

CARDIOSIM[®]: A Numerical Simulator and its Application to Study the Interactions between Mechanical Circulatory Assist Devices and the Native Cardiovascular System

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ABSTRACT

In this chapter, we will present an “in silico” simulation study of the interactions between (mechanical) Ventricular Assist Devices (VADs) and the native cardiovascular system. In this study, we aim at a quantitative description of these interactions which is based on a mathematical modeling concept and a numerical simulation approach of the underlying physiological and pathophysiological processes. Much progress has been achieved in the design of ventricular assist devices; VADs are now in position to provide patients suffering from a seriously impaired ventricular function with an excellent hemodynamic support. Although the implantation of VADs is already routinely done in clinical settings, the planning and carrying out of such interventions is nevertheless a challenging task. In the planning phase, cardiologists and heart surgeons must, among other things, consider the interactions of the ventricular assist device with the native cardiovascular system which are to be expected after the implantation of a VAD. Thus,

the specific objective of our simulation efforts is to develop an “in silico” simulation system which will provide experienced cardiologists and especially novice cardiologists with a deeper understanding and a quantitative assessment of the hemodynamic changes after the implantation of a VAD. The simulation studies were carried out by using our in-house numerical simulation system CARDIOSIM®. In the following, we will present a detailed description of CARDIOSIM®, give an overview its performance capacities, and point out and analyze important simulation results.

Keywords: Cardiovascular numerical model; Heart failure; Ventricular elastance; Ventricular assist device; Hemodynamic; LVAD; Pulsatile blood pump

INTRODUCTION

At present, the use of a Ventricular Assist Device (VAD) has already become a clinical routine measure for patients suffering from heart failure. For a deeper understanding of the interactions of a VAD with the native cardiovascular system, we carried out simulation studies of the patients' hemodynamic conditions before and after the application of a pneumatic Left Ventricular Assist Device (LVAD).

In our studies, we used our in-house simulator CARDIOSIM® [1], a modular simulation system based on a lumped parameter modeling approach. CARDIOSIM® has proven beneficial for various studies of hemodynamics in several clinical problem areas, such as mechanical ventricular assistance, application of a total artificial heart, application of a Thoracic Artificial Lung (TAL), and use of an Intra Aortic Balloon Pump (IABP). This “in silico” simulator can be used to study and to interpret the arterio-ventricular interaction in circulatory and pathophysiological conditions during pharmacological assistance, mechanical circulatory and/or ventilatory assistance [2-5].

CARDIOSIM® has a modular structure; it contains a basic core module for the native cardiovascular system and numerous supplementary modules. This structure gives the user the opportunity to assemble a simulation system for particular application areas of support devices which will precisely fulfill the user's requirements. For the simulation of the hemodynamics in these clinical areas, a considerably large number of supplementary modules already exist, such as:

- Several supplementary modules for all kinds of ventricular assist devices (for both pulsatile and continuous flow) [6-7],
- A supplementary module to reproduce the Biventricular Pacemaker (BiV) behavior [8-9]
- A supplementary module for the total artificial heart,
- A supplementary module for intra-aortic balloon pumping [10],
- A supplementary module for the Thoracic Artificial Lung (TAL) [3,11].

Moreover the effects induced by mechanical ventilatory assistance can be also simulated using CARDIOSIM® [4-7,10].

In this chapter, we will present simulation studies that refer to a patient with cardiac failure who is subject to the application of a pneumatic left ventricular assist device. The details of these investigations are described in the next section. Subsequently, we will present results of these simulation studies. We were able to show that by applying the LVAD, the hemodynamic state of the patient can be considerably improved.

MATERIALS AND METHODS

In the following, we will give an overview of CARDIOSIM® software simulator. In particular, we will outline its structure and its range of applicability. Thereafter, we will point out the methodological basis of our investigations which are aiming at quantitatively assessing the interactions between Ventricular Assist Devices (VADs) and the native cardiovascular system.

Overview of the Numerical Simulator CARDIOSIM®

The design of CARDIOSIM® is based on a lumped parameter modeling approach and a strictly modular building principle. It consists of:

- A basic core module,
- Supplementary modules which are highly specialized; they have been tailored for specific applications.

The basic core model comprises all the necessary lumped components which are needed to simulate the hemodynamics of the entire (native) cardiovascular system. The aforementioned supplementary modules are intended for very special simulation studies which involve for instance hemodynamic studies under the assumption that the patient's cardiovascular system is connected with a Ventricular Assist Device (VAD) or a thoracic artificial lung. There exists even a supplementary model for hemodynamic simulations in the case of the application of a Total Artificial Heart (TAH).

Development Process of CARDIOSIM®

CARDIOSIM® has been developed at our CNR Institute of Clinical Physiology in cooperation with Sapienza University of Rome and The National Institute for Cardiovascular Research, Bologna, Italy. Moreover, other institutions in Italy and abroad contributed to the development process. The software implementation was performed in Visual Basic language under Windows Operative System [1].

A considerable number of highly specialized modules have already been developed, and it is planned to develop further modules of this kind in the near future.

Basic Core Module of CARDIOSIM®

In the first place, the basic core module allows us to simulate the cardiovascular system of an average healthy adult. In this case, the simulations are carried out based on default values for the individual model parameters. However, we must very often consider patient-specific deviations and especially pathological changes. In our modeling approach, we take account for such deviations by assigning values to the system parameters which are different from the default values. The new values are determined such that the simulator is adapted to the aforementioned deviations in the best possible way. A circuit diagram of the basic core module is depicted in Figure 1. In this figure, the functional elements for all sections of the cardiovascular system can be seen. Subsequently, the modeling approach for all these section will be described.

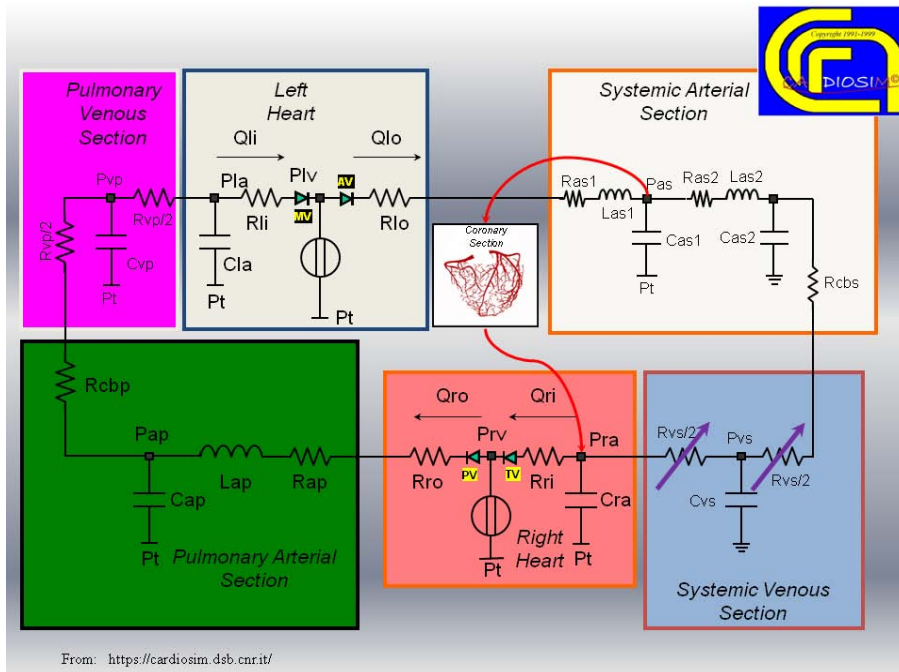


Figure 1: Electric analogue of the cardio circulatory network. Systemic (pulmonary) arterial section is modeled by characteristic resistance R_{as1} , R_{as2} (R_{cp}), by inertances L_{as1} , L_{as2} (L_{ap}) and by arterial compliances C_{as1} and C_{as2} (C_{ap}). R_{cbs} (R_{cbp}) is the variable peripheral systemic (pulmonary) arterial resistance. Systemic (pulmonary) venous section is modeled by compliance C_{vs} (C_{vp}) and by variable resistances $R_{vs}/2$. $R_{vp}/2$ are the pulmonary venous resistance. P_{as} (P_{ap}) is the systemic (pulmonary) arterial pressure, P_{vs} (P_{vp}) is the systemic (pulmonary) venous pressure, P_{lv} (P_{rv}) is the left (right) ventricular pressure, P_{la} (P_{ra}) is the left (right) atrial pressure. Q_{li} (Q_{ri}) and Q_{lo} (Q_{ro}) are the left (right) ventricular flow input and output. The combination resistance R_{li} (R_{lo}) and diode MV (AV) reproduces the mitral (atrioventricular) valve. R_{ri} (R_{ro}) and TV (PV) reproduces the tricuspid (pulmonary) valve. The compliance C_{la} (C_{ra}) reproduces the left (right) atria behavior.

Supplementary CARDIOSIM® Modules

In recent years, a considerable number of tasks using CARDIOSIM® have been completed. In carrying out these tasks, we were required to develop a number of supplementary modules; most of them are general enough to be reused for similar investigations. In the following, we will itemize the most important supplementary CARDIOSIM® modules of this kind which are now at our disposal:

- Modules for Left/Right/Biventricular Pneumatically Actuated Ventricular Assist Devices (LVAD, RVAD, BVAD) which generate a pulsatile flow [6-7];
- A module for a Totally Artificial Heart (TAH);
- A module for an Intra Aortic Balloon Pump (IABP) [10];
- A module to reproduce the Biventricular Pacemaker (BiV) behavior [8-9] ,
- A module for the Pulsatile Catheter (PUCA) pump (pneumatic pump) [3];
- A module for a rotary flow pump (Hemopump - at present, the Hemopump is no longer used, but its design principle is applied to the development of other circulatory assist devices) [4,12];
- A module for the application of a Thoracic Artificial Lung (TAL) [3,11].

In our simulations of the interactions between a LVAD and the native cardiovascular system, we used one of the aforementioned modules, viz. the module for the pneumatically actuated left ventricular assist device generating a pulsatile flow.

Left and Right Heart

We used mathematical models which describe the behavior of both ventricles. The ejection and contraction ventricular phases for both ventricles are reproduced using a variable elastance model as described in [13-15]. The left and the right atrium were described by the compliances C_{la} and C_{ra} (Figure 1) which, moreover, take on constant values (linear approximation) [2,14-15]. This representation permit to reproduce only the passive phase and not the active phase of the atrium. The connection of the ventricle to the circulatory network is realized by means of valves, which are assumed to be ideal [12,14]. The approach of Guyton [16] and Sagawa [13] wich describes the general relations of the cardiocirculatory network and the Starling's law of the heart [17-18], was considered in the development of our software simulator.

Systemic Arterial Section

The model for the systemic arterial section is modeled by two RLC circuits. The RLC circuit downstream from the left ventricle (Ras1, Las1 and Cas1) models the arteries that are situated within the thoracic cavity, whereas the arteries out of the thoracic cavity are modeled by the second RLC circuit (Ras2, Las2 and Cas2) [19,20]. In our modeling approach, we took into account

that the extramural pressure of the intrathoracic arteries is the mean intrathoracic pressure P_t (Figure 1) [10,14]. The extramural pressure of the arteries out of the thoracic cavity (modeled by Ras2, Las2 and Cas2) cannot be easily determined, since no reliable data for the abdominal pressure and the pressures in other sections of the human body can be found in the literature. Although it is an approximation, it is nevertheless justifiable to assume that extramural pressure is the ambient pressure (pressure value is equal to zero).

Systemic Capillary Bed

The capillary bed of the systemic circulation has been modeled by only one lumped parameter, viz. the resistance to flow R_{cbs} .

Systemic Venous Section

The entire systemic venous system is modeled by only one RC circuit ($R_{vs}/2$, C_{vs} , $R_{vs}/2$). The inertial forces have justifiably been neglected. The extramural pressure of the veins has been supposed to be the ambient pressure. The resistance to flow, however, has been assumed to be variable, mainly because the veins, especially the splanchnic veins, are highly compliant. Hence, their luminal cross-sectional areas (diameters) will vary considerably with the variations of the venous (transmural) pressure. As a consequence the resistance to flow will vary significantly [16].

Pulmonary Arterial Section

The systemic arterial section is modeled by a RLC circuit (R_{ap} , L_{ap} and C_{ap}).

Pulmonary Capillary Bed

The capillary bed of the pulmonary circulation has been modeled by only one lumped parameter, viz. the resistance to flow R_{cbp} .

Pulmonary Venous Section

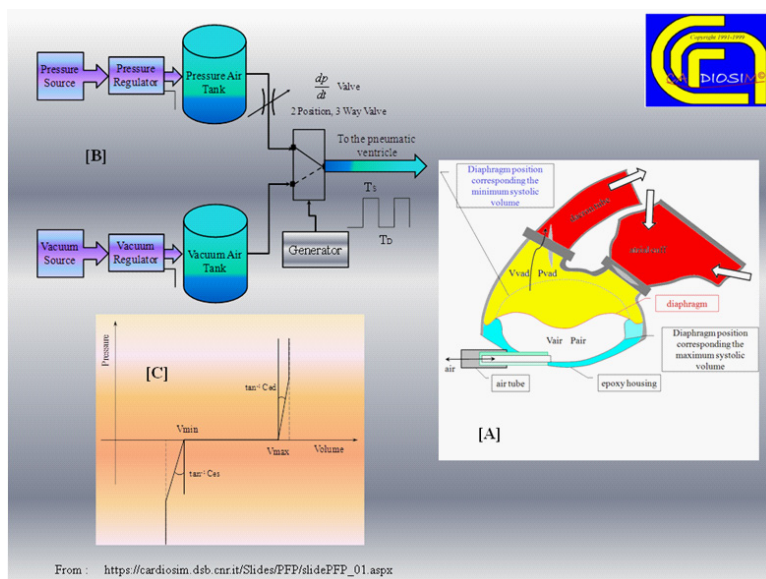
The pulmonary venous system is modeled by only one RC circuit ($R_{vp}/2$, C_{vp} , $R_{vp}/2$) [21]. As in case of the above-described systemic venous system, the inertial forces have justifiably been neglected, but, different to the systemic venous system, the extramural pressure of the veins has been supposed to be the mean intrathoracic pressure. Like in the case of the systemic venous system, however, the luminal cross-sectional areas (diameters) will vary with the variations of the transmural pressure. As a consequence the resistance to flow will vary considerably.

Coronary Circulation

Into CARDIOSIM® have been implemented various modules (more or less complex) that allow to simulate the behavior of the coronary bed. In all modules, the coronary network is implemented by lumped parameters representation based on the intramyocardial pump concept [2,22-24].

Supplementary CARDIOSIM® Module for Pneumatic Vads

Since about thirty years, ventricular assist devices of various function principles have been widely used in clinical settings, especially in intensive care units to treat patients with severely impaired heart function, such as heart failure, deep cardiogenic shock, some forms of myocarditis, and high-grade dilative cardiomyopathy. They very often fulfill the function of being a bridge to recovery or a bridge to transplantation. As mentioned in the preceding subsection, we will concentrate here on a special kind of left ventricular assist devices, viz. a pneumatically actuated left ventricular assist device. A schematic view of the structure of the device can be seen in Figure 2 (A) [25-26].



From : https://cardiosim.dsb.cnr.it/Slides/FFP/slideFFP_01.aspx

Figure 2: Panel [B] shows the general layout of the driving unit pneumatic ventricle system. TS and TD represent the systole and diastole time respectively. The pneumatic ventricle is visible in panel [A]. Dashed and continuous lines inside the ventricle represent two different position of the membrane that divide the chamber containing the blood from the one containing the air. Vair and Pair (Vvad and Pvad) represent the volume and the pressure into the part of the ventricle connected to the air tube (into the ventricle). Panel [C] shows the pressure-volume relationship which has been assumed for the pneumatic ventricle diaphragm in its end-systolic and end-diastolic positions. Ces and Ced are the diaphragm compliance at end-diastolic and end-systolic position respectively. Vmin and Vmax are the minimum and the maximum pneumatic ventricular volume respectively. The diaphragm describes the interaction between the air and the blood in the pneumatic ventricle its characteristic is shown in panel [C]. It is assumed that the pressure drop across the diaphragm is far from (Vmin) and (Vmax) values and, outside these limits, it is increasing with the compliance of the diaphragm, which may be different in the end-systolic and end-diastolic positions.

Blood pump

A schematic view of the structure of the blood pump can be seen in Figure 2(A). This ventricular assist device consists of a paracorporeal (extracorporeal) blood pump, two valves and the required cannulae. The housing of the blood pump is provided with a connection nozzle for the air inlet/outlet tubular duct and two separate connection nozzles, one for the blood inlet cannula, the other for the blood outlet cannula. The interior of the VAD is subdivided into a blood chamber and an air chamber by a flexible diaphragm. When the pressure in the air chamber increases, the diaphragm is elevated and blood will flow out of the pump. On the contrary, if the pressure in the air chamber decreases and becomes negative, blood will enter into the pump. In the so-called serial mode of operation, the inlet cannula is introduced into the left ventricle through the apex of the left ventricular wall, whereas in the so-called parallel mode of operation, the inlet cannula is inserted in the left atrium. In both the serial and parallel mode, the outlet cannula is connected with the ascending aorta.

The equations describing the pneumatic VAD behavior are reported in [25-27].

Pneumatic unit and control circuit arrangement

The pneumatic unit and its control circuit arrangement which is contained in our aforementioned supplementary module “left pneumatic pulsatile LVAD” is shown in Figure 2(B) [25-26]. The pneumatic unit consists of two tanks, one for positive air pressure and the other for vacuum. By means of an externally controlled air switching valve the inlet/outlet nozzle of the blood pump is consecutively connected with the positive air pressure tank and the vacuum tank. In the positive air pressure phase, blood is ejected into the aorta, whereas in negative pressure phase (vacuum tank) the blood pump is filled with blood. The switching valve is controlled by a dedicated control circuit arrangement which we designed for the following pumping modes:

- Synchronous mode

In the synchronous mode, the pump action is synchronized with the heart action in such a way that the blood pump of the LVAD ejects blood into the ascending aorta during the diastolic phase. As a consequence, the aortic pressure increases during the diastolic phase of the native left ventricle and with this the coronary flow will rise. The synchronization is based on the R-wave of the ECG. The control circuit arrangement receives a signal which is derived from the patient's ECG and delivers a command to the switching valve of the pneumatic unit.

- Asynchronous mode

In the asynchronous mode, the VAD operates at a fixed pump rate which is not related to heart rate of the native ventricle.

- “Full to empty” mode

In this mode, the blood pump will eject blood as soon as a sensor signals that the blood chamber is completely full. After having emptied the blood chamber completely, it will be filled again.

Thus, the LVAD will in this mode operate with a fixed stroke volume. As a consequence, the rate of pumping is only dependent on the filling rate of the blood chamber.

Methodological Basis of Our Simulation Studies

The primary goal of our simulation studies is the acquisition of a fair knowledge and a deep understanding of the hemodynamic changes which will occur when a LVAD is used to improve the patient's hemodynamics.

Overview of our simulation studies

All our simulation studies are based on the assumption of a patient with a left-sided heart failure which is a condition where the patient's cardiac output is not high enough to meet the body's demand. In the therapy of patients with advanced heart failure it is common practice to use Left Ventricular Assist Devices (LVADs), either as bridge to transplant, a bridge to recovery, or as destination therapy. Our simulation studies aimed at investigating the interactions of the impaired native cardiovascular system with a pneumatically actuated left ventricular assist device.

In carrying out our simulations, we made use of:

- The above-described “basic core module” of CARDIOSOM®. Instead of using the default parameters which refer to the average healthy adult, we assigned specific values to the parameters which are characteristic for an assumed heart failure (the ventricular elastances $El_v=0.8 \text{ cm}^3\cdot\text{mmHg}$ and $Er_v=0.5 \text{ cm}^3 \text{ mmHg}$; the left residual volume $Vl_o=25 \text{ cm}^3$, the systemic arterial compliances $Cas_1=Cas_2=3.0 \text{ cm}^3\cdot\text{mmHg}^{-1}$, and the resistance of the systemic capillary bed $R_{cbs}=1300 \text{ g}\cdot\text{cm}^{-4}\cdot\text{sec}^{-1}$). The mean Intrathoracic Pressure (Pt) and Heart Rate (HR) were fixed to $Pt=-4 \text{ mmHg}$ [5] and $HR=80 \text{ bpm}$ respectively.
- The above-described Supplementary CARDIOSIM® module for pneumatic LVADs. For our simulation studies we connected this module “in parallel” with the basic core module. We have assigned specific values for the parameters of the basic core module which are representative for the assumed heart failure. At first, we simulated the impaired function of the cardiovascular system without using the LVAD. Afterwards we carried out simulation runs to study the improvements which can be achieved by using a pneumatic LVAD. We confined ourselves to simulations in synchronous mode, synchronous mode with delay, and “Full to empty” mode. The LVAD was applied setting the driving (vacuum) pressure to 200 mmHg (-5 mmHg), the minimum (maximum) pneumatic blood volume to 10 cm^3 (80 cm^3).

Configuration of CARDIOSIM® for our simulation studies

The details of the integration of the basic core module with the supplementary CARDIOSIM® module for pneumatic LVADs can be seen in Figure 1. In parallel mode, the inlet cannula of the LVAD is connected with the left atrium (Figure 2 red dotted line), whereas in serial mode, the inlet cannula is connected with the apex of the left ventricle (Figure 3 – green line). The outlet cannula is connected with the ascending aorta (Figure 3 – red line).

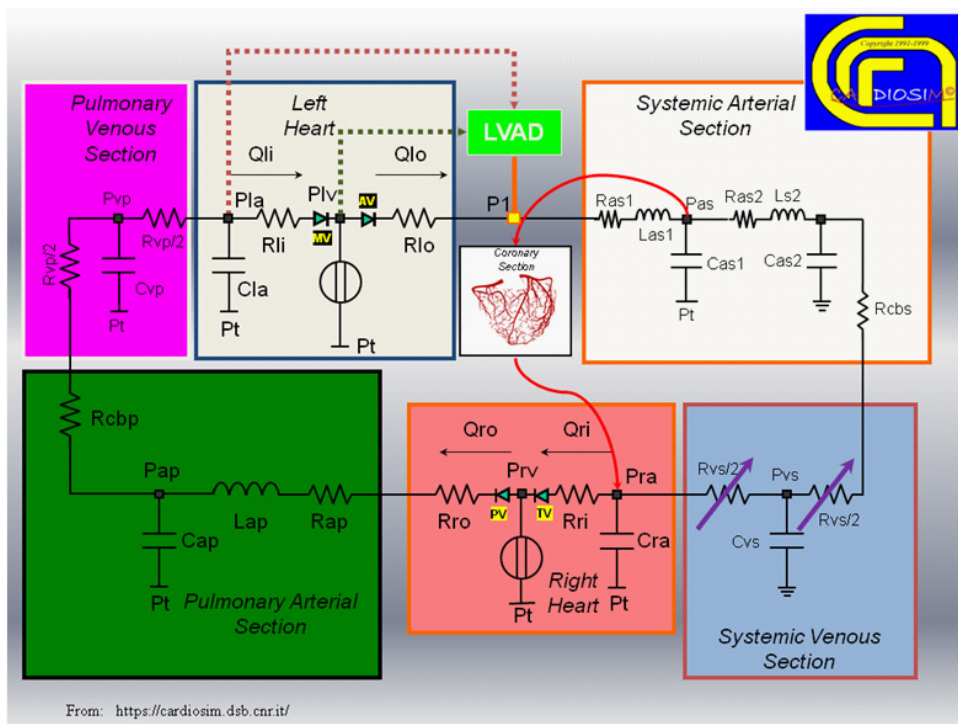


Figure 3: Electric analogue of the cardiocirculatory network with LVAD. LVAD, connected in parallel (serial) mode, takes blood from the left atrium (ventricle) and ejects blood in ascending aorta.

RESULTS

In the following, we will present the results we obtained from our simulation studies of the hemodynamic conditions of a patient with a severe heart failure:

- Before the application of a Left Ventricular Assist Device (LVAD) and
- After the application of a pneumatic LVAD in parallel mode (simulations refer to synchronous mode/ECG triggered; synchronous mode ECG triggered with time delay of 250 ms; and “full to empty” mode).

Simulation Results – Heart Failure without LVAD

As described previously, we assigned specific values to the parameters of the basic core module to adapt this module to the conditions of a severe heart failure. The results are presented in Figure 4 which depicts the results window of the graphical user interface of CADIOSIM®. In panel [A] of this window, the pressure-volume loop of the left ventricle can be seen. Panel [B] contains a pressure-time diagram of the Left Ventricular Pressure (Plv) and the Aortic Pressure (Pas). In the numerical panel [C], the simulated values of important parameters appear, such as:

- Heart rate HR;
- Mean pressure values (average of cardiac cycle) for systemic arterial pressure Pas (with, as an addition, systolic and diastolic pressure);
- Right ventricular (atrial) volumetric rate of blood Qro (Qri);
- Left ventricular (atrial) volumetric rate of blood Qlo (Qli).

Simulation Results – Heart Failure with LVAD which is Synchronized with the R-wave of the ECG

The simulation results refer to the patient whose native cardiovascular data have been simulated in the preceding paragraph. In particular the following simulation results refer to the hemodynamic state after having provided this patient with a pneumatic LVAD. The results are presented in Figure 5 which depicts the results window of the graphical user interface of CADIOSIM® as before. In panel [A] of this window, the pressure-volume loop of the native left ventricle is depicted (pathological) together with the pressure-volume loop in the assisted case. Panel [B] contains a pressure-time diagram of the left ventricular pressure and the aortic pressure for the pathological (not assisted) and the assisted case. In the numerical panel [C], the simulated values (calculated in presence of the assistance) of important parameters appear as was the previous case. It is possible to observe a reduction in the left assisted ventricular pressure-volume loop: when LVAD was applied, the left Ventricular End Systolic volume (Ves) increased from 127.32 cm³ (Figure 4) to 148.94 cm³ (Figure 5), the Ventricular End Diastolic Volume (Ved) decreased from 170.52 cm³ (Figure 4) to 161.62 cm³ (Figure 5) and the left ventricular stroke volume (SV) decreased from 43.2 cm³ (Figure 4) to 12.68 cm³ (Figure 5). The VAD activation increases the systolic (diastolic) arterial pressure from 80.0 mmHg to 106.1 mmHg (from 59.3 mmHg to 80.4 mmHg). Panel [B] (Figure 5) shows that instantaneous arterial Systemic Pressure (Pas) increases when LVAD is applied. In the presence of LVAD we define the Total Blood Flow (TBF) as the sum of Qlo and LVAD flow (QLVAD), i.e TBF=Qlo+QLVAD. In pathological conditions, the blood flow into the circulatory network (in stationary conditions) coincides with the left ventricular output flow (Qlo = 3.45 [l/min]) (Figure 4). When pneumatic ventricular assistance is applied, the TBF (in stationary conditions) is 4.85 [l/min]: in this case Qlo is 1.01 [l/min] and QLVAD is 3.84 [l/min].

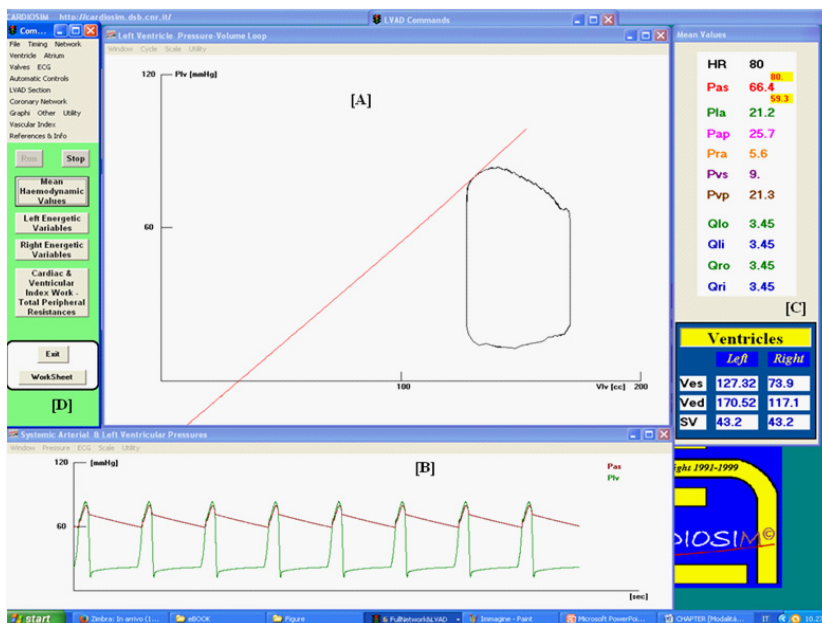


Figure 4: Graphical and numerical output produced by CARDIOSIM® software simulator of the cardiovascular system. Window [A] shows the left ventricular pressure-volume loop obtained when pathological conditions (heart failure) were simulated. Window [B] shows the instantaneous waveforms of left ventricular (Plv) and systemic arterial (Pas) pressures. In column [C] are reported the mean flow and pressures values calculated during the cardiac cycle. Pla, (Pra), Pap and Pvs (Pvp) are the left (right) atrial, pulmonary arterial and venous systemic (pulmonary) pressures. Qli (Qri) and Qlo (Qro) are the left (right) ventricular flow input and the left (right) ventricular flow output respectively. Column [D] represents the command column.

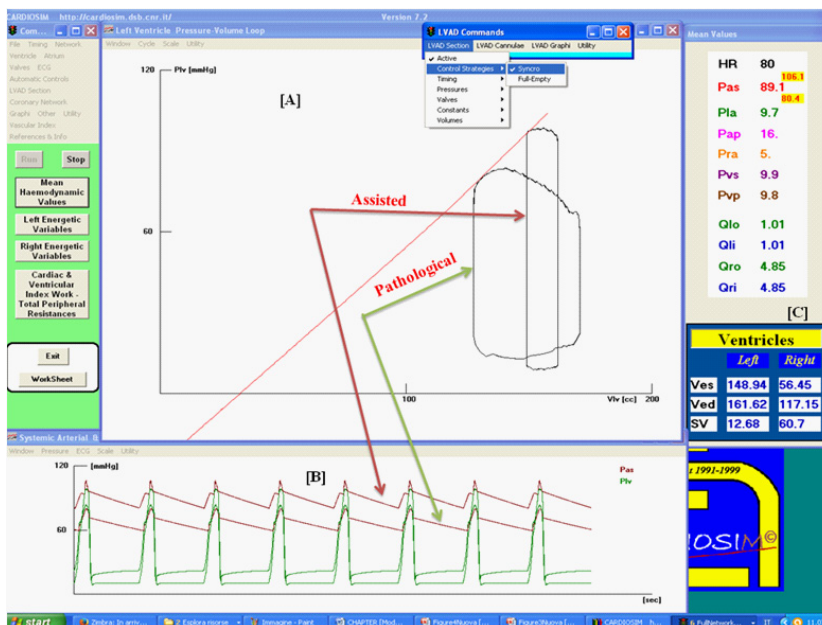


Figure 5: Graphical and numerical output produced by CARDIOSIM®. Window [A] shows the left ventricular pressure-volume loop obtained in pathological and assisted conditions. Window [B] shows the instantaneous waveforms of left ventricular (Plv) and systemic arterial (Pas) pressures in pathological and assisted conditions. In column [C] are reported the mean flow and pressures values calculated during assisted conditions. The LVAD assistance was driven in synchronized mode. Above the window [A] it is possible to observe the command section for driving the LVAD.

Simulation Results – Heart Failure with LVAD which is Synchronized with the R Wave of the ECG and in addition, with a Time Delay of 250 ms

The simulation procedure differs from those in the preceding paragraph only in the time delay of the synchronization. The simulation results are presented in Figure 6 in the same manner as before. A comparison of the instantaneous pressure diagrams in Figure 6 and Figure 5 reveals very clearly the ramifications of the time delay which are quite significant. Comparing these results with those shown in Figure 5 it is possible to observe that TBF increases up to 4.99 [l/min], the left ventricular blood flow increases from 1.01 [l/min] (Figure 5) to 1.5 [l/min] and the mean left atrial pressure decreases from 9.7 mmHg (Figure 5) to 8.3 mmHg (Figure 6). The LVAD synchronous mode can be considered similar in principle to counterpulsation.

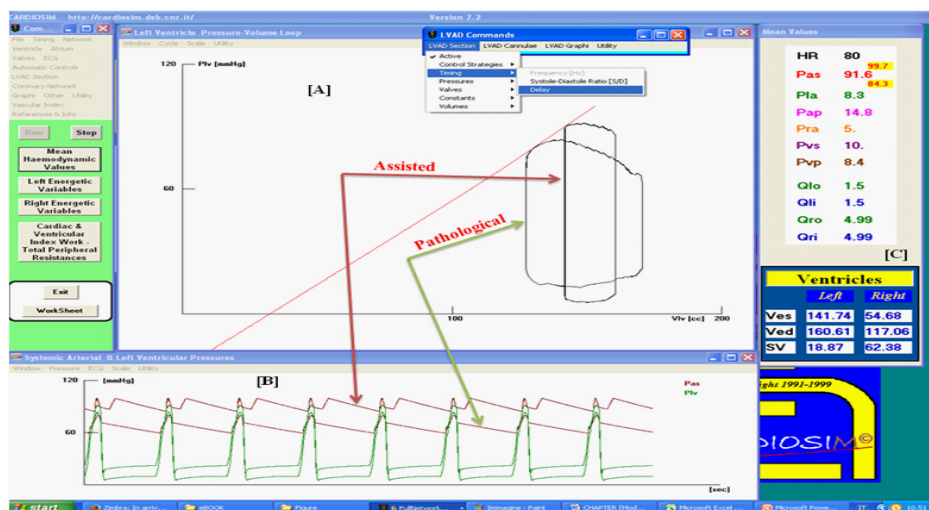


Figure 6: Window [A] shows the left ventricular pressure-volume loop obtained in pathological and assisted conditions when LVAD was driven in synchronized mode with a delay of 250 ms respect to the start of the R wave. Window [B] shows the instantaneous waveforms of left ventricular and systemic arterial pressures in pathological and assisted conditions. In assisted conditions observing the Pas waveforms it is possible to see as the LVAD ejects in delay respect to the native ventricle. In column [C] are reported the mean flow and pressures values calculated during assisted conditions.

Simulation Results – Heart Failure with LVAD in “Full to Empty” Mode

As has already been described, no synchrony exists at all. There are no time relations between the points in time where ejections of the native left ventricle and the blood pump take place, because the ejection of blood from the blood pump will take place as soon as the blood chamber of the blood pump is completely full, thus at a point of time which is completely independent of the cardiac cycle of the native heart. The simulation results are presented in Figure 7 which depicts the results window of the graphical user interface of CADIOSIM®. In panel [A] of this window, the pressure-volume curve of the native left ventricle is depicted (pathological) together with the pressure-flow curve in the assisted case. The curve in the assisted case will not be only one closed loop due to the lack of synchrony. Rather, the curve will change its shape considerably from one cardiac cycle to the next. Panel [B] contains a pressure-time diagram of the Left Ventricular Pressure (Plv) and the Aortic Pressure (Pas) for the pathological (not assisted) and the assisted case. This diagram also shows clearly the consequences of the lack of synchrony.

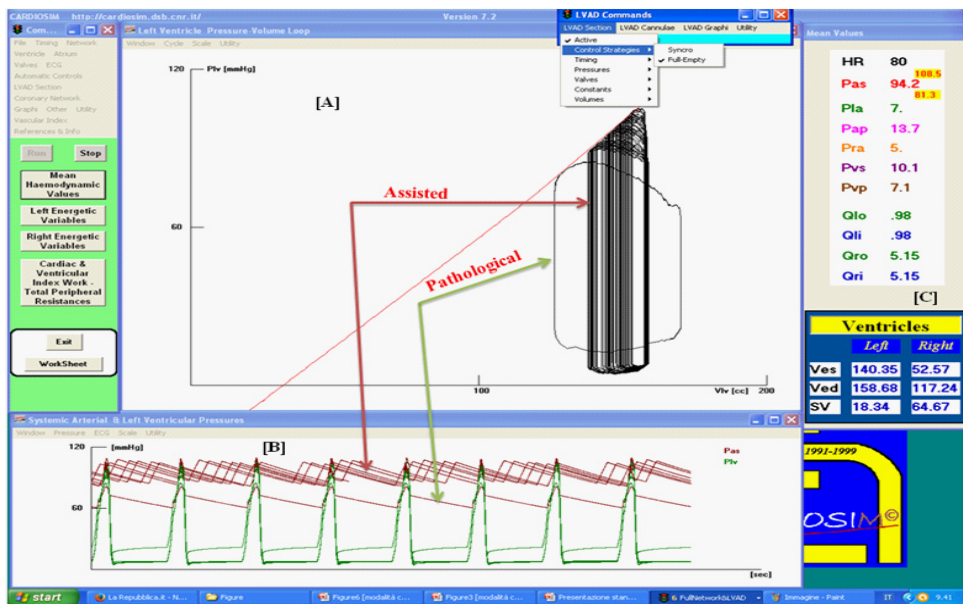


Figure 7: Graphical and numerical output obtained when LVAD was driven in the fill-to-empty mode. Window [A] shows the left ventricular pressure-volume loop obtained in pathological and assisted conditions. The fill-to-empty mode produced different cardiac loops because the ejection of the native ventricle and the ejection of the LVAD were not synchronized. Window [B] shows that the instantaneous Pas waveforms is conditioned by LVAD ejection in different time. Column [C] reports the mean flow and pressures values calculated during one generic assisted cardiac cycle, these values change in correspondence of any cardiac cycle.

CONCLUSIONS

Numerical simulation enables us to quantitatively assess physiological and pathophysiological processes. Simulation studies are thus pivotal to obtain important knowledge of the cause and the progress of diseases. In contrast to clinical experiments, simulation studies do not entail any risks, they help physicians to make more precise diagnoses, they allow physicians a better planning of therapies and, moreover, to predict their success. We handled our simulation tasks by using our in-house simulator CARDIOSIM®. The simulations presented in this paper referred to a patient with heart failure whose heart and circulation was supported by the application of a pneumatic left ventricular assist device. The studies showed clearly that, by providing the patient with such a device, we are able to improve his or her hemodynamic state considerably. However, the range of applicability of CARDIOSIM® goes far beyond the simulation studies described in this paper. Simulation studies referring to the application of some other cardiovascular and ventilatory assist devices have already been carried out. In the near future, we plan to use CARDIOSIM® in simulation studies that aim at comparisons of the performance capacities of VADs with pulsatile versus continuous flow on the one hand and “in series” versus “in parallel” mode on the other.

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