Available online at http://www.journalijdr.com

IJDR

ISSN: 2230-9926

International Journal of Development Research
Vol. 11, Issue, 03, pp.45076-45086, March, 2021
https://doi.org/10.37118/ijdr.21320.03.2021

RESEARCH ARTICLE OPEN ACCESS

CHARACTERIZATION OF THE TEMPORAL RESPONSE OF THE TEMPERATURE SENSOR OF SWAN-GANZ CATHETER

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ARTICLE INFO

Article History:
Received 10th December, 2020
Received in revised form
14th January, 2021
Accepted 24th February, 2021
Published online 15th March, 2021

Key Words:
Thermodilution,
Ejection Fraction,
Characterization, Swan-Ganz catheter

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ABSTRACT

This paper discusses concepts of thermodilution and ejection fraction associated with a typical application of the Swan-Ganz catheter, with the presentation of two models: one for the measurement of cardiac output and another for the measurement of ejection fraction. This work was developed to answer if the response to the impulse of the Swan-Ganz catheter depends on the speed of the blood flow in the right ventricle of the heart. The primary objective of this paper is to develop a method for characterizing the response of the temperature sensors in Swan-Ganz catheters so that it is possible to evaluate whether it is feasible to use deconvolution operations in measurements performed with this type of sensor.The determination of cardiac output and heart ejection fraction can help predict cardiovascular disease and more effectively estimate the risk of sudden death (ejection fraction <0.2), which are becoming more common.

Citation: ***Maxwell Diógenes Bandeira de Melo, Glenda Michele Botelho, Viviane Fernandes Moreira, Ivan Ney Alvizuri Romani, Ary Henrique Morais de Oliveira and Adson Ferreira da Rocha.*** "Characterization of the temporal response of the temperature sensor of Swan-Ganz catheter", *International Journal of Development Research*, 11, (03), 45076-45086.

INTRODUCTION

The present study is part of a larger investigation, whose goal is to explore the most relevant research on the thermodilution method and the ejection fraction measurement. After, the study proposes a methodology to develop the measurement of these parameters, allowing for catheter characterization.

Thermodilution: Cardiac output is a measure of the blood flow pumped by the heart. It is a physiological parameter of great clinical importance that is measured by the right side of the heart due to its easy access (Guyton and Hall, 2002). However, measuring methods of cardiac output are either invasive or very expensive. Within this perspective, thermodilution stands out for presenting the best characteristics related to cost, accuracy and precision. The heart, an organ composed of four separate chambers, the right and left atria (at the top) and the right and left ventricles (at the bottom), is considered to be a set formed by two pulsating two-stroke pumps (filling and emptying), responsible for the circulation of blood in the human body. Upon reaching the right atrium, blood flows to the right ventricle through the tricuspid valve. Then, it is pumped through the pulmonary valve, through the right ventricle to the pulmonary artery, usually in a unidirectional way (Burton, 1997). The thermodilution technique, initially proposed by Stewart (1987), is considered a subclass of the dilution indicator method, in which the indicator is heat itself, usually measured in a cooled saline solution that, when injected into the circulatory system, correlates with the blood flow value. However, although Stewart (1987) proved that cardiac output could be measured using thermodilution, the accumulation of the indicator substance in the patient's circulatory system imposes limitations on the number of tests that can be done. Maruschak(1985), when carrying out a study on the frequency response of fast thermistors mounted in a catheter and their influence on the measurement of the ejection fraction, concluded that the thermistor distorts the thermodilution signal in order to underestimate the real ejection fraction, suggesting that it would be possible to eliminate distortion by means of deconvolution technique. When using thermodilution and angiography techniques to measure the ejection fraction and ventricular volume of the right ventricle, Urban et al.(1987), found a satisfactory correlation between the two methods. They concluded that the use of thermodilution in ICUs leads to results with sufficient precision. Dhainaut et al. (1987), when using nuclear and thermodilution techniques to measure the right ventricular ejection fraction, concluded that thermodilution is as accurate and reproducible as nuclear techniques. Different types of indicators can be used as reagents in the indicator dilution technique. However, the most popular is the indicator proposed by (Fegler,1957a; Fegler, 1957b), which is heat itself. Fegler's method consists of injecting a small amount of cold Ringer's solution quickly into the vena cava and recording the temperature curve in the aortic arch and in the right ventricle. The injection site is usually the right atrium or the right ventricle, and the temperature is measured in the pulmonary artery. As described by Swan et al. (1972), the emergence of catheterization techniques to measure pressure in the pulmonary artery in the 1960s made it possible to insert a sensor (a small thermistor) into the pulmonary artery. This fact led to greater clinical acceptance of thermodilution, although the technique is still in the validation phase. According to Trautman et al. (1984), the pulmonary artery catheter provides important hemodynamic information and, for this important reason, is used in many patients. The invasiveness of a thermistor into a human body is negligible, and the value of cardiac output measures is immense. As the indicator is a physiologically inert solution, the measurement of cardiac output by thermodilution has become an important part in caring for critically ill patients. In order to measure cardiac output and right ventricular ejection fraction, the catheter must have a hole at the height of the atrium or right ventricle, which allows the injection of fluid (saline) into one of these chambers in the heart. When mixed with blood in the chamber, this solution will cause it to reach an intermediate temperature between the original blood temperature and the temperature of the injected liquid. Then, the obtained curve can be used to calculate the cardiac output and the right ventricular ejection fraction (Trautman, Newbower, 1988).

When evaluating the thermodilution technique in the measurement of right ventricular ejection fraction, cardiac output, ejected volume, diastolic volume and volume at the end of systole, using a fast response thermistor, Ferris and Konno(1982) found that the measured values were consistent with the theoretical ones. However, in the case of the ejection fraction, the tested range was limited. The merit of popularizing the pulmonary artery catheter, which contains a thermistor and an injection port, is widely credited to Swan and Ganz (Gans and Swan, 1972). Dos Santos (2000) developed a modified version of the da Rocha method (automatic algorithm with reduced processing time). The method showed great clinical potential, being of great value for application in the ICU (da Rocha, dos Santos, de Melo, 2005; de Melo, Araújo, da Rocha, 2016). For modeling purposes, a chamber can be used to represent the right ventricle, a tube to represent the right atrium, and a second tube to represent the pulmonary artery. In many cases, only the contraction of the ventricle is included in the model, since the atrium contraction has a secondary role in pumping, even though it helps in the final filling of the ventricles (Guyton and Hall, 2002). Considering typical values of the volume ejected by the heart, for the final diastolic volume and heart rate, which are 0.072 liters, 0.130 liters and 70 bpm, an ejection fraction of 0.55 (0.072 liters/0.130 liters) and a cardiac output of 5 liters/minute (0.072 liters x 70 bpm) are obtained (Novato,2004). When carrying out a study on thermodilution, Nelson concluded that this technique is the safest, most accurate, reproducible and least costly when compared to other existing ones (angiography, radioactive methods and echocardiography), when the objective is to evaluate the right ventricular function (Magalhães and Machado, 2004). However, when studying the effect of the temporal response of the Swan-Ganz catheter temperature sensor, da Rocha et al. (2005) identified the errors caused by the jet of the convection coefficient in the sensor region in the cardiac cycle. Thus, in order to improve the temporal response of the Swan-Ganz catheter, Melo et al. (2016) developed a technique based on deconvolution whose objective was to obtain the input signal of a linear time-invariant (LTI) system from the knowledge of its output and its response to the impulse. Melo and his collaborators developed and evaluated an algorithm to enrich the thermodilution signal. The proposed method presents an improvement of previous works in the literature in terms of precision, convergence and computational effort (de Melo et al. 2019).

Ejection fraction: Ejection fraction is the ratio between the volume ejected from the ventricle and its maximum volume at the end of diastole. Cardiac output is the total volume pumped by the heart over a period of one minute. This parameter is used clinically to assess the pumping capacity of the ventricle. Over the years, this technique has been validated by in vivo and in vitro studies, and its advantages have been observed in relation to other more traditional methods (Trautman, Newbower, 1988). After the pumping action, the pressure in the ventricle continually reduces until it is below that of the atrium. Then, the tricuspid valve opens, causing blood to flow rapidly into the ventricle. After complete filling of the ventricle, the pump is activated, initiating contraction. Next, the tricuspid valve closes, and the pulmonary valve is opened, given the pressure difference between the ventricle and the pulmonary artery. Part of the blood in the ventricular chamber is pumped into the pulmonary artery. Typically, only 55% of the volume is ejected, the ratio between the ejected volume and the total volume being called the ejection fraction (Burton, 1997). Ling et al. (1999) described the development of an adaptive system based on fuzzy logic that continuously estimates cardiac output using the blood pressure waveform as input. This system was evaluated using a set of 133 experimental data from 10 baboons. The average error between the estimated and the calculated cardiac output was less than 4%. Dr. Ling and his team tried to show in this paper all the usefulness of fuzzy logic in this field of human knowledge. Heerdt et al. (2001) published a paper whose objective was to evaluate the effect of the variability degrees of acute tricuspid regurgitation on measurements of cardiac output during changes in venous return. Anesthetized dogs were used for the respective measurements. It was also concluded that tricuspid regurgitation is a considerable barrier to accurately measuring cardiac output. Hamilton et al. 2002 developed a method to reduce the effects of breathing and a method to reduce a small slip in recording data used to estimate continuous cardiac output. Thermal data from three anesthetized pigs were recorded and compared with estimates of cardiac outputs measured on an ultrasonic flow meter. Humphrey et al. (2002) developed a system for measuring blood flow in a permanent state to the right side of the heart. The system was used to validate the thermodilution method for three known flow rates.

Zhao et al. 2003 published an interesting comparison between the bolus method for thermodilution and the Doppler method. They concluded that the Doppler method is clinically acceptable. The results were consistent with the thermodilution measurements. Kuper(2004) made an interesting comparison between the different methods of measuring hemodynamic parameters. He concluded that there is no perfect method and the choice is a complex decision. Also, Yelderman(2004) states that techniques using thermodilution are the most reliable methods for measuring cardiac output because of the elastic nature of human cardiac veins. He used a stochastic signal processing method using pseudo-random binary infusion of heat as an improvement process for noisy signals, to facilitate obtaining cardiac output for a reasonable period (five minutes) with acceptable clinical errors.

MATERIALS AND METHODS

This section will show the basic foundations of the thermodilution method and catheter temperature sensor response nature.

The Thermodilution Method: Some details of the insertion of the catheter used in the thermodilution method are illustrated in Figure 1. A Swan-Ganz catheter is inserted into the human body through a peripheral vein and advanced to the right part of the heart, running through the right atrium and the right ventricle until finally reaching the pulmonary artery. A more detailed illustration is shown in Figure 2. There is a thermistor (point 2) near the tip of the catheter. In the right atrium (point 7), there is an insertion port for the indicator substance. The catheter has other points, such as one to measure the pressure in the cardiac chambers and one to inflate a balloon located at the tip of the catheter. The main points indicated in Figure 2 are: 1) Superior vena cava, 2) Thermistor, 3) Balloon, 4) Right pulmonary artery, 5) Aorta, 6) Pulmonary artery, 7) Indicator substance injection port.

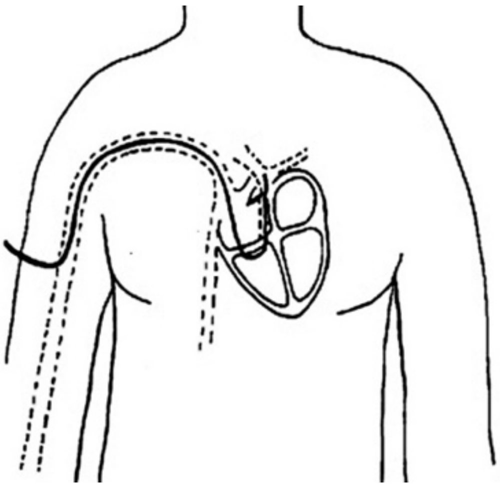


Figure 1. Illustration of the insertion of a Swan-Ganz catheter in a patient. The catheter is inserted into a peripheral vein, and is pushed towards the right ventricle, traversing the vena cava, the right atrium, the right ventricle and, finally, its tip is positioned in the pulmonary artery. Blood flow assists in directing the catheter

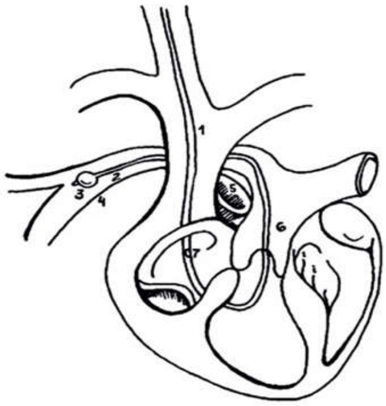


Figure 2. Detail of the insertion of the Swan-Ganz catheter in the heart. The points are: 1) Superior vena cava, 2) Thermistor, 3) Balloon, 4) Right pulmonary artery, 5) Aorta, 6) Pulmonary artery, 7) Indicator substance injection port. The indicator substance is expelled at point seven, and its temperature is measured at point two. Figure 3 shows a drawing of the Swan-Ganz catheter. The thermistor can be viewed a few centimeters from the tip of the catheter. An inflatable balloon, whose function is discussed later, can be viewed at the end of the catheter. There are two injection ports for indicator substances, the proximal and the distal port, 30 cm and 19 cm from the tip, respectively.

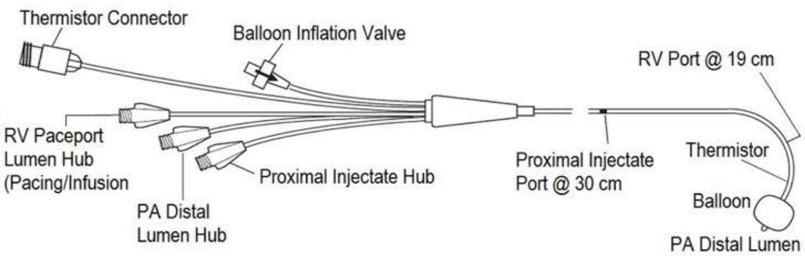


Figure 3. An outline of the Swan-Ganz catheter (model 93A-931H-7.5F, produced by Baxter Edwards)
Figure scanned from the manufacturer's manual.

The model for the thermodilution technique used in this project is approached next. This model can be understood with the aid of the functional diagram shown in Figure 4. The chamber represents the right ventricle. The tube entering the right ventricle is equivalent to the right atrium, and the tube coming out of the chamber simulates the pulmonary artery. Typical stroke volume values are 72 ml and 130 ml, which provides a cardiac output of 5L for an ejection fraction of 0.55. Currently, in many applications, the indicator is injected into the right atrium, which provides a better mixture of the indicator with blood. In the model presented here, the injection in the right ventricle will be used for simplifying the equations. It will be assumed that the measurement by the temperature sensor will be carried out in the pulmonary artery and that there is a perfect mixture between the injected bolus and the blood.

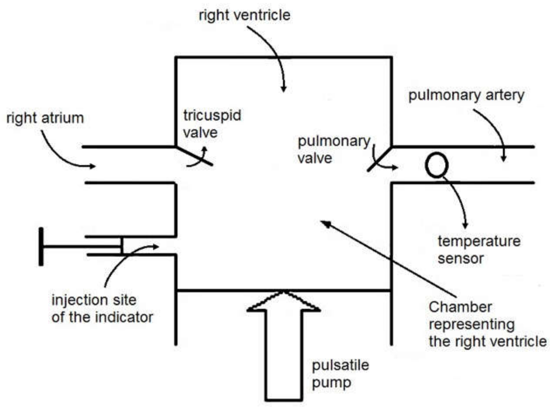


Figure 4. Simplified model for the thermodilution system proposed in this work. The main chamber represents the right ventricle. The tube entering the right ventricle is equivalent to the right atrium, and the tube coming out of the chamber represents the pulmonary artery. The small tube with an injector represents the injection site of the indicator, which in this work will be the right ventricle for reasons of simplifying the model equations. Typical stroke volume values are 72 ml and 130 ml, which provides a cardiac output of 5l for an ejection fraction of 0.55. In Figure 4 the heart beats at each time interval Δt, and the output volume is V_s, for each beat. It is assumed that the maximum volume (V_{max}) occurs at the end of diastole, immediately before contraction, and that the minimum volume (V_{min}) occurs at the end of systole, at the end of contraction. By definition $V_s=V_{max}-V_{min}$. The temperature of the injected substance is considered constant and equal to T_i. The bolus is injected according to a function called g(t), and is assumed that it will have a short duration and that the integral over time of this function is equal to the volume of the injected indicator:

$$\text{Total injected volume} = \int_{(n-1)\Delta t}^{n\Delta t} g(t)\,dt \tag{1}$$

where Δt is the duration of the pumping cycle, n is the number of pumping cycles and g(t) (L/s) is a function that describes the bolus injection.

The amount of substance injected into the blood during each pumping cycle can be defined as a discrete function:

$$I[n]=[I_{ind0},I_{ind1},I_{ind2},\dots,I_{indn}]=[I_0,I_1,I_2,\dots,I_n] \tag{2}$$

where I_{ind} or I_i is the volume of substance injected in the i-th cardiac cycle, as expressed by equation 3.

$$I_1=\int_0^{\Delta t} g(t)dt;\ I_2=\int_{\Delta t}^{2\Delta t} g(t)dt \tag{3}$$

Where Δt is the duration of the pumping cycle (in seconds) and g(t), in liters per second, is the function that describes the bolus injection. Since the injection is very fast, only one or two terms of vector I[n] are not null. At the end of diastole, the ventricle retains the maximum volume, and the blood is perfectly mixed with the indicator. A vector T[n] will be defined to represent the temperature of the successive plateaus of indicator temperature at the end of the diastole, shown in equation (4).

$$T[n]=[T_0,T_1,T_2,\dots,T_n,\dots] \tag{4}$$

The variable T_i (ºC) is the temperature of the injected substance in the i-th cardiac cycle. According to the thermodynamics laws, if there is a mixture with m fluids at different temperatures, the final temperature (da Rocha, 1997) is shown in equation (5).

$$T_{mixture}=\frac{\sum_{k=1}^{m} V_k\rho_k c_k T_k}{V_{mixture}\rho_{mixture}c_{mixture}} \tag{5}$$

Where V_k is the volume (in liters) of each component of a liquid in the mixture, Δk is the density (in Kg/m3) of each component of the mixture, ck is the specific heat (in J/ KgΔC) of each component of the mixture, Tk is the temperature (in ºC) of each component of the mixture and k is an integer. Assuming that the density (Δ) and specific heat (c) of the components are the same in all fluids, equation (6) is obtained.

$$T_{mixture}=\frac{\sum_{k=1}^{n} V_k T_k}{V_{mixture}} \tag{6}$$

Where Vk is the volume (in liters) of each component of a liquid in the mixture, T_k is the temperature (in ºC) of each mixture component, Vmixture is the mixture total volume and k is an integer. All temperatures were referenced to a base temperature. Conveniently, a transformation of variables, in order to associate the blood temperature with the zero value, was used. Thus, the initial temperature in the ventricle, before the injection of the indicator substance, is zero, as indicated in equation (7).

$$T_{ventricle} = T_0 = T_{blood} = 0°C \tag{7}$$

Developing equation (6) for the sequence of T[n], we have the following decomposition, shown in equations (8), (9) and (10).

$$T_1 = \frac{T_i I_1 + T_0 V_{\min} + T_b\left(V_{\max} - V_{\min}\right)}{V_{\max}} = \frac{T_i I_1}{V_{\max}} \tag{8}$$

where Ti (in ºC, with i = 1,2, ..., n) is the mixture temperature in the ventricle chamber at the n-th beat, T0=Tb is the initial temperature (in ºC) of the ventricle, V_{min} is the volume (in liters) at the end of the systole, V_{max} is the volume (in liters) at the end of the diastole and I_i (with i = 1,2, ..., n) is the volume of the injected substance (in liters) in each cardiac cycle and n is an integer. The temperature in the second beat is given by equation 9, and the temperature in the third beat, by equation 10.

$$T_2 = \frac{T_i I_2 + T_1 V_{\min}}{V_{\max}} = \frac{T_i I_2 + \left(T_i I_1\right)\left(\frac{V_{\min}}{V_{\max}}\right)}{V_{\max}} \tag{9}$$

$$T_3 = \frac{T_i I_3 + T_2 V_{\min}}{V_{\max}} = \frac{T_i I_3 + \left(T_i I_2\left(\frac{V_{\min}}{V_{\max}}\right) + \left(T_i I_1\right)\left(\frac{V_{\min}}{V_{\max}}\right)^2\right)}{V_{\max}} \tag{10}$$

For induction, it is possible to write the following generic equation presented in equation 11.

$$T_n = \frac{T_i \sum_{k=1}^{n} I_{n-k+1}\left(\frac{V_{\min}}{V_{\max}}\right)^{k-1}}{V_{\max}}, \tag{11}$$

Adding the temperatures of all the plateaus, expression 12 is obtained.

$$\sum_{n=0}^{\infty} T_n = \frac{T_i}{V_{\max}}\left[I_1 \sum_{h=1}^{\infty}\left(\frac{V_{\min}}{V_{\max}}\right)^h + I_2 \sum_{h=1}^{\infty}\left(\frac{V_{\min}}{V_{\max}}\right)^h + \ldots + I_n \sum_{h=1}^{\infty}\left(\frac{V_{\min}}{V_{\max}}\right)^h + \ldots\right] \tag{12}$$

Equation 12 can be rewritten as:

$$\sum_{n=0}^{\infty} T_n = \frac{T_i}{V_{\max}}\left[\left(I_1 + I_2 + \ldots + I_{last}\right)\left(\frac{1}{1 - \frac{V_{\min}}{V_{\max}}}\right)\right] = \frac{T_i}{V_{\max}}\left[\left(\sum_{p=0}^{\text{injections that are not zeroes}} \mathrm{I}_n\right)\left(\frac{1}{1 - \frac{V_{\min}}{V_{\max}}}\right)\right] \tag{13}$$

where $\sum I_n$ is the ejected total volume and $\sum T_n$ is the sum of the temperature plateau, with which we deduce that the output volume is given by equation 14.

$$V_s = V_{\max} - V_{\min} = \frac{\sum_{p=0}^{\text{injections that are not zeroes}} I_n}{\sum_{n=0}^{\infty} T_n} \tag{14}$$

Considering that all plateaus have a duration equal to Δt, it is possible to rewrite equation 14 as:

$$V_{ejected} = V_s = V_{\max} - V_{\min} = \frac{\Delta t \sum_{p=0}^{\text{injections that are not zeroes}} I_n}{\Delta t \sum_{n=0}^{\infty} T_n}, \tag{15}$$

That can be rewritten as:

$$V_s = \frac{\int_0^\infty g(t)dt}{\int_0^\infty T(t)\,dt} \tag{16}$$

If the initial simplifications regarding specific heat and density had not been made, the following equation would have been reached (Trautman, Newbower, 1984):

$$V_s = \frac{\int_0^\infty g(t)dt}{\int_0^\infty \frac{T(t)}{T(t) - \frac{c_i \rho_i}{c_{blood}\rho_{blood}}\left[T(t) - T_i\right]} dt} \tag{17}$$

Where T_i (ºC, with i=1,2,...,n) is the mixture temperature in the ventricle chamber on the n-th beat, g(t) is the rate of the injected substance (in l/s) and T(t) is the temperature (in ºC) of the injected substance, ci is the density (Kg/m3) of the injected substance, Δi is the specific heat of the mixture of the blood and injected substance, in J/KgºC. In this work, equation (16) will be used to calculate cardiac output.

Figure 5 shows the idealized curve for the blood temperature in the pulmonary artery as imposed by equation (11). This is the well-known thermodilution curve. Equation (16) is known as the Stewart-Hamilton equation, and, in summary, indicates that the area under the thermodilution curve is inversely proportional to cardiac output. Unfortunately, the temperature curve measured by the sensor is not shown in Figure 5. The thermal inertia of the temperature sensor acts as a low-pass filter. The response to the sensor impulse can be approximated by four exponential components. Figure 6 shows the real aspect of the thermodilution curve, which is a filtered version of the real curve.

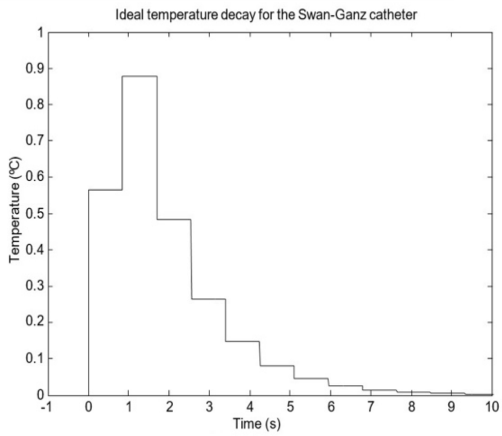


Figure 5. Idealized temperature curve of the indicator for a mixture with blood in the right ventricle. Note the decay of the curve, forming plateaus. This curve is the expected curve for the catheter response. Due to the physical characteristics of the sensor, this response is not obtained in real measurements.

The distortion shown in Figure 6, however, does not hinder the thermodilution measurement. Da Rocha et al. (2005) showed that this distortion causes a minimal error in the thermodilution measurement, since the low-pass filtering preserves the area under the curve.

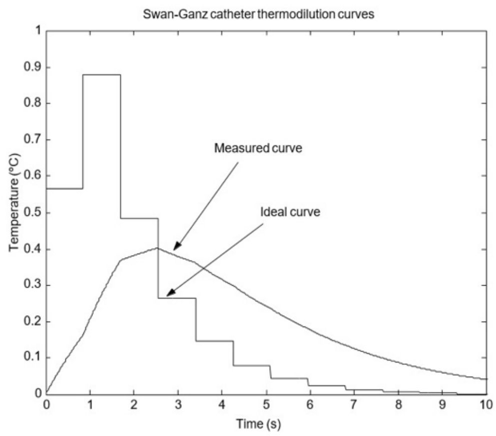


Figure 6. Thermodilution curves for the Swan-Ganz catheter.

In addition to cardiac output, it is possible to extract the ejection fraction (FE) from the thermodilution curve. This is clear from equation (11). From this equation we have that, if the injection period of the indicator has already ended, equation 18 is valid.

$$FE = 1 - \frac{T_n}{T_{n-1}} \tag{18}$$

Where T_n and T_{n-1} are temperature measurements in degrees Celsius from two successive plateaus. Thus, after the end of the ejection period, the ejection fraction is given by equation (18). This equation is commonly used to determine the ejection fraction in commercial systems. To improve accuracy, equation (18) is applied to three pairs of successive plateaus, and the arithmetic mean of the three obtained ejection fraction values is used. However, the distortion introduced by the sensor of the traditional Swan-Ganz catheter prevents measurement of the ejection fraction, since it is essential to know the amplitudes of the plateaus. Thus, the proposal of the present work is the estimation of the real curve, with plateaus, for later determination of the ejection fraction. It is possible to see the difference between the measured temperature and the actual temperature.

In this work, one of the objectives is, from the measured curve, to obtain the ideal curve. The Catheter Temperature Senson. For the proposed method to be effective, it is necessary to understand, precisely, the nature of the response of the catheter temperature sensor. The catheter temperature sensor is a thermistor, which is embedded in one of the catheter's lumens and exposed to its surface. To protect the human organism from microshocks, an epoxy layer is used to cover the thermistor. However, by protecting the organism, this layer slows down the sensor's response, as heat must diffuse through this layer before reaching the sensitive core of the thermistor. Da Rocha (1997) established that in the measure that the catheter sensor is exposed to a flow with a constant convection coefficient, it works as a linear and time-invariant system. Its response can be precisely modeled as a convolution between the response to the catheter impulse and the temperature in the convective environment, as described in Equation 19.

$$T_{measured}(t) = h(t) * T_{thermistor}(t) \tag{19}$$

Where $T_{medida}(t)$ is the temperature measured in ºC for the thermodilution curve, h(t) is the response to the catheter impulse and $T_{termistor}(t)$ is the actual temperature measured by thermistor in ºC. In the same work, da Rocha (1997) demonstrated that it is possible to use knowledge of the sensor's response characteristic to improve its functioning through deconvolution operations. This work was later perfected by dos Santos (1999; 2000), who made the algorithm faster. In da Rocha et al. (2005), it was also demonstrated that the response to the catheter impulse can be accurately modeled as a sum of four exponential components, as illustrated in equation (20).

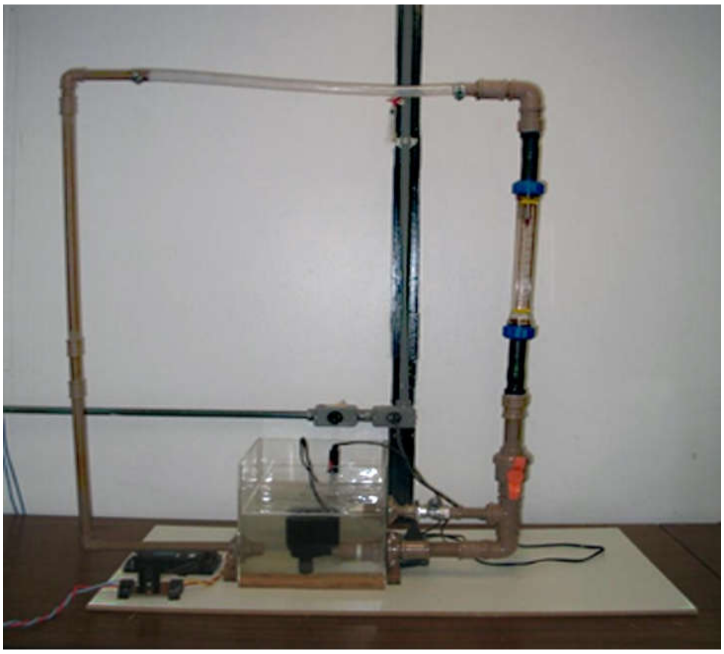
$$h(t) = K\left(e^{-at} + Be^{-bt} + Ce^{-ct} + De^{-dt}\right) \tag{20}$$

Where h(t) is the response to the catheter pulse, the parameters a, B, b, C, c, D and d, are parameters that characterize the sensor response of the Swan-Ganz catheter, and K is a constant that causes the area under the thermodilution curve to be unitary (that is, there is no DC gain). The technique proposed by da Rocha (1997)(da Rocha, 1997) involved placing the sensor in a controlled flow, and the installation of an extremely fast microthermistor next to it as a reference. Then, a very rapid injection of a liquid was performed at a temperature different from the temperature of the circulating liquid. In that work, experiments were carried out to qualify the response to the catheter pulse for flows of 0.2 to 1 m/s.

At work, a technique for determining the impulse response in the frequency domain was used. This technique, however, had a limitation: it required a very fast stimulus, which, in the frequency domain, had zeroes only at very high frequencies. In practice, this fact limited the generality of the method, and the characterization curves could present an appreciable measurement error in cases where it is not possible to generate a stimulus with a rapid frequency response.

The objective of this work is to improve the technique proposed by da Rocha (1997), removing the need for the stimulus to have, in the frequency domain, zeroes at a very high frequency. The method proposed here will follow similar principles to the da Rocha method, but will work in the time domain. However, the present method extends the flow range in the catheter characterization and increases the generality of the methodology. The first step of the methodology was the creation of a circulatory system with constant continuous flow with controlled velocity and temperature, which allowed the rapid injection of a liquid at a different temperature. In addition, there is the measurement of the temperature resulting from the liquid, downstream, by two sensors, one being with a very fast response and the other with a Swan-Ganz catheter temperature sensor. The results of the methods' characterizations are presented in section 6, and the discussions related to these results are presented in the following section.

Simulated circulatory system: The hydraulic system developed, Figure 7, consists of a 250 mm acrylic reservoir on the side, without cover, and a circuit of pipes, registers and connections in PVC arranged for water circulation in a counter-clockwise direction. This configuration is shown in Figure 7. The liquid (water) in the acrylic container is kept at 36 °C. The liquid is circulated by a submerged pump - model SB 2000 - manufactured by Sarlo Better, which provides a maximum flow rate of 1950 L/h. Figure 7. Front view of the circulatory system. The hydraulic system consists of a 250 mm acrylic reservoir on the side, without cover, and a circuit of pipes, registers and connections in PVC arranged for the water circulation in a counter-clockwise direction. The liquid circulation is made by a submerged pump model SB 2000 manufactured by Sarlo Better. It works with a voltage of 220 V, with a frequency of 60 Hz, and consumes about 30 W of power, with a maximum flow of 1950 L/h. The electrical components of the pump are fully immersed in epoxy resin, which makes it safe from electric shock. The pump was dimensioned in such a way that, with the height of the used pipe column, it was able to pump the water with the necessary flow to reach the maximum velocity required in the insertion region of catheter and thermistor, which is around 0.8 m/s. The tube attached to the upper part of the system is a transparent rubber hose, due to the need to visualize the positioning of the catheter sensor and the fast thermistor. The pump outlet is directed to a 32 mm external diameter tube that conducts the liquid in an upward direction. Part of the flow is directed to a reduction with a 20 mm external diameter tube which has the function of capturing the excess liquid necessary to obtain the desired velocity in the area of insertion of the sensors. A control was implemented for the circulating liquid temperature - in general, water - at 36ºC. For this, an NTC thermistor was placed at the beginning of the tube, to measure the temperature.



The temperature was provided to the A/D converter of the MSP430 microcontroller, which executed a PID control algorithm, which activated an on/off control system that maintained the temperature at 36oC.

For temperature measurements, an acquisition system based on Labview was used. This system was originally developed by Novato (2004) and was also adapted and used in the experiment of this work. In the experiment described in this paper, only the first two channels were used, which correspond to the two temperature channels: the fast sensor channel and the catheter sensor channel.

RESULTS AND DISCUSSION

In this section, results for a new catheter characterization algorithm are shown and practical results are discussed.

The new algorithm for characterization: The basic data for the program are the data acquired by the slow sensor and the fast sensor, acquired at a sampling frequency of 300 Hz. Initially, an estimate is made for the values of the coefficients a, B, b, C, c, D d, from equation 20. The value of K is such that the sum of all samples of h[n] is equal to 1. In the algorithm, two measurements are made, in a convective flow, by two sensors close to each other, one being the catheter temperature sensor and the other being a very fast microthermistor (40 ms time constant), assuming the temperature measured by the latter corresponds to the real temperature in the convective environment. In general, the stimulus corresponds to the rapid injection of a liquid at a temperature above or below the temperature of the circulating fluid (in the case described here, a liquid with a lower temperature, obtained by mixing ice and water, was used). The previous algorithm, proposed by da Rocha (1997) had the limitation of requiring that the reference signal had, in the frequency domain, zeros just above the frequency of 10 Hz. This limitation is removed in the present method, which makes it more general. The first step of the algorithm consists of pre-filtering the two signals (fast and slow), extracting frequencies above 50 Hz, which correspond to noise, and normalizing the two signals, so that the sum of all samples of the sampled signals is equal to 1. Once this is done, a variables change is made in which the base temperature is 0^oC, and the stimulus, instead of being negative, is positive, in order to facilitate the signal visualization. A typical example of the result of the experiments carried out is illustrated in Figure 8.

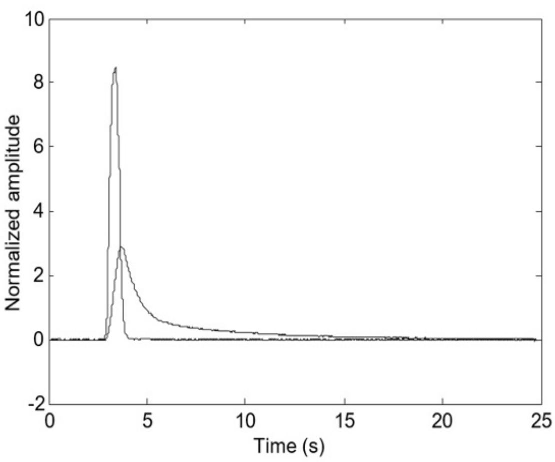


Figure 8. The signals captured by the fast (broader signal) and slow sensors. The signals were filtered and normalized. Frequencies above 50 Hz were eliminated, as they were not significant, and a variables change was made so that the base temperature was 0. The deflection was positive and the areas under the curves were unitary. It was assumed, then, that the response of the catheter has the shape shown in equation 20. Then, the

convolution is performed, using operations in the frequency domain, between the fast signal and the response, h(t), obtaining a first estimate of what the sensor's response would be. Naturally, the first approximation leads to an incorrect value, and there is a difference between the measured curve and the convolution result, since the first estimate generally does not correspond to the actual response. With this, a measure of the difference between the estimated response and the actual response is made, the point-to-point difference between the two curves is calculated, and the sum of the difference modules is calculated, which is considered as the estimate error. From that point on, a strategy begins that seeks to minimize the average rectified error between the two measures. In this strategy, for each coefficient in the expression h(t) (a, B, b, C, c, D and d), a small increment (or a "delta") is added. After that, the error is recalculated (in this case, a delta equal to 0.001 was used). The difference between the first error and the error with the delta is interpreted as an estimate of the partial derivative of the error function in the direction of the variable corresponding to the parameter that was incremented. This procedure is repeated for all variables and, at the end of it, there is an estimate of the direction of the gradient of the error function that is the direction with the greatest inclination. Then, each variable suffers a small increase in the decreasing direction of the gradient. Thus, over time, the functions become closer and closer when the error is less than a certain value, or when the results are almost visually indistinct.

EXPERIMENT RESULTS

Simulations were performed to determine the impulse responses for the flow velocities shown in Table 1.

Table 1. Flow values used in the experiments of the temperature sensor responses of the Swan-Ganz catheters were 0,109 m/s, 0,218 m/s, 0,328 m/s, 0,437 m/s, 0,546 m/s, 0,655 m/s e 0,71 m/s. The values found for the response parameters (from equation 20) to the catheter impulse at a velocity of 0.109 m/s are shown in Table 2.

Velocities (m/s)
0,109
0,218
0,328
0,437
0,546
0,655
0,710

Table 1. Flow values used in the experiments of the temperature sensor responses of the Swan-Ganz catheters were 0,109 m/s, 0,218 m/s, 0,328 m/s, 0,437 m/s, 0,546 m/s, 0,655 m/s e 0,71 m/s. The values found for the response parameters (from equation 20) to the catheter impulse at a velocity of 0.109 m/s are shown in Table 2.

Table 2. Response parameters values from equation.

Parameter	Value
a	1.5430
B	0.1628
b	0.1911
C	-2.2985
c	14.8682
D	-1,1356
d	39.9937

In Figure 9, illustrated in the same graph are the curves of the temperature measurement of the result of the sudden injection measured by the Swan Ganz catheter and the result of the convolution between the fast sensor measurement and the estimate with minimum error found with the proposed algorithm. It is clear that, visually, the curves are very close, and therefore the estimate is very good.

The test was performed for all velocities in Table 1. The curves corresponding to the responses for all flows tested are shown in Figure 11. The responses make it clear that for velocities greater than 0.437 m/s, the response to the catheter impulse depends very little on the flow velocity, and for lower velocities the response varies, but not very significantly.

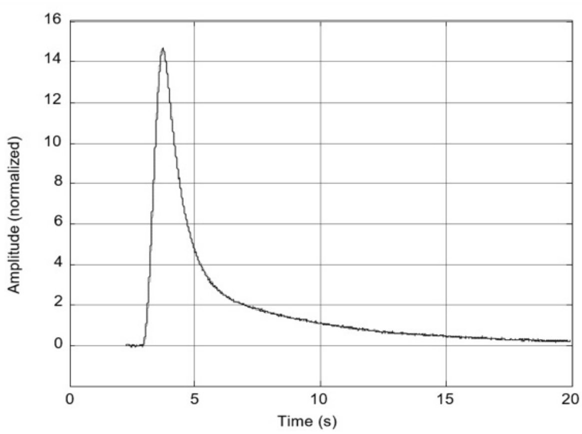


Figure 9. Temperature measured with temperature sensor of the Swan-Ganz catheter (continuous curve) and curve obtained by the convolution between the temperature measured with the rapid sensor and the optimal estimate for the catheter response (dotted curve). The difference between the two curves is practically imperceptible, which shows that the estimate is excellent. From the simulation, it is possible to plot the response to sensor impulse at a velocity of 0.109 m/s. This answer is illustrated in Figure 10.

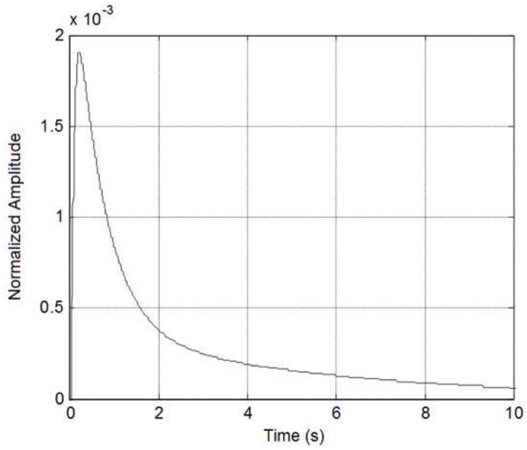


Figure 10. Response to the impulse of the temperature sensor built into the Swan-Ganz catheter.

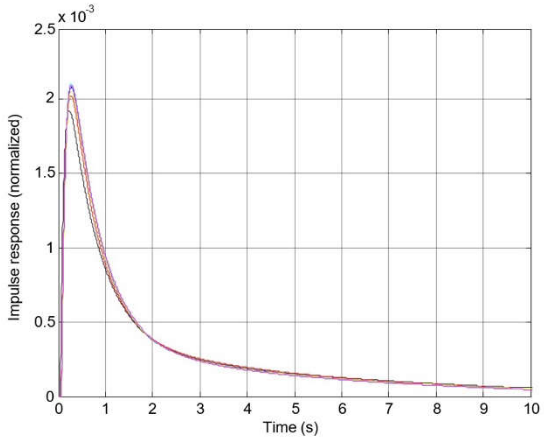


Figure 11. Summary of impulse responses from the Swan-Ganz catheter temperature sensor. The velocities range from 0.109 to 0.65 m/s, with the upper curves having greater amplitudes. The data for 0.71 m/s could not be used due to a technical problem that occurred in the acquisition process.

CONCLUSIONS

The result presented in Figure 11 extended the methodology proposed by da Rocha et al. (2005) to characterize the response of temperature sensors. One of the advances was that da Rocha's work performed the test for flows greater than 0.200 m/s, and the present work carried out tests for flows from 0.100 m/s. The results presented in this paper showed that the response of the catheter temperature sensor depends very little on the flow velocity of the surrounding fluid. The main reason for the little dependence on the catheter response in relation to the flow velocity is probably due to the presence of the protective epoxy layer. Heat must diffuse through this layer and, apparently, the removal of heat from the catheter surface occurs at a faster rate than the diffusion rate, which makes the process relatively independent of diffusion. In another paper that will be submitted in the near future, the simulations will show that this low dependence on the flow velocity becomes very important for the robustness in the deconvolution process proposed later. However, the biggest advance was the establishment of a methodology that does not require the ability to generate a stimulus that does not have zeros below a very low frequency. This methodology is not only useful in the present problem but may also be useful in several other practical situations in which it is necessary to characterize the impulse response of temperature sensors.

Acknowledgments

Financeira de Estudos e Projetos/Financial Studies and Projects (FINEP), Fundação de Amparo à Pesquisa do Estado do Tocantins/Tocantins State Research Support Foundation (FAPT), University of Brasília (UnB) and Federal University of Tocantins (UFT).

REFERENCES

Guyton AC, Hall JE 2002. Tratado de Fisiologia Médica. Editora Guanabara Koogan S.A.
Burton AC 1997. Fisiologia e Biofísica do Coração. Editora Guanabara Koogan S.A.
Stewart GN 1987. Researches on the circulation time and on the influences which affect it. IV. The output of the heart. J. Physiol; 22:159-183.

Marushak GF, Schauble JF 1985. Limitations of thermodilution ejection fraciton: degradation of frequency response by catheter mounting of fast-response thermistors. Critical Care Medicine; 138:679-682

Urban P, Scheidegger D, Gabathuler J et al. 1987. Thermodilution determination of right ventricular volume and ejection fraction: A comparison with biplane angiography. Crit Care Med; 15:652- 655.

Dhainaut J, Brunet F, Monsallier J, et al. 1987 Bedside evaluation of right ventricular performance using a rapid computerized thermodilution method. Crit Care Med 1987; 15:148-152.

Fegler G 1957a. Measurement of cardiac output in anaesthetized animals by a thermo-dilution method. Q. J. Exp. Physiol. Cogn. Med. Sci. 39:153-164.

Fegler G 1957b. The reliability of the thermodilution method for determination of the cardiac output and the blood flow in central veins. Q. J. Exp. Physiol. Cogn. Med. Sci.; 42:254-266.

Gans W, Swan HJ 1972. Measurement of blood flow by thermodilution. Am. J. Cardiol.; 29:241-246.

Trautman ED, Newbower RS 1984. The development of indicator-dilution techniques. IEEE Transactions on Biomedical Engineering, 1984; BME-3112:800-807.

Trautman ED, Newbower RS 1988. Thermodilution measurement of cardiac output. Webster JG Editor. Encyclopedia of medical devices and instrumentation. New York: John Wiley.

Ferris S, Konno, M. 1982. In vitro validation of a thermodilution right ventricular ejection fraction method Article. In: Journal of Clinical Monitoring; 81:74-80.

da Rocha AF 1997. The Dynamic Behavior of Thermistor Probes. Dissertation for degree of Doctor of Philosophy. Universityof Texas, Austin, May 1997.

Magalhães MC, Machado, GV 2004. Sistema para Medida do Débito Cardíaco e da Fração de Ejeção Baseado em Termodiluição. Universidade de Brasília, dissertação de graduação. Brasília, DF, Brazil, 2004.

dos Santos, I 2000. Proposta de um método para medição da fração de ejeção doventrículo direito. Dissertação Mestrado em Engenharia Biomédica, Departamento de Engenharia Elétrica, Universidade de Brasília, Brasília, 2000.

dos Santos I, Rocha AF. Identificação da resposta ao impulso de termistores usando uma rede neural artificial. In: IV Conferência Brasileira de Redes Neurais, 1999, S. J. dos Campos. Proceedings do IV Conferência Brasileira de Redes Neurais, 1999:314-319.

Novato, LR 2004. Projeto e Implementação de um Sistema de Termodiluição Baseado em Labview. Projeto Final de Graduação em Engenharia Elétrica, Departamento de Engenharia Elétrica. Universidade de Brasília UnB, Brazil, 2004.

da Rocha AF, dos Santos I, Nascimento FAO, de Melo MDB 2005, Haemmerich D, Valvano JW. Effects of the time response of the temperature sensor on thermodilution measurements. Physiol. Meas; 266:885-901.

de Melo MDB, Araujo HX, da Rocha, AF. 2016. A Novel Methodology for Signal Recovery in Linear Time Invariant Systems for Medicine Applications. PrzeglądElektrotechniczny, ISSN 0033-2097, R. 92 NR 010/2016.

de Melo MDB, Araujo HX, Gobira SR, Rocha, AF 2019. An Algorithm for the Enhancement of Thermodilution Signals. International Journal of Development Research; 0904:26890-26896.

Ling J, Guerra JM, Robey BL, Winter DC 1999. Continuous Cardiac Output Determination from Blood Pressure Waveforms Using a Fuzzy Logic Model. Proceedings of the First Joint BMES/EMBS Conference. 1999 IEEE Engineering in Medicine and Biology 21st Annual Conference and the 1999 Annual Fall Meeting of the Biomedical Engineering Society Cat. N, Atlanta, GA, USA; 2:931, doi: 10.1109/IEMBS.1999.804085.

Semnani RJ et al.1998. Improvement of Mini-bolo Thermodilution using Optimal Subspace-based Matrix Filter. IEEE Transactions on Computers in Cardiology; 25.

Holm C et al. 2001. Arterial thermodilution: na alternative to pulmonary artery catheter for cardiac output assessment in burn patients. Elsevier Science Ltd and ISBI.

Heerdt PM et al. 2001 Flow Dependency of Error in Thermodilution Measurement of Cardiac Output During Acute Tricuspid Regurgitation. Journal of Cardiothoracic and Vascular Anesthesia. 2001; 152:183-187.

Hamilton PS et al. 2002 Artifact Reduction for Continuous Thermodilution Measurements of Cardiac Output. Proceedings of the Second Joint EMBS/BMES Conference Houston, TX, USA, October 23-26.

Humphrey CR, Cezeaux JL, Schreiner S 2002. The Design and Fabrication of a Closed Loop Steady Flow System for Study of Thermodiluition. Proceedings of the IEEE 28th Annual Northeast Bioengineering Conference IEEE Cat. No.02CH37342, Philadelphia, PA, USA, pp. 97-98; doi: 10.1109/NEBC.2002.999483.

Zhao X, Mashikian JS, Panzica P, Lerner A, Park KW, Comunale ME 2003. Comparation of Thermodiluiton Bolus Cardiac Output and Doppler Cardiac Output in the Early Post-Cardiopulmonary Bypass Period. Journal of Cardiothoracic and Vascular Anesthesia; 172: 193-198.

Kuper M 2004. Continuous cardiac output monitoring. Current Anaesthesia& Critical Care; 15:367-377.

Yelderman, MD 2004. CEO, Monterey Medical Solutions, CA, USA. Continuous Measurement of Cardiac Output Using Stochastic System Identification Techniques. he 26th Annual International Conference of the IEEE Engineering in Medicine and Biology Society. San Francisco, CA, 5360-5362.

Carvalho VFS 2006. Caracterização da resposta temporal de sensores de temperatura dos cateteres Swan-Ganz. Trabalho Final de Gradução, Engenharia Mecatrônica, Universidade de Brasília UnB, Brasília, Brasil, dezembro de 2006.

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