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In vitro evaluation of multi-objective physiological control of the centrifugal blood pump

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Abstract

Left ventricular assist devices (LVADs) have been used as a bridge to transplantation or as destination therapy to treat patients with heart failure (HF). The inability of control strategy to respond automatically to changes in hemodynamic conditions can impact the patients' quality of life. The developed control system/algorithm consists of a control system that harmoniously adjusts pump speed without additional sensors, considering the patient's clinical condition and his physical activity. The control system consists of three layers: (a) Actuator speed control; (b) LVAD flow control (FwC); and (c) Fuzzy control system (FzC), with the input variables: heart rate (HR), mean arterial pressure (MAP), minimum pump flow, level of physical activity (data from patient), and clinical condition (data from physician, INTERMACS profile). FzC output is the set point for the second LVAD control schemer (FwC) which in turn adjusts the speed. Pump flow, MAP, and HR are estimated from actuator drive parameters (speed and power). Evaluation of control was performed using a centrifugal blood pump in a hybrid cardiovascular simulator, where the left heart function is the mechanical model and right heart function is the computational model. The control system was able to maintain MAP and cardiac output in the physiological level, even under variation of EF. Apart from this, also the rotational pump speed is adjusted following the simulated clinical condition. No backflow from the aorta in the ventricle occurred through LVAD during tests. The control algorithm results were considered satisfactory for simulations, but it still should be confirmed during in vivo tests.

KEYWORDS

fuzzy control system, left ventricle assist device, multi-objective control, physiological control

1 | INTRODUCTION

In recent years, left ventricular assist devices (LVADs) have been successfully used as a bridge to heart transplantation or as destination therapy (DT) for the treatment of congestive heart failure (CHF). Continuous flow LVADs are smaller, more reliable, and less complex than the first generation LVADs (pulsatile).[1–4]

The development of control systems which are able to adapt according to the body's metabolic demands is called physiological control. Research in this field has already been done since the early 1990s.[5–8] The purpose of LVAD control

is to provide an adequate cardiac output while sustaining appropriate pressure perfusion, thus researchers propose new methods for physiological control. In order to achieve this, research and development have been conducted to estimate pump parameters such as LVAD flow and differential pressure. Because of thrombus formation and system reliability reduction, additional sensors are not desirable.[9]

Giridharan et al[45] show a control strategy that keeps the mean pressure constant between pulmonary vein and aorta. Also, its results showed, through computational simulations that control strategy was able to keep the total flow rate under physiologically normal limits. Fu and Xu[10] propose a sensorless fuzzy controller utilizing pump motor speed and current as input variables. Results indicated that the proposed control can achieve the required pump flow, but pump flow and heart rate proportionality has been a constraint.[10] Others works propose a fuzzy system as a controller to adjust LVAD speed with preload and afterload response.[11–14] Volkron et al designed a control algorithm to obtain a linear relationship between heart rate (HR) and pump speed.[15] Gao et al show a numeric simulation of an anti-suction control based on blood assistant index of ventricle unloading.[16] Cavalheiro et al propose a supervisory control system based on Bayesian network and Petri nets to generate the algorithm control.[17] Fuzzy logic control systems were developed without the use of additional sensors. Control feedback parameters were obtained from the power and speed motor parameters (intrinsic parameters) of the blood pump. Computer simulations, in vitro and in vivo tests point to the effectiveness of the fuzzy control system.[10,11,18–20] Controllers aimed at physiological performance, but the evolution of the patient's clinical condition was not part of their rules.

More recently, researchers show new proposals for physiological control using flow, pressure, afterload, preload, pulsatility, or aortic valve status (opening), evaluated in vitro tests and simulations. Most physiological controllers use sensors for decision making. Those who do not use sensors report lower accuracy, according to the literature.[21–27] The mentioned strategies have been verified in numerical or computational simulations and in vitro tests. Results have been demonstrated that they were able to generate adequate blood flow and pressure. Therefore, it is important to design a control system capable to perform speed adjustments automatically.[6,22]

In order to achieve this, the aim of this article is to develop a multi-objective harmonious physiological control (MOPC) system, based on fuzzy logic, to adjust LVAD speed for a harmonious interaction with the physiological regulatory system,[18,28,29] during in vitro tests. Heart rate and mean arterial pressure are the main variables of the physiological system of blood pressure regulation via baroreceptor and sympathetic system, which are used as feedback for the control scheme. In the MOPC proposed controller, these variables are accessible without the use of additional sensors using intrinsic pump parameters (a design requirement).

2 | MATERIALS AND METHODS

2.1 | Centrifugal blood pump

The apical aortic blood pump (AABP) is a centrifugal blood pump with ceramic bearings, which has been developed in our laboratories at the Institute Dante Pazzanese of Cardiology, Brazil, to be used as a totally implantable LVAD for circulatory assistance in the long term. This LVAD was designed to be attached directly to the left ventricular apex and connected to the aorta through an outlet cannula.[30] Centrifugal blood pumps allow: operation at lower speed motor (approximately 2000 rpm) than axial pumps (approximately 10 000 rpm); lower rates of hemolysis, that is, less damage to blood elements; anatomically compatible dimensions and reach the estimated service life of more than 2 years of mechanical support, as bridge to transplant. Measured hemolysis mean normalized index in the AABP was 0.009 g/100 L (±0.002 g/100 L).[30–32]

Figure 1 shows the AABP's hydrodynamic performance. Figure 2 shows the AABP's electrical performance. AABP was evaluated in in vitro tests, showing satisfactory results for circulatory assistance.[30] AABP is composed of continuous-flow centrifugal blood pump, brushless direct current motor (BLDC—pump actuator) that electromagnetically couples with the pump rotor, electronic controller to drive BLDC motor, and a battery system or electrical source (10 W, approximately). A future step of this project includes multiple acute and chronic tests in vivo.

2.2 | Control strategy

The multi-objective physiological control (MOPC) of LVAD speed operates harmoniously with the physiological control system. Blood pressure regulation is performed by several physiological control systems: baroreceptors, chemoreceptors, renal system, among others.[28] Our proposal is to perform a cascade control with the physiological system. Cascading Proportional-Integral (PI) controllers are used to tune system performance to operate harmoniously with the physiological system.[33] First response is due to the physiological system, then the controller responds to this demand, thus allowing the overall response of the control system to be physiological as possible.

Fuzzy control system (FzC), on the third layer, adjusts LVAD flow, second layer flow control (FwC), in order to adjust LVAD speed, first layer. Figure 3 shows the block diagram of the MOPC technique.

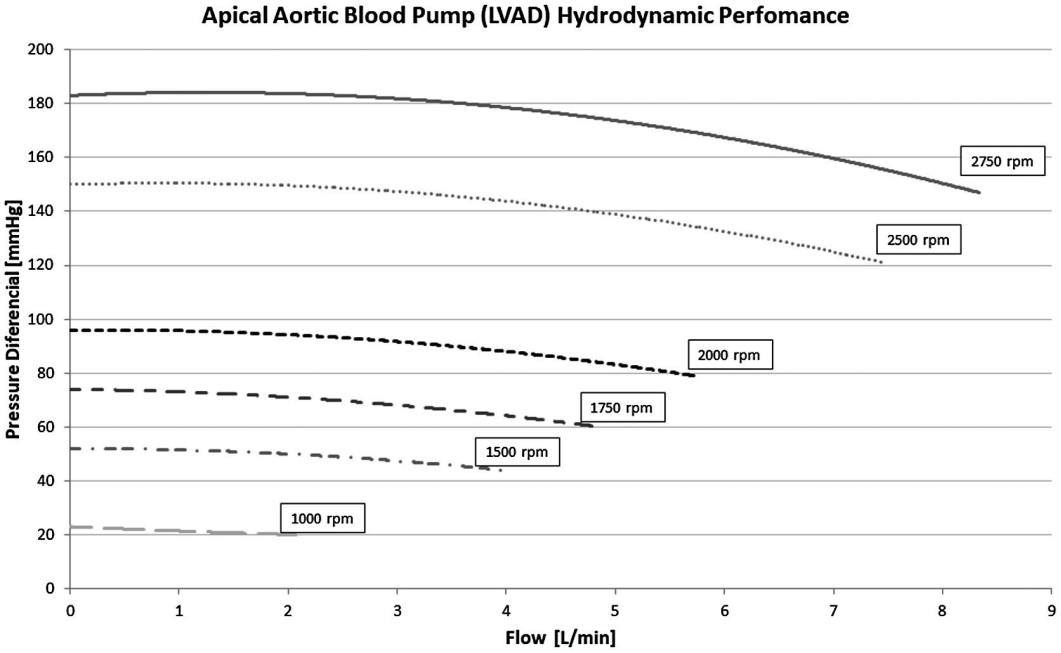

FIGURE 1 Apical aortic blood pump—hydrodynamic performance

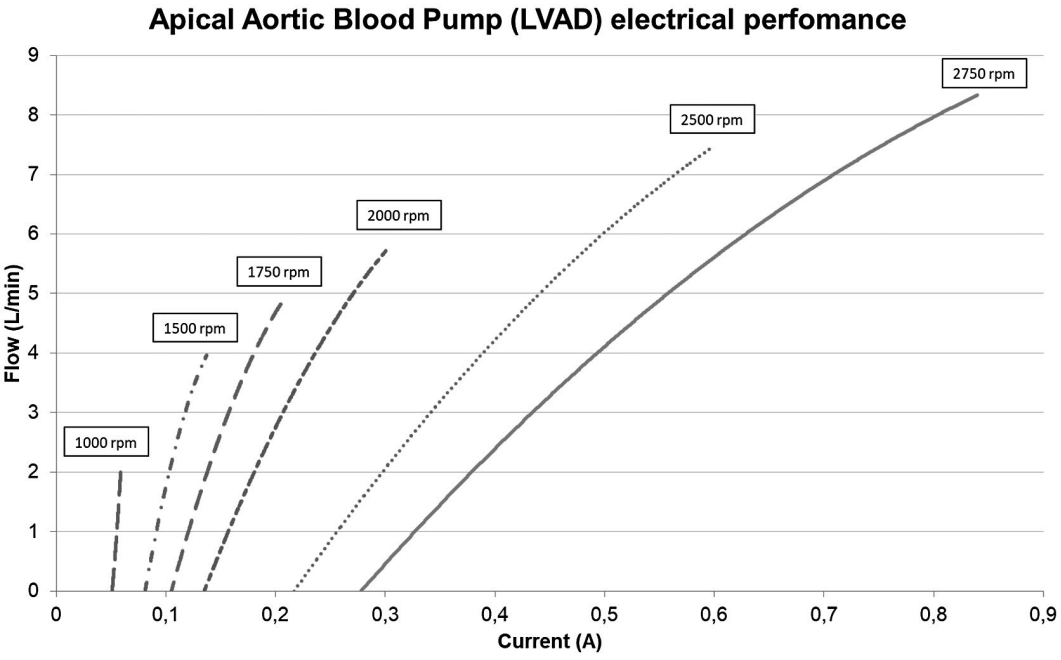

FIGURE 2 Apical aortic blood pump—electric performance

PI controller speed (first layer) adjusts electric current to maintain a set point for rotational speed, studied in preliminary works,[33–36] Figure 3. In these studies, fuzzy implementation and in vitro test was not made, therefore, the main difference and contribution to this work. Estimator block (second layer) used BLDC electric current and rotational speed for estimation of flow and differential pressure, based on look-up table 2D technique. The look-up table is determined by hydrodynamics test for range of operation. There is a correspondence between the rotation and electric current with the flow and differential pressure. This correlation is mapped via a three-dimensional surface. This technique was patented under BR1020160068363 number, in Brazil, including calibration system.[37] Its deviations are 0.23 ± 0.07 L/min for mean flow estimator; 3 ± 1 mm Hg for mean arterial pressure (MAP) estimator (with confidence interval equal to 95%). These deviations were obtained by estimation values

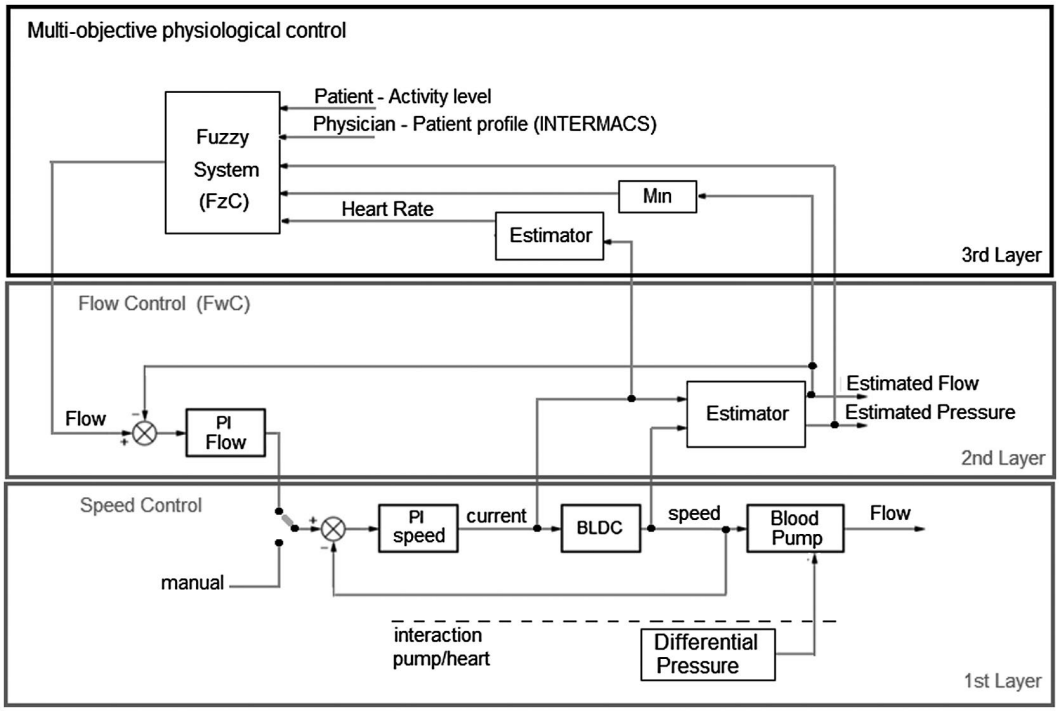

FIGURE 3 Multi-objective physiological control (MOPC) block diagram

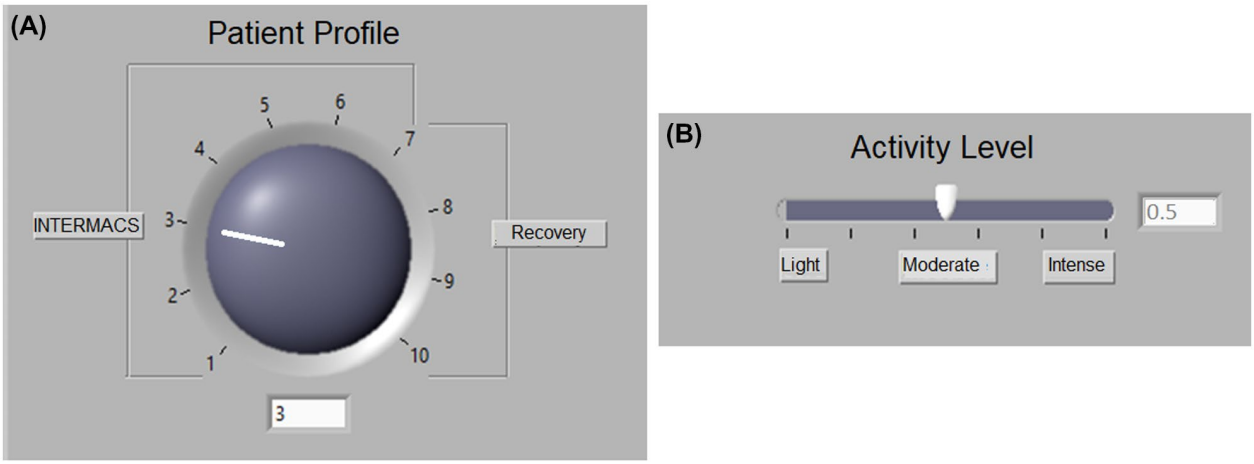

FIGURE 4 Physician and patient interface with MOPC control. A, Physician panel: patient profile. B, Patient panel: activity level [Color figure can be viewed at wileyonlinelibrary.com]

compared with ultrasonic values. MAP is estimated with differential pressure and offset parameter (experimental adjust during implant) added up. This adjust is made by comparison between patient MAP and mean differential pressure. In a future clinical trial, this offset parameter can be adjusted by blood pressure measurement, MAP during medical monitoring.

The PI flow control in the second layer adjusts pump speed in order to obtain a certain LVAD flow. The feedback signal in this layer is the estimated flow mean. PI gains were adjusted for step response (settling time), according to the harmonious response.

Heart rate (HR) estimator used fundamental frequency of the electrical current, third layer; its deviation is 1 ± 1 bpm for mean HR estimator (with confidence interval equal to 95%) for in vitro tests. To MOPC in vitro evaluation, the estimator block was adjusted 45% for hematocrit.

The input variables of the FzC are estimated minimum flow mean, estimated MAP, HR mean, patient profile according to INTERMACS profile (informed by physician) and activity physical level (informed by patient).[38] FzC has output variable: the increase or decrease of blood flow reference value. Figure 4 shows the interface MOPC control to patient

and physician. Physician sets patient profile variable following patient profile INTERMACS[38] and patient recovery, 1-7 and 7-10, respectively, Figure 4A. The rules are based on the conditions described by INTERMACS. In other words, Profile 1 is inotropic dependent progressive (very poor heart function), it implies that the LVAD will have more action. Profile 7 is NYHA III-IV (cardiac function poorly affected), it implies that the LVAD will have minor action. This characteristic is represented as set fuzzy system adjustment.

Patient sets physical activity level through the cursor, Figure 4B. Otherwise, patient adjusts physical activity level for following rest (light), moderate (walking), or intense activity (going up the stairs). MAP estimated, minimum flow estimated, and heart rate estimated are input variables (mean) in the controller.

HR and MAP are the main variables of the physiological system of blood pressure regulation via baroreceptor and sympathetic system. In the MOPC, these variables are accessible without the use of sensors (a design requirement). Fuzzy logic is particularly suited for systems that are complex and show parameter uncertainty.

2.3 | Controller design

The fuzzy control system has characteristics appropriated for assistance circulatory applications.[6,11,13,14,39,40] Figure 5 shows a membership function configuration for each variable of FzC. Membership function is adequately sets for harmonious and safe operation. Fuzzy system rules were defined according to the general circulation regulation system and were grouped according to their input variables and their interaction with the specialized systems physiological control. Circulation regulation by the nervous system, baroreceptors, backflow through the pump, pulsatility, and hypothalamic response were considered for rules constructions. Fuzzy system rules are self-validating. In case of inconsistency between inputs and hemodynamic variables, the controller assumes a safe action, that is, if the patient forgot to switch the exercise condition back to rest after they had stopped, an important contribution in this work. Optimization of the rules followed heuristic methods,[41] according to information from the specialist (physician), physiologist, and bioengineer. All information was based on knowledge of heart failure and VAD working. Table 1 summarizes fuzzy rules used in FzC.

MOPC was designed to adjust LVAD speed automatically, reference previous or physician adjusting. However, MOPC has 2 modes of control: manual or automatic. Manual mode is recommended during the implant, where hemodynamic variables are unstable and speed is adjusted by the clinician.

MOPC was implemented in Labview (2010, National Instruments, Austin, TX, USA) and in Matlab (R2010b, Mathworks, Natick, MA, USA). The data acquisition system (USB-6009 and USB-6212, National Instruments) was used for BLDC driver and for signal acquisition.

2.4 | In vitro tests—Hybrid cardiovascular simulator (HCS)

Hybrid cardiovascular simulator (HCS) was used to evaluate the MOPC under heart failure condition, Figure 6.[42] This tool allows the physical connection of AABP under evaluation. In addition, HCS allows that cardiovascular parameters can be changed in order to simulate specific heart disease: ejection fraction (EF) (15%, 20%, 30%, 35% and 40%), heart rate (HR) (50-110 bpm, range of the baroreflex system) and systemic vascular resistance (SVR) (0.9, 1.2, and 1.6 mm Hg/mL). The baroreflex system is modeled according with literature.[42] Parameters of heart failure condition were obtained from unpublished data of health service at Institute Dante Pazzanese of Cardiology.

Briefly, the baroreflex system is a complex system and key controller of MAP. Its action regulates HR and SVR to maintain MAP in physiological conditions (MAP is 100 mm Hg and HR is 75 bpm).[28,42] HCS allows operating with baroreflex system enabled, this is important to evaluate MOPC response under physiological interaction. HCS does not have a Frank-Starling response. EF variation is used as the Frank-Starling response for some analyses. EF variation reveals evolution of heart failure condition, and then this variable in MOPC is set by clinician during follow-up consults.

SVR is controlled via a flow proportional valve (valve controlled by electric current), following baroreflex response. Valve hysteresis was considered in loop control. Left ventricular contractility is controlled via drive velocity on the linear motor (BLDC). The simulator allows to enable baroreflex using HR response or with HR and SVR response, this option is important for evaluation MOPC response in SVR changing conditions. Pulmonary vascular resistance is not controlled, because in HCS this section is computational. This fact hinders a complete comparison with other authors,[9,43] however, in some configurations, it is possible. Computational (numerical) and physical sections are integrated by a signal acquisition system and actuators (pressure valves).

A solution with glycerin (50%) and water was used to simulate the blood viscosity with 45% hematocrit (3.2 mPas).[44]

To evaluate MOPC strategy, three tests were performed in HCS: Test 1 to evaluate MOPC response to EF change; Test 2 to evaluate MOPC response to activity level change; and Test 3 to evaluate MOPC response to systemic vascular resistance change.

Test 1 was carried out to assess MPOC under cardiac function recovery. To accomplish this, HCS was enabled for the baroreflex system. The initial condition was configured to 30% of EF to allow simulate cardiac function improvement (EF is 40%) or worsening (EF is 15%). In simulations, patient profile (physician input variable FzC) was adjusted in agreement with EF, in other words, to 15% EF is profiled 1 corresponds to a value of 1, and 40% EF is profiled 7, corresponds

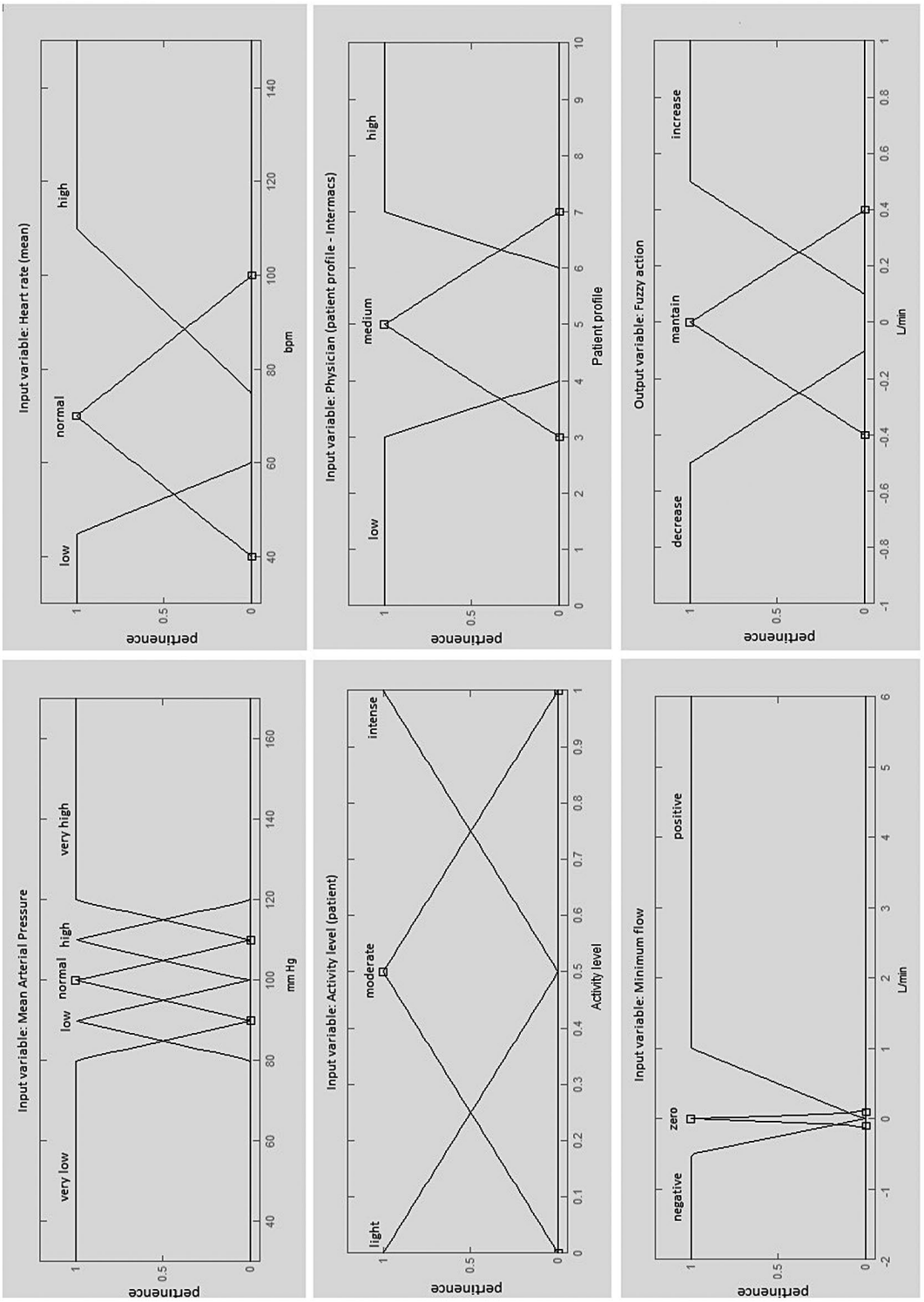

FIGURE 5 Membership function fuzzy system (FzC)

TABLE 1 Fuzzy control system rules

Rules	Description				Fuzzy action	
1	If	HR is low			then	decrease
2	If	HR is normal			then	maintain
3	If	HR is high			then	increase
4	If	MAP is very low			then	increase
5	If	MAP is low	and	AL is light	then	maintain
6	If	MAP is low	and	AL is not light	then	increase
7	If	MAP is normal	and	AL is light	then	decrease
8	If	MAP is normal	and	AL is moderate	then	maintain
9	If	MAP is normal	and	AL is intense	then	increase
10	If	MAP is high	and	AL is intense	then	maintain
11	If	MAP is high	and	AL is not intense	then	decrease
12	If	MAP is very high			then	decrease
13	If	PP is not high	and	MF is negative	then	increase
14	If	PP is high	and	MF is negative	then	maintain
15	If	PP is high	and	MF is not negative	then	decrease
16	If	PP is low	and	MF is not negative	then	increase
17	If	PP is low	and	MF is positive	then	maintain

Abbreviations: AL, patient activity level; HR, mean heart rate; LVAD, left ventricular assist device; MAP, mean arterial pressure; MF, minimal flow; PP, patient profile.

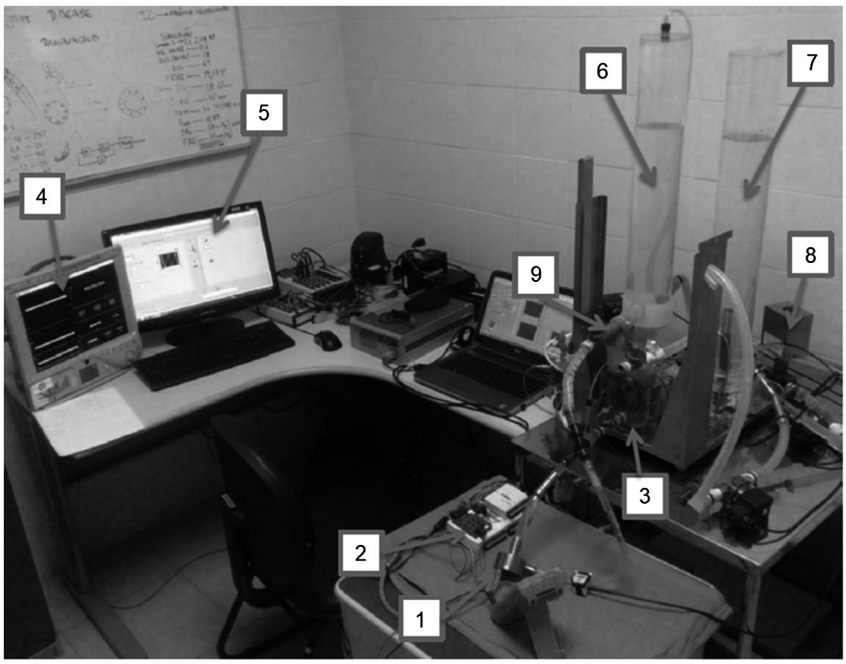

FIGURE 6 Hybrid cardiovascular simulator. 1-AABP (Apical Aortic Blood Pump) "implanted"; 2-acquisition system; 3-AABP inlet connected to apex ventricle; 4-arterial pressure monitor; 5-computational section (right heart); 6-aorta; 7-left atrium; 8-systemic vascular resistance; 9-AABP outlet connected to aorta

to a value of 7, Figure 4A. Patient's activity level was adjusted to the moderate, Figure 4B.

Test 2 was carried out to assess MOPC under activity-level change, patient input FzC, Figure 4B. This cursor was changed to light, moderate and intense levels, for 15%, 20%, and 30% of EF. Thus, patient interaction with MOPC can be observed. Activity levels were defined to provide patient comfort in daily activity, thus FzC was adjusted increase or decrease MAP in smaller portion (10 mm Hg), although it is configurable for other ranges.

Test 3 was carried out to assess MOPC under systemic vascular resistance change. SVR values for the test are: 0.9, 1.2, and 1.6 mm Hg/mL, in stepwise change. For this test baroreflex was disabled to SVR and, maintained active for heart

rate reflex. This condition is similar to vasodilators drug. EF was changed to 15%, 20%, and 30%.

3 | RESULTS

Test 1 compares the control type (MOPC and manual adjust) under cardiac function recovery. Table 2 shows MAP, cardiac output (CO), HR, SVR, LVAD flow, rotational speed, and current for each EF and control type. In Table 2, highlighted in the speed column indicates that the adjusted speed form MOPC is according to manual control (fix rotation control, defined by a physician or a specialist during in vitro test). MOPC increases LVAD flow to 4.47 L/min (when EF is 15%) and decreases to 3.13 L/min (when EF is 40%) automatically. MOPC maintains MAP between 99.5 and 104.59 mm Hg, while that MAP manual control is between 98.66 and 108.20 mm Hg for EF variations (15%-40%).

During tests, aortic valve opened above 20% EF. When the aortic valve is permanently closed (not working) CO and LVAD flow are equal. The reason for the difference between CO and LVAD flow (when EF is 15% or 20%) is according with flow estimator deviation (.23 ± .07 L/min).

SVR shows variation of the 0.33 mm Hg/mL, manual performance, and 0.26 mm Hg/mL, MOPC performance, that indicates harmonious operation between MOPC and physiological system. SVR variation occurs because baroreflex system enable.

Figure 7 shows LVAD flow performance when MOPC is active. HR is 68 bpm, CO is 5.42 L/min, MAP is 104.59 mm Hg,

TABLE 2 Recovery test results—MOPC performance

EF (%)	MAP (mm Hg)	CO (L/min)	HR (bpm)	SVR (mm Hg/mL)	Flow (L/min)	Speed (rpm)	Current (A)	Control type
15	98.66	3.98	76.4	1.24	4.15	1940	0.234	Manual
	99.50	4.19	78.0	1.30	4.47	1961	0.253	MOPC
20	99.77	4.08	75.2	1.22	4.20	1940	0.240	Manual
	99.24	4.35	78.3	1.26	4.49	1962	0.256	MOPC
30	104.07	4.74	69.0	1.09	4.17	**1940**	0.262	Manual
	101.38	4.94	73.0	1.09	4.38	**1943**	0.273	MOPC
35	106.33	5.34	65.7	0.98	4.04	1940	0.281	Manual
	102.61	5.18	71.2	1.06	3.42	1816	0.252	MOPC
40	108.20	5.77	63.1	0.91	3.85	1940	0.282	Manual
	104.59	5.42	68.2	1.04	3.13	1807	0.248	MOPC

Abbreviations: CO, cardiac output; EF, ejection fraction; HR, mean heart rate; MAP, mean arterial pressure; MOPC, multi-objective physiological control; SVR, systemic vascular resistance.

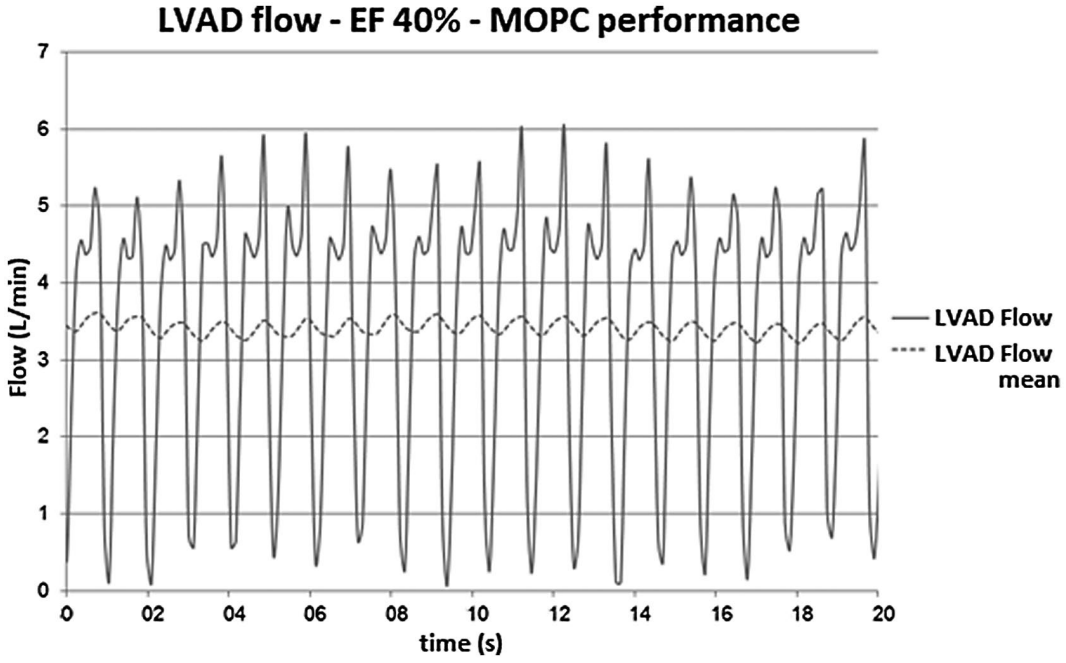

FIGURE 7 LVAD flow—EF 40%—MOPC performance

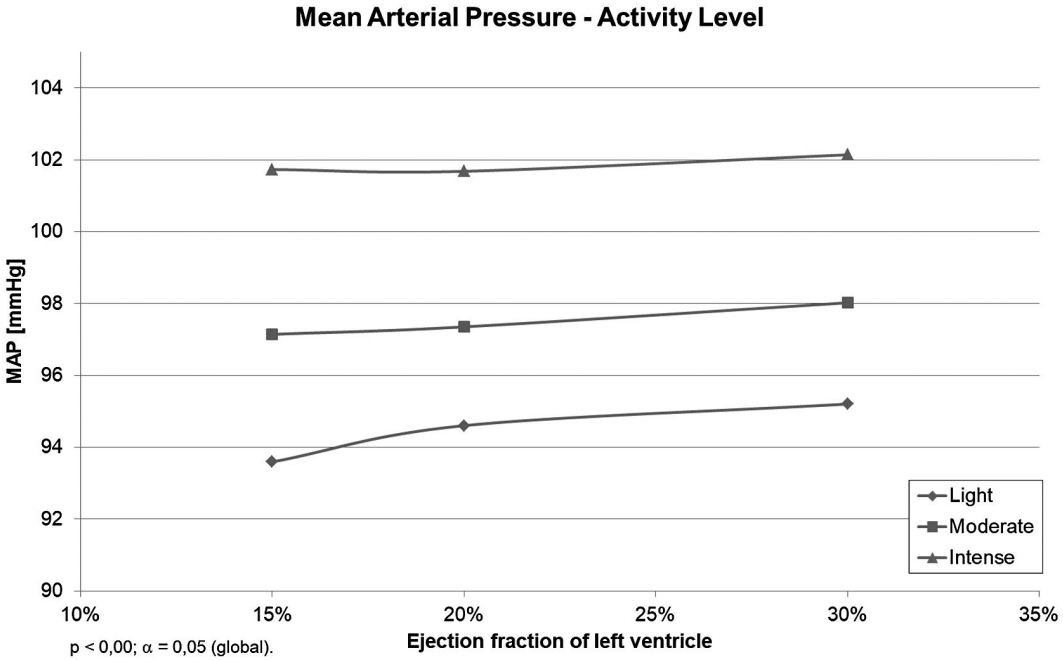

FIGURE 8 Mean arterial pressure—activity level

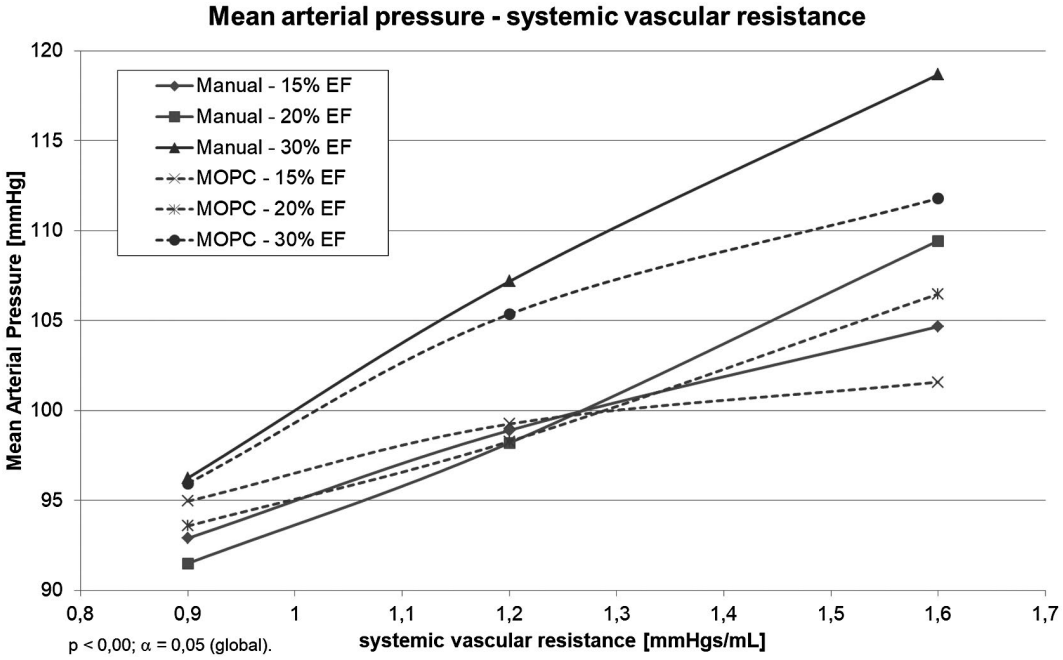

FIGURE 9 Mean arterial pressure—systemic vascular resistance

and VAD flow is 3.13 L/min. MOPC avoided LVAD backflow, even under high EF in recovering case.

Figure 8 shows MAP values by EF (15%-30%) and activity level, reported by the simulated "patient" in HCS, Test 2. This in vitro test evaluates MOPC performance regarding interaction with the patient and other physiological control systems. MOPC provides adequate flow before demand occurs through the physiological system, thus decreasing the action of regulation mechanisms and cardiac work because the patient (simulated, in vitro test) informs activity level that they are doing. This system operates as an anticipatory response in the physiological system, SVR impact. Flow and arterial pressure increases when activity level reported by the "patient" increases from moderate to intense in all EF tested (from 3.72 to 4.75 L/min, or 94 mm Hg to 102 mm Hg, in 15% EF, for example).

Figure 9 shows MOPC performance under SVR change by EF (15%-30%). When demand is generated by the physiological system locally, SVR is affected to promote regulation of circulatory pressure. SRV change compensated by MOPC is inversely proportional of variation, otherwise, LVAD speed decreases with the increase of SVR (from 2020 rpm to 1935 rpm, or from 105 to 101 mm Hg, in 15% EF, for example). This MOPC action aims to maintain MAP in a physiological pressure range.

4 | DISCUSSION

Blood pressure regulation is performed by several physiological control systems: baroreceptors, chemoreceptors, and renal system, among others. They are systems operating in cascade mode. MOPC proposal is to perform a cascade control with the physiological system. This is the harmony we seek, thinking about the patient's quality of life.

MOPC was able to set the speed automatically in agreement with manual control (equivalent physician adjusts, to in vitro test) for initial condition (30% EF), Table 2. It changes LVAD speed following the cardiac recovery function, that is, as soon as cardiac function decreases (15% EF) then MOPC automatically increases LVAD flow. On the other hand, an improvement of cardiac function (40% EF) results in MOPC automatically decreasing LVAD flow. Physiological conditions (MAP, HR, and SVR) were harmoniously kept by MOPC. This action keeps MAP in a narrow range compared to manual control, Table 2. This result corroborates with the results pointed out by Choi et al,[18] where the fuzzy system is shown as a more robust option for variations than the classic controller.

Estimator deviations were adequate for MOPC rotational speed to adjust without additional sensors usage. The estimator does not depend on opening and closing of the atrial valve or heart contraction directly, but it is affected by this variable. The look-up table relates the electromagnetic torque to the hydrodynamic torque required by the load (heart). In this sense, all the action of the heart is represented by torque. In Table 2, cardiac output is measured by an ultrasonic sensor and flow (pump) is measured by the estimator. For an ejection fraction of 15% (the aortic valve is closed), the difference between the flow values is compatible with the estimator calibration deviation. Even considering the error, MOPC was able to provide flow and pressure under physiological conditions. Giridharan and Skliar provide satisfactory results from controllers that use intrinsic parameters (estimators) for differential pressure and flow variables, in this sense, MOPC estimator is comparable with these authors.[45] Despite this, MOPC allows following clinical evolution and activity demand of patients by set rules that verify inputs and hemodynamic conditions.

MOPC avoided LVAD backflow, even under high EF in recovering case, Figure 7. Increasing the ejection fraction is an action of the Frank-Starling system. Thus, this controller action may increase the patient's exercise tolerance, as reported by other authors.[22]

MAP curves identify three levels: light, moderate, and intense, Figure 8. MOPC performs MAP as an anticipatory response in the physiological system. Level values observed in vitro test consider HCS response but can be altered (by simply changing the fuzzy system sets) according to in vivo test to indicate greater suitability (or clinical trials), which can even be performed during clinical follow-up. Patients may not feel comfortable handling LVAD adjustment, especially older patients, so in this case adjustment is disabled or performed by the physician for patient comfort. This action of MOPC can be a comfort adjustment for smaller daily variations of activity. HCS during tests does not have exercises in practice, thus there is no metabolic demand. MOPC allows adjusting MAP level but narrow range, because activity level and hemodynamic parameters are not compatible. So, anticipatory response (activity level) and safe mode of the MOPC are verified satisfactorily.

Varying baroreflex sensibility was observed in patient with heart failure and associated, nonsignificantly, with prognostic of the mortality.[46] The rules of fuzzy system mimic the baroreflex response, so varying baroreflex sensibility should not be important. This algorithm runs as self-validation with hemodynamics parameters for fault in patient and physician input.

MOPC is based in mean of flow and arterial pressure, thus systolic pressure function of left ventricular is preserved to follow physiological performance (ie, in exercise—flow up to 8-10 L/min). In this sense, this controller system is not intended to replicate the action of the Frank-Starling response.[10,12,45]

MAP variation was 93-111 mm Hg, MOPC performance, and 91-118 mm Hg, manual control, considering EF variation, Figure 9. MOPC performance allows adjusting LVAD speed automatically to changes in the hemodynamic conditions. These results corroborate the controller's robustness (safe operation) in maintaining MAP at physiological levels (narrow range), even under SVR variation (afterload evaluated independently). The slope MAP to MOPC performance is not linear, as expected, because of the set fuzzy system.

Although there was no specific test to verify ventricular suction and congestion, none of these events were observed (the events did not occur) during the test. FzC was configured with rules to consider these conditions based on physician's experience. The rules were grouped to ensure safe operation, even under conditions of patient and physician variable failure. The optimization was performed by the heuristic method.

MOPC showed satisfactory results when compared to other authors, especially in vitro performance without additional sensors.[9–11,15,43,47,48] MOPC was able to adjust LVAD speed automatically, to changes in the hemodynamic conditions. These results are limited to HCS simulated condition, that is, it mimics a heart failure patient. A fault test is in progress, however,

without validation (publication-peer recognition) of normal operation, fault analysis is compromised. More in vitro tests are necessary to test other conditions, that is, baroreflex totally disabled, ventricular suction and congestion, and besides acute and chronic in vivo tests. These new tests depend of adaptation of the HCS to perform the simulation in those conditions.

5 | CONCLUSIONS

Multi-objective physiologic control technique was able to adjust harmoniously LVAD rotational speed according to the physiological system without any additional sensors, in in vitro tests. The fuzzy logic control strategy is the main system of the MOPC and was able to allow automatic control.

It maintains MAP automatically; heart rate, cardiac output, and LVAD flow within physiological levels. MOPC was able to follow the simulated patient profile. MOPC avoided reverse flow through LVAD in test condition, even under high pulsatility (40% EF). MOPC was able to verify inputs of the controller (activity level and patient profile) and hemodynamic parameters for safe action. Tests points to controller's robustness (safe operation), that should be verified in more test conditions. These characteristics are the main differences from previous works.

Future studies in vitro and in vivo including disturbances are fundamental for the complete evaluation of the MOPC technique.

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CONFLICT OF INTEREST

The authors declare that they have no conflicts of interest with the contents of this article.

AUTHOR CONTRIBUTIONS

Tarcisio Leao: Concept/Drafting article
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