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Abstract
Reliable continuous core temperature measurement is of major importance for monitoring patients. The zero heat flux method (ZHF) 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. 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 oesophageal ($T_{es}$) and rectal ($T_{re}$) temperature. $\Delta T_{zhf} - T_{es}$ had an average bias ± standard deviation of $0.17 \pm 0.19$ °C in rest, $-0.05 \pm 0.18$ °C during exercise and $-0.01 \pm 0.20$ °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 $T_{es}$ when core temperature changed rapidly. 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.

Keywords: core temperature, zero heat flux, non-invasive, oesophageal, hyperthermia, thermometry

(Some figures in this article are in colour only in the electronic version)
1. 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 (Gisolfi and Mora 2000). Further, abnormal $T_{\text{core}}$ can indicate illness at an early stage and guide an appropriate action. Besides this, controlled $T_{\text{core}}$ manipulation has been used during surgery or as a therapeutic intervention in the past few years. For example, mild therapeutic hypothermia is thought to improve the outcome of cardiac arrest and ischemic insult to the brain (Bernard et al 2002, Sterz et al 2006). Therefore, reliable $T_{\text{core}}$ measurement is of major importance for monitoring 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 oesophageal temperature ($T_{\text{es}}$) (Parsons 1993). 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 (Daanen et al 2000, Latman et al 2001, Moran and Mendal 2002, Pušnik and Miklavec 2009, Yamakage and Namiki 2003). 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 is heated to deep body temperature and creating a region of zero heat flow from the body core to the skin (Fox et al 1973). In that way, it allows measuring $T_{\text{core}}$ at the skin surface. Typical body locations for ZHF sensors have a low skinfold thickness and few large veins (Brajkovic and Ducharme 2005) such as 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 (Parsons 1993, Fox et al 1973).

In previous research, reliability and measurement location of ZHF sensors varied. Fox et al (1973) 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 (1973) and Tsuji et al (1976) showed rather large deviations from rectal temperature ($T_{\text{re}}$) with a ZHF device at the sternum ($\Delta T_{\text{mean}}$: 0.54 ± 0.3 °C) and the forehead ($\Delta T_{\text{mean}}$: 0.9 ± 0.4 °C). Togawa (1979) got better results with a ZHF at the occipital region ($\Delta T_{\text{mean}}$: 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 $T_{\text{re}}$ was used as a reference instead of $T_{\text{es}}$.

In the past few decades little has been published concerning $T_{\text{core}}$ estimation by heat flux sensors, apart from the work of Gunga et al (2008, 2009) and Kimberger et al (2009). 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 $T_{\text{core}}$ mathematically by considering skin temperature ($T_{\text{sk}}$), 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 the measurement time. Whether the result is more easily affected by changing internal and external conditions has not been found yet (Opatz et al 2010). Kimberger et al (2009) reported some good clinical results with perioperative and intensive care patients with quite stable $T_{\text{core}}$ (98% of the heat flux measurement was within ±0.5 °C of $T_{\text{es}}$). However, in their last report on healthy subjects during bed-rest, Gunga et al...
(2009) 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 (LoA) of −0.72 and +0.55 °C).

Recently, Zeiner et al. (2010) tested a new prototype non-invasive continuous cerebral temperature (NICCT) sensor 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% LoA 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 recently brought up by Opatz et al. (2010), 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 this purpose we tested healthy subjects at 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% LoA within ±0.5 °C.

2. Methods

2.1. 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 on 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.

2.2. Protocol

The test procedure consisted of two sessions on separate days: one preparatory session lasting about 1 h and one experimental session lasting about 3 h. 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 oesophageal probe. This probe had to be introduced via the nose and was then swallowed by drinking water to enter the oesophagus. In the 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 ZHF sensors were attached to the forehead, giving them 20 min for stabilization. Then the oesophageal 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\(^{-1}\) body weight. The purpose was to increase a subject’s \( T_{core} \) by about 0.05 °C min\(^{-1}\), reaching an end temperature of around 38.5 °C. \( T_{core} \) was monitored every 2 min during the trial. If a subject’s \( T_{core} \) 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.

2.3. Measurement methods and materials

2.3.1. 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).

2.3.2. 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 (2010).

The ZHF system contains a prototype NICCT sensor (NICCT, Philips, Eindhoven, The Netherlands). This sensor consists of a patch ($40 \times 50 \times 5$ 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 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 on 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 on 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 the 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.

2.3.3. Oesophageal, rectal and skin temperature sensors. $T_{\text{es}}$ and $T_{\text{re}}$ 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 oesophageal 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 (Mekjavic and Rempel 1990) assuring that the oesophageal 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.

Skin temperature was determined using i-buttons (DS1922L, Maxim Integrated Products Inc., Sunnyvale, CA, USA) at eight locations, as described by ISO 9886 (ISO9886 2004). The i-button on the forehead was placed between the headband and the hairline. A weighted average of the eight i-buttons resulted in the mean skin temperature. A sample frequency of 0.1 Hz was used.

2.3.4. 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.

2.4. Data analysis

$T_{es}$ 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 skin temperature, both mean and forehead skin temperature ($T_{fh}$) were included in the analysis. $T_{fh}$ was of particular interest in comparison to the ZHF sensor which was also located on the forehead. Besides, an elevated forehead temperature is often seen as an indicator of fever.

2.4.1. 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 0. To check the relevance of this effect, root mean square (RMS) calculations were made in addition.

2.4.2. Bland–Altman diagram. To quantify the deviation between $T_{re}$–$T_{es}$ and $T_{ph}$–$T_{es}$ a Bland–Altman diagram (Bland and Altman 1986) 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% LoA for these measurements at two standard deviations (SDs) of the difference. We considered 95% LoA of less than ±0.5 °C as acceptable, as has been done in previous validation studies (Gunga et al 2009, Kimberger et al 2009, Suleman et al 2002).
2.4.3. 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.

3. Results

Seven subjects (four males and three females) 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.

3.1. Cycling intensity

Subjects started their cycling trial at an average intensity of $133 \pm 27$ W and finished it on average at $146 \pm 40$ W. One subject started her trial at 1.5 instead of 2.0 W kg$^{-1}$ body weight because she had recently gone through a period of inactivity. Subjects cycled at an average heart rate of $152 \pm 10$ beats per minute (bpm) and reached an average maximum of $174 \pm 11$ bpm. For individual values see table 1.

3.2. Absolute temperature patterns

An overview of the temperature patterns during the complete trial, averaged over seven subjects, is depicted in figure 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 $36.86 \pm 0.13$ °C and rose during exercise significantly to $38.41 \pm 0.41$ °C, so the manipulation to raise $T_{es}$ about $1.5$ °C succeeded. $T_{re}$ and $T_{zhf}$ also increased significantly during exercise, but the increase in $T_{th}$ was not significant ($p < 0.05$). $T_{es}$ and $T_{zhf}$ decreased significantly ($p < 0.05$) during the 10 min recovery phase, but $T_{re}$ and $T_{th}$ did not. Average $T_{sk}$ during rest was $35.24 \pm 0.56$ °C and increased till an average of $36.31 \pm 0.42$ °C during the last minute of exercise.

<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>

Table 1. Subject characteristics and intensity parameters. M = male; F = female, P = power output; HR avg = average heart rate; HR max = maximal heart rate.
3.3. Differences between measurement methods

Figure 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.

Table 2 gives the average temperature difference ± SD between different methods for the three main phases: rest, exercise and recovery. $T_{es}$ and $T_{re}$ are in all phases significantly
Table 2. Average temperature differences ($\Delta T_{\text{avg}}$) ± standard deviation comparing different methods during rest, exercise and recovery.

<table>
<thead>
<tr>
<th></th>
<th>$\Delta T_{\text{avg}}$ rest (°C)</th>
<th>$\Delta T_{\text{avg}}$ exercise (°C)</th>
<th>$\Delta T_{\text{avg}}$ recovery (°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T_{\text{zhf}}-T_{\text{es}}$</td>
<td>$0.17 \pm 0.19^a$</td>
<td>$-0.05 \pm 0.18$</td>
<td>$-0.01 \pm 0.20$</td>
</tr>
<tr>
<td>$T_{\text{zhf}}-T_{\text{es}}$</td>
<td>$0.25 \pm 0.20^a$</td>
<td>$-0.17 \pm 0.23^a$</td>
<td>$0.30 \pm 0.29^a$</td>
</tr>
<tr>
<td>$T_{\text{zhf}}-T_{\text{es}}$</td>
<td>$-0.09 \pm 0.31$</td>
<td>$0.12 \pm 0.24^a$</td>
<td>$-0.31 \pm 0.32^a$</td>
</tr>
<tr>
<td>$T_{\text{zhf}}-T_{\text{eh}}$</td>
<td>$0.97 \pm 0.58^a$</td>
<td>$0.73 \pm 0.42^a$</td>
<td>$1.06 \pm 0.60^a$</td>
</tr>
</tbody>
</table>

$^a$ Significant difference ($p < 0.05$).

Figure 3. Bland–Altman diagrams showing the differences of (A) the zero heat flux and oesophageal temperatures ($\Delta T_{\text{zhf}-T_{\text{es}}}$) and (B) the rectal and oesophageal temperatures ($\Delta T_{\text{re}-T_{\text{es}}}$). Both figures consist of all individual 5 min values for the rest phase (circles), the exercise phase (squares) and the recovery phase (triangles).

different, while $T_{\text{es}}$ and $T_{\text{zhf}}$ only differ during the rest phase ($p < 0.05$). During the whole trial there is a large difference and a large variation between $T_{\text{zhf}}$ and $T_{\text{es}}$, while measuring temperature at the same location. The RMS values gave a similar impression and have therefore been omitted.

In figure 3, Bland–Altman diagrams for $T_{\text{zhf}}-T_{\text{es}}$ and, for comparison, $T_{\text{re}}-T_{\text{es}}$ are depicted. The graphs show a mean difference of $0.00 \pm 0.20$ °C and $0.01 \pm 0.32$ °C respectively for all individual 5 min average values of the experimental trials. The 95% LoA thus ranged from $-0.40$ to $0.40$ for $T_{\text{zhf}}-T_{\text{es}}$ and from $-0.62$ to $0.64$ for $T_{\text{re}}-T_{\text{es}}$.

3.4. Delay time

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

4. Discussion

The ZHF sensor used in this study gave a reliable estimation of $T_{\text{es}}$ during both stable and changing body core temperature in a hot windless environment. The 95% LoA of
±0.40 °C are well within our acceptable level of agreement of ±0.5 °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 $T_{core}$ determination, in hospital, laboratories and at home, it is interesting to compare the results of $T_{es}$ and $T_{zhf}$. At rest, $T_{zhf}$ and $T_{es}$ performed quite similar. In line with previous studies, $T_{es}$ in rest was slightly higher than $T_{ez}$ (Daanen 2006, Houdas and Ring 1982) and so did $T_{zhf}$. During exercise and recovery $T_{zhf}$ tracked $T_{es}$ in all intervals better and faster than $T_{es}$. The rectum is often used for $T_{core}$ measurement because it consists of a large mass of deep body tissue and is not affected by environmental conditions (Easton et al 2007, Moran and Mendal 2002, Strydom et al 1965). It can be a useful measure, as it reflects the local temperature in the vulnerable abdominal cavity (Gagnon et al 2010). 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 (Easton et al 2007, Gagnon et al 2010, Molnar and Read 1974, Moran and Mendal 2002, Proulx 2008). As a result, $T_{es}$ is unreliable for monitoring quickly changing central blood temperature, reflected by $T_{es}$. This became most obvious in the recovery phase when individual differences between $T_{es}$ and $T_{ez}$ 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 $T_{es}$ had hardly started its decrease when measurements stopped. At that moment the gap between $T_{es}/T_{zhf}$ and $T_{ez}$ was still widening as can be seen in figure 1. This seems to support the suggestion that the time lag of $T_{es}$ may rise up to 20 min, depending on environmental, physical situation etc (Daanen et al 2000). Especially in emergency situations where $T_{core}$ may be fluctuating fast, this can have serious consequences. Proulx (2008) already showed that it can lead to serious hypothermia during cooling procedures after heat stress. In this respect, $T_{zhf}$ seems to be a promising alternative. In contrast to the statement of Yamakage and Namiki (2003), $T_{zhf}$ was still reliable at rates of change above 0.3 °C min⁻¹.

Figure 3 shows that the deviation of $T_{zhf}$ does not depend on the trial phase, while the deviations of $T_{es}$ are clearly grouped per trial phase, mostly as a result of the discussed delay. Nevertheless, in rest there was a significant difference between $T_{zhf}$ and $T_{es}$, 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 LoA, the current ZHF sensor seems to perform better than previous (zero) heat flux systems (Ball et al 1973, Gunga et al 2009, 2008, Togawa 1979, Togawa et al 1976, Tsuji et al 1976). Previous studies are hard to compare though, as most studies measured under different ambient conditions and at different locations. Most comparable was the setup of Gunga et al (2008), as they estimated $T_{core}$ with a heat flux device on the forehead during rest and exercise in 10, 25 and 40 °C. However, this was not a zero heat flux device and $T_{es}$ was used as reference. Our experiment shows that $T_{es}$ has an inconsistent deviation from $T_{es}$ when $T_{core}$ is not stable. This may have caused the larger LoA in their study, also in the warm and thus more comparable ambient temperature conditions ($-0.08 \pm 0.35$ and $-0.01 \pm 0.37$ °C for work and rest in 25 °C; $-0.11 \pm 0.34$ and $0.10 \pm 0.42$ °C for work and rest in 40 °C (Gunga et al 2008)). 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 $ΔT$ (0.40 °C), the LoA depicted in figure 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 (2003) 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 (2005) 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 (2010), who established LoA 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$^{-1}$) perpendicular to the sensor seems to double the SD of $ΔT_{zhf}-T_{es}$.

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 such as rectal, tympanic and oral are used in clinical. The latter two are known to be unreliable as they can easily be affected by external factors (Daanen 2006, Pušnik and Drnovsek 2005, Pušnik et al 2004, Tandberg and Sklar 1983, Terndrup et al 1989). $T_{es}$ has been shown to deviate from $T_{es}$ during $T_{core}$ changes (Moran and Mendal 2002, Proulx 2008) 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.

In conclusion, 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.

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