1

T

OF / DE

**PM-1 3½"x4" PHOTOGRAPHIC MICROCOPY TARGET
NBS 1010a ANSI/ISO #2 EQUIVALENT**

Design of a Rotary Control Valve for an Electrohydraulic Ventricular Assist Device

by

Shane D. Pinder, B. Eng.

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

Department of Mechanical and Aerospace Engineering
Ottawa-Carleton Institute
for Mechanical and Aerospace Engineering

Carleton University

Ottawa, Ontario

08 Jan 1996

copyright

1996, Shane D. Pinder

National Library
of Canada

Acquisitions and
Bibliographic Services Branch

395 Wellington Street
Ottawa, Ontario
K1A 0N4

Bibliothèque nationale
du Canada

Direction des acquisitions et
des services bibliographiques

395, rue Wellington
Ottawa (Ontario)
K1A 0N4

Your file Votre référence

Our file Notre référence

The author has granted an irrevocable non-exclusive licence allowing the National Library of Canada to reproduce, loan, distribute or sell copies of his/her thesis by any means and in any form or format, making this thesis available to interested persons.

L'auteur a accordé une licence irrévocable et non exclusive permettant à la Bibliothèque nationale du Canada de reproduire, prêter, distribuer ou vendre des copies de sa thèse de quelque manière et sous quelque forme que ce soit pour mettre des exemplaires de cette thèse à la disposition des personnes intéressées.

The author retains ownership of the copyright in his/her thesis. Neither the thesis nor substantial extracts from it may be printed or otherwise reproduced without his/her permission.

L'auteur conserve la propriété du droit d'auteur qui protège sa thèse. Ni la thèse ni des extraits substantiels de celle-ci ne doivent être imprimés ou autrement reproduits sans son autorisation.

ISBN 0-612-13849-6

Name SHANE DONALD PINDER

Dissertation Abstracts International is arranged by broad, general subject categories. Please select the one subject which most nearly describes the content of your dissertation. Enter the corresponding four-digit code in the spaces provided.

BIOMEDICAL ENGINEERING

0541 U·M·I

SUBJECT TERM

SUBJECT CODE

Subject Categories

THE HUMANITIES AND SOCIAL SCIENCES

COMMUNICATIONS AND THE ARTS

Architecture 0729
Art History 0377
Cinema 0900
Dance 0378
Fine Arts 0357
Information Science 0723
Journalism 0391
Library Science 0399
Mass Communications 0708
Music 0413
Speech Communication 0499
Theater 0465

EDUCATION

General 0515
Administration 0514
Adult and Continuing 0516
Agricultural 0517
Art 0273
Bilingual and Multicultural 0282
Business 0688
Community College 0275
Curriculum and Instruction 0727
Early Childhood 0518
Elementary 0524
Finance 0277
Guidance and Counseling 0519
Health 0680
Higher 0745
History of 0520
Home Economics 0278
Industrial 0521
Language and Literature 0279
Mathematics 0280
Music 0522
Philosophy of 0998
Physical 0523

Psychology 0525
Reading 0535
Religious 0527
Sciences 0714
Secondary 0533
Social Sciences 0534
Sociology of 0340
Special 0529
Teacher Training 0530
Technology 0710
Tests and Measurements 0288
Vocational 0747

LANGUAGE, LITERATURE AND LINGUISTICS

Language 0679
General 0289
Ancient 0290
Linguistics 0291
Modern 0401
Literature 0294
General 0295
Classical 0297
Comparative 0298
Medieval 0316
Modern 0591
African 0305
American 0352
Asian 0355
Canadian (English) 0593
Canadian (French) 0311
English 0312
Germanic 0315
Latin American 0313
Middle Eastern 0314
Romance 0314
Slavic and East European

PHILOSOPHY, RELIGION AND THEOLOGY

Philosophy 0422
Religion 0318
General 0321
Biblical Studies 0319
Clergy 0320
History of 0322
Philosophy of 0469
Theology

SOCIAL SCIENCES

American Studies 0323
Anthropology 0324
Archaeology 0326
Cultural 0327
Physical 0310
Business Administration 0272
General 0770
Accounting 0454
Banking 0338
Management 0385
Marketing 0501
Canadian Studies 0503
Economics 0505
General 0508
Agricultural 0509
Commerce-Business 0510
Finance 0511
History 0358
Labor 0366
Theory 0351
Folklore 0358
Geography 0366
Gerontology 0351
History 0578
General

Ancient 0579
Medieval 0581
Modern 0582
Black 0328
African 0331
Asia, Australia and Oceania 0332
Canadian 0334
European 0335
Latin American 0336
Middle Eastern 0337
United States 0585
History of Science 0398
Law 0615
Political Science 0616
General 0617
International Law and Relations 0814
Public Administration 0452
Recreation 0626
Social Work 0627
Sociology 0938
General 0631
Criminology and Penology 0628
Demography 0629
Ethnic and Racial Studies 0630
Individual and Family 0700
Studies 0344
Industrial and Labor 0709
Relations 0999
Public and Social Welfare 0453
Social Structure and Development 0344
Theory and Methods 0709
Transportation 0999
Urban and Regional Planning 0453
Women's Studies

THE SCIENCES AND ENGINEERING

BIOLOGICAL SCIENCES

Agriculture 0473
General 0285
Agronomy 0475
Animal Culture and Nutrition 0476
Animal Pathology 0359
Food Science and Technology 0478
Forestry and Wildlife 0479
Plant Culture 0480
Plant Pathology 0817
Plant Physiology 0777
Range Management 0746
Wood Technology 0306
Biology 0287
General 0308
Anatomy 0309
Biotistics 0379
Botany 0329
Cell 0353
Ecology 0369
Entomology 0793
Genetics 0410
Limnology 0307
Microbiology 0317
Molecular 0416
Neuroscience 0433
Oceanography 0821
Physiology 0778
Radiation 0472
Veterinary Science 0786
Zoology 0760
Biophysics 0425
General 0996
Medical

Geodesy 0370
Geology 0372
Geophysics 0373
Hydrology 0388
Mineralogy 0411
Paleobotany 0345
Paleoecology 0426
Paleontology 0418
Paleozoology 0985
Polynology 0427
Physical Geography 0368
Physical Oceanography 0415

HEALTH AND ENVIRONMENTAL SCIENCES

Environmental Sciences 0768
Health Sciences 0566
General 0300
Audiology 0992
Chemotherapy 0567
Dentistry 0350
Education 0769
Hospital Management 0758
Human Development 0982
Immunology 0564
Medicine and Surgery 0347
Mental Health 0569
Nursing 0570
Nutrition 0380
Obstetrics and Gynecology 0354
Occupational Health and Therapy 0381
Ophthalmology 0571
Pathology 0419
Pharmacology 0572
Pharmacy 0382
Physical Therapy 0573
Public Health 0574
Radiology 0575
Recreation

Speech Pathology 0460
Toxicology 0383
Home Economics 0386

PHYSICAL SCIENCES

Pure Sciences 0485
Chemistry 0749
General 0486
Agricultural 0487
Analytical 0488
Biochemistry 0738
Inorganic 0490
Nuclear 0491
Organic 0494
Pharmaceutical 0495
Physical 0754
Polymer 0405
Radiation 0605
Mathematics 0986
Physics 0606
General 0608
Acoustics 0748
Astronomy and Astrophysics 0607
Atmospheric Science 0798
Atomic 0759
Electronics and Electricity 0609
Elementary Particles and High Energy 0610
Fluid and Plasma 0752
Molecular 0756
Nuclear 0611
Optics 0463
Radiation 0346
Solid State 0984
Statistics 0984
Applied Sciences 0346
Applied Mechanics 0984
Computer Science

Engineering 0537
General 0538
Aerospace 0539
Agricultural 0540
Automotive 0541
Biomedical 0542
Chemical 0543
Civil 0544
Electronics and Electrical 0348
Heat and Thermodynamics 0545
Hydraulic 0546
Industrial 0547
Marine 0794
Materials Science 0548
Mechanical 0743
Metallurgy 0551
Mining 0552
Nuclear 0549
Packaging 0765
Petroleum 0554
Sanitary and Municipal 0790
System Science 0428
Geotechnology 0796
Operations Research 0795
Plastics Technology 0994
Textile Technology

PSYCHOLOGY

General 0621
Behavioral 0384
Clinical 0622
Developmental 0620
Experimental 0623
Industrial 0624
Personality 0625
Physiological 0989
Psychobiology 0349
Psychometrics 0432
Social 0451

Thesis Acceptance Form

M. Eng. Candidate

**The undersigned recommend to the Faculty of Graduate Studies
and Research acceptance of the thesis**

"Design of a Rotary Control Valve for an Electrohydraulic Ventricular Assist Device"

submitted by Shane D Pinder, B. Eng.

in partial fulfilment of the requirements for

the degree of Master of Engineering

Thesis Co-supervisor

Thesis Co-supervisor

Chair, Department of Mechanical and Aerospace Engineering

Carleton University

02 Feb 1996

Abstract

Mechanical circulatory assist devices are grouped in two major classes: pulsatile, in which the blood flow rate is unsteady and similar to the natural circulatory system, and: non-pulsatile, in which the blood is pumped at a steady flow rate. Ventricular assist devices or VADs assist the natural heart in pumping the blood and are therefore often pulsatile. One possibility in designing such a device is to create the pulsatile flow by alternately emptying and filling a chamber in a manner similar to the function of the natural heart. This alternation between filling and emptying of the VAD chamber can be created by using a control system to change the pump direction or by careful design of the pumping mechanism. This thesis examines the design of a rotary valve intended for use in hydraulically actuated VADs that will create an oscillating flow to empty and fill the VAD chamber from a unidirectional hydraulic pump. Note that in these devices the hydraulic fluid and the blood are separated by a flexible membrane. The valve contains a rotating element whose angular velocity is determined by the speed of the axial flow pump and appropriately sized annular gaps. This rotating element is then used to switch the flow between two sets of inlet and exit ports on the valve, one set causes the hydraulic fluid to fill the VAD chamber and the second causes the chamber to empty. Important aspects of the design include its energy efficiency, its simplicity and reliability, and a reduction in the number of parts exposed to alternating mechanical loads. A prototype of the design is described and preliminary tests described in the thesis show promising performance. The prototype was constructed to test both the feasibility of the concept and the feasibility of implementing the valve under the size and other constraints used in the design of VADs at the Cardiovascular Devices Division of the University of Ottawa Heart Institute.

Acknowledgements

The results of this thesis project could not have been achieved without the guidance of Professor Donald Russell. His constructive criticism and suggestions were greatly appreciated as well as his patience dealing with unusual requests over the last two years.

The day to day assistance of members of the Cardiovascular Devices Division of the University of Ottawa Heart Institute was certainly valued and contributed on more than one occasion to the success of this project. The efforts of Albert Hum, Mervin Valadares, and Tofy Mussivand must certainly be recognized.

The work done by members of the Science Technology Centre at Carleton University must also be acknowledged. The patience, guidance, and support of Larry Boissoneault and Peter Vorel has been greatly appreciated.

Finally, I would like to sincerely thank those at the Department of Mechanical and Aerospace Engineering at Carleton University from whom I have received continuous support. Special thanks goes out to Kris Cooper and Christie Egbert.

Table of Contents

Thesis Acceptance Form	ii
Abstract	iii
Acknowledgements	iv
Table of Contents	v
List of Tables	vii
List of Figures	viii
List of Plates	ix
List of Symbols	x
 Chapter 1	
Introduction	1
1.1 Summary of Remaining Chapters	2
 Chapter 2	
Literature Review	3
2.1 University of Ottawa Heart Institute	4
2.2 Nimbus - Cleveland Clinic	5
2.3 Research in Japan	5
2.4 ABIOMED	6
2.5 Labor für Biofluidmechanik	6
 Chapter 3	
Background	8
3.1 Glossary of Human Heart Physiology	8
3.2 Operation of a Healthy Natural Heart	12
3.3 Stroke Volume, Heart Rate, and Cardiac Output	12
3.4 Parameter Variation with Level of Activity	13
 Chapter 4	
Design	15

4.1	Aim of Valve Development	15
4.2	Reliability	15
4.3	Power Consumption	16
4.4	Size, Weight, and Geometry	17
4.5	Other Design Criteria	17
4.6	Prototype Design	18
4.6.1	Dual Axial Flow Pump	19
4.6.2	Multiple Valve System	20
4.6.3	Single Multiport Valve	20
4.6.4	Rotary Valve	22
4.7	Rotary Valve Propulsion	23
4.7.1	Hydrodynamic Transmission	24
4.7.2	Axial Flow Turbomachine	29
4.8	Detailed Design	30
4.8.1	Device Description	31
4.8.2	Axial Flow Pump Integration	35
4.8.3	Valve Rotor	37
4.8.4	Rotor Bearing Surfaces	37
4.8.5	Flexibility to Incorporate Turbine	39
4.8.6	Clearances	39
4.8.7	Material	41

Chapter 5

Testing	43
5.1 Materials and Methods	43
5.2 Experimental Procedure	44
5.3 Results	46
5.4 Discussion and Conclusions	47

Chapter 6

Conclusions and Recommendations	49
---------------------------------	----

References	51
------------	----

Appendices	A-1
Working Drawings	A-2
Data Tables	A-12
TK Solver Routine	A-14

List of Tables

Table 4.1	Increasing Probability of Failure with Number of Components	16
Table 5.1	Experimental Materials	45
Table 5.2	Experimental Data	46
Table 5.3	Experimental Calculations	47
Table A.1	Contour "A"	A-13

List of Figures

Figure 2.1	EVAD 4.0 Schematic	4
Figure 3.1	Human Heart Anatomy	11
Figure 3.2	Heart Rate Data	13
Figure 3.3	Stroke Volume Data	14
Figure 4.1	Dual Axial Flow Pump System	19
Figure 4.2	Multiple Valve System Schematic	20
Figure 4.3	Multiport Valve Schematic	21
Figure 4.4	Simplified Rotary Valve	22
Figure 4.5	Hydrodynamic Transmission	24
Figure 4.6	Flat Plate Analogy	25
Figure 4.7	Exploded View Of Rotary Valve	32
Figure 4.8	Assembly Drawing of Rotary Valve and Energy Converter	33
Figure 4.9	Front Sectional View of Rotary Valve	34
Figure 4.10	Side Sectional View of Rotary Valve	35
Figure 4.11	Simplified Half Section of Valve Rotor	36
Figure 4.12	Simplified Impeller and Cylinder	41
Figure 5.1	Experimental Apparatus	43

List of Plates

Plate 1	Parts List for Rotary Valve	A-3
Plate 2	Working Drawing of Valve Rotor	A-4
Plate 3	Working Drawing of Lower Valve Cover	A-5
Plate 4	Working Drawing of Outer Valve Housing with Internal Bearing Surface	A-6
Plate 5	Detail to Working Drawing of Outer Valve Housing	A-7
Plate 6	Working Drawing of Upper Bearing Surface	A-8
Plate 7	Alternate Working Drawing of Upper Bearing Surface with Contoured Interior	A-9
Plate 8	Working Drawing of Stator Modification	A-10
Plate 9	Working Drawing of Upper Valve Cover	A-11

List of Symbols

a,b	Layer Thicknesses	m
A,B	Integration Constants	
A	Surface Area	m²
CO	Cardiac Output	ml/min
F	Force	N
h	Reference Height	m
h,H	Surface Height	m
HR	Heart Rate	beats/min
P	Pressure	N/m²
P_s	Proportion of Impeller Geometry	
r,R	Radius	m
Re	Reynolds Number	
SV	Stroke Volume	ml/beat
t	Fin Thickness	m
T	Torque	N·m
u	Velocity in x-dir	m/s
v	Velocity in y-dir	m/s
w	Velocity in z-dir	m/s

Greek Symbols

γ	Specific Weight	N/m³
μ	Absolute Viscosity	N·s/m²
ν	Kinematic Viscosity	m²/s
ρ	Density	kg/m³
τ	Shear Stress	N/m²
φ	Cone Angle	rad
ω	Angular Speed	rad/s

Chapter 1

Introduction

Artificial heart research has been ongoing for the last forty years.¹ Devices have been used as interim support for patients awaiting transplant for several years. However, most of these devices have left patients tethered to large control systems, often the size of an average dishwasher, to avoid implantation of more parts than absolutely necessary. As a result, a patient's mobility is greatly restricted. As well, the number of donor hearts available at any one time is very small in comparison to the number of patients requiring heart replacement.² To address these problems, research has begun focusing on the development of a totally implantable, long term artificial heart for assisting or replacing the natural heart.

In designing long term artificial organs, one of the most formidable tasks is ensuring high reliability. Some failures of an artificial heart can dramatically affect the life of the patient. Redundant mechanical systems are impractical due to size limitations for implantable devices. Large numbers of parts, each with inherent reliability traits, serve to decrease the reliability of the overall system. Therefore, designs must be simple and elegant, to be sufficiently reliable. Limiting the number of parts is the most sensible way to approach this as it serves to also limit size and weight.

Intrinsic to these ideas is the concept of safety. A device that has successfully withstood demanding tests affords the recipient a greater confidence in the new technology. This directly influences the quality of one's life. These principles have been articulated by members of the Labor für Biofluidmechanik³ who have prototyped a device in observance of these criteria. Their efforts will be discussed later.

1.1 Summary of Remaining Chapters

To fully discuss the steps taken to arrive at a method of creating a pulsatile flow in a reliable manner, an investigation of alternative methods of generating the required flow was undertaken. The following chapters will deal with the background theory required to examine the various alternatives, a discussion of different possibilities with their strengths and weaknesses, and a detailed description of the chosen design with a discussion of testing results and an evaluation of the design.

Chapter 2

Literature Review

Artificial heart research is a very diverse field. One of the most fundamental divisions in artificial heart research revolves around the type of flow created: pulsatile or non-pulsatile. Non-pulsatile devices are currently used to wean donor heart recipients from heart-lung bypass. The simplest of these uses an axial pump attached to the end of a catheter¹. It is widely accepted that the most logical choice for a long term heart replacement would create a pulsatile flow because, at this point, it is impossible to predict the long term effects of changing one's blood flow so dramatically. The devices that create pulsatile flow are generally more complicated, and possess an inherent reliability problem. Any mechanical device that undergoes cyclic loading, reversing, or other repetitive motion generally has a limited lifetime. It is the challenge of pulsatile device designers to minimize this effect.

There is another fundamental problem that must be addressed: biocompatibility of propulsion devices. The cellular makeup of blood demands careful manipulation. Consequently, devices typically used to propel non-biological fluids are not practical. One way to resolve this problem is to employ a hydraulic fluid in conjunction with the propulsion device, and then transfer the pulsation using a chamber with a flexible membrane to separate blood from hydraulic fluid.

Finally, the method of providing power to a heart assist device must be examined. The following literature survey deals with methods of creating pulsatile flow in electrohydraulic heart assist devices.

2.1 University of Ottawa Heart Institute

The University of Ottawa Heart Institute has been involved in the development of a ventricular assist device for several years.⁵ In this device, hydraulic fluid is propelled using an axial pump and the motion is transferred to the circulatory system.

Figure 2.1 EVAD 3.0 Schematic

The mechanical parts of the Electrohydraulic Ventricular Assist Device (EVAD) 3.0 are depicted in Figure 2.1. The ventricle consists of a chamber with a flexible membrane separating blood from hydraulic fluid. The blood side is fitted with two one-way valves such that one is an intake valve while the other is an eject valve. Pumping of the blood

can be achieved by alternately filling and emptying the hydraulic fluid side. This is done with the use of an axial flow pump. One technique that has been used in a prototype VAD created the alternating flow by spinning the pump in one direction to fill the ventricle and reversing the pump to empty the ventricle. A volume displacement chamber (VDC) fitted with a flexible membrane is used to hold the hydraulic fluid emptied from the ventricle.

2.2 Nimbus - Cleveland Clinic

The Cleveland Clinic Foundation began working on an electrohydraulic total artificial heart(TAH) in 1988.⁶ Prior to this, extensive research and development had been carried out on a pneumatic total artificial heart.

The mechanical portion of the device consists of an electrohydraulic energy converter, pusher plates, and blood sacs. Hydraulic pressure is used to alternately compress two pusher plates which, in turn, provide pressure to two blood sacs (ventricles). The energy converter is unidirectional. Switching valves are used to direct the fluid alternately between the pusher plate actuators.⁷

2.3 Research in Japan

There are two designs of interest that are being developed at the National Cardiovascular Center in Osaka, Japan. Both are electrohydraulic devices, however one is a VAD while the other is a TAH. Both of these devices have a separately placed actuator. That is, the ventricle is located adjacent to the natural heart position while the hydraulic actuator is

located lower in the abdominal cavity. The action is transmitted through an oil filled driving tube.

The first design is a VAD that uses a linear DC motor to actuate a metal bellows which compresses and decompresses the hydraulic fluid.⁸ The blood pump is a chamber with a flexible membrane separating blood from hydraulic fluid.

The second design is a TAH designed in much the same way as the VAD with a separately placed actuator. The two ventricles are made to compress alternately using a reversing rotary actuator.⁹ This actuator consists of a large impeller rotating through an annular channel and propelling fluid tangentially. As it removes fluid from one ventricle, it transmits fluid to the other.

2.4 ABIOMED

The ABIOMED TAH consists of a centrifugal pump, a rotary switching valve, and two hydraulic fluid chambers which couple with blood pumps.¹⁰ The centrifugal pump is unidirectional, resulting in favourable reliability. The flow direction is determined by the position of an independently actuated reversing rotary switching valve.

2.5 Labor für Biofluidmechanik

Research in Germany has been approached from the perspective of designing a highly reliable device which can be demonstrated to possess a high level of safety. The concept of limiting the number of moving parts to increase the reliability of the system is utilized throughout the design procedure.

A recent development involves the use of a telescoping centrifugal pump impeller to actuate flow direction.¹¹ The impeller is able to move axially. At one extreme, the flow is directed from the blood pump to an expansion chamber. At the other extreme, the flow is reversed without reversing the direction of the motor. The axial actuator consists of a permanent magnet and a coil. This actuator need only draw power at the moment of switching. This is a result of an innovative bistable electric motor design. The motor contains two sets of permanent magnets arranged in a manner such that the coils are only stable at either extreme axial position. The result is a device with only one moving part which does not reverse.

Chapter 3

Background

To effectively understand the design criteria for heart-assist devices, the mechanics of natural heart operation must be understood.

3.1 Glossary of Human Heart Physiology

The heart is a complicated organ whose role is to receive oxygen deficient blood from all body tissues, send this blood to the lungs for re-oxygenation, and then pump the oxygen rich blood back to all body tissues. The following terms, in conjunction with Figure 3.1 form a working vocabulary of heart anatomy and physiology.¹²

Aorta	The aorta is the body's main artery, receiving oxygenated blood from the left ventricle and supplying it to all body tissues except the heart itself.
Aortic Valve	The aortic valve is a one-way valve between the left ventricle and the aorta that prevents back flow during diastole.
Atrioventricular (AV) Node	The AV node is one of two nodes in the heart's electrical conduction system that provides electrical impulses.

Coronary Artery	The coronary artery supplies oxygenated blood to the heart muscle itself.
Diastole	Diastole is the portion of the heart cycle in which refilling of the heart takes place.
Left Atrium	The left atrium receives re-oxygenated blood from the pulmonary vein.
Left Ventricle	The left ventricle pumps oxygenated blood to the entire body. It is the largest heart chamber and, consequently, the <i>most susceptible to heart problems.</i>
Mitral Valve	The mitral valve is a one-way valve located between the left atrium and the left ventricle that prevents backflow during the second stage of systole.
Pulmonary Artery	The pulmonary artery carries oxygen deficient blood from the right ventricle to the lungs.
Pulmonary Valve	The pulmonary valve is a one-way valve located between the right ventricle and the pulmonary artery that prevents backflow during diastole.

Pulmonary Vein	The pulmonary vein carries re-oxygenated blood from the lungs to the left atrium.
Right Atrium	The right atrium receives oxygen deficient blood from all body tissues.
Right Ventricle	The right ventricle pumps oxygen deficient blood to the lungs for re-oxygenation.
Sinoatrial (SA) Node	The SA node is the pacemaker of the heart's electrical conduction system. It is responsible for generating the electrical impulse that signals the heart muscle to contract.
Systole	Systole is a the portion of the heart cycle in which blood is pumped away from the heart.
Tricuspid Valve	The tricuspid valve is a one-way valve located between the right atrium and the right ventricle that prevents backflow during the second stage of systole.
Vena Cava	The vena cava carries oxygen deficient blood from all body tissues to the right atrium.

Figure 3.1 Human Heart Anatomy

3.2 Operation of a Healthy Natural Heart

In a healthy heart, blood fills the atria passively during diastole. The period of this filling is nominally twice the time required to empty the ventricles during systole. Physical activity serves to increase the rate at which blood flows in blood vessels, and consequently the heart refills more quickly. The timing of events in the cardiac cycle is controlled by several factors. The heart will beat with no outside influence so long as the blood remains minimally oxygenated. Signals from the brain can serve to increase or decrease the nominal heart rate based on one's emotional or physical state. Finally, the rate of return of blood into the atria tends to influence the rate at which it is pumped away.

3.3 Stroke Volume, Heart Rate, and Cardiac Output

Figure 3.2 depicts a collection of data representing typical variation of heart rate with the amount of blood flow supplied by the heart for various age groups¹³ and a study carried out on parameter variation in athletic hearts.¹⁴ Figure 3.3 depicts variations in the stroke volume with cardiac output for these same groups. Stroke volume is simply the amount of blood pumped in one single heart beat. Cardiac output is then simply the product of heart rate and stroke volume(3.1). Superimposed on both of these graphs are the corresponding curves for a theoretical device with a constant stroke volume of 100ml. It should be noted that the youngest heart and the athletic heart more closely approximate the constant stroke volume condition.

Figure 3.2 Heart Rate Data

This data is compiled from detailed studies regarding characteristics of the human heart^{13,14}

$$CO = SV \cdot HR \quad (3.1)$$

3.4 Parameter Variation with Level of Activity

Cardiac output is closely related to the level of activity. In many veins of the human body, pairs of one-way valves are situated in the blood vessel where it passes through muscle.

When the muscle contracts, it helps pump the blood back to the heart. As physical activity increases, this effect is emphasized, resulting in a higher flow rate.

The manner in which the heart deals with this increased flow rate is two-fold. From the figures, it can be seen that heart rate increases linearly. However, the y-intercept of the heart rate curve is not zero (ie: even at complete circulatory rest, the heart would pump). The stroke volume also tends to increase with the flow rate of blood, though not linearly. Rather, the stroke volume is horizontally asymptotic. The data also suggest that, with age, the maximum stroke volume increases.

Figure 3.3 Stroke Volume Data

This data is compiled from detail 1 studies regarding characteristics of the human heart^{13,14}

Chapter 4

Design

It was decided to design and prototype a single component that could be integrated into an existing heart assist device with the aim of improving the reliability and power consumption of the device without greatly impairing its geometry and weight. It was determined that a weakness in some heart assist devices was the reversing nature of the energy converter because it caused reliability problems in the motor bearings.

4.1 Aim of Valve Development

The aim of this project is to eliminate the reversing nature of the axial flow pump used in some heart assist devices through the incorporation of a valve to redirect fluid flow.

4.2 Reliability

The reliability of a reversing motor in heart assist devices is an area of concern. Such devices also require a complicated electronic control system to function. Each electronic component is highly reliable alone, but as the number of components rises, the reliability of the overall system decreases. It was determined that the physical size of such control systems could be reduced with a mechanical control valve, and that the overall reliability could be increased.

As a minimum requirement, heart assist devices should demonstrate 80% reliability for 2 years at a 60% confidence level.¹⁵ The minimum goal for individual components of such

devices is a reliability greater than 90% with 80% confidence level for two years duration. Although it is difficult to quantitatively assess reliability at the prototype level, qualitative assessments can be made with a reasonable level of confidence. Moving parts should be continuously separated from any stationary parts by a fluid layer to reduce wear and friction. Every effort must be made to limit the number of parts in the prototype design as this serves to increase reliability. For example, consider multiple combinations of a component that has a 0.20% chance of failure over a given period of time. Table 4.1 shows that reliability decreases as the number of components is increased. Typically, control systems that are dependent on electronics versus mechanical components require many sub-components. This can lead to unacceptable levels of reliability.

Table 4.1 Increasing Probability of Failure with Number of Components

Number of Components	Probability of Success	Probability of Failure
n	$(1-0.0020)^n$	$1-(1-0.0020)^n$
1	99.80%	0.20%
5	99.00%	1.00%
10	98.02%	1.98%
100	81.86%	18.14%

4.3 Power Consumption

In reversing heart assist devices, some power is spent on the acceleration of hydraulic fluid. The power consumed by the overall device is typically on the order of several Watts. As the goal of this design is to eliminate the reversing nature of the axial flow pump, every effort should be made to minimize the amount of hydraulic fluid that changes direction. Where these changes are necessary, the cross sectional area should be

generous, at least 1 cm^2 , so as to limit the velocity and thus the momentum of the fluid.

This value corresponds to the minimum cross sectional area in the axial pump. The power consumed to actuate the valve itself should be negligible with respect to the the overall power requirement. A value of 0.2 W is defined as the upper limit for valve power consumption.

4.4 Size, Weight, and Geometry

Detailed studies¹⁶ have been carried out to determine the anatomical fit required for heart assist devices. These studies have concluded that a system with overall dimensions of $17.5 \text{ cm} \times 10.5 \text{ cm} \times 3.8 \text{ cm}$ can adequately fit most of the population. Any changes in design must conform to these criteria. The overall volume of the valve shall be no greater than 100ml with the smallest dimension being no greater than 3.8 cm. As the weight of the overall design is to be kept as low as possible, the use of lightweight plastic material should be employed wherever possible.

4.5 Other Design Criteria

It is necessary to define a design point to determine specific physical qualities in the detailed design phase. For the purposes of valve prototype design, an average motor speed of 9000 RPM is assumed to achieve a beat rate of 60 bpm. The design point for stroke volume will be 70 ml. These values represent typical characteristics of the human heart in a resting state.

4.6 Prototype Design

Initially, a passive refill valve was considered. This valve would mount in series with the axial flow pump and would be entirely fluid actuated by rapidly decreasing the unidirectional flow through the pump until the pressure provided by the pump was below diastolic. As discussed earlier, the natural heart passively refills during diastole.

However, the natural heart is composed primarily of muscle that occupies the most volume in its relaxed state. As a consequence, the natural heart provides a small level of suction, as much as 5 mm Hg¹⁷, simply because it is made up of a highly elastic material. It was determined that the inertia of the hydraulic fluid dictated that the pump would have to be virtually stopped to lower the pressure sufficiently and take advantage of passive refill.

The design of a valve to incorporate active refill was approached from a systems point-of-view. Initially, the reversing axial flow pump was fully explored to determine its strengths and weaknesses. An electromagnetic compressor was considered and rejected due to power consumption and size restrictions. A reciprocating piston design was rejected for similar reasons. A system using two variable speed axial flow pumps was dismissed because of the number of parts required. An idea surrounding the use of a telescoping centrifugal pump impeller was dismissed as being beyond the scope of a graduate thesis. Finally, a schematic of a system of valves was conceived that would eliminate the need for a reversing axial flow pump. This schematic was then reorganized, modified, and ultimately abstracted into three dimensions to arrive at the design.

4.6.1 Dual Axial Flow Pump

The first configuration considered was a system using two axial flow pumps (Figure 4.1) that would vary in output throughout the cycle. During systole, the pump propelling fluid from the VDC to the LV would wind up while the other would slow. During diastole, the situation would reverse. The largest disadvantages to this arrangement are size, number of components, and efficiency due to the possibility of circulating flow.

Figure 4.1 Dual Axial Flow Pump System

4.6.2 Multiple Valve System

It was decided to begin with a system of valves around the axial pump to direct the flow. Figure 4.2 shows a schematic representation of this system. Using this method, the axial pump would operate unidirectionally, with a small variation in angular velocity. In practice, incorporating this system as shown would not represent a significant improvement. The reliability of switching valves does not approach the levels required for cardiovascular devices. However, schematically, this is a good first step.

Figure 4.2 Multiple Valve System Schematic

4.6.3 Single Multiport Valve

With a simple rearrangement, the system can be made into one multiport valve as shown in Figure 4.3. Schematically, this arrangement does not look particularly superior, but this shows the possibility of building a single device. The four valves in the figure could

function as one single valve with four ports and two throw positions. This valve needs to be reversed continually. Consequently, it still possesses an inherent reliability problem. The reversing nature of this valve can be eliminated through abstracting this linear arrangement into a rotary valve.

Figure 4.3 Multiport Valve Schematic

4.6.4 Rotary Valve

Abstracting into three dimensions, it is possible to realize a multiport valve in the form of a rotary valve. Figure 4.4 shows a simplified depiction of a three-dimensional rotary valve.

Figure 4.4 Simplified Rotary Valve

In the figure, hydraulic fluid flows from the VDC and is directed by the valve rotor into a chamber through a wedge-shaped window labelled *Lower*. The fluid is then drawn from the lower chamber to the upper chamber *through the centre of the valve rotor along the axis of rotation*. The fluid then travels through another window, labelled *Upper*, and is directed by the valve rotor toward the LV. As the valve rotor continues to rotate, the fluid is still drawn in the same direction through the centre of the rotor but will switch direction to flow from the LV to the VDC. The axial pump draws the fluid through the centre of the valve rotor, continuously in the same direction.

This arrangement eliminates the reliability problems inherent in a conventional switching valve, because the rotor *never stops*. At this point, the difficulties revolve around selecting a method to propel the valve rotor.

4.7 Rotary Valve Propulsion

The challenge of propelling the rotor in this valve depends on geometry and the method used to transfer torque. As a design approximation it can be assumed that fluid flow rate is roughly proportional to the angular speed of the axial flow pump, an important design criteria is realized. To make stroke volume roughly constant, the speed of the valve rotor must be proportional to the speed of the motor. In other words, the entire device could be propelled with the same energy converter, which would never need to reverse or drastically change speed.

The first situation considered recognized that the anatomical fit criteria dictate that the device should be as flat as possible. Conventional means of speed reduction, such as the use of spur gears, were considered in detail. However, it was quickly realized that a speed reduction of 300:1 was required. To do this with this sort of gear system would require several cascaded reduction gears. Space limitations and reliability concerns eliminated this as a potential method.

Wormgear and rack and pinion style reduction systems are typically used for the level of speed reduction required. However, there is an intrinsic requirement for there to be

continuous relative motion between the two components which are in contact. It was decided that this level of activity would afford the device a limited lifetime.

An abstracted concept was considered that would require no contact between the two gears. It involved using the fluid to transmit torque with the use of propellers of various pitch. Although this idea was eliminated as having no real merit, the concept of using the fluid to transmit torque led to two very promising methods.

It was decided to orient the two rotating bodies, the pump impeller and the valve rotor, along the same axis. In this manner the fluid can easily act as a torque transfer mechanism.

4.7.1 Hydrodynamic Transmission

The concept of a hydrodynamic transmission is by no means new. It is used extensively in

Figure 4.5 Hydrodynamic Transmission

the automotive industry in automatic transmission systems where it is used to provide a transfer of torque without a direct mechanical connection. It is proposed that this basic concept be extended to provide a means of providing an angular velocity reduction mechanism. Figure 4.5 depicts the fundamental layout of the proposed hydrodynamic transmission. The inner disk, labelled *Drive*, is driven by the main shaft in the axial flow pump. This disk provides torque to *Layer A* which in turn provides torque to the valve *Rotor*. However, there is an additional torque between the stationary *Housing* and the *Rotor* that is transferred through *Layer B*. As a result, the *Rotor* adopts an angular velocity between zero and the pump speed.

Figure 4.6 Flat Plate Analogy

For the purposes of analysis, the circular geometry of the hydrodynamic transmission can be unfolded into a flat plate analogy as shown in Figure 4.6. This approximation is valid provided the radius of the fluid layer is very large with respect to the thickness.¹⁸

The Navier-Stokes Equations in three dimensions are:

$$-\frac{1}{\rho} \frac{\partial}{\partial x}(P + \gamma h) + \frac{\mu}{\rho} \nabla^2 u = \frac{du}{dt} \quad (4.1)$$

$$-\frac{1}{\rho} \frac{\partial}{\partial y}(P + \gamma h) + \frac{\mu}{\rho} \nabla^2 v = \frac{dv}{dt} \quad (4.2)$$

$$-\frac{1}{\rho} \frac{\partial}{\partial z}(P + \gamma h) + \frac{\mu}{\rho} \nabla^2 w = \frac{dw}{dt} \quad (4.3)$$

where: ρ is the density of the fluid;

P is the fluid pressure at a given point;

γ is the specific weight of the fluid;

h is the height, of a given point, above a reference elevation;

μ is the absolute viscosity of the fluid;

u is the velocity of the fluid, at a given point, in the x-direction;

v is the velocity of the fluid, at a given point, in the y-direction, and;

w is the velocity of the fluid, at a given point, in the z-direction.

To analyze the fluid mechanics of this arrangement, the following assumptions were made:

1. the viscosity of the fluid is constant throughout;
2. there is no pressure gradient in fluid layer;

$$\frac{\partial P}{\partial x} = \frac{\partial P}{\partial y} = \frac{\partial P}{\partial z} = 0 \quad (4.4)$$

3. changes due to reference height are negligible;

$$\frac{\partial \gamma h}{\partial x} = \frac{\partial \gamma h}{\partial y} = \frac{\partial \gamma h}{\partial z} = 0 \quad (4.5)$$

4. axial velocity of the fluid is zero, and;
5. there is no net external torque on the valve rotor.

The symmetry of the rotary valve makes the final assumption reasonable, however effects such as the Bernoulli effect at areas of high fluid speeds, could invalidate this assumption. The contribution of such effects will be discussed later.

The fourth assumption and the geometry of the arrangement yield:

$$v = 0 ; w = 0 ; \nabla^2 u = \frac{\partial^2 u}{\partial y^2} \quad (4.6)$$

For steady state flow the Navier-Stokes equations reduce to:

$$\mu \frac{\partial^2 u}{\partial y^2} = 0 \quad (4.7)$$

Integrating once with respect to y yields a constant value for shear stress in the fluid:

$$\tau_{xy} = \mu \frac{\partial u}{\partial y} = A \quad (4.8)$$

Integrating again yields an equation with constants of integration, A and B :

$$u = \frac{Ay + B}{\mu} \quad (4.9)$$

Applying boundary conditions, the constants of integration are found:

$$y = 0, u = 0 \rightarrow B = 0 \quad (4.10)$$

$$y = a, u = u_1 \rightarrow A = \frac{u_1 \mu}{a} \quad (4.11)$$

From (4.8) that this result is known to be equal to the shear stress acting on the *bottom* of the plate:

$$\tau_{1b} = \frac{u_1 \mu}{a} \quad (4.12)$$

To calculate the shear stress acting on the *top* of the plate, a similar approach can be employed, resulting in:

$$\tau_u = \frac{(u_2 - u_1)\mu}{b} \quad (4.13)$$

Consequently, the total force acting on the middle plate would be equal to zero if:

$$\frac{u_1}{a} = \frac{(u_2 - u_1)}{b} \quad (4.14)$$

Going back to **Figure 4.5**, if the angular velocity of the *Drive* is known, and the desired velocity of the *Rotor* is known, then the ratio of the required gaps for *Layer A* and *Layer B* can be found to achieve the desired speed ratio using fluid mechanics alone. The sizes of these gaps can then be set knowing that they must be small enough to maintain laminar flow. This will be discussed later.

4.7.2 Axial Flow Turbomachine

Another method of torque transfer was considered which would make use of the tangential component of velocity imparted by the impeller. In theory, it is possible to include fins on the interior of the rotor which would interact with the fluid so as to provide the required torque. However, it was found that in order to achieve the required speed ratio, the angle of attack of the valve rotor blades would have to be below 1.0 degrees. The fluid angle at the exit of the axial flow pump was found to be approximately 50 degrees. The level of accuracy required for this method would be higher than

economically feasible, requiring angular tolerances below 0.1 degrees.¹⁹ Changes in back pressure serve to change the tangential component of fluid velocity which, in turn, would change the speed of the valve rotor. As much as this provided the possibility of controlling the speed of the valve rotor through blood pressure, it was decided to evaluate the potential of the hydrodynamic transmission and leave this alternative for possible future research.

4.8 Detailed Design

Although it would be preferable to build a valve that could actually be used in practice, it was decided that the prototype would be built as a proof of concept. Consequently, wherever it was found to be justifiable, compromises could be made. Existing components such as an energy converter and stator from previous prototypes were used as a starting point for the design. Each other component was designed to mate with these parts. The designers of the motor have determined that a non-reversing motor would be 25% smaller than the motor used here. Design of a motor and stator specifically for this application would reduce the overall size of the valve.

An overall dimension of five centimeters (two inches) was chosen for the valve housing. It is expected that this could be reduced by 25% after redesign of the motor. This would result in a valve that is within the criteria set out for the anatomical fit.

From the calculations in Appendix 3, it is known that the critical tolerances occur on parts which border the fluid layers. These fluid layers have a thickness on the order of a couple

of thousandths of an inch. Consequently, tolerances on these pieces is of particular importance. As some mating parts are required to be concentric, the tolerances on the mating surfaces are also particularly important. These tolerances are specified in the working drawings at plus or minus 0.0125 mm (0.0005 inches).

As the clearance between the rotor and the bearing surface was so small, it was decided to build the potential for adjustment into the device. This was done using an 80° conical surface on both the bearing surfaces and the mating surface on the rotor. In this way, an axial displacement of the bearing surface of five thousandths of an inch results in a change in the fluid layer of less than one thousandth of an inch.

The method of maintaining concentricity involved using dowel pins tightly fitted into holes on each mating part. The pieces were to be machined using these dowel pins to locate the part on the lathe. In this way, all parts requiring concentric machining have a common reference point.

4.8.1 Device Description

Detailed working drawings of the rotary valve are found in Appendix 1. An exploded view of the overall design and an assembly drawing are shown in Figure 4.7 and Figure 4.8 respectively.

Figure 4.7 Exploded View of Rotary Valve

Figure 4.8 Assembly Drawing of Rotary Valve and Energy Converter

Figure 4.9 Front Sectional View of Rotary Valve

Figure 4.9 and Figure 4.10 depict, respectively, a simplified front sectional view and a side sectional view of the rotary valve. For simplicity, the valve rotor is not shown. During the portion of the cycle shown, hydraulic fluid enters the valve from the VDC. The rotor directs the flow into the lower chamber. From here, the axial flow pump propels the fluid from the lower chamber to the upper chamber. The rotor then redirects the fluid to the LV.

The height of these chambers was determined in conjunction with the size of the windows to ensure the minimum cross sectional area was no less than 1.0 cm^2 . This is equal to the minimum cross sectional area in the axial flow pump and therefore does not place any additional restrictions on the pressure of the hydraulic fluid. The upper chamber also contains a portion of the axial flow pump. The stator was machined to mate with the upper bearing surface and chamber. As stated earlier, the minimum cross sectional area in the axial pump is 1.0 cm^2 . Both tube fittings and chamber windows are designed

observing this minimum cross sectional area. As a result, no single point is more restrictive than the axial pump, but multiple restrictions serve to increase the pressure loss in the valve and thus the power consumption. This factor was weighed with the knowledge that the size of the valve would have to increase to allow larger orifices. This was not acceptable. It should be noted that all ports also fall within 45° of rotor sweep angle. This ensures that the portion of the cycle which does not provide pumping is minimized.

Figure 4.10 Side Sectional View of Rotary Valve

4.8.2 Axial Flow Pump Integration

The axial flow pump is positioned inside the valve itself. It was decided that the impeller of the pump should be situated in the center of the valve. Two specific design problems needed to be addressed to accommodate the pump. The existing stator that was used for

the rotary valve prototype was machined with a contoured outside diameter. The stator fits inside the valve upper bearing surface. Designs for both a contoured bearing surface and a conical bearing surface were prepared. It was decided that the loss in efficiency using the conical bearing surface was not significant enough to warrant the added expense in building the contoured bearing surface.

It was necessary to design a method of seating the motor in the upper cap of the valve while maintaining a hydraulic fluid seal. An o-ring seat was machined into the upper cap such that the motor could easily be slid into the valve while maintaining the seal. This provides added convenience in the assembly of the device.

Figure 4.11 Simplified Half Section of Valve Rotor
Single digit numbers indicate point identifications used in the equations to follow while two digit numbers indicate surfaces, or values pertaining to surfaces such as fluid layer thicknesses. ϕ indicates the cone angle of the rotor bearing surface.

4.8.3 Valve Rotor

Figure 4.11 is a simplified cutaway view of the valve rotor. Single digit numbers indicate point identifications used as subscripts in the equations to follow. Two digit numbers indicate surfaces, or values pertaining to surfaces such as fluid layer thicknesses.

4.8.4 Rotor Bearing Surfaces

To efficiently incorporate the existing stator, and to maximize the smooth motion of the rotor, the rotor bearing surfaces were made conical. To determine the torque between these bearing surfaces and the rotor surface, begin with equation (4.12) in differential form.

$$dF = \frac{\mu u dA}{a} \quad (4.15)$$

Multiplying by r , the radius of a given point on the surface:

$$dT = \frac{r \mu u dA}{a} \quad (4.16)$$

The tangential velocity of a given point on the surface can be expressed in terms of the rotor angular speed, ω_r , and the radius:

$$u = r \omega_r \quad (4.17)$$

The differential area can be expressed in terms of the radius at a given point and the cone angle of the rotor:

$$dA = \frac{2\pi r}{\cos \phi} dr \quad (4.18)$$

Combining (4.16), (4.17), and (4.18) and applying bounds:

$$T = \int_{r_2}^{r_3} \frac{2\pi\mu\omega r^3}{a_{23}\cos \phi} dr \quad (4.19)$$

Integrating with respect to r :

$$T = \frac{\pi\mu\omega r}{2a_{23}\cos \phi} (r_3^4 - r_2^4) \quad (4.20)$$

This result holds so long as the Reynolds number is less than 1400, corresponding to a typical limiting value to maintain laminar flow.²⁰

It was mentioned earlier that it is expected that the symmetrical quality of this device should result in a zero external torque condition. However, when the rotor passes through the chamber windows and ports, pressure variations are induced. If all of the velocity head of the fluid were lost across a chamber window, a entire rotor fin would be subjected to an unbalanced pressure difference of 2.6 mmHg. This would result in a

maximum net torque of 0.0013 N-m. This corresponds to a torque 10 times stronger than that generated by the hydrodynamic transmission. As a result, the rotor may stall when transiting the windows or ports. As well, under normal conditions each half of the valve rotor experiences a different pressure. As a result, a net force will be exerted through the centre of the rotor against the bearing surfaces. This would result in a situation similar to that in a journal bearing and would affect the torque provided by the hydrodynamic transmission. A possible solution to this would be to double the number of fins and ports in a second symmetrical plane. This would result in a zero external force situation. This would also increase the size of the device as twice as many chamber windows would be required. Consequently, it would only be applied in the event that the results of testing demanded its use.

4.8.5 Flexibility to Incorporate Turbine

The centre of the rotor was designed with a small enough diameter relative to the smallest portion of the cone such that a shoulder exists. Not only does this allow modification of its inside diameter without affecting other properties, it would also allow the incorporation of an alternative method of providing torque to the rotor. The position of the impeller could be adjusted axially to allow for insertion of a bladed annulus inside of the rotor. In this way, the turbomachine option could be exercised.

4.8.6 Clearances

Clearances between the valve rotor and stationary members are particularly important. The amount of clearance between the rotor bearing surface and the rotor compared with

that between the rotor and the impeller serves to control the speed of the valve rotor. To find the torque delivered to the valve rotor from the pump impeller, continue from equation (4.16). Here, the tangential velocity of the pump impeller relative to the rotor is described in terms of the angular speed of the pump impeller, ω_i , the angular speed of the valve rotor, ω_r , and the radius of the surface, r :

$$u = r(\omega_i - \omega_r) \quad (4.21)$$

The differential area is expressed in terms of the radius of the surface and the differential height of the surface:

$$dA = 2\pi r_1 dh \quad (4.22)$$

As before, equations (4.16), (4.21), and (4.22) are combined and the result integrated:

$$T = P_s \frac{2\pi\mu h_{11} r_i}{a_{11}} (\omega_i - \omega_r) \quad (4.23)$$

Initially, the impeller is treated as a cylinder as in Figure 4.12. The constant, P_s , is a proportionality constant due to geometry that is equal to the proportion of the cylindrical area actually occupied by the impeller blade tips. Because of edge effects and the effect of swirling fluid, it was decided to evaluate P_s experimentally. Clearances were calculated by equating the torque at the rotor bearing surface with that transferred from the impeller.

The simultaneous equations were solved with the iterative solver *TK Solver*¹ Release 2.0.²

The routine appears in Appendix 3.

Figure 4.12 Simplified Impeller and Cylinder

It is also possible that a laminar layer will not be created between the impeller tips and the rotor. The ability to widen the inside diameter of the rotor means that the impeller could easily be fitted with a ring around its circumference to essentially make $P_g = 1.0$.

4.8.7 Material

Originally, the material proposed for this project was not influenced by the weight limitations. Because it was a proof of concept, it was felt that the sacrifice of increased flexibility of polymers would serve no purpose in exploring the feasibility of the hydrodynamic transmission. However, with the added weight constraint, it was decided to

¹*TK Solver* is a Trademark of Universal Technical Systems, Inc.

²*TK Solver Release 2.0* is a product of Universal Technical systems, Inc.

choose a polycarbonate material known commercially as Lexan™. The volume of the valve components was calculated to be 54 cm³. The predicted mass of the valve was calculated to be 65 g. This was considered acceptable as it falls well within the design criteria. However, Lexan™ has a modulus of elasticity of 62 MPa. Consequently, it was anticipated that the thickness of various members may be an area of concern due to flexibility. It was decided that the device would be built as designed and tested at a reduced speed for the purposes of the proof of concept. In theory, the stroke volume of a system built of perfectly rigid components is independent of the operating point. In practice, the flexibility of the device would negate this conclusion.

Chapter 5

Testing

Figure 5.1 Experimental Apparatus

5.1 Materials and Methods

Figure 5.1 depicts the experimental apparatus used to evaluate the design of the rotary valve. This arrangement offers no pressure gradient except that created by fluid inertia. This was chosen as an initial test condition. In the event that the device would operate

acceptably, a suitable load could be incorporated. Table 5.1 lists the materials used in the testing apparatus. The electric motor is equipped with three Hall effect sensors. These sensors can locate the motor shaft at each of three instants as it rotates. The multimeter used this information operating in its frequency measurement mode to determine the angular speed of the motor. The device was connected to two pieces of Tygon tubing. Each of the ends were placed in separate 250 ml beakers. A large beaker was used to ensure that the electric motor was continuously immersed in hydraulic fluid. Because the motor is not sealed, failure to do this results in air entering the device. The working fluid used was decamethylpolysiloxane with a specific gravity of 0.854 and viscosity of 1.0 centistoke.

During operation, all aspects of the testing were recorded using a video camera. The recording was viewed on a video cassette recorder to determine the beat rates.

5.2 Experimental Procedure

Several copies of the rotor were built for the purpose of testing. It was then possible to modify each rotor based on the results of testing. It was discovered early that the rotor required Teflon™ inserts to keep it suspended between the two bearing surfaces. These were machined directly into the rotors. Without these inserts, the rotor would cling to one bearing surface keeping the rotor fixed.

Initial problems were encountered with eccentricity of mating components. An additional piece was fitted on the upper cap to make fine adjustments of the motor position possible. It was essentially a collar with three set screws evenly spaced around the circumference.

At this point, it was determined that the valve rotor would spin easily, but the impeller failed to provide sufficient torque. Several thin brass rings were built to place on the circumference of the impeller with a light press fit. This final modification served to make the assumptions made in the previous analysis much more plausible and resulted in acceptable operation of the device.

Table 5.1 Experimental Materials

Materials	
Description	Serial Number
Philips PM 2525 Multimeter	D4000780
DH1715 Model Dual Regulated Power Supply	891040
EVAD 4.0 Controller Interface Card	not applicable
EVAD 4.0 Energy Converter	not applicable
Rotary Valve Assembly	not applicable
experimental stand apparatus with clamps	not applicable
1/2" ID Tygon tubing	not applicable
2x250ml beakers, 1x1000ml beaker	not applicable
2"x18" 1/2" glass tray	not applicable

5.3 Results

It was found that changing the speed of the motor drastically affected the operation of the valve. At low speeds, the rotor would often spin quickly though not approaching the speed of the motor. At high speeds, the rotor would stop entirely, often stalling when the fins were located in the middle of the windows. At intermediate speeds, it was occasionally possible to locate an operating point which functioned with limited stability. The data in Table 5.2 are taken from a video of the most acceptable operation of the device. The fluid layer between the brass ring and the rotor was set at a nominal value of five thousandths of an inch. The motor was operating at 1820 RPM. During this operation, the rotor acted erratically, but with a predictable cycle. It would rotate through 180° at high speed, then through the rest of the cycle at a smooth speed. The stroke volume of this observation was 60-65 ml.

Table 5.2 Experimental Data

Data	
Hall Sensor Frequency	0.091 kHz
Power Supply Configuration	Voltage Displayed, Current Limited
Power Supply Voltage	9.13 V
Stroke Volume	60 - 65 ml

Beat Rate Data: Left Beaker		Beat Rate Data: Right Beaker	
Period	103 s	Period	154 s
Number of Cycles	13	Number of Cycles	19

Table 5.3 Experimental Calculations

Calculated Values			
Average Beat Period	8.0 s	Average Beat Rate	7.5 beats/min
Energy Converter Angular Speed		1820 RPM	
Theoretical Beat Rate at Motor Speed = 15000 RPM (if linear)			62 beats/min

5.4 Discussion and Conclusions

The results refute one of the initial design assumptions. The symmetry of the valve configuration was expected to result in a zero external torque. However, the rotor fins passed through the windows and the tube fittings, where fluid velocity is highest, in a manner consistent with the Bernoulli effect. This resulted in stalling when the device was operated at motor speeds above 3500 RPM. It is proposed that this problem be rectified with the use of a larger cross sectional area for the windows and the tube fittings. Added mass in the construction of the rotor would increase its inertia and tend to maintain its angular speed. To limit the angle that these holes occupy, they would have to be machined as rectangular slits as opposed to circular holes.

The flexibility of the rotor bearing surfaces may have been responsible for some problems. It removed the possibility to quantitatively assess the operation of the hydrodynamic transmission as the thickness of the conical fluid layer could not be determined. However, the tests indicate that there is no need to design these components as cones. Added thickness in these components is demanded to limit flexibility because of the pressure gradient existing within the valve.

The realization that the hydrodynamic transmission would only provide propulsion when the impeller was fitted with a shroud is somewhat enlightening. There is little difficulty in machining the brass ring for this application. Because this serves to increase the surface area and thus the torque supplied, the clearances can be made much larger. This serves to eliminate, or at least diminish the problems associated with eccentricity.

Chapter 6

Conclusions and Recommendations

The design described herein proposes that a rotary control valve be used to redirect the flow of a unidirectional axial flow pump. The pulsating flow offered by this arrangement is used to propel the blood pump in a heart assist device. The method of propelling the valve rotor involves the use of a hydrodynamic transmission between the pump impeller and the valve rotor. This method has not been used before in this application.

The device that has been designed contains two moving parts, an axial flow pump impeller and a valve rotor, which rotate continuously without reversing. With continued long term development, the reliability of such a device should be able to meet the requirements typically used in the design of heart assist devices.

The prototype device is very sensitive to the mechanical flexibility of its components. It is also difficult to achieve repeatable results, likely due to eccentricity of components.

However, correcting these flaws should not pose an insurmountable problem. Several design recommendations are put forward for future work.

The velocity of the fluid at critical points must be reduced. However, the angle occupied by valve ports must be maintained or reduced. To overcome both of these limitations, it is proposed that valve ports be machined as rectangular slits. The mass of the rotor should

be increased to provide a steady angular speed through inertia. These modifications are expected to refine the rotary valve concept.

A major limitation in the design of this valve was the use of the existing pump. The pump was designed to be a reversing axial flow pump. Design of a pump specifically meant for use with the rotary valve should improve performance and reduce the size of the valve by more than 25%.

It is recommended that future designs have only one member act as the rotor bearing surface. This eliminates problems associated with mating these concentric parts. The impeller should be machined as a single piece with a solid circumference in order to maximize the hydrodynamic driving surface. It is recommended that all parts associated with the hydrodynamic transmission be manufactured with a constant diameter using materials or processes where high tolerances can be achieved.

It is expected that the proposed design changes would make the valve sufficiently reliable to act as a control centre of a heart assist device. The use of a single non-reversing actuator is a clear advantage of the design as it demands a sufficiently low amount of power. Based on this assessment, the final recommendation is to investigate long-term development of this device for incorporation into existing heart assist devices.

References

1. Akutsu, T. and Kolff, W.J.: Permanent substitutes for valves and hearts. *ASAIO Transactions*, 4: 230, 1958.
2. *UNOS Annual Report of the US Scientific Registry of Transplant Recipients and the Organ Procurement and Transplantation Network Transplant Data, 1988-1991*. US Department of Health and Human Services, 1993.
3. Affeld, K., Bailleu, A., Buß, A., Dihuweit, J., Friedrichsen, U., Gadischke, J., Hanitsch, R., Hetzer, R., Huber, A., Kähler, J., Koehler, M., Käufer, H., Mrasek, R., Ramöller, A., Schichl, K., Schiessler, U., Siekmann, H., Strowich, S., Svejda, D., Teut, J., Thelen, W., Theobald, J.N., Weber, K.-U., Wittjen, C., Walter, J., Ziemann, A., and Zimmer, R.: A New Electrohydraulic Energy Converter for a Left Ventricular Assist Device. *Artificial Organs*, 18(7): 479-483, 1994.
4. Casimir-Ahn, H., Lönn, U., Peterzén, B., et al: Clinical Use of the Hemopump Cardiac Assist System for Circulatory Support. *Annals of Thoracic Surgery*, 59(2 Suppl): S39-45, 1995.
5. Diegel, P.D., Mussivand, T., Holfert, J.W., Juretic, J.T., Miller, J.A., Maclean, G.K., Szurmak, Z., Santerre, J.P., Rajagopalan, K., Dew, P.A., Hansen, A.C., Sinnott, M.M., Olsen, D.B., and Keon, W.: Electrohydraulic Ventricular Assist Device Development. *ASAIO Journal*, 38: M306-M310, 1992.
6. Harasaki, H., Fukamachi, K., Massiello, A., Chen, J.-F., Himley, S.C., Fukumura, F., Muramoto, K., Niu, S., Wika, K., Davies, C.R., McCarthy, P.M., Kiraly, R.J., Thomas, D.C., Rintoul, T.C., Carriker, J.C., Maher, T.R., and Butler, K.C.: Progress in Cleveland Clinic-Nimbus Total Artificial Heart Development. *ASAIO Journal*, 40: M494-M498, 1994.
7. Irié, H., Massiello, A., Kiraly, R., Chen, J.F., Crouch, R., Fukamachi, K., Jacobs, G., Harasaki, H., Golding, L., Thomas, D., Carriker, W., and Butler, K.: Initial *In Vivo* Tests of an Electrohydraulic Actuated Total Artificial Heart. *ASAIO Journal*, 38: M497-M500, 1992.
8. Yagura, A., Taenaka, Y., Takano, H., Nakatani, T., Tatsumi, E., Sekii, H., Sasaki, E., Akagi, H., Goto, M., Akutsu, T., Oh-umi, T., Tsukahara, K., and Tsuchimoto, K.: Chronic *In Vivo* Evaluation of a Portable Electrohydraulic Ventricular Assist System with a Linear Actuator. *ASAIO Journal*, 36: M284-M287, 1990.
9. Masuzawa, T., Taenaka, Y., Kinoshita, M., Nakatani, T., Akagi, H., Takano, H., Fukui, Y., Sasagawa, H., and Takahashi, K.: A Motor Integrated Regenerative Pump as the Actuator of an Electrohydraulic Totally Implantable Artificial Heart. *ASAIO Journal*, 38: M232-M236, 1992.

10. Yu, L., Finnegan, M., Vaughan, S., Ochs, B., Parnis, S., Frazier, and O.H., Kung, R.T.V.: A Compact and Noise Free Electrohydraulic Total Artificial Heart. *ASAIO Journal*, 39: M386-M391, 1993.
11. Ibid. 3, p. 480.
12. Fung, Y.C., *Biodynamics: Circulation* (New York: Springer-Verlag New York Inc., 1984), p. 27.
13. Rodeheffer, R.J., Gerstenblith, G., Becker, L.C., et al: Exercise Cardiac Output is Maintained with Advancing Age in Healthy Human Subjects: Cardiac Dilatation and Increased Stroke Volume Compensate for a Diminished Heart Rate. *Circulation*, 69: 203-213, 1984.
14. Wallace, A.G.: The Heart in Athletes. In *The Heart, 6th Edition* (J.W. Hurst, ed., New York: McGraw Hill Book Company, 1986), pp. 1398-1403.
15. Unpublished communication. Mussivand et al. University of Ottawa Heart Institute.
16. Mussivand, T., Masters, R.G., Hendry, P.J., Rajagopalan, K., Walley, V.M., Nahon, D., Hicks, A., and Keon, W.J.: Critical Anatomic Dimensions for Intrathoracic Circulatory Assist Devices. *Artificial Organs*, 16(3): pp. 281-285, 1992.
17. Ibid. 12, p. 197.
18. Munson, B.R., Young, D.F., and Okiishi, T.H., *Fundamentals of Fluid Mechanics* (New York: John Wiley & Sons, Inc., 1990), p. 377.
19. Lieblein, S.: Experimental Flow in Two-dimensional Cascades. In *Aerodynamic Design of Axial-flow Compressors* (Johnsen, I.A. and Bullock, R.O., eds., Washington, D.C.: National Aeronautics and Space Administration, 1965), pp. 183-226.
20. Ibid. 18, p. 375.

Appendices

Appendix 1
Working Drawings

Parts List for Rotary Valve

PART NAME	REQUIRED	MATERIAL
Valve Rotor	1	Lexan™
Lower Valve Cover	1	Lexan™
Outer Valve Housing	1	Lexan™
Dowel Pin	2	Shop Discretion
Upper Bearing Surface	1	Lexan™
Stator (modification)	1	Tecoflex™
Upper Valve Cover	1	Lexan™
O-Ring AS 568-018 (3/4"ID)	1	Nitrile
1/8-40 UNC-2A Hex Socket Cap Scr	4	Steel
Brushless Motor (EC4)	1	Provided by OHI
Impeller III	1	Provided by OHI
Nylock™ Nut	1	Provided by OHI

NOTES:

Concentricity: Outer Valve Housing, Upper Bearing Surface, Upper Valve Cover
 Maintained by Dowel Pins
 Maximum Eccentricity: 0.001"

Bearing Surface angle 80 degrees: Valve Rotor, Outer Valve Housing, Upper Bearing Surface

Angular Tolerance: within 1 degree overall

Variation between complementary members: within 0.1 degree

Valve Rotor should turn freely on axes

Precision Running Fit

Lubricant: hydraulic fluid (1.5 cstk, SG=0.854)

Stator to be machined to a Close Sliding Fit with Brushless Motor

Plate 1 Parts List for Rotary Valve

Plate 2 Working Drawing of Valve Rotor

Plate 3 Working Drawing of Lower Valve Cover

Plate 4 Working Drawing of Outer Valve Housing with Internal Bearing Surface

Plate 5 Detail to Working Drawing of Outer Valve Housing

Plate 6 Working Drawing of Upper Bearing Surface

Plate 8 Working Drawing of Stator Modification

Plate 9 Working Drawing of Upper Valve Cover

Appendix 2

Data Tables

Table A.1 Contour "A"

X	R	X	R
0.000	0.509	0.300	0.453
0.039	0.508	0.311	0.450
0.054	0.507	0.322	0.447
0.069	0.506	0.334	0.443
0.081	0.503	0.345	0.440
0.096	0.501	0.357	0.437
0.110	0.499	0.369	0.434
0.122	0.	0.380	0.431
0.135	0.494	0.392	0.428
0.148	0.492	0.404	0.426
0.160	0.489	0.417	0.423
0.173	0.486	0.429	0.420
0.185	0.483	0.442	0.417
0.197	0.480	0.455	0.415
0.209	0.477	0.468	0.413
0.220	0.474	0.481	0.410
0.232	0.471	0.495	0.408
0.243	0.468	0.509	0.407
0.255	0.465	0.523	0.404
0.266	0.462	0.538	0.403
0.277	0.459	0.552	0.402
0.289	0.456	0.600	0.402

Appendix 3
TK Solver Routine

VARIABLE SHEET				For Academic Use Only
Input	Name	Output	Unit	Comment
20	a_{s1}		thou	Fins & Outer Wall
20	a_{s1}		thou	Fins & Ceiling/Floor
	a_{s2}	0.86824089	thou	Bearing Surface
	a_{11}	3.5	thou	Drive
900	R_3		thou	Outer Radius of Fin
600	R_4		thou	Inner Radius of Fin
	R_{s1}	540.74311	thou	Major Radius of Bearing Surface Cone
	R_5	538.0982	thou	Major Radius of Rotor Cone
442	R_2		thou	Minor Radius of Cones
404	R_1		thou	Radius of Driven Surface
400.5	R_0		thou	Radius of Impeller
585	H_3		thou	Half Height of Fins
45	H_2		thou	Height of Bottom of Bearing Surface
40	H_1		thou	Half Height of Driven Surface
80	H_0		thou	Height of Overlapping Surface
10	ϕ		deg	Cone Angle
40	t		thou	Fin Thickness
	ω_v	35.225592	RPM	Valve Rotor Speed
9000	ω_1		RPM	Impeller Speed
0.4	P_r			Blade Tip Proportion of Circumference
	Re_1	0.71806415		Bearing Surface Reynolds Number
	Re_2	555.25833		Drive Reynolds Number < 1400: Laminar
	T	0.00007118	N-m	Drive Torque
	T_1	0.00000028	N-m	Torque at Wall
	T_2	0.00000001	N-m	Torque at Ceiling/Floor
	T_3	0.00007027	N-m	Torque at Bearing Surface
	T_4	0.00000053	N-m	Torque at Sleeve Ends
	μ	0.00128104	N-s/m ²	FLUID CHARACTERISTICS
1.5	ν		cSt	
854	ρ		kg/m ³	
3.14159265	π			

RULE SHEET	For Academic Use Only
Rule	
$T = T_1 + T_2 + T_3 + T_4$	
$T_1 = 4 * R_1^2 * \omega_v * \mu * H_1 * t / a_{23}$	
$T_2 = 4 / 3 * (R_1^3 - R_2^3) * t * \omega_v * \mu / a_{23}$	
$T_3 = \mu * \omega_v * \pi * (R_1^4 - R_2^4) / \sin(\phi) / a_{23}$	
$T_4 = (R_1^4 - R_2^4) * \omega_v * \mu * \pi / a_{23}$	
$T = P_s * 2 * R_1^3 * (\omega_i - \omega_v) * \mu * \pi * H_0 / a_{11}$	
$\mu = p * v$	
$Re_1 = \omega_v * R_1 * a_{23} / v$	
$Re_2 = \omega_v * R_1 * a_{11} / v$	
$a_{11} = R_1 - R_0$	
$R_1 = R_2 + (H_1 - H_2) * \tan(\phi)$	
$R_{2a} = R_2 + (H_1 + a_{23} - H_2) * \tan(\phi)$	
$H_2 = H_1 + a_{23} / \sin(\phi)$	

END

2 2 1 1 9 6

FIN