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Investigation of a non-invasive venous blood flow measurement device

Using thermal mass measurement principles

Abstract: Venous blood circulation can be restricted due to various conditions commonly indicating a related medical condition. However, current non-invasive methods for determining venous blood flow are limited to be either very inaccurate or expensive. Alternatively, a method to measure sap flow non-invasively in trees is through thermal mass measurement principles. This paper investigates applying the thermal mass flow measurement principle to determine venous blood flow.

A simplified finite element model (FEM) and simulation are created to determine the operating behavior and expected response of a thermal mass flow meter with venous blood flow under the skin. An initial prototype of a thermal mass venous blood flow meter is designed using a Peltier-element and RTD thermistors. Initial tests were done on $N = 8$ subjects identifying the presence of blood flow and, testing the devices basic functionality and performance.

The simplified FEM model of venous blood flow proved the thermal mass blood flow device is feasible, and determined the initial characteristics of the first prototype. The initial prototype proved to be functional detecting rises in temperature downstream of $+1.4\text{ K}$ ($0.8 - 1.8$ inter-quartile range) when the blood flow was released ($t = 90\text{ s}$ after release), compared to when blood was not flowing.

The initial prototype proved to be able to detect the presence of blood flow in all subjects. However, further work is required to increase the differences in temperature values or gradient measured for a change in flow rate so the actual flow rate can be determined.

Keywords: venous blood flow, thermal mass measurement, vascular thrombosis, circulatory system.

<https://doi.org/10.1515/cdbme-2019-0045>

1 Background

The circulatory system is fundamental to the body's functionality transporting necessary nutrients and oxygen to cells within organs and peripheral limbs. Therefore, dysfunctions in the circulatory system are commonly an indication or cause of specific medical conditions or diseases. Specifically, changes in venous flow can indicate shock [1], dehydration [2], vein thromboses [3], and a general decrease in peripheral perfusion [4]. In addition, restricting blood flow to peripheral limbs has been shown to induce muscle hypertrophy and thus increase strength gain with low intensity resistance training (KAATSU training) [5]. However, during KAATSU training the blood flow rate is not usually measured, only assumed from pressure cuff measurements [5].

Currently, vascular Doppler ultrasound is the most common technique for determining venous blood flow. However, the Doppler ultrasound is currently expensive, depends heavily on probe orientation and requires the assumption of vessel cross sectional area and average velocity to determine the mass flow rate. Additionally, other techniques such as the capillary nail re-fill test only give limited information about the venous flow rate.

Previous studies have shown the principle of a thermal mass (thermo-dilution) measurement system to be able to monitor the venous flow rate to within 5 % accuracy invasively within the vein [6], [7]. However, these studies required a catheter insertion and saline injected as a thermal mass. Alternatively, the sap flow in trees is able to be determined by heating a plant stem externally and measuring the temperature difference up and down stream [8]. Thus, non-invasively monitoring the flow within the stem.

This paper investigates creating a venous blood flow measurement device using principles of thermal mass measurement similar to those seen in thermo-dilution and sap measurements. A finite element model (FEM) is developed to prove the concept, and from these results, a first prototype is presented and initial tests are performed.

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2 Methods

2.1 FEM model simulation

A simplified FEM has been created to determine the operating behavior and expected response of a thermal mass flow meter by simulations. The simplified model consists of bone ($r = 20$ mm), muscle (20 mm thick), a vein ($\varnothing = 0.25$ –4 mm) tangential to the skin surface, skin (1.4 mm thick [9]), and an aluminum thermal element directly in contact with the skin surface (25x25x10 mm). Note, adipose tissue is not considered due to there likely being a negligible amount present in the areas of measurement. Thermal properties were taken from the COMSOL database (COMSOL Multiphysics®, Stockholm, Sweden) [10]. The complete geometry can be seen in Figure 1.

Figure 1: FEM model consisting of bone, muscle, skin, blood and heat source.

Venous blood flow through vein is simulated as continuous Laminar flow [6] ranging from 0 – 20 cm.s^{-1} . The temperature is evaluated downstream from the thermal element (distance ranging from 0 – 20 mm). Note, the thermal influence of the temperature measurement probe is ignored. The skin surface was additionally modelled with free heat convection to account for ambient temperature cooling. The initial temperature of all of the tissues and the venous blood was set to 37 °C. The skin surface was initially set to 24 °C and ambient temperature set to 23 °C.

Due to the bodies heat regulation mechanisms for cooling and heating [11], heating was chosen to induce a thermal mass into the blood flow. To maximize the thermal mass transported the element temperature was set to the maximum safe temperature possible before cell damage occurred, 42 °C [12].

2.1.1 Device parameter's simulation

The characteristics of the thermal mass input device were investigated in simulation. Specifically, the length of the heating element and the distance of skin temperature measurement downstream were varied to determine the parameters that would result in the largest change in measured temperature for a change in venous flow rate. For these

simulations, a vessel diameter of 4 mm and velocity of 10 cm.s^{-1} was used. All values are evaluated at steady state ($t = 90$ s) and compared to the no heating and/or no venous blood flow (control) scenario.

2.1.2 Physiological parameters simulation

In addition, the variation due to expected inter-patient physiological differences are simulated. Specifically, the venous blood flow velocity and vessel diameter are investigated concerning the variation in the measured skin surface temperature. From these results, the required measured temperature resolution and accuracy can be determined. The parameters are investigated over the range typical and/or expected range to be used with the designed device. All values are evaluated at steady state ($t = 90$ s) and compared to the no heating and/or no venous blood flow (control) scenario.

2.2 Initial prototype

Based on the results of the FEM simulation an initial prototype device has been made. The prototype consists of an aluminum block with a Peltier element placed on top, and RTD temperature sensors (TSic 506-T092 Temperature sensor, B+B Thermo-Technik GmbH) placed 10 mm away from the edge and inside of the aluminum block. The temperature sensors are calibrated for 5–45 °C and are accurate to ± 0.1 K [13]. The device is held together with a 3D printed housing and controlled with a Raspberry Pi 3B+. The complete device can be seen in Figure 2.

Figure 2: Initial prototype of the thermal mass venous blood flow measurement device. (Left) CAD model (Right) Actual prototype.

2.2.1 Functionality test

The device's functionality and the results of the FEM simulation are initially evaluated through a flow vs no flow test. The flow vs no flow experiment consists of the following steps: 1. Placing the device (42 °C) on the blocked vein (pressing on the targeted vein upstream). 2. Wait for steady state temperature ($t \sim 90$ s). 3. Release vein. 4. Wait for steady state temperature ($t \sim 90$ s).

The experiment is performed on $N = 8$ different subjects. The device was placed on visible veins found on the subject's forearm while they were sitting rested at a desk. The change in

temperature (from no flow steady state) is then compared between subjects and to the previous simulation results in terms of absolute difference and approximate rise time.

3 Results

3.1 FEM model simulation

3.1.1 Device parameters simulation

Figure 3 shows that in the scenario of no blood flow, as the distance increases the temperature decreases rapidly. In addition, when the venous blood flow is increased the temperature of the skin further away from the heat source increases as a result of the added thermal mass to the blood. The point of largest difference in temperature between heated no flow and flow, and where the influence of thermal conduction of the heat source through the skin is approximately negligible is found to be at approximately 10 mm. Therefore, this position was chosen for the temperature sensor and further simulations.

Figure 3: FEM simulation comparing measurement distance from the heating element to the measured temperature for an expected flow (10 cm/s) and no flow. The purple region indicates the region covered by the heating element.

The heating element size (length over the blood vessel) on the measured temperature seen downstream was also observed to increase the measured temperature. Therefore, a trade-off of device usability (Overall device size) and the amount of heat able to be input is present. Due to the typical length of visible veins being not longer than approx. 30 mm and to reduce device complexity a length of 25 mm (the size of the standard Peltier element) was chosen.

3.1.2 Physiological parameters simulation

Based on the previous simulation results, the following simulations used a temperature sensor 10 mm away from the heating element and a heating element 25 mm in length.

From Figure 4 it can be seen that the change in temperature (gradient) decreases rapidly as the flow rate of the venous blood is increased. A large change in temperature ($0.9 - 0.95 \text{ K}$) is seen when comparing to the no flow case, however, only very small differences ($<0.1 \text{ K}$) in temperature are seen between differing flow rates above 4 cm/s .

Figure 4: FEM simulation comparing the temperature measured for differing flow rates ($0\text{-}20 \text{ cm/s}$) in 4mm diameter vein.

Various diameter blood vessels ($\varnothing = 0.25\text{-}4 \text{ mm}$) were simulated with 10 cm/s blood velocity. The temperature could be seen to vary up to 0.25 K in diameters less than 1.5 mm . However, in diameters greater than 1.5 mm (visible veins under the skin) the variation was reduced to 0.07 K .

3.2 Functionality test

The flow vs no flow experiment was successfully performed on 8 subjects and their results compared. The normalized temperature changes from releasing the blocked vein can be seen in Figure 5. Across all subjects a median rise in temperature of 1.4 ($0.8\text{-}1.8$ inter-quartile range) was observed 90 s after the vein was unblocked.

Figure 5: Flow vs. no flow experimental results ($N = 8$) using the initial designed prototype. Prior to $t = 90 \text{ s}$ the vein is blocked. After $t = 90 \text{ s}$ the vein is released. The heating element is applying 42°C during the entire experiment.

4 Discussion

The FEM simulation was shown to behave as expected heating the venous blood flow and transporting the thermal mass downstream to the measurement point (Figure 3). Figure 4 indicated that a relatively large difference in temperature would be observed between flow and no flow, however, only very small differences ($<0.05 \text{ K}/\text{cm s}^{-1}$) could be seen between velocities greater than 5 cm s^{-1} . This is likely the result of very little time being spent underneath the heating element and temperature sensor, thus less heat able to be transferred into and out of a given volume of blood. In general, limiting the ability for any ‘standard’ temperature sensor to define these specific velocity values. In addition, the inter-subject variation in temperature that could occur due to differing vessel diameter proved to be a significant factor. As the variation is equal to or greater than the change in temperature from flow rate changes, it is not possible to separate what the cause of the change in temperature is when measuring only absolute temperature values.

The initial prototype was shown to function as expected, adding and detecting the thermal mass in the venous blood vessels. From the flow vs no flow experiments (Figure 5) an increase in temperature could be seen for all patients due to transported thermal mass. However, a large variation response (inter quartile range 0.8–1.8 K) showed that the behavior could not be generalized across subjects.

In both the FEM simulation and experiment a relatively large increase in temperature was observed when the flow was present (FEM 0.7 K vs Exp. 1.4 K), assuming 20 cm s^{-1} and 4 mm diameter vein was present in the subjects measured. Therefore, showing that the presence of flow could be detected with a low accuracy temperature sensor.

In addition, in both the FEM simulation and initial functionality test a large time constant was observed with the system not reaching steady state until at least 90 s, or even longer for some subjects (Figure 5), had passed. This is likely due to the thermal capacitance properties of skin and muscle considerably reducing the system response time. As a result, the response time of such a device is significantly limited and unsuitable for any current potential applications.

When using the initial prototype a few unforeseen issues were encountered. For functional tests, it proved difficult to find a straight, $>20 \text{ mm}$ long, visible vein to ensure the vein was heated and measured appropriately. Visibility was required to ensure the temperature sensor was on the vein. In addition to the positioning issues, it was difficult to place the device on the subject ensuring contact of the heating element and thermistor, but not compressing the target vein to be

measured. It should however be noted that no reports of discomfort due to heating was recorded by subjects.

Currently, the device is able to detect the presence of blood flow. Further work looking into potential pulsing of temperature or a combination of cooling and heating is required to create larger differences in temperature values or gradients able to quantify the venous flow rate.

Overall, a simple FEM model was created to show the potential application of a thermal mass flow meter to measuring venous blood flow. In addition, an initial prototype was created to confirm simulation results and prove device functionality. Initial experimental results showed the prototype was able to detect the presence of blood flow but was unable to distinguish small changes in flow rate. Further work is required to increase the differences in temperature values or gradient measured for a change in flow rate.

Author Statement

Research funding: The author states no funding involved. Conflict of interest: Authors state no conflict of interest. Informed consent: Informed consent has been obtained from all individuals included in this study.

References

- [1] J.-L. Vincent and D. De Backer, “Circulatory Shock,” *N. Engl. J. Med.*, vol. 369, no. 18, pp. 1726–1734, Oct. 2013.
- [2] J. Gonzalez-Alonso, J. Calbet, and B. Nielsen, “Muscle blood flow is reduced with dehydration during prolonged exercise in humans,” *J. Physiol.*, pp. 895–905, 1998.
- [3] F. Rosendaal, “Venous thrombosis: a multicausal disease FR,” *Lancet*, vol. 353, pp. 1167–73, 1999.
- [4] A. Lima and J. Bakker, “Noninvasive monitoring of peripheral perfusion,” in *Applied Physiology in Intensive Care Medicine*, Berlin, Heidelberg: Springer Berlin Heidelberg, 2006, pp. 131–141.
- [5] T. Abe, C. F. Kearns, and Y. Sato, “Muscle size and strength are increased following walk training with restricted venous blood flow from the leg muscle, Kaatsu-walk training,” *J. Appl. Physiol.*, vol. 100, no. 5, pp. 1460–6, 2006.
- [6] C. Clark, “A local thermal dilution flowmeter for the measurement of venous blood flow in man,” *Med. Biol. Eng.*, vol. 6, no. 2, pp. 133–142, 1968.
- [7] P. Andersen and B. Saltin, “MAXIMAL PERFUSION OF SKELETAL MUSCLE IN MAN,” *J. Physiol.*, vol. 366, pp. 233–249, 1985.
- [8] D. M. Smith and S. J. Allen, “Measurement of sap flow in plant stems,” *J. Exp. Bot.*, vol. 47, no. 12, pp. 1833–1844, 1996.
- [9] S. B. Wilson and V. A. Spence, “A tissue heat transfer model for relating dynamic skin temperature changes to physiological parameters,” *Phys. Med. Biol.*, vol. 33, no. 8, pp. 895–912, 1988.
- [10] “COMSOL Multiphysics®,” COMSOL AB, Stockholm, Sweden.
- [11] T. H. Benzinger, “ON PHYSICAL HEAT REGULATION AND THE SENSE OF TEMPERATURE IN MAN,” 1959.
- [12] A. R. Moritz and F. C. Henriques, “Studies of Thermal Injury: II. The Relative Importance of Time and Surface Temperature in the Causation of Cutaneous Burns,” *Am. J. Pathol.*, vol. 23, no. 5, pp. 695–720, Sep. 1947.
- [13] B+B Thermo-Technik GmbH, “TSIC™ Digital semiconductor temperature sensor TSIC 506,” 2014.