# An Axial Flow Left Ventricular Assist Device: Modelling and Control

GUILLERMO CAMPOMAR [1], MATIAS SILVESTRINI [1], CARLOS D'ATTELLIS [1,2]
[1] Grupo de Estudio de Sistemas de Control – G.E.Si.C.
Universidad Tecnológica Nacional, Facultad Regional San Nicolás.
[2] Universidad Favaloro, Facultad de Ingeniería
Av. Belgrano 1723, 1093 - Buenos Aires
ARGENTINA

*Abstract:* - In the last ten years, axial flow left ventricular assist devices (LVAD) of small size were developed. This paper proposes a model for a new LVAD that reproduces the characteristics of the signals recorded in ex-vivo experiences in calves, through the interaction with a cardiovascular system model. Given the validity of the model, an approach to control is presented in order to adapt the speed of the pump to the physiological demand, keeping a pulsatility index of aortic pressure and avoiding risky situations to the patient.

*Key-Words:* - Axial Flow LVAD, Bilinear Model, Simulation, Control.

## 1 Introduction

The cardiovascular disease is one of the main health problems with epidemic characteristics; as such, it has an important social and economics impact. In severe cases, when pharmacological treatments are unsuitable to hold an adequate blood circulation, to transplant the heart is the method generally accepted. But in general, the demand exceeds the number of donors, and as consequence, patients that need a heart have to wait a long term to receive one. In these cases a ventricular assist device is considered a bridge to cardiac transplant [1-3].

The improvement in design and reliability of circulatory assist devices have made possible to use them in large periods of time. This fact allows to glimpse the use of mechanical assistance as an alternative to the heart transplant in chronic cases [4-6].

Several types of blood pumps were designed: pulsatile or continuous flow, centrifugal or axial flow. At the end of the eighties, trying to overcome the drawbacks existing in pulsatile systems, a generation of axial-flow blood pumps were developed [5,6]. Following this approach a new pump of axial-flow was designed (IMPSA/FF, see [7] for details). The results about modelization and control presented in this paper correspond to it.

Despite clinical trials have been done with this type of devices, the modelling and control problems are far to be solved.

The structure of this paper is as follows. In Section 2 a brief description of the IMPSA/FF pump is presented, including ex-vivo tests.

The mathematical model of the pump is developed in Section 3. In Section 4 the coincidence between the results obtained with the mathematical model and ex-vivo test are shown. The analysis of the interaction of the pump with a cardiovascular model is described in Section 5. An approach to the pump control is introduced in Section 6. Finally, conclusions are enumerated in Section 7.

## 2 Axial-Flow Blood Pump

The pump is made up by a housing divided in two compartments for both titanium rotors with magnets of rare earth. Above them the magnetic field of the statoric windings acts to generate the movement [7].

The output rotor rotate in opposite way to the input rotor of the pump, and by design, both must keep a constant relation of speed:

$$\frac{n_1}{n_2} = k_v \quad (1) .$$

The pump determines a flow of 5lt/min with a pressure difference of 100 mmHg in static conditions, for input and output rotors speed of $n_1 = 8400$ rpm and $n_2 = 3900$ rpm respectively [7].

### 2.1 Ex-vivo evaluation

To evaluate the interaction between the pump an a normal heart, during 2002 ex-vivo tests were made in the Favaloro Foundation (Buenos Aires, Argentina); the electro-mechanical device was externally connected to the body of a calf [7]. (See Fig. 1)

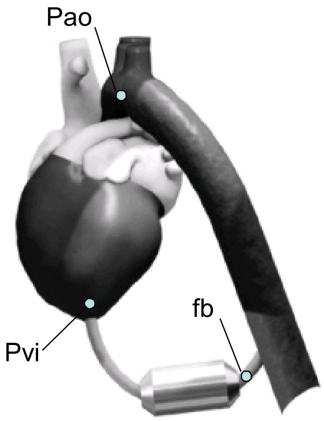
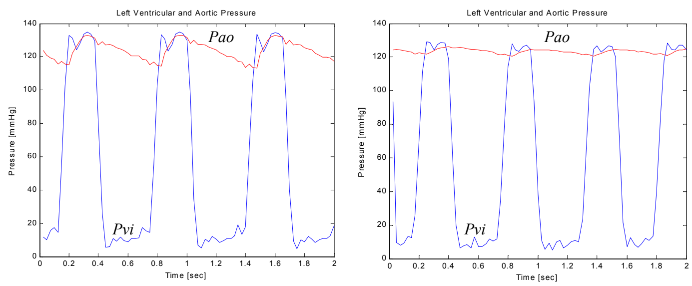



Fig. 1. $P_{ao}(t)$ : aortic pressure; $P_{Vi}(t)$ : left ventricular pressure; $f_b(t)$ : pump flow.

The signals recorded in the ex-vivo test are shown in Fig. 2; they correspond to two different rotor speeds.



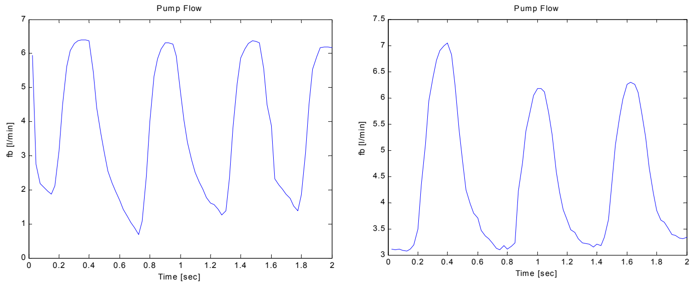




Fig. 2. $P_{ao}(t)$, $P_{Vi}(t)$ and $f_b(t)$ recorded from the ex-vivo tests, corresponding to $n_1 = 7200$ rpm (left side) and $n_1 = 8100$ rpm (right side).

## 3 The mathematical model

Several models were developed in order to describe the behavior of LVAD [8,9]. A new model is proposed in this paper for the axial-flow pump analyzed:

$$\frac{df_b(t)}{dt} = k_1.H(t) + k_2.f_b(t).\omega_1(t) + k_3.\omega_1(t) \qquad (2),$$

where the inputs are $H(t) = P_{Vi}(t) - P_{ao}(t)$, and $\omega_1(t) = \frac{2.\pi.n_1(t)}{60}$.

The parameters $k_1, k_2$ and $k_3$ could be considered as constants in the range $6000 < n_1 < 8000$ rpm.

The bilinear differential equation (2), considering $\omega_1(t)$ as a constant, becomes linear equation. This linear equation can be expressed in discrete-time, and the parameters estimated using the recursive least squares algorithm with constant forgetting factor [10,11].

## 4 Validation of the model

The comparison between the pump flow recorded during the ex-vivo tests and the output of the mathematical model are shown in Fig. 3.

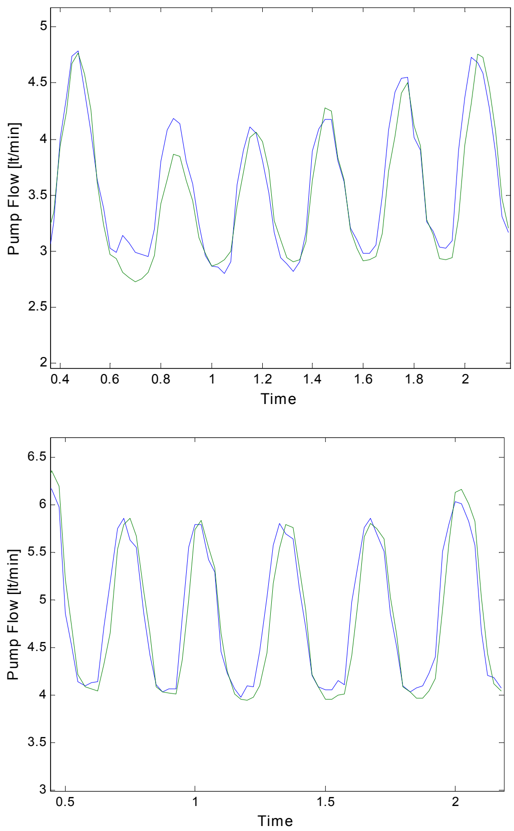



Fig. 3. Pump flows corresponding to $n_1 = 6700\,\text{rpm}$ (top) and $n_1 = 7900\,\text{rpm}$ (bottom)

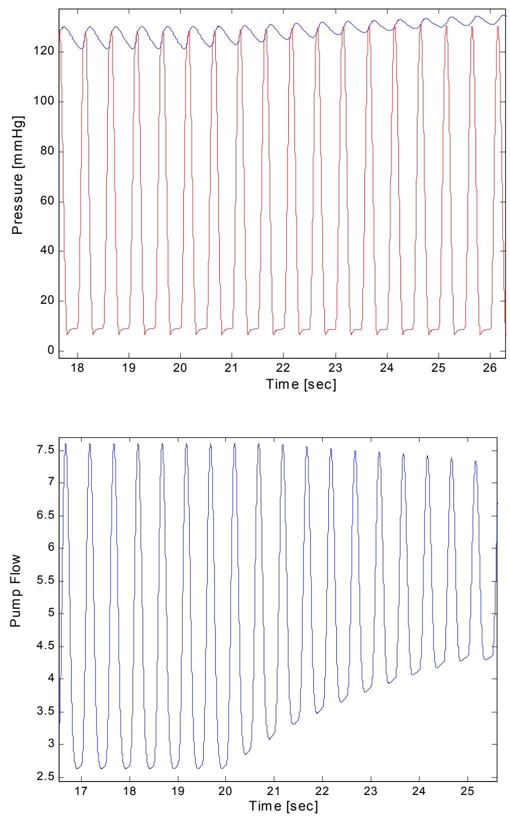



Fig. 5. $P_{ao}(t)$, $P_{Vi}(t)$ and $f_b(t)$ corresponding to a linear variation of $n_1$ between 6000 and 8000 rpm

## 5 Interaction with a cardiovascular system

In order to evaluate the interaction between the proposed pump model and the cardiovascular model introduced in [12], the system depicted in Fig. 4 was simulated.

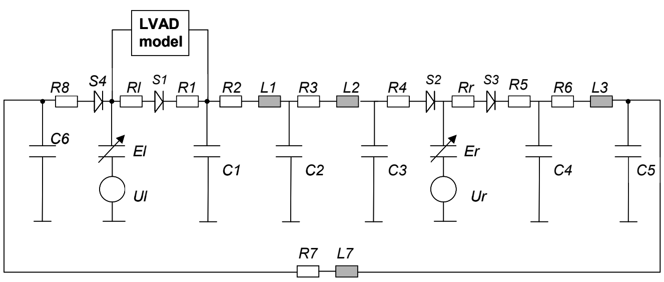



Fig. 4. LVAD and cardiovascular system models

The parameters of the electric circuit have been adjusted by computer simulations, to match the results to those obtained in ex-vivo tests. The results are shown in Fig.5 .

## 6 An approach to the control problem

The objective of the controller is to set the speed according to the physiological conditions, avoiding risky situations for the patient. Several control strategies were recently proposed for axial flow pumps [8,13,14].

A modification of the index defined in [8] is presented in this paper (see Fig. 6).

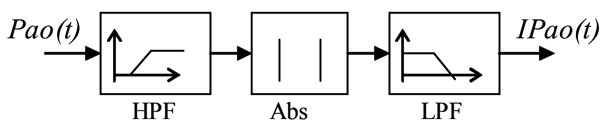



Fig. 6. Scheme for calculation *IPao(t)*. Second-order Butterworth filters, cut-off frequency of 0.5Hz.

The variation of the pulsatility index *IPao(t)* for a linear function of $n_1(t)$ is shown in Fig.7.

These results matched those obtained in ex-vivo experiences.

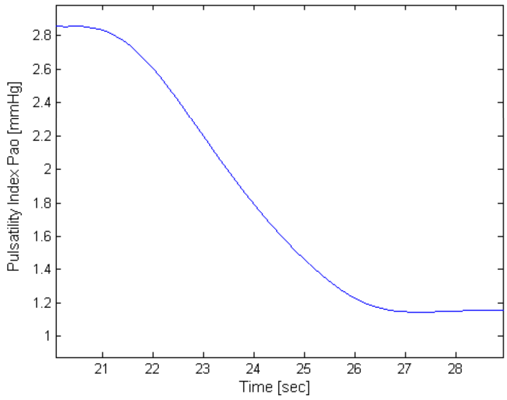



Fig. 7 Pulsatility index *IPao(t)*

The control strategy is as follows:

1) To choose the reference pulsatility index $IPao_{ref}$,
2) to calculate $.e(k) = IPao_{ref} - IPao(k)$ ,
3) to define a proportional-integral controller $\omega_1(k) = k_p.e(k) + k_i.e(k-1) + \omega_1(k-1)$ ,
4) to set the constants $k_p, k_i$ for minimizing the cost $J = \sum_{k=0} |e(k)|.k$ , with a second-order system in Fig. 8, which behavior is similar to that of the complete model.

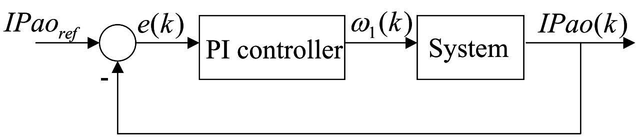



Fig. 8 Block diagram for calculation $k_p$ and $k_i$.

In Fig. 9 the results obtained from a closed-loop system simulation with the complete model are shown.

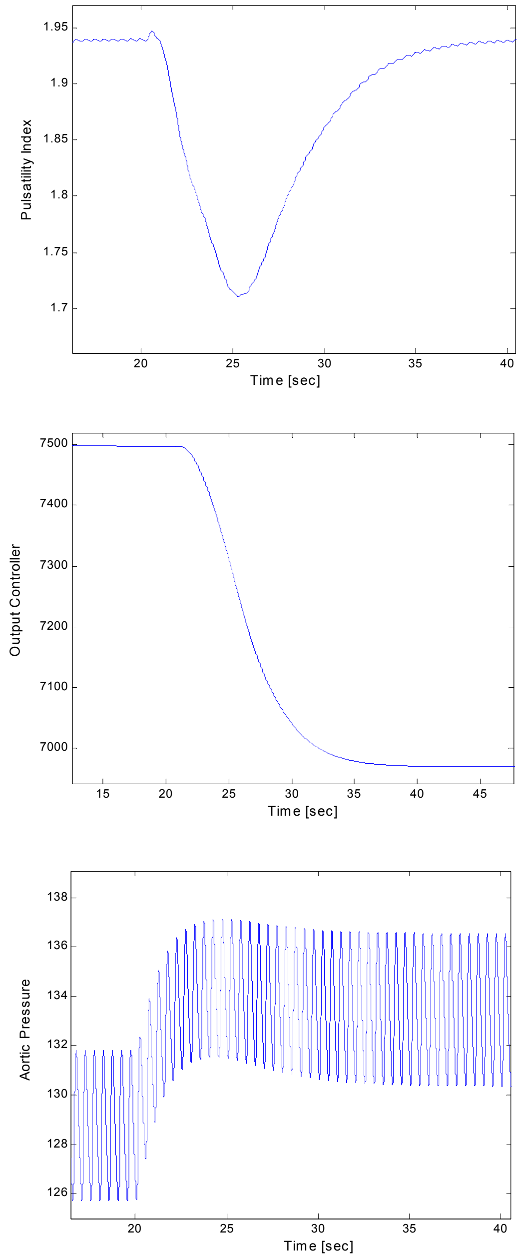



Fig. 9. *IPao(t)*, $n_1(t)$ and *Pao(t)* when R3 (Fig. 4) rises 20% in 20sec. $IPao_{ref} = 1{,}94$ .

## 7 Conclusion

A new dynamical model of LVAD was introduced. Its interaction with a simple model of the cardiovascular system could be considered as a first step to test control algorithms. The next step will be to probe these algorithms on physical models.

**Acknowledgement**

This work corresponds to the project "Mathematics in Medicine and Biology" supported by Fundación Antorchas.

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