A STUDY FOR A PORTABLE IR SENSOR TO DETECT THE BLOOD TEMPERATURE DURING CORONARY BYPASS IMPLANTATION

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Abstract

The objective of this research was to investigate the possibility of using an infrared prototype device for the detection of the blood temperature during a surgical operation for coronary bypass implantation. The correlation between the fluid temperature profile and the fluid flow rate was demonstrated. Currently, an expensive infrared (IR) camera placed in a corner of the operating room is used for the qualitative detection of these data, but a lot of problems arise regarding noise generated by external heat sources (medical staff, instrumentation...). The idea was to design a low cost portable device to be placed near the region of interest. Each blood vessel acts like a thermal wave emitter, so the amount of heat is proportional to the blood flow detected by the IR sensor. The first step in our research was the implementation of a coronary system model for predicting its operation and carrying out punctual and profile measurements. We chose a pyroelectric sensor for its high quality-cost ratio. Because this kind of sensor detects only a variable infrared source, we used an electromechanical chopper for modulating the radiation. It consists of an electronic shutter whose opening speed is controlled by an astable multivibrator. The output signal was analysed using a dedicated electronic circuit including a bandpass filter and an amplifier; then an acquisition board is employed for capturing and displaying the signal using a PC.

The prototype assessment was made using a laboratory equipment. Further steps will concern in vivo measurements during surgical operation for coronary bypass implantation.

Keywords

Blood temperature; coronary bypass; pyroelectric sensor

1. Introduction

The medical applications of infrared technology involve a variety of tasks: oncology, pain, vascular disorders, arthritis, rheumatism, surgery, tissue viability, dermatological disorders, monitoring the efficacy of therapeutic drugs, etc.

In particular, application of thermograph in cardiac surgery provides important information regarding coronary flow, coronary anatomy and myocardial perfusion and function [1].

Today, infrared imaging is proving its ability to assist the surgeon in making real time decisions during the open heart surgery procedure. It gives valuable information during the operation by showing the change in temperature patterns on the heart as various procedures are performed.

In cardiac surgery, thermographic infrared cameras contain optical system which scan the field of view at a very high speed and focus the IR radiation on a detector that converts the radiation into an electrical signal. Using IR cameras, some experiences were made in order to find a relation between blood flow and temperature in the coronary.

In 1985 Papp et al. [2] demonstrated the correlation between coronary flow and epicardial temperature by quantitative infrared thermography of rabbits' hearts.

Cardiac thermograms were taken with an infrared camera and a hydraulic occluder was used to constrict gradually the left common coronary artery, then the heart perfusion was observed.

In the isolated rabbit's heart the relation of the decreasing flow to the decrease of mean epicardial temperature was very strict and quasi-linear.

In 1987 Adachi [3] studied the myocardial perfusion of ten dogs by an infrared imaging system. The experiment regarded continuous monitoring of beating heart of anesthetized dog by an infrared camera and in the same moment measuring the blood flow in the myocardium wall. As in Papp's experiment, the left coronary artery was occluded to create an ischemia for 90 minutes, then it was opened and the temperature was recorded for 210 min of reperfusion.

As a result of the experiment, the authors found an empiric relation between blood flow in the coronary and superficial temperature.

In an other experiment [1] with dogs a cool saline solution was injected in the aortic artery to decrease the blood temperature and to observe in which way the system was returned to normal condition.

However, since in cardiac surgery a high value of sensitivity is required, very expensive thermocameras are necessary in all of these experiments.

The aim of this paper is to describe the design of a low cost portable device that could be placed near the region of interest during a surgical operation for coronary bypass implantation. We started our research with the implementation of a coronary system model for predicting its operation and carrying out punctual and profile measurements. For the prototype design, the basic theory that we used was the fact that each blood vessel acts like a thermal wave emitter, so the amount of heat is proportional to the blood flow detected by the IR sensor. We chose a pyroelectric sensor for its high quality-cost ratio. The amplitude of the detected signal is proportional to the energy of infrared radiation emitted by the object, so each peak of this signal corresponds to the relative temperature of the object. The prototype assessment was made using a laboratory equipment. This paper describes an experimental study conducted according to a physical model, so further steps of this research will concern in vivo measurements during surgical operation for coronary bypass implantation.

2. Theory

2.1 IR radiation and blood flow

In general, IR sensors detect the thermal energy emitted by the human body in the form of electromagnetic radiation at a wavelength of infrared region (0.76-100μm) [4].
Total energy emitted and the temperature are related by the Stefan-Boltzman formula:

$$W = \varepsilon \sigma T^4 \tag{1}$$

where W is the radiant flux density (W/m^2), ε is the emissivity factor, σ is the Stefan-Boltzman constant and T is the absolute temperature. The value of ε for human skin at room temperature is unity, so the human body acts like a black body and the wavelength of the energy peak is at 10μm.
The other important physical phenomenon is the convective heat-transfer between the artery wall and the blood, given by the following equation:

$$\overset{*}{Q} = hA(T_b - T(t)) \tag{2}$$

where $\overset{*}{Q}$ is the heat flow (J/s), h is the convective heat-transfer coefficient, A is the internal surface area which is in contact with the blood, T_b is the blood temperature and T (t) is the arterial wall temperature in time t.

In general, blood flow in artery is laminar: neglecting the effect of pulsatile flow [1], the Reynolds number can be calculated by the follow equation:

$$\mathrm{Re} = \frac{v d \rho}{\eta_{blood}} \tag{3}$$

where v is the blood flow velocity, d is the vessel diameter, ρ is the blood density and η_{blood} is the viscosity, we can find that it's smaller than the critical value of 2000 [5].

In a coronary artery with a d of 3mm, a v of 0.3 m/s, considering a blood ρ of 1050 Kg/m^3 and a η_{blood} of 4 cp, using Eq. (3) results in a Re of 236.35, so the blood flow can be considered a laminar flow.

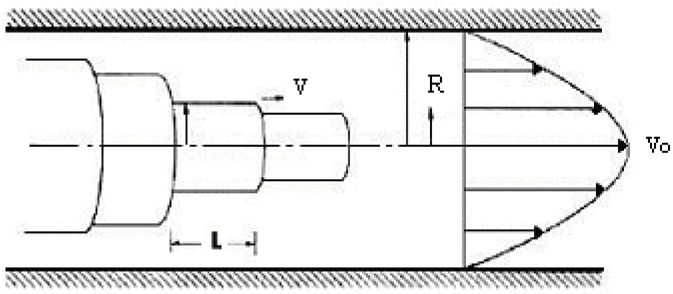

Figure 1: Parabolic profile of velocity in a laminar flow.

In the coronary the peak of flow occurs during the diastolic phase: in this case the profile of velocity vectors is parabolic because the flow is laminar [5] as you can see in Figure 1.

2.2 Pyroelectric sensors

Pyroelectric sensors act as a thermal transducer, detecting a temperature variation generated by an incident IR radiation through changes in material properties.

A pyroelectric sensing element converts the thermal radiation into the output measurable quantity like charge, voltage or current by pyroelectric effect that can be measured by pyroelectric coefficient (p) defined by the following expression:

$$p = \left(\frac{\partial P}{\partial T} \right)_{E,\sigma} \tag{4}$$

where P is the electrical polarization, T the temperature and the derivate is calculated with the electric field E and the elastic stress σ constant.

The main advantage of these detectors is that they can work at room temperature without external system of cooling that is necessary using photon detectors (they must be cooled to cryogenic temperatures to minimize background noise) [6].

Moreover, the pyroelectric detectors have a high quality-cost ratio, high sensitivity and low cost manufacturing; for these qualities, they have found many applications such as intruder alarms, fire alarms, chemical analysis, laser detectors, and human information detection.

A pyroelectric device consists of a window, a sensing element, and a readout integral circuit. The window acts like an optical high-pass filter that blocks unwanted wavelengths such as visible lights. In our sensor (IRA-E900 Murata) [7] the window is a 5 μm long pass Silicon filter.

Our infrared sensor uses lead zirconate titanate (PZT), a ferroelectric ceramic, as the sensing element. The pyroelectric element has on its faces two electrodes: this structure forms a capacitor that varies its charge when the detector is heated by incident radiation.

In order to detect this small charge a simple FET amplifier (with an appropriate load resistor) with low noise and high impedance is necessary.

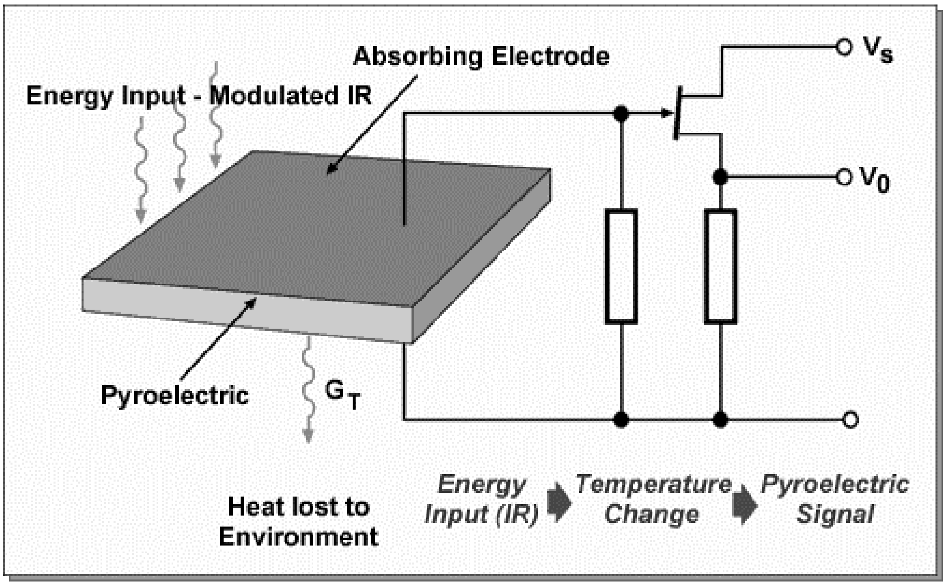

Figure 2: The infrared-voltage relation in a pyroelectric sensor and its internal circuit.

A change in temperature ΔT produces a transient change in the surface charge causing a displacement current I to flow in an external circuit connected to the pyroelectric material:

$$I = pA\frac{d(\Delta T)}{dt} \tag{5}$$

where p is the pyroelectric coefficient and A is the detector area.

In general, pyroelectric sensors detect only modulated radiations so for using them to measure a continue emission by an infrared source is necessary to modulate the incident radiation with a mechanical chopper, because the sensing element reacts only to changing heat flux.

Generally, the incident IR is periodically chopped in order to generate a continuous output voltage or current.

It can be possible to use a lens to focalize the infrared radiation on pyroelectric sensor.

3. Materials and methods

3.1 The chopper and the lens

Since the pyroelectric detectors detect light only when a temperature change in the element occurs, it is necessary to use an optical chopper.

For our application, emission of IR radiation from the coronary in open heart surgical operation, the incident IR radiation is periodically chopped in order to generate a continuous output voltage or current.

As first solution for the chopping system, we chose a stepping motor with rotating slotted disc; however, a lot of problems arose because the chopping system is large in size, requires high power and shows low reliability. Thus we proposed an electronic shutter that establishes the chopping frequency: it is composed by an electromagnet which when an electric current flows on it, the piston in it draws back. So a little lever can push a mechanical part that opens the sliding blinds: when the current stops to flow into the coil of electromagnet the lever comes back in the original position and closes the blinds.

We drive the shutter with a square wave with a frequency of 1 Hz provided by a timer circuit: at this chopper frequency the responsivity of the sensor is max.

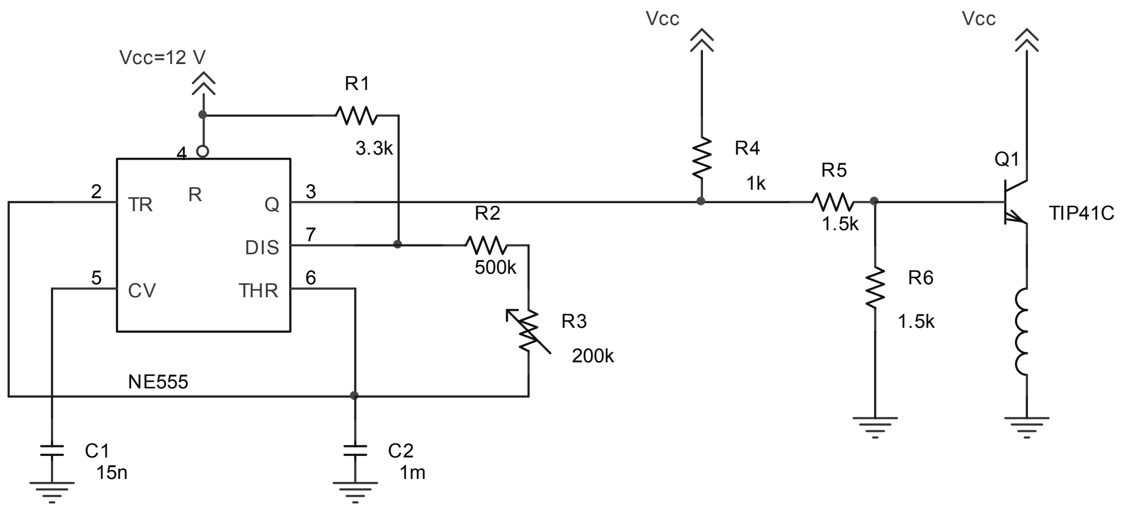

Figure 3: Driver electronic circuit for the chopper.

The coil of the shutter absorbs quite high current and it generates heat which can disturb the measure, so we use a little impeller in order to cool the sensor surrounding zone.

A cheap spherical lens was adopted to condense the infrared ray into the element areas. The curvature, diameter and position of the lens were determined to optimize the sensing areas and obtain high output.

The lens we used (ZnSe) has a focal length of 20 mm so the distance from the detector must be 40 mm and the distance from the area of interest must be equal, according the equation:

$$\frac{1}{p}+\frac{1}{q}=\frac{1}{f} \qquad (6)$$

where p is the distance between the detector and the lens, q is the distance between the lens and the interesting area and f is the focal length of the lens.

In order to locate the system in the optimal way we used a laser pointer system: in this manner we can place the detector in a fixed support in the right position.

The pointer system is constituted by two laser pointers: the two laser beams coincide only in one point that is the focal point of the lens. The user must move the system until the two beams coincide in one point, must fix the position of the support and then must switch off the laser (in fact we saw that a long exposition of the area by laser can be dangerous). Figure 4 shows the final prototype.

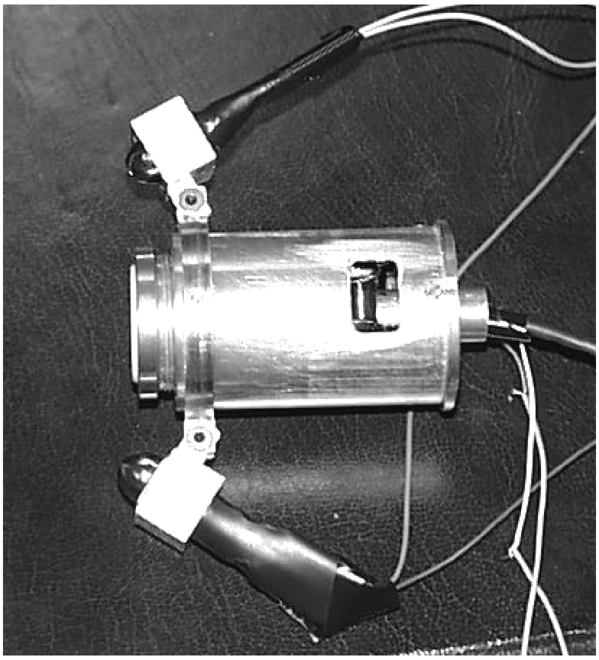

Figure 4: The final prototype

3.2 Electronic design

The chopper is driven with a 555 Timer at a 1 Hz frequency (this chopping frequency corresponds to the one where the relative responsivity of the sensor is maximized). Since the coil impedance is very low, it is necessary to amplify the output current and this is accomplished using a NPN silicon power transistor (Figure 3).

Regarding the acquisition of the signal generated by the sensor (1 Hz wave with an amplitude in the order of the mV), as shown in Figure 5, we use a band-pass filter centred at 1 Hz (-3 dB bandwidth: 4 Hz; centre band gain: 45 dB) to eliminate noise outside the frequency range of our signal.

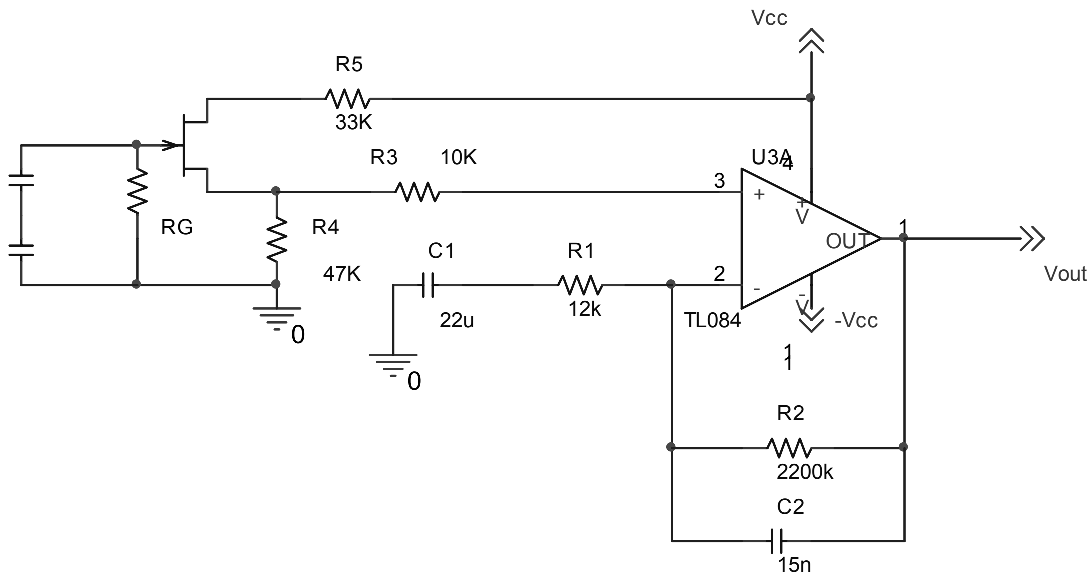

Figure 5: Readout electronic circuit for voltage response of the detector.

For the acquisition and the visualization of the sensor signal we used a PCI-6024E board (National Instruments) and LabVIEW (Laboratory Virtual Instruments Engineering Workbench, National Instruments) software.

3.3 Flow simulation by pump

In order to test our prototype we need a flow simulation: there are many software coronary blood flow simulator in literature but these are very complex because they join the fluidodynamic principles and the geometric simulation ones.

Moreover these simulation software don't provide us a valid instrument for laboratory test of the device.

Due to the complexity of this problem it's difficult to implement fluidodynamic simulator in vitro which are reliable and refer to the clinical case for this scope.

For these reasons we focussed our attention to model a laminar flow into a flexible tube with dimension as possible as close to the first part of the left coronary artery one, and to control the temperature of the flow which flows into the pipe.

The result, obviously, is far to simulate the coronary circulation but it's indispensable to test the device and to evaluate the predicted theory.

The device that we used is constituted by a water pump, with a thermostat, which pushes the fluid at fixed temperature into a silicon tube with a diameter of 4 mm and a thickness of 1 mm.

As the silicon has low thermal conductivity (15 W/m°C) and low emissivity (0.3) rather than blood vessel, we used a copper hollow tube with the same dimension of the previous pipe. Copper, in fact, has an emissivity (0.98 if it's painted matt-black) closer to thermal characteristics of human tissues: this material is often used to construct manikins in order to analyse the heat exchange during surgery hypothermia [8].

4. Results and discussions

4.1 Laboratory tests

First of all we found the in/out characteristic of the sensor: we used the pump to vary the temperature of the water in the tube from 25 to 42 °C (step of temperature variation: 1 °C) and the LabVIEW program to visualize the voltage responsivity on PC monitor.

The results are shown in the Figure 6 and they refer to the final configuration with chopper on, after the calculation of peak by the peaks detector program.

As we can note, the relation between the voltage responsivity and the temperature is almost linear.

In the voltage responsivity we included even the offset voltage, depending by the FET polarization, that for the sensor we have used is about 1 V.

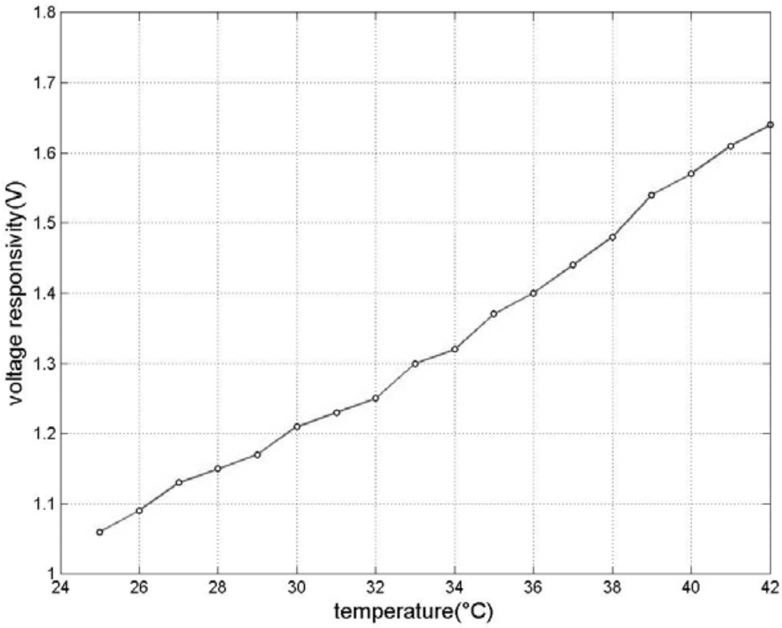

Figure 6: Correlation between voltage responsivity and temperature.

Figure 7 shows the voltage responsivity of the sensor in the temperature range of 30-37 °C with a step of temperature variation of 0.2 °C. Again, the correlation between the responsivity of the sensor and the temperature of the fluid is very similar to linear trend.

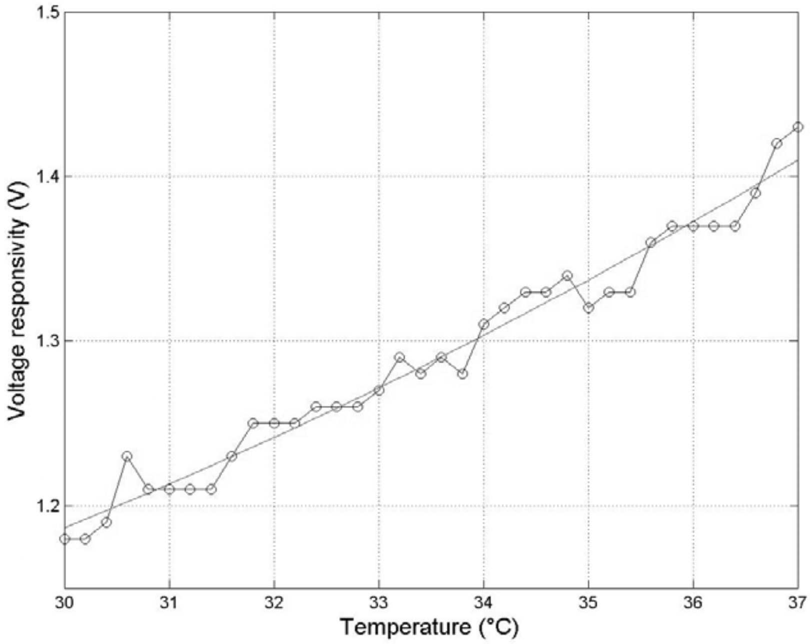

Figure 7: Correlation between voltage responsivity and temperature in the range of major interest.

Another test consisted in the measurement of the sensor responsivity after a flow interruption that simulates an artery ischemia: starting from a temperature of 37 °C we closed the tube for 5 minutes.

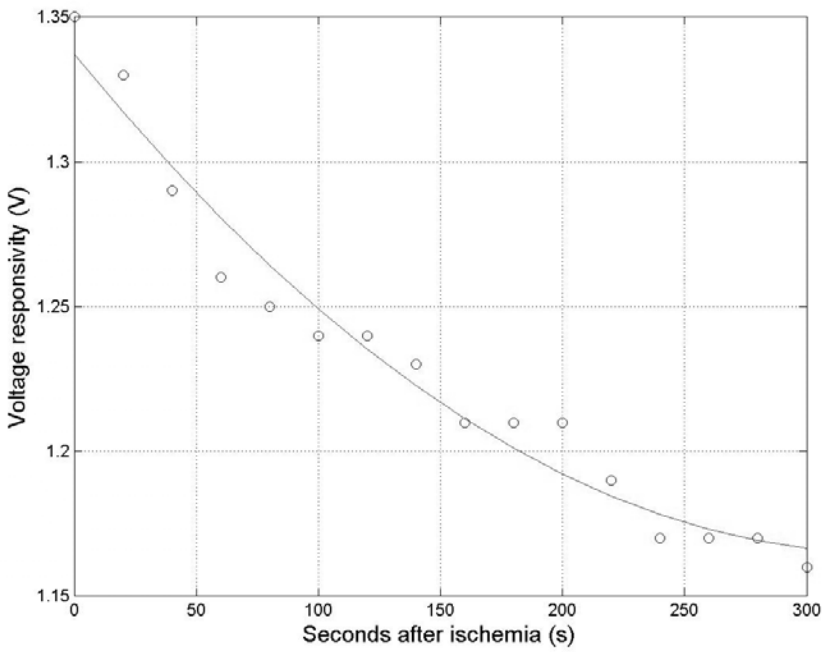

Figure 8: Correlation between voltage responsivity and time after ischemia.

As shown in Figure 8, we measured a rapid voltage responsivity drop due to the temperature decreasing of the tube up to room temperature.

Final laboratory test consisted of the measurement of the responsivity after reperfusion (Figure 9): starting from empty tube condition (at room temperature) we switched on the flow with a temperature of 37 °C to evaluate the time of the sensor for reaching this temperature.

As we can note, during early 10 seconds the sensor responsivity almost reached the value corresponding to the temperature set by the pump.

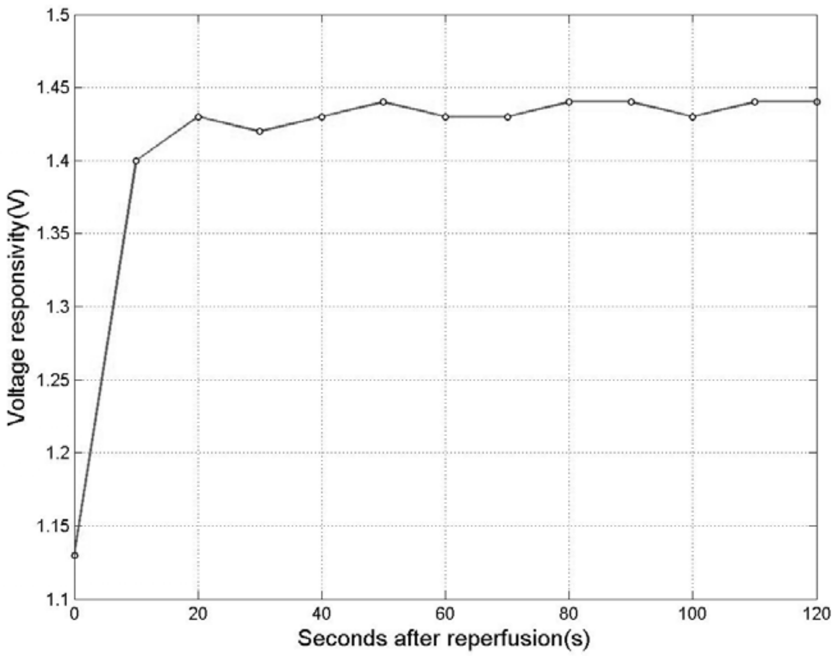

Figure 9: Correlation between voltage responsivity and time after reperfusion.

4.2 Future developments

A further development of this device can be the design of a detector which uses a linear array of pyroelectric sensors.

Currently we're working on a prototype with four sensors: we're using for this a chopper that consists of a stepping motor and a "sash-window" to modulate the incident radiation in the same way for all sensors. So it's possible to detect the temporal variation of temperature in four different points of blood vessel and then to follow perfusion of the zone of interest.

In this way we hope to rebuild the profile of blood flow from an initial condition (for example, empty vessel or introduction of cold saline solution) to normal condition.

Both hardware and software developments could be necessary and also a possible use of other types of infrared sensors, in order to obtain a better output response, will be evaluated.

5. Conclusions

The present article describes the study of a low cost portable device for the detection of the blood temperature during a surgical operation for coronary bypass implantation. Because of blood vessels act like a thermal wave emitter, the amount of heat is proportional to the blood flow detected by the IR pyroelectric sensor. The amplitude of the sensor output signal is proportional to the energy of infrared radiation emitted by the vessel, so each peak of this signal corresponds to the relative temperature of it. Using a thermostated water pump, we obtained a complete characterization of the device, in particular the relation between its responsivity and the temperature of the area of interest. This paper describes an experimental study conducted according to a physical model, so our research will require further work for performing biological experiments.

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