CONTINUOUS CORE BODY TEMPERATURE ESTIMATION VIA SURFACE TEMPERATURE MEASUREMENTS USING WEARABLE SENSORS: IS IT FEASIBLE?

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Abstract: Core body temperature is an important indicator of well being of humans. The commonly used methods and sites of measurement do not lend well for continuous measurement at home. However, continuous monitoring using non-invasive, small, low cost sensors could have many applications like detection of hypothermia and fever in low birth weight neonates in rural settings. We investigate the feasibility of making such measurements using only skin temperature sensors. Our initial sensor prototype is composed of flexible materials, with embedded high precision thermistors and is based on dual heat flux technique. Our initial experiments show that the reliable estimation of core temperature under varying ambient conditions and at different measurement sites is a challenge, but promising. Further work is needed to combine results from experimental measurements and theoretical simulations to provide good insights and enable accurate estimation of core body temperature for long term monitoring at home.

1 INTRODUCTION

Even though pulmonary artery temperature is considered to be the gold standard for core temperature measurement, it is not used for general clinical practice. Rectal temperature is close to this but poses severe constraints due to invasiveness and concerns about hygiene even though continuous monitoring probes are available. Digital thermometers for oral and axillary temperature measurements have replaced the traditional mercury in glass thermometers and provide relatively quicker and easier measurements. However, these are meant for manual use and do not suit continuous monitoring needs. Recently, tympanic thermometers based on infrared measurements have become popular as they are fast and easy to use. However, different studies show different results regarding their accuracy in comparison to the pulmonary artery temperature (Hooper & Andrews 2006). There have been some attempts to integrate it into a head gear similar to headphones (Boano, Lasagni & Roemer 2013). Other techniques like radiometric sensing (Bonds, Gerig, Weller & Herzig 2012) have been evaluated. Conductive textiles have been used for monitoring temperature of neonates in Neonatal Intensive Care Units (Chen, Dols, Oetomo & Feijs 2010). However, none of these are suitable for continuous temperature monitoring of neonates at home due to inconvenience in use, complexity of the device or power requirements. Skin temperature is the easiest to measure in a non-invasive manner using simple sensors. However, single point measurements of skin temperature vary significantly with the ambient conditions. As the thermoregulation of neonates is not well developed, using fixed offsets to skin temperature would add to the errors. Hence there is a need for reliable estimation of core temperature from skin temperature measurement to develop low cost sensors that are accurate and reliable for remote monitoring applications.

2 CORE BODY TEMPERATURE ESTIMATION FROM BODY SURFACE MEASUREMENTS

One of the earliest efforts to develop a sensor for core body temperature measurement from surface measurements has been based on zero heat flow method (Fox, Solman, Isaacs & MacDonald 1973). The method uses a heater to create a zone of zero heat flux such that the skin temperature under the sensor reaches the core temperature. Studies have shown the effectiveness of this method except during exceptional rapid cooling or heating as the response time is around 15 to 20 minutes when applied to forehead (Togawa 1985). There have been other attempts to evaluate the effectiveness of this scheme for different applications (Zeiner et.al. 2010) (Teunissen et.al. 2011). However, the use of a heating element creates problems for applications which involve mobility and require low power consumption. New technique which does not involve the use of heater and is based on a double sensor has been reported for monitoring heat strains (Gunga et.al. 2008) and also for space applications (Gunga et.al. 2009). The design uses two temperature sensors separated by an insulating layer whose thermal conductivity is known. By measuring the skin temperature and the temperature at the upper sensor and knowing the thermal conductivity of human tissue, the core temperature can be calculated. However, this technique still requires the correct knowledge of the thermal conductivity of the epidermal tissues where the sensor is placed. Another solution has been proposed using two heat flow channels in parallel (Kitamura, Zhu, Chen & Nemoto 2010). This work avoids the dependence on knowledge of the thermal properties of tissue below the sensor. Its performance has been compared to the zero heat flux method. Further work has been presented to improve the response time (Sim, Lee, Baek & Park 2012). These present interesting ideas for use in continuous monitoring applications. However, additional constraints on materials of the device to conform to body contour have to be added for it to be acceptable for long term use on neonates. Also, correspondence between experimental measurements and theoretical correctness and assumptions has to be established. We explore these aspects by considering sensors for neonates. We study the accuracy under varying conditions and at different sites of measurement using both simulations and experiments.

2.1 The Device Structure

For the device to be easily acceptable to parents for long term use, it has to be such that it does not leave any mark on the neonate’s skin. Also, it cannot be taped to the skin as regular application and removal of tapes create redness. Using a hard metal contact at the skin interface or an inflexible material for the sensor cover, would create problems. Hence using a soft material that bends to conform to the body contour and also ensure that the sensor is in contact with the skin is a good option. PolyDimethylsiloxane (PDMS) can be used as the material for the device. PDMS provides both flexibility and biocompatibility. Hence it can be placed on the delicate skin of neonates over long periods of time. Another major advantage provided by PDMS is that it is highly hydrophobic. This makes it very good for usage where it is likely to be exposed to water or urine.

We use Dow Corning’s Sylgard 184 elastomer kit for the purpose. Two different ratios of base and curing agent have been used for two different layers of the sensor (shown in Fig 1). For the layer that is contact with the skin, base elastomer to curing agent ratio of 10:0.4 is used while and for the top layer, a ratio of 10:1 is used. The top layer is prepared first and the lower layer is added after partial curing of the first layer. This ensures that the two layers bind together. By following this method, the need for complex surface modification techniques like use of oxygen plasma to improve adhesion between the layers is avoided. The lower layer provides adhesion to the skin surface and the upper layer provides the rigidity needed for the sensor. Also, the lower layer is softer and does not leave any mark when placed on the skin. Experiments were performed with different ratios of base and curing agent. It was observed that a ratio of 10:0.2 provides better adhesion over a longer time. But the layer is too soft and can get easily damaged. The ratio of 10:0.4 provides adhesion while being strong enough for long term use. However, the adhesion of the sensor is not enough to hold it for a long time over prolonged removal and placement cycles especially in the presence of oil or dirt on the skin surface. Hence a belt has been provided on the side of sensor. In the presence of a belt, the adhesion is enough to ensure contact of the sensor with the skin to get reliable measurements with low pressure from the belt on the skin. Instead of using IC based temperature sensors as used in prior work, we use high precision thermistors (MF51E103E3950). Since the device can be made in very controlled manner with respect to composition and size of moulds, the thermal properties can be ensured for every device and extensive calibrations for each sensor would not be needed after it is in final form. The device also has an integrated SOC (BLE112) from Bluegiga Technologies for data acquisition and Bluetooth® low energy communication with a gateway device.

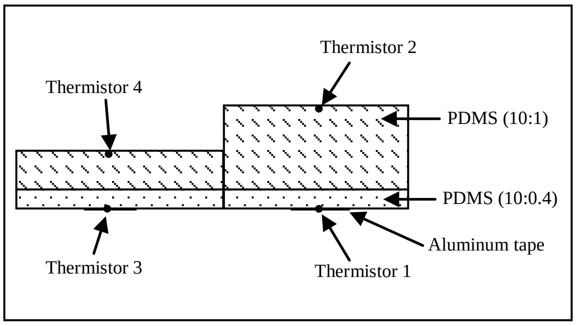

Fig 1: Device Structure

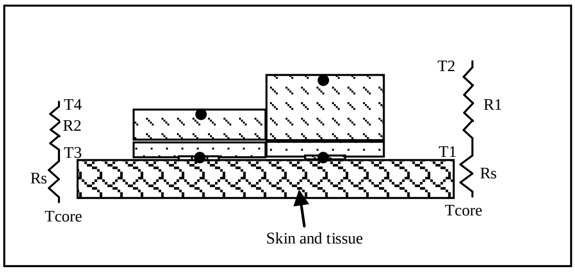

Fig 2: Schematic of device placed on skin and analogous equivalent circuit with two parallel heat flow channels and thermal resistances network

Following the concept as proposed by Kitamura et.al., heat flow in the two channels can be modelled using an equivalent thermal resistance network with two parallel heat flow paths where R1 and R2 are the respective thermal resistance of the two channels and Rs is the thermal resistance of skin and subcutaneous tissue.

$$\frac{Tcore - T1}{Rs} = \frac{T1 - T2}{R1} \qquad \text{-- (1)}$$

$$\frac{Tcore - T3}{Rs} = \frac{T3 - T4}{R2} \qquad \text{-- (2)}$$

When the temperatures T1 and T3 (as shown in Fig. 2) are measured close to each other as is the case here, the core temperature can be calculated as

$$T_{core} = T1 + \frac{(T1-T2)*(T3-T1)}{(T1-T2)-(T3-T4)*K} \qquad \text{-- (3)}$$

where K = R1/R2. Hence the dependence on knowledge of thermal resistivity of skin and subcutaneous tissue can be eliminated. The equation in the published work by Kitamura and used in some of the work that used the concept has slight error in the denominator term. We have used a corrected equation (equation 3). K can be obtained experimentally by applying a fixed temperature for Tcore and measuring T1, T2, T3, and T4 as

$$K = \frac{(Tcore-T3)*(T1-T2)}{(Tcore-T1)*(T3-T4)} \qquad \text{-- (4)}$$

2.2 Experimental setup and measurements

To verify the concept, experiments have to be performed to simulate different core temperatures. To simulate the effect of skin and human tissues experiments were tried using different phantoms. Phantoms made from gelatine (obtained from porcine cells) as well as poly vinyl alcohol were tried. However, when these were exposed to continuous heating especially above 36 degrees for long durations (30-40minutes), the physical properties started changing due to loss of water content. This made it difficult to conduct the experiments. Hence PDMS (10:1) was used instead as it has thermal conductivity of 0.15 W/m K which is very close to that of skin and fat. The sensor was placed on a PDMS layer of thickness 5mm which in turn was placed on a well controlled hot plate to simulate the core temperature.

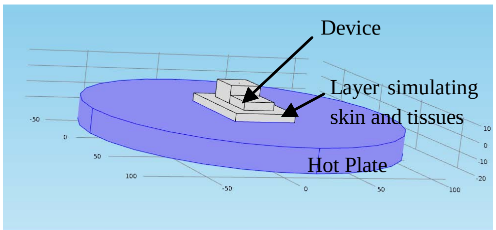

Fig 3: Experimental Setup

Three temperature points of 36°C, 36.5°C and 37.5°C were used to cover the normal body temperature range. Based on the measurements as shown in Fig 4, the value of K was obtained as 3.5 for all three temperature points.

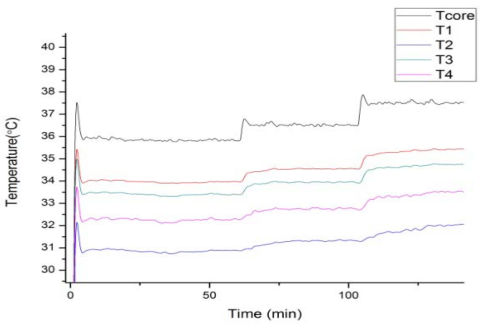

Fig 4: Measured Temperatures during experiment

Calculation of Tcore is very sensitive to the temperature difference T3-T1. Hence, it justifies the use of high precision thermistors instead of IC based temperature sensors as used in earlier work.

2.3 Heat Transfer Simulation

A three dimensional sensor model was created using COMSOL™ as shown in Fig. 3 and heat transfer simulation was performed to verify the feasibility of using the concept. In the model, the sensor was composed of PDMS blocks with four domain point probes placed at the bottom and top surfaces similar to that in fig 1. The simulation was performed by placing it on another PDMS layer similar to the experimental setup. The bottom boundary of the lower PDMS layer was assigned the temperature equivalent to Tcore. Radiation and convection was specified at the top surface. The convective heat transfer coefficient was kept very low to have simulation equivalent to still air in the room. The parameters were specified as follows and ambient temperature was specified as 25°C which was same for experimental measurements.

Parameter	**Value**
Thermal Conductivity (k)	0.15 W/(m°K)
Density (ρ)	0.97 Kg/m^3
Heat Capacity (Cp)	1460 J/(Kg°K)
Surface Emissivity (ε)	0.9
Convective heat transfer coefficient (h)	2.5 W/(m^2K)

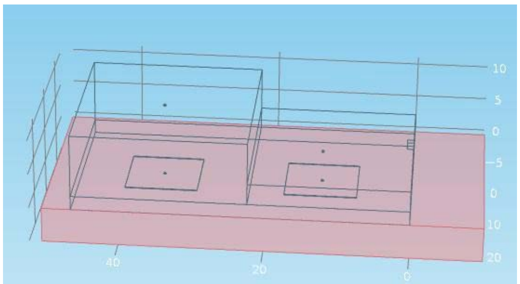

Fig 5: COMSOL model

3 RESULTS AND DISCUSSION

Using the value of K as 3.5 obtained in the experiment, the applied core temperature was estimated. The mean error was less than 0.2°C for the experimental temperature range. The results are shown in Fig 4. Intentionally, smoothing is not performed to show that there are minor fluctuations in temperature as the core temperature itself has fluctuations due to the involved PID control of the hot plate. The response of the sensor is quite fast to step changes in temperature and hence it can track varying temperatures with a delay of less than 5 minutes even for step changes.

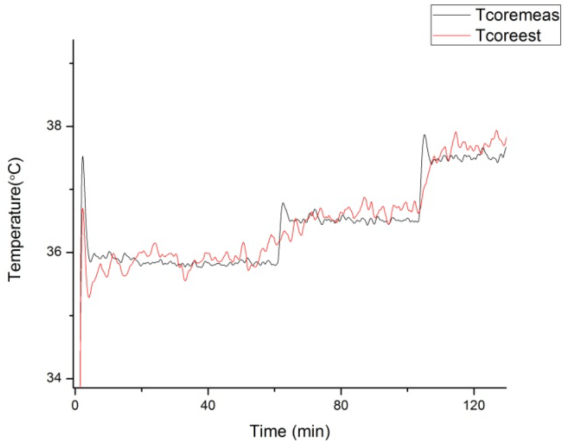

Fig 6. Comparison of applied Tcore and Estimated Tcore

The sensor was placed on the forehead on four different days and timings to measure the performance. It takes approximately 20-30minutes for the temperature to stabilise. This is similar to the time mentioned in earlier work. Hence, the choice of PDMS as the material does not increase the measurement time.

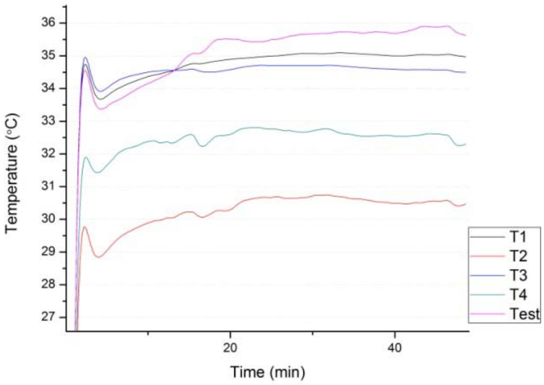

Fig 7: Measurement on forehead

It is important to compare the measured temperature to other standard sites of measurement used in regular clinical use. Tympanic temperature was measured using an infrared tympanic thermometer (Omron MC510) and the oral and axillary temperatures were measured using a digital thermometer (Neclife). The results show that the obtained temperatures using dual flux method is usually lower compared to all three measurements.

However, the variation is consistent across different temperatures. However, concrete conclusions cannot be derived based on these measurements as no temperatures in the hypothermic and fever ranges were measured.

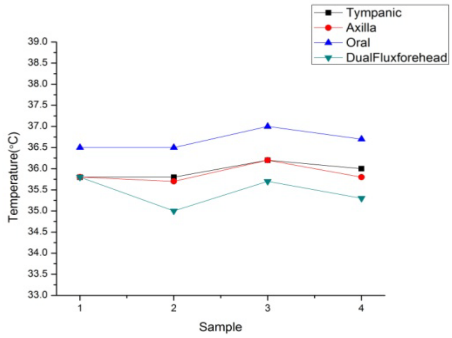

Fig 8: Comparison of temperature measurements using different methods and sites of measurement

Comparison of experimental results and theoretical simulations can help in improving the design of the sensor. Hence, a comparison of COMSOL simulation results and experimental measurements for the different temperatures measured within the sensor is presented in Fig 9. T2 and T4 match very closely for all three points. However, T1 and T3 are higher is simulation than in experiment. One of the reasons could be the contact resistance between the hot plate and the lower PDMS layer.

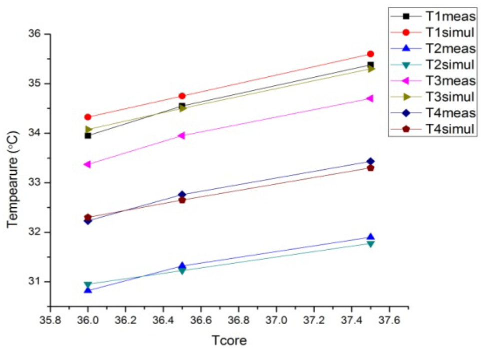

Fig 9: Comparison of measured temperatures within the device and COMSOL simulations

The temperature at different boundaries can be seen in the simulation results in Fig 11. During simulation, it can be easily noticed that the presence of a metallic contact (aluminium foil in this case) at the bottom layer impacts the absolute value of temperatures even though it does not significantly affect the difference in temperatures. Also, there is significant impact of selection of convection boundaries.

Since simulation assumes, perfect contact, this can impact the values. The K value obtained in simulation 2 as expected. In simulation this value does not vary significantly over ambient temperatures ranging from 15°C to 32°C as seen in fig. 10. However, as the ambient temperature increases, T1-T3 becomes very small and hence in experimental measurements, inaccuracies in the thermistor calibration can have a significant impact on the estimated temperature.

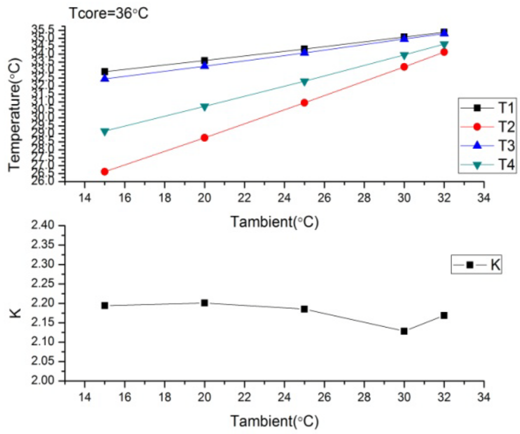

Fig 10: Temperatures T1, T2, T3, T4 and the values of K with varying ambient temperatures

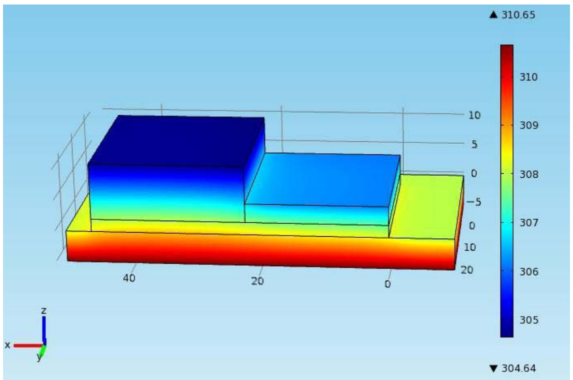

Fig 11: Temperature at different boundaries in the COMSOL simulation

4 CONCLUSIONS AND FUTURE WORK

The choice of PDMS as the material and a simple geometry does not adversely affect the sensor characteristics compared to results mentioned in earlier work. However, it is required to understand in details the reasons for differences in experimental values and theoretical simulations including the possibility to overcome the effects of contact resistance which is dependent on contact resistance without affecting the flexibility of the sensor. Also, the effect of ambient variations and convection are important for accuracy of the sensor. A much more detailed model is required to compensate for these as compared to the simple thermal resistance network model. It might be useful to use dynamic models to infer the change in ambient conditions and, if possible, the context of measurements to reduce the possibility of false alarms. Another important aspect that has not been studied so far is the use at a measurement site other than the forehead. If the sensor can be used at a location between the abdomen and chest, it can be coupled with other vital parameter measurements on a single sensing device. However, the layer of adipose tissues is significantly lesser for neonates and hence data obtained on adults and neonates could vary.

Another very important aspect for future work is to develop a good model incorporating multiple tissue layers and blood perfusion using Penne's Bio-heat equations instead of simple heat transfer models. It could be interesting if some of the parameters can be learnt from data obtained through continuous measurements and be able to predict conditions like increased blood perfusion due to exposure to cold and compensate for differences in core temperature measurements. It would also help in validating the assumption that the effective thermal resistance of tissue below the sensor is same for both parallel heat flow channels.

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