ABSTRACT

NARAYAN, PRAVEEN. Core Body Temperature Sensors - A Review. (Under the direction of Dr. Michael Daniele).

Core body temperature (CBT) is a vital physiological parameter that reflects the true health status of an individual. Monitoring of core body temperature is of utmost importance across various domains, most crucially in clinical practice. Temperature monitoring is typically carried out by measuring skin temperature due to the convenience of these methods. These methods however, suffer from a lack of accuracy. On the other hand, conventional CBT measurement methods have good accuracy but are too invasive and almost exclusively carried out only in critical care units. There is thus a necessity for the development of highly accurate and comfortable CBT sensors for easy and precise measurements both for use clinically as well as in daily living. This survey thesis explores the importance of core body temperature sensing and its various applications. It then looks at how skin surface temperature is measured and the important requirements for temperature sensing. Conventional CBT measurement methods are explained and the theory behind the measurement techniques are discussed in detail. Many interesting research studies and experiments using each of these techniques are also analyzed. The tools to engineer a wearable core body temperature sensor are explored including commercial IC chips, important design considerations as well as many low power techniques. Important research studies involving different CBT prediction methods have been summarized and the operational parameters and specifications of these systems have been reported. Finally, the state of the art in wearable core body temperature sensing has been assessed and reviewed in this thesis, and some ideas for future directions for research and development in this field have been suggested.

Core Body Temperature Sensors - A Review.

By
Praveen Narayan

A thesis submitted to the Graduate Faculty of
North Carolina State University
in partial fulfillment of the
requirements for the degree of
Master of Science

Electrical Engineering

Raleigh, North Carolina
2024

APPROVED BY:

Dr. Michael A. Daniele
Committee Chair

Dr. Alper Bozkurt

Dr. Edgar Lobaton

DEDICATION

I dedicate this work to my family, close relatives and friends.

BIOGRAPHY

Praveen Narayan was born May 8th, 1996, in Fairfax, Virginia. At the age of 7, he and his family moved to Chennai, India where he did his middle and high school. He pursued a Bachelor of Engineering in Electronics and Communication Engineering at SSN College of Engineering (Anna University). During this time, he got an opportunity to intern at the Council of Scientific and Industrial Research (CSIR). His final capstone project was the development of Smart Windshield Wipers based on ANFIS. With a genuine interest in the Biomedical field and in Embedded systems, Praveen decided to pursue his Master of Science degree in Electrical Engineering with a specialization in Bioelectronics at North Carolina State University, Raleigh. He currently works at Medtronic as an R&D Electrical Engineer.

ACKNOWLEDGEMENTS

I would like to thank my advisor Dr. Michael Daniele for his support, insights and accommodation and for helping me complete this thesis fully even as I grappled with several health and financial issues. I would also like to express my gratitude to Dr. Bozkurt and Dr. Lobaton for being a part of my committee and for their feedback.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

1. INTRODUCTION

1.1 SIGNIFICANCE OF CORE BODY TEMPERATURE

Core body temperature is the temperature of the internal organs that are located deep within the body (such as heart, liver, brain, and blood). It is maintained within a very narrow range from 36.5˚C - 38.5˚C (97.7- 101.3 Fahrenheit) [1]. It is within this range that the body's metabolic activities, enzymatic reactions, and cellular functions required for survival take place most optimally [2]. Core body temperature also fluctuates following illnesses, physical activity, circadian cycles, and other physiological processes. It also varies due to several other biological and environmental factors such as time of day, age, gender, specific measurement site, etc. [2]. The process of thermoregulation is responsible for keeping the body's core internal temperature within the normal range. Thermoregulation is also responsible for the fighting of infections and diseases through its various defense mechanisms [2,3]. There are relations between thermoregulation, thermal management and anesthesia in the context of intensive care medicine [3]. The hypothalamus of the brain controls thermoregulation and sends signals to various parts of the body in the event of temperature changes taking place. It prevents hyperthermia (significant core temperature elevation; core temperature > 40.5˚C) and hypothermia (significant core temperature depression; core temperature < 35˚C) [2]. Physiologically, hyperthermia leads to excessive sweating, lightheadedness, fatigue, headache, tachycardia and many other symptoms and if left untreated could lead to severe agitation, nausea, muscle cramps, confusion, seizures and coma [2]. Hypothermia on the other hand is considered to be much more dangerous, with deaths being twice as frequent when compared to hyperthermia [2]. Some of the physiological impacts include respiratory distress, cardiac arrhythmias, shivering, muscle dysfunction and other symptoms [2]. If the core temperature is not brought back within normothermic range, it could lead to cardiac arrest or coma. The effects of core blood temperature on perspiration, shivering and vasoconstriction/vasodilation are also being actively researched [3]. Core body temperature also plays a major role in sleep regulation [4]. S. Haghayegh et al [5] describes Borbely's 2 component model for sleep regulation in their research study. The model consists of a homeostatic process S and a circadian process

C. The homeostatic process S is a measure of 'sleep pressure', whereby it gradually increases during the day/during the period of wakefulness, and decreases in the same manner during sleep. It is thus at its maximum right before sleep onset and is at its lowest at the end of sleep. Hence it plays a role in setting the stage for sleeping and awakening [5]. The circadian process C also determines the states of sleep onset and sleep offset. Core body temperature plays a key role in this process C. According to the research, core body temperature is at its lowest point during the period of deepest sleep, rises slightly right before wakening (sleep offset), gradually rises throughout the day (apart from a slight dip mid-day), peaks about 4 hours before sleep onset, and declines substantially during the onset of sleep [5]. As core body temperature decreases, it was found that sleep pressure (which is a measure of process S) starts to peak. The rise in core body temperature coincided with decreases in sleep pressure. Melatonin synthesis is also an aspect of process C. The combination of these 2 processes C and S are overall responsible for the 24hr transition between sleeping and wakefulness [5]. Thus, core body temperature may be seen as a biomarker of process C, and as an extension of this, is directly responsible for the sleep-wake cycle. Peripheral/skin temperature is a measure of the heat on the outermost surface of the body [6]. The skin makes up about 15% of a human's body weight, and there are thus regional variations in skin temperature even during stable ambient conditions. Elevated or decreased skin surface temperatures may be indicative of an underlying illness or abnormality within the body [6]. This temperature is also affected by a number of external environmental factors such as ambient temperature and humidity. Changes in peripheral temperature may be a reflection of the core temperature changes. While peripheral temperature variations over a broad range of temperatures are tolerated, even small deviations in core temperature have strong physical and mental effects on the body [7]. For the development of thermal management practices in intensive care/operative settings, it is important that efficient core temperature measurement methods are developed and introduced. This would lead to improvement in clinical patient outcomes. Thus, core body temperature and its continuous monitoring is vital for determining the health status of an individual.

1.2 APPLICATIONS

Some of the applications for continuous monitoring of core body temperature include:

1.2.1 MONITORING OF INFANTS/PRETERM INFANTS IN HOSPITALS

Monitoring of vital signs in neonates has become increasingly noninvasive to prevent possible perturbations as well as to avoid disruption of the sleep-wake cycle [8]. While noninvasive systems for the detection of heart rate, respiratory rate, blood pressure, oxygen saturation have now become commonplace in intensive care units, measurement of core body temperature still remains less accessible and predominantly invasive. Core temperature measurements for preterm neonates are of utmost importance because they require thermal life support using external devices such as convective incubators and radiant heaters. Such measurements help in the monitoring of thermoregulation and prevent medical negligence or mishaps [9].

1.2.2 MONITORING OF PATIENTS IN CRITICAL CARE HEALTH UNITS

This is important to enable early deterioration detection and screening. Core temperatures outside the normal range could have severe effects on biochemical and cellular functions [10]. This could lead to systemic inflammatory responses, eventually progressing to multi-organ failure. There is often a high frequency of temperature abnormalities among the critically ill, making it all the more important to monitor and regulate their core body temperatures [10].

1.2.3 MONITORING OF ATHLETES

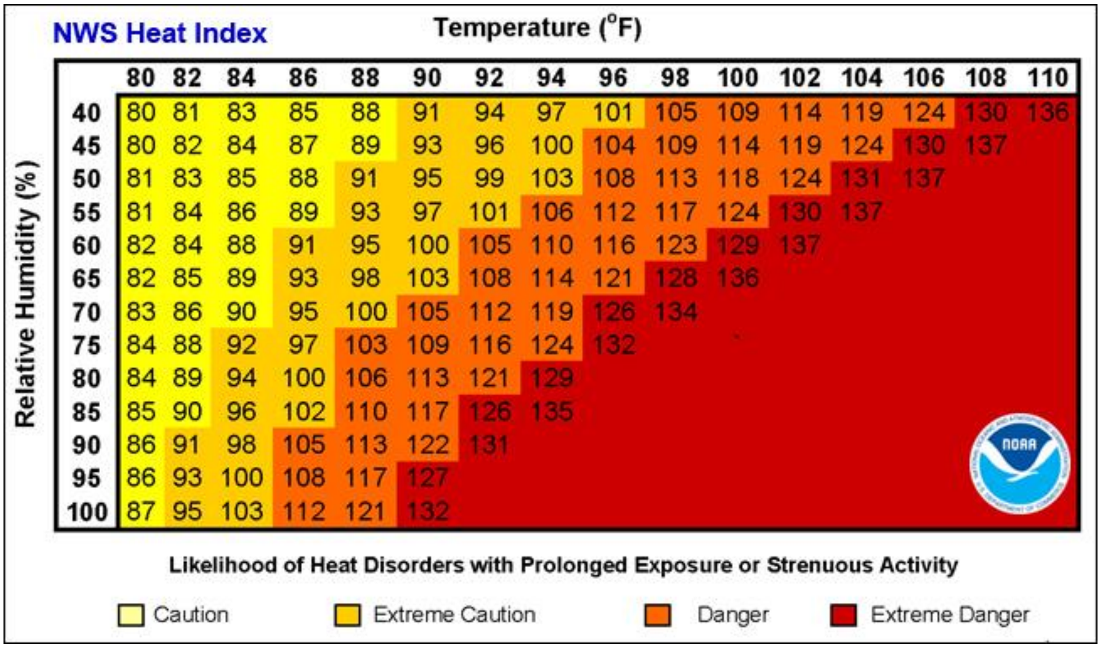
This is necessary to help improve performance, reduce injuries, and prevent dangerous physical conditions. Exercising results in an elevation of core body temperature and studies [11] have indicated that a steep rise in temperature decreases performance and power output, with more blood being directed to the skin for cooling. It also puts the athlete at risk of heat stroke/stress. A constant monitoring of core body temperature may help athletes make better decisions with respect to training, cooling down, and planning for racing events [11].

1.2.4 USAGE IN THE EARLY DIAGNOSIS OF NEURODEGENERATIVE DISEASES AND OTHER HEALTH/SLEEP ISSUES

Diseases may leave traces in a person's circadian cycle. One simple way to monitor the circadian cycle is to measure core body temperature [12]. Changes in the circadian cycle could point to many health issues. Physiologically, the circadian rhythm of core body temperature is due to the heat gain and heat loss mechanisms by the Suprachiasmatic nuclei (SCN) in the hypothalamus. Q. Huang et al [4] monitored the human circadian temperature dynamics during the daily routine of subjects to better understand the circadian rhythm and help integrate chronotherapy concepts in daily medicine. Chronotherapy is the science of optimization based on timing to maximize an intended therapeutic effect [13]. For example, in the concept of disease/cancer treatment, it involves administering drugs at different times of the circadian cycle to target receptor/gene expression and thereby maximize the therapeutic effect and at the same time minimize potential side effects [13]. In sleep science and in the treatment of insomnia or disorders such as Delayed sleep phase disorder (DSPD), it involves moving a patient’s bedtime to a later time each day, until the patient is at a normal sleeping schedule [14]. In this study [4], the temperatures of subjects were measured using both a chest surface body temperature wearable as well as a core body temperature sensor wearable. It was hoped that real time information of the circadian rhythms of patients could be transmitted to hospitals to enable them in the early diagnosis of diseases as well as to monitor any improvements or deterioration in patients [4]. The Core body temperature was thus used as a circadian biomarker. Active research is being carried out involving core body temperature sensing to investigate Parkinson’s, Alzheimer’s, insomnia, narcolepsy, and jet lag [15].

1.2.5 PHYSIOLOGICAL MONITORING OF CONSTRUCTION WORKERS AND PEOPLE WORKING IN HIGH-TEMPERATURE/HIGH-HUMIDITY ENVIRONMENTS

Workers in such industries are often exposed to extreme levels of heat for prolonged periods of time. The industries often rely on indirect risk identification methods to ensure safety such as the creation of safety protocols and the monitoring of environmental conditions in the work vicinity. However, such methods do not account for individual worker heat tolerance and adaptation, and thus a system involving the continuous measurement of every individual worker's core temperature can help in increasing safety and reducing costs [16]. Shown below in figure 1 [17] is the heat index issued by the CDC to assess the risk of developing heat disorders based on ambient temperature and humidity.



	80	82	84	86	88	90	92	94	96	98	100	102	104	106	108	110
40	80	81	83	85	88	91	94	97	101	105	109	114	119	124	130	136
45	80	82	84	87	89	93	96	100	104	109	114	119	124	130	137	
50	81	83	85	88	91	95	99	103	108	113	118	124	131	137		
55	81	84	86	89	93	97	101	106	112	117	124	130	137			
60	82	84	88	91	95	100	105	110	116	123	129	137				
65	82	85	89	93	98	103	108	114	121	128	136					
70	83	86	90	95	100	105	112	119	126	134						
75	84	88	92	97	103	109	116	124	132							
80	84	89	94	100	106	113	121	129								
85	85	90	96	102	110	117	126	135								
90	86	91	98	105	113	122	131									
95	86	93	100	108	117	127										
100	87	95	103	112	121	132										

Figure 1: Heat index issued by the CDC to assess the risk of developing heat disorders based on ambient temperature and humidity [17].

1.3 PERIPHERAL BODY TEMPERATURE MEASUREMENTS

As briefly mentioned previously, the peripheral temperature is a measure of the heat on the outermost surface of the body [1]. Body temperature is commonly measured on the surface of the skin (periphery) at various sites. This temperature differs based on the site of measurement, due to the vast surface volume of the skin. It is also heavily influenced by exposure to the external environment, activities of the nervous system as well as changes in blood flow below the skin [6]. The local skin temperature could also be used as an indicator of how much a person sweats, and thereby their fat percentage/body weight. A research study was conducted by M. Nakayoshi et al [18] to further understand the processes of human thermal physiology in an outdoor summer setting. Among many observations, one was that the subjects who had a greater body weight/size tended to sweat more. And those who sweated more had lower skin temperatures compared to those who sweated less. This was also a reflection of the difference in the evaporative cooling effect by the sweat. Observing the patterns of changes in the skin temperature could also be useful to establish a personal baseline, and this could in turn help indicate any abnormalities within the body. Several modern-day health/fitness trackers, wearables, smartwatches and bands carry this out, and provide actionable insights to the wearer. However, such insights cannot be used as a substitute for clinical diagnosis. For example, most fevers elevate body temperature, which is reflected at the surface of the skin. But all elevations in temperature observed at the skin need not be indicative of a fever. The most common sites of measurement include under the arm (axillary temperature), within the mouth (oral temperature), and skin on the forehead (temporal temperature) [19].

1.3.1 AXILLARY MEASUREMENTS

Axillary measurements are preferred due to their ease of access and the levels of comfort they provide to the patient during temperature measurements. They are still one of the most common methods used worldwide [20]. However, these measurements are heavily influenced by external environmental temperatures as well as sweating and perspiration and typically take some time to give reliable/stable readings. Shorter measurement times using this method may result in readings with a higher degree of error [20]. The effects of sweating and perspiration cause the skin temperature to be lower than the core body temperature. In the case of neonates, these measurements require supervision to prevent displacements leading to poor cost effectiveness with respect to nursing time [20]. Digital and mercury-based thermometers are the most widely used instruments for these measurements.

1.3.2 TEMPORAL MEASUREMENTS

Temporal measurements are taken using IR/Non-contact thermometers. These thermometers have commonly been used to check the temperature of travelers at public places during the Covid-19 pandemic and accept/deny them entry if elevated temperatures were detected. These thermometers are also used for tympanic (within the ear) measurements or can simply be pointed at the subject for temperature readings. Some research has even indicated that these thermometers may be as accurate as rectal thermometers [21], which has traditionally been the most preferred means of measuring core temperature. While these types of sensors are advantageous in that they can quickly measure/display temperatures of a large number of people and are simple to use, they have sometimes been found to misread normal temperatures as elevated and even fail to identify elevated temperatures. Moreover, these readings may also be affected by external factors such as sunlight, heat, and perspiration on the forehead [22,23]. Conflicting research has indicated that such measurements cannot adequately replace rectal measurements except for simple screening purposes [22]. In addition to IR thermometers, reusable (or disposable) plastic-encased liquid crystals for forehead application may be used for temperature measurements. The crystal substance changes color as the temperature rises. They are most suitable for home use and are preferred due to their ease of use, comfort as well as rapid results [24]. However, several studies have shown that these devices show inaccurate readings and frequently record normal temperatures despite an elevated body temperature [25,26]. Shown below in figure 2 [27] is a liquid crystal forehead thermometer.

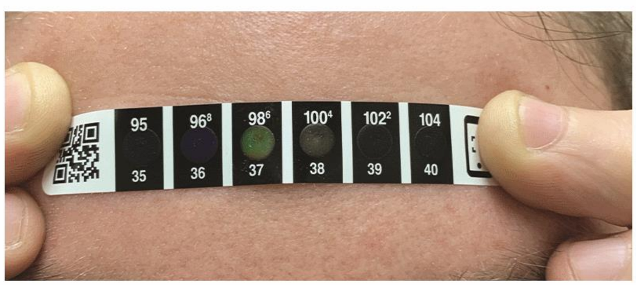


Figure 2: Liquid Crystal forehead thermometer [27].

1.3.3 ORAL MEASUREMENTS

Oral measurements are carried out using either digital or mercury-based thermometers. They are less affected by ambient temperature and are thus more accurate than axillary measurements [24]. Another advantage is the ease of accessibility, and the chances of better accuracy based on patient compliance. However, these measurements may be inaccurate if the patient has just ingested hot/cold foods and liquids or smoked. This method is also not suitable for intubated/comatose patients as well as for infants and there are also chances of mouth lacerations or oral infections occurring due to this method. Similar to temporal methods, this method is also unsuitable for continuous temperature measurements over longer periods of time [28].

All the above methods are advantageous in that they are non-invasive and offer a high degree of comfort for the patients and convenience to doctors during temperature measurement. These methods are also affordable and hygienic. While these advantages of non-invasiveness, comfort, etc. are widely agreed upon in the medical community and in several research studies, there has been a lot of conflicting research on what remains the most accurate peripheral method. To this end, a study was conducted by S. Asadian et al [6] to determine and compare the overall accuracy and precision of the above 3 peripheral temperature measurement types (5 sites- Axillary, oral, 2 temporal and tympanic). These measurements were compared to that of the readings obtained by a central nasopharyngeal measurement which they considered to be 'standard'. This standard method involves the insertion of a probe into the external opening of the nose and directing it to the pharynx. 237 patients in the ICU were observed for this study. All 4 peripheral measurements were taken simultaneously on a patient, and the nasopharyngeal measurement was made immediately afterwards. Axillary and Oral measurements were carried out using a standard mercury thermometer manufactured by Jiangsu Company, while the tympanic measurement was performed using a tympanic thermometer manufactured by Jinus Company. The forehead temperature was measured using the BC200 model strip thermometer. The nasopharyngeal measurement was obtained using a combination of a sensor, monitor and a disposable probe thermometer from Space Labs Medical America. The study found that the tympanic methods had a degree of precision, sensitivity and specificity/accuracy. They also had high correlation and agreement with the standard nasopharyngeal method. The oral method was found to have acceptable precision but not a good level of accuracy and also had a lower sensitivity and specificity.

The axillary method was found to have acceptable precision and a good correlation with the standard method but had low sensitivity and specificity. The temporal method had acceptable precision and a high correlation to the standard method but had poor sensitivity and specificity. The paired t-test was used to assess precision. For accuracy, sensitivity and specificity, positive predictive value (PPV), negative predictive value (NPV), and kappa coefficient tests were carried out. The tympanic methods also had the highest/very good agreement with the standard method as compared to the rest of the methods. This can be observed with their high kappa coefficient values. It was concluded that the tympanic methods were the best peripheral temperature methods, especially for use in the ICU, due to their high accuracy and precision, and their good correlation and agreement with the standard nasopharyngeal method [6]. The following were some of the results obtained from the testing and results described above.

Table 1: Accuracy of the 4 types of peripheral temperature measurements (Values closer to 1 have greater accuracy) [6]

Method	**Accuracy (Correlation between each method and the standard method) Correlation Coefficient (r)**
Tympanic (Within the ear) (Right and left)	0.938 (Right) 0.932 (left)
Oral (Within the mouth)	0.829
Axillary (Under the arm)	0.776
Temporal (On the forehead)	0.891

Table 2: Fever (≥38°C)/Afebrile detection by the 4 types of measurement methods (There were 74 patients who actually had fever, while 163 patients did not have fever) [6]

Method	Detected fever with actual presence of fever	Detected fever in absence of fever	Detected no fever with actual presence of fever	Detected no fever in absence of fever	False Positive and False Negative (%)
Tympanic (Within the ear) (Right and left)	62 (Right) 60 (Left)	0 0	12 14	163 163	0% and 0.162% 0% and 0.189%
Oral (Within the mouth)	48	0	26	163	0% and 0.351%
Axillary (Under the arm)	39	0	35	163	0% and 0.472%
Temporal (On the forehead)	40	3	34	160	0.018% and 0.459%

Table 3: Sensitivity, Specificity, PPV and NPV as measures of accuracy and precision of the different methods compared to the Standard nasopharyngeal method [6]

Method	Sensitivity	Specificity	Positive Predictive Value (PPV)	Negative Predictive Value (NPV)	Kappa Coefficient
Tympanic (Right and left)	83 (Right) 81 (Left)	100 100	100 100	93 93	88 85
Oral	64	100	100	86	72
Axillary	52	100	100	82	60
Temporal	54	98	93	93	59

1.4 IMPORTANT REQUIREMENTS FOR TEMPERATURE MEASUREMENTS

Although the measurement of skin temperature is important for the assessment of healthy skin function, it is typically not a clear indicator of the internal temperature of the body. All peripheral temperature measurements are about 1 ˚C -2 ˚C lower than the actual core body temperature [29] and thus do not represent the 'true' temperature of the body's internal organs. Moreover, the peripheral measurement methods described in the previous section are not suitable for continuous monitoring over a long period of time. Therefore, there is a necessity to measure core body temperature rather than peripheral temperature. Shown in figure 3 [30] are the distributions of peripheral and core temperature across the human body.

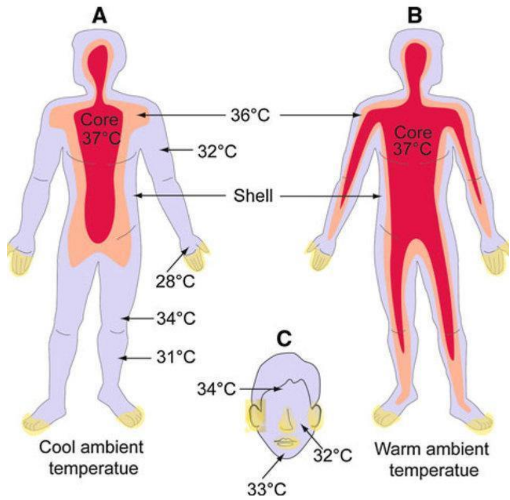


Figure 3: Representation of core and peripheral temperature on the human body. Reproduced from [30] with permission from Springer Nature.

Some of the important requirements for these measurements are explained in the following sections.

1.4.1 ACCURACY AND PRECISION

The values obtained must be close to the 'true'/established value of temperature for accuracy, while a repeatable set of values must all be close to each other for precision. In clinical practice, a measurement method is reliable when the 'standard deviation' is 0.3 ˚C - 0.5 ˚C [3]. Standard deviation is a measure of the amount of variation or dispersion in a set of values. It quantifies how much individual values in a dataset differ from the mean (average) of the dataset. A low standard deviation indicates that the values tend to be close to the mean, while a high standard deviation suggests that the values are spread out over a wider range. A smaller standard deviation implies more precision and less variability, while a larger standard deviation indicates greater variability or dispersion in the data. Temperature signal processing, sensor sensitivity, and sensor resolutions must be optimally designed to ensure a high level of accuracy and precision.

1.4.2 RESPONSE TIME

Short measurement times are required regardless of application. However, for some applications, a shorter measurement time may be more critical. Emergency measurements and instant entry/exit point screenings require times of <10s. A faster response rate would mean that the temperature sensor is able to respond to temperature changes more quickly and thus provide an accurate reading [31]. For continuous and long-term monitoring, longer measurement times are acceptable. Often, tradeoffs may need to be made between meeting other system requirements and obtaining optimal response times, all while maintaining accuracy.

1.4.3 OPERATING RANGE

Typical measurement ranges are between 25 ˚C and 45 ˚C, and unless in extreme situations, the sensing is typically expected to be continuous over a narrow range/window.

1.4.4 CONVENIENCE/HANDLING

The measurement method must be easy to use, safe, and convenient for the user, thereby improving acceptance. It should ideally be as minimally invasive as possible and must be free of artifacts. Most importantly, the method of measurement must be comfortable enough to be used by patient and practitioner. Depending on how these measurements are deployed, (invasive/non-invasive, contact/non-contact), the types of measurement would have various levels of convenience. Ideally, such measurement methods must be housed in the form of wearable form factors which limit user actions/inputs. Thus, user engineering aspects must also be taken care of.

1.4.5 LOW POWER

Core body temperature is a vital sign that needs to be measured continuously and over a very long period of time. Thus, measurement methods which use low power sensors/wearables are important and must ideally consume low power while also requiring a lesser number of charge cycles and consequently longer lasting (weeks/months depending on the application). In essence, they require long-term power autonomy with extremely limited power. Some examples of low power design techniques and calculations have been included in greater detail in section 6.

1.4.6 INTERFERENCE

The measurement must not be influenced by the ambient temperature regardless of whether it is the external environment (such as in non-invasive/minimally invasive methods) or by other bodily fluids/tissues within the body.

Typically, a trade-off will need to be made between all these requirements due to the necessity of continuous measurement and an appropriate level of invasiveness. Knowledge of natural thermal differences across various parts of the body, benefits and limitations of different measurement sites, as well as purpose and conditions of measurement must also be considered when deciding upon the best possible method of measurement or when designing a new system for measurement. While vital signs such as pulse rate, O2 saturation, blood pressure and respiration rate have solutions compact enough for use in wearables for continuous monitoring, core body temperature wearables for continuous monitoring remain the missing puzzle piece. It is the most difficult of vital signs to measure non-invasively. To this end, there has been a lot of research and development of core temperature sensors and their integration into wearable form factors. The shift in the measurement of vital signs from hospital settings to ambulatory settings and in daily life has been propelled by the popularity of wearable devices. Aside from all the requirements explained above, other important parameters that designers need to consider include the potential form factors (sticker medical patches, wrist worn bands, smart watches, embedded clothing, ear wear, etc.), size, cost, and thermal sensitivity. Thermal sensitivity is especially important because for clinical/medical grade applications, even the slightest of transient heat/power (of the order of microwatts) can cause highly inaccurate readings. The type of interface (whether analog or digital) is also an important consideration as this dictates the number and type of associated components required to be part of the system.

2. CORE TEMPERATURE MEASUREMENTS

Core temperature measurements have been carried out in a number of ways and sometimes require the insertion of a surgical probe in a hospital setting. The following are some of the methods that have been traditionally used for the measurement of core body temperature.

2.1 RECTAL TEMPERATURE

This has traditionally been the most preferred measurement of core body temperature and is usually associated with a high level of accuracy. The readings are usually stable due to the presence of a large mass of deep body tissue [32,38]. A standard thermometer is inserted into the rectum and the readings are typically about 1 ˚C higher than oral/axillary temperatures [33]. For continuous monitoring, a rectal probe, within which a flexible thermistor is present, is used as it helps with easier access and stays in place over longer periods of time. A minimum insertion depth of about 8 cm is accepted, while depths > 10cm are typically preferred for studies, as depth could affect temperatures by 0.1 ˚C - 0.2 ˚C [34]. The main disadvantage is the inconvenience / invasiveness of the thermometer and temperatures also do not usually display accurately when the body is in shock. There are also considerable delay times between 10-20 min during rapidly changing body temperatures and known delays for up to an hour depending on the contents of the patient's rectum, thus making this an unsuitable method when detections for rapid changes are required. Securing the probe for long term continuous monitoring is practically difficult. It is also unsuitable for patients with gastrointestinal bleeding, rectal infections and those who have undergone rectal surgery. Rectal temperature measurements are also contraindicated for premature infants [35,42]. Figure 4 [36] shows the Philips reusable adult rectal temperature probe with a single thermistor.

Figure 4: Philips reusable adult rectal temperature probe [36]

2.2 TYMPANIC TEMPERATURE

Usage of IR sensors to measure the temperature of the tympanic membrane of the ear is another commonly used method for the calculation of core body temperature. Measuring temperature through the ear is far less invasive when compared to other measurement methods. Moreover, the internal carotid artery supplies blood to both the tympanic membrane as well as the hypothalamus, which in turn is responsible for thermoregulation. Thus, it is believed that a correlation exists between the 'true' core temperature of the body, and the temperature measured at the tympanic membrane [37]. However, it has recently been found that this method is not as reliable as rectal measurements, and that ear infections or physical activity may affect results. Measurements are prone to error and inaccuracy [38]. The sensor must ideally measure the temperature of the eardrum, but due to inadequate insertion or manufacturing designs to prevent far enough insertion down the ear canal, this is often not the case. This results in the sensor actually displaying the temperature of the external ear canal, which is not a representation of core temperature [39]. For direct temperature measurements, cotton swab probes may be brought into direct contact with the eardrum, and the accuracy has been found to be similar to that of esophageal measurements. However, it is still difficult to perform and causes significant discomfort and pain to the patient, thus also increasing the risk of tympanic perforation. Shown in figure 5 [40] is the Novatemp tympanic temperature sensor.

Figure 5: Novatemp tympanic temperature sensor [40]

2.3 ESOPHAGEAL TEMPERATURE

This is preferred because of the location and blood supply of the esophagus, as well as the high responsiveness of this method. It is also a highly accurate method for core temperature measurement due to instant responses to central blood temperature changes and there are low chances of dislocation in anesthetized patients [38]. The probe is inserted through either the mouth or nose, and all the way to the esophagus, wherein the preferred location is the region between the 8th and 9th thoracic vertebrae [38,41]. However, insertion of a probe into the esophagus causes discomfort to the patients especially if they are conscious. Patients with a strong gag reflex cannot tolerate the probe and this method has been found to be inaccurate in situations where the patient has just ingested food/drinks and if certain amounts of saliva have been swallowed during the measurement. This method is usually used as an alternative when more invasive methods such as pulmonary artery measurements cannot be carried out on the patient [42]. Figure 6 [43] shows the Philips esophageal temperature probe.

Figure 6: Philips esophageal temperature probe [43]

2.4 URINARY BLADDER TEMPERATURE

This method was initially used based on the premise that the temperature of the urine corresponded to the core body temperature. This was however soon found to be inaccurate, as the temperature of urine was found to also change based on the rate of urination. This method of measurement is carried out in patients who already require urinary catheters, and there are no additional risks involved in the insertion of these catheters. The response time is better than that of rectal measurements, but delays occur in situations when there are fluctuations in body heat/thermally unstable conditions [38]. Shown below is the Smiths Medical Foley Catheter with a temperature measurement probe [44].

Figure 7: Bladder Catheter with temperature probe [44]

2.5 PULMONARY ARTERY TEMPERATURE

This is considered to be the gold standard for core temperature measurements. It involves the venipuncture of blood vessels to insert a Swan-Ganz Catheter in the pulmonary artery. Despite the invasiveness and the inconvenience caused to the patient, such measurements are carried out where the accuracy of readings is of utmost importance, such as in critical care units and in specific clinical interventions and is almost exclusively reserved for hemodynamic monitoring. The pulmonary artery carries blood directly from the 'core' of the body, and thus this method is considered to be the most accurate representation of core body temperature [38]. This method may not be used in patients with tricuspid or pulmonary valve mechanical prosthesis or those with tricuspid or pulmonary valve endocarditis. Shown below in figure 8 [45] is an arterial line catheter equipped with a thermistor.

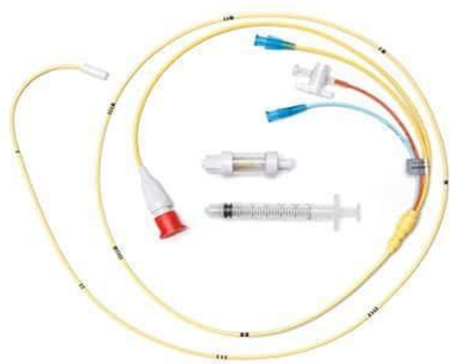


Figure 8: Arterial line catheter with thermistor [45]

For practical applications, the choice of core temperature measurement method also depends upon factors such as the requirements of either discrete or continuous/extended period measurements, the level of accuracy required (simple fever screening is less complicated to obtain accurate measurements as opposed to those in critical care units), the level of invasiveness (directly influenced by the state of consciousness of the patient), as well as the probability that the temperature being measured may remain stable or undergo rapid changes, which in turn influences the choice of quicker or slower measurement methods.

J.Y. Lefrant et al [46] compared the first 4 methods along with the inguinal method to the Pulmonary Artery temperature method. 42 ICU patients requiring pulmonary artery catheters were involved in the study. All temperatures were simultaneously monitored, and the Bland-Altman method was used for comparison. The Bland-Altman plot is a graphical method used to assess the agreement between two quantitative measurements. It provides a visual representation of the agreement in measurements by plotting the differences between two methods against their means. The pulmonary arterial temperature ranged from 33.7°C to 40.2°C. 529 measurements were obtained in total (about 9-18 for each patient). In the initial 20 patients, there were 252 measurements made among esophageal, rectal, inguinal, axillary, and pulmonary artery temperature measurements. Subsequently, in the last 22 patients, a total of 277 comparisons were conducted which included urinary bladder temperature in addition to the other methods. The results from the study are summarized in the table on the next page. It can be seen that for critically ill patients, electronic thermometers placed in the urinary bladder and esophagus exhibit greater reliability compared to electronic rectal thermometers. The electronic rectal thermometer, in turn, proved to be more reliable than the thermometers placed in the inguinal and axillary regions for measuring core temperature.

Table 4: Comparison of different core temperature measurement methods with Pulmonary Artery method [46].

Method	Mean Difference between Pulmonary Artery temperature and method
Esophageal	0.11±0.30°C
Rectal	-0.07±0.40°C
Axillary	0.27±0.45°C
Inguinal	0.17±0.48°C
Urinary Bladder	-0.21±0.20 °C

3. TECHNIQUES FOR CORE TEMPERATURE MEASUREMENTS

3.1 RESISTANCE-BASED THERMOMETRY

These techniques use thermal elements such as thermistors, RTDs (Resistance temperature detectors), thermocouples, digital temperature ICs and analog temperature ICs.

3.1.1 THERMISTORS

Thermistors are passive elements whose resistances change based on temperature. There are mainly 2 types of thermistors namely: NTC (Negative temperature coefficient) and PTC (Positive temperature coefficient). With NTCs, the resistance of the element decreases as temperature rises, while for PTCs, the resistance of the element increases with increasing temperature. NTCs are commonly made of semiconductor materials such as metallic oxides, using Cobalt, Manganese or Nickel, while PTCs are made of Lead Oxides, Barium or Strontium. The resistance of these materials is known, and the shape of the non-linear 'curve' between temperature and resistance depends upon the properties of the materials used in the thermistor [47,51]. At a fundamental level, in NTC thermistors, the increased temperature and thus thermal energy excites electrons in the semiconductor material, allowing them to move more freely. This increased mobility of electrons results in a decrease in resistance. In PTC thermistors, the crystalline structure undergoes changes (thermal lattice agitations) as temperature increases. These changes lead to an increase in resistance due to reduced mobility of charge carriers within the semiconductor material. Thermistors can be integrated into simple measurement circuits. They only require a resistor to create a voltage divider network, and the change in resistance corresponds directly to the change in voltage, which can be measured using an analog-to-digital converter [47]. They may be used in everyday appliances such as ovens, fire alarms, refrigerators, and digital thermometers as well as in applications spanning the medical, aerospace, automotive, food handling, instrumentation, and several other industries. They offer high precision and accuracy and can be easily miniaturized for use on the skin as well as within body cavities, blood vessels and tissues [48]. These thermometers are typically electronic, and the sensor is housed within casings with good thermal transmission properties. Thermistors are typically available in several configurations such as the hermetically sealed flexible thermistor (HSTH), the bolt-on/washer type, and the self-adhesive surface-mount style and are constructed in types such as the Bead, Disk, Rod, Chip, and glass encapsulated. Figure 9 below shows the construction of a disc, bead and rod thermistor [48].

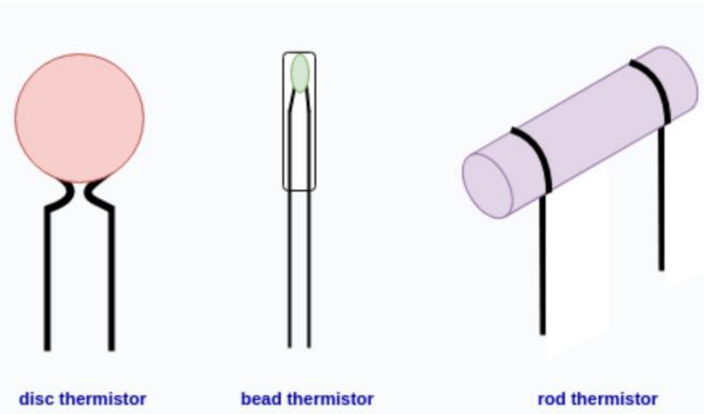


Figure 9: Different types of thermistors namely disc, bead and rod [48]

The Exergen medical thermometer is an example of a commercial device that uses an infrared sensor combined with a thermistor to measure body temperature by scanning the temporal artery on the forehead. This contactless thermometer is often used in clinical settings. The following are some of the advantages for the usage of thermistors in temperature sensing:

High Sensitivity: Thermistors exhibit a significant change in resistance for a small change in temperature. This high sensitivity makes them suitable for precise temperature measurements.

Wide Temperature Range: Thermistors are available in various types and configurations, allowing them to cover a wide range of temperatures. Different types of thermistors are designed for specific temperature ranges, from extremely low temperatures to high temperatures.

Accuracy: With proper calibration and characterization, thermistors can provide accurate temperature measurements within their specified temperature range [47].

Small Size: Thermistors are compact and can be integrated into small spaces or attached to surfaces, making them suitable for applications where space is limited.

Fast Response Time: Thermistors typically have a fast response time, allowing them to detect temperature changes quickly [48].

Cost-Effectiveness: Compared to some other temperature sensing technologies, thermistors are often more cost-effective, making them accessible for a wide range of applications.

Thermistors are often used in voltage divider circuits. A thermistor is placed in series with a fixed resistor (usually called a "bias resistor") and connected across a voltage source. The voltage across the thermistor is measured using an Analog-to-Digital converter (ADC), and by knowing the characteristics of the thermistor and the bias resistor, the temperature can be deduced. Figure 10 [49] shows an NTC thermistor connected in a voltage divider circuit.

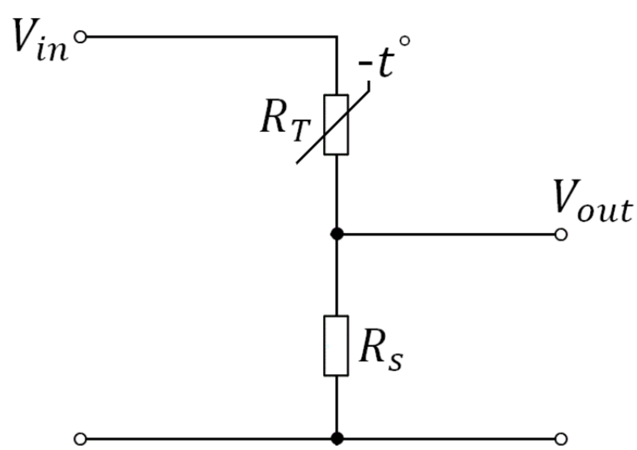


Figure 10: NTC thermistor connected in a voltage divider circuit [49]

$$V_{out} = (R_S/(R_S + R_T)) * V_{in}$$

Equation 1: Voltage Divider Equation. 'V_{out}' is the output voltage of the circuit. 'R_S' is the resistance of the bias resistor, 'R_T' is the resistance of the thermistor, and 'V_{in}' is the input voltage fed into the circuit.

Equation 1 is the equation used to determine the output voltage of the circuit. Once the output voltage is determined using an ADC or a voltmeter, this value can be fed into the equation, and thermistor resistance can be determined. Since the resistance-temperature relationship is non-linear, calibration and correction are necessary to convert resistance readings into accurate temperature measurements. This can involve using lookup tables, Steinhart-Hart coefficients, or polynomial equations to calculate the temperature. The Steinhart-Hart equation provides a way to approximate the behavior of a thermistor over a range of temperatures. This equation is a mathematical model that approximates the non-linear relationship between resistance and temperature for thermistors. The coefficients A, B, and C are determined through calibration and characterization of the specific thermistor being used. The accuracy of the equation depends on how well these coefficients are determined for the particular thermistor's characteristics. The equation is given as follows:

$$1/T = A + Bln(R) + C(ln(R))^3$$

Equation 2: Steinhart-Hart Equation. 'T' is the absolute temperature in Kelvin (K). 'R' is the resistance of the thermistor at temperature 'T'. 'A', 'B' and 'C' are the coefficients specific to the thermistor's material and design.

3.1.2 RESISTIVE TEMPERATURE DETECTORS (RTDs)

Resistive Temperature Detectors (RTDs) are similar to thermistors in that their resistances change with temperature. However, unlike thermistors they use a coil of wire wrapped around a core made from ceramic or glass. The coil of wire is typically made of pure material (such as Copper, Nickel, or Platinum) that has an accurate and known resistance-temperature relationship. This allows for easy determination of the measured temperature. Platinum RTDs are the most common type of RTDs to be used in industrial applications, due to their excellent stability, wide range, and resistance to corrosion. Nickel RTDs are cheaper than Platinum and also have good corrosion resistance [50,51]. However, it loses accuracy at higher temperatures and is thus limited to a lower temperature range. Copper RTDs have the best linearity and are low cost. Copper, however, oxidizes at higher temperatures, thus making these RTDs unsuitable for a higher temperature range. RTDs may be constructed in one of 3 ways namely: wire wound, coiled element, and thin film. S. Han et al [52] designed a battery-free wireless sensor that uses an RTD for the continuous monitoring of surface temperature and full-body pressure. This device is skin-like and makes use of Near field communication (NFC) power delivery and data communication to send a continuous stream of data to a central acquisition system. This data may then be assembled into spatiotemporal color plots based on areas on the body which have higher temperature and pressure gradients. The plots can then be used in the early detection of pressure sores, monitoring the circadian rhythm of the patient, and thus helping in finding predictive information to identify and prevent diseases. The system consists of 65 epidermal NFC temperature sensors. The subsystem of these sensors consists of modules for NFC communication, temperature sensing, analog-to-digital conversion, piezoelectric pressure sensors, a resistor and an energy harvesting module. Figure 11 [52] depicts these subsystems within the device and the layout of the components.

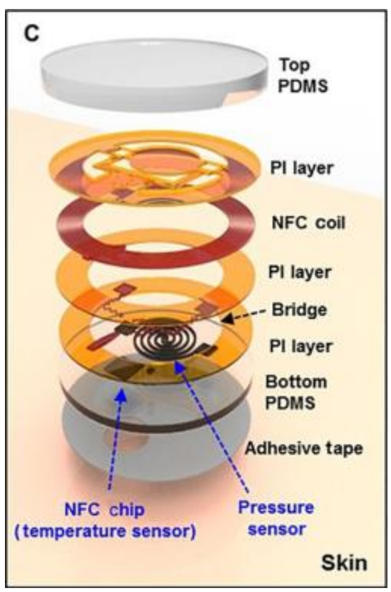


Figure 11: Exploded view of the device structure designed by S. Han et al. Reproduced from [52] with permission from Science Translational Medicine, The American Association for the Advancement of Science.

As already mentioned, the temperature sensor uses an RTD (SL13A) integrated into the NFC chip. The temperature sensor is defined by the accuracy and precision of measurement, the effective thermal mass of the overall device, and the response time. To define the precision of the temperature sensor, a calibration process was carried out and the recordings of these measurements that were wirelessly transmitted were compared to the readings obtained by an IR thermal camera. The differences in temperature between the 2 were less than 0.04°C. Even during transient conditions, the differences in temperature obtained were within 0.2°C of each other.

3.1.3 THERMOCOUPLES

Thermocouples are used for the measurement of higher temperatures and in situations where larger temperature ranges exist. They work on the principle of the Seebeck effect, wherein a conductor subjected to a thermal gradient generates a small voltage [53]. The magnitude of voltage generated depends on the type of metal. Thermocouples are usually created using combinations of alloys driven by factors such as chemical properties, stability, availability, and cost. The composition of various metals is influenced by the sensitivity and temperature range involved in the specific application [51]. The main limitation with the use of thermocouples is the lack of accuracy, as system errors of less than 1 ˚C may be hard to achieve. Thermopiles, also known as thermoelectric sensors or thermoelectric modules, are devices that generate an electrical voltage in response to a temperature difference between two of their surfaces [54]. They are usually composed of several thermocouples connected in series. Thermopiles are used to sense temperature differences between the hot and cold sides of the device. One side is exposed to the heat source, while the other side is kept at a reference temperature. The temperature difference between these sides generates the voltage output. Thus, the main difference between thermistors and thermocouples is that in thermocouples, the voltage induced between the 2 metals is the temperature sensing parameter, while in thermistors the temperature sensing parameter is resistance. While thermistors are constructed using semiconductor materials, thermocouples are made up of 2 different metals. While a thermistor is a passive sensor, thermocouples are active sensors. Thermocouples also have mostly linear responses unlike thermistors which have a non-linear response [54]. One end of these metals is connected at a point designated as the 'hot junction' while the other end is connected to the 'cold/reference junction'. The hot junction temperature is assessed with respect to the reference junction. When heat is applied to the hot junction, the temperature difference between the 2 metals causes a thermoelectric potential to be generated. The voltage thus produced across the terminals is directly proportional to the temperature difference between the junctions. The voltage is then scaled to temperature readings. Figure 12 [54] shows the construction of a thermocouple circuit schematic.

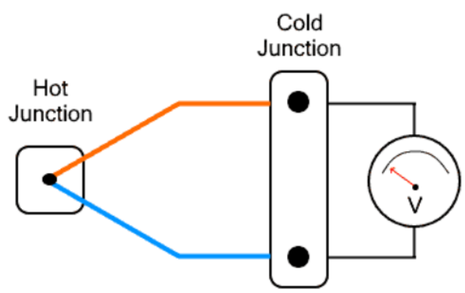


Figure 12: Simple schematic of a thermocouple [54]

Thermocouples are used in temperature measurement devices and non-contact IR thermometers. E. J. S. Chaglla et al [55] proposed the development of a system to measure tympanic core body temperature using Graphene-Inked IR thermopile sensors. A commercial IR thermopile MLX90614-DCA suitable for use in medical applications of ±0.2 °C in the range from 36 °C to 38 °C was used for the CBT measurement. Graphene, which boasts of the best electrical and thermal conductivity and high mechanical strength and surface area was used to improve the accuracy of the temperature measurements. To this end, Graphene platelets were coated onto the lens surface of the IR thermopile sensors using a drop-casting technique. The form factor of the device was an ear hook type enclosure designed to attach to the auricle. 10 participants were used in the experiments. CBT measurements were obtained during rest and also during physical activity. During rest, temperature was measured once every 60 seconds over a period of 10 minutes. An IR tympanic thermometer (IRT6520) was used as reference while the readings were also obtained using both the original MLX90614-DCA as well as the designed Graphene-coated thermopile. A Cosinuss one commercial body monitor was also used for the experiment involving exercise. During exercise, the measurements were made over a period of 25 minutes. For the ‘rest’ experiment, the average temperatures were calculated using stabilized data for ten subjects, both with the original thermopile (Tth) and the thermopile coated with graphene (TGN). These measurements were compared against the reference temperature obtained from the IRT6250 (Tref). Lin’s Concordance Correlation Coefficient (CCC) was utilized for comparison. The calculated CCC values were 0.150 for Tth compared to Tref, and 0.163 for TGN compared to Tref. Despite weak correspondence with the reference thermometer for both devices, the graphene-inked thermopile exhibited slightly better concordance, indicating improved performance. Bland–Altman analysis was conducted, showing the difference between CBT measurements obtained with Tref and those registered with Tth and TGN. The average temperature measured using the original thermopile was 0.51°C below Tref, with a ±0.62°C 2 SD (Standard Deviation). Conversely, the average temperature measured with the graphene-inked thermopile was 0.36°C below Tref, with a ±0.40°C 2 SD. The Bland–Altman analysis results reveal that the original thermopile underestimates core body temperature (CBT) by 0.51°C, whereas the graphene-inked thermopile underestimates CBT by 0.36°C. Consequently, the graphene-inked thermopile mitigates the underestimation of CBT by 0.15°C. Thus, the data from the Bland–Altman analysis suggested that the application of graphene coating could enhance the transmission of infrared (IR) radiation through the lens of the thermopile sensor [55].

For the experiment during physical activity, it was observed that during the initial four minutes of physical activity, CBT experienced a decline from 36.8°C to 36.6°C due to the body's thermoregulation response [55]. This response involved increased blood flow to the skin and skeletal muscles, leading to heat dissipation and a minor core temperature reduction. However, with continued physical activity, accumulated heat leads to an anticipated rise in CBT. After 8 minutes of exercise, CBT reaches around 37.3°C. It can thus be concluded that the graphene-inked sensor effectively detected fluctuations in CBT.

This graphene-inked thermophile also yielded slightly higher temperatures when compared to the original MLX90614-DCA thermopile. The experiment also showed that the graphene-inked sensor performed better when compared to the commercial body monitor. The Cosinuss One consistently reported a temperature 1.5°C lower than the reference CBT throughout the test. Unlike the Cosinuss One, the graphene-inked sensor demonstrated reduced susceptibility to airflow and ambient temperature influences, contributing to its more reliable measurements [55].

3.1.4 ANALOG TEMPERATURE SENSORS

Analog temperature sensors may be used as an alternative to thermistors with a fixed-value resistor in a voltage divider circuit. These Analog ICs provide linear output voltage over certain temperature ranges and are very easy to use. They do not require linearization or other accompanying circuitry. A commercial example of this type of sensor would be the LMT70 Precision analog temperature sensor by Texas Instruments. This sensor can act as a great replacement for RTDs and precision NTC/PTC thermistors [56].

3.1.5 DIGITAL TEMPERATURE SENSORS

Digital temperature sensors are highly accurate and can help simplify overall circuit design due to the analog-to-digital conversion that takes place inside the thermometer IC itself, and thus eliminating the need of a separate microcontroller for this purpose. Just like Analog temperature sensors, they don’t require additional linearization circuitry. A commercial example of this type of sensor would be the TMP112X digital temperature sensor by Texas Instruments. This sensor may also be used for requirements involving NTC/PTC thermistors [57].

Some examples of commercial temperature sensor ICs have been described in section 4.

3.2 RADIANT-HEAT/IR BASED THERMOMETRY

This is a non-invasive and convenient measurement method, wherein IR technology may be used to determine temperature at the eardrum or the forehead, which is then extrapolated to obtain core temperature. The temperature of the object is determined by measuring the IR radiation from its surface. An IR sensor based smart system to estimate core body temperature based on convection and body resistance methods has been proposed by Pratim Ray et al [58]. This system runs on a 5V DC source and is energy efficient (consumes only 165mW). The system uses the MLX90614 IR based temperature measurement chip which contains a digital signal processing unit along with an AT Mega 328 breakout board. The IR sensor was placed between 1.5cm to 30cm away from the forehead and the temperature was found to differ by 0.7 ˚F when compared to Terumo's CM-210 deep body temperature sensor. In general, readings may easily be affected by uneven heating of the detector or the probe, incorrect positioning onto the measurement surface and this method is unsuitable for long-term continuous monitoring [59]. The use of IR sensors for the measurement and estimation of core body temperature from the forehead (temporal) and within the ear (tympanic) have already been discussed previously. IR thermal imaging and near IR spectroscopy are other techniques that have recently been considered. IR thermography involves the use of a thermal imager to detect and capture the IR radiation emitted by the body. This method has been studied for usage in the determination of core temperature from thermal images of the face/inner canthus of the eye. It is based on the theory that they are supplied by the internal and external carotid arteries which are thought to provide a good estimation of core temperature [60]. Near Infrared spectroscopy (NIRS) is a non-invasive tool used for the continuous measurement and monitoring of brain tissue oxygenation. It is used in critical care units and is seen as a potential tool to estimate core temperature, though there hasn't been active clinical research yet [61].

3.3 RADIO WAVE/PILL-BASED THERMOMETRY

This method is commonly called the temperature pill telemetry. It is based on the transmission of low-frequency radio waves, which vary with temperature. It involves ingesting thermistors/sensors in capsule form, which allows the temperature inside the digestive tract to be measured and transmitted to an external receiver. This method allows for a convenient non-invasive mechanism of core temperature measurement. Though there are commercial telemetry pills available in the market, this method has primarily been used in field and laboratory settings. Small, single-use thermistors have been used in athletes on a trial basis to determine temperatures of the alimentary canal [62]. There have also been many studies to validate the accuracy of telemetric monitoring systems and compare it to that of more widely accepted measurement methods. An experiment was carried out by McKenzie et al [63] to compare the VitalSense telemetric monitoring system (the system consisted of a receiver/monitor and a thermistor-based injectable capsule) with the rectal based measurement method consisting of a standard thermistor probe inserted at a depth of 11cm. Results showed that there were no significant differences between both measurement methods and that the telemetry pill method had fewer compliance issues and better acceptability as compared to the rectal probe. However, research undertaken by Monnard et al [64] using the CorTemp telemonitoring system (which is similar to VitalSense) has indicated that its efficacy for 24hr continuous core temperature monitoring is questionable. The research concluded that given the very narrow range within which core temperature is regulated, any estimation method needed to be highly accurate and precise, and this was lacking in the ingestible capsules, which had inherent measurement errors and temperature drift issues. It has been recommended that the pill be swallowed 5-6 hours before any measurements/data points are to be collected to prevent food/drinks from affecting measurements and avoid sensor expulsion prior to data collection. This method is also unsuitable for infants, patients with a history of gastrointestinal disorders and those with pacemakers (or other implantable devices) to prevent any potential electromagnetic interference. Nuclear magnetic resonance, microwaves and ultrasound are other newer methods that involve the use of radio waves to estimate core body temperatures. MRIs have primarily been used as a potential tool to determine and monitor brain tissue temperature during imaging tests, but these devices are bulky and unsuitable for patients with implants. Microwave radiometry has a better penetration depth than IR thermography but is prone to electrical interference [65]. Ultrasonic measurements work on the principle that the speed of ultrasound penetration is influenced by temperature.

3.4 HEAT FLUX THERMOMETRY

Zero Heat Flux (ZHF) thermometry is a minimally invasive method to evaluate core body temperature based on thermal flux measurement. A deep body temperature thermometer using this method was invented by Fox and Solman way back in 1971 [66]. It is based on the principle of heat flow from warmer to cooler environments, wherein the magnitude of flow would depend on the thermal gradient and properties of the medium through which heat flows. A state of thermal equilibrium is established between the skin and an active heating element through insulation, thus causing the skin surface to be heated to deep body temperature and this allows the calculation of core body temperature at the surface. The sensors are preferably located at regions of low skinfold thickness such as the sternum or the forehead. The active heating element requires a power source, and thus this system may be uncomfortable as a wearable for the patient, due to the heat that is generated. Zero Heat Flux sensors are commercially available, but their use is not as widespread, despite their high levels of accuracy. Teunissen et al [67] made use of a ZHF system containing a non-invasive continuous cerebral temperature sensor manufactured by Philips. This sensor is in the form of a flexible patch that can be placed on the forehead and consists of a layer of thermal insulation and the electronics. Neoprene was used to achieve flexibility while the electronics were mounted onto a Kapton layer. The heating element is controlled by a PI controller. The ZHF temperature was compared to esophageal temperatures as the gold standard and was found to correspond to within +/- 0.5 ˚C. While the ZHF method may be suitable for longer-term continuous monitoring, the need of a heater element has made this an energy hungry technique, thus making them unsuitable for battery operated systems [67]. Active thermal flux thermometers and their passive counterparts are based on thermistor technology. Shown below in figure 13 [66] is the schematic of the control circuit and probe cross-section of the deep body thermometer designed by Fox and Salmon.

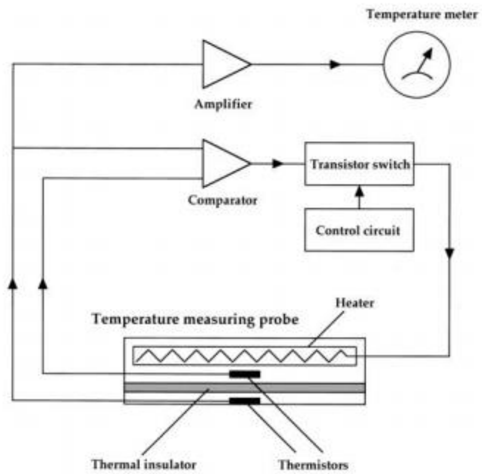


Figure 13: Deep Body thermometer designed by Fox and Salmon. Reproduced from [66] with permission from Springer Japan KK.

Anuar et al [68] proposed the development of a wearable temperature sensor based on the Single Heat Flux concept. The device is able to transmit information wirelessly using a module and would thus be suitable for remote and continuous monitoring. The device's reliability was validated by comparison of its measurements with a commercial IR contact thermometer. The single heat flux and double heat flux (explained in this section subsequently) methods eliminate the need for a heater. This is because an additional parameter (temperature of the device) is taken into account. A main difference between the 2 methods is that the single heat flux method uses a single temperature sensor for each skin temperature and device temperature while the double heat flux method uses 2 sensors for each parameter. The model for the single heat flux system of the proposed wearable is shown in figure 14 [68].

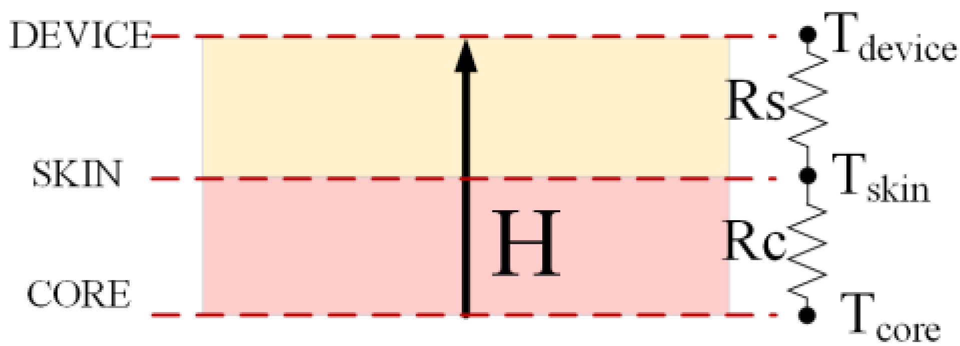


Figure 14: Single heat flux system model. Reproduced from [68] with permission from IEEE.

The system consists of 2 thermistors (one outside of the skin and one on top of the device). The equation below can be used to determine the core body temperature.

$$T_{core} = T_{skin} + (R_C/R_S)(T_{skin} - T_{Device})$$

Equation 3: Equation to determine core body temperature, where R_S is the thermal resistance of the device (between the 2 sensors) and R_C is thermal resistance of the core body (area beneath bottom sensor). R_S and R_C are determined from thermal conductivity k of the respective materials (human skin and in this system EVA foam).

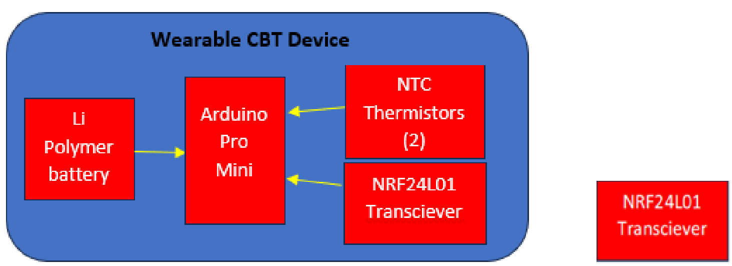


Figure 15: Wireless CBT measurement System [68]

As shown above in figure 15, the system consists of a transmitter and a receiver. The transmitter part of the system includes an Arduino microcontroller, 2 NTC thermistors and an NRF24L01 transceiver wireless module. The data obtained from the thermistors is wirelessly sent to the receiver part of the system, which is another NRF24L01 wireless module. The module may transfer data to a device such as a laptop.

The experiment [68] involved 5 participants, whose body temperature was measured at multiple places (forehead, behind the ear and wrist). As a reference (T_{ref}), a commercial IR thermometer (FS-201) was used to measure temperature at these same sites. The ambient temperature and humidity were also constantly monitored to ensure that it remained relatively constant over the course of the experiment. It was observed that the sensor took longer time to settle on a stable/equilibrium reading for the hotter sites, as compared to the less hot sites. The forehead recorded the hottest temperature, followed by behind the ear and then the wrist, which settled on an equilibrium temperature the fastest. The difference between the reference temperature measured at each site compared to the core temperature measured by the designed system is summarized below.

Table 5: Difference between measured reference temperature and measured CBT at different sites [68]

Site	Mean Temperature Difference (T_{ref}-T_{site})
Forehead	0.05°C
Behind the ear	0.15°C
Wrist	0.37°C

A derivative of ZHF, called Dual Heat Flux thermometry, eliminates the need for a heater element. It is a passive core temperature measurement technique that makes use of temperatures established across 2 paths of unequal but known thermal resistances [69]. They are thus more comfortable as compared to ZHF systems. Kitamura et al [69] developed an instrument whose probe consisted of dual heat flow channels with different thermal resistances, with each channel having a pair of IC sensors attached on its top and bottom. The instrument is driven by a battery, thus overcoming the necessity of AC power supply. Core body measurements were taken using both ZHF as well as dual heat flow methods and were found to be 37.02 °C and 37.08 °C respectively. These results showed that the dual-heat flux method was as accurate as the ZHF method and is suitable for long-term continuous monitoring due to it being battery powered [69]. Due to a latency of several minutes before thermal equilibrium is reached, these methods are unsuitable for rapid screening applications. But once equilibrium is reached, rapid changes in core temperatures can be measured with high accuracy. The passive nature of this method makes it more susceptible to environmental artifacts. J. Feng et al [70] developed a miniaturized core body temperature wearable that was based on this dual heat flux principle. The temperature measuring probe of this device consisted of 2 round blocks, each 34 mm in diameter with 1 block having a greater height than the other. The temperature probes were attached to the forehead, and the entire system was in the form factor of an elastic hairband. Internally, the system consisted of 5 subsystem units as shown below in figure 16 (a) and (b). The wearable made use of low power design techniques such as incorporating controllable power switches and Bluetooth low energy technology. The temperature probe consisted of 4 Negative temperature coefficient thermistors.

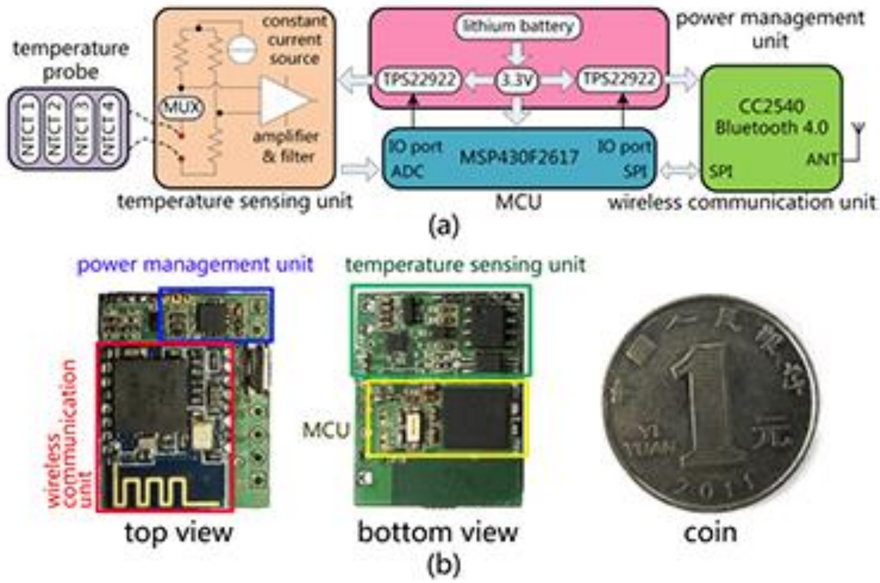


Figure 16: (a) Internal block diagram of the wearable. (b) Top and bottom view of the PCB. Reproduced from [70] with permission from IOP Publishing.

For the heat transfer medium of the probe, PDMS was used, due to its highly flexible, hydrophobic, viscoelastic and biocompatible nature. To help reduce thermal equilibrium time, Calcium Carbonate powder was mixed with the PDMS in a certain ratio. Figure 17 shows the model based on which the core body temperature was estimated. The orange block represents the skin/subcutaneous tissue layer of the body, while the blue blocks represent the temperature probe. A temperature gradient always existed between the body's deep tissue and the skin's surface in the vertical direction, as CBT was typically higher than the skin temperature. The equation to determine the CBT was determined as shown below.

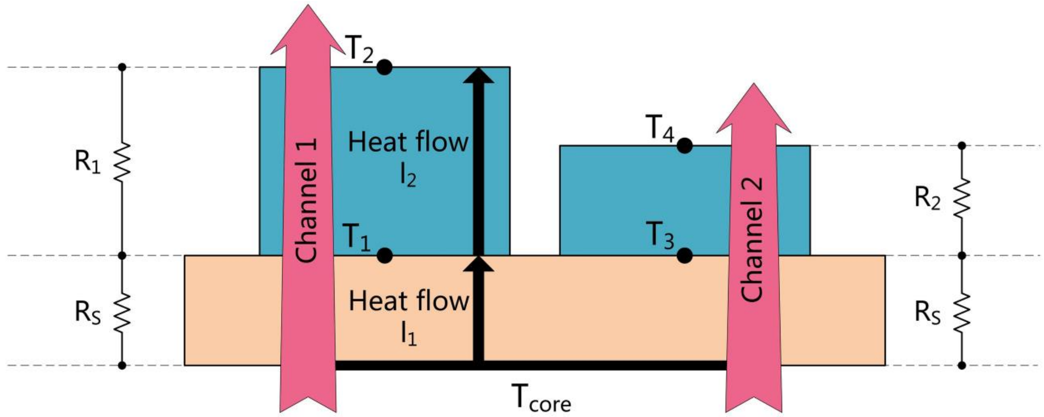


Figure 17: Model for CBT measurement based on the Double Heat Flux method. Reproduced from [70] with permission from IOP Publishing.

$$T_{core} = T_1 + ((T_1 - T_2)(T_1 - T_3))/(K(T_3 - T_4) - (T_1 - T_2)), K = R_1/R_2$$

Equation 4: CBT calculation equation, where 'K' is the ratio of the thermal resistances R_1 and R_2. T_1 is the temperature at the temperature probe-skin interface at channel 1, T_2 is the temperature on top of the probe at channel 1, T_3 is the temperature at the temperature probe-skin interface at channel 2, T_4 is the temperature on top of the probe at channel 2.

This study [70] also involved finite element analysis, experiments on a hot plate and also tests involving human participants. For the hot plate in the temperature range of 37-41°C, the system's measurement error as compared to a high accuracy thermometer was 0.08 ± 0.20 °C. In the human trials, the system's measured temperature during rest (34 subjects) showed a difference of 0.13 ± 0.22 °C compared to sublingual temperature. Additionally, during exercise trials (30 subjects), a significant temperature increase of 1.36 ± 0.44 °C was observed from rest to jogging. What these results showed was that this method did hold promise for reliable continuous measurement of core body temperature during rest as well as the tracking of CBT variations during active periods of exercise.

M.J. Buller et al [71] also presented an algorithm to estimate real-time core body temperature from heart rate variability (sequential/minute by minute) measurements of heart rate for first responders wearing different levels of PPE. The basis behind this study was to enable the development of a real-time monitoring system for these individuals, who are often at risk of heat exhaustion and cardiac stress, due to their reduced ability to thermoregulate while wearing PPEs. The core temperature estimation algorithm (ECTemp algorithm) is based on the premise that heart rate serves as a "noisy" indicator of core temperature. The algorithm leverages the fact that heart rate reflects both metabolic rate and blood flow to the skin, encompassing information on heat generation and heat transfer to the environment. The relationship between oxygen consumption and cardiac output, as defined by the Fick principle [72], is integral to estimating metabolic rate using heart rate. Additionally, variations in heart rate associated with increased skin perfusion from hypothalamic temperature-driven vasodilation provide insights into heat dissipation. The algorithm assumes that the equilibrium between these processes is manifested in heart rate and uses it to approximate the level of heat storage. The estimated core temperature was compared to the core temperature measured by rectal and esophageal probes. A high level of accuracy was found between the heart rate method and the conventional methods. A major limitation of this work was that the efficacy of this algorithm was only tested in a homogeneous group of participants (US army volunteers) in terms of age and fitness levels. Thus, its effectiveness is still unknown for older and less fit individuals [71].

4. COMMERCIAL TEMPERATURE SENSOR IC CHIPS

While there are several factors that go into the design of a core temperature type system/temperature monitoring system, one of the most important requirements is that the temperature sensor must be able to provide accurate readings of its external surroundings, regardless of whether it used within the body, on/near the body, or for environmental monitoring. This section highlights a few examples of commercially available temperature sensor IC chips commonly used in wearable devices.

Melexis MLX90632 FIR sensor [73]: This is a miniature SMD thermometer IC for accurate non-contact temperature measurements. It has a standard measurement resolution of 0.02°C and a medical measurement resolution of 0.01°C. It requires a supply voltage of 3.3V and supply current of 1mA. It is most suitable for high precision non-contact temperature measurements such as for body temperature, IOT applications, etc. It also has an accuracy of +-0.2°C within the narrow temperature range of 35°C to 42°C for medical applications. The standard/commercial grade version of this sensor is used in smart thermostats while the medical grade version is used in ear and forehead thermometers as well as in wearables. This sensor must be in thermal equilibrium and under isothermal conditions and is not recommended for use in heat transfer and transient condition applications, due to the inherent nature of IR sensors to be susceptible to thermal gradient errors. Figure 18 [73] shows the Melexis MLX90632 IC, while figure 19 [73] depicts the medical accuracy table of this IC.

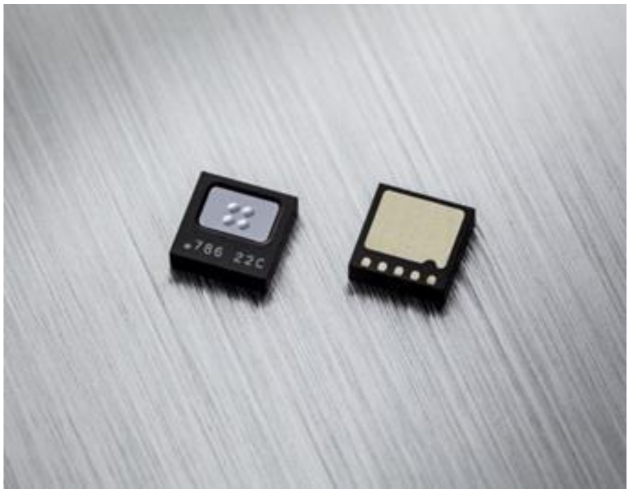


Figure 18: Melexis MLX90632 FIR Sensor [73]

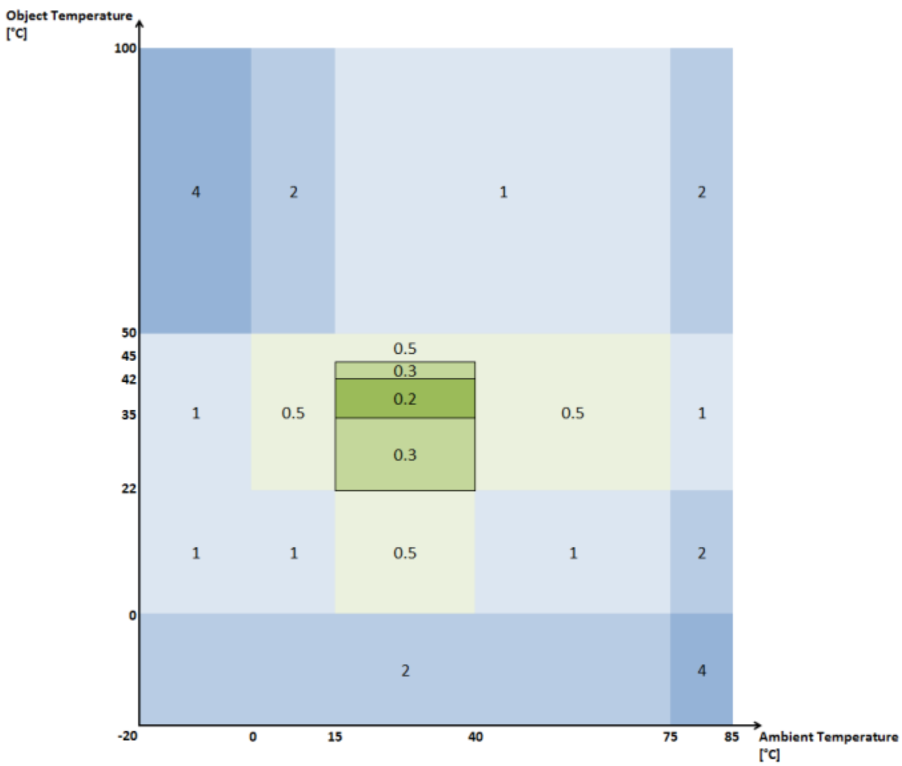


Figure 19: Melexis MLX90632 Medical accuracy table [73]

TI TMP114 Ultrathin digital temperature sensor [57]: This is a highly accurate I2C compatible digital temperature sensor that is housed in an ultrathin (0.15 mm) 4-pin package. It has an accuracy of ±0.3 °C maximum from –10 °C to 85 °C and ±0.5 °C maximum from –40 °C to 125 °C. It consumes very low power between 1.08V to 1.98V. The small size and low height of the TMP114 package makes it suitable for systems where volume is a constraint that enables the placement of the sensor under other surface mount components. It is suitable for use in wearable fitness and activity monitors, mobile phones, etc. Figure 20 [57] shows the Melexis TI TMP114 IC, while figure 21 [57] depicts the average accuracy of the IC at 1.2V and 1.8V.

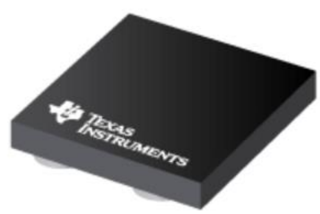


Figure 20: TI TMP114 digital temperature sensor [57]

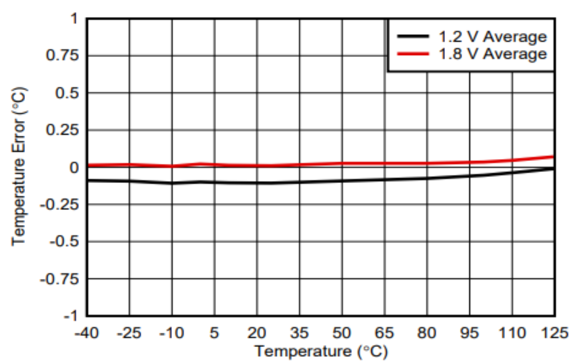


Figure 21: TI TMP114 Average Temperature Accuracy [57]

Analog Devices ADT7320 [74]: This is a digital temperature sensor that has an accuracy of +-0.20°C from −10°C to +85°C at 3.0 V and ±0.25°C from −20°C to +105°C at 2.7 V to 3.3 V. It doesn't require temperature calibration by the user and no linearity correction is required. It contains a 16-bit analog-to-digital converter (ADC) to monitor and digitize the temperature to a resolution of 0.0078°C. It is rated for operation over the −40°C to +150°C temperature range. Its high accuracy makes it ideal for medical instrumentation applications and consumes low power (Minimum of 2.7V and a maximum of 5.5V). Figure 22 [74] shows the Analog Devices ADT7320 IC, while figure 23 [74] shows the Temperature Accuracy at 3V (left) and at 5V (right).

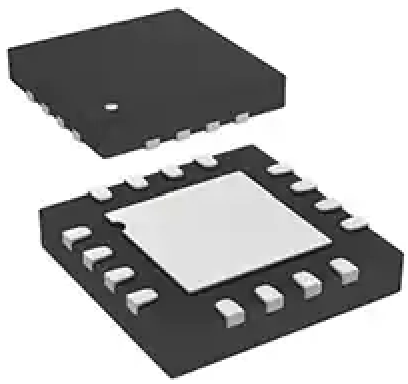
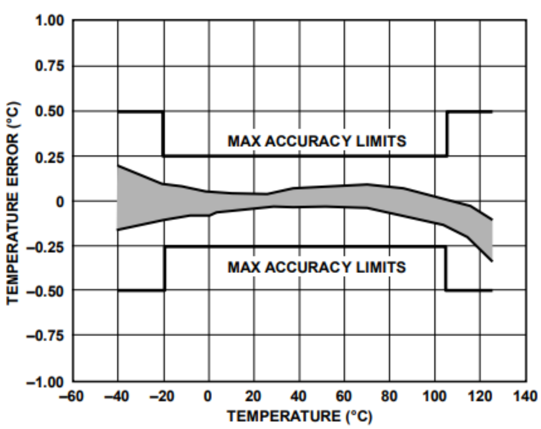


Figure 22: Analog Devices ADT7320 digital temperature sensor [74]



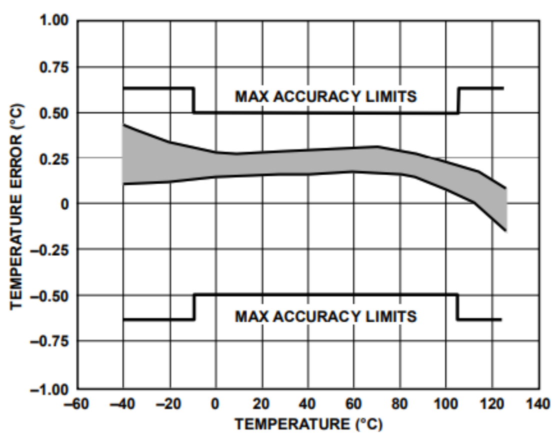


Figure 23: Temperature Accuracy at 3V (left) and at 5V (right) [74]

Maxim MAX30208 [75]: This is a digital temperature sensor that has an accuracy of ±0.1°C from +30°C to +50°C and ±0.15°C accuracy from 0°C to +70°C. The devices measure 2 x 2 x 0.75 millimeters (mm) and come in a thin 10-pin LGA package. The ICs operate off a supply voltage ranging from 1.7 to 3.6 volts and consume less than 67 microamps (µA) in operation and 0.5 µA in standby. They can easily be connected to a microcontroller using flat flexible cables (FFC) or flat printer cables (FPC). They are suitable for use in battery-powered devices such as smartwatches and medical patches. Figure 24 [75] shows the Maxim MAX30208 IC, while figure 25 [75] depicts the block diagram.

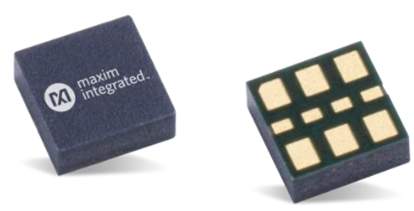


Figure 24: Maxim MAX30208 digital temperature sensor [75]

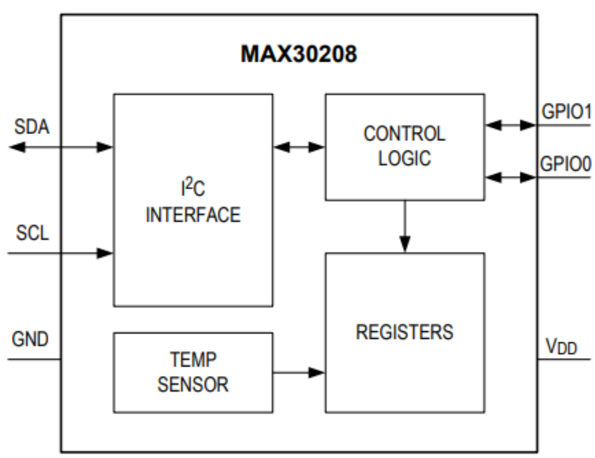


Figure 25: Maxim MAX30208 block diagram [75]

TI LMT70 [56]: This is a precision analog temperature sensor that has an accuracy of ±0.05°C from 20°C to 42°C and ±0.36°C accuracy over its wide temperature range from -55°C to +150°C. The device is considered ultra-small, with a 0.88mm by 0.88mm 4-bump package. It has a wide power supply range of 2V to 5.5V and is useful for applications that require high precision and low power, with its low power supply current requirement of 9.2uA to 12uA (max.). This IC is also considered to be a great replacement for RTD and precision NTC/PTC thermistors. It is ideal for use in medical thermometers, human body temperature monitoring solutions, as well as in medical/fitness equipment. Figure 26 [56] shows the TI LMT70 IC, while figure 27 [56] shows the accuracy of this IC at different temperatures.

Figure 26: TI LMT70 Analog temperature sensor [56]

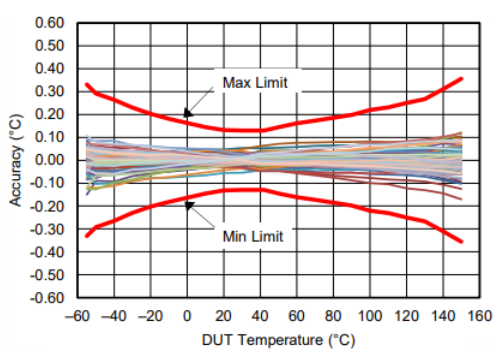


Figure 27: TI LMT70 Accuracy vs Temperature [56]

Table 6: Comparison of device specs and operational parameters of the 5 temperature sensor ICs

Parameter	Melexis MLX90632 [73]	TI TMP114 [57]	Analog Devices ADT7320 [74]	Maxim MAX30208 [75]	TI LMT70 [56]
Type of Sensor	FIR Sensor	Digital Temperature Sensor	Digital Temperature Sensor	Digital Temperature Sensor	Analog Temperature Sensor
Temperature Range	-20°C to +100°C	-40°C to +125°C	-40°C to +150°C	0°C to 70°C	-55°C to +150°C
Accuracy	±0.2°C in the human body temperature range	±0.2°C (20°C to 50°C), ±0.3°C (-10°C to +80°C), ±0.5°C (-40°C to +125°C)	±0.2°C (-10°C to +85°C at 3V), ±0.25°C (-20°C to +105°C at 2.7V-3.3V)	±0.1°C (30°C to 50°C), ±0.15°C (0°C to 70°C)	±0.2°C (-20°C to 90°C), ±0.23°C (90°C to 110°C), ±0.36°C (-55°C to +150°C)
Resolution	0.01°C	16-bit resolution: 0.0078°C	16-bit resolution: 0.0078°C	16-bit resolution: 0.005°C	Depends on ADC Resolution
Supply Voltage range	3.3V	1.08V to 1.98V	2.7V to 5.5V	1.7V to 3.6V	2V to 5.5V
Power Consumption	1000uA, 1.5uA (sleep current)	0.7uA, 0.16uA (shutdown current)	210uA, 2uA (shutdown current)	67uA, 0.5uA (standby current)	9.2uA, 0.05uA (shutdown current)
Response Time	Refresh rate of 0.5s	300ms	Less than 2s	0.5s	1.5s
Package Size	3x3x1mm	0.76x0.76x0.15mm	4x4mm	2x2x0.75mm	0.88x0.88mm
Package Type	QFN, SFN	Picostar (DSBGA)	LFCSP	Thin LGA	WLCSP (DSBGA)

5. TEMPERATURE SENSOR DESIGN AND CONSIDERATIONS FOR WEARABLES

A temperature sensor system involves various components that work together to measure the body's internal/skin temperature and interpret it. The following are some of the parts of a typical temperature sensor system:

1) TEMPERATURE/CORE BODY TEMPERATURE SENSOR:
    - Sensor Element: This is the physical component that responds to changes in temperature and directly measures the core body temperature. This could be a thermistor, thermocouple, or other temperature-sensing technology suitable for internal temperature measurement.
2) SIGNAL CONDITIONING:
    - Amplification: The raw signal from the temperature sensor might be weak, so an amplifier can be used to strengthen the signal and improve accuracy.
    - Filtering: Sometimes, noise or unwanted signals might be present in the sensor output. Filtering helps remove these unwanted components of the signal.
3) ANALOG-TO-DIGITAL CONVERTER (ADC):
    - Conversion: The conditioned analog signal from the temperature sensor is converted into a digital signal so that it can be processed by digital components.
4) MICROCONTROLLER/PROCESSOR:
    - Processing Unit: The microcontroller or processor receives the digital signal from the ADC and can perform further processing or calculations, if necessary.
    - Memory: Temporary storage for data and program instructions.
5) DISPLAY/OUTPUT INTERFACE:
    - Display Unit: This could be an LCD, LED, or any other type of display to show the core body temperature value.

- Output Interface for External Systems: The temperature value might also be output through other means, such as communication ports (UART, I2C, SPI) or wireless communication (Bluetooth, Wi-Fi) to external devices and systems.

6) TEMPERATURE CALIBRATION/CORRECTION:
   - Calibration Circuitry: This ensures accurate temperature readings by calibrating or correcting the sensor output.
7) POWER SUPPLY:
   - Power Management: This ensures that the components receive the required power and manages power consumption effectively.
8) USER INTERFACE:
   - User Controls: If the temperature sensor system has user-configurable settings, there might be buttons or controls for the user to interact with the system.
9) ALARM SYSTEM:
   - Alarm Unit: If the core body temperature exceeds or falls below a predefined threshold, the system can be designed to trigger an alarm or notification.

5.1 SENSOR'S OWN TEMPERATURE INFLUENCE

It must be ensured that the sensor's own temperature doesn't influence the measurement reading of a wearable device. The heat from the sensor IC eventually leads to the sensor die from the PCB and through the package. This heat is conducted through a metal thermal pad located under the package and can cause parasitic heating. This leads to thermal conduction in and out of other pins and thus interferes with temperature measurements. Designers may employ certain techniques to counter parasitic heating such as the use of thin traces to minimize thermal conductivity away from the sensor IC [76]. They may also measure temperature at the top of the package, as far away as possible from the IC pins, instead of the underside. The electronic components which may contribute heat to the temperature monitoring system can also be placed as far away from the sensing element as possible. This helps minimize their impact on temperature measurement data.

5.2 SYSTEM-USER THERMAL CONSIDERATIONS

System-to-user thermal design considerations must also be taken into account in addition to ensuring thermal isolation from heat sources. One possible solution is to ensure that the sensor is as close as possible to the target temperature to be measured. Another solution is to incorporate flex or semi-rigid PCBs into wearable designs and medical patches. The thickness of the flex board may also be minimized as much as possible as thinner boards can flex more efficiently and enable better contact [76].

5.3 CALIBRATION

Several legacy temperature sensors need to be calibrated in the field or recalibrated once a year. However, factory calibrating these sensors is a solution that helps eliminate this need. It also has several advantages such as bypassing the need to develop software to linearize the output, as well as simulate and fine-tune the circuit [76]. The risk of impedance mismatches is also reduced, and precision components are no longer required.

5.4 EVALUATION SYSTEMS

Robust evaluation systems for easy development and testing of sensors are also an important consideration. Typically, these systems consist of a microcontroller and its dedicated shield to interface with the sensor IC or a flex PCB to hold the sensor. The designer may only need to connect the evaluation hardware to a PC using a USB cable. The necessary device drivers are then automatically installed onto the system and the evaluation kit software may be downloaded. Many mobile or wearable devices can measure body temperature at multiple locations. Thus, these evaluation systems are modeled in such a way as to create profiles of both local and whole-body temperature. The microcontroller achieves this by polling each temperature sensor at regular intervals.

6. TECHNIQUES USED FOR LOW POWER TEMPERATURE SENSORS AND WEARABLES

The most important consideration that must be taken into account when designing temperature sensors and wearables is power consumption. A wide range of techniques may be employed, and data analysis performed, to reduce the power consumed by the temperature sensor and thereby the overall power drawn by the system. A previous section discussed different techniques for core temperature measurements with mentions of both Digital temperature sensor ICs and Analog Temperature sensor ICs. Techniques to optimize power consumption in fact start from here. In the case of digital temperature sensors, the sensing element typically consists of an analog frontend along with an analog-to-digital converter that are integrated onto a single chip. There is always a small amount of time/latency during which the system measures temperature, and thus it is possible to allow the system to remain in this 'idle' state every time measurements are made. The frequency of these measurements being made can in turn be programmed (by making the use of duty-cycling methods), and thus lead to reduced current consumptions. It is possible to allow the system to remain in these 'idle' states without compromising on accuracy, as temperatures do not typically change quick enough as compared to the time taken for these measurements. Temperature sensor chips always have specified tolerances and margins of error. The type of interface that the digital temperature sensor uses also influences power consumption. 2 common interfaces include I2C and SMBus.

The I2C (Inter-Integrated Circuit) is a 2-wire serial interface (Serial Data Line (SDA) and Serial Clock Line (SCL)) that is intended for communications between ICs within a single device and uses a 'Master-Slave' communication scheme [77]. This allows multiple devices to communicate with each other on the same bus. The master device initiates communication, while the slave devices respond to the master's commands/requests. The SDA carries the actual data between devices while the SCL carries a clock signal that synchronizes the data transfer. Each device on the I2C bus has a unique 7-bit or 10-bit address. The master specifies the address of the slave it wants to communicate with. Figure 28 [77] shows the basic schematic of the I2C protocol.

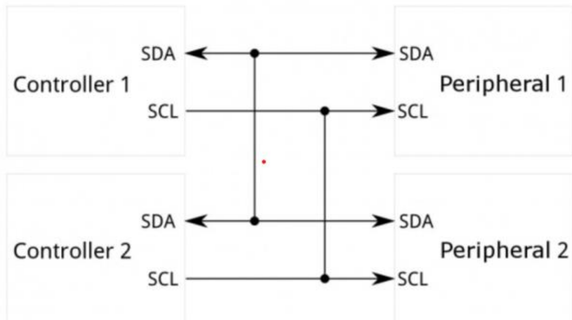


Figure 28: I2C Protocol [77]

The SMBus (System Management Bus) is a derivative of I2C that also uses a 2-wire interface and similar communication schemes, but at a lower maximum data transfer rate of 100kbit/s. It has a more strict and standardized protocol to ensure interoperability between different manufacturer devices [78].

Both interfaces make use of 'pullup' resistors on their signal lines to create logic high in the absence of devices pulling down on the bus. The higher the resistance of these pullup resistors, the higher the current consumed. There is thus a usage of current at every bus transaction. Lowering overall power consumption requires tradeoffs between frequency and resistance [79]. Another consideration is the type of computation which may either be performed on the device itself, or sometimes on a peripheral device. In general, it is good practice to use components in their sleep mode whenever possible and write energy aware code (without the use of polling loops, making use of interrupts, checking Vcc before running, etc.). Thus, power budgets are impacted by a myriad of factors such as efficiencies of each component, duty cycling, data processing and transmission of information.

It may sometimes be more economical to perform the steps required for preprocessing of signals inside of the sensor rather than performing these steps after transmitting all raw data out. For example, a measurement may be passive and may require lesser power. However, it requires more signal processing, and we need to consider if this needs to be done on the device or on a peripheral, which in turn affects the design. We need to consider efficiencies of each component, efficiencies related to how we run the device (such as duty cycling/processing, sleep modes, interrupts, software design optimizations, etc.), whether we need to constantly transmit information out of our device, how much data to be transmitted and the type of data (e.g.: digital voltages every 5 min such as in CGMs or analog signals continuously such as in ECGs), and thus these factors affect the operation of the device. They also play a factor in how we claim/design our wearable lifetime operation. Clearly, power requirements have a significant effect on the design and operation of wearable biosensors. Some of the commonly used techniques are as follows.

6.1 COMMONLY USED TECHNIQUES

6.1.1 SAMPLING FREQUENCY AND CONVERSION RATE

This is the frequency with which the sensor measures the temperature. Temperature sensors allow you to configure the sampling frequency or measurement interval. Increasing the measurement interval (reducing the frequency of temperature readings) can reduce the overall power consumption. This intermittent sampling involves waking up the sensor at predefined intervals, taking measurements and putting the sensor back to sleep. It is possible to reduce sampling frequency significantly in applications where the temperature changes slowly. ADCs are integrated in digital temperature sensors, and each sample/measurement of temperature is caused by an analog-to-digital conversion. Each measurement is retained until it is retrieved, or until a subsequent conversion overwrites it. A large amount of power may also be wasted if the conversion rate is higher than the polling rate (explained below), as this results in making measurements that are never obtained/retrieved from the temperature sensor [80]. An appropriate sampling rate may be determined based on the application. The TI TMP114 digital temperature sensor [57] has adjustable conversion time and period, with an Active conversion time of 6.4ms. The sensor is in 'continuous conversion mode' during this time. The Conv_Period [1:0] bits in the configuration register control the conversion rate. The sensor typically consumes 68uA during conversion and 0.26uA during the low power standby period. The average current consumption in the continuous mode can be reduced by decreasing the rate of conversions.

$$Average\ Current = ((I_{ACT} * t_{ACTIVE}) + (I_{Standby} * t_{Standby}))/t_{Convperiod}$$

Equation 5: Average Current consumed by the TI TMP114 in continuous mode. t_{ACTIVE} is the Active Conversion Time, $t_{Convperiod}$ is the Conversion Period and $t_{Standby}$ is the standby time between conversions calculated as $t_{Convperiod}$-t_{ACTIVE} .

6.1.2 POLLING RATE

This is the frequency with which data is obtained/retrieved from the temperature sensor. If the sensor is not in use all the time, a considerable amount of power savings may be achieved. Higher polling rates are usually associated with lesser power savings [80]. A large amount of power is wasted if the polling rate is higher than the conversion rate, as resources are spent retrieving the same data repeatedly.

6.1.3 SERIAL COMMUNICATION FREQUENCY

This is a term used in the context of the timing of bus transactions. It involves optimizing the data transfer rate of the communication interface (such as I2C, SPI, UART, SMBus, etc.). Higher polling rates require higher communication frequencies, and it correlates to less time spent on each transaction. I2C protocols require smaller pullup resistors to enable higher communication frequencies. This however causes an increase in the amount of instantaneous current consumed during logic low. Lowering the I2C bus speed decreases the frequency of clock pulses, thus reducing the power consumption associated with I2C communication [79,80]. This can be particularly effective if high-speed communication is not necessary in the application. Unnecessary bus activity can also be minimized by limiting the frequency and volume of data transmitted over the I2C bus. This can be achieved by optimizing the communication protocol and avoiding unnecessary transactions. Optimization of the protocol can be done by using burst mode transmission, where multiple data values are sent in a single burst, minimizing the time the sensor and communication interface need to be active. This can be more power-efficient compared to sending individual values separately. Some communication protocols such as Bluetooth Low Energy (BLE) and ZigBee consume less power when compared to Wi-Fi. The TI TMP114 [57] is I2C and SMBus compatible. Shown below are the equations used to determine the minimum and maximum possible value of the pullup resistor [79,81]. Once these limits have been determined, the value of the pullup resistor on the I2C bus can be selected based on the trade-off between power consumption and speed. t_r, C_b and V_{OL} are known parameter values that can be obtained from the I2C specifications table for the sensor. These values are obtained based on the operating mode (Standard, Fast or Fast Plus). I_{OL} value depends on values of V_{OL} and V_{CC}.

$$R_P(min) = (V_{CC} - V_{OL}(max))/I_{OL}$$

Equation 6: Equation to determine minimum value of pullup resistor for the I2C bus for TMP114, where $V_{CC}$is the Amplitude, V_{OL} is the Low-level output voltage and I_{OL} is the Low-Level output current.

$$R_P(max) = t_r/(0.8473 * C_b)$$

Equation 7: Equation to determine maximum value of pullup resistor for the I2C bus for TMP114, where $t_r$is the Rise time of both SDA and SCL signals and C_b is the capacitive load for each bus line.

6.1.4 ONE SHOT

This is an optional feature found in most digital temperature sensors. The least amount of power is consumed by the sensor when it is set to shutdown mode. No measurements are made by the device during this mode, as all circuitry is idle. Sometimes, a single command referred to as a ‘one-shot’ command may be issued to allow one temperature measurement to be made and then return to the shutdown state. The TI TMP114 has a one-shot conversion mode [57].

6.1.5 LOW-POWER AND SLEEP MODES

Many modern temperature sensors and microcontrollers come with low-power modes. These modes allow the device to operate with reduced power consumption when high accuracy is not required. By placing the sensor in a low-power mode between measurements, overall energy consumption can be significantly reduced. There are also microcontrollers with ultra-low-power features for processing and data management. Deep sleep modes may also be utilized during idle periods to minimize power consumption. The microcontroller must be able to efficiently manage the sensor's sleep/wake-up cycles. These sensors are often optimized for wearable applications. The TI TMP114 [57] has a low-power shutdown mode. If a temperature conversion is in progress when the Mode bit in the configuration register is set to enter shutdown mode, the device stops the conversion and discards the partial result. All active circuitry is powered down and this mode can be used in conjunction with the One_Shot bit to perform One-shot temperature conversions. The typical power consumption in the shutdown mode is only 0.16uA.

6.1.6 INTERRUPTS

Rather than continuously polling the temperature sensor for new readings, an interrupt-driven approach may be followed. The sensor can trigger an interrupt signal to the microcontroller when new data is ready. This approach allows the microcontroller to remain in a low-power state until needed.

6.1.7 DYNAMIC VOLTAGE AND FREQUENCY SCALING (DVFS)

Some microcontrollers support DVFS, which allows adjusting the operating voltage and clock frequency dynamically. By scaling down the voltage and frequency based on the workload, overall power consumption can be reduced during periods of low activity. This is a type of adaptive communication protocol that can dynamically adjust communication frequency based on the required data accuracy and urgency of data acquisition. For example, the sensor could communicate at a lower frequency during periods of stable temperature and increase the frequency during temperature transitions. This would differ based on the user's activity level or the time of day. Along the same lines, adaptive wake-up thresholds may be set to optimize the wake-up mechanism.

6.1.8 SELECTIVE POWER GATING

Power gating involves completely shutting off power to specific components/sections of the device when they are not actively processing data. This prevents unnecessary power consumption and extends the overall battery life. If there are multiple sensors involved in the system on different parts of the body, selectively activating only the necessary/relevant sensors at a given time based on monitoring requirements can help save power.

6.1.9 HYSTERESIS

Hysteresis may be introduced in the temperature threshold settings. Hysteresis prevents frequent transitions between active and sleep modes when the temperature is near a threshold. This can minimize unnecessary power cycles. The TI TMP114 [57] has an alert status register with individual high and low thresholds along with adjustable hysteresis values. The default hysteresis value for both the high and low limits is 5°C.

6.1.10 TEMPERATURE AVERAGING

Similar to intermittent sampling, instead of taking a single temperature reading, multiple readings may be averaged over a certain period. This can help reduce the impact of noise and short-term fluctuations in the measurements, allowing the sensor to operate at a lower power level while maintaining accuracy.

6.1.11 ENERGY EFFICIENT ANALOG FRONT-END AND HARDWARE

A low-power analog front-end for signal conditioning may be implemented. Components with a low quiescent current and components such as programmable gain amplifiers to adjust the sensor dynamic range may be used. Energy harvesting techniques to power the core body temperature sensor such as body heat or motion can supplement or replace battery power. Making use of hardware features such as automatic clock stretching or peripheral DMA (Direct Memory Access) that can optimize power consumption during data transfer is also important. It is also equally important to turn off unused features. Many temperature sensors have additional features such as high-resolution modes which may be disabled to reduce overall power consumption. Combining data from multiple sensors (Sensor fusion) to enhance accuracy and reduce power consumption can also be implemented.

6.1.12 ENERGY AWARE SOFTWARE AND ALGORITHMS

Optimized signal processing and data analysis algorithms that strike a balance between accuracy and computational efficiency help reduce overall power consumption. Using an interrupt-driven approach, avoiding polling loops, using fixed-point arithmetic instead of floating-point, duty cycling strategies, dynamic energy-aware algorithms (algorithms that adapt their behavior based on energy state of the device), approximation (usage of a simpler/faster algorithm that produces less accurate results but consumes less energy), compression (reducing size/complexity of input or output data to save energy), parallelization and pipelining (dividing a computational task into smaller and independent subroutines that can be executed simultaneously/in parallel to reduce execution time and energy), and caching (storing frequently used or recently accessed data in a memory location that allows faster retrieval and minimizes the need to repeatedly access the original source of data to help reduce latency, improve performance and reduce energy) [82].

6.2 IMPORTANT PARAMETERS TO DESCRIBE CURRENT CONSUMED BY DEVICE

6.2.1 ACTIVE CURRENT

Active Current is the current consumed while the digital temperature sensor is making a temperature measurement [80]. During Analog to Digital conversion, this current may also be described as Quiescent current (Iq). The time taken for the digital temperature sensor to measure temperature is called Conversion time. Active current is consumed during Conversion time. Most modern digital temperature sensors typically have a conversion time of 100 ms or less. It is not required for these devices to operate continuously to provide updated temperature measurement results, especially at a frequency of 4Hz or faster. These devices spend time either in a standby or shutdown mode when they are not making measurements.

6.2.2 SHUTDOWN CURRENT

Shutdown current is the current consumed while the digital temperature sensor is completely idle [80]. In this mode, no measurements are made by the device. It can be described as leakage current and is typically less than 1 µA.

6.2.3 STANDBY CURRENT

Standby current is the current consumed by the digital temperature sensor while it is neither active nor in the shutdown mode [80]. Devices that do not have a shutdown mode still count time during this ‘idle’ phase to consider calculations for making periodic measurements. During this phase, current is only consumed by the active parts of the system, typically an oscillator and a digital counter. The standby current is always higher than the shutdown current.

6.2.4 AVERAGE CURRENT

Average Current is the current consumed by the device in active mode integrated over a time span [80]. The time that the device spends in active and standby modes is included as part of the average current.

Another possible solution that may be thought of to reduce overall power consumption is to simply disconnect the power from the temperature sensor when not in use. However, there is still an amount of leakage current during this shutdown phase even as the internal circuitry is completely idle. Some systems may have additional circuitry to include a power-off switch, and this could in turn lead to further leakage currents comparable in magnitude to the current due to the shutdown phase.

7. CORE BODY TEMPERATURE PREDICTION METHODS

While all the methods described in the previous sections for the measurement of core temperature are being used today, with each method having its own sets of pros and cons, they are all invasive to various degrees, which leads to the need to be able to measure core body temperature in a much less invasive way. To this end, there has been active research on finding methods of estimating/predicting core body temperature from surface temperature and through using non-invasive methods.

7.1 PROTOCOL 1 (6-PARAMETER APPROACH: SKIN TEMPERATURE, HEART RATE AND HEAT FLUX TO PREDICT CBT)

Human subject studies conducted by Niedermann et al [83] involved the usage of non-invasive methods such as skin temperature, skin heat flux and heart rate. 2 studies were conducted in differing ambient conditions (hot environment and cold environment). The study used a multi-parameter approach and identified different physical and physiological parameters like skin temperature, heart rate and particularly skin heat flux, that needed to be considered for a reliable prediction of core body temperature in different environmental and working conditions. It concluded that multiple physical and physiological parameters at different body sites needed to be measured for reliable prediction of core body temperature. The study also showed that no additional environmental data within the range of environmental conditions chosen for these studies was required to predict core body temperature. The only data needed for the calculations were the physiological data obtained non-invasively. Table 7 shows the mean and standard deviation of this data obtained for the various physiological parameters.

Table 7: Mean and standard deviation of the data obtained from the 6 physiological parameters. T: Temperature, HF: Heat Flux, bmp: beats per minute [83]

Physiological parameter	Mean	Standard Deviation
$T_{upper\ arm}$ (°C)	34.831	3.095
$T_{lower\ arm}$ (°C)	34.472	2.389
T_{thigh} (°C)	35.185	2.681
Heart Rate (bmp)	108.230	29.891
HF_{chest} ($W\ m^{-2}$)	201.118	104.627
HF_{back} ($W\ m^{-2}$)	328.077	122.751

The hot environment study involved an ambient temperature of 30°C, while the cold ambient temperature had a temperature of 10°C. The subjects were dressed in a 2-layered clothing system. 10 healthy and physically active males were selected for this study with the following characteristics determined after a maximal graded exercise test on the treadmill: Age= 23.0±3.9 years, height= 180±9 cm, weight= 74.3±8.3 kg, maximal oxygen consumption= 57.8±5.3 ml kg−1 min−1 and maximal heart rate= 193±11 bpm. A second submaximal exercise test with a constant treadmill speed was also performed to set the exercise intensity to 40% and 60% of V02 peak through adjustment of the inclination. All subjects were non-smokers, free of any cardiovascular, metabolic and intestinal diseases, and were not under any medications.

The studies were conducted over a few days, and at the same time each day to avoid any effects that differing circadian rhythms at different times of the day could cause to the core body temperature [83]. After entering the chamber, the subjects were first made to stand stationary, and then had to walk at 40 %V O2 peak for 40 min in a hot environment (30.0±0.2 °C). This was followed by a 20 min sitting break outside the chamber (25.7±0.2 °C). After the break, the subjects re-entered the chamber and after being made to stand stationary for another 2 min on the treadmill, they were made to complete a 40 min walk at 60 % V O2 peak. This was followed by a 40 min sedentary rest period outside the chamber. Two trials were conducted for each subject and involved the same experimental protocol, with the only difference being the presence/absence of additional thermal radiation from the front. This was done to simulate an additional heat source. A panel of 25 150W red bulbs was used for this purpose, and the amount of thermal radiation was held constant at 500 W m−2. This was measured using a heat flux sensor.

For the cool environment study [83], 10 healthy physically active male students aged 24.6±2.0-years (height= 180±5 cm, weight= 75.1±9.1 kg, maximum oxygen consumption= 60.2±5.9 ml kg−1 min−1, maximal heart rate= 193±9 bpm) participated and underwent the same set of screening tests as the subjects in the hot study. The subjects were dressed in a 3-layer clothing system. They were made to remain stationary for 10 min, walk at an inclination and speed corresponding to 60% VO2 peak and finally a 60 min rest phase. The chamber was set up to mimic military conditions of activity and rest periods and had a temperature of 10.1±0.2 °C. Intestinal temperature (Tint) was measured using a telemetric pill system (CorTemp, HQInc). Rectal temperature (Tre) was measured using a rectal thermistor (MSR, Henggart). Skin temperature was measured at 9 sites (head, chest, upper back, upper arm, lower arm, hand, abdomen, thigh, and calf) using iButtons (DS1922L, Maxim Integrated Products). Additionally, two temperature sensors (iButtons, DS1923) were placed between the first and the second layer of the clothing system mentioned earlier. These sensors were placed on the chest and the upper back of the right side of the body and fixed on the T-shirt with surgical tape. All temperature data were recorded at 10s intervals. Skin heat flow was measured using flat heat flux sensors (Captec, Lille) on the chest and upper back fixed with surgical tape. The sensors were thin, light, and flat copper-based plates. The data were recorded at 10s intervals. Heart rate was recorded at 5s intervals using a heart rate monitor (Polar RS800, Polar Electro). All data obtained in these experiments was reduced to 1 min intervals by picking one sample per minute [83].

7.1.1 STATISTICAL ANALYSIS

The data sets obtained from these experiments [83] were divided into 2 parts randomly. Some data was used for model development while the rest of the data was used for model validation. Given that this was now a multiple regression analysis problem (the value of a variable needs to be predicted based on the value of 2 or more other variables), uncorrelated components were required. A principal component analysis (PCA) was conducted for this reason. The correlation matrix was applied to the standardized data and an eigenvalue higher than 1.0 was a prerequisite for the components to be included into the model. Multiple iterations were performed to exclude parameters showing communalities lower than 0.6 or complex structure (two or more components with a loading higher than 0.4 for each distinct parameter). In the first iteration, communalities of parameters were considered. The second iteration involved the loading of the parameters. For each independent group of dependent factors, the extracted parameters were converted into a factor score. Thus, this allowed the independent principle components to be calculated from the physiological measures. Finally, multiple linear regression of the core temperature could be carried out using these components. The equation below was used to calculate Factor scores for each group.

$$factor\ score = \sum coefficient_{\text{variable}} * (x_{\text{variable}} - x'_{\text{variable}})/\text{sd}_{\text{variable}}$$

Equation 8: Factor scores for each group calculated using a coefficient of each included variable, where the coefficient was obtained from PCA. xvariable and x'variable are the measured and mean value of the variable respectively, while sdvariable is the standard deviation of the data. Standardized variables were calculated using the mean and corresponding standard deviation.

Validation of this model was carried out by calculating the root mean square deviation (rmsd) and using it to measure fitment between the measured and calculated core body temperature. The acceptance criteria was set at 0.5°C based on another study conducted by Yokota et al. A model bias was calculated based on the equation below to determine if the model had either overestimated or underestimated the core body temperature.

$$bias = \Sigma(x_{\text{predicted}} - x_{\text{measured}})/\text{Number of measurements}$$

Equation 9: Model Bias

The subject data that had been excluded from the model development was used to validate the model. The predicted core body temperature was compared with the measured core temperatures (rectal and intestinal), and the rmsd and model bias were compared to the standard deviation of the measured core temperatures (rectal and intestinal). Additional validation was performed by using a data set from another study conducted by Maikenen et al. After the calculation of factor scores based on equation 8, the scores were integrated in the multiple regression model as independent standardized variables as shown below in equation 10.

$$Core\ body\ temperature\ =\ 0.278\ *\ factor\ score1\ +\ 0.275\ *\ factor\ score2\ +\ 37.644$$

Equation 10: Core Body Temperature

Thus, a multi-parameter statistical approach was chosen to develop a model predicting core body temperature in changing environment and activity levels [83]. With the help of PCA to extract three skin temperature sites, heart rate and two skin heat flux sites, a multiple regression model to predict core body temperature using these extracted parameters in hot and cool conditions was calculated. While the experiments to generate this model involved subjecting the participants to 2 different exposures (hot and cold), the factors extracted from PCA turned out to be different if only one type of exposure was considered (hot or cold). Extraction of factors using PCA also depends on the number of skin temperature sites being measured during the experiment. Based on the results, it is recommended that core body temperature be predicted using at least 3 skin temperature sites, 2 heat fluxes and also heart rate. Skin temperature at the thigh, heat flow measurement from the chest and upper back were found to be of particular importance. In addition, heart rate is important as it provides information about the metabolic heat generated by the body. It is also strongly recommended for the parameters to be measured as close to the skin as possible, as results become unreliable farther away. This model is however limited currently to a very specific group of subjects who all had similar fitness levels and body composition, and to the specific environmental conditions simulated in the chamber [83]. Therefore, more research may be needed to further improve the prediction of core body temperature using this model and non-invasive measurement approach and make it applicable to a wider range of subjects who differ in all aspects of body composition (such as height, weight, body surface area, body fat percentage, oxygen consumption, heart rate, etc.), age, gender, and physical fitness levels. A broader range of environmental conditions must also be simulated. The more sites at which skin temperature and skin heat flux is extracted, the more accurate would be the prediction of core body temperature.

7.2 PROTOCOL 2 (COMPARISON OF AN 18 PARAMETER AND 2 PARAMETER APPROACH WITH THE NIEDERMANN APPROACH TO PREDICT CBT)

Another study by Eggenberger et al [84] aimed to compare the model developed by Niedermann et al [83], with two novel multi-parameter core temperature prediction models. The study involved the prediction of rectal temperature under two different exercise and clothing conditions in a hot and moderately humid environment, as well as at rest at normal room temperature. The study also tried determining the extent to which the number of parameters measured could be reduced, without affecting the results obtained in their model. It was determined that a prediction model with only two input measurements (heart rate and insulated skin temperature at the scapula) had similar prediction ability when compared to two more comprehensive models. Among the two comprehensive models, the new model comprising 18 input parameters [84] did not outperform the previously developed model by Niedermann et al [83]. which included only six measures and no insulated skin temperatures. It could thus be concluded that the model with only 2 input measurements provided comparable validity and superior practicality for estimating rectal temperature in young adult males.

A homogenous group of males was used in this study due to the possible influence that age, gender, and health status may have on thermoregulation [84]. These males were non-smokers, healthy, and physically active. 13 healthy male participants (age 30.9 ± 5.4 years) participated in two experimental sessions. Both experiments consisted of 15 min baseline seated rest (23.2 ± 0.3◦C), 15 min of seated rest and cycling in a chamber for 20-60 min (35.4 ± 0.2◦C), and finally a 30 min seated rest outside the chamber. The differences in session 1 and session 2 were the clothing worn by the participants and the heart rate at which these participants exercised. During session 1, the participants wore light athletic clothing (t-shirt and shorts) and exercised at 75% of their heart rate maximum (HR max). During session 2, the participants wore protective firefighter clothing (jacket and trousers) and exercised at 50% HR max. As a reference, rectal temperature was measured using a rectal thermometer (DS18B20, MSR Electronics) 10cm past the anal sphincter. Insulated skin temperature using the same MSR thermistors was measured at 4 different sites, uninsulated skin temperature at the same sites, iButtons (DS19221, Maxim Integrated) at 6 different sites, skin heat flux at 3 different sites using gSKIN heat flux sensors (XM 26 9C, greenTEG), heart rate using a 2-lead ECG belt (Unico swiss tex) and Faros loggers (Bittium Bio signals). All the data was recorded at a sampling rate of 0.1 Hz.

7.2.1 STATISTICAL ANALYSIS

Similar to the study undertaken by Niedermann et al, the dataset was divided into 2 groups, where one set was used for model development while the other set was used for model validation [84]. Prior to this, the data was processed using moving average filters, second order low-pass Butterworth filters and cross-correlation analysis. The validation dataset (data participants set aside for this purpose) was applied to the same prediction model as Niedermann et al, and component scores were calculated using equation 8. PCA was performed on the 18 non-invasive parameters using the dataset that was reserved for model development with orthogonal rotation, while the sampling adequacy was verified using the 'Kaiser-Meyer-Olkin' measure. This model was termed the 'Max-Input Model'.

A 'Min-Input Model' was developed in which a multiple linear regression equation was built based on the development dataset using 7 uncorrelated parameters obtained from a first reduction of the 18 input parameters [84]. These 7 parameters were heart rate, insulated temperature at the scapula, non-insulated temperature at the radial artery, arm, and forehead as well as heat flux at the sternum and scapula. A further stepwise reduction of the number of parameters (while also ensuring that the final regression equation still produced 'acceptable' predictive values) finally resulted in 2 parameters (heart rate and insulated temperature at the scapula). The following were the equations obtained for the max-input and the min-input model.

$$\text{Core Body Temperature} = 0.2978 * \text{factor score1} + 0.2471 * \text{factor score2} + 37.2539$$

Equation 11: Max-Input Model Core Body Temperature

$$\text{Core Body Temperature} = 0.0100 * \text{heart rate} + 0.0837 * T_{\text{ins scapula}} + 33.1735$$

Equation 12: Min-Input Model Core Body Temperature. $T_{\text{ins scapula}}$ **is the insulated temperature at the Scapula.**

While the model developed by Niedermann et al. was found to have had an R^2 adjusted value of 0.708 and a SEE (Standard error of estimate) of 0.276°C, the Max-Input model had an R^2 adjusted value of 0.703 and a SEE (Standard error of estimate) of 0.278°C, and the Min-Input model had an R^2 adjusted value of 0.677 and a SEE (Standard error of estimate) of 0.290°C. It is based on this that it can be concluded that the Min-input model which uses just the heart rate and insulated skin temperature at the scapula provided comparable validity and superior practicality for estimating rectal temperature in young adult males. The usage of heart rate either as a single measure or as part of a prediction model was also demonstrated in other research, including that of Laxminarayan et al., 2018 [85]. This importance of heart rate could be based upon the relationship between metabolic activity and heat transfer to the skin. Elevated temperature could also have an effect on the sinoatrial and atrioventricular nodal cardiac cells. Skin temperature at the Scapula could be based on the observation that placement of temperature sensors on a bony site could better reflect core temperatures as compared to measurements on muscle tissues. It could also be due to the superiority of measurements at proximal sites as compared to distal sites (such as calf, thigh, hand, and forearm) [84,85]. To further validate this prediction method, it is suggested that the experiments replicating the min-input model be carried out on a more diverse group of participants as well as carrying out different types of exercises in different environments and clothing. The comparison of parameters used in the 3 models has been summarized and included in table 8.

Table 8: Comparison of parameters used in the 3 models. T: Temperature, HF: Heat Flux, ins: Insulated, HR: Heart Rate [83,84]

Parameter	Niedermann et al., 2014b	Max-Input Model	Min-Input Model
Temperature non-insulated			
T scapula		X	
T forearm	X	X	
T radial		X	
T thigh	X	X	
T calf		X	
T hand		X	
T arm	X	X	
T sternum		X	
T rib		X	
T ins rib		X	
T forehead		X	
Temperature insulated			
T ins radial artery		X	
T ins scapula		X	X
T ins sternum		X	
Heart Rate			
HR	X	X	X
Heat Flux			
HF scapula	X	X	
HF rib		X	
HF sternum	X	X	

8. COMMERCIAL WEARABLE CORE BODY TEMPERATURE SYSTEMS

8.1 3M BAIR HUGGER TEMPERATURE MONITORING SYSTEM

This is an accurate, non-invasive, and continuous core temperature monitoring system. It is said to offer the accuracy of invasive methods (such as pulmonary artery blood temperature) along with the benefits of non-invasive monitoring. It is unaffected by ambient conditions and is deployed as a single patient use probe [86]. This system makes use of the zero-heat-flux thermometry method explained in section 3.4. A thermal insulator is applied to the skin to help closely approximate the area of skin covered by the insulator with the tissue temperature beneath the skin. The right amount of heat is applied to the skin surface to oppose the heat transfer and thus create a zone of perfect insulation. The system thus establishes an isothermal pathway to the patient's core by applying a small amount of heat within the sensor to help warm the measurement site until the heater and skin temperature are the same. This sensor is placed on the forehead of the patient. [86,87,88] Some studies have also shown the effectiveness of placing this sensor on the neck region to monitor core body temperature [89]. The control unit continuously displays core body temperature, trends 2 hours of patient temperature data, and is connected to a patient monitor via a simple cable. An important use-case of this system is to manage normothermia through the perioperative journey. During Pre-op, Intra-op and post-op situations, the patient's core body temperature is actively measured and monitored using the system [86,87,88]. Figure 29 (a) [87,88] shows the exploded view of the 3M Bair Hugger temperature sensor patch.

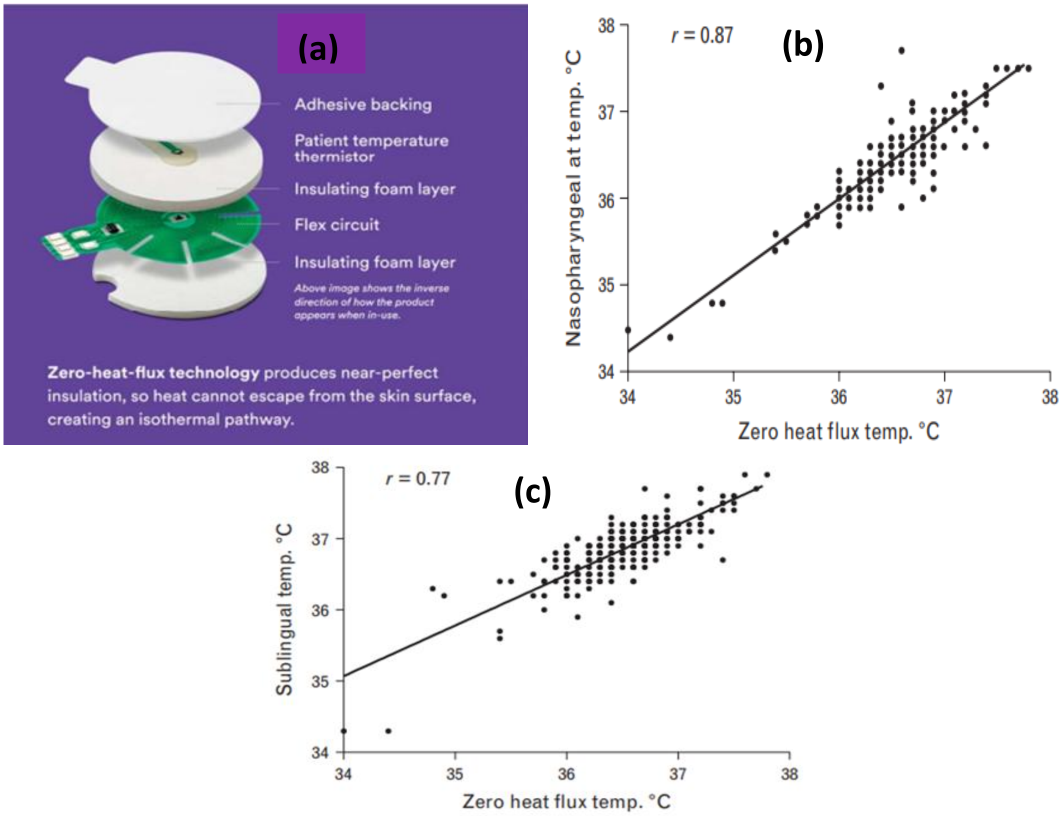


Figure 29: (a) Exploded view of 3M Bair Hugger temperature sensor patch [87,88]. (b) Scatter plot for ZHF vs Nasopharyngeal temp measured at 15, 45 and 75 min in the T. Iden study. (c) Scatter plot for ZHF vs Sublingual temp measured at 15, 45 and 75 min in the T. Iden study. Reproduced from [90] with permission from Wolters Kluwer Health Inc., European Journal of Anaesthesiology.

T. Iden et al [90] conducted a study to assess the Bair Hugger (forehead) system and compare it with sublingual and nasopharyngeal sensors for measuring core body temperature. CBT measurements were taken at 15, 45, and 75 minutes after the induction of anesthesia using nasopharyngeal and sublingual probes, as well as Bair Hugger sensors attached to the forehead using the ZHF method. Sublingual temperature was monitored by lifting the tongue, and nasopharyngeal temperature was monitored by carefully inserting the sensor through a nostril. The study included 83 patients, and the correlation between sublingual and ZHF as well as nasopharyngeal and ZHF, was analyzed using Spearman's rank correlation analysis. The Bland–Altman method was employed to interpret accuracy, precision, and limits of agreement. The temperatures measured by the Bair Hugger sensor using the ZHF method were found to be slightly higher than nasopharyngeal temperatures (mean difference of 0.07°C) and slightly lower than sublingual temperatures by 0.35°C. Spearman rank correlation analysis showed strong correlations between ZHF and nasopharyngeal pairs (r-values of 0.85 at 15 min, 0.88 at 45 min, and 0.89 at 75 min) and moderate to strong correlations between ZHF and sublingual pairs (r-values of 0.69, 0.76, and 0.84). Figure 29 (b) and (c) [90] show the scatterplots for ZHF vs Nasopharyngeal temperatures and ZHF vs Sublingual temperatures at 15, 45 and 75 minutes post anesthesia. Bland–Altman analysis indicated biases of 0.07°C, 0.05°C, and 0.10°C for ZHF vs. nasopharyngeal group at 15, 45, and 75 minutes, respectively. The ZHF vs. sublingual group showed biases of -0.37°C, -0.36°C, and -0.33°C. It could thus be concluded that the Bair Hugger sensor demonstrated good agreement with nasopharyngeal measurements and acceptable agreement with sublingual thermometry in terms of correlation, accuracy, and precision [90].

Y. Eshraghi et al [91] evaluated the ZHF 3M Bair Hugger thermometer for use in cardiac surgical patients by comparison to the gold standard in core body temperature measurement- pulmonary artery catheters. This was when the 3M Bair Hugger thermometer was still in its prototype phase and called the SpotOn. This method was seen as an alternative to uncorrected skin surface temperature measurements. The principle of working of this prototype was entirely based on the ZHF method explained in section 3.4. 105 patients undergoing Cardiopulmonary Bypass (CPB) were enrolled in this study. A proportional-integral-derivative (PID) scheme was used to sustain the zero-heat-flow condition across the probe. The external hardware included two components: (1) circuitry for controlling the heater and limiting current, and (2) circuitry for regulating voltage and measuring data for 3 zero-heat-flux probes and 2 standard clinical thermistor probes. The heater current limiting circuitry was implemented as a precautionary measure to ensure that the skin temperature under the probe does not surpass 41°C in case of software or hardware failures. The Bair Hugger probe was placed on the forehead and another Bair Hugger on the neck.

Uncorrected skin surface temperature was also measured at the forehead with a skin probe from Medtronic. Temperatures were recorded at 1-minute intervals during surgery and also for 4 hours post operation. An acceptable agreement of 0.5°C was set between pulmonary artery and ZHF measurements. Agreement was assessed using the Bland and Altman method. Mean bias, standard deviation, and limits of agreement for both were estimated using the same method and resampling algorithm, taking into consideration the varying number of observations for each patient and their correlation. The ZHF measurements that were within ±0.5°C of the corresponding pulmonary artery measurement were computed using bootstrap resampling and Lin's concordance correlation coefficient was computed. The average overall difference between forehead ZHF and pulmonary artery temperatures was −0.23°C (95% limits of agreement of ±0.82); 78% of the differences were ≤0.5°C. The average temperature during surgery difference was −0.08°C (95% limits of agreement of ±0.88); 84% of the differences were ≤0.5°C. The average postoperative difference was −0.32°C (95% limits of agreement of ±0.75); 84% of the differences were ≤0.5°C. The average overall difference between neck ZHF and pulmonary artery temperatures was −0.30°C (95% limits of agreement of ±0.88); 75% of the differences were ≤0.5°C. The average temperature during surgery difference was −0.15°C (95% limits of agreement of ±0.84); 81% of the differences were ≤0.5°C. The average temperature during surgery difference was −0.40°C (95% limits of agreement of ±0.84); 81% of the differences were ≤0.5°C. Uncorrected forehead skin temperatures were 3.2°C ± 1.4°C lower than pulmonary artery temperature. It could be concluded that the ZHF 3M Bair Hugger system was as accurate as the Pulmonary artery method, and thus a reliable method to be used during cardiac surgery [91].

8.2 GREENTEG (CORE - CALERA SENSORS)

These sensors are non-invasive and allow for the continuous measurement of core body temperature. They are light/compact and can easily be worn. They can be deployed in different form factors such as a chest strap, a wrist-worn smartwatch, a medical monitoring patch on the upper arm, or on the torso. They also have wireless connectivity capability and can be controlled using a smartphone app/smartwatch. The sensor is also a white label product and supports OEM integrations with wearables in medical applications. Shown in figure 30 [93] is the wearable CORE CBT sensor. Some of the use cases for this sensor include patient monitoring (hospitals or at-home), sleep monitoring, fertility tracking, neurodegenerative disease detection, baby monitoring, heat stress prevention and cooling-based/heat training by athletes [92]. Some of the commercial products using this sensor are summarized in the table below.

Table 9: Commercial products using the CORE/Calera Sensor and their form factor

Company/Product	**Form Factor**
Recon Health Virtual Care Patch	Medical Patch on upper arm
Withings Hybrid Scan Watch	Smartwatch on wrist
C-Detect's medical monitor	Wearable on bicep
Corsano Health's Cardiowatch 287	Wristband
Myant Skiin	Smart Garments

Figure 30: CORE CBT Sensor [93]

This sensor also works on the principles of resistance and heat-flux based thermometry. It is made up of a heat flux sensor working on the principle of the Seebeck effect explained in section 3.1.3. As shown in figure 31 (a) [94], the sensor contains a Thermocouple. The thermocouple material and the number of thermocouples used influences the sensitivity of the sensor. The thermopiles of the thermocouple are highly integrated into the substrate of the sensor [94].

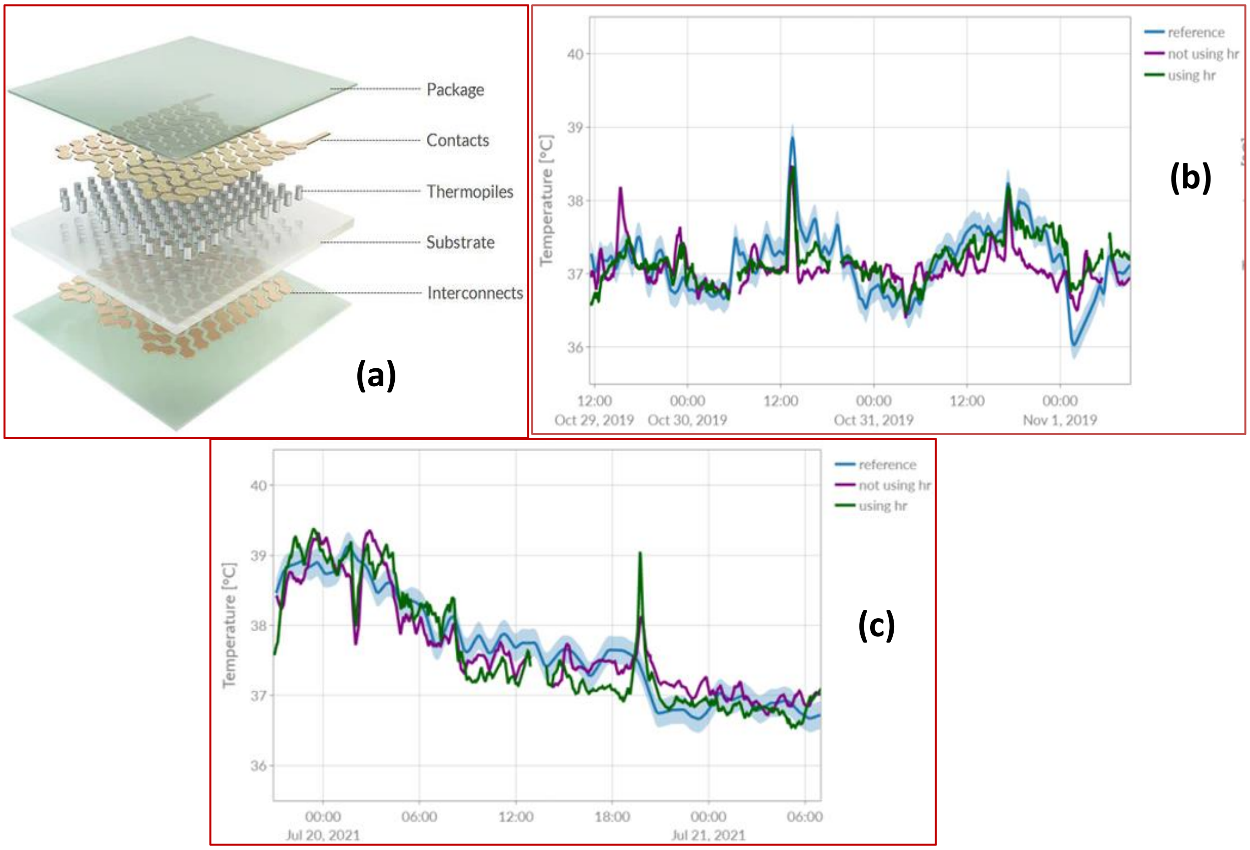


Figure 31: (a) Exploded view within the CORE sensor [94]. (b) Time series comparison on free-living measurements for algorithm 1 (without heart rate) and 2 (with heart rate) [96]. (c) Time series comparison on fever measurements for algorithm 1 and 2 [96].

The CORE body temperature monitor includes two integrated algorithms for determining precise core body temperature readings in various physiological and environmental situations. The 'Everyday Living mode' and the 'Intense Endurance Sports Activity mode'. These algorithms are specifically designed to provide accurate core body temperature values in different scenarios, to address the demands of both routine daily activities and intense endurance sports. The intense endurance sports algorithm makes use of heart rate data as well [95].

A validation study was conducted by S. Etienne et al [96] at the University Hospital of Basel. The goal was to accurately measure core body temperature on the upper arm in free living adults as well as adults with fever. Once again, BodyCap ingestible radio-pills were used as a reference. 63 healthy adults who were given a Covid-19 booster were part of the study. 27 participants developed fever. The average measurement duration for the fever patients was 38 hours while it was 66 hours for the free-living patients. Statistical analysis for both algorithms (with and without heart rate) was carried out and the results are summarized below in table 10. Comparison to the 3M Bair Hugger sensor (which had a 95% interval of ±0.88°C and bias of 0.23°C) described in section 8.1 seems to show that this CORE sensor is more accurate.

Figure 31 (b) and (c) [96] show time series comparisons between the 2 algorithms in free-living measurements and separately in fever measurements. It was observed that the algorithm with heart rate improved prediction accuracy in free-living measurements as compared to the algorithm without heart rate. The heart rate algorithm allows the model to react faster to spikes in physical activity. In the fever measurements, the type of algorithm had no significant effect on the accuracy of measurements.

Table 10: Results of validation study of CORE algorithms on patients [96]

	Algorithm 1 (Without Heart rate)	**Algorithm 2 (With Heart Rate)**
Bias	-0.053°C	0.044°C
Mean Absolute Error	0.341°C	0.323°C
Standard Deviation	0.43°C	0.41°C
Correlation Coefficient	0.635	0.713
Limit of Agreement	±0.86°C	±0.82°C

Another validation study to demonstrate the effectiveness of the CORE sensor in measurement of core body temperature during daily life (free living) was carried out by the CORE internal research team [97]. The sensor was worn on the chest in this study. The BodyCap ingestible radio-pills were used as a reference. The initial reading was taken after 4 minutes, and more than 6 million data points were obtained during activities such as sleeping, working, eating and commuting to work, over a 3-day period. The adults in the study ranged from ages 23 to 63, with 75% of subjects being male. The overall mean absolute error was found to be 0.21°C. The SD was 0.28°C, limit of agreement 0.56°C and the correlation factor was 0.92.

CONCLUSIONS & FUTURE DIRECTIONS

In this survey thesis, the importance of temperature measurements and in particular core body temperature measurements were first established. Several applications ranging from medical monitoring (clinical) to sports performance improvement to using core temperature as a diagnostic tool as well as workplace monitoring were mentioned. Commonly used methods to measure skin surface (peripheral) temperature were compared. These however lack accuracy and are easily influenced by external factors. They are also unsuitable for longer term, continuous monitoring. This made it necessary to measure core body temperature instead, which is a better reflection of the state of the body and its 'true' temperature. The standard core temperature measurement methods were then compared with the Pulmonary Artery method being the gold standard. Different techniques to measure core temperature with particular importance to Resistance-based and Heat-Flux based methods were researched. Some commercially available temperature sensor chips used in wearable devices were described and compared. Important considerations for optimal design and implementation of core temperature sensors as well as best practices for low power consumption have also been discussed. While the core temperature measurement methods discussed were accurate, they are typically very invasive and are almost always used only in a clinical setting or in ill patients. To this end, the results of several research studies aiming to improve upon these measurement and prediction methods have been summarized and reported. The estimation and prediction of core body temperature from surface temperature has been the way forward. The need for continuous monitoring (in clinical and non-clinical settings), comfort, non-invasiveness and at the same time high accuracy has been the driver behind the development of wearable sensors. Wearable core body temperature sensors are on the cutting edge of technology, and there are less than a handful of commercially available surface sensor devices that measure core temperature. The state of the art in wearable core body temperature sensing has subsequently been reviewed towards the end.

Some limitations were identified in the latest research studies conducted, and it is hoped that experiments with testing protocols be carried out along those lines.

a) Compare the performance of different commercial wearable core body temperature sensors.

b) Include a more diverse group of participants (age, gender, physical fitness levels) and who differ in body composition (BMI, body fat %, subcutaneous and visceral fat, muscle mass, bone mass, basal metabolic rate, heart rate, oxygen saturation, etc.).

c) Compare the effects of different environmental conditions, clothing and exercises/sports.

d) Determine if motion artifacts have an effect on temperature readings.

e) Compare commercial CORE sensor with Fitbit and digital thermometer over a 3 week period (Daily living and physical activity).

REFERENCES

[1] NSW, SafeWork. “Core Body Temperature v Peripheral Body Temperature.” *SafeWork NSW*, 20 Sept. 2018, www.safework.nsw.gov.au/resource-library/heat-and-environment/working-in-extreme-heat-the-facts/accordians/core-body-temperature.

[2] Cheshire, William P. “Thermoregulatory Disorders and Illness Related to Heat and Cold Stress.” *Autonomic Neuroscience*, vol. 196, Apr. 2016, pp. 91–104.

[3] Sessler DI. Temperature monitoring and perioperative thermoregulation. Anesthesiology. 2008 Aug;109(2):318-38

[4] Q. Huang, S. Komarzynski, M. Bolborea, B. Finkenstadt, and F. A. Levi, "Telemonitored Human Circadian Temperature Dynamics During Daily Routine," (in English), Front Physiol, vol. 12, May 10, 2021.

[5] S. Haghayegh, M. H. Smolensky, S. Khoshnevis, R. C. Hermida, R. J. Castriotta, and K. R. Diller, "The Circadian Rhythm of Thermoregulation Modulates both the Sleep/Wake Cycle and 24 h Pattern of Arterial Blood Pressure," *Compr Physiol*, vol. 11, no. 4, pp. 2645-2658, Oct 12, 2021.

[6] Asadian S, Khatony A, Moradi G, Abdi A, Rezaei M. Accuracy and precision of four common peripheral temperature measurement methods in intensive care patients. Med Devices (Auckl). 2016 Sep 1; 9:301-8.

[7] Uth, Marc-Florian, et al. “Body Core Temperature Sensing: Challenges and New Sensor Technologies.” Procedia Engineering, vol. 168, 2016, pp. 89–92.

[8] Anton O, Dore H, Rendon-Morales E, Aviles-Espinosa R, Seddon P, Wertheim D, Fernandez R, Rabe H. Non-invasive sensor methods used in monitoring newborn babies after birth, a clinical perspective. Matern Health Neonatol Perinatol. 2022 Nov 22;8(1):9.

[9] Smith J, Alcock G, Usher K. Temperature measurement in the preterm and term neonate: a review of the literature. Neonatal Netw. 2013 Jan-Feb;32(1):16-25.

[10] Nierman DM. Core temperature measurement in the intensive care unit. Crit Care Med. 1991 Jun;19(6):818-23.

[11] Lorenzo S, Halliwill JR, Sawka MN, Minson CT. Heat acclimation improves exercise performance. J Appl Physiol (1985). 2010 Oct;109(4):1140-7, Epub 2010 Aug 19.

[12] Reid KJ. Assessment of Circadian Rhythms. Neurol Clin. 2019 Aug;37(3):505-526. doi: 10.1016/j.ncl.2019.05.001.

[13] Sonia Ancoli-Israel, Chapter 103 - Sleep and Fatigue in Cancer Patients, Editor(s): Meir H. Kryger, Thomas Roth, William C. Dement, Principles and Practice of Sleep Medicine (Fourth Edition), W.B. Saunders, 2005, Pages 1218-1224.

[14] Ozaki N, Iwata T, Itoh A, Kogawa S, Ohta T, Okada T, Kasahara Y. Body temperature monitoring in subjects with delayed sleep phase syndrome. Neuropsychobiology. 1988;20(4):174-7.

[15] Mattis J, Sehgal A. Circadian Rhythms, Sleep, and Disorders of Aging. Trends Endocrinol Metab. 2016 Apr;27(4):192-203. doi: 10.1016/j.tem.2016.02.003. Epub 2016 Mar 3.

[16] Acharya P, Boggess B, Zhang K. Assessing Heat Stress and Health among Construction Workers in a Changing Climate: A Review. Int J Environ Res Public Health. 2018 Feb 1;15(2):247.

[17] US Department of Commerce, NOAA. “What Is the Heat Index?” Www.weather.gov, www.weather.gov/ama/heatindex.

[18] M. Nakayoshi, M. Kanda, R. Shi, and R. de Dear, "Outdoor thermal physiology along human pathways: a study using a wearable measurement system," (in English), *Int J Biometeorol*, vol. 59, no. 5, pp. 503-515, May 2015.

[19] Fever Temperatures: Accuracy and Comparison. Myhealth.alberta.ca, myhealth.alberta.ca/Health/Pages/conditions.aspx?hwid=tw9223.

[20] Lodha R, Mukerji N, Sinha N, Pandey RM, Jain Y. Is axillary temperature an appropriate surrogate for core temperature? Indian J Pediatr. 2000; 67:571–574.

[21] Batra, Prerna, and Sudhanshu Goyal. "Comparison of Rectal, Axillary, Tympanic, and Temporal Artery Thermometry in the Pediatric Emergency Room." Pediatric Emergency Care, vol. 29, no. 1, Jan. 2013, pp. 63–66.

[22] Mogensen CB, Wittenhoff L, Fruerhøj G, Hansen S. Forehead or ear temperature measurement cannot replace rectal measurements, except for screening purposes. BMC Pediatr. 2018 Jan 26;18(1):15.

[23] Piccinini F, Martinelli G, Carbonaro A. Reliability of Body Temperature Measurements Obtained with Contactless Infrared Point Thermometers Commonly Used during the COVID-19 Pandemic. Sensors. 2021; 21(11):3794.

[24] El-Radhi AS, Barry W. Thermometry in paediatric practice. Arch Dis Child. 2006 Apr;91(4):351-6. doi: 10.1136/adc.2005.088831.

[25] Dart RC, Lee SC, Joyce SM, Meislin HW. Liquid crystal thermometry for continuous temperature measurement in emergency department patients. Ann Emerg Med. 1985 Dec;14(12):1188-90.

[26] Scholefield JM, Gerber MA, Dwyer P. Liquid crystal forehead temperature strips. A clinical appraisal. Am J Dis Child. 1982 Mar;136(3):198-201.

[27] Liquid Crystal Forehead Thermometers Now Available as Part of Palmer Wahl's Fever Alert Line | 2020-06-04 | the National Provisioner. Www.provisioneronline.com, www.provisioneronline.com/articles/109459-liquid-crystal-forehead-thermometers-now-available-as-part-of-palmer-wahls-fever-alert-line.

[28] Types of Thermometers, Their Accuracy, and How to Use Them. Healthline, 15 Dec. 2020, www.healthline.com/health/types-of-thermometers/taking-oral-temp

[29] What Is Core Temperature? How Is It Measured? - ONiO. www.onio.com, www.onio.com/article/what-is-core-temperature-how-is-it-measured.

[30] White, Matthew & Bosio, Catharine & Duplantis, Barry & Nano, Francis. (2011). Human body temperature and new approaches to constructing temperature-sensitive bacterial vaccines. Cellular and molecular life sciences: CMLS. 68. 3019-31. 10.1007/s00018-011-0734-2. License Number: 5719080279846. *Copyright © 2011, Springer Basel AG.*

[31] An Engineer's Guide to Temperature Sensing-Temperature sensor design challenges and solutions., page11, www.ti.com/lit/eb/slyy161a/slyy161a.pdf?ts=1705404661884

[32] Y. Seo, T. Dileo, J. B. Powell, J. H. Kim, R. J. Roberge, and A. Coca, "Comparison of estimated core body temperature measured with the Bio Harness and rectal temperature under several heat stress conditions," (in English), J Occup Environ Hyg, vol. 13, no. 8, pp. 612-620, 2016.

[33] Del Bene VE. Temperature. In: Walker HK, Hall WD, Hurst JW, editors. Clinical Methods: The History, Physical, and Laboratory Examinations. 3rd edition. Boston: Butterworths; 1990.

[34] Miller KC, Hughes LE, Long BC, Adams WM, Casa DJ. Validity of Core Temperature Measurements at 3 Rectal Depths During Rest, Exercise, Cold-Water Immersion, and Recovery. J Athl Train. 2017 Apr;52(4):332-338.

[35] David S. Greenes, Gary R. Fleisher, when body temperature changes, does rectal temperature lag? The Journal of Pediatrics, Volume 144, Issue 6, 2004, Pages 824-826.

[36] "Philips Reusable Adult Esophageal/Rectal Temperature Probe with Single Thermistor, ,21075A for Sale – Temperature Probe Manufacturer from China (106238959)." Hkbiocaremed-Com.sell.everychina.com, hkbiocaremed-com.sell.everychina.com/p-106238959-philips-reusable-adult-esophageal-rectal-temperature-probe-with-single-thermistor-21075a.html.

[37] Childs C, Harrison R, Hodkinson CTympanic membrane temperature as a measure of core temperature Archives of Disease in Childhood 1999;80:262-266.

[38] Hymczak H, Gołąb A, Mendrala K, Plicner D, Darocha T, Podsiadło P, Hudziak D, Gocoł R, Kosiński S. Core Temperature Measurement-Principles of Correct Measurement, Problems, and Complications. Int J Environ Res Public Health. 2021 Oct 10;18(20):10606.

[39] Barnett BJ, Nunberg S, Tai J, Lesser ML, Fridman V, Nichols P, Powell R, Silverman R. Oral and tympanic membrane temperatures are inaccurate to identify Fever in emergency department adults. West J Emerg Med. 2011 Nov;12(4):505-11. doi: 10.5811/westjem.2011.2.1963.

[40] NOVATEMP Tympanic Temperature Sensors | NOVAMED USA. Www.novamed-Usa.com, www.novamed-usa.com/tympanic_temperature_sensors.html.

[41] Mekjavić IB, Rempel ME. Determination of esophageal probe insertion length based on standing and sitting height. J Appl Physiol (1985). 1990 Jul;69(1):376-9.

[42] Al-Mukhaizeem F, Allen U, Komar L, Naser B, Roy L, Stephens D, Read S, Kim C, Schuh S. Comparison of temporal artery, rectal and esophageal core temperatures in children: Results of a pilot study. Paediatr Child Health. 2004 Sep;9(7):461-5.

[43] Philips - Esophageal/Rectal Temperature Probe Disposable - HCM1837A. Philips, www.usa.philips.com/healthcare/product/HCM1837A/esophageal-rectal-temperature-probe-disposable-9-fr--sterilized--continuous-monitoring-sensor.

[44] “Smiths Medical Foley Catheter Temperature Sensor.” Www.shopcatheters.com, www.shopcatheters.com/p-smiths-medical-foley-catheter-temperature-sensor.html.

[45] “Thermodilution & Pulmonary Artery Catheters.” Merit Medical, www.merit.com/cardiac-intervention/thermodilution-and-accessories/thermodilution-and-accessories/thermodilution-pulmonary-artery-catheters/.

[46] Lefrant, JY., Muller, L., de La Coussaye, J.E. et al. Temperature measurement in intensive care patients: comparison of urinary bladder, oesophageal, rectal, axillary, and

inguinal methods versus pulmonary artery core method. Intensive Care Med 29, 414–418 (2003)

[47] Wavelength Electronics. “THERMISTOR BASICS | Wavelength Electronics.” Teamwavelength.com, 5 Jan. 2018, www.teamwavelength.com/thermistor-basics/.

[48] “Thermistor: Construction, Working Principle, Types and Applications - Sensors and Transducers - Teachics.” Teachics.org, teachics.org/sensors-and-transducers/thermistor-construction-working-types/.

[49] Vis, Peter J. “Voltage Divider Circuit Calculator - for NTC Thermistor.” Petervis.com, 2021, www.petervis.com/electronics%20guides/calculators/thermistor/thermistor.html.

[50] Helle. “RTD Temperature Sensors - the Fundamentals.” Prelectronics.com, 2013, www.prelectronics.com/the-fundamentals-of-rtd-temperature-sensors/.

[51] Davis, Nick. “Introduction to Temperature Sensors: Thermistors, Thermocouples, RTDs, and Thermometer ICs - Technical Articles.” Www.allaboutcircuits.com, 16 Sept. 2022, www.allaboutcircuits.com/technical-articles/introduction-temperature-sensors-thermistors-thermocouples-thermometer-ic/

[52] Han S, Kim J, Won SM, Ma Y, Kang D, Xie Z, Lee KT, Chung HU, Banks A, Min S, Heo SY, Davies CR, Lee JW, Lee CH, Kim BH, Li K, Zhou Y, Wei C, Feng X, Huang Y, Rogers JA. Battery-free, wireless sensors for full-body pressure and temperature mapping. Sci Transl Med. 2018 Apr 4;10(435): eaan4950. License number: 1445512-1. *Copyright © 2018, The American Association for the Advancement of Science*.

[53] Awati, Rahul. “What Is the Seebeck Effect?” SearchNetworking, www.techtarget.com/searchnetworking/definition/Seebeck-effect.

[54] Technology, Electrical. “What Is the Difference between Thermistor and Thermocouple?” ELECTRICAL TECHNOLOGY, 26 Dec. 2021, www.electricaltechnology.org/2021/12/difference-between-thermistor-thermocouple.html.

[55] E. J. S. Chaglla, N. Celik, and W. Balachandran, "Measurement of Core Body Temperature Using Graphene-Inked Infrared Thermopile Sensor," (in English), Sensors-Basel, vol. 18, no. 10, Oct 2018.

[56] “LMT70.” Ti.com, 2014, www.ti.com/product/LMT70.

[57] “TMP114 Data Sheet, Product Information and Support | TI.com.” Www.ti.com, www.ti.com/product/TMP114.

[58] Ray PP. An IR Sensor Based Smart System to Approximate Core Body Temperature. Journal of Medical Systems. 2017 Aug;41(8):123.

[59] Chen H-Y, Chen A, Chen C. Investigation of the Impact of Infrared Sensors on Core Body Temperature Monitoring by Comparing Measurement Sites. Sensors. 2020; 20(10):2885.

[60] Kesztyüs D, Brucher S, Wilson C, Kesztyüs T. Use of Infrared Thermography in Medical Diagnosis, Screening, and Disease Monitoring: A Scoping Review. Medicina (Kaunas). 2023 Dec 9;59(12):2139.

[61] Moerman A, Wouters P. Near-infrared spectroscopy (NIRS) monitoring in contemporary anesthesia and critical care. Acta Anaesthesiol Belg. 2010;61(4):185-94.

[62] O’BRIEN, CATHERINE, et al. “Telemetry Pill Measurement of Core Temperature in Humans during Active Heating and Cooling.” Medicine & Science in Sports & Exercise, vol. 30, no. 3, Mar. 1998, pp. 468–472.

[63] J.E. McKenzie, D.W. Osgood, Validation of a new telemetric core temperature monitor, Journal of Thermal Biology, Volume 29, Issues 7–8, 2004, Pages 605-611, ISSN 0306-4565.

[64] Monnard, Cathriona R., et al. “Issues in Continuous 24-h Core Body Temperature Monitoring in Humans Using an Ingestible Capsule Telemetric Sensor.” Frontiers in Endocrinology, vol. 8, 13 June 2017.

[65] Raiko J, Koskensalo K, Sainio T. Imaging-based internal body temperature measurements: The journal Temperature toolbox. Temperature (Austin). 2020 May 29;7(4):363-388.

[66] Yamakage, Michiaki and Akiyoshi Namiki. “Deep temperature monitoring using a zero-heat-flow method.” Journal of Anesthesia 17 (2003): 108-115. License number: 1445200-1. Copyright © 2003, Springer Japan KK.

[67] Teunissen, Lennart & Klewer, J & Haan, Arnold & de Koning, Jos & Daanen, Hein. (2011). Non-invasive continuous core temperature measurement by zero heat flux. Physiological measurement. 32. 559-70. 10.1088/0967-3334/32/5/005.

[68] H. Anuar and P. L. Leow, "Non-invasive Core Body Temperature Sensor for Continuous Monitoring," (in English), 2019 Ieee International Conference on Sensors and Nanotechnology (Sn), pp. 61-64, 2019. Copyright © 2019, IEEE.

[69] Kei-Ichiro Kitamura, Xin Zhu, Wenxi Chen, Tetsu Nemoto, Development of a new method for the noninvasive measurement of deep body temperature without a heater, Medical Engineering & Physics, Volume 32, Issue 1, 2010, Pages 1-6, ISSN 1350-4533.

[70] J. J. Feng, C. C. Zhou, C. He, Y. Li, and X. S. Ye, "Development of an improved wearable device for core body temperature monitoring based on the dual heat flux principle," (in English), Physiol Meas, vol. 38, no. 4, pp. 652-668, Apr 2017. License number: 1445521-1,1445525-1. Copyright © 2017, IOP Publishing.

[71] Mark J. Buller, William J. Tharion, Cynthia M. Duhamel & Miyo Yokota (2015) Real-time core body temperature estimation from heart rate for first responders wearing different levels of personal protective equipment, Ergonomics, 58:11, 1830-1841.

[72] Yartsev, Alex. “Fick’s Principle of Cardiac Output Measurement | Deranged Physiology.” Derangedphysiology.com, 30 June 2015, derangedphysiology.com/main/cicm-primary-exam/required-reading/cardiovascular-system/Chapter%20811/ficks-principle-cardiac-output-measurement.

[73] “Miniature Digital Infrared Thermometer IC in Surface-Mount Technology.” Melexis, www.melexis.com/en/product/MLX90632/Miniature-SMD-Infrared-Thermometer-IC.

[74] “ADT7320 Datasheet and Product Info | Analog Devices.” Analog.com, 2015, www.analog.com/en/products/adt7320.html.

[75] “MAX30208 Datasheet and Product Info | Analog Devices.” Analog.com, 2019, www.analog.com/en/products/max30208.html.

[76] Ahmad, Majeed. “Quickly Integrate Clinical-Grade Temperature Sensing into Portable, Wearable Medical Designs.” DigiKey, DigiKey’s North American Editors, 24 June 2020, www.digikey.com/en/articles/quickly-integrate-clinical-grade-temperature-sensing-into-medical-designs.

[77] “I2C - Learn.sparkfun.com.” Sparkfun.com, 2018, learn.sparkfun.com/tutorials/i2c/all.

[78] “What Is SMBus?” Total Phase Blog, 13 Jan. 2021, www.totalphase.com/blog/2021/01/what-is-smbus/.

[79] “Pull-up Resistor and Pull-down Resistor Explained.” Basic Electronics Tutorials, 30 Aug. 2016, www.electronics-tutorials.ws/logic/pull-up-resistor.html.

[80] Schackmann, Ren. Low-Power Design Techniques for Temperature-Sensing Applications Application Report Low-Power Design Techniques for Temperature-Sensing Applications. 2019.

[81] Parks, Michael. “Pull up Your Pins: How to Size Pull-up Resistors | Bench Talk.” Mouser.com, 2018, www.mouser.com/blog/pull-up-your-pins-how-to-size-pull-up-resistors.

[82] “How Can You Use Algorithms to Optimize Energy Usage on Mobile Devices?” www.linkedin.com, www.linkedin.com/advice/0/how-can-you-use-algorithms-optimize-energy-usage-vsfjc.

[83] Niedermann R, Wyss E, Annaheim S, Psikuta A, Davey S, Rossi RM. Prediction of human core body temperature using non-invasive measurement methods. Int J Biometeorol. 2014 Jan;58(1):7-15.

[84] Eggenberger P, MacRae BA, Kemp S, Bürgisser M, Rossi RM, Annaheim S. Prediction of Core Body Temperature Based on Skin Temperature, Heat Flux, and Heart Rate Under Different Exercise and Clothing Conditions in the Heat in Young Adult Males. Front Physiol. 2018 Dec 10; 9:1780.

[85] Laxminarayan, Srinivas, et al. “Individualized Estimation of Human Core Body Temperature Using Noninvasive Measurements.” Journal of Applied Physiology, vol. 124, no. 6, 1 June 2018, pp. 1387–1402.

[86] “3M Bair Hugger Temperature Monitoring | Infection Prevention | 3M United States.” Bairhugger.com, MMM-ext, 2024, www.bairhugger.com/3M/en_US/bair-hugger-us/core-temp/.

[87] Own the Zone: Bring Visibility to a Critical Vital Sign with Continuous Core Temperature Monitoring. 3M, https://multimedia.3m.com/mws/media/1568726O/paper-on-the-importance-of-temp-monitoring.pdf/?utm_term=hcbg-ipd-warming-en_us-lead-ownthezone-own-3mcom-temp_monit_wp_lp-whitepp-na-jul18&elqTrackId=b44e0f9072da476fb15a1a2ed27e851d&elqaid=7604&elqat=2

[88] “3M Bair Hugger Temperature Monitoring | Infection Prevention | 3M United States.” Bairhugger.com, MMM-ext, 2024, www.bairhugger.com/3M/en_US/bair-hugger-us/core-temp/.

[89] Tachibana S, Chida Y, Yamakage M. Using the Bair Hugger™ temperature monitoring system in neck and chest regions: a pilot study. JA Clin Rep. 2019 May 16;5(1):32.

[90] T. Iden, Timo; Horn, Ernst-Peter; Bein, Berthold; Böhm, Ruwen; Beese, Janne; Höcker, Jan. Intraoperative temperature monitoring with zero heat flux technology (3M SpotOn sensor) in comparison with sublingual and nasopharyngeal temperature: An observational

study. European Journal of Anesthesiology 32(6): p 387-391, June 2015. License Number: 5719490642669. *Copyright © 2015, European Society of Anaesthesiology*.

[91] Eshraghi Y, Nasr V, Parra-Sanchez I, Van Duren A, Botham M, Santoscoy T, Sessler DI. An evaluation of a zero-heat-flux cutaneous thermometer in cardiac surgical patients. Anesth Analg. 2014 Sep;119(3):543-549.

[92] "Patient Monitoring - Greenteg." www.greenteg.com, www.greenteg.com/en/cases/patient-monitoring.

[93] "CORE Body Temperature Sensor." CORE, corebodytemp.com/.

[94] "The Technology behind CORE - Miniaturized Thermal Energy Transfer Sensors." CORE, corebodytemp.com/blogs/news/the-technology-behind-core-thermal-energy-transfer-sensors.

[95] "Core Body Temperature Algorithm - Greenteg." Www.greenteg.com, www.greenteg.com/de/faqs/detail/core-body-temperature-algorithm.

[96] "Validation Study: Upper Arm Fever Algorithm." Greenteg.com, greenteg, cms.greenteg.com/uploads/20220421_Validation_upper_Arm_a75831c9f6.pdf.

[97] "Validation Study: Real Life Condition Chest Algorithm." Greenteg.com, greenteg, cms.greenteg.com/uploads/20222104_Validation_Chest_Real_Life_Conditions_3d9a69dba0.pdf.