AN ELECTROMECHANICAL PUMPING SYSTEM FOR THE CARDIOVASCULAR SIMULATOR

Jeison Fonseca, jfonseca@dantepazzanese.org.br
Aron Andrade, aron@dantepazzanese.org.br
Institute Dante Pazzanese of Cardiology

Daniel Legendre, dlegendre@dantepazzanese.org.br
Institute Dante Pazzanese of Cardiology

Eduardo Bock, ebock@dantepazzanese.org.br
Institute Dante Pazzanese of Cardiology

Juliana Leme, juleme@dantepazzanese.org.br
Institute Dante Pazzanese of Cardiology

José Biscegli, biscegli@dantepazzanese.org.br
Institute Dante Pazzanese of Cardiology

Julio Lucchi, jclucchi@uol.com.br
Technological Institute of Aeronautics

Abstract. *This work presents the development of the electromechanical pumping system to be used in our Cardiovascular Simulator. The Cardiovascular Simulator that has been developed in our laboratories is a mock loop automatically controlled by computer, which main goal is to provide an auxiliary tool to develop Ventricular Assist Devices (VAD) control strategies. The simulator system is composed basically by four modules: an electromechanical pump; two mechanical valves - to drive the flow direction; a compliance chamber – to provide compliance effect to mock; and a tourniquet to provide hydraulic resistance. In order to pump the working fluid, a Brushless Direct Current Motor (BLDCM) has been used to drive a planetary roller screw, changing the motor rotation into linear displacement of a diaphragm. A BLDCM electronic controller is made in two levels: the first level is analogical and drive the rotor spinning movement based on the signal of three Hall sensors assembled at the motor structure; the second level is digital and can make modulation of roller screw movement, based on other three Hall sensors – two of these is used to impose the limits for roller screw displacement, and the third to get fillment information of pump chamber. The pump chamber fillment is passive, and happens according to reverse venous pressure. Once pump chamber is completely full, the electronic controller driver the BLDCM to eject fluid from chamber – the pumping effect. Since roller screw displacement can be modulated, the internal chamber volume can be adjusted changing "heart elastance". This effect is important to simulate some pathology like Congestive Heart Failure (CHF). The results that has been obtained from the simulator with BLDCM in terms of pressure and flow measurements have shown the pump stability, therefore the reproducibility and repeatability conditions could be improved.*

Keywords: *Cardiovascular Simulator, Mock Loop, LabVIEW®, Brushless DC Motor.*

1. INTRODUCTION

Nowadays, heart disease has been the main issue of deaths around the world. In advanced cases the only solution is the heart transplant. However, transplant is hard to active because of some difficulties on organ capitation system or because of great amount of patients waiting for a transplant. Therefore, the development of artificial hearts is important, and the study of a simulation tool can help this development. Simulation provides a good way to optimize the control of this artificial heart pumps.

Thus, this work presents the development of an electro-fluid dynamic simulator that will be used as tool in order to optimize the assistant artificial pump devices.

The simulation of the human cardiovascular system has been studied for years. In Physbe (1966) - a Physiological Simulation Benchmark Experiment presented by McLeod the cardiovascular system model is divided in nine compartments: right heart, lungs, left heart, aorta, head, arms, legs, trunk and cava vein. It was assembled in Simulink (MatLab®) environment.

In Lucchi, 1999 is shown a simulation of the cardiovascular system based on electrical circuitry assembled on PSPICE - Simulation Program with Integrated Circuit Emphasis. In that work, the cardiovascular simulation was made based on concentrated parameters considering a pulsatile flow. The system was assembled with three models: the heart (ventricle and active atria chamber); the cardiovascular system (peripheral circulation, pulmonary circulation, coronary circulation and the bronchial shunt); and the baroreceptor reflex model. In that simulator, it is possible to change some variable values like elastances, compliances and pulmonary or peripheral resistances, allowing analyzing some patho physiological conditions. The Congestive Heart Failure (CHF) condition can be simulated through changes in ventricular elastance value. The tool assessment was obtained by simulating the ventricular assistance with two kinds of devices: with continuous or pulsatile flow (Lucchi, 1999).

In Bustamante(2004) a physical cardiovascular simulator was presented, where some device parameters can be controlled by computer and some sensors can acquire pressure signals from some points of the system. Simulation is represented by three modules: a pneumatic pump with cardiac valves in order to simulate the heart pump; a circulatory system which includes a compliance chamber and vascular resistance; and a control and instrumentation system. That simulator is controlled by computer and, when activated a diaphragm is displaced, generating a curve with a sinusoidal behavior that makes the fluid inside the chamber goes out to vascular loop. In that model, there are two compliances chambers with rubber plate on its basis that can expand depending on amount of fluid inside the chamber. The circulatory system resistance can be modeled in two ways: making shorter the tubes internal diameter or by computer that drives a tourniquet coupled to a step motor, controlling the tube occlusion. Control system and data acquire system are made by a computer running LabVIEW®. LabVIEW® adjusts the "heart rate" by an electro-valve that controls the air flow from a pneumatic compressor.

In Felipini *et. all.*(2008) and in Legendre *et. all.*(2008) a physical cardiovascular simulator is presented. It consists of four components: pump system (left ventricle), circulatory system, test compartment module and acquisition and analysis monitoring system. This simulator use a Direct Current motor (DC) and a piston to push the diaphragm as pumping system, creating pulsatile flow.

2. MATERIALS AND METHODS

Our system can simulate the left ventricle and the peripheral circulation considering concentrated parameters. The connection diagram is showed in Fig. 1. It was assembled with four main modules: a reservoir, working as a "passive atria", where the ventricle pre-charge can be adjusted by the height of reservoir; a pumping system composed by a BLDCM, a planetary roller screw, a polyurethane diaphragm, a chamber, and a couple of mechanical valves; a compliance chamber; and an adjustable clamp. Figure 2 show the assembled system.

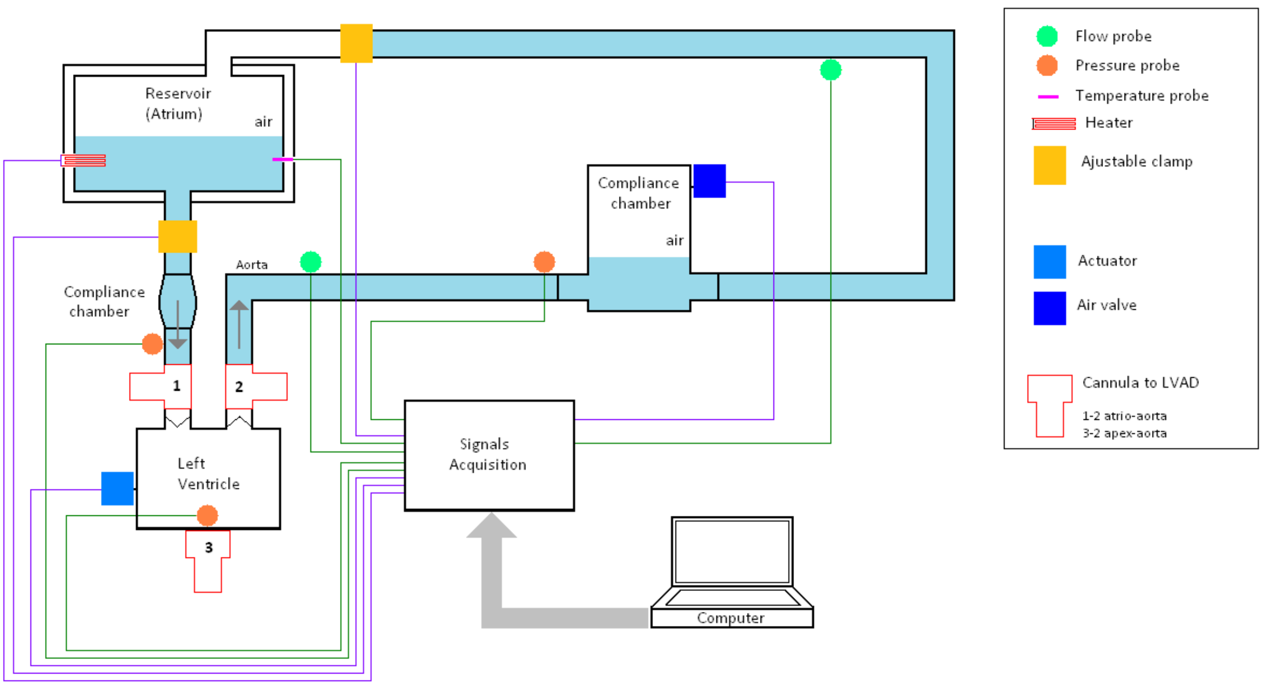


Figure 1. Diagram of cardiovascular simulator

The motor drives the roller screw in both directions, making systolic and diastolic function. At diastolic phase the diaphragm returns passively to its initial position – following the Frank-Starling law. In order to get information from end points of movement there are two Hall Effect sensors.

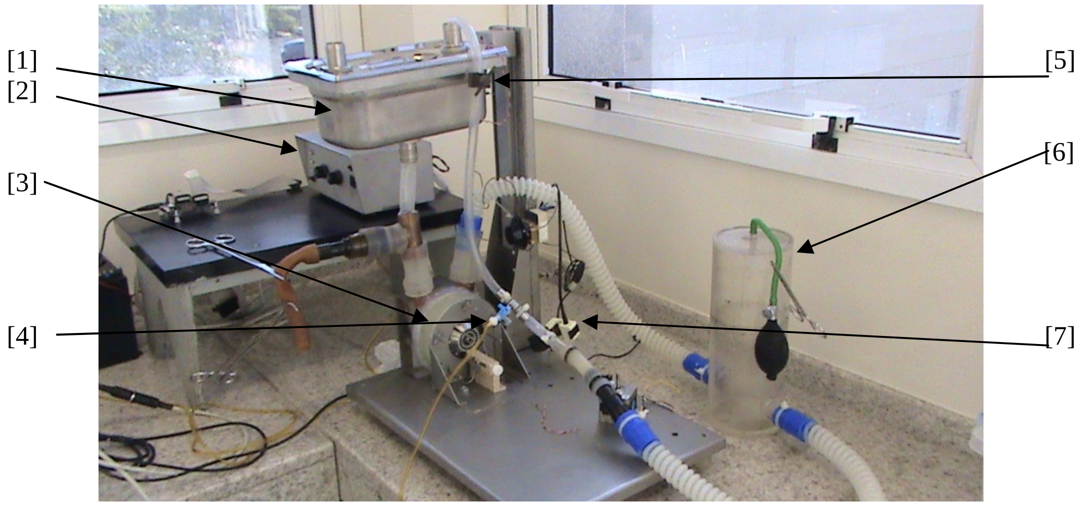


Figure 2. Assembled system: [1] Reservoir; [2] Electronic controller; [3] BLDC Pump; [4] Pressure Probe; [5] Tourniquet; [6] Compliance chamber; [7] Flow Probe.

The motor speed and the systolic volume can be adjusted by the computer program. The adjustment of desired systolic volume is automatically made by computer which set the hall sensor position. Therefore, the elastance of heart can be changed.

In order to evaluate the simulator behavior, the pressure system (Truewave, Edwards, USA) was calibrated according to a reference from a previously calibrated multi-parametric monitor (Dixtal, DX2020, BRA). The pulsatile and average flow signal was measured by an ultrasonic flowmeter (Transonic,HT110,USA) .

The acquired data was concentrated at the connector block (BNC-2110, National Instruments,TX). Therefore, four channels were acquired: intra-ventricular pressure, systemic pressure, pulsatile and average flow on systemic circulation;

The signals were sent to a board installed in the computer (PCI-6036, National Instruments,TX) and a software developed in LabVIEW® (National Instruments,TX) environment, called virtual instrument, can show the signals and save them in files, Fig. 3.

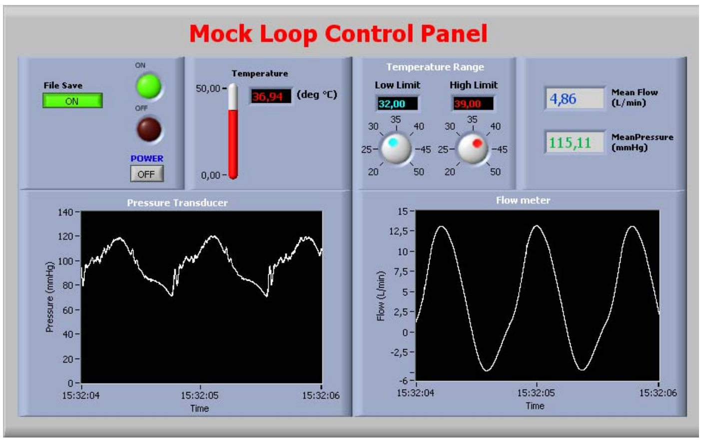


Figure 3. Cardiovascular Simulator Panel (LabVIEW®)

After initial calibration, the next step was, while keeping the ventricle in 90 bpm (beats per minute), to adjust the “aortic” pressure at 120mmHg by 80mmHg. Those pressures were obtained by adjusting the clamp and the air volume inside the compliance chamber.

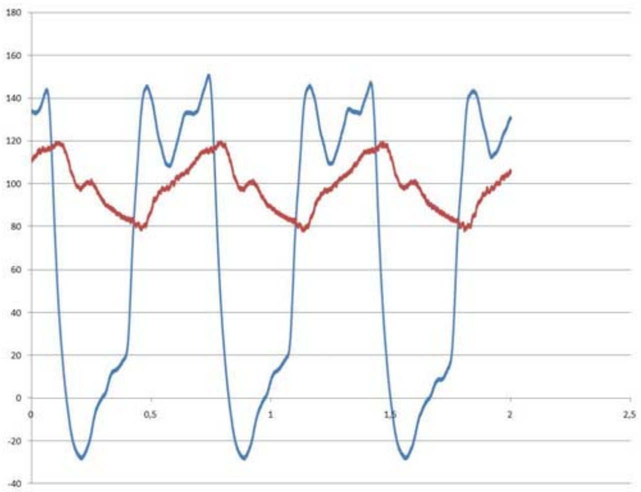
3. RESULTS AND CONCLUSIONS

The experimental results are shown in the Fig. 4. Figure 4 (a) shows intravascular pressure (blue signal) and the aortic pressure (red signal). Figure 4 (b) shows the pulsatile flow (green) and the average flow (blue) in aorta.

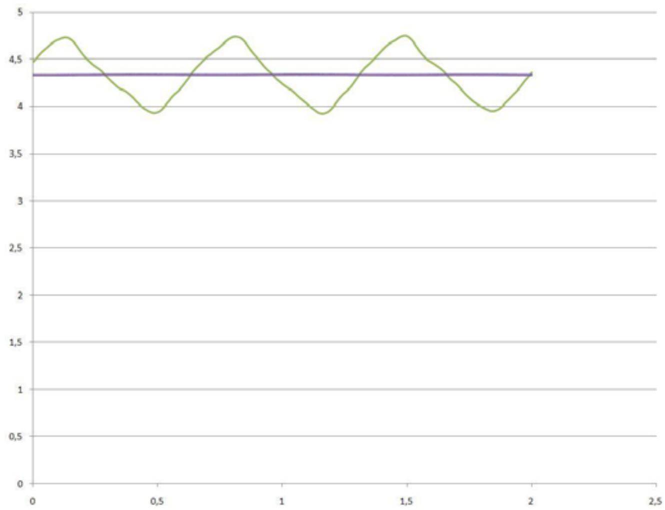
From Figure 4(a) is possible to conclude that the peaks in intraventricular pressure occurs because the ventricular chamber is rigid and do not have a dumper effect. The instants at the valves open and closes are observed from the same figure. When the ventricular chamber pressure increases over the aortic pressure, the aortic valve opens and the aortic pressure starts to increase. However, in the next instant, the pressure inside the chamber starts to decrease, so the aortic valve closes and the aortic pressure starts to decrease and its slope depends of aortic compliance value. At this moment the “mitral valve” opens allowing the fluid flows from reservoir (atria) to ventricular chamber, and another loop occurs.

From the systemic flow pattern, Figure 4 (b), is possible to conclude that the obtained pulsatile flow was different from physiological pattern. It happened because the ventricular pre-load (pressure in the ventricle inlet) was so high that the system worked continuously and no waiting time for passive chamber filling was detected.

The heart rate was set in 90bpm (beats per minute) in order to keep the mean flow rate in 4.0lpm (liters per minute).



(a)



(b)

Figure 4. Obtained results for pressure (a) and flow (b).

4. ACKNOWLEDGEMENTS

The authors would like to express their gratitude to FAPESP – Fundação de Amparo à Pesquisa do Estado de São Paulo by collaboration to realize this research.

5. REFERENCES

Bustamante, John, Barros, J. F., Roldán, A., García, S. 2004. “Modelo físico Del sistema cardiovascular – DYNASIM”, Rev. Col. Cardiol., Vol. 11, pp. 150-156.

Felipini, C.L., Andrade, A., Lucchi, J.C., Fonseca, J., Nicolosi, D.E.C, 2008, “An Electro-fluid-dynamic Simulator for the Cardiovascular System”, Artificial Organs, Vol. 32, pp. 349-354.

Legendre, D., Fonseca, J. ; Andrade, A., Biscegli, J., Manrique, R., Guerrino, D., Prakasan, A, Ortiz, J., Lucchi, J. 2008, “Mock Circulatory System for the Evaluation of Left Ventricular Assist Devices, Endoluminal Prostheses, and Vascular Diseases”, Artificial Organs, Vol. 32, p. 461-467.

Lucchi, Júlio Cesar, 1999, “Simulação Elétrica Aplicada à Investigação Hemodinâmica Da Assistência Ventricular”, PhD Thesis in Electronic and Computer Engineering, Technological Institute of Aeronautics ITA, Sao Jose dos Campos, Brazil, 154p.

PHYSBE, 1966. “A Physiological simulation benchmark experiment using SIMULINK”. 22 May 2009, <http://www.mathworks.com/products/demos/simulink/physbe/ >.

5. RESPONSIBILITY NOTICE

The authors are the only responsible for the printed material included in this paper.