechanisms and causes $T_{ac}$ to stop its increase. Therefore, the cause of the $T_{ac}$ cooling does not seem to originate in whole brain cooling but in more local phenomena.
Numerous studies applied face fanning during rest, active heating or passive heating, while measuring $T_{ty}$. Nearly all of them found a similar fast and distinct cooling response of $T_{ty}$, which exceeded the response of $T_{es}$ significantly (4-8; 10; 52-54). Although $T_{ty}$ is not equal to $T_{ac}$, it is plausible that face fanning affects $T_{ty}$ in a similar way as $T_{ac}$ was affected in this study. So, in agreement with several of the referred studies (4-8; 10; 52-54), it seems not appropriate to use $T_{ty}$ results during face fanning as evidence for the existence of selective brain cooling.

**Conclusion**

In conclusion, this study showed that changing ambient temperature severely affected the relationship of $T_{ac}$ with $T_{es}$ and $T_{re}$. Covering the auricle attenuated this effect, but not sufficiently to allow for reliable predictions. Wind also altered the relationship of $T_{ac}$ and its reference core temperatures. As this effect was fast and independent of covering the auricle, local vascular cooling mechanisms seem to be involved. During exercise and recovery in warm and stable ambient conditions, $T_{ac}$ allowed for acceptable group predictions of core temperature. However, for reliable individual predictions, methodological improvements are required.

**REFERENCES**


Chapter 5

Telemetry pill versus rectal and esophageal temperature during extreme rates of exercise-induced core temperature change

Teunissen LPJ, de Haan A, de Koning JJ, Daanen HAM
Physiol Meas 2012; 33(6): 915-924
ABSTRACT

Purpose and methods
Core temperature measurement with an ingestible telemetry pill has been scarcely investigated during extreme rates of temperature change, induced by short high-intensity exercise in the heat. Therefore, nine participants performed a protocol of rest, (sub)maximal cycling and recovery in 30°C during which pill temperature ($T_{pill}$) was compared to rectal temperature ($T_{re}$) and esophageal temperature ($T_{es}$).

Results
$T_{pill}$ corresponded well to $T_{re}$ during the entire trial, but deviated considerably from $T_{es}$ during the exercise and recovery periods. During maximal exercise the average $\Delta T_{pill} - T_{re}$ and $\Delta T_{pill} - T_{es}$ were $0.13 \pm 0.26°C$ and $-0.57 \pm 0.53°C$ respectively. Response time from the start of exercise, rate of change during exercise and peak temperature were similar for $T_{pill}$ and $T_{re}$. $T_{es}$ responded 5 min earlier, increased more than twice as fast and its peak value was $0.42 \pm 0.46°C$ higher than $T_{pill}$.

Conclusion
Also during considerable temperature changes at very high rate, $T_{pill}$ is still representative of $T_{re}$. The extent of the deviation in pattern and peak values between $T_{pill}$ and $T_{es}$ (up to $>1°C$) strengthens the assumption that $T_{pill}$ is unsuited to evaluate central blood temperature when body temperatures change rapidly.
INTRODUCTION

The last two decades, radio telemetry has become an increasingly popular method for core temperature determination in operational settings and field studies. Radio telemetry uses a ‘sensor pill’ which is swallowed and transmits FM signals reflecting gastrointestinal temperature. Because of the wireless and comfortable nature of this measurement method, it is regularly used during exercise, for subjects wearing protective clothing and for prolonged monitoring.

It has been recognized for ages that different body sites may be subject to different temperatures (1). The measurement method may even increase variation (2). For monitoring purposes, it is important to determine whether $T_{pill}$ can be used to estimate accepted core temperatures as $T_{re}$ and $T_{es}$ when direct measurement is not feasible. $T_{re}$ is thought to give an indication of the temperature in the vulnerable abdominal cavity and is an adequate index of whole body temperature in rest or steady state exercise (3). $T_{es}$ reflects central blood temperature and responds fast when the body is gaining or losing heat. Both measures are regularly used for clinical monitoring. In addition, during severe hyperthermia, $T_{re}$ and $T_{es}$ may indicate the temperature of the body structures that are most at risk: the gut and the brain respectively (4; 5). In that respect, it would be useful if $T_{pill}$ could provide a reliable estimation of either of these measures.

Most previous studies comparing pill temperature ($T_{pill}$) to either esophageal ($T_{es}$) or rectal temperature ($T_{re}$) showed acceptable levels of agreement (6). As a result, Byrne and Lim (6) concluded in their review that $T_{pill}$ is a valid index of core temperature. However, most of the reviewed studies only looked at slowly and slightly changing core temperatures, due to circadian rhythm, immersion or low intensity exercise (7-14).

Regarding the few studies that applied high intensity exercise, Easton et al. (15) showed very good agreement with $T_{re}$ during 16 km maximal cycling in 30°C ambient temperature, as well as Gant et al. (16) during 4x12 min intermittent shuttle running in 15°C. However, these studies did not include $T_{es}$ in their measurements and exercise duration limited the maximal rate of change in core temperature. Kolka et al. (17) compared $T_{pill}$ to both $T_{re}$ and $T_{es}$ during 3x5 min cycling at 80% $VO_{2peak}$ (30°C ambient temperature), inducing a high rate of temperature change. They concluded that the rate
of change in $T_{es}$ was twice as high as $T_{pill}$ and five times as high as $T_{re}$. There was also a significant difference in response time, $T_{pill}$ being slower than $T_{es}$, but faster than $T_{re}$. This difference between $T_{pill}$ and $T_{re}$ is not in line with Gant et al. (16) and Easton et al. (15), who found similar responses. Although all three studies suggest that $T_{pill}$ is at least not slower than $T_{re}$, clear evidence about the relationship between $T_{pill}$ and $T_{re}$ during large high-rate temperature changes is warranted to properly judge its value in those conditions. Further, because of the short intermittent protocol in the study of Kolka et al. (1993), temperature changes were small and differences did not reach the hyperthermic range (>38°C). So it could not be established to which extent $T_{pill}$ deviates from $T_{es}$ during larger high-rate temperature fluctuations.

Because of these limitations and inconsistencies in previous studies, the main purpose of this study is to directly compare $T_{pill}$ to both $T_{es}$ and $T_{re}$ during short high intensity exercise in the heat inducing substantial temperature changes at a very high rate. To get more insight in any differences between high intensity exercise at submaximal and maximal level, the protocol consisted of both a submaximal and a maximal cycling exercise bout. In addition, stabilization and recovery periods were monitored. Following Easton et al. (15) and Kolka et al. (17), ambient temperature was set at 30°C to induce a quick and substantial rise in core temperature. We hypothesized that $T_{pill}$ would track $T_{re}$ and give a delayed and attenuated image of $T_{es}$.

**METHODS**

The study was approved by the Research and Ethics Committee of TNO, The Netherlands.

**Participants**

Nine healthy and fit participants (six males and three females, exercising at least two times a week at recreational level) with a mean age of 26 ± 4 years, a mean weight of 72.9 ± 4.6 kg and tolerant of the esophageal probe, participated in this study. Participants were requested to follow their usual diets and lessen physical activities the last day before each trial. Each participant was fully informed of the purposes, protocol,
Telemetry pills during extreme rates of core temperature change

experimental procedures and any associated risks and benefits before giving their written consent to participate.

Protocol

Four hours before their experimental session, participants swallowed a temperature pill with water. As food, drinks and saliva might affect temperature measurements as long as the pill is located in the stomach, the pill has to be swallowed several hours before the start of measurement in order to reach the intestinal tract (6). Just before the measurement, participants redressed, attached a heart rate sensor and inserted a rectal and esophageal probe themselves. After about 5 min, when the esophageal temperature had stabilized, the measurement protocol in the climatic chamber started (Table 5.1). The submaximal exercise test was performed on a bicycle ergometer and started with an intensity of 130 W. Intensity was, if necessary, increased till participants reached a heart rate of about 150 beats per minute (bpm) in the last three minutes. During the subsequent 8-min maximal exercise trial, participants were instructed to cover as much distance as possible and received distance feedback from a display.

Table 5.1. Experimental protocol.

<table>
<thead>
<tr>
<th>Time (min)</th>
<th>Activity</th>
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</tr>
</thead>
<tbody>
<tr>
<td>0-20</td>
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</tr>
<tr>
<td>20-30</td>
<td>Submaximal exercise</td>
<td>HR ~150 bpm</td>
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<tr>
<td>30-32</td>
<td>Break</td>
<td></td>
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<tr>
<td>32-40</td>
<td>Maximal exercise</td>
<td>self-paced</td>
</tr>
<tr>
<td>40-50</td>
<td>Recovery</td>
<td></td>
</tr>
</tbody>
</table>

Measurement methods and materials

Experiments were carried out in a custom made climatic room (Weiss Enet, Tiel, The Netherlands). Temperature was set at 30°C with 50% relative humidity and still air, to quickly induce hyperthermia ($T_{es}>38°C$), while having sufficient possibility for evaporative cooling. Exercise was performed on a Lode Excalibur bicycle ergometer (Lode, Groningen, The Netherlands).

Gastrointestinal temperature was measured using Jonah ingestible core body temperature capsules (Philips Respironics, Mini Mitter, Bend, Oregon), which were
logged by the Equivital Life Monitor (Equivital Hidalgo Ltd, Cambridge, UK). The Equivital Life Monitor (ELM) is a wireless chest-mounted system that uses an array of sensors to assess someone’s physiological status. Heart rate data were collected by the ELM as well. Data were collected each 15 s interval and displayed and stored instantly on a notebook via Bluetooth®.

$T_{es}$ and $T_{re}$ were measured using thermistors (Yellow Springs Instruments 400 and 700 series respectively, Yellow Springs, OH, USA). Thermistors were calibrated before data acquisition in a thermal water bath (Tamson TLC-15, Tamson instruments, Bleiswijk, The Netherlands) using a certified Pt100 calibration thermometer (P650, Dostmann electronic, Germany) with resistance temperature sensor (PD-13/S, Tempcontrol, Voorburg, The Netherlands). The insertion depth of the esophageal probe was based on sitting height (18), assuring that it was located at the T8/T9 level, close to the left ventricle. The rectal probe was inserted to a depth of 10 cm beyond the anal sphincter and fixed with tape. The esophageal and rectal sensors were attached to a custom-made data acquisition system (VU University, Amsterdam, The Netherlands), consisting of a datalogger with a medical power supply and Labview software (National Instrument, Austin TX, USA). Sample frequency was set at 1 Hz. The mass of the participants was determined on a weighing scale prior to exercise (Sartorius F300S, Göttingen, Germany).

**Data analysis**

$T_{es}$ data were processed with a gating routine to remove the negative peaks due to swallowing relatively cool saliva. Then individual and group averages per 30 s were calculated for all temperature parameters, as well as individual averages per 5 min. These values have been used for statistical analysis in SPSS statistical software (SPSS 17.0, SPSS Inc, Chicago IL, USA).

ANOVA for repeated measures was used to determine significant temperature changes of all sensors in response to a phase transition. For that purpose the final 30 s of a trial phase was compared to the final 30 s of the next phase. T-tests for paired comparison, with Bonferroni adjustment for multiple comparisons, were performed to check for significant differences between $T_{pill}$ and $T_{es}/T_{re}$ at different intervals. Differences between $T_{re}$ and $T_{es}$ were additionally calculated as well.
As an indication of response time of \( T_{pill} \), \( T_{es} \) and \( T_{re} \), time for 0.1°C change from the start and end of exercise have been calculated (17). The average rate of temperature change (the average temperature change per minute for a specific interval) has been calculated for both the submaximal and maximal exercise bouts, as well as for the recovery phase. Further, peak values of the different measurement methods and maximal differences at a discrete time point have been compared by paired t-tests. Values are expressed as means ± standard deviation (SD). Calculated averages are arithmetic averages. Significance level for all tests was set at \( p<0.05 \).

**RESULTS**

For one participant, the rectal sensor failed and for another one heart rate was not recorded, so analyses involving \( T_{re} \) and HR are based on eight participants. Further, all trials had a complete dataset. No temperature drops in \( T_{pill} \) were observed when participants drank cold water prior to the experiment, confirming the 4-hour ingestion time was sufficient.

**Exercise intensity**

Average power output over the complete submaximal exercise interval was 153 ± 13 W with an average HR of 134±15 bpm and an average maximal HR of 156 ± 16 bpm. During the maximal exercise test, participants cycled on average at 247±61 W with an average HR during the complete interval of 168 ± 14 bpm and an average maximum of 181 ± 9 bpm.

**Temperature patterns during the trials**

Figure 5.1 shows the averaged temperature patterns for \( T_{es} \), \( T_{re} \) and \( T_{pill} \). Table 5.2 shows the exact temperatures at the end of each phase, as well as the total temperature change during exercise (exercise strain). Nearly all phase transitions resulted in significant temperature changes. The increase in \( T_{pill} \) during exercise was significantly smaller than \( T_{es} \), but similar to \( T_{re} \).
Figure 5.1. Average esophageal temperature ($T_{es}$), pill temperature ($T_{pill}$) and rectal temperature ($T_{re}$) patterns during the experimental trial. Submax = submaximal; max = maximal.

Table 5.2. Pill temperature ($T_{pill}$), rectal temperature ($T_{re}$) and esophageal temperature ($T_{es}$) at the end of each phase. Values are averaged over the last 30 s of rest, submaximal exercise (submax), maximal exercise (max) and recovery. In addition, the last row reports exercise strain, indicating the total temperature increase during the exercise phases.

<table>
<thead>
<tr>
<th></th>
<th>$T_{pill}$ (°C)</th>
<th>$T_{es}$ (°C)</th>
<th>$T_{re}$ (°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>End rest</td>
<td>37.11 ± 0.28</td>
<td>36.81 ± 0.32*</td>
<td>36.92 ± 0.41</td>
</tr>
<tr>
<td>End submax</td>
<td>37.27 ± 0.33†</td>
<td>37.54 ± 0.39†*</td>
<td>37.10 ± 0.39†*</td>
</tr>
<tr>
<td>End max</td>
<td>37.91 ± 0.39†</td>
<td>38.72 ± 0.60†*</td>
<td>37.76 ± 0.46†</td>
</tr>
<tr>
<td>End recovery</td>
<td>38.19 ± 0.45</td>
<td>37.62 ± 0.39†*</td>
<td>38.06 ± 0.54†</td>
</tr>
<tr>
<td>Exercise strain</td>
<td>0.80 ± 0.20</td>
<td>1.91 ± 0.63†*</td>
<td>0.75 ± 0.54†</td>
</tr>
</tbody>
</table>

†Significantly different from previous phase ($p<0.05$)

*Significantly different from $T_{pill}$ ($p<0.05$)
Differences between measurement methods

Figure 5.2 shows the temperature differences ± SD between measurement methods, averaged over each 5-min interval (for maximal exercise 4-min intervals) and over subjects. $T_{pill}$ and $T_{es}$ were significantly different at all intervals, except for the ‘sub2’ and ‘rec1’ period, when the difference was changing from positive to negative and vice versa. At the ‘max2’ period, $\Delta T_{pill}$-$T_{es}$ amounted $0.72 \pm 0.63$°C. $T_{pill}$ and $T_{re}$ did not show any significant difference except at the ‘max1’ interval ($0.15 \pm 0.15$°C). $T_{re}$ and $T_{es}$ differed significantly during the last four phases of the trial (maximal exercise and recovery).

![Graph showing temperature differences between measurement methods](image)

Figure 5.2. Averaged temperature differences between the different measurement methods - pill ($T_{pill}$), esophageal ($T_{es}$) and rectal ($T_{re}$) - during each 5-min interval of the experimental trial (for maximal exercise 4 min intervals). Sub = submaximal exercise; max = maximal exercise; rec = recovery period. *Significantly different from 0.

Over the entire trial, $T_{pill}$-$T_{es}$ and $T_{pill}$-$T_{re}$ show a mean difference of $0.04 \pm 0.52$°C and $0.14 \pm 0.26$°C respectively for all individual 5-min average values of the experimental trials. The 95% limits of agreement thus ranged from -0.97 to 1.05°C for $T_{pill}$-$T_{es}$ and from -0.38 to 0.65°C for $T_{pill}$-$T_{re}$.
Response time and rate of change

Table 5.3 displays response time and rate of change of $T_{pill}$, $T_{es}$ and $T_{re}$. $T_{es}$ responded significantly faster to the start of exercise than $T_{pill}$, increasing 0.1°C about 5 min earlier. $T_{pill}$ did not respond significantly different from $T_{re}$. After stopping exercise, it took about three minutes before $T_{es}$ had decreased 0.1°C. $T_{pill}$ and $T_{re}$ had not yet returned to end exercise temperatures after 10 min recovery in all except one participant. During submaximal exercise, as well as maximal exercise and recovery, the rate of $T_{pill}$ change did not differ from $T_{re}$, but was significantly lower than the rate of $T_{es}$ change (Table 5.3).

Table 5.3. Response time and rate of change of pill ($T_{pill}$), esophageal ($T_{es}$) and rectal ($T_{re}$) temperatures. Response time is expressed as time for 0.1°C increase and decrease from start and end of exercise respectively. Rate of change is expressed as average temperature change per minute during the submaximal exercise (sub), maximal exercise (max) and recovery (rec) phase.

<table>
<thead>
<tr>
<th></th>
<th>$T_{pill}$</th>
<th>$T_{es}$</th>
<th>$T_{re}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Response time 0.1°C increase (min)</td>
<td>8.8 ± 1.7</td>
<td>3.9 ± 2.1*</td>
<td>9.4 ± 2.6</td>
</tr>
<tr>
<td>Response time 0.1°C decrease (min)</td>
<td>&gt;10</td>
<td>3.0 ± 2.8</td>
<td>&gt;10</td>
</tr>
<tr>
<td>Rate of change sub (°C/min)</td>
<td>0.016 ± 0.013</td>
<td>0.072 ± 0.017*</td>
<td>0.018 ± 0.010</td>
</tr>
<tr>
<td>Rate of change max (°C/min)</td>
<td>0.065 ± 0.022</td>
<td>0.143 ± 0.073*</td>
<td>0.070 ± 0.045</td>
</tr>
<tr>
<td>Rate of change rec (°C/min)</td>
<td>0.028 ± 0.046</td>
<td>-0.110 ± 0.041*</td>
<td>0.029 ± 0.028</td>
</tr>
</tbody>
</table>

*Significantly different from $T_{pill}$ (p<0.05)

Peak values and delay

$T_{pill}$-$T_{es}$ peak temperature difference was -0.42 ± 0.46°C, with individual extremes to -1.21°C. Due to the delay of $T_{pill}$, the average maximal $T_{pill}$-$T_{es}$ difference at a discrete time point around the end of maximal exercise was -1.01 ± 0.66°C, with individual extremes to over -2°C. Differences between $T_{re}$ and $T_{es}$ showed a similar pattern. Peak values of $T_{pill}$ and $T_{re}$ did not show a significant difference (0.16 ± 0.31). Maximal temperature differences between $T_{pill}$ and $T_{re}$ did differ (0.47 ± 0.29), but substantially less than compared to $T_{es}$. 

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DISCUSSION

This study compared the response of gastrointestinal temperature measured by a temperature pill ($T_{pill}$) to the responses of $T_{es}$ and $T_{re}$ during rapid core temperature changes due to short high intensity exercise in the heat. Results indicate that even in these extreme conditions, $T_{pill}$ provides a reasonable estimation of $T_{re}$ and subsequently gives an indication of the temperature of vulnerable abdominal organs with a similar thermal delay (3; 19). $T_{pill}$ increasingly deviates from $T_{es}$ when body heat content changes rapidly. Responsiveness, rate of change and peak values all differ to such an extent from $T_{es}$ that any inference to $T_{es}$ is out of place.

Rest measurements

In rest, $T_{pill}$ was consistently higher than $T_{es}$, which agrees with some previous studies investigating $T_{pill}$ in rest and low intensity exercise (12; 14). This is in line with the fact that $T_{re}$ has generally been found to be slightly (0.2°C) higher than $T_{es}$ in rest as well (20; 21). The positive bias between $T_{pill}$ and $T_{re}$ in this study has also been reported before (9; 16; 22), but this finding is not consistent in literature (6). The decreasing trend in rest of all temperature profiles is probably caused by cold peripheral blood returning to the body core, as perfusion of the periphery increases on entering the warm climatic chamber due to vasodilation. After a longer rest period, temperatures would be expected to stabilize or slightly increase again.

Exercise and recovery

In agreement with previous studies applying longer exercise protocols with moderately high rates of temperature change (15; 16), $T_{pill}$ and $T_{re}$ followed the same pattern during exercise and recovery. Except for the small and practically meaningless difference at the ‘max1’ interval, there was no significant bias, although $T_{pill}$ tended to be somewhat higher than $T_{re}$. At the end of maximal exercise and during the recovery phase, individual variation in $\Delta T_{pill-Tre}$ increased. This is probably due to an individually different thermal delay of the intestinal tract. $T_{es}$ showed an increasing difference with $T_{pill}$ during exercise, up to an average bias of >0.7°C. In the course of the recovery phase, the difference increased in the opposite direction to significant and meaningful proportions. Apparently, the previously reported deviation of $T_{es}$ from $T_{pill}$ during exercise induced
core temperature changes (17; 22), is enlarged with the magnitude of the temperature change.

**Response time, rate of change and peak values**

Response time was similar for $T_{pill}$ and $T_{re}$ (~9 min), increasing 0.1°C after the start of exercise about 5 min later than $T_{es}$. Response times of $T_{es}$ and $T_{pill}$ (3.9 and 9.4 min respectively) were larger than reported by Kolka (17), who found 1.8 and 3.8 min at 80% VO$_2$max in a similar ambient temperature. However, their participants had already performed a low intensity cycling protocol before. $T_{pill}$ and $T_{re}$ reached their peak value/plateau about 5-6 min after the end of exercise. In line with Kolka (17), the rate of change in $T_{es}$ during (sub)maximal exercise was more than two times higher than $T_{pill}$ and $T_{re}$. On the contrary, the current study did not find Kolka et al.’s (17) significant difference in rate of change between $T_{pill}$ and $T_{re}$. Possibly this is due to the longer ingestion time and thus different pill location in this study, although several human studies found a similar $\Delta T_{pill}-T_{re}$ at different time points after ingestion (7; 23). Alternatively, the different exercise protocol may have affected the results.

The slower response of $T_{pill}$ compared to $T_{es}$ can first be explained physically, as it requires a great amount of energy to change the temperature of the entire intestinal tract (15; 19; 24-26). In addition, blood supply plays a role during moderate to heavy exercise. Sympathetic activity is increased and parasympathetic activity reduced, leading to vasoconstriction of the vessels in the gastrointestinal tract. There is a linear reduction in splanchnic blood flow with increasing exercise intensity from 2.8 l/min at rest to 0.5 l/min during heavy exercise (27; 28). It has been reported that maximal exercise even leads to gastric ischemia (23; 29). As the local temperature of the gastrointestinal tract depends on its blood supply, this reduction in blood flow results in a substantial delay in temperature response.

Peak values of $T_{pill}$ and $T_{re}$ did not differ, but were substantially lower than $T_{es}$, sometimes more than 1°C. On a discrete time point, measurement differences of over 2°C were even possible. So it has to be recognized that there are substantial temperature differences across the body when a high rate of heat storage or heat loss is present. Viewing the temperature patterns (Figure 5.1), $T_{pill}$ and $T_{re}$ can actually be considered as
a low pass filtered version of $T_{es}$. As exercise duration increases and/or exercise intensity decreases, temperature patterns are expected to become gradually more similar.

In view of their similar response, $T_{re}$ might be preferable to $T_{pill}$ in situations where practical and comfort motives are less important. Temperature pills require considerable ingestion time and are possibly influenced by gastrointestinal motility (17). Besides pills are more expensive than rectal probes, are for some subjects difficult to swallow and can suffer from electromagnetic interference.

**Conclusion**

Radio telemetry is a useful tool for continuous core temperature determination in an operational setting. $T_{pill}$ has been shown to reflect both $T_{re}$ and $T_{es}$ reliably when changes in core temperature are small and/or gradual and has been suggested to track $T_{re}$ better than $T_{es}$ at higher rates of change (6). This study proves that also during extreme rates of temperature change, induced by short maximal exercise in the heat, $T_{pill}$ is representative of $T_{re}$. So in those conditions, $T_{pill}$ provides a valuable indication of the thermal stress imposed on the vulnerable abdominal organs (3). Further, having quantified the extent of the deviation between $T_{pill}$ and $T_{es}$ (up to >1°C), this study confirms and strengthens the assumption that $T_{pill}$ is of no use for evaluating central blood temperature when body temperatures change rapidly. As central blood temperature is thought to approximate the temperature perfusing the brain (30), brain temperature should not be monitored by $T_{pill}$ in these conditions.

**REFERENCES**


