

Computer Modeling of Interactions of an Electric Motor, Circulatory System, and Rotary Blood Pump

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The innovative ventricular assist systems (IVAS) is the next generation ventricular assist device for use as a permanent implantable device. Its practical application depends upon control of the electric motor and interactions of the electric motor, blood pump, and circulatory system. Computer modeling and simulation are necessary to investigate and evaluate the interactions and feasibility of sophisticated control algorithms. In this paper, a computer model of the complete system, including the cardiovascular system, blood pump, and electric motor, is proposed. The model is obtained based on an electric circuit model of the cardiovascular system, a parametric model of the blood pump, and a dynamic model of the electric motor. The cardiovascular system uses nonlinear parameters to simulate the time-varying property of the ventricles, and the cannula collapse effect caused by over-pumping. The blood pump model can be obtained either from pump design data, or test data. The motor control can be operated with closed-loop regulation, depending upon physiologic requirements. Different operation modes (current or speed) of the electric motor can be simulated. The computer model is implemented using MATLAB. Various motor operation modes are simulated and their effects are evaluated. By adjusting the motor input, the pump can achieve proper output so that normal physiology can be obtained. In addition to evaluating existing operation modes and their effect on the physiologic system, the computer simulation results show that this computer model can contribute significantly to the development of new physiologic control algorithms. It is demonstrated that, using this motor-pump-physiology interaction model, development of an innovative ventricular assist system can be greatly facilitated. *ASAIO Journal* 2000; 46: 604–611.

A continuous flow, rotary blood pump has been considered an effective solution to cardiovascular disease. A rotary blood pump has numerous advantages for the next generation of left ventricular assist devices (LVAD): small size, high efficiency, and simplicity of implantation.^{1–3} However, these and other advantages are bought at the expense of inherently greater demand for external control. Unlike conventional displacement pumps, which can adapt to filling volume, the rotary blood pump is a passive device whose output is not adapted to

the filling volumes, and external control is much more complicated. Furthermore, flow and pressure sensors are not intended, yet pump flow must be adjusted according to physiologic demand.

To investigate the performance characteristics of the blood pump and its interaction with the pulsating heart, a computer model is necessary to predict and analyze performance before any *in vitro* or *in vivo* tests. New control algorithms can also be derived based on the computer simulation.

Computer simulation of the circulatory system has been described in the literature for many years.^{4–6} A general approach is to simulate the cardiovascular system as an electric circuit, because the governing equations of the cardiovascular system are similar to those of an electric circuit, and physiologic parameters can be obtained by solving the circuit equations. Ventricles can be modeled by time-varying components. Left ventricular failure can be easily modeled by choosing appropriate time-varying components corresponding to the left ventricle.⁵

Very little literature has discussed the modeling of interactions between the circulatory system and heart assist pumps. Mitsui *et al.*⁵ modeled a nonpulsatile left ventricular bypass pump using a constant flow source. Schima *et al.*⁶ used second order polynomials for centrifugal pumps, which were obtained by curve-fitting to the pressure-flow data from the pump characteristics. However, pump-physiology interactions for rotary blood pumps are very sophisticated, and effects of different motor-pump operations on physiology are still unclear; this has affected development of real-time physiologic control algorithms.

The purpose of this study was to develop a computer simulation model of the circulatory system with an LVAD assist device. Self-regulated effects by baroreceptor mechanisms in response to physiologic dynamics are considered in the model. Mathematical models of the electric motor and the blood pump are proposed, and effects of certain motor-pump operation modes are discussed. Based on this model, sophisticated, intelligent control algorithms can be developed to adapt to physiologic demands.

Electric Circuit Model of the Cardiovascular System

Electric Circuit Model

The electric circuit model of the circulatory system is shown in **Figure 1**. Vascular resistance, vascular compliance, and inertia of blood in the vessels are represented by electrical resistance (R), capacitance (C), and inductance (L). The ventricles are modeled by time-varying parameters, and all the valves are modeled by diodes that can only carry unidirectional

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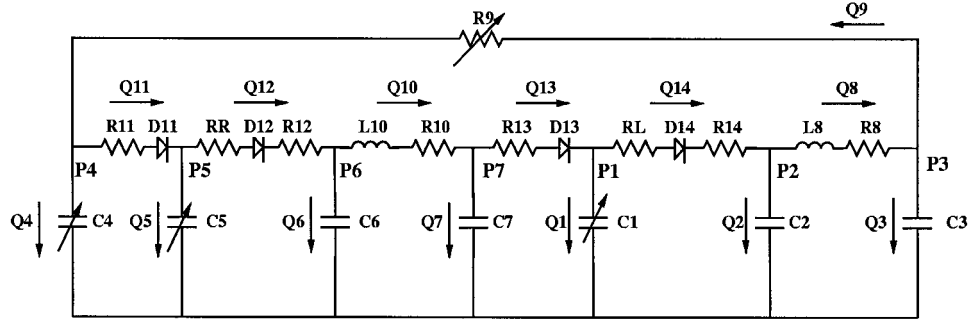


Figure 1. Electrical model for the cardiovascular circulatory system. 1, left ventricle; 2, aorta; 3, peripheral vessels; 4, systemic vein and right atrium; 5, right ventricle; 6, pulmonary artery; 7, pulmonary vein and left atrium; 8, aorta; 9, systemic peripheral vessels; 10, pulmonary peripheral vessels; 11, tricuspid valve; 12, pulmonary valve; 13, mitral valve; 14, aortic valve.

tional current flow. Systemic and pulmonary circulations are described by systemic and pulmonary peripheral resistances and inductances representing viscous and inertial properties of the blood flow, with capacitances corresponding to the elastic properties of the vessel walls.

With this model, by choosing appropriate parameters, normal hemodynamics can be simulated. Parameters used in simulation are listed in **Tables 1** and **2**. Left ventricular failure can be simulated by decreasing the amount of compliance change of the left ventricle and, in this paper, it is achieved by increasing the diastolic compliance, C_{1d} . Features of self-regulated control of the circulatory system and its interaction with the blood pump can be attained by adding additional elements to this model.

Baroreceptor Feedback Control

Baroreceptor effects on heart rate (HR) and peripheral resistance (R_g) are nonlinear functions of the arterial pressure with a negative slope, which corresponds to negative feedback in the closed loop system.⁴ **Figures 2** and **3** show the barorecep-

tor feedback loop characteristics of HR and R_g over mean arterial pressure (PAS).⁴

Transition characteristics of the baroreceptor feedback loop has been experimentally derived.⁴ The heart rate response can be written as follows:

$$HR = HR_o + HR_M \left(1 - \frac{PASN^{\mu_{HR}}}{1 + PASN^{\mu_{HR}}} \right) \quad (1)$$

where HR_o and HR_M are threshold values of the heart rate, $PASN$ is the normalized arterial pressure ($PASN = PAS/100$), and μ_{HR} is a constant ($\mu_{HR} = 8$).

Similarly, the peripheral resistance response can be written as:

$$R_g = R_{g_o} + R_{g_M} \left(1 - \frac{PASN^{\mu_R}}{1 + PASN^{\mu_R}} \right) \quad (2)$$

where R_{g_o} and R_{g_M} are threshold values of the peripheral resistance, $PASN$ is the normalized arterial pressure, and μ_R is a constant ($\mu_R = 6$).

It is also shown that variations in heart rate and peripheral resistance caused by the pressure changes are delayed by the efferent pathway from the vasomotor center to the effector cells at the heart muscle. Real time responses of the heart rate and peripheral resistance are expressed as follows:

$$HR(t) = \int_0^t e^{-\frac{t-\tau}{T_N}} \left[HR_o + HR_M \left(1 - \frac{PASN(\tau)^{\mu_{HR}}}{1 + PASN(\tau)^{\mu_{HR}}} \right) \right] d\tau \quad (3)$$

$$R_g(t) = \int_0^t e^{-\frac{t-\tau}{T_R}} \left[R_{g_o} + R_{g_M} \left(1 - \frac{PASN(\tau)^{\mu_R}}{1 + PASN(\tau)^{\mu_R}} \right) \right] d\tau \quad (4)$$

Table 1. Parameters of the Circulation Model

Component	Value	Unit
C_1	12 (systole C_{1s}) -0.3 (diastole C_{1d})	ml/mm Hg
C_2	0.8	ml/mm Hg
C_3	1.0	ml/mm Hg
C_4	600 (systole C_{4s}) -400 (diastole C_{4d})	ml/mm Hg
C_5	16 (systole C_{5s}) -1.2 (diastole C_{5d})	ml/mm Hg
C_6	3.0	ml/mm Hg
C_7	5.0	ml/mm Hg
L_8	0.01 mm Hg ² /ml	
R_8	0.5	mm Hg/ml
R_9	1.0	mm Hg/ml
L_{10}	0.002	mm Hg ² /ml
R_{10}	0.12	mm Hg/ml
R_{11}	0.01	mm Hg/ml
R_{12}	0.01	mm Hg/ml
R_{13}	0.01	mm Hg/ml
R_{14}	0.01	mm Hg/ml
R_{15}	0.05 ($P_{LV} > -3$ mm Hg) -0.25 ($P_{LV} < -3$ mm Hg)	mm Hg/ml
C_{15}	0.2	ml/mm Hg
R_{16}	0.05	mm Hg/ml
R_L	0.0175	mm Hg/ml
R_R	0.0175	mm Hg/ml

Table 2. Parameters of the Regulated Model

Component	Value	Unit
HR_o	50	1/min
HR_M	150	1/min
T_N	5	s
μ_{HR}	8	—
T_{HR}	28	s
K_{HR}	0.6	1/min/W
R_{g_o}	0.85	mm Hg/ml
R_{g_M}	1.5	mm Hg/ml
μ_R	6	—
T_R	26	s

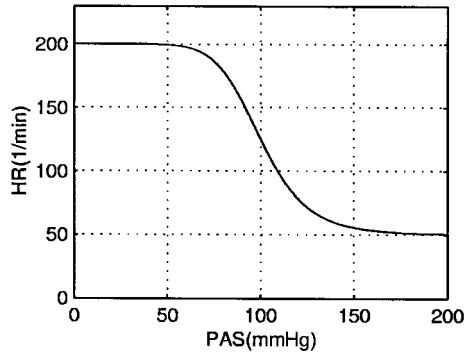


Figure 2. Baroreceptor feedback loop characteristic: heart rate (HR) over mean arterial pressure (PAS).

where T_N and T_R are delay time-constants for $HR(t)$ and $R_9(t)$, respectively.

Regulated Cardiovascular System

Pressure control with the aid of the baroreceptor reflex is achieved by adjusting the heart rate and peripheral resistance as described above. The block diagram of the physiologically regulated closed-loop cardiovascular system is shown in **Figure 4**, where the cardiovascular circulatory system model calculates the physiologic parameters based on the electric circuit model. The baroreceptor sensor senses the arterial pressure and feeds it to the regulator to produce the heart rate and peripheral resistance for the model. This closes the self-regulated baroreceptor feedback control loop.

Interaction of the Circulatory System and the Blood Pump

The electric circuit model with the blood pump is shown in **Figure 5**. Because the rotary blood pump is designed to connect from the left ventricle to the aorta, a new branch is added between the left ventricle and the aorta. R_{15} and R_{16} represent the inlet and outlet cannula resistances, and C_{15} represents the cannula capacitance. R_{15} is a nonlinear resistance to simulate the suction effects, which are normal (0.05) for positive ventricular pressure and significantly higher (0.25) when ventricular pressure is negative (indicating ventricular suction).

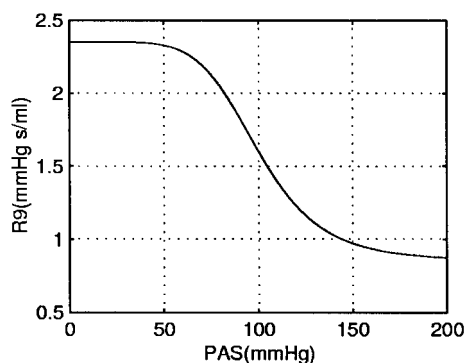


Figure 3. Baroreceptor feedback loop characteristic: peripheral resistance, R_9 , over mean arterial pressure, PAS.

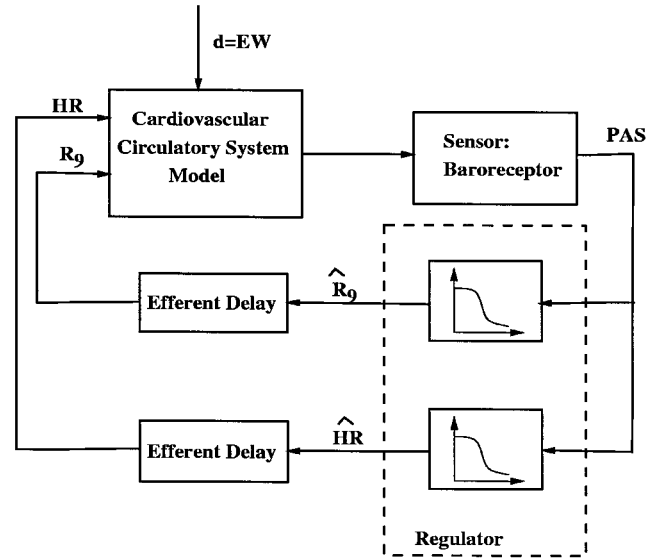


Figure 4. Block diagram of the regulated closed-loop cardiovascular system model. PAS: mean arterial pressure; HR, heart rate; R_9 , peripheral resistance; d , disturbance (EW, ergometric work load).

Equations for the Simulation Model

From **Figure 5** we can obtain the following governing equations:

Equations for the Circulatory System. The volumes can be obtained through the compliances as follows:

$$V_i = \int Q_i dt \quad (5)$$

where $i = 1, 2, \dots, 7$.

Pressure drops across branches 8 through 14 can be written as:

$$P_1 - P_2 = (R_L + R_{14})Q_{14} \quad (6)$$

$$P_2 - P_3 = L_8 \frac{dQ_8}{dt} + R_8 Q_8 \quad (7)$$

$$P_3 - P_4 = R_9 Q_9 \quad (8)$$

$$P_4 - P_5 = R_{11} Q_{11} \quad (9)$$

$$P_5 - P_6 = (R_R + R_{12})Q_{12} \quad (10)$$

$$P_6 - P_7 = L_{10} \frac{dQ_{10}}{dt} + R_{10} Q_{10} \quad (11)$$

$$P_7 - P_1 = R_{13} Q_{13} \quad (12)$$

Flows are balanced by the following node equations:

$$Q_1 = Q_{13} - Q_{14} - Q_{15} \quad (13)$$

$$Q_2 = Q_{14} + Q_{16} + Q_8 \quad (14)$$

$$Q_3 = Q_8 - Q_9 \quad (15)$$

$$Q_4 = Q_9 - Q_{11} \quad (16)$$

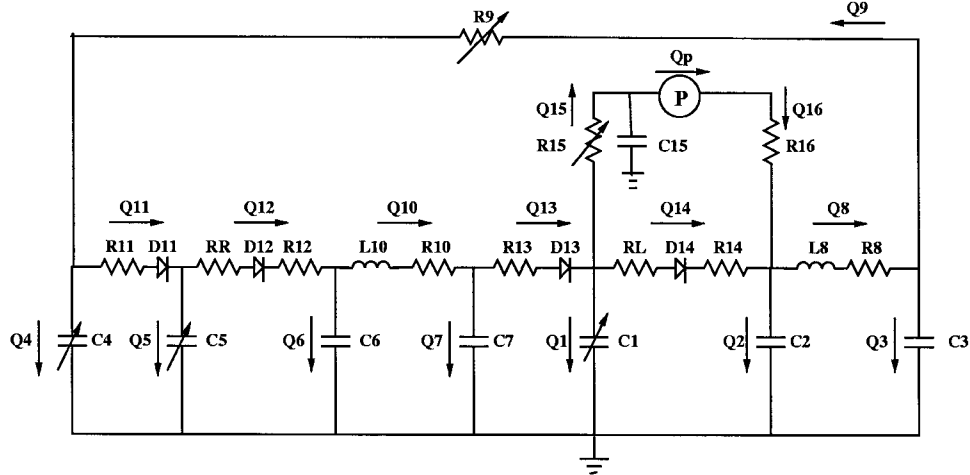


Figure 5. Electrical model of the circulatory system with the blood pump.

$$Q_5 = Q_{11} - Q_{12} \quad (17)$$

$$Q_6 = Q_{12} - Q_{10} \quad (18)$$

$$Q_7 = Q_{10} - Q_{13} \quad (19)$$

The volumes can also be written through compliances and pressures:

$$V_i = C_i P_i \quad (20)$$

where $i = 1, 2, \dots, 7$.

Equations for the Blood Pump. The pump branch including inlet, outlet, and the blood pump is characterized by

$$V_{15} = C_{15} P_{15} \quad (21)$$

$$P_1 - P_{15} = Q_{15} R_{15} \quad (22)$$

$$P_{16} - P_2 = Q_{16} R_{16} \quad (23)$$

$$C_{15} \frac{dP_{15}}{dt} = Q_{15} - Q_{16} \quad (24)$$

and pump pressure and flow are related to the circulatory variables by

$$P_{\text{pump}} = P_{16} - P_{15} \quad (25)$$

$$Q_{\text{pump}} = Q_{16} \quad (26)$$

The pump characteristics can generally be obtained through tests. Two groups of characteristic curves can be extracted from a typical pump test, as shown in **Figures 6** and **7**.

These characteristics can be built into the simulation model as look-up tables. To further simplify the simulation process, mathematical equations can be obtained by curve-fitting.⁷ The characteristics in **Figure 6** can be approximated by

$$Q^2 = a_1 w^2 + a_2 \Delta P + a_3 \quad (27)$$

where w is the pump speed, ΔP the differential flow of the pump outlet to inlet, and a_1 , a_2 , and a_3 are constants. The characteristic curves in **Figure 7** can be expressed by

$$Q = a_4 \frac{I}{w} + a_5 \quad (28)$$

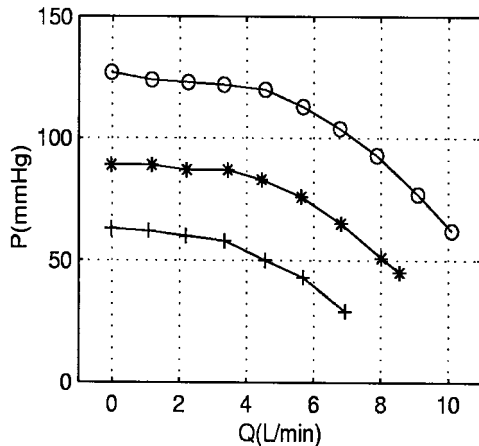


Figure 6. Pump pressure vs. flow. (o) 3500 rpm; (*) 3000 rpm; (+) 2500 rpm.

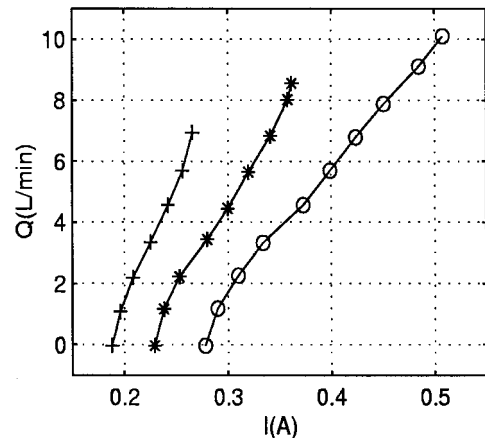


Figure 7. Pump flow vs. motor current. (o) 3500 rpm; (*) 3000 rpm; (+) 2500 rpm.

where I is the motor current, and a_4 and a_5 are constants.

Total blood volume is kept constant. To eliminate the round-off error, this can be a further constraint to modify the variables at every step

$$V(k+1) = \sum_{i=1}^7 V_i(k+1) + V_{15}(k+1) \quad (29)$$

and the venous pressures should be modified at each step by

$$P_4^*(k+1) = P_4(k+1) + \frac{V_0 - V(k+1)}{C_4 + C_7} \quad (30)$$

$$P_7^*(k+1) = P_7(k+1) + \frac{V_0 - V(k+1)}{C_4 + C_7} \quad (31)$$

Modeling of the PM Motor Using a Physiologic Time Base

The time-varying property of the physiologic system is very different from that of the transient electrical system. Usually the time-varying period in the physiologic system is on the order of seconds,^{4,5} whereas for a PM motor, especially when its size is very small, the electrical and mechanical time constants are in microseconds and milliseconds. Therefore, the transient process of the PM motor can be neglected in investigating motor-pump-physiology coupling effects, and steady-state models for the PM motor can be used. In addition, the dynamics of the current and speed can be neglected, and the steady-state voltage equation of the PM motor can be written as equation 32

$$\mathbf{V} = \mathbf{E} + \mathbf{I}(R + \mathbf{j}X) \quad (32)$$

where V is the motor terminal voltage vector, E is the motor back EMF vector, I the motor current vector, and R and X are motor winding resistance and reactance, respectively. The back EMF E is related to the motor speed by

$$E = K_e w \quad (33)$$

where K_e is the motor back EMF coefficient, w is the motor speed.

For a PM motor the inductance is usually very small, therefore X can be neglected. Furthermore, by realizing field oriented vector control, the motor current I is in phase with E , and equation 32 can be written as a scalar equation:

$$V = K_e w + IR \quad (34)$$

With closed loop speed and current regulation, because the motor dynamics can be neglected for the physiologic cycles, the motor current I and speed w are assumed to always follow the command current and speed. There are basically two fundamental operation modes for the electrical motor as described below.

Speed Mode Operation

Speed mode operation is achieved by having motor speed, w , always follow a reference speed, no matter what the load. It is seen that the speed is the most direct motor-pump coupling variable, because the pump runs at the same speed as the motor. Under speed mode operation, the motor current varies depending on the pump load.

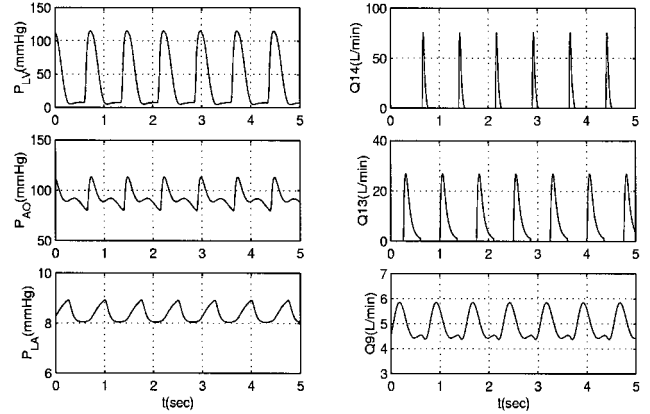


Figure 8. Pressure and flow of a healthy heart. P_{LV} , left ventricular pressure; P_{AO} , aortic pressure; P_{LA} , left atrial pressure; Q_{14} , left ventricular output flow; Q_{13} , left ventricular filling flow; Q_9 , system flow.

Current Mode Operation

Under current mode operation, the motor phase current, I , should always follow a reference current no matter what the load is. In current operation mode, the motor speed varies depending upon the load.

The speed and current mode operations are the terms in the electric motor interface that do not have specific corresponding physiologic meanings. Generally either the speed or the current mode operations will incur pressure and flow variations in the blood pump, and the variations can be very different, based on the characteristics of the pump. For the rotary blood pump under investigation, with the characteristics shown in **Figures 6 and 7**, the current mode operation is more likely to achieve stable flow. Therefore, it is preferable to use current mode operation in the physiologic controller to achieve normal physiology.

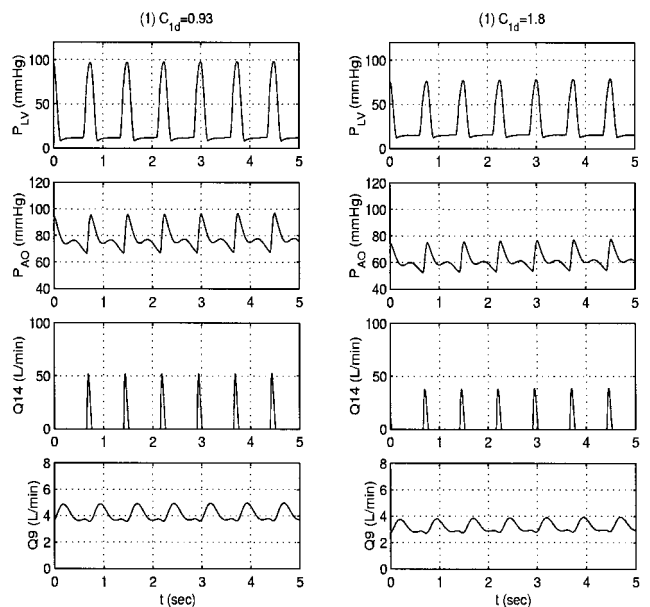


Figure 9. Pressure and flow of sick hearts. Left: $C_{id} = 0.93$; Right: $C_{id} = 1.8$.

Table 3. Aortic Pressure and Systemic Flow Corresponding to Different C_{1d}

C_{1d} (ml/mm Hg)	Aortic Pressure (mm Hg) Max/Min/Mean	Systemic Flow (L/min) Mean
0.3	115/80/94	5.0
0.63	105/80/94	4.4
0.93	97/63/77	4.0
1.2	90/58/71	3.7
1.8	78/48/60	3.1
2.8	62/37/47	2.4
4.0	50/30/38	1.8
6.0	38/22/29	1.3

Result of the Circulatory Model

Although a simple numeric method is used to solve the numeric equations, as long as the time step is sufficiently small, the results converge. The non-regulated system can be simulated by a fixed heart rate and peripheral resistance. In the simulation, when we choose $HR = 80$ beats/min, $R_9 = 1.0$ mm Hg/ml, and a time step of 5 ms, the results converge well; total blood volume was kept at 4500 ml. **Figure 8** shows the pressure and flow waveforms of a healthy heart. Left ventricular failure is modeled by increasing the diastolic compliance (C_{1d}) of the left ventricle. **Figure 9** (left) shows the pressure and flow waveforms of a sick heart, which is simulated by making $C_{1d} = 0.93$, and **Figure 9** (right) shows a sick heart when making $C_{1d} = 1.8$. **Table 3** displays the aortic pressure and systemic flow for a nonregulated model corresponding to different degrees of sickness. By using the baroreceptor feedback characteristics, the regulated closed-loop circulatory system can be simulated. **Figure 10** shows the pressure and flow waveforms of the same model as in **Figure 8**. The system was stabilized to a heart rate of 73 beats/min,

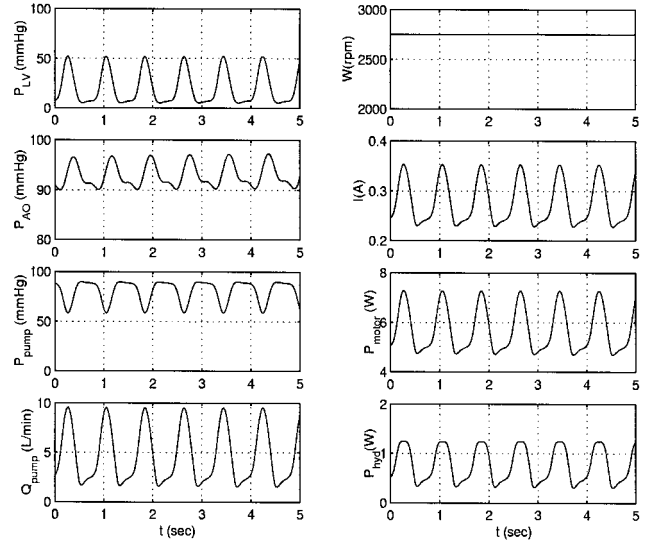


Figure 11. Physiologic and electrical parameters under speed mode operation, $w = 2700$ rpm.

peripheral resistance of 1.1 mm Hg/ml, mean aortic pressure of 93 mm Hg, and mean systemic flow of 4.6 L/min.

Hemodynamics of the circulatory system under blood pump assist are also obtained, and different operation modes are simulated. **Figure 11** shows the result of speed mode operation for a mildly sick heart. The pump speed is kept at 2700 rpm, and the pump produces an output flow, Q , of 4.73 L/min. Under blood pump assist, the vital parameters become normal. **Figure 12** shows the same situation, but the motor is in current mode operation; the motor current is kept at 0.27 A, and the

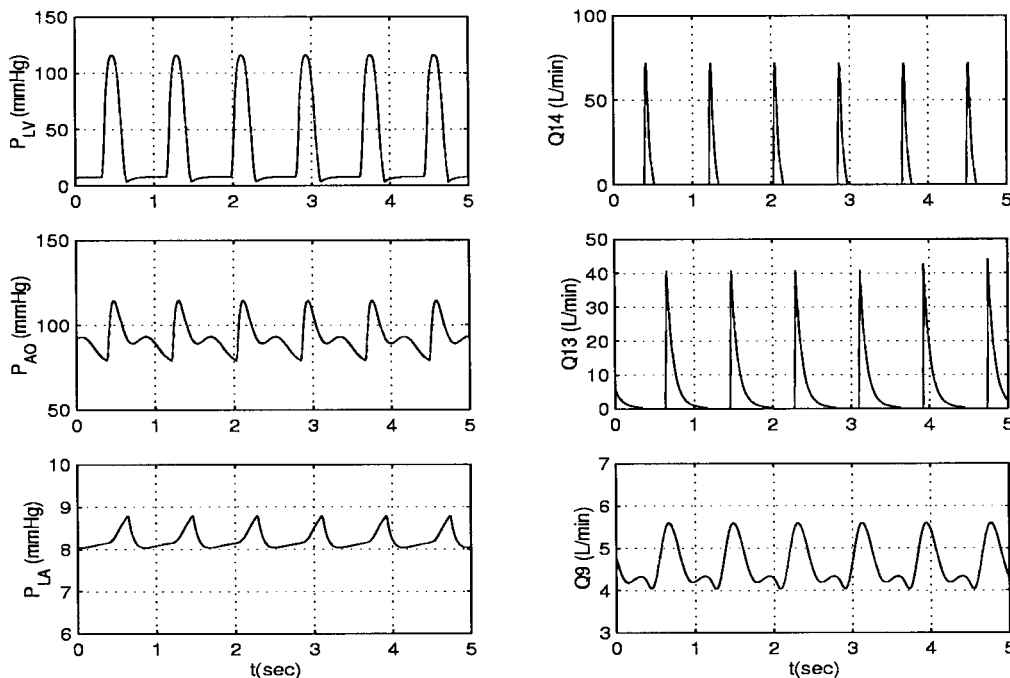


Figure 10. Pressure and flow of a healthy heart with the regulated model.

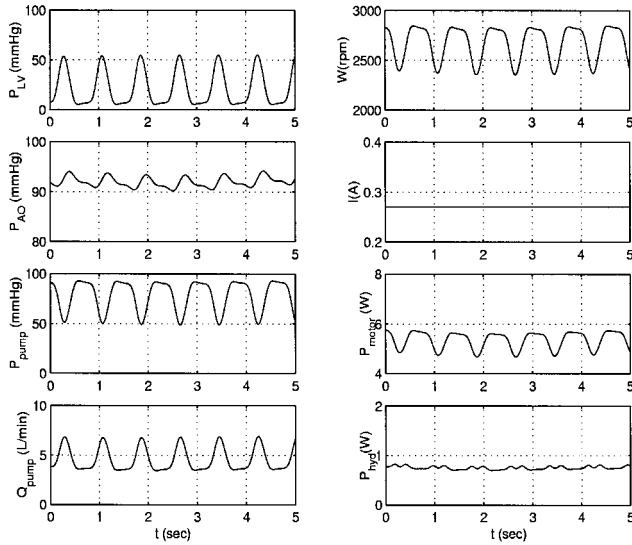


Figure 12. Physiologic and electrical parameters under current mode operation, $I = 0.27$ A.

pump produces an average output flow of 5.04 L/min. The risk of excessive draining can also be simulated. **Figure 13** displays ventricular suction due to overpumping in speed mode operation, by excessively increasing the motor speed, and **Figure 14** shows ventricular suction in current mode operation, induced by excessively increasing the motor current. From **Figures 13 and 14**, it is seen that overpumping causes the inlet pressure to drop to very low or even negative values, which causes suction, making the pump flow highly pulsatile (average is low). Selection of nonlinear resistance, R_{15} , is critical when simulating suction effects. In real life the pump flow may go to zero during suction, and the main limitation of the simulation model is the selection of R_{15} .

Conclusions and Discussions

A computer simulation model has been developed for a regulated circulatory system under LVAD assist. Interaction between

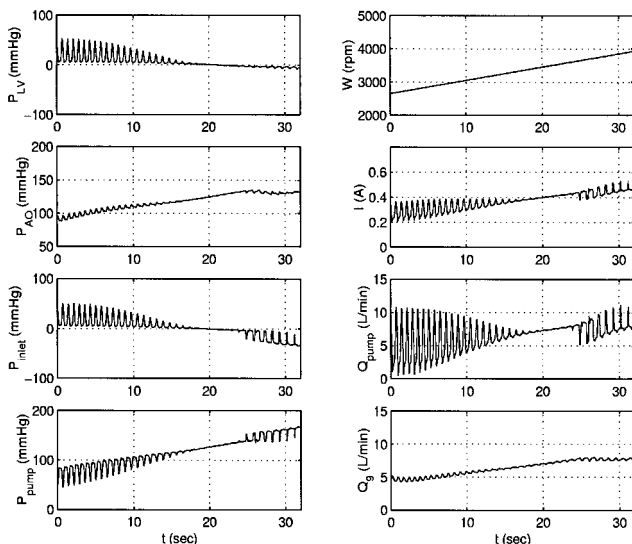


Figure 13. Ventricular suction in speed mode operation.

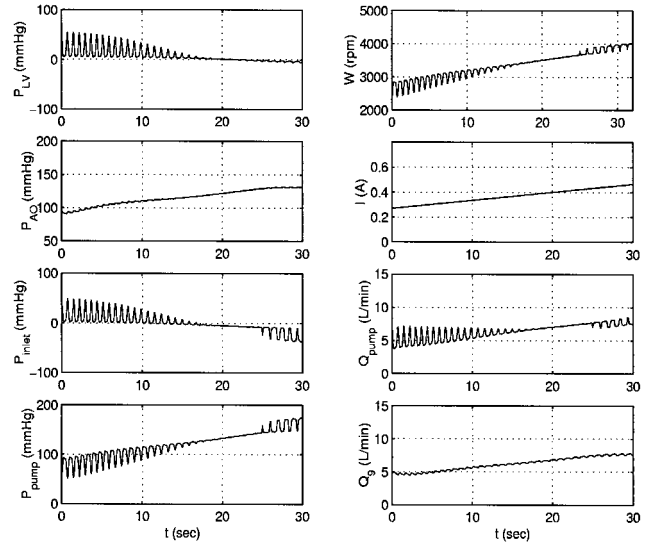


Figure 14. Ventricular suction in current mode operation.

the physiologic system and the blood pump is discussed. Simulation results show that output of the simulation model reasonably reflects the hemodynamics. The model can also describe inappropriate operations, such as ventricular suction.

1. The electric circuit model gives reasonable vital parameters of the system. Results indicate that the approach is convenient and feasible.
2. Modeling of the pump and motor is also discussed. Experimental data and certain simplification can be undertaken due to the slow process of the physiologic system.
3. The model can also simulate ventricular suction caused by overpumping. It thus provides information for researchers searching for approaches to prevent such faulty conditions.
4. By computer simulation, the pump-physiology interaction can be easily modeled. System dynamics can be simulated that expose information on the relationship between the motor variables (speed and current) and pump variables (flow and pressure). These relationships lead to approaches to sensorless diagnosis and control of the blood pump, only using motor variables, which will be discussed in a different article.
5. By using this model, sophisticated intelligent control algorithms can be developed and evaluated to ensure optimal performance.
6. The results obtained indicate that the proposed model provides a convenient and effective tool for development of real time controllers for rotary blood pumps. Thus, control algorithms can first be implemented in the simulation model, and performance can be evaluated before going to costly and time-consuming *in vitro* and *in vivo* tests.

Parameters of the Computer Model

Parameters used in this paper are listed in Tables 1 and 2.

References

1. Smith WA, Golding LAR: Progress on the CCF rotary pump. *Proc Cardiovasc Science Technol Conf*, 1991, p. 226.
2. Butler KC, Maher TR, Borovetz HS, *et al*: Development of an axial flow blood pump LVAS. *ASAIO J* 38:M296–M300, 1992.
3. Golding LAR, Crouch RD, Stewart RW, *et al*: Postcardiotomy centrifugal mechanical ventricular support. *Ann Thorac Surg* 54: 1059–1064, 1992.
4. Moeller D, Popovic D, Thiele G: *Modelling, Simulation and Parameter Estimation of the Human Cardiovascular System*. Braunschweig, Vieweg, 1983.
5. Mitsui N, Fukunaga S, Koura Y, *et al*: Numerical simulation of non-pulsatile left ventricular bypass. *Artif Organs* 20:632–636, 1996.
6. Schima H, Honigschnabel J, Trubel W, Thoma H: Computer simulation of the circulatory system during support with a rotary blood pump. *Trans ASAIO* 36: M252–M254, 1990.
7. Medvedev A: IVAS system modelling: Development of the basic performance prediction model. *IVAS Report*, 1998.