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INFLUENCE OF PULSATILE CATHETER PUMP SYNCHRONIZATION ON HAEMODYNAMIC VARIABLES: NUMERICAL SIMULATION

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ABSTRACT
Severe cardiovascular diseases can be treated using left ventricular assist devices (LVAD). One of the possible LVADs is the Pulsatile Catheter (PUCA) pump that consists of a hydraulically or pneumatically driven membrane pump connected to a valved catheter. In this work a numerical model of the cardiocirculatory system and of the PUCA have been developed in order to study their interaction. In the numerical simulator a pathological condition of the left ventricle has been reproduced and successively the effects of the PUCA on the haemodynamic variables applied were studied. Different functioning modes were tested by changing the ratio between the pump frequency and the heart beat rate (HR) as 1:1, 1:2 or 1:3 and by introducing a delay time between the cardiac and the PUCA cycle. The performance of the pump was evaluated in terms of cardiac output, PUCA and coronary flows and it was studied for different HR values.
Results show a good resemblance between the model and literature data and indicate that different synchronization and timing can influence the functioning of the pump. In particular, the frequency ratio and the time delay of the pump cycle can contribute to optimize the performance of the PUCA.

KEY WORDS
Cardiovascular system, Haemodynamics, Modelling, Simulation, Ventricular assist device

1. Introduction

A pathological condition of the left ventricle (i.e. severe heart failure) can be solved using a left ventricular assist device (LVAD). Assist devices can be driven in different ways to produce the better support for the patient. One of the devices for assistance is the Pulsatile Catheter (PUCA) pump. This LVAD is introduced into the left ventricular cavity through an easily accessible artery or directly through the thoracic aorta during open-chest surgery. When positioned, the pump catheter (triggered with the ECG signal) aspirates blood from the left ventricle (LV) and ejects it into the ascending aorta during the diastolic phase.

Numerical cardiovascular models can be useful tools to study the effects of PUCA pump when it is applied in the case of left ventricular failure. Numerical models can be used to analyze the interactions among the LV, the pump and the cardiovascular system from haemodynamic and/or energetic point of view.
A PUCA numerical model inserted (as LVAD) in the numerical simulator of the human cardiovascular system CARDIOSIM© is presented in this paper. The numerical device is synchronized with the ECG signal. It is possible to synchronize the frequencies of the PUCA model according to the heart rate of the patient with a synchronization ratio of 1:1, 1:2 or 1:3.
The aim of the work was to analyze the interactions between the cardiovascular system and the catheter pump in order to evaluate the trends of the left ventricular flow, the pump flow and the coronary flow in pathological and assisted conditions. The analysis was conducted changing the time delay between the pump aspiration beginning and the ventricular ejection, in order to find the best temporization of the pump.

2. Materials and Methods

2.1 The Cardiovascular Model

CARDIOSIM© is the numerical model of the cardiovascular system developed using a lumped parameter model for the different circulatory sections [1]. The software has a modular structure and is composed of seven different sections that represent: systemic arterial (pulmonary) circulation, left (right) heart, systemic (pulmonary) venous circulation and coronary circulation. In the basic configuration of the software (Fig. 1) the systemic arterial section is represented by a modified windkessel (Rcs, Cas and Ls) with a peripheral resistance (Ras). As shown in Fig. 1 the systemic venous section is reproduced by Rvs and Cvs components. The pulmonary arterial section is implemented by windkessel model (Rcp, Cap and Lp) with the peripheral resistance (Rap). The pulmonary venous section is implemented by a single compliance [2]. The behaviour of each section can be modelled using more or less complex representations [3]. The coronary circulation was implemented by a lumped parameter model [3], [4]. The behaviour of each ventricle (atrium) was described by a variable elastance model synchronized with the ECG signal [5], [6], [7]. The model reproduces the Starling's law of the heart. The left ventricular time-varying elastance *elv(t)* is a function of the left ventricular systolic elastance (*Elvs*), left ventricular diastolic elastance (*Elvd*) and left activation function *alv(t)*:

$$elv(t) = Elvd + \frac{Elvs - Elvd}{2} alv(t) \qquad (1)$$

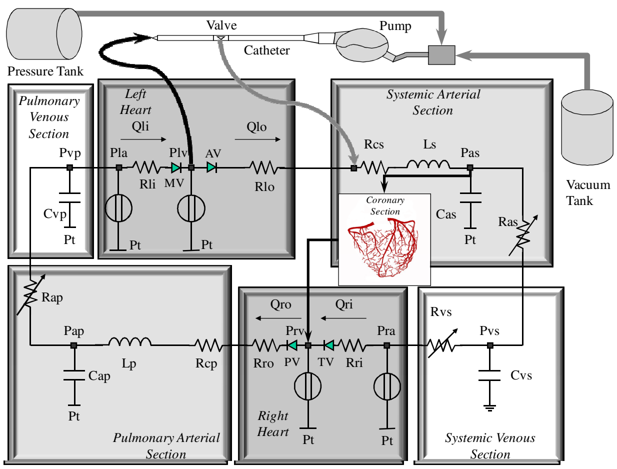

Figure 1. Electric analogue of the cardiovascular system. TV is the tricuspid valve, PV is the pulmonary valve, MV is the mitral valve and AV is the aortic valve. The pump is connected alternatively to two different source tanks. The first one is the pressure source tank and the second one is the vacuum source tank. The coronary circulation is implemented by "lamped parameters model". Table 1 reports the symbols used in figure.

In equation (1) the left activation function describes the contraction and the relaxation changes in the ventricular muscle:

$$alv(t) = \begin{cases} 1 - \cos\left(\dfrac{t}{T_T}\pi\right) & 0 \le t \le T_T \\ 1 + \cos\left(\dfrac{t - T_T}{T_{TE} - T_T}\pi\right) & T_T < t \le T_{TE} \\ 0 & T_{TE} < t \le T \end{cases} \qquad (2)$$

Where T_T is the T wave peak time in ECG signal, T_{TE} is the T wave ending time in ECG signal and T is heart period. Changing T_T and T_{TE}, (Eq.2) it is possible to determine when ventricular ejection starts and its duration. This permits to synchronize ventricular mechanics with ECG signal. The right ventricular model is similar to that for the left side, except for the values of parameters.
The time-varying atrial elastance *ea(t)* (for both atria) is also a function of the characteristic elastance during atrial systole (*Eas*), atrial diastole (*Ead*) and activation function *aa(t)*:

$$ea(t) = Ead + \frac{Eas - Ead}{2} aa(t) \qquad (3)$$

The activation function describes the contraction and the relaxation changes in the atrial muscle:

$$aa(t) = \begin{cases} 0 & 0 \le t \le T_{PB} \\ 1 - \cos\left(\dfrac{t - T_{PB}}{T_{PE} - T_{PB}} 2\pi\right) & T_{PB} < t \le T_{PE} \\ 0 & T_{PE} < t \le T \end{cases} \qquad (4)$$

Where T_{PB} is P wave beginning time in ECG signal and T_{PE} is the P wave ending time in ECG signal. In equation (4) changing T_{PE} and T_{PB}, it is possible to determine when atrial active ejection starts and its duration. This permits to synchronize the atrial mechanics with ECG signal.

2.2 The PUCA Pump Model

To simulate the functioning of the PUCA pump a lumped parameter model has been developed in CARDIOSIM© according to [8], [9]. This model permits to simulate the pressure losses in the catheter and membrane pump due to the inertia of blood, to the inflow and outflow valves and to the resistance of the catheter:

$$P_{DRIVE} - P_1 = \rho a L + 1/2 \rho v^2 (4\lambda L/d + Kv + Kc) \qquad (5)$$

P_{DRIVE} is the aspiration pressure (P_{ASP}) or the ejection pressure (P_{EJE}) of the pump. P_1 is the aortic (Pas) or left ventricular pressure (Plv), ρ is the blood density (1.060 g/cm^3), μ is the dynamic viscosity (0.0014 Pa·s) of the blood that is considered as a Newtonian fluid, d and L are respectively the diameter (d=6.8 mm) and the length (L=40 cm) of the catheter [8]. Because the flow in the catheter is assumed to be turbulent, it has been considered the flow resistance factor as $\lambda=0.0791\cdot(Re^{-1/4})$. Kv and Kc are two constants that represent the flow resistance caused by the valves and the flow resistance caused by diameter change from membrane to catheter, respectively. Kc is 1.0 during aspiration and 0.5 during ejection; Kv is 3.0 during aspiration and 3.2 during ejection [8]. *a* is the blood acceleration and *v* is the blood velocity inside the catheter. The pump has a maximum filling volume that represents the PUCA capacity and that can be set according to the specific pump performance. It is also possible to set PUCA frequency with the synchronization ratio of 1:1, 1:2 or 1:3 in relation to HR [10].
It is also possible to define the beginning and duration of both ejection and aspiration phases of the pump in accordance to the ECG of the patient. This permits to synchronize the pump functioning with the cardiac cycle and to relate the duration of the systolic phase of the heart with the aspiration of the device and the diastolic phase of the heart with the ejection of the PUCA into the aorta

(counterpulsation). Different modalities of functioning have been implemented.
In the work the attention was focused on the temporization and on the air and vacuum pressures of the device without modelling anything else with the only exception of the catheter that has been represented not as a ideal tube but with its flow resistance.

Table 1

Parameter	*Symbol*	*Unit*
	Left (right) heart	
Rli (Rlo)	Left input (output) valve resistance	[mmHg·cm^{-3}·sec]
Rri (Rro)	Right input (output) valve resistance	[mmHg·cm^{-3}·sec]
Pla (Pra)	Left (right) aerial pressure	[mmHg]
Plv (Prv)	Left (right) ventricular pressure	[mmHg]
Qli (Qlo)	Left input (output) flow	[l·min^{-1}]
Qri (Qro)	Right input (output) flow	[l·min^{-1}]
	Systemic (Pulmonary) Arterial Section	
Rcs (Rcp)	Systemic (pulmonary) characteristic resistance	[mmHg·cm^{-3}·sec]
Ls (Lp)	Systemic (pulmonary) inertance	[mmHg·cm^{-3}·sec^{2}]
Cas (Cap)	Systemic (pulmonary) arterial compliance	[cm^{3} ·mmHg^{-1}]
Ras (Rap)	Systemic (pulmonary) arterial resistance	[mmHg·cm^{-3}·sec]
Pas (Pap)	Systemic (pulmonary) arterial pressure	[mmHg]
	Systemic (Pulmonary) Venous Section	
Rvs	Systemic (Pulmonary) resistance	[mmHg·cm^{-3}·sec]
Cvs (Cvp)	Systemic (Pulmonary) venous compliance	[cm^{3} ·mmHg^{-1}]
Pvs (Pvp)	Systemic (Pulmonary) venous pressure	[mmHg]
Pt	Thoracic pressure	[mmHg]

The pump synchronization is a critical issue.
In the case of a 1:2 synchronization ratio, there are two cycles: in the first one the pump aspirates blood from the ventricle, while in the second one the pump continues to aspirate and ejects blood into the aorta.
Starting from the above considerations the results of the simulations presented in this paper were finalized to analyze: the left ventricular flow, the pump flow and the coronary flow in pathological and assisted conditions. In the first step a pathological condition was reproduced setting: the right ventricular systolic elastance (*Ervs*=0.6 mmHg·cm^{-3}), the right ventricular diastolic elastance (*Ervd*=1.028·10^{-2} mmHg·cm^{-3}), *Elvs*=0.5 mmHg·cm^{-3}, *Elvd*=0.14 mmHg·cm^{-3}, Ras= 2000 mmHg·cm^{-3}·sec and Cas = 2 cm^{3}·mmHg^{-1}. In the second step two different assistances were realized. During the first assistance the pump was synchronized by setting the ratio between the frequency of the pump and HR as 1:1. The aspiration and ejection PUCA pressures were set to P_{ASP}=-30 mmHg and P_{EJE}= 400 mmHg, respectively. The maximum pump volume was set to V_{PUCA}=90 cc. During the second assistance the PUCA was synchronized by setting the ratio between the frequency of the pump and the HR as 1:2.

Finally, in this second assistance, a time delay between the pump aspiration beginning and the ventricular ejection was introduced, in order to find the best temporization of the pump and analyzing its effects on mean coronary blood flow (CBF). In all the simulations the LVAD was ECG-triggered and the HR was changed between 70 and 250 bpm [8].

3. Results

Results show the mean values (during the cardiac cycle) of the left ventricular output flow in pathological and assisted conditions and of the flow produced by the assistance (Fig. 2). The PUCA pump was synchronized as 1:1 ratio. The pathological conditions were reproduced as described above.

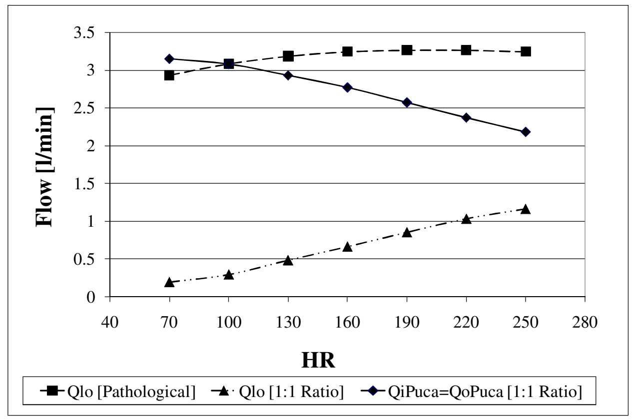

Figure 2. The flows presented were obtained for different HR values. Qlo [Pathological] is the left ventricular output flow in pathological conditions (without the assistance). Qlo [1:1 Ratio] is the left ventricular output flow when the assistance is active. QiPuca and QoPuca are the flows produced by the LVAD. The total flow in assisted condition is the sum of Qlo [1:1 Ratio] and QoPuca [1:1 Ratio].

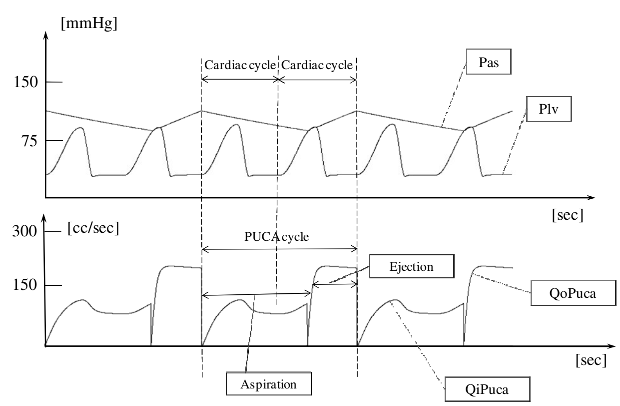

Figure 3. Instantaneous waveforms pressures and flows during PUCA assistance synchronized as 1:2 ratio. The waveforms represent the output produced by the CARDIOSIM© software.

Figure 3 shows the pressure and flow waveforms produced by the software simulator when the assistance (applied starting from the pathological condition described) was synchronized as 1:2 ratio. This condition guarantees a good efficiency of the pump even if it is synchronized with a higher HR [8]. The pressures presented, in the upper panel, are the instantaneous systemic arterial pressure (Pas) and the left ventricular pressure (Plv). In the lower panel of Figure 3 the waveforms of the input (QiPuca) and output (QoPuca) flows of the LVAD are reported. During the simulation HR was set to 100 bpm, the aspiration and ejection PUCA pressures were set to P_{ASP}=-30 mmHg and P_{EJE}= 400 mmHg, respectively.

Figure 4 shows the mean flow values produced by LVAD at different HR values and permits to compare 1:1 and 1:2 PUCA synchronization ratios.

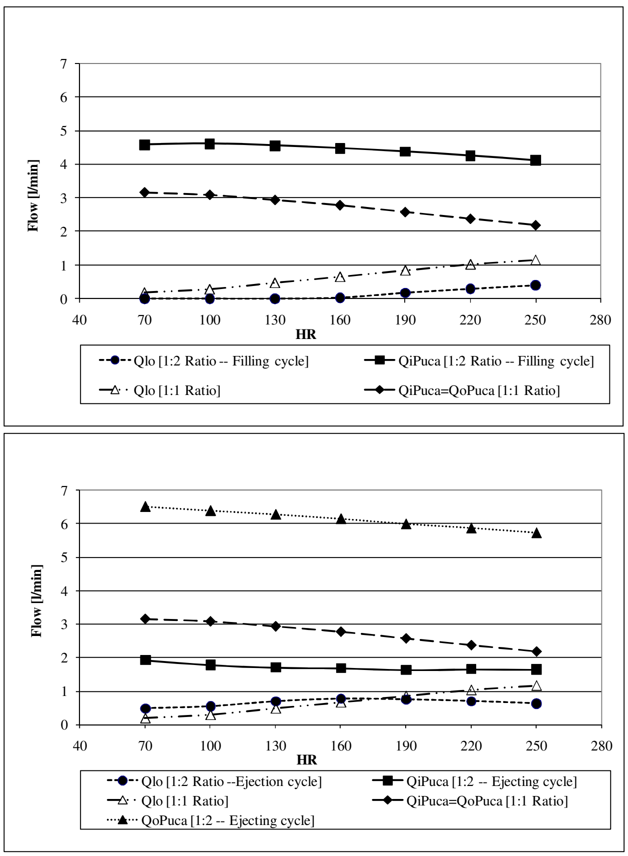

Figure 4. The upper panel shows the left ventricular output flow (Qlo) and the PUCA flow for both synchronizations of the pump [1:1 and 1:2 ratios] during the first cardiac cycle. The lower panel shows the left ventricular output flow (Qlo) and the PUCA flow for both synchronizations of the pump [1:1 and 1:2 ratios] during the second cardiac cycle. When the LVAD is synchronized with 1:1 ratio the output flow of the pump (QoPuca) is equal to the input flow (QiPuca). When the LVAD is synchronized with 1:2 ratio the QoPuca is zero during the first cardiac cycle (upper panel).

The two panels of Figure 4 reproduce the two next cardiac cycles of the 1:2 synchronization ratio (corresponding to a single PUCA cycle (Figure 3)). The upper panel shows the mean flows during the first half of the PUCA cycle while the lower panel shows the mean flows during the second half. Each half PUCA cycle corresponds to one cardiac cycle.

PUCA performance was evaluated comparing pump mean flow (during the pump cycle) in 1:1 and 1:2 synchronization ratio. Considering a PUCA cycle corresponds to consider a single cardiac cycle for the 1:1 synchronization ratio and two cardiac cycles in the case of the 1:2 synchronization ratio. Figure 5 shows the mean PUCA flows calculated in the same conditions as above.

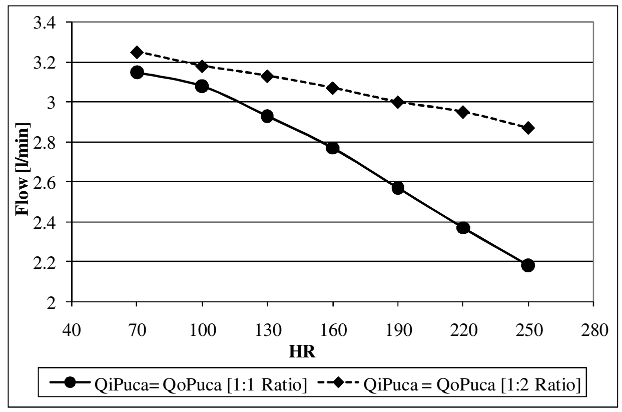

Figure 5. PUCA mean flows produced by the assistance for 1:1 and 1:2 synchronization ratio.

Figure 5 shows that the differences of the two performances are more evident for higher HR. This is in accordance with the literature data that assess that for higher HR the pump works better in 1:2 or even in 1:3 than in 1:1 synchronization ratio [8].

This part of the work was devoted to study the effects of different time delay of pump aspiration beginning on CBF. Figure 6 shows the pressures and flows waveforms produced in the same condition of Figure 3. In this case the PUCA aspiration started with a delay of 200 msec from the beginning of left ventricular ejection.

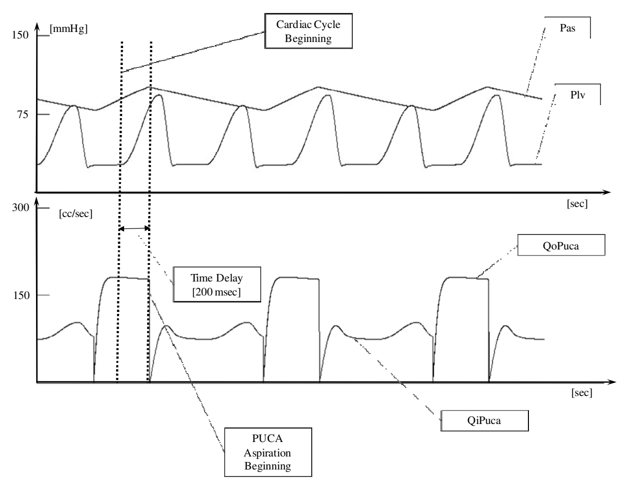

Figure 6. Instantaneous pressure and flow waveforms during PUCA assistance synchronized as 1:2 ratio. The PUCA aspiration starts with a delay of 200 msec from the left ventricular ejection beginning.

Finally Figure 7 shows the mean CBF during the first and the second half of the PUCA cycle in 1:2 synchronization ratio for different time delays.

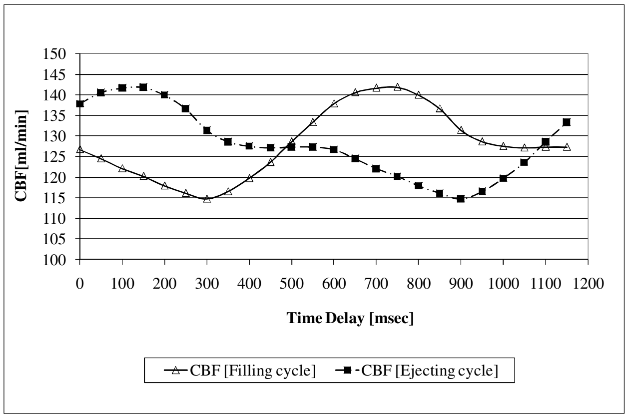

Figure 7. Mean coronary blood flow (CBF) in the first and second cycle of the PUCA (1:2 synchronization ratio). The mean values were calculated during the cardiac cycle.

In this case the cardiac cycle duration was 600 msec (HR=100 bpm) while the PUCA cycle duration was 1200 msec. The delay goes from 0 msec (that means that the PUCA starts to aspirate at the same time of the left ventricular systole) to 1150 msec (at the end of the second cardiac cycle) with a step of 50 msec. The mean CBF assumes different values during the PUCA filling and ejection cycle.
Figure 7 shows that the mean CBF is maximum for time delays of 0-50 msec and of 600-650 msec (considering the sum of the mean CBF in both filling and ejecting cycles). These time delays correspond to set the PUCA functioning in counterpulsation modality so that the PUCA ejects blood in the aorta during diastole. Results show that the counterpulsation modality assures a better coronary perfusion.

4. Conclusion

The model permits to study the interactions between the PUCA and the cardiovascular system for different functioning modes. Results show that the pump contributes to the ventricular unloading and to increment the CBF. In particular, the frequency ratio between the PUCA and the cardiac cycle influences the performance of the pump in terms of the cardiac output and PUCA flow. Especially for higher HR a synchronization ratio of 1:2 can assure a better performance than 1:1 ratio. The other parameter that can influence the pump performance is the time delay between the cardiac and the PUCA cycle: different ejection beginning respect to the cardiac cycle can improve the mean CBF especially in counter pulsation timing. This study may have consequence for the optimization of the PUCA performance in order to obtain the best haemodynamic support conditions for the patient.

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