MODEL OF A CALORIMETRIC SENSOR FOR MEDICAL APPLICATION

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A calorimetric sensor has been developed to measure surface heat dissipations in the human body. An experimental prototype has been built in order to study its performance and a simple model that represents acceptably the experimental system has been proposed.

Keywords: *calorimetric model, heat-conduction calorimetry, medical calorimetry*

Introduction

The instrumentation for the thermal measure has been developed mainly in the thermal analysis area. In this area, countless devices have been constructed: one for every need and with the typical characteristics of the process that is intended to be studied [1, 2]. In calorimetry, it is necessary for the energetic process that is intended to be measured to be located in a controlled zone where the detector system output can be clearly related to the power and/or energy developed in this zone.

The aim that has been proposed is to measure the heat dissipations developed in localized zones of the human body; in this case, the dissipation intended to be measured is not located in the inner part of the measurement instrument but in the outer part, that is the reason why the measurement process is out of the calorimetry standards. However, it has been constructed an experimental prototype whose operating principle is based on the heat transference by conduction law. Besides, as this instrument has a thermostat that remains at a constant temperature, isothermal calorimeters can be included within the heat conduction [3].

In this paper, it is shown an experimental prototype and a model that represents acceptably its performance. It is made a study of the influence of each one of the parameters of the model on the sensitivity of the device. The direct application of the sensor is to measure power variation and energy levels developed in localized zones of the human body. Finally, it is indicated the importance that this instrument has in medical applications and it is made a proposal of necessary improvements for clinical experimentation.

Experimental prototype

The calorimetric sensor is made up of a TEC1-12704 thermopile of 40×40×5 mm and 127 thermocouples. This thermopile is located between the thermostat and a 70×70 mm aluminium plate of 1 mm of thickness that is leaned on the zone where the dissipation takes place, i.e., the zone which is intended to be studied. The system is surrounded by a thermal insulating body. The thermostat is constituted by a 40×40×5 mm small aluminium block leaned against a heating plate which is in contact with a cooling thermopile that reduces the ambient temperature. The cooling thermopile has an embedded dissipator with the corresponding fan. In Fig. 1, it is shown a diagram of the experimental system.

A programmable power supply (Agilent E3640A) feeds the heating plate that dissipates the power necessary to keep the desired temperature, a simple PID controller determines the power to be dissipated. The temperature is measured by a HP34401A multimeter through a RTD 100W30 by Omega that is located in the 40×40×5 mm aluminium block. The thermostat temperature remains constant and the oscillations are lower than 0.01 K in stationary state as well as when a calibration measurement is being carried out.

The calorimetric signal is directly read by a HP34401A multimeter (10 nV of resolution). For the calibration, a constant resistance (77 Ω) is placed behind the aluminium plate as it is shown in Fig. 1. A programmable power supply (HP6284A/59501B) feeds the calibration resistance and determines the dissipated power at every instant, other HP3478A multimeter reads the voltage in a 1 Ω resistance standard that is in series with the calibration resistance. All the instruments are controlled by a computer through the GPIB bus. The sampling period of the measures is 1 s.

The manufactured experimental device is a laboratory prototype that is not still suitable for clinical research. With this prototype, it is intended to simulate its operation in medical application. In Fig. 1, it is distinguished between part A which represents the measuring device and part B which represents the zone where the thermal process to be

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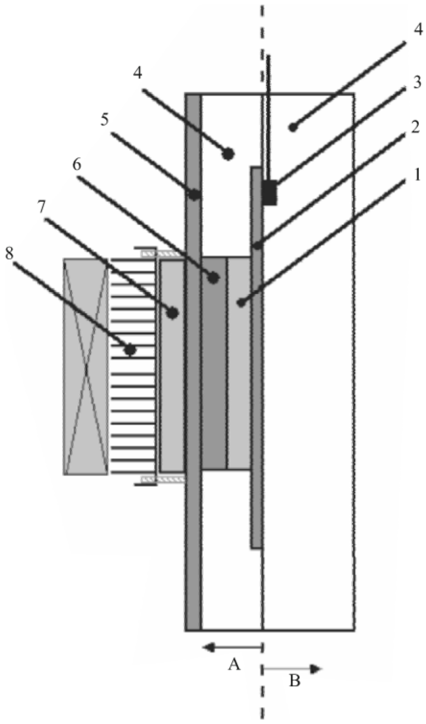

Fig. 1 Experimental prototype: 1 – measurement thermopile , 2 – aluminium plate, 3 – calibration resistance, 4 – thermal insulation, 5 – heating plate, 6 – aluminium block with RTD sensor, 7 – Peltier element, 8 – dissipator with fan. Part A with all its elements represents the Calorimetric Sensor and part B the dissipation zone

measured is generated. In medical application, part B represents a zone of the human body where it is being developed a process whose calorific energy is intended to be studied.

Model

For the modellization of the experimental system, it has been chosen the method ‘at localized constants’ [4], this method has given good results for explaining the performance of different calorimeters [5, 6]. The size of the model is determined with some previous electrical calibrations, in which it is checked that a model given by a two-pole transference function is enough to rebuild the calorimetric curves acceptably. This is the reason why it is proposed a two-body model. Figure 2 shows a diagram of the model where C_1 and C_2 are the heat capacities of the sensor and the dissipation zone, T_1 and T_2 are the temperatures of each body. P_1, P_2 and P_{12} are the thermal conductivities. T_{01} is the programmed temperature of the thermostat and T_{02} is the temperature in the dissipation surroundings (inner part of the human body).

Figure 2 shows the model at localized constants and its relationship with the main elements of the

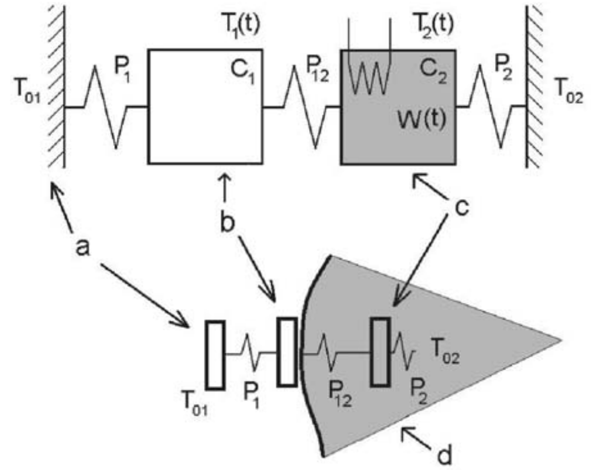

Fig. 2 Model of a calorimetric sensor. a – thermostat, b – calorimetric sensor, c – dissipation zone, d – inner part of the human body

experimental device which are the sensor (C_1) and the thermostat (T_{01}) represented in Fig. 2 by letters a and b. In Fig. 2, letter c represents the zone of the human body where the thermal process to be studied takes place, for example, the one corresponding to a tumour. This zone (C_2) is in contact with the sensor through the thermal coupling of the thermal conductivity P_{12}. It is also in contact, through the thermal coupling of the thermal conductivity P_{12}, with the rest of the human body which is at a temperature T_{02}.

The energetic balance for each domain provides the model equations:

$$\begin{aligned} 0&=C_1\frac{\mathrm{d}T_1(t)}{\mathrm{d}t}+P_{12}(T_1-T_2)+P_1(T_1-T_{10}) \\ w(t)&=C_2\frac{\mathrm{d}T_2(t)}{\mathrm{d}t}+P_{12}(T_2-T_1)+P_2(T_2-T_{20}) \end{aligned} \tag{1}$$

The power developed is $w(t)$, and the output is $y(t)=k(T_1-T_{01})$, where k=0.1 V K^{-1}. It is easy to relate the power with the output through a differential equation of the type:

$$w(t)-P_2(T_{01}-T_{02})=a_2\frac{\mathrm{d}^2y(t)}{\mathrm{d}t}+a_1\frac{\mathrm{d}y(t)}{\mathrm{d}t}+a_0y(t) \tag{2}$$

where

$$\begin{aligned} a_2&=C_1C_2/P_{12} \\ a_1&=C_1+C_2+C_1P_2/P_{12}+C_2P_1/P_{12} \\ a_0&=P_1+P_2+P_1P_2/P_{12} \end{aligned}$$

In this equation, the term $P_2(T_{01}-T_{02})$ affects the baseline that is corrected after carrying out the measurement. If it is made $w^*(t)=w(t)-P_2(T_{01}-T_{02})$, the transference function can be exposed in the following way:

Table 1 Model parameters represented in Fig. 2 and in Eqs (1) and (3). The calorimetric output is $y(t)=k(T_1-T_{01})$ where k=0.1 V K^{-1}

P_1/W K^{-1}	P_2/W K^{-1}	P_{12}/W K^{-1}	C_1/J K^{-1}	C_2/J K^{-1}	S/mV W^{-1}	τ_1/s	τ_2/s
0.2176	0.1880	0.0840	31.5504	4.1690	112	116	15

$$TF(s)=\frac{Y(s)}{W^*(s)}=\frac{S}{(1+s\tau_1)(1+s\tau_1)} \quad (3)$$

being $Y(s)$ and $W^*(s)$ the Laplace transforms of $y(t)$ and $w^*(t)$; S is the sensitivity; τ_1 and τ_2 are the time constants.

Calibration and model identification

The electrical calibration has been the dissipation of a pulse train of 250, 500 and 100 mW during 500 s each pulse (Fig. 3). Besides, the dissipation position has been changed, placing it in different points of the aluminium plate. For the determination of the model parameters, it has been used the simplex search algorithm method by Nelder and Mead [7] using the software MatLab [8], the error criterion chosen to minimize was the standard deviation. Table 1 shows the identification results, i.e., the values of the obtained model parameters. Depending on the dissipation place, the coupling P_{12} ranges from 0.095 to 0.073 W K^{-1} providing a sensitivity of 112±7 mV W^{-1}. In the adjustment between the experimental curve and the calculated, it has been obtained a standard deviation that ranges between 0.3 and 0.9 mV, the number of points used in the adjustment was 10800. It has to be taken into account that the baseline of the experimental curves has some low-frequency oscillations of ±0.5 mV due to the thermal oscillations of the laboratory.

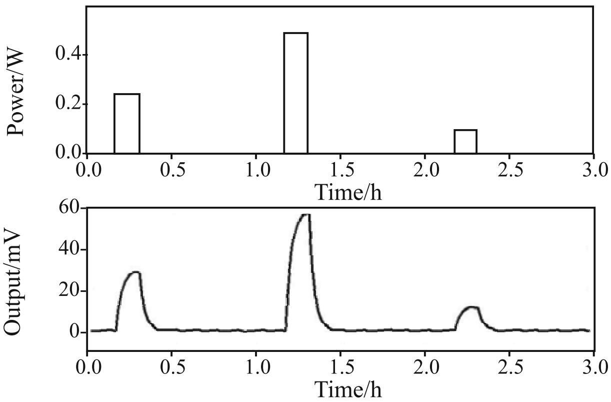

Fig. 3 Calibration measure

Results and discussion

The model parameters have been determined. Logically, the sensor fixed parameters are P_1 and C_1, the rest of the parameters are characteristic of the measurement zone: proximity of the dissipation zone to the sensor (P_{12}), thermal insulation of the dissipation from the rest of the human body (P_2) and the extension of the dissipation zone (C_2).

If all the model parameters are fixed, it is observed that the increase of the thermal conductivity between the dissipation and the sensor (P_{12}) produces an increase of the sensitivity. Experimentally, the sensor has been calibrated by locating the dissipation at different points of a same 7×7 cm^2 surface obtaining a sensitivity of 112±7 mV W^{-1}.

A decrease of P_2 produces an increase of the sensitivity, i.e., when the majority of the heat goes to the sensor, the sensitivity increases. In this case, the thermal insulation of the dissipation zone to the outer part must be increased. An experimental check of this fact has been carried out, reducing the thermal insulation between the aluminium plate where the dissipation is leaned and the outer part. The model for this new situation has been identified by obtaining a value of P_2=0.2509 W K^{-1}, the increase of P_2 has produced a sensitivity reduction of 20% (89.4 mV W^{-1}). Moreover, increasing P_2 has made the oscillations of the baseline be bigger due to the term $P_2(T_{01}-T_{02})$ of the Eq. (2), thus, the oscillations width of the baseline changed from ±0.5 mV (for P_2= 0.1880 W K^{-1}) to ±1.2 mV (for P_2=0.2509 W K^{-1}).

An increase in the mass affected by the dissipation would produce an increase of C_2. The simulation with this model predicts that an increase of C_2 in 100%, keeps constant the sensitivity S, the first time constant τ_1 increases slightly from 116 to 118 s, but the second time constant τ_2 would have a relatively significant increase from 15 to 30 s.

The experimental prototype has intended to represent the real situation that can take place in a direct measurement of the human body. The dissipation has been placed behind an aluminium plate of 1 mm in thickness, thus representing the surface tissues. The average temperature of the dissipation zone in stationary state has some oscillations of ±0.02 K, similar to the ones that the surface of the human body has.

Conclusions

As a conclusion, it can be pointed out that the proposed model represents acceptably the sensor operation, and the developed sensor has a promising future in the medical field. The direct application of the sensor is to measure powers and energies developed in localized zones of the human body but, moreover, it can provide information about the dissipation extension through the value of τ_2.

From the measures carried out, it can be concluded that the instrument is capable of measuring dissipations within an area of 50 cm^2 around the sensor. For the final configuration, there is a sensitivity of 112 mV W^{-1} with an uncertainty of 6%. Taking into account that the noise of the calorimetric output is ±0.5 mV, the detectable minimum power is 4.5 mW and, to reach 99% of its value, it is necessary a time of $-\tau_1\ln(1-0.99)=534$ s, what is equivalent to a minimum detection, in these conditions, of $534\cdot4.5\cdot10^{-3}=2.4$ J.

Finally, it is shown that it is necessary to propose some technique to allow the introduction of an artificial dissipation in the proximity of the zone to be measured in order to carry out a specific calibration for each case. On the other hand, it is also pointed out that it is essential to simplify the instrumentation required by the sensor in order to be able to use the equipment in clinical experimentation.

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