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Speed Control of the Implantable Centrifugal Blood Pump to Avoid Aortic Valve Stenosis: Simulation and Implementation

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***Abstract*— This paper presents a computational simulation and implementation of electromechanical actuator performance of the Implantable Centrifugal Blood Pump (ICBP) as part of a speed controller study to avoid aortic valve stenosis. The ICBP as Left Ventricular Assist Device (LVAD) is an electromechanical device designed for long-term assist left heart in performing its functions. The centrifugal pumps are controlled by varying the rotor (impeller) speed. ICBP successful operation depends on an appropriate rotational speed control system, ensuring: 1) no reverse flow through the pump during left ventricle diastolic phase, and 2) aortic valve correct opening, avoiding later valve stenosis. A computational model of the actuator done in Matlab / Simulink (R2010b, Mathworks, Massachusetts, USA) was used in the simulations. Control and signal processing was used Labview (National Instruments, Austin, USA). Signals equivalent to intraventricular pressure and a variable rotational reference were used to evaluate motor and speed controller performance. Speed values were chosen so that pressure pump exceeds intraventricular pressure only after the opening of aortic valve. The proposed controller is Proportional-Integral (PI) type. The simulation results were satisfactory, no steady error in response speed. Practical tests showed satisfactory results to follow speed reference signal, as simulated. Future studies will evaluate the bands control.**

I. INTRODUCTION

A novel Implantable Centrifugal Blood Pump (ICBP) has been developed in our laboratories, to be used as Ventricular Assist Device (VAD) for long term circulatory assistance with a unique impeller design concept. This feature was called dual impeller because it combines a spiral-shaped cone with vanes to improve blood flow characteristics around the top inflow area to avoid blood clot due to stagnant flow. This work is part of ICBP muti-institutional project of a VAD for long term application [1].

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 represent the majority of research currently developed, which allows: 1)operation at lower motor speeds (approximately 2,000 rpm) than the continuous flow axial pumps; 2)obtain lower rates of hemolysis, i.e., less damage to blood elements; 3)have compatible anatomically dimensions and 4)reach the estimated life together in support of 2 years. Figure 1 shows the motor and ICBP.

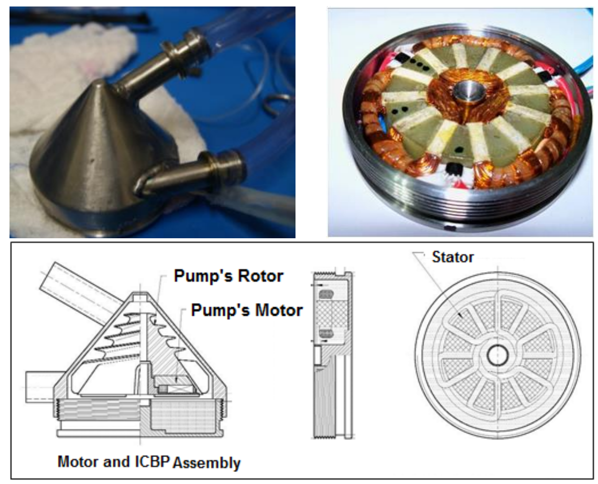

Figure 1. *Picture and Draw of the ICBP and BLDC Motor assembly.*

The ICBP as Left Ventricular Assist Device (LVAD) is an electromechanical device designed for long-term assist left heart in performing its functions. The centrifugal pumps are controlled by varying the rotor (impeller) speed. ICBP successful operation depends on an appropriate rotational speed control system, ensuring: 1) no reverse flow through the pump during left ventricle diastolic phase, and 2) aortic valve correct opening, avoiding later valve stenosis.

The brushless motors, BLDC, have been the main component of propulsion in the development of most of the VAD. Among the characteristics that make use in implantable pumps, there is the absence of brushes, which avoid wear observed in other electrical motors, and intolerable in implantable systems. The operation at high rotational speeds and small size are also factors that support this use [2].

A computational model was created to represent dynamics and control applied to actuator and driver of the ICBP. A reliable model is an important tool to set up parameters of ICBP controller [3]. Future applications may be integrate the model to a cardiac simulator [4]. The Matlab/Simulink (MATHWORKS, R2010b, Massachussets, USA) platform was chosen as virtual environment for simulation because it allows data and results integration with

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other software, including Comsol Multiphisycs, that is being used for modeling of ICBP.

The dynamic model is necessary to study transients of the motor drive system and steady state. The instantaneous currents are crucial for power computation and electromagnetic torque is important for drive system performance evaluation. These features become a significant factor in appliances such as VAD [5].

The goal of long-term devices for destination therapy, to reduce the dependence of the constant clinical monitoring, and to allow the return to home and the consequent improvement in the quality of life for patients, involves the use of control theory in order to achieve this goal. Challenge of long-term care is the responsibility of control algorithms that respond to the changing demands of patients and the difference between patients with heart failure [6].

Parnis *et al*, shows a proportional speed controller of VAD Jarvik 2000 ®, using the motor current fundamental frequency as heart rate input for the controller. The linear relationship between heart rate and rotation is the major limitation of this control technique [7].

Giridharan *et al*, propose a PI controller considering rotation set to differential pressure. Heart failure conditions were computationally simulated to evaluated controller perform and showed satisfactory results. The use of implantable sensors is a limiting factor. To overcome this limitation was proposed an estimator based on Kalman filters, but with worse results than sensor system [8].

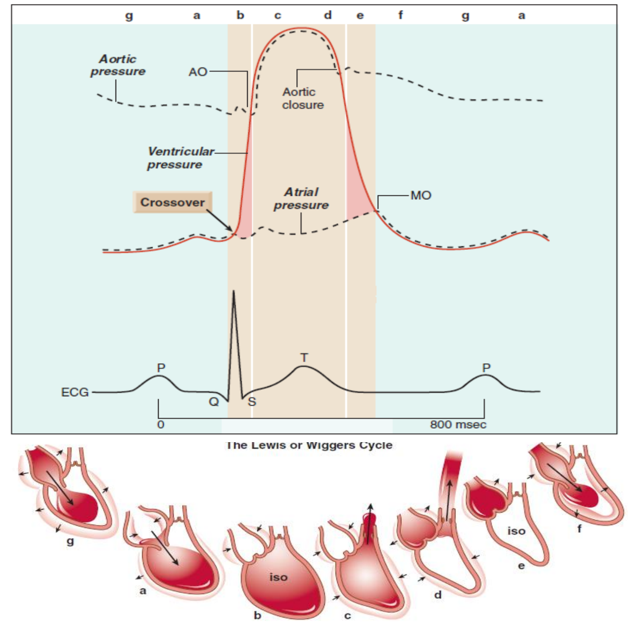
This paper has been divided into three main parts; the first consist of mechanisms of cardiac contraction and relaxation; the second part that describes the virtual implementation with help of Matlab / Simulink blocks diagram to represent the electromechanical actuator and physiological parameters; and third part that shows the computational model results to speed control, implementation results and discussion about physiological impact in ICBP perform. Furthermore, the main contribution of this paper is to study BLDC speed control with torque and speed variables under actual physiological conditions.

II. MECHANISMS OF CARDIAC CONTRACTION AND RELAXATION

Braunwald [9] shows the cardiac cycle. Figure 2 shows Left Ventricular (LV) events. The three basic events are:

1) LV Contraction (letters b and c); 2) LV Relaxation (letters d and e); and 3) LV Filling (letters f, g and a) shown in fig. 2. Although similar mechanical event occur in right ventricle this work focused on left ventricle. The cycle conceived by Wiggers and fully assembled by Lewis is important information on the temporal sequence of events in the cardiac cycle.

In figure 2 was showed cardiac cycle to normal person, but ICBP is indicated to person with heart failure. However, person with heart failure has cardiac cycle similar to normal person, only with amplitude pressure modified (in most cases). Therefore simulations in this paper used cardiac cycle data to normal person.



Legend. a = atrial systole or booster; b = isovolumic contraction; c = maximal ejection; d = start of relaxation and reduced ejection; e = isovolumic relaxation; f = rapid phase; g = slow filling; AO = aortic valve opening; MO = mitral valve opening; ECG = electrocardiogram. Cycle length of 800 miliseconds for 75 beats/min.

Figure 2. *The mechanical events in the cardiac cycle. (Braunwald, 2012. Adapted).*

Soleimani *et al* shows a work in which the observed incidence of aortic insufficiency in about 9% of patients with continuous flow left VAD support with an average of 374 days of support [10].

III. SIMULINK MODEL

The BLDC model implemented in MATLAB/SIMULINK used blocks of the SimPowerSystems toolbox. The BLDC was simulated with a block of Permanent Magnet Synchronous Machine (PMSM) with a trapezoidal back electromotive force (BEMF) signal. Electrical and mechanical parts of the machine are represented by a second-order state-space model.

The BLDC is connected to an inverter and supplied by a variable source of Direct Current (DC). This source is adjusted by a Proportional Integral (PI) control with feedback of motor speed. A measures block was included to measure stator current, i.e., power of BLDC. Figure 3 show BLDC block diagram implemented. This model was validated to represent BLDC motor as electromechanical actuator of ICBP [11].

PI parameters were selected by Ziegler-Nichols method. Gains 1 and 0.1, respectively, proportional and integral.

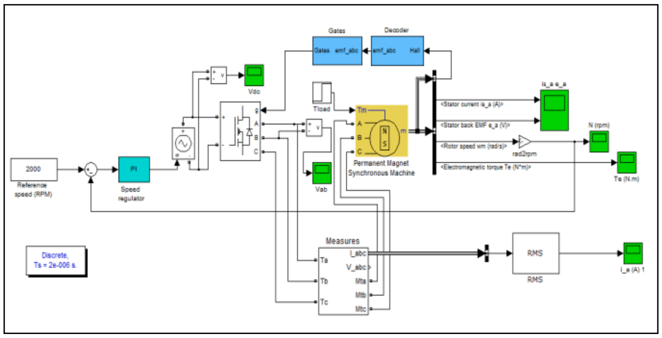
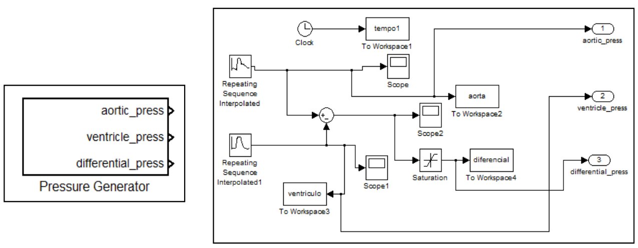


Figure 3. *Block Diagram BLDC.*

Subsystem was made to generate pressure ventricle and aortic signal (75 beats/min), Fig. 4. Differential pressure between aortic and ventricle pressure was used as torque signal. In Fig. 3, Tload block was replaced by Pressure Generator subsystem. Load torque value has been obtained from tests previously carried out.



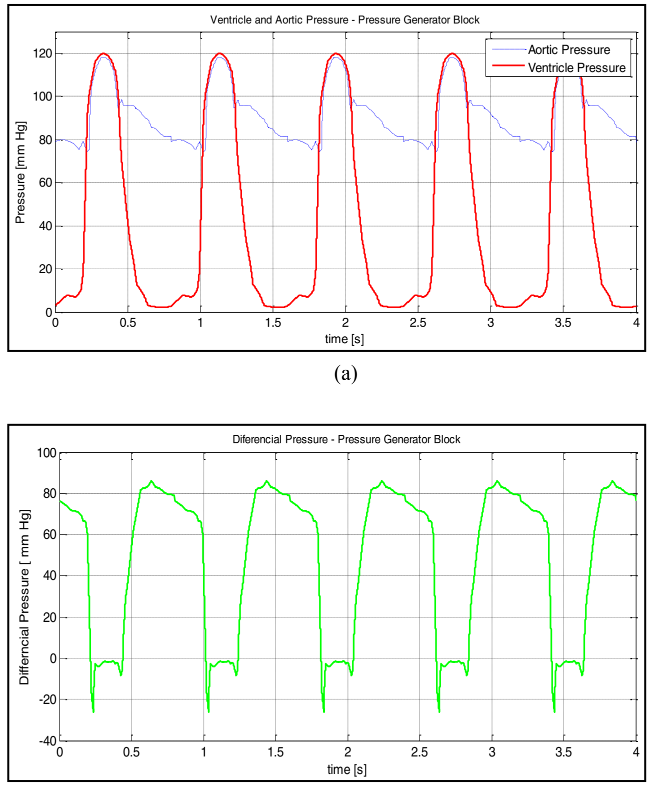
(a) (b)

Figure 4. *Pressure Generator. (a) Subsystem block; (b) Internal Diagram Block.*

Figure 5 show waveform aortic, ventricle and differential pressure generated by subsystem.

Figure 6 presents differential pressure signal as torque signal to simulate pressure variation in pump inlet. A saturation block was used to limit negative torque. It because BLDC model works as motor when receives positive torque signal and as generator when receives negative torque. Furthermore, negative pressure means pressure loss through aortic valve, condition that is not of interest to this paper.

Speed signal was proposed in pulses so that ensure aortic valve opening, avoiding aortic stenosis. Pressure pulse was synchronized with the cardiac cycle, while maintaining low-speed in start of systole (b phase in Fig. 2) and rising speed after start of relaxation (d phase in Fig. 2). Speeds used in simulations are typical VAD speeds found in centrifugal pumps, values between 1800 and 2200 rpm [12, 13].



(a)

(b)

Figure 5. *Signal pressure generated by subsystem. (a) Ventricle and Aortic Pressure; (b) Differential Pressure.*

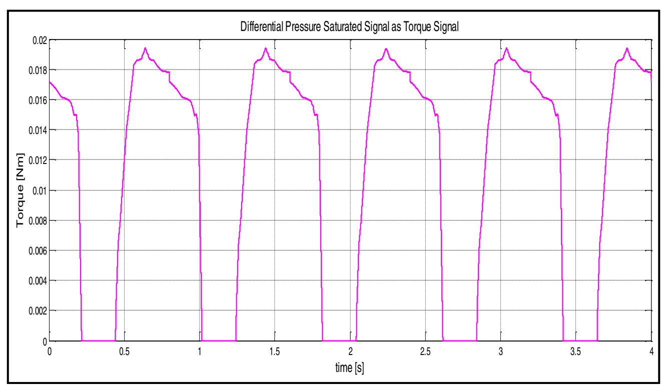
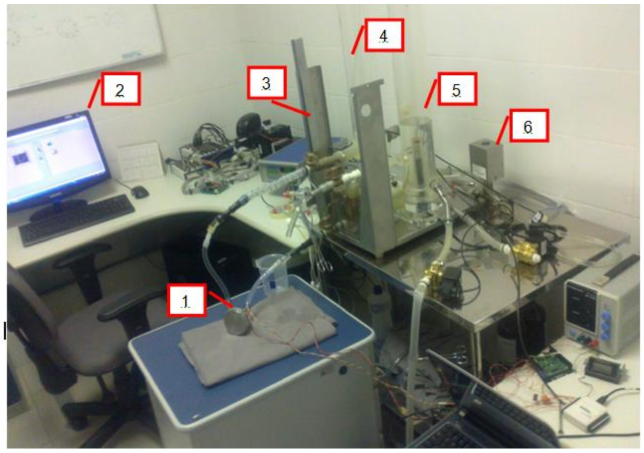


Figure 6. *Differential Pressure Signal as Torque Signal.*

Implementation of control was assessed in a cardiovascular hybrid simulator (CHS). The parameters of the simulator were maintained in: 75 bpm heart rate, 100 mmHg mean aortic pressure, and ejection fraction of 40%.

Figure 7 show CHS during tests.



Legend: 1-ICPB; 2-computer to simulated right heart; 3-left heart (ventricle); 4-Aortic compliance; 5-left atrium; 6-peripheral vascular resistance.

Figure 7. *Cardiovascular Hybrid Simulator(CHS).*

IV. Results

Figure 8 shows simulation result about speed control. Blue dashed line represents speed reference (synchronized with cardiac cycle). Red solid line represents speed BLDC response with PI controller. Green dotted line represents ventricle pressure signal (without scale, only to check synchronism). The steady speed error was less than 2%.

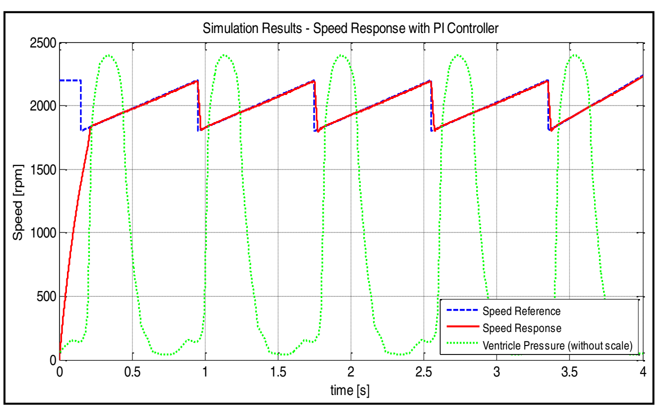


Figure 8. *Speed controller response.*

In speed pump of 1800 rpm the ICBP works only for no reverse flow. In this case, ventricle performs blood ejection through aortic valve. With speed increasing ICBP generates blood flow to aortic artery.

Figure 9 shows speed response panel after control implementation with Labview (National Instruments, Austin, USA) and data acquisition hardware (DAQ USB-6009, National Instruments, Austin, USA).

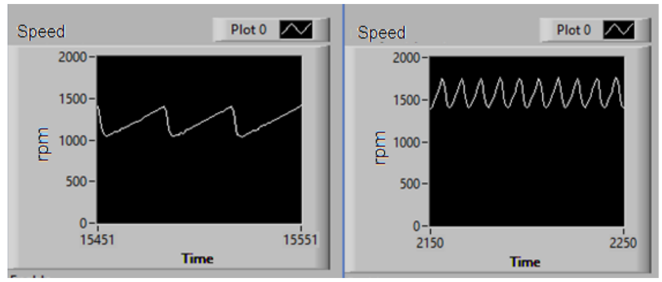


Figure 9. *Speed response panel.*

Due to dynamics of the ICBP there is signal attenuation of the speed reference to values above 80 bpm (beats per minute) in 30%. A compensator should be designed to minimize this attenuation.

V. Conclusion

Simulink computational model was considered satisfactory to represent BLDC dynamics and to generate physiological signals (aortic and ventricle pressure).

PI control was considered satisfactory to speed adjust, even with speed variable reference and torque variable. Steady speed error was less that 2%, it is appropriated to generate flow through ICBP.

Practical results are consistent with simulations, especially under actual load conditions. To evaluate the bands control is important to ensure synchronism with heart rate.

The controller results are promising for its use as a pulsatile flow VAD synchronized with the natural heart rate, especially to guide its construction.

Future work will implement the controller in "In Vivo" tests.

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