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IN VITRO FLOW CONTROL PERFORMANCE OF THE APICAL-AORTIC BLOOD PUMP IN LEFT VENTRICULE FAILURE CONDITIONS

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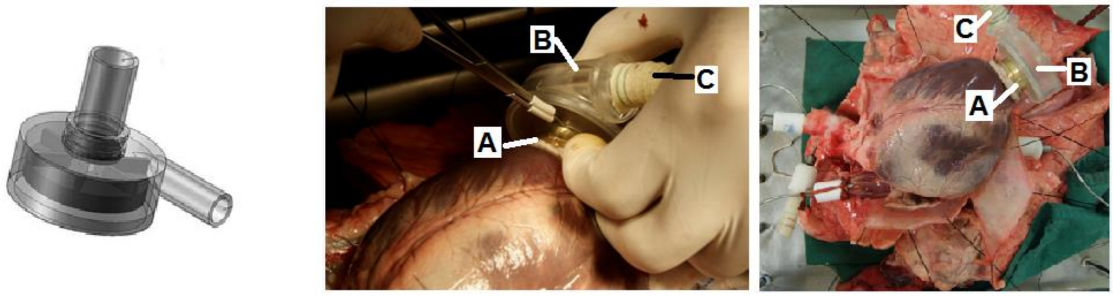
Abstract. *This paper shows the Blood Flow Control (FwC) performance to adjust rotational speed of an Apical-Aortic Blood Pump (AABP) in order to provide an adequate flow to left ventricle in different patient conditions. AABP is a Left Ventricle Assist Device (LVAD) developed for long-term assistance totally implantable. This LVAD is implanted directly in apical left ventricle and connected aorta by cannula. FwC uses (Proportional – Integral) PI Control to adjust rotational speed in order to provide blood flow. FwC does not use sensor for feedback, there are an estimation system to provide blood flow measurement. Control strategy has being studied in a Hybrid Cardiovascular Simulator (HCS) as a tool that allows the physical connection of AABP under evaluation. In addition, HCS allows changes of some cardiovascular parameters in order to simulate specific heart disease: Ejection Fraction (10 – 25%) and heart rate (50 – 110 bpm). FwC was able to adjust blood flow with steady error less than 2%. Results demonstrated that FwC is adequate to LVAD control in different left ventricle failure conditions.*

Keywords: *Flow Control, Left Ventricle Assist Device, Centrifugal Blood Pump, Heart Failure, Artificial Organs.*

1. INTRODUCTION

The Apical Aortic Blood Pump (AABP) is a centrifugal blood pump with ceramic bearings, Fig. 1, that has been developed in our laboratories to be used as left ventricular assist device (LVAD) for long term circulatory assistance totally implantable. This LVAD was designed to be attached directly to the left ventricular apex and connected aorta by an outlet cannula (da Silva, *et al*, 2013). Initial studies in similar centrifugal pump were conducted by Bock *et al* (2011).



(a) (b)

(a) Apico Aortic Blood Pump (AABP) design; (b) AABP implant position. A = left ventricular apex; B = AABP; C = outlet cannula.

Figure 1. Apico Aortic Blood Pump.

This device is composed of: continuous flow centrifugal blood pump, Brushless Direct Current (BLDC) motor, controller to drive the motor and battery system. Centrifugal pumps represents the majority of research currently developed and applied, which allows operation at lower motor speeds (approximately 2,000 rpm) than the continuous flow axial pumps (approximately 10,000 rpm); obtain lower rates of hemolysis, i.e., less damage to blood elements, have anatomically compatible dimensions and reach the estimated life together in support of 2 years. AABP shows mean normalized index of hemolysis was 0.009 g/100 L (±0.002 g/100 L) (da Silva, *et al*, 2013).

The centrifugal pumps are controlled by varying the rotor (impeller) speed. AABP successful operation depends on appropriate rotational speed control system, ensuring: 1) no reverse flow through the pump during left ventricle diastolic phase, usually occurs when pump speed is set too low causing flow from aorta to left ventricle through the pump; and 2) aortic valve correct opening, avoiding later valve stenosis. In order to achieve these requirements is necessary to control the VAD flow (Leão, *et al*, 2013).

The goal VAD control is to provide a satisfactory cardiac output while sustaining an appropriate pressure perfusion. To achieve adequate control system, development has been carried out to estimate the pump parameters as VAD flow and differential pressure. Sensors is not desirable, because can result in thrombus formation and system reliability reduction. Systems that detected the changes in the body's metabolic demands is one of the important objective in designing VAD control system (AlOmari, *et al*, 2013).

In study by Giridharam *et al* (2004), they show control strategy of maintaining a constant average pressure difference between the pulmonary vein and aorta. Results also showed that control strategy was able to maintain total flow rate to its physiologically normal limits.

Fu and Xu (2000) propose a sensorless fuzzy controller utilizing pump motor speed and current. Results indicated that proposed control can achieve required pump flow, but proportionality between pump flow and heart rate has been limited factor.

The blood Flow Control (FwC) adjusts rotational speed of an Apical-Aortic Blood Pump (AABP) in order to provide an adequate flow to left ventricle. The FwC is part of a novel technique for automatic control of speed of VAD, which operates at harmoniously with pressure control physiological system. This technique adjusts VAD speed via FwC and fuzzy control system (FzC). Figure 2 shows block diagram of the novel technique, where FwC is in 2nd Layer (Leão, *et al*, 2015). The flow estimator is in patent process; its error is 0.23 ± 0.07 L/min (confidence interval 95%).

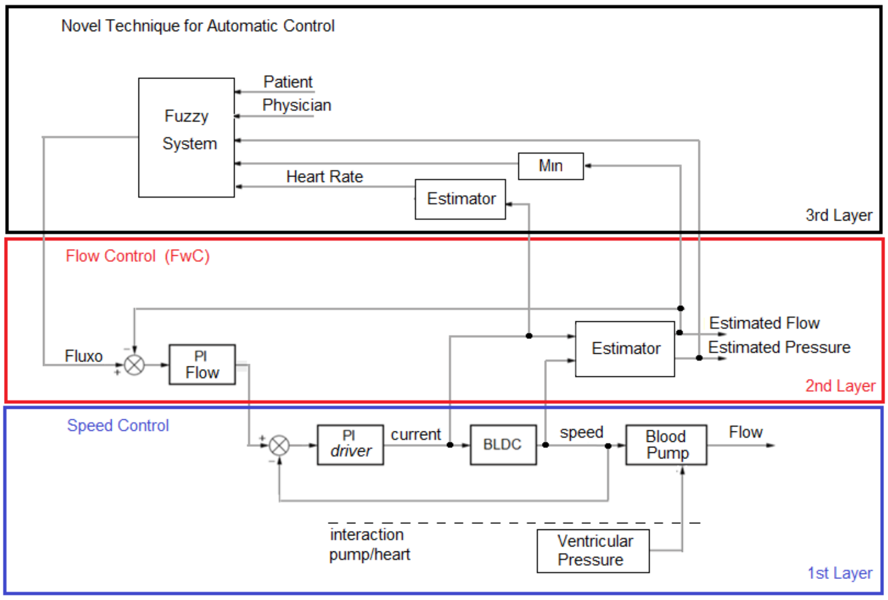
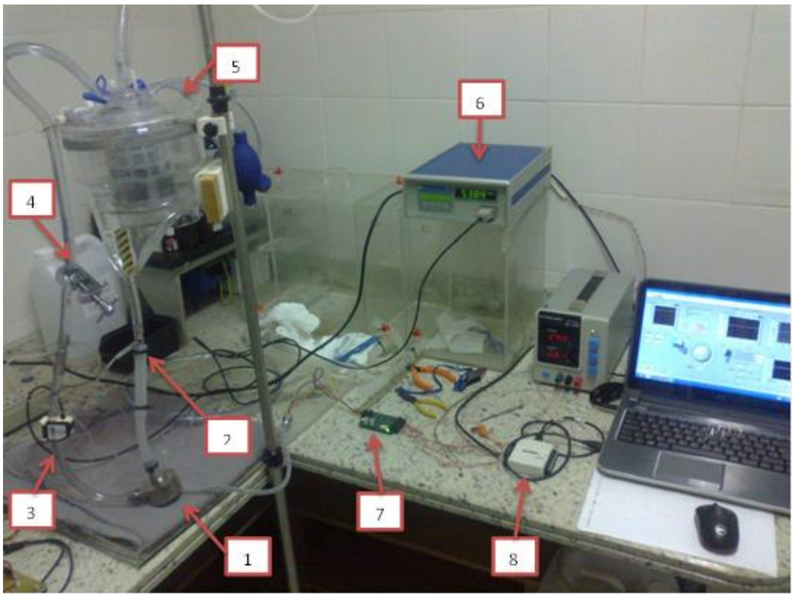


Figure 2. Novel Technique for Automatic Control of speed of VAD.

This paper has been divided into two main parts; the first consist of response dynamics analysis of 1st and 2nd layers; the second part shows FwC performance analysis. Thus, the main contribution of this paper is FwC validation under heart failure conditions.

2. METHOD

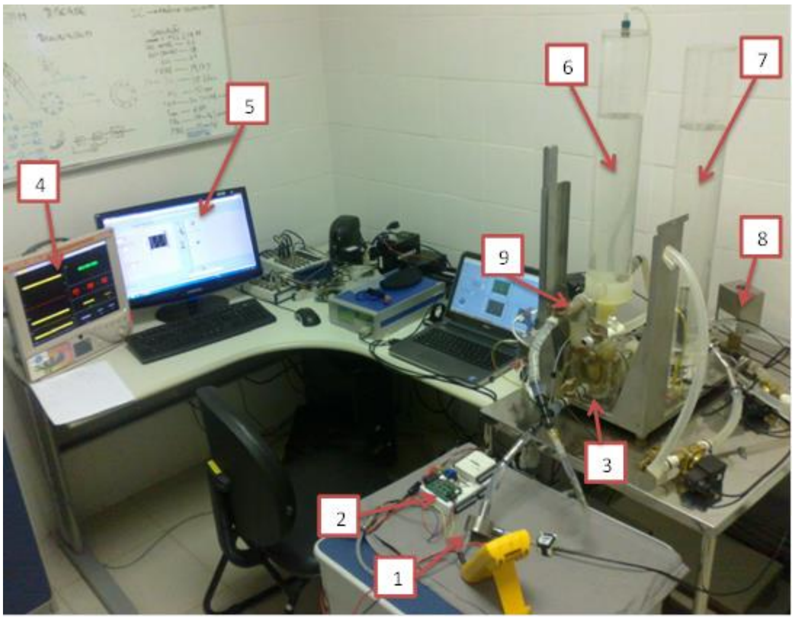
To analyze system dynamics was used a mock loop to simulate a ventricle without contraction, Fig. 3 (Leão *et al*, 2012). Flow is monitored by an ultrasonic flowmeter (HT110, Transonic, New York, USA). AABP input and output pressures are measured by a pressure monitor (DX2020, Biomedical Dixtal, São Paulo) (Leme *et al*, 2011). Circuitry consists: 5 L acrylic reservoir; silicone flexible tubes 3/8" to the connection between the reservoir and the AABP; tourniquet to adjust flow; and a solution of distilled water (50%) and glycerine (50%) to simulate the viscosity of blood with hematocrit of 45%, i.e., 3.2 mPas. The temperature is maintained at 25 ° C to simulate the viscosity under physiological conditions (Legendre, 2009).



1-AABP–Apical-Aortic Blood Pump; 2-pressure connection; 3-flow transducer; 4-tourniquet; 5- acrylic reservoir; 6-flow monitor; 7-BLDC driver; 8-acquisition system.

Figure 3. Mock loop.

To analyze FwC under heart failure condition was used a Hybrid Cardiovascular Simulator (HCS), Fig. 4 (Fonseca, *et al*, 2011). This tool allows physical connection of AABP under evaluation. In addition, HCS allows changes of some cardiovascular parameters in order to simulate specific heart disease: Ejection Fraction (10, 15 and 25%) and heart rate (50, 80 and 110 bpm).



1-AABP-Apical-Aortic Blood Pump "implanted"; 2- Acquisition system; 3-AABP inlet connected to apex ventricle; 4-arterial pressure monitor; 5-computational section (right heart); 6-Aorta; 7- left atrium; 8-sistemic vascular resistance; 9-AABP outlet connected to Aorta.

Figure 4. Hybrid Cardiovascular Simulator

Heart Failure conditions were obtained unpublished data from health service Institute Dante Pazzanese of Cardiology in Sao Paulo.

FwC was implemented in software Labview (2010, National Instruments, Austin, USA) and acquisition system (USB-6009 e USB-6212, National Instruments, Austin, USA). This equipment was used for to obtain transfer function in 1st and 2nd layers.

3. RESULTS

Equation (1) shows approximated transfer function (TF) (Laplace dominion) experimentally obtained (identified) ($G_{(s)}$) at mock loop tests with AABP, Fig. 3. TF input is BLDC rotational speed and TF output is VAD mean flow. BLDC driver has minimum rotational speed, because sensorless commutation, then TF input was calculated by Eq. (2).

$$G_{(s)} = \frac{0{,}00192}{s^2 + 0{,}99\,s + 0{,}64} \tag{1}$$

$$RPM_{G_{(s)}} = RPM_{VAD} - 513 \tag{2}$$

Equation (3) shows flow control (PI) TF ($Gpi_{(s)}$), its input is flow error and output is reference voltage to BLDC driver. Close loop TF ($Gcl_{(s)}$) was calculated according 2nd layer (FwC, Fig. 2), Eq. (4), its input is VAD flow reference and output is VAD mean flow (actual).

$$Gpi_{(s)} = \frac{0{,}6\,s + 0{,}05}{s} \tag{3}$$

$$Gcl_{(s)} = \frac{0{,}384\,s + 0{,}032}{s^3 + 0{,}99\,s^2 + 1{,}024\,s + 0{,}032} \tag{4}$$

Figure 5 shows G(s) step response with simulated and experimental data. Figure 6 shows Gcl(s) step response with simulated and experimental data.

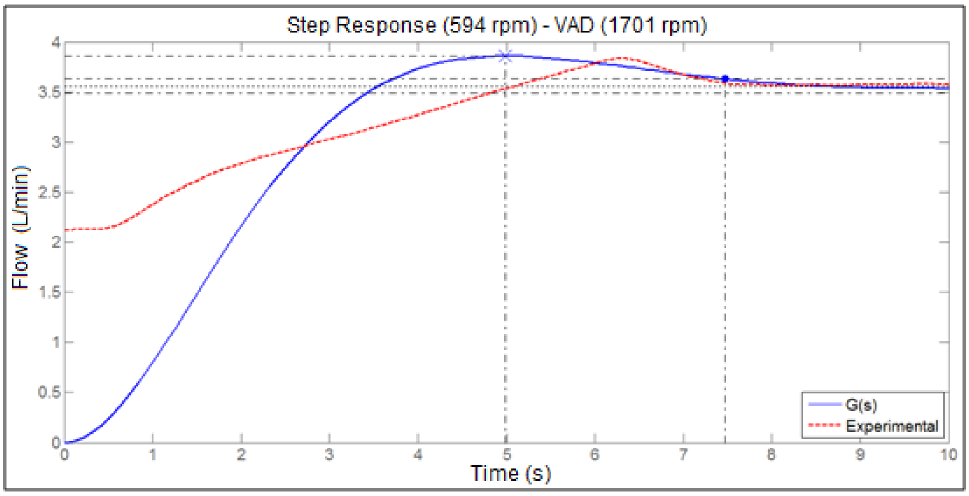


Figure 5. Step Response G(s).

Approximated TF was able to represent dynamics system, morphology and steady value indicate sufficient precision to calculate mean flow. Initial conditions were different between simulated and experimental data, because flow equal zero is not possible, due sensorless commutation.

Table 1 shows actual and estimated mean flow, with FwC adjusted to set point at 5.0 L/min, in heart failure conditions.

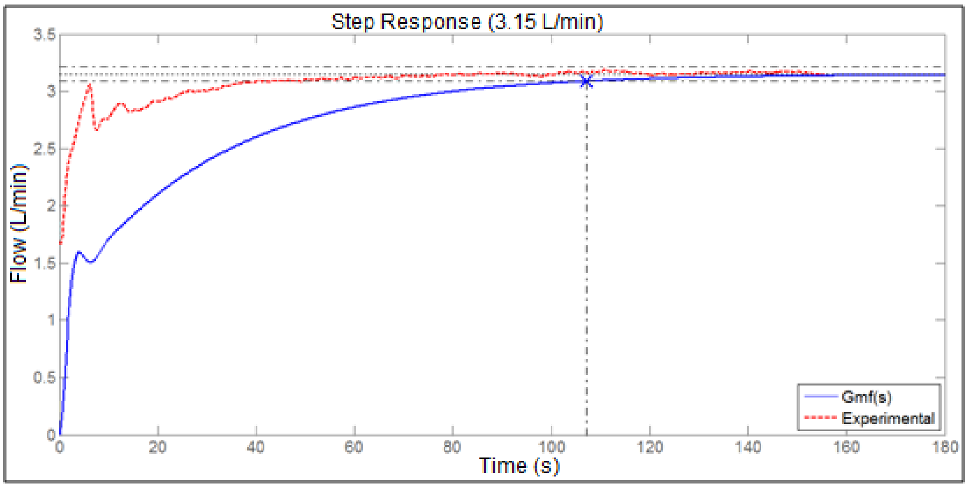


Figure 6. Step response Gcl(s).

Table 1. Flow Control (FwC) performance in Heart Failure conditions (Set-point: 5.0 L/min).

EF	HR (bpm)	Actual Mean Flow (L/min)	Estimated Mean Flow (L/min)
10%	50	4.83 (4.77 -4.87)	5.00 (4.96 - 5.03)
	80	4.85 (4.80 - 4.92)	5.00 (4.98 - 5.02)
	110	4.82 (4.78 - 4.87)	5.00 (4.97 - 5.03)
15%	50	4.84 (4.75 - 4.91)	5.00 (4.95 - 5.04)
	80	4.78 (4.72 - 4.85)	5.00 (4.97 - 5.04)
	110	4.67 (4.61 - 4.73)	5.00 (4.96 - 5.03)
25%	50	5.19 (5.05 - 5.32)	5.00 (4.90 - 5.06)
	80	5.18 (5.05 - 5.30)	5.01 (4.96 - 5.06)
	110	4.98 (4.91 - 5.07)	5.02 (4.96 - 5.08)

Note: EF – Ejection Fraction; HR – Heart Rate

4. CONCLUSIONS

Dynamic system is adequate to adjust VAD rotational speed, considering others control layer and physiological effects in simulated patients.

FwC was able to maintain mean flow at reference, even under heart rate different conditions. Steady error value is less than 2%.

Future work will use the FwC to provide a control system that operates harmoniously with the physiological system.

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