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**DEVELOPMENT OF A DYNAMIC MODEL OF A VENTRICULAR ASSIST
DEVICE FOR INVESTIGATION OF CONTROL SYSTEMS**

by

SEAN TAN

B.Eng. (Aerospace Engineering)

A thesis submitted to
the Faculty of Graduate Studies and Research
in partial fulfillment of
the requirements for the degree of

**Master of Engineering
in Aerospace Engineering**

Ottawa-Carleton Institute for Mechanical and Aerospace Engineering

Department of Mechanical and Aerospace Engineering

Carleton University

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**DEVELOPMENT OF A DYNAMIC MODEL OF A VENTRICULAR ASSIST
DEVICE FOR INVESTIGATION OF CONTROL SYSTEMS**

submitted by

Sean Tan, B.Eng. (Aerospace)

in partial fulfilment of the requirements for
the degree of Master of Engineering.

Chairman.

Department of Mechanical and Aerospace Engineering

Thesis Co-Supervisor

Thesis Co-Supervisor

Carleton University

September 1996

ABSTRACT

The design of a ventricular assist device (VAD) is a complex process. The design evaluation often requires a working prototype for *in vitro* (using circulatory loop) and *in vivo* (using lab animals) testing. To improve the evaluation process, a generic VAD-circulatory system simulation model has been developed. The advantages of using this model over the conventional method are that it has the ability to: 1) perform dynamic parameter evaluation, 2) evaluate control strategy efficiently and inexpensively, 3) to produce a "computer simulated" prototype and 4) to accommodate modifications for future studies.

The simulation model developed uses the bond-graph modeling technique. This technique consists of subsystems, composed of compliance, resistance and inertance elements, linked by power bonds. The VAD-circulatory system model derived from the bond-graph is simulated by using Matlab.

The following conclusions and observations were made from an evaluation of the performance of the VAD-circulatory system model: 1) Within the normal allowable systemic parameter limits, the fixed rate controller is found to be more tolerant to changes than the full-fill/full-eject controller, 2) The full-fill/full-eject control mode is more sensitive to control parameter changes than the fixed rate mode, 3) The full-fill/full-eject control mode tends to adjust itself to the working environment by changing its heart rate and pressure, and 4) The length of the cannulae does not have a great impact on the dynamics of the circulatory system.

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LIST OF SYMBOLS AND ACRONYMS

A	= piston area
bpm	= beat per minute
C	= compliance
C_{at}	= arterial compliance
C_{bl}	= blood compressibility
C_{ht}	= compliance of VAD blood chamber
CVD	= Cardiovascular Devices Division of Ottawa University Heart Institute
C_{vn}	= venous compliance
E	= pressure source
FFF	= full-fill/full-eject control mode
I	= inertance
I_{enot}	= inertance of output cannulae
I_{enin}	= inertance of input cannulae
I_{ht}	= VAD diaphragm inertia
I_{pe}	= inertance of peripheral circulation
P	= pressure
P_t	= additional input force for driving the VAD
P_{tb}	= base force for driving the VAD
P_r	= remnance beating pressure
Q	= flow rate
R	= resistance
R_{ht}	= VAD diaphragm resistance
R_{in}	= input valve resistance
R_{ot}	= output valve resistance
R_{pe}	= resistance to peripheral circulation
TAH	= total artificial heart
t	= time in second
V	= volume
VAD	= ventricular assist device
w	= remnance beat rate

STATE VARIABLES

p_2 = pump piston momentum
 q_6 = heart chamber blood volume
 q_{10} = output cannulae blood volume
 p_{12} = output cannulae momentum
 q_{14} = arterial blood volume
 p_{17} = peripheral momentum
 q_{19} = venous blood volume
 p_{21} = input cannulae momentum
 q_{23} = input cannulae volume

1.0 INTRODUCTION

Chronic heart disease is the leading cause of death in most of the developing countries. The disease is especially lethal if heart failure occurs. Currently, there are no effective widely used treatments for patients dying from heart disease. Each of the ongoing studies of alternative treatments, including heart transplantation, have major clinical limitations. They are often costly and yield poor quality of life for the patient. One of the solutions for end-stage heart disease is the use of circulatory assist device or artificial heart. Clinical experiments and trials have proven the artificial heart technology to be a viable solution (Mussivand, 1996).

Two major types of circulatory assist devices are the total artificial heart (TAH), which replaces the heart and the ventricular assist device (VAD), which is designed to aid the ventricles as they pump blood through the circulatory system. Although both devices essentially pump or circulate blood, it has been clinically demonstrated that VADs can effectively treat about 2/3 of patients requiring artificial hearts.

The VAD is also found to be a more feasible device for development. The reasons are:

1. In most cases, especially in elderly patients, it is the left ventricle that fails first due to heavily worked heart muscle and wear in the chamber. By using a VAD to assist circulation, the implantation procedure is simplified when compared to that for the TAH.
2. The device may be more cost effective due to its simpler design.
3. There may be psychological benefits for the patient since the natural heart is still present.

To increase understanding of the interaction between the VAD and the circulatory system and to reduce the experimental costs for *in vivo* testing of the device, a mathematical model with an accompanying simulator is needed.

With the aim of providing a better tool for developing the Left Ventricular Assist Device (LVAD), a model which simulates the interaction between the VAD and the circulatory system is developed for parameter determination and design testing. The final VAD-circulatory simulation model will provide a fast and easy way for determining the required parameter values (compliance, resistance and inertance), to examine the interaction of the VAD and the rest of the circulatory system, and to evaluate various design modifications. This model has the following advantages over the existing development tools which involve using a static model and *in vitro* and *in vivo* testing since the model has the ability:

1. to determine control parameters dynamically,
2. to produce a "computer simulated" prototype efficiently,
3. to evaluate control strategy efficiently,
4. to model different devices without major modification, and
4. to accommodate modifications for future studies.

In addition to the advantages listed, the simulation model developed is a generic model made up of basic compliance (C), inertance (I) and resistance (R). Such a model can be used to determine the design parameters of a wide variety of VAD designs under consideration. Together with the chosen control method, this model will allow for quick prototyping of a design.

1.1 Cannulae Effects Study

The cannulae are the low porosity woven Dacron tubes that connect the VAD to the left ventricle and the VAD to the aorta for conducting blood into and out of the device. The effects of both the length and the compliance of the input and output cannulae has an effect on the dynamic performance of the VAD as both parameters affect the blood pressure and blood flow. Since the compliance of the cannulae is fixed due to its material characteristics, studying the effects of cannulae length will help the developers and surgeons determine its optimal geometric length without taking into account the anatomical fit, hydraulic losses and thrombogenesis.

1.2 Dynamic Determination Of Controller Parameters

To be able to determine the controller parameters (parameters that vary with the design) dynamically is both beneficial and important. This is because the effect of each parameter on the system can be observed. Widely used methods for parameter determination include static mathematical models and mock circulation studies for testing the performance of the prototype. These methods can provide parameter values as the starting point of the design. However, in a real life situation, the parameters are interacting dynamically and how the parameters will affect the system cannot be predicted by using a static model. Thus, a dynamic simulation model will be a more efficient tool for determining the desired parameter values for the design and for examining the effects of each parameter.

1.3 Evaluation Of Control Method

There are two types of control method under consideration in this thesis. The first method is the fixed rate mode and the second is the full-fill/full-eject (FFFE) mode. The fixed rate is easy to design and to control. The VAD is set to beat at a certain rate, which may not be meet the physiological need of the recipients. The full-fill/full-eject mode overcomes this problem by switching systole and diastole based on a filling and ejecting condition, and the switching point is determined by the preset VAD volume limits. The reference VAD used for this model employed a sensor for determining the device volume. The full-fill/full-eject mode allows the VAD to mimic the real ventricle more realistically, but the dynamics and control schemes are far more complicated than in the fixed rate mode. Although both methods can be adapted to a design, it would be expensive and complicated to produce different sets of live subject experiments to evaluate each method as opposed to using a computer simulated model. The developed dynamic model of the VAD and circulation system allows the developer to switch between the two control modes easily and quickly to examine how each method will affect VAD operation.

1.4 Residual Beat Synchronization

The model can also be modified to examine the effects of residual beating of the artery (Windkesel effect) for possible synchronization of the VAD to the residual pulse. This effect is quite difficult to evaluate using the existing mock circulation loops since residual pulse and pressure waves cannot be readily introduced into the system. With a computer model, a combination of pressure amplitude and oscillation frequency can be added to the arterial pressure to simulate any residual beating for a synchronization study.

1.5 The Ventricular Assist Device - Circulatory System Model

To make the computer model more general and more applicable to various designs, the system simulation is built by using basic resistive, compliance and inertance elements. The system is modeled based on the bond-graph method (Karnopp et al, 1990). The model can be expanded or reduced easily as required by the complexity of the analysis. This method ensures that energy is conserved from element to element. The bond-graph method is also powerful in the way that it can reduce the model directly to first order differential state-space equations. This helps minimize the complexity of reducing higher order differential equations into more manageable first order equations when employing conventional mathematical modeling methods. The bond-graph method can also provide a visual model that summarizes energy and information flows in the system.

With the base mathematical model derived using the bond-graph technique, the VAD-circulatory model is then simulated by using Matlab. The Matlab program written for this purpose is modular, which will allow fast and easy modification. This simulation program produces system pressure, flow rate and volume plots. It also calculates the input power and system efficiency which can be used to evaluate design feasibility.

This thesis describes the model that had been built by using the bond-graph method to simulate the VAD and the circulatory system. It presents the parameter and control method studies performed for evaluating the

performance of a VAD design under consideration. This thesis will describe the limitations of the model and the working ranges of critical parameters that make up the entire system simulation. A discussion is included to provide recommendations for future improvements and uses of the developed model.

1.6 Model Assumptions

For developing the VAD-circulatory system model, there are a few assumptions made to make the model easier to handle:

1. There is no distributed friction in the blood flow.
2. the model is developed based on a healthy human body, and
3. the pulmonary circulation (lung circulation) is not taken into account.

To further make the development of the model possible, the model uses lumped parameters such that the aorta and arteries are treated as one artery, the circulation through the internal organs are treated as the peripheral flow and the vessels returning blood to the heart and the right chambers of the heart are treated as one vein. To improve the details and accuracy, distributed parameters can be used, but this will add complexity to the model and to the analysis required for evaluation of the system.

1.7 The Scope

The scope of this thesis is to develop a VAD-circulatory system model for dynamic determination of design parameter of a VAD and to allow performing simulation on the system performance coupled with different control methods.

2.0 SURVEY OF PREVIOUS WORK ON VENTRICULAR ASSIST DEVICES AND TOTAL ARTIFICIAL HEART

This chapter summarizes the work of many of the researchers around the world who are actively developing circulatory assist devices either as VADs or TAHs. Regardless of the type of design being considered, the goal is to build a device that would take the place of the damaged heart in an attempt to extend the patient's life with minimal sacrifice to the quality of the patient's life. Since a development of a research program for building a circulatory assist is very expensive, most of the research is done jointly between a major research center and associated industrial partners or sponsors. A list of the major research centers and their area of concentration is shown in table 2.1.

This chapter discusses current VADs and TAHs under development around the world. Information relating to previously developed devices is also included in this chapter as a reference on the progress of research in this field.

RESEARCH CENTER	INDUSTRIAL PARTNER/SPONSOR	TYPE OF DEVICE DEVELOPED
Texas Heart Institute, Dallas, TX	Abiomed, Danbury, MA	VAD, TAH
Penn State University, Hershy, PA	3M/Sarns, Ann Arbor, MI, Thoratec	VAD, TAH
Cleveland Clinic Foundation, Cleveland, OH	Nimbus Inc, Rancho Cordova, CA	VAD, TAH
University Of Utah, Salt Lake City, UT	None	TAH
Jarvik Research Institution	Jarvik Research Institute, NYC, NY	VAD
Milwaukee Heart Institute, Milwaukee, WI	Bradley Foundation, Milwaukee, WI	TAH
Texas Heart Institute, Dallas, TX	Aerospatiale and Sagem, Paris, France	TAH
Baylor College of Medicine, Houston, TX	NASA	VAD, TAH
Free University of Berlin, Berlin, Germany	Mercedes-Benz Inc., Stuttgart, Germany	VAD, TAH
University of Tokyo, Tokyo, Japan	Unknown	VAD, TAH
National Cardiovascular Research Center, Osaka, Japan	Government of Japan	VAD, TAH
Gifu University, Gifu, Japan	Unknown	TAH
University of Kyoto, Kyoto, Japan	Unknown	VAD, TAH
Hokkaido University, Sapporo, Japan	Unknown	TAH
Innsbruk University Hospital Austria	Unknown	VAD
University of Aix-Marseille II France	Unknown	TAH
Bratislavske Lekarke Listy Czechoslovakia	Unknown	TAH
Institute for Transplantation & Artificial Organs Moscow, Russia	Government of Russia	VAD, TAH
University of Vienna Vienna, Austria	Unknown	VAD, TAH
Seoul National University Seoul, Korea	Unknown	VAD
Shanghai Medical Institute Shanghai, China	Unknown	TAH
National University of Taiwan Taipei, Taiwan	Unknown	TAH
CVD of Ottawa Heart Institute, Ottawa, ON	CAE, WHC, Gouvernement of Canada	VAD

Table 2.1 Research Centers Actively Involved in VAD or TAH Development

2.1 Introduction And History Of VAD Development

In the late 1800s and the early 1900s, researchers were already searching an device to replace the natural heart. The idea at that time was to produce a pump that will circulate the blood inside a human body, but due to the lack of knowledge and lack of technology, the ideas never went beyond concept and basic models. It was not until December 1957 that Dr. W. J. Kolff and Dr. Tetsuzo Akutsu implanted the first artificial heart in a dog and maintained its circulation for 90 minutes. That particular heart was made of polyvinyl chloride and its control scheme was simply based on Starling's Law. This means that if the venous pressure or atrial pressure rises, the cardiac output would be automatically increased (Kolff, 1990).

The next step in the development of the artificial heart was the implantation of the first artificial human heart by Denton Cooley. This artificial heart remained functioning for three days. Then in 1975, a left ventricular assisted device (LVAD) was tested with *in vivo* experiments (Kolff, 1990).

The next step in the history of artificial hearts came about in 1979 when a Jarvik-7 artificial heart successfully kept a calf alive for 221 days. To further prove the promising nature of the artificial heart, Dr. William DeVries implanted the Jarvik-7 as a permanent heart in Dr. Barney Clark in 1982. With careful treatment and intense monitoring by the research team and medical staff, Dr. Barney Clark lived for 112 days and finally died in 1983. This implant was enough to show that the artificial heart will no doubt be an alternative to natural heart transplant for the terminally ill heart diseased patient (Kolff, 1990).

Today, there are many research teams around the world developing various types of artificial heart with different control philosophies and basic designs. Artificial hearts can be divided into two major categories: the ventricular assist devices and the total artificial hearts.

The ventricular assist device (VAD) is designed to assist the natural ventricle in blood circulation. For most people, as one ages, it is usually the ventricles and especially the left ventricle that first fails. This is because the left

ventricle is the one chamber that works the hardest by pumping blood to the rest of the body. The idea of a VAD is that it can use it as an aid to the left ventricle for pumping blood without physically removing the entire heart, which is otherwise still working normally. Although the VAD is sufficient for most patients, there are still a large number of people who require the entire heart to be replaced and thus it is also necessary to develop a device that can replace the natural heart.

The total artificial heart (TAH) is a device that is intended to be used as a replacement for the diseased heart as either a bridge to heart transplant or as a permanent solution. When using a TAH, the natural heart must first be removed. This would often be necessary since the heart being removed is usually not functioning. Presently, most TAHs, such as the Jarvik-7, are being used as bridges to heart transplants. It is the goal of research teams to make the TAH a permanent alternative to heart transplant because the number of donated hearts is only a fraction of the number required to save the thousands of people who suffer fatal heart disease.

2.2 Ventricular Assist Device

A ventricular assist device, as mentioned before, is a device which helps the weakened natural ventricle to circulate blood (for a basic understanding of heart physiology, refer to chapter 3 of this thesis). It can be used to support either the left ventricle or the right ventricle, or both ventricles. In left ventricular assist, the input cannulae is inserted into either the left atrium or left ventricle depending on the design. A separate output cannulae is inserted into the ascending aorta or abdominal aorta to maintain the blood flow needed by the human body and maintain normal physiological conditions. For most right ventricular assist, the process is similar except the input cannulae is inserted into the right atrium and the output cannulae is inserted into the pulmonary artery to assist pulmonary circulation (Koshal et al, 1991). In biventricular assist, the combination of the two techniques is used. The techniques for cannulation are shown in figure 2.1 (Vaska , 1991).

Figure 2.1 LVAD Cannulation Technique (Vaska, 1991)

There are two major designs of VAD, and due to the differences in design philosophy, most research institutes only concentrate on one design. The first design is the continuous flow pump and the second is the pulsatile pump.

The continuous flow pump designs include roller pumps and centrifugal pumps. This type of design is called a continuous flow pump because a nonpulsatile, continuous flow is provided by the device. The first type of continuous flow pump is the roller pump. This pump uses rollers to compress blood-filled tubes and force blood forward into the aorta. This pump type is the most commonly used device for temporary support in most open chest surgeries. This pump is relatively inexpensive when compared to other designs. The disadvantages of this device are hemolysis (the escape of hemoglobin from the interior of the red blood cell into the surrounding medium caused by the disruption of the integrity of the cell membrane) caused by compression of red blood cells and thrombosis (the formation of a clot in an unbroken blood vessel) caused by damaged platelets which can cause occlusion

of blood tubes (Vaska, 1991). Another disadvantage of the roller pump is the absence of pulsatile flow.

The second type of continuous pump is the centrifugal pump. This pump uses a turbine like head that is coupled magnetically to an electric controller and uses centrifugal forces created by a flow vortex to propel blood. The advantage of such a pump is that flow output is limited when the input blood pressure is low allowing simpler control of cardiac output. The disadvantages of the device include platelet and red blood trauma, and the absence of pulsatile flow as provided by the natural heart (Vaska, 1991).

The second VAD design type is the pulsatile pump. This type of pump provides a periodic pulsatile flow just like the normal heart does. The major advantage of this kind of pump design is its pulsatile nature. Since it provides the same type of flow as the natural heart, it may be more physiologically compatible to the natural human body than the centrifugal pump. The disadvantage of such a pump is its high development cost and complexity in design.

Depending on the requirements and design, the pulsatile VAD can be controlled in two major different modes. The fixed rate mode causes the VAD to operate at a preset rate independent of the natural heart rate beating. The second method is the full-fill/full-eject mode (considered as the synchronous mode). This control mode is dependent on the filling and emptying rate of the VAD which in turn depends on the natural heart. VADs of this design will tend to follow the natural heart rate depending on the preset volume limit. Other synchronous modes may use the electrocardiogram (ECG) or some other signal as a trigger to synchronize the pump with the cardiac cycle.

2.2.1 Continuous Flow VAD

Although the continuous flow VAD development is a major interest of many institutions, the VAD-circulatory model in this thesis is focused on pulsatile VAD. Since the simulation model can be modified to accommodate continuous flow VADs, they should be investigated as part of the future research.

As mentioned, there are two types of continuous flow circulatory assist devices. Of the two, the centrifugal pump VAD is the more noticeable one since it is being developed as a permanent heart disease solution rather than as a temporary aid during operations.

The development of centrifugal pumps dates back to the mid-sixties. An example of this type of VAD is the Biomedicus Biopump (Vaska, 1991) and it is shown in figure 2.2. This pump consists of a set of four stacked cones mounted on a spindle shaft supported on a bearing. The centrifugal pump head is magnetically coupled to an electric motor controller for creating the desired vortex flow. The blood would enter the device at the tip of the cone. With the cones spinning, the blood is being pumped out of the device at the large diameter end of the cones.

Figure 2.2 Biomedicus Centrifugal Flow Pump (Vaska, 1991)

Another centrifugal pump is the Baylor Gyro Pump developed at the Baylor College of Medicine in Texas (Ohara et al., 1993). This pump was designed

based on a gyroscopic principle that supports and stabilizes the rotation of the rotor impeller. Driving power was provided by a brushless DC motor. The aim of the pump design was to support both the left and right ventricles. To meet the demand requirements, two prototypes are being made with one of them incorporating a stronger motor. The operating conditions to be met for right ventricular support are blood flow rate of 4 L/min and output pressure of 60 mmHg at pump rotation of 1800 rpm. For left ventricular assistance, the pump should provide a flow of 6 L/min with a blood pressure of 240 mmHg.

Figure 2.3 Baylor Gyro Pump (Ohara et al., 1993)

After the testing, the two pumps were found to have satisfactory performance. Although the performance was acceptable, both pumps had pump efficiencies less than 6%, which is too low. Improvements are being performed on the pumps by optimizing the gap size of the stator and the rotor of the DC motor.

From the same institution, another pump, the Baylor-Nikkiso centrifugal pump is being developed jointly with Nikkiso Co. Ltd. in Japan (Nanto et al. 1993). This pump is very similar to the ones previously mentioned and the design schematic is shown in figure 2.4. For *in vitro* testing, the pump is to maintain the following conditions:

Revolution speed.	2100 rpm
Inlet pressure :	0 - 5 mmHg
Outlet pressure.	100 mmHg
Flow rate (water)	6 L/min

The design has proven to be feasible and following the *in vitro* testing, *in vivo* testing was also performed. This pump was capable of 2 weeks of continuous operation and was determined by Baylor College to be suitable for long term usage.

Figure 2.4 Baylor-Nikkiso Centrifugal Pump (Nanto et al. 1993)

Another centrifugal design is the device developed by Kijima et al.(1993) at a Research and Development Center in Japan. The goal of the design was to develop a "straight path" centrifugal pump (Capiox centrifugal pump as shown in figure 2.5) which requires the lowest pump speed to produce the same blood flow rate of 5 to 8 L/min under the same pressure loads. This is done in an attempt to overcome the hemolysis and temperature rise problem associated with centrifugal pumps. After the testing was done, the pump was proven to deliver the promised solution while maintaining the required flow rate.

Figure 2.5 Structure Of Capiox Centrifugal Pump (Kijima et al, 1993)

a disadvantage of centrifugal pumps is their inability to deliver pulsatile flow. To overcome this problem, Qian (1990) at Shanghai Second Medical University in China has devised two different model solutions. The first solution is to change pump rotational speed periodically. From *in vitro* model evaluations, the idea proved to be acceptable. The second model produces pulsatile flow by varying the gaps between impeller and cap on the inlet pipe. This is achieved by oscillating the constant speed rotating impeller axially like a piston pump. A theoretical analyses has shown that this idea also works. By employing

these two ideas, the advantages of the centrifugal pumps and pulsatile pumps can be combined. These ideas are not yet being pursued in other major research centers.

2.2.2 Pulsatile Flow VAD

It is the pulsatile flow VAD that is the main interest of the Cardiovascular Devices Division of the Ottawa Heart Institute (CVD) and intended focus of the VAD-circulatory simulation model in this thesis. In general, a pulsatile flow VAD is like a piston pump. While the piston is pushing, blood is ejected from the outlet valve with the inlet valve shut tightly. When the piston direction is reversed, the outlet valve is closed to prevent backward flow and blood enters through the inlet valve to fill the chamber.

There are several methods to move the piston, and in many designs, a flexible diaphragm is used in place of the piston. One widely used design is the pneumatic actuator VAD, which depends on air pressure to drive a diaphragm inside the VAD chamber. The institutions actively involved in this type of design are University of Utah, (Smulders et al., 1995 and Kolff, 1991), National Cardiovascular Center Research Institute in Osaka, Japan (Nakatani et al., 1993) and Thoratec Laboratories jointly with California Pacific Medical Center (Farrar et al., 1993).

Figure 2.6 Soft, Compressible VAD Design By Dr. Kolff Of Utah (Kolff, 1991)

The general configuration of a pneumatically driven VAD consists of a flexible diaphragm housed inside a two valve hard shell. Power to drive the device is usually provided by an external driving unit. The Thoratec VAD shown in figure 2.7 is an example of this type of device. During systole, the air pressure increases above the normal arterial pressure, which is generally 100 mmHg, to pump blood into the circulatory system. At diastole, the air pressure drops below the lowest ventricular pressure (5 mmHg) for filling the VAD chamber. Since an external power source is being used by this type of VAD, a fixed rate controller is generally used for controlling purpose. The goal of VAD performance is generally to maintain a flow rate between 3 to 8 L/min and output pressure between 40 mmHg up to 300 mmHg depending on the experiment being performed.

Figure 2.7 The Thoratec VAD And Its External Drive Console
(Farrar et al. 1993)

Similar to the pneumatic VAD in design is the Electrohydraulic Ventricular Assist Device (EVAD) developed at the Cardiovascular Devices Division of the University of Ottawa Heart Institute (OHI) (Mussivand et al., 1996) as shown in figure 2.8. This VAD uses a diaphragm to pump the blood. Instead of using air pressure with an external unit to drive the device, the driving force is provided by pressured silicon oil. During the systolic phase, oil is pumped from a reservoir into the VAD chamber by a high speed axial flow pump driven by a brushless DC motor. The turbine then reverses its direction during diastole. This VAD also differs from a pneumatic VAD in that it is a unified unit which does not need an external driving unit. The oil reservoir is built on the back of the device and connected to the VAD chamber by a conduit which contains the axial flow pump. This unit has a stroke volume of 70 ml and can operate in fixed rate mode and in full-fill/full-eject mode. Under normal operating condition, the EVAD can maintain a flow rate up to 7 to 11 L/min against an output pressure of 100 mmHg with an input pressure of 5 to 15 mmHg.

Figure 2.8 University of Ottawa Heart Institute EVAD
(Mussivand et al. 1996)

Similar in design is the Pusher Plate VAD developed by Penn State University (Tsach et al., 1989) (Klute et al., 1992) as shown in figure 2.9. The Penn State VAD uses a rollerscrew, with a pusher plate fixed at the end, to pump blood. By using a mechanical drive system, the VAD does not need an external driving unit. One of the major advantages of using the rollerscrew driver is its adaptation to full-fill/full-eject controller. Since the drive shaft is rigid, the position of the pusher plate can be easily determine, thus the VAD chamber volume can be determined also. Through various experiments, this device demonstrated satisfactory performance by maintaining cardiac output of about 6 L/min and output pressure between 88 and 107 mmHg. The peak electric energy required for the device varies from 8.77 joules to 53.54 joules, but the efficiency is only between 9% and 12%.

Figure 2.9 Penn State Pusher Plate EVAD (Klute et al., 1992)

There are two major reasons why this unit is particularly relevant to this thesis. The first reason is that it also uses a bond-graph model of the design and the second reason is this is one of the few research teams that publishes all of their design parameter values.

Symbol	Description	Value	Unit
A_p	Pusher plate area	4260	$\{\text{mm}^2\}$
C_{ao}	Aortic compliance	1330	$\{\text{mm}^3/\text{mmHg}\}$
C_{ei}	Atrial and venous compliance	4850	$\{\text{mm}^3/\text{mmHg}\}$
f	Motor efficiency	0.8	$\{\text{dimensionless}\}$
I_{ao}	Aortic inertance	2.35×10^{-6}	$\{\text{kg}/\text{mm}^4\}$
J	Armature moment of inertia	1000	$\{\text{kg}\cdot\text{mm}^2\}$
K_e	Back emf constant	0.54	$\{\text{volt}\cdot\text{s}/\text{rad}\}$
K_t	Motor torque constant	540000	$\{\text{kg}\cdot\text{mm}^2/(\text{amp}\cdot\text{s}^2)\}$
K_c	Rotary to linear conversion factor	6.784	$\{\text{mm}/\text{rad}\}$
K_{pf}	Pressure conversion factor	0.133	$\{\text{kg}/(\text{mmHg}\cdot\text{mm}\cdot\text{s}^2)\}$
M_p	Mass of pusher plate	0.115	$\{\text{kg}\}$
R_m	Motor armature resistance	4.9	$\{\text{ohms}\}$
R_{av}	Aortic valve resistance	0.00005	$\{\text{mmHg}\cdot\text{s}/\text{mm}^3\}$
R_{iv}	Mitral valve resistance	0.000024	$\{\text{mmHg}\cdot\text{s}/\text{mm}^3\}$
R_{cap}	Systemic resistance	0.007-0.00136	$\{\text{mmHg}\cdot\text{s}/\text{mm}^3\}$

NOTE: R_{cap} is equal to 0.001 $\{\text{mmHg}\cdot\text{s}/\text{mm}^3\}$ for calculating optimal control input

Figure 2.10 Numerical Values Of Model Parameters For Penn State EVAD
(Tsach et al., 1989)

An important feature of the Penn State bond-graph model is its discontinuity between systole and diastole. That is, the Penn State model requires two separate models for the two phases of a heart cycle. From examining the model, an explanation of the usage of two separate models by Penn State University is that during diastole, there is no work being done on the blood. This means the VAD and the rest of the circulatory system are energetically decoupled at this point creating a discontinuity in the model. The Penn State model served its purpose by allowing the research team to develop a functional unit.

2.3 Total Artificial Heart

Developed as a solution to end stage heart disease, the TAH is designed to replace the natural heart rather than to assist the weakened heart. The TAH is still being developed to meet the generally accepted standard performance rating of:

Output flow = up to 15 L/min

Average output pressure = 100 mmHg

Average pulmonary pressure = 5 - 25 mmHg

Modern development begins with the Soft-Shell Mushroom Heart design by W. J. Kolff in 1968 (Kolff et al, 1991). This is a pneumatically drive device which depended on a supply of pressurized gas for pumping blood. When the manufacturing was taken over by Dr. Clifford Kwan-Gett, this device was further modified to become the Jarvik Total Artificial Heart and was implanted in Dr. Barney Clark in December 1982.

Figure 2.11 Sketch Of Soft-Shell Mushroom Heart (Kolff et al, 1991)

Presently, TAH development is being carried on all over the world. Most of these TAHs are designed to be pulsatile except for the Minnesota Heart centrifugal pump. A variation of the centrifugal TAH is the Flow Transformed Pulsatile TAH developed by University of Tokyo (Imachi et al, 1992). This TAH uses a centrifugal pump as the driving unit. To achieve pulsatile flow, two three-way valves are added to the design. With much effort, the final device can produce maximum output flow off 7.8 L/min while maintaining an output pressure of 100 mmHg.

Figure 2.12 Flow Transformed Pulsatile TAH (Imachi et al. 1992)

Another TAH that is based on a centrifugal pump to achieve pulsatile flow is built by Abiomed Inc. in Danvers Massachusetts (Yu et al. 1993). By using a rotary valve, the flow generated by a miniature centrifugal pump (50% efficiency) is changed directionally for alternate left and right ventricular filling

Figure 2.13 The ABIOMED TAH (Yu et al. 1993)

Pulsatile TAHs can be generally divided into two different categories depending on the driving mechanism used. The first category is the

electromechanically powered TAH and the second is the pneumatically powered TAH. In general, electromechanically powered TAHs rely on either pusher plates or rigid moving actuators. TAHs of this design are usually small enough to be made of one piece without other external driving source, such as a pressurized air pump, except for the electrical power source.

The Milwaukee Heart is an example of pusher plate heart that uses a brushless DC motor to drive the pusher plate through a lever arm (Gao et al. 1992). To reduce energy demand and prolong device life, the motor of this device is designed to operate unidirectionally with the help of the drive link and lever arm. The physical characteristics of this TAH are as following:

Total volume	= 490 ml
Total weight	= 600 ml (nominal)
Actuator weight	= 280 g
Stroke volume	= 60 - 65 ml
Heart Output	= 4 - 10 L/min
Power requirement	= 1.2 W/L
Overall efficiency	= 20%

Figure 2.14 The Milwaukee Heart (Gao et al. 1992)

Besides developing a VAD, the Pennsylvania State University has also developed a Dual-Pusherplate Pump TAH (Snyder et al, 1992, Reibson et al, 1993). This TAH is a natural extension of the pusher plate VAD. Instead on one blood chamber with one pusher plate, this device uses two of each. The alternate filling of the right and left chambers is achieved by reversing the rollerscrew rotation.

Figure 2.15 The Penn State TAH (Snyder et al, 1992)

Similar in design to the Penn State TAH is the electromechanical TAH, as shown in figure 2.12, built by Baylor College of Medicine (Orime et al, 1993, Orime et al, 1992, and Takatani et al, 1992). This TAH also uses pusher plate, controlled by a roller screw, for alternate right and left chamber filling. It has undergone a four month accelerated endurance test and is demonstrated to have long-term durability.

Figure 2.16 The Baylor College TAH (Orime et al., 1993)

Two TAHs, with different driving designs from the previously mentioned ones, are being developed by the Helmholtz-Institute for Biomedical Engineering at the RWTH, Aachen, Germany (Kautmann et al., 1992), as shown in figure 2.17, and by Seoul National University in Korea (Kim et al. 1992, Min et al. 1993, and Min et al. 1991), as shown in figure 2.18. Both of these TAHs use rigid actuators to pressurize the blood sac for pumping blood. The main difference between the two TAHs is that the German TAH uses two semi-circular actuators, one on each side of the device, whereas the Koreans use a single circular actuator, moving like a pendulum, to achieve left and right filling.

10000 - Schematic drawings of the artificial heart: 1) rigid housing, 2) sewing cuff, 3) flat disc valve, 4) diaphragm, 5) pusher plate, 6) displacement axis, 7) pump chamber, 8) motor, 9) gear unit, 10) electronic control unit

Figure 2.17 The German TAH (Kaufmann et al, 1992)

Figure 2.18 Seoul National University TAH (Kim et al, 1992)

Unlike the electromechanically powered TAHs, which have many different designs, fluid driven TAHs are either driven electrohydraulically or by using gas. With either driving medium, the basic TAH design consists a flexible diaphragm that separates the blood from the actuating fluid. The diaphragm's motion is controlled by the supplied fluid. When one side of the heart is filling, the fluid is transferred from the other side for pumping. Major research centers actively involved in developing this type of TAH are: University of Utah (Rowles et al, 1992, Tatsumi et al, 1992) as shown in figure 2.19, Nimbus Inc. jointly with Cleveland Clinic Foundation (Rintoul et al, 1993, Irie et al, 1992) as shown in figure 2.20, University of Vienna General Hospital in Vienna, Austria (Rokitansky et al, 1991) and Institute of Transplantology and Artificial Organs in Moscow, U.S.S.R. (Ivanovich et al, 1991).

Figure 2.19 University Of Utah TAH (EHTAH) (Rowles et al, 1992)

Figure 2.20 Nimbus Cleveland Clinic Total Artificial Heart
(Rintoul et al, 1993)

There are two other TAHs, being studied, that have unusual design philosophies. A TAH is being developed by the National Cardiovascular Center Research Institute in Osaka, Japan (Masuzawa et al, 1992), as shown in figure 2.21. This device is an electrohydraulic unit using silicon oil as the actuating medium and a regenerative pump (a turbine pump) as the actuator. The major different between this TAH and the other electrohydraulic TAH is that the blood pump and the actuator are placed separately in the thorax and in the abdominal region, respectively. This design is meant to overcome the size limitations determined by the size of the patient's thorax and problems with durability.

Figure 2.21 The Osaka Regenerative Pump TAH (Masuzawa et al, 1992)

The other interesting TAH is the skeletal muscle actuator TAH developed by Gifu University School of Medicine in Gifu, Japan (Sasaki et al, 1992), as shown in figure 2.22. This is a pusher plate TAH in design, but it uses skeletal muscle power as the actuator to drive the device. The claimed advantages of this system are: 1) the muscle power is transmitted effectively, 2) the system can be installed with minimal surgery and without interruption in blood flow to the muscle, 3) the muscle is free from high intraventricular pressures that cause ischemia (a lack of sufficient blood to a part due to an obstruction of circulation) and 4) the muscle can be set to the proper length in diastole in order to provide strong contractions.

Figure 2.22 Skeletal Muscle Actuator TAH (Sasaki et al. 1992)

By examining the other VADs and TAHs developed by various research centers, it is possible that all of them can be modeled by basic resistance, inertance and compliance elements. By transforming the actual unit into a model consisting of these basic elements, and with modifications to model a TAH, these devices can all be simulated by the VAD-circulation simulation model in this thesis. By doing so, new modifications to the devices can then be evaluated efficiently and relatively inexpensively when compared to modifying an actual unit.

3.0 THE PHYSICAL SYSTEM AND THE MODEL

It was important that a VAD-circulatory system model should be a good representation of the natural system. Within a certain error range, the model should produce all the expected responses. In order to have an accurate model, the physical circulatory system and normal physiology must be understood.

This chapter provides a brief description of the human circulatory system. The mathematical basis for the simulation model is also included as part of the discussion.

3.1 The Blood Vessels Of The Circulatory System

The human cardiovascular system is made up of blood vessels of various sizes. Each type of vessel has its own structure depending on its function and location. The major peripheral blood vessels leading out from the left ventricle are: aorta, artery, arteriole, capillary, venule, vein and vena cava. Of these major vessels, the ones which lead oxygenated blood out of the heart to the rest of the body are known in general as the arteries. The ones which returned the deoxygenated blood back into the heart are known as the veins. A more detailed diagram of location and the name of each blood vessel is shown in figure 3.1 (Tortora et al, 1984).

Figure 3.1 The Human Circulatory System (Tortora et al. 1984)

3.1.1 The Aorta And The Arteries

The arteries are the main vessel branches that carry the blood away from the heart to various parts of the body. They have three layers, which are known as tunics, and a hollow core. The inner layer was made up of simple epithelial cells which are in contact with blood. The middle layer (tunica media), consisting of elastic fibers and smooth muscle, is usually the thickest layer and is mainly responsible for the elasticity and contractility of the vessels. The outer protective coat, the tunica externa, is composed of elastic and collagenous fibers (Tortora et al, 1984).

The elastic and contractile properties of the arteries are responsible for continual moving the blood down the vessels to other parts of the body. When the heart contracts and ejects blood into the large arteries and the aorta, the arteries expand to contain the extra blood. When the left ventricle relaxes, the elastic recoil of the arteries forces the blood onward. This recoil action continues down the vessels for circulation of blood and it is also this action that is responsible for the pulses that are felt at the wrist and the neck. This action is known as the Windkessel Effect (Fung, 1984).

The large arteries are referred to as elastic or conducting arteries, which include the aorta and major arteries which branch off from the aorta. The walls of the aorta and the elastic arteries are relatively thin as compared to their diameters and the middle layer contains more elastic fibers and less smooth muscle. The large number of elastic fibers in the aorta and large arteries are responsible for the recoil of the vessels for a more continuous flow to compensate for the pulsatile and discontinuous flow nature of the heart.

The medium size arteries are called muscular or distributing arteries. These are the vessels that branch off from the conducting arteries. Compared with the large arteries, their middle layer contains more smooth muscle than elastic fibers. They are also capable of vasoconstriction (decrease in diameter) and vasodilatation (increase in diameter) to control the volume of blood to suit the needs of various tissues and body parts (Tortora et al. 1984).

3.1.2 The Arterioles, The Capillaries And The Venules

The arterioles, the capillaries and the venules are responsible for circulating blood to various tissues inside the body. These vessels are smaller in size than the arteries and the veins. As the arteries branch, the blood flows through the arterioles, into the capillaries and then into the venules which converge into the veins.

Arterioles are small arteries that have an inner tunic similar to that of the arteries. The middle layer is composed of more smooth muscle than elastic fibers and the outer layer is made up of elastic and collagen fibers. As the arterioles get closer to the capillaries, the vessels are little more than a layer of epithelial cells surrounded by a few scattered smooth muscle cells. Just like the arteries, the arterioles can also contract or expand and these are key actions that regulate blood flow.

Capillaries are microscopic vessels that are composed of only epithelial cells and are usually so small that only one blood cell can flow through at a time. They are found near almost every cell in the body, especially in the high activity regions such as the organs and the muscles. The primary function of capillaries is to allow exchange of nutrients, oxygen and wastes between the blood and the surrounding tissues and cells.

Venules are several capillaries that join together to collect blood and drain it into the veins. The venules near the capillaries have both the inner layer and outer layer and as they approach to the veins, the vessels might also have the middle layer as found in the arteries (Tortora et al. 1984).

3.1.3 The Veins And The Vena Cava

Veins, which usually appear bluish-green when viewed through the skin, carry deoxygenated blood back into the heart. They are composed of the same three layers as the arteries except that there is more smooth muscle than elastic tissue and more fibrous tissue on the outside. Even though there is less elastic tissue, they are still flexible enough to adapt to volume and pressure

changes. There are two major differences between the veins and the arteries: pressure and structure. The pressure in the veins is much lower than that of the arteries thus when blood leaves a cut vein, it flows evenly unlike the rapid "jet" seen in a cut artery. Because of the low pressures in the veins, there can be trouble overcoming gravity to push the blood up into the heart, this is especially evident in the lower extremities. To prevent back flow, the veins in many parts of the body, especially in the lower limbs, have valves that prevent flow away from the heart.

Vena cavae are the large veins that drain the blood from the veins into the right atrium of the heart. They have the same structural layout as the veins except they do not have the valves found in some of the large veins (Tortora et al, 1984).

3.2 The Heart

The human heart is a precisely controlled pump that circulates blood around the human body to provide essential substances such as oxygen and nutrients to various parts of the body. The heart is hollow, muscular and it weighs roughly 340 grams (Tortora et al, 1984). A normal heart beats from 40 times to 160 times per minute depending on the body's condition, age, physical activities and etc. It pumps blood through over 90000 kilometer of blood vessels (Tortora et al, 1984). The heart is shaped like a blunt cone pointed downward with about 2/3 of its mass to the left of the body center line. Its rough dimensions are 12 cm long, 9 cm wide and about 6 cm thick (Tortora et al, 1984).

The heart can be divided into the right heart and the left heart with two chambers on each side. The right side is responsible for taking deoxygenated blood from the veins and circulating it through the lungs for oxygenation, then, the left ventricle circulates the oxygenated blood through the entire body. Each side of the heart is further divided into two chambers: the atrium and the ventricle. The main function of an atrium is to receive blood either from the veins or the pulmonary veins and a ventricle is for pumping the blood out of the heart into either the lungs or the rest of the body.

As blood is circulating through the heart, the right atrium receives blood from the vena cava during diastole(expansion of the heart). It then pumps the blood through the tricuspid valve into the right ventricle, where during systole (contraction of the heart), the right ventricle pumps the blood through the pulmonary valve, through the pulmonary arteries and into the lungs. After blood has been oxygenated, the left atrium receives the blood from the pulmonary vein, then it empties the blood through the bicuspid valve into the left ventricle ready to be pumped into the rest of the body. Out of the four chambers, the left ventricle works the hardest since it must provide enough pressure to overcome the systemic resistance and force the blood through the entire body. It is this chamber that usually fails first (Tortora et al, 1984).

Figure 3.2 Position Of The Heart And The Heart Chambers
(Tortora et al. 1984)

Figure 3.3 Structure Of The Heart (Tortora et al, 1984)

3.3 The Mock Circulatory Loops

During the development of an artificial heart, it is required to do extensive testing and evaluation of the design device. Since it is not possible to use live specimens at all times, a physiologically comparable mock circulatory device or loop is highly desirable. Most research centers in the world use mock loops for evaluating the operation of artificial hearts. They also use them to study the relationship between the parameters such as heart rate, stroke volume, cardiac output and peripheral resistance. In order for a mock circulatory system to be considered effective, it must meet a few operating requirements (Arabia et al, 1984):

1. The system must be a relatively accurate replication of the physical cardiovascular system with matching hydraulic characteristics.
2. The system must have adjustable resistance and compliance elements for simulation of various physiological conditions.
3. The system must mimic the physical attachments and environment as seen *in vivo* testing and implantation.
4. The system must be simple to operate and easy to maintain to enhance the accuracy and reliability of the system.
5. The system must be reliable (Rosenberg et al, 1981).

For a mock circulatory system to be considered useful, it must be matched, from the hydraulic standpoint, to the compliance and resistance parameters of the real physical system. Since there are so many blood vessels in a human body, to make the system effective and easy to analyze, the major groups of vessels were lumped together to form several individual vessel system. The major systems used for most mock circulatory system are: arterial flow, systemic circulation and venous return flow. Each of these has its own adjustments for compliance and/or resistance and the values for the parameters to be used are obtained from normal physiological values. Table 3.1 shows the typical parameter values of major blood vessels in a human body. From these values, the compliance, resistance and inertance for the system can be obtained.

The ideal resistance for a blood vessel, with the blood flow assumed to be uniform, can be found by using the following equation (Fung, 1984):

$$R = \frac{\Delta P}{Q} \quad (3.1)$$

Where:

- R = vessel resistance [mmHg s/mm³]
- ΔP = change in blood press [mmHg]
- Q = volume flow rate of blood [mm³/s]

To determine the volume flow rate with uniform flow and constant cross sectional area, the following equation can be used (Fung, 1984):

$$Q = Av \quad (3.2)$$

Where:

- Q = volume flow rate of blood [mm³/s]
- A = total cross-sectional area of the blood vessel [mm²]
- v = velocity of the blood flow [mm/s]

To determine the ideal compliance of the blood vessel, the following equation is used (Rosenberg et al, 1981):

$$C = \frac{\Delta V}{\Delta P} \quad (3.3)$$

Where:

- C = compliance of the blood vessel [mm³/mmHg]
- ΔV = change in volume of blood in the blood vessel [mm³]
- ΔP = change in blood press [mmHg]

The ideal fluid inertance is defined as the change in fluid pressure due to a change in the volume flow rate. For modeling purposes, the fluid inertance is important since the systemic pressure is affected by the blood flow rate, which is constantly changing due to the body's needs. For determining the ideal fluid inertance of blood flow in a vessel, the following equation can be employed (Rosenberg et al, 1981):

$$I = \frac{\Delta P}{dQ/dt} \quad (3.4)$$

Where:

- I = inertance [kg/cm⁴]
- P = change in blood press [mmHg]
- dQ/dt = change in blood volume flow rate [mm³/s²]

However, the blood inertance in the cannulae depends on the cannulae geometry. The assumptions used for determining the inertance are: the density of the blood remained constant and the cross sectional area was constant for the entire length of the cannulae. For determining the blood inertance in the cannulae due to the cannulae geometry, the following equation is used (Rosenberg et al, 1981):

$$I = \frac{\rho L}{A} \quad (3.5)$$

Where:

- I = inertance [kg/cm⁴]
- L = cannulae length [cm]
- A = cannulae cross sectional area [cm²]
- ρ = density of blood [kg/cm³]

By determining these parameters, the mock circulatory system then can be calibrated to match the hydraulic characteristics of the real circulatory system. These values are also very important to any mathematical and computer models based on the mock loop. Any variations in these parameters

may well be key in the design, operation and calibration of the final working artificial heart.

3.3.1 The Donovan Mock Circulatory Loop

The mock circulatory loop, currently used for development at CVD of University of Ottawa Heart Institute, was originally designed by Dr. F. M. Donovan Jr. in the mid 70's and was modified in 1989 to suite the needs. It is a simple, easy to use device that contains a venous compliance chamber, arterial compliance chamber, a peripheral loop with variable resistance setting and actual size attachments for the artificial heart. For each of the two chambers, the compliance can be adjusted according to the needs and to the desired simulation conditions. The air above the working fluid of the system acts as the compliance controller (Donovan, F. H. Jr., 1974) and figure 3.4 shows a Donovan loop used for development and testing of the electrohydraulic ventricular assist device.

Under normal working conditions, water is usually used to simulate blood since the densities of the two fluids are approximately equal. In order to have the mock circulate system represent the real system more accurately, a volume of 4 to 5 litres of water is usually used. By carefully controlling the amount of water in each of the two chambers (the venous and arterial compliance chambers), the desired compliance can be obtained. To make the system work better, the tubes, which are used to mimic the peripheral flow, have variable restricters around them to control the resistance. From this loop, the usual measurements obtained for analysis are drive line pressure, heart pressure, heart rate, cardiac output, venous pressure, arterial pressure and supplied power.

For a typical mock circulatory system, the arterial compliance is set at 1.32 cm^3/mmHg and the venous compliance was set at 19.2 cm^3/mmHg . The arterial pressure is set at 120 for systole and 80 for diastole and the heart rate is usually set at 60 beats/min with a cardiac output of 5 litres/min. However, these parameters can be varied as desired to match physiological limits. The

normal dimensions of the whole system are 34 cm x 20 cm x 38 cm (width x depth x height).

Figure 3.4 The Original Donovan Mock Circulatory Loop System
(Donovan, F. H. Jr., 1974)

3.4 The Bond-Graph Model For Simulation

To simulate the cardiovascular system and the importance of the parameters for designing a new control system, the bond-graph modeling technique was employed. From the bond-graph model, state space equations governing the model were derived and the resulting equations were used to create a simulation using Matlab. All of the derived equations were first order differential equations which were simpler to model and analyze than the higher order equations that are often the result of employing conventional mathematical methods. The key elements used for the bond-graph model were resistance (R), inertance (I) and compliance (C) (Karnopp et al, 1990).

The bond-graph model of the VAD was divided into three major sections: the power source for the VAD, the VAD itself with the blood chamber and the normal physiological cardiovascular system with arterial, peripheral and venous circulation. The entire model contains nine energetic elements thus giving nine first order state-space equations for describing the system. Figure 3.5 shows the bond-graph model of the current circulatory system model with a VAD hooked into the system.

The power section for the model was composed of an effort source that provides force to the VAD. Attached to the pressure source were the resistance to motion of the pump, inertia of the pump (e.g., piston or diaphragm) and a transformer transforming the input force into a pressure acting on the blood in the heart.

Attached to the power section was the VAD model. This model consisted of a compliance chamber, which the blood from the input cannulae entered through the inlet valve, and the output valve and the output cannulae. The cannulae were modeled as semi-flexible tubes with limited compliance and inertia governed by their geometric shape. Other important elements of the VAD section were the input and output valves. These valves were passive valves which would open or close due to the pressure difference across the valve. During diastole, the input valve would open for filling the blood chamber. As soon as the pressure rose above the pressure in the input cannulae, the input valve would shut and the output valve would open to let

the blood to be pumped into the aorta. These two actions are not coupled because the opening and closing of the valves depend on the pressure gradient across the valves and since it takes time for the pressure to build up inside the VAD, thus there is an overlapping period where the two valves are either both open or close. The valves in this section are modeled with two distinct resistance values for the open and close states. For the modeling purpose, the closed valve was just the open valve with a much larger R value such that essentially no back flow would occur.

Attached to the VAD section was a model that simulated the actual physiological circulatory system. This section grouped all the blood vessels into three major circulatory systems: the arterial circulation, the peripheral circulation and the venous circulation. The arterial circulation was modeled as a compliant vessel that represented the combination of the aorta and the large arteries. It was directly linked to the peripheral circulation, which represented a collection of all of the micro circulation in every part of the body. The peripheral circulation was modeled by an inertance element and a resistive element. It could be modeled by those two elements because the capillaries were very small in diameter and did not have the compliance of the larger blood vessels. Moreover, since the peripheral circulation covers the entire body, the resistance of the system was very important to the entire circulatory loop. From the peripheral circulation, the blood then entered the venous circulation which was modeled as a compliance vessel just like the arteries.

With these elements and subsystems all linked together, the entire circulatory model contained nine energy storage elements resulting in nine first order differential equations as the governing state-space equations. The state-space equations are shown below. In the equations, the 'p' variables represent force or pressure and the 'q' variables represent velocity or flow rate. The numbered subscripts correspond to the bond number as defined in the bond-graph.

1. $dp_2/dt = F - R_{ht}(p_2/I_{ht}) - A(q_6/C_{ht})$
2. $dq_6/dt = A(p_2/I_{ht}) - (1/R_{ot})(q_6/C_{ht}) - (1/R_{in})(q_6/C_{ht})$
 $+ (1/R_{ot})(q_{10}/C_{bl}) + (1/R_{in})(q_{23}/C_{bl})$
3. $dq_{10}/dt = (1/R_{ot})(q_6/C_{ht}) - (1/R_{ot})(q_{10}/C_{bl}) - (p_{12}/I_{cnot})$
4. $dp_{12}/dt = (q_{10}/C_{bl}) - (q_{14}/C_{at}) + P_r \cos(wt)$ (3.5)
5. $dq_{14}/dt = (p_{12}/I_{cnot}) - (p_{17}/I_{pe})$
6. $dp_{17}/dt = (q_{14}/C_{at}) - R_{pe}(p_{17}/I_{pe}) - (q_{19}/C_{vn}) - P_r \cos(wt)$
7. $dq_{19}/dt = (p_{17}/I_{pe}) - (p_{21}/I_{cnin})$
8. $dp_{21}/dt = (q_{19}/C_{vn}) - (q_{23}/C_{bl})$
9. $dq_{23}/dt = (p_{21}/I_{cnin}) - (1/R_{in})(q_{23}/C_{bl}) + (1/R_{in})(q_6/C_{ht})$

The symbols used for defining the parameters of the model were as the following:

- F = pressure source
- R_{ht} = VAD diaphragm resistance
- I_{ht} = VAD diaphragm inertia
- A = piston area
- C_{ht} = compliance of VAD blood chamber
- R_{ot} = output valve resistance
- C_{bl} = blood compressibility
- I_{cnot} = inertance of output cannulae
- C_{at} = arterial compliance
- I_{pe} = inertance of peripheral circulation
- R_{pe} = resistance to peripheral circulation
- C_{vn} = venous compliance
- I_{cnin} = inertance of input cannulae
- R_{in} = input valve resistance
- w = reminance beat rate
- P_r = reminance beat pressure
- t = time

The state variables were defined as the following:

- p_2 = pump piston or diaphragm momentum

- q6 = heart chamber blood volume
- q10 = output cannulae blood volume
- p12 = output cannulae momentum
- q14 = arterial blood volume
- p17 = peripheral momentum
- q19 = venous blood volume
- p21 = input cannulae momentum
- q23 = input cannulae volume

Figure 3.5 Bond Graph Model Of The VAD Circulatory Model

In the developed model, the main function of the physiological circulatory section was to simulate the real life system as closely as was possible. Once the parameters were determined, the values would remain fixed. On the other hand, the VAD section and the power section were modeled so the parameters could be altered for determining the effect of the parameters on the entire circulatory system. Out of the eight parameters defined in the VAD section, there were a few that would remain fixed. The resistance of the output and input valves would remain fixed because these valves were commercially available and the specifications for these valves were fixed. The inertance of the output and input cannulae might remain fixed since the cannulae had fixed diameter. The anatomical fit study, which was performed with a VAD placed into the human thoracic cavity for determining the device placement position, had already determined the position of the VAD so that the length was also fixed (Diegel et al, 1992). Unless there are other locations inside the human body that are better suited for the implantation, the slight differences in the cannulae lengths due to the size of a person should not greatly affect the model. The compliance of the cannulae was also a fixed value. This value was determined by the material from which the cannulae were made. Unless there is a change in material, the cannulae compliance is fixed.

The design parameters, namely the VAD diaphragm resistance, piston inertance, piston area and the VAD compliance, were the key parameters that were adjusted and tuned for the best operation of the VAD. In the simulation and design process, these values were adjusted, as outlined in chapters 5 and 6 to determine the compatibility of the VAD to the rest of the system. Besides the design parameters, the control parameters, which were the input force, systolic fraction and heart volume limits, were also determined by tuning the system performance. The best results from the simulations determined the parameters so that the model performed closest to the experimentally obtained physiological data or known performance (Tortora, 1984, Fung, 1984, and Guyton, 1993). The important performance parameters that were to be matched were the cardiac output (flow rate), the arterial pressure, the venous pressure, the heart rate and the pressure characteristics of the valve operation. Other parameters that were also important to the design of the control system for the VAD were the VAD blood chamber pressure, the system

efficiency and the power required to drive the VAD to achieve the desired state of operation.

The state-space equations in (3.5) derived from the bond-graph model could be used to solve for the time response of the model. The nine simultaneous differential equations would be very difficult to solve analytically thus they were not very useful for design. With the help of Matlab, a program was written using predetermined parameter values and initial conditions. The model was simulated digitally by numerically integrating the equations over a specified time interval. The program was divided into four different sections each responsible for a special task. The main program (VADSIM5.M) defined the controller and system parameter values, and initial conditions and the time interval for which the simulation was to be conducted. The second section (VADN.M) of the program was a function to evaluate the state-space equations of the model using the input force as determined by the VAD controller as required for the analysis by using Runge-Kutta integrating method. The third section (VADIN.M / FFFE.M) of the program was used to model the controller with predefined period and systolic fraction for the fixed rate controller, or volume limits for the full-fill/full-eject controller. The last section of the program (OPTN.M) combined all the results obtained from the previous parts and displayed the information in a graphical form including parameters such as the flow rate and the pressures in the circulatory system. The output graphs could then be used to determine the validity of the model and the design by comparing them to known published data (Tortora et al, 1984, Fung, 1984). Other modules that were also incorporated for the analysis were: systemic blood volume graphs, energy calculation and power calculation. An additional stand alone program was written for determining the eigenvalues of the model. This program was written to plot the positions of the poles as the parameter values were being altered. These results would help clarify the effects of parameter changes on performance and the sensitivity of the model to variations in parameters. For the complete listing of the programs, refer to appendix 1.

3.5 Parameter Values For Evaluating The Model

In the model, it was very important that the values used for various parameters were as close to the normal physiological values as possible. However, although the parameters such as the pressure in various vessels and the dimensions of the vessels can be determined and easily obtained from various sources, the parameters such as the resistance, flow rate, compliance and inertance were difficult to find since they represented lumped models of complex continuous effects.

Even though the compliance, inertance and resistance values cannot be obtained directly from the physiology references, they can be calculated and estimated based on parameters such as pressures, estimated flow rates and dimensions. For the bond-graph model, these parameters were approximated using equations 3.1, 3.2, 3.3 and 3.4. For the complete set of parameter values used in the development of the bond-graph model and in the simulation refer to table 3.1 and table 3.2.

Other physiological parameters that were also important in the compatibility of the VAD and were important for simulating the natural physiological states were as follows (Tortora et al, 1984, Fung, 1984):

1. Cardiac output (CO)	=	3.5 L/min at rest - 15 L/min exercising
2. Heart rate (HR)	=	40 -160 beats/min
3. Preload pressure	=	5 -20 mmHg
4. Afterload pressure	=	60 - 160 mmHg
5. Blood volume inside a body	=	3.8 L. for woman, 5.5 L. for man
6. Systolic fraction	=	0.27 - 0.41
7. Cardiac cycle period	=	0.8 - 1.0 sec

	Aorta	Artery	Arteriole	Capillary	Venule	Vein	Vena Cavae
Diameter (mm)	25 ¹	4	0.03	0.008	0.02	5	30
Velocity (cm/s)	40 ¹ 40 ²	10-40 ¹ 10-50 ²	0.1	0.03	0.3	0.3-5	5-20 ¹ 20 ²
Total Blood Volume (%)	3	13	3	4	3	51	10
Volume (L)	0.15	0.65	0.15	0.2	0.15	2.55	0.5
Total Cross- Sectional Area (cm ²)	2.5	20	40	2500	250	80	8
Mean Cross- Sectional Area (cm ²)	2.5	12.6	7e-6	5e-7	3.1e-6	0.2	7
Pressure (mmHg)	80-120 ¹ 80-115 ² 80-120 ³	40-100 ¹ 45-100 ² 70-110 ³	25-40 ¹ 30-40 ²	12-25 ¹ 10-25 ³	8-12 ¹ 10-12 ³	5-10 ¹ 7.5 ² 4-8 ³	2-5 ¹ 3.75 ² 2-5 ³

NOTE: The parameters which do not have a superscript number were parameters obtained from Tortora et al, 1984.

Table 3.1 Table Of Published Physiological Parameters For The Circulatory System (1. Tortora et al, 1984, 2. Fung, 1984, and 3. Guyton, 1993)

	Aorta	Artery	Arteriole	Capillary	Venule	Vein	Vena Cavae
Volume Flow Rate (cm ³ /s)	100	200-1000 600*	4	75	75	24-400 212*	40-160 100*
Resistance (mmHg.s) (cm ³)	0.4	0.1	0.27	0.2	0.089	0.024	0.03
Compliance (L/mmHg)	0.00375	0.0108	0.01	0.0133	0.0375	0.51	0.167
Length Estimated (cm)	50	30	-----	30	-----	30	50
Inertance (kg/cm ⁴)	2e-2 2.35e-2 1	1.5e-3	-----	1.2e-5	-----	3.75e-4	6.25e-3

NOTE: * = Mean values

Table 3.2 Physiological Parameters Required For The System Model
(Tsach et al, 1989)

4.0 VENTRICULAR ASSIST DEVICE AND CONTROLLERS HEART-PARAMETER DETERMINATION

A VAD-circulatory simulation model, as mentioned, should be easily modified to suit the needs of VAD developers. These needs include performing studies on different types of driving mechanism, on the effects of variations in controller parameters and on VAD control methods.

The ability to modify the model to adapt different types of driving mechanism was important since there are many different types of VAD drivers as discussed in chapter 2. Although there were several different methods for driving the VAD (i.e., piston, pusher plates, pneumatic diaphragm using forced air, electrohydraulic), they can be grouped into two major categories: fluid drivers and mechanical drivers. The VADs using fluid drivers used silicon oil or pressurized carbon dioxide to push on a diaphragm and generate enough pressure to pump the blood into the circulatory system. An example of the fluid VAD was the OHI electrohydraulic VAD, which this VAD-circulatory model was based on. This device, as described in chapter 2, used a high speed DC motor to propel silicon oil to drive the diaphragm. Since this type of VAD relied on the flexibility of the diaphragm, the VAD chamber compliance parameter played an important role.

The mechanical driver, as used in the Penn State VAD described in chapter 2, acted directly on the blood chamber of the VAD. Since the driving piston or the pusher plate acted directly on the blood chamber and the pusher plate is very rigid, the heart compliance parameter for the model would be set to zero for modeling this type of device.

Different sets of controller parameters would produce different VAD designs, thus it was crucial to perform studies on how the controller parameters would alter the behaviour of the device. Since the properties or the driving device might change characteristics over time, it is useful to know how the device would work under these conditions. This VAD-circulatory system model would also be very useful for examining the operation of the device under a variety of different control methods.

There were two generally accepted control methods examined for the VAD: the fixed rate control mode and the full-fill/full-eject mode (which considered as the synchronized mode). These two modes differed mainly in how the VAD was to be controlled for pumping the blood. From the design point of view, the synchronized mode was the most complicated to implement due to the number of parameters involved and the difficulties in matching the VAD to the natural heart beat. The goal of this type of control was to synchronize the VAD rate to the natural heart rate. There were a few ways of synchronizing the VAD to the natural heart. One of the ways of synchronizing the VAD to the natural heart rate might be to use the Electrocardiogram (ECG) to trigger the VAD. Since the electric potential would change between diastole and systole, it could be used as the signal for controlling the VAD to switch between the two phases. Other proposed method included using the cardiac output as the feedback signal for controlling purposes or using various biological feedback signals for matching the VAD to the natural heart state. Although there exist many possible synchronization methods, the fixed rate and full-fill/full-eject methods remained as the preferred controlling methods. The fixed rate mode, on the other hand, was much simpler to implement since the VAD would beat at the same rate regardless of the natural heart rate. In between the two methods was the full-fill/full-eject method in which the control was based on the chamber filling rate and the pressure difference across the valve (the detailed description of the two controllers is found in sections 4.1 and 4.2).

Due to the ease of implementation, only the fixed rate mode and full-fill/full-eject controllers were incorporated in the developed VAD-circulatory model. To make the model accept both controllers and to evaluate the control methods accurately, the following parameters (as described in chapter 3) were included: input force, base force, heart piston area, heart compliance, heart piston inertia, heart resistance, heart beating frequency, systolic fraction, heart volume limits, input and output valve resistances and input and output cannulae sizes and compliances. Most of the mentioned parameters could be chosen and modeled to meet the desired operating conditions. The parameters that were fixed were the valve resistance and the cannulae compliance and inertia since these were not easily and practically changed. Very often a fixed cannulae length was required for the anatomical fit.

The input and output valve resistances were normally predetermined since the valves were often commercially available. These valves were generally widely accepted and their design was already proven in clinical tests. Using commercially available valves would reduce VAD development time and cost and it may help the research team to have a more successful VAD design.

The parameters relating to the input and output cannulae were also accepted as fixed values with low variability. This was especially true for cannulae compliance since the compliance of the tubes depends on the nature of the material from which they were manufactured. The cannulae in use clinically had a fixed compliance value and unless the material used was changed or redeveloped, the cannulae compliance will be accepted as this fixed parameter value. The cannulae inertia was more variable than the compliance. This was because that inertia was a function of the cannulae length, which can be varied to a certain extent. Although the cannulae length can be picked to obtain a desired cannula inertia value, physiological constraints mean that the surgeons may not have a choice but to use a cannulae of a fixed length and thus a fixed inertia. This was due to the restrictions on the location of the device within the chest cavity.

The parameters described, could be further grouped into three types of parameters: 1) control parameters, 2) design parameters and 3) systemic parameters. The control parameters were the ones that would be selected and adjusted after the construction of the device for controlling the behaviour of the VAD and achieve the required performance. These parameters were:

1. input forces (P_i , P_{fb})
2. systolic fraction
3. heart rate
4. chamber volume limits

The design parameters were the ones selected before the actual construction. These parameters would determine the physical characteristics of the device such as the weight and size. The parameters in this category were:

1. heart compliance
2. heart resistance
3. heart piston inertia
4. heart piston area
5. input and output valve resistances (these values were fixed since they were commercially available)
6. cannulae compliance (fixed by the material)
7. cannulae inertia (depended on the physical length)

The systemic parameters were the uncontrollable physiological parameters that varied according to the body's condition. For the construction of the VAD-circulatory model, these values were kept at their nominal values as found in chapter 3 and they were only varied during the simulation to evaluate the sensitivity of the VAD designs to parameter variations. These parameters were:

1. arterial compliance
2. venous compliance
3. peripheral inertia
4. peripheral resistance

In this chapter, the methodology of each control method is discussed. Also included in this chapter is a discussion of the methods used for determining nominal parameter values for each controller.

4.1 Fixed Rate Mode Control Method

The fixed rate controller controlled the VAD to beat with a fixed heart rate and systolic fraction set by the operator. The main goal was to generate a specified average blood flow rate which would be monitored by the operators. With the heart rate set between 30 to 150 beats per minute (bpm) and the systolic fraction between 25% and 50%, the VAD would operate at this same working point regardless of the natural heart beat. For normal body conditions, the VAD would be set at 60 bpm and systolic fraction at 30% to maintain an average blood flow rate of about 6 L/min. For increased activity, the heart rate would

be increased to increase the blood flow rate and during periods of low body activity, the VAD could be set to beat at about 50 bpm to meet the reduced blood flow demand.

The advantage of the fixed rate control method was that it can operate without feedback parameters from the physiological system since it works independently of the natural heart beat. This would make the device simpler to design and to construct.

The heart rate and systolic fraction were the two parameters that would be adjusted while the VAD was in operation and the input forces and chamber volume limit would be fixed. With the correct VAD design, the combination of beat frequency and systolic fraction could make the VAD to provide adequate blood flow rate and pressure to accommodate the physical state of the patient. In normal operation, the physician could only adjust these two parameters to maintain proper blood flow rate and heart pressure as required by the condition of the patient. The other two parameters were also important to the VAD operation since the controller and the VAD together made up an entire circulation assisted device. The combination of control and design parameters would determine the power requirements of the VAD, its physical size and its complexity. In a general sense, the higher the input forces, the higher the power requirements would be to drive the device. The driving was could also used to control the heart pressure and the heart beat frequency.

The design parameters were important to the general behaviour and the physical design of the VAD. The chamber compliance would be one of the determining factors of the VAD volume. For a more flexible chamber, a change in pressure would require a bigger change in volume than a chamber with stiffer membrane. An optimal compliance value was therefore required to satisfy the size and response constraints. It would be good to have low heart compliance (high stiffness) since this could increase the VAD sensitivity to volume change. Also, a chamber with low compliance would generally require a smaller volume than a flexible chamber to achieve the same response speed.

The heart piston inertia would directly affect the design weight of the device. With a higher inertia, the entire VAD weight would be increased. It would be ideal to design a device with minimal weight since a light weight device would be more suitable for long term implantation. As part of the design consideration, it would also be ideal to design a VAD to have a natural frequency at the nominal heart rate of 60 bpm. If this could be achieved by tuning the heart compliance and inertia, the VAD would be most efficient at that operating condition since it would require the least amount of power to drive the device at this operating point.

The heart resistance in the design would act to resist the motion of the piston under normal operating conditions. The lower the resistance, the higher the efficiency of the device. It was unreasonable to have zero resistance. Together with the compliance and inertia, the resistance could be used to tune the heart frequency. Furthermore, with higher resistance, the heart pressure could be easily build up and the resistance could be used to control the piston motion. However, the cost of having restrictive control was its high energy dissipation characteristic. This would mean that higher power was required to offset the energy lost to the resistance. There would be a trade off while deciding a resistance value to be used and selection requires careful analysis.

The fixed rate controller may be suitable for use in VADs that were designed to be used as a bridge to heart transplant. The reason was that, while waiting for a donor heart, the patient was normally under intensive care at the hospital. During that period of time, the patient was not expected to perform tasks that require extensive physical movement or sudden abrupt action. A fixed rate VAD was quite adequate to provide the required blood flow rate and since only the heart rate and the systolic fraction could be changed, it would be much easier for the physicians to make necessary adjustments to meet the patients needs. Even if the patient was required to leave the bed, as long as the motion was not very abrupt or fast, the fixed heart rate would be adequate. With this VAD, the physician could adjust the heart rate to meet the patient's physiological requirements without difficulty.

4.2 Full-Fill/Full-Eject Control Method

In comparison with a fixed rate controller, the full-fill/full-eject controllers behaved in a way that more closely resembled the natural circulatory system. This was because the rate at which the VAD filled or emptied depended on the pressures in the blood at the VAD inlet and outlet. For example, a higher inlet pressure would result in more rapid filling of the VAD, while a higher outlet pressure would lower the rate at which blood flowed out of the VAD. In this way, a VAD using the full-fill/full-eject control could respond to changes in the body due to, for example, dilation of the capillaries in response to cold or, possibly, limited beating of the natural heart. Dilated capillaries would reduce peripheral resistance to blood flow and thus reduce the outlet pressure and increase the flow rate. For a VAD using this control method, the switching between the systole and diastole would be determined with a preset VAD volume limit. At the end of diastole, when the VAD chamber was filled to the preset maximum volume, the VAD would switch to systole. When the VAD reached the minimum volume limit, the VAD would then switch back to the systolic phase and the whole cycle would repeat again. Since this controller depended on the VAD filling rate which in turn depended on the inlet and outlet pressure, the VAD with a full-fill/full-eject controller may have the tendency to follow the natural heart beat and the beat rate could vary constantly.

The fundamental parameters that were required to design a VAD with full-fill/full-eject controller were similar to those of the fixed rate controller except that the switching between diastole and systole for the fixed rate mode depends on preset beat frequency and systolic fraction whereas for the full-fill/full-eject mode it depends on the chamber volume limits.

The combination of the control and design parameters listed were important to the operation of the VAD with a full-fill/full-eject controller. The switching point for systole and diastole depended on the end heart volume and the heart compliance. This was because that with a smaller heart volume or lower heart compliance (high stiffness), the end volume limit would be reached much faster than higher volume limit and compliance (low stiffness) since the entire diaphragm would move uniformly. Also, the larger the values, the

larger the physical size of the VAD would be. The heart piston area would also affect the heart volume since a larger area piston would reduce the traveling distance of the piston, for a given stroke volume. At the same time, the larger the piston, the lower the heart rate would be since the larger the surface area, the smaller the pressure would be for the same input force. Due to this effect, a longer time would be required to build up the VAD pressure to operate the valves for switching from systole to diastole and vice versa.

Although the piston area had an effect on the heart rate, the key parameters were the input forces and heart resistance. With a larger total force, the heart rate would increase along with the heart pressure. Besides the input forces, the heart resistance plays an important role in fine tuning the heart rate and heart pressure. With higher resistance, the heart diaphragm and piston would take longer to move through a complete cycle thus slowing the beat rate. But with higher resistance, the blood would tend to accumulate within the blood chamber thus causing the heart pressure to increase. Generally, it would be ideal to have the lowest resistance to improve the system efficiency. In a real practical design a higher than theoretically preferred resistance may be chosen since resistance could be used to control the VAD response and it could even be used to fine tune the natural frequency of the device.

The effect of heart inertia on the VAD when using full-fill/full-eject control was similar to the fixed rate controller. The inertia could be chosen to achieve the desired heart response and oscillating harmonic frequency. By matching the natural frequency of the device to the natural heart rate would increase the system efficiency. With higher inertia, the VAD would beat more smoothly, but at the same time, this would add weight to the device and it also requires a higher driving force if the VAD motion was to be kept at the same frequency.

The VAD system that incorporates the full-fill/full-eject controller may be better suited for long term usage than one with a fixed rate controller. The full-fill/full-eject controller might be a better choice because it may respond to physiological changes by altering its beat rate to match that of a natural heart to a certain extent. This ability to respond to changes in load was important for normal activities of patients who were using such a device. As

long as the activity was within a predetermined limit, the VAD should function quite normally. In a way, the full-fill/full-eject controller may even be more practical than a true synchronous controller since it would respond to the changes in load without the complexity involved in incorporating a synchronous controller. It would let the circulatory system determine the necessary operating rate and would react accordingly. Synchronous control would require elaborate and complicated sensors for determining various parameters. This control method should provide a desirable solution to the VAD control problem with reasonable complexity and cost.

4.3 Device Parameter Determination

When designing a VAD, it was important to select reasonable values for heart resistance, heart piston inertia and heart chamber compliance. Only with the right combination of these values could the input forces and piston area be chosen to ensure proper operation of the device. The conventional approach to obtaining these values was to use a static method. The heart compliance was determined based on the change in VAD chamber volume due to a change in VAD pressure relationship. With the determined compliance value, the heart inertia was then chosen based on the total weight of the device and other physical limits. The heart resistance was usually set at the lowest possible value that could be built to reduce energy loss caused by high resistance. After selecting the three basic parameters, input forces were selected and a model was built to determine the feasibility of the design. This approach did not include the consideration of dynamic interaction among the parameters and with the circulatory system.

The dynamic method of using the VAD-circulatory model developed in this project has an advantage over the conventional method. By using this model, all parameters could be simulated and evaluated dynamically. The influence of the parameter set on the system performance could be studied efficiently and a set of parameter envelopes could be determined as a limiting range for the design. One definite advantage of this model over the conventional static method was its ability to evaluate the feasibility of either the fixed rate mode

or the full-fill/full-eject mode by simply selecting the desired mode. This model was used to perform studies on the effects of cannulae length and the use of residual arterial pressure variations for use in synchronization of the VAD to the rest of the circulatory system.

To determine the key parameters for the model, the heart compliance value was used as a starting point. A heart compliance value was chosen based on actual heart compliance and values obtained from other circulatory assist devices that were similar in design to the Ottawa Heart Institute EVAD (5 to 300 cm^4/N) (Klute et al. 1992, Tsach et al. 1989). The piston inertia was picked to keep the VAD weight close to 200g. With the heart resistance set at a reasonably low value, all values along with a set of input forces were incorporated into the simulation for evaluation. The goal was to have a combination of heart parameters and nominal systemic parameters to achieve the following:

1. Arterial pressure = 80 to 140 mmHg
2. Venous pressure = 5 to 40 mmHg
3. Blood flow rate = 3.5 to 15 L/min
4. Heart rate = 40 to 120 bpm
5. Heart valve to operate with pressure differences
6. The parameter graphs resemble generally accepted cardiac cycle pressure graphs.

When the nominal values of the heart parameters were determined, a range of values were tested to determine when the VAD would stop functioning normally to establish a working condition limit. All of the studies described above were done for both fixed rate mode controller and full-fill/full-eject mode controller.

The method to determine the cannula compliance was to use a piece of cannula that has been previously implanted inside a calf during an *in vivo* evaluation. A used cannulae was required since a new cannulae was porous and could not hold fluid during a compliance measurement. Only an implanted piece would be feasible because blood would coagulate and seal the pores so when air was injected into the cannulae, changes in volume due to a change

in pressure was observable. After obtaining the cannulae, one end was sealed with a plug and the other end was connected to a pressure source. Since the cannulae diameter was fixed and was supported with plastic rings, the change in volume would be a function of change in length. Simply by measuring the length change and pressure change, the cannulae compliance could be determined. During the experiment, a sphygmomanometer (an instrument for measuring arterial blood pressure) was used as both a pressure source and as a pressure gauge. As for measuring the cannulae length, a ruler was used. The data in table 4.1 was collected for both pressure and length and since the cannulae diameter was fixed at 2.5cm, the volume was calculated. With all the data plotted, the cannulae compliance was determined (see figure 4.1). This method could produce a compliance value greater than it should be since the cannula was allowed to expand freely during the experiment and it would be restricted when implanted inside the human body.

Pressure (mmHg)	Length (cm)	Volume (cm ³)
0	13.5	66.27
60	16.5	81.00
210	21.0	103.08

Table 4.1 Changes In Cannulae Volume Due To Changes In Pressure

GRAPH FOR DETERMINING C OF THE CANNULA

Note:

- 1) The compliance $C = \frac{\Delta V}{\Delta P}$
- 2) The cannula used is a 22mm diameter low porosity woven Dacron cannula with rings on the outside. The cannula has been used for *in vivo* testing and has dried blood on it.

Figure 4.1 Graph For Determining The Compliance Of The Cannulae

From the initial simulation studies, the nominal parameters were determined for both the fixed rate mode and full-fill/full-eject mode. These values were used as the basis for further studies and as the foundation for building a working device.

The parameters that could be shared among the two control methods were:

1. Cannulae compliance (C_{blo} , C_{bli}) = $44 \text{ cm}^5/\text{N}$
2. Input cannula inertia (I_{cin}) = $0.0000132 \text{ Ns}^2/\text{cm}^5$
3. Output cannula inertia (I_{cot}) = $0.0000263 \text{ Ns}^2/\text{cm}^5$
4. Heart piston area (A) = 19.27 cm^2

The following table included the range of design and control parameters used for determining the feasibility of the model and the design.

	Input Force (N)	Base Input Force (N)	Max VAD Volume (mL)	Min VAD Volume (mL)	VAD Cht (cm^5/N)	VAD Iht (100kg)	VAD Rht (Ns/cm)
FIXED RATE MODE	20 - 40	0 - 10	----	----	4 - 100	0.0020 - 0.0040	0.15 - 0.47
FFFE MODE	15 - 30.5	0 - 10	60 - 100	0 - 10	10 - 250	0.0020 - 0.0040	0.10 - 0.25

Table 4.2 Table Of Nominal Controller Parameters Range

By setting the design and control parameters to their nominal values, the performance of the VAD using both the fixed rate mode and full-fill/full-eject mode was studied. The results obtained were shown in table 4.3.

	Fixed Rate Mode	Full-Fill/Full-Eject Mode
Efficiency (%)	45.6	15.22
Peak input power (W)	24.9	48.41
Average input power (W)	2.717	5.681

Table 4.3 Table Of VAD Performance Results

The performance of the VAD are also plotted in terms of pressure, flow rate and volume. Figure 4.1 shows the performance of the VAD by using fixed rate mode and figure 4.2 shows the VAD performance by using full-fill/full-eject mode. The performance graphs are further subdivided into six graphs. For the output valve graph, the solid line represents the pressure difference across the output valve in mmHg and the dotted line represents the blood flow rate across the valve in L/min. For the artery plot, the solid line with higher oscillation amplitude is the output cannulae pressure and the other solid line is the arterial pressure, and both are in mmHg. The dotted line on the bottom is the blood flow rate in L/min. For the heart pressure and motor velocity graph, the solid line with the highest peak is the VAD pressure in mmHg. The solid line in the square wave form is the input force in N and the dotted line is the motor velocity in cm/s. For the input valve graph, the solid line is the pressure across the input valve in mmHg and the dotted line is the blood flow rate in L/min. In the Veneous graph (Vein), the solid line with high amplitude is the input cannulae pressure and the other is the venous pressure, and both are in mmHg. The dotted line on the bottom represents the venous flow rate in L/min. The peripheral graph on the right side bottom shows the blood flow rate through the peripheral system in L/min.

The circulation volume curves are also plotted to verify the conservation of blood volume. Figure 4.3 shows the volume curve for VAD operating in fixed rate mode and figure 4.4 shows the volume curve for VAD operating in full

fill/full eject mode. The total volume represented by the two curves does not include the pulmonary circulation.

Q = flow rate (L/min)
P = pressure (mmHg)
V = velocity (cm/s)

Figure 4.2 Performance Graph Of VAD Operating In Fixed Rate Mode

Q Flow rate (L/min)
P pressure (mmHg)
V velocity (cm/s)

Figure 4.3 Performance Graph Of VAD Operating In Full-Fill/Full-Eject Mode

Figure 4.4 Volume Graph Of VAD Operating In Fixed Rate Mode

Figure 4.5 Volume Graph Of VAD Operating In Full-Fill/Full-Eject Mode

5.0 CONTROL METHOD EVALUATION

It was important to decide on the control method to be incorporated into the VAD design. A VAD with fixed rate controller might be easier to implement and easier to operate since all heart parameters were fixed and only the heart rate and systolic fraction could be altered upon demand to meet the body requirement. Problems which this design might encounter were its inability to adapt to changes in its operating environment. Since a VAD with fixed rate mode controller would work almost independent of the working environment, it might be more reliable than the one using full-fill/full-eject controller. The fixed rate mode might be more suitable for using in a patient who was not very active. This control mode might also be useful in experimental VAD to examine the integrity and reliability of the entire VAD design under controlled working condition.

On the other hand, the VAD with the full-fill/full-eject mode controller might be more suitable for long term usage since it behaves in a way that more closely resembles the natural circulatory system. By using this control method, the VAD could adapt itself to various physiological conditions continuously. But, because it depended on conditions in the physiological system, the VAD system was quite sensitive to parameter changes, thus making the design process more difficult. Also, since it was sensitive to operating condition variations, it might have limited working range for a fixed set of heart parameters. Unless the heart parameters could be changed, the full-fill/full-eject controller might be less reliable than fixed rate mode controller for extreme fluctuations operating conditions.

For a design to be valid and feasible, it was important to study the effect of variations in controller parameters under fixed nominal physiological condition. With different sets of heart parameters, the VAD might respond very differently under various conditions. Furthermore, with different heart parameters, the physical dimensions and characteristic of the VAD might also vary. With the developed VAD-circulatory model, one could efficiently determine different combinations of heart parameters. With one set of optimal parameters selected to be the basis of VAD design for both control

methods, each individual controller parameter could be varied to examine how the VAD response would change. This study would allow the designers to predict VAD performance affected by changes in controller parameters due to either manufacturing deviation or component property changes. It could also be used for tuning the VAD to be used for various patients. Since the developed VAD-circulatory model was a generic simulation model, researchers could use it to do quick prototyping to explore various ideas with changes in controller parameters or even control methods.

The developed model was not only useful for determining the controller parameter and the two implemented method, but it could be modified to incorporate other controlling schemes. The simulation program was written in a modular fashion so that a separate module containing a new control method could be easily used with the simulation. By doing so, a new set of control parameters could be determined and from this, a working model then could be built after the validation of the functionality by the new simulation model.

Another important study performed by using the developed VAD-circulatory simulation model was the possibility of synchronization of a VAD with full-fill/full-eject mode to the residual arterial beating. In a normal human body, the aorta and the arteries have a natural residual pulsatile beating that helps to push blood down the vessel. This residual pulse usually matches the heart rate. At the nominal condition, it would be about 60 beats per minute. It would be of some importance to study if the VAD would synchronize to the residual pulse since if it was achieved, the efficiency of the system could be improved. On the other hand, the residual pulse might disrupt the performance of the VAD and cause it to fail at unexpected operating conditions. Furthermore, if the VAD with full-fill/full-eject controller could be made to synchronize to the residual pulse, it might be then possible to use the knowledge to develop a fully synchronized VAD.

This chapter discusses the sensitivity of the two control methods to variations in controller parameters. Analysis was performed to determine the limits of the variations before the VAD would fail to function. This chapter also

includes discussion of the study performed to examine the synchronization of the full-fill/full-eject controlled VAD to the residual beating of the arteries.

5.1 Sensitivity To Design Parameters

As mentioned before, the design parameters were not only important to the physical design of the VAD. Two sets of design parameters could produce two completely different VADs in both dimension and design philosophy. The design parameters also greatly influence the VAD performance.

The effect of varying the design parameters was examined by fixing the systemic parameters and varying the controller parameters within a reasonable range. The parameters that were fixed at their nominal values were, input and output valve resistance, arterial compliance, peripheral resistance, peripheral inertia and venous compliance. The nominal values for these parameters are shown in table 5.1. The design parameters under considerations were:

1. VAD compliance
2. Piston inertia
3. VAD resistance
4. Cannulae inertia
5. Cannulae compliance
6. Piston area
7. Valve resistance (fixed parameters due to commercial availability)

The sensitivity study was done by varying one parameter at a time while keeping the other parameters fixed at their nominal values. It was found from the simulation that the VAD performance was quite tolerable to variations in design parameters.

It was also determined from the sensitivity study that the full-fill/full-eject controller was more sensitive to design parameter changes than the fixed rate mode. This was because that full-fill/full-eject controller depended more on the heart compliance, piston inertia and heart resistance combination to

control the heart performance and the heart rate. Since the fixed rate controller utilized a timing algorithm to control the heart rate and systolic fraction, even with variation in control parameters, the VAD would beat at the same rate. Only other variables such as the pressures, piston velocity, input power and efficiency would cause deviations in design parameters to match to the new VAD operating environment.

5.1.1 Fixed Rate Method

For the design parameter study, the systemic parameters were kept at their nominal values and the input forces remained constant. The input forces were kept constant was because that these input force values worked well with the parameters set at the nominal condition, and by keeping them constant, the nominal conditions could be used as the reference for parameter variation study. The ranges of variation of the design parameters and their nominal values were shown in table 5.1.

Input Forces P _I = 40 P _{Ib} = 0	Nominal Value	Variation Range (* factor)
VAD Compliance (cm ⁵ /N)	28	0.2 - 5.0
VAD Inertia (100kg)	0.0038	0.2 - 5.0
VAD Resistance (Ns/cm)	0.47	0.2 - 5.0
Output Cannulae Inertia (Ns ² /cm ⁵)	0.000263	0.1 - 10.0
Input Cannulae Inertia (Ns ² /cm ⁵)	0.000132	0.1 - 10.0
Cannulae Compliance (cm ⁵ /N)	44	0.2 - 5.0
Piston Area (cm ²)	19.27	0.2 - 5.0

NOTE: The output and input cannulae have the same compliance

Table 5.1 Nominal Design Parameters And Range Variation For Fixed Rate Mode

For the simulation, the heart rate was kept at 60 beats per minute and the systolic fraction at 0.30. It was found from the simulation that it did not matter how the parameters were varied, the beat rate and systolic fraction remained constant.

It was observed that the VAD with a fixed rate controller was quite tolerable to changes in design parameters except for input cannulae changes. The effect of an input cannulae would be expected to be significant since if the input pressure or the input flow rate were either too high or too low, the VAD with a preset beat rate might not accommodate the variations. The result could be either be an increase in arterial pressure, if the input pressure was higher than expected, or a decrease in pressure if the condition was reversed. In that case, an adjustment would be necessary for the accommodation. Other than input cannulae effects, the VAD performed within the expected ranges.

The description of the effects of changes in the design parameters on the VAD's performance is presented qualitatively. The reason is that by using qualitative description, the general response trend could be used as a guide for design purposes rather than trying to replicate the exact performance.

l_{cnn} (input cannulae inertance)

The VAD functioned well when the l_{cnn} was varied within the range 0.5 to 3 and the performance failed at 0.1 and 10. This meant that the allowable cannulae length would be between 2.5 cm and 15 cm. In general, if the input cannulae was shortened, the amplitude of cannulae pressure oscillation would decrease and the VAD efficiency would increase. This was found to be the opposite when the cannulae length was lengthened, or l_{cnn} was increase. This indicated that it would be beneficial to have as short as input cannulae as possible.

Figure 5.1 Graph Of Venous Circulation At $l_{cnn}=0.5$

Figure 5.2 Graph Of Venous Circulation At $l_{cnn}=3.0$

Cbli (Input Cannulae Compliance)

The performance check for cannulae compliance variation would only be meaningful if the cannulae material was changed. But for future reference, the VAD was found to work properly when the compliance was varied between 0.2 to 2 times the nominal value. The compliance affected the pressure oscillation. It was found that with a stiffer cannulae (smaller C_{bli}), the oscillation frequency and amplitude increase. At the same time, the input pressure would be increased slightly. The opposite was found when C_{bli} was at double the nominal value. This meant that with a change in material, at the preset nominal conditions, as long as the cannulae compliance lies within the described range, the VAD could most likely function with the new design.

Figure 5.3 Graph Of Venous Circulation At $C_{bli} \cdot 0.2$

Figure 5.4 Graph Of Venous Circulation At $C_{bli} \cdot 2.0$

Lenot (Output Cannulae Inertance)

The output cannulae length did not seem to have a great effect on the VAD performance. The range was varied between .1 to 10 times the nominal value and the main effect would be on the oscillation frequency and amplitude of the arterial pressure. At low Lenot, the frequency was high with small amplitude and vice versa for high Lenot value (long cannulae). The interesting effect was that at 10 times the nominal length, the heart pressure wave form was smoother. This would indicate that the length of output cannulae may be selected to tune the performance of the VAD.

Figure 5.5 Graph Of Arterial Circulation At Lenot=0.1

Figure 5.6 Graph Of Heart Pressure At Lenot=10.0

Cblo (Output Cannulae Compliance)

The change in output cannulae compliance had more influence on the VAD than the input cannulae compliance. The reason was that the change in compliance would affect the change in pressure in the output cannulae. This would then affect the output valve operation and thus VAD operation. The acceptable change in Cblo was from 0.5 to 2 times the nominal value. With lower Cblo, the cannulae oscillation amplitude became large and the arterial pressure lowered, but the system would continue to function. With higher Cblo, the oscillation amplitude would be lower. But with any change in Cblo, the VAD pressure wave would have sharp peaks. This meant that the VAD would not be working as smoothly as the natural heart would be.

Figure 5.7 Graph Of Arterial Circulation At Cblo=0.2

Figure 5.8 Graph Of Arterial Circulation At Cblo=1.0

Cht (VAD compliance)

The effect of VAD compliance variations was mainly the shape of the heart pressure curve. For the variation ranges from factor of 0.2 to 5, the VAD performed normally. At a lower Cht value, the arterial pressure and flow rate would increase along with efficiency. At the same time, the average input power was lowered. The heart pressure curve with low Cht value was very rough with many peaks. This would be caused by faster pressure oscillations due to the stiffer diaphragm. With higher Cht value, the heart pressure wave was quite smooth, but the efficiency, heart pressure, flow rate and arterial pressure all dropped. This would be because with a more flexible diaphragm, more power would be needed to keep the pressure up. This was evident from an increase in average input power.

Figure 5.9 Graph Of Heart Pressure At Cht=0.2

Figure 5.10 Graph Of Heart Pressure At Cht=5

Iht (VAD Inertia)

The variations in Iht from factor of 0.2 to 5 did not affect the overall VAD performance. All parameters remained nearly constant including the efficiency. The observable effects were in the heart pressure curves. At lower Iht value, there was more oscillation, this would be caused by the ease of setting the VAD into motion with the same input force. On the other hand, when Iht was increased, the mass was increased. By doing so, the piston would require a higher force to be moved, and a lower oscillation frequency was observed.

Figure 5.11 Graph Of Heart Pressure At Iht=0.5

Figure 5.12 Graph Of Heart Pressure At Iht=5.0

Rht (VAD Resistance)

The VAD resistance had significant influence on the VAD operation. With the Rht set at 0.2 of its nominal value, the heart pressure increased thus causing the arterial pressure and flow rate to increase as well. With lower Rht, the system efficiency increased since there would be less restriction on the diaphragm movement and more input energy was directed to pumping blood. When Rht was set at 5 times the nominal value, the heart pressure decreased thus causing the flow rate, arterial pressure and efficiency to decrease. The low heart pressure could be avoided by increasing the input forces. By doing this, the efficiency would be reduced further, but the VAD could be tuned to operate properly.

Figure 5.13 Graph Of Heart Pressure At $R_{ht} \times 0.2$

Figure 5.14 Graph Of Heart Pressure At $R_{ht} \times 5.0$

A (Piston Area)

The VAD was found to be operational when the piston area was varied within the factor of 0.5 to 2. At half of the nominal A value, the heart pressure increased and arterial pressure, venous pressure and flow rate to increase. The increase in heart pressure would be caused by the decrease in area while applying the same force. When A was increased, the effect was opposite to decreasing the piston area. Since the effects were quite significant, returning of the VAD would be required if the piston area was to be changed.

Figure 5.15 Graph Of Heart Pressure At $A=0.5$

Figure 5.16 Graph Of Venous Circulation At $A=0.5$

2

PM-1 3 1/2" x 4" PHOTOGRAPHIC MICROCOPY TARGET
NBS 1010a ANSI/ISO #2 EQUIVALENT

PRECISIONSM RESOLUTION TARGETS

From the studies made, it was found that the VAD with a fixed rate controller would be quite tolerant to design parameter changes except for the input cannulae and the piston area. This would mean that a designer would have great freedom to select parameters to meet physical requirements or financial requirements

5.1.2 Full-Fill/Full-Eject Method

The same method of studying the design parameter effects was also used for the full-fill/full-eject controller. The parameters under examination and their nominal values and variation factors were found table 5.2

Input Forces Pf = 30.5 Pfb = 5	Nominal Value	Variation Range ($\pm$ factor)
VAD Compliance (cm ⁵ /N)	220	0.2 - 5.0
VAD Inertia (100kg)	0.0025	0.2 - 5.0
VAD Resistance (Ns/cm)	0.17	0.2 - 5.0
Output Cannulae Inertia (Ns ² /cm ⁵)	0.000263	0.1 - 5.0
Input Cannulae Inertia (Ns ² /cm ⁵)	0.000132	0.5 - 2.0
Cannulae Compliance (cm ⁵ /N)	44	0.2 - 5.0
Piston Area (cm ²)	19.27	0.5 - 2.0

NOTE: The output and input cannulae have the same compliance

Table 5.2 Nominal Design Parameters And Range Variation For Full-Fill/Full-Eject Mode

For this part of the study, all systemic parameters were kept constant at their nominal values. The other design parameters were set at the nominal values

It was observed that for a change in parameters that would affect the pressure, the heart rate would be changed while parameters not directly affecting the pressure did not have significant effects. This is because the operation of the VAD depended on the pressure gradient which in turn determined the blood flow rate and direction. It was also found that the full-fill/full-eject system was less tolerable to piston area changes as was found with the fixed rate controller.

The effects descriptions are also qualitative as in the previous section.

l_{cnn} (input cannulae inductance)

The VAD functioned well when l_{cnn} was varied within the range from 0.5 to 2 cm. This meant that the cannulae length was varied between 2.5 cm and 15 cm there was no significant observable changes VAD operation. The only parameters affected are the input cannulae pressure oscillation. With lower l_{cnn} , the oscillation frequency increased and vice versa for higher l_{cnn} . This meant that the length of the input cannulae could be chosen based solely on anatomical fit and the overall VAD operation would not be affected.

Figure 5.17 Graph Of Venous Circulation At $l_{cnn}=0.2$

Figure 5.18 Graph Of Venous Circulation At $l_{cnn}=2.0$

6.0 SYSTEMIC PARAMETER STUDIES AND SENSITIVITIES

The VAD-circulatory simulation model was also used for evaluating sensitivities of the system to variations in systemic parameters. By setting the systemic parameters at their nominal values as found in chapter 3, a set of design and control parameters was determined (see chapter 5). In normal human activities, the systemic parameters vary. The variations might be due to an increase or decrease in activity, emotional factors, environmental changes or many other reasons. However, for a VAD design to function properly, the VAD should work when the parameters are varied within physiologically plausible ranges and, if the VAD fails, the reasons for failure should be determined. In general, the full-fill/full-eject control mode is more sensitive to physiological changes than the fixed rate control mode. This was because the full-fill/full-eject mode depends on the interaction with the rest of the system more than the fixed rate mode. On the other hand, a VAD operating with fixed rate mode would be functioning with preset conditions and, only when necessary, would the operating setup be altered by trained personnel.

Besides studying the parameters and sensitivities of the VAD to parameter variations, the eigenvalues of the entire system were also investigated. The eigenvalue study was important for the design and control of the VAD. By examining the natural frequencies and damping factors of a system, the system response could be predicted in a general way. The major use of the eigenvalues was to gain an intuitive feel for the effects of changing the parameters on the response of the entire coupled system. Although eigenvalues could be used for general response prediction, they were not truly representative since the system was nonlinear due to changes in valve resistance between open and close positions. For the eigenvalue study, the nominal parameter values were used. Changes in the nominal system eigenvalues were then used as an indication of how the parameter changes would affect the system behavior.

Lenot (Output Cannulae Inertance)

The output cannulae length did not seem to have a great effect on the VAD performance. The range was varied between 0.1 to 5 times the nominal value and the main effect as on the output cannulae oscillation frequency and amplitude. At low lenot, the cannulae natural frequency was high with small amplitude and vice versa for high lenot value (long cannulae). With extremely long output cannulae, it had a more significant effect on the arterial pressure oscillation and this was not observed until the length was increased to over 6 times of its original value.

Figure 5.21 Graph Of Arterial Circulation At Lenot*0.1

Figure 5.22 Graph Of Arterial Circulation At Lenot*5.0

Cblo (Output Cannulae Compliance)

The change in output cannulae compliance had more influence on the VAD than the input cannulae compliance. The reason was that the change in compliance would affect the pressure in the output cannulae. This would then affect the switching between systole and diastole due to the time required for filling and emptying the VAD. The acceptable changes in Cblo are from 0.5 to 2 times the nominal value. With lower Cblo, the cannulae oscillation amplitude became large and the arterial pressure lowered, but the system would continue to function after going through approximately three cycles. The heart rate would increase by lowering Cblo. With higher Cblo, the oscillation amplitude are lowered. At the same time, the heart rate is decreased. From the observations made, it would indicate that as long as the Cblo lies within the examined range, the VAD should function normally.

Figure 5.23 Graph Of Arterial Circulation At $Cblo \cdot 0.5$

Figure 5.24 Graph Of Arterial Circulation At $Cblo \cdot 2.0$

Cht (VAD compliance)

The effect of varying Cht was a change in heart rate. In general, with a stiffer diaphragm (lower Cht), the heart rate would increase along with arterial pressure. When Cht was increased the heart rate would decrease. It was also found that the more flexible the diaphragm was, the lower the heart pressure and the efficiency would be. With higher Cht, the heart pressure curve would be smoother than if Cht was low. This was because at high Cht a change in volume would not cause a large change in pressure thus the heart pressure would increase gradually.

Figure 5.25 Graph Of Heart Pressure At Cht*0.2

Figure 5.26 Graph Of Heart Pressure At Cht*2.0

Iht (VAD Inertia)

The variations in Iht from factor of 0.2 to 5 did not affect the overall VAD performance. All parameters remained nearly constant including the efficiency. The observable effects were in the heart pressure curves and piston velocity. When Iht was increased from 0.2 to 5 times its nominal value, the piston velocity lowered and the heart pressure curve became rounder. This was because with a heavier piston, more effort was required to move it, thus the piston velocity would be expected to be slower with the same input force.

Figure 5.27 Graph Of Heart Pressure At Iht*0.2

Figure 5.28 Graph Of Heart Pressure At Iht*5.0

Rht (VAD Resistance)

The VAD resistance had significant influence on the VAD operation. The VAD was only operable when Rht was between half to double of its nominal value. At lower Rht, the motion of the piston and diaphragm was not highly restricted, thus the heart rate increased and so did heart pressure, arterial pressure and blood flow rate. When Rht was doubled, the heart rate, the arterial pressure and blood flow rate all decreased. This meant that with a set of controller parameters, Rht could be selected to tune the VAD to achieve desired performance.

Figure 5.29 Graph Of Heart Pressure At $Rht*0.5$

Figure 5.30 Graph Of Heart Pressure At $Rht*2.0$

A. (Piston Area)

Since the full fill/full-eject controller depended much on the controller parameter, the changes in piston area had a very large influence on VAD operation. When A was decreased, so did the heart rate and arterial pressure. It was the opposite when A was increased. Though the general behavior of the VAD as a result of changes in A could still be observed, it was so sensitive that any changes in A would require a new set of controller parameter values.

Figure 5.31 Graph Of Heart Pressure At $A*0.7$

Figure 5.32 Graph Of Heart Pressure At $A*1.5$

While comparing the behavior of the two controllers, it was found that the fixed rate mode was more tolerable to design parameter changes. The main difference between the two was the operation of the full-fill/full-eject controller would be affected more by parameters that were causing changes in pressure than the fixed rate controller. This meant that with proper parameter selection, a VAD with a full-fill/full-eject controller would be more flexible in design when a certain performance was required.

5.2 Synchronization To Residual Beating

The initial synchronization study was done exclusively for the full-fill/full-eject controller to examine the possibility of the VAD self-synchronizing to any residual arterial beating. This study was done to examine the effect and was not intended for trying to tune the VAD to synchronize to the natural circulatory system. Studying this effect could provide a guide to predicting how the VAD would behave when the arterial characteristic changes.

Although the initial study was done by varying both the oscillating frequency and residual beating pressure, the results were inconclusive. Further study should be performed to examine the possible effects of residual beating on the operation of the VAD.

6.0 SYSTEMIC PARAMETER STUDIES AND SENSITIVITIES

The VAD circulatory simulation model was also used for evaluating sensitivities of the system to variations in systemic parameters. By setting the systemic parameters at their nominal values as found in chapter 3, a set of design and control parameters was determined (see chapter 5). In normal human activities, the systemic parameters vary. The variations might be due to an increase or decrease in activity, emotional factors, environmental changes or many other reasons. However, for a VAD design to function properly, the VAD should work when the parameters are varied within physiologically plausible ranges and, if the VAD fails, the reasons for failure should be determined. In general, the full-fill/full-eject control mode is more sensitive to physiological changes than the fixed rate control mode. This was because the full-fill/full-eject mode depends on the interaction with the rest of the system more than the fixed rate mode. On the other hand, a VAD operating with fixed rate mode would be functioning with preset conditions and, only when necessary, would the operating setup be altered by trained personnel.

Besides studying the parameters and sensitivities of the VAD to parameter variations, the eigenvalues of the entire system were also investigated. The eigenvalue study was important for the design and control of the VAD. By examining the natural frequencies and damping factors of a system, the system response could be predicted in a general way. The major use of the eigenvalues was to gain an intuitive feel for the effects of changing the parameters on the response of the entire coupled system. Although eigenvalues could be used for general response prediction, they were not truly representative since the system was nonlinear due to changes in valve resistance between open and close positions. For the eigenvalue study, the nominal parameter values were used. Changes in the nominal system eigenvalues were then used as an indication of how the parameter changes would affect the system behavior.

In this chapter, the eigenvalue study is discussed. This chapter also includes discussions on the effects of systemic parameters on the VAD system incorporating both controllers.

6.1 Eigenvalue Study

The main reason for the eigenvalue study was, as mentioned before, to gain an intuitive feel for the effects of parameter variations on the system response. This study can also be used as a tool for tuning the system response in an attempt to increase device efficiency. The goal was to tune the system parameter so that one of the VAD's natural frequencies was close to the natural heart oscillation frequency. By achieving this, the VAD should require less power to drive. This technique was found to be especially useful for tuning the full-fill/full-eject controller. The eigenvalues for nominal parameter values are shown in figure 6.1 and figure 6.2.

Figure 6.1 Graph Of Fixed Rate Controller Nominal Parameter Eigenvalue

Figure 6.2 Graph Of Full-fill/Full-Eject Controller Nominal Parameter Eigenvalues

From the eigenvalue plots, the nominal parameter eigenvalues are listed in table 6.1. The method for determining the which eigenvalue was most closely related to which parameter was to vary the parameters by $\pm 15\%$ of the nominal value and examine which eigenvalues were affected the most. An example is the variation of the VAD incertance (lht) as shown in figure 6.3. It was found that the eigenvalues at $-117.4 \pm 53.69i$ was affected the most and thus those ones were determined most closely related to lht. The graphs showing the eigenvalue plots with parameter variations can be found in appendix 2.

Figure 6.3 Eigenvalue Plot With 15% Variation In lht

Most Closely Related Parameter	Fixed Rate Mode	Full-Fill/Full-Eject Mode
Ipe	-0.5233	-0.5057
Cvn	-0.1284	-0.2194
Cat	-4.671	-5.11
Icnin, Cbli	-4.468 - 34.24i -4.468 + 34.24i	-7.374 - 29.05i -7.374 + 29.05i
Icnot, Cblo	-10.61 - 35.68i -10.61 + 35.68i	-13.16 - 37.42i -13.16 + 37.42i
Cht, lht	-117.4 - 53.69i -117.4 + 53.69i	-44.88 - 8.447i -44.88 + 8.447i

Table 6.1 Nominal Parameter Eigenvalues

The eigenvalues were only meaningful if the system was steady and if it was linear. For this purpose, the input and output valves were both set open. Although the eigenvalues were not truly meaningful, nonetheless, they gave a rough indications to the response of the system. As mentioned before, the eigenvalue study was found to be more useful for predicting the behavior for full-fill/full-eject controller than for fixed rate controller. This was because that the operation of the full-fill/full-eject controller depended much more on the design parameters than the fixed rate controller.

One of the most noticeable effects was the change in VAD rate due to changes in Cht and lht and their affected eigenvalues. When the Cht was set at 3500 cm⁵/N, with lht set at 0.35 kg and Rht at 0.027 Ns/cm, the VAD natural frequency was at about 4 rad/sec. This value was close to the desired 1 rad/sec making the VAD natural frequency to be close to 60 beats per minute. Although the desired oscillation frequency was achieved, the design was proven to be not feasible. This was because that with Cht at 3500 cm⁵/N the VAD diaphragm was too flexible and it would not require much effort to push the diaphragm. With the piston pushing on the blood and the blood filling rate being much slower than the piston velocity, to achieve the 70 ml volume limit, the piston had to have unreasonably long travel distance of 3 m! In

order to keep the piston travel distance short, the entire diaphragm had to be moving uniformly. With increased surface area, the volume limit could be reached with shorter piston travel distance. To achieve this, C_{ht} was chosen to be $220 \text{ cm}^5/\text{N}$, which was a stiffer diaphragm design. At this C_{ht} value, the VAD performance was found to be satisfactory and the piston travel distance was decreased to 13 cm. It was therefore decided that it was not feasible to try to design the VAD, with the given conditions, to have a low oscillating frequency.

Trying to match the VAD's natural frequency to that of the natural heart could increase the overall system efficiency. When the efficiency was increased, power consumption would tend to decrease, thus making the design more desirable. Another use of the eigenvalues for determining the response of the system using the fixed rate controller was for examining the heart pressure curve. At higher frequency (higher eigenvalues), the pressure curve produced by the simulation would have a few fast oscillations. These oscillations were not desirable since they contribute to reduce the system efficiency. This reduction in system efficiency could be caused by the loss of energy to the high frequency oscillation in VAD's motion. By tuning the system (observing the eigenvalues), these oscillations could be lowered to match to the natural heart frequency as close as possible in an attempt to increase the system efficiency.

From the eigenvalue study, it was found that within the 15% parameter variation range, the system remained to function normally. It was also found that the systemic parameters (C_{at} , I_{pe} , R_{pe} , C_{vn}) had smaller effects on the eigenvalues than the other parameters directly relating to the VAD system. This meant, at the nominal design parameter values, the VAD would be quite tolerant to physiological changes. It also showed that the VAD system should remain operational within the allowable physiological variations.

The effects of cannulae parameter variations on the eigenvalues were not very significant. When changing C_{blo} , I_{enot} , C_{bli} and I_{enin} , the effects were limited to the eigenvalues relating to these parameters. This suggests that changing either the length or compliance of the cannulae would not greatly affect the VAD operation.

From the eigenvalue study, it was found that the design parameters had the greatest influence on the system as a whole. By changing Ch_t , Iht , Rht and A , all eigenvalues would be affected. This indicated that changing design parameters would change the entire system response. It also showed that the 15% change in design parameters had a larger effect than an equivalent change in systemic parameters. This suggested that the response of the VAD-circulatory system would depend more on the controller and VAD design rather than on physiological changes.

6.2 Systemic Parameter Effects For Fixed Rate Mode

The parameter study for the fixed rate mode controller was important since the VAD was controlled to operate at a particular heart rate and systolic fraction. Due to systemic variations, the preset control parameters may not be suitable and, thus, a set of working limits must be established to advise the physicians when the operating state must be altered. For the systemic parameter effect study the parameters under consideration and their variation ranges were shown in table 6.2. Although the valves used in the VAD did not belong to the natural circulatory system, their values were provided as a fixed variable, not as design variables, thus they were included in this part of the study.

	Nominal Value	Variation Range (* factor)
Output Valve Resistance (Ns/cm ⁵)	0.0008	0.2 - 10.0
Arterial Compliance (cm ⁵ /N)	133	0.2 - 5.0
Peripheral Resistance (Ns/cm ⁵)	0.013	0.2 - 5.0
Peripheral Inertia (Ns ² /cm ⁵)	0.0275	0.2 - 10.0
Venous Compliance (cm ⁵ /N)	3750	0.2 - 5.0
Input Valve Resistance (Ns/cm ⁵)	0.0008	0.2 - 10.0

Table 6.2 Nominal Systemic Parameters And Range Variation For Fixed Rate controller

For evaluation purposes, the design and control parameters were set at their nominal values and only the systemic parameters were altered to study their effects on VAD operation and the circulatory system. Also, to make the study clearer to evaluate, only one parameter was altered at a time. The nominal parameters used were:

1. Input force = 40 N
2. Base input force = 0 N
3. Heart rate = 60 beats/min
4. Systolic fraction = 0.30
5. VAD compliance = 28 cm⁵/N
6. Piston inertia = 0.0038 * 100kg
7. VAD resistance = 0.47 Ns/cm
8. Piston area = 19.27 cm²
9. Input cannulae inertia = 0.0000263 Ns²/cm⁵
10. Output cannulae inertia = 0.0000263 Ns²/cm⁵
11. Input cannulae compliance = 44 cm⁵/N
12. Output cannulae compliance = 44 cm⁵/N

At the nominal condition, the performance of the VAD was satisfactory. The VAD had an efficiency of 45.6% with maximum input power of 24.9 W and average input power of 2.717 W. For the evaluation purposes, the efficiency was defined as follows:

1. Efficiency = total systemic energy loss / total input energy
2. Total systemic energy loss = energy loss due to output valve resistance + energy loss due to output valve resistance + energy loss due to peripheral resistance
3. Energy loss = $\int \text{pressure} \times \text{flow rate} \, dt$

Figure 6.4 shows the performance of the VAD at steady state. The total circulation volume was 3.712 L. This volume was lower than that expected normal blood volume inside a human body and this was because that the pulmonary circulation (blood resident in the lungs) was not accounted for in this simulation model. The volume plot and the percentage by volume of blood in each part of the circulatory system is shown in figure 6.5.

Q = flow rate (L/min)
P = pressure (mmHg)
V = velocity (cm/s)

Figure 6.4 Plots Of VAD Performance With Fixed Rate Controller

Figure 6.5 Blood Circulation Volume For Fixed Rate Controller

From the parameter evaluations, it was found that in general, the VAD functioned within the described variation range. It was found that the VAD was most sensitive to peripheral resistance, which was expected, since, if the resistance was too high, it would restrict blood flow regardless of the VAD design. There would also be a problem in maintaining system pressure if the resistance was too low.

The effects of each parameter on the system performance are described in this section. The description is qualitative in order to understand trends in the system response.

Rin and Rot (Input And Output Valve Resistance)

It was found that with the variation of valve resistance between 0.2 and 10 times its nominal value, the VAD would still function as long as the input force is adjusted. By using the simulation model, it was found that the input valve affected venous return and output valve affected the artery pressure. There was no observable interaction between the two sides of the VAD. This could be because the VAD controller was dominating the response which would not be greatly affected by either parameter.

In general, with lower R_{in} , the venous pressure would drop with a larger venous oscillation amplitude. It was the opposite for higher R_{in} . Also by changing R_{in} , the rest of the circulatory system did not seem to be affected. R_{ot} only affected the arterial pressure and the output valve operation. With lower R_{ot} , the output flow was increased and the arterial pressure increased with larger oscillation amplitude. At higher R_{ot} , a higher input force would be required to pump the blood out of the VAD. Otherwise, the flow would be low and the arterial pressure would decrease.

Figure 6.6 Graph Of Venous Circulation At $R_{in}=0.2$

Figure 6.7 Graph Of Venous Circulation At $Rot*10.0$

Figure 6.8 Graph Of Arterial Circulation At $Rot*0.2$

Figure 6.9 Graph Of Arterial Circulation At $Rot*10.0$

Cat (Arterial Compliance)

Variations in arterial compliance could due to many factors including mental state, activity and environment. This parameter would change continuously since this would be one natural way of maintaining systemic pressure and keeping the blood circulating. At Cht of 0.2 times the nominal value, the heart pressure curve was very smooth and there was a large difference between the systolic and diastolic arterial pressure. At this operating point, the other parameters did not have significant changes. At the other extreme end with 2 times the nominal Cat, the pressure difference between the systolic and diastolic arterial pressure was smaller and the oscillation amplitude was increased. With the increase in Cat, the VAD efficiency decreased and so did venous pressure. At this operating condition the VAD would still have satisfactory performance.

Figure 6.10 Graph Of Arterial Circulation At Cat*0.2

Figure 6.11 Graph Of Arterial Circulation At Cat*2.0

I_{pc} (Peripheral Inertia)

With changes in I_{pc} , the venous pressure, arterial pressure and heart pressure were all influenced. But the effects were not very great and the VAD functioned properly when the variation was between a factor of 0.2 to 5.0. When I_{pc} was lowered, the peripheral flow had a small oscillation. Other than an increase in VAD efficiency, there were no other significant differences in performance. When I_{pc} was increased, both the arterial and VAD pressures increased. The difference between the systolic and diastolic arterial pressure was smaller. Also, the peripheral flow was increased thus causing the venous pressure to decrease slightly. Overall, I_{pc} was not a major factor in the VAD performance.

Figure 6.12 Graph Of Peripheral Circulation At $I_{pc}=0.2$

Figure 6.13 Graph Of VAD Performance At $I_{pe} \times 5.0$

R_{pe} (Peripheral Resistance)

The peripheral resistance played a major role in the functioning of the entire circulatory system. R_{pe} could be affected by factors like organ blood supply requirement, blood viscosity, blood liquid content and other factors relating to body activity and environment. Since R_{pe} acted as a restricter of the blood flow, if R_{pe} was too high, regardless of the VAD design, the system would fail. Although for the parameter sensitivity study, the variation range factor for R_{pe} was from 0.1 to 5, the system only functioned properly with a factor between 0.5 and 2. This range would fall within the normal physiological allowable range thus there would not be any concern that the VAD would cease to operate within the limit. When R_{pe} was lowered by half, the flow rate increase. At the same time, the venous pressure increased and the arterial pressure decreased. Although input forces can be increased to push the arterial pressure up, this would increase both the flow rate and venous pressure beyond allowable limits. This operating point was considered as the bottom range limit. When R_{pe} was doubled, the flow rate decreased to nearly the lowest allowable limit of about 3 L/min. The arterial pressure increased but remained within the normal physiological limits, but the difference between the systolic and diastolic arterial pressure decreased.

Figure 6.14 Graph Of Peripheral Circulation At R_{pe}=0.5

Figure 6.15 Graph Of Peripheral Circulation At $R_{pc} \times 2.0$

C_{vn} (Venous Compliance)

The variations in C_{vn} did not have a great significant on the system performance, except when C_{vn} was very high. At 5 times the nominal C_{vn} value (very flexible), the arterial and peripheral system worked as usual, but because of the high venous flexibility, the venous pressure was not high enough to fill the VAD chamber. Eventually, the VAD would fail to operate. For ranges between a factor of 0.2 to 2, the system continued to perform normally. With a lower C_{vn} value (stiffer), the venous pressure increased with larger venous pressure oscillations. When the value was increased, the flow rate increased and the venous pressure decreased, but other than those effects, no other apparent significant effects were observed.

Figure 6.16 Graph Of Venous Circulation At C_{vn}*5.0

Figure 6.17 Graph Of Venous Circulation At C_{vn}*0.2

Figure 6.18 Graph Of VAD Performance At Cvn*2.0

Overall, the VAD and the circulatory system would function satisfactorily when the systemic parameters were varied within the normal physiological limits. The input force may have to be adjusted to compensate for some of the extreme operating conditions, which would also affect the power consumption and the efficiency. From the observations made, peripheral resistance would be the dominant parameter that could cause the VAD to cease to operate regardless of the design. Otherwise, VAD design could be altered to accommodate other operating limits.

6.3 Systemic Parameter Effects For Full-Fill/Full-Eject Mode

The systemic variations were important for altering the operation of the fixed rate mode controller, but it was found that the full-fill/full-eject controller was much more sensitive to the changes. This was because the operation of the full-fill/full-eject controller depended more strongly on the parameter selections.

For the systemic parameter sensitivity study, the parameters under consideration and their variation ranges were shown in table 6.3. As was done for the fixed rate controller, the input and output valve resistances were included as part of the systemic parameters.

	Nominal Value	Variation Range (* factor)
Output Valve Resistance (Ns/cm ⁵)	0.0008	0.2 - 5.0
Arterial Compliance (cm ⁵ /N)	133	0.2 - 2.0
Peripheral Resistance (Ns/cm ⁵)	0.013	0.2 - 5.0
Peripheral Inertia (Ns ² /cm ⁵)	0.0275	0.2 - 5.0
Venous Compliance (cm ⁵ /N)	3750	0.2 - 2.0
Input Valve Resistance (Ns/cm ⁵)	0.0008	0.2 - 5.0

Table 6.3 Nominal Systemic Parameters And Range Variation For full-Fill/Full-Eject Controller

As was also done for the fixed rate controller, the design and control parameters were set at their nominal values and only one of the systemic parameters was varied at a time. The nominal parameters used were as follows:

1. Input force = 30.5 N
2. Base input force = 5 N
3. Maximum VAD Volume = 70 mL
4. Minimum VAD Volume = 5 mL
5. VAD compliance = 220 cm⁵/N
6. Piston inertia = 0.0025 * 100kg
7. VAD resistance = 0.17 Ns/cm
8. Piston area = 19.27 cm²
9. Input cannulae inertia = 0.0000263 Ns²/cm⁵
10. Output cannulae inertia = 0.0000263 Ns²/cm⁵
11. Input cannulae compliance = 44 cm⁵/N
12. Output cannulae compliance = 44 cm⁵/N

At the nominal condition, the VAD performance was satisfactory. The efficiency was 15.22% with a maximum input power of 48.41 W and average input power of 5.681 W. The low efficiency, as compared to the efficiency when fixed rate mode was used, was caused by extra input energy required to drive the more flexible VAD diaphragm used (which was less sensitive to volume changes). Figure 6.19 shows the performance of the VAD at steady state. The total circulation volume was 4.485 L. The volume plot and the percentage by volume of blood in each part of the circulatory system is shown in figure 6.20.

Q = flow rate(L/min)
P = pressure (mmHg)
V = velocity (cm/s)

Figure 6.19 Plots Of VAD Performance With Full Fill/Full Eject Controller

Figure 6.20 Blood Circulation Volume For Full-Fill/Full Eject Controller

From the parameter evaluations, it was found that, in general, the VAD functioned within the described variation range. But, it also was found that the VAD was quite sensitive, overall, to systemic parameters. The effects were more prominent with the arterial and venous compliance. Since the compliance would directly affect the pressure, it would greatly affect the VAD operation which was based on volume flow rate and blood pressure gradients. It was also found that by varying the parameters, the heart rate also changed. The explanation was that since the diastole and systole depended much on pressure differences across the valves, any changes that would affect the systemic pressure and flow rate would also affect the heart rate.

The VAD was found to adjust itself to the body's needs. For example, as when the arterial compliance and peripheral resistance were lowered, the heart rate of the VAD increased. This would be necessary for the real heart since the lowering of compliance and resistance would mean that the body requires more circulation for supplying oxygen and nutrients to other parts of the body. By increasing the VAD pumping rate, these demands could be met automatically without external adjustments. This would also stand true on the opposite end of the body requirements. When a person was relaxed, the compliance and resistance would increase since there would be no need for increased blood flow. At this condition, the VAD would also slow down as the natural heart would have done in the same situation.

The effects of each parameter on the system's performance are described in this section. The description is a qualitative description as in the previous section.

R_{in} and R_{ot} (Input And Output Valve Resistance)

The variations in valve resistances seemed to have the least effects on the VAD when compared to other systemic parameters. Both valve resistances were varied by factors from 0.2 to 5. For the input valve resistance, when it was set at the lower limit, the arterial pressure increased and so did the venous pressure amplitude. The heart rate increased from 60 beats per minute to 70, and the flow rate also increased. But with increased valve resistance, the heart rate and arterial pressure would both drop. The venous oscillation amplitude also decreased at the same time.

When the output valve resistance was lowered, the effects were quite small apart from larger arterial pressure oscillations. With higher R_{ot} values, the heart rate and arterial pressure would decrease. In effect, the output had a smaller influence than the input valve but both parameters would not greatly affect the VAD operation.

Figure 6.21 Graph Of Heart Pressure At $R_{in}=0.2$

Figure 6.22 Graph Of VAD Performance At $Rm^*5.0$

Figure 6.23 Graph Of Arterial Circulation At Rot=0.2

Figure 6.24 Graph Of Arterial Circulation At Rot=5.0

Cat (Arterial Compliance)

The VAD was found to be very sensitive to arterial compliance changes. The functional range was between 0.5 to 1.2 times the nominal value. This was because that the arterial pressure would affect how the VAD behaves. At lower Cat values, the venous pressure and flow rate would increase. The arterial pressure was lowered. At this condition, the VAD would have to go through many cycles before a steady state could be reached.

When the Cat was increased to 1.2 times the nominal value, the VAD operated much like when Cat was set at its nominal value but the flow rate decreased slightly. The system would also go through about two cycles before reaching steady state.

Figure 6.25 Graph Of VAD Performance At Cat 0.5

Figure 6.26 Graph 01 Arterial Circulation At Cat 12

Ipc (Peripheral Inertia)

It was found that when the Ipc value was varied from 0.2 to 2 times the nominal value, the VAD functioned quite well. With lower Ipc, the peripheral flow rate was oscillating. But other than that, there were no significant effects. When Ipc was doubled, the arterial pressure increased and venous oscillation frequency increased but the efficiency went down. What was interesting was that if Ipc was increased to more than double its nominal value, the system response changes drastically. At a factor of 5, the arterial pressure differences was unacceptably small. Also, the heart rate decreased. At this point and beyond, the VAD would cease to operate normally.

Figure 6.28 Graph Of Peripheral Circulation At $I_{pc} \times 0.2$

Figure 6.28 Graph Of VAD Performance At Ipe 2.0

Figure 6.29 Graph of VAD Performance At $I_{pe} \times 5.0$

R_{pc} (Peripheral Resistance)

Unlike the I_{pc}, the limit for R_{pc} variation factor was between 0.5 and 2 but this limit was close to normal body allowable R_{pc} deviation. It was observed that at a lower R_{pc}, the heart rate would increase. The flow rate and venous pressure also increased, but arterial pressure decreased. When R_{pc} was increased to 2 times the normal value, the heart rate decreased to 50 beats per minute and the flow rate was also decreased. Although the average arterial pressure was increased, the difference between peak to peak values was decreased.

Figure 6.30 Graph Of VAD Performance At $R_{pe} \times 0.2$

Figure 6.31 Graph Of VAD Performance At $Rpe \times 2.0$

C_{vn} (Venous Compliance)

It was found from the observations that the full-fill/full-eject controller was very sensitive to changes in venous compliance. This was because the venous compliance would affect the venous pressure, which dictated the filling rate and thus the VAD operating cycle. The VAD was found to be greatly affected even when the C_{vn} was only at 1.2 times the nominal value. At this value, the heart rate was decreased to 30 beats per minute and both the arterial pressure and flow rate decreased considerably. This suggested that C_{vn} variations must be examined carefully while designing a VAD with a full-fill/full-eject controller.

Figure 6.32 Graph Of VAD Performance At $Cn=1$

Overall, the VAD and the circulatory system would function satisfactorily when the systemic parameters were varied within the normal physiological limits with the exception of the venous compliance. To accommodate the extreme variations in systemic parameters, the VAD controller parameters might have to be modified considerably before satisfactory performance could be obtained.

6.4 Summary Of Systemic Parameter Effects On The System

From the systemic parameter studies with the design and control parameters fixed, it was found that the operating range for a fixed rate controller was greater than that of the full-fill/full-eject controller. This finding was as expected since the beating of the VAD using fixed rate control depended mostly on the preset heart rate and systolic fraction and the full-fill/full-eject control depended on the variations in the parameters. Table 6.4 outlined the approximate working limits of each control method in terms of multiple factor of the nominal systemic parameter values.

	Fixed Rate (x nominal value)	Full-Fill/Full-Eject (x nominal value)
Output Valve Resistance (Ns/cm ⁵)	0.2 - 5.0	0.2 - 5.0
Arterial Compliance (cm ⁵ /N)	0.2 - 2.0	0.5 - 1.2
Peripheral Resistance (Ns/cm ⁵)	0.5 - 2.0	0.5 - 2.0
Peripheral Inertia (Ns ² /cm ⁵)	0.2 - 10.0	0.2 - 2.0
Venous Compliance (cm ⁵ /N)	0.2 - 2.0	0.2 - 1.2
Input Valve Resistance (Ns/cm ⁵)	0.2 - 10.0	0.2 - 5.0

Table 6.4 Variation Limit Factors Of Systemic Parameters

While determining the effects of the systemic parameters variations, the VAD system would be considered to fail when the VAD ceased to function normally. This meant that either the VAD was not providing high enough systemic pressure or the systemic flow was too low or the VAD was not beating at the desired heart rate (this only applied to full-fill/full-eject mode). It was also found from this study that although the design and control parameters could be kept at the nominal values, the VAD could work better if the input forces were to be adjusted, as in the case of increased valve resistance.

As a comparison between the two controlling method, it was found that fixed rate controlled VAD was much more tolerant to physiological changes. The fixed rate controller would also be quite easy to operate if the systolic fraction or the heart rate must be changed to accommodate extreme conditions. This does not mean that full-fill/full-eject would not be as good as the fixed rate controller. On the contrary, since the heart rate of the VAD was self adjustable to physiological changes, as observed, a carefully designed and tuned VAD with a full-fill/full-eject controller, may be a closer "clone" of the real heart than a VAD with a fixed rate controller was used.

7.0 CONCLUSION AND RECOMMENDATIONS

7.1 Thesis Goal And Background Summary

The goal of developing the VAD-circulatory model was to build a simulation model as a basis for dynamic determination of circulatory assist device design parameters and to examine a coupled control approach for controlling the operation of the VAD.

The model can be used as the foundation for modeling the physical system and testing various design ideas quickly and easily without the conventional restraints of static models. The selection of parameters and evaluation of the designs were based on the following criteria:

1. Highest possible energy efficiency defined as the ratio of average power in to the average power to the blood (presently achievable value is 12% to 30%),
2. reasonable results for maintaining the normal functioning of the human body (nominal blood pressure of 80 to 120 mmHg, average blood flow rate of 5 L/min and nominal heart rate of 60 beats per minute),
3. quick and inexpensive method (as compared to live subject experiments) of modeling the natural circulatory system, and
4. reasonable and manufacturable VAD design parameter values (e.g., heart piston inertance of 200g and piston area 19 cm²).

The developed simulation model would also allow the research teams to determine the geometric effect of the cannulae and the feasibility of various control methods. The major advantages of this model over other conventional static models are its ability to perform dynamic parameter studies and its ability to produce a "computer simulated" prototype efficiently. By performing dynamic analysis, the interactions between the basic controller parameters and the physiological parameters could be easily evaluated. The computer simulated prototype will save time and money for the development of new concepts.

The VAD-circulatory system simulation model was developed by using the bond-graph technique and was based on the Matlab package. The simulation model is a generic VAD-circulatory system composed of basic compliance, inertance and resistance elements. By using the bond-graph technique and these basic elements, the model was described by nine first order differential equations in state space form. The simulation program was written to be run on a UNIX version of MATLAB. It was written in a modular fashion so that the controller modules and the output modules could be tailored to suit the design requirements defined by the designer (i.e., the values defined in chapter 4). A listing of the program can be found in Appendix 1.

The simulation model is based on two types of controller: fixed rate and full-fill/full-eject. The fixed rate controller forced the VAD to operate at a preset heart rate and systolic fraction, which are independent of the natural physiological state. The full-fill/full-eject controller depends on the working state of the natural circulation system. The switching between systole and diastole was controlled by the blood filling rate and preset VAD volume limits. The filling rate was in turn determined by the design parameters, as listed in chapter 4, and the natural heart rate.

7.2 Conclusions

To evaluate the feasibility of the design and to determine the working limits of each controller, the VAD-circulatory simulation model was setup to examine its response to each control method over various parameter ranges. The studies performed were:

1. Eigenvalue studies for determining the sensitivity of the system performance to various parameters,
2. systemic parameter effect studies on the controllers for determining the working limit of the device,
3. studies on the sensitivity of the control methods to the controller parameters to evaluate design range of the VAD parameters,
4. a study of the effect of the cannulae on system performance to determine reasonable cannulae lengths, and

5. a study of the synchronization of full-fill/full-eject controller to residual arterial beating to evaluate the controller's performance.

For building the simulation model, the Penn State University VAD model was consulted for determining the starting points for selecting design and systemic parameters. The systemic parameters were then compared to known published physiological values, as found in chapter 3, for refinement. The control parameters were largely based on the OHI VAD and the Penn State University VAD. The OHI VAD provided the basic circulatory loop configurations and the basic VAD design. The VAD also provided the parameters required for determining the geometric influence of the cannulae. From the Penn State University VAD, the starting point for choosing the VAD compliance, resistance and inertance was determined and these values are found in chapter 2. The selected parameter values were then used in the simulation and were adjusted until reasonable physiological conditions were met. These parameter values can be found in chapter 4.

From the various studies performed, several conclusions can be made:

1. From the eigenvalue study, the VAD, with either control method, was found to be stable over the allowable parameter ranges.
2. From the systemic parameter study, it was found that both control methods were feasible and robust within normal limits. But it was also found that the fixed rate mode would be the more reliable of the two since it was more tolerant to parameter variations.
3. From the control parameter study it was found that within reasonable variation limits both control methods worked properly. Again, the fixed rate might be more reliable since it kept the VAD operating at 60 beats per minute regardless of physiological changes. The heart rate for full-fill/full-eject control fluctuated with the parameters and the input pressures and initial conditions must be carefully tuned to obtain steady results.
4. For the same working environment, the fixed rate controller seemed to be more efficient than the full-fill/full-eject controller.
5. From the residual beat study, the results were inconclusive. Further study is required to examine this effect.

6. From the cannulae effect evaluations, it was found that the length of the cannulae affected only the pressure oscillations and, as long as the length values, are kept within factor of five of typical values, there would be no significant effects on the VAD performance.

From the evaluations performed, it can also be concluded that the developed model did match the goal of the project by providing a dynamic analysis and fast prototyping. This model should be sufficient for most design studies with minor modifications to suit different requirements. Although this model can also be modified to accommodate other VAD designs around the world, those parameters are proprietary information and thus verification of the model against other designs is not possible at this point.

7.3 Future Improvement And Recommendations

In order to make the existing VAD-circulatory model more complete and more versatile, the following can be used as a guideline for future modification.

1. Modify the model to accommodate TAH designs by adding heart valves and additional chambers. This can possibly be done by treating the TAH as two separate chambers, each with a certain compliance value and inertance value, and treating the pulmonary circulation as a combination of inertance and resistance.
2. Add pulmonary circulation for evaluation of its effect which sometimes is crucial to maintaining normal physiological conditions.
3. Add details, such as motor control and feedback control, to the controller modules to allow investigations with synchronous controllers and continuous flow pump designs.
4. Improve the VAD-circulatory model by adding more basic elements, such as the compliance and resistance of different organs, to simulate systemic circulation more accurately.
5. Add components to model specific actuators for improving the level of detail of the simulation.
6. Perform studies on the possible effect of residual heating on the performance of the VAD. The present model already contains a force

source at the artery which simulates residual beating of the artery. The study should be performed with small oscillation and pressure variations to examine possible synchronization of the VAD to the residual beat.

The suggestions above are mainly aimed at improving the accuracy and versatility of the model. The modified model will give an even better representation of the physiological system but at the same time, the complexity of the model will be greatly increased. The increased complexity may be offset by the increased accuracy and the advantages should be weighed to produce the most useful model.

The program for the simulation model can also be improved. Firstly, the program can be done in an object oriented fashion to make the model more modifiable. Secondly, it would be ideal if the program can be written to run more efficiently. Presently, due to speed and complexity constraints, the simulation must be run on a UNIX platform with high speed processor for satisfactory performance. The program might be modified or rewritten in other programming languages, if necessary, so it can be run on a 486PC computer.

As a final recommendation, a working prototype based on the VAD-circulatory model should be built to prove the success of the development of this simulation model.

REFERENCES

1. Antaki J. F. et al. 1993, In Vivo Evaluation of the Nimbus Axial Flow Ventricular Assist System: Criteria and Methods, ASAIO Journal, 39:M231-236
2. Arabia M., Akutsu T., 1984, A New Test Circulatory System for Research in Cardiovascular Engineering, Annals of Biomedical Engineering, vol. 12, p.29-48
3. Auslander D. M., Lobdell T. E., Chong D., 1972, A Large-Scale Model of the Human Cardiovascular System and its Application to Ballistocardiography, Transactions of the ASME, September, p.230-238
4. Brooks J., 1993, The Heart of the Matter: Dalton Camp and His Controversial Transplant, Canadian Medical Association Journal, Oct.1, V. 149(7), p.996-1003
5. Cohen D. J. et al., 1992, Effect of Synchronous and Asynchronous Pulsatile Flow During Left, Right, and Biventricular Bypass, Artificial Organs 16(6):614-622
6. Davis P. K. et al, 1989, Current Status of Permanent Total Artificial Hearts, Annals of Thoracic Surgeons, 47:172-178
7. Diegel, P. D., Mussivand T. et al., 1992, Electrohydraulic Ventricular Assist Device Development, ASAIO Journal, 38: M306-M310
8. Donovan F. H. Jr., 1974, Design of a Hydraulic Analog of the Circulatory System for Evaluating Artificial Hearts, The Universities Center - Jackson.
9. Farrar D. J., Hill J. D., 1993, Univentricular and Biventricular Thoratec VAD Support as a Bridge to Transplantation, The Society of thoracic Surgeons, Nov. 55:276-82
10. Fung Y. C., 1984, Biodynamics: Circulation, Springer-Verlag, New York
11. Gao H., et al., 1992, In Vitro Assessment of the Milwaukee Heart and Right to Left Balance, ASAIO Journal 38:M722-M725
12. Guyton A. C. M.D., 1993, Human Physiology and Mechanisms of Disease, Philadelphia: W.B. Saunders Company, 5th edition
13. Imachi K. et al., 1992, The Second and Third Model of the Flow Transformed Pulsatile Total Artificial Heart, ASAIO Journal, 38:M717-M721
14. Irie H. et al, 1992, Initial In Vivo Tests of an Electrohydraulic Actuated Total Artificial Heart, ASAIO Journal 38:M497-M500

15. Ivanovich S. V. et al, 1991, Present and Future of the Artificial Heart Driving System. Artificial Organs 15(5):376-380
16. Jarvis P. et al, 1991, An In Vitro Evaluation of an Artificial Heart. Trans AM Soc Artif Internal Organs XXXVII:21-32
17. Johnson K. E. et al, 1992, Use of Total Artificial Hearts: Summary of World Experience, 1969-1991. ASAIO Journal, 38:M486-M492
18. Johnson K. E. et al, 1990, Summary of the Clinical Use of the Symbion Total Artificial Heart: A Registry Report. Journal of Heart Transplantation, Sept-Oct, p.103-116
19. Kantrowitz A., 1993, Early Mechanical Left Ventricular Assistance. ASAIO Journal Oct.-Dec.:834-839
20. Karnopp D. C. et al, 1990, System Dynamics: A Unified Approach. Second Edition, John Wiley & Sons, Inc., New York
21. Kaufmann R. et al, 1992, Electromechanical Artificial Heart wit a New Gear Type and Angled Pump Chambers. The International Journal of Artificial Organs 15 no. 8:481-487
22. Kijima T. et al, 1993, A Straight Path Centrifugal Blood Pump Concept in the Capiox Centrifugal Pump. Artificial Organs 17(7):593-598
23. Kim H. C., Min B. G., 1992, Cardiac Output Regulation in the Moving Actuator Total Artificial Heart Without a Compliance Chamber. ASAIO Journal 38:846-850
24. Klute G. K. et al, 1992, An Optimal Controller for an Electric Ventricular Assist Device: Theory, Implementation, and Testing. IEEE Transactions On Biomedical Engineering, Vol. 39, No. 4, April, 394-403
25. Kolff A. et al, 1992, The Soft-Shell Mushroom Heart Remembered. Artificial Organs 15(3):225-240
26. Kolff W. J., 1993, The Dawn of Counterpulsation Muscle and Pneumatic Powered LVADs. ASAIO Journal Oct.-Dec.:825-827
27. Kolff W.J., 1990, The Invention of the Artificial Heart. The International Journal of Artificial Organs 13 no. 7:396-403
28. Konstantinov B.A. et al, 1991, Total Artificial Heart without Valves: Principles of Design and Implantation Technique. Artificial Organs 15(5):369-371
29. Koshal A. et al, 1991, Mechanical Bridge to Cardiac Trnasplant: Where Do We Stand in 1990? Canadian Society of Cardiovascular and Thoracic Surgeons 35 No. 6 December:578-580

30. Lick S. et al, 1993, Use of the Symbion Biventricular Assist Device in Bridging to Transplantation. Annals of Thoracic Surgeons, 55:283-287
31. Masuzawa T. et al, 1992, A Motor Integrated Regenerative Pump as the Actuator of an Electrohydraulic Totally Implantable Artificial Heart. ASAIO Journal 38:M232-M236
32. Miller G. E. et al, 1993, Evaluation of a Multiple disk Centrifugal Pump as an Artificial Ventricle. Artificial Organs, 17(7):590-592
33. Min B. G. et al, 1991, Development of a New Moving-Actuator Type Electromechanical Total Artificial Heart. Artificial Organs 15(2):144-146
34. Min B. et al, 1993, Different Stroke Volumes for the Left and Right Ventricles in the Moving-Actuator Type Total Artificial Heart. The International Journal of Artificial Organs 16 no. 1:45-50
35. Mussivand T. V., Personal Correspondence, 1996
36. Mussivand T. V. et al, 1996, Totally Implantable Intrathoracic Ventricular Assist Device. The Society of Thoracic Surgeons, 61:444-447
37. Naito K. et al, 1993, Development of the Baylor-Nikkiso Centrifugal Pump with a Purging System for Circulatory Support. Artificial Organs, 17(7):614-618
38. Nakatani T. et al, 1993, An Implantable Left Ventricular Assist System with an Electrical Impedance Monitoring and Control System. ASAIO Journal 39:M644-M648
39. Nardi G. et al, 1981, Modular Hydropneumatic Mock Circulatory system for the Evaluation of Cardiovascular Prostheses. ASAIO Journal, Vol. 4, October/December, p.139-148
40. Ohara Y. et al, 1993, Baylor Gyro Pump: A Completely Seal-less Centrifugal Pump Aiming for Long-Term Circulatory Support. Artificial Organs, 17(7):599-604
41. Olsen D. B., 1992, Artificial Organs of the Future. ASAIO Journal 38:M134-M138
42. Orime Y. et al, 1992, Versatile One-Piece Total Artificial Heart for Bridge to Transplantation or Permanent Heart Replacement. Artificial Organs, 16(6):607-613
43. Orime Y. et al, 1993, The Baylor-ABI Electromechanical Total Artificial Heart Accelerated Endurance Testing. ASAIO Journal 39:M172-M176
44. Pantalos G. M., 1993, Artificial Heart: Past, Present, and Future. Artificial Organs, 17(10):826-827

45. Qian K.X., 1990, New Investigations of a Pulsatile Impeller Blood Pump, ASAIO Transactions 36:33-35
46. Reibson J. D. et al, 1993, An Annular Compliance Chamber for the Pennsylvania State University Electric Total Artificial Heart, ASAIO Journal, 39: M415-M418
47. Rintoul T. C. et al, 1993, Continuing Development of the Cleveland Clinic-nunbus Total Artificial Heart, ASAIO Journal 39:M168-M171
48. Rosenberg G. et al, 1981, Design and Evaluation of the Pennsylvania State University Mock Circulatory System, ASAIO Journal, April/June, p.41-43
49. Rowles J. R. et al, 1992, Development of a Totally Implantable Artificial Heart, ASAIO Journal 38:M713-M716
50. Rowles J. R. et al., 1993, Ventricular Assist and Total Artificial Heart Devices for Clinical Use in 1993, ASAIO Journal Oct.-Dec.: 840-855
51. Rokitsansky A. et al., 1991, The New Small Viennese Total Artificial Heart: Experimental and First Clinical Experiences, Artificial Organs 15(2):129-135
52. Ruchti T. L. et al., 1993, Identification Algorithm for Systemic Arterial Parameters with Application to Total Artificial Heart Control, Annals of Biomedical Engineering 21:221-236
53. Sasaki E. et al., 1992, A Skeletal Muscle Actuator for an Artificial Heart, ASAIO Journal 38:M507-M511
54. Skalak R. et al, 1987, Handbook of Bioengineering, USA:McGraw-Hill Book
55. Smulders Y. M. et al., 1991, Concept of a Soft, Compressible Artificial Ventricle Under Evaluation, Artificial Organs 15(2):96-102
56. Snyder A. J. et al., 1992, Noninvasive Control of Cardiac Output for Alternately Ejecting dual-Pusherplate Pumps, Artificial Organs 16(2):189-194
57. Snyder A. J. et al., 1992, An Electrically Powered Total Artificial Heart, ASAIO Journal 38:M707-M712
58. Takano H. et al., 1993, Clinical Experience With Ventricular Assist Systems in Japan, Annals of Thoracic Surgeons, 55:250-256
59. Takatani S. et al., 1992, Development of a Totally Implantable Electromechanical Total Artificial Heart: Baylor TAH, Artificial Organs 16(4):398-406

60. Tatsumi E. et al., 1992, A Blood Pump with an Interatrial Shunt for Use as an Electrohydraulic Total Artificial Heart, ASAIO Journal 38:M425-M430
61. Tortora G. J., Anagnostakos N. P., 1984, Principles of Anatomy and Physiology, Fourth Edition, Harper Row Publishers, New York
62. Trinkl J. et al., 1993, Control of a Potary Pulsatile Cardiac Assist Pump Driven by an Electric Motor Without a Pressure Sensor to Avoid Collapse of the Pump Inlet, ASAIO Journal 39:M237-M241
63. Tsach U. et al., 1989, A Novel Output Feedback Pusher Plate Controller for the Penn State Electric Ventricular Assist Device, Journal of Dynamic Systems, Measurement, and Control, Vol.111, March
64. University of Utah, 1993, Proposal for a Technology License and/or Industrial Partnership, Artificial Heart Research Laboratory Institute for Biomedical Engineering, University of Utah, Salt Lake City, Utah, May
65. Uyama C., Akutsu T., 1991, Three-Dimensional Interface Geometry of the Human Heart with the Artificial Heart, ASAIO Transactions, 37:608-614
66. Vakamudi A.K., Akutsu T., 1994, Control Systems for the Total Artificial Heart, In: p. 372-381
67. Vaska P. L., 1991, Biventricular Assist Devices, Critical Care Nurse Vol. 11 No. 8:52-60
68. Weiss W. J. et al, 1993, A Completely Implanted Left Ventricular Assist Device: Chronic In Vivo Testing, ASAIO Journal 39:M427-M432
69. White K. R. et al, 1991, Control Systems for the Electric Heart, Artificial Organs 15(5):427-429
70. Yamazaki K. et al, 1993, Development of a Miniature Intraventricular Axial Flow Pump, ASAIO Journal 39:M224-M230
71. Yoshizawa M. et al, 1992, An Automatic Control Algorithm for the Optimal Driving of the Ventricular-Assist Device, IEEE Transactions on Biomedical Engineering Vol. 39 No. 3:243-251
72. Yu Long-Sheng et al., 1993, A Compact and Noise Free Electrohydraulic Total Artificial Heart, ASAIO Journal 39:M386-M391

```

x0(11)=0;
else %FFFE MODE
x0=zeros(11,1);
x0(1)=0.0; % pump momentum 0
x0(2)=111.2; % heart chamber volume (cm^3) 111.2
x0(3)=289.4; % output canula volume (cm^3) 289.4
x0(4)=0.0; % output canula momentum 0
x0(5)=715.9; % arterial volume (cm^3) 715.9
x0(6)=1.9; % peripheral momentum 1.9
x0(7)=3169; % venous volume (cm^3) 3169
x0(8)=0.0; % input canula momentum 0
x0(9)=134.9; % input canula volume (cm^3) 134.9
x0(10)=5.55; % piston position 5.55
x0(11)=0;
end

```

```

%INTEGRATION USING RUNGE-KUTTA 23 METHOD

```

```

if mode == 1
[tt,xx]=ode23('vadr5',t0,t1,x0);
else
[tt,xx]=ode23('vadr5',t0,t1,x0,1e-07);
end

```

```

%HEADING TO BE DISPLAYED ON GRAPH

```

```

if mode == 1
ts='AT NOMINAL';
p='FIXED RATE';
W='Rot 10';
else
p='FFFE MODE';
W='NOMINAL';
end

```

```

%CALLS OUTPUT ROUTINE

```

```

optn5
%flow

```

```

%INITIAL CONDITION FOR NEXT TIME SET SIMULATION

```

```

xx(max(length(xx)),:)
%pause;
%yy(.8)

```

VADSIMS.M

%Setup part for VAD model with in-out cannulae taken into account

%constant variables

% August 8,1995

clear

global A Cva Rpe lpe Cat Cht btoff atoff ocanoff vnoff icanoff lcnof lcnin lht Rot Rin Pf Pfb Rht Cblo
Cbli rm beat sysfrt tst piston

global push max_bt_vol min_bt_vol mode force count P w p W ped pcs told ht_vol voltest diastole systole

%all units are in cm base and in SI units

%CONTROL METHOD 1=fixed rate 2=full-fill/full-eject

mode = input('enter: 1 for fixed rate, 2 for full-fill/full eject -');

pcd=0; %DIASTOLIC COUNTER

pcs=0; %SYSTOLIC COUNTER

count=0;

systole = 1;

diastole = 0;

%INITIAL INPUT FORCE CONDITION FOR FIXED RATE MODE

beat = 60; %[beats/min] % NUMBER OF HEART BEATS PER MINUTE

sysfrt = 0.30; %[*period] % SYSTOLIC FRACTION

%DETERMINING THE REQUIRED INPUT FORCE

if mode == 1

Pf = 40; Pfb = 0; %FIXED RATE MODE=40,0

else

Pf = 30.5; Pfb = 5; %FFFE MODE=31.5

end

%PARAMETER FOR AORTIC OSCILLATION

P = 0.0; %OSCILLATION AMPLITUDE [N]

w = pi*2.33; %BEAT RATE 60=2pi,30=pi,90=3pi,120=4pi

% 40=1.33,70=2.33,80=2.67,100=3.33

%FULL-FILL/FULL-EJECT MODE INITIAL CONDITION AND CONTROL MODE

push = systole; %INITIALLY PUSHING CONDITION

max_bt_vol = 70; %MAXIMUM HEART FILLING VOLUME 90

min_bt_vol = 5; %MINIMUM HEART EMPTYING VOLUME 5

%DESIGN PARAMETER WITHIN THE VAD

if mode == 1 %FIXED RATE MODE

Cht= 28; %[cm^5/N]182

lht= 0.0038; %[*100kg] Rht=0.35,Cht=135,lht=0.0046 will work!!

Rht = 0.47; %[Ns/cm] Rht=0.47,Cht=28,lht=0.0038 will work!!

else %FFFE MODE

Cht= 220; %[cm^5/N]

lht= 0.0025; %[*100kg] Rht=0.17,Cht=220,lht=0.0025 will work!!

```
Rht = 0.17,    %(Ns/cm)
end
```

%HEART GEOMETRIC CHARACTERISTIC

```
A = 19.27,    % used 19.27 [cm^2]
```

```
piston = (max_ht_vol-min_ht_vol)/A; %(cm) HEART PISTON TRAVELLING LENGTH
```

%PRE-DEFINED PARAMETERS USED BY THE EVAD

%CANNULAE

```
knot = 0.000263, % normally 0.00055 [Ns^2/cm^5]0.000263
```

```
lcnn = 0.000132; % normally 0.00025 [Ns^2/cm^5]0.000132
```

```
Cblo = 44,    %44
```

```
Cblt = 44,    %44
```

% VALVE RESISTANCES

```
Rin = 0.0008,    % used 0.003325 [Ns/cm^5]
```

```
Rot = 0.0008,    % used 0.00665 [Ns/cm^5]
```

```
m = 6,    % resistance multiplier (exponent)
```

%CIRCULATORY SYSTEM PARAMETERS

% COMPLIANCES

```
Cin = 3e-05 [cm^5/N]3750
```

```
Cat = 133e-05 [cm^5/N]133
```

% COMPLIANCE OFFSETS

```
acoff = 4.13,    % arterial C offset-4.13
```

```
vcoff = 0.4,    % venous C offset-0.4
```

```
ocanoff = 5.24,    % output cannula offset-5.24
```

```
icanoff = 2.02,    % input cannula offset-2.13
```

```
hroff = -0.178,    % heart C offset-0.178
```

% PERIPHERAL CIRCULATION

```
Rpe = 0.013,    % used 0.013 [Ns/cm^5]0.01813
```

```
lpe = 0.0275,    % used 0.0275 [Ns^2/cm^5]0.00021
```

```
format short ;
```

%SETTING UP CHARACTERISTIC MATRIX

```
a=zeros(9,9);
```

```
a(1,1)=-Rht/lht;
```

```
a(1,2)=-A/Cht;
```

```
a(2,1)=A/lht;
```

```
a(2,2)=-(1/Rot)+(1/Rin))/Cht;
```

```
a(2,3)=1/(Rot*Cblo);
```

```
a(2,9)=1/(Rin*Cblt);
```

```
a(3,2)=1/(Rot*Cht);
```

```

a(3,3)=-1/(Rot*Cblo);
a(3,4)=-1/lcnol;
a(4,3)=1/Cblo
a(4,5)=-1/Cat;
a(5,4)=1/lcnol;
a(5,6)=-1/lpe;
a(6,5)=1/Cat;
a(6,6)=-Rpe/lpe;
a(6,7)=-1/Cvn;
a(7,6)=1/lpe;
a(7,8)=-1/lcnn;
a(8,7)=1/Cvn;
a(8,9)=-1/Cbli;
a(9,2)=1/(Rin*Chi);
a(9,8)=1/lcnn;
a(9,9)=-1/(Rin*Cbli);

```

```

% DETERMINING EIGEN VALUES
damp(a);

```

```

% DETERMINING TIME INTERVAL SCALING
if beat == 60
    trac = 60;
else
    trac = beat;
end

```

```

% SETTING UP TIME INTERVAL
t0 = 0.0; tf = 60*trac;

```

```

% 1L = 1000 cm^3
% Total blood volume = 5L
% Arteries contains 13% volume = 0.65L = 650 cm^3
% Veins contains 64% volume = 3.2L = 3200 cm^3
% Heart chambers contains 9% volume = 0.45L = 450 cm^3

```

```

% SETTING UP INITIAL CONDITIONS

```

```

if mode == 1 % FIXED RATE MODE
    x0=zeros(11,1);
    x0(1)= 0.0; % pump momentum 0.0 0.0
    x0(2)= 10.8; % heart chamber volume (cm^3) 10.8
    x0(3)= 288.9; % output canula volume (cm^3) 288.9
    x0(4)= 0.0; % output canula momentum 0.0 0.0
    x0(5)= 707.3; % arterial volume (cm^3) 707.3
    x0(6)= 2.3; % peripheral momentum 2.3 2.3
    x0(7)= 2511; % venous volume (cm^3) 2511
    x0(8)= 0.0; % input canula momentum 0 0
    x0(9)= 127.4; % input canula volume (cm^3) 127.4
    x0(10)= 0;

```



```

x0(11)=0;
else
    %TTFE MODE:
    x0=zeros(11,1);
    x0(1)=0.0;    % pump momentum      0
    x0(2)=111.2;  % heart chamber volume (cm^3) 111.2
    x0(3)=289.4;  % output canula volume (cm^3) 289.4
    x0(4)=0.0;    % output canula momentum 0
    x0(5)=715.9;  % arterial volume (cm^3) 715.9
    x0(6)=1.9;    % peripheral momentum 1.9
    x0(7)=3169;   % venous volume (cm^3) 3169
    x0(8)=0.0;    % input canula momentum 0
    x0(9)=134.9;  % input canula volume (cm^3) 134.9
    x0(10)=5.55;  % piston position 5.55
    x0(11)=0;
end

%INTEGRATION USING RUNGE KUTTA 23 METHOD
if mode==1
    [t,xx]=ode23('vads3',t0(1),x0);
else
    [t,xx]=ode23('vads3',t0(1),x0,1e-6);
end

%HEADING TO BE DISPLAYED ON GRAPH
if mode==1
    ts='AT NOMINAL';
    p='FIXED RATE';
    W='Kor 10';
else
    p='TFE MODE';
    W='NOMINAL';
end

%CALLS OUTPUT ROUTINE
optn5
%flow

%INITIAL CONDITION FOR NEXT TIME SET SIMULATION
xx(max(length(xx)),:)
%pause,
%yy(.8)

```

VADN5.M

% FUNCTION FOR EVALUATING VAD/CIRCULATORY SYSTEM MODEL
% August 2,1995

function d = vadm5(t,x)

global Rout Rinm Rot Rin Rht lht htoff atoff ocanoff icanoff vnoff A Cht lcnct lcnin Cat lpe Rpe Cvn
Cblo Cbli rm

global max_ht_vol min_ht_vol push mode ht_vol force count piston P w pc told diastole systole

%INPUT FORCE:MODE 1 FOR FIXED RATE, MODE 2 FOR FULL FILLA-FULL EFFECT

if mode == 1

u = vadm(t);

%FIXED RATE

elseif mode == 2

ht_vol = x(2)-(x(10)-piston)*A; %HEART CHAMBER VOLUME

u = ffe(ht_vol,max_ht_vol,min_ht_vol); %FFE MODE

end

% [x(11),u]

%CALCULATION OF PRESSURE DIFFERENCE ACROSS THE VALVES

dp_out = x(2)/Cht - x(3)/Cblo + htoff - ocanoff;

dp_in = x(9)/Cbli - x(2)/Cht + icanoff - htoff;

% VALVE SWITCHING MODE

Routn = Rot * 10^(rm*((dp_out)<0));

Rinn = Rin * 10^(rm*((dp_in)<0));

% CHARACTERISTIC MATRIX

a=zeros(11,11);

a(1,1)=-Rht/lht;

a(1,2)=-A/Cht;

a(2,1)=A/lht;

a(2,2)=-((1/Routn)+(1/Rinn))/Cht;

a(2,3)=1/(Routn*Cblo);

a(2,9)=1/(Rinn*Cbli);

a(3,2)=1/(Routn*Cht);

a(3,3)=-1/(Routn*Cblo);

a(3,4)=-1/lcnct;

a(4,3)=1/Cblo;

a(4,5)=-1/Cat;

a(5,4)=1/lcnct;

a(5,6)=-1/lpe;

a(6,5)=1/Cat;

a(6,6)=-Rpe/lpe;

a(6,7)=-1/Cvn;

a(7,6)=1/lpe;

a(7,8)=-1/lcnin;

a(8,7)=1/Cvn;

a(8,9)=-1/Cbli;

a(9,2)=1/(Rinn*Cht);

a(9,8)=1/lcnin;

a(9,9)=-1/(Rinn*Cbli);

$a(10,1) = 1/lh;$

% FORCE INPUT MATRIX

$b = \text{zeros}(11,1);$

$b(1,1) = 1.0;$

$b(11,1) = 1.0;$

% COMPLIANCE OFFSETS MATRIX

$c = \text{zeros}(11,1);$

$c(1,1) = -A*htoff;$

$c(2,1) = (1/Rom)*ocanoff + (1/Rinn)*icanoff - ((1/Rom) + (1/Rinn))*htoff;$

$c(3,1) = (htoff - ocanoff)/Rom;$

$c(4,1) = ocanoff - atoff + P*\sin(w*t);$

$c(6,1) = atoff - vnoff - P*\sin(w*t);$

$c(8,1) = vnoff - icanoff;$

$c(9,1) = (htoff - icanoff)/Rinn;$

% SPACE STATE EQUATION

$\dot{d} = a*d + b*u + c;$

VADIN.M

**% FUNCTION FOR INPUT FORCE AND FORCE PERIOD FOR FIXED RATE
CONTROL**

% August 2,1995

function u = vadin(t)

global Pf Pfb beat sysfrc

period = 60/beat;

% each period is one heart cycle

systole = sysfrc * period;

% systolic fraction

u = Pf*(rem(t,period)<systole) + Pfb; % force input to the VAD

FFFLM

% FUNCTION FOR FULL-FILL/FULL-EJECT MODE
 % THE FILLING AND EMPTYING IS BASED ON THE MAX AND MIN HEART VOLUME
 % August 2,1995

function u = fffc(ht_vol,maxht,minht)

global Pf Pfb max_ht_vol min_ht_vol push pcs pcd count told systole diastole

```
%SYSTOLIC PHASE
if push == systole
    pcs = pcs + 1;
    if ht_vol > minht
        push = 1;
        u = Pfb + Pf;
    else
        push = 0;
        u = Pfb;
    end
%DIASTOLIC PHASE
elseif push == diastole
    pcd = pcd + 1;
    if ht_vol < maxht
        push = 0;
        u = Pfb;
    else
        push = 1;
        u = Pf - Pfb;
    end
end
end
```

OPTN5.M

%Function for outputting graphs and results of the simulation
% August 2,1995

```

disp('optn5'); %DISPLAY ENTRANCE POINT

yy(:,1) = xx(:,3)/Cblo + ocanoff ; % OUTPUT CANN PRESSURE
yy(:,2) = xx(:,9)/Cbli + icanoff ; % INPUT CANN. PRESSURE
yy(:,3) = xx(:,2)/Cht + htoff; % HEART PRESSURE
yy(:,4) = xx(:,5)/Cat + atoff; % ARTERIAL PRESSURE
yy(:,5) = xx(:,7)/Cvn + vnoff ; % VENOUS PRESSURE

yy(:,8) = xx(:,1)/lit; % PISTON VELOCITY
yy(:,9) = xx(:,6)/lpc; % PERIPHERAL FLOWRATE
yy(:,13) = xx(:,4)/lcnat - yy(:,9); % ARTERIAL FLOWRATE
yy(:,14) = yy(:,9) - xx(:,8)/lcmin; % VENOUS FLOWRATE

%SETTING MAX AND MIN HEART VOLUME LIMIT
if mode == 1
    maxvol = max(xx(:,2)-(xx(:,10)-piston)*A);
    minvol = min(xx(:,2)-(xx(:,10)-piston)*A);
else
    maxvol = max_hr_vol;
    minvol = min_hr_vol;
end

% DETERMINING FORCES AND POWER
for i = 1:length(xx(:,1))

    Rout = Rot* 10^(min*(yy(i,3) - vv(i,1))-0); %OUTPUT VALVE SWITCHING
    Rinn = Rin* 10^(min*(yy(i,2) - yy(i,3))-0); %INPUT VALVE SWITCHING

    vv(i,6) = (yy(i,3) - yy(i,1))/Rout; %OUTPUT CANNULA FLOW RATE
    yy(i,7) = (yy(i,2) - yy(i,3))/Rinn; %INPUT CANNULA FLOW RATE
    yy(i,10) = (yy(i,3) - yy(i,1)); %PRESSURE ACROSS OUTPUT VALVE
    yy(i,11) = (yy(i,2) - yy(i,3)); %PRESSURE ACROSS INPUT VALVE

    %CALCULATING e*F OF INPUT AND OUTPUT VALVE
    otpower(i,1) = ((yy(i,10)).^2)/Rout/100;
    mpower(i,1) = ((yy(i,11)).^2)/Rinn/100;

    %CALCULATING INPUT FORCES
    if mode == 1
        yy(i,12) = vadin(u(i)); % FIXED RATE MODE INPUT FORCE
        voltest = xx(i,2)-(xx(i,10)-piston)*A;
        vv(i,1) = voltest/100;
    end

    voltest = xx(i,2)-(xx(i,10)-piston)*A; %DETERMINING HEART VOLUME
    vv(i,1) = voltest;
    hplot(i,1) = maxvol; %SET MAX AND MIN HEART VOLUME
    hplot(i,2) = minvol;

end

```

```

if mode == 2
    %FFFE MODE POWER CALCULATION
    d_x11 = diff(xx(:,11));
    d_u = diff(u);
    f_input = d_x11./d_u;
    f_input(length(u)) = f_input(length(u)-1);
    yy(:,12) = f_input;
end

%CALCULATING THE HEART PISTON POSITION AND VAD VOLUME
for m = 2:length(xx(:,1))
    hitvol(m,1) = trapz(u(1:m,1),xx(1:m,1))*A/hit;
end

%CALCULATION FOR POWER = e*f
pepower = (Rpe*(yy(:,9))^2)/100; %PERIPHERAL POWER
hipower = (Rht*(yy(:,8))^2)/100; %HEART POWER
power = yy(:,12).*yy(:,8)/100; %INPUT POWER = F*Vmotor;
%powerwo = yy(:,16).*yy(:,8)/100;

l = length(power);
H = fix(0.7*l);

impower = max(power(1:H)); %INSTANTANEOUS POWER
avpower = powercal(power,u); %AVERAGE POWER

%POWER TO THE BLOOD = flow * (Hypressure - Aypressure)
blpower = ((yy(:,5)-yy(:,1)).*yy(:,6))/100;
avbpower = powercal(blpower,u); %AVERAGE OUTPUT POWER

%CALCULATING ENERGY COMSUPTION
% Energy=int(e*f dt)
meng = energy(power,u); %TOTAL INPUT ENERGY
hteng = energy(hipower,u); %HEART CONSUMPTION
oveng = energy(otpower,u); %OUTPUT VALVE COMSUMPTION
peeng = energy(pepower,u); %PERIPHERAL CONSUMPTION
iveng = energy(impower,u); %INPUT VALVE CONSUMPTION
engloss=oveng+peeng+iveng; %TOTAL SYSTEMIC ENERGY LOSS

%mengtwo = energy(powerwo,u);

%DISPLAYING ENERGY
disp(' ');
disp('Energy ');
disp(' ');
disp(['IN = ',num2str(meng),' HT = ',num2str(hteng),' OV = ',num2str(oveng),' PE = ',num2str(peeng),' IV = ',num2str(iveng)]);

```

OPTN5.M

```

disp(' ');
disp(['SYS= ',num2str(engloss),' TOT = ',num2str(engloss+heng)]);
disp(' ');
disp(' ');
disp('Parameter: ');
disp(' ');

%DETERMINING EFFICIENCY OF VAD
%eff=avpower in / av power to blood
eff=engloss/meng*100;

%CONVERTING POWER TO PRINTABLE FORMAT
maxpower=num2str(max(power));
avgpwr=num2str(avpower);
maxpht=num2str(75*max(yy(11:1,3)));
minpht=num2str(75*min(yy(11:1,3)));
maxq=num2str(0.06*max(yy(11:1,9)));
maxpat=num2str(max(vv(11:1,4))*75);
minpat=num2str(min(yy(11:1,4))*75);
avgblp=num2str(avbpower);

%DETERMINING SYSTOLIC FRACTION
if mode == 1
    syst = systc*100; %SYSTOLIC FRACTION FOR FIXED RATE
else if mode == 2
    syst = pct(tpcd+pcst)*100; %SYSTOLIC FRACTION FOR FILE
end

%DISPLAYING POWER AND PRESSURE PARAMETERS
disp(['Maximum Pm(W)= ',maxpower,' Avg Power(W)= ',avgpwr,' Max Pht(mmHg)= ',maxpht]);
disp(' ');
disp(['Min Pht (mmHg)= ',minpht,' Max Qot(l/min)= ',maxq,' Max Pat(mmHg)= ',maxpat]);
disp(' ');
disp(['Min Pat(mmHg)= ',minpat,' AvBloodPower(W)= ',avgblp,' Eng2= ',num2str(meng)]);
disp(' ');
disp(['Efficiency= ',num2str(eff),' %', ' sysfrac= ',num2str(syst),' %']);
disp(' ');
%disp([' max vol= ',num2str(max(vv(:,1))), ' min vol= ',num2str(min(vv(:,1)))]);

%pause
clg

%CALLS GRAPH PRINTING ROUTINE
ht_plot2

%PLOTTING HEART PISTON POSITION AND VOLUME
figure(2)

```



```

clf;
hold;
adjht(:,1) = xx(:,2)+piston*A-hrvoff(:,1);
plot(t,xx(:,10),'r');
plot(t,vv(:,1)/10,'c-');
plot(t,yy(:,12)/10);
plot(t,hplot(:,1)/10,'m');
plot(t,hplot(:,2)/10,'m');
legend('piston disp','ht_vol','force','vol_limit',-1);
grid
print volumeplot.ps

```

```

figure(2)

```

```

clf;

```

```

flow

```

```

% plot (power,t,otpower,t,mpower,t,pcpower)%t,htpower)
% legend('inenergy','outputvalve','input valve','peripheral','heart',-1)
% title('2)
% plot(htpower)
% legend('heart R',-1)
% title('3)
% text(1.0,8,'Avg Power = ');
% text(4.0,8,num2str(avpower));
% text(1.1,1,'Max Power = ');
% text(4.1,num2str(max(power)));
% text(1.0,6,'Avg bpower = ');
% text(4.0,6,num2str(avbpower));
% text(1.0,4,'Efficiency = ');
% text(4.0,4,num2str(ef));
% adjht(:,1) = xx(:,2)+piston*A-hrvoff(:,1); % calculating heart volume

```

HE_PLOT2.M

```
%plotting heart simulation results
% August 2,1995
```

```
figure (1);
```

```
clf
```

```
orient landscape %PRINTING SIDEWAYS
```

```
% PLOTTING OUPUT VALVE PARAMETERS
```

```
subplot(321)
plot(t,yy(:,10)*75,'y-',t,yy(:,6)*0.06,'c-')
axis([0,max(t),-150,50])
title('OUTPUT VALVE'),ylabel('Q , P'),grid
%legend('Pout','Qout',-1)
```

```
% PLOTTING INPUT VALVE PARAMETERS
```

```
subplot(322)
plot(t,yy(:,11)*75,'y ',t,yy(:,7)*0.06,'c-')
title('INPUT VALVE'),ylabel('Q , P'),grid
axis([0,max(t),-120,30])
%legend('Pinv','Qinc',-1)
```

```
% PLOTTING ARTERIAL PARAMETERS
```

```
subplot(323)
plot(t,yy(:,13)*.06,'c-',t,yy(:,4)*75,'y-',t,yy(:,1)*75,'m-')
%hold
%axis_plot
title('Artery'),ylabel('Q , P'),grid
axis([0,max(t),-20,150])
%legend('Qat','Pat','Pocn',-1)
```

```
% PLOTTING VENOUS PARAMETERS
```

```
subplot(324)
plot(t,yy(:,14)*.06,'c-',t,yy(:,5)*75,'y-',t,yy(:,2)*75,'m-')
%hold
%axis_plot
title('Vein'),ylabel('Q , P'),grid
axis([0,max(t),-20,50])
%legend('Qvn','Pvn','Pcn',-1)
```

```
% PLOTTING VAD PARAMETERS
```

```
subplot(325)
plot(t,yy(:,3)*75,'y-',t,yy(:,8),'c-',t,yy(:,12),'m-')
title('Heart pressures and motor velocity'),xlabel('time'),ylabel('P , V'),grid
axis([0,max(t),-30,150])
%legend('Phr','Vpis','Fin',-1)
```

```
% PLOTTING PERIPHERAL FLOW
```

```
subplot(326)
plot(t,yy(:,9)*.06,'c-')
title('Peripheral'),xlabel('time'),ylabel('Q'),grid
axis([0,max(t),-2,10])
text(-3,-6,p)
```

HT_PLOT2.M

```
text(0,-6,W)
```

```
%legend('Flow',-1)
```

```
%gtext('NOMINAL VALUES')
```

```
%gtext(['Q] = L/min , [P] = mmHg , [V] = cm/s')
```

```
%GRAPH LAYOUT
```

```
% (321) (322)
```

```
% (323) (324)
```

```
% (325) (326)
```

```
print htplo1 ps
```

FLOW.M

```
%plotting the circulation volume
% August 2,1995
```

```
clf
orient landscape
```

```
% ADJUSTING VAD VOLUME ACCORDING TO PISTON POSITION
adjht(:,1) = xx(:,2)+piston*A-hivoff(:,1);
```

```
% SUMMING TOTAL VOLUME
gg(:,1) = (xx(:,3)+xx(:,9)+adjht(:,1)+xx(:,5)+xx(:,7))*0.001; %*0.001 to convert from ml to l
tv=num2str(max(gg(:,1)));
```

```
subplot('position',[0.1 0.1 0.6 0.8]),
hold
```

```
% PLOTTING VOLUME GRAPH
plot(t,gg(:,1),'c-'), % total volume
title('FIFE MODE CIRCULATION VOLUME'),ylabel('volume (l)'),xlabel('time (s)'),end
```

```
plot(t,adjht(:,1)/1000,'p-'), % heart volume
```

```
plot(t,xx(:,3)/1000,'w-'), % output cannula volume
```

```
plot(t,xx(:,5)/1000,'y-'), % arterial volume
```

```
plot(t,xx(:,7)/1000,'m-'), % venous volume
```

```
plot(t,xx(:,9)/1000,'r-'), % input cannula volume
```

```
% plot(t,xx(:,5)/1000+xx(:,7)/1000,'r-') %at+vn
```

```
% CALCULATING VOLUME FRACTION FOR EACH PART OF THE SYSTEM
ocf=max(xx(:,3))/max(gg(:,1)*1000)*100;
icf=max(xx(:,9))/max(gg(:,1)*1000)*100;
htf=max(adjht(:,1))/max(gg(:,1)*1000)*100;
atf=max(xx(:,5))/max(gg(:,1)*1000)*100;
vnf=max(xx(:,7))/max(gg(:,1)*1000)*100;
```

```
%text(max(t)+.2,mean((xx(:,7)+xx(:,5))/1000),'Art+Ven')
```

```
%ADDING TEXT EXPLANATION TO THE GRAPH
text(max(t)+.2,mean(gg(:,1)),'Total Vol')
text(max(t)+1.8,mean(gg(:,1)),'.')
text(max(t)+1.2,mean(gg(:,1)),num2str(tv))
```

```
text(max(t)+.2,mean(xx(:,7)/1000),'Venous')
```

FLOW.M

```

text(max(tt)+1.1,mean(xx(:,7)/1000),num2str(vnf))

text(max(tt)+.2,mean(xx(:,5)/1000),'Artery')
text(max(tt)+1.1,mean(xx(:,5)/1000),num2str(af))

text(max(tt)+.2,max(xx(:,3)/1000)+.05,'OutCan')
text(max(tt)+1.1,max(xx(:,3)/1000)+.05,num2str(ocf))

text(max(tt)+.2,mean(xx(:,9)/1000)-.05,'InpCan')
text(max(tt)+1.1,mean(xx(:,9)/1000)-.05,num2str(icf))

text(max(tt)+.2,mean(adph(:,1)/1000)+.05,'Heart')
text(max(tt)+1.1,mean(adph(:,1)/1000)+.05,num2str(hnf))

text(max(tt)+0.2,max(gp(:,1))+0.2,'% By Volume')

print flowplots

```

ENERGY.M

% Energy calculation function
% Total energy = $\int cf \, dt$

function eng = energy(pwr,t)

eng = trapz(t,pwr);

POWERCAL.M

% Power and average power calculation function

function calc = powercal(powerin,t)

calc = trapz(t,powerin)/(t(length(t))-t(1));

EIGEN.M

% PROGRAM FOR STUDYING THE EIGENVALUES OF VARIOUS PARAMETERS
% ALL CRITICAL PRAMETERS WILL BE VARIED FROM -20% TO +20% OF THE NOMINAL.

```
clear
global A Cvn Rpe lpe Cat Cht hioff atoff ocanoff vnoff icanoff lcnof lcnin lht Rot Rin Pf Pfb Rht Cblo
Cbli rm ee
```

%all units are in cm base and in SI units

%DESIGN PARAMETER WITHIN THE VAD

```
Cht= 220; %[cm^5/N]182
lht= 0.0025; %[kg]          %Rht=19.27,Cht=28,lht=0.08 will work!!
Rht = 0.17;
```

```
A= 19.27; % used 19.27 [cm^2]
```

%PREDEFINED PARAMETERS USED BY THE EVAD

```
% Capacitance
lcnof= 0.0000263; % normally 0.00055 [Ns^2/cm^5]0.000263
lcnin= 0.0000132; % normally 0.00025 [Ns^2/cm^5]0.000132
Cblo = 44;
Cbli = 44;
```

% Valve resistances

```
Rin= 0.0035; % used 0.003325 [Ns/cm^5]
Rot = 0.0035; % used 0.00665 [Ns/cm^5]
```

%CIRCULATORY SYSTEM PARAMETERS

% Compliances

```
Cvn= 3750; % [cm^5/N]3750
Cat= 133; % [cm^5/N]133
```

% Peripheral circulation

```
Rpe= 0.013; % used 0.013 [Ns/cm^5]0.01813
lpe= 0.0275; % used 0.0275 [Ns^2/cm^5]0.00021
```

format short ;

%setting up characteristic matrix

```
a=zeros(9,9);
```

```
a(1,1)=-Rht/lht;
a(1,2)=-A/Cht;
a(2,1)=A/lht;
a(2,2)=-((1/Rot)+(1/Rin))/Cht;
a(2,3)=1/(Rot*Cblo);
a(2,9)=1/(Rin*Cbli);
a(3,2)=1/(Rot*Cht);
a(3,3)=1/(Rot*Cblo);
```


EIGEN.M

```

a(3,4) = -1/lcnot;
a(4,3) = 1/Cblo;
a(4,5) = -1/Cat;
a(5,4) = 1/lcnot;
a(5,6) = -1/lpc;
a(6,5) = 1/Cat;
a(6,6) = -Rpe/lpc;
a(6,7) = -1/Cvn;
a(7,6) = 1/lpc;
a(7,8) = -1/lcnin;
a(8,7) = 1/Cvn;
a(8,9) = -1/Cbli;
a(9,2) = 1/(Rin*Cht);
a(9,8) = 1/lcnin;
a(9,9) = -1/(Rin*Cbli);

```

```

damp(a);

```

```

ee = eig(a);

```

```

ee1 = num2str(ee(1));
ee2 = num2str(ee(2));
ee3 = num2str(ee(3));
ee4 = num2str(ee(4));
ee5 = num2str(ee(5));
ee6 = num2str(ee(6));
ee7 = num2str(ee(7));
ee8 = num2str(ee(8));
ee9 = num2str(ee(9));

```

```

clf;
figure;
factor = 0.55;

```

```

%for i = 1:17

```

```

%factor = factor + 0.15;
factor = 1;
Chum = Cht*factor;

```

```

a = zeros(9,9);

```

```

a(1,1) = -Rh/Ult;
a(1,2) = -A/Chum;
a(2,1) = A/Ult;
a(2,2) = -(1/Rot)+(1/Rin)/Chum;
a(2,3) = 1/(Rot*Cblo);
a(2,9) = 1/(Rin*Cbli);
a(3,2) = 1/(Rot*Chum);
a(3,3) = -1/(Rot*Cblo);
a(3,4) = -1/lcnot;
a(4,3) = 1/Cblo;

```

EIGEN.M

```

a(4,5)=-1/Cat;
a(5,4)=1/lcnot;
a(5,6)=-1/lpc;
a(6,5)=1/Cat;
a(6,6)=-Rpc/lpc;
a(6,7)=-1/Cvn;
a(7,6)=1/lpc;
a(7,8)=-1/lcnin;
a(8,7)=1/Cvn;
a(8,9)=-1/Cbli;
a(9,2)=1/(Rin*Chun);
a(9,8)=1/lcnn;
a(9,9)=-1/(Rin*Cbli);

subplot('position',[0.4 0 1 0.6 0 8]);
plot (eig(a),'x');
axis([-50,0,-60,60]);
title('THE EIGEN VALUE@ NOMINAL'),xlabel('REAL'),ylabel('IMAGINARY'),grid

hold;

% end

text(5,-20,ez1);
text(5,-15,ez2);
text(5,-10,ez3);
text(5,-5,ez4);
text(5,0,ez5);
text(5,5,ez6);
text(5,10,ez7);
text(5,15,ez8);
text(5,20,ez9);

print eigensalue.ps

```

APPENDIX 2

Graphs Showing the Changes in Eigenvalues Due to Variations in Parameters

POSSIBLE SELECTIONS

- 0.5223
- 0.1284
- 4.671
- 4.468-34.24i
- 4.468+34.24i
- 10.61-35.68i
- 10.61+35.68i
- 117.4-53.69i
- 117.4+53.69i

FIXED RATE EIGENVALUE COLLECTION

-3.5233
 -0.1284
 -4.671
 -4.468-34.24i
 -4.468+34.24i
 -10.61-35.69i
 -10.61+35.69i
 -17.4-53.69i
 -17.4+53.69i

-0.5233
 -0.1234
 -4.671
 -4.468-34.24
 -4.468+34.24
 -10.61-35.68
 -10.61+35.68
 -117.4-53.69
 -117.4+53.69

-0.5233
 -0.1284
 -4.671
 -4.468-34.24
 -4.468+34.24
 -10.61-35.68
 -10.61+35.68
 -117.4-53.69
 -117.4+53.69

[illegible]

FIXED RATE EIGEN VALUE @ CDR-10%

- 0.5233
- 0.1284
- 4.671
- 4.468-34.24i
- 4.468+34.24i
- 10.61-35.68i
- 10.61+35.68i
- 117.4-53.69i
- 117.4+53.69i

FIVED BATE BISEL 1.1. 19 1990

-0.5233
 -0.1264
 -4.671
 -1.468-34.24i
 -1.468+34.24i
 -10.61-35.68i
 -10.61+35.68i
 -117.4-53.69i
 -117.4+53.69i

FIXED RATE EIGEN VALUES (CFO) - 1955

-0.5233
 -0.1284
 -4.671
 -4.468-34.24i
 -4.468+34.24i
 -10.61-35.68i
 -10.61+35.68i
 -117.4-53.69i
 -117.4+53.69i

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-0.5233
-0.234
-0.671
-1.469-34.241
-1.469-34.241
-10.61-35.691
-10.61+35.691
-17.4-53.691
-117.4+53.691

3 OF/DE 3

PM-1 3 1/2" x 4" PHOTOGRAPHIC MICROCOPY TARGET
NBS 1010a ANSI/ISO #2 EQUIVALENT

PRECISIONSM RESOLUTION TARGETS

FIXED RATE/GE/VAL. EG. 10.10.10.

-0.5233
-0.1284
-4.671
-4.468-34.24i
-4.468+34.24i
-10.61-35.68i
-10.61+35.68i
-117.4-53.69i
-117.4+53.69i

FFFF EIGEN VALUES NORMAL

-0.5057
 -0.2194
 -5.11
 -7.374-29.05i
 -7.374+29.05i
 -13.16-37.42i
 -13.16+37.42i
 -44.88-8.447i
 -44.88+8.447i

FREE EIGEN VALUES OF M-571

- 0.5057
- 0.2194
- 5.11
- 7.374-29.05i
- 7.374+29.05i
- 13.16-37.42i
- 13.16+37.42i
- 44.88-8.447i
- 44.88+8.447i

12 13 14

FFFF EIGENVALUES 00-1000

- 0.5057
- 0.2194
- 5.11
- 7.374-29.05i
- 7.374+29.05i
- 13.16-37.42i
- 13.16+37.42i
- 44.88-8.447i
- 44.88+8.447i

FREE SPACE MODE

-0.505
-0.219
-5.11
-7.374-29.0
-7.374+29.0
-13.16-37.4
-13.16+37.4
-44.88-8.44
-44.88+8.44

FREE EIGEN VALUES 2016-1974

-0.5057
 -0.2194
 -5.11
 -7.374-29.05i
 -7.374+29.05i
 -13.16-37.42i
 -13.16+37.42i
 -44.88-8.447i
 -44.88+8.447i

2000

FREE EIGENVALUES OF A

-0.5057
-0.2194
-5.11
-7.374-29.05i
-7.374+29.05i
-13.16-37.42i
-13.16+37.42i
-44.88-8.447i
-44.88+8.447i

FREE EIGENVALUES

- 0.5057
- 0.2194
- 5.11
- 7.374-29.05i
- 7.374+29.05i
- 13.16-37.42i
- 13.16+37.42i
- 44.88-8.447i
- 44.88+8.447i

SEEK SEP 10 1963 10:00:00

-0.5057
 -0.2194
 -5.11
 -7.374-29.05i
 -7.374+29.05i
 -13.16-37.42i
 -13.16+37.42i
 -44.88-8.447i
 -44.88+8.447i

FREE EIGENVALUE PROBLEM

-0.5057
 -0.2194
 -5.11
 -7.374-29.05i
 -7.374+29.05i
 -13.16-37.42i
 -13.16+37.42i
 -44.88-8.447i
 -44.88+8.447i

END

1 9 | 0 6 | 9 7

FIN